

# MEC '14

## Redefining the Norm

*University of New Brunswick's*  
Myoelectric Controls/Powered Prosthetics  
Symposium

August 19 – 22, 2014  
Fredericton, New Brunswick, Canada

Symposium Proceedings



**Institute of Biomedical Engineering**  
University of New Brunswick



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University of New Brunswick  
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August 19-22, 2014**

## Welcome to MEC '14

On behalf of the organizing committee and the staff of the Institute of Biomedical Engineering at the University of New Brunswick, we would like to welcome you to **MEC '14**. We are pleased to present a diverse and thought-provoking assortment of scientific papers and discussions relating to the field of upper limb prosthetics. Our theme for this year's symposium is "**Redefining the Norm**", an important topic for clinicians, researchers, and prosthetic manufacturers alike.

This conference has grown significantly over the years, and this year, we have made the jump to a new location to accommodate a larger delegation. We did this with mixed emotions. People who have been attending MEC for many years have valued the close-knit "family" that this conference has created, even in our humble beginnings in classrooms on the University campus. In this move to a new venue, however, we will continue to strive to offer the same quality experience and sense of camaraderie. We are excited to see new and familiar faces at every MEC, and with our growing numbers, we are looking forward to growing our "family".

This year's keynote speakers will highlight many of the advancements in research and technology pertaining to upper limb prosthetics. The keynote speakers are Paul Marasco, Dan Blocka, and Linda Resnik.

**Paul Marasco, PhD**, is the Director of Amputee Services at the VA Medical Centre in Cleveland, Ohio. His research focuses on brain organization and neural plasticity.

**Dan Blocka, BSc, CO(c), FCBC**, is Professor at George Brown College and Past President of the International Society of Prosthetics and Orthotics (ISPO).

**Linda Resnik, PT, PhD**, is a Research Health Scientist at the Providence VA Medical Centre. Her research focuses on the development and testing of rehabilitation outcome measures.

The goal of the symposium is to share information, generate discussion, and inspire future research, which, in the end, will benefit all upper limb amputees.

We hope you will join us for the conference's social events on Tuesday and Thursday, August 19<sup>th</sup> and 21<sup>st</sup>. Social events are an important part of MEC, as they allow time for informal networking and discussion of the day's events, while experiencing some of Fredericton's warm hospitality.

Once again, welcome to **MEC '14**. Please don't hesitate to ask questions to any of our staff members.

**Wendy Hill, BScOT**

**Erik Scheme, PhD, PEng**  
**Co-Chairs MEC '14**

## MEC '14 ORGANIZING COMMITTEE

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## Poster Sessions

There will be two poster sessions: **Session A will be held on Tuesday, August 19** and **Session B on Thursday, August 21**.

On Tuesday, during the morning break and lunch, delegates will be able to view **Session A** posters. At 2:30 p.m., the Poster Session will begin with each presenter having one minute at the podium to describe their work. Presenters will then proceed to their posters, where they will be available until 3:30 p.m. to answer any questions.

On Thursday, we will again follow the same format, allowing Poster Presenters one minute at the podium to begin the Poster Session.



## Social Events

### *Welcome Wine & Cheese Reception*



On Tuesday, August 19, a Wine & Cheese reception, sponsored by **Touch Bionics Inc.**, will be held at Government House. Government House is conveniently located adjacent to the Delta Fredericton on the St. John River. Entertainment will be provided by local award winning fiddler/violinist, Katherine Moller.

### *Banquet Dinner & Dance*

On Thursday, August 21, a Banquet Dinner & Dance, sponsored by **Otto Bock**, will be held at the University's Student Union Building. This will be a casual evening, dining on lobster, salmon, and roast beef, while enjoying music provided by "Jill Harmonic", a local acoustic rock and folk band. Transportation to and from the Delta Fredericton and UNB Campus will be provided.



## **Notice Regarding Audio/Video Recording and Photography of Events**

University of New Brunswick Institute of Biomedical Engineering (UNB IBME) may elect to take photographs of people and events during the MEC'14 Workshops, Symposium, and Networking Events from August 18 to 22, 2014. By attending MEC'14, you agree to permit UNB IBME to use your likeness in these photos in promotion of the conference. The release checked off when registering indicated that you agree that UNB IBME shall be the copyright owner of the photographs and may use and publish these photographs. UNB IBME is released from any and all claims and causes of action that you may have now or in the future based upon or in connection with photographs and UNB IBME's use of the photographs in any manner. All rights granted to UNB IBME by you in the Release are irrevocable and perpetual. You waive all rights to any equitable relief in connection with the Release and the subject matter of the Release.



## **Education Credits**

For each morning and afternoon session, a sign-up sheet will be at the Registration Desk. A *Certificate of Attendance* from IBME will be mailed to delegates in the fall.

## **H. Clifford Chadderton 1919-2013**

***H. Clifford Chadderton will be remembered as a devoted man. For more than 40 years, he tirelessly served the needs of Canadian amputees, both young and old.***

Cliff Chadderton was renowned as Canada's most influential developer of innovative programs and services for war, civilian and child amputees, and as a tireless advocate for veterans.

A D-Day veteran, Mr. Chadderton lost part of his right leg in October 1944 while in command of a company of the Royal Winnipeg Rifles battling for the Scheldt Estuary in Belgium and Holland. He joined The War Amps on returning to Canada in 1944. An active member of the Association, in 1965 he was appointed Executive Secretary (later Chief Executive Officer). Under his leadership, the Association transitioned from a solely veteran oriented organization to a charitable institution which effectively represents all amputees, and particularly child amputees.

Chief among his greatest accomplishments was founding the War Amps internationally-renowned Child Amputee (CHAMP) Program, which assists thousands of amputee children across Canada with the cost of artificial limbs and education, and provides counselling and regional seminars. He also established several other programs including PLAYSAFE, to promote child safety with a "kids-to-kids" approach, Matching Mothers, to bring together new and experienced CHAMP families for advice and support, and JUMPSTART, which ensures that multiple amputee children have the computer skills they need for an independent future.

Prior to his appointment as Executive Secretary at The War Amps in 1965, Cliff Chadderton held several impressive positions: Advisor to the Minister of Labour in veterans' rehabilitation, National Secretary of the Army Benevolent Fund, and Director of the Canadian Army Financial Welfare Program.

As Chairman and, at the time of his passing, Honorary Chairman, of the National Council of Veteran Associations of Canada, Cliff Chadderton had a long list of credits in his struggle for veterans' rights. On behalf of both the NCVA and The War Amps, he appeared before hundreds of tribunals established by Veterans Affairs Canada in the pursuit of innovative pension benefits and allowances on behalf of individual veterans, their families and their children, with particular focus on the prioritization of the seriously disabled veteran.

During Mr. Chadderton's life and career, he received numerous awards. He considered the creation of the CHAMP Program, however, to be his greatest achievement and will stand as his lasting legacy.

## FINANCIAL SUPPORT

The Institute of Biomedical Engineering and the MEC' 14 Organizing Committee would like to recognize the following organizations for their contributions to the symposium:

New Brunswick  
Health Research  
Foundation



Fondation de la  
recherche en santé  
du Nouveau-Brunswick

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**Thank you for making this week a success!**



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## TOWARDS IMPROVING PARTIAL HAND PROSTHESES: THE EFFECTS OF INTRINSIC MUSCLE EMG AND WRIST MOTION ON MYOELECTRIC PATTERN RECOGNITION

Adenike A. Adewuyi<sup>1</sup>, Levi J. Hargrove<sup>2</sup>, Todd A. Kuiken<sup>2</sup>  
*<sup>1</sup>Northwestern University, <sup>2</sup>Rehabilitation Institute of Chicago,*

### ABSTRACT:

In the United States, over 90% of upper limb amputations occur distal to the wrist (i.e., partial-hand amputations) [1]. Recent development of powered and independently-functioning prosthetic digits offers the possibility of a wide range of functional hand grasps and individual finger motions not previously available to partial-hand amputees. Unlike individuals with higher-level amputations, partial-hand amputees usually possess residual intrinsic hand muscles from which additional information-rich EMG data may be extracted and used for improved prosthesis control. Intrinsic muscle EMG may be more robust to changes in wrist kinematics than extrinsic muscle EMG. Here, we (1) quantify the contribution of EMG data from intrinsic hand muscles to pattern recognition-based classification of hand and finger movements and (2) determine how wrist motion affects classification performance. Nine electrode pairs were placed around the proximal and distal forearms of 9 non-amputees and 1 bilateral partial-hand amputee. Twelve electrode pairs were placed on non-amputees' hands and four were placed on the amputee's residual hand. A computer-guided data acquisition system prompted subjects to perform 19 hand grasps and individual finger motions. Subjects also performed 2 hand grasps either with the wrist held in different static positions or while dynamically moving the wrist. A well-characterized pattern recognition system, based on time-domain features and classified by linear discriminant analysis, was trained using (1) extrinsic EMG signals, (2) intrinsic EMG signals, or (3) a combination of extrinsic and intrinsic EMG signals. Training with a combination of extrinsic and intrinsic EMG signals resulted in a reduction in classification error of 67% for hand grasps.

## **SIMULTANEOUS, PROPORTIONAL WRIST AND HAND CONTROL FOR NATURAL, DEXTEROUS MOVEMENTS OF A PHYSICAL PROSTHESIS BY AMPUTEES**

Sebastian Amsüss<sup>1</sup>, Peter Goebel<sup>2</sup>, Bernhard Graimann<sup>3</sup>, Dario Farina<sup>1</sup>

<sup>1</sup>*Dpt. of Neurorehabilitation Engineering, UMG Georg-August University, Göttingen, Germany* <sup>2</sup>*Otto Bock Healthcare Products, Vienna, Austria*

<sup>3</sup>*Otto Bock HealthCare, Duderstadt, Germany*

### ***ABSTRACT***

In this study a simultaneous, proportional wrist control for hand prosthesis was combined with sequential control of hand functions (open, close lateral, close opposition). The system integrates estimations from both a regression and a classification model to one final output for stable single-DOF and also multi-DOF control. The system was implemented in a realtime capable environment. Two amputees were fitted with a custom made prosthesis socket housing 8 electrodes by a professional orthopedic technician. An Otto Bock Michelangelo hand prosthesis with wrist flexion/extension and rotation units was attached to the socket. The subjects were then asked to complete some tests for evaluating the control. Both subjects were able to use simultaneous wrist control and also the hand functions. To our knowledge, this is the first time, that simultaneous, proportional control was tested with amputees in a real life setup using a physical prosthesis.

## REGAINING HIGH FUNCTIONAL, MULTIPLE DEGREES OF FREEDOM HAND CONTROL FOLLOWING BIONIC RECONSTRUCTION

Sebastian Amsüss<sup>1</sup>, Aidan D. Roche<sup>2</sup>, Peter Göbel<sup>3</sup>, Bernhard Graimann<sup>4</sup>,  
Dario Farina<sup>1</sup>, Oskar C. Aszmann<sup>2</sup>

<sup>1</sup>*Dpt. of Neurorehabilitation Engineering, UMG Georg-August University, Göttingen, Germany*

<sup>2</sup>*Christian Doppler Laboratory for Restoration of Extremity Function, Division of Plastic & Reconstructive Surgery, Department of Surgery, Medical University of Vienna, Austria*

<sup>3</sup>*Otto Bock HealthCare Products GmbH, Vienna, Austria*

<sup>4</sup>*Otto Bock Healthcare GmbH, Duderstadt, Germany*

### INTRODUCTION

Avulsion injuries affecting the junctions between the spinal cord and peripheral nerves may result in a complete, flaccid paralysis of the denervated limb. As a result, both the motor and sensory functions may be lost. Nerve transfer operations can be attempted to reinnervate the upper limb, but have poor outcomes for hand function [1], [2]. Amputation of the affected extremity and replacement with an artificial hand (bionic reconstruction) has been proposed as a viable solution for this problem [3]. The hypothesis is that even a simple mechatronic gripping device, only allowing opening and closing, would still yield more functionality than an insensate and atrophic human hand. Therefore, replacing the natural but functionless hand by a prosthesis could increase the quality of life of the patient. Amputations of this kind are referred to as “elective” since the patient and clinical team jointly have to actively decide this step.

In this study, a novel method is proposed to increase the functional outcome of this intervention even further. By acquiring the multichannel electromyographic (EMG) signal of the remnant muscles, which still exhibit voluntary excitability after surgical operations, as well as of reinnervated muscles, the movement intent can be extracted accurately using machine learning methods and be used for controlling a prosthesis. Using this technique, a complex prosthetic system offering various actuated degrees of freedom (DOF) can be controlled intuitively and with little mental effort [4]. The functionality can thus be greatly enhanced with respect to simple gripper prostheses or even the denervated natural hand, thus supporting the drastic step of amputation from a functional point of view.

Over the past decades, several pattern recognition systems have been proposed to solve the problem of prosthetic control by EMG (see [5] for an extensive overview). In this study, the development of a novel method specifically developed for the problem at hand is presented. It enforces maximal discriminability between EMG patterns

with force proportional output, and is thus able to distinguish between subtle pattern differences in suboptimal signal conditions, such as present after the above described nerve injuries. We will show that with this approach, subjects who underwent bionic reconstruction are able to control 2 degrees of freedom of the wrist and 2 grasping types for performing complex functional tasks in an effortless way, regaining high functionality of the hand.

### MATERIALS AND METHODS

The method implemented applies the technique of common spatial patterns (CSP) [6], [7] with specific modifications and extensions. This technique is most often used in signal pre-processing as a spatial raw signal filter, facilitating the extraction of more discriminative signal features in later steps of the signal processing chain. By applying linear combinations, it transfers an input vector  $x \in \mathbb{R}^c$  to a vector  $y \in \mathbb{R}^d$  in component space, in which the signal variance is maximized for one class and minimized for another class.

$$y = W^T x \quad (1)$$

The transformation matrix  $W \in \mathbb{R}^d$  is found by solving the following function

$$\max_W \frac{W^T \Sigma_1 W}{W^T \Sigma_2 W} \quad (2)$$

which can easily be achieved by computing the generalized eigenvalue decomposition of , with being the covariance matrix of class .

Alternatively to using CSP as a spatial filter for raw signals, we propose to apply the above technique to force related features extracted from the EMG. This yields two qualities which are highly desirable in the concept of a control methodology to drive a prosthetic device: The linear combination of features according to (1) results in an output which is proportional to the exerted force and the

optimization criterion in (2) maximizes the discriminability of outputs from different classes. Therefore, the obtained method allows proportional, selective and robust control. In this application of the CSP, the correlation matrices are used instead of the covariance matrices in (2), since centering the input and output data is not desired (on average no input/output).

The obtained estimator, termed CSP-PE (CSP proportional estimator) for the remainder of this work, exhibits beneficial mathematical properties for robust, selective and proportional control for inferring movement intents from EMG to control a prosthetic hand. Since the method is designed for pair-wise class discrimination, a one-versus-one extension was applied.

## EXPERIMENTAL VALIDATION

### Subjects:

The proposed control method was tested on two subjects who had undergone elective amputations between 2010-2011. Patient A suffered from an electrocution accident in 2008. After initial restoration of the left hand, the nerve damage proved to be too substantial and 2 years after the accident, the poorly functioning hand was electively amputated at the transradial level. Patient B was involved in a motorcycle accident, in which he suffered from avulsion injuries to C8/T1. Also in this case, right hand function was poor indicating elective amputation transradially 10 years after the initial accident. The experimental setup of this study on multifunctional prosthetic control in subjects of bionic reconstruction was approved by the local ethics committee and both subjects gave written informed consent to the participation.

### Experimental setup:

For this study, both subjects were fitted with a customized, experimental prosthetic socket, housing 8 bipolar electrodes (13E200AC, Otto Bock HealthCare Products GmbH, Vienna, Austria). A Michelangelo hand prosthesis from the same manufacturer was attached to the sockets. The prosthesis could actively perform the following movements: wrist pronation, wrist supination, wrist flexion, wrist extension, lateral grasp, tripod pinch grip and hand opening. For training the proposed supervised algorithm, EMG data corresponding to this set of motions were recorded while the subjects performed the according phantom limb movements (plus the rest movement). In one run, each movement was repeated at 3 force levels (low, medium, high - visualized to the subject on a computer screen for biofeedback). In total, 3 runs were recorded, in low, medium and high arm positions for increased recognition stability during reaching tasks. The training lasted approximately 15 minutes.

### Functional tests:

In order to assess the control quality achieved with the proposed CSP-PE algorithm, subjects had to complete 3 functional tests of increasing difficulty. First, in the box and blocks test, subjects were asked to transfer as many wooden cubes of 2.5 cm edge length from one box to another within 60 s [8]. It thus only required simple opening and closing of the hand (one degree of freedom). In the clothes pin test, subjects had to pick up a clothes pin from a horizontal bar, rotate it, and place it on a vertical bar [9] (two degrees of freedom). The last test was the most complicated and required control over all the 3.5 degrees of freedom of the prosthesis. The task was to pick up a flat wooden block from a shelf at shoulder level, rotate it, and place it like a book in a shelf at waist level, release and grab it again with the lateral grasp, and put it back down in its initial position. This test is referred to as block turn test and cannot be executed with classic grippers since it requires the activation of all the degrees of freedom of the experimental Michelangelo hand. In both the clothes pin and the block turn test, the time required for successful completion was evaluated. All tests were repeated 3 times and the average was reported. Both subjects completed the test scenario with the proposed CSP-PE method and the experimental prosthesis. Prior to the amputation, a functional assessment of the denervated hand was performed.

## RESULTS

In the initial session before the amputation, the subjects were asked to attempt various tasks of pick-and-place of the Action Research Arm Test (ARAT, [10]) and the Southampton Hand Assessment Procedure (SHAP, [11]). [3]. Different objects such as spheres and cubes of various sizes should be picked up and transferred to another location – however both subjects performed dismally during these tests, showing limited or absent hand function.

With the multifunctional test prosthesis and the CSP-PE control method implemented as described above, both subjects were able to complete all 3 tests of varying difficulties in this study. Simple gripping functionality, as required in the box and blocks test, was equally achieved as the control over more complex movements, like the ones necessary for the other two tests conducted in this study. The direct control of all functions by performing the desired phantom limb movements was described as very intuitive by the subjects and required little training times. Amputee A transferred on average 27 blocks in 60 s, required 13 s for completing the clothes pin test and 31 s for the block turn test. Amputee B achieved a score of 24 blocks, 44.7 s for the clothes pin transfers and 21 s for the block turn test. None of these tests could have been completed by the subjects with their denervated hands and it is also worth noting that the investigated tasks of this study would be difficult or

impossible to complete with classic myocontrol systems, since the direct controllability is limited to two functions in these devices. The tests in this study required control over up to 7 functions (3.5 degrees of freedom) in daily-life tasks, which would not be controllable with conventional control schemes in an intuitive manner.

## DISCUSSION AND FUTURE WORK

A novel prosthetic control scheme was investigated in the context of bionic reconstruction. While the amputation of an anatomically intact extremity may appear as an extreme measure, the complete lack of functionality of the limb and the possibility to recover such functionality justify this intervention. In this study, it was shown that by using state of the art prosthetic hardware combined with an innovative control paradigm, the lost functionality can effectively be restored at a high level. This implies complex wrist control and the most relevant functional grip tasks.

The intuitive control of a prosthetic hand with many degrees of freedom, as explored in this work, results in dexterous, natural movements, which allows the successful integration of the device into the patient's body image and daily life usage. Almost any task of typical activities of daily living can be executed, as proved by tests that included all the degrees of freedom of the prosthetic hand. High functionality of course justifies in an even stronger way the elective amputation procedure. To this end, we are currently intensively working on facilitating also simultaneous control over multiple degrees of freedom of the prosthesis. This important stride will additionally increase the naturalness of the movements substantially, and thus constitutes an important step towards the goal of optimal bionic reconstruction.

## ACKNOWLEDGEMENT

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## A COMPARISON OF THE “PERCEPTION OF DISABILITY” IN BILATERAL PROSTHETIC WEARERS, BILATERAL HAND TRANSPLANTS AND BILATERAL TOE-TO-HAND TRANSFERS

Diane Atkins  
*Baylor College of Medicine*

### **ABSTRACT:**

#### **Introduction:**

To date there has never been a national or international effort designed to compare the “disability experience” of individuals who have been fit with advanced electric multi-articulating hands, and those who have experienced hand transplantations or toe-to-hand transfers. The purpose of this study is to present the experience of the bilateral amputee, and their “perception of disability”, as it relates to these interventions. As significant, functional achievements have been possible with advanced electric prosthetic hands, as well as with hand transplantation surgery, the self-reported, subjective experience of the individual who has lost both hands is an equally important outcome to review.

#### **Method:**

The subject population included 3 study groups, 3 bilateral transradial users of electric multi-articulating hands, 4 bilateral hand transplant subjects, and 1 bilateral multiple toe-to-hand transfer subject. Each subject was asked to complete the Disabilities of the Arm, Shoulder and Hand (DASH) questionnaire.

#### **Results:**

Because of the differences in the subject group size, the data was calculated by Cohen’s d-effect size, to represent the group differences. A large effect size ( $d = .91$ ) was calculated in the DASH scores between the bilateral hand transplants and bilateral prosthetic hand users. Bilateral transradial users of electric multi-articulating hands scored a lower perception of disability (mean = 39.83) when compared to individuals who had undergone bilateral multiple hand transplant surgery (mean = 53.25). The DASH score of the individual with bilateral toe-to-hand transfers was the least at 27.

#### **Conclusion:**

As dramatic advances are being made in the field of electric multi-articulating hand technology, the field of reconstructive microsurgery and hand transplantations has also produced significant functional results. While “success” is measured objectively in a variety of outcome measures, the subjective experience is important to measure as well.

Although the functional outcomes are similar in these subject groups, the individuals who utilized electric multi-articulating hands perceived themselves as “less disabled” when compared to the hand transplant subjects. The individual with bilateral multiple toe-to-hand transfers considered himself the “least disabled” of any of the 8 subjects. This preliminary study requires further investigation as advances and options become available for the individual who has lost both hands, so that a prospective patient can compare not only the objective outcomes, but the subjective experience as well.

## REDEFINING THE NORMS OF THE CLINIC ENVIRONMENT: THERAPY “HITS THE ROAD”

Kerstin Baun MPH, OTR/L, MacJulian Lang CPO, FAAOP, Christopher Bollinger, MOT, OTR/L  
*Advanced Arm Dynamics, Northwest and Northeast Centres of Excellence*

### ABSTRACT

Many elements factor into the success of an individual presenting with upper limb absence or amputation. Research and development of new technology that provides ever improving component options and features is one. Optimal adjustment, correct fit, myo testing, training and simulation are others [5]. Helena Burger in her MEC keynote address in 2011 stated that teamwork was the key to success [4]. Foundational to all of these important factors is the environment—the context in which our patients use their prostheses. The impact of the environment on function has been established in occupational therapy literature for decades [1,8,9,10,11,12]. It is recognized by the World Health Organization as a key component of the international classification of functioning, disability and health (ICF). Learning theories speak to its importance as it influences our ability to attend to, process and integrate new understanding.

How does this impact our therapeutic interventions? Clinic environments are the most common setting for prosthetic services and training. Early intervention in this setting is very appropriate for ensuring prosthetic fit and developing basic prosthetic skill. Complexity can be controlled and the patient is able to focus on discreet objects and tasks. As basic control is consolidated, skill generalization to more challenging environments is indicated. The patient’s ability to function well with their prosthesis in natural environments—those in which the patient needs to use their prosthesis—is the ultimate indicator of success. Regardless of a client’s age and the nature of their prosthesis, ensuring that they are able to integrate it into their daily lives builds confidence and the likelihood of long-term prosthetic acceptance. Case studies are presented which demonstrate use of natural environments and their impact on the prosthetic training process.

### ENVIRONMENTAL CONSIDERATIONS

#### Environment—A legal and therapeutic perspective

According to the World Health Organization’s International Classification of Functioning, Disability and Health, “environmental factors make up the physical,

social and attitudinal environment in which people live and conduct their lives”. This definition includes the natural and built environment such as a client’s home, their community and workplaces. It includes the climate and geography of where they live and travel as well as their family, friends and colleagues. Even health professionals are considered part of the environment [2]. Furthermore, the environment interacts with body functions/structures, participation and activity. It can impact these either positively or negatively (Figure 1).

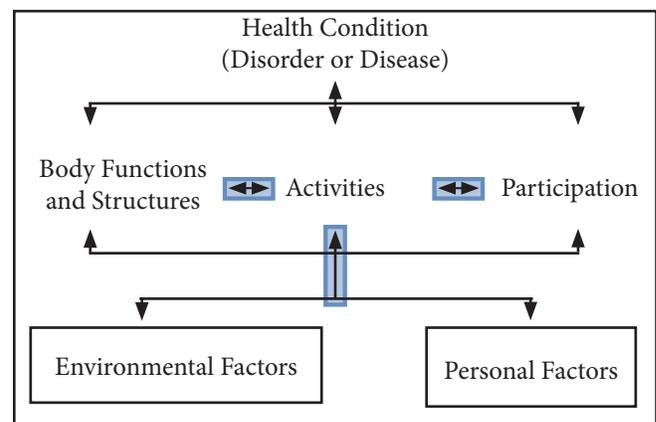


Figure 1: Interactions between the components of ICF

In the United States, according to the Americans with Disabilities Act of 1990, assessment and intervention by providers is to take place in the natural and least restrictive environments that support the client’s successful participation. The AOTA position paper *Use of Environments and Context to Support Health and Participation in Occupations* states that “interventions and recommendations focus on selecting and using environment and contexts that are congruent with clients’ needs and maximize participation in daily life occupations.” [1] For new prosthetic users, this may mean that the best environment—the least restrictive environment—for learning how to use their prosthesis may be a quiet clinic room where they are encouraged to try out and explore their new device with simple, repetitive tasks that build consistency, skill and confidence. However, as the client gains competence, the environment can be “engineered” or chosen to provide less structure, more challenges and fewer

supports with the goal of increasing client function. As abilities become consistent and predictable, the environment most conducive to skill consolidation and generalization should be similar to the environment in which they will ultimately be using their prosthesis. If a patient plans to use their prosthesis in a work environment, ensuring that they are able to generalize their newly developed skills to that setting is key to building confidence. For an outdoor enthusiast, a trial using their prosthesis to engage in their recreational pursuit will be the ultimate test of its success. Although a clinic may be a good starting place, eventually the client will want to ensure that given natural environmental factors, they feel comfortable using their new prosthesis. In addition, by shifting the treatment environment to natural contexts, social demands of an activity will be affected. This provides opportunity for the therapist to observe and assist the client in developing psychosocial skills related to prosthetic use in the community. By providing training in the appropriate environment, we can ensure that the client will be successful.

Kielhofner, the developer of the Model of Human Occupation, conceptualizes the environment as 4 concentric layers (Figure 2). “The core layer consists of objects, the materials and artifacts of daily life. The next layer corresponds to tasks, or the projects and activities that comprise play, work and self-care, and determine one’s use of objects. Surrounding the layer of tasks are social groups and organizations. These groups delineate certain roles, relationships between roles and essential tasks necessary to the group’s functioning. The final layer, culture, consists of the beliefs that tie together and govern the actions of groups of people. Together, these layers represent an environmental hierarchy which influences both the decisions to encounter ones surroundings and subsequent performances in these surroundings.” [10]

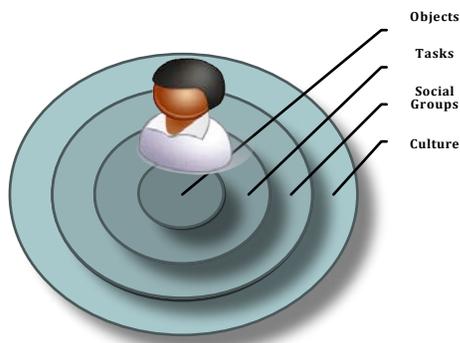


Figure 2: Environmental layers and the human system

In a clinic, we can quite easily and appropriately address the object and task layers of the environment. For example, we can have a patient practice packing (task) their computer into

its carrying case (objects). We may even be able to address some social and cultural aspects pertinent to prosthetic use. Building on the prior example, we can role play discussions with co-workers that a patient may encounter when returning to work. However, for the patient to truly acclimate and excel in use of their prosthesis in their work environment, they will be best served with a therapy evaluation and intervention in that setting. Objects and tasks may be different from those in a clinic setting but more importantly, often the individual must integrate function with their prosthesis amid peer interactions to meet the expectations of their work role. This adds complexity and challenge beyond simply handling items and completing tasks.

#### Educational Impact of the Environment

Most practitioners specializing in the field of upper limb prosthetics and rehabilitation may not view themselves as educators. However, we are providing education and teaching our patients in almost every interaction that we have with them. In working with children, we know that play based learning is often the best approach for skill development and prosthesis use. However, what do adults need to learn best? Adult learning theories suggest that the environment is an important component in developing new skills and understanding as an adult. This point of view is also supported by principles of neuroplasticity.

Adult learning theories (including andragogy, self-directed learning, transformational learning and experiential learning) propose that adults are internally motivated and bring life experiences and knowledge to bear on new learning experiences. They are goal oriented, relevancy oriented and practical.[6,7] Addressing these principles through not only selecting appropriate learning objects and tasks but also physical and social environments, expedites our client’s learning process and improves the experience for them. When ensuring that our interventions are relevant, goal directed and applicable to “real life”, clients will feel confident in their ability to successfully use their prosthesis in the environment(s) for which they need it. Optimizing practice through judicious and creative use of the environment can speed skill acquisition and patient learning.

Although full discussion is beyond the scope of this paper, it is worth noting that neuroplasticity, “the brain’s lifelong ability to reorganize neural networks as a result of new experiences”[3], is the basis of all learning. The environment plays a key role in influencing plasticity; the brain is shaped not only by genetics but also by the characteristics of a person’s environment and what that person does in their environment [14]. Sensory demands of the environment can have a tremendous effect on function. From basic inputs such as noise level and visual distractions to complex elements such as emotional connections to person

or places, function will be affected. The enriched nature of real life environments necessitates sensory filtering to allow for focus on salient elements—in this case, prosthetic function. Furthermore, natural environments are often emotionally loaded and activate limbic system involvement in task completion [3].

## CASE STUDIES

### Examples of Therapeutic Use of the Environment

Practitioners in the field of upper extremity prosthetics routinely work with their clients in a clinic setting to ensure fit and function. For early training activities, the clinic is often the ideal setting for client's to gain ability and skill. Once basic control is mastered, we can simulate work or ADL activities in the clinic setting through use of a variety of items to mimic tasks in the client's natural environment. However, based on the legal, therapeutic and educational elements discussed above, it is in our client's best interest to explore function of their prosthesis in the most natural environment possible. Below are examples of therapeutic use of the environment to consolidate prosthetic skill and ensure successful skill generalization.

Case Study #1: Our patient, an avid long distance swimmer and bicycling enthusiast was struck by a boat and sustained right dominant side transhumeral and left partial hand amputations. He desired a prosthesis that would allow him to return to cycling safely. An activity specific prosthesis with seal-in liner and elevated vacuum suspension with "The Arm" bike specific elbow and quick disconnect release handle adapter were designed and fit. At completion of fitting, he was initially able to try out the new biking arm on a bike trainer using his bicycle which he'd brought along.



Figure 3: The bike arm in action

This allowed for socket and alignment changes and provided a safe way to assess his comfort and build confidence with this device. However, he enjoys cycling in his community and needed to be sure that his prosthesis would hold up in road conditions and be comfortable for longer rides. We proposed and completed a 12 km test ride that included dealing with traffic, hills, sharp turns and obstacles such as railroad tracks. The ride itself was a graded activity, beginning with a flat stretch of road with wide shoulder and low traffic to ensure initial success. Later in the ride more difficult elements were introduced, all of which he mastered. By completing this community ride, we were able to determine that his prosthesis fit comfortably on a variety of terrain, that he could easily navigate obstacles and maintain symmetrical posture with minimal compensatory movements. We also learned that his left partial hand fatigued quickly and that this would be the limiting factor that he would need to keep in mind when planning rides. Upon returning home, he stated that he completed several of his old training route rides and that he was very satisfied with the performance of his prosthesis. The DASH recreational activity component scores showed dramatic improvement post fitting and training, from 87.5 prior to fitting to 50 post fitting (general population norms for males: mean 9.17, SD 20.72 [13]).

Case Study #2: An energetic, active toddler, our patient was fit with bilateral myoelectric prostheses in October of 2013. He uses Ottobock system 2000 hands in electric voluntary opening (or "cookie crusher") mode and custom designed passive elbow joints. After initial fitting and training in our office, he was seen in his home environment in Spokane Washington to evaluate skill carry over. Additionally, on subsequent visits to the office for adjustments we went to a nearby park to play. Use in this environment resulted in an extended wear time with functional use on several playground elements.



Figure 4: Prosthetic use practice on the playground.

In addition to everyone enjoying the park environment, his parents gained strategies for how to set him up successfully on equipment they may encounter at parks near their home. They also learned current limitations including their child's dislike of swings and his need for repeated assist prior to independent follow through. For example, to grasp onto the merry go round pictured below, we used the "mommy switch" on multiple tries before the patient initiated volitional opening to grasp the bar.

Although this park setting was very quiet and we were the only ones present, social demands of typical playgrounds were discussed with the parents. Strategies for dealing with unwanted questions and pushy peers were addressed. Speaking in terms of Kielhofner's model of the environment, this shifted the focus from objects and tasks to the social groups and cultural layers.

**Case Study #3:** An individual with quadramembral amputations, our patient moved internationally to take on a new work position and live in close proximity to our clinic. He had mastered standing and walking on his lower limb prostheses but was able to use only passive upper limb prostheses due to fit and function difficulties of his myoelectric systems which he had received prior to his move. He was dependent in donning and doffing of all prostheses. At our clinic, he was fit with upper limb prostheses which he can independently don and doff with ease and training in use of his previously fit iLimb Ultra hands and new ETDS commenced. We began in our clinic with simple tasks. The patient made tremendous gains very quickly including abilities in self-feeding, writing on vertical and horizontal surfaces (an important workplace skill), opening containers

and managing small manipulatives. He required occasional cueing for grip selection and repositioning. However, the patient reported less success at home and not wearing his myoelectric prostheses to work at all. Home and work place evaluations and intervention were clearly indicated.

In the home environment, he was instructed in functional mobility and adaptive devices were discussed which would maximize his independence. Positioning at his desk, where he often works from home, was evaluated and recommendations were provided to decrease compensatory movement and allow for extended prosthetic use.

An initial workplace evaluation revealed that the client would benefit from multiple ergonomic improvements in his immediate workspace. Adaptive equipment and technique options were discussed specifically related to his unique workplace demands. Most importantly, this visit highlighted increased difficulty with myoelectric control while completing tasks in the work setting. The additional environmental demands of the office appeared to decrease this individual's focus and accuracy. The workplace evaluation revealed that there was a lack of generalization of skill from the clinic to the client's work environment. Further training and possible programming changes are indicated including the addition of social demands and challenging physical settings to foster skill transfer to the workplace. Training and evaluation in the workplace setting will optimize success and outcome measures will be used to evaluate effectiveness of intervention.

## CONCLUSION

Multiple factors are involved in our patient's ability to succeed with their prosthesis. The environment—physical, social and attitudinal—impacts prosthetic outcomes. By ensuring skill generalization from the clinic to the natural environment in which the patient must function with their prosthesis, we build our patient's confidence and can address any issues that could limit their abilities. By engaging clients in tasks in the most natural setting, we move beyond the immediate environmental layers of objects and tasks and factor in the social and cultural demands that impact prosthetic use. The power of the environment to influence success as described by learning theories, the World Health Organization and therapy models, along with the legal directive to provide intervention in least restrictive settings compels us to venture beyond the clinic and empower our patients to succeed in their natural environments.

## ACKNOWLEDGEMENTS

Our sincere thanks to the patients that agreed to share their stories and pictures.

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## A SUBJECTIVE VIEW TO THE QUESTION OF SENSATION VERSUS FUNCTION

Claribell Bayona  
*NYU Langone Medical Center*

### ***ABSTRACT***

#### **Introduction :**

Sense of touch represents an important component during the performance of any functional activity. The loss of sensation in an upper extremity will impact the use of the arm and hand dexterity. There is limited data that has been collected on the subjective experience of individuals with upper limb loss and hand transplantations regarding how they feel about sensation versus function in their affected limb. As technology and surgical procedures continue to advance it is important to consider these experiences and their effect on outcomes.

#### **Methods:**

Twenty-seven upper limb subjects participated in a study designed to test performance outcomes using objective and subjective measures. The group consisted of 15 myoelectric prosthetic users, 6 toe-to-hand transfer recipients and 6 hand transplant recipients. A questionnaire was developed that included 5 questions relating to their thoughts on having sensation versus having function in their affected side(s). The questionnaire presented 4 questions with scaled responses and 1 question asking for a description of specific experiences where they felt that one was more important than another.

#### **Results:**

The responses for the 4 scaled questions were compared between the 3 subject groups. The myoelectric users and hand transplant recipients preferred having function over having the sense of touch 80% and 100% respectively. The hand transplant group values the return of sensation but does not choose it over overall function. When asked for their comments on their feeling on sensation and function during activities, the myoelectric group reflects on having sensation and being able to take off their prosthesis and “feel” however when it comes to choosing one over the other, this group prefers having function versus having sensation. The toe to hand group was divided in choosing either sensation or function, however they were clearly the most satisfied with the function of their transplanted toes even if they did not have a full sense of touch.

#### **Conclusion:**

The responses of the 3 subject groups reflect that function is very important and often preferred over having sensation, especially if function would be compromised. As prosthetic technology and surgical procedures become more sophisticated it will be important to take into consideration the importance of sensation, and continue this dialogue with users/recipients in our attempts to improve hand surgery procedures, prosthetic technology and therapeutic protocols.

## **A COMPARISON OF PATTERN RECOGNITION AND TARGETED MUSCLE REINNERVATION (TMR) CONTROL SCHEMES USING COMMERCIALLY AVAILABLE SYSTEMS: A CASE STUDY**

David Beachler, Caitlin Dennison  
*Walter Reed National Military Medical Center*

### **ABSTRACT:**

Myoelectric upper extremity prosthetics have been used for decades. Many recent improvements with myoelectric elbows, wrists and hand systems have made upper extremity prostheses more acceptable to patients. As these systems become more advanced and articulating joints increase, controlling them intuitively can be challenging when using conventional two site EMG signal processing. Multiple site EMG control schemes have brought intuitive control to the forefront of prosthetics. Both TMR and pattern recognition control schemes are alternative avenues to using myoelectric prosthetics. These control schemes are designed to improve the patient's ability to intuitively control their prosthesis.

This case study looks at the recorded functional outcome measures of one unilateral transhumeral amputee after receiving Targeted Muscle Reinnervation (TMR) surgery. This study compares the use of a commercially available myoelectric prostheses specifically designed for a TMR simultaneous control scheme versus a commercially available pattern recognition sequential control scheme. In both prosthetic fittings, the same socket, elbow, wrist rotator and terminal devices were used. Only the control schemes were changed. The patient was first fit with the TMR control scheme, trained in occupational therapy for nine weeks and functionally assessed using the box and blocks task, the nine hole peg test, the South Hampton Assessment Procedure (SHAP) and the Jebsen-Taylor hand function test. At nine weeks the second control scheme, pattern recognition, was introduced to the patient. The patient was trained for an additional nine weeks in occupational therapy and was tested again using the same functional assessments.

## GRASP AND FORCE BASED TAXONOMY OF SPLIT-HOOK PROSTHETIC TERMINAL DEVICES

Joseph T. Belter, Bo C. Reynolds, and Aaron M. Dollar  
*Yale University, Department of Mechanical Engineering and Material Science*

### INTRODUCTION

Although hundreds of prosthetic terminal devices are available, ranging from task specific devices to almost indiscernible cosmetic replicas of human hands [1], the voluntary opening split-hook (shown in Fig. 1) is widely accepted as the most commonly utilized functional terminal device [2]. Even with advanced multi-fingered and multi-DOF myoelectric terminal devices now available on the market, many prosthetists and amputees still turn to the body-powered split-hook due to its proven robustness, performance, low cost, and light weight [3], and many amputees who have learned to utilize the split hook reluctant to switch to a newer technology with a long learning curve.

In this paper, we present two sub-classifications of split-hook prosthesis use to enable a better understanding of the functional capabilities and usage of this class of terminal devices. While the specific pose or shape of the hook with relation to the object may help to define the grasp type, it was also observed that within different grasp types there are multiple limitations to the forces that can be exerted on the object. For this reason, we separated our look at the split-hook into two taxonomies. The first is based on the nature of contact with an object, while the second is based on the types of force exertion that can be utilized.

There has been relatively little work related to thorough classifications of split-hook usage that focuses on the function of the device instead of the task or object. While the majority of studies on prosthesis use to date have involved primarily written surveys to users (e.g. [4]), others have supplemented with follow up visits to the amputee's home to observe the use of the device in the normal and unstructured environment [5]. As far as the authors are aware, all of these studies focus on high-level function (e.g. usage frequency, classes of tasks) without detail on the specific ways in which the device is used. The most relevant study that we were able to identify was performed by Fraser in 1998, in which 66 amputees were videotaped at their homes using their terminal device while performing a set of common everyday tasks [6]. The video was then analyzed and each action involving the terminal device was categorized into a manipulative or non-manipulative category. The secondary categories included,

grip, release, hold, transfer, support, steady, etc. We are interested in studying the specifics of split-hook type terminal device usage with enough detail to identify specific design shortcomings and areas that can be improved, and therefore would like a detailed classification scheme.

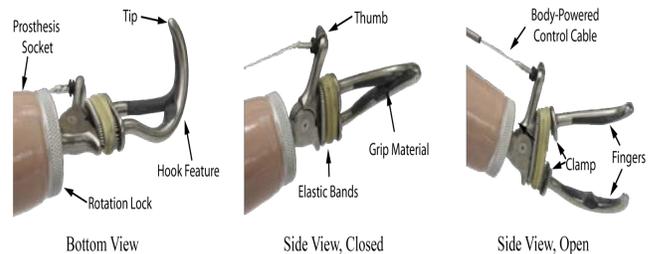


Figure 1: Standard Hosmer Body-powered split-hook

### METHODS

#### Terminology

In order to remove ambiguity, we have defined the terms that will be used within the taxonomy.

- Prehensile – The object is intended to be fully supported within the terminal device without the requirement of gravity to hold the object in place.
- Contact – Point of interaction between the terminal device and the environment.
- Grip Security – Ability for the grasp to be held with increasing amounts of external disturbances applied to the object.
- External Force – A force applied to the external world that would require an equal force to support the terminal device. These types of loads would include pushing, pulling, or lifting.
- Internal Force – A force that is created between the fingers of the terminal device. These types of forces include pinching and grasping in a prehensile fashion.

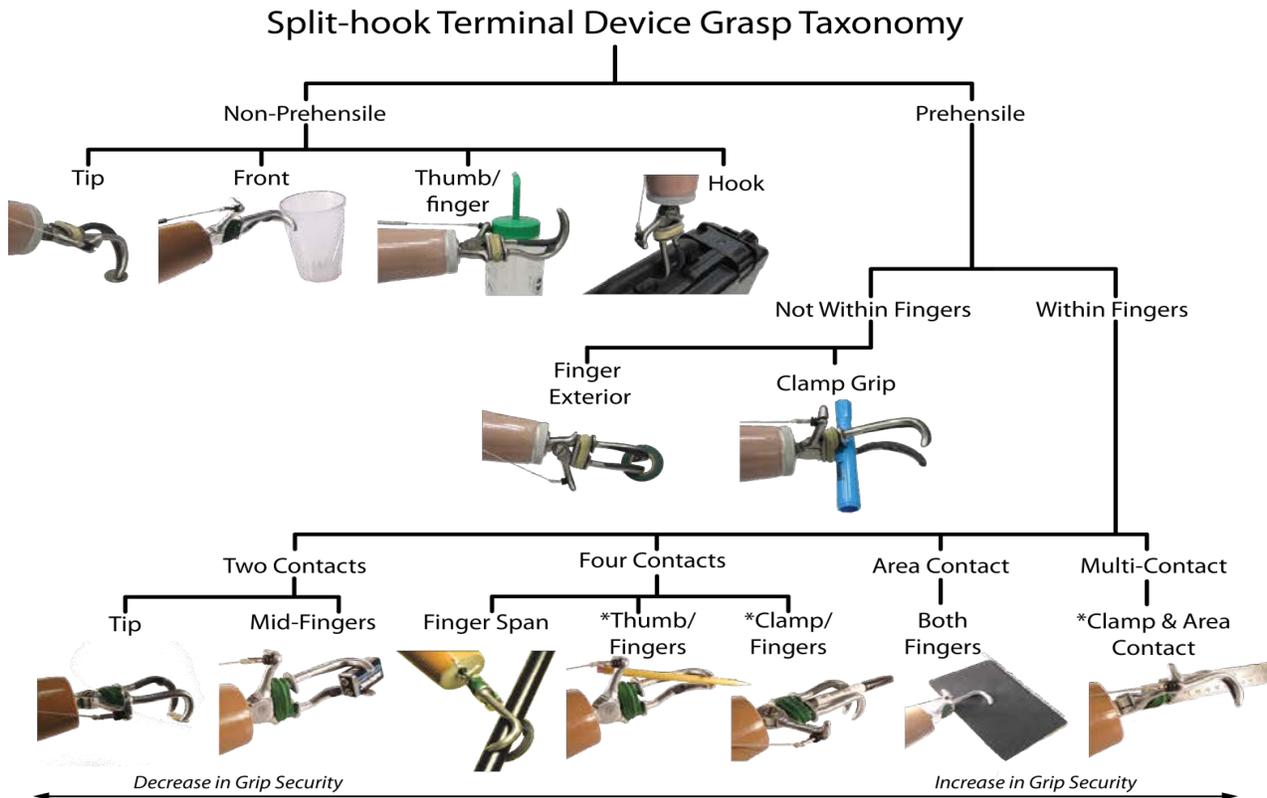


Figure 2: The Split-Hook grasp taxonomy shows all observed uses for the voluntary-opening split-hook as controlled by a body-powered control cable and harness. \*Grasps only observed by placing the object within the hook using the able hand.

### Method of Establishing Taxonomies

The presented taxonomies were generated after thorough discussions between the authors, one of which is a 10-year user of a Hosmer Dorrance 5X-Ti, voluntary-opening body-powered split-hook. In addition, over 5 hours of video footage was acquired of that author's hook use during numerous activities, including hygienic tasks, laundry, cooking/eating meals, and standard office style work. The video footage was gathered using a head mounted webcam and video recording system similar to the system used in [7], and the user was recorded while in his natural home setting. This protocol was approved by Yale University's Institutional Review Board. The video data was studied by all the authors and used to validate the taxonomies, with the authors categorizing each grasp or interaction seen in the video according to the developed classification schemes to ensure that no additional categories were required. The examples shown in Fig. 2 and Fig. 3 are based on those seen within the recorded video.

## RESULTS

### Grasp-based Split-hook Taxonomy

The grasp taxonomy (Fig. 2) attempts to capture any way the split-hook is used to interact with the environment.

We first divided the interactions into a non-prehensile or prehensile category, similarly to the non-manipulative and manipulative category used by [7]. The non-prehensile category was then subdivided by the locations of contact on the hook. It was shown that almost all surfaces of the hook including the tip, front, thumb, and inner surface of the hook were used in a non-prehensile manner. These types of interactions were used to provide stabilizing actions and often in assistance to the able hand. In standard presentations of terminal device grasping capabilities, these important aspects are often overlooked. The rigidity of the hook and the shape help it to perform the non-prehensile type interactions in a predictable manner.

The prehensile category is first sub-divided based on whether the fingers are used or not. Although the fingers are the main prehensile feature of the split-hook, grasping objects with the outer surface of the fingers was also observed (e.g. the "finger/exterior" grasp, with a roll of tape), as well as within the "clamp". For prehensile grasps with contacts occurring between the fingers, they were again sub-divided based on the number of contacts between the object and the split-hook. Generally, these are limited to two contacts (usually for "precision grasps" on small objects), four

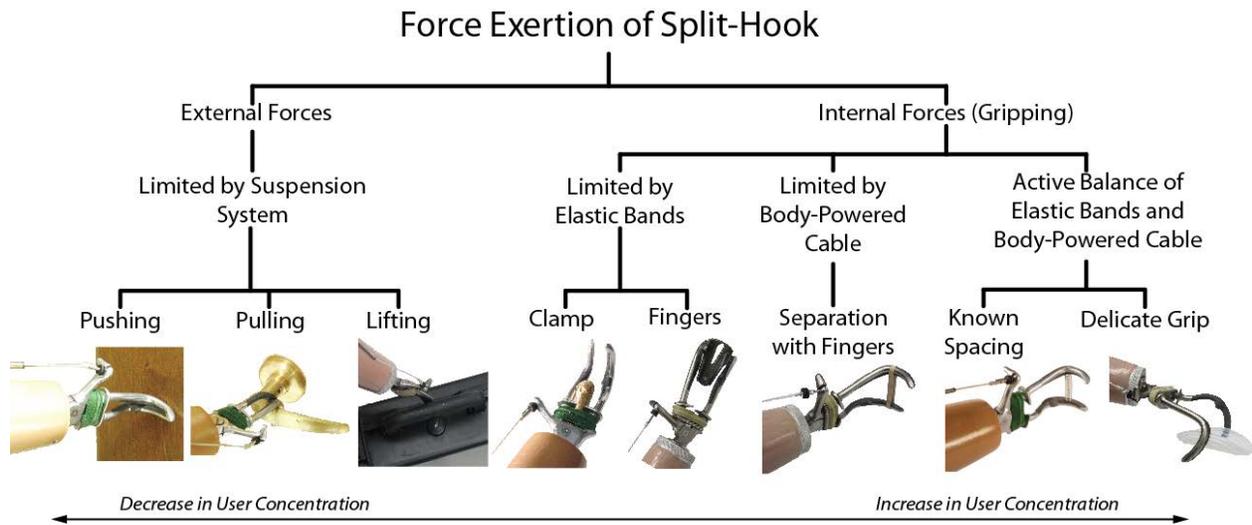


Figure 3: The split-hook force taxonomy shows the methods of force exertion utilized by split-hook users. The organization is based on the limitation to the maximum force exerted and the amount of user concentration required to achieve the task.

contacts (often for long, thin objects, and often involving both the fingers and an additional feature such as the “thumb”), or area contacts, where a large portion of the entire grip material surface was in contact with the object, which is sometimes used in combination with the “clamp”. The entire taxonomy is organized from left to right based on an increase in object grasp security which is strongly correlated to the number of contacts between the hook and the object.

Note that a few particular grasps, marked with an asterisk in Fig. 2, typically cannot be achieved without the use of the contralateral hand or terminal device. Partly due to their complexity and the geometry of the objects, these grasps generally require proper position of the object with respect to the split-hook and were not achievable when trying to pick up an object directly from a table or drawer. For example, the common “Thumb/Fingers” grasp, used to hold an eating or writing utensil, requires specific positioning of the utensil that is not generally achievable without placing it in the grip using the other limb.

#### Force-based Split-hook Taxonomy

The split-hook force exertion taxonomy (Fig. 3) shows the different methods and limitation of exerting load on the environment. The first categorization separates the force exertion based on external or internal forces. This division is very similar to the division of non-prehensile or prehensile functions since all prehensile grasps require internal forces. The second sub-categorization is based on the factor that limits the amount of force that can be exerted on the environment. This includes the prosthesis suspension system, the elastic bands, and the control cable.

The suspension system consists of the entire apparatus used to fixate the prosthesis to the amputee, including the prosthetic socket and body powered harness. Since a trade-off exists between how tightly the prosthesis is attached and the level of comfort, the amount of loading during a pushing, pulling, or lifting is limited. The achievable loads within this category are also limited based on the strength of the user.

The internal gripping forces of the voluntary-opening split hook are generally determined by the elastic bands (the number and stiffness of which are chosen by the user according to their preferences, needs, and abilities), but can also be modulated by a delicate balance of tension in the body-powered control cable and the elastic bands. When an object is grasped and all tension is released from the control cable, the maximum force exertion is dependent on the elastic bands, the location of the object within the hook, and the size of the object. For a standard amount of elastic band tension, the grip force at the fingers is between 45-70 N for 1-3 cm sized objects. The grip force of a 1 cm sized object in the clamp can reach 140 N. In Fig. 3, the hook is shown crushing a peanut within the clamp feature. The user can also overpower the strength of the elastic bands and exert a load with the outside of the fingers. The limiting factor in this type of grasp is the tension in the body powered control cable, minus the force required to open the fingers due to the elastic bands, as shown in Fig. 3 separating a stiff rubber band.

Often more delicate and fine movements of the fingers are required. In this case, the user can alter the position and force on the control cable such that a much smaller grip force is placed on the object than that of the elastic bands alone. Ultimately, the grip force exerted on the object is

the difference between the load in the control cable and the tension stored in the elastic bands. With a high degree of concentration, the user is able to grasp delicate objects like a small spring, or a compliant drink lid, as shown in Fig. 3 (far right). However, due to the required tension on the cable, it can be difficult for the user to keep that pose while moving their arm in space, due to body-powered harness.

The force-based taxonomy is sorted from left to right by the level of user concentration required to perform each category of force exertion.

## DISCUSSION

The two taxonomies presented in Fig. 2 and 3 show that the split-hook is used to perform a wide array of grasp and interaction types. Although the grasp taxonomy was not based on the objects being grasped, it can be seen that the achievable prehensile grasps encompass a large set of common object sizes and geometries. Furthermore, the array of achievable grasp types hold many objects in multiple orientations, depending on the requirements of the task intended to be performed.

### Object Acquisition versus Holding

A major feature of grasp utilization relates to the difference in acquisition methods for each particular grasp type. Some of the grasps are unachievable without careful repositioning of the object prior to closing the split-hook. For example, the four-contact, thumb/finger grip, as seen in Fig. 2 (bottom) securing a fork, is only achieved if the able hand positions the fork within the hook. This is the case for most grasp types involving multiple contacts with the thumb or clamp. This is observed in human hands where a separate grasp type is used to pick up the object before some type of within-hand manipulation transitions the object into a more stable or useful grasp. Although in the case of the split-hook, no within hand manipulation can take place. This shows that not only is the grasp important, but the method of acquiring it in order to determine its relevance and practical usage to both unilateral and bilateral amputees doing unilateral and bilateral tasks.

The necessity for careful repositioning prior to achieving a useful grasp position may lead to the high incidence of non-prehensile uses of the split-hook or other terminal devices. In fact, the most common grasps seen in the video of the split-hook user were non-prehensile grasps, with within-finger two-contact grasps the next most common. Similar results were found by Fraser [7]. If a unilateral amputee requires the use of the able hand to assist the terminal device in holding or using an object, then it may be easier to simply perform the task completely with the able hand. Perhaps since non-prehensile functions do not require this additional effort and

require very little user concentration, they are performed more frequently.

One major limitation to the capabilities of the split hook is the lack of a robust medium- and large-diameter power grasp (often called “wrap grasps” as they involve a large amount of contact between grasper and object to support large loads). These power grasps make up a large portion of grasps for human hands, yet are not capable of being performed using the standard split-hook. The lack of a large power grasp was made up for by utilizing the split-hook wearers’ body and often sandwiching large cylindrical items between the hook or prosthesis socket and the chest or stomach.

When looking at the force exertion capabilities of the split-hook, it is clear that there are limitations based on the nature of any voluntary opening terminal device, the most important of which is the inability to close tighter on an object than the limitation imposed by the elastic bands.

## CONCLUSIONS

Using the split-hook taxonomy, we can easily compare the functional use of the hook with both human hand and other prosthetic hand use. Although it has the ability to perform a wide variety of functions, it lacks in the ability to acquire all the grasps without assistance from the able hand and the ability to exert forces is limited by the nature of the voluntary opening control strategy. We believe that the presented taxonomies can allow for more detailed grasp analysis in order to compare and better understand the function and overall utility of numerous terminal devices. Since many non-prehensile functions of the terminal device were observed we believe that researchers should look toward understanding the entire utility of a terminal device instead of just a wide range of grasped objects or hand postures.

## ACKNOWLEDGEMENTS

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## PROGRESS TOWARDS THE YALE BODY-POWERED ANTHROPOMORPHIC PROSTHETIC HAND, MECHANICAL COUPLING METHODS

Joseph T. Belter, and Aaron M. Dollar

*Yale University, Department of Mechanical Engineering and Material Science*

### INTRODUCTION

The current state-of-the-art in anthropomorphic prosthetic hands including the iLimb, Bebionic, and Vincent designs is to have an individual actuator for each finger in order to enable multiple grasping behaviours and postures. The ability to utilize various grasp types helps us to improve the hold on various shaped objects or to position the hand in the appropriate posture to facilitate as wide a range of tasks as possible. Although each company has their own strategy, these devices rely on myoelectric sequences, co-contractions or patterns to preselect the type of grasp. Many researchers have studied possible strategies to give quick and easy grasp selection including state-space trees and highly tuned pattern recognition software. The large variation in grasp types is what gives these myoelectric hands increased utility over single degree of freedom myoelectric hands.

Despite achieving various grasp types, it is still difficult for users of myoelectric hands to modulate grip force due to the lack of feedback from the hand. Body-powered devices, since actuated through the upper arm or shoulders, are much easier to operate in terms of modulating grip force. This is largely due to the “feel” of the grasp as a result of the force exerted on the shoulder from the harness. They are also simpler and more robust than multi-degree of freedom myoelectric devices. Until now, body powered devices have been restricted to operating a single degree of freedom, as in the body powered split hook, or to open and close an anthropomorphic hand in a single grasp motion.

A major difficulty in the actuation of body-powered anthropomorphic prosthetic hands is the distribution of force from the body-powered cable to the five fingers. The simplest method is to couple all the fingers together into a single combined motion. This results in a single degree of freedom terminal device that is capable of performing a single grasp type. Although it is mechanically simple, other coupling methods can be used that allow for multiple grasping types and adaptive grip behaviour. Underactuation has shown advantages in robotic grasping including better power grasping, more adaptive behaviour to various objects types and shapes, and an increase in the number of contacts



Figure 1: The unique balance bar coupling mechanism allows for both adaptive power grasping and precision grasping between the index finger and the thumb.

made on objects [1]. Many researchers have turned to relying on underactuated adaptive transmission systems to distribute load from a single actuation source to multiple fingers of the hand [2].

In this paper we present the progress toward the Yale body-powered anthropomorphic prosthetic hand. The first topic will be the development of a prototype hand featuring a mechanical coupling mechanism that enables proper force distribution from a body-powered cable to the four fingers of the hand for both a power and precision grasp. This variation in grasping behaviour is selected through a mechanical lever placed on the outside of the hand. By placing this lever on the side of the hand, the user can simply switch between power and precision grasp situations using an able hand or by contact with another object in the environment such as a table edge. The second topic is a discussion of the progress toward a mechanical coupling method tuned for lateral, precision, and power grasps similar to those achieved in the current state-of-the-art prosthetic hands but achieved mechanically for use in body-powered devices.

### ADAPTIVE GRASPING BEHAVIOR

By coupling the fingers together through a differential mechanism, the finger motion is not restricted when one

finger or link makes contact. Instead, all the fingers are able to close until contact is made. This behaviour has shown

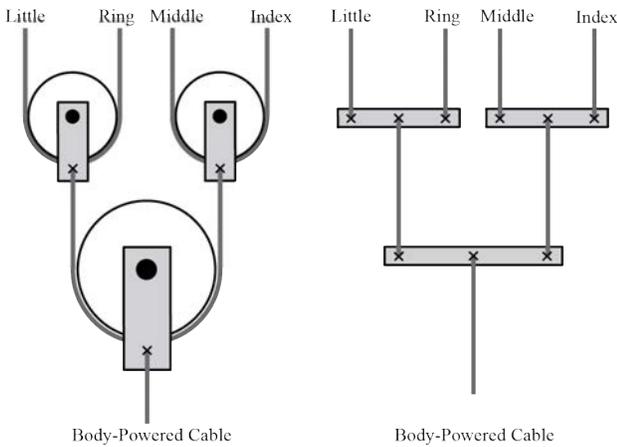


Figure 2: Typical underactuated differentials for tendon driven hands use floating pulleys (left) or waffle trees (right) to achieve force distribution that is independent of finger movement.

to be extremely helpful in power grasping and grasps that envelope object. Even state-of-the-art myoelectric hands with actuators on each finger use a similar grasp closing method. The iLimb, Bebionic, and Vincent hands rely on current control to achieve the proper force distribution during power grasping. The present strategy is to run each of the motors to stall which allows each of the fingers to make contact on an irregular shaped object. The various grasp types are achieved by selectively altering the rest position, and speed of closing for each finger. The same method of driving each finger until contact is made can be achieved mechanically in body-powered hand through the use of a differential coupling method.

Force Distribution Methods in Anthropomorphic hands

The most common way to distribute force from a single input such as a body-powered cable is a direct coupling of the five fingers. Underactuation can be used between the fingers to adapt to various shaped objects. A review of these types of distribution mechanisms can be found in [2]. All of these methods distribute a single input to four or more outputs. Fig. 2 illustrated the common methods of achieving an underactuated grasp between the fingers of a tendon driven hand. An example of this type of coupling can be seen in [3,4]. The first method is by creating a series network of floating pulleys. Regardless of the position of any of the finger tendons, the force is equally distributed across all four tendons. The second method, shown in Fig. 2 (right) is a waffle tree arrangement. Here each of the floating bars is free to tilt to accommodate various finger positions at contact.

With straight waffle tree bars, the force is again distributed equally across all fingers.

Although equal distribution of the gripping force across all four fingers of the hand is beneficial for power grasping,

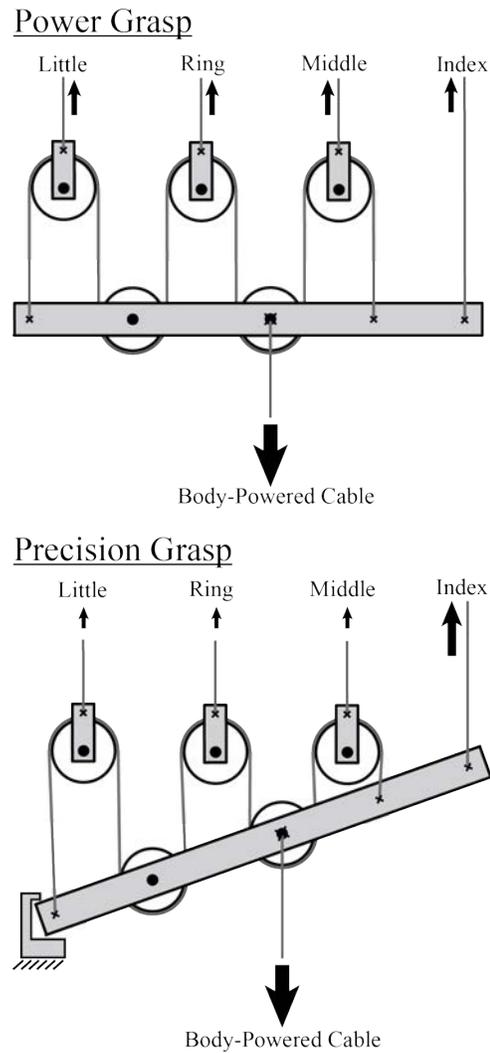


Figure 3: The proposed differential system uses a combination of a single balance bar and pulleys to achieve equal force distribution during power grasping. When the bar is tilted and locked in the precision grasp position, the middle, ring, and little finger remain balanced while the index finger is directly coupled to the position of the body-powered cable.

when performing a precision grasp, or any other grasp that involves any finger not making contact with the object this is undesirable. For example, when performing a precision grasp, it is better to directly control the motion of the index finger as a function of the body-powered cable position. This gives better control and feel for the force being place on the

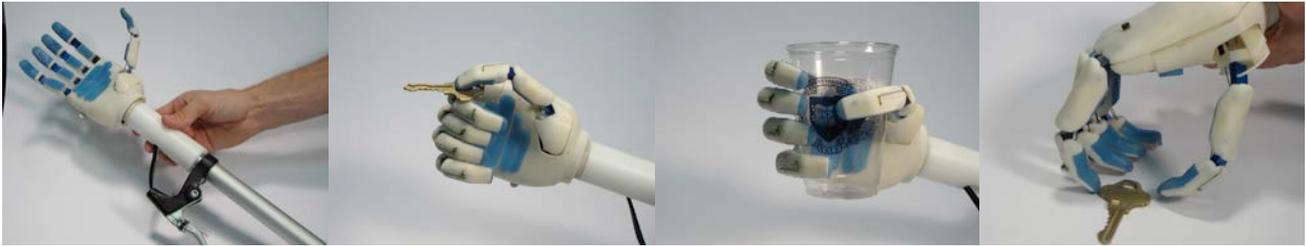


Figure 4: The prototype hand was evaluated using an able body simulator that mimicked the movement of the body-powered actuation cable. Here the hand is shown accomplishing a wide variety of lateral, power, and precision grasps.

single finger if contact is only occurring between the index finger and thumb.

Our proposed coupling design, shown in Fig. 3, is a combination of a single balance bar and floating pulleys, connected to the body-powered cable at the center. The index finger actuation cable is fixed to one end of the bar. The actuation cables for the middle, ring, and little finger are attached to floating pulleys. All of the floating pulleys are coupled to the balance bar with a single tendon that spans two additional pulleys attached to the balance bar. Any difference in position of the middle, ring, and little finger can be taken up through movement of the pulleys on the tendon. Any difference between the position of the index finger and the average position of the middle, ring, and little finger results in the entire floating balance bar tilting in either direction. At any point in this motion, the force is still distributed equally among the four fingers.

When a precision grasp is desired, a small protrusion on the side of the hand is pulled downward and latched. This motion locks the left side (the side opposite the index finger) of the balance bar in the most downward position. The result is shown in Fig. 3 (bottom). Since the bar is now constrained to pivot about the latching point, the movement of the index finger is now a direct function of the position of the body-powered cable. In addition, since the index finger and the set of middle, ring, and little fingers are now decoupled, more force from the body powered harness is now transferred into the index finger instead of the force being shared equally between the four fingers. Table 1 shows the force distribution between the four fingers in relation to the body powered harness for both the power and precision grasp configurations. The precise distribution of forces can be altered by changing the spacing between the output tendons on the balance bar.

Table 1: Finger tendon force ratios of proposed coupling method

Grasp Type	Body-powered Cable Force	Little Finger tendon force	Ring Finger tendon force	Middle Finger tendon force	Index Finger tendon force
Power Grasp	x	0.25x	0.25x	0.25x	0.25x
Precision Grasp	x	y	y	y	0.57x - 1.29y

### IMPLEMENTATION IN PROTOTYPE HAND

A prototype hand was fabricated to test and evaluate the proposed coupling strategy. Fig. 1 shows the prototype hand in the power and precision grasp postures with the front and back cover removed to expose the balance bar coupling mechanism. This prototype hand has two joints in each finger with a single tendon spanning both joints of each finger. The joints of the hand are made of a urethane flexure material similar to those used in the OpenHand [5]. The use of flexure joints helps to improve the adaptability by adding additional out of plane compliance to each finger. All hand components were made from ABS plastic with the balance bar and floating pulleys fabricated from aluminium.

#### Coupling of Thumb Movement

A common observation is that various grasp types require different motion paths of the thumb. For the prototype hand, the thumb was placed on a passive circumduction axis that allowed to user to place the thumb in one of three positions to perform a lateral, precision, or power grasp. The thumb flexion actuation tendon was directly fixed to the main body powered cable without any adaptability. This was done to ensure a force balance across the objects being grasped in the hand during both power, precision, and lateral grasps.

#### Testing and Evaluation

Fig. 4 shows an example of the ability of the prototype hand in power grasp to adapt to a wide variety of object shapes. The prototype hand was tested with able body

subjects using a simulator that mimicked the actuation of a single body powered cable (see Fig. 4 left). Although this is not a true measure of hand function, the system allowed the authors to better study the grasping behaviours of the hand with a single input tendon. This system was also evaluated using the SHAP test to give a wider range of objects and ADL tasks. The results of this test are positive but strictly qualitative since it was performed by an able person without a true body-powered harness. The alteration of the coupling method was important to maintain a stable and predictable precision grasp.

### INTEGRATED GRASP SELECTOR FOR MULTIPLE GRASP TYPES

After showing the benefits of the grasp specific coupling strategy, the authors have developed a body-powered anthropomorphic hand capable of achieving lateral, precision, and power grasping through the use of an integrated mechanical coupling mechanism. Prior to grasping, the user simply places the thumb in the position associated with the desired grasp type (similar to what was required in the first prototype). The movement of the thumb (which can be achieved by an able hand or through contact with the environment) acts like a mechanical selector to alter the pre-grasp position, closing speeds, and overall force distribution from the body-powered cable to the five fingers of the hand. Others have used mechanical selectors to change the grasp type [6] but none have associated this change with a movement of the thumb. Fig. 5 shows the prototype hand with the three distinct position of the thumb which internally affects the coupling strategy from the body-powered harness to the five finger. These properties of the force distribution

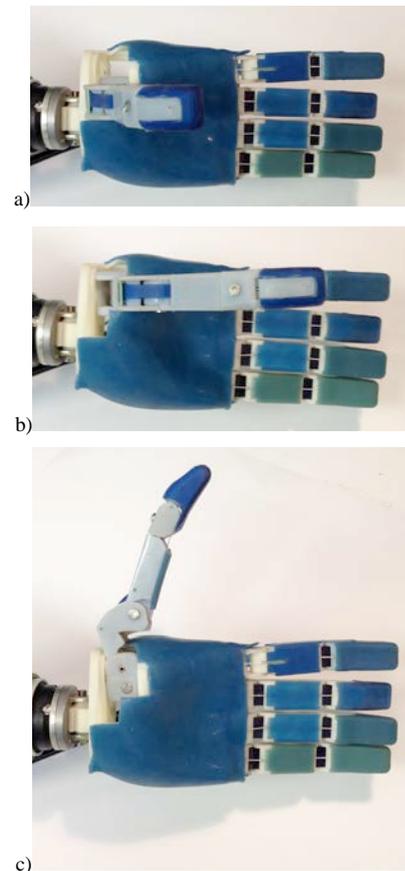


Figure 5: The three distinct positions of the thumb, power grasp (a), precision grasp (b), and lateral grasp (c) each alters the internal coupling of the body-powered cable to optimize for each grasp type.

and closing rates were experimentally tailored to give the best possible finger behaviours for each individual grasp type.

### DISCUSSION

After testing the hand designs, we were able to show a high degree of adaptability during power grasping with additional control of the index finger during precision and lateral grasping. In addition, we found that latching the balance bar allowed the user to have an index finger point. This was useful for delicate tasks such as typing and pushing buttons. Without the altered coupling strategy, precision grasping was difficult since the middle, ring, and little finger would continue to reconfigure even after the precision grasp was established on the object.

The simple and lightweight coupling methods have the potential to improve the utility of body-powered anthropomorphic hands and deserve further testing with body-powered prosthesis users. The authors plan to continue

this work to properly compare this mechanical coupling method to other existing body-powered terminal devices and even existing myoelectric systems.

### ACKNOWLEDGEMENTS

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## DESIGN OF THE GENERATION 3 VANDERBILT MULTIGRASP HAND

Daniel A. Bennett, Skyler A. Dalley, Don Truex, Michael Goldfarb  
*Department of Mechanical Engineering, Vanderbilt University*

### INTRODUCTION

Upper extremity prosthetic devices have traditionally been limited to a single degree of freedom, and are capable of performing only one grasp. This is a significant abstraction of the human hand, which has approximately 20 degrees of freedom and can execute a wide array of grasps and postures. As mechatronics technology has advanced in recent years, there have emerged a number of devices that incorporate multiple actuated degrees of freedom in order to more accurately model the lost limb. While still an abstraction of the native hand, these “multigrasp” hands have the potential to offer enhanced functionality to upper extremity amputees. Several examples of these prostheses are presented in [1-8]. These hands contain between one and six independent actuators and between eight and sixteen joints, where in each device, the discrepancy between the number of actuators and the number of joints is accommodated by differential, kinematic, or compliant coupling. The configuration of each of these hands (i.e., the number and allocation of DOF's, number and allocation of actuators, and type and extent of coupling), as described in [9], varies considerably. Specifically, the manner in which to allocate and configure the DOF's, actuators, and coupling in a multigrasp prosthesis is highly variable, and is highly dependent upon the functional objectives of the hand and the nature of the user interface that controls it.

The authors have previously described the Generation 2 Vanderbilt Multigrasp Hand (VMG), which contained sixteen DOF's actuated by four actuators [7]. Experience with that device informed several design improvements that were incorporated in the Generation 3 VMG, which will be described herein. To the authors' knowledge, the Generation 3 VMG is a unique configuration that has not been previously described in engineering literature. This paper briefly describes the configuration of the Generation 3 VMG and the design philosophy that motivated the improvements from the Generation 2 hand. Finally, an experimental characterization of hand performance and functionality is presented.

### PERFORMANCE AND FUNCTIONAL OBJECTIVES

The design objectives for the hand described in this paper are similar to those presented by the authors in [7]. Many objectives remain the same, and are summarized below, but the most significant departures in objectives will be described in the following sections. The objectives from [7] can be summarized as:

- Achieve six grasp types (tip, lateral, tripod, cylindrical, spherical, and hook).
- Achieve point and platform hand postures
- Provide fingertip forces commensurate with typical ADLs; namely maximum index and thumb fingertip forces of approximately 25 N, and maximum combined fingertip forces of approximately 12 N for the remaining digits.
- Provide joint angular velocities of at least 4 rad/s, which corresponds to a half-range of motion bandwidth of 1.5 Hz.
- Total hand mass <500g.

#### Grasps and Postures

Grasps made by an intact hand can be grossly classified as either a precision or a conformal grasp. Generally, precision grasps are used for grasping objects whose size is much smaller than the hand, where dexterity and precise digit motion is required. Conversely, conformal grasps are generally used for grasping objects whose size is commensurate with the hand itself, where stability and relative digit motion is preferred. Common precision grasps are tip, tripod, and lateral pinch grasps, and common conformal grasps are cylindrical (power), hook, and spherical grasps. The vast majority of grasps used during the activities of daily living (ADLs) by healthy individuals are one of these 6 types [10, 11]. In addition to these grasps, two postures, point and platform, are important for creating a complete grasp taxonomy [12]. The point posture is useful for operating many modern technology interfaces (keyboards, cell phones, touch screens), and the platform is useful for carrying flat objects or for reaching into confined spaces (e.g. a clothing pocket), or donning clothing over the hand and arm. A major objective of the Generation 3

VMG as opposed to the Generation 2 VMG is to incorporate the natural dichotomy between precision and conformal grasps into the fundamental design of the hand, thereby optimizing its performance in executing each type of grasp.

### Power Consumption

In contrast to the Generation 2 hand, the Generation 3 hand is intended to be fully self-contained, with not only the actuators, but also the servo controller circuitry contained within the envelope of the hand. In order to be useful as an untethered prosthetic device, the hand should be capable of operating for a full day between battery charges. Surveys have shown that a majority of amputees use their prostheses more than 8 hours per day; a substantial proportion use them more than 12 hours per day [13, 14]; and some amputees use prostheses up to 16 hours per day [15]. As such, a battery charge should provide for at least 12 hours of use, and ideally 16.

## MULTIGRASP HAND DESIGN

### Allocation of Actuation for Grasping

An essential design objective for the hand prosthesis is to provide both precision and conformal grasps. Recall that a distinguishing feature of the latter is the ability to conform to an object being grasped, thus maximizing the area of contact between the hand and object. In such grasps, one would like the shape of the object to determine the configuration of the hand. Conversely, precision grasps are generally used to handle or manipulate objects that are much smaller than the size of the hand. Such grasps are non-conformal (indeed, the notion of conforming to an object much smaller than the hand is not well posed), and therefore the hand must determine its grasp configuration independently of the object shape. In such cases, under-actuation should be avoided. In order to provide such functionality, the prosthesis described here incorporates 4 independent actuators configured as shown in Figure 1.

Note that degrees of actuation (DOA) 1-3 all correspond to a single degree of freedom (DOF), allowing the configuration

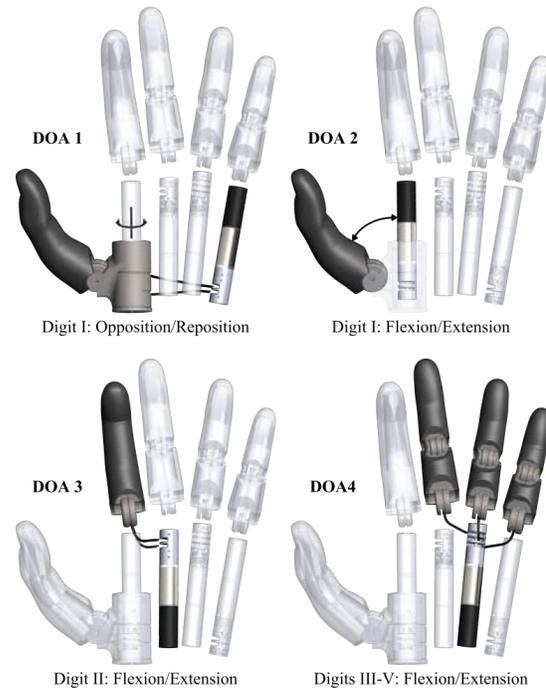


Figure 1. Allocation of actuation in hand prosthesis.

of those digits to be controlled precisely, which is essential for precision grasps. Conversely, DOA 4 controls all 6 DOFs of digits III-V, allowing the configuration of those digits to adapt in the presence of external forces, which is essential for conformal grasps. This is a significant departure from the Generation 2 hand, in which every digit was underactuated. This configuration excelled in conformal grasps, but had difficulty producing stable precision grasps.

### Tendon Actuation and Series and Parallel Elasticity

The full control of digits I and II in flexion and extension that is required for precision grasps is provided by bidirectional tendon actuation in the three respective DOF/DOAs of digits I and II. Each motor unit drives a pulley which has two tendons spooled onto it in different directions, so that positive and negative motor rotation correspond to flexion and extension of the digits, respectively. For each of these digits, the extension tendon is affixed to a linear spring in the fingertip which serves to maintain tension in the tendons throughout the entire range of motion, even if the extension and flexion path lengths differ.

Digits III-V, on the other hand, are only actuated in flexion. Torsional springs located in the metacarpal phalangeal and proximal interphalangeal joints provide the extensive force for each digit. Three tendons are spooled onto the motor pulley in the same direction, and are affixed to linear springs in the fingertip of each digit. For these digits, the spring is responsible for allowing relative motion of the digits with respect to each other, an essential characteristic

of conformal grasps. If an external force is applied to one or more digits, the tendon will compress the spring in the affected digits, while the other digits are still free to move.

Embedded System Design

An embedded system was developed for the hand in order to enable fully self-contained control of all DOAs of the hand. The system is powered by a 14 v battery, which is



Figure 2. Hand with cover removed, showing embedded system and motor units.

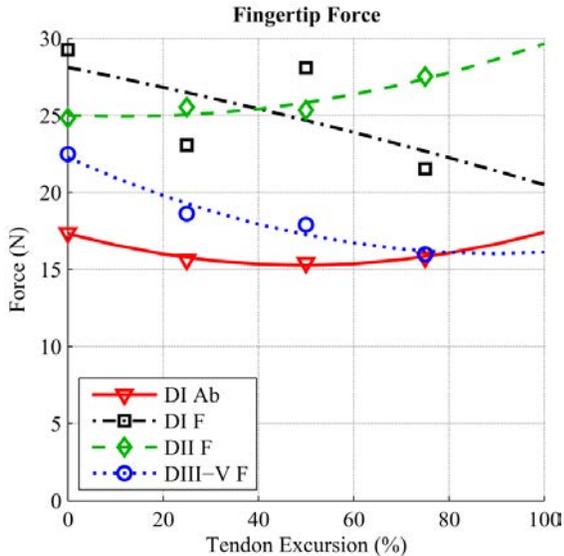


Figure 3. Fingertip forces corresponding to each motor unit

located elsewhere on the prosthetic arm. Using a controller area network (CAN) serial interface, the board accepts position and/or force commands from a high-level controller and executes a low-level PID loop to control the motors. Each motor is controlled by a 20 kHz pulse-width-modulation (PWM) servoamplifier, and each channel is capable of a maximum continuous current of 3.5 A. All four servoamplifiers are controlled by a single microcontroller

(Microchip dsPIC33). A picture of the system as it fits within the hand is shown in Figure 2.

**CHARACTERIZATION OF HAND PERFORMANCE**

Hand Size and Mass

The mass of the hand, including the embedded system encased in the palm, is 546g. The major dimensions of the hand are 8.9 cm across the widest portion of the palm, and 20 cm from the base of the palm to the tip of digit III. Based on anthropometric norms as given in [16], these dimensions correspond to the breadth and length of a 35<sup>th</sup> percentile and 85<sup>th</sup> percentile male hand, respectively. Note that while the breadth is constrained by the layout of the palm, the length of the hand is not as substantially constrained, and could be shortened without difficulty by decreasing the length of the fingers. Shortening the finger length by 1 cm, which is well within the dimensional constraints of the finger design, would render the overall dimensions equivalent to that of a 35<sup>th</sup> percentile male (9 cm breadth, 19 cm length). The hand dimensioned as such would further correspond to a hand breadth and length of a 99<sup>th</sup> and 85<sup>th</sup> percentile female hand, respectively.

Fingertip Forces and Motion Bandwidth

Fingertip forces were measured using an Exttech Instruments 475044 force gauge attached orthogonally to each

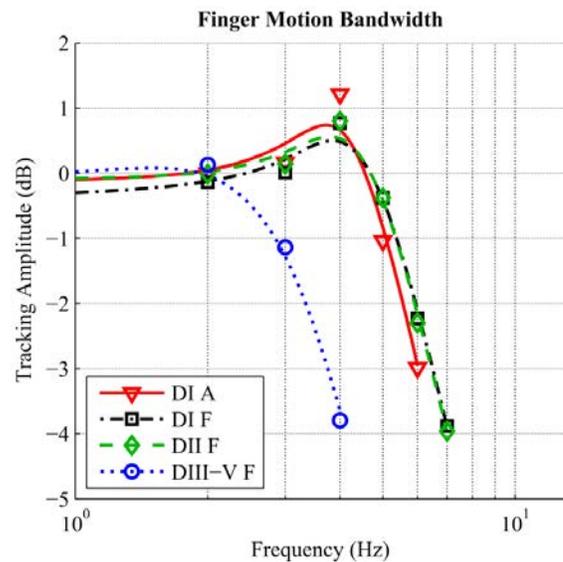


Figure 4. Bandwidth Corresponding to each motor unit.

fingertip. The motors were then supplied 2.5 amps for a duration of 1 second. For each measured tendon displacement, 3 trials were taken and averaged. The results of these measurements are shown in Figure 3. Assuming that during a typical precision grasp the index finger would become

almost fully flexed (~80% excursion), and the thumb would remain almost fully extended (~0% excursion) a composite tip grasp force can be estimated at approximately 29 N. For a conformal grasp, the forces would be a combination of digits II – V, and the tendon excursion would vary depending on the shape of the grasped object. Assuming a nominal excursion of 50%, the composite grasp force for a conformal grasp would be approximately 45 N. Not that these grasp forces are well within the ranges required for activities of daily living, as discussed in [7].

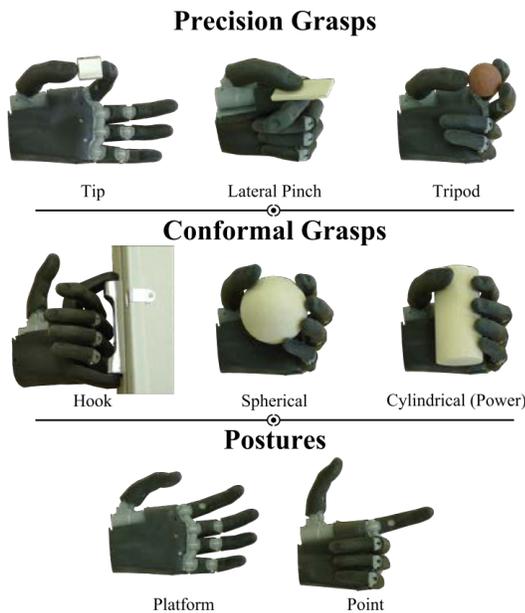


Figure 5. Grasps and postures provided by the hand prosthesis

The motion bandwidth of the fingers was measured during unloaded movement by applying a sinusoidal position command of varying frequencies to each motor, with a magnitude of half the total range of motion of the associated digits. The gain between the commanded and measured signals for each motor is shown plotted in Figure 4. Note that the direct drive DOAs each have a -3 dB bandwidth of approximately 6 Hz, while the underactuated (unidirectional) DOA exhibits a -3 dB bandwidth of approximately 3 Hz. Both are well above the 1.5 Hz bandwidth that nominally characterizes activities of daily living, as previously discussed.

#### Hand Postures and Grasps

Figure 5 shows the Generation 3 VMG achieving the 8 prescribed grasps and postures. The control method used to achieve these grasps and postures using a standard two-site myoelectric interface is described in [17].

#### Battery Life

The electrical power required by the hand was characterized by measuring the power consumed by the two basic activities performed by the hand: unloaded movement (or changing the posture of the hand) and grasping objects. To test unloaded movement, the hand was commanded to move through the complete set of canonical grasps, and the current was measured using a current probe (Agilent model 1146 A). Note that any given posture change is a subset of the complete set of canonical grasps, so this measurement serves as an upper bound for any given transition. To measure the grasping force, the hand was commanded to grasp a 500 mL water bottle filled to a mass of 500 g (approximately full) with sufficient force to securely lift the bottle, and then release the bottle. Each of these tests were performed 10 times, resulting in current requirements of 2.18 A-s and 2.25 A-s, respectively. With the battery used in the test (14 v, 1.35 A-h lithium polymer with a mass of 133 g), the hand could perform approximately 2100 power grasps or approximately 2300 movement sequences, or some combination of these activities, on a single charge.

#### CONCLUSION

The authors describe here the design of the third generation of the Vanderbilt Multigrasp Hand. The hand has 9 DOFs, and is actuated by 4 motors through the use of tendon. Incorporating lessons learned from previous designs, this hand focuses on explicitly providing a capability to perform precision and conformal grasps. This is done by including a combination of under-actuated and fully-actuated digits, rather than relying entirely on under-actuation. Additionally, this iteration was designed to be entirely self-contained, with all actuation and control of the motors fitting within the envelope of the hand. The hand was shown to achieve the set of canonical grasps required for the majority of activities of daily living, while also achieving the required forces and speeds.

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## REDEFINING BIOFEEDBACK TRAINING: PATIENT AND CLINICIAN PERSPECTIVES ON CURRENT AND ALTERNATIVE TRAINING SYSTEMS

Chris Bollinger, Janice Hsu, Joby Varghese  
*Advanced Arm Dynamic*

### ABSTRACT:

#### Introduction:

Biofeedback plays an important role in myo-site training during prosthetic rehabilitation with upper limb (UL) amputees [1]. However, a need for more customized and extensive training has been identified, and it has been suggested that the use of virtual training simulations and games may be advantageous [2].

#### Methods:

According to self-determination theory, intrinsic motivation is driven by three fundamental human needs: competence, autonomy and relatedness [3]. These needs can be satisfied through engaging in video games, making them an intrinsically motivating learning method [4]. The evolution of virtual environment training with patient and prosthesis specific goal-directed tasks may also enhance motivation and motor learning.

#### Results:

A literature review addressing biofeedback, motivation, motor learning and gaming technology, as well as current and potential biofeedback systems available to UL prosthesis users, was completed. Electric prosthesis user and UL prosthesis clinician questionnaires were developed in order to evaluate responses to current systems and perceived value in a wider variety of biofeedback options.

#### Conclusions:

Literature results and clinical interactions in an exclusively UL prosthetic rehabilitation setting show UL prosthetic professionals' agreement that prosthetic rehabilitation training must progress to match rapidly advancing component and controls technology. Although data collection is ongoing, preliminary results indicate that overall, UL prosthesis users and clinicians believe that the use of biofeedback is valuable for electric prosthesis training and utilization of gaming technology and virtual environments would improve patient engagement. Further research is ongoing to determine prosthesis user and clinician preferences to guide development of training technologies to

meet patient preferences, motivational dynamics and motor learning methodologies for advanced prosthesis technologies.

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## COMPARING DIFFERENT TYPES OF TRAINING OF THE MYOSIGNAL

Raoul Bongers<sup>1</sup>, Bernhard Terlaak<sup>1</sup>, Hanneke Bouwsema<sup>1,2</sup>, Corry van der Sluis<sup>3</sup>

1. *University of Groningen, University Medical Center Groningen, Center for Human Movement Sciences, Groningen, The Netherlands*

2. *Centre of Expertise in Rehabilitation and Audiology, Adelante Rehabilitation Centre, Hoensbroek, The Netherlands*

3. *University of Groningen, University Medical Center Groningen, Department of Rehabilitation Medicine, Groningen, The Netherlands*

### ABSTRACT

With the coming on the market of devices that require dexterous myocontrol for controlling, the demands on training the myosignal get higher. The current study examined three different types of virtual training of the myosignal: a) presenting the produced myosignal on a screen, b) control a virtual prosthesis, and c) train with a computer game. Although it might have been expected that training with a game would have been the most effective training, we did not find significant differences between training schedules.

### INTRODUCTION

Myocontrol is the control of an external device through electromyography (EMG) signals derived from the action potentials produced by the muscles and is used to control for instance myoelectric prostheses [1]. Recent advances in prosthetics resulted in advanced prosthetics coming on the market. These advanced prostheses required more profound myocontrol, which puts high demand on the training of the myocontrol. The current paper examined different types of virtual training to improve myocontrol.

We focussed on virtual training methods because these can start much earlier after amputation than conventional training methods because wounds do not need to be healed. Therefore, virtual training may exploit neuroplasticity processes that are active immediately after amputation. Moreover, virtual training is cost effective since a prosthesis does not need to be attached to a customized socket.

The different types of virtual training examined in the current paper fall each fall into one of the three broad categories of virtual training available. First, a basic class in which EMG signals are presented on the screen. Second, the presentation of a virtual prosthesis. Third, a computer game. To be able to determine whether different types of myocontrol required a different training, we tested discrete and continuous myocontrol<sup>2</sup>.

### METHODS

Thirty-six able-bodied right-handed participant were studied, average age was 22.5y. The study was approved by the local medical ethics committee (METc UMCG, NL39792.0342.12) and informed consent was given before participation. Participants were assigned to one of the three training groups, with 10 participants in each group.

#### Design

Each group trained with an other method included in the Ottobock PAULA (Prosthetists Assistant for Upper Limb Architecture) software. The first group (Myo) trained with their myosignals displayed as feedback on a computer screen. The second group (VH) trained with a virtual myoelectric prosthetic hand presented on a screen that was controlled in the same way as an actual prosthesis, and the third group (Game) trained with a computer game in which they controlled two cars through myocontrol.

The experiment was conducted on three consecutive days. On the first day, the participants performed a pre-test to determine baseline myocontrol skills. Subsequently, myocontrol was trained on three consecutive days. Each of the groups trained 6 sessions of 2 minutes each day, with a 30 second break between each session. After training on the third day, the same tests as in the pre-test were administered as a post-test.

#### Materials and procedures

Ottobock PAULA<sup>a</sup> software, in conjunction with 75M11Myoboy with active socket electrodes (13E200 Myobock electrodes) connected through USB to a pc, was used for training and electrode placement. One electrode was placed on the wrist flexor muscles and the other on the wrist extensor muscles.

#### Pre-test and post-test measurements

Custom software was developed to test discrete and continuous myocontrol abilities in separate tests. The test of discrete myocontrol used a virtual prosthetic hand with proportional control. Participants had to open and close the virtual prosthetic hand to full aperture at either the slowest,

moderate or fastest controllable velocity in separate trials (cf. Bouwsema et al.). All velocities were executed three times in a random order (total of 9 trials).

Testing of continuous myocontrol used myosignal feedback and a semi-random graph to which the participants had to match their myosignals over a 30 second period. The test was performed twice, with either the flexor or extensor muscles, used to control closing and opening of the hand respectively.

**Training sessions**

PAULA software was used for the training, as it contained direct myosignal feedback training, a virtual myoelectric prosthetic hand and a computer game training mode. Participants of all groups were instructed to consciously influence and improve the control over their myosignals. The Myo group, training with their produced myosignals as instantaneous feedback (Figure 1), was explained that stronger contraction would lead to higher signals and were told to train 2 minutes per round without further specific instructions. The VH group, training with a virtual myoelectric prosthetic hand (Figure 2b), was explained that flexor and extensor contractions led to closing and opening of the hand on the screen and stronger contractions led to faster movements. They were told to train hand opening and closing for 2 minutes per round without further specific instructions on the exact movements to make. The Game group had to control the vertical movements of two cars with their myosignals (each muscle group controlling movements of one car), and steer these through gaps in oncoming walls by producing myosignals of the correct height (Figure 2c) for 2 minutes per round. One car had to be steered through the gap per wall, switching cars each wall.

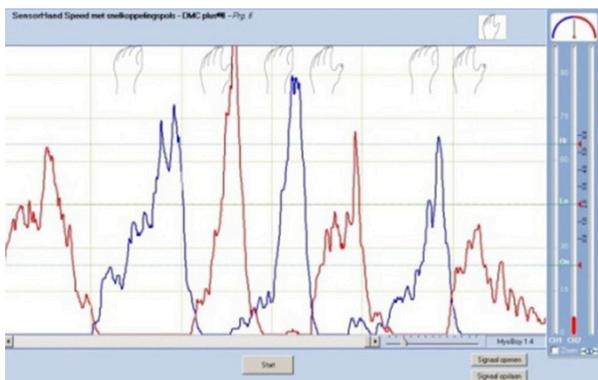


Figure 1: Screen of the Myogroup

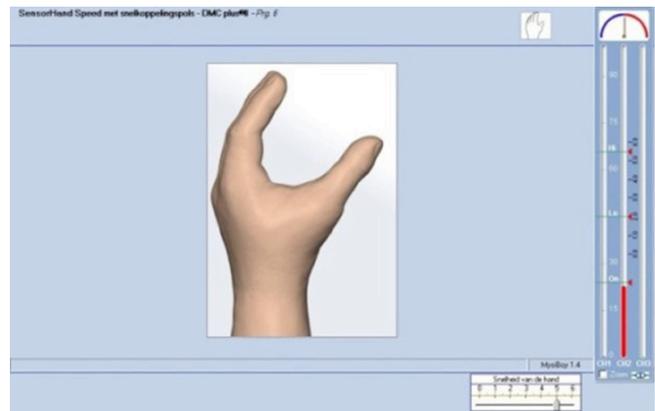


Figure 2: Screen of the VH group

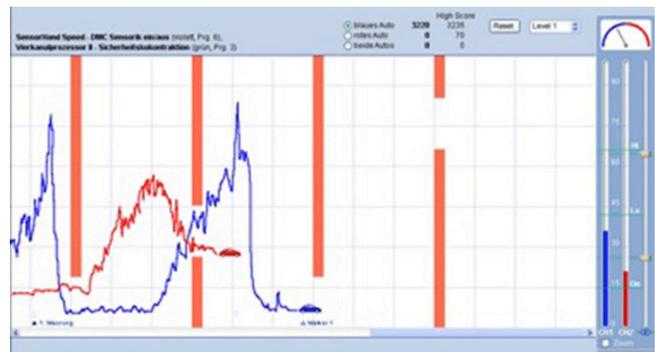


Figure 3: Screen of the Game group

**Data analysis**

Custom-written scripts in Matlab were used to compute the following dependent variables: mean velocity of the hand opening and closing for each required velocity (slow, moderate, fast) for the discrete test; the error between the produced myosignals and the predefined graph for the continuous test. Repeated measures ANOVAs were performed to determine significant differences.

**RESULTS**

**Discrete myocontrol**

Training groups did not differ significantly in their learning effect on discrete myocontrol. As expected, a large effect of velocity was found  $F_{(2,60)} = 562.42, p = .00, \eta^2_g = .68$ . In the fast condition, the participants reached the highest velocities (500.23 mm/s (11.32) (mean[SE])), whereas the lowest were reached in the slow condition (186.82 mm/s (8.03)). No significant effect of test or any significant interaction was found.

**Continuous myocontrol**

No significant differences between the learning effects of the three training groups were found on the error. The main

effect of test showed that participants improved significantly from pre-test to post-test  $F_{(2,60)} = 87.13$ ,  $p = .00$ ,  $\eta^2_g = .20$  (pre-test: .1304 V (.0028), post-test: .1027 V (.0020)).

## DISCUSSION

The current study showed that virtual training of myoelectric control using direct feedback of the produced myosignals projected on a computer screen, a virtual prosthetic hand, or a computer game does not differ for discrete and continuous aspects of myocontrol. As such, in the clinical practice, the available method can be used based on the needs and preferences of the patient.

It is argued that training in a gaming environment leads to more motivation to keep training and thus would lead to more learning.[3] One reason why we might not have found a larger training effect for the Game group might be that the duration of the training was too short. Effectively, participants trained for 36 minutes over 3 days, which may be too short to take advantage of the benefits of training with a computer game. As literature was unable to provide adequate information on the required training time for an optimum in duration of training, we based the current design on our earlier study.[2]

Interestingly the effects of training on discrete myocontrol were not the same as those on continuous myocontrol, that is, we found an effect of training on continuous myocontrol but not on discrete myocontrol. The origins of this differences require further investigation of different processes underlying discrete and continuous myocontrol. For instance, it might be the case that effects of virtual training the myosignal shows up in an other test of discrete myocontrol but not in the current test.

The current study is limited in that able-bodied participants were used. However, this allowed us to examine more participants than would have been possible if we had used novice amputees that learn to use a prosthesis. This larger number of participants results in more reliable measurements and results. However, caution should be taken when generalizing the current findings to amputees.

Finally, the current study showed an effect of virtual training of the myosignal of a continuous test of the myosignal. However, from the current study it is not clear whether and how this effect transfers to functional tasks with a prosthesis. Anyhow, the current study showed that some aspects of the

myosignal can be trained and that different types of virtual training schedules show the same effect.

## ACKNOWLEDGEMENTS

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## A COMPUTER VISION-BASED APPROACH TO HIGH DENSITY EMG PATTERN RECOGNITION USING STRUCTURAL SIMILARITY

Alexander Boschmann, and Marco Platzner

*Department of Computer Science, University of Paderborn, Germany*

### ABSTRACT

The displacement of electrodes during usage significantly degrades robustness and usability of pattern recognition-based myoelectric control systems. Image representations from high density electromyographic (EMG) signals offer high spatial resolution and only change slightly during electrode shift, preserving structural information. In this paper, we propose a simple one-versus-one classifier based on the Structural Similarity Index (SSIM). SSIM quantifies visual similarity of two images as the product of three components: luminance, contrast and structure. Our experimental results show that an SSIM-based classifier outperforms a traditional pattern recognition-based classifier like LDA using structural information from images of high density EMG data in terms of absolute classification accuracy and robustness to electrode shift

### INTRODUCTION

Pattern recognition based control schemes are an active research area and can potentially enable the amputee to intuitively operate multiple degrees of freedom [1,2]. One challenging factor in this context is variation in electrode recording placement. This effect has been previously studied but remains an unsolved problem [3,4].

High density EMG recordings offer unprecedented spatial resolution of muscular activity and contain structural information suitable to use for pattern recognition of myoelectric control schemes. Analyzing image data extracted from high density EMG recordings it is notable that structural information remain mostly unchanged during electrode shift.

In the field of computer vision, many different methods are utilized to quantify similarity between images based on their structural information. A prominent method is the Structural Similarity Index (SSIM) [5] that decomposes two images into the components luminance, contrast and structure and offers an objective method to quantify their similarity. Based on this method, we propose a simple one-versus-one classifier to experimentally distinguish between 10 hand and

wrist movements only using image representations from high density EMG recordings.

This paper is structured as follows. The experimental setup and structure of the proposed SSIM-based classifier are presented in section II. The experimental results are shown in section III while section IV offers a short discussion.

### METHODS

#### Data Acquisition and Feature Extraction

For this experiment, EMG data corresponding to 10 hand and wrist motions were acquired from one healthy normally limbed 30 years old male subject. The data were collected from an array of 96 electrodes consisting of 4 rows of 24 electrodes wrapped around the forearm. Each electrode had a diameter of 1 cm and the center-to-center distance between adjacent electrodes was about 1 cm. A TMS International REFA 128 high density EMG system was used for data acquisition. Each contraction was held for 5 seconds, followed by a 2 seconds rest period. During the experiment 10 different trials were recorded, each consisting of 12 repetitions of the same contraction. After each trial a one-minute rest period was included to avoid muscle fatigue effects. From each contraction, 4 seconds of data from the steady state phase were extracted. In total  $12 \times 4 \text{ sec} = 48$  seconds of data were recorded for each movement class. The first 24 seconds were used for training the classifier; the remaining 24 seconds were used for classification. Fig. 1 shows a grid representation of the EMG data during the acquired movements.

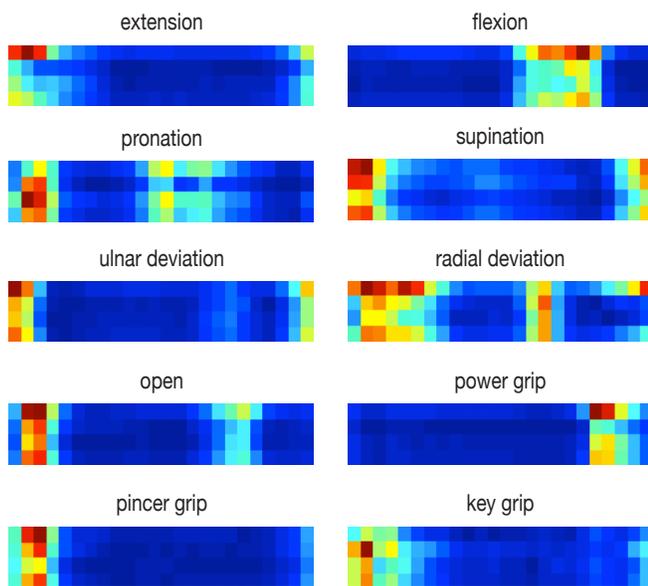


Figure 1: Grid representation of EMG movement data. Each channel was smoothed using a 100 ms sliding window. High rms areas are colored red, low areas blue.

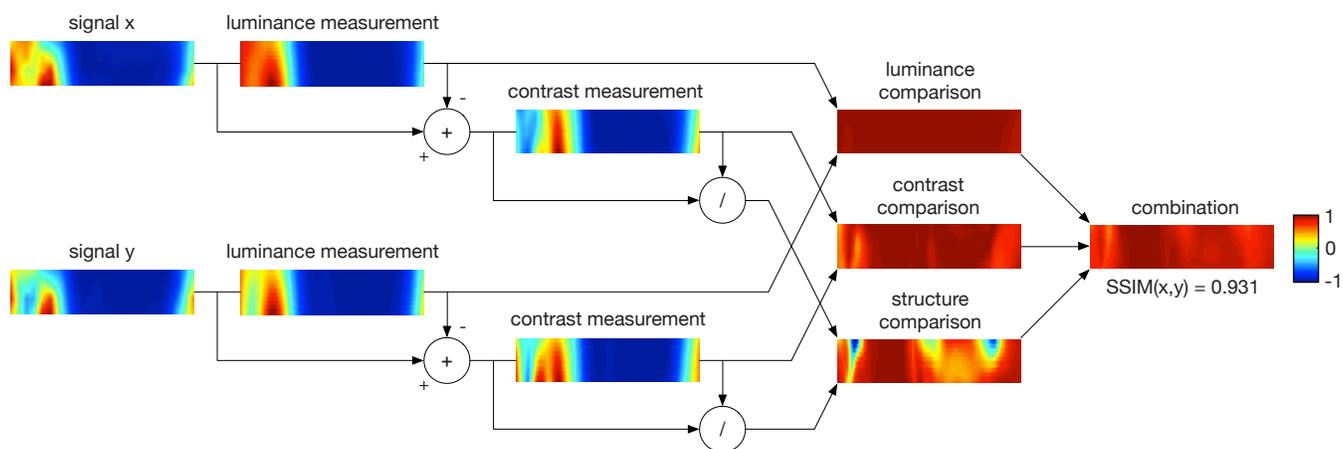


Figure 2: Decomposition and SSIM comparison of two similar signals representing the same movement class (extension), resulting in a relatively high SSIM index value. The input signals were linearly interpolated with factor 3

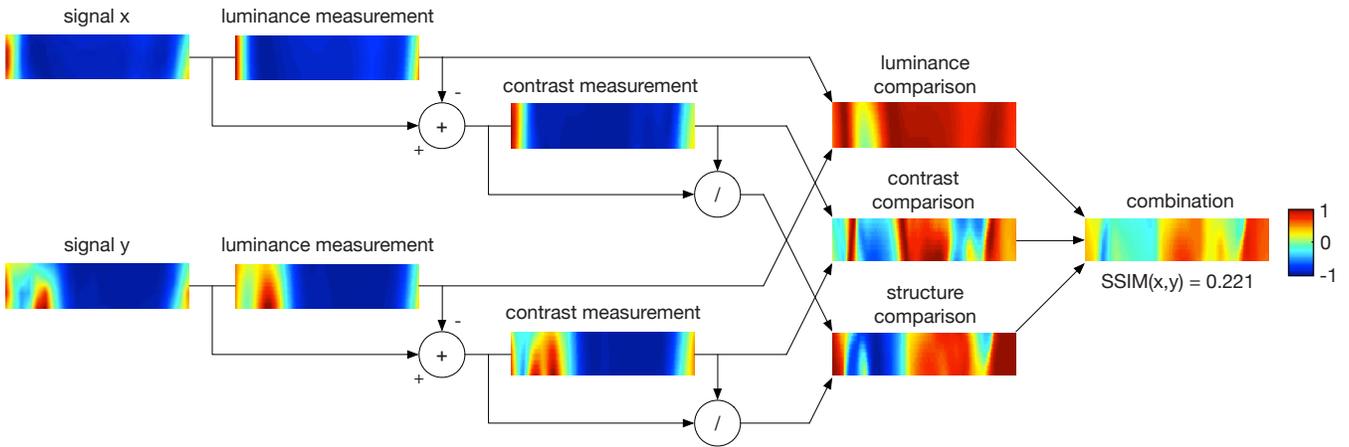


Figure 3: Decomposition and SSIM comparison of two dissimilar signals representing different movement classes (ulnar deviation and extension), resulting in a relatively low SSIM index value..

### The Structural Similarity (SSIM) Index

In order to calculate the SSIM index [5] of two images represented by the non-negative signals  $x$  and  $y$ , the images are decomposed into three components: luminance, contrast and structure. The components are then compared separately. First, the luminance  $\mu$  of both signals is estimated as the mean intensity:

$$\mu_x = \frac{1}{N} \sum_{i=0} x_i \quad (1)$$

The luminance comparison function  $l(x,y)$  is a function of  $\mu_x$  and  $\mu_y$ :

$$l(x, y) = \frac{2\mu_x\mu_y + C_1}{\mu_x^2 + \mu_y^2 + C_2} \quad (2)$$

$C_1$  and  $C_2$  are constants to avoid dividing by zero. The mean intensity is then removed from the signal and the standard deviation (square root of variance) is estimated as the signal contrast  $\sigma$ :

$$\sigma_x = \left( \frac{1}{N-1} \sum_{i=0}^{N-1} (x_i - \mu_x)^2 \right)^{\frac{1}{2}} \quad (3)$$

The contrast comparison function  $c(x,y)$  is a function of  $\sigma_x$  and  $\sigma_y$ :

$$c(x, y) = \frac{\sigma_x\sigma_y + C_2}{\sigma_x^2 + \sigma_y^2 + C_2} \quad (4)$$

Furthermore, structure comparison  $s(x,y)$  is conducted:

$$s(x, y) = \frac{\sigma_{xy} + C_3}{\sigma_x\sigma_y + C_3} \quad (5)$$

with  $C_3=C_2/2$  and  $\sigma_{xy}$  estimated as the covariance of the signals:

$$\sigma_{xy} = \frac{1}{N-1} \sum_{i=0}^{N-1} (x_i - \mu_x)(y_i - \mu_y). \quad (6)$$

Finally, the three comparisons (2), (4) and (5) are combined in the SSIM index between signals  $x$  and  $y$ :

$$SSIM(x, y) = [l(x, y)]^\alpha \cdot [c(x, y)]^\beta \cdot [s(x, y)]^\gamma \quad (7)$$

where non-negative  $\alpha, \beta$  and  $\gamma$  are used to adjust the relevance of the three components. In this paper, we use  $\alpha = \beta = \gamma = 1$ . The SSIM index is locally calculated using an 11-pixel Gaussian sliding window filter. It can range from -1 to 1, with 1 being the index value for two identical images.

For illustration purposes, Figures 2 and 3 depict image decomposition and SSIM index calculation of high density EMG data used in the experiments. Fig. 2 shows the comparison between two similar but not identical signals representing the same movement class. The difference in luminance and contrast between the input signals are almost zero, while there are minor differences in image structure due to the slightly differently performed movement. In this example, his results in a relatively high SSIM index of 0.931.

Fig. 3 shows comparison between images representing slightly different movement classes, resulting in a relatively low SSIM index of 0.221. While the difference in luminance is only minor, there are significant differences in contrast and structure.

Due to a minimum amount of pixels necessary for calculating the SSIM index, input signals ( $4 \times 24$  pixels) are linearly interpolated with factor 3 when used as input for SSIM index calculation.

Training Phase

As previously indicated, the first 24 seconds of data representing each contraction class are used to train the pattern recognition system while the remaining 24 seconds are used for classification. Fig. 4 illustrates the training phase of the experiment. For each movement class,  $n$  frames of  $4 \times 24$  pixels are extracted from the high density EMG raw data using a 100 ms wide, 50 ms overlapping rms filter window. These  $10 \times n \times 4 \times 24 = n \times 960$  values will be used as feature vectors for training of a traditional LDA classifier that we use as a reference for comparing against the SSIM-based classifier in the Results section.

The extracted  $n$  frames are then averaged into 1 frame for each movement class. The resulting 10 frames of  $4 \times 24$  pixels form the training model for the SSIM-based classifier.

Test Phase

In this phase we use SSIM as a simple one-against-one classifier as depicted in Fig. 5. For this purpose, we extract  $n$  test frames from the test data in the same manner as the frames for the training model. A frame representing EMG data of an unknown movement class is interpolated and SSIM-compared with each of the 10 interpolated frames from the training model. The highest resulting SSIM index decides the class.

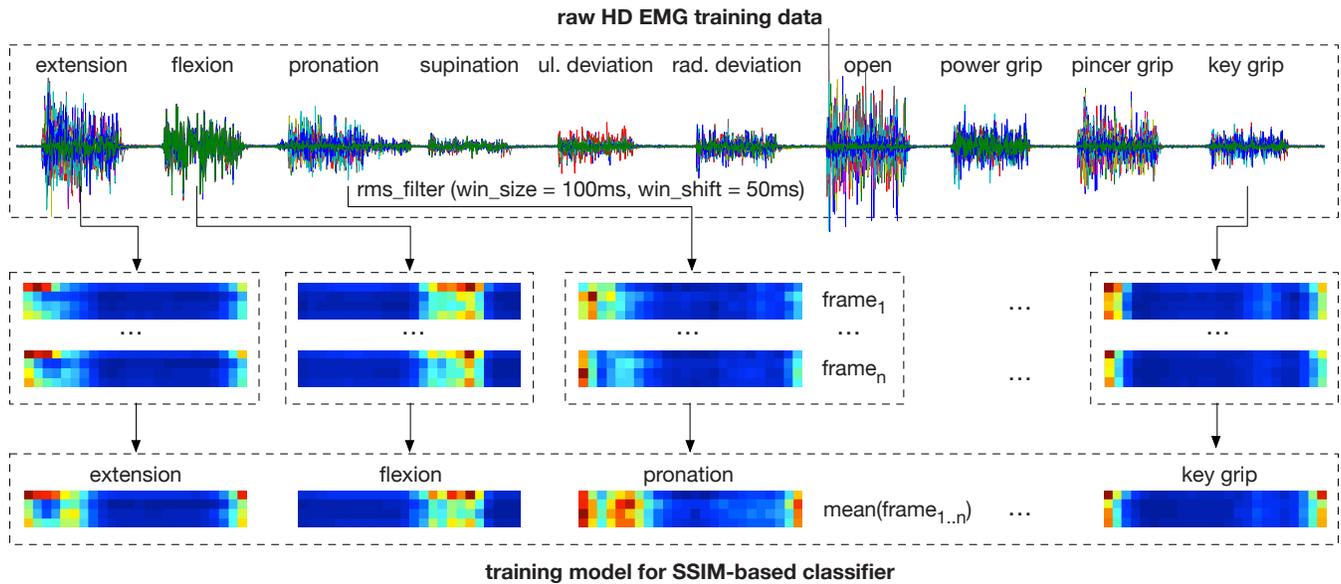


Figure 4: During training phase,  $n$  frames are extracted from raw training data and averaged to one frame per class.

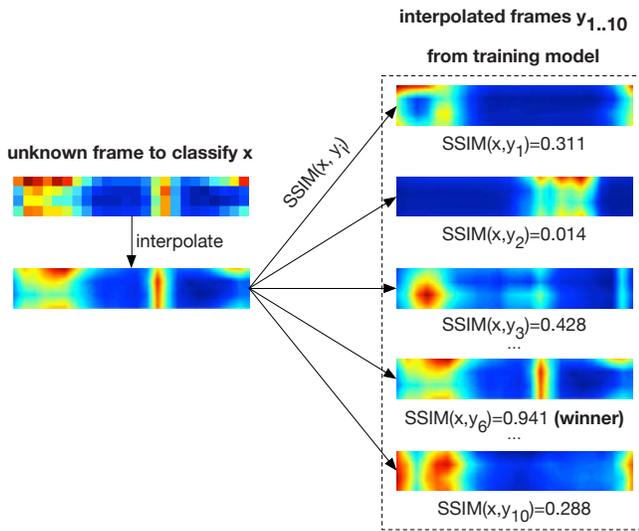


Figure 5: Test phase of the experiment. To determine the class of an unknown frame, it is pairwise SSIM-compared with the training frames from the training model.

To simulate the electrode shift effect, the representation of the electrode array in the test data was horizontally shifted by 1 cm in the software simulation. This is equivalent to the amount of electrode displacement that is likely to occur during everyday prosthesis wear.

## RESULTS

We have performed the experiment to answer two specific questions. First, can a computer vision-based classifier that treats high density EMG data as images and classifies them based on structural information outperform a traditional pattern recognition-based classifier? Second, since electrode shift has only little influence on image representation of EMG data, is a computer-vision based classifier more robust to it?

To evaluate the performance of the SSIM-based classifier and LDA when classifying unshifted EMG data we have trained and tested both classifiers in all 24 possible positions successively. Then, we have averaged the performance in terms of classification accuracy. Fig. 6 shows that the SSIM-based classifier had around 95% correct decisions while LDA was correct around 80% (black bar).

To determine the classification performance when dealing with slightly shifted data, we trained both algorithms in all possible positions successively and in each position we shifted the test data horizontally 1 cm left and right (Fig. 6 white and gray bars). We found that while LDA's accuracy was decreasing to an average of 46%, the SSIM-based classifier's accuracy went only down to 92%.

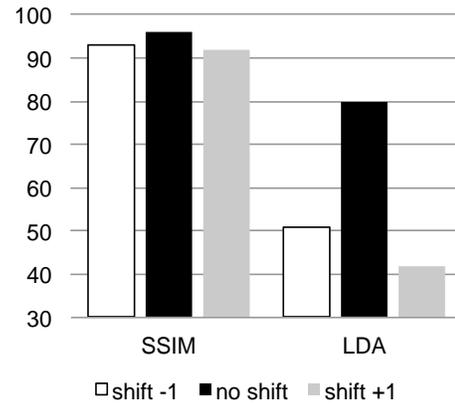


Figure 6: Experimental results. Averaged classification accuracy rate without shifting of test data (black), shift of 1cm left (white) and 1cm right (gray)

## DISCUSSION

The Structural Similarity Index quantifies the visual similarity between two images as the product of three components: luminance, contrast and structure. Our results indicate that a simple one-versus-one classifier based on SSIM seems to outperform a traditional pattern recognition-based classifier like LDA using structural information from images of high density EMG data in terms of absolute classification accuracy and robustness to electrode shift.

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## DEVELOPMENT OF AN INNOVATIVE TEST-PROSTHESIS: AN IMPORTANT TOOL IN THE DECISION MAKING PROCESS IN PROVIDING PATIENTS WITH AN UPPER LIMB PROSTHESIS

Michael A. H. Brouwers<sup>1</sup>, Ingrid E. M. Roeling<sup>1</sup>, Iris van Wikjk<sup>1</sup>, Ellen P. H. Mooibroek-Tieben<sup>1</sup>,  
Marieke W. Harmer-Bosgoed<sup>1</sup>, Dick H. Plettenburg<sup>2</sup>

<sup>1</sup>De Hoogstraat Rehabilitation, <sup>2</sup>Department of Biomechanical Engineering, Delft University of Technology

### ABSTRACT

#### **Objective:**

To share experiences with an innovated test-prosthesis to help patients in their decision making process.

#### **Background:**

De Hoogstraat Rehabilitation provides over 500 children and adults with upper limb deficiencies with ongoing specialized care. It is important to inform new patients about the benefits and disadvantages of wearing a prosthesis and the characteristics of the different types of prostheses. Besides the availability of videos, written information and experienced patients, who demonstrate their prosthesis and its possibilities, potential users lack the possibility to test different types of prostheses. Therefore a test- prosthesis was developed.

#### **Prototype:**

The first prototype of the test-prosthesis has been used since 2012. It helps patients to have more realistic expectations of the wearing and usage of a prosthesis. This first design is based upon a vacuum technique and is suitable for children and adults with a trans radial congenital or acquired limb deficiency. These patients are able to test myo-electric or body-powered control. The test-prosthesis is re-usable. Therapists are able to fit patients with the test-prosthesis within 15 minutes without the intervention of a technician.

#### **Innovations:**

Further development of the test-prosthesis has been done in order to be able to use it without the prosthesis being connected to the vacuum pump. The new version of test-prosthesis aims for an improved fit in patients with a short trans radial amputation or congenital longitudinal deficiency.

#### **Conclusion:**

The test-prosthesis is an important tool in the decision making process in providing patients with an upper limb prosthesis; does the patient want a prosthesis or not, which terminal device and which control system, myo-electric versus bodypowered complies best with the request. Experiences

with the test-prosthesis will be shared. Further experiences should reveal whether the test-prosthesis may help to reduce the high rejection rates of upper limb prostheses.

## UPPER LIMB FUNCTION AFTER UPPER LIMB AMPUTATION

Helena Burger

*University Rehabilitation Institute, Republic of Slovenia*

### INTRODUCTION

Persons after upper limb amputation may have problems at several activities and at participation (1). Using upper limb prosthesis is one way to decrease these problems. There is lack of evidence to establish clinical guidelines regardless prosthetic prescription and treatment (2).

The aim of our study was to find out upper limb function in persons following upper limb amputation.

### METHODS:

We included all subjects 15 years old or older that visited outpatient clinic at our Institute in one year having a functional prosthesis and willing to participate.

We collect clinical data and tested hand function by Southampton Hand Assessment Procedure – SHAP (3).

Study was approved by ethical committee of our Institute. Results were statistically analyzed by SPSS.

### RESULTS:

Twenty-three subjects, fifteen men, 15 to 64 years old fulfill inclusion criteria.

All SHAP scores were significantly worse with the prosthesis than with non-amputated hand. All SHAP scores for non-amputated hand were worse than are norms for healthy Slovene population.

The only factor influencing SHAP scores was time from amputation to fitting of the first prosthesis.

### DISCUSSION:

Similar to other studies (4) we found out that time from amputation to fitting the first prosthesis influence the function of prosthesis user.

The surprising result of our study was that also SHAP scores of non-amputated hand were worse than scores of healthy persons. That has never been described before and

needs further studies on greater number of subjects from different countries.

The main limitation of our study is small number of included subjects.

### CONCLUSIONS:

We can conclude that it is important to fit the first prosthesis soon after amputation.

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## COMPARISON OF MULTIPLE LIMB LOSS MILITARY PATIENTS AND UPPER BODY PROSTHETIC USE, A RETROSPECTIVE ANALYSIS

Joe Anthony Butkus

*Walter Reed National Military Medical Center, Bethesda, MD, USA*

### ABSTRACT

Health care for patients suffering from limb loss at Walter Reed National Military Medical Center has changed remarkably since the initial injured service members began to return from the battlefields of Iraq and Afghanistan in 2003. Occupational therapy and upper extremity prosthetics have improved by research, analysis, patient feedback, and trial and error. Patients and staff are very fortunate to have access to extensive support to benefit the patient without managed care concerns. The concentrated patient population allows staff to develop extensive experience and casually examine trends among upper extremity prosthetic users.

### INTRODUCTION

In 2013 we received 3 patients with very severe, but similar injuries involving 3 or more limbs and all resulting in bilateral upper extremity amputations. Each patient sustained one trans-humeral and one trans-radial amputation. Therapy staff prepared with accumulating resources, seeking out former patients and developing treatment plans to find ways to best assist these patients and other staff. Staff communicated frequently to resolve issues, find creative solutions, seek support, and share information. The patients themselves also shared information, adaptive equipment, emotional support, and not to be underestimated, vibrant senses of humor.

### RESULTS

As therapy progressed and patients achieved greater levels of independence, staff began noticing some similarities and differences of the individuals' results. Adaptive equipment that worked for some patients did not work for all. Occasionally tasks were performed uniformly as if they all practiced together. Some patients were motivated to perform certain tasks and worked to solve the problems when others did not see a problem they needed to address. Strategies to perform the same tasks contrasted greatly. Patients were very eager to try strategies recommended by other patients with similar injuries and always were interested in comparing notes on how they completed tasks. The staff began to see

some patterns emerge that were based on the relative need to perform specific tasks, the individuals' personality, and how patients desires effected prosthetic choices.

### TRENDS

Overall there were a number of items related to prosthetics and therapy that were universal among this small group of patients. Patients did not select a myo-electric elbow for use on a daily basis. Patients demonstrated an affinity for wearing myo-electric prosthetics on their trans-radial sides. Use of cell phones was of utmost importance at all times and many types of interfaces and set ups were trialed. Patients preferred use of the five function wrist with lockable rotation on a body powered arm over more traditional set ups for the above elbow side. Patients did not use the more recently released advanced myo-electric terminal devices and preferred the typical myo hands or the ETD. Patients universally found the sensor in the sensor hand speed interfered with function more than it assisted. These items were some of the more accepted trends for all of the included patients.

Despite many similarities, there were frequent differences for patients as well. A variety of adaptive equipment items were used for activities of daily living (ADLs) that did not require an arm for certain tasks. Patients varied on what types of equipment they chose to use if they chose to use the equipment at all. Patients received customized containers or bags to assist with carrying items they used frequently and the organization of these varied greatly. Prosthetic leg systems and donning/doffing were entirely unique. Patients contrasted with each other over which tasks they would perform independently and tasks for which they would receive assistance. These items will be reviewed in greater detail in the presentation. This experience reinforced how client centered care is essential in therapy even with patients with very similar impairments as each client is entirely unique.

## CONCLUSION

The concentration of patients and resources available to the military limb loss patient offers a unique perspective to review the prosthetic use and trends for the larger population as a whole. There are many issues for the limb loss patient that need further research to identify use. The review of trends above are based on experiences with a handful of patients and needs to be formally researched to ensure accuracy and guide evidence based practice.

## EMG-BASED PREDICTION OF MULTI-DOF ACTIVATIONS USING SINGLE-DOF TRAINING: A PRELIMINARY RESULT

Claudio Castellini and Markus Nowak  
*Robotics and Mechatronics Center  
DLR – German Aerospace Center  
Weßling, Germany*

### INTRODUCTION

In this paper we propose an advancement to the problem of simultaneous, proportional myocontrol of hand/wrist prostheses [1,2,3]. In particular, we address the prediction of simultaneous activations of multiple degrees of freedom (DOFs) by training a machine learning method on single-DOF activations only – for example, correctly predicting simultaneous flexion of the index and thumb by training on index flexion and thumb flexion only. In myoelectric control this is a very desirable property, since training on single-DOFs only will in general not correctly predict multiple-DOF activations; on the other hand, directly gathering multiple-DOF activation data from the subject quickly becomes unfeasible as the number of DOFs grows.

So far, to the best of our knowledge, the only successful approach to this problem is represented by the application of Non-negative Matrix Factorisation to two/three DOFs of a prosthetic wrist [4]; we hereby propose an alternative approach which is able to solve the problem for single-finger activations. Surface electromyography (sEMG) data are firstly collected for single-finger forces; the data set is then augmented with artificial sEMG clusters representing multiple-DOF activations; lastly, a machine learning method is trained on the augmented data set. The augmentation procedure works by linearly combining the single-DOF sEMG clusters and is therefore called Linearly Enhanced Training (LET).

To preliminarily validate the procedure, an experiment was conducted on seven intact subjects engaged in the production of a simple, repetitive single-DOF activation pattern (thumb adduction, index flexion and little finger flexion). The results are very promising.

### PROBLEM STATEMENT

#### Simultaneous and proportional myocontrol

Simultaneous and proportional myocontrol of a prosthetic or rehabilitation device [6] entails that a (disabled) human subject can control its  $m$  DOFs independently, at the same time, and in a “graded” fashion, that is, according to the desired level of activation. For instance, if each DOF can be controlled in torque, we must build a human-machine interface consisting of  $d$  sEMG electrodes and  $m$  approximant functions  $f$  such that  $\tau \approx f(\mathbf{x})$  where  $\tau \in \mathbb{R}$  is the required torque at the DOF and  $\tau \in \mathbb{R}^d$  is the reading from the electrodes. (In this simplified framework we intentionally blur the distinction between the muscular activation and the DOF activation expressed as a torque command.) Machine learning is usually employed in the regression mode (e.g., Support Vector Regression [7,8] or Non-negative Matrix Factorisation [4]) to build the approximant functions from a set  $X = \{\mathbf{x}_i, \tau_i\}_{i=1}^n$  of (sample, target) pairs previously collected from the subject – the so-called *training set*. (Notice that in the training set one needs to have one target value per each DOF, hence  $\mathbf{x}_i \in \mathbb{R}^d$  and  $\tau_i \in \mathbb{R}^m$ .)

The usage of machine learning has the advantage of allowing *natural* control. The training set  $X$  is built by inducing the subject to activate one DOF (e.g., flexing a finger, pronating the wrist, etc.) and recording the corresponding sEMG values. If the input/output relationship is fairly represented by the values in  $X$ , then each  $f$  will correctly approximate the torque  $\tau$  required to control the corresponding DOF; moreover, since  $X$  was collected from the subject while engaged in performing the actions corresponding to the activations of each DOF, the resulting approximant will command to each DOF the intended torque – hence the term *natural control*, or *intent detection*.

Clearly, in order for this approach to be feasible, an appropriate sampling of the input space *for each DOF considered* is required; target values can be either gathered using a torque/force sensor, or more realistically, they can

be arbitrarily set at 0 or 1 whenever, in turn, a DOF is not active or maximally active. (Recall that in general the subject is an amputee who cannot produce any reliable ground truth in principle.) Following the “realistic approach” outlined in [5], this corresponds to such a training set:

$$X = \{(X_0, \tau_0), (X_p, \tau_1), \dots, (X_m, \tau_m)\}$$

where each subset  $(X_k, \tau_k)$  corresponds to the (sample, target) pairs collected when only the  $k$ th DOF is active, and all others are inactive. (The subscript 0 denotes the resting state, in which all DOFs are inactive.) For example, consider the case in which the selected DOFs are the flexion of the index (I) and little (L) finger; the corresponding training set, denoted with  $sf$  for single-finger activations, is

$$X_{sf} = \{(X_0, (0\ 0)), (X_I, (1\ 0)), (X_L, (0\ 1))\}$$

where  $X_I$  and  $X_L$  denote sEMG samples collected, in turn, when either the index or the little finger was maximally active. Given an appropriate machine learning method, two functions  $f_I(\mathbf{x})$  and  $f_L(\mathbf{x})$  trained on  $X_{sf}$  (namely,  $f_I(\mathbf{x})$  would be trained using the first component of each  $\tau_k$  as target values, and  $f_L(\mathbf{x})$  would be trained on the second) will return a sensible approximation of the torques required at the index and little finger *whenever either of the two fingers, or none of them, is active*.

### Multi-DOF activations

Indeed, the above method will not generalise to the case in which both DOFs are active at the same time – simultaneous flexion of the index and little finger: the sEMG signal corresponding to a multi-DOF activation, call it  $X_{IL}$  in the example above, has, in general, no trivial relationship to those obtained for the single-DOF activations it is composed of. Nevertheless, being able to estimate multi-DOF activations is very desirable: e.g., while grasping, many fingers are active at the same time; while reaching with the aid of a prosthetic wrist+hand, the device must flex, pronate and grasp simultaneously.

Traditionally (see, e.g., [9]), this problem has been solved by directly gathering from the subject the sEMG signals corresponding to the required multi-DOF activation(s) – in the above case,  $X_{IL}$  would be available, and a new training set (denoted  $mf$  for multi-finger)

$$X_{mf} = \{(X_0, (0\ 0)), (X_p, (1\ 0)), (X_L, (0\ 1)), (X_{IL}, (1\ 1))\}$$

could be used to determine the  $f$ s. This method will yield the expected approximants, but becomes quickly unfeasible as the number of DOFs,  $m$ , grows, since the number of possible combinations grows exponentially with it. (The most advanced hand prosthesis in the world at the time of writing, the *i-LIMB Ultra Revolution* by Touch Bionics, see [www.touchbionics.com](http://www.touchbionics.com), has  $m = 6$ , which becomes 7 or 8 if a

self-powered prosthetic wrist is additionally used.)

An alternative way is that of estimating  $X_{IL}$  from  $X_I$ ,  $X_L$  and/or  $X_0$ ; that is, trying to build a machine which will *generalise to multi-DOF activations although it has been trained on single-DOF activations only*. The only attempt so far at solving this problem, as far as we know, appears in [11] for two DOFs of the wrist plus hand opening/closing. In this work, *Non-negative Matrix Factorisation* trained on  $X_{sf}$  yields a model acting both as a linear predictor of the required activations *and* as a linear “un-mixer” of multi-DOF activation signals into single-DOF ones. Although we have no comparative results so far, we speculate that NMF will hardly generalise to the case of single fingers, for which a linear approach has been shown to produce unacceptably low prediction accuracy [5].

We rather propose to artificially augment  $X_{sf}$  in order for it to enable the desired generalisation by any machine learning method trained on it (possibly non-linear). We therefore look for a function  $\mathcal{F}$  such that  $\mathcal{F}(X_0, X_I, X_L)$ . If such a function is available, then an “enhanced” training set  $X'$  can be built out of  $X_{sf}$ ,

$$X' = \left\{ \begin{array}{l} ((X_0, (0\ 0)), (X_I, (1\ 0)), (X_L, (0\ 1))), \\ (\mathcal{F}(X_0, X_I, X_L), (1\ 1)) \end{array} \right\}$$

such that training on  $X'$  will yield the required approximants. Notice that  $X'$  is built with *no* explicit knowledge of  $X_{IL}$ , avoiding the above-described exponential blowup of training time and effort.

### LINEARLY ENHANCED TRAINING (LET)

A very simple idea to build such an  $\mathcal{F}$  is that of considering the multi-DOF activation signal as a *linear combination* of the single-DOF signals involved in it. This hypothesis seems reasonable since both sets of motor units involved in the single-DOF activations must participate simultaneously in the multi-DOF activation, to different degrees; we will also assume that the multi-DOF activation samples lies somewhere on the vector in  $\mathbb{R}^d$  bisecting the two vectors corresponding to the single-DOF activations:

$$\mathcal{F}(X_0, X_I, X_L, \alpha) = \{x \mid x = \alpha[(x_I - x_0) + (x_L - x_0)], \\ \forall x_0 \in X_0, x_I \in X_p, x_L \in X_L\}$$

where  $\alpha$ , for which we assume, must be found by exhaustive search. This procedure adds to the original training set one cluster of linearly-built artificial sEMG samples per each DOF combination, and is therefore called Linearly Enhanced Training (LET). Notice that LET is in principle applicable to any  $k$ -ary combination of single-DOF activations (not only pairs), and to an arbitrary number  $m$  of them, in which case  $2^m$  parameters  $\alpha$  must be found;

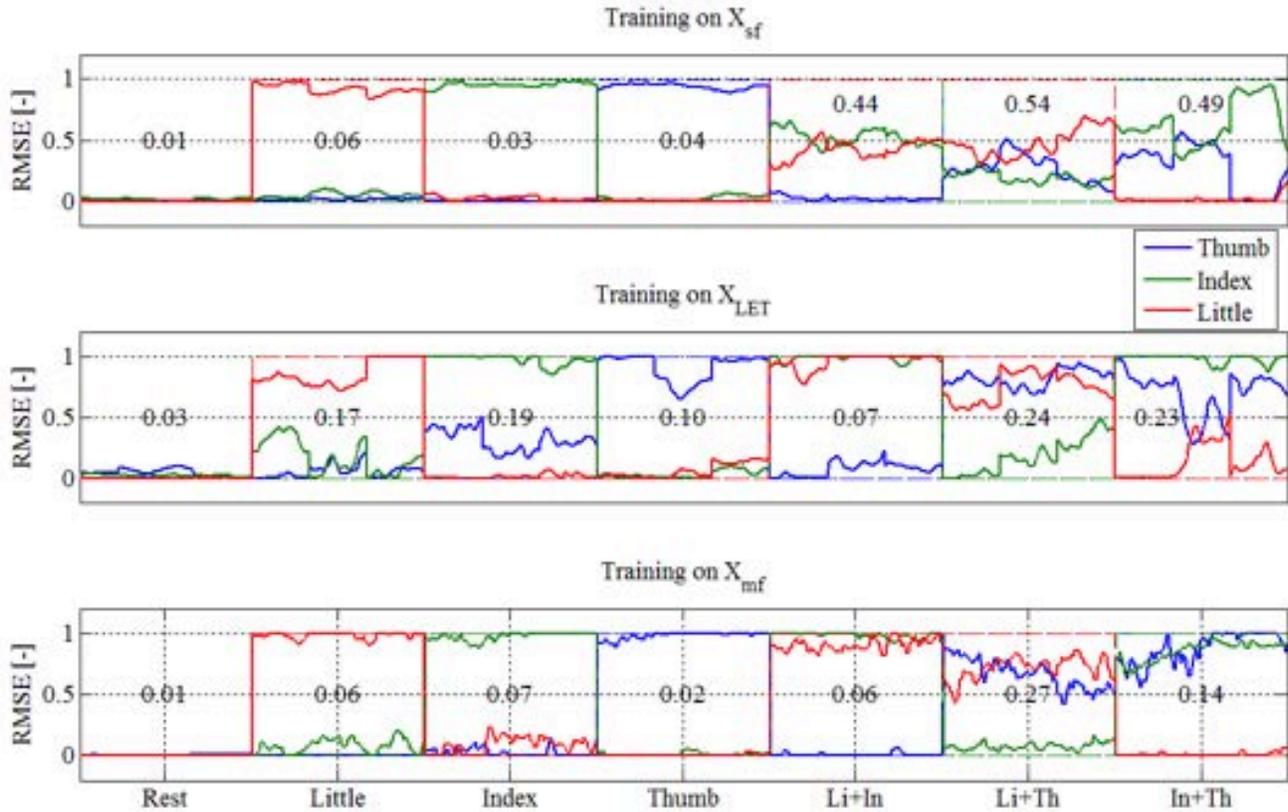


Figure 1: NRMSE for a typical subject for each DOF, DOF combination and training set. Each block shows the error evaluated over the three repetitions of the related single- or multi-DOF activation.

moreover, it is independent of the machine learning method of choice. Notice, however, that the LET-enhanced training set  $X_{LET}$  is exponentially larger than  $X_{sf}$  (but still just as large as  $X_{mf}$ ), which could be problematic in case the training heavily depends on its size.

## EXPERIMENT DESCRIPTION

In order to partially validate the LET procedure, we set up a simple psychophysical experiment, stimulating human subjects to apply 3 single-DOF activations, plus all *pairs* of them, while recording their sEMG signals; we then compared the prediction accuracy of a known regression method trained, in turn, on  $X_{sf}$  and  $X_{LET}$ ; for further comparison, the accuracy obtained by training on  $X_{mf}$  was also evaluated. We expected the performance obtained using  $X_{LET}$  to lie somehow in the middle between those obtained using  $X_{sf}$  and  $X_{mf}$ .

Notice that in this preliminary experiment we *do* use the explicit knowledge of  $X_{IL}$  in order to estimate  $X_{LET}$ , with the hope of finding that the required coefficients  $\alpha$  can be treated as invariants across multi-DOF activations and subjects.

## Subjects

Seven healthy human subjects (age 23÷42yrs, 6m/1w) were recruited for the experiment. Each subject received a thorough description of the experiment; informed written consent was obtained from all participants. Experiments with sEMG have been approved by the Ethical Committee of the DLR.

## Materials and methods

The sEMG signal was measured using ten *MyoBock 13E200* electrodes by Otto Bock ([www.ottobock.com](http://www.ottobock.com)), uniformly placed around forearm close to the elbow, using an elastic biocompatible adhesive bandage. These electrodes provide an amplified, band-pass filtered and rectified signal. To reduce noise a Butterworth filter of 1<sup>st</sup> order is applied with cut-off frequency of 1.5Hz. The sEMG data was collected at approx. 46Hz using a standard analog-to-digital conversion card connected to a Windows machine via Ethernet. (The setup closely follows that of [5] – the interested reader is referred once again to that paper.)

### Experimental protocol

Each subject was comfortably seated in front of a table with a large monitor, on which two 3D hand models were shown, one acting as a visual stimulus (i.e., what the subject was required to do) and the other showing the predicted forces as finger flexions. The experiment was divided into three sessions (rest was allowed in between sessions).

The first session consisted of three repetitions of, in turn, little finger flexion, index finger flexion, thumb adduction, little and index finger, little and thumb, thumb and index – that is, three single-DOF activations and three multi-DOF ones. Data gathered during the single-DOF activations are while the union of  $X_{sf}$  and data gathered during the multi-DOF activations are  $X_{mf}$ . The collected data was used to determine the three coefficients – one for each multi-DOF activation – by minimising the Euclidean distance between the artificial samples in  $\mathcal{F}$  and the “true” samples in  $X_{mf}$ . For example, for the index and little finger

$$\alpha_{iL} = \arg \min_{\alpha} \|\mathcal{F}(X_0, X_I, X_L, \alpha) - X_{iL}\|^2$$

Using the  $\alpha$ s found this way,  $X_{LET}$  was built and a non-linear, incremental regression method was then trained using, in turn,  $X_{sf}$ ,  $X_{mf}$  and  $X_{LET}$ . The chosen method was Ridge Regression with Random Fourier Features [12], which we have already successfully employed in [5]. This method requires finding three hyperparameters  $\lambda$ ,  $D$  and  $\sigma$ , two of which ( $\lambda$  and  $D$ ) were set at standard values of 1 and 700 (see [5] again), whereas one optimal value of  $\sigma$  for each training set was found by grid search and 3-fold leave-one-repetition-out cross-validation over each related training set.

In the second and third session, the prediction was started using, in turn, the model obtained by training on  $X_{sf}$  and on  $X_{LET}$ ; each subject was then again shown, using the stimulus hand, the same DOF activations as in the first session, and instructed to have the prediction hand reproduce them and to keep them stable for 3 seconds. Online testing on  $X_{mf}$  was neglected in order to keep the experiment as short as possible; rather, the performance using  $X_{mf}$  was evaluated offline by training on the first two repetitions and testing on the third.

### EXPERIMENTAL RESULTS

The optimal values of  $\alpha$  were determined to lie in the range  $0.7 \pm 0.2$  across all multi-DOF activations and subjects (mean plus/minus one standard deviation). The Root Mean-Squared Error (RMSE) was calculated for each of the 7 different activations (rest, 3 single-DOF, 3 multi-DOF) for each subject, for each training set and for each single- or multi-DOF activation. As it happens in [5], the ground truth

is represented by the visual stimulus values, ranging from 0 to 1; the RMSE is therefore expressed in arbitrary units.

Figure 1 shows the results for one typical subject. As expected, the prediction error on single-DOF activations (and rest) is good to excellent in all three cases with a slightly worse result obtained while training on  $X_{LET}$ , whereas the error on multi-DOF activations is high when using  $X_{sf}$  (0.44, 0.54 and 0.49 for little+index, little+thumb and index+thumb in turn) and reasonably good when using  $X_{LET}$  (0.07, 0.24 and 0.23). Surprisingly, the error when using  $X_{mf}$  is on average just a little better than when using  $X_{LET}$  (0.06, 0.27 and 0.14).

Figure 2 shows the results averaged over all subjects. The trend is confirmed: with respect to training on  $X_{sf}$ , training on  $X_{LET}$  makes the error slightly worse for single-DOF activations but largely better for multi-DOF activations; and surprisingly, training on  $X_{mf}$  does not yield a considerably large improvement.

### CONCLUSIONS AND DISCUSSION

The LET procedure, presented in this paper, enables in principle any machine learning method to predict multi-DOF activations using data collected during single-DOF activations only. LET works by approximating the multi-DOF activation sEMG signals, which are unfeasible to gather directly, using a linear combination of the related single-DOF signals. In a psychophysical experiment, using the LET technique, a standard machine learning method was able to obtain prediction error values on multi-DOF activations similar to those obtained on single-DOF activations.

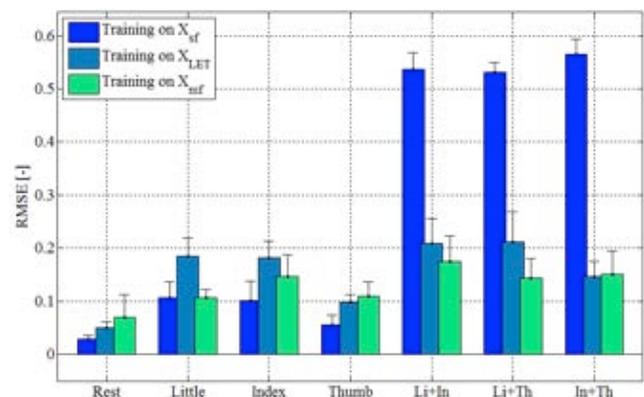


Figure 2: RMSE per training set per DOF  
Mean and standard error of the mean across all subjects

We consider this a preliminary result, both because the number of subjects and DOFs considered is low, and because multi-DOF activations were explicitly used to build. However, as the values of the s lie in a quite close range for all subjects, we plan to enforce LET in a completely automatic

way, and compare again its performance against the usage of and . Although initial though, this result looks promising and future work includes chacking whether it generalises to, e.g., the (combined) DOFs of the wrist, possibly in combination with a few grasping postures.

## ACKNOWLEDGEMENTS

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## A DEXTEROUS HAND PROSTHESIS BASED ON A GENEVA DRIVE: PRELIMINARY DESIGN

Marco Controzzi, Francesco Clemente, Sara Dragoni, and Christian Cipriani  
*The BioRobotics Institute, Scuola Superiore Sant'Anna, V.le R. Piaggio 34 56025 Pontedera, Italy*

### ABSTRACT

This paper presents the preliminary design of a new dexterous hand prosthesis provided with an innovative mechanical transmission based on a Geneva drive coupled with a four bar mechanism. This solution has been designed in order to reduce the number of actuators (and, in turn, weight and power consumption) as much as possible still allowing flexion of the index independently from the thumb and the last three fingers. This would allow the hand to perform the main grips and functions used in the daily life.

### INTRODUCTION

People who have suffered the amputation of one or both their hands are unable to perform motor tasks which are extremely important and essential for conducting life autonomously. Traditional myoelectric prostheses – one degree of freedom (DoF) grippers – partially restore some of these motor functions, however they lack of dexterity. This can be largely attributed to the lack of independently driven digits and of the abduction/adduction movement of the thumb, which is crucial in order to perform a wide range of prehensile patterns. For this reason several dexterous anthropomorphic prosthetic hands were recently developed [1], [2], [3], few of them reaching the market. Among these, the *iLimb* (by Touch Emas Ltd) and the *BeBionic* (by RSL Steeper Ltd) prostheses are five-fingered hands with independent flexion/extension of all the digits (i.e. five independent actuators). The main drawback of the *BeBionic* and *iLimb* hands is that the abduction/adduction movement is not automatically driven. The individual wearing the prosthesis must rotate the thumb manually in either the opposition (thumb facing the fingers as to perform a power grasp) or the reposition (thumb facing the lateral aspect of the index as to perform a lateral grasp – the key grip) configurations. Although the abduction/adduction movement is automatic in the latest version of the *iLimb* hand (namely *iLimb ultra revolution*), this is obtained by further increasing the number of actuators (for a total of six). Such a solution, could not be the optimal, as it results in an increase of the prosthesis weight and power consumption. In addition, one could argue that the reduced space available within the palm of the *iLimb* forced the designers to place

the thumb abduction/adduction axis in an anatomically incorrect orientation, which might result in a poor cosmetic appearance.

The *Michelangelo* hand (by Ottobock) still being five-fingered, adopts a completely different mechanical architecture with respect to the previous examples, as it is actuated by just two motors [4], [5]. The main drive of the hand, embedded within the palm, allows for the simultaneous flexion/extension of all the digits with two different transmission ratios, one with respect to the thumb and the other with respect to the fingers. These transmission ratios switch based on the direction of rotation of the drive. More in detail, considering a left handed prosthesis, a clockwise rotation of the main drive produces a predominant flexion of the fingers (used to execute a lateral grasp), whereas when the drive rotates counter-clockwise fingers and thumb flex in a similar manner (in order to perform a power grasp). The thumb abduction/adduction movement is actuated by an electrical motor housed within the thumb itself. In our opinion the main drawback of this architecture is due to the fact that all digits are practically rigidly coupled, i.e., they flex/extend simultaneously. This feature disables to perform useful and common activities of daily living: for example it is not possible to flex the index when the flexion of the thumb is locked, as required to press the trigger of a drill while holding it. For the same reason it is not possible to take the index apart the last three fingers, as required to press buttons. This function instead is available in *BeBionic* and *iLimb* hands. In short, although in the last decade classical robotic knowledge has produced significant results in the field of prosthetic limbs, and several research groups are investigating new and more advanced devices [2], [3], current clinically available myoelectric hands still present design limitations that prevent for a wide diffusion.

In this work we present the concept and design of a new motorized hand, with three actuators and five digits, aimed to overcome the mechanical limitations of currently available myoelectric prostheses. The hand is designed around a miniaturized transmission mechanism – a Geneva drive – that allows flexion of the index finger and abduction of the thumb. This compact mechanism allows to reduce the number

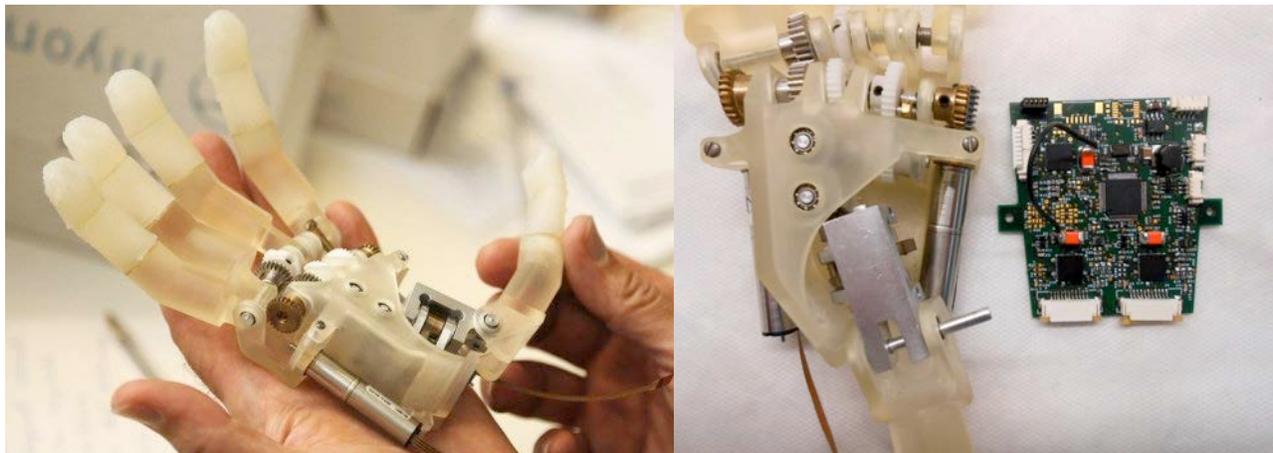


Figure 1: Prototype of the hand prosthesis and the electronic board developed

of actuators still allowing flexion of the index independently from the thumb and the last three fingers.

### SYSTEM OVERVIEW

#### Functional requirements

Functional requirements for our design were set according to the results of surveys from the amputees community [6] and to the approximate percentage of utilization of the main grips in the Activities of Daily Living (ADLs) [7]. A prosthetic hand should allow amputees to perform the following functions: i) lateral grasp, ii) cylindrical grasp, iii) pinch grasp, iv) index pointing-up (e.g. to press the lift button), vi) index pointing-down (e.g. to press a keyboard button), and the vii) neutral position (i.e. the hand assuming a small form factor in order to easily don/doff a coat, shirt etc.). In addition to function (and cosmetic appearance) a prosthesis should match robustness constraints, required for its practical use. Thus we decided for a functional and robust (i.e. using a reduced number of components) architecture comprising the following features: i) independent flexion/extension of the three last fingers by means of a single actuator (in the palm), ii) independent flexion/extension of the thumb actuated by a motor inside the thumb itself, iii) semi-independent flexion/extension of the index finger and adduction/abduction of the thumb using a Geneva drive 0 actuated by a third motor (in the palm). The three motors are brushless. The hand includes an embedded printed circuit board hosting the controller and touch sensors based on FSR (Force Sensing Resistors) in order to allow for sensory feedback to the user (future development). The prosthesis was designed in order to allow transradial hand fittings (cf. Figure 1).

#### Transmission design

The main novelty of the prosthetic hand presented in this work is the transmission that couples the flexion/extension

of the index and the abduction/adduction of the thumb thus allowing both movements by using a single motor (the concept is depicted in Figure 2). The transmission is composed of a four-bar mechanism which is attached to the actuator on the crank (1) side and to the flexion joint of the index finger (3) on the rocker (2) side. Lengths of the links composing the planar quadrilateral linkage were designed in order to obtain a crank-rocker configuration (Figure 2–A). Considering a static analysis, the closer the transmission angle (i.e. the angle between the crank (1) and the coupler (0)) is to 90° deg, the larger is the force transmitted to the index finger (and in turn to the grasped object). We thus synchronized the four bar mechanism and the index finger so that the above-mentioned point would correspond to the index half-flexion. In turn, one of the two dead points of the mechanism (i.e. the configurations in which the crank (1) and the coupler (0) are aligned) happened to be when the index is fully extended (the other dead point is never reached). The crank (1) is connected in series with a Geneva drive (4)–(5) which output shaft is attached to the adduction joint of the thumb (6). The Geneva drive translates a continuous input rotation into an intermittent rotary output motion. During the rotation of the drive wheel (4), a pin of which it is provided with enters into a slot of the driven wheel (5) advancing it by one step. When the pin of the drive wheel (4) leaves the slot of the driven wheel (5), the latter is locked in its current position thanks to an elevated circular disc on the former. The Geneva drive was designed to produce an intermittent movement of the thumb abduction between two positions (Figure 2-T2): open (used, e.g., in the lateral grasp) and closed (used, e.g., in the power grasp). This movement was synchronized with one of the dead points of the four-bar mechanism (the one corresponding to full index extension), so that the index finger movement is minimized during the thumb adduction (Figure 2-T2). Finally, a rotary potentiometer was attached to the Geneva drive in order to measure the position.

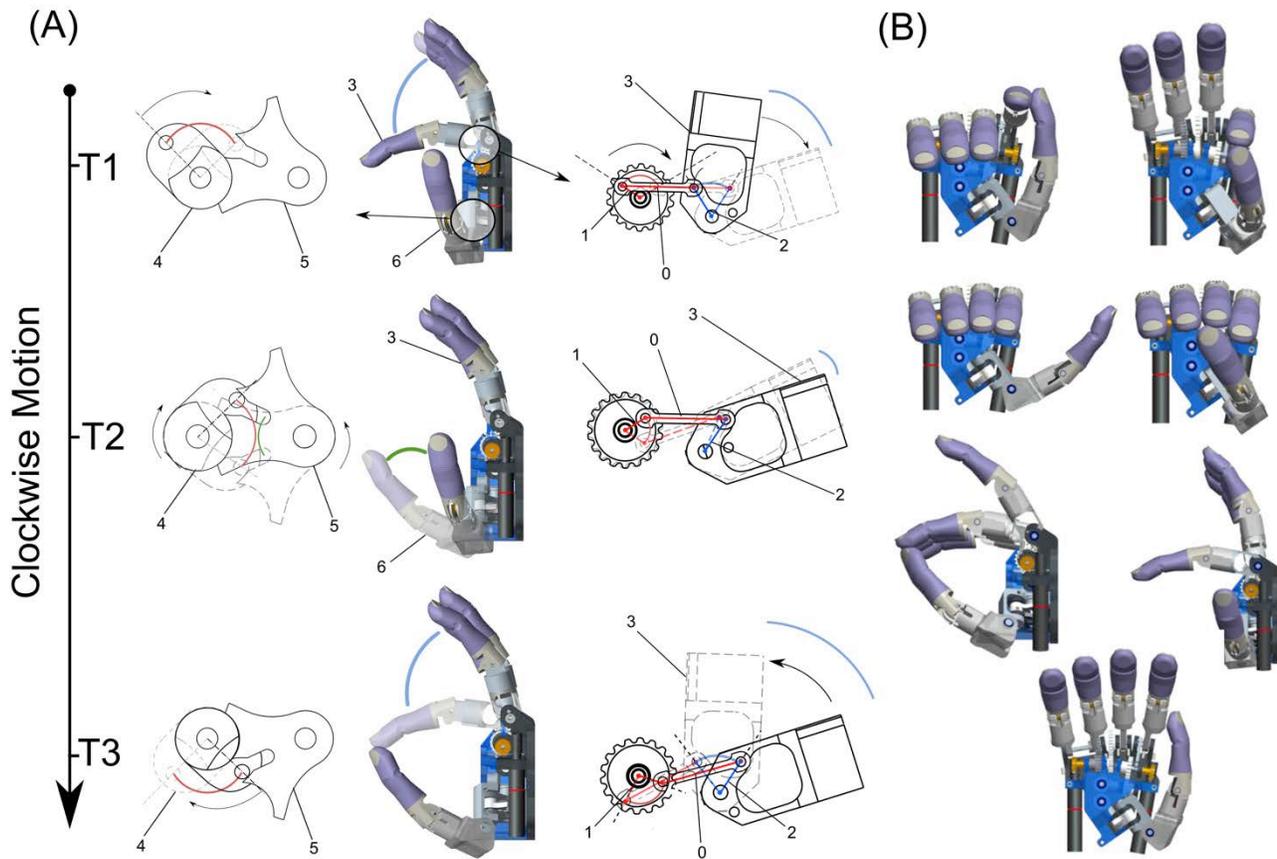


Figure 2: (A) Sequence showing the actuated movements of the hand: from T1 to T3 the clockwise motion of the motor in three phases highlighting the transmission connecting the index flexion joint (3) by means of the four bar mechanism (1) – (2) to the thumb abduction joint (6) through a Geneva drive (4) – (5) is shown. T1 Index opening independently from the thumb abduction; T2 Thumb abduction movement and slight movement of the index flexion; T3 Closing movement of the index finger independently from the thumb abduction. (B) Some of the postures available with the current kinematical architecture.

The middle, ring, and little fingers of the hand are rigidly connected and actuated by a single motor housed within the palm. The thumb flexion is achieved by means of a worm-gear transmission actuated by a motor housed within the thumb. All actuators are non back-drivable. The non-back-drivability was obtained by means of worm gear transmissions. This allows the actuators to maintain a stall torque without energy consumption. Therefore it is possible to switch off the power once a stable grasp is achieved.

#### Embedded control system

The core of the electronic controller is the 16-bits based Digital Signal Controller (DSC, dsPIC33FJ128, Microchip), optimized for the real-time management of digital signals (Figure 3). The DSC uses a standard serial bus (RS232) in order to communicate with an external human-machine interface, or HMI (or with a PC for debug). The brushless motors are controlled by commercially available brushless motor drivers integrated circuits (L6235Q, ST) and by the

necessary circuitry (in particular, digital to analog converters – DAC – and current shunt monitors in order to measure the current flowing in the motor) (Figure 3). Touch sensors and Geneva drive position readouts (Thumb Abduction Sensor) are also collected by the DSC and used in the control strategy.

The embedded controller was designed to be driven by an EMG pattern recognition control scheme able to decode the user's desired hand posture. To this aim the DSC implements a Finite State Machine (FSM – Figure 4) in order to receive external commands from the HMI and accordingly, to implement low-level motor control functions. Each state of the FSM corresponds to one of the postures and grasps physically achievable by the hand. As depicted in Figure 4, the hand can switch to one posture/grasp from the *open state* only. The transition from one grasp to another one that requires changing the thumb abduction position consists of two phases: in the first phase the Geneva Drive rotates in order to correctly position the thumb; in the second phase

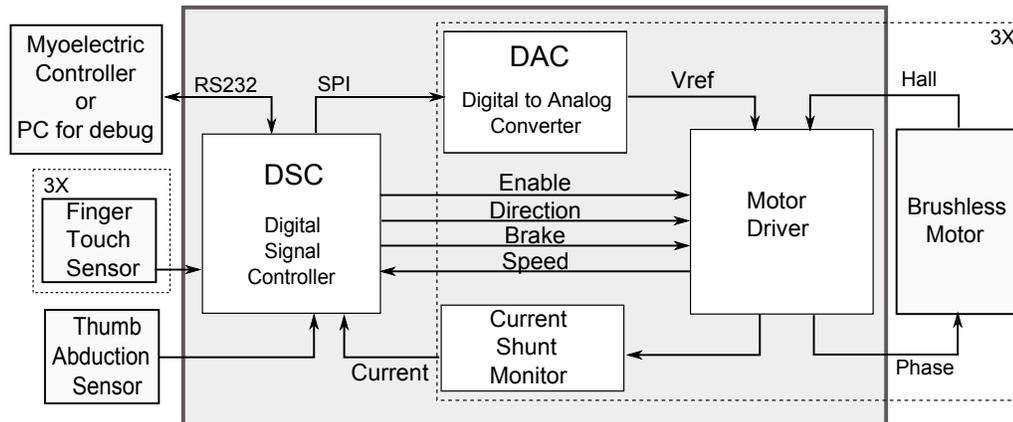


Figure 3: Control architecture of the hand.

the motors flex the involved fingers according to a torque (or speed) control.

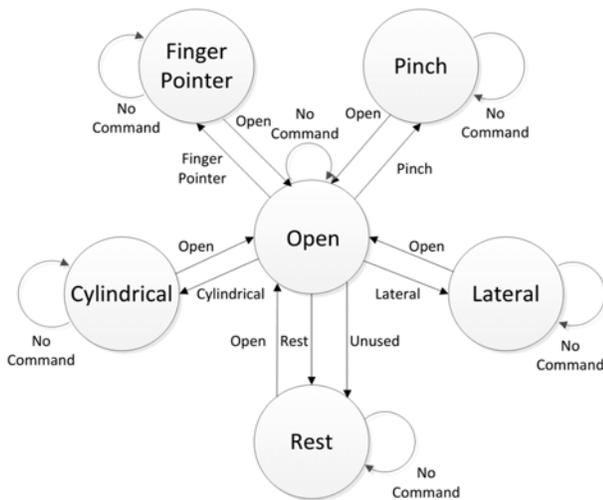


Figure 4: Finite State Machine diagram regulating the operation of the hand.

#### Expected performance

All of the actuated joints are driven by 8 Watt DC brushless micro motors (Maxon Motor, model EC10) with an integrated planetary gearhead (64:1). Each motor fits into a 10 mm diameter and 48 mm length cylinder and weighs 21 g. The unit provides a 70 Nmm continuous torque with peaks up to 150 Nmm (limited by the gearhead). With the actual mechanical transmission the fingers will be capable to apply up to 30 N at fingertip and to move with a maximum speed of 100 deg/s. Concerning the power consumption for this prosthesis, the current required to the three motors during the closure movement is about 450 mA for 0.8 s (i.e. the time to completely close at maximum speed). During the grasping phase this value can rise up to 2.4 A (stall current)

for 0.2 s (maximum estimated time required to securely grasp an object). So considering these values, the average energy consumption for a single grasp is 0.25mAh. Therefore two commercial Otto Bock 757B8 NiCad rechargeable batteries (6V, 250mAh), connected in series, could ensure an autonomy of about one thousand grasps for the prosthesis. The expected weight is less than 400 g and will be achieved employing aluminium alloy and reinforced thermoplastic materials.

#### ACKNOWLEDGEMENTS

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## REDEFINING NORMS FOR UPPER LIMB PROSTHETIC TREATMENT THROUGH INTERNATIONAL COLLABORATIONS

Dan Conyers CPO, FAAOP, John Miguelez CP, FAAOP  
*Advanced Arm Dynamics, Inc.*

### ABSTRACT

The global marketplace is rich with opportunities for international and cross-cultural collaborations that can stimulate rapid improvements in upper limb prosthetic patient care around the world. Beyerlin et al, (2003)<sup>1</sup> provides perspective on marketplace evolution by comparing the patterns of just a few decades ago when businesses supplied goods and services to customers based on their proximity. In that economic model, only large multinational companies engaged in global trading practices whereas today, there is intellectual and technical cross-pollination in a vast array of businesses. The specialized world of upper limb prosthetics is an excellent example of how international scientific collaboration improves processes and outcomes for researchers, manufacturers, clinicians and patients. From the inception of new ideas to the commercialization of new products, collaboration accelerates results. Innovations in treatment can often be applied to the delivery of care and products to patients in regional markets early in the collaborative process. Additionally, collaboration may quickly bring new methods, materials or devices into global standard operating procedures since multiple professional viewpoints can be expressed in regional and international publications and presentations. A successful collaboration generates the potential for future collaborations by others based on the information that is shared in publications or presentations. The awareness and experiential history that the initial collaborators have of each other's working styles and successes may lead naturally to other projects among these same collaborators.

### INTRODUCTION

Numerous studies provide evidence that collaborative interaction is a natural tendency that originates in early childhood.<sup>2, 3, 4</sup> These universal experiences provide the context to understand the more specific upper limb prosthetic collaborations outlined in this paper. The seven stages of collaboration presented herein are based on the authors' experiences of how to effectively establish collaborative relationships, define the essential elements, and deliver on the collective outcomes. We hypothesize that the globalization

of information technology is contributing to earlier market penetration of newly developed products, materials and treatments. We believe that domestic and international collaborations are a catalyst for rapidly disseminating multi-perspective information to the global upper limb prosthetics industry. The "Results" section of this paper specifies topics that have been presented at the Myo Electric Controls Symposium (MEC) that involved the work of some of the collaborative teams that the authors have participated in.

### BACKGROUND

Over the past two decades the authors have engaged in international and regional scientific collaborations that include clinical partners, research partners, manufacturing partners, and patient participants from Scotland, Norway, England, Italy, Austria, Germany, Columbia, China, UAE, Sweden, India, Canada, the United States, and many smaller regional areas. With each successive experience we have come to better understand the principles outlined in this paper and have made refinements to improve on our own collaborations. Experience has shown that being selective and analytical regarding which opportunities to pursue is an essential initial step. Also critical is the identification of deliverable benefits to the collaborators and the patients who stand to receive our clinical care and/or new products. A simplistic outline of this process includes seven key stages: 1. Establish a relationship 2. Identify a mutual desire and ability to collaborate 3. Build trust 4. Define collaboration topic(s) to address 5. Design workflow and responsibilities 6. Agree on deliverables 7. Identify benefits to the collaborators and to the patients

### METHODS

#### Establish a Relationship

Collaborative relationships in upper limb prosthetics begin like many other relationships—there is shared interest or mutually beneficial focus. Publications, meeting attendance and lectures are incubators for ideas that capture the interest of potential collaborators. Internationally staged meetings (ISPO, MEC, TIPS, OT World Congress) along with internationally distributed publications (*Prosthetics*

and *Orthotics International*, *International Journal of Surgery*, *The International Journal of Robotics Research*, *International Journal of Engineering Science*, etc.) create opportunities to present ideas globally and learn about ideas that may become the foundation for collaboration. These presentations and publications provide the earliest momentum for collaboration—awareness. Becoming aware of a concept in upper limb prosthetics, or connecting with someone else's viewpoint on a topic that is already a passion for you and your group, can be the spark for taking action. Reaching out to an international author or presenter can be the first step in exploring whether the individuals or groups share a sustainable passion for a particular subject. Connecting further, on a social level, will indicate whether the personalities and interests of the individuals lend themselves to the collaborative process. Shared passion for a topic will provide the initial impetus required to further explore opportunity for collaborative possibilities.

#### Identify a Mutual Desire and Ability to Collaborate

Topics that initially brought the potential collaborators together are elaborated on, discussed and some are eliminated. This process, which is usually done in person and of a conversational nature, further reveals each participants' resources and capacity for collaboration. Considerations for this phase of the process include: language capabilities, time zone differences, availability to meet electronically and in person, access to cooperative and appropriate patient subjects, and various physical and intellectual resources. It is also important to discuss the time and resources going into the project and the specific benefits coming out of the project for each participant. Shared passion alone will be insufficient to create a lasting positive collaboration and yet that passion must exist as a baseline for the group to be willing to overcome the challenges associated with international collaboration. Making the work "more fun than fun" is discussed in Bennis and Biederman's book, *Organizing Genius*<sup>5</sup>, and supports our experiences of bringing our collective excitement to the process. Motivation and enthusiasm from all members for entering into the collaboration is predictive of participation and engagement moving forward.

#### Build Trust

Building trust is critical to each member's ability to follow through on a desire to collaborate. Equally important to sustainable collaboration is to achieve an understanding of the basic assortment of personalities that will be involved in the collaboration. It is preferable to do this in a business casual social setting, over a series of meals or drinks where information is shared about each other's interests and passions both inside and outside the specific area of upper limb prosthetics. It is not imperative that all parties share similar interests, and is in fact more stimulating and productive when a wide variety of experiences and passions are brought to the

collective table. In our experience there is strong evidence that building trust requires the potential collaborators to protect their separate intellectual property in a formal manner. This is accomplished by creating mutually agreed upon documents in the form of a non-disclosure agreement (NDA) or a mutual non-disclosure agreement (MNDA). This seemingly simple step establishes an environment for the open exchange of ideas that the collaborators jointly agree will not be disseminated outside of the group, or with the industry at large, without the express agreement of the collaborators. This sets the stage for reciprocal communication and participation. As in all relationship building, an element of trust is the highly critical piece of the equation which must be mutually determined, resolutely protected and not biased to any one participant unless agreeable to all.

#### Define Collaboration Topic(s) to Address

Once collaboration potential is affirmed, and the future collaborators have begun to attain trust, topics can be further narrowed to fill a specific need of the patient population. It is important to consider the potential benefits. For clinicians, this could be improved treatment options or outcomes; for researchers or manufacturers it could be new or improved products or materials. For patients, the benefits could be a range of well-developed treatment choices. Additionally, there is robust potential for multi-perspective development of collateral materials to assist in disseminating, adopting, training and utilizing the concept or product in the practical setting of the global marketplace. The agreed upon collaboration should include a discussion of timing for the project so that realistic expectations are formulated. It is important for the collaborators to understand which aspects of the project will fall under the NDA or MNDA, and which aspects of the project are based on general knowledge or prevalent awareness already existing in the industry or global marketplace. The authors have been involved in multiple new product developments and commercialization processes, as well as improvements to existing products and treatment options. We have experienced a range of successes and failures in accurately documenting expectations. This stage of collaboration is paramount to the long-term success of the group. When each member feels protected, we have experienced the nearly immediate mutual benefits of comparing regional materials, knowledge and techniques that were previously unknown or poorly understood to one or both collaborators. Seemingly simple regional solutions are often based on culturally familiar items, or processes, locally sourced materials or climatically influenced solutions for dealing with specific challenges. It has been our experience that locally sourced solutions are often taken for granted based on a non-global perspective or a misunderstanding of their availability and usefulness in other regions. These solutions may translate almost immediately to other similar

regions, or with modifications, be adapted for other diverse regions.

#### Design Workflow and Responsibilities

Once a topic for collaboration is identified, it is important to spend time working through a myriad of details. This includes beginning to determine perceived and actual responsibilities, and the expectations for realistic timing of each step for all members of the group. The authors have spent considerable time in discussions with collaboration partners to make certain that each party understands and expresses their desires, their expectations, and their responsibilities for the collaboration. These discussions are the framework where goals begin to form and participants begin to see the true feasibility of the collaboration. Once goals are established, a workflow can be designed to provide a blueprint for the stages of development, testing and feedback, and to define the timelines and the deliverables. Elevating the standards for these discussions and information sharing is something that Beyerlein, et al (2003)<sup>1</sup> describes in detail. For research and manufacturing collaborations this includes timelines for design work, prototyping, patient testing, feedback, modifications, redesign and improvements, beta testing, and manufacturing. The final phase of timelines addresses moving the product into pre-product launch preparations that culminate in marketplace distribution. For purely clinical collaborations, this includes identification of materials, techniques or components to be tested, numbers of subjects, types of subjects, outcome measures that will be used or created, and types of feedback that will be reported and shared. Agreement on who will represent the findings, and in what venue or format, is important to this stage of collaboration. Definition of shared credit for the collaboration is essential to fairly recognize all of the contributors and to build upon the foundation of trust in the present collaboration.

#### Agree on Deliverables

This stage of the collaborative process is crucial from the perspective of aligning the expectations of the collaborators. Deliverables include tangible items such as products, materials, collateral materials, timelines and costs. This stage specifies the participants specific investments of time, energy and money related to the future costs that will be associated with providing the results of the collaboration to the upper limb prosthetic industry. It is vital to determine if the results of collaboration are intended to benefit the general prosthetics industry or are instead designed to create a competitive advantage in the industry for the researchers, manufacturers or clinical caregivers. Clear, written language describing the deliverables will yield a contract that all parties can agree to. This contract will reduce misunderstanding or misuse of the results of collaboration.

#### Identify the Benefits to Collaborators and to Patients

It is the authors' experience that it is essential to identify and communicate the specific benefits to the collaborators and to the patients. By defining this information clearly, each member is afforded the opportunity to clearly visualize the "win" for themselves and for each other, as well as for the upper limb prosthetic patient. When this information is included in the agreement, it helps to define the expectations for shared expense, timeliness, effort, obligation and reward. The rewards can take many forms including the creation of further opportunities for research; manufacturing and release of a new product; new materials or innovative techniques; improvements to an existing product, material or technique; and the potential for future shared projects that build on the results of the initial collaboration.

### **RESULTS**

The authors have repeatedly experienced that a well-executed upper limb prosthetic collaborative project can be highly beneficial to the industry, the clinician, the researcher or manufacturer, and the upper limb prosthetic patient. Specific examples of successful scientific collaborations span the past 20 years and include pattern recognition<sup>6, 7</sup>, implantable controls<sup>8</sup>, multi-articulating hands<sup>9, 10</sup>, powered digits, materials science<sup>11</sup>, control algorithms, socket designs, implantable suspensory techniques<sup>12</sup>, surgical techniques for upper limb amputee management, training protocols and techniques<sup>13</sup>, and other products and materials that are integrated into the industry as standard operating procedure for many upper limb prosthetic specialists worldwide. Our particular perspective is that working in collaboration is exponentially more powerful than working in the isolation of a specific profession, company, region or country. Involving actual patients and collecting thoughtful feedback from a multi-disciplinary clinical team provides a platform for more rapid development of ideas and a more thorough dissemination of the results.

Further, these scientific collaborations have come full circle, disseminating results and new information in the same venues where the collaboration opportunities were originally incubated. This has led to the creation of additional opportunities for the authors and their collaborative partners that continue to build over time with each successful experience. We suggest that the guidelines of this paper be considered by readers desiring to develop their own collaborative focus and contribute to the globalization of ideas, materials, treatment techniques and componentry.

### **CONCLUSION**

Multi-national, multi-profession collaboration can induce the rapid globalization of ideas, materials and

techniques for bringing advanced scientific solutions to patients who seek upper limb prosthetic rehabilitation. The benefit of globalization is seen in the development, dissemination and adoption of *new ideas* into the standard operating procedures for care and treatment around the world. Globalization has a secondary influence on the acceptance of new upper limb prosthetic solutions by providers, patients and payers as evidenced by their integration within the worldwide prosthetics industry. Upper limb prosthetic collaboration that includes researchers, manufacturers, clinicians and patients can cross international boundaries and overcome cross-cultural constraints when the participants adopt a well-structured plan for thoughtful collaboration. The dissemination of each new idea to the global market perpetuates the process of future scientific collaborations. “No one person, no one alliance, no one nation, no one of us is as smart as all of us thinking together.” – James Stavridis<sup>14</sup>. These words are quite simply, a well-phrased justification to apply repetitive multi-national, multi-regional, multi-discipline, multi-center collaborative thinking and action to explore, develop, promote and disseminate ideas, materials, treatments or product concepts from idea to reality that lead to benefits for all involved and rapidly, directly benefit the upper limb patient end users whose lives we seek to improve.

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## ADVANCED SILICONE TECHNIQUES TO ADDRESS VARYING UPPER LIMB PROSTHETIC PATIENT NEEDS

Gene Douglas Dawson, Chris Lake  
*Lake Prosthetics and Research, Euless TX*

### ***ABSTRACT***

Upper limb prosthetic patients present with a variety of needs. On one side of the spectrum, these can be related to the trauma or anomaly that resulted in the need for a prosthetic replacement. For these patients there may be a significant emphasis placed on protection of a fragile residual limb. On the other side of the spectrum, specific activities and/or work-related tasks that the patient desires to accomplish can warrant highly customized socket interfaces. Of all the materials available to the upper limb prosthetist and technician, no other material has the versatility in application as does silicone. The purpose of this paper is to discuss five silicone application principles that our clinical practice has developed in order to serve this highly individualized patient population.

The silicone application principles described will include:

- 1) The construction of air bladders for enhanced suspension, ease of donning, and increased residual limb stability;
- 2) The blending of a lower durometer room temperature silicone with higher durometer vulcanized silicone to provide higher coefficients of friction against the skin while enhancing the structural integrity.
- 3) The construction of defined tunnels for dacron or cable tension mechanisms;
- 4) The construction of suspension wedges and studs as alternatives to more traditional pin lock mechanisms; and,
- 5) The construction of a tube through the silicone that allows the point of air expulsion from the socket interface to be distal with a remotely placed valve;

The clinical applications of the above techniques will be exemplified in a variety of patient cases. These cases will encompass levels of amputation ranging from partial hand to transhumeral as well as incorporating various residual limb presentations; those that are heavily scarred to those with heterotopic ossification. In addition to the technical and

clinical principles, various tips and tricks will be discussed to help reduce the silicone learning curve an oft cited barrier to entry for this material.

## DEVELOPMENT OF THE BENTO ARM: AN IMPROVED ROBOTIC ARM FOR MYOELECTRIC TRAINING AND RESEARCH

Michael R. Dawson<sup>1</sup>, Craig Sherstan<sup>2</sup>, Jason P. Carey<sup>3</sup>, Jacqueline S. Hebert<sup>1,4</sup>, Patrick M. Pilarski<sup>2,4</sup>

<sup>1</sup>*Glenrose Rehabilitation Hospital, Alberta Health Services*, <sup>2</sup>*Department of Computing Science, University of Alberta*,

<sup>3</sup>*Department of Mechanical Engineering, University of Alberta*, <sup>4</sup>*Division of Physical Medicine and Rehabilitation, University of Alberta.*

### INTRODUCTION

Myoelectric training is a key step in transitioning upper limb amputee patients to a successful myoelectric prosthesis fitting. Although training was found to be important in fittings with children [1], a gap in the literature exists for linking training systems to improved outcomes in adults. In order to close this gap additional studies must be performed to more closely evaluate the effect of different kinds of training on clinical outcomes. Our group hypothesizes that improvements in training pre and post fitting will help improve a patient's confidence in myoelectric control and more closely align their expectations to the current capabilities of myoelectric technology.

A review of myoelectric training devices was previously performed to identify the strengths and weaknesses of existing systems [2]. In general existing commercial systems were found to only allow an amputee to control a single degree of freedom (i.e. hand open/close) using a pair of EMG electrodes placed over antagonistic muscles on the residual limb. Key improvements for future systems included integrating the ability to perform more functional tasks, tracking relevant outcome metrics over time, accommodating both conventional and pattern recognition controllers, and employing a variety of training methods such as signal strength display, video games, virtual reality, and robotic arms.

To address these issues the Myoelectric Training Tool (MTT) was designed to assist in the training and assessment of amputee patients in advance of prosthesis fitting [3]. The MTT was also designed as a research platform for testing novel EMG controllers based on machine learning methods and sensory feedback systems. The original research version of the MTT included a desk mounted off-the-shelf AX-12 smart robotic arm (Crustcrawler, Inc), EMG acquisition system, EMG controller, and graphical user interface. It allowed control of up to two degrees of freedom simultaneously for targeted muscle reinnervation

(TMR) patients. More recently a simulated version of the robotic arm and a myoelectric Tetris video game have also been incorporated into the system and the physical robotic arm was upgraded to the slightly stronger AX-18 smart arm. Since 2010 the research version of the MTT has successfully been utilized in subject trials with 15+ able-bodied subjects and 5 upper limb amputee subjects [3-6].

As we pushed the limits of the training tool in research and training applications we discovered certain limitations with the AX-12 and AX-18 smart arms. As seen in Fig. 1 the arm employs two actuators at the elbow and wrist flexion joints to increase the amount of torque available. However, when fully extended and moving at the low speeds typical for myoelectric control the elbow servos would tend to overheat and eventually shut themselves down. This was partly attributed to each actuator in the elbow having its own independent controller which caused synchronization issues, but also the AX-12 and AX-18 actuators were in general underpowered for use in a myoelectric training platform. In practise this meant limiting the payloads used in experiments to less than 50g and the trial times to less than 10 minutes. Some of the other areas we identified for improvement included the shoulder joint, which tended to move jerkily, and the general non-anatomical appearance of the arm, which made it more difficult for subjects to imagine it as a prosthesis. To overcome these issues we decided to build an improved robotic arm, the Bento Arm, designed specifically for our myoelectric training and research applications.

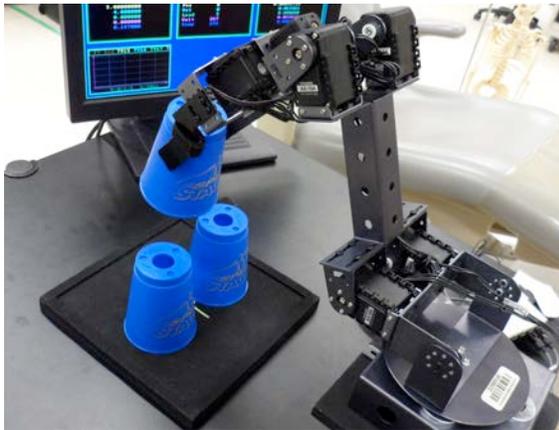


Figure 1 - The AX-18 Smart Arm being used with myoelectric control in a cup stacking task

**DESIGN SPECIFICATIONS**

The preliminary design specifications for the Bento Arm are summarized in Table 1 and Fig. 2a.

Table 1: Design specifications for Improved Robotic Arm

Item	Design Specification
Size	Full scale with anatomically correct proportions
Mass (excluding stand)	≤ 1.5 kg
Payload Capacity	≥ 0.3 kg
Degrees of Freedom	<ul style="list-style-type: none"> <li>➤ Shoulder rotation</li> <li>➤ Elbow (flexion/extension)</li> <li>➤ Wrist rotation</li> <li>➤ Wrist (flexion/extension)</li> <li>➤ Hand (open/close)</li> </ul>
Actuators	Include position and velocity control and feedback
Workspace	22cm along the surface of the table
Modular wrist connector	Compatible with commercial terminal devices and multi-articulated hands
Prototyping Cost	≤ \$5000

To more closely mimic upper limb prostheses the new arm was specified to be 1:1 scale with anatomical proportions to more easily allow the patient to visualize it as an arm. The mass of the arm including terminal device was planned to be less than 1.5 kg to keep it in the same range as myoelectric prostheses with powered elbows. The payload capacity of the arm was specified as 0.3 kg or greater to allow the option to use heavier objects in functional tasks. This payload capacity applied to when the arm was stationary or moving, did not include the weight of the terminal device, and would be sustainable indefinitely without overheating.

The available degrees of freedom were specified to match those of commercially available components (elbow flexion, wrist rotation, hand open/close) and provide some degrees of freedom that may be available in the future (shoulder rotation and wrist flexion). To accommodate conventional and novel EMG controllers the actuators required sensors to allow velocity and position control. If possible, the stronger MX series in the Dynamixel line of actuators (Robotis, Inc.) would be used to enable the reuse of interfacing electronics and software already developed for the AX-18 smart arm. A workspace of 22cm was specified to allow compatibility with the modified box and blocks task originally developed for the AX-12 smart arm [3]. A modular wrist connector was desired to allow custom designed and commercial terminal devices to be interchanged, allowing patients to test different prehensor options. The total prototyping cost was specified to be \$5000 or less in order to keep the arm inexpensive relative to its capabilities and allow it to be more easily duplicated in additional setups.

For the detailed design of the arm the specifications were further subdivided into mechanical, electrical, and software categories. Whenever feasible off-the-shelf or open source components were selected to reduce the design time and allow the prototype to be more easily duplicated.

**MECHANICAL DESIGN**

Initially, a basic kinematic model of the robotic arm was developed in SolidWorks 2014 to determine the combination of link length and joint range of motion required to meet the 22cm work space and anatomical proportion requirements. These lengths and joint ranges were then used with known component weights in iterative calculations to estimate the torques required to provide the desired payload capacity. From previous testing of the MX series actuators it was determined that they can support up to approximately half of the rated stall torques when moving an object at speeds typical of myoelectric prostheses. In order to accommodate the higher torques at the proximal elbow and shoulder joints the heavier and more powerful MX-106T and MX-64T actuators were selected respectively. An iglide PRT slewing ring bearing (Igus, Inc.) was used at the shoulder to isolate the MX-64T actuator from the weight of the arm and allow for smooth rotation. The lighter and smaller MX-28T actuators were chosen for the wrist and hand joints in order to decrease the distal bulk and mass of the arm. Through testing it was determined that a bearing would not be required at the wrist rotation joint.

The SolidWorks assembly model was updated with the selected components and the links, brackets, and adapters to interface between them. The associated aluminium brackets manufactured especially for the MX series were selected for

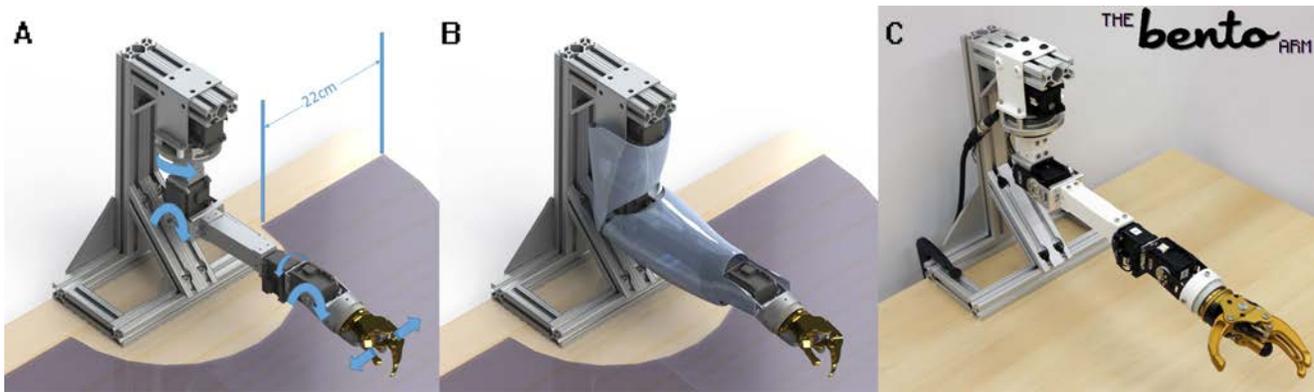


Figure 2 – a) Design specifications showing degrees of freedom and desktop workspace, b) the Bento Arm with removable 3D printed arm shells, and c) the 3D printed prototype of the Bento Arm (shown here without arm shells)

the joints. The remaining custom designed square tubing links, brackets, and adapters were specified to be 5052 aluminium when bending was required and 6061 aluminium otherwise. Notches were added to the square tubing links to allow for easy routing of actuator and sensor cables and an 8020 beam (80/20 Inc.) stand was designed to enable the arm to be rigidly mounted or clamped to a table or desk and to allow for mounting of electronic enclosures.

To improve the visual appearance of the arm and allow it to be more easily identified as a prosthesis, arm shells were designed as seen in Fig 2b. A 3D scanner was used to scan the arm of one of the authors and the resulting CAD file was converted to a solid body and shelled out in SolidWorks. The upper arm and forearm shells were each split into an upper and lower section and secured to the square tubing to allow for easy assembly and removal. If desired, the realism of the shells could be further improved by painting in skin tones or covering with fabric, gel, or silicone materials.

### ELECTRICAL DESIGN

The MX series actuators are designed to be daisy chained together on a bus and controlled by digital packets via a serial communication protocol. The protocol allows a control computer to send position and velocity commands to each actuator on the bus as well as poll for feedback on position, velocity, temperature, and load. A 12VDC, 150W power supply (Iccnexergy, Inc.) and a power switch (CW Industries) were specified to power the bus and allow it to be turned on and off independent of the control computer. An USB2dynamixel controller (Robotis, Inc.) was selected to interface between the actuators and a control computer via USB. A custom designed controller developed for the AX-12 arm is also available that allows the MX bus to interface with a RS-232 port. An electronics enclosure mounted in the stand houses all of the controllers and the power switch. Cables are

routed through the square tubing whenever possible and kept neat with 3D printed guides or zip ties.

To allow the wrist to be compatible with commercial terminal devices a quick disconnect wrist (Ottobock, Inc.) was acquired. A 7.5VDC, 30W power supply (Triad Magnetics, Inc.) was specified to provide power and a custom control interface was designed to allow a commercial prehensor to be connected to two analog output channels on a data acquisition card. The interface was tested with a SensorHand Speed (Ottobock, Inc.) as shown in Fig. 2c, but should be compatible with any myoelectric terminal device that can run off a 7.0-8.0 V battery pack and that expects an analog voltage from EMG electrodes between 0 and 5V.

### SOFTWARE DESIGN

One of the advantages with choosing the Dynamixel line of actuators is that Robotis has openly published their serial communication protocol. Thus, in addition to their own software interface which support Microsoft Windows and Linux in a variety of languages there is a strong community built around the actuators that have developed additional interfaces for platforms such as Mac OS X, Robotic Operating System (ROS), and Arduino Boards. In theory any of these could be used to control the arm. However, to allow the arm to be used in myoelectric training and research, we developed the following modular software interfaces that interface the arm with a Bagnoli-8 EMG acquisition system (Delsys, Inc), National Instruments data acquisition cards, conventional or machine learning based EMG controllers, and graphical user interfaces (GUI).

#### MATLAB's xPC Target

A software interface utilizing MATLAB's xPC Target real-time operating system has previously been developed for the MTT [3]. With xPC Target a backend computer runs

in real-time and a frontend computer allows signals to be displayed and parameters to be controlled. The main features of the backend controller include a signal processing module for EMG, a modified RS-232 driver to communicate with the Dynamixel bus, and a conventional EMG controller. The frontend GUI was built in Microsoft Visual Studio 2008 and graphically displays the EMG signal strength and allows the operator to change and save all the threshold, gain, mapping, and robotic arm settings.

### Robotic Operating System

Recently, we developed a control interface for the Bento Arm based on the Robot Operating System (ROS), an open source software architecture for robotics that includes a wealth of existing packages and tools. ROS is built on a robust messaging architecture with packages for data logging and playback, visualization, dynamic simulation, graphical interfaces, kinematics and motion planning, as well as support for numerous commercial sensors and actuators. While ROS is not a real-time operating system, it has the advantage that it may be possible to run both controller and GUI on a single laptop or embedded computer, or conversely in a distributed manner.

The graphical interface is built on rqt, a ROS GUI builder. It communicates with the actuators via the USB2dynamixel controller and includes features to pause, turn torque on or off and return the arm to a home position. A proportional EMG controller was also designed with adjustable parameters such as channel threshold, max, and gain. Feedback on EMG signal strength as well as the currently selected joint for each channel pair is also displayed in the GUI.

### **PROTOTYPE & SOCKET INTEGRATION**

To accelerate development custom 1" square tubing links, brackets, and adapters were 3D printed in white PLA on a Replicator 2 desktop 3D printer (Makerbot, Inc.) as seen in Fig 2c. An infill percentage of 50% was used in order to optimize the effective strength of the 3D printed parts while saving as much weight as possible [7]. After assembling the arm, functional testing with the ROS interface was able to proceed rapidly. During assembly several improvements in adapter and bracket design were identified and will be incorporated into the final aluminium parts. Initial testing suggests that arm should be able to achieve all of the design specifications listed in Table 1.

During development, it became apparent that the Bento Arm could potentially be mounted to a socket and used as an experimental prosthesis in a research setting. To test the feasibility of this idea an acrylic adapter was mounted to a clear thermoplastic socket (Fig 3). In addition to the research benefits of having a wearable arm, since the PLA version of

the arm only weighs 500g from the elbow down (excluding terminal device), it could potentially allow earlier functional training for patients who cannot yet tolerate the full weight of a commercial myoelectric prosthesis.

### **FUTURE WORK & CONCLUSIONS**

While the 3D printed PLA parts were sufficient for pre-prototyping and functional testing we are not confident they will endure long-term testing with the higher payloads and potential impacts with the environment. Consequently, we will proceed with machining the final components out of aluminium. Other improvements to the hardware setup will include developing an array of custom grippers for situations when commercial terminal devices are not available and designing an integrated embedded controller and battery power system to allow subjects to use the arm untethered.

Moving forward with the software systems, we will build a simulation in Gazebo (a ROS integrated robotics simulator) for the Bento Arm as well as integrate it with Rviz (a ROS visualization tool). Additionally, we will create an interface to allow motion planning via MoveIt! (a ROS motion planning tool).



Figure 3 – The Bento Arm integrated with a test socket

In conclusion, the Bento Arm is an affordable, fully sensorized, experimental robotic arm designed specifically for myoelectric training and research. The Bento Arm provides an early training option to potentially help improve clinical outcomes for myoelectric prosthesis fittings. Future research will focus on using the arm to investigate clinical training protocols, novel machine learning controllers, and sensory feedback systems.

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## OPINIONS OF PAEDIATRIC UPPER LIMB AMPUTEES

Sarah Day

*National Centre for Prosthetics and Orthotics, GB*

### ABSTRACT

#### Summary:

A 21 question survey was distributed through the UK charity Reach. Children aged 8-16 with upper limb absence or amputations were invited to answer questions about their views on disability, prosthetic usage and activities of daily living.

#### Introduction:

There is currently very little evidence regarding the actual usage rates of upper limb prosthetic devices. Previous studies have collected data from adult amputees registered at limb-fitting centres. As a result, little is known about paediatric populations and those who do not regularly limb wear. The aim of this study was to investigate the opinions and prosthetic wearing patterns of children with upper limb deficiencies.

#### Methodology:

A 21 question survey was devised to cover several different topics, including treatment methods, usefulness of prosthetic devices in everyday activities, opinions about disability and preferred terminology. Questions which investigate terminology for the residual limb and prosthesis were also included. Pre-testing of the questionnaire was conducted, and the study was granted approval by the University of Strathclyde Ethical Committee.

Members of the UK charity Reach were sent an invitation email with a link to the online survey. Children aged between 8 and 16 who live in the UK and have an upper limb absence were invited to participate. Parental consent was sought prior to the child's involvement.

#### Results:

As the survey will remain open until 16/3/14, the results are not yet available. To date we have received 44 responses.

#### Conclusion:

Preliminary analysis of the data shows that the paediatric upper limb population have strong views about disability and limb use. Results will be compared with similar studies

conducted on adult populations. The results of this study may be used, in combination with other studies, to influence the treatment and prescription criteria for children with limb absence.

## INVESTIGATING THE SIZE WEIGHT ILLUSION IN UPPER LIMB AMPUTEES

Sarah Day<sup>1</sup>, Gavin Buckingham<sup>2</sup>

<sup>1</sup>National Centre for Prosthetics and Orthotics, GB; <sup>2</sup>Heriot Watt University, GB

### ABSTRACT

#### Introduction:

The size weight illusion occurs when a person underestimates the weight of a larger object in relation to a smaller object of the same mass. This is well documented in normal populations but there has only been one study looking into its presence in populations with limb loss. This project aims to expand on the work by Wallace comparing the performance of a larger sample of upper limb amputees with able-bodied persons to investigate whether the size weight illusion exists in amputees and whether it is of the same magnitude as in the normal population.

There are several potential benefits to this study. Currently what causes the size weight illusion is unknown although there are several theories. Testing with active prosthetic users allow the researchers to eliminate certain variables such as sensory feedback as current prostheses used do not provide sensory feedback from the fingers to the user. The findings from this study also provide a greater understanding of what information individuals with a prosthetic limb use to judge the weight of objects, which may have consequences for the environmental ergonomics of this population.

#### Methodology:

During our investigation we compared a group of upper limb amputees using prosthetic devices to a group of people with normal upper limb function. Approval for the study was granted by the University of Strathclyde Ethical Committee.

Participants were tested one at a time in separate testing sessions. Participants were asked to lift objects of varying size and weight and ask them to rate them as a number, with a larger number indicating a larger weight. This number was then used to determine if the participants were experiencing the size weight illusion. The order in which the objects were offered to the participants was random and different for each participant.

The data was analysed using T-tests and ANOVA with the SPSS software package.

#### Results:

The data collection phase is due to be completed on 31/3/14. To date we have collected data on 5 amputee and 5 control subjects.

#### Conclusion:

Preliminary analysis of the data shows that the amputee group did experience the size weight illusion. This supports the previous findings by Wallace. Interestingly, the magnitude of the illusion and sensitivity to weight appear to be different according to the subject groups, although this will be confirmed upon completion of the data collection

## FLEXIBLE AND STATIC WRIST UNITS IN UPPER LIMB PROSTHESIS USERS: AN EXPLORATIVE STUDY

Marieke Deijs<sup>1</sup>, <sup>2</sup>Raoul Bongers<sup>1</sup>, Natascha Ringeling - van Leusen<sup>3</sup>, Corry Van der Sluis<sup>1</sup>

<sup>1</sup>University of Groningen, University Medical Center Groningen, Department of Rehabilitation Medicine

<sup>2</sup>University of Groningen, University Medical Center Groningen, Center for Human Movement Sciences

<sup>3</sup>Revant Rehabilitation Center Breda

### ABSTRACT

#### Introduction:

Wrist movement is an important requirement in upper limb prosthesis design for successfully employing the prosthesis in activities of daily living (ADL). Several prosthetic hands with wrists that have multiple passive motion capabilities, such as flexion/extension and radial/ulnar deviation, are on the market. However, only limited research has been performed on the surplus value of flexible wrist units compared to conventional wrist units that can only rotate.

#### Purpose:

To systematically assess the value of two flexible wrist units compared to static wrist units using a range of tests covering all factors of the domains "Body functions and Structures", "Activities" and "Participation" as described in the International Classification of Functioning and Health.

#### Methods:

Eight transradial amputees using a myo-electric prosthesis participated in a cross-over design. Participants tested two prosthetic hands with flexible wrists: the Flex-wrist (Otto Bock) and the Multi-flex wrist (Motion control). Each wrist was used for two weeks in its flexible condition and another two weeks in its static condition (no flexion/extension allowed). Five measurements were performed: at the start of the project and after each two week period. Functionality was assessed using SHAP and Box & Block test. Manual functioning was assessed using UEFS 2.0 (OPUS) questionnaire. Participants' satisfaction was assessed using D-QUEST and TAPES questionnaires, and open-ended questions on advantages and disadvantages of the prosthetic hand and wrist. To assess compensatory movements, shoulder movements were measured using an Inertial Magnetic Measurement System (MTw™, Xsens Technologies) during the execution of six ADL tasks (lifting object, handling cutlery, closing zip, turning door handle, lifting crate, stirring).

#### Results:

Functionality, manual functioning and satisfaction did not reveal significant differences over the four wrist conditions (n=6). From the open-ended questions, participants' satisfaction tended to be in favor of flexible wrists. Participants stated that flexible wrist units allowed a less restricted way of moving, made handling things easier, and required less awkward movements and postures which is less burdening for the shoulder. In the two analyzed ADL tasks (lifting object and handling cutlery), use of a flexible wrist indicated a smaller range of shoulder angles (n=4). Results may alter slightly, since not all data have been collected and analyzed yet.

#### Conclusions:

Functional tests and questionnaires on manual functioning and satisfaction do not indicate positive effects of flexible wrists. Users' satisfaction, derived from open-ended questions, is greater for flexible wrists. Flexible wrist units in prostheses suggest a decrease of compensatory movements.

## REDEFINING THE NORM IN TRANSRADIAL PROSTHETIC SOCKET COMFORT THROUGH USE OF SILICONE MATERIALS

Rob Dodson, Maryia A Cameron  
*Advanced Arm Dynamics*

### ABSTRACT

#### Introduction

Persons after upper limb amputation may have problems at several activities and at participation. Introduction

The advancements of silicone technology, socket construction techniques and implementation of silicone as a routine component of prosthetic design have evolved rapidly over the last two decades. Redefining the “norm” for interface design in an effort to increase comfort is proving to be critical for improved patient care. Multiple articles have been written showing the healing effects of silicone and the improved range of motion provided when using this material.

#### Methods:

In the last two years our clinical setting has fit 23 patients with a High Temperature Vulcanized (HTV) silicone based Transradial Anatomical Contoured (TRAC) socket interface for use with their current prosthetic device. Each of these patients will be surveyed post delivery of the new silicone socket design to determine their perceived level of comfort when utilizing this interface material as compared to the use of flexible thermoplastic interface material through the following categories:

#### 1. PHYSICAL TOLERANCE:

- a. Donning ease
- b. Pain experienced within the socket
- c. Hours worn per day
- d. Protection of fragile skin
- e. Accommodation of bony prominences

#### 2. RANGE OF MOTION:

- a. Restriction
- b. Flexibility

#### 3. DEPENDABILITY:

- a. Suspension
- b. Activity of Daily Living (ADL) performance consistency.

#### Results:

All 23 patients report an increase in the level of comfort through the use of silicone as the material interface within their prosthetic design based on clinical notation and observation.

#### Conclusion:

We believe the data obtained through this new, more extensive survey will support the hypothesis that using silicone as the primary socket interface material improves comfort and enhances the overall prosthetic wearing experience.

## ADAPTIVE SWITCHING IN PRACTICE: IMPROVING MYOELECTRIC PROSTHESIS PERFORMANCE THROUGH REINFORCEMENT LEARNING

Ann L. Edwards<sup>1,2</sup>, Michael R. Dawson<sup>3</sup>, Jacqueline S. Hebert<sup>2,3</sup>,  
Richard S. Sutton<sup>1</sup>, K. Ming Chan<sup>2</sup>, Patrick M. Pilarski<sup>1,2</sup>

<sup>1</sup>Department of Computing Science, University of Alberta, Edmonton, AB, Canada;

<sup>2</sup>Division of Physical Medicine and Rehabilitation, University of Alberta, Edmonton, AB, Canada; <sup>3</sup>Glenrose Rehabilitation Hospital, Alberta Health Services, Edmonton, AB, Canada

### INTRODUCTION

Myoelectrically controlled prostheses are a class of assistive device that use electrical signals generated by muscle activation. These electromyographic (EMG) signals are used to control one or more electromechanical actuators that move prosthetic joints. Myoelectric control signals are typically measured with electrodes on the surface of the skin, with one pair of electrodes over each muscle site. In this manner, each muscle site directly controls one motion of the prosthesis, and various methods of switching can be used as needed to control additional motions of the prosthesis [1] [2] [3].

Some state-of-the-art myoelectric hands currently used by amputees have over a dozen possible grip patterns that can be manually selected by the user. Despite increasing possible control options, a robotic arm with so many available motions presents a problem, since there exist more degrees of freedom than there are available control signals from the human user [1] [4] [5]. One solution to circumvent this problem is for the user to switch between all available joints or grip patterns in a predesigned, optimized order. As another option, the amputee and their prosthetist may selectively reduce the number of available control options (i.e., the amputee will have access to and switch between only a small subset of the device's available functions during regular use). Both of these options require trade-offs between switching effort and device functionality.

While switching between functions continues to be used in clinical settings to extend prosthesis functionality, it can be laborious. Switched or gated control is considered to be slow and non-intuitive, requiring both time and sustained cognitive effort on the part of the user [1] [4]. Non-intuitive control in fact represents one of the main reasons amputees stop using their myoelectric prostheses [1] [2] [3]. These limitations have been a driving force for more advanced control paradigms such as pattern recognition [1] [3]. However, as functionality increases and control

becomes more challenging, one acknowledged solution is for prostheses to begin to assume more autonomy in interpreting and executing a user's intended movements.

Previous work by our group has therefore examined ways to streamline and optimize prosthetic control interfaces such as the switching system indicated above, potentially increasing the number of available and accessible modes or functions through the use of machine intelligence [5] [6] [7] [8]. In particular, our prior work showed how predictions about sensorimotor signals, such as signals pertaining to arm movements, could be learned and maintained using a reinforcement learning technique known as *General Value Functions* (GVFs) [9]. GVFs are temporally extended predictions about a signal of interest that have been applied to building up real-time anticipatory knowledge in relation to human-machine interactions [5] [6] [7]. We have shown in experiments using reinforcement learning offline (prior to use in prosthesis control) that GVFs may offer a way to help streamline control interfaces with robotic arms. In particular, we demonstrated the use of GVFs and reinforcement learning to predict which joint of a robotic arm an amputee user intends to actuate next, and proposed the idea of an adaptive or situation-specific switching list [6]. A natural extension of this work would be to apply predictions to actual human interaction with artificial limbs with the intent of improving control. Applying predictions to human machine interaction is consistent with the knowledge that, similar to GVFs, the human brain makes motor predictions of its own, using both knowledge of context and immediate sensory input [10].

In the current paper we extend our prior studies to present preliminary evidence that our method of adaptive switching does in fact provide benefit during the operation of a robotic arm by a myoelectric user. This work is the first simple demonstration of the use of prediction learning in real time to improve the control of a prosthetic device during its use by an amputee subject. Predictions are learned and used in real time by the control system to reduce the burden of

control on the user, making it easier and faster to switch to the user's intended next joint or function.

## METHODS

In order to implement and assess adaptive switching, three subjects—two transhumeral amputees and one able-bodied subject—were recruited to perform a simple, semi-repetitive task using an experimental robotic arm. Because of the similarity between the data sets, in the interest of space only one representative data set is presented in this paper. The subject was a body-powered prosthetic user and had no experience using myoelectric control or using our specific robotic arm. We attached surface electrodes to the skin over his wrist extensor muscle on the intact arm, which provided control signals for switching between robot joints. Separate sets of electrodes were also attached to the biceps and triceps muscle of his residual limb. Those electrodes became the source of control signals for flexing and extending selected joints of the robot arm. An 8-channel Bagnoli EMG system (Delsys, Inc.) was used in the acquisition of EMG control signals from the experimental subject, at a frequency of 1 kHz. The subject gave informed consent to participate and the trial was approved by the human research ethics board at the University of Alberta.

We used a custom-built robot arm known as the Myoelectric Training Tool (MTT) in our experiments [11]. The MTT includes an AX-18 smart robotic arm (Crustcrawler, Inc.) that has five degrees of freedom and can be controlled via EMG signals by both amputees and able-bodied subjects. In addition, it can be used as a training tool for amputees preparing to use a myoelectrically controlled prosthetic arm, as it was designed to be functionally similar to commercial prostheses. Figure 1 shows the amputee subject using the MTT to perform a simple task.



Figure 1: Amputee participant performing simple tasks with the robot arm using myoelectric control signals

The subject was given time to become familiar with the MTT. After familiarization, the subject was presented with a specific task that involved a subset of the available joints (specifically hand open/close, wrist flexion/extension, and shoulder rotation). The task was chosen to be functionally comparable to other tasks of daily living—for instance, picking up a dish and placing it on a shelf. The instruction given to the subject in both the non-adaptive and adaptive trials was to manipulate the MTT to grasp an imaginary object on one side of the shoulder space, rotate the shoulder to the opposite side, wave with the wrist joint, and rotate the shoulder back to the other side. Each trial involved repeating this task for a total of 3 minutes.

Two types of trials were performed in order to test the predictive capabilities of our design compared with conventional switching methods. In the non-adaptive trial, the subject switched their myoelectric control between four joints in a fixed switching order: hand, wrist, elbow, and shoulder. In contrast, in the adaptive trial, the joints were continuously reordered in the switching list based on their likelihood of being used next. This was done in an ongoing fashion throughout the course of the task through the use of GVs. Three 3-minute trials were done each for non-adaptive and adaptive switching.

As described in Pilarski et al. (2012), GVs represented predictions about the subject's situation-specific use of each joint in the switching list [6]. These predictions were learned during the subject's use of the robot arm and continuously

ranked based on their relative magnitudes. In the current work, with adaptive switching turned on, the system learned to predict the intended joint for the given task in advance of the switch signal from the user. When a switch signal was received by the system, the highest-ranked joint in the adaptive switching list became the active joint, with the remaining joints filling in the new switching list in decreasing order of prediction strength. All GVF learning was implemented as per Pilarski et al. [6].

In order to build up real-time predictions about the intended active joint, we combined ongoing sensorimotor data from the robotic arm with EMG data from the human user. Each of the AX-18 motors that make up the joints of the MTT relayed a number of useful sensorimotor outputs, including angular position, angular velocity, load (current), temperature, and voltage. We used a select number of these motor observations as features, or information about the current state, in the learning system. The included observations were angular position and angular velocity of each joint. Features based on the current state of the arm enable the system to build up expectations about future switching decisions made by the user. The machine learning system was re-initialized at the beginning of each trial—GVFs started each trial with no stored knowledge (predictions) about the user or the task in question.

## RESULTS AND DISCUSSION

Figure 2 compares the number of switches required per event for non-adaptive switching (top) with the number of switches required during adaptive switching (bottom) for the subject. Each switching event was considered to begin when the user triggered a joint switch, and end when the user initiated movement of any of the MTT joints. Therefore, all switches made while shifting control to a new joint are counted as a single switching event. As shown in Figure 2, there was a significant difference between non-adaptive switching and adaptive switching. With adaptive switching enabled, after an initial period of learning by the system (i.e. the first several switching events), typically only one switch was required by the user to select the most appropriate joint.

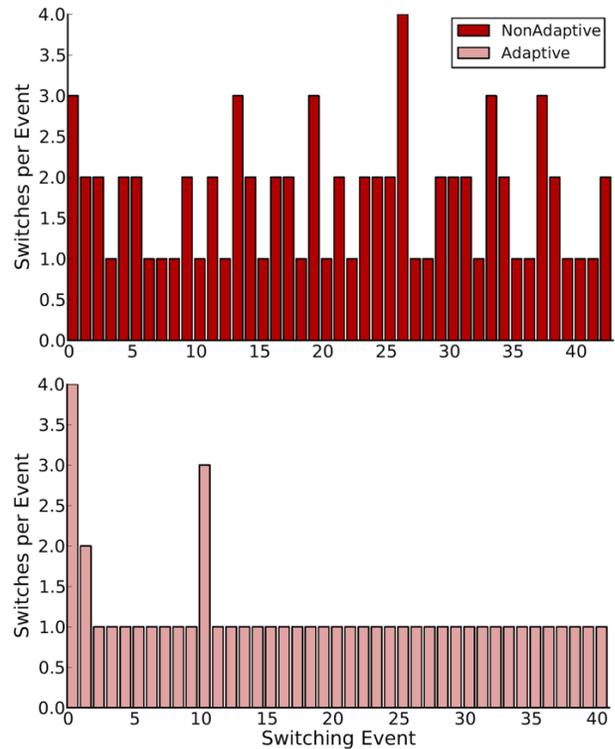


Figure 2: Number of voluntary switches initiated by the amputee subject per switching event over the course of a single 3 min trial. Shown for both non-adaptive (top) and adaptive control (bottom) approaches

The decrease in the number of switches is also reflected in Figures 3 and 4. Figure 3 shows the average amount of time (measured in seconds) dedicated to switching, calculated over the three non-adaptive trials and the three adaptive trials. Adaptive switching showed a large decrease in time spent switching compared with non-adaptive. Thus, for each 3-minute trial, the subject saved an average of about 20 seconds when adaptive switching was enabled. Figure 4 is the total number of switches averaged over three trials. The decrease in the amount of time spent switching is also illustrated in the decrease in the total number of switches per trial. Furthermore, the median time per switching event was consistently more than 1 second for all non-adaptive trials, and consistently under 1 second for adaptive trials. Not only was the median time per event lower, but in some trials the total number of switching events completed in a task was also greater when adaptive switching was enabled.

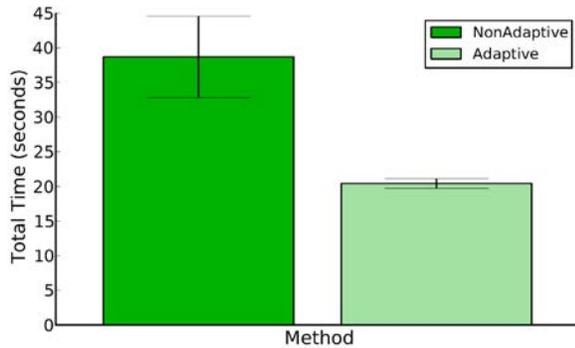


Figure 3: Average time the amputee subject spent switching per trial when using non-adaptive and adaptive switching (left and right, respectively, average over 3 trials)

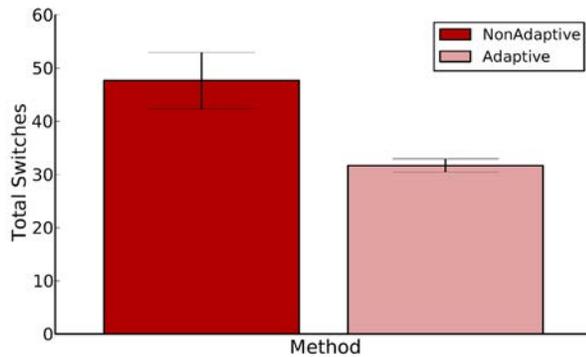


Figure 4: Average number of switches made by the amputee subject per trial when using non-adaptive and adaptive switching (left and right, respectively, average over 3 trials).

These results suggest there are efficiencies with adaptive switching, and agree with our expectations regarding the simple task presented to the subject: there were clear regions of the task space that corresponded to the use of specific joints. For this task, it would have been possible to hand-code several different switching lists in response to the different positions of the shoulder actuator. The simplicity allowed us to easily verify the correctness of the adaptive switching options proposed by the learning system. However, a key observation from the present work is that situation-specific switching orders do not need to be hand-coded; our system learned situational delineations as the robotic arm was being used, and without prior information about the user or their task. Furthermore, we observed that as the task changed or became more complex (and thus increasingly hard to engineer situation-specific switching lists) the learning system scaled up naturally and easily without the need for manual tuning.

## CONCLUSION

The primary contribution of this paper is a concrete demonstration of adaptive switching in an applied setting. This study is the first time that real-time prediction learning has been used to improve the control interface of a robotic device during un-interrupted use by an amputee subject. Our experiments with an amputee subject showed that for simple tasks, enabling adaptive switching on a robotic arm significantly decreased the time spent switching. This is consistent with and extends previous studies using pre-recorded (non-real-time) data that indicated the potential merit of adaptive switching.

We believe that adaptive switching would help to decrease the cognitive load required by amputees during more complex tasks and real-world functional situations involving wearable prostheses. In particular, in our future work we will study the use of adaptive switching in tasks with multiple solution pathways—i.e., situations where many possible (and user specific) movement sequences could be used to achieve the task's objective.

## ACKNOWLEDGEMENTS

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## DESIGN AND TESTING OF A CABLE DRIVEN TACTOR SYSTEM FOR SENSORY FEEDBACK IN UPPER LIMB AMPUTEES

Katherine R. Evans<sup>1</sup>, Michael Rory Dawson<sup>2</sup>, Kent Herrick<sup>1</sup>

Jacqueline S. Hebert<sup>1</sup>, Jason P. Carey<sup>1</sup>

<sup>1</sup>University of Alberta; <sup>2</sup>Glenrose Rehabilitation Hospital

### ABSTRACT:

Myoelectric prostheses have recently undergone extensive developments in their complexity and movement patterns, yet controlling these devices can be quite difficult as they lack sensory feedback. With targeted muscle reinnervation surgery, motor nerves that used to supply the hand are relocated to muscle sites in the residual limb of amputee patients, and recorded electromyography signals are used to intuitively control a prosthetic device. Sensory nerves also reinnervate the skin so that when touched on part of the reinnervated skin, the patient feels as though they are being touched on their missing limb. Previous work has demonstrated that this restored hand map can be harnessed to provide feedback to the patient such that when they grip something with a robotic hand, a tactor pushes into their reinnervated skin and they feel as if they are gripping the object directly. The aim of this work is to take the next step in integrating a practical sensory feedback device into an actual prosthesis for clinical trials. There are various clinical challenges to overcome when integrating a feedback system into a socket, such as maintaining a proper suction seal and avoiding excessive cost, weight, and bulk. Preliminary design work has been completed to develop a reduced profile horizontal cable driven tactor for socket integration. This investigation aimed to improve the functionality of the tactor, integrate multiple tactors into a test apparatus, and conduct preliminary trials with able-bodied and amputee participants. Subject testing incorporated a designed experiment using customized rigid tactor cuffs to investigate different parameters including tactor head shape and diameter, the effect of the presence of a liner between the skin and tactor head, and the optimal application of incremental force steps. Response variables included correct identification of the number, location, and applied force of the tactor(s), as well as overall comfort of the system. Testing provided valuable insight into optimal system parameters and layout in order to address the issue of space limitations when using multiple tactor heads. The results of this testing will feed into future prosthetic socket designs to provide sensory feedback in clinical research trials.

## TRANSLATING SOFT ROBOTICS TECHNOLOGIES TO PROSTHETICS: EARLY RESULTS WITH A SOFTHAND PROSTHESIS

Sasha B. Godfrey<sup>1</sup>, Arash Ajoudani<sup>1</sup>, Matteo Bianchi<sup>1</sup>, Manuel Catalano<sup>1</sup>, Giorgio Grioli<sup>1</sup>,  
Karen L. Andrews<sup>2</sup>, Alycia S. Gailey<sup>3</sup>, Marco Santello<sup>3</sup>, Kristin D. Zhao<sup>2</sup>, Antonio Bicchi<sup>1</sup>

<sup>1</sup> *Istituto Italiano di Tecnologia*; <sup>2</sup> *The Mayo Clinic*; <sup>3</sup> *Arizona State University*

### ABSTRACT:

The Pisa/IIT SoftHand is a novel under-actuated, synergy-driven robotic hand. It is a single degree of actuation (DOA), anthropomorphic hand with a single tendon that drives the 19 joints of the fingers and thumb in adduction and flexion along the first grasp synergy, as defined by the principal component of the covariance of statistical data collected from hand movements in common grasping tasks. Because of its grasping versatility, anthropomorphic appearance and behavior, control simplicity, and robustness, the design principles of the SoftHand offer a potential for the development of radically new prosthetic devices. Here, we report on the early developments of a new SoftHand-based prosthesis resulting from a joint research project between IIT and INAIL, and on advances in the understanding and optimization of its interface with people with limb loss, done in collaboration with ASU and the Mayo Clinic.

Current myoelectric prostheses generally fall into two categories: single grasp, three-digit hands and multi-grasp anthropomorphic hands. While the former are often easier to control, lighter in weight, and more affordable than their more advanced counterparts, their single, rigid grasp limits function. The SoftHand presents an intermediate alternative to current myoelectric hands. The under-actuated, anthropomorphic design allows the SoftHand to adopt various grasps as it molds around objects, while the single DOA, and thus single motor, design allows for simple control without the need to consciously select a specific grasp. We will present recent progress towards a SoftHand-based prosthesis, including sizing, portability, compatibility, and end-user testing. We describe a new control scheme, using only two EMG electrodes, that allows the user to drive both the position and stiffness of the SoftHand via the co-contractions of an antagonist pair of muscles, and can be implemented on an embedded microcontroller. We will also discuss sizing issues, power requirements and motor specifications, and interfaces with readily-available commercial prosthetic solutions, such as Otto Bock electrodes. We will report on preliminary biomechanical and clinical tests that have been performed to evaluate a SoftHand-based prosthesis prototype. These results suggest the SoftHand is a clinically-relevant

alternative to the prosthetic options currently available and will serve to guide future improvements in the development process.

## RESTORING SENSATION IN AMPUTEES: CLINICAL CONSIDERATIONS

Emily L Graczyk, Matthew A. Schiefer, Daniel Tan, Michael Keith,  
J. Robert Anderson, Joyce Tyler, Dustin J. Tyler

*Louis Stokes Cleveland Department of Veterans Affairs Medical Center; Case Western Reserve University*

### ABSTRACT

#### Introduction:

Regaining sensory feedback is important to upper limb amputees. Previous approaches to restore sensation in the missing limb include sensory substitution, targeted sensory reinnervation, and intrafascicular nerve stimulation. We present the clinical experience using extraneural peripheral nerve cuff electrodes with percutaneous leads for more than 18 months in subjects with upper limb loss.

#### Methods:

Two unilateral upper limb amputees participated in this study to date. Subject S102 had a right wrist disarticulation resulting from trauma 18 months prior to implantation surgery. Subject S104 had a right transradial amputation resulting from trauma 8.5 years prior to implantation surgery.

Eight-contact Flat Interface Nerve Electrodes (FINEs) were implanted on the median and ulnar nerves of both subjects, and on the radial nerve of S104. A four-contact CWRU spiral cuff electrode was implanted on the radial nerve of S102. The cuffs were implanted in the forearm in S102 and in the upper arm in S104. The leads were routed subcutaneously to a connector site in the axilla side of the upper limb. Spring-and-pin connectors attached the cuff leads to percutaneous open helix leads that were routed to exit sites over the medial deltoid. S102 had 20 percutaneous exit sites and S104 had 24 exit sites. Surgeries were conducted at the Louis Stokes Cleveland VA Medical Center under an Investigational Device Exemption.

After surgery, the arm was immobilized for one week. Subject follow-up occurred at least twice per month, during which the nerves were electrically stimulated through the implanted electrodes. Between visits, the subjects kept the percutaneous sites covered with a waterproof bandage. Subjects were evaluated throughout the study for infection, unexplained paraesthesia, pain at the implantation and exit sites, skin irritation at the exit site and under the socket, and phantom pain.

#### Results:

The implantation surgeries lasted 5.25 and 4.1 hours. The electrodes and percutaneous leads have been in place for 22 months (S102) and 14 months (S104) with no serious complications or unwanted side effects. All evaluations of pain, infection, and irritation were negative. Although the subjects reported mild to severe phantom pain at initial screening, both reported cessation of phantom pain since implantation. On three occasions a blocked pore near the percutaneous leads required expression. Both subjects report that neither the electrodes nor percutaneous leads interfere with daily activities.

#### Conclusions:

Nerve cuff electrodes implanted above and below the elbow, leads crossing the elbow, and percutaneous leads are well-tolerated long-term.

## FROM FEEDBACK TO FEEDFORWARD: THE ROLE OF SENSORY FEEDBACK IN THE CONTROL OF MYOELECTRIC PROSTHESES

Cornelia Hartmann, Strahinja Dosen, Andrei Ninu, Dario Farina

*Department of Neurorehabilitation, University Medical Center Göttingen, Georg-August University, Germany*

### INTRODUCTION

Closing the loop in upper limb prosthetics is a current challenge. There is a variety of prosthetic devices on the market, but none of them include a system to provide somatosensory feedback to the user. Moreover, although there is a general agreement that closing the loop in active prostheses is beneficial, the exact role and benefits of feedback are largely unclear [1].

Routine grasping is a fast activity. Closing the prosthesis to grasp an object takes approximately 2 to 3 seconds. Moreover, there is a lag in the prosthesis response due to delays in signal processing (artificial controller) and prosthesis inertia. Therefore, the feedback of the prosthesis aperture and grasping force might not be as useful for the online corrections during the actual execution of the grasping action, which is performed mainly in a feed-forward manner [2]. However, the feedback can still be important for learning this feed-forward task since it provides information about the final outcome of the movement. In this context, the feedback information promotes a trial-to-trial learning of the system dynamics, allowing more consistent generation of the grasping forces and providing an input for the subject to build an internal model of the system he/she operates. Theoretically, once this model is optimized, the user would be able to control the system by relying mostly on the feed-forward commands, i.e., the feedback would not be needed anymore. This view is also in agreement with physiological motor control [3].

In the current work, we preliminarily evaluated this concept by using a prosthesis training scenario which simulated a daily living activity, namely the grasping and manipulation of a raw egg. For this task, reliably generating a certain grasp force is the crucial factor for success.

The aim was to investigate if the subjects were able to learn the model of the prosthesis and if the amount of sensory information provided during the training scenario had an influence on either the training duration or the quality of the post-training prosthesis use.

### METHODS

Six able-bodied volunteers participated in these preliminary experiments (4 male, 2 female, age  $31 \pm 6$ ). All participants signed an informed consent approved by the local ethics committee. Using two bipolar EMG channels recorded at the flexor and extensor muscles of their forearms, subjects controlled the grasping and opening of a prosthetic hand (Michelangelo, Ottobock, AT) which was mounted on a stand. An egg-shaped dummy was mounted so that the fingers of the prosthesis closed on it when a grasp was initiated by the subject (Fig. 1). The subjects could observe the prosthesis and its interaction with the egg dummy throughout the experiment. A visual feedback in the form of a bar proportional to the grip force exerted with the prosthesis was provided to half of the participants (AV group), while the other half (NV group) did not receive this additional sensory feedback. The visual feedback was displayed on a standard 17" computer screen placed behind the prosthesis.

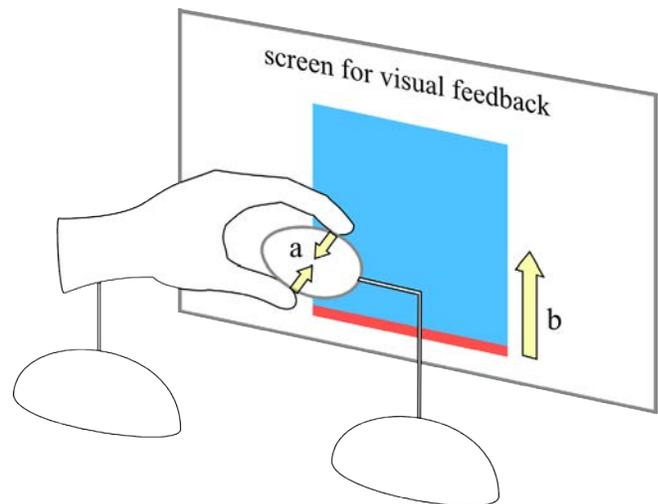


Figure 1: Setup of prosthetic hand and egg dummy. The bar for the visual force feedback changed height (b) with changing contact force (a) on the egg dummy (AV group only)

The experiment was divided into two phases: a training phase and a post-training phase. During training, subjects were asked to repeatedly grasp the egg dummy

by generating appropriate forearm contractions to control the prosthesis. After each grasp they were informed by the investigator whether the attempt would have been successful or unsuccessful if a real egg had been used. Unsuccessful in this context means that the egg was either not gripped strong enough, and would have slipped during manipulation, or would have been crushed due to excessive force. The selected force range for successful trials was 10-25 N. The training consisted in a series of consecutive trials and was interrupted when the subject achieved seven successful grasps consecutively. The two groups received the same verbal information at the end of each trial and the AV group also obtained the direct force feedback visually. The force feedback was removed immediately after the training ended.

In the post-training phase, both groups were asked to perform additional 30 grasping trials without any feedback. The post-training performance was the percentage of successful grasps after training.

## RESULTS

Figure 2 shows the exerted forces from the training and post-training phases of one NV and one AV group subject. For this NV subject (Fig. 2 a), the training phase was comparatively short and already the first attempts were very close to the allowed force windows. Force convergence was achieved after few minor over- and undershoots. In the post-training phase, a drift of the exerted force towards higher values could be observed after several attempts. For the AV subject (Fig. 2 b), the training phase is characterized by few larger force overshoots in the initial phase, and convergence of force within the allowed force range at the end. In the post-training phase, most attempts were successful, the amplitude of over- and undershoots was rather small, with a maximum deviation of 8 N from the allowed range.

The performance results for all subjects are reported in Table 1. For the majority of subjects the performance increased in the post-training phase. However, each group had a subject with decreased performance in the post-training phase when feedback on the trial-by-trial performance was no longer conveyed.

Table 1: Individual subject performance during training (last seven attempts excluded) and post-training phase.

	NV Group			AV Group		
Training Performance (%)	54	71	44	63	47	50
Post-Training Performance (%)	80	57	63	77	80	30

The average training duration was comparable for both groups (NV:  $29 \pm 6$  trials; AV:  $32 \pm 19$  trials). At the

transition from training phase to post-training phase, a difference in the subjective self-confidence of the subjects could be observed depending on the group association. When the additional visual feedback was switched off for the AV group and subjects had to continue without the proportional force display, they felt more insecure and doubtful about their expectable performance than the NV group subjects (subjective feedback). However, the post-training performance of the two groups was similar (NV:  $67 \% \pm 12 \%$ , AV:  $62 \% \pm 28 \%$ ).

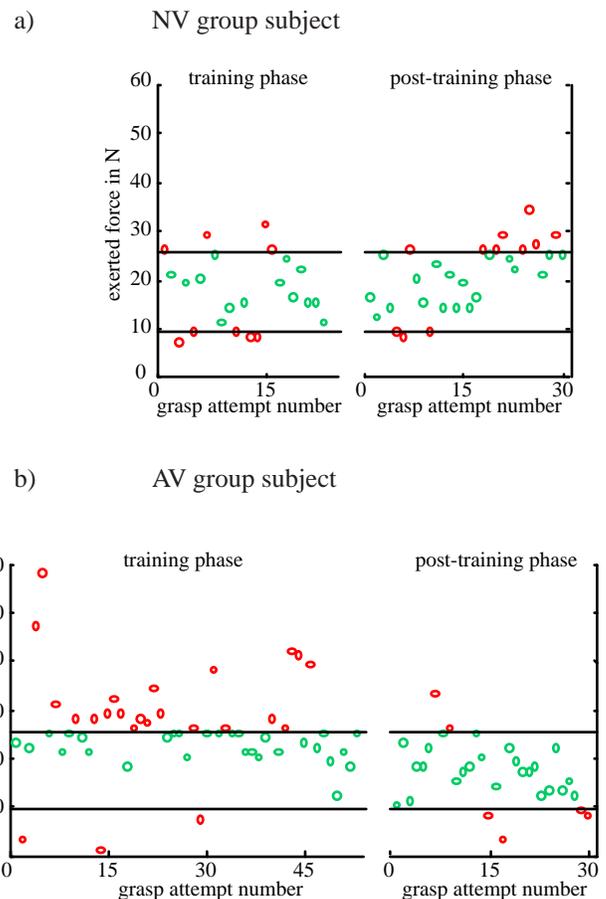


Figure 2: Example of the training phase and subsequent post-training performance. a) NV group subject. b) AV group subject. Area between black bars shows the success area. The green and red dots indicate successful and unsuccessful grasps, respectively.

A difference between the two groups could be observed during the end of the training phase. Subjects of the AV group were more consistent in producing the force for the seven last trials. The average standard deviation over the seven last training trials was 4.5 N for NV subjects and 2.7 N for the AV group. However, in the post-training, the average inter-attempt force variability was comparable for both groups.

Considering the entire subject sample without grouping, there was a tendency for a positive association between training duration and performance after training, so that, e.g., the subject with longest training had post-training performance of 80% while the subject with shortest training showed the worst post-training performance (30%).

## DISCUSSION

The results of this preliminary evaluation demonstrate that the subjects were able to learn the model of the system by relying on the provided feedback information. However, the success rates were variable (57 to 80%) and one subject was not successful (30%). Furthermore, it also seems that the amount (fidelity) of feedback information did not have a major influence on the learning process. The group that received continuous and precise visual information about the actual grasping force (bar plot) performed essentially the same as the group provided with only the information on the success of the trial.

The presence of an additional analogue visual feedback representing the invisible physical variable (force) did also not appear to influence the training duration, i.e., the training phase for the AV group was neither consistently shorter nor longer than for the NV group. The same applies for the subjects' performance in the post-training phase. In general, subjects in both groups were able to achieve good post-training performance, hence it can be concluded, that a simple "binary" feedback (success / no success) after each attempt was sufficient for the subjects to calibrate their internal control/system model. This feedback mimics a natural interaction with the environment and is always present in feed-forward prosthesis control.

However, a better force repeatability could be observed during the end of the training phase for the AV group, which disappeared again in the post-training phase. This demonstrated that an "analogue" force feedback was useful in trial-to-trial corrections and more refined calibration of the model of the system.

## CONCLUSION

When interpreting the results of this study, and when addressing the use and benefits of feedback in prosthesis

control in general, the feed-forward aspect of the investigated system may have to be taken into account as well. For example, in the preliminary experiments described in this paper, an unknown influence of the inert mechanical properties of the prosthesis and/or the robustness of EMG control must be assumed. This is why in a follow-up study, the role of feedback should also be investigated against the background of an ideal (virtual) prosthesis, as well as other control interfaces.

Furthermore, as the training scenario was intentionally designed to be realistic, i.e., subjects were able to see and hear the prosthesis, all subjects also received audio feedback on the prosthesis activation. Thus, representing the grip force via the visual channel may have not added additional information for the user in building his internal control model. The information obtainable via the visual channel was anyway already very rich (e.g., velocity of fingers closing on the egg, point of contact, deformation of prosthetic fingers after contact). Subjects may consciously or subconsciously derive from these information conclusions about the exerted force, so that an explicit display of the force via the visual channel may not contribute to the training. Future work shall now investigate if providing additional feedback via a new modality which is not already represented in the default training scenario, may add advantageous information, thus resulting in shorter training durations or improved post-training performance and consistency. Such feedback could, for example, be achieved by including the tactile modality.

In conclusion, the results of these preliminary tests suggest that the feed-forward mode of control common to all commercial prosthetics systems is very rich of feedback information and this information can be used for very fine force exertion even without any additional direct feedback on force. These results also prompt the need for further research on larger subject samples and more experimental conditions to identify the conditions, if any, in which direct feedback on physical variables, such as force and speed, is needed.

## ACKNOWLEDGEMENTS

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## A FLEXIBLE TEST BENCH FOR CLOSED-LOOP PROSTHESIS CONTROL

Cornelia Hartmann, Strahinja Dosen, Marko Markovic, Dario Farina  
*Department of Neurorehabilitation Engineering , University Medical Center Göttingen*

### ABSTRACT

In the last few decades and in particular in recent years, important efforts have been devoted to the development of active prosthetic systems, which has resulted in many innovations. These developments open the challenge of evaluating and comparing a multitude of different approaches and technologies with respect to performance and robustness. Here, we present a standardized test bench framework implemented in Simulink/Real-Time Windows Target that allows simple configuration and evaluation of feed-forward and feedback human prosthesis control systems under real-time conditions. The framework comprises six blocks (placeholders) connected into a generic closed-loop system. Each block can be configured by selecting the desired component from a list of implementations available in the library. The generic blocks are: 1) the input interface, as the source of control signals, 2) the control module, implementing the control logic, 3) an interface to the actual system to be controlled, and three optional components for setting up an 4) experimental task, 5) information encoding and 6) a feedback interface. Due to a standardized input/output interface design, all implemented components are interoperable and the whole framework is easily extensible.

The applicability of the framework was demonstrated in two pilot experiments. The first experiment evaluated and compared different feedback encodings for prosthetic applications. For this purpose, electrotactile feedback, encoded either spatially or via intensity modulation, was applied during a pendulum stabilization task. The second experiment investigated the influence of myoelectric input signals (commercial state of the art) as compared to joystick signals (ideal noiseless interface) on grip force control of a physical prosthetic hand. The first experiment demonstrated that spatial coding was more intuitive for the subjects initially but intensity coding allowed for higher precision after sufficient training. The second experiment revealed that the accuracy of force control was poor and comparable for the two input interfaces (myoelectric vs. joystick control) since the limiting factor to accuracy were the intrinsic mechanical characteristics of the prosthetic limb rather than the control source. Importantly, this simple experiment demonstrated that

each link in the control chain imposes its own limitations, so that elaborate advancements only in one component may not always yield the desired improvements in the performance of the overall system. Conveniently, such limitations can be identified using the proposed test bench since it allows testing and comparing specific component implementations in an idealized system environment.

This framework will be shortly made publicly available to the broader scientific community.

## KINEMATIC DATA FOR MODIFIED BOX AND BLOCKS TEST: NORMATIVE AND PROSTHETIC COMPARISONS

Jacqueline Hebert<sup>1</sup>, Albert H. Vette<sup>1</sup>, Justin Lewicke<sup>2</sup>

<sup>1</sup>University of Alberta; <sup>2</sup>Glenrose Rehabilitation Hospital, Alberta Health Services

### ABSTRACT

For upper limb prosthetic users, performance measures do not always take into account the compensatory motions that are required to accomplish a task. In particular, prosthetic users employ excessive trunk and shoulder motion to compensate for a lack of wrist motion. This has been observed with common tests of motor performance such as the Box and Blocks (BB). The limitation of using a gross motor performance test is that the measurement may not take into account body compensations. Similarly, improved prosthetic components that may reduce body compensations by providing more degrees of freedom may not impact overall task completion speed, and standard measures would therefore not detect an improvement. For these reasons, we previously used a modification of the BB combined with motion capture in a prosthetic user to compare performance with different prostheses. The purpose of the current report is to further investigate the utility of this new metric by presenting a normative data set, and to compare data from other prosthetic users to normative kinematics. The overall goal is to provide a standard reliable method of assessing upper limb and trunk motion as well as prosthetic performance.

We modified the BB test by placing 16 blocks in a 4 x 4 grid with pre-determined sequence to specifically target reach and grasp function. We used an upper limb marker set for motion capture, and collected kinematic data in 16 able-bodied participants. Four specific block cycles of the modified BB test were chosen for analysis, to represent 4 quadrants of the workspace: near-far and medial-lateral. These block cycles were analyzed to establish kinematic data for upper limb and trunk motion. Trajectories were compared to those of transradial and transhumeral prosthetic users.

In able-bodied participants, no differences were found between right and left hand performance other than task completion time. Small but significant differences were found for standing and seated performance, with slightly greater ranges in standing for axial trunk rotation, medial-lateral sternum displacement, and anterior-posterior hand displacement. The kinematic trajectories were very

consistent among able-bodied participants. The altered kinematic trajectories in the prosthetic participants were clearly discernable in comparison to the normative data.

## COMPARATIVE SENSORY OUTCOMES FROM THREE TRANSHUMERAL TARGETED REINNERVATION CASES

Jacqueline S. Hebert<sup>1,2</sup>, K. Ming Chan<sup>1</sup>, Michael R. Dawson<sup>2</sup>

*1Division of Physical Medicine and Rehabilitation, University of Alberta,*

*2Glenrose Rehabilitation Hospital, Alberta Health Services, Edmonton AB, Canada*

### INTRODUCTION

Targeted reinnervation has demonstrated success in improving motor control signals for both shoulder disarticulation and transhumeral levels of amputation [1-3]. Targeted reinnervation has also shown promising sensory outcomes in that redirected sensory afferents from the median, radial and ulnar nerves reinnervate the skin in the residual limb, creating an expression of the missing hand map [4]. When these patients are touched on the reinnervated skin, they feel as if they are being touched on the missing limb [4-6]. This provides a potential portal to restore sensory feedback from the prosthetic device that could be anatomically matched to the missing limb.

Restored cutaneous sensory percepts have been well described in subjects with targeted reinnervation that have developed hand maps on chest skin after reinnervation. A wide range of sensory modalities have been shown to be restored including near normal touch, temperature, pain sensation to electrical stimulation [4] and vibrotactile thresholds [7]. These findings were demonstrated in a subject with chest reinnervation due to local denervation of skin overlying the muscle (thinning of subcutaneous tissue) as well as in a subject with end-to-side coaptation of the supraclavicular nerve to the ulnar nerve [1, 4].

The sensory outcomes for the transhumeral targeted reinnervation procedure have not been as thoroughly reported. Dumanian et al [3] reported on six subjects with transhumeral targeted muscle reinnervation procedures. The nerve transfers yielded four electromyographic signals separate and distinct from adjacent muscles, controlled by four different nerves. However they did not report any sensory outcomes. Sensinger et al [8] reported one transhumeral reinnervation subject in whom “the distal end skin was purposefully denervated”. This subject was reported to have transfer sensation of the palm and palmar aspect of digits 1-3 when touched on the medial aspect of the upper arm, and was able to discriminate between gradations of force applied to these areas of skin reinnervation. Marasco et al [9] reported on embodiment responses in 2 subjects with transhumeral amputation who had undergone “the sensory component” of

targeted reinnervation and who had clear sensory percepts to touch projected to the missing limb. Our group recently reported a subject with transhumeral amputation who underwent a new fascicular end-to-end technique by coapting individual sensory fascicles of the median and ulnar nerves to target cutaneous areas [10]. That subject developed distinct median and ulnar hand maps in two separate cutaneous areas, separated from the motor sites in the residual limb.

Based on these reports, we categorized the techniques employed for sensory reinnervation [11] into 3 approaches; 1) targeted reinnervation with skin denervation over the muscle site; 2) targeted reinnervation with end-to-side cutaneous nerve transfer; and 3) targeted reinnervation with fascicular end-to-end nerve transfer. Given the increasing interest in pursuing methods of incorporating physiologically natural matched sensory feedback from prosthetic limbs, a more detailed comparison and critique of targeted sensory reinnervation approaches may be instructive. The goal of this report is to compare the sensory outcomes of 3 different targeted reinnervation techniques used in our center on transhumeral amputees. The intent is to inform future planning of optimal surgical techniques to maximize sensory restoration with reinnervation.

### METHODS

At our center we have performed 7 targeted reinnervation procedures on transhumeral amputees since 2008. We performed detailed sensory testing on 3 subjects, each with a different sensory reinnervation approach. Informed consent was obtained with approval from the health ethics research board at the University of Alberta.

#### Sensory Reinnervation Techniques

- 1) **Subject TH1** had the standard targeted muscle reinnervation approach, with thinning of subcutaneous tissue overlying the muscle sites causing local skin denervation;
- 2) **Subject TH2** had targeted muscle reinnervation and an end-to-side coaptation of medial brachial cutaneous nerve to the median nerve;

3) **Subject TH3** underwent a new end-to-end fascicular sensory reinnervation technique. A sensory fascicle of the median nerve was coated to the intercostobrachial cutaneous nerve, and a sensory fascicle of the ulnar nerve was coated to the axillary sensory branch. The precise intra-operative technique has been previously reported [10].

Motor Testing

We used an 8 channel Bagnoli-8 EMG Acquisition system (Delsys, Inc.) to record surface EMG from the muscles of the residual limb. The subject was asked to think of performing a specific limb movement relating to the elbow, wrist, or hand. During the visualization, the muscles were first palpated to feel the contraction then the electrode was placed over the muscle to record the signal strength. A graphical user interface displayed EMG signal strength and controls for adjusting mappings, signal gain, and thresholds. The process of identifying the number of separable muscle signals was an iterative process consisting of exploring electrode placement and adjusting signal gain and thresholds, as per the standard approach to myoelectric signal site testing. The number of discretely separable motor signals was confirmed by matching the signal to operation of a robotic arm with 5 available degrees of freedom [12].

Sensory Testing

Skin pressure sensitivity and anatomical sensory mapping was performed on the residual limb. Sensory threshold for just discernable pressure stimuli was established using a 20 monofilament testing set. Light touch stimulation with a cotton ball and discrete touch with a Semmes-Weinstein monofilament was used to test for the presence of hand or digit sensation. The subject was asked to report where they felt the sensation when the residual limb was touched. To test reliability and consistency, each point was checked 3 times in random order and the subject’s sight was obscured to remove any visual cues.

**RESULTS**

Subject characteristics are listed in Table 1. All subjects were male with unilateral transhumeral amputation.

Motor Outcomes

All subjects successfully reinnervated the target muscles, with 4 discretely separable motor signals for hand close (medial biceps), hand open (lateral triceps), elbow flexion (lateral biceps) and elbow extension (medial triceps). The signals could be used to simultaneously operate two degrees of freedom of a robotic arm. In addition, subject TH3 had a discretely separable wrist flexion signal in brachialis muscle, which was used for a switching function. TH2 had palpable muscle contraction in brachialis with imagined wrist flexion, but it could not be discretely separated from the biceps hand close signal.

Clinically, all three subjects were able to operate a myoelectric prosthesis with simultaneous motion of elbow and hand. TH1 wore a myoelectric prosthesis that included a Dynamic Arm elbow and Greifer terminal device (Ottobock, Inc.) exclusively until last year, when component repairs kept the prostheses away for the greater part of a year. TH2 operated a myoelectric prosthesis with a Boston Elbow (Liberating Technologies, Inc) and an ETD hook (Motion Control, Inc.) when working at office duties, and a body powered hook for working in the field. In the last year he discontinued myoelectric use due to inconsistent performance of components, electrode contact issues and donning time, but continued to wear the body powered prosthesis daily. Both of these subjects are in the process of being fit with new updated myoelectric components. TH3 is a full time daily myoelectric user with a Boston Elbow and BeBionic hand (RSI Steeper, Inc).

Table 1: Targeted reinnervation subject characteristics

Subject	*Age	Time from initial amputation to reinnervation	Residual limb length	Motor reinnervation technique	Sensory reinnervation technique	Time of follow up
TH1	62	29 months	25 cm	**Standard	Thinning of subcutaneous tissue	5 ½ years
TH2	29	20 months	28 cm	Standard + median n. to brachialis	End-to-side: MBC to median n.	6 years
TH3	20	10 months	26 cm	Standard + ulnar n. to brachialis	Fascicular end-to-end: IBC to median n. and axillary cutaneous to ulnar n.	15 months

\* Age at time of targeted reinnervation surgery; \*\* “Standard” motor reinnervation = reinnervation of median n. to medial biceps; distal radial n. to lateral triceps; MBC = medial brachial cutaneous; IBC = intercostal brachial cutaneous.

### Sensory Outcomes

**1) Subject TH1:** At 8 months post reinnervation, light touch on the anterior distal residual limb elicited referred sensation to the volar tips of digits 1, 2, and 3. The sensation was described as a strong pins and needles sensation; with deeper pressure, the sensation would extend to multiple digits and the palm in some areas. Posteriorly on the residual limb, there were 3 areas along the incisional scar where the subject experienced referred sensations to the dorsal aspect of digit 2, 4 and 5. These findings were consistent with the location of the motor nerve transfers.

At 5½ years post reinnervation, the hand map was greatly reduced on the residual limb. Sensibility was poor at 4.56 g monofilament threshold throughout. Touch to the very distal end of the residual limb at different points elicited sensation of touch to the volar aspect of the thumb, a brushing sensation across digits 2-4, and tingling in digits 3-5 (shown in Figure 1). There was one location on the lateral arm that referred to the tip of digit 2.

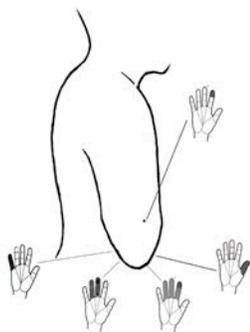


Figure 1: Sensory hand map of TH1, 5 years post reinnervation. There were several points clustered at the distal end of the limb.

**2) Subject TH2:** At 6 months post reinnervation, light touch to the anteromedial distal residual limb elicited a tingling sensation to the volar aspect of digit 1-3, and localized tingling to the distal finger and base of the fingers. This was consistent with the median nerve transfer. Stimulation to the posterior distal limb referred sensation primarily to the hypothenar eminence and digits 4/5. Sensation was best elicited with “moving touch” or a stroking movement across the skin.

At 6 years post reinnervation, subject TH2 had generally poor sensation in the residual limb, with a monofilament sensibility threshold of 4.17 to 4.56g throughout. Precise localization of stimulation was poor, and the subject often localized the sensation occurring several inches proximal to the actual applied area on the arm. He also reported the sensation of “telescoping” of his phantom limb, with

experienced hand position immediately at the end of the distal limb.

Hand mapping with a 4.56 g monofilament identified a number of locations where the stimulation referred to the palm of the hand, and radiated up the ulnar border of the hand, mostly described as “pressure” or sharp tingling sensation (shown in Figure 2).

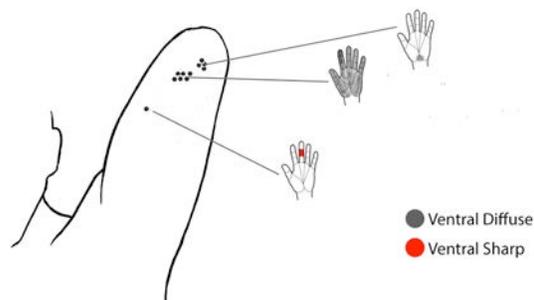


Figure 2: Sensory hand map of TH2, 6 years post reinnervation. There were several areas on the anteromedial limb that corresponded to sensation in the palm and heel of the hand, illustrated as a cluster of dots in this figure.

**3) Subject TH3:** At 4 months post reinnervation, pressure sensitivity was diminished in the cutaneous denervated territories, with a 6g monofilament threshold. However, mapping of hand sensation was consistent with the nerve transfers. Specifically, all median nerve digit sensation corresponded with the intercostobrachial cutaneous nerve territory, and all ulnar nerve sensation corresponded with the axillary cutaneous nerve branch territory (Figure 3).

Mapping results at 15 months post reinnervation have been previously reported [10]. The representations of the digit sensations became widely spread with multiple locations for each digit within the cutaneous territories. Sensitivity threshold improved to 0.4g monofilament. Subjectively, the subject reported that in the areas where single digits were felt, it was a feeling of his digit being touched with increasing pressure when greater force was applied. In areas where the subject could feel 2 or 3 digits the sensation was reported as “brushing” in quality. Force level discrimination was 88% and 100% for 3 levels of force in the index and small finger areas respectively [10].

## DISCUSSION

The first two reinnervation techniques initially restored limited areas of cutaneous sensory percepts corresponding to the hand map. The percepts found within the first 8 months were located in the skin overlying the muscle, as expected. In contrast, the third subject with the fascicular sensory end-to-end technique developed widespread topographic

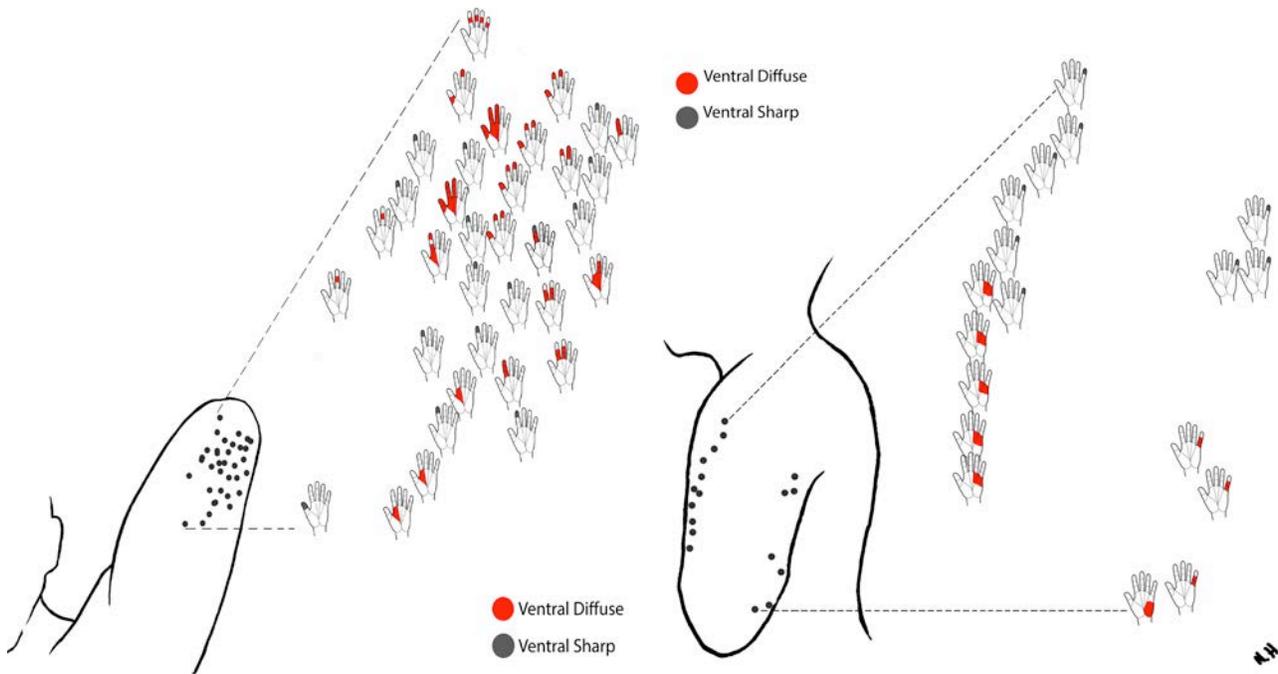


Figure 3: Sensory hand maps of TH3, 4 months post reinnervation. (a) Sensory points in the intercostobrachial cutaneous area corresponded to median n. digit sensation. (b) Sensory points in the axillary area corresponded to ulnar n. sensation. By 15 months, these areas had filled in with additional widespread representation of the median and ulnar hand maps respectively [10].

representation of the digits with discrete separation of the median and ulnar hand maps.

It should be noted that detailed sensory threshold testing was not performed prior to the reinnervation surgeries in the first two subjects, as the focus was on the muscle reinnervation procedure. Therefore they may have had impaired sensation on the residual limb prior to reinnervation, accounting for the poorer post-operative sensory recovery. For our most recent case (TH3), detailed sensory testing assured us that he had normal sensitivity thresholds prior to the planned sensory reinnervation, which likely greatly influenced the success of the restored sensory percepts. Nonetheless the discrete hand maps of the median and ulnar nerves in separate cutaneous areas, not overlying the muscle sites, confirms that surgical control over the cutaneous sensory restoration is possible.

Comparing our results to those presented in the literature, detailed sensory mapping results have only been reported for reinnervation subjects at shoulder disarticulation and short transhumeral amputation levels. The end-to-side technique in subjects with sensory reinnervation over the chest has demonstrated discrete transferred sensation of the hand map with excellent sensibility [4], comparable to our end-to-end fascicular sensory reinnervation subject. However, the somatotopic organization of the hand map in the end-to-side sensory reinnervation subject was intermixed with both median and ulnar afferents in the same regions [1].

In contrast, in our subject TH3, there was clear exclusive separation of median and ulnar hand maps in two separate cutaneous areas. In the literature, no other detailed hand maps for subjects with the transhumeral reinnervation procedure have been reported.

Another interesting observation is found in the fact that the first two subjects showed attenuation in the hand maps over time. The maps were not stable after 5 years of follow up. This change may be due to reduced sensitivity from full time prosthetic use, as patients using a prosthesis have been shown to have significantly poorer touch-pressure sensitivity in the residual limb compared to non-users [13]. Another possibility is lack of functional use of the sensory percepts. Factors influencing the strength and maintenance of sensory percepts will need to be examined in future study. In particular, sensory training and the use of relevant sensory feedback within a functional prosthesis may have an impact on sensory thresholds by strengthening cortical representation of the missing limb.

## CONCLUSIONS

Based on the findings in our three cases, we have suggestions for future consideration. First, detailed sensory testing should routinely be performed preoperatively prior to reinnervation surgery. Secondly, to optimize sensory outcomes, handling of the sensory nerves and the sensory

approach should be as carefully and deliberately planned as the motor nerve transfers. Lastly, the effects of sensory training on strengthening sensory percepts after reinnervation should be further investigated.

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## TEST-RETEST RELIABILITY AND RATER AGREEMENTS OF THE ASSESSMENT OF CAPACITY FOR MYOELECTRIC CONTROL VERSION 2.0

Liselotte Maria Norling Hermansson<sup>1</sup>, Helen Y. N. Lindner<sup>1</sup>, Ann Langius-Eklöf<sup>2</sup>

<sup>1</sup>*School of Health and Medical Sciences, Örebro University, Örebro, Sweden*

<sup>2</sup>*Department of Neurobiology, Care Sciences and Society, Karolinska Institute, Stockholm, Sweden*

### ABSTRACT

#### Introduction:

The Assessment of Capacity for Myoelectric Control (ACMC) is an observation-based tool that evaluates ability to control a myoelectric prosthetic hand [1]. Validity evidence led to ACMC version 2.0, but test - retest reliability and minimal detectable change (MDC) of ACMC have never been evaluated. For instruments that have an evaluative purpose, such as ACMC, the MDC is a useful clinical value to suggest whether a change is due to measurement error or true change. Investigation of rater agreements in this version was also needed because it has new definitions in certain rating categories and items.

#### Methods:

Upper limb prosthesis users (n=25, 13/12 male/female, 15/10 congenital/acquired; mean age 27.5, range 7-72, years) performed one standardized activity twice, 2–5 weeks apart. Activity performances were video-recorded and assessed by two ACMC raters. The item raw scores were converted to Rasch interval ability measures. Ordinal data were analyzed by weighted  $\kappa$ ; interval data were analyzed by intraclass correlation coefficient (ICC) and Bland–Altman limit of agreement (LOA) method.

#### Results:

For test–retest reliability, ICC<sub>2,1</sub> was 0.94. Average weighted  $\kappa$  was 0.76 and percentage agreement (PA) was 85%. In individual items, weighted  $\kappa$  agreements were fair to excellent (0.52–1.00) and PAs were  $\geq 66$ –100%. MDC<sub>95</sub> was  $\leq 0.55$  logits (1 rater) and 0.69 logits (2 raters). All MDC<sub>95</sub> values were  $\leq 5\%$  of the total ability logit range.

In the Bland–Altman plot the upper and lower LOA were 0.86 and -0.88 respectively. All except one participant were within the 95% LOA.

For inter-rater reliability, weighted  $\kappa$  agreements were fair to excellent in both sessions (0.44–1.00), and ICC<sub>2,1</sub> was 0.95 (test) and 0.92 (retest).

Intra-rater agreement (rater 1) was excellent (ICC<sub>3,1</sub> 0.98). The weighted  $\kappa$  values of the test session were all  $>0.80$  and the PAs for each item were  $\geq 96\%$ .

#### Conclusion:

The results of the present study demonstrate different aspects of the reliability of ACMC 2.0. Based on these results, we can recommend ACMC as a tool to follow the progress of users in controlling their myoelectric prostheses. The MDC is clinically useful for ACMC raters as a guideline when following the client's changes over time.

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## SWEDISH NORMATIVE DATA FOR THE SOUTHAMPTON HAND ASSESSMENT PROCEDURE – SHAP: PRELIMINARY FINDINGS

Liselotte Maria Norling Hermansson, Martin Andrén, David Johansson  
*School of Health and Medical Sciences, Örebro University*

### ABSTRACT

#### Introduction:

Today there is an increasing awareness of the need for validated instruments in upper limb prosthetics outcome assessments. One of the instruments suggested is the Southampton Hand Assessment Procedure (SHAP). In order for the clinicians to know how to interpret the results from the assessments, norms for different age groups are needed. Previously, normative data for SHAP has been reported for an English sample and work is underway from Slovenia. However, norms are only available for people up to 75 years of age, and no normative data is available for Sweden. Hence, the aim was to collect data also from people age 75 and older, and from Sweden.

#### Methods:

*Participants and procedure* — 58 persons (md age 52.5, range 20-92 years) participated in the study. Participants were recruited among students at Örebro University, staff at Örebro University Hospital, and people attending a community-based centre for senior citizens. Inclusion criteria were self-determined normal hand-function and no hand-impairment at time of testing. Data was collected according to the standardized procedure in the manual, and by one single “assessor” an OT-student.

*Instrumentation* — The SHAP consists of 26 tasks: 6 abstract objects in both lightweight and heavyweight form, and 14 simulated ADL-tasks. All tasks are representing one out of six grip-patterns. The subject is instructed to start and stop the timer before and after performing the task. An overall score, the Index of Functionality (IOF), is calculated based on the resulting times.

#### Results:

Mean IOF was 96.7. One-way ANOVA with index of functionality as dependent variable showed a statistical significant difference in dominant hand function between age groups.

## INVESTIGATING THE USE OF IMITATION TO SUPPORT LEARNING TO USE A MYOELECTRIC PROSTHESIS - A PILOT STUDY IN ANATOMICALLY INTACT SUBJECTS

Kelly Hesselink<sup>1</sup>, Laurence Kenney<sup>2</sup>, Sibylle Thies<sup>2</sup>,  
Adam Galpin<sup>2</sup>, John Head<sup>2</sup>, Emma Gowen<sup>3</sup>

<sup>1</sup>University of Twente, Netherlands; <sup>2</sup> University of Salford; <sup>3</sup>University of Manchester

### ABSTRACT

Controlling a myoelectric prosthesis is fundamentally different from controlling an intact hand and hence appropriate training is required (Bongers 2012). One promising approach to training new motor skills is based on the use of imitation (Celnik 2008). Recent work has shown that brain activity patterns more closely reflect those seen in normal movement planning and execution when amputees imitate a prosthesis user, compared with imitation of subject with anatomically intact arm (Cusack 2012). A very recent study suggests that imitation of skilled prosthesis users may also improve motor performance, when compared with imitation of an intact person (Cusack 2014).

In this study we built on the initial findings of Cusack and colleagues to develop and pilot a protocol to investigate imitation of an expert prosthesis user as training tool. In contrast to Cusack, we developed a protocol involving practice of transitive actions (use of the prosthesis to perform a common functional task) and intransitive actions (use of the prosthesis to perform actions or gestures, not associated with acquiring an object). We also extended the work of Cusack's group by examining gaze behaviour during task practice.

The pilot study was a 2 group experimental design. The intervention group received imitation-based training and the control group learned to perform two tasks using a conventional training method. The imitation group were shown transitive and intransitive videos, which they imitated. The control group received written instructions, based on which they performed the tasks. For both groups the tasks involved participants, starting from a pre-defined pose reaching to grasp a vertical cylinder, moving it to a tube, releasing it then returning the hand to the starting location. Task difficulty was manipulated by the orientation of the target tube (easy — vertical; hard — horizontal). The pilot measurement protocol was tested with five participants. Most of the outcome measures, including percentage of tasks successfully completed, duration of reach phase and duration of manipulation phase showed some evidence of a learning effect over the protocol in some participants, although peak velocity did not. No difference was observed between groups.

The results of testing the measurement protocol showed the potential to extend the study, but also identified problems which first should be solved.

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## A NEED FOR A MORE USER-CENTERED DESIGN IN BODY POWERED PROSTHESES

Mona Hichert, Dick H. Plettenburg, Alistair N. Vardy

*Delft Institute of Prosthetics and Orthotics, Delft University of Technology, The Netherlands*  
*m.hichert@tudelft.nl, www.dipo.3me.tudelft.nl*

### INTRODUCTION

Users of body powered prostheses (BPP) complain about too high operating forces, leading to pain and/or fatigue during or after prosthetic operation. In the worst case nerve and vessel damage can occur [1, 2], leading to non-use of prostheses. Smit et al. investigated cable forces and displacements required to operate commercially available voluntary closing and voluntary opening hands and hooks [3, 4]. The capacities of prosthetic users to operate these terminal devices remain unknown. Taylor reported in 1954 forces and displacements measured with 50 'normal' subjects for arm flexion ( $280 \pm 24$  N;  $5.3 \pm 1.0$  cm), shrug ( $270 \pm 106$  N;  $5.7 \pm 1.5$  cm) and arm extension ( $251 \pm 29$  N;  $5.8 \pm 1.7$  cm) (mean $\pm$ SD) [5]. Unfortunately, the measurement procedure is unclear. Moreover, the study reported forces and displacements from isolated movements instead of combinations of movements typically used for BPP operation. Our recent pilot experiments on 10 male subjects ( $28 \pm 2$  years old) also without arm defects using a BPP harness revealed average values of 475 N and a peak value of 970 N for one subject. Although these values are higher, it remains unclear if these force levels are sufficient to comfortably operate a BPP, or too low leading to non-use. Importantly, knowing the capacities and limitations of prosthetic users will aid in choosing and redesigning future BPPs to prevent non-use.

The goal of this study is to investigate the maximum cable operating forces prosthetic users can develop on the control cable. These maximum forces will be compared to the cable forces required to operate commercially available BPP based on the measurements of Smit et al. [3, 4]. Furthermore, this study addresses the question, whether it is possible to predict maximum cable operation forces by the anthropometric data of users in terms of shoulder width, upper arm length and upper arm circumference (serving as a measure of muscle volume), facilitating the prosthesis fitting procedure and preventing the need for costly measurement equipment.

### METHOD

This study was approved by the medical ethical committee of University Medical Centre Groningen (UMCG).

The subjects were recruited from University Medical Centre Groningen, Erasmus Medical Centre, Rotterdam, and the rehabilitation institute De Hoogstraat, Utrecht.

#### Subjects

In this study 25 adults (13 females and 12 males, age:  $49 \pm 13$  years, height:  $175 \pm 8$  cm, weight:  $75 \pm 14$  kg, mean $\pm$ SD) with a trans-radial deficiency participated. All participants were free of neurological, muscle, joint or motor control problems concerning the upper extremity or the torso (exclusion criteria). A total of 16 participants had a left deficiency, and 9 had a right deficiency, 15 had a congenital defect, and 13 had experience with BPP.

#### Equipment

##### *Anthropometric data*

The subjects shoulder width, upper arm length and remaining lower arm length was measured with an anthropometer (GPM - Model 101). For measuring the upper arm circumference a sewing tape measure was used. The subjects' length was measured by a tape measure connected to the wall. Body weight was taken by Soenle Scale.

##### *Maximum force measurements*

For measuring maximum cable operation forces, a prosthetic simulator was used (Figure 1), consisting of a thermoplastic shell with a 3.5 mm neoprene cover at the inside. With Velcro straps the simulator can be fitted on the hard socket of the subject's prosthesis. A 1.5 mm steel cable was used as operating cable running from the prosthetic simulator to the shoulder harness interrupted by a force sensor (S-Beam load cell ZFA 100kg). Cable excursion was disabled in this setup. The shoulder harness was adjustable to the subject's dimensions. The force sensor was amplified (Scaime, CPJ) and sampled (NI USB-6008), and finally stored using a custom LabVIEW programme (LabVIEW 2012 version). Cable forces were recorded with a sampling rate of 333 Hz.

##### *Procedure*

Prior to the measurements subjects were requested to read the information letter and sign an informed consent form. Personal data (gender, age, dominant and amputated

side, experience in prosthetic use, currently and previously used prostheses, cause of deficiency) were recorded and body measures (height, weight, shoulder width, upper arm length, remaining lower arm length, upper arm circumference affected and sound side) were taken. Anthropometric data were taken following the instructions of the NASA Reference Publication 1024 [6]. Shoulder width was taken according to “103. Biacromial Breadth”, upper arm length of amputated side according to “751. Shoulder-Elbow Length”, upper arm circumference according to “113. Biceps Circumference, Relaxed”, remaining lower arm length according to “381. Forearm-Hand Length”, where the fingertips are represented by the far end of the subjects’ stump.

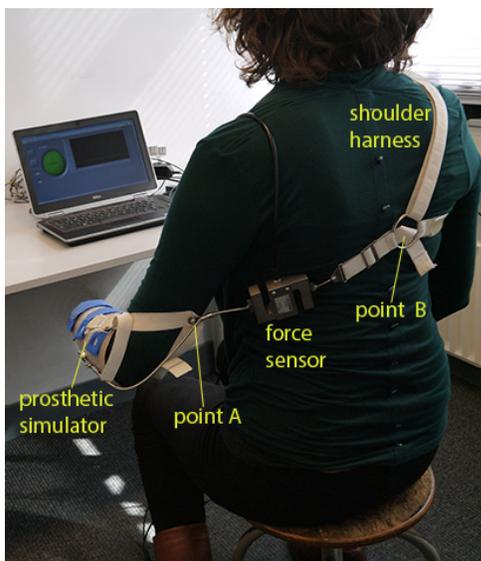


Figure 1: Measurement set-up.

#### Fit of equipment

A prosthetic simulator was connected to the subjects’ prosthesis. For two subjects (one male and one female), which did not possess an own prosthesis, the simulator was placed on a temporary WILMER Open Fitting [7]. For two subjects (one male and one female) the simulator was placed on the remaining arm. The straps from the prosthetic simulator were fitted in a way that point A (Figure 1) was on approximately 1/3 of the upper arm length above the elbow. The harness ring was placed lateral to the spinal cord on the affected side at the level of the shoulder blade (point B in Figure 1). When the subject was standing upright, raising the sound arm to a 90 degree angle with the thorax, neither tension nor sag of the control cable occurred.

The end of the control cable was fixated to the prosthetic simulator and cable displacement was disabled. Once the equipment was fitted, the subject was instructed to use shoulder protraction of the sound side, humeral abduction

and antelexion on the affected side simultaneously to create cable forces. Next, after the measurement program was started by the experimenter, the subject delivered his/her maximal force level, i.e. cable force for 3 seconds. This procedure was repeated 3 times.

#### Data analysis

Data analysis was performed using Matlab (version 2013b), inspected visually and the maximum over the three trials was determined.

The upper-arm circumferences of 5 subjects with a BMI (weight [kg]/ (height [cm])<sup>2</sup>) higher than 30 kg/cm<sup>2</sup> were removed from the analysis as their data would almost certainly be affected by fat depositions.

#### Statistics

For statistical analysis SPSS version 20 was used, and a significance level of  $\alpha=0.05$  was maintained. A three-way ANOVA was used to evaluate the effects of *gender* (male vs. female), *experience* (prior BBP experience vs. no BBP experience), and *defect type* (congenital vs. other causes). Correlations between maxima and anthropometric data were analysed using the Pearson correlation coefficient.

## RESULTS

The maximum cable operation force averaged over all subjects was  $267 \pm 123$  N. The maxima deviated from 87 to 538 N over all subjects resulting in a range of 451 N. Forces created by female subjects ( $194 \pm 86$  N) were significantly lower than those of males ( $346 \pm 108$  N) ( $F_{1,21}=10,647$ ,  $p=0.004$ ). No significant effect of experience was found, experienced BPP-users ( $285 \pm 106$  N), non-experienced BPP-users ( $247 \pm 141$  N) ( $F_{1,21}=2,313$ ,  $p=0,143$ ). Finally, maxima of subjects with a congenital deficiency ( $222 \pm 76$  N) showed no significant difference compared to the maxima of subjects with acquired arm defects ( $334 \pm 151$  N) ( $F_{1,21}=3,459$ ,  $p=0.077$ ). However, a striking difference in the range of maximum delivered forces must be reported (260 N for subjects with congenital arm defects versus 451 N for subjects with an acquired arm defect).

These maximum operating forces of potential users were compared to the required operation forces for commercially available voluntary opening (VO) BPP, when realizing a hand opening of 50 mm and voluntary closing (VC) BPP, when creating a pinch force of 15 N [3,4].

Tables 1 and 2 show the number (percentage) of subjects which are able to operate a certain prosthesis with their full strength. Monod reported that the value for the critical force, the force that humans can conduct without fatigue effects during continuous isometric contractions, lies between

15 and 20% of the maximum voluntary contraction [8]. Hence, Tables 1 and 2 also show the number (percentage) of subjects, which are able to operate the devices with 20% of the measured maxima. Summarized, Tables 1 and 2 show, that 3 out of the 7 VC and 2 out of the 14 VO devices cannot be operated by all subjects with the highest force they can create on the control cable. When considering the non-fatigue level at 20% of the maximum operation force, none of the VC and VO devices can be operated by all subjects.

Shoulder width, upper arm length, upper arm circumference of the affected and the sound side were correlated with the maximum operation forces of subjects. Pearson correlation coefficients are shown in Table 3. The correlation coefficients were found to be significant for shoulder width, upper arm circumference of the affected and the sound side. Additionally, the correlation coefficients show a positive linear trend. However, the relatively low coefficients represent a large deviation of the correlated data points.

Table 1: Subjects able to operate voluntary closing BPP

VC Prosthesis	required cable force for a 15 N pinch [3]	subjects able to create required cable force	subjects able to create required cable force with 20% of max. force
	N (mean±std)	number of subjects (percentage of subjects)	number of subjects (percentage of subjects)
Hosmer APRL hand, 52541 (L) size 8	61±0.6	25 (100%)	9 (36%)
Hosmer APRL hook, 52601	62±0.0	25 (100%)	8 (32%)
Hosmer soft hand, 61794 (R) size 7 3/4	131±0.7	22 (88%)	0 (0%)
Otto Bock, 8K24 (L) size 7 3/4, frame	78±0.3	25 (100%)	3 (12%)
Otto Bock, 8K24 (L), size 7 3/4, frame and inner glove	90±0.9	24 (96%)	3 (12%)
Otto Bock, 8K24 (L) size 7 3/4, frame + inner glove, and cosmetic glove	98±0.5	24 (96%)	2 (8%)
TRS hook, Grip 2S	33±0.2	25 (100%)	19 (76%)

Table 2: Subjects able to operate voluntary opening BPP

VO Prosthesis		required cable force for 50 mm prehensor opening [4]	subjects able to create required cable force	subjects able to create required cable force with 20% of max. force
		N (mean±std)	number of subjects (percentage of subjects)	number of subjects (percentage of subjects)
Hosmer Model 5XA Hook	1 band	25 ± 0.3	25 (100%)	22 (88%)
	2 bands	50 ± 0.2	25 (100%)	14 (56%)
	3 bands	71 ± 0.2	25 (100%)	7 (28%)
Hosmer Sierra 2 Load VO Hook	Set. 1	40 ± 0.3	25 (100%)	15 (60%)
	Set. 2	82 ± 0.1	25 (100%)	3 (12%)
RSL Steeper Carbon Gripper	Set. 1	43 ± 0.3	25 (100%)	14 (56%)
	Set.2	48 ± 0.1	25 (100%)	14 (56%)
Otto Bock Model 10A60 Hook (2 × 2 Springs)	Set. 1	32 ± 0.5	25 (100%)	20 (80%)
	Set. 2	94 ± 0.3	24 (96%)	2 (8%)
Hosmer BeckerImperial Hand (ungloved)		63 ± 0.4	25 (100%)	8 (32%)
Hosmer Sierra VO Hand	Gloved	70 ± 0.6	25 (100%)	7 (28%)
Hosmer Soft VO Hand	Gloved	104 ± 0.9	23 (92%)	1 (4%)
RSL Steeper VO Hand	Gloved	81 ± 0.7	25 (100%)	3 (12%)
Otto Bock VO Hand	Gloved	79 ± 0.5	25 (100%)	3 (12%)

Table 3: Pearson correlation coefficient

		maximum force
shoulder width	Pearson correlation	0,594**
	Sig. (2-tailed)	0,002
	N	25
upper arm length	Pearson correlation	0,232
	Sig. (2-tailed)	0,264
	N	25
upper arm circumference sound arm	Pearson correlation	0,543*
	Sig. (2-tailed)	0,013
	N	20
upper arm circumference affected arm	Pearson correlation	0,449*
	Sig. (2-tailed)	0,047
	N	20

\*.Correlation is significant at the 0.05 level (2-tailed)

\*\*Correlation is significant at the 0.01 level (2-tailed)

## DISCUSSION

In this study 25 subjects with a trans-radial deficiency participated. On average they created a maximum cable force of 267±123 N. Males created significant higher forces than females ( $F_{1,21}=10,647, p=0.004$ ). No significant

differences were found for experienced BPP-users versus non-experienced BPP-users ( $F_{1,21}=2,313$ ,  $p=0,143$ ). In addition, forces created by subjects with a congenital arm defect versus by subjects with acquired arm defects showed no significant differences ( $F_{1,21}=3,459$ ,  $p=0,077$ ). Comparing these results to the study of Taylor (arm flexion ( $280\pm 24$  N;  $5.3\pm 1.0$  cm), shrug ( $270\pm 106$  N;  $5.7\pm 1.5$  cm) and arm extension ( $251\pm 29$  N;  $5.8\pm 1.7$  cm)) the order of magnitude of the maxima is the same, although isolated movements of 'normal' subjects were measured [5]. It might be that the increase in length and strength over the past 60 years is compensated by the fact that isolated movements of 'normal' subjects were measured or there was never a difference between subjects with versus without arm deficiency. In that case the trial experiments as mentioned earlier are not representative for a large population.

The results of Table 1 and 2 showed that 3 out of the 7 VC devices and 2 out of the 14 VO devices cannot be operated by all 25 users with the exertion of their full capacities. None of the devices can be operated when correcting the subject's maximum forces with a fatigue level (20% of the maximum force). This represents the poor match between user capacities and user demands the prosthetic devices offer. Ideally, the prosthesis must be operated without pain nor fatigue [1, 2]. It seems that the user demands have not been heard the past 25 years [4].

Note that estimations of fatigue presented in Table 1 and 2 are based on theoretical values of Monod [8], who reported a critical force between 15 and 20% of the MVC, thus with 20% the conservative value was taken. Furthermore, the required cable operation forces are only representing the prehensors and are not taking into account any friction losses due to the Bowden cable transmission. The reported efficiencies of Bowden cables in BPP-use can decrease to 60%, depending on the curvature of the cable and the material the cable is made of [9]. Even so, the pinch force level of 15 N set as a measurement requirement for voluntary closing prehensors in Smit and Plettenburg's study is only an estimation [3].

This study addressed the possibility of predicting maximum cable operation forces by the anthropometric data of users in terms of shoulder width, upper arm length, and upper arm circumference of both arms. Significant Pearson correlation coefficients were found for shoulder width, upper arm length, upper arm circumference of affected and sound arm. Shoulder width and upper arm circumference seem to have a predicting quality, even though it is a weak one. The exact maximum cable operation force cannot be predicted for a specific user by taking the anthropometric data, but due to the dimensions of shoulder width and upper arm circumference the user can at least be categorized (e.g. in S, M, L, XL). However, the upper arm circumference, as

a measure of muscle volume, cannot be applied for users where large fat deposits interfere with the muscle estimate. As such, participants with a BMI > 30 kg/cm<sup>2</sup> were excluded from the analysis. The significant correlations are a useful insight for designing prostheses in the future. The CPO may base the choice of device based solely on an relatively easy anthropometric measurement.

#### *Study limitations & recommendations*

This study did not evaluate the maximum cable excursions BPP-users can achieve. Additionally, the isolated operation movements have not been measured. A future study should address these questions.

Before exerting the maximum forces on the cable the subjects did not have any training. They were only instructed in which movements they should perform. This might partly explain the deviations in maxima. As a result of training, the maximum forces might be even higher. However, no significant differences were found between experienced and non-experienced BPP-users.

## CONCLUSIONS

The goal of this study was to investigate the maximum operating forces prosthetic users can create on the control cable. The created maximum forces were compared with the cable forces required to operate commercially available BPP based on the measurements of Smit et al [3,4]. Furthermore, the question whether it is possible to predict user capacities in terms of maximum cable operation forces by the anthropometric data of users was addressed in this study.

On average cable forces of  $267\pm 123$  N were created. With the measured maxima 3 out of the 7 VC devices and 2 out of the 14 VO devices could not be operated by all 25 subjects. When correcting the measured cable forces for fatigue effects during continuous operation (20% of the maximum force) none of the VC and VO devices can be operated by all 25 potential users. Significant Pearson correlation coefficients for shoulder width, upper arm circumference of affected and sound side versus the maximum cable operation force show a positive linear trend. However, with the anthropometric data of users it is not possible to predict maximum forces, but for a categorization of users strength the anthropometric data seems to be an appropriate measure.

Summarized, this study proves quantitatively that the forces commercially available BPP require are too high, with the result of not being applicable for all prosthetic users. The provided data helps us to understand how a BPP must be designed and serves as design requirements for new *user-centred* prosthesis design.

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## DESIGN FOR INDEPENDENT POWERED WRIST MOTION FOR TRANSRADIAL MYOELECTRIC PROSTHESIS USERS

Ali Hussaini, Peter Kyberd

*Institute of Biomedical Engineering, University of New Brunswick, Canada*

### INTRODUCTION

Current transradial myoelectric prostheses that incorporate a wrist rotation unit do not allow for simultaneous control of wrist rotation and terminal device. Selecting an active output is achieved by co-contraction (i.e. mode switching), physical switches (toggle, pushbutton) or specialized cable-harness mechanisms.

Independent control of hand and wrist function has been achieved in the past by utilizing embedded mercury switches, 3-state controllers, and fibre optic bend sensors. Control of wrist units is also offered commercially by using the surface electrode signals that drive the terminal device, though the hand cannot be used at the same time.

This research investigates simultaneous control of the wrist and hand function by reengineering the prosthetic socket interface and augmenting the available control input from the user. The objectives include the complete removal of mode-switching, suspension without a harness, and simultaneous motion of the hand and wrist unit. The primary focus is to determine if a viable control input can be created from the user's forearm rotation (e.g. pronation, supination) alongside surface EMG signals that drive hand function. These inputs can then be used to control two degrees of freedom simultaneously.

### BACKGROUND

A common prescription for a unilateral transradial amputation is a dual-site prosthesis that controls the opening and closing of an electric hand [1]. By individually contracting the flexor and extensor muscles of the forearm, the user can create two EMG signals which can be used to open or close a hand. Alternate control strategies for operating the electric hands exist for users that have only one distinguishable control site.

The inclusion of a wrist rotation unit adds a layer of complexity which involves balancing the physical nature of the new componentry with the perceived benefit of an additional degree of freedom. The wrist rotator can add

weight, bulk, length, and increase the moment arm. Previous studies have shown these to be some of the primary reasons for prosthesis rejection in general [2].

If the physical characteristics associated with a wrist rotator are overcome, the cognitive effort required to use the wrist rotator is another challenge. Conventionally, the control input signal is shared between the wrist and hand. As a result, only one can be controlled at any given time. This is not intuitive and may necessitate excessive concentration to control effectively. Often the user will decide that the required effort is too great and find a simpler strategy. The user will decide to over-rotate the shoulder to position their hand instead of using the powered wrist [3]. The time it takes to select the wrist rotator function, send a control signal to rotate the wrist, and then select the hand function again are all deterrents to rotator use. They may also decide to use their opposite intact hand to perform more complex motions [4].

#### Previous Solutions

The problems associated with controlling a wrist unit have parallels in the control of some of the more advanced prosthetic hand options that are being offered today. These advanced terminal devices require the correct combination of contractions (EMG amplitude, rate, instances) to switch control state [5].

These mode switching methods can be employed with the prosthetic wrist as well but can lead to muscle fatigue and require greater cognitive effort and time. For this reason, secondary sources of control (wrist or hand) should be investigated so that the prosthesis can function independently.

Additional control inputs for wrist rotation have been looked at previously which have included the UNB 3 state controller [6], mercury switches [3][7], fibre optic bend sensors [8], and commercial Otto Bock wrist control units [9], with varying degrees of success. The mercury switch solution augmented the surface electrodes with a second dedicated input source that controlled the wrist rotator. It was also small enough to be contained inside the prosthesis without increasing the overall length. This approach required the intact hand to perform a calibration, and used shoulder

abduction and adduction to activate the device (if sufficient forearm rotation did not exist).

### Pattern Recognition and TMR

Emerging pattern recognition (PR) methods may offer a viable input source for wrist control as well, as classification algorithms can decode the intended wrist motion from a network of surface electrodes. PR algorithms have shown success in detecting signal patterns, synergies and user intent, but clinical application and integration inside a rigid prosthetic socket introduces variables and challenges that have yet to be resolved [10]. When surface electrodes are forced to deal with loading forces and residuum motion inside a dynamic environment, PR accuracy can decline, especially when attempting simultaneous control. What they do offer is *seamless* sequential control when used with a physical socket. This can be visualized as a prosthesis user moving from hand to wrist motion, without have to perform a co-contraction to switch from hand mode to wrist mode. A recent study at the Rehabilitation Institute of Chicago (RIC) has tested simultaneous pattern recognition control by training motion classes separately and then combining the results to form a compound output (e.g. elbow extension and hand opening), with results indicating a user preference to use simultaneous control strategies [11]. However, forearm rotation was not one of the tested motions and a virtual arm was used for analysis.

Targeted muscle reinnervation (TMR) surgery has been used to offer simultaneous control to myoelectric users in the past [12], with success. This procedure provides a more intuitive control strategy, as the nerves that are involved are used to drive the same output in the prosthesis (e.g. electric hand motion) as the nerves did before the amputation took place. This results in a direct control strategy [13] which is conducive to a simultaneous output, as output motions of the prosthesis have independent control inputs. This also allows for proportional strategies to be more easily implemented, giving the user more capability in directing the prosthetic devices function. TMR surgery has so far been limited to amputations at levels proximal to the transradial amputation) [12][14]. Performing the TMR surgery at the transradial level, or any level distal to the elbow has the potential to increase the available (and distinct) user inputs to drive both a multi-axis wrist unit and an articulating hand. When coupled with a PR strategy, the ability to distinguish further movements/ intentions from the user may increase. This is an important area of research that has yet to be explored, where PR algorithms extract data from a TMR foundation in the forearm.

### Design Criteria

The aim of this project was to overcome these different effects and decouple the input for wrist motion from that of

hand function. This creates dedicated control inputs for wrist and hand motions. The motion of forearm rotation (pronation and supination) will be the physical point of control for the prosthesis user. The input will be rotation of the remnant forearm. Ideally, direct simultaneous control will result which utilizes intuitive control from the user, reflecting the natural motion (forearm rotation) of the anatomical limb.

The solution is in two parts. The first is to create the conditions for rotation with the patient's remaining residual limb. The second involves measuring that movement. Both of these aspects need to be solved together, they are not mutually exclusive as the way in which the residuum is allowed to rotate can dictate how that rotation is measured. This research is targeted to patients that retain a sufficient length of their forearm to allow rotation of their residuum to be effective, as forearm rotation becomes increasingly more difficult the shorter the limb is [15]. Users of two-site myoelectric prosthesis are the prime candidates.

## **IMPLEMENTING A SOLUTION**

The rigid nature of inner sockets in transradial prostheses drastically reduces or completely eliminates any potential forearm rotation that may exist. Therefore, the socket must be re-designed to allow for forearm rotation, whilst ensuring an intimate fit and adequate load bearing. It must also seem natural to the user.

A flexible liner (gel or silicone) would appear to be a solution for this design, as the innate compliance allows the liner to conform to the residual limb without restricting rotation. Thus a prosthesis that truncated the distal half of the inner socket and provided an opening for the liner encapsulated limb to pass through, solves the problem of donning the socket. Measurement of forearm rotation could then be achieved using an array of Hall Effect sensors and magnets, potentiometers, accelerometers or other detection devices. But there are problems with this approach.

The first problem involves the surface electrodes. Maintaining a rigid inner socket allows for electrode packages (contacts and preamplifiers) to remain in the socket. However, the liner will place a barrier between the electrodes and the skin. This will require creating an opening in the liner or placing remote electrodes inside. Cutting openings into the liner is not favourable as it introduces the potential for shear and tearing forces. The use of remote electrodes is not viable either as this will require the liner to stay in the socket. Donning a flexible liner while it is within the socket is impossible for a user. Pulling the liner out is restricted as the electrodes leads (remote electrodes) are of a given length. Remote electrode leads may also lead to additional electric noise as they would be rotating with the forearm section

of the prosthesis. This can potentially lead to unintentional opening and closing of the prosthetic hand.

The second factor to consider involves the loading forces placed on the residuum as it moves in space. If the outer forearm section is taken as a reference (specifically the axis of rotation of a wrist unit in the lamination ring), the distal end of the residual limb translates in a plane transverse to the longitudinal axis of the prosthesis when the forearm is rotated. This means that the distal end of the residuum does not track the distal end of the prosthesis exactly. This is a result of allowing the forearm to rotate within the socket. The envelope of translation needs to be restricted. Even without rotating the forearm, this translation occurs. The weight of a grasped object (or simply the weight of the terminal device) causes a deflection of the distal prosthesis with respect to the distal end of the residuum. This problem will intensify when an object is lifted and the elbow is flexed. As a result of the inner socket translating upward with respect to the distal end of the forearm section, a large force is localized at the top of the residual limb. Figure 1 illustrates this concept, and it can be seen that the localized force is at the point that makes contact with the ceiling of the forearm section. If this is a bony prominence (radial bone), the pressure may become excessive and unbearable for the user.

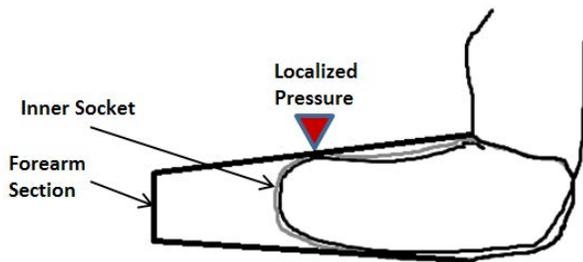


Figure 1: Localized pressure when elbow is flexed

Figure 1 also presents a secondary problem in terms of forearm rotation. If the distal portion of the residual limb is forced to carry the weight of the prosthesis, the potential rotation in the forearm is lost immediately. For example, the biceps brachii muscle cannot be used to supinate to the same degree because there is an increased demand for its role as an elbow flexor. There is also an effect from the cam created with forearm rotation from the neutral position (90° elbow flexion, thumb pointed upward). As soon as the user pronates or supinates, the forearm section of the prosthesis descends, as it tracks the relative height of the forearm inside. Rotation back to neutral requires that the user's forearm essentially lift up the prosthesis as it goes through its rotation. To solve this problem, the distal end of the residual limb needs to be linked to the forearm section where the wrist unit resides. This will solve two problems. The prosthesis will track with elbow

flexion and the localized pressure at the top of the residuum will decrease to an acceptable level.

### Current Design

Figure 2 shows the inner socket of a research prosthesis currently being developed. The liner is made of Seaflex 200 (North Sea Plastics) with circumferential fenestrations that are distal to the position of the surface electrode in the inner socket.



Figure 2: Liner and inner socket allowing rotation

This design allows for the distal portion of the liner to rotate with respect to the proximal end which resides inside the inner socket. Openings for the electrodes have been made in the liner to allow for skin contact. The Seaflex 200 material is rigid to the point that it is not susceptible to tearing. Figure 3 shows a test jig apparatus that links the rigid inner socket to the distal wrist rotator and hand with two aluminium uprights.

The system is powered with a 7.2 V (1200mAh) lithium battery and controlled with a microcontroller (Arduino UNO, Arduino). Two 3-axis accelerometers are used to calculate relative rotation and elbow flexion.

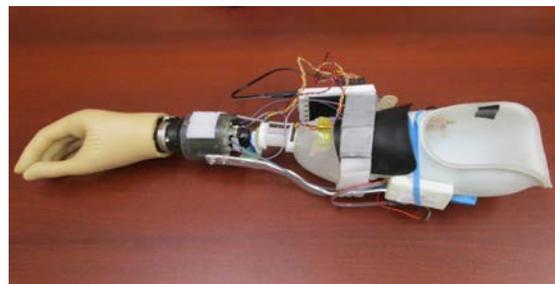


Figure 3: Testing apparatus

Figure 4 shows the physical link between the distal end of the residuum and the proximal end of the wrist unit. The link is made with a silicone coupler (Dragon Skin®, Smooth-On). This restricts the lateral movement of the residuum but allows for torsion. When the elbow is flexed at 90° the user is able to control the wrist rotator and hand independently.

This becomes difficult when the elbow is flexed beyond 90° and shoulder flexion and abduction are introduced. The silicone coupler also stretches along the longitudinal axis of the forearm when rotating away from the neutral position, indicating that the distance between the distal end of the residuum and the wrist rotator is not static. This is a result of the way in which the radial bone rotates around the ulna. This pistoning action may need to be accommodated for in future revisions.

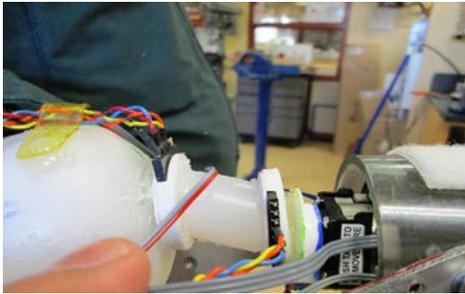


Figure 4: Coupling between residuum and forearm

The use of this particular silicone for the coupler mechanism allows for compliance and makes the liner easy to rotate, but this is at the cost of allowing unacceptable levels of deflection. A more rigid alternative needs to be found that will allow for torsion while allowing for the wrist unit to follow the trajectory of the residuum more closely.

## DISCUSSION

The concepts explored in this paper illustrate the key design considerations for improving transradial prostheses with independent and simultaneous wrist and hand motion for the unilateral user. The prosthesis should be able to operate independently from the opposite hand. Decoupling the hand input from the input for a wrist unit has the benefit in creating two dedicated input sources, which can lead to simultaneous and direct control, which is the preferred method of control.

Future work on this project will focus on the mechanism designs that will lock the distal end of inner socket with the forearm section, increasing the accuracy of detecting a forearm control signal. Performance measures and assessment tests will also need to be defined to test the efficacy of independent wrist control. This will likely include timed assessment tests and motion capture studies that will provide quantitative data that can be used for comparison. Previous work with transradial prosthesis users that analyze compensatory motions have been performed, but the number is limited [16][17][18]. It is also difficult to compare compensatory studies if standardized protocols for testing and measurement are not used. It is suggested that compensatory testing involving transradial prosthesis users begin moving to standardized

testing methods where time to complete tasks and limb kinematics can be recorded. The Box and Blocks test [19] and a clothespin relocation task [20] are two options that offer ease of testing, portability, repeatability, and allow for new prosthesis designs to be tested and compared reliably across different research institutions.

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## PORTABLE PATTERN RECOGNITION TRAINING SYSTEM

Tom M. Idstein, Laura A. Miller, Michael C. Stephens, Dat H. Tran, Levi J. Hargrove  
*Center for Bionic Medicine, Rehabilitation Institute of Chicago*

### ABSTRACT

Many multi-articulated myoelectric prostheses have been introduced in recent years. With the advent of commercially available pattern recognition systems, patients now have the ability to intuitively control a growing number of degrees of freedom (DOF) [1]. Patient acceptance and optimal user control is largely dependent on professional training provided prior to the use of an actual prosthesis — i.e., pre-prosthesis training [2]. Many myoelectric prostheses are abandoned by users due to the lack of easy to use, pre-prosthesis, training tools [3]. We have developed a portable, self-administered pre-prosthesis training system to improve the efficacy of pattern recognition (PR) control. The system includes a portable, self-administered surface electromyographic (EMG) signal acquisition system and an intuitive graphical user interface (GUI), which can reside on a personal computer. The user can control several DOFs on an avatar in one of three modes. The Practice Mode allows the user to freely move each DOF with only visual feedback to monitor success. The Targeted Achievement Mode and the Motion Completion Mode are two distinct modes in which the user must complete movements of certain DOFs. These modes provide both visual and quantitative feedback on control ability [4]. Users are able to increase difficulty to work towards more efficient control. The goal of this system is to provide patients with a tool that they can use in their own time and environment, to become familiar with and more competent at PR control. This may allow a patient to identify which control strategies or muscle movements work best, and to practice, thereby potentially increasing the muscle strength and endurance required to operate each prosthetic DOF. Allowing the patient to become familiar with PR control prior to clinic visits would maximize the clinical time available for the patient to learn about the prosthesis instead of learning control strategies or fighting fatigue. This system may also minimize the time necessary to perform outcome measures, or for the prosthetist/orthotist to collect pertinent data to indicate improved control prior to commercial purchase; such data may help justify reimbursement. A portable, self-administered, pre-prosthesis training system that provides patients with the opportunity to become familiar with PR control in their own time will both reduce clinic time and enhance prosthesis function.

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## FOCUS ON WELLNESS: A FORMAT FOR PROVISION OF MENTAL HEALTH SCREENING IN OUTPATIENT PROSTHETICS CLINICS

Warren T. Jackson, Tiffany A. Ryan  
*Advanced Arm Dynamics*

### ABSTRACT

beyond and, if indicated, to mobilize provision of mental health services by appropriate providers in their locale.

#### Methods:

The Wellness Inventory (WI) is a short battery of seven validated screening instruments that measure resilience (J. Block & Kremen, 1996), health-related quality of life (OPUS), pain (SF-36/12), depression (Kirkcaldy & Tynes, 2006), alcohol use (AUDIT-C), drug use/misuse, and posttraumatic anxiety (PC-PTSD).

#### Results:

During the initial evaluation of 123 adult prospective prosthetics recipients, the WI was administered by the examining OT. During the informed consent process, the participants were advised that the wellness screening did not constitute a formal psychological evaluation, no patient-doctor relationship was formed with the consulting psychologist, and no diagnosis would be rendered. The WI was administered orally to the participants who read the large-print items and response choices along with the examiner. In order to minimize participant burden inherent in completing paper-and-pencil measures, the examiner recorded participant verbal responses on a standardized response sheet. The results were then scored and summarized in a brief summary report by a licensed psychologist with separate sections for the insurer, referring physician, and clinical staff with recommended topics to review with the participant. Results: The sample of participants was 70.7% Caucasian with mean age of 43. Of the sample, 73.2% had experienced traumatic amputation with an average of 7.9 years since injury. Over 50% reported that pain interfered moderately to extremely with their ability to perform their normal work. Over 50% screened positive for depression and 22.8% screened positive for PTSD. As indicated by their WI responses, participants were provided with referrals to the appropriate mental health care providers.

#### Conclusions:

High prevalence of mental health concerns in this sample of participants confirms the need to include screening at time of initial contact. The results of the WI are designed to promote patient self-understanding during treatment and

## MOTOR ADAPTATION OF TRANSRADIAL AMPUTEES AND ABLE-BODIED SUBJECTS USING EMG CONTROL

Reva E. Johnson<sup>1,2</sup>, Konrad P. Kording<sup>1,2</sup>, Levi J. Hargrove<sup>1,2</sup>, and Jonathon W. Sensinger<sup>2,3</sup>

<sup>1</sup>*Rehabilitation Institute of Chicago, Chicago, IL, USA*

<sup>2</sup>*Northwestern University, Evanston, IL, USA*

<sup>3</sup>*University of New Brunswick, Fredericton, NB, Canada*

### INTRODUCTION

Sensory feedback remains an elusive goal for powered prosthesis users. Clinicians and researchers have explored feedback strategies since the early nineteen sixties with varying degrees of success, but none are used in clinical prostheses and an understanding of what feedback would be most effective is lacking [1], [2]. Some studies show that additional sensory feedback improves performance on discrimination tasks [3], [4] or in tasks where vision is removed [5], but many others show no clear improvement [2]. We don't understand whether amputees continuously rely on feedback during prosthesis movements, or if they are able to coordinate such movements using feedforward control while using feedback for error corrections. To guide sensory feedback strategies for prostheses, we need a better understanding of how amputees use feedback during movement planning and execution.

The influence of feedback on movement planning and execution can be predicted with Bayesian models of sensorimotor adaptation. These models assume that the brain predicts the next state of the body (e.g. position and velocity) and corrects this prediction using sensory feedback [6]. These two information processes—feedforward state prediction and feedback correction—are continually updated by the brain during movement. But the brain cannot perfectly predict the next state because of fatigue, variability of neuron firing rates, and other factors [7]. The brain also cannot perfectly evaluate feedback information because of the limited resolution of the sensory system. These imperfections introduce *uncertainty* in both the feedforward and feedback estimations. The uncertainty determines the integration of the two processes: if the feedforward state prediction is estimated with high uncertainty, the brain will rely more on feedback information during adaptation.

Prosthesis users may experience high levels of uncertainty in both feedforward and feedback information. The variability of electromyographic (EMG) signals used for prosthesis control is much greater than the variability of force and position signals used by able-bodied subjects to control objects [8], [9]. Thus, feedforward uncertainty may be

increased in amputees. Feedback uncertainty is presumably also higher in amputees because prosthesis users are limited to visual feedback and incidental feedback transmitted through the socket [10]. These factors affect adaptation and influence how feedback is used during movement.

Amputees have an intact central nervous system, which suggests that sensorimotor adaptation should follow similar patterns as in able-bodied subjects. Cortical reorganization is minimal for amputees using EMG-controlled prostheses [11] and adaptation is displayed during reaching tasks with body-powered prostheses [12]. To accurately describe and improve prosthesis control, we need to investigate the sensorimotor adaptation of amputee and able-bodied individuals using EMG control.

In this study, we compared adaptation behavior during EMG control in three conditions: amputee subjects using their residual limb, amputee subjects using their intact limb, and able-bodied subjects. We investigated trial-by-trial adaptation to visual perturbations with two levels of feedback uncertainty (similar to [13]).

### METHODS

Three subjects with transradial amputations and eight able-bodied subjects participated in this experiment, which was approved by the Northwestern University Institutional Review Board. Amputee subjects were between 59 and 75 years old (one female, two male) and able-bodied subjects were between 23 and 32 years old (three female, five male).

For able-bodied subjects, the dominant arm was strapped into a modified elbow brace that restricted motion. EMG activity during elbow extension was measured by a bipolar electrode positioned over the lateral head of the triceps brachii and was used to control a virtual cursor on a display screen. EMG signals were high-pass filtered at 0.1 Hz, rectified, low-pass filtered at 5 Hz, normalized, and mapped to cursor velocity [14]. Dynamics were chosen to match those of a typical prosthetic arm [15].

Amputee subjects completed the experimental protocol once using the residual limb and once using the intact limb, which were both strapped into a modified wrist brace that restricted motion. The two protocols occurred on the same day in randomized order with a rest break in between. For both the residual and intact limb conditions, EMG activity of the wrist flexors was processed in the same manner as able-bodied subjects and used to control the virtual cursor.

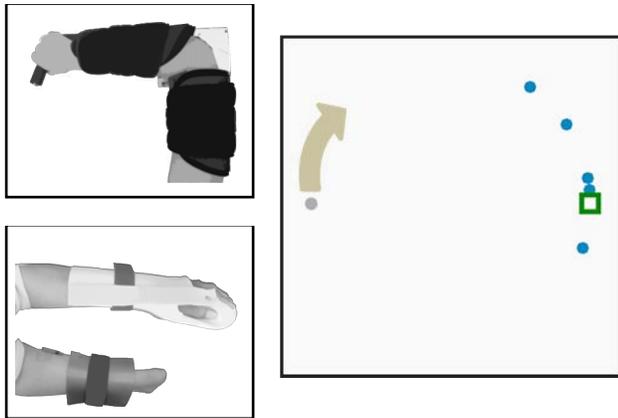


Figure 1: Experimental setup. Able-bodied subjects used EMG activity of the elbow extensors (top left) and transradial amputee subjects used EMG activity of the wrist flexors (bottom left) to control the virtual cursor. The cursor moved in a single-degree-of-freedom circular track and endpoint feedback of the movement was shown as either one dot or five dots (right)

The cursor moved along a single degree-of-freedom circular track with a radius of 13 cm. The starting position of the track was located at 180 degrees (left side of the circle) and a target was located at 0 degrees (right side of the circle). For each trial, subjects moved the cursor from the starting position to the target. Movement duration was limited to 3 seconds.

There were three phases of each experiment: familiarization, training, and testing. The familiarization phase consisted of 10 trials, in which the cursor was displayed as one dot that was unperturbed and visible throughout the movement. In the training phase, the cursor was still unperturbed and displayed as one dot, but visual feedback was taken away 0.5 seconds into the movement time. The cursor reappeared after the movement to give 100 ms of terminal feedback. Training continued until the subject was able to complete 10 trials with an average error of under 20 degrees. In the testing phase, subjects were given only terminal visual feedback. The testing phase included 4 blocks of 75 trials each, with approximately 2 minutes of rest between blocks.

During the testing phase, visual perturbations were applied to the displayed cursor endpoint. Perturbations were randomly distributed between -40, 0, and 40 degrees. Subjects were encouraged to hit the target as accurately as possible, and were instructed that the visual feedback (the dot) represented their true cursor position.

Two levels of feedback uncertainty were created by displaying the cursor as either one dot or five dots. When subjects saw the cursor as one dot, feedback uncertainty was low. When subjects saw five dots, feedback uncertainty was high. The location of the five dots was drawn from a Gaussian distribution with the mean as the cursor position and the standard deviation as 40 degrees. The feedback uncertainty for each trial was randomly distributed.

## RESULTS

We compared adaptation performance of three groups: amputee subjects using their residual limb, amputee subjects using their intact limb, and able-bodied subjects—all using EMG control. Subjects used EMG control to move a cursor towards a stationary target. We studied trial-by-trial adaption to visual perturbations with two levels of feedback uncertainty. Terminal visual feedback was displayed as one dot (low feedback uncertainty) or five dots (high feedback uncertainty).

Each subject displayed trial-by-trial adaptation to the visual perturbations (Fig 2). A visual perturbation in one direction typically elicited a correction in the opposite direction on the following trial. The average amplitude of the correction response, or the slope of the regression between error ( $n$ ) and perturbation ( $n-1$ ), is a measure of adaptation rate. The slope of each regression was negative; however, for the purposes of this paper we will report adaptation rates as positive values.

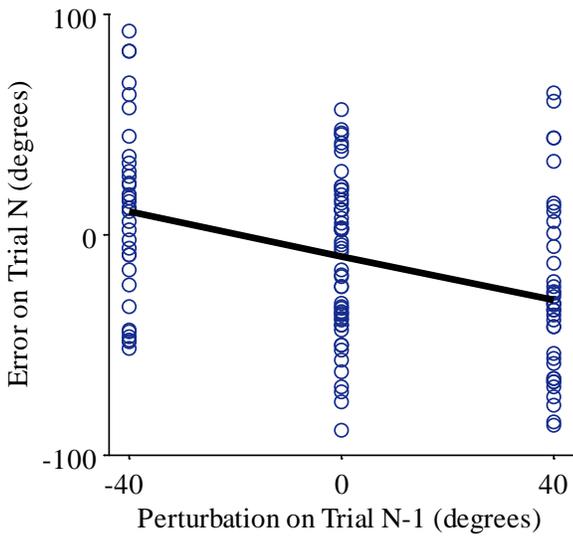


Figure 2: Representative plot of data from one subject using EMG from the residual limb during the low feedback uncertainty condition. Each circle represents an individual trial. Adaptation rate is calculated as the slope of the regression line (bold solid line) between the endpoint error of trial N and the perturbation on trial N-1. Trials following one-dot terminal feedback were separated from those following five-dot terminal feedback and adaptation rate was calculated separately for each feedback condition (one-dot condition shown). Note: the slope of each regression is negative; however, adaptation rates are reported here as positive numbers

Adaptation rate decreased as feedback uncertainty increased (Fig 3). For able-bodied subjects, there was a statistically significant difference between adaptation rates in the one-dot (low feedback uncertainty) and in the five-dot (high feedback uncertainty) conditions ( $p < 0.01$ , one-way ANOVA). For amputee subjects, we observed a difference between feedback uncertainty conditions but were unable to perform statistical analyses because of the small sample size.

Adaptation rates of amputee subjects were similar to those of able-bodied subjects (Fig 3). Within amputee subjects, adaptation rates were slightly lower when using the intact limb, as compared to the residual limb (Fig 3).

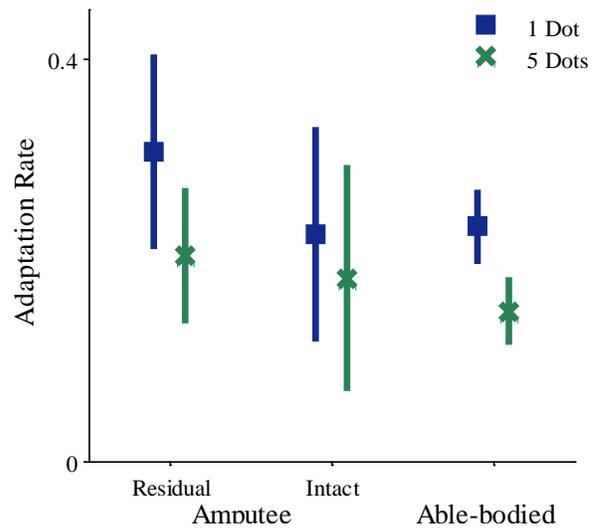


Figure 3: Mean adaptation rates across 3 amputee subjects and 8 able-bodied subjects. Bars represent standard errors of the mean

The mean error of able-bodied subjects was lower than that of amputee subjects (Fig 4). Within amputee subjects, mean error was similar in both residual and intact limb conditions for two subjects, but was higher in the intact limb condition for the third subject.

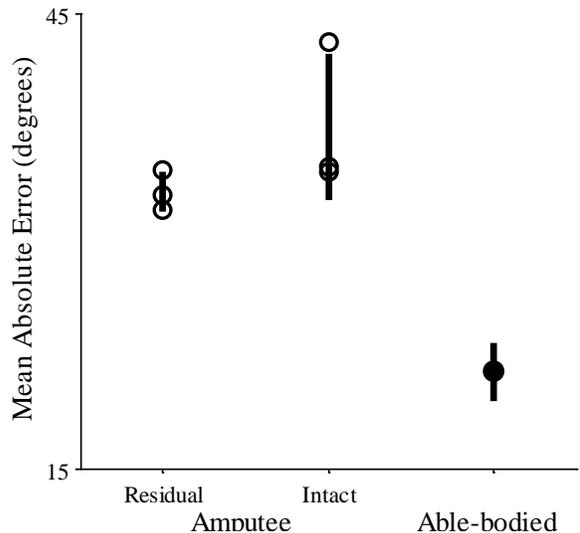


Figure 4: Mean absolute endpoint error across subjects. Open circles indicate the mean error for each amputee subject (for residual or intact limb). The closed circle indicates the mean error across all able-bodied subjects. Bars represent standard errors of the mean

## DISCUSSION

In this work we investigated trial-by-trial adaptation behavior of three different groups using single-DOF EMG control: amputee subjects using their residual limb, amputee subjects using their intact limb, and able-bodied subjects. We found that adaptation rates were influenced by feedback uncertainty, that adaptation was similar across all three groups, and that mean error was higher for amputee subjects.

These results are a strong first step towards characterizing motor adaptation with EMG control, however, many questions remain. Subjects in this study used single degree-of-freedom EMG control, but many powered prostheses feature multi-site control schemes [16], [17], which likely alter motor adaptation behavior. These results were obtained with a virtual environment, whereas physical prosthesis use involves a more complex adaptation task.

Both amputee and able-bodied subjects showed development and adaptation of a simple internal model (Fig 3). Furthermore, adaptation rates decreased in the presence of higher feedback uncertainty (Fig3). These trends confirm that Bayesian models are appropriate for describing the motor control of amputees using EMG control. Previous studies have suggested that prosthesis users are capable of internal model development [18], [19], and our results provide convincing evidence in support of this theory. The similarity of adaptation rates across groups also supports the use of able-bodied subjects as an approximation of prosthetic control in amputees, when necessary.

The mean error of amputee subjects was higher than that of able-bodied subjects, but adaptation rates were similar (Fig 3 and 4). This finding is surprising, since higher errors are presumably linked to higher feedforward uncertainty, which increases adaptation [13]. However, the relationships between mean error, feedforward uncertainty, and adaptation rates may vary for different control scenarios (e.g. amputees vs. able-bodied subjects). The groups of amputee and able-bodied subjects were not age-matched, and there were very likely differences in reaction time and focus. The two groups also used different muscles—amputee subjects used wrist flexors, whereas able-bodied subjects used elbow extensors. Future work will determine if differences in mean error were influenced by experimental factors, or if differences should be attributed to age and ability factors.

These results provide strong motivation for future applications of sensorimotor paradigms to powered prosthesis control. We found that both amputee and able-bodied subjects adapted to perturbations with behavior that confirms Bayesian predictions of sensorimotor adaptation. Further characterization of adaptation will help in designing

sensory feedback strategies that can most effectively reduce uncertainty for amputees using powered prostheses.

## ACKNOWLEDGEMENTS

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## THE DESIGN OF AN ANTHROPOMORPHIC HAND PROSTHESIS WITH AUTOMATED GRIP SELECTION

Ben J. Jones, Peter J. Kyberd

*Institute of Biomedical Engineering, University of New Brunswick, Canada*

### INTRODUCTION

The first myoelectric operated hands were of a pincer type design by Viennatone and Otto Bock [1] (p. 10). These designs were based on that of the Russian Hand [2] from around 1960. The design uses a single motor to actuate two non-articulating fingers and an opposable thumb, linked to the same drive mechanism. The design has been refined over the years but remains largely unchanged today, with products like the System Electric Hand and Sensor Hand Speed being the mainstay of Otto Bock's portfolio. There have been many research prototype hand designs produced in the past. Attempts have been made to improve on functionality, lifelike appearance and natural movement, with designs such as the Belgrade Hand [3], the SVEN and ES hands [4] and the Southampton Hand [5]. These early designs incorporated adaptive mechanisms to allow the fingers to articulate and conform to the shape of objects, to give a more natural and stable grasp. Most were not made commercially available, with the exception of the ES Hand which was available circa 1980, but was deemed to be too heavy and lacking in cosmetic appearance [6]. Surveys into myoelectric hand prosthesis use by Kyberd [7] and Bidiss [8] both concluded that weight was a key factor in user dissatisfaction. Other factors such as, life-like function and appearance, durability, and cost were also deemed important. To this end several adaptive hands have now become commercially available: First with the Touch Bionics iLimb Hand in 2007, then the BeBionic Hand from RSL Steeper, and most recently the Vincent Evolution 2 Hand. The Michelangelo Hand from Otto Bock, though not a truly adaptive hand, as it lacks curling fingers, is also now available. Though improvements in functionality and appearance have been made, weight is still an issue as most of these hands weigh in the region of approximately 500g.

### GOALS

The goal of this research was to design and build a prototype hand capable of multiple grip patterns, which could be automatically selectable without the use of the uninjured side. The aim was to improve on the current adaptive hand designs in terms of functionality, weight, and aesthetic

appearance. It was deemed necessary to make the hand so it could be used with more advanced control strategies such as pattern recognition.

### HAND DESIGN

The hand has an internal aluminium frame for strength, and a plastic hard shell to give lifelike shape and appearance and to protect inner components. The main drive consists of a single brushless d.c. (BLDC) motor with worm and wheel gearing, connected to a ball screw. This drives a whiffletree mechanism, to balance the grip force across all four fingers, to give an adaptive grasp. Each finger is fitted with an actuated locking pin to lock the individual digits in either the open or closed position, thus allowing the whiffletree mechanism to operate on the remaining unlocked digits. The thumb is actuated by two more BLDC motors to give rotation and flexion. The ABS plastic shell casings were manufactured using a Fused Deposition Modelling (FDM) rapid prototype (RP) machine. The prototype is shown in Figure 1.



Figure 1: Prototype Hand

### Whiffletree Design

The whiffletree was designed in a vertical configuration (see Figure 2) to save on longitudinal space. It acts on all four fingers. Balancing springs were added to the whiffletree to alleviate the effect of stiction which can cause the digits to uncurl in an unnatural manner when the hand is opened [5].

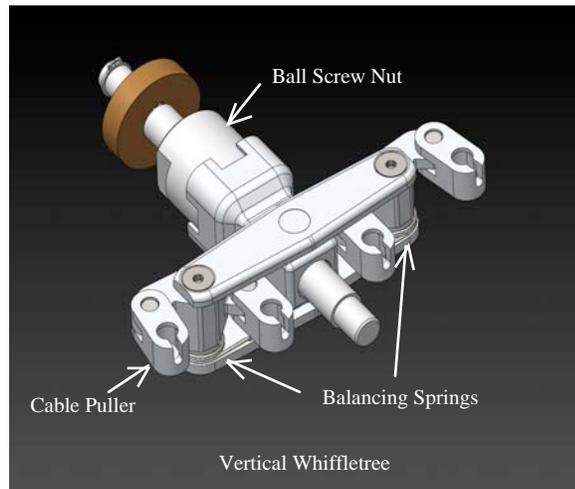


Figure 2: Whiffletree Design

### Digit design

The digits were designed to be easily replaceable. Cables within the digits link the proximal and distal phalanges, to give under-actuated motion at the proximal interphalangeal (PIP) joint in order to give a natural curling motion. The distal interphalangeal (DIP) joints were fixed and an angle of 30 degrees. The cables are able to slide within the digit shell to allow the distal phalange to be pressed closed by an external force applied to the back of the digits, to prevent accidental damage to the digit and cables. The digit locking pins are driven by small piezoelectric motors. The pins engage in to a slot in the back of the rotating part of the metacarpophalangeal (MCP) joint to lock the digit in either the open or closed position (see Figure 3)

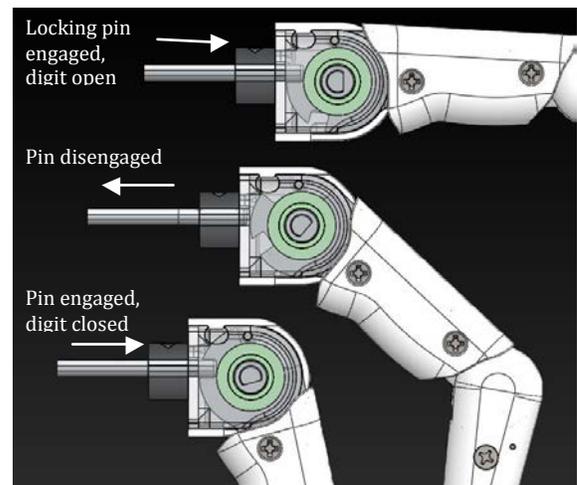


Figure 3: Digit Locking Pins

### Thumb

The Thumb is able to rotate and flex at the Trapezo-metacarpal (TM) joint, allowing the thumb to be moved into positions opposing the digits, to form useful grip patterns. The mechanism uses worm and wheel gears, combined with a sensorless BLDC motor and planetary gearbox.

### Sensors

Position sensors were fitted to each of the MCP joints of the fingers and to the flexion and rotation mechanisms of the thumb. A longitudinal soft pot potentiometer was fitted to the inside the palm, with a wiper pin attached to the ball screw nut, to measure the longitudinal position of the main drive. Force sensitive resistors (FSRs) were placed in the finger tip of each digit, under a silicone tip pad, to give a relative measure of the force applied at the finger tip. Two more FSRs were placed in the thumb tip. A strain gauge was fitted to the internal frame of the thumb to detect forces applied to the thumb at any location.

## COSMESIS DESIGN

### Cosmesis

A cosmetic covering was produced using a rapid prototyped mould. Silicone was injected in to the mould to produce a 1.75mm thick covering in the shape of the hand design. This covering proved to be too thick in the joint regions and substantially limited the movement of the joints, decreasing the range of motion and the achievable grip force. For the purpose of demonstrating the prototype's ability to form grip patterns, the cosmesis was cut at the joints and material removed to allow efficient movement of the joints. A new concept cover design was made using corrugations to allow the joints to flex. This concept has not been implemented into a new cover at this time.

## CONTROLLER

### Controller Hardware

The hand was controlled using a dsPIC33FJ256GP710 microcontroller mounted on an Explorer 16 development board. Two sensorless (BLDC) controllers were used to run the two motors used for thumb rotation and flexion. A BLDC controller with hall sensors was used for the main motor drive. Sensor readings were interfaced to the analogue to digital modules of the microcontroller. The digit lock piezoelectric motors were controlled via dedicated controller integrated circuits, interfaced to the main microcontroller via an I<sup>2</sup>C bus. The control hardware was mounted externally to the hand mechanism and connected via a bundled cable tether.

### Control Inputs

A grip selection switch was interfaced to the controller to allow the user to select individual grip patterns. A push button was added to enable the grip pattern once the pattern had been selected. This method of grip selection was used for development. The intent is for grip patterns to be ultimately selected by pattern recognition software. Two FSRs were initially used to simulate the myoelectric control signals to open and close the hand, with the speed being proportional to the force applied. Two Otto Bock electrodes were later used for the control signals, once the hand had been fitted to a bypass socket for testing.

### Controller Software

The microcontroller was programmed using MPLAB Integrated Development Environment (IDE), with the C30 compiler. Once the user has selected a grip pattern using the grip selection switch and the grip enable button, the controller moves the thumb and main drives to position the hand in a predefined neutral position, before then moving to the new desired grip pattern position. Once the desired grip pattern has been reached the hand switches from position control to a proportional speed control algorithm to open and close the hand in that grip pattern, using either the FSR or Otto Bock electrodes as the control input.

## RESULTS

### Grip Patterns Achieved

The following grip patterns can be selected: Power (cylindrical and spherical), Tripod (remaining fingers open), Tripod (remaining fingers closed), Tip (remaining fingers open), Tip (remaining fingers closed), Lateral, Diagonal Volar, Hook, Extension and Adduction. The hand can also form useful postures such as a neutral thumb position for cosmesis removal, a flat palm, and a finger point posture.

### Grip Force Measurement

A Jamar Plus + hand dynamometer and digital pinch gauge were used to measure grip forces. An adapted grip handle was made from a straight ABS plastic bar to replace the curved grip handle of the standard dynamometer. The prototype could not fit all four fingers into the curve of the standard grip handle because the fingers have a fixed adduction angle, which does not allow them to move as closely together as a natural hand. The hand controller uses current measurement for each of the motors to prevent the high current at stall from damaging the motors. The present over current algorithm limits the grip force slightly at stall, as can be seen by the two readings for the power grip in Table 1. This could be optimised to reduce the loss of grip force while maintaining sufficient over current protection.

Table 1: Measured Grip Forces

Grip Pattern	Measured Force (N)
	<i>Hand Prototype</i>
Power (With over current protection)	50.7
Power (No over current protection)	54.6
Lateral	7.5
Tripod (Digits Open)	18.3
Tip (Digits Open)	10.2
Tripod (Digits Closed)	18.5
Tip (Digits Closed)	10.8

The prototype falls short of the power of an actual hand [9], but meets the distributed grip forces required for grasping most objects [10]. The adaptive nature of the design allows a firm grasp of most of the objects used in testing. The lateral grasp however, proved to be below expectations. The BLDC motor used was in theory able to exert a grip force of 16N at the thumb tip. Due to the method of sensorless commutation, the motor controller was however inefficient at delivering the full expected stall torque. The controller used relies on the measured back emf of the motor to control the phase of excitation of the coils. When the motor approaches stall, the controller is unable to read sufficient back emf and so shuts off power to the motor, reducing the stall torque. For the main drive motor, where the phase of commutation is measured with hall sensors, the motor controller is able to deliver full power at a much slower speed and consequently delivers a much greater torque just before the motor stalls.

### SHAP Test

Initial trials with A Southampton Hand Assessment Procedure (SHAP) test kit [11] were carried out to evaluate the prototype hand's ability to grasp everyday objects. The hand was mounted onto a bypass socket which allowed an able bodied user to operate the arm using Otto Bock electrodes held in position by a silicone and fabric band (See Figure 4).



Figure 4: Prototype hand fitted to bypass socket

The hand was able to grasp most of the test objects satisfactorily with the exception of the heavy lateral object, where a tripod grip had to be used in place of the lateral grip due to the poor grip strength in the lateral grip pattern. Other task which could not be completed to the test specifications were: the ‘Tray’ task due to the external cable tether interfering with the test kit case, and the jug pour and carton pour, due to the risk of damaging the exposed electronic circuitry of the controller by an unwanted spillage. A direct comparison could not therefore be made with the test results of other hand prostheses at this stage.

#### Cloths Pin Test

A clothes pin relocation task [12] was also carried out. Three red clothes pins from an ‘Original Rolyan Graded Pinch Exerciser’ were moved from a horizontal bar to a vertical bar and timed. Moving in the reverse direction was also timed. The test was completed 6 times and an average score for moving the 3 clothes pins up and down was recorded.

Table 2: Cloths Pin Test Results

Trial	Task Time (Sec)	
	<i>Move 3 pins up</i>	<i>Move 3 pins down</i>
1	12.03	15.29
2	11.68	24.49
3	13.19	12.18
4	10.72	12.53
5	10.82	10.91
6	10.94	11.40
Average	11.56	14.47

#### FURTHER WORK

In order to properly validate the design, an improved and robust prototype would need to be developed with integrated electronics, which could be fitted to users for extensive testing. Areas for improvement are: the thumb grip force, the mounting of the digit locking pin actuators to prevent lateral loading on the motors causing damage, wiring across joints

for reliability, weight optimisation to improve usability, noise reduction in the ball screw bearings and palm shell materials.

#### ACKNOWLEDGEMENTS

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## A THEORETICAL ANALYSIS OF PROSTHETIC HAND ASSESSMENT TOOLS BASED ON DIFFICULTY OF PERFORMING THEIR TASKS.

Isamu Kajitani<sup>1</sup> and Jumpei Oba<sup>2</sup>

*1 National Institute of Advanced Industrial Science and Technology*

*2 Kobe Gakuin University, Faculty of Rehabilitation*

### INTRODUCTION

This report introduces a theoretical analysis of prosthetic hand assessment tools. We focus on difficulty of performing tasks in the assessment tools based on the idea of Fitts' law [1], which is a model of human motor system and defines index of difficulty (ID). The ID is a measurement of theoretical difficulty of performing a reaching task.

Difficulty of the task is also affected by relationship between a prosthetic hand and its user, so we define two types of relationships. Basic hand motions which are required to perform the assessment tasks are classified into three groups.

Positive (increase difficulty) and negative (decrease difficulty) factors to the difficulty of controlling prosthetic hand are discussed. Suggestions from those discussions include that therapist can control the level of difficulty of assessment tools by combining those positive or negative factors. For engineers, a newly developed hand can be evaluated in terms of decreasing or increasing levels of difficulty.

### METHODS

#### Index of difficulty (ID)

The Fitts' law was proposed to discuss speed and accuracy for reaching movement tasks [1], and has been applied to various fields, such as human computer interface [2] and controlling of robot manipulators [3]. In Fitts' law, ID is basically expressed as equation (1), where  $A$  is amplitude of movement and  $W$  is width or tolerance of a target region (as shown in Figure 1), which means that larger movement or smaller target area lead to difficult tasks.

$$ID = \log_2 \quad (1)$$

There are varieties of equations for various applications of the Fitts' law, angular movements, such as wrist rotation, are also expressed as similar equation (2). [4]

$$ID = \log_2 (2\theta_A/\theta_w) \quad (2)$$

where  $\theta_A$  is movement amplitude of the angular motion,  $\theta_w$  is target width.

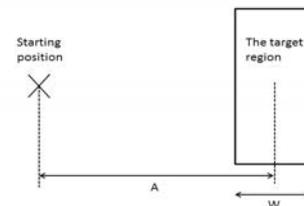


Figure 1: The Fitts' law experimental setting. [1]

When we apply this relationship to the prosthetic hand control, reaching to small target objects or pinching fragile object require precise hand control, and they have much difficulty.

The ID is also applied to pointing tasks on a computer monitor, whose edges are easy to reach with a mouse pointing device; because the mouse pointer stops on the edge, so we can make a long quick movement without the fear of overshooting the target. In this case, the target regions on the edge are considered as having infinite width [5]. In prosthetic hand case, this indicates that pinching non-fragile objects is similar to pointing objects on the edge of computer monitor, because hand closing motions ought to stop when fingers attach to the hard object.

#### Human machine relationship

Two types of human machine relationships are defined in this report, which affect difficulty of prosthetic hand control.

**Direct:** When you use a forearm prosthetic hand, the hand location directly follows motion of the forearm, because the hand is attached to the forearm by using the prosthetic socket. This means that hand location and forearm motion has very close relationship. Location of the hand is also changed by movement of other body parts, such as compensatory movements of a body trunk or lower extremity movements. Manipulation of a body-powered hook has this relationship. Those types of close and direct relationships are referred to as "direct" in this report.

Indirect: In the case of a myoelectric prosthetic hand, hand open-close function is operated by using forearm muscle activities through myoelectric signals. This means that their relationships are not so close to each other, compared to the “direct” relationship. We call this relationship as “indirect”.

Assessment tools used in prosthetic hand training

There are various assessment tools, which are applied in the prosthetic hand trainings, such as an APMC (Assessment of Capacity for Myoelectric Control) [6], a SHAP (Southampton Hand Assessment Procedure) [7] and a BBT (Box and Block Test) [8]. Most of their tasks are classified into two groups, (1) relocation of abstract target objects, which we focus on in this report, and (2) various ADL activities.

The relocation tasks consist of reaching and grasp-release motions. The reaching motions are divided into two spatial hand movements of moving hand location and changing hand direction. The graspe-release motions are actually controlling of hand open width. Those hand movements have different level of difficulty which relate with positive and negative factors as shown in the following section.

**RESULTS**

Table 1 shows classified positive and negative factors for the difficulty of hand motions listed in the previous section, based on the Fitts’ law. Controlling hand open width relates to size and fragility of the target object. Small object requires much attention to deal with, so it can be positive factor. Fragile object require precise operation of grasping, whereas non-fragile object can be regarded as it is placed on the edge of the moving range, where the target width set to be infinite.

Because the prosthetic hand has lower degree of freedom compared to a sound human hand, precise control of hand direction is required when the target object is directional or has complicate shape, such as a cube or other ADL objects. Arraignment of the target objects or assigned target placement direction also require higher attention to the hand direction control.

Reaching to or relocation of the target object in a long distance has higher difficulty. When we have obstacles in reaching or relocation path, it can be a positive factor to the difficulty of moving hand location.

Table 2 shows positive and negative factors to the level of difficulty for human machine relationship and some assessment settings. The “direct” relationship is negative factors to the difficulty of operation, whereas the “indirect”, by contrast, is regarded as a positive factor. Having a time restriction and complex task are also positive factors.

Some of the assessment tools have sets of repetitive tasks with uniform target objects, such as same size, shape or directions, in one set of task. In this case, an amputee can easily get the technique of handling the objects, therefore uniform target objects can be negative factors of difficulty. In the case of repetition of non-fragile uniform target object, what the amputee learns might be limited to appropriate control timing for appropriate open width.

Table 3-5 show examples of positive and negative factors in the BBT and the SHAP (spherical and tripod objects). We can easily compare the differences between the BBT and the SHAP, such as the number of repetition or the direction of object placement. Difference with target object shapes (spherical or tripod objects) are also clear. The SHAP tripod object test is difficult in the hand direction control, compared to the SHAP spherical object test.

Table 1: Positive and negative factors for basic hand movements in the assessment tools.

Basic hand movements in assessment tools.	Property of target objects.	Positive factor	Negative factor
Controlling hand open width	Size	Small	Large
	Fragility	Fragile	Non-fragile
Changing hand direction	Shape	Directional / Complicate	Non-directional / Simple
	Target placement	Assigned / Arraignment	Arbitrary
Moving hand location	Distance	Long	Short
	Obstacle	Yes	No

Table 2: Positive and negative factors to the level of difficulty for assessment settings.

Assessment settings	Positive factor	Negative factor
Human machine relationship	Indirect	Direct
Time restriction	Yes	No
Complexity of the task	Complicate	Simple
Variation of target objects	Various	Uniform

**DISCUSSIONS**

This report introduces the positive and negative factors to the difficulty of prosthetic hand control.

We can change the difficulty of the assessment task to fit the level of the amputee, by adding the positive factors to

increase difficulty, and vice versa. For example, when we want to focus on the training of hand open close operation, it is better for us to use non-directional objects, like a spherical object, because directional objects, such as cubic, require hand direction control, so they need to be paid attention to both of hand open-close and hand direction control.

Engineers can use those positive and negative factors in explaining their developed prosthetic hand. For example, multi grip-pattern terminal device can reduce the difficulty of hand direction control or a flex fingers can reduce difficulty of grasping fragile objects. In developing training simulator, we can utilize those factors as well.

Table 3: Positive and negative factors: BBT

<b>BBT (Box and Block Test)</b>		
Controlling hand open width	Positive factors	-
	Negative factors	Non-fragile target objects Repetitive tasks with uniform size objects.
Changing hand direction	Positive factors	Directional target objects Repetitive tasks with arbitrary object direction
	Negative factors	Arbitrary target placement
Moving hand location	Positive factors	Obstacle in the relocation path
	Negative factors	-

Table 4: Positive and negative factors: SHAP spherical object

<b>SHAP Spherical object</b>		
Controlling hand open width	Positive factors	-
	Negative factors	Non-fragile target objects
Changing hand direction	Positive factors	Assigned target object placement
	Negative factors	Non-Directional target objects
Moving hand location	Positive factors	Obstacle in the relocation path
	Negative factors	-

Table 5: Positive and negative factors: SHAP tripod object

<b>SHAP tripod object</b>		
Controlling hand open width	Positive factors	-
	Negative factors	Non-fragile target objects
Changing hand direction	Positive factors	Directional target object Assigned target object placement
	Negative factors	-
Moving hand location	Positive factors	-
	Negative factors	-

This report is limited to the theoretical and qualitative analysis of positive or negative factors to the difficulty of performing assessment tasks. Quantitative analyses are required in future works.

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## DIFFICULTY OF PERFORMING ACTIVITIES OF DAILY LIVING WITH THE MICHELANGELO® MULTIGRIP HAND AND TRADITIONAL MYOELECTRIC HANDS

Andreas Kannenberg, Eva Proebsting  
*Otto Bock HealthCare*

### ABSTRACT

#### Introduction:

Most myoelectric prosthetic hands offer only the tripod grip to allow for pinch grasping of objects. This is a dramatic limitation of function as compared to the sound human hand. The purpose of this study was to investigate whether the Michelangelo® multigrip hand (Otto Bock HealthCare, Duderstadt) improves function and reduces perceived difficulty of performing activities of daily living (ADL) in comparison to single grip myoelectric hands.

#### Methods:

16 experienced users of regular transradial prostheses gave informed consent and participated in this cross-over observational study. The validated Orthotics and Prosthetics User Survey - Upper Extremity Functional Status (OPUS-UEFS) was used as the primary outcome measure in its original and revised version with 23 and 19 ADLs, respectively (2). As secondary outcome measure the same 23 and 19 ADLs were rated using the scoring system of the Prosthetic Upper Extremity Functional Index (PUFI) (3) for the way of doing an activity (function) and usefulness of the prosthesis. Patients completed the OPUS-UEFS and PUFI at baseline for their existing device as well as after a minimum of 4 weeks of use of the Michelangelo hand. Statistical analysis was conducted using the Wilcoxon signed rank test with a power of 80%. Differences with p-values.

## DEALING WITH CHANGING CONTEXTS IN MYOELECTRIC CONTROL

Anna J. Koop<sup>1</sup>, Alexandra Kearney<sup>1,2</sup>, Michael Bowling<sup>1</sup>, Patrick M. Pilarski<sup>2</sup>

<sup>1</sup>*Department of Computing Science, University of Alberta, Edmonton, AB, Canada*

<sup>2</sup>*Division of Physical Medicine and Rehabilitation, University of Alberta, Edmonton, AB, Canada*

### ABSTRACT

Myoelectric prostheses approximate the motion and flexibility of biological limbs, especially when compared to their mechanical counter-parts. Machine learning enhances the functionality of these devices; however, in an ever-changing environment, the effectiveness of conventional approaches is impeded. We introduce Partition Tree Learning—a method for learning predictions in an ongoing fashion whilst being able to identify and adapt to new contexts automatically. We compare the performance of PTL to that of a stochastic gradient descent learner on a stream of data from a participant actuating a myoelectrically controlled robot arm. In a consistent context both learners' predictions are comparable. After a context switch, PTL is able to adapt to the change and outperform the gradient descent learner. These preliminary results indicate that PTL may effectively deal with change in real-world prosthetic use, lending its ability to learn over varying situations to the constantly changing environment of powered prosthetics.

### INTRODUCTION

For ongoing, every-day use, it would be ideal for a prosthetic device to adapt to the unavoidable changes in the user and environment [1-4]. These changes take many forms: muscle fatigue degrading the control signal; shifts in the position of the residual limb while performing a single pattern of movement; the changing task profiles as a user moves from driving a car to putting away groceries. These changes, operating at different semantic and temporal scales, create challenges for control-related machine learning techniques that have been developed mainly for stationary environments—where the signals are coming from the same distribution and the learner aims for a universal best-fit solution [5]. In particular, changing circumstances are known to be problematic for pattern recognition in myoelectric control [1-3]. A solution that is learned a priori may be robust in general but unable to adapt to the particular circumstances of the moment. On the other hand, an ongoing learner can adapt as needed but may not provide the stability necessary for both user acceptance and effective control [2, 4, 6, 7].

Recent work has focused on using techniques to adapt across several of the many contexts that arise in myoelectric control. Sensinger et al. compared supervised and unsupervised adaptive approaches to improve pattern recognition for a single user across multiple sessions, where previous work had looked only at single-session performance [1]. Tommasi et al. were concerned with the transition from stable pre-trained models to customization for a particular user, and used adaptive combinations of the pre-trained models to reduce training time [4]. Prior work by our group has explored the use of real-time machine learning to adapt and improve both prediction and control policies during ongoing use by a single user [8-10]. Together these approaches to adaptive learning form a solid basis to approach problems inherent in building myoelectric control schemes for specific users and their varied patterns of use.

In this paper we are concerned with the changing contexts that arise from task switches during persistent use of an assistive device. Specifically, we present initial findings on one method by which a system learns automatically during ongoing multi-context use. Our meta-learning approach, termed Partition Tree Learning (PTL), is able to adapt to changing contexts without requiring pre-processing or explicit context identification. PTL therefore promises to complement existing learning methods and further expand the adaptability, robustness, and functionality of myoelectric human-machine interfaces.

### PARTITION TREE LEARNING

Many machine-learning algorithms are developed primarily for use in stationary environments: expecting that either the task of interest does not change, or the state representation (that is, the features the learner uses to predict) is detailed enough that every context is uniquely identified. This is a necessary simplification that is sometimes sufficient, however, the many contexts a prosthetic user naturally encounters are diverse and too complex to represent in a single computationally efficient predictor. Thus, we are interested in studying online or continual learning systems where the learner is able to adapt to the specific context.

Moreover, we aim to create a learning system that adapts to and identifies changing contexts automatically.

To that end, we are developing Partition-Tree Learning (PTL)—a meta-learning algorithm. Through PTL we can adapt existing learning algorithms to improve their performance in non-stationary environments, such as those encountered in myoelectric control. PTL increases the base learner’s accuracy without drastically increasing their computational complexity. This is a direct extension of the Partition-Tree Weighting algorithm for probabilistic modeling by Veness et al. [11], which provides theoretical guarantees on the performance for minimal computational and memory costs.

A key aspect of PTL is that it coordinates learners across different time scales. These learners are designed to converge on the best prediction over the long term and are unable to adapt to local context. PTL allows these stationary learners to be used effectively in non-stationary environments by coordinating their predictions and limiting the data over which they learn.

PTL uses a binary partition tree (visualized in Fig. 1) to split the data into discrete binary segments. Each node of the tree represents a distinct learner over a specific segment. The root learner, sitting at the top of the tree, operates on all the data (up to  $2^d$  time-steps): it behaves identically to the base learner. The leaf nodes operate over a single time-step, and therefore their predictions are mainly determined by the initial settings of the learning algorithm. On the levels in between, each node at depth  $i$  operates on  $2^i$  time-steps.

Because PTL operates online it never has to store or compute the entire tree at once. Instead, it keeps a list of  $d$  learners, and up to  $d$  statistics summarizing the error for each completed subtree. At each time step PTL updates each of the  $d$  learners: from the learner at the leaf node monitoring only the current time-step, through all intermediate learners with their longer segments, ending with the root node. If this is the first time-step of a new partition for any depth (as is always the case at the leaf nodes), it will create a new learner at that node and update its own records for the newly completed subtree.

When making a prediction, PTL consults each of the  $d$  currently active learners and reports the weighted combination of their predictions. Each prediction is weighted according to a prior that considers long-term learners more likely than short-term, the statistics for the relevant completed subtrees, and the performance of the particular learner over its particular segment. This allows PTL to adjust automatically to changes in the environment: when the environment changes such that the shorter-term learners predict better than

the long-term learner, the weight shifts to favour predictions from the short-term learner. When the environment is stable, the weight is mainly on the long-term learner and PTL makes predictions accordingly.

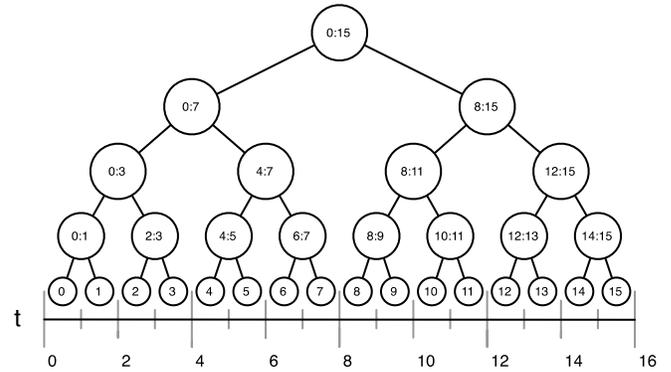


Figure 1: A full binary partition tree, showing all the binary intervals over the time 0...16.

## METHODS

Interactive data was gathered from multiple able-bodied subjects—participants without amputations. We used a myoelectrically controlled robot arm which replicated the functionality of a commercial prosthetic device. Informed subject consent was acquired as per ethics approval by the University of Alberta Health Research Ethics Board. The experiment was composed of a simplified conventional prosthetic training task: a square movement. Each subject moved in the square pattern for 8 minutes.

For the duration of the trial the subjects moved the arm in a square pattern, where the learner predicted the joint activity of the robot’s arm. If the elbow joint was in motion on any given time-step, then the value of the joint activity would be one, when stationary, the value would be zero.

For comparing the learners we used a 2048-timestep-length segment from the middle of the trial. To create a clear domain switch, we presented this time series twice to the learning agents: first where the signal to be predicted was the sum of the joint activation signal over a horizon of 50 time-steps, and second where the target prediction was the sum of the negation (so that if the joint was moving the one time-step signal would be -1, otherwise zero).

The base learner was a simple stochastic gradient-descent learner that used tile coding over the trace signals on the activation of the two joints together with the position data on each joint. These settings are similar to those used in Pilarski et al. [8–10], but rather than using a discounted sum of future signals we used a fixed-horizon sum for the

gradient descent learner. The weights were initialized to 0 and the learning rate  $\alpha$  was set to 0.05, which was the best on the square task.

PTL used the same learner and parameters, with a complete reset at the segment boundaries: each new learner was re-instantiated with its weights starting at 0. The performance of the base learner was by measuring the cumulative prediction error. We repeated trials across several different users and the results were consistent: one is singled out for discussion in detail.

## RESULTS & DISCUSSION

The cumulative error is shown in Figure 3. Before the task change, the two algorithms made near-identical predictions for all parameter settings tested. After around 300 time steps the rate of error accumulation has levelled off. This can also be seen in the profile of the predictions, shown in Figure 2, where both lines overlap and closely track the target before the switch. After the switch point, they both predict as before but PTL has lower peaks, more quickly compensating for the change. The long-term benefit from the predictions immediately following the switch can be seen in the cumulative error, where there is clear separation between the cumulative errors.

At the switch point, there is a sharp penalty visible in both the cumulative error graph and the prediction graph. The error rate re-stabilizes but takes slightly longer for the gradient descent learner than the initial learning phase, approximately 500 time-steps compared to 300. After this stabilization period the predictions of both algorithms again overlap.

The weight visualization in Figure 2 provides insight into how PTL handles the context switch. Before the switch, PTL places the most weight on the long-term learners, shown. There is a deviation from this around the 1024 time-step mark. At that point, and more consistently after the switch, the weight is distributed across more learners. The medium-term learners are given weights equal to the long-term learner. These shorter learners, not being misled by previous experience in the pre-switch domain, are able to adapt faster to the new signal. As the task continues without another switch, the long-term learner eventually catches up, and the weight again shifts to favour the long-term learner. Introducing another or more frequent switches will shift the weight more towards the short-term learners.

In this preliminary work, we used artificially imposed switch points to understand how the PTL algorithm behaves during contextual shifts. As part of our ongoing work, we are investigating the performance of PTL in a variety of tasks where switching boundaries were naturally occurring in the data stream.

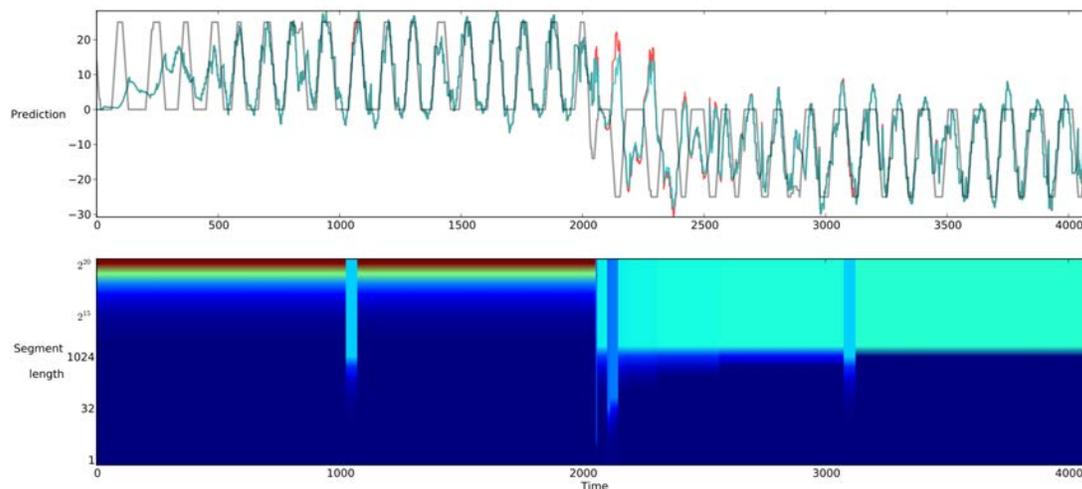


Figure 2: Predictions made by the gradient descent learner (red) and PTL (blue) compared to the true target (grey). The lower figure visualizes the weight on each of the segment lengths over time.

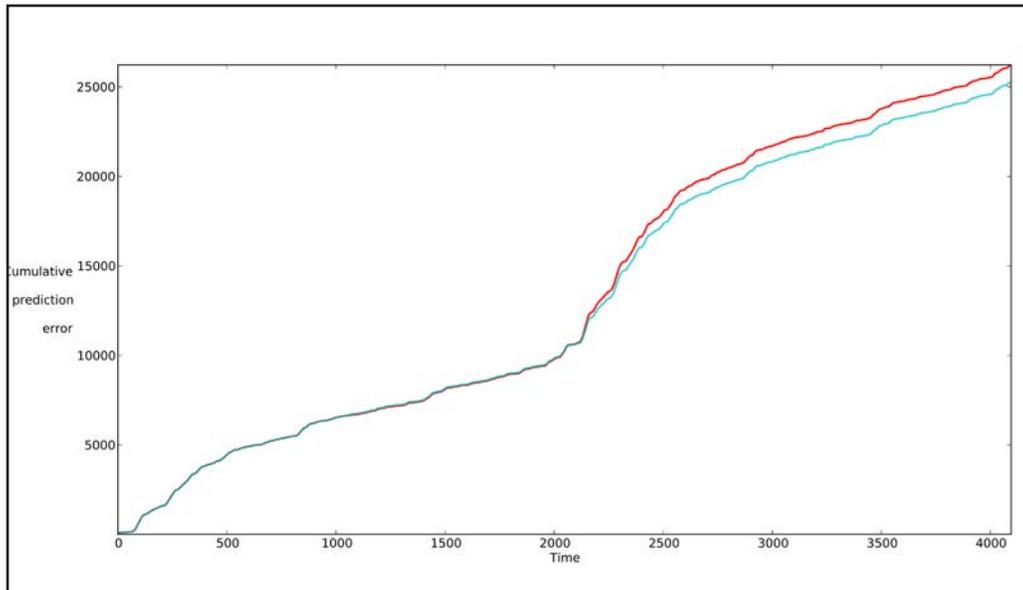


Figure 3: Cumulative error for the gradient descent learner (red) and PTL (blue) over time

## CONCLUSION

In this work we introduced Partition Tree Learning—a method for learning predictive information during ongoing myoelectric control. This approach helps to maintain consistency while still providing the flexibility to adapt to changes in the user and their situation. Our results suggest that PTL is a beneficial way to learn and adapt during long term, contextually varying prosthesis use. PTL is capable of adapting to changing situations without requiring explicit contextual identification. As PTL is a meta-learning approach, it is also complementary to many existing control optimization techniques that function both prior to and during the control of a myoelectric device. PTL therefore promises a new approach to enhancing the versatility and utility of future myoelectric devices.

## ACKNOWLEDGEMENTS

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## FIRST CLINICAL FITTING OF AN INDIVIDUAL AFTER BILATERAL TMR WITH INTUITIVE PATTERN RECOGNITION CONTROL

Todd Kuiken MD, PhD<sup>1,2</sup>, Kristi Turner OTR/L<sup>1</sup>, Nicole Soltys CP<sup>1</sup>, Gregory Dumanian, MD<sup>2</sup>

<sup>1</sup>Rehabilitation Institute of Chicago, Chicago, Illinois, USA

<sup>2</sup>Northwestern University, Chicago, Illinois, USA

### INTRODUCTION

Targeted Muscle Reinnervation (TMR), a surgical procedure to enhance the control of upper limb prostheses [1] has been available for just over a decade, and has been performed in more than 100 individuals. Pattern recognition-based myoelectric control, which has been the topic of research for decades, recently became commercially available. These two technologies have great synergy: TMR provides increased access to motor control data, and intuitive pattern recognition (IPR) algorithms decode this information to enable intuitive control of many degrees of freedom (DOFs).

For high-level bilateral amputees, TMR has previously been performed on only one limb; patients use a myoelectric, motorized prosthesis on the TMR side and a body-powered device on the other. In 2011 we were presented with a unique patient who required extensive bilateral surgical revisions, which made bilateral TMR appropriate. He was subsequently successfully fit with bilateral 'Complete Control' IPR control systems (Coapt, Inc.). This case study describes the surgery, which took place at Northwestern Memorial Hospital (NMH), Chicago, and the prosthetic fittings and occupational therapy (OT) sessions, which took place at the Rehabilitation Institute of Chicago (RIC).

### CASE STUDY

A 43-year-old male lineman sustained a severe electrical burn injury in September of 2010. His injuries required a left-side amputation at the shoulder- disarticulation level and right-side amputation at the transhumeral level. While in acute care, the patient was placed in a medically induced coma for approximately three weeks and underwent 14 surgeries, including multiple wound debridements and skin grafts. Once medically stable, the patient was admitted to an inpatient hospital and discharged home in November, 2010.

In May 2011, the patient was evaluated for TMR surgery at the RIC and NMH. He had extensive heterotopic ossification (HO) of his left scapula that precluded lying on his back and interfered with sleep. On the right side he had

lost over 90% of his biceps and had a split thickness skin graft over the remaining muscle and much of his humerus.

In January 2012, revision surgery was performed on the patient's left side. The HO over the scapula was removed and the latissimus dorsi was transferred to provide soft tissue coverage over the scapula. TMR surgery involved transfer of the musculocutaneous and median nerves to the upper and lower pectoralis major, respectively; transfer of the ulnar nerve to pectoralis minor, which was moved laterally out from under the pectoralis major; and coaptation of the radial nerve to the long thoracic nerve to reinnervate the serratus anterior.

In May 2012, TMR surgery and soft tissue reconstruction was performed on the patient's right side (Figure 1). Since he did not have biceps muscle available, the gracilis muscle was surgically transferred from his right leg as a free flap. The musculocutaneous and median nerves were transferred to proximal and distal motor points, respectively, on the gracilis muscle, which was covered with a split-thickness skin graft. The distal radial nerve was transferred to the denervated lateral head of the triceps.

The patient was fit with his left-side, direct (conventionally) controlled prosthesis in August 2012, at which time he began occupational therapy training with the device. His prescribed device included a locking shoulder (LTI) with a rocker switch (Ottobock) to lock/unlock the shoulder; a Boston Digital Arm™ (LTI), and a wrist rotator, hand, and electric terminal device (ETD) (Motion Control). He used 4 independent myoelectric sites to control elbow flexion, elbow extension, hand open, and hand close. An FSR was used to switch between hand and wrist control.



Figure 1: The patient before revision surgery (left) and after both TMR surgeries and tissue reconstruction surgery (right)

Because the patient did not live in the Chicago area, therapy was limited to three, one-week visits that involved approximately 90 minutes of daily occupational therapy, follow up medical appointments, and prosthetic modifications. Therapy focused on myosite strengthening, development of a home exercise program, providing education on prosthesis functions, improving prosthetic control, repositioning, and grasp/release in various planes of movement. The patient identified functional goals including drinking from a cup or bottle, feeding himself, and performing yard work. Functional ability was limited by poor socket fit, difficulty in obtaining isolated EMG signals, muscle fatigue, and the need to frequently modify gains and thresholds. The patient's definitive left-side prosthesis was delivered in October 2012. At discharge from OT, the patient demonstrated isolated control of all left upper extremity prosthesis functions in 7 out of 10 trials while standing. He was able to pick up and release light objects on a table; however, he was unable to feed himself or drink from a cup or bottle.

### INITIAL TRAINING FOR INTUITIVE PATTERN RECOGNITION CONTROL

In September 2012, approximately 4 months after TMR surgery on the patient's right side, the patient was introduced to IPR control, on his right side, at the RIC through custom software (CAPS) [1] and a Virtual Reality (VR) system. Training included establishing repeatable, unique movements and practice in a VR environment. [2] The patient was introduced to one DOF with pattern recognition control (i.e. elbow flex/extend); this was increased to three DOFs (elbow flex/extend, hand open/close, wrist supination/pronation) as control and understanding improved. To avoid confusing the patient, initial training in use of the research IPR system on the right side was performed after the patient completed his therapy for direct control of his left-side prosthesis. The patient thus had approximately a total of 5 hours of VR training using the research IPR system at the end of each of his final two weeks of direct control training. Although the control sites created by TMR generated fairly small signals,

the research IPR controller successfully distinguished between different movements, and the patient was able to control all three virtual DOFs. The patient was given a home exercise program to perform to assist in establishing unique, repeatable movements that could be used to operate each DOF.

In October 2012 the patient was fitted with his right-side prosthesis. This device included a custom silicone interface with LTI domes, an embedded research IPR controller, (Figure 2) a laminated frame and chest strap for suspension, a Boston Digital Arm (LTI), and a wrist rotator, hand, and ETD (Motion Control). This prosthesis was connected to his left-side, direct control prosthesis such that both devices would be worn together. The patient received approximately 5 hours of pattern recognition training for the right side, spread throughout one week.



Figure 2: Top: Experimental APR system embedded in socket. Bottom: Final laminated system with commercial Complete Control APR controller

The focus of therapy was again to identify unique and repeatable muscle movements with which to control each DOF. In addition, he learned the timing and movement sequence required for prosthesis-guided calibration [2] (Figure 2). Using the research IPR controller, the patient was able to perform individual movements of his right-side prosthesis with minimal or no unintended movements for all DOFs. However, practice with the research IPR system was limited due to muscle fatigue and the continued problem of poor socket fit and consequent inconsistent electrode contact.

In February 2013, the patient was sent home with an IRB-approved research prototype of the IPR controller embedded in his right-side prosthesis. He continued to use direct control for his left-side device. Weekly phone calls with the OT involved discussions on how to achieve repeatable and unique movements to improve control reliability and reduce unintended movements of the prosthesis. Photographs were emailed to the patient to illustrate modifications to the home exercise program as needed. During the initial home trial with the research IPR prosthesis, the patient focused on improving his control of the prosthetic functions and building up tolerance to the weight of the prosthesis. He was unable to perform functional tasks due to inadequate socket suspension (i.e., the weight of the prosthesis continued to pull the socket distally) and inconsistent electrode contact. The patient reported wearing the prostheses 3-4 hours a day, every other day. He indicated that inconsistent use of the prostheses was due to his personal schedule, muscle fatigue, and poor socket fit. Between February 2013 and November 2013 when he received the commercial IPR controller, the patient returned to the RIC two times, in which right-side socket fit issues were repeatedly addressed. During the last visit, socket fit issues for the right-side transhumeral prosthesis were resolved by changing from the custom silicone interface to a roll-on gel liner with Motion Control dome electrodes. With this new interface, the patient began to report performance of functional activities to the therapist. Using the research IPR system, the patient was able to reach and grasp items at various levels, including getting mail from the mailbox, pushing a wheel barrel, and picking up sticks from the ground. He was able to perform yard work tasks (an initial goal identified in August 2012) using the research IPR-controlled prosthesis, something he was not able to achieve using direct control. The patient spent a total of 3 months at home with this research IPR prosthesis before he received the commercial system.

#### FITTING OF COMMERCIAL INTUITIVE PATTERN RECOGNITION CONTROL SYSTEM

In November of 2013, both of the patient's prostheses were fitted with Coapt Complete Controllers—commercially available IPR systems. The research IPR controller was

removed from the right transhumeral prosthesis and replaced with the commercial system and the left shoulder disarticulation prosthesis was retrofitted with a second commercial system. The patient received occupational therapy each day for four days, during which he used IPR control on his left side for the first time. Because the socket fit issues had been addressed, the patient was able to begin therapy immediately. On the first day that the patient used both prostheses with the commercial IPR, he was able to perform isolated movements of all three DOFs on each side. The patient had not been able to utilize the left direct control prosthesis well enough to perform functional activities, such as drinking, due to lack of elbow control. Therefore, he required cueing to utilize the elbow functions during tasks, for improved body mechanics, rather than bending forward. With the improved, intuitive control provided by the commercial IPR system, the patient was easily able to control elbow function. During this visit, the patient demonstrated proficiency with retrieving and placing crushable cups in cupboard, opening the refrigerator, retrieving and placing various containers from the refrigerator to table and back to the refrigerator, and picking up items from the floor. He was able to perform functional tasks including combing his hair, feeding himself finger foods, eating with a fork, folding towels, holding and carrying a laundry basket, writing his name, pouring water from a bottle to a cup, drinking from a water bottle, eating with a fork, and brushing his teeth. The patient became so proficient at using his prostheses that he was able to perform the same movements at the same time on both sides (e.g., hand open with the right and left ETD at the same time) as well as opposing movements (e.g., wrist supination on the right at the same time as wrist supination on the left).

During this visit in November, the patient was also able to complete outcome measures with the commercial IPR control including the Box and Blocks test (Figure 3),

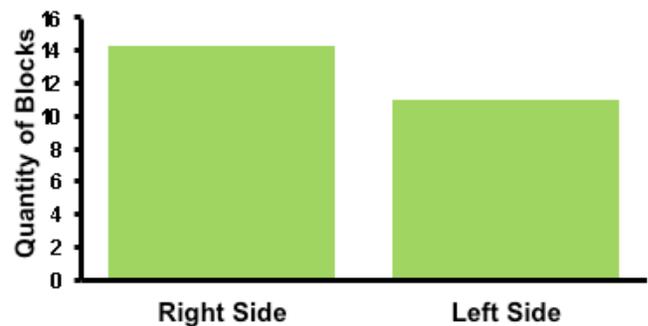


Figure 3: Box and Blocks Test: Number of blocks moved in two minutes using the right-side transhumeral or left-side shoulder disarticulation device, with commercial Complete Control APR controller. Data shown are averages of three trials.

the Clothespin Relocation Test (Figure 4), and a Block Stacking Task (Figure 5).

In the box and blocks test, even though this was his first week using pattern recognition on his left side, he was able to move an average of over ten blocks using this device (Figure 3). This number is similar to that achieved using his right side, with which he had experience using the research IPR system. In the Clothespin Relocation Task (Figure 4), the patient completed the test in a similar amount of time when using his left-side prosthesis as when using his right-side device during two of the three trials, with an average time separation of one second. In both these two tests, the patient demonstrated a comparable control of function on both sides using IPR, despite the different amputation levels.

For the Block Stacking Task, the patient was given 3 minutes to stack as many 1-inch blocks on top of one another as possible (Figure 5). Using the commercial IPR system, he was able to stack an average of over eight blocks using his left-side prosthesis after minimal practice. This task demonstrated both his confidence and degree of control of his prosthesis: he was able to bring his terminal device very close to the stack to release the block without fear of unintentional movements that might cause the tower to fall.

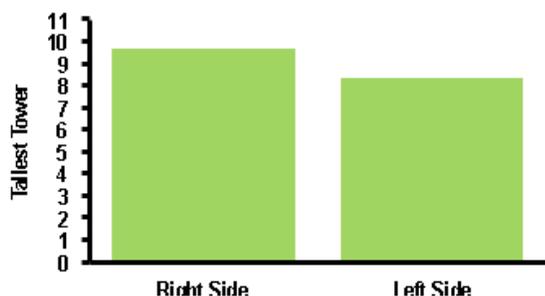


Figure 5: Block stacking: Number of 1-inch blocks stacked by patient in 3 minutes using the right-side transhumeral or left-side shoulder disarticulation device, with commercial Complete Control APR. Data shown are averages of three trials

## DISCUSSION AND CONCLUSIONS

This case study outlines the successful clinical fitting and functional outcomes of a bilateral upper limb amputee with IPR-controlled prostheses. The time from initial pattern recognition training to a definitive IPR-controlled transhumeral prosthesis for his right side, was approximately 13 months (September 2012 to November 2013). During this time the patient and clinical team overcame many challenges, including socket fit issues and muscle fatigue. Once fit issues were resolved, the patient verbalized the ability to use the experimental pattern recognition-controlled prosthesis to

perform functional tasks including yard work, which he was unable to perform using his direct controlled prosthesis.

The patient had not used pattern recognition on his left side until he was fit with the Coapt Complete Control IPR system, which was not available at the time of his introduction to pattern recognition on his right side in September 2012. The more secure socket fit on his left shoulder disarticulation side, and improved muscle strength from using his direct control prosthesis made the fitting and training process more streamlined than for the right transhumeral side. Within the four days of the clinical fitting, the patient was able to perform a variety of functional and bilateral tasks in different planes of movement. These included drinking from a cup or bottle and feeding himself; goals which he was unable to achieve using direct control. The ability to reach into various planes of movement (reaching items from floor), to manage crushable objects, and to perform bilateral functional tasks are typically extremely challenging tasks for patients with bilateral high level amputations using direct control. These tasks are typically due to problems with signal isolation, poor proportional control, the need for mode switching, and lack of intuitive control. It would also be expected that performance with the left-side shoulder disarticulation prosthesis would be less proficient than with his right-side transhumeral prosthesis, but his performance was similar for all three outcome measures. Because he was using physiologically correct movements to control the device with the IPR system, he was able to perform equally well on both sides. The patient's ability to perform the same movements and opposite movements bilaterally at the same time for all DOFs demonstrates the intuitive control provided by the IPR systems. Such movements would be very challenging to perform due to the cognitive load imposed by direct control.

Since the patient has been home with his bilateral prostheses, he reports frequent device use and satisfaction with the commercial IPR control system. He continues to use the prostheses for functional tasks including yard work, retrieving items from the refrigerator, and drinking from cups and bottles.

## ACKNOWLEDGEMENTS

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## A SIX DEGREE OF FREEDOM OPEN SOURCE HAND FOR EVALUATING MYOELECTRIC CONTROLS

Nili Eliana Krausz<sup>1</sup>, Ronald Rorrer<sup>2</sup>, Richard Weir<sup>2</sup>  
*<sup>1</sup>Northwestern University, <sup>2</sup>University of Colorado Denver*

### ABSTRACT

A six degree-of-freedom (dof) open source hand was developed with the goal of providing students, inventors, and researchers with a platform for testing of myoelectric control strategies. A significant challenge in controlling myoelectric prosthetic hands has been the limited number of anatomical myoelectric inputs in a residual limb that are available for use as a control signals. Thus most upper-limb prostheses have a minimal number of dofs (and generally only a single degree of control), and some means of dimensionality reduction is required to allow for the control of more complex prosthetic hands. Many researchers are investigating this problem, and there are multiple potential strategies that have been suggested for solving this problem. However, much of the research that has been done in the past and is currently being performed in this area utilizes virtual hands to simulate the performance of a prosthesis, without any actual implementation on a real hand. It is not clear that control schemes that are effective on a virtual hand will be effective on a physical prosthesis. In addition to inertial effects that can affect both the myoelectric signals and the dynamics of the control system, when using a virtual hand there is often additional visual feedback on screen that will not be present when using a physical device. Therefore, the authors postulate that the use of a physical hand would help to better establish the clinical relevance of a given control strategy, and thus the development of such a hand should be quite beneficial for researchers in the field. Two of the major goals of the project were that the hand be inexpensive and “open source” The hand has a single flexion/extension dof for each finger and a thumb with a flexion/extension dof and an abduction/adduction dof. Additionally, the hand was designed to utilize off-the-shelf motors and power transmission components, with the finger and hand shells being designed for fabrication with a 3D printer. The hand that was developed is within the size range of anatomical hands, however, the authors plan to reduce the size and weight and improve the robustness of the hand in the future. Additionally, the authors intend to add additional sensors to the hand with the goal of enabling research focused on providing sensory feedback to amputees.

## OPPORTUNITIES AND CHALLENGES WITH RE-EMERGING EXTERNALLY POWERED UPPER-LIMB ORTHOTIC TECHNOLOGY; PART II: CURRENT PATIENT CASES AND FUTURE DIRECTIONS

Chris Lake, CPO, FAAOP<sup>1</sup>; Jeremy Sherman, BSME<sup>1,2</sup>

<sup>1</sup>Lake Prosthetics and Research, Euless TX

<sup>2</sup>Baylor College of Medicine, Houston TX

### ABSTRACT

The use of external powered techniques for upper limb prosthetics has enjoyed a steady progression in technology since the 1960s. Inputs, microprocessors, mechanical and electric components, socket interface techniques, and therapeutic modalities have all improved as the field as a whole gained increased experience. The field of upper limb prosthetics is now recognizing the benefits of implantable electrodes, multi-articulated hands, and is on the verge of realizing commercially available pattern recognition systems. With these strides in technology, the clinical prosthetist will sometimes meet individuals with a relatively non-functional arm intact. This meeting usually involves a discussion of whether or not this individual should have their non-functional arm amputated in favor of the placement of an externally powered prosthesis. While this speaks highly of the advances of upper limb prosthetics it does beckon the question of why this technology hasn't transferred in application to upper limb orthotics? This paper will highlight several clinical cases to show their individual progression across a continuum of externally powered orthotic care as well as discuss future direction and development.

### COMMERCIALLY AVAILABLE EXTERNAL POWERED UPPER LIMB ORTHOTIC OPTIONS

Since the 1960s various institutions have discussed the use of external power in upper limb orthotics. [1 -6] Both the University of New Brunswick and the University of California at Los Angeles have done large scale reviews and academic efforts related to this endeavor. While the reported cases had favorable results and the mechanical engineering behind the developments were well thought out, significant clinical momentum was not enjoyed. Many of the early papers discussed the tedious nature of the fabrication and fitting as well as the varied and difficult patient presentations encountered.

While many different designs have been discussed over the last 50+ years, only two designs are commercially available today. The MyoPro from Myomo, Incorporated (Figure 1) is an externally powered elbow orthosis that was primarily designed for the stroke population. Lately, its

clinical application has been expanded to include individuals with brachial plexus injury, spinal cord injury, and other neurological deficits or diseases such as an amyotrophic lateral sclerosis. This device primarily utilizes myo-electrodes as input device though it has been designed to accept the inclusion of linear transducers.



Figure 1: "MyoPro" by MyoMo

The Power Grip wrist hand orthoses from Broadened Horizons, Incorporated (Figure 2) is a wrist hand orthosis



Figure 2: "Power Grip" by Broadened Horizons

that utilizes a mounted linear actuator to provide the second and third finger flexion about the metacarpophalangeal joint. The thumb is fixed in a position to allow three point pinch as the second and third finger are flexed. The power grip orthosis can utilize either switches or myo-electrodes as input devices. This device has primarily been used with the spinal cord injury population.

### CLINICAL POPULATION AND PATIENT CASES

#### Demographics

The patient population is vast and can include common traumatic mechanisms to rare cerebellar disorders. For the purposes of this paper, we will focus on the more common presentations which include cerebral vascular accident, spinal cord injury, brachial plexus and associated peripheral nerve

injuries, as well as neurological diseases such as amyotrophic lateral sclerosis. The following demographic information was taken from the respective patient associations related to each pathology and/or presentation.

Per the WHO, 15 million individuals worldwide have a cerebral vascular accident yearly. One third of those individuals recover fully, one third of those individuals die, and one third of those individuals are left with permanent disability. In most cases it is generally accepted that 5 million individuals a year have upper limb deficits secondary to stroke. In the US, stroke contributes to 795,000 cases per year with approximately 4.5 million stroke survivors in the population at any given time. This population has a very specific presentation that is highly influenced by neural plasticity and is generally older in age. The other aforementioned presentations include patient populations of lower incidence affecting a much younger population:

- There are an estimated 250,000 individuals in the US living with the functional deficits of the spinal cord injury. It is estimated that there are 12,000 new spinal cord injured individuals a year.
- There are 1.4 million traumatic brain injuries per year. Of this population, 80,000 to 90,000 present with permanent disability.
- The prevalence of brachial plexus injury is 1.2% for North America. This includes both birthing trauma and high velocity trauma occurring later in life. Considering that in 2008 there were 528.7 million individuals in North America that would suggest that there are over 6 million individuals with brachial plexus injury that undoubtedly have an upper limb deficit.
- Amyotrophic lateral sclerosis is a devastating and insidiously progressive neurological disease. It has a high mortality rate, and at any given time there are 30,000 individuals in the US living with this disease. Every year 5,600 individuals are diagnosed with this disease in the U.S.

#### Patient Cases

*The following patients provided informed consent for research participation and for image and video use.*

- (1) Megan suffered a Cerebral Vascular Accident after giving birth to her first child. This case chronicles her use of the MyoPro orthosis and the neuroplastic and rehabilitative musculature changes that occurred with the usage of this device.
- (2) Patrick presents with a spinal cord injury secondary to the failed implant of a cervical spinal cord

stimulator. Immediately following the surgery he presented as a quadriplegic but was able to regain reasonable function of all his limbs except his left upper limb. This case highlights the application of the MyoPro orthosis and its use to augment weakened and easily fatigued biceps musculature.

- (3) Jess presents with a brachial plexus injury secondary to a high velocity injury over 10 years ago. Jess considered having his affected arm amputated so that he can take advantage of upper limb prosthetic technology. This case chronicles his initial usage of the MyoPro device which resulted in documented increase in bicep strength. This ultimately led to further orthotic treatment.
- (4) Clay presents with amyotrophic lateral sclerosis and due to the course of the disease process relies on an electric wheelchair for mobility. This case highlights the significant upper limb deficits caused by the disease process and how a continuum of external powered upper limb orthotic care enhances the function and quality of life for this individual.

#### **FUTURE DIRECTIONS**

The use of external powered techniques for upper limb prosthetics has enjoyed great progress within the last 20 years. Much, if not all of the progress made in upper limb prosthetics has immediate application to upper limb orthotics. The challenges for the orthotic population are very similar to those of the prosthetic population. In coming years there will be a great need for similar therapeutic modalities as well as functional and reconstructive surgical techniques. The saving grace of this endeavor will be the fact that the orthotic patient population dwarfs the upper limb prosthetic population. In fact, the size difference and the technical challenges may lead one day to the trickle-down of technology from orthotics to prosthetics instead of from prosthetics to orthotics. To effectively treat the orthotic population the clinician must be proficient in upper limb prosthetic techniques and exercise extreme patience and command over the available materials.

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## REIMAGINING THE PEDIATRIC PROSTHESIS – A CASE STUDY OF SOLUTIONS TO BILATERAL CONGENITAL ABSENCE AT THE ELBOW

MacJulian Lang CPO, FAAOP, Kerstin Baun MPH, OTR/L  
*Advanced Arm Dynamics*

### ABSTRACT

The decision to fit a child with a prosthesis at an early age is a complicated matter. Varied rates of acceptance and rejection have been reported<sup>1-3</sup> and a variety of factors that influence usage have been discussed. Studies by Postema, et al.<sup>2</sup> and Wagner, et al.<sup>1</sup> indicated lack of function and lack of comfort as reasons for rejection. When presented with the challenge of fitting a child with bilateral absence at the elbow the authors strove to design a prosthesis system that would address these two concerns. The case study is a description of the steps taken to ensure maximum potential function and comfort.

### INTRODUCTION

When the decision to fit a young child with a prosthesis is made the question then becomes what type of prosthesis to fit and at what age. There are varying opinions and research results on this matter. Shida-Tokeshi et al.<sup>4</sup> found “there are no conclusive indicators as to which type of prosthesis contributes to continued prosthetic wear and use”. In a review of studies evaluating connection between age at first fitting to rejection rate Meurs, et al.<sup>5</sup> concluded that “only little evidence was found for a relationship between fitting of a first prosthesis in children with congenital upper limb deficiency and rejection rates or functional outcomes.” Additionally, most studies investigating rejection rates in pediatric prosthesis users have looked at unilateral absence, most frequently at the transradial level. This is likely due to the insufficient number of bilateral patients to find significant relationships.

This paper presents a case study of a boy with bilateral congenital absence at the elbow. At the time of initial evaluation he was four and half months old and all prosthetic options were presented to his parents. The rationale behind whether to fit or not was discussed. Wanting to provide their child with additional options for function the parents described a strong desire to have their child fit with prostheses. Additionally, the parents indicated that they would like to have him fit with myoelectric prostheses as early as appropriate.

In a review of relevant literature the authors found little guidance on what to fit for this patient’s presentation. Peers polled provided experience and perspective that helped direct the care plan. It was decided that in order to provide the highest likelihood for continued use of the prostheses that function and comfort would be paramount in the design of the prostheses.



Figure 1: 4.5 month old patient



Figure 2: Patient at 7.5 months with passive prostheses

The family lives approximately 350 miles from our clinic which created a challenge for fitting and follow-up. Expectations for function while wearing the prostheses were clearly discussed. The parents committed to traveling as needed for adjustments, new fittings, and intensive therapy in and around the clinic. The parents also pledged to be vigilant with a consistent home therapy program.

### PROSTHETIC FITTING

#### Passive Prostheses

At seven and a half months the patient was seen for initial fitting of passive prostheses. The design criteria used in the creation of the passive prosthesis were: ease of donning and doffing, secure suspension with minimum of harnessing, maintenance of shoulder active range of motion, positioning

of prosthetic elbow hinge center at appropriate anatomical location, ability to adjust volume for growth, provision of internal and external rotation, passive grasp, and finally appearance. In our estimation this combination of attributes would provide the widest range of potential function and comfort.

The prostheses utilized roll-on silicone liners and lanyards for suspension. The lanyards were attached distally to the liners and passed to the exterior through a posterior channel and then up and through a side rectangle loop. The strap was attached to itself with Velcro. This method allowed the parents to quickly don the liners and secure the prostheses. This style of suspension removed the necessity of a harness. The addition of socks over the liners accommodated for future growth as they could be removed as the socket became overly tight. The terminal devices utilized were Life Touch Micro terminal devices manufactured by Therapeutic Rehab Systems, Boulder, CO. They provided passive grasp into which the parents were able to place objects.

The biggest challenge in the design of the prostheses was the length of the patient's limbs. Pediatric elbow components available did not provide an adequate solution. There were no options available that would maintain an appropriate prosthetic elbow hinge center while at the same time providing for internal and external rotation. This dilemma was solved using a pair of passive positional joints with detent resisted motion. These joints were previously designed for use as adult finger joints by one of our international collaborators. The diminutive size of the joints allowed them to be mounted to the outside of a female threaded coupler. The corresponding male thread was positioned on the outer portion of the frame. The design thus allowed for rotation of the hinge assembly around the humeral axis while maintaining a hinge center very similar to theoretical anatomical joint location.

#### Myoelectric prostheses

At 18 months of age the patient returned for initial fitting of myoelectric devices. The design criteria for the myoelectric devices stayed very similar to that of the passive devices with the additional complication of the electric components. The adjustability of the sockets also became a larger concern as it was important for these devices to be usable for a longer period of time. Donning accuracy was also an inherent concern as skin and electrode contact was necessary.

Initially a wet fit suction suspension design was explored due to the necessity of electrode contact but this was quickly abandoned. The alcohol based donning lubricant was met with a very negative reaction by the patient. In the long run volumetric adjustments would have been more difficult due to the lack of easy adjustment. The design reverted to lanyard

suspension. In order to minimize build height of the frame the lanyards were affixed to the sides of the liners in lieu of distal attachment. Lateral proximal and distal channels were created to allow the strap ends to be tensioned together creating a continuous loop. Holes were cut into the roll on liners to allow for skin contact with the electrodes.

In order to maximize volume adjustability the socket and frame system was highly modified from a standard design. Free floating panels of rigid frame were created from the anterior, posterior, and medial frame walls and the flexible socket under the anterior and posterior panels were also trimmed free. The medial side of the inner flexible socket was left attached and the carbon panel affixed. The panels over the electrodes were tethered to the distal frame to help position the panels and protect the electrode wires. Thus the socket and frame walls would fold out of the way during donning but would relocate during tensioning.

A BOA™ mechanical lacing system was utilized to create the closure. The resultant socket and frame system was very flexible and adjustable while also being firm and stable during use. Ease of donning and doffing was significantly improved due to the flexibility of the frame and the increased visibility to the channels for the lanyard. The final benefit of the design was the ability for the parents to visually confirm the location of the holes in the liner and manually position the electrodes during the donning process. This drastically improved the repeatability of electrode contact. As with the passive prostheses a harness was not necessary for suspension and the patient maintained normal shoulder active range of motion in the prosthesis.

The elbow hinge system was again a custom feature. The joints utilized in the passive prosthesis did not have sufficient motion resistance to accommodate the increased weight of the terminal devices. A new joint was designed to provide a similar detent resisted motion. Larger joint heads were laser cut from stainless steel and motion resistance was derived from press fit ball nose spring plungers. To allow for passive internal and external rotation two attachment plates were likewise laser cut from stainless steel. A single attachment screw allowed for variable friction resisted motion limited by end stops. Approximately 45 degrees of rotation were allowed in both internal and external rotation from the sagittal plane.



Figure 1: Fully expanded socket/frame system



Figure 2: "Mommy Switch"

Terminal devices used were size 5 Otto Bock System 2000 hands with 7 in 1 controllers. The 7 in 1 controllers offered the ability to wirelessly monitor myoelectric signals and make adjustments to the programming while the prostheses were on the patient. The ability to quickly and easily change the control strategy as the patient became more volitional in his control was also beneficial. The parents especially appreciated the ability to see the electrode signals to better understand how the system functioned. One aspect that was not available with the 7 in 1 controller was a "mommy switch" or manual input into the system for opening of the hands. In order to achieve this we soldered a force sensitive resistor in line with the extensor electrode wire which created a shared input. This provided much improved function during therapy. Ottobock Myolino Wrist 2000 wrist units were used to allow for flexion, extension, deviation and rotation of the wrists. This provided a wide range of angles and positions to better approach objects to be grasped or held.

## PROSTHETIC TRAINING

### Passive Prostheses

For the passive prostheses the primary goal of training involved teaching the parents how to don and doff the prostheses and how to interact with their child while the prostheses were on. The importance of maintaining a very positive approach was emphasized. Parents were instructed in developmentally appropriate activities to try such as oral exploration and self-feeding. For example, parents were shown how to passively position the prostheses to hold a teething toy and teething biscuits. The ability to passively rotate the forearm sections accommodated for differences in size of objects as well as to aim the hands towards his mouth. The patient could then independently move the items into

his mouth using shoulder flexion, forward neck flexion or a combination of both.

It was recommended that parents reserve high preference activities and foods for prosthetic wear times to maintain a highly positive environment. Parents demonstrated excellent ability to manage donning and doffing and followed through on suggested activities to engage the patient in. Although bilateral wear times were incorporated into the patient's program, parents preferred to don one prosthesis at a time, allowing for touch exploration with the other arm. The patient would use his limb to explore the passively opening hand and "pinch" his arm with it playfully.

### Myoelectric Prostheses

With the introduction of myoelectric prostheses came new challenges and new opportunities. Positive reinforcement during the donning process was critical to initial success. A food motivator was found that allowed for sufficient time to make adjustments and practice donning strategies. The flexibility of the socket and frame combination in the myoelectric devices drastically increased the ease and accuracy of donning the prostheses as well as decreasing the occasional frustration of the parents.

Use of the "mommy switch" allowed for demonstration of the hands' ability to open and close to hold onto preferred toy items and reach for favorite foods. At this point in time, the patient was able to open the terminal device but it appeared this was not volitional. Given these new tools and the developmental abilities of the child, the parents were instructed in an expanded repertoire of activities that they could now engage their child in. These included holding and scribbling with markers, holding food items for self or others to eat, and holding onto pull and push style toys, among others.

The parents continued to prefer unilateral wear of the prostheses and the patient continued to explore the terminal device and would observe the active movement of his prosthetic hands. The passive humeral rotation allowed for the terminal devices to be positioned towards midline in unilateral wear and he would often put his contralateral arm into the terminal device and allow it to close on him. During bilateral wear the forearms were externally rotated to enable bimanual grasp of objects.

On subsequent visits to our office for adjustments and further training, volitional use of the hands was clearly demonstrated by taking "turns" with an adult amputee model opening and closing a TD. The patient was able to demonstrate effective reaching and positioning of his hands as well as active and passive use of his prostheses during play both in the clinic and on a playground. Again the passive

positioning available at wrist and elbow allowed for a more ergonomic and secure grasp on the playground equipment. His bilateral wear time was increased to approximately 2 hours. Using the SIRS (Skills Index Ranking Scale) which measures development of myoelectric prosthetic control<sup>6</sup>, the patient is functioning at a level 9 of 14 as he is able to use the active grip function while supporting the weight of his prostheses bilaterally.



Figure 1: Patient at 22



Figure 2: Self-exploration months of age

Permeating the prosthetic training process was play. Prior to fitting, play was used to build a therapeutic relationship and explore preferred toys and interests. Throughout the fitting process, play was used to engage the patient in favorite activities and thereby develop first passive and subsequently active use of his prostheses. Indoor and outdoor play activities were chosen to consolidate skill development and parents were provided with modeling and instruction on ideas to encourage follow through in their home environment.

## DISCUSSION

The decision to fit a young child is ultimately that of the parents. In this case the parents of a young child with bilateral upper limb loss were highly motivated to have their child fit at an early age. It has been reported that one factor in prosthesis rejection is lack of function and comfort.<sup>1,2</sup> The goal in the design of these specific prostheses was to allow for the best chance for meaningful function and thus the greatest chance for continued wear. To achieve this we sought to combine the most passive positioning possible in both the passive prostheses as well as the active motion devices. This was combined with a system where other potential roadblocks for wear such as difficulty in donning, repeated trips to the clinic for adjustments to accommodate growth, or discomfort during wear were reduced or removed. The prosthetic designs have been tolerated well to date and the patient is using them

in an increasingly functional manner. Continued use of these prostheses is not guaranteed. However, by following this patient closely and continuing to create solutions that allow for maximum potential function and comfort the best likelihood of long term use will be ensured. Thus we hope to provide him with prosthetic options that he will have the ability and desire to use to for years to come.

## ACKNOWLEDGEMENTS

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*The parents of the subject gave written permission to include the photos in this paper.*

## THE IMPORTANCE OF A TEAM APPROACH TO ACHIEVE THE BEST OUTCOMES WITH THE UPPER LIMB AMPUTEE

Debra Latour, M.Ed., OTR/L<sup>1</sup>, Laura Katzenberger, CP<sup>1</sup>, Claribell Bayona, OTR/L<sup>2</sup>,  
<sup>1</sup>Handspring, <sup>2</sup>NYU Rusk Rehabilitation Institute

### INTRODUCTION

Outside of the traditional rehabilitation center approach to amputee care, it has become more challenging to address the complex needs of the upper limb amputee utilizing a comprehensive team approach. Incorporating the nurse case manager adds another critical ingredient to the overall process and ultimate success in achieving optimal outcomes in the upper limb amputee. The purpose of this study is to explore the benefits associated with early prosthetic and occupational therapy intervention, holding monthly team meetings, collaborative appointments with the occupational therapist and prosthetists as well as the condensed fitting model.

#### Methods

This study is retrospective and follows the care of a 29 year old adult female who sustained a traumatic injury at work that led to a transradial amputation of her non-dominant hand. Her care was managed by a catastrophic case management company who pulled together a team to treat their patient. This patient was evaluated two days after release from the hospital and was fit with a myoelectric prosthesis incorporating a terminal device with multi-articulating digits, a passive functional aesthetic prosthesis and a sports adaptive prosthesis within 6 months of injury. A monthly team meeting allowed all members to report on the current status of the patient's care under them. Time was allotted around each of these team meetings for an occupational therapy session with the prosthetists present to facilitate necessary adjustments for optimal function of the prosthesis during therapeutic activities. A 4 day condensed fitting occurred 3 months after injury to fit the patient with a preparatory prosthesis and begin the fitting for the passive functional aesthetic prosthesis. She received ongoing prosthetic training and rehabilitation with integration of skills to professional and personal life activities.

#### Results

The results of this case study suggest that there are functional benefits to providing team-approached collaborative care for the catastrophic upper limb amputee. In this case, the patient's progress utilizing multiple prostheses

was accelerated significantly due to team collaboration and communication. This also established a higher level of success with more advanced and complicated terminal device technology incorporating multi-articulating digits. Once the myoelectric prosthesis was provided, the patient became more independent in bimanual activities, reducing overuse of her sound side during functional tasks. Use of diverse prosthetic technologies accompanied by integral and collaborative training from the team of stakeholders enhances patient functional outcomes.

#### Discussion

A case study is typically thought to describe the care of an individual. However this case study provides a reflective sample of industry best practice standard and how communication transcending disciplines and organizational stakeholders can impact client care for successful outcomes that include functional independence, personal satisfaction and perceived quality of life. Highlighted is the concept that *no single prosthesis can address the multiple deficits associated with upper limb loss*. Early prosthetic and occupational therapy integration, along with this patient's full return to prior work engagement highlights the "*Golden Period*" the value of early intervention. The importance of implementing Occupational Therapy services early was critical, ensuring that the patient regained lost AROM, improved shoulder stabilization and strength, and prepared the residual limb for prosthetic fitting and tolerance. Patient education demonstrating one-handed techniques increased patients' independence prior to receiving prosthesis. Individual training to use the prostheses was critical to the patient's ability to functionally incorporate her affected limb during all tasks, reducing the potential overuse of her sound side. The vitally important role of the NCM interfacing with the comprehensive team approach resulted in optimal outcomes in functional independence, return to work and recreational pursuits.

**FIGURES AND IMAGES**

Table 1: *Team Approach*

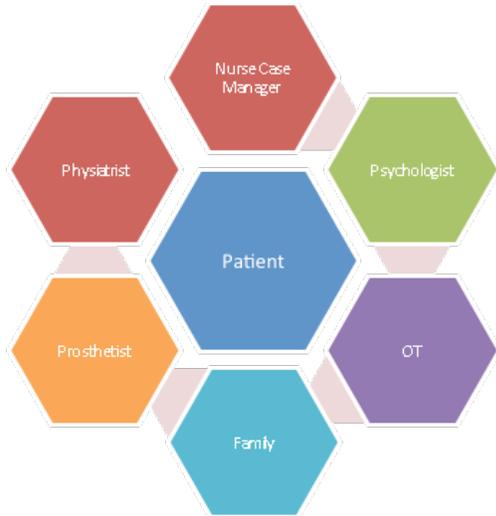


Table 2: *Patient Well-Being*

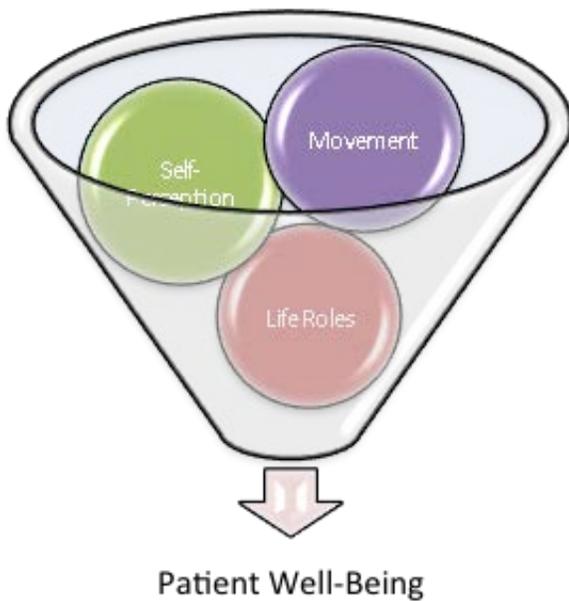


Figure 1: *Example of passive aesthetic function*



Figure 2: *Example activity-specific technology*

**ACKNOWLEDGEMENTS**

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## CREATIVE HARNESSING SOLUTION: CUTANEOUS ANCHOR TECHNOLOGY TO ACCESS FUNCTION

Debra Latour, M.Ed., OTR/L  
*Single-Handed Solutions, LLC*

### INTRODUCTION

Individuals with upper limb deficiency who choose to wear and use prosthesis technology do so for many reasons. Evidence suggests that these reasons relate to personal, social and functional preferences. With age, these individuals often experience difficulties with prosthetic fitting and use due to anatomical presentations that may affect both of their upper limbs. This problem is likely to be magnified in the person who has acquired limb loss due to trauma. The non-amputated side (previously known as the 'uninvolved' side) may be affected by a number of debilitating issues including arthritis, nerve damage, scar tissue, brachial plexus involvement, or rotator cuff injury to name a few. These problems may impact the user's ability to access power from available body movements, or to access myo-signals to control externally-powered technology. The end-result is that the prosthesis-user experiences challenges to use the non-amputated yet 'involved' upper extremity as well as the prosthetic extremity in order to complete bilateral activities necessary for independent function.

Traditionally, a body-powered prosthesis is activated by a harness system, using the contralateral shoulder as the power source. Many users complain of harness-related discomfort in the axilla, at the O-ring, upper body asymmetry, pain in the contralateral shoulder, difficulty performing bilateral tasks and diminished cosmesis. For these reasons, many individuals reject use of prostheses. Occupational therapists like me help clients develop skills to live with maximal independence and to improve quality of life. I have invented an alternative method to capture body power without use of a traditional harness that may provide a solution to these complaints.

The Cutaneous Anchor technology derives its primary source of control from the scapula on the same side of the limb deficiency. The terminal device is operated by the ipsilateral shoulder. Because the harness is eliminated, the benefits have the technology have been reported to include more symmetrical bilateral upper extremity development, increased function, greater comfort and improved cosmesis. Although it was originally developed for use for individuals with involvement at the trans-radial level, derivatives of the

technology have been used to suspend and control prosthetic technology at all levels including trans-humeral and partial hand. This technology and its method of use are patented with the US Patent Office. The Anchor has been used in patient treatment since August 2006. Pediatric and adult patients appear to derive benefit, satisfaction and improved function of their upper extremity prosthesis using this device as measured by the U-BET and the PSI during initial studies. The Cutaneous Anchor Technology

Creative Solutions to Accessing Power addresses case solutions for problems associated with accessing power and/or control of the prosthesis using simple technology advances to complement the more complex technology used in the design of the prosthesis. Reflective case studies are discussed which include initial presentation with consumer-stated problems and concerns, solutions offered and training provided to the user from the perspective of the occupational therapist. Occupational therapists are concerned with the abilities of our clients to attain the skills vital for maximal functional independence that include self-care, vocational and leisure-time activities. Proficiency in these areas fosters an enjoyable and positive perception of quality of life.

### METHOD

Subjects: Four subjects are identified for the purpose of this reflection:

- a. A is an 70 year-old male with L trans-humeral loss acquired in an industrial accident 40+ years ago. He is a long-time user of body-powered technology however, due to injuries of his involved UE including fracture of the residual humerus combined with insertion of a pacemaker in his R chest wall, he can no longer sustain an axillary harness. His prosthetist fit him with OttoBock technology for both elbow and terminal device control; using a chest strap for suspension. The chest strap rode up on the client's chest and he resorted to using his intact hand to stabilize the strap in place, thus rendering the hand unavailable for functional tasks in order to use the prosthesis assistively.

- b. B is a 31 year-old male with recent acquired loss of his dominant index finger at the PIP joint due to an industrial accident. B is a construction worker who uses tools, climbs ladders onto scaffolding and carries heavy objects. He plans to return to work. His residual anatomy precludes him from being a candidate for externally-powered technology. His case reflects his personal preferences to eliminate the cuff to activate body-powered technology appropriate for his partial-hand anatomy.
- c. C is a 45 year old male who presents with L congenital trans-radial deficiency and severe R medial and lateral epicondylitis accompanied by R shoulder and wrist pain. He has not used a prosthesis in over 30 years, having abandoned use (by self-report) due to discomfort attributed to the harness.
- d. D is a 61year old male with a L trans-radial deficiency due to trauma 40 years ago. He is a long-time user of body-powered prosthetic technology, but is now experiencing surgery and rehabilitation to his R shoulder due to a rotator cuff injury. During the rehab process he is unable to use his harness-driven prosthesis and requires a different solution.

Apparatus:

Each client utilized a diverse form of cutaneous anchor technology to meet their individual needs, given the varying anatomical presentations and demands of the prosthetic technology in use.

Procedures:

Each client was fitted with the technology and instructed to use in guided trials as well as independent use during home programs.

Data Analyses:

Outcome measures were used, pertinent to the clinical settings. These included completion of the DASH pre- and post-intervention, video-graphed documentation of functional tasks as described by the UNB as well prosthetic satisfaction survey.

**RESULTS**

Preliminary data reflects overall satisfaction with the cutaneous anchor technology as an alternative to traditional harnessing. This data will be further explored with final outcomes to be reported at this event.

**DISCUSSION**

The Cutaneous Anchor Technology is simplistic in design, is durable, affordable and easily available. It presents as a compelling alternative to existing technology.

The potential benefits of this device appear to result in increased satisfaction, increased prosthesis wear and use (as related to tolerance, frequency and spontaneity) because it allows for improved comfort, cosmesis and intuitive access to movement during functional activity. These benefits appear regardless of limb length or time of loss (congenital vs acquired). Implications continue to project use toward dynamizing upper limb orthoses in clients with loss of function but not necessarily loss of limb. Future study toward this end is planned.

**FIGURE AND TABLES**



Figure 1: Example of cutaneous anchor technology for individual with trans-humeral involvement.



Figure 2: Example of cutaneous anchor technology for individual with partial hand involvement.

## ACKNOWLEDGEMENTS

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1. The owners and staff at Handspring, NY for their support in the development of this study;
2. The owners and staff at Attain Therapy and Fitness, MA for their support in the development of this study;
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4. My husband Michael for investing with me in our company, Single-Handed Solutions, LLC to further develop this and other technologies to benefit individuals like me;
5. TRS, Boulder CO for distributing this technology.

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## CREATIVE SOLUTION WITH EXTERNALLY-POWERED TECHNOLOGY

Debra Latour  
*Single-Handed Solutions, LLC*

### ABSTRACT

Primary goals of an occupational therapist are to facilitate client's abilities toward maximal functional independence and quality of life. It is not uncommon for individuals with upper limb deficiency/loss to experience complex problems that interfere with prosthesis use or for which completing the task independently with a prosthesis is challenging. These challenges are oft experienced regardless of the sophistication of the prosthetic technology. To achieve these goals for our clients with upper limb deficiency, we create innovative solutions using simple technologies. This paper offers a retrospective case study examining the problem faced by a client to fill a diabetic insulin pump, and details the successful interventions.

#### Method:

**Subject:** The patient is a 45 year old woman with recent acquired loss of her distal palm and digits of her dominant hand. Co-morbidities include recently diagnosed carpal tunnel syndrome of her L wrist and diabetes which is managed by insulin via pump. She was recently fit with Touch Bionics ProDigits. Her goals were to become independent in all aspects of self-care including managing the insulin pump.

**Apparatus:** Attempts were made to utilize her existing technology to complete the task including application of the device and insertion of the needle. These attempts failed due to limitations in reach and surface area of the digits' tips.

**Procedures:** An elongated digit was fabricated of low-temperature thermoplast and molded to the shape of the TB digit. The length of the attachment was approximated to accommodate the patient's anatomy.

#### Results:

Preliminary data reflects overall satisfaction with the adaptive digit. The patient was able to demonstrate independence to complete the task. The device fits on other digits to accommodate proximal and distal reach across the abdomen in order to alternate sites.

#### Discussion:

Although many tasks can be managed either with the use of a prosthesis, or even without it; some tasks are particularly difficult. Many users of prosthetic technology desire to be independent in all aspects of self-care, particularly those requiring access to more intimate areas of anatomy. Our senses of self-esteem, self-worth and productivity are often wrapped up in our abilities to complete these tasks without the assistance of others. Use of simple adaptations can enhance the quality of life for our patients and alter the self-perception from "de-powered" to "em-powered".

## THE APPLICATION OF OPTIMAL FORAGING THEORY TO THE QUANTITATIVE EVALUATION OF SOMATOSENSORY FEEDBACK SYSTEMS IN PROSTHETIC LIMBS

Rock Lim<sup>1</sup>, Zachary Thumser<sup>2</sup>, Paul D. Marasco<sup>1</sup>

<sup>1</sup>*Cleveland Clinic*

<sup>2</sup>*Louis Stokes Cleveland Department of Veterans Affairs Medical Center*

### ABSTRACT

Integration of somatosensory feedback with prosthetic limbs requires a way to quantifiably evaluate changes in limb performance and control. Individual measurements of speed, accuracy, or reliability could lead to an incomplete performance evaluation; necessitating a more thorough analysis. Optimal Foraging Theory (OFT), a mathematical model for predicting predator behavior, integrates these factors into a single summary statistic, allowing for a more comprehensive sense of the effects of tactile feedback on prosthesis operation.

The OFT equation utilized in this study is the form presented by Eric L. Charnov in 1976:

$$R = \frac{E_n}{T = E / (T_h + T_s)}$$

The rate of energy intake (R) is equal to energy obtained from prey (E) divided by the sum of handling time (T<sub>h</sub>) and searching time (T<sub>s</sub>).

In this experiment an Ottobock MyoHand mounted with a tactile feedback system was used to sort visually-identical rubber blocks (prey) by hardness, simulating one significant aspect of the day-to-day activities required of a prosthesis user. A system for tactile feedback, which seems to be the most important factor in hardness assessment (Srinivasan, MA & LaMotte, RH 1995), was integrated into the prosthesis. The system transmits force from a load cell located on the thumb of the prosthesis to proportionate force on the palm near digits #1 and #2.

Correctly-sorted blocks (E) were scored as analogous to successfully consumed prey, while searching time (T<sub>s</sub>) referred to the time spent selecting blocks, and handling time (T<sub>h</sub>) represented time spent assessing and sorting blocks. To remove effects of differently valuable prey types and prey density, blocks were arranged in a grid and assigned equal value. When performing the task with tactile feedback active, subjects exhibited higher OFT scores than without, suggesting that tactile feedback can improve users' ability to discriminate between objects of different hardness. Through application of OFT multiple significant factors have

been integrated into one cohesive value that can concisely evaluate the potential improving effects of tactile feedback on prosthesis users' ability to more easily and intuitively perform routine object manipulation.

Other versions of OFT equations can focus on more specific aspects of foraging, such as relative prey value, patches of varying prey densities, or risk (Pyke, GH 1984), or focus on specialized foraging scenarios. These equations could be similarly applied to studies on other scenarios analogous to foraging, such as studying texture discrimination, on tasks emphasizing acquisition speed, or tasks requiring dexterous manipulation of objects.

## SIMPLIFIED APPROACH TO MYOTESTING AND ELECTRODE PLACEMENT FOR SUCCESS IN CLINICAL PATTERN RECOGNITION

Blair A. Lock, MS, PE, and Frank D. Cummins II  
*Coapt LLC*

### INTRODUCTION

For decades, the subject of applying pattern recognition algorithmic technology to powered, upper-extremity prosthetics has been researched, evaluated, and refined. Thanks to this global development effort, pattern recognition technology has finally entered into the upper-extremity marketplace. Pattern recognition control of powered, upper-extremity prostheses promises a more intuitive method of use and has the potential to provide numerous other ancillary benefits to the patient. Training individuals to use pattern recognition in a clinical fitting has only recently started to be documented [1-3]. For the clinician, however, the approach to myotesting and electrode placement for pattern recognition has not been well documented. With pattern recognition the process of myosite location can be performed somewhat differently than with a traditionally-controlled myo prosthesis fitting. Therefore, the combined sections of this contribution are intended as a reference guide to clinicians in the early stages of performing a pattern recognition fitting for their patients.

### BACKGROUND

Traditional myotesting can loosely be described as the art of clinically locating the areas on the residual limb where suitable surface electromyogram (sEMG, EMG) prosthesis control signals are present. For traditional methods of myoelectric control, it is important to locate sites where EMG signals can be detected reliably with significant amplitude. When intending to use more than one site, it is important to locate signals that are isolated and independent; i.e. one that remains at lower amplitude while the other is active and vice versa. Often, sites with the highest overall amplitudes may not but used – superseded instead by sites giving superior isolation. Benchtop tools such as the MyoBoy® from Ottobock and the Myolab from Motion Control are commonly used by the clinician to aid in this process. [4]

For amputees having undergone targeted muscle reinnervation (TMR) surgery, myotesting to locate 4-5 isolated sites is common and can require significant clinical iteration [4].

### OBJECTIVE

This narrative describes the in-clinic approach to myotesting and electrode placement for pattern recognition users. In many ways, these processes are different yet greatly simplified over the traditional control fittings because of the nature of how the pattern recognition algorithms work. Pattern recognition utilizes the full “concert” of information contained in a number of EMG signals whereas traditional myoelectric control schemes rely on comparative amplitude information from singular EMG signals. In this manner, pattern recognition control is less sensitive to EMG electrode placement and can be achieved with quasi-generic placement of electrodes [5, 6]. Furthermore, variances in factors such as inter-electrode spacing, orientation, and electrode contact size have all been shown to be accommodated by pattern recognition [7].

### MYOSITE LOCATION FOR PATTERN RECOGNITION

When planning to employ pattern recognition control of a prosthesis, some aspects of pre-prosthesis myosite location differ from the myotesting methods of traditional control. Pattern recognition control does not depend on isolated or independent EMG signals; therefore tool-based location of those sites is commonly not required. In general, these following steps can be followed for the new practice of pattern recognition myosite location:

#### 1) Discussion

Because pattern recognition is a form of intuitive user control, it is important to first discover what control motions the patient will find intuitive. Asking the patient what postures/feelings of their phantom/missing hand, wrist, elbow, etc. are discernable is a good place to start. Begin with the most physiological: i.e. asking the patient if they can they feel as though they can make a full hand close or a fist. Ask if they feel all fingers spread open for hand opening; if they feel as though they can rotate their wrist in either direction; if can they do that with the phantom hand relaxed, etc. Through this exploratory questioning, the **goal** is to determine:

- *what the patient can feel that is intuitive to them;*
- *what feels repeatable each time they do it;*
- *actions that feel different for each motion.*

Often, a patient will indicate they cannot feel the whole hand during open or close, or that many fingers are “stuck”, or that the perception of wrist rotation is faint, etc. Here it is important to focus on what they can feel that, again, is intuitive, repeatable, and separable. Many patients may have good perception of a specific finger or two and they may be able to focus on using those perceptions for open/close. Similarly, involving the feeling of their thumb leading the pronation and supination can often help.

In some cases, there may be considerable time between early interactions with the patient and the myotesting/fitting phases. It is often helpful to generate a common vocabulary and a home exercise program for these patients to practice their pattern recognition control motions. The program can be based upon the motions that were discovered during this discussion phase. It can include printed images, strengthening techniques, and other reference material to help the patient focus and develop.

## (2) Palpation

Using both hands to grasp and cover as much of the patient’s residual limb as possible, a clinician can feel much of the overall general activity – similarly to how the pattern recognition algorithm does – by sensing the patterns of activity from multiple sites at the same time. This should be done while continuing the discussion and discovery of the patient’s intuitive, repeatable, and separable control motions. As the patient is asked to perform the different motions and to relax in between, the clinician will attempt to feel the different patterns of muscle contraction activity for each of their motions. Throughout this task the clinician is not attempting to locate areas of strong, isolated contractions – instead, the **goal** is to *feel for areas with any underlying muscle activity related to the control motions*, especially those areas that seem to provide unique and repeatable contractions. Since pattern recognition does not only depend on strong muscle signals, areas where slight but unique-to-each-motion contractions are felt should not be ignored and should be noted for potential inclusion as a pattern recognition myosite.

## (3) Planning

Planning the placement of EMG electrode contacts for pattern recognition should follow the concept of covering the areas of interest that were noted from palpation and discussion. In contrast to traditional control methods, no electrode amplifier “pre-amps” are required for pattern recognition.

Consequently, a pair of electrode contacts rather than a packaged electrode “pre-amp” will constitute a “myosite”. A pattern recognition system commonly employs 8 of these myosites [7]. The relative placement and positioning of the pair of contacts is variable and accommodating. (More on this in the PLACEMENT OF ELECTRODE CONTACTS section of this paper). The muscle signals picked up in the general area underneath each myosite will be considered, together, by the pattern recognition algorithm. Since the **goal** is to *generally cover the area of interest with pattern recognition electrode contacts*, generic and sometimes symmetrical myosite location is often possible. There are, however, a few common sense rules that should be followed:

- Avoid areas that will lose electrode-to-skin contact during use – Like traditional myoelectric control, electrodes need to remain in good contact with the skin in order to provide good pattern recognition control. Take caution to plan myosite locations at areas where the skin and electrodes maintain robust contact.
- Stay within planned socket trimlines – Keep the planned socket coverage area in mind when palpating and planning. If a contraction area of interest is discovered that is outside, or near planned trimlines, simply consider planning that myosite close to the edge of the socket and some of this signal will still be captured and used in pattern recognition
- Avoid areas that have no underlying muscle – Electrode contacts are placed in order to pick up electrical activity from muscle contractions. Areas over bone and skin only will not provide much useful signal and therefore should be avoided. In addition, electrode contacts can become uncomfortable over bony prominences. An exception to this guideline is in placing a reference or ground electrode so long as it is comfortable to the patients.
- Avoid sensitive skin areas – Even if the strongest palpated muscle signal is present at an area that is sensitive (scar tissue, wound areas, etc.), it is not worth the discomfort to the patient to plan a myosite at that location. One of the benefits of pattern recognition is that highly isolated, independent myo signals are not paramount and therefore a myosite in this situation can be planned for nearby, non-sensitive skin areas.
- Avoid areas of socket loading – Similar to the avoidance of sensitive skin areas for comfort reasons, it is generally good practice to avoid myosites at areas of high socket loading. In addition, specific areas of loading – such as the proximal brachioradialis of transradial amputees and the

deltoid of transhumeral amputees – are sometimes important to avoid as these areas of muscle can often be active for limb positioning.

All amputee patients will present with unique situations which heightens the importance of this planning stage. Here, we offer a notes on a few specific cases:

#### *Patients new to myoelectric prostheses*

Patients who are recent amputees or have not been able to be fit with traditional myoelectric prostheses before may become candidates for pattern recognition myoelectric control. When planning myosite location with these patients it is generally encouraged to use the palpation method in place of extensive use of a myosite finding tool such as the MyoBoy® or Myolab. Palpation will provide a quick sense of contraction areas-of-interest and these are good locations to plan the first few pattern recognition myosites. Remaining pattern recognition electrode contacts can then be spaced generally in between these areas of interest.

#### *Existing users of myoelectric prostheses (retro-fits)*

When adding pattern recognition to a prosthesis for an existing myoelectric user, it is usually beneficial to note the location of any one or more existing myosites. Each of these are typically good areas to place a pair of electrode contacts constituting a pattern recognition myosite – i.e. someone has determined in the past that this location provides useful control signal(s) and reliable electrode-to-skin contact. After covering these existing sites, the remaining pattern recognition electrode contacts can be planned as described above.

#### *Targeted Muscle Reinnervation patients*

TMR patients benefit by having a rich set of control information present for surface EMG detection and, therefore, are typically ideal candidates for pattern recognition myoelectric control. When planning myosite placement for these patients, it is important to cover the residual limb areas that will elicit this rich pattern of information. For TMR patients who have been fit with a traditional myoelectric prosthesis, follow the note above about planning a subset of the pattern recognition electrode sites at the previously-determined, traditional-control electrode sites. Extra care must also be taken with the TMR population during the discussion and palpation phase – because of their reinnervation, muscle contractions that are valuable to control can be often be sensed at unique, and unexpected locations. Remember that it is a goal to locate unique patterns of underlying activity. Finally, some TMR recipients may also have reinnervated

sensation. It can be best practice to avoid these sensitive areas for patient comfort.

### **PLACEMENT OF ELECTRODE CONTACTS FOR PATTERN RECOGNITION**

The discussion, palpation, and planning activities are all intended to help determine the locations on the patient's residual limb that are of interest to be used for pattern recognition control. At the end of the planning, EMG areas of interest should be marked on the patient's skin or loosely transferred to a preliminary check socket. When fabricating the electrode contacts into the system, a number of simplifying considerations can be made. These are what make placement of electrode contacts for pattern recognition forgiving and comprehensive.

#### Single reference contact

Most traditional electrode “pre-amp” or “remote” packages include an electrode contact area that is the reference or “ground” for each myosite. With pattern recognition, only one reference contact for the entire system is required. In Figures 1, 2, and 3, the single reference contact is denoted by the annotated “R”. Placement of this contact is somewhat arbitrary but should follow the general consideration of placement achieving reliable electrode-to-skin contact and not causing the patient discomfort.

#### Alignment direction

A pair of electrode contacts that make up a pattern recognition myosite do not always have to be aligned to match the direction of the underlying muscle fibers. In many placement situations, it may be advantageous to cover areas of multiple underlying muscles or, simply, because of size or shape constraints. Figure 1 shows 5 pattern recognition myosites on a transradial interface where 2 of the 5 are not aligned with muscle fiber direction.

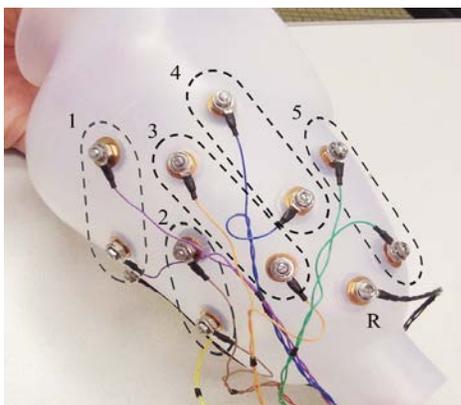


Figure 1: Example of pattern recognition myosite placement for a short transradial: looking at the exterior of the transradial inner-socket interface with the distal end at the bottom right of the image. Here we see 5 of 8 pattern recognition myosites (a pair of electrode contacts for each) and the single electrode contact for the system ground/reference. Note that not all myosite electrode contact pairs are aligned with the muscle fiber direction (specifically myosites 1 and 2).

Inter-contact spacing

Pattern recognition performance improves when as much muscle contraction signal as possible is considered. Widely spacing the pair of electrode contacts for a pattern recognition myosite can ensure this desired condition. As a general guideline, inter-spacing the two contacts anywhere from 30-55mm apart is appropriate. For example, myosite 3 in Figure 1 has the contacts separated by 47mm while myosite 2 has contacts 32mm apart. In Figure 3, however, an interface for a large transhumeral is shown and some of the myosites have inter-electrode spacing of 55mm or greater.

Contact Sharing

For very short or space-constrained residual limbs, an acceptable and common practice in placing myosites for pattern recognition is to share some electrode contacts between myosites. Figure 2 illustrates this in an example where myosites 2 and 3 both employ the same distal electrode contact.

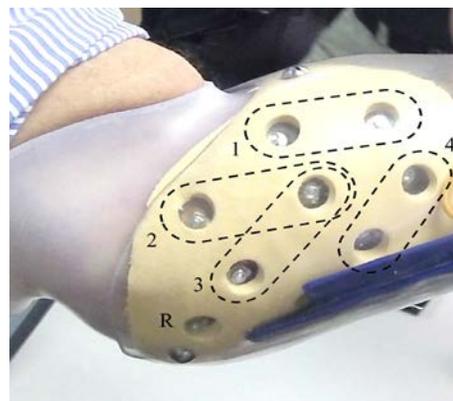


Figure 2: Example of pattern recognition myosite placement for a transradial: looking at the exterior of the transradial socket as assembled. Here we see 4 of 8 pattern recognition myosites (a pair of electrode contacts for each). Note that myosites 2 and 3 are sharing a common electrode contact.

General Distribution

In many pattern recognition fittings, it is acceptable to place the myosites at a very generic or symmetrical layout. Commonly, this can be a good approach to ensuring coverage of all residual limb muscle signals while simplifying planning and fabrication. Many patients will enjoy the same performance from their prosthesis when this placement practice is used in place of iterative, time-consuming myotesting. Take caution, however, and use the discussion, palpation, and planning phases modify the symmetric placement as required. Figure 3 illustrates myosite placement for a transhumeral TMR amputee – 8 myosites have been symmetrically placed with a slight gap to relieve an area of humerus loading under shoulder flexion.

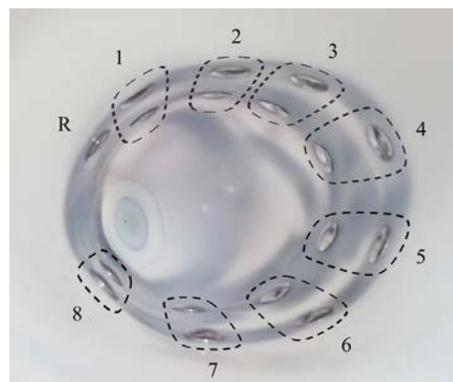


Figure 3: Example of symmetric pattern recognition myosite placement: looking into a transhumeral socket with the medial side at the bottom of the image. Here we see a generally-symmetric orientation of 8 pattern recognition myosites (a pair of electrode contacts for each) and the single electrode contact for the system ground/reference. Note the slight avoidance of the superior humerus loading zone.

## CONCLUSION

The recent commercial availability of pattern recognition control to upper-extremity prosthetics brings great promise for improved control, device acceptance and reach to more potential myoelectric patients. Pattern recognition, however, necessitates learning some new, clinically-practical approaches to myosite location and electrode placement. While pattern recognition control requires an increased number of electrode sites, it offers the concession that locating and placing these can be forgiving and simplified.

## ACKNOWLEDGEMENTS

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## THE “INTERNET OF THINGS” TO QUANTIFY UPPER LIMB PROSTHETIC USE: A NOVEL OUTCOME TOOL

Luca Lonini, Jose Mauricio Ochoa, Chaithanya Krishna Mummidisetty, Arun Jayaraman  
*Rehabilitation Institute of Chicago*

### ABSTRACT

Despite significant technological advancements in upper limb prosthetic designs, there still exist a lack of quantitative information on how these advanced devices are used at home and in the community. Currently, a number of outcome measures exist, which attempt to characterize different domains (function, activity, participation) and aspects (quality of life, capability of operation, etc.) of the prosthesis use. Traditionally, these measures are questionnaires or scoring tests, which are subjective, qualitative and affected by rater bias and recall bias. This limits the ability to specifically quantify everyday home and community prosthetic device use. We propose a novel outcome tool to characterize upper limb prosthetic use based on the “Internet of things”: a network of wireless motion or proximity sensors that can be attached to different objects is used to detect when the user is performing a specific ADL using a prosthesis. Twelve upper limb prosthesis users (using a two degree of freedom wrist) and twelve able-bodied are asked to execute a set of three ADLs (e.g. preparing breakfast, brushing teeth, doing the laundry) in a simulated home environment. The wireless sensors, attached to objects and to the participants arm, collect data about number of times the ADLs are performed, time spent to perform each activity, in addition to quantifying the movement characteristics (e.g. degree of freedom). This information is used to compare performance across different users and across the two groups, as well as with the score obtained with traditional tests (e.g. OPUS). We are able to reliably detect and time the performed ADLs and evaluate the proficiency of each user based on the movement data. Our results indicate that the “Internet of things” is a promising tool to characterize everyday home use of a prosthetic device and allows collecting information over extended periods of time. Future work will involve testing of this application for users at home.

## COMPUTER VISION FOR AUTOMATIC CONTROL OF ORIENTATION AND PRESHAPE IN A HAND PROSTHESIS WITH AN ACTIVE WRIST

Marko Markovic<sup>1</sup>, Bernhard Graimann<sup>1</sup>, Strahinja Dosen<sup>2</sup> and Dario Farina<sup>2</sup>

<sup>1</sup>*Translational Research and Knowledge Management, Otto Bock HealthCare GmbH, Duderstadt, Germany. Email: {marko.markovic, bernhard.graimann}@ottobock.de*

<sup>2</sup>*Department of Neurorehabilitation Engineering, University Medical Center Göttingen, Georg-August University, Göttingen, Germany. Email: {strahinja.dosen, dario.farina}@bccn.uni-goettingen.de*

### ABSTRACT

We present here a method for the automatic control of orientation of a prosthetic hand with an active wrist. The orientation controller was integrated into a previously developed system for the semi-autonomous control of preshaping of the hand prosthesis. The overall system utilizes computer-vision algorithms coupled with wearable augmented reality glasses and a simple myoelectric control in order to provide automatic preshaping and rotation of a multi-degree of freedom prosthetic hand. The control of orientation was tested in 5 able-bodied subjects that used the system to grasp 10 differently oriented objects in two consecutive sessions (100 trials in total). The overall mean error in orientating the prosthesis with respect to the target object was  $9 \pm 5^\circ$ . Importantly, in all the trials, the control of orientation was precise enough for the subjects to accomplish the task without correcting the system. Therefore, these results demonstrate the general feasibility of the proposed control concept.

### INTRODUCTION

The available flexibility of the modern hand prostheses [1], [2] is largely unexploited as the commercial state-of-the-art human machine interfaces cannot easily accommodate the control of multiple degrees-of-freedom [3]. In order to tackle this problem some novel, unconventional control interfaces were recently proposed [4].

We have previously developed a system for closed-loop semi-autonomous control of prosthesis preshape [5]. The system utilized point-cloud analysis of the scene observed through the stereo glasses worn by the user in order to extract information about the targeted object size and shape in 3D space. This information was then processed by an artificial controller, which used a set of heuristic rules to select the hand preshape (grasp type and size) suitable for grasping the object.

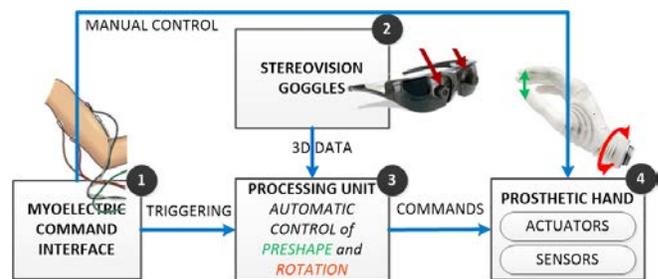


Figure 1: System structure depicts connections and workflow between key components. The system controls both prosthesis preshape (green arrow) and rotation (red arrow), but also allows manual override thus effectively implementing the semi-autonomous control.

In the current study, we present a novel method for the automatic control of an active prosthetic wrist (rotator). Therefore, the artificial controller now implements autonomous rotation control, thus yielding not only hand preshape but also posture which is convenient for the grasp. We then assessed the performance of the orientation control by conducting an experiment in able-bodied subjects that used prosthesis to grasp differently oriented target objects.

### METHODS

The system consists of the following components (Fig. 1.): 1) *Simple two-channel myoelectric command interface* (Myobock, OttoBock Healthcare GmbH) for the triggering of the automatic operation and for manual control; 2) *Augmented reality (AR) glasses* (Vuzix, iWear920AR) with embedded stereo cameras continuously feeding the system with the stream of stereo images; 3) *Processing unit* (standard laptop), analyzing the acquired images, segmenting and modeling the targeted object and applying heuristic analysis in order to determine appropriate prosthesis preshape and orientation; 4) *Michelangelo prosthetic hand* (OttoBock Healthcare GmbH,

Germany) implementing lateral and palmar grasps and equipped with additional rotation unit (pronation/supination).

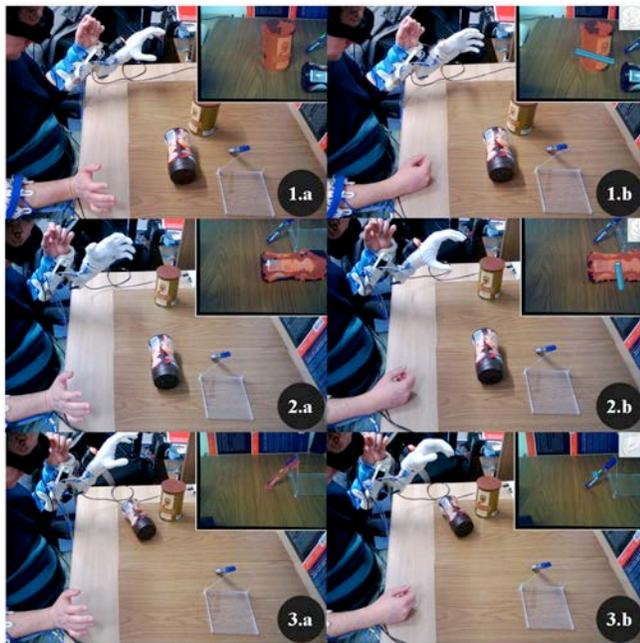


Figure 2: The user operating the system in a cluttered environment consisting of three differently oriented objects. Upper right corner of each sub-picture depicts which object was being targeted (via the AR glasses). Each of the panels (1 to 3) shows one grasping trial. The left panels (1-3.a) capture the moment user triggered the system by extending the contralateral hand. The right panels (1-3.b) depict the outcome of the automatic control (see text).

In the test setup for able-bodied subjects, the user wore the AR glasses, together with the prosthetic device attached to his left/right arm using a custom-made orthopedic splint. Two pairs of dry bipolar electrodes were placed over the flexor/extensor muscles of the contralateral arm. Contralateral arm was used here only for convenience, i.e., to avoid movement artefacts that could arise if the electrodes were under the splint. The operation sequence started with the user looking into an object of interest and then generating a simple myoelectric trigger signal (activation of the forearm extensor muscles), after which the system responded by automatically adjusting the prosthesis hand preshape and wrist rotation. After this automatic adjustment, the user resumed full manual control over the prosthetic device. He/she could therefore use myoelectric interface to close the hand and grasp the object, or fine tune and/or correct the automatic decisions (i.e., changing the grasp type, size, or orientation). Finally, after grasping and releasing the object, the operation sequence restarted.

In order to determine the hand orientation (wrist rotation), the artificial controller employed computer vision to represent the target object using a set of predefined geometrical models (box, cylinder, line, sphere [5]). Having this contextual information available allows for a more sophisticated identification of the object surface patch that is going to be grasped (e.g. object facets that are occluded by the other object, or that are facing away from the user are automatically ruled out). With it the controller could estimate the orientation of the target object in space, and based on the current orientation of the hand, calculate the angle of wrist rotation. Several examples of automatic rotation and preshaping are illustrated in Figure 2, in which the user operated the system to grasp several objects placed on the table surface. First (Fig. 2[1.a-b]), the user targeted a large cylindrical object placed vertically on the table surface. The user triggered the automatic control (Fig. 2[1.a]) and the system responded by preshaping the hand into the palmar grasp, opening it with a large aperture, and rotated the wrist until it became approximately perpendicular to the table surface (Fig. 2[1.b]). The hand was therefore ready for the grasp, i.e., correctly preshaped and aligned with the object axis. The user then targeted another large cylindrical object now placed horizontally (Fig. 2[2.a]). The system again preshaped the hand into a palmar grasp with wide aperture. However, it rotated the wrist until the hand became approximately parallel to the table surface (Fig. 2[2.b]). Finally, a thin pen inclined at 45° was targeted (Fig. 2–[3.a]). Since the initial position of the hand was horizontal, the system responded by rotating the wrist for approximately 45° and it selected the lateral preshape with a small aperture (Fig. 2[3.b]).

Five able-bodied subjects (26±2 years) participated in the experiment and signed an informed consent that was approved by the local ethics committee. The subjects were seated in a chair facing the desk surface on which the objects were presented. The prosthesis was mounted on the left forearm, resting at the beginning of each trial at a predefined location (parallel to the table surface). The experimental protocol consisted of ten trials repeated twice, thus yielding 100 trials in total (5 subjects x 10 trials x 2 repetitions). After each trial, the prosthesis automatically returned into the neutral position, parallel to the table surface. In each trial, the subjects were presented with an object in one of the three possible orientations, i.e., 0°, 45°, or 90° with respect to the table surface. They were instructed to look towards the object, while keeping the prosthesis at the resting pad, then trigger the system, wait for the prosthesis to preshape/rotate, and then finally grasp and transport the object to the designated location. If the users judged that the prosthesis was not oriented properly, they were instructed to correct the orientation manually (using myocontrol). For each trial, the outcome measure was the absolute difference between the

actual prosthesis rotation, as selected and implemented by the automatic controller, and the desired (ideal) prosthesis orientation which was equal to the orientation of the target object, measured in degrees.

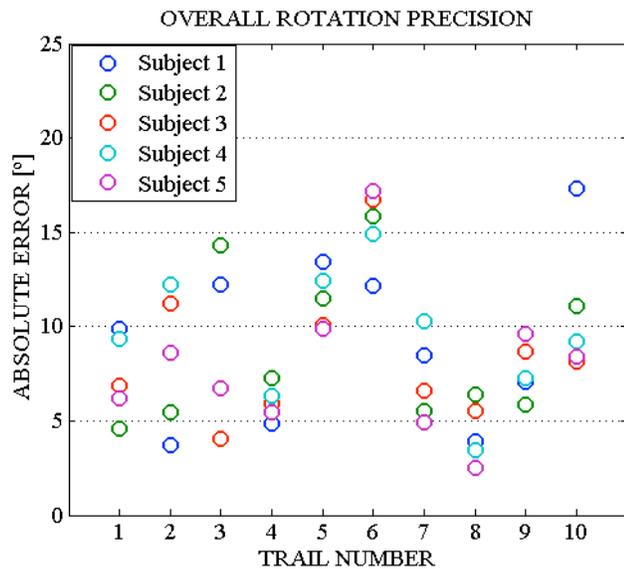


Figure 3: System precision in adjusting the hand rotation across the trials. The error was calculated as the absolute value of the difference between the prosthesis orientation selected by the automatic control and the actual object orientation. Note that the error never surpassed 20°.

## RESULTS

The subjects successfully accomplished the task (grasp, transport and release) in all the trials. The average orientation error (mean±standard deviation) of the automatic control across all subjects and objects was  $9^{\circ}\pm 5^{\circ}$ . The results from individual trials are presented in Figure 3. Note that in 90% of the trials the orientation error was less than 15°. The highest registered error was 18°.

## DISCUSSION

The ultimate goal of this research is to simplify the control of a multiple degree-of-freedom prosthesis by endowing the artificial controller with the ability to sense the external environment and make autonomous decisions. More specifically, we have demonstrated how an artificial controller can be enriched with an additional, non-conventional information source (stereo cameras) and a high level processing (cognitive-like reasoning) to achieve fully automatic control of both prosthesis preshape and rotation. Therefore, high-level tasks can be accomplished automatically, which decreases the cognitive burden from the

user, with the final goal of making the grasping a routine, effortless activity.

The overall results demonstrated good system reliability, i.e., in 90% of the trials the orientation error was lower than 15°. Importantly, in all the trials, the error was low enough for the subjects to successfully accomplish the task without the need to implement any corrections. The performance can be further improved by employing more sophisticated computer vision algorithms. Moreover, when necessary the erroneous system decisions can be manually corrected using myoelectric interface (semi-autonomous control).

The presented control concept can be easily scaled-up to more complex systems. For example, in the case of a high-level amputation (i.e., full upper limb prosthesis), the 3D contextual data obtained through computer vision could be fused with inertial sensors that are tracking the prosthesis orientation in space. In this scenario, the user would only need to trigger the system in order to perform both hand preshaping and orientation as well as reaching for the target object.

## ACKNOWLEDGEMENTS

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## LOW-COST, COMPLIANT CONTACT SENSOR FOR FRAGILE GRASPING WITH REDUCED COGNITIVE LOAD

Blaine Matulevich<sup>1</sup>, Vikram Pandit<sup>1,2</sup>, Chia-Hsien Lin<sup>1</sup>, Gerald E. Loeb<sup>1,2</sup>, and Jeremy A. Fishel<sup>1</sup>  
<sup>1</sup>SynTouch, Los Angeles, CA, USA, <sup>2</sup>University of Southern California, Los Angeles, CA

### INTRODUCTION

Myoelectric prosthetic hands have a long history of impressive technological development but low functional utilization and acceptance [1]. Substantial research has gone into the development of technologically advanced prostheses; but while costs of care have skyrocketed over the decades [2], no observable progress has been made in improving the practical functionality of these devices, their user satisfaction, or quality of life for their users [3]. Before adding even more technology, it is important to understand problems that lead to this unsatisfactory situation:

1. Operating a myoelectric hand requires a lot of mental effort, especially for tasks requiring precision (e.g. grasping delicate objects); variance in performance in these tasks creates a lack of confidence in the hand's performance.
2. Adding technology makes any device more complex, hence expensive and prone to breakage.

To be successful, new technologies must address problem 1 without adding to problem 2. Problem 1 arises from the limited sources of command signals that can be used to control a prosthesis. As prostheses become more anthropomorphic with multiple degrees of freedom (iLimb, Touch Bionics; BeBionic, RSL Steeper; Michelangelo, OttoBock; etc.), the limitation in command signals becomes a bottleneck in their functional utility, with little improvement in performance seen over simpler devices [4]. Targeted muscle innervation to expand the number of command signals shows substantial promise [5], but may prove to be too costly and invasive for widespread adoption, particularly in less-severe amputations. As it has been reported that acceptability of prosthetic hand technology is more dependent on the required attention than the success in grasping [6], making prosthetic devices more intuitive to control should remain a primary objective.

Fragile and precise grasping are among the most difficult and cognitively demanding tasks for prosthetic hand users. With currently available technologies, even inconsistent performance handling fragile objects requires substantial patience and visual attention, resulting in high cognitive

load. To perform well, EMG signals must be precisely timed when grasping these objects; even small errors can result in incomplete grasps or undesired high stalling forces when the fingers close. Consequently, unilateral amputees prefer to use their intact hand for most tasks, especially those involving fragile objects. Able-bodied subjects have no difficulty in grasping fragile objects due to the wealth of tactile feedback available during these tasks [7], [8]. Conversely, even the fully intact human hand with its high level control is almost useless in the absence of tactile feedback [9]. Various sensing technologies have been developed to bring human-like tactile sensing to robotics [10]-[12], yet few sensors meet the unique specifications demanded in prosthetic applications. This study presents a biologically inspired method to enable fragile grasping of objects by combining compliant tactile sensors with a biomimetic contact detection reflex.

### METHODS



Figure 1: Left - a BioTac® tactile sensor; Right - a low-cost NumaTac® prototype tactile sensor.

In previous work we had explored the benefits of human-like tactile sensing in prosthetic hands for reflexive grip control [13] and tactile perception [14] with the BioTac sensor (Figure 1, SynTouch LLC, Los Angeles). The BioTac is a finger-like compliant tactile sensor capable of sensing much of what human fingertips can sense: normal and shear forces [15], [16], point of contact [17], vibrations [18], [19], and temperature [20]. While the complexity and cost of this device make it poorly suited for a commercial prosthetic hand, it was useful as a research tool to identify which of these sensory modalities could enhance prosthetic hand function.

Our findings indicated that the performance and reliability of grasping fragile objects could be greatly improved while simultaneously reducing the cognitive load (addressing problem 1 as stated above) using only a small subset of the BioTac's capabilities (compliance and sensitivity to contact) [13]. To address problem 2, we developed a simplified version of this sensor to provide these specific capabilities.

#### The NumaTac Sensor

The NumaTac (Figure 1) is a low-cost and compliant tactile sensor that provides sensitive contact detection. It consists of a rigid bone-like core covered with open-cell reticulated foam. The foam is self-skinning and sealed with a polyvinyl fluoride to trap the air inside the sensor. A pressure sensor embedded into the core and sealed with a silicone gasket records the pressure inside the foam. When the sensor makes contact with an object, the sensor detects the resulting pressure increase inside the foam. The NumaTac possesses similar sensitivity to contact as the human fingertip and the BioTac [18], but cannot resolve the location or direction of contact. It can also be molded into almost any desired shape.

#### Contact Detection Reflex

Humans are capable of quickly grasping objects without excessive forces. This is enabled by specialized cutaneous receptors and spinal circuitry that can detect contact and inhibit further activation of the muscles closing the fingertips [7]. Excessive forces are typically not a concern to prosthesis users when handling rigid non-fragile objects (operators typically send large EMG signals, letting the motors stall on the object). When handling fragile objects, however, the user must close the fingers slowly with small EMG signals until stable contact can be confirmed visually. Relatively large command signals may be required to overcome friction and the amplitude of the user-generated EMG signal tends to be noisy, making this process slow, difficult to control, and heavily reliant on visual feedback and attention. This can be greatly simplified by artificially mimicking the above-described inhibitory reflex [13].

To achieve this desired reflexive behavior, a state change was implemented to reduce the gain of the EMG signal delivered to the prosthetic controller upon sensing contact in opposing fingertips (Figure 2). Fluid pressure was used to detect this contact for each sensor (liquid in the BioTac, and air in the NumaTac). When there was no contact, control signals had unity gain to make the hand more responsive and easier to close at faster speeds. After contact was detected, this gain was reduced to 0.3 (determined by user preference). For low EMG levels this would cause the motor to stall on the object with low but predictable force, dependent on the EMG level, closing speed before contact, a small delay in the feedback loop and the compliance of the fingertips. As EMG signals were not abolished upon contact, the prosthesis

operator maintained full control over the stalling force and was capable of closing with high forces with elevated EMG signals. While simplistic in nature, this approach was found to dramatically improve speed and reduce performance variability in repeated tasks [13].

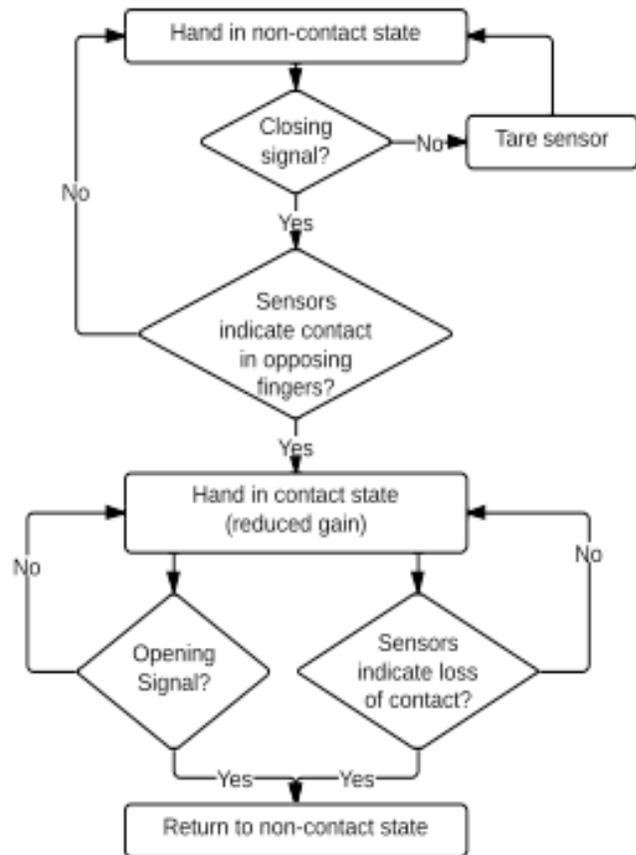


Figure 2: Control algorithm with contact detection

#### Hardware

Similar methods to those described in [13] for evaluating the BioTac in fragile grasping performance were used to also evaluate the NumaTac sensor in this study. Relevant aspects of these methods are summarized herein.

In previous work, specialized mechanical adapters were fabricated to install the BioTacs onto a commercially available 1-DOF myoelectric hand (MC Hand, Motion Control). Fixtures to attach the NumaTac sensors to these same adapters were made to facilitate switching between sensors. In all tests a sensor was placed in the thumb and index finger to detect opposing contact. The cosmesis was removed along with the passively coupled ring and pinky fingers and a non-functioning fingertip was installed on the middle finger, although if desired another sensor could be used with no major changes to the algorithm.

EMG signals were taken directly from the pair of electrodes in the subject's prosthetic socket used to control his regular prosthesis (13E200 MYOBOCK® Electrode, OttoBock). The electrodes have adjustable gain and filtering developed by OttoBock, designed to provide a DC voltage in proportion to muscle activation to control the prosthetic hand. The contact detection algorithm, previously developed in computer software, was programmed onto an electrical board to improve portability and reduce latency.

#### Experimental Comparison

Three experiments were designed to test the speed, accuracy, and ease with which fragile grasping activities could be performed. These tests utilized simple objects that a prosthesis user could expect to encounter in everyday scenarios, as identified by our subject. A fourth experiment was designed for evaluating performance when handling rigid objects to evaluate whether this controller might impede non-fragile grasping tasks. The following tests were performed (Figure 3):

- i. Pick up ten foam packing peanuts from a table and place them into a container as quickly as possible. Peanuts gripped with excess force ( $\sim 3\text{N}$ ) would break and would not count towards the total.
- ii. Grasp ten crackers handed to the user by the experimenter, and place them into a container as quickly as possible. Two variants were run with importance placed on either speed or accuracy. In the speed trials, crackers that were broken ( $\sim 5\text{N}$ ) did not count towards the total. In the accuracy tests, broken objects resulted in a failed trial and the entire trial would be repeated.
- iii. Move nine eggshells between cartons as quickly as possible. Broken eggs ( $\sim 25\text{N}$ ) did not count towards the total. In a variation with distraction, the subject was asked to simultaneously spell a series of words.
- iv. Grasp and move ten unopened soda cans across a table as quickly as possible. This activity was designed to compare performance on rigid grasping.

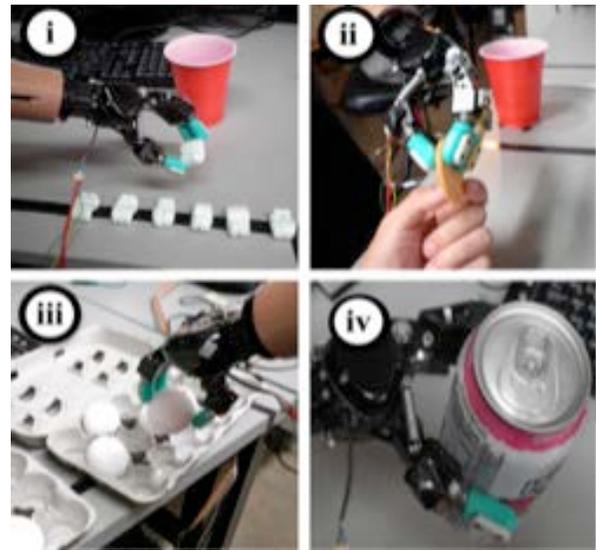


Figure 3: Grasping Experiments

Performance was tested with one subject, a 22 year-old male, congenital, unilateral, trans-radial amputee and myoelectric prosthesis user. The subject was compensated for his time during testing and development. Each task was performed by the subject with 1) his own prosthesis (VS, VariPlus Speed, OttoBock), 2) the BioTac-equipped hand with contact detection algorithms (BT), 3) the NumaTac equipped hand with contact detection algorithms (NT), and 4) his intact contralateral dominant hand (DH). For each experiment, the subject was allowed to train until his performance became steady, then 5 trials were recorded.

## RESULTS

In every timed fragile-grasp task (i-iii), the subject's personal prosthesis without compliance or tactile sensing (VS) had the worst performance (Figures 4-6). The BioTac equipped hand with contact detection (BT) and the NumaTac equipped hand with contact detection (NT) were always better than the subject's personal prosthesis and neared the performance of his dominant hand (DH).

### A. Performance on Timed Grasping Tasks

The performance index normalized by the time to complete the task with the subject's dominant hand (DH) is presented in Table 1. The subject's personal prosthesis (VS) scored as poorly as 4.82 times slower than the dominant hand and was never better than 2.45 times slower at fragile grasp tasks. Additionally, the subject repeatedly broke objects with his personal prosthesis (VS): 1.2 foam peanuts per trial, 2.8 crackers per trial on the speed test (and 12 failed trials on the accuracy test), and .4 eggs per trial both with and without distraction.

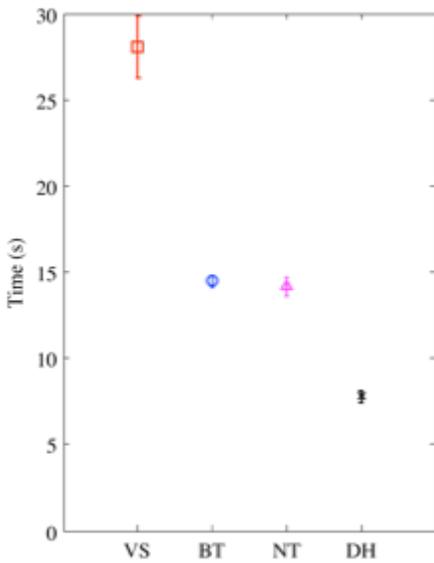


Figure 4: Foam Experiment

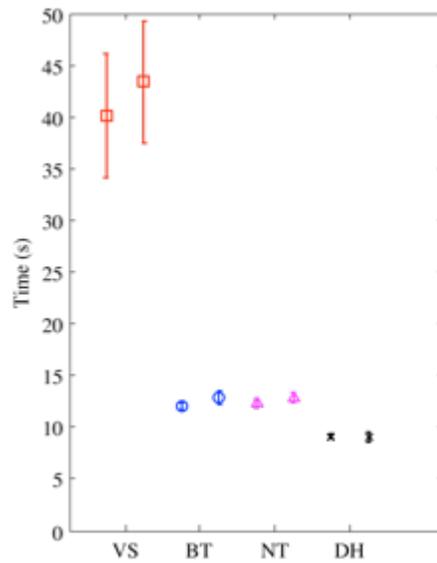


Figure 5: Cracker Experiment  
Left = speed trials  
Right = accuracy trials

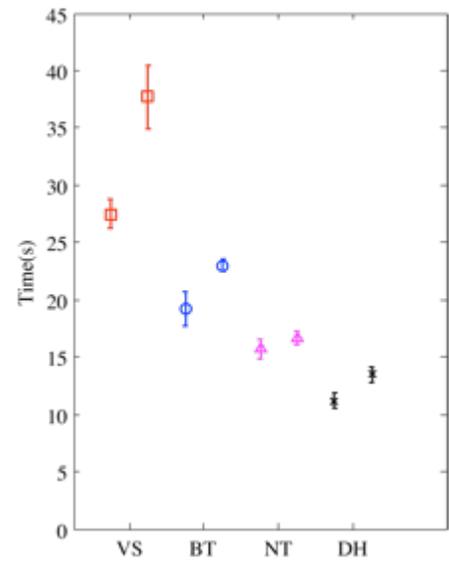


Figure 6: Experiment.  
Left = trials without distraction  
Right = trials with distraction

Compliant sensors with contact detection brought performance closer to biomimetic performance, with one trial requiring only 1.25x the time for the dominant hand. Additionally, the variance in the subject's performance was much lower with contact detection than without for all fragile grasping tasks using both the BioTac and the NumaTac sensors. With contact detection, the subject broke only one object during testing (a cracker, using BioTacs).

A one degree-of-freedom ANOVA test was carried out to verify the improvements each method provided. Both methods of contact detection outperformed the VariPlus Speed hand at a high confidence level ( $P < 0.01$ ) for every timed fragile-grasp task. None of the tested prosthetic systems were significantly different ( $P > 0.10$ ) on the soda test, suggesting that no performance was lost with either sensor on rigid grasping tasks.

Table 1 – Summary of Results

	VS	BT	NT	DH
Foam	3.59	1.85	1.82	1
Crackers - Speed	4.41	1.32	1.36	1
Crackers - Accuracy	4.82	1.43	1.43	1
Eggs - No Distraction	2.45	1.71	1.40	1
Eggs - Distraction	2.79	1.70	1.24	1
Soda	1.86	1.86	1.76	1

## DISCUSSION

Compliant contact detection sensors not only provided significant speed improvements when compared to the subject's regular prosthesis, but also reduced the variance of performance to levels near that of the subject's dominant hand. Variance is an important factor as a hand with inconsistent performance reduces operator confidence. The subject reported that the compliant sensors made stable grasps much easier to achieve due to the reduced need for precision and the mitigation of force overshoot. Because of this, he felt confident to move more swiftly during grasping activities after minimal training.

The results of the distraction task show that the cognitive burden of the subject was greatly reduced by our contact detection algorithm. While his performance worsened on all four control strategies (evidence that he was distracted),

with his own prosthesis the relative performance change was drastic and the variance became much greater, while with contact detection the amount of performance loss and the variance were similar to that of his dominant hand (Figure 6). This suggests that using the prosthesis with contact detection substantially reduced the cognitive burden of the operator when compared to his personal prosthesis.

Despite its simplicity, the NumaTac sensor proved to be a good substitute for the BioTacs. The performance with NumaTacs was significantly better than with BioTacs on both egg grasping tasks and statistically the same on other tasks. The difference in performance between the sensors may be due to their physical properties. It was observed that the fluid-filled BioTac has a short range of very high compliance that decreases abruptly after the skin contacts the rigid core; the foam-filled NumaTac has a larger range in which the stiffness slowly increases as the foam is compressed. Further studies will be needed to evaluate.

Too little progress has been made on facilitating prostheses' most important task – grasping objects. Here we have presented two simple ideas that appear to offer a real improvement in the usability of prosthetic technology. Compliance is a biomimetic property that can easily be applied to nearly any existing prosthesis, at a significant gain of function. Contact detection reflexes similar to biological reflexes can also be used to improve performance and consistency during everyday tasks, allowing the prosthetic hand to be both quick and delicate as well as intuitive and natural for the user. These principles can also help make prosthetic technology more affordable by providing superior functionality with inexpensive actuator technology and avoiding the need for tactors or haptic displays. In a low-cost prosthesis, we were able to obtain performance that is unattainable to date in many expensive research hands. Next steps will be to attach a smaller version of the NumaTac directly to the metal skeleton of the prosthesis under the cosmesis so that it can be tested under normal field conditions by a larger number of subjects.

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## WIRELESS IMPLANTABLE MULTICHANNEL MYOELECTRIC SYSTEM

Daniel McDonnall, Daniel Merrill, Christopher Smith, Kenneth Shane Guillory  
*Ripple*

### ABSTRACT

Control of prosthetic arms has been limited by the small number of inputs that are used to control multiple degrees of freedom in the limb. We are developing an implantable multichannel myoelectric device to detect signals from multiple residual muscles that will be sent wirelessly to the prosthetic limb. This approach offers the advantages of recording more channels of isolated muscle signals and providing access to deep muscles that cannot be detected with surface electrodes. We report the results of work to verify the in vitro performance of the system and an in vivo trial to validate device function in an animal model.

The implant was constructed on a circuit board with a bioamplifier ASIC and additional discrete components. The implant was inductively powered by an external transceiver, and digitized signal data were sent from the implant by reflected impedance modulation. Each implant included four pairs of electrodes in epimysial disc, intramuscular bands, and fine wire configurations. The electronic components and ASIC die were coated with a conformal electronics sealant, and the entire assembly was coated in silicone.

Benchtop performance was verified in a dry configuration and while the devices were soaked in saline. The amplifier was shown to have an input-referred noise of  $2.2 \mu\text{VRMS}$ , a common mode rejection ratio greater than 55 dB, and neighboring channel isolation averaging 66 dB.

These prototype implants were validated in a six-dog study at the University of Utah. Two four-channel devices were implanted bilaterally in the front limb by placing the electronics package behind the shoulder with electrodes implanted in deltoideous and lateral head of triceps. One week following implantation, each animal was fitted with a backpack carrying an external transceiver coil and a battery-powered data acquisition system, and the dogs were allowed to freely walk down a hallway. EMG recorded from each animal as it walked down a hallway had very low noise and, in conjunction with recorded video, clearly indicated swing/stance phases of gait.

This study demonstrates this design can be used to amplify and transmit muscle signals. This approach has the potential to substantially improve the control of prosthetic limbs by providing simultaneous, multi-degree of freedom control, especially if used with advanced prosthetic arms/hands, targeted muscle reinnervation patients, and pattern recognition algorithms.

## REDEFINING THE NORM: OBJECTIVE FUNDING DEVELOPMENT IN AN EVER CHANGING PAYER ENVIRONMENT

John Miguelez, Tiffany A. Ryan  
*Advanced Arm Dynamics*

### ABSTRACT

#### Introduction:

To fund development of and support reimbursement for advanced upper limb prosthetic technologies, it is incumbent upon upper limb (UL) prosthetic industry leaders to provide evidence based performance data to payer and referral sources. Clinically relevant outcome measure results influence payers' motivation to reimburse for current technologies. Clinicians specializing in UL prosthetic rehabilitation are experts in the intricacies of UL technologies and treatments and most qualified to serve as payer source advisors. An industry standard, definitive outcome measures to evaluate successful UL prosthesis use remains elusive. [1, 2] To address the deficiency of performance data and the efficacy of UL prosthetic terminal devices technologies, Advanced Arm Dynamics initiated a study utilizing objective measures to evaluate patient performance with, and perceptions of, electric terminal devices.

#### Methods:

This longitudinal study includes outcomes measurement of 30 subjects with transradial amputation utilizing the following electric terminal devices; hook, "standard" tripod grip hand or multiarticulating hand. Qualitative and quantitative measures utilized are the DASH-Disability of the Arm Shoulder, TAPES-Trinity Amputation and Experience Prosthesis Scales-Revised, Box and Blocks, SHAP-Southampton Hand Assessment Procedure, and internally developed clinical measures. These measures were administered at the preparatory and definitive fitting stages.

#### Results:

Results indicate the electric hook then the multiarticulating hand with a quantitative advantage over the standard hand. Qualitative results for the multiarticulating and standard hands are deemed as preferable to those of the electric hook

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## EVALUATION AND RECORDING OF USE IN A TRANSHUMERAL TMR HOME-TRIAL

Laura A. Miller, PhD, CP,<sup>1,2</sup> Blair A. Lock, MS, PE,<sup>1,3</sup> Levi J. Hargrove, PhD, PE,<sup>1,2</sup> Suzanne Finucane, PTA,<sup>1</sup> Kristi Turner, OTR/L,<sup>1</sup> Jon Sensinger, PhD<sup>4</sup>  
(1) *Center for Bionic Medicine, RIC*, (2) *Northwestern University*,  
(3) *Coapt LLC* (4) *University of New Brunswick*

### INTRODUCTION

Many above-elbow myoelectric prosthesis users have low tolerance of their prostheses citing poor control. [1, 2] Targeted Muscle Reinnervation (TMR) is a surgical technique well suited to improve the ability for upper-limb amputees to intuitively control myoelectric devices, especially when pattern recognition is applied[3].

This work follows up on another recent study in which we reported statistics for pattern recognition control usage in single subject [4]. In this study we compare pattern recognition (PR) to direct control (DC) in a second subject, based on actual motor thresholds.

### METHODS

The subject gave informed written consent for a research study approved by the Northwestern University's Institutional Review Board. His prosthesis comprised a powered elbow (Boston Digital Arm), powered wrist rotator (Motion Control), and electronic terminal device (Otto Bock hand). A custom microcontroller was used that could provide DC or PR control. For the DC phase of the study, 2 EMG sites were used for hand open and close and 2 used for elbow flexion and extension. The subject switched between hand and wrist operation by providing a quick elbow extension signal. Once in wrist rotation, he could switch back to hand or the system would time out back to hand after 10 seconds. The location of the sites was evaluated and agreed upon by 3 certified prosthetists. The configuration of the control was set by 2 certified prosthetists. For the PR phase, 8 EMG sites were used as input to a pattern recognition algorithm. The subject could "recalibrate" the device at home as often as he felt necessary[5, 6]. The custom microcontroller then sent messages through the Boston Digital Arm to drive the motors.

The subject had the arm at home for approximately six weeks with each control method (DC first, PR second). Usage/log data was recorded and stored on the controller's memory for later analysis. Data recorded included the total time the system was powered on and for each powered on

cycle, the counts of actuation commands, binned into 5% speed groups (0-100% available output command speed), for the 6 motions. During the DC phase, the number of switch events during each powered on cycle was also recorded. Each on-cycle resulted in one line of data. Data were binned continuously and the updated bin counts were recorded to the file on the controller once per minute. Therefore, if the arm was turned off, the most data that could be lost would be the previous 59 seconds.

### DATA PROCESSING

Motor commands were processed by the custom controller and then sent to the intrinsic controller of the Boston Digital Arm. The settings of the Boston Digital arm remained constant throughout the home trials.

Due to intrinsic inertia of the motors, there was a threshold of movement in the motor commands. That is, a certain level of signal was required before the motors would actually start to move. In order to process the data accurately, the prosthesis was controlled manually through the data processing system. Motor commands of various motor percentages were sent to the prosthesis until it began to move. These values were recorded and used to threshold the data.

Table 1: Motor power thresholds vs. data processing thresholds

Degree of freedom	Actual manual motor drive threshold (%) *	Data processing threshold (%)
<b>Elbow flexion</b>	17-18	20
<b>Elbow Extension</b>	9	10
<b>Pronate</b>	19	20
<b>Supinate</b>	18	20
<b>Hand Open</b>	19-20	20
<b>Hand Close</b>	24	25
* A range in value indicates threshold varied over position (e.g. full extension vs 90 degrees)		

Only data logs from prosthesis wear times exceeding 10 minutes were considered (this resulted in the exclusion of 50 minutes of DC data and 47 minutes of PR data compared

to over 250 hours of use with each). As described above, actuation commands less than those measured were grouped with the No-motion data, as these were not likely to cause prosthesis movement. For each phase there was a window of time when the arm did not function and was in repair. The subject was also seen in the laboratory for outcome measure evaluation before and after each phase. These times were not considered in the evaluation. Given that data are only evaluated on one subject, no statistical analysis has been done.

**RESULTS**

During the DC phase, the prosthesis was used for an average of 189 minutes per donning and a total wear time of 263 hours spread out over 52 days. The prosthesis was actuated 8.64 % of the time worn. On average, there were 5.67 impulse switching events per hour.

During the PR phase, the prosthesis was used for an average of 208 minutes per donning and a total wear time of 255 hours spread out over 57 days. The prosthesis was actuated 4.44 % of the time worn.

Figure 1 shows the relative actuation of each of the motions for PR control. This graph shows, for each degree of freedom, the total actuation decisions for that movement divided by the total actuation signals for all movements, greater than the threshold. Therefore, the hand was the most used degree of freedom (accounting for over 50% of the total actuation signals for Hand open and Hand close), followed by wrist rotation, and then elbow.

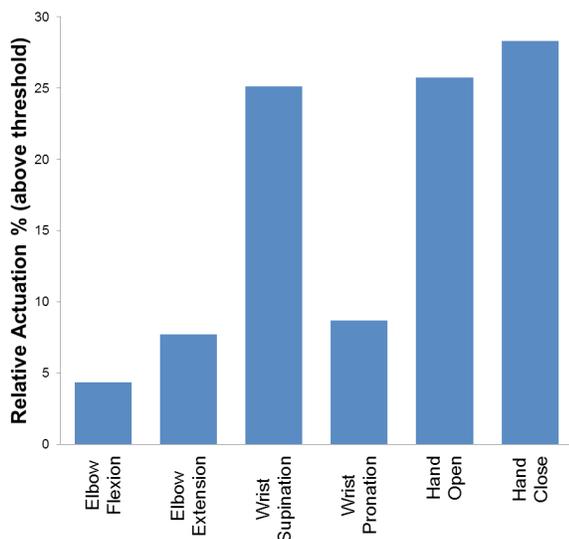


Figure 1: Relative amount of actuation of each motion for the total amount of actuation when using PR control

The same graph for DC control shows an excessive amount of Hand Open (>80% relative actuation). This corresponds to the subject’s feedback during weekly calls that he often had inadvertent opening of the hand.

Figures 2 and 3 show the distribution of the actuation speed for PR and DC directly. For each DOF, each “bin” of speed was divided by the total number of commands for that DOF above the threshold. Both PR and DC show a distribution with a tendency toward slower actuations, indicating good control. However, for DC control, there is a spike in speeds for Hand Open and Wrist Supination (also technically a Hand Open signal). This also correlates to the subject feedback that he often had inadvertent opening of the hand.

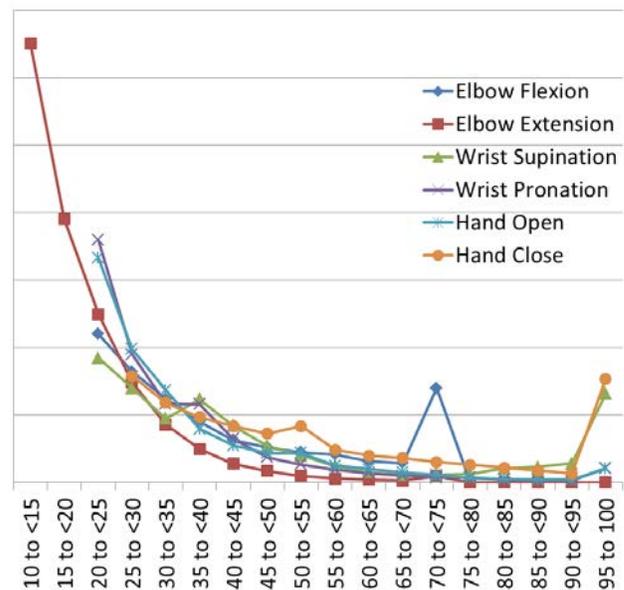


Figure 2: Relative distribution of motor command signal magnitude when using PR control.

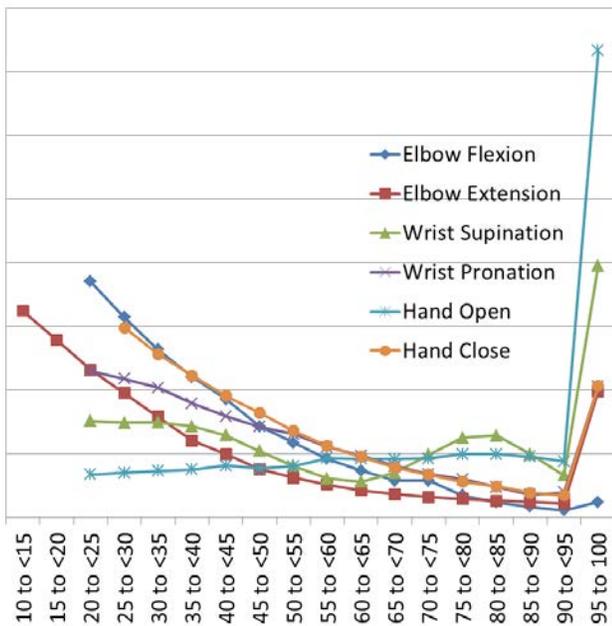


Figure 3: Relative distribution of motor command signal magnitude when using DC control.

## DISCUSSION

The data recorded are the motor command, not the actual motor movement. For each of the 3 DOFs, one motion was actuated more than its antagonist counterpart. This imbalance may have multiple causes. It may indicate that median control speeds for opposite motions are not related. Or, the user may have continued to actuate the DOF once reaching the limit for actual functional reasons (e.g. generating a hand close once closed in order to generate grip force) or due to a lack of feedback (not noticing that the elbow is at the maximum range of motion and continuing to drive in that direction). It may also mean that one control is easier than the other (wrist pronation vs supination).

For direct control, the user frequently complained of inadvertent hand opening. This is reflected in the usage data. The cause of this inadvertent hand open was partly due to the inability of the user to control the prosthesis. This was a difficulty that he had during use of his own prosthesis and during the initial fitting with the research prosthesis.

It was noted that when he returned for final outcome measures after the direct control phase, there was intermittent noise on the hand open channel. Noise on an EMG channel typically presents as a 100% signal through the processor and would be recorded as such. Prior to collection of outcome measures, the nut of this electrode dome and the ground dome were tightened, reducing the noise. However, the subject did

still have inadvertent hand open, especially when eliciting an elbow extension signal.

## CONCLUSION

Though there are limitations to the data collected, this type of investigation into usage statistics can help with diagnostics of hardware and control. This data can also help with validation of home logs in tracking use.

## ACKNOWLEDGEMENTS

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## EFFECT OF A WRIST FLEXION UNIT ON TORSO MOVEMENT IN A TRANSHUMERAL PROSTHESIS USER DURING THE BOX AND BLOCKS TEST

Laura A. Miller, PhD, CP, Kimberly Ingraham, BE, Kristi Turner, OTR/L  
*Center for Bionic Medicine, Rehabilitation Institute of Chicago*

### INTRODUCTION

There are multiple timed outcome measures that have been used to evaluate prosthesis use and the effect of additional components. However, these timed tests do not necessarily evaluate the quality of movement or compensatory movements necessary due to restrictions of component design. The use of motion analysis to evaluate upper limb prosthesis use has recently begun to explore these compensatory movements in individuals who use an upper limb prosthesis. [1-5] It has been shown that individuals who use a prosthesis use more compensatory movements (increased overall trunk movement) compared to non-amputees [1]. Motion analysis has also been used to look at the effect of TMR control on compensatory movement when completing a modified box and blocks test [5]. However, little has been done to compare the effect of specific component functions on compensatory movements.

### METHODS

One individual using a transhumeral prosthesis with a Boston Digital Arm, Motion Control wrist rotator, and ETD terminal device with a manual wrist flexion unit was evaluated using an Optotrack motion analysis system. The subject gave informed written consent for a research study approved by the Northwestern University's Institutional Review Board.

The individual completed two 120s trials of the Box and Blocks test [6] with the wrist unit in the neutral position (Figure 1) and one trial with the wrist unit fully flexed. Motion capture data collection began when the subject was prompted "Go" to begin movement.

Markers on the distal sternum, the acromion of the prosthetic side, and on the distal anterior side of the socket were used to calculate two angles, shown in Figure 2. The first angle, the global shoulder angle, was defined as the angle between the global vertical axis and the line connecting the distal sternum marker to the prosthetic side acromion. This angle combines the effect of both shoulder elevation and lateral trunk lean. The value of this angle during the two static trials was 76.5 degrees. This angle would be smaller

the more the subject elevated his prosthetic side shoulder and/or leaned towards the sound side.



Figure 1: Photograph of subject performing box and blocks with motion sensors attached

It was hypothesized that with wrist flexion the subject would be able to remain more upright, using less trunk and shoulder compensation to complete the task, which would result in a larger average global shoulder angle with less variability.

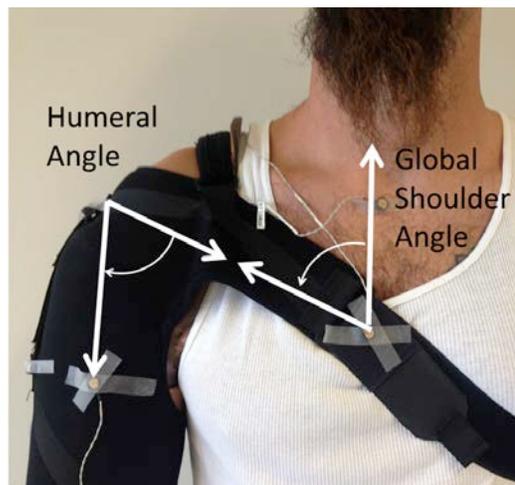


Figure 2: Photograph showing the vectors (lines) used to calculate the angles for global shoulder movement and humeral angle

The other angle calculated was the angle in the global coordinate system between the line connecting the prosthetic acromion to the distal sternum and the line connecting the prosthetic acromion to the distal anterior socket marker. This angle was used to measure compensatory shoulder abduction/flexion during the task. It was hypothesized that with wrist flexion, less abduction (decreased humeral angle) would be observed. The value of this angle during the static trials was 52.5 degrees.

## RESULTS

With the wrist flexion unit in the neutral position the subject moved 14 blocks in the first trial and 11 blocks in the second trial. With the wrist fully flexed, he moved 21 blocks.

Figure 3 shows the global shoulder angle for all three trials. The two trials without wrist flexion are shown in the lighter weight line and the dashed line. The trial with flexion shows an overall shift in the positive direction, indicating that the subject was able to remain in a more upright posture. The average ( $\pm$ SD) angle for the two trials without wrist flexion was 43.5 ( $\pm$ 6.3) degrees. The average angle with wrist flexion was 57.6 ( $\pm$ 3.6) degrees.

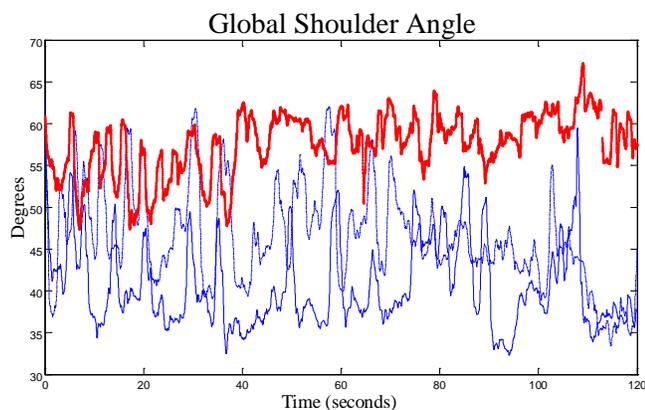


Figure 3: Global shoulder angle for the three trials. The trial with wrist flexion is the heavier weighted line.

The humeral angle was also calculated but did not show large differences. The average ( $\pm$ SD) angle without wrist flexion was 46.7 ( $\pm$ 3.6) degrees. The average angle with wrist flexion was 43.0 ( $\pm$ 5.5) degrees.

## DISCUSSION

The use of wrist flexion has the ability to alleviate the need for upper limb prosthesis users to rely on compensatory movements to complete a task. A case study analysis of the effects of using wrist flexion shows that there is a reduced

amount of torso lean as well as reduced variability during the repetitive movement.

This paper is a very basic analysis of how compensatory movements change by including a wrist flexion unit. The calculation of torso movement presented combines the effect of both shoulder elevation and actual torso lean. Ideally, the superior sternal marker would have been used in the analysis to be able to separate these conditions. However, it was found that the subject's facial hair obstructed the marker for a majority of the trials, impeding its use. Also, the calculation of humeral angle combines humeral flexion and humeral abduction (movement in the sagittal plane and in the frontal plane). A more in-depth analysis would break down the contribution of each movement to the total angle.

In this case, the subject's performance (i.e., number of blocks moved) improved with the addition of wrist flexion. However, in a case where timed test results do not show differences between conditions, motion analysis may show improvements in the quality of the movement that are not appreciated otherwise.

## ACKNOWLEDGEMENTS

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## A COMPARISON BETWEEN DIFFERENT CONFIGURATIONS OF HAND/WRIST PROSTHESES

Federico Montagnani, Marco Controzzi, and Christian Cipriani

*The BioRobotics Institute, Scuola Superiore Sant'Anna, V.le R. Piaggio 34 56025 Pontedera, Italy*

### ABSTRACT

Traditional myoelectric transradial prostheses severely limit their users in performing a wide range activities in the daily life. In the last few decades, while significant efforts were spent in designing artificial dexterous hands, little work has been done with regards to wrist design. Nevertheless the human wrist contributes significantly when performing upper limb motor task. We hypothesized that a single DoF hand equipped with a 2-DoF active wrist allowed performance comparable with a highly performant multi DoF hand coupled with a 1 DoF wrist rotator. To assess this hypothesis we compared four emulated architectures of hand-wrist prostheses using the SHAP test. Our preliminary results show indeed that shifting the dexterity from the hand to the wrist could enhance the ability by transradial amputees in performing tasks of daily living. Hence, this study, suggests that larger attention should be paid to the development of artificial wrists.

### INTRODUCTION

Restoring the human hand motor function with an artificial prosthesis has been and still is today one of the grand challenges in bioengineering. To achieve this it is necessary to develop the artificial limb, physically capable of performing motor tasks, as well as the human-machine-interface (HMI) able to record and decode the intentions of the individual. The design of artificial limbs is of interest for this work, in particular the development of prostheses for the transradial amputees.

Traditional myoelectric transradial hands, clinically available since the early 1990's, are 1 Degree of Freedom (DoF) grippers. These can be integrated with a 1-3 DoFs wrists. Although able to restore certain motor functions, these prostheses limit the individual in performing a wide range of tasks useful in the daily life. The person is often forced to perform compensatory movements of the upper arm that may cause injuries in the long term [1],[2]. These are some of the reasons why a significant percentage of myoelectric hand users abandon their 1 DoF prosthesis [3].

To address this problem, in the last few decades, researchers have spent significant efforts in designing artificial hands with enhanced dexterity, trying to mimick the complex mechanical architecture of the human hand. Several multi-fingered and multi-grasp hands have seen the light in research labs around the world and few designs have actually reached the clinic [4], [5] and the markets (e.g. Michelangelo by Ottobock, I-limb Ultra by Touch Bionics) .

In contrast to the advances of hand design, wrist design has drawn much less interest among the researchers and has modestly progressed so far. There is only one motorized wrist commercially available, which actuates the hand pronation/supination (10S17 Electric Wrist Rotator, by Ottobock). All other commercial wrists are passive (not motorized) and include 1 DoF, flexion/extension wrists (e.g. MyoWrist Transcarpal 10V38 by Ottobock), 1 DoF flexion/extension compliant wrists (e.g. Michelangelo wrist, by Ottobock) or 2 DoFs (flexion/extension and abduction/adduction) compliant wrists (e.g. Multiflex, Touch Bionics Inc., Livingstone, Scotland). Rotation of these wrists around the prosthetic socket (hand pronation/supination) is also usually available. In research labs very few new designs have been presented so far. Kyberd et al., designed and actuated a 2 DoFs wrist that could be controlled by EMG Pattern Recognition [6]. Montagnani and colleagues proposed a 2 DoFs wrist with switchable stiffness [7]. Weir et al. presented a 3 DoFs anthropomorphic wrist actuated by three servomotors [8] while Thayer et al., and Lovchik et al., presented 2 DoF wrists actuated by means of linear actuators [9], [10]. None of these designs were clinically assessed.

The modest progress reached in prosthetic wrist design is quite surprising. Indeed the wrist with its movements (practically) always contributes to the execution of a arm motor task, thus it stands to reason to consider it as important as the hand. In support of this statement Bertels and colleagues [11] showed that even a single DoF wrist coupled with a conventional prosthetic hand reduces the compensatory movements by the amputee required to reach for objects. Previously, similar studies in the orthopaedic field anticipated the same finding and quantified the extent of compensatory movements of the upper limb when the

flexion/extension of the wrist was locked [12], [13]. Overall, although the importance of the DoFs of the wrist seems evident, very few studies in the field of prosthetics have been carried out so far.

In this work we preliminary assessed and compared four emulated architectures of hand-wrist prostheses using the SHAP (Southampton Hand Assessment Protocol) [14]. The hand-wrist configurations, were emulated using custom built orthoses worn by able bodied subjects. The four configurations differed depending on which hand/wrist movements were allowed and which ones were physically blocked. In this way we mimicked the biomechanics of different prosthesis solutions, under the control of the *ideal controller*, namely, the unimpaired sensorimotor and muscular system. We mimicked: (i) a 1 DoF hand (open/close) coupled with 1 DoF wrist (rotation), (ii) a multi grasp hand coupled with a 1 DoF wrist (rotation), (iii) a 1 DoF hand coupled with 2 DoFs wrist (rotation and flexion/extension), and (iv) the ideal prosthesis: the unconstrained hand and wrist. In other words, the objective was to evaluate the ability in executing activities of daily living (ADLs), with different levels of *shared dexterity* between the wrist and the hand.

We hypothesised that a single DoF hand equipped with a 2 DoF active wrist allowed performance comparable to a highly performant multi DoF hand coupled with a 1 DoF wrist rotator. Our preliminary results show indeed that shifting the dexterity from the hand to the wrist could enhance the ability by transradial amputees in performing tasks of daily living. Hence, this study, suggests that larger attention should be paid to the development of artificial wrists.

## MATERIALS AND METHODS

### Participants

Five right-handed able-bodied subjects (3 males and 2 females, aged  $31.2 \pm 2.0$  years old) participated in this preliminary study. They were asked to perform the SHAP in the four hand/wrist configurations, wearing the developed orthoses. Informed consent according to the Declaration of Helsinki was obtained before conducting the experiments.

### Hand/wrist configurations

Three wearable orthoses were developed in order to lock specific DoFs of the human hand and wrist, thus emulating different prosthesis configurations (Table 1, Fig. 1). The four configurations are described in the following.

**1 DoF wrist (pronation/supination), 1 DoF hand (open/close).** This configuration corresponds to a typical myoelectric fitting (e.g. Ottobock Speed Hand coupled with a wrist rotator). It was obtained using a commercial wrist orthosis coupled to two splints fixed to the hand which locked

the ab/adduction movement of the thumb and coupled the fingers (Fig. 1-A). The subjects were allowed to open/close the hand and to pronate/supinate the wrist, when they wore this orthosis.

**2 DoFs wrist (pronation/supination, flexion/extension), 1 DoF hand (open/close).** This configuration although investigated in a number of research works (e.g. [6]) is not currently clinically available. The orthosis had a revolute joint aligned to the wrist, allowing its flexion/extension (Fig. 1-B).

**1 DoF wrist (pronation/supination), multi DoFs hand.** This configuration emulated the *ideal* hand prosthesis, i.e. the one perfectly matching the human hand biomechanics. Examples of robotic hands that attempted to do so are the DARPA Revolutionizing Prosthetics Proto 2 [8], and in a simplified manner the Vanderbilt Multi-grasp hand [14] and the SmartHand [4] to cite a few. In our experiment this *ideal* prosthetic hand (mimicked using the natural hand) was coupled to an orthosis that allowed for just the wrist rotation (Fig. 1-C ). A simplified version of this configuration is clinically available by fitting a multi-grasp hand (like the *BeBionic* or *iLimb*) with a motorized wrist rotator.

Table 1: Distribution of the available DoFs between the wrist and the hand across configurations.

	DoFs in the wrist	DoFs in the hand	Total DoFs #
A	Pronation/supination	Open/close	2
B	Pronation/supination Flexion/extension	Open/close	3
C	Pronation/supination	22 (natural hand)	23
D	Pronation/supination Flexion/extension Radial/ulnar deviation	22 (natural hand)	25



Figure 1: Four hand/wrist configurations achieved with custom built ortheses.

**Ideal hand/wrist prosthesis.** In this configuration subjects had no constraints and could move their hand normally (Fig. 1-D). This configuration mimicked the ideal hand/wrist prosthesis, restoring all the DoFs lost due to the amputation. It is clinically unavailable and was included as a control for the other configurations.

#### Experimental protocol

The ability to execute Activities of Daily Living (ADLs) with the four hand-wrist configurations was evaluated using the SHAP. This is a standardized protocol for the evaluation of the hand functional range [15]. The protocol is divided into two sessions: in the first one the subject is required to grasp and manipulate abstract objects (cylinders, tabs, spheres, etc.); in the second session the subject is required to perform 26 ADLs like turning a door handle, picking up coins, moving containers, etc. Details can be found elsewhere [15].

Each subject performed the original version of the SHAP four times, one for each hand-wrist configuration, in a single session (day). The first trial was done with the sound limb, whereas the following three trials were performed wearing one of the ortheses in a randomized order across subjects, in order to reduce possible learning effects that could bias the results.

The SHAP is a time based protocol: subjects were asked to complete the tasks as quickly as possible. The duration of the tasks were used to calculate scores which described the index of function of the subject. Thus, the best performance equated to the fastest task execution. The results of the SHAP were resumed with a *global* Index of Function (IoF) and six *partial* IoFs related to the six main grasp types involved in the test. We did not do any statistical analysis on the results,

given the small number of subjects involved in this pilot study. It should be noted, however, that differences of the IoF across conditions greater than 2 denote a statistically different performance, as recalled in [16].



Figure 2: Experimental set up including the SHAP materials.

## RESULTS

The subjects learned quickly how to master the tasks in the different configurations and the duration of the whole experiment (four trials) was about 1.5 h. The performance was  $98.6 \pm 0.8$  (mean  $\pm$  standard error of mean) for configuration D (ideal hand-wrist, i.e. the control condition)  $96 \pm 0.9$  for configuration B (2 DoFs wrist, 1 DoF hand),  $95.5 \pm 0.9$  for configuration C (1 DoF wrist, multi-DoFs hand),  $91.4 \pm 0.9$  for configuration A (1 DoF wrist, 1 DoF hand, i.e., the typical myoelectric fitting), as measured by the global IoFs (Fig. 3). The same trend of the global IoF was shown by the partial IoFs related to the six main grasp types (Fig. 4). It is worth noting that although the results from configurations B and C were globally similar, the subjects dramatically changed the way of performing the same tasks depending on the type of orthosis worn, as observed during the sessions.

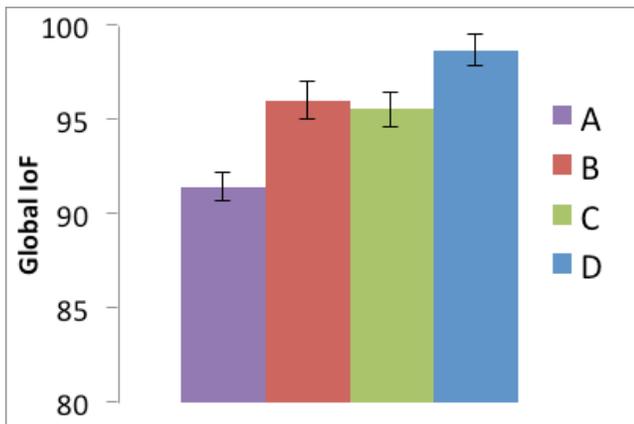


Figure 3: Global Index of Function in each hand-wrist configuration (error bars denote the standard error of mean).

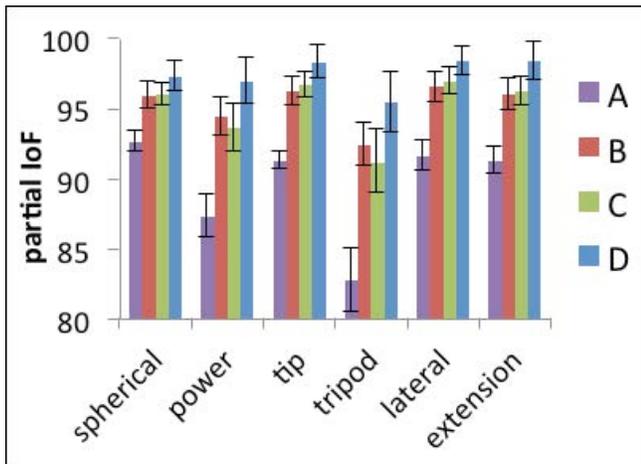


Figure 4: Partial Indexes of Function for the six grasp types, in each hand-wrist configuration.

## DISCUSSION

If we compare the results from configuration A with the results available from the literature, achieved by amputees wearing the equivalent prosthesis (1 DoF wrist, 1 DoF hand), a large difference can be found [19]. In fact, our results were significantly better (90 vs. 70-80 in [17]). It seems obvious that this difference is due to the fact that our subjects had unimpaired sensorimotor control. Thus our results should be read for comparison across configurations and not in absolute terms.

It stands to reason that the less constrained (i.e., most dexterous) configuration (D) allowed for the best performance, while the one with less DoFs (i.e., 2 – configuration A) resulted in the worst performance. The interesting result is that configuration B and C, although B allowed for just 3 DoFs while C allowed for 23, resulted in a very similar

performance. A 2 DoFs wrist (pronation/supination plus flexion/extension) coupled with a 1 DoF hand (open/close) performs similarly to a natural-like dexterous hand (22 DoFs) coupled with a 1 DoF wrist. In other words an additional DoF in the wrist accounts for the performance achievable by 21 DoFs in the hand.

This very important outcome should be used to guide bioengineering research. In fact replicating the biomechanics of the wrist with an artificial one, is a much easier task for engineers with respect to replicating the complexity of the human hand. Equivalently, developing a HMI for the few DoFs of the wrist is today more feasible than for the several DoFs of the hand [18], [19]. The consequence of our finding is also very important in the domain of prosthetic fittings. Our results suggest that a complex (and expensive) multi-grasp hand cannot be fully exploited if it is fitted on a simple rotator. A much cheaper 1 DoF hand on a slightly more complex wrist would result in a similar performance, as measured by the SHAP.

The SHAP provided us with a performance metric based on the time to execute the task. However, nothing could be claimed with regards to the way of performing the tasks. For this reason our future study will focus on the analysis of the extent of compensatory movements in the proximal joints, in the four configurations.

## ACKNOWLEDGEMENTS

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## THE EFFECT OF UNILATERAL VERSUS BILATERAL UPPER EXTREMITY PROSTHESES USE FOR SERVICE MEMBERS WALKING WITH QUADRILATERAL AMPUTATIONS

Annemarie E. Orr, Elizabeth M. Nottingham, James W. Vandersea,  
Kerry S. Quinn, Barri L. Schnall  
*Walter Reed National Military Medical Center*

### ABSTRACT

Since the beginning of the wars in Iraq and Afghanistan there have been a total of 1634 U.S military Service Members (SM) who have sustained major limb amputations [1], five of whom have sustained quadrilateral amputations (QA). Limited objective data is available on the effect that the use of upper extremity prostheses (UEP) have on gait, especially in cases of bilateral upper extremity loss. This case series looks at two SMs with QA and analyzes the effect of walking with unilateral or bilateral UEPs has on gait patterns. These SMs sustained traumatic injuries as a result of an improvised explosive device, and sustained transfemoral and knee disarticulation lower extremity amputations and transradial (TR) and transhumeral upper extremity amputations. SMs wore their preferred prosthetic components which included microprocessor controlled prosthetic knees. Both SMs selected to use a myoelectric prosthesis on the TR side and a body powered prosthesis on the transhumeral side. Each used the TR side myoelectric prosthesis when walking with unilateral UEP. While participating in occupational and physical therapy, one service member preferred to use only the TR side UEP while the other preferred to use both UEPs. The SMs visited the gait laboratory as they were near the peak of the rehabilitation process, when they were independent community ambulators. Motion capture data was collected for both SMs as they walked at self-selected velocity while wearing bilateral UEPs and again while using only the TR side prosthesis. Data shows SMs walked at a faster self-selected walking velocity while wearing the TR side prosthesis (1.08m/s vs. 1.02m/s). The average range of trunk flexion throughout the gait cycle was similar (23.4° while wearing the TR side UEP vs 23.3° with bilateral UEPs) between the two conditions. Though average data was similar between conditions, data showed a 2° reduction in trunk flexion for each SM, which corresponded to the UEP setup used in training. With two participants limited conclusions can be drawn however, preferred UEP use during training surely coordinates with comfort during walking. More research must be done to confirm exactly how this affects other key aspects of gait. Furthering knowledge on this topic will be clinically beneficial to therapists and prosthetists

as they design rehabilitation protocols for SM who have sustained multiple limb loss.

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## A PERMANENT, BIDIRECTIONAL, OSSEOINTEGRATED INTERFACE FOR THE NATURAL CONTROL OF ARTIFICIAL LIMBS

Max Ortiz-Catalan<sup>1</sup>, Rickard Branemark<sup>2</sup>

<sup>1</sup>*Chalmers University of Technology and Centre of Orthopedic Osseointegration,  
Sahlgrenska University Hospital*

<sup>2</sup>*Centre of Orthopedic Osseointegration*

### ABSTRACT

Although myoelectric prostheses have been clinically implemented since the 1960's, they are still far from the functionality of their biological counterpart. Currently, the lack and instability of physiologically appropriate control signals has been observed as the bottle neck towards an intuitive prosthetic control. Implantable neuromuscular electrodes could provide sufficient and long-term stable bioelectric signals. However, the permanent trans/percutaneous communication between the implanted electrodes and the artificial limb is a major obstacle that has hindered the exploitation of this approach for decades. In order to overcome this problem, we have developed a permanent bidirectional interface into the human body based on the principle of osseointegration.

We have enhanced the OPRA Implant System with a series of electric feedthrough connectors to permanently communicate with several implanted electrodes (up to 12 contacts). A modular design was conceived so that each component can be replaced or upgraded (i.e., with more selective electrodes) without the need of explanting the others. More importantly, the original implant system has kept its mechanical integrity; the osseointegrated and percutaneous components, as well as the implantation protocol and skin interface, were preserved in order to maintain a successful osseointegration and prosthetic coupling.

This novel system was implanted in January 2013 in a trans-humeral amputee and continues to be used without complications up to date (> 1 year). The conventional surface electrodes used by the patient's myoelectric hand were replaced by epimysial electrodes with a custom designed analog controller. The controllability of the prosthesis is no longer restricted to environmental conditions (temperature) and limb position. It is not affected by motion artifacts or electromagnetic interference, thus there is no involuntary actuation. Consequently, the functionality provided by the prosthesis has increased considerably. This control system is

used daily by the patient at home and at work. The patient wears the prosthesis as long as he is awake and has reported occasionally sleeping with it, as there are no superficial components causing discomfort. Additionally, the feasibility of myoelectric pattern recognition to intuitively control 3 degrees of freedom in real-time, and long-term reproducible tactile sensory perception elicited via neurostimulation to a permanently implanted cuff electrode, have been demonstrated.

This is the first clinical demonstration of the proposed technology which addresses two major issues in the field of artificial limbs, namely stable attachment and natural control. Further research and clinical investigations will be conducted to fully validate this approach.

## AN OPEN SOURCE PLATFORM FOR PROSTHETIC CONTROL ALGORITHMS BASED ON BIOELECTRIC PATTERN RECOGNITION (BIOPATREC)

Max Ortiz-Catalan<sup>1</sup>, Rickard Branemark<sup>2</sup>

<sup>1</sup>*Chalmers University of Technology and Centre of Orthopedic Osseointegration,  
Sahlgrenska University Hospital*

<sup>2</sup>*Centre of Orthopedic Osseointegration*

### ABSTRACT

Bioelectric signals processing and pattern recognition algorithms (SP-PRAs) have been extensively studied as they hold potential for providing more intuitive prosthetic control. Although authors seem to agree on reporting a common performance metric (prediction accuracy), there is a significant amount of study-dependent variables that hinder inter-study comparisons. BioPatRec is an open source effort to provide a common research platform for the development of prosthetic control strategies based in SP-PRAs, as well as for benchmarking in a shared repository of bioelectric data.

BioPatRec is a modular platform implemented in MATLAB that allows a seamless integration of a variety of algorithms in the fields of signal processing; feature selection and extraction; pattern recognition; and, real-time control. It includes a virtual reality environment and all the necessary functions for the myoelectric control of a virtual arm, computer games, and external devices; from data acquisition to real-time evaluations. Statistical and biologically inspired PRAs for the predication of individual and simultaneous movements have been included. Moreover, PRAs can be configured in several dedicated topologies due to an implementation framework that also allows the seamless addition of others. Additionally, a repository of myoelectric recordings related to hand and wrist motions is provided for individual and simultaneous movements; 11 and 27 classes from 20 and 17 subjects, respectively.

BioPatRec functionalities are easily accessible via graphical user interfaces, which allow for non-experts to explore the potential of a variety of SP-PRAs for the prediction of motion intent and control. Detailed instructions for use and development are provided in the online project hosting platform (<http://code.google.com/p/biopatrec/>), which includes issue tracking and an extensive “wiki”. This transparent implementation has shown to facilitate utilization, but more importantly, collaboration.

The modular design of BioPatRec allows researchers from different fields to seamlessly benchmark their algorithms by applying them in prosthetic control. For example, a pure

artificial intelligence researcher can easily add a pattern recognition algorithm without necessarily knowing how to obtain and process bioelectric signals, or how to produce and evaluate physically meaningful outputs.

BioPatRec is fundamentally a tool to aid the development and benchmarking of algorithms applied in prosthetic control. It has been made openly and freely available with the hope to accelerate, through the community contributions, the development of better algorithms that can potentially impact the patient’s quality of life.

## FIRST IN MAN DEMONSTRATION OF FULLY IMPLANTABLE MYOELECTRIC SENSORS TO CONTROL AN ADVANCED PROSTHETIC WRIST AND HAND

Paul F. Pasquina<sup>a,d</sup>, M.D.; Melissa Evangelista<sup>b</sup>; Antonio J. Carvalho<sup>a,c</sup>; Joseph Lockhart<sup>b</sup>;  
Sarah Griffin<sup>a,c</sup>; George Nanos<sup>d</sup>, M.D.; David Hankin J.D.<sup>b</sup>; Patricia McKay<sup>a,d</sup>, M.D.;

<sup>a</sup>Uniformed Services University of the Health Sciences; <sup>b</sup>The Alfred Mann Foundation; <sup>c</sup>The Center for Rehabilitation Sciences Research, Henry M. Jackson Foundation; <sup>d</sup>Walter Reed National Military Medical Center

### ABSTRACT

Over the past decade, the development of electromechanical prosthetic devices has outpaced the advancement of methods to control them. Existing systems rely on electromyography (EMG) signals recorded from the surface of the skin to enable user control. These surface recordings are often inconsistent and unreliable and are limited to recording from large muscles close to the surface of the skin. The constraints that this places on the operability, functionality, and reliability of the prosthetic systems and the lack of a viable alternative have contributed significantly to the high rates of prosthetic abandonment that persist amongst individuals with amputation(s). Users require a prosthetic system that allows them to have stable and consistent control over the manipulation of their prosthesis. Implantable sensors have been carefully examined as a method by which to directly obtain EMG signals from residual musculature. By placing the sensors inside the musculature, it is believed that it would be possible to eliminate noise and interference issues that result from recording from the skin surface and greatly expand user operability by allowing direct control of prosthetic function. Implantable Myoelectric Sensors (IMES<sup>®</sup>), developed by the Alfred Mann Foundation, are tiny electrodes that, when inserted into muscle, can wirelessly detect and transmit EMG from muscle contraction to an electromechanical wrist and hand via an electro-magnetic coil built into the prosthetic socket. We are currently conducting the first FDA -approved human feasibility study of the IMES<sup>®</sup> System at Walter Reed National Military Medical Center with individuals who have unilateral transradial amputations. To date, the first subject has been implanted and has demonstrated advanced control of all of the functions of a three degree of freedom (DOF) myoelectric wrist and hand, both individually and simultaneously. In addition, the subject has shown steady improvement and sustained performance in all functional assessments over a six-month period. A second study subject has recently been implanted and successfully began utilizing the IMES<sup>®</sup> System in-clinic.

### INTRODUCTION

The first commercially available myoelectric prosthetic arm reached the market nearly fifty years ago. Despite this, body powered prostheses, first developed during the American Civil War, remain the most popular form of upper arm prosthesis to this day. It is a highly complex challenge to establish intuitive control of a prosthetic for an individual who has suffered an amputation, especially a limb as dexterous and precise as the human hand. Decades of development and advancement in myoelectric prosthetic science have not been able to significantly improve the reality that is realized in clinical care. It is no surprise, therefore, that prosthetic abandonment rates remain high; by one survey, nearly 30% of soldiers from the Vietnam War and 22% of soldiers from Operations Enduring Freedom and Iraqi Freedom abandoned prosthesis use altogether, complaining of weight, discomfort, pain, lack of functionality, and poor fit.<sup>1</sup> With a rapidly increasing number of individuals with limb loss (estimated to reach 2.2 million individuals in the United States alone by 2020) and the particular needs of service members returning from United States military action in the Middle East, who are more likely to have higher level and more complex injuries and multiple amputations, it is imperative that steps are taken to address the needs of this population. While the short-term effects of abandoning prosthetic device use may be obvious, individuals who rely on only one arm/hand for daily use are also at a much higher risk of developing overuse injuries and arthritis of their neck, upper back and remaining limbs, further influencing long-term morbidity and quality of life.<sup>3</sup>

Advanced upper-limb prosthetic devices as they exist are capable of a wide range of precise function that mimics the human musculoskeletal system. The most severe challenge to accessing the full potential of these technologies is finding a way to provide control over all of these functions. Systems to translate user intent to these devices are limited. Currently, these electromechanical devices rely on EMG signals recorded from the surface of the skin and pair each signal to a particular prosthetic function. While these surface myoelectric signals have proven to be a sufficient input for controlling the movement of a powered prosthesis, they have significant limitations.<sup>4</sup>

First, the surface of the skin itself presents fundamental challenges to recording EMG signals from the musculature. Surface signals are particularly susceptible to noise introduced by the environment, interference from muscle co-activation (present in most upper-limb articulation), movement between the electrode and the skin, and changes in the skin impedance due to perspiration<sup>5</sup>. This can be exacerbated if the prosthesis is incorrectly donned, as this would alter the positioning of the electrodes on the skin surface and thereby the muscles they record from. These issues make control inconsistent and unreliable, which can frustrate users. Second, due to limited sampling depth, the number of unique sensing sites is limited to larger muscles close to the surface of the skin. The pre-amputation function of these muscles may not correlate with the functions their signals now control in the prosthetic hand. Surface sensing can also limit the pure number of unique sensing sites, as the use of too many can lead to cross-talk and incorrect activation. For a powered prosthesis that offers more functions than available sensing sites, the current muscles must control more than one function by employing switching techniques like rapid co-contraction to signal mode changes. The result is non-intuitive, sequential prosthetic joint manipulation rather than intuitive, simultaneous control. Limited dexterity of control is often cited as the primary reason for abandonment of myoelectric prostheses.<sup>6</sup>

The Implantable Myoelectric Sensor (IMES<sup>®</sup>) System is a prosthetic system developed by the Alfred Mann Foundation that provides simultaneous three degree of freedom (DOF) prosthetic wrist and hand control by detecting EMG signals from the interior of residual muscles. The system utilizes tiny wireless electrode sensors (IMES<sup>®</sup>) that can be implanted directly into the belly of whatever residual muscles are desired and are capable of detecting and wirelessly transmitting EMG data. An electro-magnetic coil built into the prosthetic arm provides power to the IMES<sup>®</sup> and receives the EMG data. The localization of these devices in the muscle provides numerous advantages that translate directly to the user. First, the number of accessible muscles compared to conventional systems is significantly expanded. This means that prosthetic function can more appropriately be linked to residual muscle function to greatly improve intuitive control. Second, placing them within the muscle removes many concerns about the number and position of sensors causing interference and provides the opportunity to access a more stable and reliable EMG signal, providing more consistent control over a greater number of functions than is possible with existing interface methods. Furthermore, by improving the number and quality of distinct signals that can be communicated, more precise control of limb function is possible.

Our team at the Walter Reed National Military Medical Center (WRNMMC), in conjunction with the Alfred Mann Foundation for Scientific Research, Advanced Arm

Dynamics, and the Uniformed Services University of the Health Sciences has recently initiated the first FDA-approved clinical trial to examine the feasibility of the IMES<sup>®</sup> System to record and transmit EMG information generated by muscle contraction from the interior of residual muscles to provide user control of a prosthetic device by individuals with transradial amputation. The study is currently being conducted at WRNMMC under an Institutional Review Board (IRB) approved protocol, with full human use ethics review and patient informed consent.

## STUDY OVERVIEW

The IMES<sup>®</sup> System is a group of components that function together as an integrated prosthetic control system. The System registers, records, and transmits the electrical impulses generated during muscle contraction and then processes this information to affect motors that move the joints of a myoelectric prosthesis.

The IMES<sup>®</sup> System currently allows for simultaneous control of up to three distinct movements, or degrees of freedom (DOF). These include wrist pronation/supination, hand open/close, and thumb abduction/adduction. Two IMES<sup>®</sup> are required to control each DOF, one for each of the opposing motions, such that six IMES<sup>®</sup> are used.

Our study is a prospective, non-blinded, single group, interventional, feasibility study being conducted at Walter Reed National Military Medical Center. Up to ten subjects with unilateral transradial amputations who are current myoelectric device users will be recruited with three being implanted with up to eight IMES<sup>®</sup> devices each. Prior to surgical implantation, each subject must undergo a baseline screening examination to confirm eligibility. This includes a needle EMG exam and several functional assessments completed with their existent myoelectric device. The needle exam measures EMG signal in the target muscles to ensure that sufficient volitional control remains to allow operation of the IMES<sup>®</sup> System and also allows mapping of the target muscle locations for surgical implantation. In addition, it can be used to rule out neuromuscular disorders that would exclude the subject from participation. Functional tests include the Assessment of Capacity for Myoelectric Control (ACMC), Box and Blocks Test (BBT), and Southampton Hand Assessment Procedure (SHAP).

Following a two week recovery period and post-operative examination, subjects initiate a month of in-clinic pre-prosthetic training with a bench-top device under the supervision of a prosthetist and occupational therapist, 4-5 times a week. This month of training allows the subject to become accustomed with how the IMES<sup>®</sup> System functions and to train themselves to contract the appropriate muscle

to operate the System as they desire. During this time, the subject also works with study staff to determine an optimal set of programming parameters. Programming is an iterative process that involves adjusting a set of signal processing parameters (gain, degree of signal smoothing, and threshold) such that the resulting electromechanical wrist and hand movements are consistent with user intent.

After they complete pre-prosthetic training, subjects are provided with a fitted prosthetic arm of their own to take home and utilize throughout the study. For the following month, they train in-clinic with study personnel 4-5 times a week before transitioning to once every two weeks in-clinic for the next five months. Their at-home usage is also tracked. Functional progress is monitored with a monthly repeat of the assessments utilized pre-surgery (the ACMC, BBT, and SHAP) along with an Accuracy Test, an in-house evaluation of the subject's ability to execute a series of different hand movements that test independent and simultaneous control over the three DOF offered. The SHAP, BBT, and ACMC are widely used clinical assessments and training tools for developing and tracking prosthetic control and performance. These take place once a month beginning with an Initial Assessment when the subject receives their definitive prosthesis, through month six, with a follow-up at one and two years.

The first subject was implanted and underwent several months of prosthetic training before a second subject was recruited, to ensure the safety and reliability of the system prior to enrolling a second subject.

## RESULTS

To date, two patients have been recruited and enrolled in the study. The first suffered a right transradial amputation secondary to trauma as a result of an IED explosion injury while serving as a member of the United States armed forces in 2012. Following recovery the subject was equipped with a surface-sensor myoelectric prosthetic device, of which the subject was a high performance and frequent user. The second subject suffered a left transradial amputation as the result of trauma sustained during a firefight in 2007, also while serving as a member of the United States armed forces. Both have been implanted with eight IMES® devices into specific muscles of their residual forearms. No complications were encountered during surgery or thus far in post-operative treatment. The muscles were chosen based on their natural anatomical function and paired with the prosthetic function with which they correlated. This was done in order to provide the subjects with the most intuitive control possible. With the first subject, minimal issues associated with swelling and edema post-surgery were treated with standard procedures to reduce swelling, including the application of compression

garments. No further issues related to the surgical procedure have been encountered. Both were able to operate the prosthetic system without issue aside from muscle fatigue beginning at the first visit two weeks post-surgery.

From the first day of pre-prosthetic training, the first subject was able to exert control over all three degrees of freedom offered by the electromechanical hand and wrist, both individually as well as simultaneously. While this was promising, continued training was necessary as the subject had difficulty differentiating between different muscle groups and their associated functions, often unintentionally engaging two muscles and thus two prosthetic functions at the same time. With time, adjustment of programming parameters, and training, the first subject's ability to specifically initiate and operate the functions of the IMES® system has substantially improved and the incidence of this co-activation has significantly decreased.

Even as the frequency of in-clinic training has decreased, the subject has sustained the consistent improvement seen throughout the initial six month period in the monthly functional assessments. (Table 1)

Table 1. Subject 001 Functional Assessments Scores

### A. SHAP Functionality Profile Scores

Visit	Functionality Profile Scores						IOF
	Spherical	Power	Tip	Tri.	Lat.	Ext.	
Baseline	31	31	13	6	37	55	<b>34</b>
Initial	50	15	29	37	10	28	<b>32</b>
Month 1	61	43	52	34	19	36	<b>45</b>
Month 2	71	45	54	46	28	37	<b>52</b>
Month 3	60	69	67	41	26	56	<b>57</b>
Month 4	71	57	64	46	29	51	<b>61</b>
Month 5	82	54	63	53	40	51	<b>60</b>
Month 6	72	66	41	58	45	64	<b>61</b>

\*IOF: Index of Function, Tri.: Tripod, Lat.: Lateral, Ext.:Extension, Sph.: Spherical

**B. Box and Blocks Test**

Visit	Average Score*
Baseline	12
Initial	4
Month 1	19
Month 2	23
Month 3	22
Month 4	24
Month 5	28
Month 6	25

\*Based on three attempts per visit

Similar to the first subject, the second had no issue individually isolating and operating the three DOF offered by the prosthetic system even just two weeks post-surgery. The post-operative surgical assessment showed no complications and healthy healing of the surgical entry points. While the second subject has yet to engage in the functional assessments following the surgery, the subject's progress with muscle isolation thus far has tracked well with (if not better than) that of the first subject.

**DISCUSSION**

To date, the IMES® System has functioned exactly as intended: it has reliably and accurately transmitted EMG information recorded from implanted target muscles to allow precise user operation of the prosthetic device. The consistency of control and the sustained association between targeted muscle groups and prosthetic function indicates that the IMES® electrodes have not migrated and their positions remain stable. Additionally, the first subject has shown steady improvement in all functional assessments and has reached a high level of control over the prosthesis.

The first subject has reported no issues operating the devices overhead or while sweating, areas of particular concern in which most myoelectric devices malfunction or become difficult to operate due to the change in position of the limb within the socket and the change in impedance at the surface of the skin (which can interfere with surface-signaling).

It is impossible to quantify the impact of consistent and intuitive dexterous control in prosthetic devices and that contribution to overall patient health. McFarland and colleagues reported that the average time to abandonment amongst service members who have suffered an upper-limb amputation as a result of military action in Operations Iraqi Freedom and Enduring Freedom (OIF/OEF) was a mere eight months. The most common reasons cited for their abandonment were the short length of their residual limb, pain, weight of the device, too much fuss, and inability to

control the device<sup>1</sup>. Those who abandoned their devices were more at risk for the development of cumulative trauma disorder (CTD), fatigue, arthritis, bursitis, and tendonitis. By increasing the consistency and stability of control for users, it may be possible to ameliorate issues associated with inability to control the device and the amount of fuss associated with their operation. By reducing prosthetic abandonment, it would thereby also be possible to reduce the development of further health issues and improve quality of life in individuals with amputations.

While this study is not yet completed, the results to date and the satisfaction of the participating subjects gives optimism about the ability of this technology to provide intuitive, stable, and dexterous control for all levels and types of amputation. In addition, the consistency of signal and accuracy of transmittance could facilitate continual user improvement and development, to achieve significant control and dexterity beyond that which has been seen thus far. By reducing the fuss associated with prosthetic systems and pairing prosthetic function with the same processing by which the function would be naturally controlled, it may be possible to significantly impact quality of life and improve the difficult transitions and challenges individuals who have suffered limb loss face.

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# REDEFINING THE NORMS OF TRANSHUMERAL AMPUTEE CARE THROUGH TMR, SKELETAL MODIFICATION AND SPECIALIZED UPPER LIMB PROSTHETIC REHABILITATION

Jennifer K. Peterson<sup>1</sup>, Pat Prigge<sup>1</sup>, Brian T. Carlsen<sup>2</sup>, John Miguelez<sup>1</sup>

<sup>1</sup>*Advanced Arm Dynamic, Redondo Beach, CA, USA*

<sup>2</sup>*Mayo Clinic, Rochester, MN, USA*

## INTRODUCTION

Historically, a transhumeral amputee would be fitted with a prosthesis that is contoured around the post-amputation residual limb anatomy, often making secure prosthetic suspension and rotational stability difficult to achieve. Surgical revision of the soft tissue may improve prosthetic fit; however, maintaining suspension on a transhumeral limb continues to present a challenge to the prosthetist. In addition, the patient's remaining biceps and triceps provide only two signals to myoelectrically control the elbow, wrist, and hand resulting in inefficient, non-intuitive, sequential control of the prosthetic components.

Surgical techniques such as Targeted Muscle Reinnervation (TMR), bone revision, and soft tissue reconstruction procedures paired with specialized prosthetic rehabilitation are improving the overall outcome and acceptance of prosthetic devices for transhumeral amputees. Thus the norms in transhumeral prosthetic care are being redefined. The transhumeral amputee now has the opportunity to have more secure prosthetic suspension, an increased functional envelope, and more intuitive control of myoelectric components, while reducing the need for restrictive harnessing.

## TEAM APPROACH

Ideally, a patient who needs to undergo a transhumeral amputation would have the opportunity to meet collectively with a surgeon, prosthetist and occupational or physical therapist prior to the amputation.<sup>1</sup> If this pre-surgical meeting is not possible, the collaboration between the surgeon, prosthetist, therapist, and patient should occur shortly after the amputation. During the team meeting, the patient's skin and musculoskeletal structure can be assessed, surgical and prosthetic options can be discussed, and the patient's potential future functional abilities can be projected.

Once the amputation has occurred, the team approach continues throughout the phase of healing, the prosthetic fitting and the ongoing rehabilitation process to ensure that the patient has the best possible functional outcome.

## TARGETED MUSCLE REINNERVATION

Since the introduction of TMR in 2004<sup>2</sup>, it has been found to be helpful for shoulder disarticulation and transhumeral amputees. TMR is a surgical technique which transfers residual peripheral nerves that are no longer controlling distal muscles to intact muscles that have been denervated.<sup>3-5</sup> The nerves re-innervate the "target or host" muscles and generate muscle contractions that can be read by electrodes which activate prosthetic movements. This increases the number of EMG signals available for controlling the prosthetic components. This control is intuitive since motor commands for hand movement produce prosthetic hand movements, and the natively innervated biceps and triceps are left to control the elbow component. An additional prosthetic signal for wrist component control could potentially be created by inserting the remnant ulnar nerve into the denervated brachialis muscle.

Patient studies document the prosthetic success after TMR, although large outcome studies are still lacking.<sup>6-10</sup> After TMR surgery, prosthetic movements are more efficient and simultaneous control of prosthetic components is possible.

## RESIDUAL LIMB BONE REVISION

Most transhumeral residual limbs have a cylindrical presentation with a straight remnant humeral bone and little or no distal end flare. This makes secure prosthetic suspension difficult since there are no anatomical contours around which a prosthetic socket can take hold. In addition, the skin around the humerus is very mobile, such that there is rotational instability in the transverse plane. To improve socket rotational stability, the socket has traditionally required anterior and posterior wings that limit shoulder ROM. With this socket design, the patient is unable to actively rotate at the shoulder to effect a lateral and medial movement of the terminal device. To further complicate the prosthetic fitting, the remnant humeral bone may be short and make secure prosthetic suspension even more challenging.

Bony revision may help to improve the length and contour of the humeral bone to improve prosthetic suspension. An angular osteotomy may be performed during which the distal end of the remnant humeral bone is angled to provide an orienting “hook” for the socket to take hold. The humerus can also be lengthened through distraction, through an autograft, or through an allograft. These operative techniques are beyond the scope of this paper. However, a technique performed by one of the authors (BC) can improve prosthetic fitting using a humeral allograft to lengthen and orient the residual limb bone. The operation utilizes a cadaveric distal humerus which is fixed to the patient’s residual humerus. This technique provides the residual limb with a bone structure the shape of humeral condyles (figure 1) but with adequate space allowed for fitting a prosthetic elbow. These condyles allow the prosthetist to contour a prosthetic socket around the condyles and narrow the socket above the condyles thus preventing slippage of the socket. This skeletal structure sets the stage for successful prosthetic fitting.



Figure 1: Allograft with distal humeral condyle shape

### UNIQUE PROSTHETIC SOCKET DESIGN

The challenges of prosthetic socket design in high level upper limb amputations have been previously reported<sup>11, 12</sup> and the prosthetic limitations to shoulder ROM are well known. Goals for prosthetic design for transhumeral levels are to capture all of the available shoulder motion, allow for the prosthesis to be used in any position away from the body, maximize comfort, and minimize harnessing. These goals are lofty but achievable through the surgical creation of an ideal residual limb and a carefully designed socket interface. The inner socket material must be flexible enough to allow donning and doffing with consideration of the distal shape of the humerus which has been remolded by means of osteotomy or cadaveric allograft. This is similar in nature to challenges presented in an elbow disarticulation socket design. The key difference is that a reconstructed humerus can be sculpted to an appropriate length to ensure proper matching of length for the prosthesis to the length of the sound side. Materials such as rolled silicone with direct lamination technique allow the

flexibility required to don and doff. However, an additional component is needed to ensure that secure suspension is maintained. BOA closure systems (figure 2) have been implemented for use in many prosthetics applications. These systems have the ability to mechanically modify the tension around the socket and residual limb with a continuous cable that can be directed to increase pressures in desired areas.

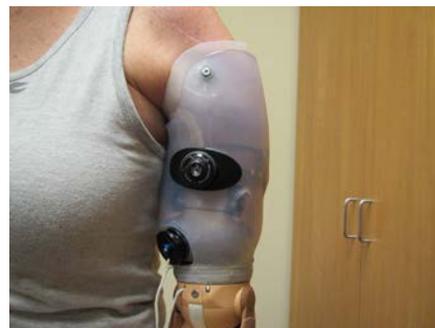


Figure 2: Rolled silicone inner socket with Boa Closure System

### SPECIALIZED PROSTHETIC THERAPY

The functional success of an upper limb prosthesis wearer is improved through prosthetic training.<sup>13-15</sup> Due to the complex nature of transhumeral upper limb prosthetic rehabilitation, specialized prosthetic therapy is essential especially if the patient has undergone a TMR procedure. Upper limb prosthetic therapy should begin prior to the amputation or soon after the amputation. Pre-surgery, the patient is provided with amputee, prosthetic and rehabilitation resources to educate the patient and to manage the patient’s expectations for post-surgical care and function. The patient is instructed in necessary strength, ROM, and endurance exercises.

Following surgery, the patient needs to follow precautions to prevent disruption of the healing bone or transferred nerves but this does not preclude the patient from participating in pre-prosthetic therapy. Pre-prosthetic therapy prepares the patient for wearing and controlling a prosthetic device. The patient is instructed in residual limb desensitization activities, edema control, a Graded Motor Imagery Program<sup>16</sup> and one handed techniques or adaptive equipment use to assist the patient with understanding how to perform ADLs after the amputation and before being fitted with a prosthetic device. Following TMR surgery, it is difficult to know which nerve fibers will re-innervate the targeted muscle. Thus it is important for the patient to work on motor imagery for all the possible movements that the reinnervated muscle may intuitively control. Specialized TMR exercises are started 3 weeks after the TMR surgery and are progressed as the host muscles re-innervate.<sup>10</sup> This helps to strengthen

the reinnervated muscles in order to produce contractions strong enough to be detected by electrodes for myoelectric prosthetic control. The upper limb prosthetic therapist must possess a thorough understanding of peripheral nerve distribution and the actions the residual nerves produce, such that the therapist is able to assist the patient in determining movement commands that will produce the most optimal EMG reading for prosthetic myoelectric control.

The therapist assists with finding electrode sites that produce strong EMG signals with adequate signal separation from other electrode signals. Once the patient is fitted with the prosthetic device, the therapist works closely with the prosthetist and patient to determine the most functional prosthetic alignment and to make software adjustments to produce clean control of the prosthetic components. Prosthetic therapy continues through the intermediate phase and then through the advanced phase of upper limb prosthetic therapy<sup>17, 18</sup> and outcome measures are utilized during each phase to monitor progress. Throughout the entire therapy process, proper posture and proper body mechanics are emphasized in order to prevent future acute and chronic musculoskeletal injuries.

### CASE EXAMPLE

A 50 year old female underwent a left transhumeral amputation following a motor vehicle accident. She had previously been very physically active at work and at home. In an attempt to continue with her previous life activities, she struggled to perform the activities using one hand and compensatory body movements. She presented with scar tissue with invaginations of redundant tissue on her left residual limb. The patient was able to isolate her left biceps and triceps muscle contractions and thus was an excellent candidate for utilizing a myoelectric prosthesis.

The client was fitted with a full myoelectric prosthetic system including a Dynamic Arm Plus elbow, an electric wrist rotator and a DMC Variplus myoelectric hand. Although the prosthesis improved her function, use of the prosthesis was slow and non-intuitive due to her having only two muscle control sites, biceps and triceps, to control the prosthetic components. She was required to use switching methods to cycle amongst the prosthetic components resulting in sequential control of each component to position and utilize the prosthesis for functional tasks.

In addition, she had short humeral bone length and needed to wear a harness to assist with prosthetic suspension. The bone length did not allow the socket to maintain contact on the lateral surface of the residual limb during shoulder abduction. The suspension and socket design were further complicated by the loose, redundant tissue at the distal end

of the residual limb. To stabilize the prosthesis from external rotational forces, the socket confined the shoulder anteriorly and posteriorly. This resulted in lack of active control of internal and external shoulder rotation of her anatomical shoulder. The patient was limited to approximately 80 degrees of shoulder flexion and abduction due to the harness and socket design restrictions.

The patient along with the surgeon, prosthetist and physical therapist considered the client's current prosthetic challenges and functional abilities; discussed potential surgical, prosthetic and therapy options; and developed a comprehensive upper limb prosthetic rehabilitation plan. The patient underwent a surgery for TMR and soft tissue revision. Bone revision to lengthen and broaden the distal end of the humerus with a distal end humeral bone allograft was also performed.

After a period of recovery and preprosthetic therapy, the patient was refitted with the previous elbow, wrist and hand prosthetic components. The removal of redundant soft tissue as well as the increased length and the shape of the humeral bone allograft allowed the development of a socket system that contoured around the flared shape of the distal end of the humerus. The socket system included a custom rolled silicone socket interface and a BOA closure system. With this suspension, the harness was able to be removed and the patient demonstrates the ability to flex and abduct the shoulder to 170 degrees (figure 3), thus improving her prosthetic functional envelope as well as her comfort. Her functional envelope was further improved by the fact that the secure suspension allows her to actively control her left shoulder internal and external rotation so that the prosthetic hand moves laterally and medially during function. This amount of active shoulder ROM is not typical for a transhumeral amputee wearing a traditional transhumeral socket suspension design. Traditionally, there would be no active internal and external shoulder ROM and the allowed shoulder flexion and abduction would be limited to no more than 90 degrees.



Figure 3: Overhead prosthetic placement

The TMR surgery resulted in the patient having direct intuitive control of hand open and close. This intuitive control decreased the cognitive demands of controlling the prosthesis, eliminated the need to produce switching signals to cycle between hand and elbow control, and improved the speed of function. Once the patient demonstrated isolated control of the elbow and hand components, the electric wrist rotator was activated and the patient was instructed on how to use the component to supinate and pronate the prosthetic hand through a 4 channel switching method.

Prior to and following surgery, the patient participated in extensive prosthetic therapy program including TMR exercises, prosthetic controls training, and functional training with proper posture and body mechanics. The patient, therapist, prosthetist and surgeon remained in close contact throughout the patient's rehabilitation.

## RESULTS

Due to the bone revision, soft tissue revision, TMR, advanced prosthetic fitting techniques, and specialized upper limb prosthetic therapy, the patient has reported a marked improvement in prosthetic comfort and function. The patient appreciates the fact that a harness is not necessary for secure suspension. The rolled silicone socket is comfortable and the BOA closure system prevents socket migration over the allograft humeral condyles.

The patient's functional envelope has improved with her ability to control the prosthesis overhead and out to the side away from her body. She also has active control over shoulder internal and external rotation eliminating the need for her to reach across to move the prosthetic hand manually with her intact right hand.

The patient wears her prosthesis 11 hours a day in her home and work environments. She has intuitive control of her prosthetic hand and elbow and has adapted well to controlling the wrist pronation and supination through 4 channel switching. Although her prosthetic component movements are not yet simultaneous, she has increased responsiveness of the prosthetic components which translate into more efficient functional use.

## CONCLUSION

It should no longer be customary to fit a transhumeral amputee without first considering the patient's functional needs, physical anatomy, surgical options, prosthetic options, and the necessity for specialized rehabilitation care. With advances in surgical techniques, advances in prosthetic devices and fitting techniques and the development of upper limb prosthetic therapy programs, the norm in upper

limb amputee care has changed. By preparing the residual limb, the patient will have the best possible foundation for functional prosthetic success. This foundation will also prove to be useful when advances such as osseointegration, implantable electrodes, and pattern recognition become more widely available in the clinic to propel the patient towards further functional success.

Note: The subject of this case study gave informed written consent for research and to having her image used, which was approved by the Western Institutional Review Board (WIRB).

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## DESIGN OF A COSMETIC GLOVE STIFFNESS COMPENSATION MECHANISM FOR TODDLER-SIZED HAND PROSTHESES

Ronald A. Bos, Dick H. Plettenburg\*

*Delft Institute of Prosthetics and Orthotics, Delft University of Technology, The Netherlands  
d.h.plettenburg@tudelft.nl, www.dipo.3me.tudelft.nl*

### INTRODUCTION

The missing of an upper limb has dramatic consequences on one's physical abilities and mental health [1-3]. An upper limb prosthesis is designed to improve those factors and can do so with a perfect fulfilment of the three main requirements in prosthesis design: cosmesis, comfort and control [4]. However, reports show that a lot of users are dissatisfied with their prosthesis and rejection rates range from 20%–40% [5]. Overuse injuries to the remaining limb, as well as back- and neckpain often account for rejection. If not rejected, active prostheses sometimes end up being used only for cosmesis [4-6]. Clearly, current prostheses do not meet all requirements.

Among body-powered hand prostheses the mechanical efficiency is low, whereas hooks often show much better performances [7-10]. The activation forces for hands are high and range from 60–130 N for a small 15 N pinch force [10], while the comfortable limit is at 40 N [11]. High activation forces also negatively influence the proprioceptive feedback. The addition of a cosmetic glove is the most prominent cause of these high forces. The viscoelastic behaviour of the material adds a large amount of stiffness and energy losses due to hysteresis [12-15]. Moreover, it limits the usability of voluntary closing devices (a prosthesis that closes upon activation by the user), which are advantageous due to their increased mechanical efficiency and proprioceptive feedback compared to their voluntary opening counterparts [8; 16]. This problem increases in severity for toddler-sized prostheses, as the relative thickness of the glove increases and children are less capable of producing such forces [17]. It appears that the desire for a natural looking hand, i.e. improved cosmesis, counteracts the comfort and control of the prosthesis – being the main advantages of a body-powered prosthesis.

Several solutions are possible in order to address this problem, e.g.: cosmetic glove omission; cosmetic glove modification; prosthesis modification; or, cosmetic glove stiffness compensation. However, the cosmetic glove is indispensable [18], alternative materials are very hard to find [19-23] and glove and prosthesis modification can only provide for partial solutions [12]. Consequently, compensating the glove stiffness is left as solution. This

ideology has already led to a series of mechanisms and methods [12-14; 24-26], but none of them have resulted in a working concept due to challenges that lie in the non-linear behaviour of the glove stiffness, high occurring forces and small working volume.

This study aims to design a new glove stiffness compensation mechanism for a toddler-sized hand prosthesis. It should fit into the wrist of the prosthesis and its goal is to reduce activation forces to a minimum. Consequently, emphasis is put on reducing energy dissipation within the mechanism and maximising energy density. This is done by completely omitting the use of bearings and any sliding contact by using the concept of rolling link mechanisms [24] to reduce friction losses to a minimum, leading to the development of a Rolling Stiffness Compensation Mechanism (RSCM). As a result, the apparent presence of the cosmetic glove stiffness can be decreased.

### METHODS & MATERIALS

#### Design criteria

The main objective for the compensation mechanism was to reduce activation forces down to the comfortable limit of 40 N [11]. Furthermore, it should passively open the prosthesis and create a voluntary closing device, due to the advantages in enhanced feedback. A toddler-sized WILMER WHD-4 prosthesis, with a mass of 69 g, was used as a reference point [27]. Consequently, the compensation mechanism needed to be light-weight and fit into a compatible wrist, which may range from 30–38 mm in diameter [28]. At such small scale, the use of hinged and sliding joints becomes impractical and can introduce coefficients of friction of up to  $f = 0.2$ . For this reason, the concept of rolling link mechanisms was used, which is a method that uses only rolling friction and is able to reduce the friction coefficient down to  $f < 0.001$  [24].

#### Design approach

The stiffness of the cosmetic glove-prosthesis combination can be compensated for by adding a negative stiffness element in parallel to the prosthesis mechanism, such that the addition of the two gives a reduced, resultant stiffness. A prosthesis with cosmetic glove generally

possesses a progressive stiffness characteristic [12-14]. By mimicking the shape of this stiffness characteristic, but acting in the opposite direction, the resultant stiffness can be reduced to a minimum. Moreover, a voluntary closing device can be acquired with overcompensation (Figure 1).

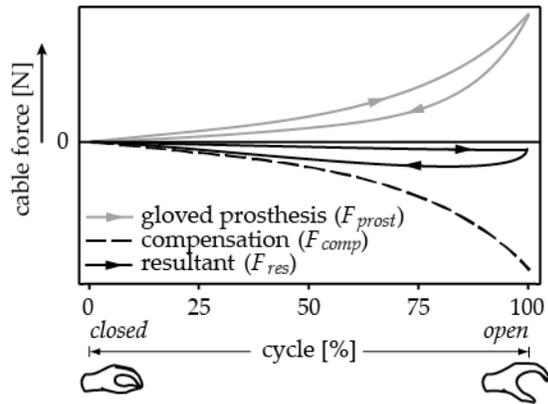


Figure 1: An ideal case of overcompensation of the glove characteristic. The operation of the resultant characteristic is reversed and turns an otherwise voluntary opening device into a voluntary closing device.

### Conceptual design

The combination of linear helical springs as energy storage and a rolling link mechanism as mechanical linkage, led to the design of a Rolling Stiffness Compensation Mechanism (RSCM). The RSCM's overall shape and how the parts connect through stabilisation bands is shown in Figure 2a. Its working principle is shown in Figure 2b in three steps, where the force symbols correspond to those in Figure 1. The three steps are:

1. The glove stiffness has the tendency to close the hand, resulting in a force from the gloved prosthesis ( $F_{prost}$ ). The compensation force from the RSCM ( $F_{comp}$ ) counteracts this force. Because  $F_{comp} > F_{prost}$ , the hand passively opens.
2. As the user pulls on the RSCM ( $F_{res}$ ),  $F_{comp}$  decreases and the glove stiffness will cause the hand to close ( $F_{res} + F_{prost} > F_{comp}$ ). As the hand closes, elastic energy from the cosmetic glove is transferred to the springs in the RSCM.
3. The hand is closed and the glove is relaxed ( $F_{prost} \rightarrow 0$ ). The springs in the RSCM are fully loaded while  $F_{comp}$  is minimal. Increase in  $F_{res}$  will now only increase grip strength. Because  $F_{comp} > F_{prost}$ , the hand will passively open again if  $F_{res}$  becomes zero, returning the system to step (1).

### Stabilisation

Although rolling link mechanisms show very low values for energy dissipation, they need to be stabilised by a stabilisation band [24]. This is done by cross-weaving steel bands through the rolling elements and fixating their ends under tension with micro spot welds (see Figure 2a).

In order to prevent misalignment and asymmetry during operation, the part on which  $F_{res}$  operates (see Figure 2b) is guided between two axes. These axes are fixed on the lower intermediate body and fitted with plain bearings. This same part also prevents the rollers from rolling inwards too much, which can put the mechanism in a form-lock.

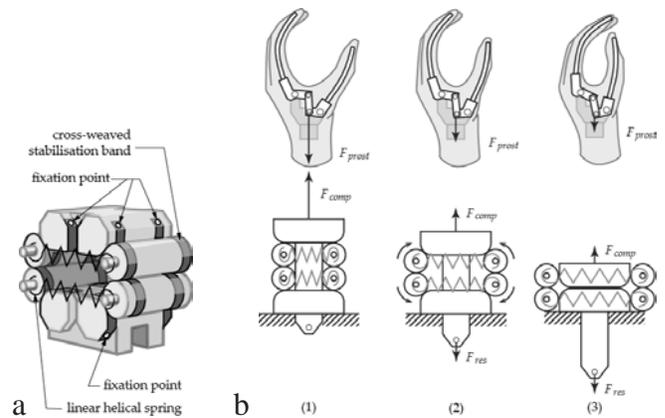


Figure 2: Conceptual design of the Rolling Stiffness Compensation Mechanism (RSCM) with (a) its overall shape and how the parts connect through cross-weaved stabilisation bands, indicating the visible fixation points, and (b) its working principle in combination with the WILMER WHD-4 mechanism.

### Data acquisition

Stiffness characteristics were measured of the gloved prosthesis ( $F_{prost}$ ), the prototype ( $F_{comp}$ ) and the combination of the two ( $F_{res}$ ). A custom-built test bench was used to obtain the stiffness characteristics by measuring absolute force and displacement [9; 10; 13-15]. The direction of force corresponded to the situation as shown in Figure 2b. In the test bench, the measurand was fixed into position and connected to a cable which inflicted displacement. The force at the end of the cable was measured by a load cell (model: B3G-C3-50kg-6B, Zemic, Etten-Leur, The Netherlands) and the position by a linear position transducer (model: LCIT 2000, S/N: J 0069, Schaevitz, Hampton, VA). Both force and position measurements were fed to a data acquisition (DAQ) device (model: NI USB-6008, 12-bit, 10 kS/s, National Instruments, Austin, TX) and into the computer, using LabView version 10.0.1 (National Instruments, Austin, TX) for visualisation and storing of the measured values.

All measurements were repeated five times. Before each series of measurements, the stiffness of connective elements (e.g. cable) was measured separately and eventually corrected for during data processing.

#### Data processing

The measurements involving the prototype were processed by a moving average. The raw data were divided into separate windows of 0.1 mm covered distance, in which a weighted average was calculated using a Gaussian function. Also, using the same weighting factors, a weighted standard deviation was calculated.

For all tests, the amount of input energy was determined by calculating the surface area beneath the loading curve. Hysteresis was determined by calculating the surface area enclosed by the cycle. The efficiency of the RSCM combined with prosthesis was determined by calculating the ratio between the loading and unloading curve.

All data processing was performed in Matlab 2010b (Mathworks, Natick, MA).

### RESULTS

The resulting prototype is shown in Figure 3, where additional protrusions were added to the overall shape to guide the stabilisation bands. The outer dimensions (length×width×depth) are equal to 33×18.6×19 mm when the springs are relaxed (see Figure 3) and 33×26.2×19 mm when the springs are fully tensed. The total mass of the mechanism is equal to 26 g.

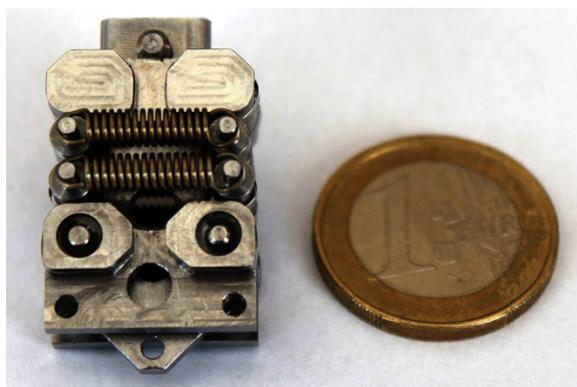


Figure 3: The manufactured RSCM prototype, alongside a Euro coin [Ø 23mm] for scale.

The absolute resultant forces of the RSCM combined with the gloved prosthesis are shown in Figure 4, where a comparison is provided with the original - uncompensated - gloved prosthesis. It can be seen that the operation forces are not necessarily lowered, but the peak forces are greatly

reduced. The resultant value for input energy ( $E_{in}$ ) is 229 Nmm and for hysteresis ( $E_{hyst}$ ) 108 Nmm. These values include the hysteresis of both the gloved prosthesis and the compensation mechanism. They are, however, not equal to the addition of the two, indicating a dependence on operation force.

The stabilisation bands are large contributors to the mechanism's hysteresis. Thinner bands greatly reduce input energy and hysteresis. In the prototype bands of 20 µm thickness were used, and are able to reduce maximum resultant forces down to 40 N, where the RSCM shows an average efficiency of 68% and a combined efficiency with the gloved prosthesis of 52%.

### DISCUSSION

The RSCM is successful in having a progressive negative stiffness characteristic and is therefore suitable as a novel compensation mechanism. However, the mechanism's performance is largely restrained by the stabilisation bands: their thickness causes large hysteresis loops and inaccuracies during assembly may cause misalignment of parts. Nonetheless, it is able to reduce operation forces of a gloved prosthesis and even reverse its working principle with overcompensation, making it possible to turn a voluntary opening device into a voluntary closing device.

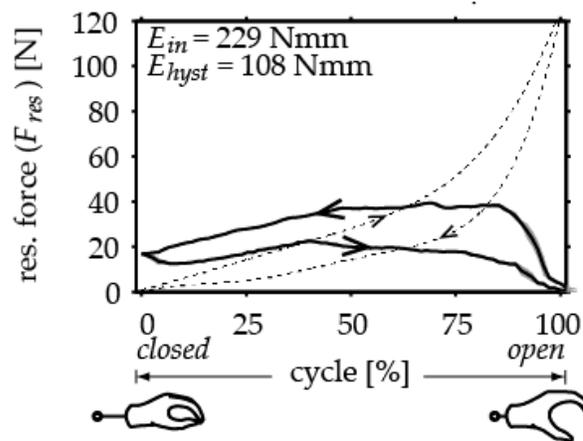


Figure 4: Figure showing the measured absolute resultant force of the gloved prosthesis with compensation mechanism (solid) and compared to the original gloved prosthesis (dotted). Input energy and hysteresis are also shown for the compensated prosthesis. Arrows indicate the direction of the curves, distinguishing loading and unloading curves. Notice how the direction of the curves have been reversed, creating a voluntary closing prosthesis.

A toddler-sized prosthesis fitted with an Otto Bock silicone glove shows a peak force of 120 N and input energy of 277 Nmm. It is shown that the RSCM is able to reduce

the peak force down to the comfortable limit of 40 N [11] and lower the input energy down to 229 Nmm. This is done by using springs with a stiffness of  $1.05 \text{ Nmm}^{-1}$  and stainless steel stabilisation bands with a thickness of 20  $\mu\text{m}$ . Compared to the large reduction in peak force, the input energy is only slightly reduced, indicating that the current RSCM prototype mainly redistributes the necessary input energy.

The small dimensions of the RSCM allow it to be used inside the wrist of a toddler-sized prosthesis. It requires a minimum inner diameter of 33 mm, which may easily fit into a 38 mm diameter wrist. In the future the RSCM can also be integrated with the prosthesis mechanism, further reducing the length of the mechanism. The low mass minimally affects the overall mass of the prosthesis, bringing it to a total of 95 g.

The performance of the mechanism is mostly influenced by the stabilisation bands and misalignment of parts. Specifically, thicker stabilisation bands add more rigidity to the system, introducing a higher stiffness due to elastic deformation and energy dissipation due to plastic deformation. Thinner bands, however, are increasingly difficult to assemble. Their fixations (spot welds) become weaker, inaccuracies are more likely to occur and less pretension can be added. This causes the mechanism to become more sensitive to external factors and individual parts to become misaligned, resulting in non-parallel axes of the rolling elements – one of the larger sources of rolling friction [24]. Improved and more automated assembly strategies should reduce these side-effects, which would allow for thinner bands without compromising in robustness of the mechanism.

Apart from the presented configuration, i.e. the used geometry and springs, other configurations are also possible. In general, the ratio between the roller's radii defines the shape of the RSCM's negative stiffness characteristic. Even non-circular shapes are possible to implement. The springs and stabilisation bands then determine the magnitude and attended losses. This modular principle makes the RSCM suitable to be designed for other types of gloves and even other applications.

In conclusion, the RSCM is a novel negative stiffness element with a large possible area of application. At current stage, its efficiency leaves something to be desired. However, it is believed that its performance can be further increased by improving stabilisation band assembly and constraining out-of-line movements.

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## REDEFINING NORMS SURROUNDING PROSTHESIS ACCEPTANCE AND REJECTION RATES

Kasey R. Poole, OTR/L, MOT, Tiffany Ryan, OTR/L, MOT, Kerstin Baun, OTR/L, MPH,  
Dan Conyers, CPO, FAAOP John Miguelez, CP, FAAOP  
*Advance Arm Dynamics*

### ABSTRACT

The need for objective criteria to identify upper limb (UL) prosthesis acceptance and rejection rates is well documented.<sup>1,2</sup>

This preliminary study serves to evaluate queries representative of factors recognized as influencing prosthesis acceptance. Validated and clinical outcome measures responses of persons receiving comprehensive upper limb prosthetic rehabilitation are utilized.

Preliminary survey results indicate UL prosthesis users participating in a comprehensive, specialized prosthetic rehabilitation setting have lower rejection rates as compared to industry wide published results. Research indicates the upper limb prosthetic patient population will be better served by healthcare professionals cognizant of the defining factors influencing prosthesis acceptance and incorporating this awareness into specific standards of care for this patient population.

### INTRODUCTION

For decades, professionals within the UL prosthesis industry identified and sought to address the need to better quantify prosthetic acceptance, rejection and abandonment. However, an industry consistent criterion to define prosthesis acceptance and rejection still remains obscure. Bidiss and Chau's twenty five year review of surveys states that, "trends and inferences regarding prosthesis usage and acceptance have remained particularly elusive" as a result of the "heterogeneous population base and methodological differences that restrict comparisons between studies."<sup>3</sup> A wide variance of acceptance and rejection rates among adult upper limb amputees with a range of 0-100% rejection. This variance leads to a diminished understanding of the factors truly affecting acceptance, use and rejection of an UL prosthesis. Biddis' and Chou's thorough comparison of survey results within the framework of Anderson's model for health services to categorize factors influencing prosthesis acceptance.<sup>3,4</sup> The UL amputee results in this study are

representative of a portion of the factors identified in the above survey.

### METHODS

The preliminary study pool consisted of upper limb amputees of varying levels of amputation, presenting for UL prosthetic rehabilitation at multiple Advanced Arm Dynamics (AAD) centers throughout the United States giving informed written consent for this study, approved by Western Institutional Review Board (WIRB). Inclusion criteria also include: English speaking; 18-70 years of age, with varying levels of unilateral and bilateral limb deficiency.

Participants received UL prosthetic rehabilitation intervention via standardized protocols throughout all AAD centers. Upper limb prosthetic rehabilitation specialist teams in each center include certified prosthetists, occupational or physical therapists on-site throughout the rehabilitation process. Ongoing support services to this group include psychological screening, guidance and oversight by a board certified neuropsychologist, access to peer support and coordination of UL rehabilitation health services.

Outcomes measurement tools were utilized at specified fitting stages, from pre-prosthetic to post-definitive. Outcome measure results utilized for this study included the Comprehensive Arm Prosthesis and Rehabilitation Outcomes-Revised (CAPROQ-R) and Trinity Amputation and Prosthesis Experience Scales- Revised (TAPES-R). This study is an analysis of preliminary data collected at various stages of prosthetic rehabilitation. These two outcomes were utilized at different points in the rehabilitation process, per standard clinical care protocols and will be explained separately.

### CAPROQ-R

CAPROQ-R: an internally developed UL amputee specific questionnaire outcome measure undergoing steps for validation. Specific factors influencing individual patient's perception of prosthesis acceptance include functional use and overall satisfaction with their prosthesis.

CAPROQ-R data was assimilated from 159 data sets consisting of 70% male and 30% female; age 18-70 (mean = 61 yrs.); 40% right, 48% left, 12% bilateral UL deficiency. Cause of UL deficiency results: injury 70.5%, disease 16.7, congenital 12.1%, other .08 %. Amputation levels represented: partial hand 29%, transmetacarpal 2%, wrist disarticulation 5%, transradial 39%, elbow disarticulation 1%, transhumeral 16%, shoulder disarticulation 5%, other 1%. Of the subjects reporting currently participating in the prosthesis fitting process (n=99), 72% reported having a secondary prosthesis.

#### CAPROQ-R Variables

The following variables influencing acceptance were evaluated: 1) type of prosthesis or prostheses, 2) wear time, 3) wear patterns, 4) reasons for non-wear, 5) comfort, 6) prosthesis satisfaction

#### TAPES-R

“The TAPES-R is a multidimensional assessment designed to facilitate examination of psychosocial processes involved in adjusting to prosthesis and the specific demands of wearing a prosthesis.”<sup>5</sup> This measure was originally intended to be used for lower limb amputees, however, has since been validated for use with upper limb amputees.<sup>6</sup>

#### TAPES-R Variables

Variables evaluated include subscales for these scale categories: 1) Psychosocial adjustment, 2) activity restriction and 3) prosthesis satisfaction.

## RESULTS

#### CAPROQ-R:

##### *Prosthesis Wear and Use*

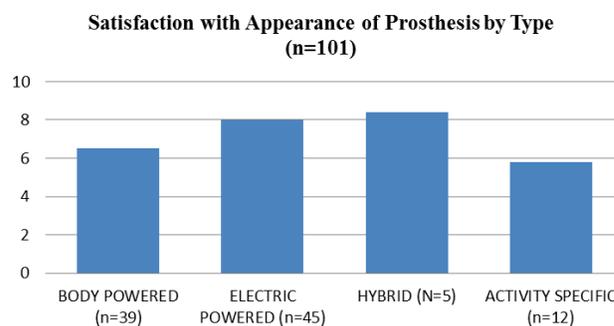
Preliminary data rendered 89.2% of patients wore their prosthesis. Subjects reporting daily prosthesis wear were 53.8% with wear time ranging from 1 hour to 18 hours and an average wear time of 8.94 hours. Respondents reporting hours of active use of component features during wear time (n=70); 0-4 hours 61%, 5-9 hours 21%, 10-14 hours 15%, 15-19 hours 1%.

Of the respondents answering as to their level of current wear (n=91), 12 subjects or 13% reported stopping prosthesis wear. The five most common reasons for not wearing a prosthesis on a daily basis by those responding (n=29) were: the prosthesis was not comfortable 37.9%, subject was just as, or more, functional without a prosthesis 34.5%, prosthesis was too hot 20.7%, prosthesis was too heavy 24.1%, prosthesis was difficult or tiring to use or control 20.7%.

##### *Prosthesis Appearance and Need*

CAPROQ-R respondents also ranked satisfaction with the appearance of their prosthesis (es) by type (n=101). See Table 1 below. The mean number of years this group has had a prosthesis is 7.92 yrs.

Table 1: Satisfaction with Prosthesis Appearance 0-10 by Type; CAPROQ-R



When patients were asked to rate their need for a prosthesis on a scale of 0-10 with 0 being “not at all for daily life” and 10 being “absolutely essential for daily life” (n=130), the average rating of need for a prosthesis for daily life was 7.86.

#### TAPES-R:

The TAPES-R subject pool included responses from 53 subjects, age 18-70 with mean age of 47 years; inclusion criteria identical as those indicated in CAPROQ-R above. Causes of limb deficiencies for the group were 85% trauma, 11% congenital and 4% medical and amputation levels were partial hand 23%, above elbow 21%, below elbow 56%, shoulder disarticulation 2%. The group mean for years since amputation 15.51 years, years using a prosthesis 11.2 years, and years in current prosthesis 2 years.

##### *Prosthesis Wear, Use and Satisfaction*

On average, patients reported wearing their prosthesis an average of 6.46 hours per day. The mean score for functional satisfaction was 11.04 on a scale of 5-15 possible points, a high score indicates satisfaction. Aesthetic satisfactions score mean was 6.71 of 3-9 possible points scale with a high score indicating satisfaction. The mean group score to the question asking subject to rank best description of how satisfied they are with their prosthesis 7.84 on 0-10 positive scale.

##### *Prosthesis Psychosocial Adjustment*

The TAPES-R Part I Scale for psychosocial adjustment with 1 being strongly disagree to 4 being strongly agree are listed in the Table 2.

Table 2: TAPES Psychosocial Adjustment to Prosthesis Subscales, Positive Ranking 1-4

TAPES Psychosocial Adjustment to Protheses (n=53)	
General Adjustment	3.24
Social Adjustment	3.32

The mean score representing adjustment to limitation was 1.56. A high score on a scale of 1-4 indicates a positive adjustment. This scoring category queries subjects as to whether having a prosthesis interferes or limits the amount, type and kind of work they can do and if having a prosthesis makes them more dependent on others and limits their ability to do what they want to do.

### DISCUSSION

As defined by Merriam Webster, “accept” means “to give admittance or approval to” or to “endure without protest or reaction” not to be confused with the term “success”.<sup>7</sup> Regardless of wear pattern, it is our conclusion that a patient accepts their prosthesis if they wear it at all and if it allows them to accomplish what they desire to accomplish. Acceptance knows no time trajectory—a patient might only use their prosthesis once a month to perform a specific activity, which means they have accepted it to serve that purpose.

In terms of data comparison surrounding acceptance and rejection, this study pool’s rejections rates and wear are lower than those found by Biddiss and Chau in 2007. See Table 3 below.

Table 3: Mean Rejection Rate for All types of Protheses

Mean Rejection Rate for All Types of Protheses	
Bidiss & Chau <sup>1</sup>	29.3%
CAPROQ-R	13%

CAPROQ-R results indicate 47.9 percent of users had a secondary prosthesis. It is likely that having a secondary prosthesis improves prosthesis acceptance. A secondary prosthesis allows a patient to be able to perform a wider range of activities as well as providing the user an alternative in the event the need for repair of the primary prosthesis.

Improved CAPROQ-R and TAPES-R rates may be attributable to aspects of comprehensive rehabilitation services which are in alignment with the factors influencing acceptance as outlined by Biddis and Chou. Subjects in this study pool participated in extensive evaluation of patient needs and desires, education to address expectations based on realistic functions of current prosthetic systems and components, comprehensive treatment plan which includes a

UL amputee specific therapy/training regimen throughout the rehabilitation process to ensure integration of the prosthesis into ADLs, IADLs, work and community reintegration; a thorough outcome measures protocol and lastly, ongoing clinically relevant, routine follow-up as a standard of care. Follow-ups have been documented as an important factor affecting prosthesis acceptance with delayed follow-ups are associated with higher rates of rejection.<sup>8</sup>

### LIMITATIONS

This is a preliminary study and further statistical analysis to discover underlying causes for acceptance, patterns of wear, and preferences among prosthetic users is warranted. The variance in population and query format for each measure does not allow for direct comparison of responses.

### CONCLUSION

These findings suggest upper limb amputees may have a higher rate of prosthetic acceptance than previously postulated, specifically when they receive comprehensive UL prosthetic rehabilitation with regularly scheduled follow up. Ongoing data collection will render more long term results to compare with this early data. The inclusion of additional data representing psychosocial factors impacting acceptance may enhance the depth of knowledge. It is our expectation that with a properly designed and implemented comprehensive rehabilitation plan, further results will reveal a positive trend in prosthesis acceptance rates.

Continued research to identify key factors defining prosthetic acceptance is necessary. Obviously, a pertinent aspect in defining prosthesis acceptance must include a thorough analysis of UL prosthesis users’ perspective on the topic. As these factors are identified, patients will be best served with utilization of industry accepted validated measures of these factors.

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## MUSCULOSKELETAL COMPLAINTS IN MAJOR UPPER LIMB DEFECTS IN THE NETHERLANDS: PREVALENCE, INFLUENCE ON HEALTH STATUS AND WORK AND RISK FACTORS

Sietke G. Postema<sup>1</sup>, Raoul M. Bongers<sup>1</sup>, Helena Burger<sup>2</sup>, Liselotte M. Norling Hermansson<sup>3</sup>, Michiel F. Reneman<sup>1</sup>, Michael A. H. Brouwers<sup>4</sup>, Pieter U. Dijkstra<sup>1</sup>, Corry K. van der Sluis<sup>1</sup>

<sup>1</sup>University Medical Center Groningen, University of Groningen; <sup>2</sup> University Rehabilitation Institute

<sup>3</sup>Örebro University, Örebro County Council; <sup>4</sup>Rehabilitation Center de Hoogstraat, Utrecht

### ABSTRACT

#### Objectives:

(1) To compare the prevalence of self-reported musculoskeletal complaints (MSC) in individuals with major upper limb defects (ULD) in the Netherlands with a control group, (2) to explore the influence of MSC on health status and work and (3) to assess predictors of MSC, disability and work productivity in ULD.

#### Methods:

A national survey among individuals with ULD and controls was performed, using the databases of rehabilitation centers and orthopedic workshops in the Netherlands. A questionnaire was designed based on known risk factors for MSC, and it included validated (subscales of) existing questionnaires, such as SF36 and the Pain Disability Index (PDI). Inclusion criteria were  $\geq 18$  years and major ULD at or proximal to the carpal level. Controls were recruited by convenience and matched on age and sex.

#### Results:

Of the 263 individuals with ULD that completed the questionnaire, 42% had a congenital transversal defect and 58% had an amputation. The mean age was  $50.7 \pm 16.7$  years and 60% was male. A prosthesis was used by 79%. In total 108 controls were included (mean age  $50.6 \pm 15.7$ ; 65% male).

Year prevalence of MSC (lasting for at least four consecutive weeks) was 65% in individuals with ULD, compared to 34% in controls.

The most common location of MSC was the dominant or non-affected limb (46% in patients and 17% in controls), followed by upper back/neck (43% in patients and 19% in controls). Presence of MSC was associated with lower scores on scales of general health perception, mental health, work productivity and higher scores on disability. Prosthesis use

did not differ between individuals with and without MSC. Predictors for presence of MSC were deficiency of the right limb, higher upper extremity work demands and being divorced or widowed. More pain, lower mental health and higher age were associated with a more severe disability. Predictors for lower work productivity were presence of MSC and more pain.

#### Discussion:

Presence of MSC is a common problem in individuals with ULD. Mostly affected were the non-affected limb and upper back/neck. More research on employment of the affected and non-affected limb, and its relation with MSC, is therefore warranted. Interestingly, presence of MSC was not related to prosthesis use. Associations with disability and work productivity add extra relevance to the study results, because of its relevance for individuals and society.

#### Conclusion:

Prevention and treatment of MSC deserves an important role in rehabilitation medicine of individuals with ULD.

## **PARTIAL HAND AMPUTATION - REDEFINING THE “NORM” IN MULTIDISCIPLINARY PROSTHETIC CARE AS NEW TECHNOLOGY EMERGES AND AS WE ARE ABLE TO COLLABORATE WITH MORE PROGRESSIVE SURGEONS.**

Pat Prigge<sup>1</sup>, Brian Waryck<sup>1</sup>, John Miguelez<sup>1</sup>  
<sup>1</sup>*Advanced Arm Dynamics, Inc.*

### **ABSTRACT**

In clinical practice, partial hand amputation cases challenge the surgical and rehabilitation team to develop a treatment plan that gives the patient the best possible outcomes functionally. Surgically, preserving length at the partial hand level still remains important but as more prosthetic options become available for the partial hand amputation level, the need for further multidisciplinary dialogue becomes critical. We now know that some amputation levels are ideal given specific recent advances in technology. If the surgeon is not equipped with this knowledge, the results of an amputation or reconstructive surgery can be sub-optimal as it relates to prosthetic fitting. Creating a collaborative dialogue can help ensure that the patient, surgeon, prosthetist and therapy team have considered all available prosthetic options, the most current prosthetic technologies, as well as the advantages and disadvantages of each design for the patient.

### **INTRODUCTION**

According to studies<sup>1</sup>, there are approximately 16,000 amputations performed at a level distal to the wrist every year. This same study suggests that the number of amputations at the wrist and proximal to the wrist, number around 1,600 per year. Until recently, the vast majority of available prosthetic components and technology advancements have focused on the smaller subgroup. There still remains a considerable need for the larger subgroup of people who, following a partial hand amputation, are left with diminished functional capacity. The functional benefit of prosthetic usage for patients with multiple finger amputations proximal to the DIP joint has been demonstrated both in terms of measurable “improvements in 3-finger-pinch and grip strengths” and the “global subjective improvement in the ability to perform activities” as reported by patients themselves<sup>2</sup>. We also know that studies indicate several factors for abandonment. Those include comfort, training, experience of the clinic team, being part of the decision process, function and others<sup>3</sup>. Including recent advancements in partial hand prosthetic technologies, considerable opportunity exists for those with amputations distal to the wrist to have a well fitted, secure and functional prosthesis.

In today’s competitive health care arena, it is increasingly difficult to establish a comprehensive team that can function collaboratively for the benefit of the patient. Typically, the surgical team is the first to be involved with the care of this patient population, given the specific nature of hand trauma. Identifying a hand surgeon that is interested in these discussions opens the door for education, regarding the latest prosthetic options, fitting techniques and training and sets the stage for dialogue on the subject of ideal residual limb length and remnant musculature for prosthetic fitting. In some parts of the country, there are teams within large medical facilities that can gather these resources quickly for the benefit of the patient. In other settings, it will typically resemble a network of independently employed specialists, with both a willing interest and a comprehensive understanding of their specialty and the role that plays in supporting the collaborative management of these cases. Once this team is in place, it also presents an opportunity to manage these cases going forward, in the event that cases are seen with a less than ideal residual limb presentation, which require a discussion around surgical revision.

A thorough clinical evaluation with a team including the patient, prosthetist, therapist and surgeon will give a rounded approach to the options both surgically and prosthetically. Once the optimum surgery has been performed and the limb is left in an idealized situation, the process of pre-prosthetic training and fitting begins. Depending on the specific presentation of the limb and the functional goals of the patient, the specifics of the design are determined.

Socket design is paramount for the partial hand amputation level. It must be designed to maintain suspension, maximize functional wrist utilization and enable all functions remaining in the hand. Selecting the components that fulfil the patient’s functional needs and utilizing an expedited fitting procedure is necessary to see quick achievement of optimized fittings.

### **CASE STUDY 1**

Patient is a five y.o. male with bilateral partial hand loss secondary to a fall into a day old camp fire pit at 13 months of

age. The original presentation was bilateral transmetacarpal partial hands with no fingers or thumbs present. At age four the parents opted to have a toe transfer done on his right hand. The second toe from his right foot was transplanted to his right hand. Currently he presents with left hand transmetacarpal amputation and right hand transmetacarpal with toe transfer in the thumb position. Upon evaluation it was noted that he had minimal grasping capability and the toe could not tolerate extension force without collapse. The physician in charge of his treatment suggested that additional toes could be transferred to give him some grasping capability. The parents were reluctant because of concern for his overall functional level and mobility as an active child and wanted to provide the best potential outcome for him. They also discussed the potential for hand transplants when he is old enough and that it would require life threatening immunosuppressive medications. Since he was too young for a transplant and the parents were fearful of additional toe transplants impacting his mobility, they opted to pursue prosthetic solutions.

The clinic team assembled and discussed the prosthetic options available to him. A passive device was recommended that replicated the form of a hand but had ratcheting joints in the fingers so that the fingers could be pre-positioned and make use of the innate power of his “transferred toe” thumb to grasp and achieve pinch force against the fingers. This would also allow for quick repositioning of the fingers for various sized and shaped objects.

The device was fitted and comprehensive therapy was provided to maximize his function with the device. A preparatory device was fabricated to facilitate this training and to optimally work out the design and alignment of the fingers. With this device he learned very quickly how to pre-position the device for various tasks.



Figure 1: Patient repositioning device



Figure 2: Fingers in extension with thumb

The socket was fabricated from PETG and designed to maximize his wrist range of motion. The toe to hand thumb had a path of movement that was not typical of an intact thumb so special consideration was made in the design of the socket to accommodate this motion. The fingers were single joint ratchet mechanisms that can be repositioned with the other hand or by pressing them against objects. The glove is a PVC passive glove that was matched to the size of his hand and also what fit best onto the socket. The two were bonded together and a wrist strap was added around the base of the thumb to enhance suspension force. Once the alignment was final the socket was laminated with a thin PETG inner surface to keep the lamination away from the skin, ease donning and doffing and produce the lowest profile socket possible.

Currently he uses the device daily at school to independently accomplish his educational requirements. He can quickly don the device to aid in writing or other tasks where opposition support is needed. He can also efficiently remove the device when he is more comfortable completing the task without the device. Prior to the delivery of the device he could not write except by holding a pencil with both hands and having someone hold the paper. With the device he wrote out his name and signed in cursive. Everyone was impressed at what he was quickly able to do with the device.

Based upon the positive outcome of this fitting, the physician decided to discuss additional surgeries with the family. One surgical option was to consider a toe transfer for the left side to duplicate what has been accomplished on the right side. A second surgical option was a tendon transfer on the right side, with the goal to strengthen the opponens force that is lacking in the transferred toe. The parents are encouraged to have these options available to their child until he is old enough for hand transplants, which at this time are their ultimate goal.

## CASE STUDY 2:

Patient is a 57 year old male with bilateral partial hand amputations, MCP level at fingers and thumbs, secondary to severe frostbite in 2012. In 2013 he underwent a great toe transfer from his right foot to his right hand. The surgery was successful but he had no ability to grasp anything with the thumb because there was nothing to oppose it. The CMC joint of the thumb remained with full ROM but the MCP and the IP joints of the thumb lacked any active ROM. The physician considered doing other toes but his health and physical conditions of the vessels supplying his toes were not in good enough condition to consider transferring a smaller toe. This left a dilemma as the toe transfer alone did not produce the functional outcome desired.

The physician asked what could be accomplished prosthetically to provide function for the individual. Many options were discussed but the clinical team selected a passive device with four double ratcheting finger joints to mimic the MCP and the PIP joints. Doubling the joints was required because of the limited ROM of the thumb joints.

A preparatory device was fabricated with PETG and the joints were bonded directly to the socket with quick set epoxy. Injected silicone was used to give the fingers some semblance of anthropomorphic shape in order to fill a cosmetic hand shell. A strap was configured that allowed him to adjust the tension on this device without the use of velcro. This facilitated independent donning and doffing as well as adjustment of the strap. A definitive prosthesis was then fabricated from custom rolled silicone with a direct lamination technique which secured it directly to the frame. A Michelangelo glove was used to provide the cosmesis which fit very well over the mechanism.

Intensive therapy was performed to educate the patient on his prosthesis and residual limb care, repositioning of the joints, functional skills and education in proper posture and body mechanics.

With the prosthesis the patient was more independent with his ADL's. He demonstrated two handed activities that required less assistance with his contralateral side. Activities such as holding a drink, that he otherwise would use both hands for, were done single handed. He was able to zip his coat, use the device for eating with regular utensils and generally found the device to be very useful. The physician after seeing the functional advantages of this device has scheduled a toe transfer for his other hand.



Figure 3: Ratcheting joint prosthesis



Figure 4: Repositioning little finger

## CONCLUSION

Case studies demonstrate how this collaborative approach has led to unique information and idea exchanges between surgeons and prosthetics teams to explore detailed surgical approaches such as toe to hand transfers that subsequently require the most current socket design and advanced prosthetic technology to successfully fit partial hand amputations. The synergy between the teams produces results that individuals acting alone cannot achieve.

All participants gave informed written consent for this research study and to having their image used, which was approved by Western Institutional Review Board (WIRB).

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# HIGH RESOLUTION MUSCLE PRESSURE MAPPING FOR UPPER LIMB PROSTHETIC CONTROL

Ashkan Radmand, Erik Scheme, Kevin Englehart  
*Institute of Biomedical Engineering, University of New Brunswick*

## ABSTRACT

This work investigates whether surface pressure changes caused by the contraction of the forearm muscles can be used as the primary information source for robust prosthetic control. A technique called Muscle Pressure Mapping (MPM) is used to detect patterns in the muscle pressure between the residual limb and socket using a high density array of custom pressure sensors. Based on the achieved results, MPM is shown to significantly outperform an EMG based system in detecting different wrist and hand gestures. Eight classes of hand motion were classified with 99.67% accuracy using MPM technique. Comparatively, a standard 6-channel EMG based approach yielded a classification accuracy of 89.38%.

## INTRODUCTION

Several multi-degree-of-freedom upper limb prostheses have recently been developed, with the promise of highly dexterous control. Inadequate controllability of these devices, however, has limited the adoption of these devices. It is expected that improving existing control strategies or introducing novel robust methods of control will result in higher acceptance rates.

Surface electromyography (EMG) has long been used as one of the major neural control sources for powered upper limb prostheses. It contains useful information about the neuromuscular activity from which it originates and can be used to extract the user's intent [1]. Various EMG signal processing methods have been used to extract movement intent. Conventional myoelectric control systems use an estimate of the signal variance to quantify the intensity of contraction in muscles as the control input. Although such control schemes have been widely used commercially, they are incapable of controlling more than one or two degrees of freedom (DOF) [1,2]. Pattern recognition-based myoelectric control is an intelligent signal processing technique that can potentially be used to control multiple DOFs. In this approach, a set of features containing spatial and temporal information about the acquired signals are extracted and forms an input to a pattern classifier which determines the user's intended movement [1,3-5]. Because of consistent

improvements in the clinical performance of this technique, pattern recognition-based myoelectric control has found acceptance and has recently been commercially deployed for the first time [6]. EMG, however, is a noisy and stochastic signal in nature, and therefore its robustness to things like changes in electrodes locations and residual limb movements remains a concern. This has motivated the investigation of an alternative control inputs that may offer a higher degree of robustness and less susceptibility to variations in usage conditions.

Many movements of the hand are controlled by muscles in the forearm, known as extrinsic hand muscles. These muscles, located in the anterior and posterior compartments of the forearm, control movements of the wrist, hand, fingers and thumb. When the forearm muscles contract to move the hand, some muscles bulge outward while some others recede inward from the surface. This results in pressure changes observable between the surface of the forearm skin and the socket. The aim of this work was to study whether this pressure pattern can be used as the primary information source for prosthetic control. The hypothesis is that the pressure patterns generated by various hand motions are distinct enough to differentiate the various motions from each other.

Other groups have previously measured pressure differences using targeted pressure sensors to extract movement intent. Phillips and Craelius [7], who first proposed this idea, used pressure sensors to produce topographic maps of the pressure exerted against the hard prosthetic socket. They used these pressure patterns to distinguish volitional finger flexion/extensions. Yungher and Craelius [8] then used an array of force sensors to measure pressure changes on the skin and discriminated 6 different grasps with high accuracy. Li et al. [9] recently used an array of 32 FSR sensors combined with a SVM classifier for finger motion recognition based on pressure distribution maps. They were able to identify 17 different finger motions in within-session validation. Other groups [10] have used pressure sensors in attempts to minimize the effects of pressure on EMG but they have never used it as the primary information source for control

In this work, the pressure patterns are distinguished in a similar technique which we refer to as Muscle Pressure Mapping (MPM). Although similar to previous methods, MPM monitors the shape of the limb by completely encircling it using a high density grid of custom force sensors. This advantage enables MPM to accurately sense even small changes in the surface pressure map of the limb. Also, with greater coverage and resolution, it requires no custom placement of electrodes within the socket. MPM uses pattern recognition algorithms to discriminate the patterns and determine intended motion classes, suited for prosthetic function. The results of this technique are compared to those of a commonly reported pattern recognition based EMG control approach [4] using the same experimental protocol.

## METHODOLOGY

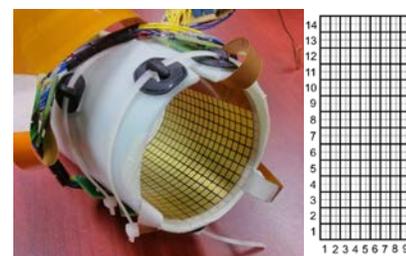
### Population and data acquisition

Because of the coverage of the MPM, concurrent EMG collection was unfeasible. Consequently, two identical experiments were performed; one for collecting pressure data and one for collecting EMG data. Pressure maps corresponding to eight classes of motion were collected from ten right-handed, healthy, normally-limbed male subjects within the age range 25 to 33 years. Also, EMG data corresponding to the same classes of motion were collected from eight right-handed, healthy, normally-limbed subjects (7 male, 1 female) within the age range of 19 to 32 years. All experiments were approved by the University of New Brunswick's Research Ethics Board (REB# 2013-030).

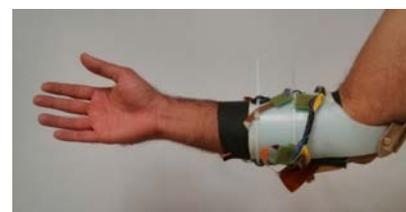
The pressure maps were recorded using a custom high-density grid of 126 (14 rows and 9 columns) pressure sensors of size 1 cm × 1 cm mounted inside an adjustable prosthetic socket as shown in Figure 1-a. The flexible sensor grid covered the dominant forearm, centred at the position with largest muscle bulk, with the sensors encircling it. Two zip ties, one at the top and one at the bottom of the socket, were utilized to adjust the socket size for each subject (Figure 1-b). Because of the density and coverage of the grid, targeted placement of the sensors was not required. The 126 channels of pressure data were sampled at 20Hz by a custom data collection system. Data from the pressure sensors generate a pressure map image, composed a single amplitude value for each of the 14 x 9 pixels, for each reading.

To collect surface EMG data, a Trigno Wireless System (Delsys Inc., USA) was used. Six wireless electrodes were equally spaced placed around the dominant forearm, proximal to the elbow, at the position with largest muscle bulk. The six channels of EMG were band-pass filtered (20-500Hz Butterworth) and sampled at 1 kHz by a custom data collection system.

Subjects were prompted to elicit contractions corresponding to eight classes of motion including wrist flexion/extension (WF/WE), wrist supination/pronation (WS/WP), chuck grip (CG), power grip (PG), hand open (HO), and no movement (NM). Each contraction was sustained for three seconds and a three second rest was given between subsequent contractions. Subjects were instructed to perform contractions at a moderate and repeatable force level and given rest periods between trials to avoid fatigue. The data were collected while subjects were seated in an armchair and held the arm in a fixed position with the elbow resting on the chair's arm. Four sets of contractions were collected for each subject and the average duration of the experiment was approximately 20 minutes per subject.



(a)



(b)

Figure 1: (a) Adjustable pressure sensing socket; each 2x2 array of cells forms a single pressure sensor (b) Placement of the socket with zip ties used to adjust its size

### Data Processing and Classification Methods

No specific data processing was performed on the pressure data. Consequently, each decision was computed from an analysis window corresponding to 50ms (a 20Hz sampling rate). Depending on the amount of pressure placed on pressure sensors, the reading value was between 0 and 1 for each sensor and these amplitudes were the only input to the classification stage.

EMG data were digitally notch filtered at 60Hz using a 3rd order Butterworth filter in order to remove any power line interference. Data were segmented for feature extraction using 200ms windows, with processing increments of 100ms. Simple time-domain (TD) features described by Englehart

and Hudgins [4] were extracted from the EMG data for each processing window.

To recognize the acquired pressure patterns and EMG features of different motion classes, a linear discriminant analysis (LDA) classifier [4] was used as an effective real-time control scheme for prosthetic control. This approach has been widely accepted [11] because of its relative ease of implementation and high performance.

Fourfold cross validation was used to test the performance of each technique. For each fold, the data from three of the four repetitions of each motion were used for training, and the data from the fourth repetition was used for testing. The classification results reported represent the average across all four folds.

## RESULTS

### Acquired pressure maps

Figure 2 illustrates examples of the pressure maps acquired for each of the eight motion classes computed from a representative subject. In these pressure maps, the areas of low and high pressure are clearly visible for each motion, resulting in a distinct image for each motion.

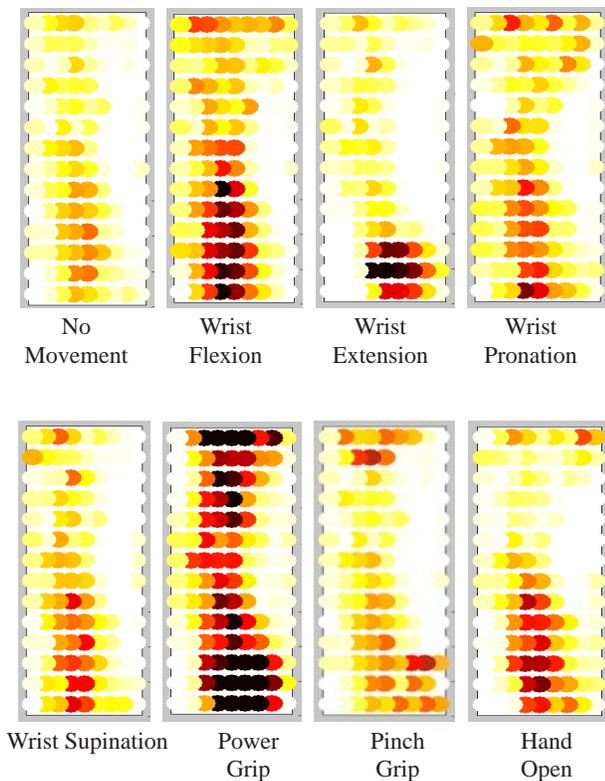


Figure 2: Examples of acquired pressure maps for the motion classes (Darker areas correspond to higher pressure).

### Hand motion classification

Hand motion classification was performed for eight classes of motion using both of the control methods (MPM and EMG), with the users arm in a fixed static position. As shown in Figure 3, the mean overall error was 0.33% using pressure maps while it was 10.62% using EMG signals. The MPM significantly outperformed the EMG based control ( $p < 0.01$ ) in classification accuracy using an analysis of variance with a probability threshold of 0.05. Furthermore, the MPM approach outperformed the EMG based approach within each validation fold for all users.

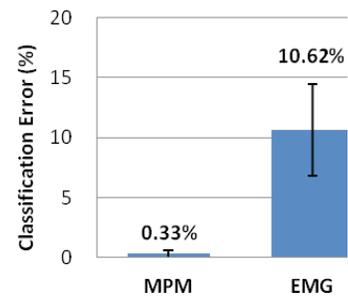
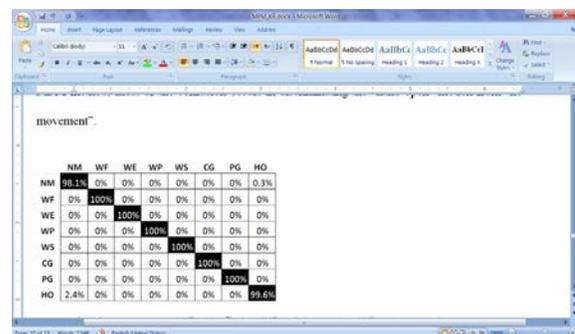
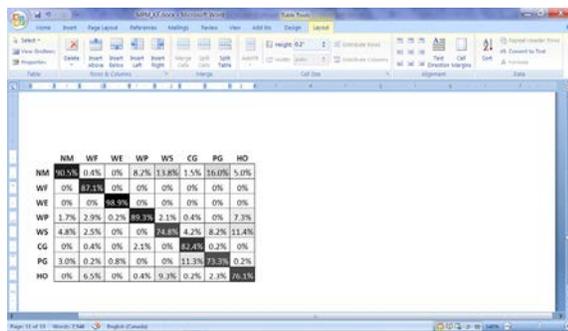


Figure 3: Average classification errors of the MPM and EMG control methods for eight hand motion classes. The error bars indicate the standard error across all subjects.

Inter-class confusion matrices for both methods are shown in Figure 4. Entries of the matrices represent the average accuracy on the diagonal and the average error everywhere else across all subjects and repetitions for the indicated class of motion. A perfect classification result would yield 100% on the diagonal and 0% everywhere else. The achieved results show that, using MPM method, only a small amount of confusion occurred between the “hand open” motion and “no movement”.



(a)



(b)

Figure 4: Inter-class confusion matrices for (a) MPM, and (b) EMG control methods, averaged across all subjects and repetitions. The color coding of the main diagonal entries has been inverted, so that a perfect classification result would yield 100% on the diagonal and 0% everywhere else, and a completely white matrix.

## DISCUSSION

The results of hand motion classification illustrated in Figure 3 showed that MPM control technique is highly accurate in detecting different wrist and hand gestures.

Compared to EMG, this method not only generates significantly higher classification accuracy; it may also have several advantages. With the high resolution and full coverage of socket, MPM does not require targeted electrode location. In addition, pressure sensors are very light and thin, making them comfortable and unobtrusive for the wearer. Existing EMG sensors, such as stainless steel electrodes, apply substantial and uneven pressure on the skin, often creating pain points or skin irritation. The pressure sensor components of the MPM are not even required to be in direct contact with the skin (it could be worn over thin clothing), minimizing any risk of skin reaction. Also, the performance of MPM relies on muscle physiology which is relatively stable over short periods. While fatigue and sweat can change the characteristics of EMG, the physical dimensions of the limb do not change appreciably over moderate time periods. Electrode lift is another problem with EMG control as it changes the EMG signals characteristics introducing substantial noise. This problem, however, does not exist with the proposed MPM method, because contact is not needed; lift simply results in a 0 reading level, similar to that seen during the rest motion.

The analysis window of MPM control (50ms in this work) is much shorter than that of EMG control (200ms in this work). Furthermore, because the proposed method does not rely on temporal information, the length of analysis window is only limited by the scanning frequency of custom

hardware. Although not employed in this work, a majority voting post-processing scheme could be employed within the window length of the EMG approach, further improving the relative performance of the MPM. Finally, MPM generates pressure maps that are similar to a digital image, and uses pattern recognition algorithms to decode the user's intent. Because they are images, the significant body of shift/rotation tolerance in the image processing literature could potentially be employed to accommodate sensor displacement or shift that may occur during or between uses of the device. Ongoing work is therefore investigating the robustness of the MPM approach to confounding factors such as electrode shift and residual limb movement.

## CONCLUSION

Although there is a significant body of research describing different techniques for EMG-based prosthetic control, their clinical robustness is still being improved and novel methods of control are still desired. In this work, we investigated whether pressure patterns between the forearm and a socket can be used as a primary source of information for prosthetic control. The technique, referred to as MPM, uses a high density array of custom pressure sensors and pattern classification algorithms to sense pressure patterns. It was demonstrated that, compared to multi-channel EMG control, this approach is significantly more accurate in recognizing multiple wrist and hand motions for all subjects. Future work will focus on validating the presented technique with amputees and developing sensor shift/rotation and residual limb movement resilient techniques for robust, real-time pattern recognition based control.

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## EFFICACY OF FLEXIBLE, CONDUCTIVE FABRICS FOR ELECTROMYOGRAPHY

Tim Reissman PhD, Beth Halsne CPO, Max Shepherd, Robert Lipschutz CP, Todd Kuiken MD PhD  
*Rehabilitation Institute of Chicago, Northwestern University*

### INTRODUCTION

With the recent development of flexible, conductive fabrics and miniature sensors, researchers have begun to incorporate wearable technology for clinical applications in physical medicine and rehabilitation [1]. In particular, conductive fabrics with elastic properties seem well suited for use as embedded electrodes and signal transmission lines for electromyographic (EMG) signal recording in a prosthetic interface, such as an elastomeric gel liner [2]. However, there is a dearth of information in the literature as to how useful and robust these materials are when used as either electrodes contacting the skin or as conductive leads. Here we report the results of testing eight flexible, conductive fabrics, thereby adding to the literature empirical data to better inform future development of such systems.

### MATERIALS AND METHODS

We identified eight flexible, conductive fabrics to compare in this study. The fabrics, listed in **Table 1**, consisted of two flexible stainless steel meshes, one coarse (fabric #1) and one fine (#2), and six different types of silver-coated fabrics (#s 3-8), listed in order of increasing Nylon content. Each sample was ironed onto a prosthetic liner fabric (Alps South, St. Petersburg, FL) and consisted of a continuous 2.5 x 0.3 cm strip of fabric, which was cut wider at one end to allow a 1 cm diameter silicone dome to fit underneath. By forming a fabric topology similar to a passive, single-ended electrode with a 2.5 cm lead (see **Figure 1**) we could measure electrical resistance (a quantitative indicator for conductivity) between the dome head and a uniform length of fabric for several testing conditions.

We identified two conductivity failure modes that are likely to occur for these fabrics in a gel liner: abrasion (i.e., material stresses due to friction between the fabric and the skin) and corrosion (i.e., tarnishing of conductive materials in the presence of perspiration). We used three tests to examine these failure modes under accelerated conditions (i.e., typical parameters were exaggerated) to determine failure points in a shorter period of time. The test conditions included (i) continuous dry abrasion (in absence of perspiration); (ii)

corrosion (repeated exposure to perspiration), and (iii) wet abrasion, which combined both the abrasion and perspiration conditions. For each of the fabrics, we used a sample size of at least  $n=3$  per test.

For the abrasion tests, we used an orbital sander (Skil Model 7492-02, Robert Bosch, Inc.) together with a custom apparatus designed to apply constant pressure and mimic the frictional forces between the fabrics and skin. The maximum pressure applied to the samples was 10 psi; this value was based on previous work that examined the pressure on a limb from the prosthesis socket during ambulation [3, 4]. We used 2000 grit wet/dry sandpaper with a particle size of 12.6  $\mu\text{m}$ . This sandpaper was chosen to replicate the highest dynamic coefficients of friction for skin, ranging between 0.2-0.7 [5, 6]. We estimated that one rotational cycle of the orbital sander, which operated at approximately 60 Hz, represented the shear friction and displacement between a socket/liner interface and the skin during one step travelled with a lower limb prosthesis [7].

Table 1. Conductive Fabrics

Fabric Number	Description (Product Number)	Manufacturer
1	T304 Stainless steel mesh – 230 wires per inch (230X230S0014W48T)	TWP, Inc (Berkeley, CA)
2	T304 Stainless steel mesh – 325 wires per inch (325X325TL0014W48T)	TWP, Inc (Berkeley, CA)
3	Medical grade silver plated 76% Nylon, 24% elastic fiber fabric (A321)	Less EMF, Inc (Latham, NY)
4	Medical grade silver plated 76% Nylon, 24% elastic fiber fabric with acrylic coating (A321-ac)	Less EMF, Inc (Latham, NY)
5	Silver plated 78% Ag/Nylon, 22% elastomer (1150902130)	VTT/Shieldex Trading USA (Palmyra, NY)
6	Medical grade silver plated 92% Nylon, 8% Dorlastan fiber (E251)	Less EMF, Inc (Latham, NY)
7	Medical grade silver plated 92% Nylon, 8% fabric (23967-01)	Inventables, Inc (Chicago, IL)
8	Silver plated “spandex” fabric – exact composition unknown (N/A)	Nobel Biomaterials, Inc (Scranton, PA)

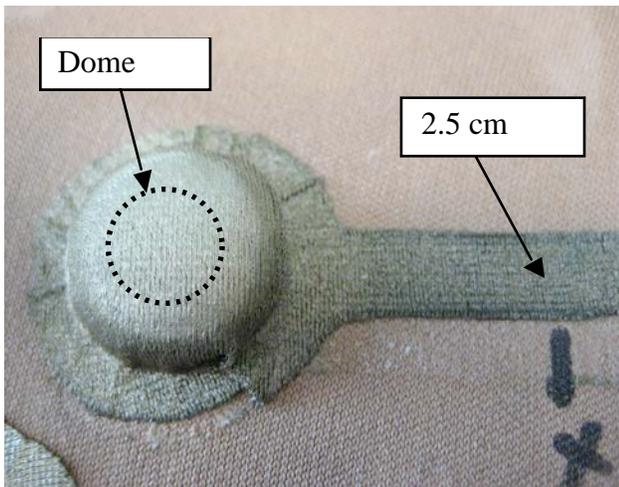


Figure 1: Example experimental setup showing a silver-coated fabric with silicone dome placed underneath. Electrical resistance was measured between the dome area and a point on the fabric 2.5 cm away.

For the dry abrasion tests, we placed the samples under the sander, pausing every 30 minutes to take three resistance measurements with a multi-meter (Model #110 True-rms digital multimeter, Fluke); the average resistance was then calculated. This procedure was repeated until no conductivity was detected, i.e. an open-circuit condition existed, which we recorded as 1000 ohms, an arbitrary and relatively large value compared to any previous measurements. For the corrosion tests we applied 1 mL of artificial eccrine perspiration (Part #1700-0020, Pickering Laboratories) to the sample, thoroughly soaking the entire fabric. Resistance measurements were taken approximately every 2-3 hours, making sure beforehand that the fabric was completely dry. Three resistance measurements were recorded each time and averaged, and this process was repeated until the open-circuit condition was reached. For the wet abrasion tests, we placed samples under the sander for 30-minute intervals and applied 1 mL of artificial perspiration at each interval, as the samples dried more quickly during the cyclic abrasion. Three resistance measurements were taken every 15 minutes during this test; this increased sampling rate was necessary as the resistance increased faster under this condition. Once again, measurements were recorded until an open-circuit condition was reached, and average values were calculated.

## RESULTS

Among the three testing conditions, wet abrasion resulted in the quickest conductance failure (i.e. open-circuit condition) time. For the dry abrasion experiment, none of the fabric samples reached an open-circuit condition, even when exposed to over 500 minutes of abrasion. In fact, only small

increases in resistance over initial values for all eight fabric types were observed, ranging from a fractional difference to a 2x increase.

For the corrosion experiment, we observed different trends in resistance increases between the stainless steel meshes and the silver-coated fabrics (see **Figure 2**). The stainless steel meshes had better corrosion resistance than the silver-coated fabrics, as they did not reach the open-circuit condition after twelve testing cycles. However, resistance measurements from the stainless steel meshes fluctuated significantly, from 20 ohms to 100 ohms, depending on the location and pressure applied by the multi-meter leads, perhaps due to the in-plane rigidity of the meshes. As a result of this observation, we attempted to keep these two variables as consistent as possible among measurements. We observed failure for all six silver-coated fabrics within 12 repeated applications of perspiration. This result was consistent either with or without the silicone domes underneath the fabric, thereby eliminating the possibility that the silicone affected these results. Each silver-coated fabric displayed an exponential time-evolution trend with respect to the electrical resistance increase (as depicted by a sample result shown in **Figure 2**), which resulted in eventual open-circuit conditions.

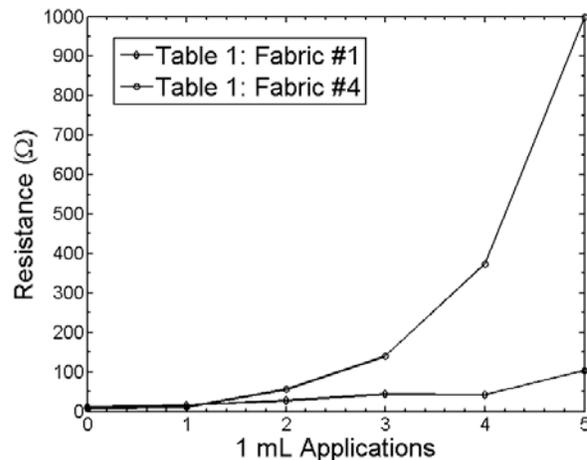


Figure 2. Representative time-evolution trends for a steel mesh (#1) and silver-coated fabric (#4) during the corrosion test. Values are averages of three measurements.

For the wet abrasion experiment, the time-evolution trends for all eight fabric types were similar to the exponential trend observed for Fabric #4 in **Figure 2**. To compare the different fabrics, we compiled the data into box-plots representing the approximate oscillation number when the open-circuit condition was observed (see **Figure 3**). As can be seen, the stainless steel meshes performed better than all but one type of silver-coated fabric. Upon inspection,

shearing could be observed in which the mesh was torn apart, creating a gap that precluded electrical conductivity.

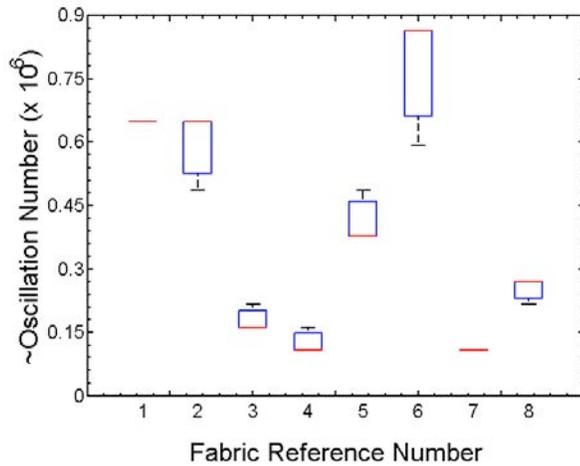


Figure 3. Approximate oscillation number at which open-circuit condition was reached for each fabric during wet abrasion

Among the silver-coated fabrics, a build-up of residue from the perspiration, perhaps from the sulphur content, was observed in the form of a ring around the dome head, which finally formed a complete insulating barrier and resulted in an open-circuit condition. Conductivity was maintained from the outside this residue ring, but not from within the ring, leading to the possibility that tarnishing was the failure mode. No correlation was observed between performance and Nylon content, suggesting that the silver-coating process was more influential in determining corrosion resistivity.

## DISCUSSION

For all three testing conditions of these fabrics, the results indicated an expected increase in electrical resistivity over time with usage in prosthetic liners. The stainless steel meshes were more robust in maintaining electrical resistance values than most of the silver-coated materials. However it should be noted that the initial resistance of the stainless steel fabrics was higher, and the fabrics' in-plane rigidity seriously affected the ability to obtain consistent measurements. The conductance of these meshes failed primarily due to shear effects under abrasion, whereas for the silver-coated fabrics, failure occurred primarily due to the corrosive effects of perspiration. Our conclusion is that none of the tested materials are suitable as long-term EMG electrode signal recording. However, these fabrics may be suitable as flexible conductive leads, given proper isolation from direct abrasion with the skin and corrosive effects of perspiration.

## ACKNOWLEDGEMENTS

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## INTERMANUAL TRANSFER IN UPPER LIMB MYOELECTRIC PROSTHETIC TRAINING USING A PROSTHETIC SIMULATOR

Sietske Romkema  
*University Medical Center Groningen*

### ABSTRACT

#### Introduction:

The aim of this study is to determine differences in intermanual transfer effects after performing different training tasks, executed with a myoelectric prosthetic simulator that can be attached to a sound arm. Prosthetic training should start within the first month after amputation for the best results. To start training immediately intermanual transfer can be used (Romkema, et al. 2013). Intermanual transfer implies that motor skills learned at one side of the body, transfer to the other side. This suggests that by practicing the unaffected arm, in the period between amputation and prosthetic fitting, the affected arm will also improve. To determine which tasks have the highest effect of intermanual transfer, different training tasks will be examined.

#### Methods:

Able-bodied right-handed participants (N=60) were randomly assigned to four experimental or one control group. The experimental groups performed a five-day training program with a simulator. Each group had a training program with different training tasks (reaching, grasping, force control, functional tasks) while the control group followed a sham training. To determine the effect of the training tasks on improvement in skill a pretest, posttest, retention test design was followed. All groups were tested on reaching, grasping, force control and functional test tasks. The training program was performed with one arm; tests were performed with the other arm. Movement times, kinematics of the grasp pattern and reach, and force control were measured.

## OUTCOME MEASURE DEVELOPMENT FOR PERSONS UTILIZING UPPER LIMB PROSTHESES

Tiffany A. Ryan  
*Advanced Arm Dynamics*

### ABSTRACT

#### Introduction:

The need for outcome measures to assess the use of upper limb (UL) prostheses is well documented. There are many useful measures but no one performance assessment meets the needs of all UL prosthesis users. Ongoing patient care interactions are the impetus to develop a relevant, clinically based outcome measure to inform individual and cohort UL prosthetic rehabilitation services.

#### Methods:

A literature review from compendiums of outcome measures for UL injury and prosthesis users was completed. These findings were categorized for intended measurement, applicability to the patient population and sensitivity to detect clinically relevant changes in performance.

Common functional grasp patterns with the hand as a primary mover or assist were matched to activities frequently assessed in performance outcome measures. Use of a prosthesis in all functional planes of movement was a key consideration in task choice.

Consequentially, 10 functional tasks were selected and trialed in Advanced Arm Dynamics (AAD) outpatient UL prosthetic rehabilitation centers with amputees utilizing various terminal devices. The tasks were evaluated for redundancy, floor and ceiling effect, and representation of functional grasp patterns.

Administrator/participant instructions and scoring criteria were developed and trialed. The pilot trial was administered by an AAD therapist, with a transradial amputee utilizing an Otto Bock Sensor Hand Speed and a transhumeral amputee utilizing a hybrid prosthesis with a RSLSteeper Bebionic3 hand. A group of three AAD and three non-AAD occupational therapists scored the subjects' performance.

#### Results:

Patient and therapist feedback is promising. Both the experienced transradial amputee and inexperienced transhumeral amputee report: testing burden as low, tasks

are relevant to their lives and illuminate areas for improved prosthesis use. AAD and non-AAD therapists identified minor improvements for the scoring criteria and instructions. Steps to analyze inter rater reliability are in process.

#### Conclusions:

Outcome measure development is ongoing. It is our intention to develop a quantitative measure for UL prosthetic clinicians to be utilized in the clinic setting along with qualitative measures to concisely ascertain the efficacy of comprehensive prosthetic rehabilitation interventions.

## A ROBOT HAND TESTBED FOR ENHANCING EMBODIMENT AND FUNCTIONAL NEUROREHABILITATION OF BODY SCHEME IN UPPER LIMB AMPUTEES

Veronica J. Santos, Randall B. Hellman, Eric Chang, Justin Tanner, Stephen I. Helms Tillery  
*Arizona State University*

### ABSTRACT

The learning and relearning of coordinated sensorimotor actions require a body scheme founded upon strong action-perception relationships. Mismatches between commanded actions and sensory feedback can lead to lack of acceptance of a prosthetic limb as part of one's own body. Furthermore, upper limb amputees who experience phantom limb pain often refer to their missing limbs as feeling "paralyzed" because their efferent commands no longer generate expected afferent signals, such as proprioceptive feedback associated with changes in limb configuration. The body scheme can be adapted by providing amputees with sensory feedback that is spatiotemporally consistent with their actions. To this end, we have developed an anthropomorphic robot hand testbed that can be used to investigate the artificial sensory feedback associated with and generated by physical interactions between artificial fingers and objects. A remote actuation system was developed for the modular control of tendon-driven artificial hands, such as the "BairClaw" introduced here. Motor-actuated tendons are routed over cantilever-based load cells for direct measurement of tendon tensions. The index finger of the BairClaw has four degrees-of-freedom (three flexion-extension, one adduction-abduction) and features Hall effect sensors at each joint for direct measurement of joint angles. The BairClaw was designed around a biomimetic multimodal tactile sensor that can measure temperature, vibration, and skin deformation associated with finger-object interactions. The testbed has been used to replay human and nonhuman primate fingertip motions and forces in three dimensions. The testbed could be used to drive invasive or non-invasive neural interfaces with rich tactile and proprioceptive data in order to enable amputees to develop consistent action-perception relationships in real-time. Such sensory feedback may also enhance an amputee's sense of embodiment and reduce phantom limb pain. Moreover, the testbed could be used for neurorehabilitation interventions similar to mirror box training. By providing subjects with the ability to control and visualize physical finger-object interactions, it may be possible to extend current motor imagery therapy approaches that focus on the mirroring of postural changes alone. Finally, the anthropomorphic robot hand testbed could be used for human-in-the-loop haptic exploration,

sensory event-driven grasp and manipulation, and simulation of proprioceptive and tactile impairments to test theories of sensorimotor integration.

## A MULTI-DIGIT TACTOR ARRAY SYSTEM FOR PHYSIOLOGICALLY RELEVANT SENSORY FEEDBACK FOR PROSTHETIC LIMBS

Julio Santos-Munné<sup>1</sup>, Paul D. Marasco<sup>2</sup>

<sup>1</sup>*HDT Robotics, Inc.*; <sup>2</sup>*Cleveland Clinic*

### ABSTRACT

Artificial limbs are difficult for an amputee to control efficiently because they do not provide physiologically relevant sensory feedback. Amputees must instead rely on vision, not touch, to manipulate grasped objects. Lack of appropriate feedback is cited as a reason why motorized (myoelectric) devices are often rejected. Cable-driven, body-powered prosthetics are often preferred because the actuator cables provide some sensory feedback through harness pressure and cable excursion.

We are developing a simple and robust sensory feedback system (Tactor Array) for retrofitting to a Targeted Reinnervation amputee's everyday prosthetic limb. A tactor is a small robot that transfers touch information, picked up at the terminal device, to reinnervated sensory receptors that are connected to the afferent pathways once serving the missing limb. Touching the sensors on the prosthesis activates the tactors which then give the amputee the distinct, physiologically relevant, perception that their missing hand is being touched. This array system will provide contact and proportional pressure feedback from the three active digits (D1-3) of a myoelectric terminal device. The two remaining passive fingertips (D4&5) and the palm will provide contact feedback.

The tactor array system is designed to retrofit to existing limb systems. All power and processing is self-contained and located on-board the prosthesis. The devices are packaged and implemented for robust use as the amputee's primary prosthesis in the home. The tactor array system at 35 x 21 x 13 mm is 50% smaller than previous devices from the Revolutionizing Prosthetics Initiative (RPI) 2007 and RPI 2009 efforts. To facilitate fitting for everyday home-use this tactor package size is only slightly larger than an off-the-shelf Ottobock™ Myo-Electrode. The tactor array components are simplified with 1 Degree of Freedom (proportional pressure, contact, and texture). The devices are daisy chainable, with 4-conductor ribbon cable, and have an embedded Controller Area Network (CAN) Motor Controller for device communication and control. Each tactor unit has a 12 mm range of movement. The devices can generate up to 5N peak

force with an acceleration of 4Gs and with velocity up to 0.4 m/s. The total mass of each tactor is  $\approx 30$  g and the domed tactor contact head is 8 mm in diameter. The tactors in the array are configured for blister-forming into prosthetic socket mounts.

Evaluations after 1 yr. home-use will examine the spatial resolution of the reinnervated skin, limb ownership, and reliance on vision.

## MYOELECTRIC PARTIAL HAND PROSTHESES – REQUIREMENTS FOR A SOPHISTICATED PROSTHETIC TREATMENT

Michael Schaefer, CPO; F.Muders,CPO; S.Kunz,CPO  
*POHLIG GmbH, Traunstein-Heidelberg, Germany*

### ABSTRACT

The treatment with prosthetics after partial hand losses remains a somewhat overlooked area within the field of technical orthopedics. Through the end of the first decade of the present century, the options for a functionally adequate prosthetic response to this type of amputation or malformation remained very limited. While body-powered prosthetic solutions did in some cases offer good functional results [2,3,4,5], albeit with aesthetic shortcomings, highly aesthetic partial hand prostheses, such as silicon partial hand prostheses, could meet the aesthetic needs of their users, but with purely passive functionality [1,15].

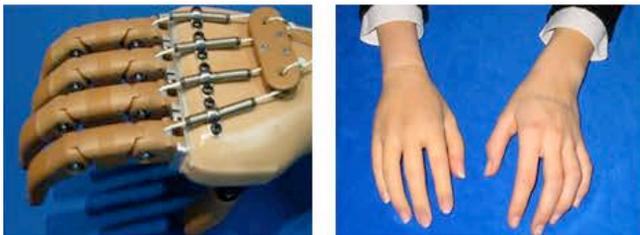


Fig.1: Body-powered and passive partial hand prosthesis

The highly varied levels of the partial hands demand a highly individualized approach tailored to the specific situation. While individual functioning finger rays or residual functioning of the carpus are often present, which must be strengthened and supported, the challenges of technical realization of the prosthetic structure and the socket are significant. For approximately the past seven years, innovative prosthetics manufacturers have been offering myoelectric finger components intended to increase the functionality of the amputated limbs. First successful fittings were publicized worldwide. Nonetheless, in the authors' view, certain questions and performances need to be discussed:

Which amputation levels are suited to the fitting of a myoelectric partial hand prosthesis, and which are not?

Are overlengths of the type often seen in practice acceptable?

What demands should be made of a modern and, most importantly, functional design of such prosthetic constructions?

How can sensory responses in remaining portions of the hand be taken into account in a modern socket design?

Which cosmetic approaches can be taken into account with myoelectric partial hand prostheses, and, in particular, realized over the long term?

These concrete questions must be addressed and answered in the course of diagnosis and ascertainment of a patient's medical history in designing modern myoelectric partial hand prostheses. The following article represents an attempt to examine the demands and exigencies of this complex method of treatment and depict potential approaches explored over 7 years' experience in the fitting of myoelectric fingers in partial hand prostheses.

### INTRODUCTION

Experience has shown that the treatment situation for partial hand prosthetics is concentrated in most countries on a small number of dedicated centers and certified Prosthetists who have deeply specialized in this field. This is not least a consequence of the highly varied array of amputation types and anatomical malformations which must be taken into account as carefully as possible in fitting a prosthesis.



Fig. 2: Longitudinal , Transversal, and Central Defects

In no other field of treatment of the upper extremity does the importance of taking into account and incorporating residual functioning of an amputated or malformed limb in a modern prosthesis design. With this in mind, it makes sense

to classify the various hand situations into levels (Fig. 2) and evaluate them according to the possibilities they present and the demands placed upon them by their users.

Over the past 25 years, certified prosthetists around the world have grappled with the possibilities presented by the integration of myoelectric technology into the treatment palette of partial hand prostheses [7,8,11,12]. Standard prostheses were broken down into their functional units and employed with modifications, or new components were developed to better address the need for minimization of structural height while retaining the maximum degree of grip strength. Unfortunately, this was accomplished with significant concessions in terms of retaining the physiological geometry and form of the hand, as well as in the overall dimensions of the prosthesis. These efforts should thus be recognized as important research contributions to the state of knowledge in this new field of treatment, but it should be emphasized that these devices only very rarely met both the functional and aesthetic requirements for a prosthetic treatment suitable for daily life.

Only with the development of independent prosthetic finger components did an adequate implementation of myoelectric prosthetic systems in the field of partial hand prosthetics become possible [10,12,13,14,16]. Accordingly, in 2007 and 2008, first experiences with prosthetic treatment could be gathered with these systems, initially in the context of clinical evaluation.

## METHODS

At the start of the treatment, it must first be determined whether suitable stump conditions are present. Particularly in the case of partial hand amputations in which individual fingers have survived, fully taking into account the geometry of the hand is absolutely vital. Excessively long partial hand stumps resulting in changed length relationships in the remaining hand are unacceptable (Fig. 3).



Fig. 3: Partial hand situations for which treatment with myoelectric finger units is not appropriate due to inadequate installation space or consequent excessive length.

Rather, in the functional structure of the hand, both the grip radii and the actual length relationships of the finger/

hand must be correct in order to ensure the opposition and interaction of the fingers. Excessively long finger structures or a deformation of the architecture of the hand, even to a small degree, are not acceptable (Fig. 4). If needed, finger components may be shortened, or shorter finger units used; however, the size of the hand must take the actual circumstances into account.



Fig. 4: Geometry of the hand (right-hand image); disproportionate length relationships (left-hand image) are unacceptable.

To address these requirements, the decision was taken at an early stage in the authors' workshops to employ the Vincent finger system marketed by the eponymous German company, Vincent Systems. In comparison with its competitor product on the market, its decisive advantages were its small structural height and the more robust design of its components, which are exposed to significant stresses. The prosthetic design generally incorporates three to four functional units (Fig. 5):

1. The HTV silicone inner socket, which – depending on the clinical conditions – may be realized in a closed, full-contact form, or as a framework construction.
2. A prepreg carbon fiber frame which, rigidly attached to the silicone inner socket, forms the load-bearing and protective component to which the finger units and the electronic components are attached. This frame socket is also required to transfer stresses that arise, and should thus be firmly attached to the HTV silicone socket using a sandwich construction method.
3. An aesthetic and water-resistant partial hand glove, which – depending on the requirements of a specific case – can have various finishes, including individual aesthetic skin tone, standard skin tone, or functional. As part of the individual partial hand treatments, an individual master model must be made for each glove treatment in order to ensure the optimal functioning of the finger components.
4. A functional forearm cuff. This is necessary in the case of long partial hand stumps and the consequent lack of space to receive the electronic components

such as the controller and battery of the finger system. The functional forearm cuff may be made of various basic materials, depending on the use characteristics desired, including neoprene (see Fig. 3, middle), silicone, or carbon fiber. Care must be taken to ensure that the routing of the electronic cables across the wrist joint be done using a hose element that will prevent elongation. In addition, the electronic components on the bandage must be protected from impact.



Fig. 5: Partial hand components: HTV silicone socket (left) with carbon-prepreg frame, reinforced neoprene bandage

## RESULTS

Between 2007 and 2014, both socket technology and the design of the components of the system could be greatly optimized. While the Vincent finger system has now entered its 6th generation, some different socket designs and combinations of materials have been tested too. Since 2012, the clinical trials phase has been completed, and definitive prosthesis treatments have been made available to users. One particular challenge has since been mastered, namely that of integrating all technical components into the hand structure for amputees with short partial hand stumps, permitting the accompanying forearm cuff to be dispensed with (Fig. 6). For their users, these types of treatment represent a significant functional improvement to the quality of daily life.



Fig. 6: Test version of a myoelectric partial hand prosthesis with integrated controller and battery

It became clear early in the course of this project that such complex prosthetic treatments require intensive therapeutic care during the initial phase of treatment. Thus, standard practice in the authors' firm is for an individually tailored therapeutic plan to be drawn up early in the process, at the time of the production of the first test prosthesis. This

plan takes into account manual capabilities, training of proportional control techniques, and various ADL activities.

As part of the German Bionic Hand research project, work is currently underway on an interactive, computer-supported training program which the user can use in parallel with the control training units in order to perfect the use of the myoelectric partial hand.

## ACKNOWLEDGEMENTS

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### **AUTHOR**

*Michael Schaefer, CPO, CEO  
POHLIG GmbH  
Grabenstaetter Str. 1  
83278 Traunstein  
GERMANY  
Mail: [m.schaefer@pohlig.net](mailto:m.schaefer@pohlig.net)*

## CHARACTERIZING FACTORS AFFECTING VIBRATION INDUCED MOVEMENT ILLUSIONS: TOWARD KINESTHETIC SENSORY FEEDBACK

Jonathon S. Schofield<sup>1</sup>, Michael Dawson<sup>2</sup>, Jason Carey<sup>1</sup>, Jacqueline Hebert<sup>1</sup>

<sup>1</sup>University of Alberta; <sup>2</sup>Glenrose Rehabilitation Hospital

### ABSTRACT

Dexterous hand movement is dependent on closed loop control comprised of efferent motor output and afferent sensory feedback. Control is significantly altered in those with upper limb amputation, as sensations of touch and movement are inherently lost. This absence of sensory feedback impedes efficient prosthetic use and is highlighted as a major factor contributing to rejection of myoelectric devices. A unique solution may lie in exploiting vibration induced movement illusions (the kinesthetic illusion); a physiological phenomenon whereby vibration introduced to musculotendinous regions of a limb can induce sensations of limb movement. Although prominent in literature, this phenomenon has yet to be explored in prosthetic users.

Applying the kinesthetic illusion in prostheses requires a fundamental understanding of how to consistently elicit and manipulate movement sensations. Specifically, literature suggests vibration amplitude, frequency, and degree of stretch in a vibrated muscle impact movement sensations. Yet little agreement exists on the importance of each or the nature of their effects. This study uses a full factorial approach to comprehensively characterize the effects of amplitude, frequency, and muscle stretch on the kinesthetic illusion.

Movement illusions were induced in 12 able-bodied participants. For each, 18 combinations of amplitude (0.1 to 0.5 mm), frequency (70 to 110 Hz), and limb position (two positions resulting in different states of muscle stretch) were introduced to the triceps and biceps. Following each combination, participants were asked a series of questions to quantify the strength of illusion (SOI), range of motion (ROM), and illusionary velocity. Using a factorial ANOVA, results indicated vibration amplitude has the most significant effect on all three output variables SOI, ROM and velocity in both the biceps and triceps.

## FIRST CYBERNETIC PARTIAL HAND PROSTHESES IN ANATOMICAL SIZE

Schulz S., Valencia R.

VINCENT Systems GmbH - Medical Technics, Karlsruhe, Germany

### SUMMARY

A partial hand amputation presents a particular challenge for the orthopaedic technician with regard to functional prosthetic restoration. Anatomy and residual function of the partial hand differ to a high degree. For the first time, a new generation of prostheses is enabling the functional replacement of single fingers and the thumb with active electrically driven prosthetic fingers corresponding in size to the anatomy of the human hand. The following article presents the externally powered partial hand system, the *VINCENTpartialhand*, which can be adapted to the most varied clinical conditions and restore an active and adaptive grasping function to the patient concerned (Fig.1).



**Figure 1:** Example of partial hand system with four active long fingers and an active thumb with a biomechanically optimized shaft structure.

### INTRODUCTION

Restorative treatment following a partial hand amputation can vary greatly and requires highly individual and variable prosthetic solutions. The spectrum ranges from missing individual finger phalanges to entire sections of the metacarpus and the thumb. The situation is potentially complicated further by stiffening of joints, scars and folds in the skin, and the associated functional impairments, as well as hypersensitivity in the remaining sections of the hand. Alongside a sophisticated shaft design, the prosthetic components must satisfy this set of requirements.

### History of powered single finger prostheses:

A project entitled “*Externally powered fingers*”, administered in the USA by the “*VA Lakeside Medical Center, Chicago*” and supported by the “*Department of Veterans Affairs Rehabilitation R&D*” and the “*National Institute on Disability and Rehabilitation Research (NIDRR)*”, was concluded as early as 1989 [1]. Within the scope of this project, R. Weir at the “*Northwestern University of Chicago, Department of Biomedical Engineering*” constructed a prosthesis in which motors and gearboxes were integrated fully into individual fingers and in the thumb for the first time [2][3]. These efforts can be seen as the forerunner of all powered partial hand systems in operation today. In practical treatment, this principle has been implemented by the Scottish company *Touch Bionics* since 2007 in their “*ProDigits*” product [4]. The limitations of these systems lie in particular in their shape and size. Since 2009, the German company Vincent Systems has provided the first anatomically and biomechanically optimised system *VINCENTpartialhand* [5] [6].

### GENERAL DESIGN

The new partial hand system design aims to enable as far as possible the anatomically and biomechanically correct restoration of the active gripping and holding function. Cosmetic aspects, weight, holding force, gripping speed and noise generation were further aspects considered during this new development.

### SINGLE FINGER PROSHESIS

At the heart of the system being presented is an electrically driven single finger prosthesis. It consists of an active prosthesis powered by an electric motor and gearbox integrated into the middle finger. The drive unit moves the proximal phalange and the middle finger phalange, which is coupled elastically to the proximal phalange via a spring, actively and bi-directionally (Fig.2). The finger can also be returned to its original position even when no voltage is applied and is protected against overload by a mechanical overload coupling.



**Figure 2:** Two axes anatomical motor driven single finger prosthesis.

Adaptation to prosthesis shaft:

The precise positioning, alignment and connection of the powered finger to the prosthesis shaft without the use of additional aids is time-consuming and complicated. A sturdy aluminium frame assumes positioning, stabilisation and electrical contact of the single finger in series or along a curve, allowing for a slight abduction of the finger at a pre-set angle (Fig.3). The frame may already contain parts of the electronic control unit and sensors. Electrical contact is by means of spring contacts set into the circuit board in the frame and which touch the gold-plated contact surfaces of the prosthetic finger.



**Figure 3:** Modular, anatomically size partial hand system: frame with slight abduction for four finger and an active thumb, left hand.

**PROSTHETIC THUMB**

The thumb is of paramount importance in grasping and holding using the cylindrical or lateral grips as well as with regard to assuming an opposition position to the fingers especially for the precision grips: the pincer grip and the three-point grip. In contrast to the fingers, the loss of the thumb cannot be compensated by the remaining fingers; a significant functional impairment is the result.

Passive thumb:

With regard to restoration in situations in which a thumb stump is insufficiently long or mobile or is lacking other adaptation options for functioning cosmetic treatment of the thumb, a mechanically operated thumb can be a solution (Fig.4).



**Figure 4:** Passive thumb with stepless proximal joint and ratchet-type distal joint, soft thumb-tip.

The thumb is equipped with two movement axes. The proximal axis allows the thumb to rotate inwards and outwards, whereby the motion resistance can be set steplessly via a tensioning screw directly on the axis. The second, distal axis, positioned at right angles to the proximal axis, features a ratchet-type mechanism (Fig.5).



**Figure 5:** Two axes scope of movement of the passive thumb, the proximal joint with motion resistance (ri.) and the distal joint with ratchet-type (le.).

This permits stepless adduction and movement in the opposite direction is prevented by the interlocking teeth. Opening of the thumb is by tensile force at the distal end of the thumb and simultaneous reverse rotational movement.

Active thumb:

In situations where treatment involves the use of powered fingers and the necessary peripherals such as control unit, sensors and battery system are implemented, it can be advantageous to also actively control the thumb, if added functionality for the patient can be achieved as a result (Fig.6).



**Figure 6:** Active thumb with powered distal joint and stepless passive proximal joint, soft thumb-tip.

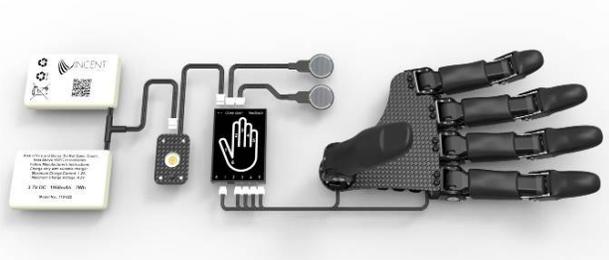
The construction of the proximal passive joint of the thumb is almost identical to that of the manual thumb. The distal powered thumb joint is constructed in a similar way to a single finger, but without a moveable fingertip. Instead, the thumb is sheathed in an elastic PU covering (Fig.7).



**Figure 7:** Two axes scope of movement of the active thumb, the manual proximal joint (ri.) and the motor driven distal joint (le.).

### CONTROL OF THE PROSTHESIS

A number of different components must be integrated in the prosthesis shaft in order to operate the prosthetic finger. The control unit has six slots for four active prosthetic fingers and one thumb with up to two active movement axes, a slot for the vibration motor of the force feedback system and two ports for various sensors.



**Figure 8:** Overview of all component of *VINCENTpartialhand* system: 4 single finger, active thumb, frame, control unit, 2 FSR sensors, magnetic charging unit, 2-cell-LiIon-accu with a capacity of 1300mAh.

The power supply is connected via an additional slot. Prismatic LiPo cells are generally used in this case, with a total voltage of 8.4V and capacities of 1300mAh up to 2000mAh. The accumulator cells are charged via a magnetic contact charger, on which the system's on-off switch is also located. The electronic control is connected via Bluetooth connection to a tablet PC in order to enter patient-specific settings. The prosthesis wearer can choose between a number of different grip types by means of variable switchover signals. A corresponding 2-finger system (Fig.8, Fig.10 and Fig.11).

### Sensors:

Control of the prosthesis is either by one or two sensors. The user may choose between EMG sensors, touchpads and bend sensors, whereby a combination of sensors can also be used [7]. If these options are not sufficient, the system can be extended considerably by the addition of further input devices via the Bluetooth module integrated in the 6-channel control unit.

### EVALUATION

The concept for the new prosthesis system emerged from treatment situations in practice. Experienced orthopaedic technicians were involved at every stage of development. Numerous patients were treated in parallel with development and this had a decisive influence on the optimisation process [5][6][8][9][10].

### Example of treatment:

Figure 9 shows an example of treatment of a young woman with a congenital malformation of the hand. Until now, conventional treatment has been based on a rigid forearm shaft made of a PMMA-glass-carbon compound in combination with silicone shaft technology (left side Fig.9).



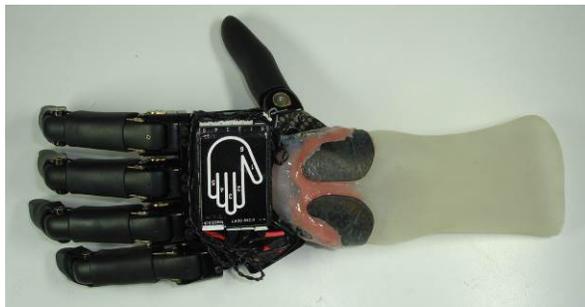
**Figure 9:** Restorative treatment, left: previous conventional prosthesis; right: example of treatment from Pohlig GmbH with *VINCENTpartialhand* components.

A conventional electric hand system is rigidly connected to the shaft without permitting rotation of the wrist. Forearm rotation was severely restricted and movement of the wrist blocked completely. Shape and size of the hand did not correspond to the natural hand of the wearer. The prosthetic hand weighed 702 g. The new prosthesis concept was aimed at fully retaining clinically recognizable, existing freedom of movement of the forearm and wrist and at the same

time realising a multi-articulating hand with a variety of supplementary functions.

Structure of the prosthesis:

The new design of the prosthesis originated in collaboration with Pohligh GmbH from Germany. The new prosthetic concept is based on minimized shaft technology. Connection of the prosthesis is via an individual HTV-silicone liner in combination with an extremely short CFC hand shaft in the region of the carpus (Fig.10).



**Figure 10:** Restorative treatment option, inside – top view at the control.

The minimal carbon-prepreg frame comes into operation solely in the region of the important guide bones. Depending on the skin situation, larger surfaces can be embedded in silicone and adaptively or even freely embedded. As a result, the prosthesis wearer obtains the best possible freedom of movement in the region of underarm, wrist and hand.



**Figure 11:** Restorative treatment option, separate frame with single finger.

The main challenge lay in integrating all system components, such as 2-cell-li-ion-1300-mAh battery, magnetic charging socket, control unit and sensors, into the hand structure, thereby eliminating the need for the associated sleeve on the forearm. *VINCENTpartialhand* components made up the cybernetic element of the whole. Four motor-driven biaxial long fingers and a motor-driven thumb with passive base joint, connected to an aluminium frame make up the core of the biomechanical-anatomical hand structure (Fig.3 and Fig.11). Control of all grip functions is maximal

distally, directly at the hand. Two FSR touchpad sensors, firmly integrated in the silicone shaft, detect the slightest movement in the metacarpus. Thanks to the biomechanically optimal connection, the new design of the prosthesis affords the wearer unlimited rotation of the forearm and a reduction of max. 10% in the mobility of the wrist. The hand, with a total weight of 361g, corresponds in size and shape to the naturally slender hand of the wearer (Fig.12). In comparison with the natural hand of the prosthesis wearer the prosthesis, weighing approx. 250g, comes close to a biomechanically natural weight ratio for the first time.



**Figure 12:** Fully integrated partial hand with 4 active driven single fingers and an active driven thumb allows a fine manipulation.

Benefits for the prosthesis wearer:

The prosthesis wearer describes the benefits of the new partial hand system as follows: The new structure of the hand brings with it great freedom of movement thanks to unrestricted forearm rotation and wrist mobility. For the first time, it is also possible to use the palm of a prosthetic hand. Alongside many other new grip types, the index finger can be used to type on a keyboard (Fig.13) and together with the thumb for grab small objects (Fig.12 and Fig.14).



**Figure 13:** The index finger can be used to type on a keyboard.

Also for the first time, the new hand allows the wearer to put on even tight-fitting clothing without any problems, and anatomy and habitus correspond to that of a small, female hand. Its low weight, in addition to the high degree of wearing comfort, has a very positive effect on posture and the muscles of the arm and neck.



**Figure 14:** The precision grip can be used to grab and hold small objects.

### ACKNOWLEDGEMENTS

Particular thanks are due to our clinical partners in Germany, Austria and the USA, especially to the orthopaedic technicians involved in development. We would like to single out in particular Michael Schäfer / Pohlig GmbH, GER and Jack Uellendahl / Hanger Clinic, US.

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### AUTHOR

*Dr. Dipl.-Ing. Stefan Schulz, BioRobotLab at KIT and CEO of Vincent Systems GmbH – Medical Technics, Karlsruhe, Germany, [www.vincentsystems.de](http://www.vincentsystems.de), [www.handprothese.de](http://www.handprothese.de), [info@vincentsystems.de](mailto:info@vincentsystems.de)*

## NEW STRATEGIES FOR MYOELECTRIC CONTROL OF MULTI-ARTICULATING HAND AND PARTIAL HAND PROSTHESES

Schulz S., Spengel S., Probst P.  
*VINCENT Systems GmbH - Medical Technics, Karlsruhe, Germany*

### SUMMARY

The movement abilities of a modern, cybernetic, multi-articulating prosthetic hand are many and varied. Conventional methods of myoelectric control of all available functions of the prosthesis push the cognitive capacity of the prosthesis wearer to the limit. In order to switch between grip types and grasping patterns, prosthetic control units currently in operation make use of transponders, buttons or remote controls. Taking the control logic of the *VINCENTevolution2* (Fig.1) as an example, this article presents a possibility of reproducing in a very short time a selection of 12 grip types, more than 20 hand positions and random intermediate positions solely with two EMG signals and without additional aids. Furthermore, with the new generation of prosthetics, the morse-code established in information technology is adapted to the control of a multi-articulating hand. It is developed further into a priority-based control concept which allows a fast, direct and flexible enhancement of grip control.



**Figure 1:** Multi-articulating prosthetic hand; six motors actively drive the ten bi-directional movement axes of the *VINCENTevolution2*.

### INTRODUCTION

Multiarticulating prosthetic hands are often capable of a wide range of movements due their actively moveable axes [1][2]. If only the end positions of the individual finger joints are considered, 64 different hand positions alone can be deduced in the case of six motors in the hand. Even if the number of meaningful, useful grips will lie far below this figure, the direct and proportional controllability of the new

prosthetic hands constitutes a huge challenge, coupled with the same degree of potential.

The objective is to develop a control strategy which is easy to learn and insensitive to interference and which makes the scope of movement of new prosthetic systems available in a practical, everyday form.

### CONTROL OF THE PROSTHESIS

The new control models are demonstrated using the example of the multi-articulating prosthetic hand

*VINCENTevolution2*. Equipped with six motors, the hand enables active movement of each individual finger and the thumb in ten bi-directionally driven movement axes. The bow springs between the proximal and distal joints also permit an adaptive grip with basic tension – in line with the natural muscles and ligaments of the human hand. Thanks to the lateral movement of the opposable thumb to the ring finger, this hand is currently the most actively mobile prosthesis on the market, which makes it particularly interesting in terms of the development of new control strategies (Fig.2).



**Figure 2:** Lateral movement of the *VINCENTevolution2* and opposable thumb to the ring finger.

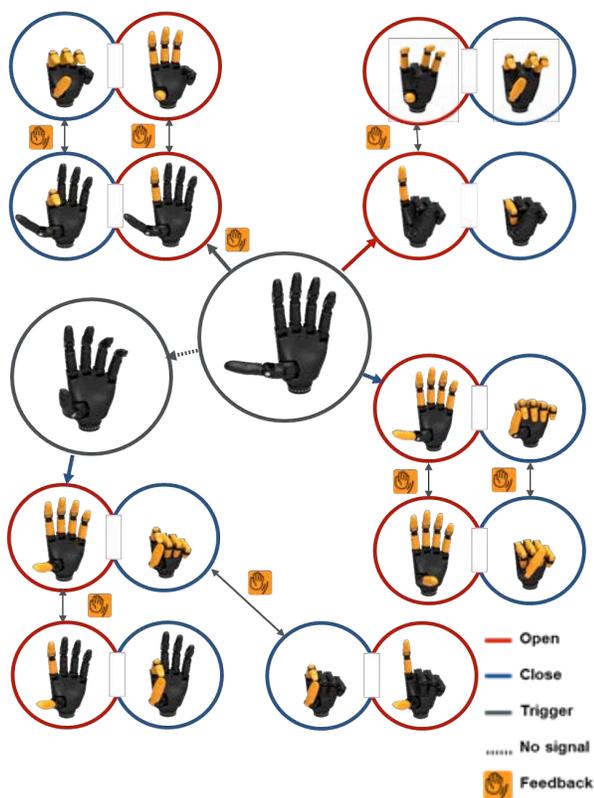
Until now, the prosthesis could be controlled by means of one to two EMG sensors via grip pattern primitives between which the prosthesis wearer switches. Grip patterns could be selected using simple control commands, such as one to three short contractions to the “open” or “close” signals, one or a number of co-contractions, or the signal edge variation. With this method, however, it is only possible to switch directly between a few grip patterns. Additional grip groups must be selected by a button, transponder or remote control.

### SINGLE TRIGGER CONTROL

The development objective was to enable direct selection of a large number of grip pattern primitives without the use of additional technical aids. This grip selection should not present any major cognitive challenge to the prosthesis wearer. By means of the fixed arrangement of all available grips in the system, physical imprinting of the sensorimotor cortex of the prosthesis wearer’s cerebral cortex should stimulate intuitive incorporation into his or her own body image.

#### Mode of operation:

The basic idea behind the control model is that the user need only memorize a single switching signal – the so-called *Trigger*. The Trigger signal in this case can be a randomly chosen signal or a signal sequence such as Peek or Double-Peek or a co-contraction. To enable selection, this single Trigger signal is combined with the natural “open”, “close” and signal pause.



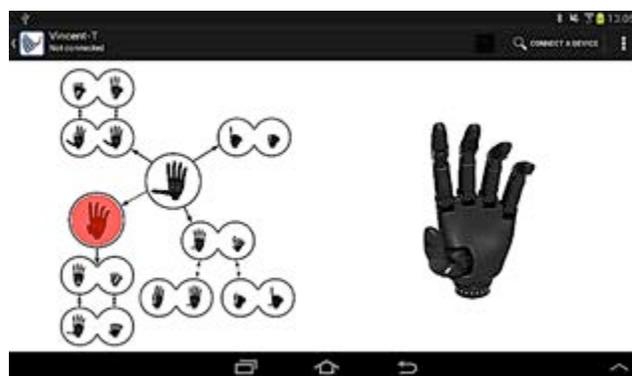
**Figure 3:** State Machine - the flat hand is the central state. Get there by entirely opening the prosthesis and holding open. The colored fingers mark the proportionally controllable joints.

The first five grip groups are directly attainable from a central hand position. The associated grip types in each case

are reached from there with the Trigger signal. By means of a long “open” signal, it is possible to jump from any grip type directly into the central hand position. If the “open” signal is held open, the control jumps into the index finger grip. If, however, a “close” signal is given, the lateral grip is chosen. If the Trigger signal is sent instead, the flat hand is activated. Where no signal is given, the hand automatically returns to the natural position in which the power grip is active. Ideally, the Trigger signal is set to the “open” electrode. In this way, the “close” signal can be used without delay time exclusively for closing the hand (Fig.3).

#### Control feedback and training:

To support the training process, vibrotactile feedback can be activated in the *VINCENTevolution2*. Here, a vibration motor signals the detection of the Trigger signal and hence the attainment of the different grip type in each case, as well as the central hand position. A special training app helps the prosthesis wearer when learning how to use the control system. The app displays the grip selected in each case virtually and in its logical position within the tree (Fig.4).



**Figure 4:** VINCENTtraining-app: the interactive training program communicates with the prosthesis control unit and displays the executed grip in each case and the control principle virtually.

### MORSE CODE CONTROL

From a technical point of view, the control model presented can be extended beyond the 12 grip patterns by further “individual grips”. This, however, would quickly reveal the limits of dynamic grip selection, in terms of the user’s cognitive capacity and the adherence to acceptable delay times between grips. Therefore, for the *VINCENTevolution2* an enhancement of the control concept has been developed which in particular features a comprehensive command set to permit efficient communication between the user and the prosthesis. The core of this development is the adaptation of the Morse alphabet, used for transmitting information in the form of letters, numbers and symbols, whose coding

format is based primarily on the solid, efficient transmission of information [4].

Technical implementation:

Several signal forms are possible for the adaptation of Morse code for prosthesis control. Commands for control of the prosthesis can be constructed by means of one to two EMG sensors, touchpads or any other sensor consisting of two control characters [3]. These differ either in the assignment to a signal source or by the varying signal lengths or signal amplitudes.

**Table 1:** Example of the assignment of prosthesis functions to a control command, consisting of one to three control characters. In practice, the *dash* sign can represent a long muscle-, the *dot*, a short muscle contraction.

No.	Order	Code	Function
1	T	–	Neutral hand position
2	E	•	Rotate thumb
3	M	--	Index finger
4	N	--•	Hook grip
5	A	•–	Lateral grip
6	I	••	Three-point grip
7	O	---	Pincer grip
8	G	---•	Close thumb
9	K	--•–	Hold writing instrument
10	D	--••	Hold fork
11	W	•--	Hold knife
12	R	•–•	Hold spoon
13	U	••–	Hold toothbrush
14	S	•••	Reset to neutral position

A signal variant or a combination of different variants can be selected, depending on the motoric and cognitive abilities of the prosthesis wearer. A control command consists in this case of one or more control characters. In Morse code, two control commands with one control character, four control commands with two characters, eight control commands with three control characters, etc. are available. In prosthesis control, frequently used grip patterns, such as cylindrical grip, pincer grip or thumb movement, are selected using a short control command consisting of two control characters. Other, less frequently used grip patterns, such as the index point, are selected with a somewhat longer control command consisting of three control characters. Seldom used, but nevertheless functional, grip patterns, which enable for example individual secure holding or clamping of cutlery, writing instruments or tools, can be assigned control commands of four or more characters, where the shorter control commands have already been assigned functions. Table 1 shows an example of such an assignment. This priority-dependent gradation of the complexity of control command structure grants the prosthesis wearer highly efficient control of the prosthesis

and, similarly, offers the user the possibility of continually extending the range of functions, as training progresses.

**FORCE FEEDBACK CONTROL**

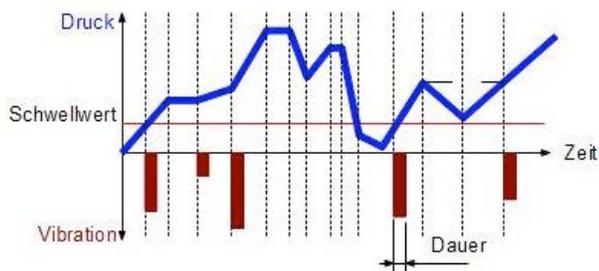
To be able to grip an object securely and to assess its characteristics, we use not only our eyes, but primarily our sense of touch via the hands. Prosthesis wearers lack this sensory information provided by the fingertips. In addition, the sense of touch in the arm or hand stump is also considerably limited due to the prosthesis shaft. The day-to-day realization of a sense of touch in a prosthesis presents the specific problem of relaying this sensory information from the prosthesis back to the wearer. Irrespective of the type of feedback mechanism, both the receptors in the skin and our brain adjust to this “alien stimulus” and react over time by suppressing this “malfunction”. Feedback can no longer be perceived in a differentiated manner.



**Figure 5:** Thanks to a force feedback system, objects can be gripped precisely even without direct eye contact: hand in *Vincent-handlegrip*.

For the first time, a force feedback system has been developed for inclusion as standard in the *VINCENTevolution2* prosthetic system which was suitable for everyday use and considerably mitigated this habituation effect (Fig.5). The habituation effect is particularly strong when the receptors in the skin are exposed to constant, unchanging stimuli. The newly developed feedback system is based on the fact that only differentiated information relating to changes in force or touch is relayed to the prosthesis wearer. This can happen in a number of ways, which we call “modes”. In “Mode 1”, for example, information about a steadily increasing force at the tip of the prosthetic finger can be transmitted to the wearer as follows: the information on the force is divided into a number of clearly differentiated stages. In the same way, information is classified into categories such as “Initial contact”, “Increasing force” and “Decreasing force”. As each stage is reached, a feedback signal is sent to the wearer in the form of single pulses – the number of pulses depends on the level of the force stage reached. Only when there is a change from one stage to another is a signal given, indicating

the type and level of the change. No feedback is given at points in between. In “Mode 2”, for example, information on the current stage level can this time be generated by means of the modulation of frequency of the signaling device, instead of via the number of pulses. “Mode 3” differentiates by amplitude, etc. (Fig.6). The signaling device for the force feedback system can in this case be of various types: vibration, electro stimulation, light, sound, etc. A vibration actuator is used in the *VINCENTevolution2*.



**Figure 6:** Example for an vibrotaktile feedback reaction (red line) with modulation of amplitude of a force signal (blue line).

## EVALUATION

Clinical practice has shown that prosthesis wearers were able using the Single Trigger Control system to quickly and easily achieve all grip types offered in the *VINCENTevolution2* State Machine after just a few minutes of training (Fig.7). It is perceived to be highly intuitive and easy to learn. Enhancement of the system by means of the Morse Code Control system is currently being trialed on a group of test users to establish its suitability for everyday use. It is not yet available as a standard feature of the prosthesis. The adaptation of Morse code is of particular importance for the control of multi-articulating prosthetic hands. Coding which is efficient, resistant to interference and easy to learn opens up new possibilities for direct selection and proportional control of a host of different grip types. As training progresses, the control concept offers the user a scope of functions which can be continually extended and customized. The force feedback system is already included as standard in the *VINCENTevolution2*. Levels of acceptance among prosthesis wearers are high. The system supports the training process and enhances the operational features of the prosthesis. Clinical studies on the effect of the system on the sensorimotor cortex are currently being performed as part of a research project.



**Figure 7:** Training with the hand prosthesis, using the *Vincent-app*: the interactive training program communicates with the prosthesis control unit: training of tablet-, *Vincent-cupholder*-, zylinder- and indexgrip at work.

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## AUTHOR

*Dr. Dipl.-Ing. Stefan Schulz, BioRobotLab at KIT and CEO of Vincent Systems GmbH – Medical Technics, Karlsruhe, Germany, [www.vincentsystems.de](http://www.vincentsystems.de), [www.handprothese.de](http://www.handprothese.de), [info@vincentsystems.de](mailto:info@vincentsystems.de)*

## INTEGRATED SYSTEM OF NEW ELECTRIC TD AND WRIST COMPONENTS - STAGE 1 DEVELOPMENTS

Harold H. Sears PhD; Edwin Iversen MSME; Jeffery Christenson MSBME; Tony Jacobs, BUS.  
*Motion Control, Inc., Salt Lake City, UT*

### INTRODUCTION

Developments of prosthetic components should, ideally, be driven by the needs of the consumer population. However, the prosthetic marketplace usually drives development in small increments, given the small size of the UE prosthetic population.

Government, or other agencies, can be effective in driving more large-scale projects, where more resources can be focused on a longer-term goal, *if* the goals are truly based upon consumer need.

In 2011, a major government research agency, Congressionally Directed Medical Research Programs (CDMRP), announced a project to produce next-generation devices, based upon the expressed needs of military personnel with UE limb loss, and the particularly high needs of those with bilateral loss.

The CDMRP specifications included the directives to:

1. Evolve a new ultra-rugged TD, as well as three degree-of-freedom (3-DoF) powered wrist modules. The components are to be used individually, and in useful combinations.
2. Designs should leverage the benefits of body-powered hooks, and make no concessions to hand-like aesthetics.
3. The 3-DoF wrist will include powered flexion, for 1 and 2 DOF, and powered pro/supination.
4. Added to the approved project was a new Universal Quick Disconnect (UnivQ/D), and a new passive wrist flexion module.

### ELEMENTS OF THE PROJECT (TO DATE)

#### 1. *Universal Q/D – derived from Open-Source project.*

To achieve and improve the interchangeability of modular components, an improved quick disconnect (Q/D) was proposed. Even prior to the CDMRP project an ‘open source’ project was formulated by other researchers [1] to

advance a new standard, which could be adopted by any willing manufacturer. Within our project, the goals became:

- A quick disconnect component that is ‘universal’, i.e., usable by any TD and wrist component on the market, with shorter axial length, maximum water resistance, higher overload limits (in all directions), and smaller size.
- Lock/unlock capability in pro/sup direction, which is easily activated by users.
- Electrical connections at least matching the current standard.
- Easy transition to Bus-type communication.

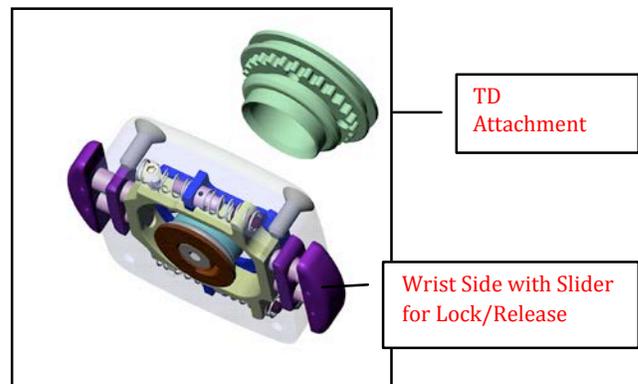


Figure 1 – Universal Q/D drawing showing detail of Slider which locks & unlocks, as well as releases TD.

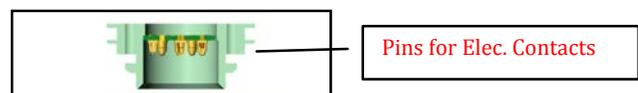


Figure 2 – Detail of TD Attachment, showing spring-loaded pins used for electrical contacts.

The design of the UnivQ/D, is available to any manufacturer who chooses to join the group, and contributions have been made by many participants, including Motion Control, RIC, UNB, LTI, TAD, Steeper, Touch Bionics.



Figure 3 - Universal Q/D installed onto ETD.

As the design has evolved, the specific features of the Prototype UnivQ/D, have been accomplished:

The breakaway force in an axial direction is over 35 kg (70 lb.) which is at least 25% higher than the current standard. Transverse breakaway force is similarly increased.

The sideway push button will have two stable positions, 1, unlocked and free to rotate, and 2, locked to rotation. If the push button is held down (against a spring force) the TD may be removed easily, even if a 2<sup>nd</sup> hand is not available to pull it.

The length of the assembly satisfies the initial goal of ~12 mm (1/2 in.) of length, reducing the required Q/D length by about 50%. The minimum diameter of the mechanism is 47 mm, or ~1.85 in.

Water resistance has been accomplished by the positioning of O-ring seals, so that water will not reach either the electrical contacts nor mechanism of the Q/D.

Another benefit will be the relative economy of the small, technically simple distal-side element, which will be manufactured with each TD used with the system. Even body-powered devices may be interchanged with the electric hands, or hooks, or sport devices which will become more practical with a water-resistant attachment.

The current standard for Q/D of electric TDs, (originally an Otto Bock innovation) has a simpler side as well, but it is on the proximal, or wrist side. Thus, at present the more complex (and expensive) side must be part of every TD used with the system.

## 2. New Passive Self-Returning (SR) Wrist Flexion

A passive flexion joint is an opportunity to add DoF to the prosthesis with minimal space and weight. Current versions are popular but not water-resistant, and sometimes cumbersome to lock/unlock.

A new design approach utilizes a unique passive one-way lock, allowing the wearer to position the wrist by simply pushing in the flexion direction, while the lock remains smoothly engaged, preventing motion in extension. The range of motion allows up to 85° in flexion and up to 45° extension. To unlock, an accessible push button is prominent enough to push easily with the other hand or prosthesis, or against any surface. Once unlocked, the wrist is spring-loaded to return to neutral, becoming a very compliant joint. As has been learned with earlier wrists, a compliant wrist improves comfort due to its shock absorption, and also allows grip security while the TD is moving, as in driving and steering a cart or bicycle.

As with other components proposed in the CDMRP project, the SR Flexion Wrist is designed to be water-resistant, using a tight enclosure, while keeping the weight and size comparable with existing flexion wrist components.



Figure 4 – the Self Returning (SR) Flexion Wrist component allows up to 85° of flexion. Once locked, the wrist prevents extension, and needs only to be pushed to the desired flexion position. Note the wrist enclosure which prevents water and/or dirt exposure.

## 3. New Powered Wrist Flexion

The CDMRP specifications place a high priority on development of a powered flexion drive. The motion of wrist flexion has rarely been available in a powered component. Although passive devices are now widely distributed (e.g., Motion Control Flexion Wrist, and Multi-Flex Wrist are used with MC TDs, as well as other brands of TDs). A powered flexion device is needed for high-level prostheses, and coordinated motions such as combined flexion and rotation, which will be available with a controller operating both powered motions. With a 2-DoF flexion device, combined with powered pro/supination, a high level of positionability will be available in an UE prosthesis for high level, and bilateral prostheses. Eating, dressing, self-care, and other

everyday activities will be made more possible when the prosthesis allows the TD to move into every possible orientation, and through a wider work envelope.

The 1-DOF wrist flexion is one of the highest priorities of the entire project, and will be utilized with the ETD (and other TDs) first, then later in the project applied with a 2<sup>nd</sup> flexion DoF and/or wrist rotation, and the advanced ETD2.

The size, weight, and torque output of the Powered Flexion has been shown to be comparable with existing powered wrist devices, i.e., it will be no longer than 5 cm., weigh no more than 140 gm, and move any TD a range of at least 120 deg, within 1.0 sec., and at lower speeds apply torques of approximately 24 in.lb ( 2.7 Nm).

At present, it is feasible to install powered flexion either proximal to the UnivQ/D, or distal, although the practicalities of the prosthetic market might require that the Q/D be distal, so that various TDs might be interchanged, and require only a single flexion device.

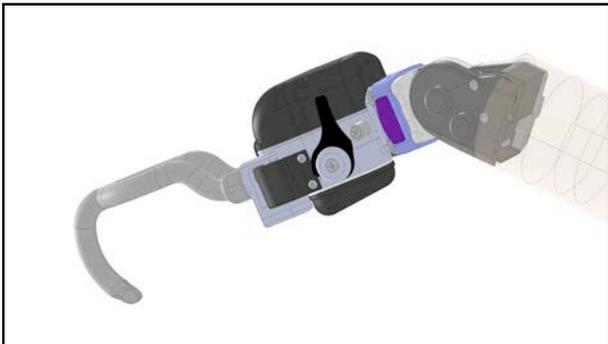


Figure 5- CAD drawing of the powered 1-DoF Flexion Wrist component, with the ETD shown attached via the UnivQ/D distal to the flexion component. Note that all components in this combination will be water resistant, including potentially the entire forearm.

#### 4. Future Plans

The components presented in this Progress Report are all expected to be in small-scale field trials (under IRB oversight) by the time of the MEC conference. It is the intent of the CDMRP-funded project that commercial products be produced, which is the most practical way for research and development to reach the consumer.

These three Stage 1 projects are the first to reach this transition, but the stated goals of a new ultra-rugged TD and a 3-DoF powered wrist will build upon this experience, and is expected to be market-ready within the foreseeable future.

## ACKNOWLEDGEMENTS

1. Funding has been provided for these projects by a PRORP Technology Development Award, part of the Congressionally Directed Medical Research Programs (CDMRP).

2. Previous funding for the earlier products, upon which some of these projects are based, including the MC Electric Terminal Device (ETD), MC Flexion Wrist and Multi-Flex Wrist, has been provided by NIH, NSF, and NIDRR (Dept.of Education).

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## FORCE LIMITING AUTO GRASP – F.L.A.G. - A USER-INITIATED METHOD FOR GRIP SECURITY

Harold H. Sears  
*Motion Control, Inc.*

### ABSTRACT

Background: High pinch force electric hands (and hook-type TDs) are capable of crushing a fragile, object and pinching up to 20-25 lb (88-110 N).

New myoelectric TD wearers typically do not have well-developed proportional control. They must be cautious in gripping fragile objects, and often are timid about using their TD around family members and other people. The goal of this project was to develop a simple, easily enabled force limit technique so the wearer of a high-force myoelectric hand can be assured of limiting their pinch force to a desired level.

- Specific Criteria:
  - The wearer must be able to turn the feature on/off at will
  - Low pinch must be easily limited, and maintained
  - An additional feature should include auto-grasp, i.e., automatic response to electrode slip or loss of contact with skin.

The accomplishments of the development, to date:

- The FLAG feature may be provided with the Motion Control terminal devices which contain the ProPlus controller, including the ProPlus ETD, and the ProPlus MC Hand. The wearer completes the following sequence to utilize the FLAG feature:
  - FLAG is enabled with a “Hold-Open” command for 3 continuous seconds. An audible beep signals the wearer that the F.L.A.G. feature is enabled.
  - Wearer intentionally grips object - the TD will stop gripping at 2-3 lb grip force (9-13 N). A “beep” signals the wearer that the grip force has limited, and the motor turned off.
  - Pulse the Grip Force step-by-step to desired force - each command pulse raises force +3 lb. Each pulse is accompanied by a “beep”.

- To disable F.L.A.G feature, Hold-Open Command again, for 3 sec. A double “beep” indicates feature is turned off.

- Auto-Grasp feature is also provided, while the FLAG feature is enabled. A very sudden EMG signal to open the TD automatically will trigger a single pulse to close the TD.
- A small-scale field trial is underway, which will provide feedback from actual wearers. Data is being collected to evaluate the benefit of FLAG to wearers, and to give indications on the :
  - The situations when FLAG is utilized.
  - The frequency of FLAG utilization.
  - The ease with which the wearers are able to enable or disable FLAG

## FUNCTIONAL ASSESSMENT OF TRANSRADIAL AMPUTEES WITH A MYOELECTRIC POSTURAL CONTROLLER AND MULTI-FUNCTIONAL PROSTHETIC HAND

Jacob Lionel Segil<sup>1</sup>, Stephen Huddle<sup>2</sup>, Richard F. Weir<sup>2</sup>

<sup>1</sup>University of Colorado at Boulder; <sup>2</sup> University of Colorado at Denver / Anschutz Medical Campus

### ABSTRACT

A technological bottleneck in the development of multi-functional prosthetic hands remains the myoelectric controller. The myoelectric controller should provide a human-machine interface that deciphers user intent in real-time and be robust enough to operate in daily life. Here we describe a functional assessment of transradial amputees with a myoelectric postural controller and multi-functional prosthetic hand. The postural controller used three myoelectric signals to drive a cursor in a two-dimensional domain, the postural control domain, and output a continuously varying hand posture. The postural controller did not require training and/or classification; instead it implemented a simple linear transformation based on the location of the cursor in the postural control domain in order to determine the desired hand posture. Functional grasps were arranged in intuitive locations with respect to the orientation of the three EMG signals within the postural control domain. Transradial amputees performed the Southampton Hand Assessment Procedure with a modified Bebionic hand and a postural controller. Able-limbed subjects also performed the test with the identical prosthesis and controller for comparison. The results describe that the transradial amputees and able-limbed subjects achieved the same performance indicating that the postural controller is a valid myoelectric control system after transradial amputation. The transradial amputees restored 55% of typical hand function on average. These results prove the efficacy of the postural controller for transradial amputees and compare favorably to other myoelectric controllers in the commercial and research realms.

## INITIAL EXPERIENCE WITH THE RIC VO/VC TERMINAL DEVICE

Jon Sensinger<sup>1</sup>, Jim Lipsey<sup>2</sup>, Ashley Thomas<sup>2</sup>, Kristi Turner<sup>2</sup>  
*University of New Brunswick, 2) Rehabilitation Institute of Chicago*

### ABSTRACT

We have developed a lightweight simple terminal device that can switch between voluntary opening and voluntary closing modes. This device combines the advantages of both modes, but retains a reasonable size, weight, complexity, and ruggedness. In this paper we describe the design concept, design embodiment, performance specifications, and results of a group of pilot subjects using the device to complete the Southampton Hand Assessment Protocol. We found that the majority of subjects (6/7) obtained better Index of Function IOF scores when they could choose between modes than when they had to use only one mode.

### INTRODUCTION

Body-powered prostheses are preferred by the majority of persons with a unilateral amputation [1] due to a variety of reasons including a lightweight rugged design, low cost, and extended physiological proprioception [2], [3]. However, the cable-actuated harnessing system can only pull; it can't push. Devices accordingly fall into one of two categories: voluntary-opening (VO) devices or voluntary-closing devices (VC). VO devices are similar to clothes-pins, in which a spring keeps the device closed. A common example in prosthetics is the Hosmer #5 hook. VC devices are similar to bicycle calliper brakes or wrenches; a common example in prosthetics is the TRS Grip. Both types of devices have advantages and weaknesses, which make one of them better than the other for a certain task, but neither of them universally better across all tasks. Many persons accordingly wish a terminal device could switch between the two modes [4].

Several groups have tried to create a device that transitions between the two modes across the range of Bowden cable movement (e.g. [5]–[9]). However, because the fingers must both open and close for the same required Bowden cable movement, the net gear ratio, and thus the pinch force, must mathematically be weaker to accommodate the increased distance travelled. This class of devices is therefore unlikely to see adoption in clinical prosthetics use.

Several other groups have designed devices that can switch between two modes [10], [11]. In these designs full Bowden cable excursion is used for a given movement (maintaining high pinch forces), and the user can switch whether Bowden cable excursion opens or closes the fingers. The key challenge of this switching strategy is to design a device that can maintain the same thumb position in both modes despite the fact that both the finger position and direction of movement reverse, while maintaining a design that is rugged, simple, lightweight, and small—indeed, all of the properties that make body-powered prostheses so popular in the first place. This goal has so far remained elusive.

We have recently developed a simple design that easily switches between modes [12]–[14]. In this paper we present the basic concept, some performance specifications of the device, and results of a pilot study tested on subjects using the device. We conclude by highlighting a take-home field trial that started in April 2014.

### DESIGN CONCEPT

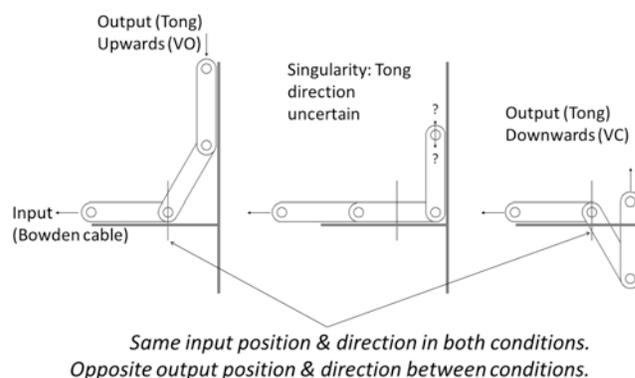


Figure 1: A linkage singularity can change the output position and movement direction, for a given input position and movement direction

The basic design concept employs a linkage singularity (Figure 1). Linkage singularities are commonly used in a variety of objects, such as car crankshafts and piston rods

on train wheels. In our use, we exploit the fact that a given output can have a different initial position and movement direction on either side of the singularity. By using a switch to determine which path is chosen, we can easily switch between the two modes while retaining the same input position and movement direction in both modes.

### DESIGN EMBODIEMENT

This design concept can be embodied in a prosthetic terminal device, as shown in Figure 2. The thumb position stays the same in both modes, but the left tong changes position and movement direction between modes. It is important to note that it is always the lateral tong that moves in both modes – an important design feature for object manipulation.

There are two springs in the RIC VO/VC design. One is attached to the thumb; the other is attached to the moving tong. In this way the spring forces add together in VO mode (to provide a higher force), and subtract in VC mode (to decrease the force that must be overcome by the user to apply their own cable tension).



Figure 2. The RIC VO/VC design

### DEVICE PERFORMANCE

The device is the same size as the APRL VC terminal device. It weighs 352g. 12N of cable force is needed to position the device in VC mode, and the device can deliver 6N of pinch force in VO mode. A hysteresis curve for both modes is shown in Figure 3. From this curve it can be seen that both modes are efficient (little hysteresis is observed), and that the VO mode generates greater cable force than VC mode.

### OUTCOME MEASURES

Five able-bodied subjects and two subjects with an amputation (1 transradial and 1 transhumeral) participated in a study that used the Southampton Hand Assessment Protocol (SHAP) [15]. All subjects gave informed consent in a study approved by the Northwestern University Institutional Review Board.

Subjects completed each task of the SHAP test four times. The first trial used VO or VC mode (in randomized order). The second used the alternative mode. In the third trial, they were asked to choose which they preferred for that particular task. In the fourth trial, they had to use the opposite mode (to assess whether there was a learning affect across mode).

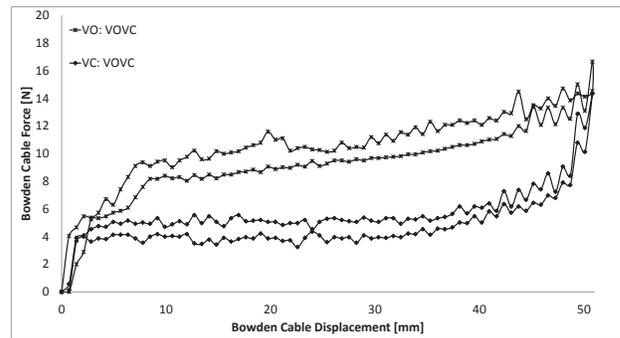


Figure 3. Bowden Cable Hysteresis curve

Able-bodied subjects had an average IOF of  $50.2 \pm 7.3$  in VO mode and  $53.8 \pm 11$  in VC mode. Importantly, they had an average improvement of 5.5 IOF in their preferred mode over being forced to use VO mode across all tasks ( $p=.02$ ) and an average improvement of 7.0 compared to being forced to use VC mode across all tasks ( $p = .18$ ). No statistically significant learning effect was observed in the fourth trial ( $p=.25$  and  $.20$ ). Both subjects with an amputation had improved performance when they could switch modes (51 and 55) compared with VO mode (40 and 43) or VC mode (44 and 47).

### DISCUSSION

The RIC VO/VC design was appreciated by all of the users – particularly the person with a transhumeral amputation, who preferred to use VC devices but was constrained to use VO devices in order to preserve elbow control. Many subjects expressed enthusiasm regarding how easy it was to switch between modes.

The pinch-force in VO mode was low (6N) compared with other VO devices (e.g. 20N for #5 Hosmer or 16N for Sierra 2-load), and the majority of subjects wished it was higher. We had originally designed the device with a low VO pinch force, hypothesizing that subjects would prefer VC mode for tasks that required higher pinch forces (thus saving on harnessing strain for the majority of VO movements at a lower force). However, it became evident that VO was preferred for several tasks that required higher pinch forces. These included tasks such as knife cutting (in which the thumb should be distal when the fingers are closed), and tasks

that required multi-joint movement (in which it was difficult to maintain consistent pinch force using VC mode).

A new version of the device has been designed that increases the pinch force in VO mode. This device is being sent home for month-long field-trials, after which standardized outcome measures will be performed [16].

### ACKNOWLEDGEMENTS

The contents of this research were developed under a grant from the U.S. Department of Education, NIDRR grant number H133G120059. However, those contents do not necessarily represent the policy of the Department of Education, and you should not assume endorsement by the Federal Government.

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## INITIAL EXPERIENCES WITH THE RIC ARM

Jon Sensinger<sup>1</sup>, Jim Lipsey<sup>2</sup>, Tom Sharkey<sup>2</sup>, Ashley Thomas<sup>2</sup>, Laura Miller<sup>2</sup>, Kristi Turner<sup>2</sup>, Jose Ochoa<sup>2</sup>, and Tom Idstein<sup>2</sup>

*University of New Brunswick, 2) Rehabilitation Institute of Chicago*

### INTRODUCTION

RIC has developed a small, lightweight, modular prosthetic arm that comprises an elbow, wrist rotator, wrist flexor, and hand with powered thumb. This arm was designed to be appropriate for a 25<sup>th</sup> percentile female. This paper discusses the enabling design features of the device, along with feedback obtained during in-laboratory use. This arm is part of a larger take-home field trial.

### MOTIVATION

The majority of existing prosthesis users rightfully want more—more speed, more torque, more degrees of freedom, etc...[1]. But only 50-70% of persons with an upper-limb amputation use a prosthetic device[2]. There is thus a substantial white-space that could be filled if the needs of this untapped ground of end-users could be filled. However, the needs of this group are very different, and indeed can be characterized by wanting less, not more. Specifically, persons with an amputation who reject the use of existing prostheses want less weight, in order to have better comfort, and they want less size, in order to preserve cosmesis [2]. There is a key gap in design options for a lightweight prosthesis that encourages this substantial subset of the population to use prosthetic technology.

Over the last five years, RIC has developed a lightweight, modular prosthetic arm targeted to fit the anthropomorphic weight and dimensions of a 25<sup>th</sup> percentile female. Cosmetic material and pylons can be extended to make this device appropriate in size and weight for 87% of the population. This arm has been achieved by looking closely at the engineering requirements of key design features, in light of the clinical realities of prosthesis use.

### ENABLING FEATURES OF THE ARM

**Elbow.** The elbow uses a custom exterior-rotor motor; a type of motor that achieves high torque and efficiency compared with conventional robotic motors [3]. Particularly, this type of motor is optimal for the ballistic motions seen in prostheses [4], in which a prosthetic limb starts at rest,

quickly ramps up to speed, and is at rest at the other end of its range of motion within a scarce 0.4 seconds. The huge accelerations incurred during these rapid movements, and the resulting forces caused by inertial components such as gear transmissions, have a large effect on the energy efficiency of the mechanism, and RIC has developed motors that are tuned to the daily requirements of activities of daily living.

This motor is connected through a planetary gear and a non-backdrivable clutch to a differential roller screw. Differential roller screws combine the high efficiency of ball screws with the ruggedness of lead screws. This differential roller-screw in turn drives a four-bar polycentric elbow joint, which provides a nonlinear gear ratio optimized to provide optimal torque and speed over the range of flexion.

Of particular note, the elbow captures the carrying angle found in human elbows, which allows the elbow to reach the midline when flexed without the need for humeral rotation (Figure 1).



Figure 1. Human elbows (left) have a natural carrying angle, that causes the hand to be more lateral than the elbow when fully extended. This carrying angle has an important effect during elbow flexion, causing the hand to move closer to the midline. In contrast, conventional prosthetic elbows (shown on the right side of the image) flex the forearm straight up, no closer to the midline, and thus require humeral rotation to reach the midline.

**Wrist Rotator and Flexor.** The wrist rotator and flexor both contain the same actuator design, which comprises a custom exterior-rotor motor, planetary gear, non-backdrivable clutch, and a custom cycloid gear. Cycloid gears offer a large gear ratio, rugged design, and a compact size. If designed properly, they can achieve high gear ratios—even at low torques, where conventional gears have poor efficiency [5]. Their design can be scaled to different sizes using a framework to properly transform the geometry [6]. The resulting wrist actuators are extremely quiet.

**Hand.** The hand drives all of the fingers together with a commercial off-the-shelf brushless motor, connected through a spur gear to a differential roller screw, which is in turn connected to a four-bar finger linkage. The fingers themselves are four-bar linkages, providing coupled MCP/PIP flexion that achieves wrap-around grasp.

The thumb is independently powered by a brushless motor connected to a custom off-axis helical gear. The axis of rotation of this helical gear was chosen after numerous iterations with clinicians to achieve a variety of grasps. It is worth noting that it was very difficult to find a grasping posture that stayed within a 25<sup>th</sup> percentile female envelope, yet was able to grasp large objects, specifically cans/drinking glasses, encountered in every-day life. The final design was able to achieve these goals, as evidenced by successful interaction with all of the tasks in the SHAP protocol [7].

**Other Essential components.** The RIC arm uses the PDCP Can bus standard [8], enabling it to communicate digitally amongst its own actuators and with other devices. Using a communication bus enables us to substantially reduce the number of wires. The wrist uses the latest version of the universal quick disconnect [9], developed by Motion Control. This quick disconnect enables the device to generate large wrist-rotator torques without unscrewing, and provides uninterrupted bus communication even when the wrist is manually rotated.

Pattern recognition is embedded in the master controller. Custom low-level motor controllers drive each of the brushless motors. The lithium-ion battery for the arm is housed in the forearm.

**PERFORMANCE METRICS**

Performance metrics are provided below (Table 1).

Table 1: Performance metrics

**HAND**

Close time (s)	0.4
Pinch force (N)	84+
Mass (g)	365

**WRIST ROTATOR**

Speed (deg/s)	900
Torque (Nm)	0.9
Mass (g)	170

**WRIST FLEXOR**

Speed (deg/s)	450
Torque (Nm)	1
Mass (g)	155

**ELBOW**

Speed (deg/s)	80
Torque (Nm)	12
pass torq (Nm)	68
Mass (g)	808

A rendering of the design is provided in Figure 2.

Figure 2. Design of the Beta-3 arm.

## FIRST IMPRESSIONS

This arm has been fit in a laboratory setting on one person with a transhumeral amputation. The subject gave informed written consent for a research study approved by the Northwestern University's Institutional Review Board.



Figure 2. Design of the Beta-3 arm



Figure 3. Small size of the Beta-2 arm is evident from the picture

## NEXT STEPS

This arm will be used in the third phase of an on-going study testing the effects of pattern recognition vs. direct control, in which subjects with a transhumeral amputation have already taken commercially available arms without a wrist flexor home for field-trials. Subjects will take the device home for a month, after which their performance on standardized tests will be measured and feedback will be obtained to improve the final device.

## ACKNOWLEDGEMENTS

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## OPPORTUNITIES AND CHALLENGES WITH RE-EMERGING EXTERNALLY POWERED UPPER-LIMB ORTHOTIC TECHNOLOGY; PART I: AN OVERVIEW

Jeremy Sherman<sup>1,2</sup>, BSME; Chris Lake<sup>1</sup>, CPO, FAAOP

<sup>1</sup>Lake Prosthetics and Research, Euless TX

<sup>2</sup>Baylor College of Medicine, Houston TX

### ABSTRACT

Since the 1960's, myoelectric control has received a significant amount of attention from various institutes globally. Commercially available orthotics produced from this global interest utilizing this technology has been available in one form or another since this period. While initially these devices proved the feasibility of myoelectric control, attempts to optimize the products for patients needs were met with significant and difficult to overcome challenges. Eventually this led to the conclusion that myoelectric control alone was insufficient for orthoses, and inferior to the cheaper and easier to produce mechanical alternatives.

Since that time, orthotics utilizing myoelectrics have re-emerged periodically in response to advances in the technology, but haven't solidified in everyday clinical practice. A wide array of patient presentations benefit from upper limb externally powered myoelectric orthoses, such as spinal cord, brain and peripheral nerve injuries, progressive degenerative neurological diseases and cerebrovascular accidents. As these patient populations continue to grow, it is becoming increasingly important as a profession to approach care in a more standardized and comprehensive manner.

### INTRODUCTION

Since the 1960's, myoelectric control has received a significant amount of attention from various institutes globally. Commercially available orthotics produced from this global interest utilizing this technology has been available in one form or another since this period. While initially these devices proved the feasibility of myoelectric control, attempts to optimize the products for patients needs were met with significant and difficult to overcome challenges. Eventually this led to the conclusion that myoelectric control alone was insufficient for orthoses, and inferior to the cheaper and easier to produce mechanical alternatives. [1]

Since that time, orthotics utilizing myoelectrics have re-emerged periodically in response to advances in the technology, but haven't solidified in everyday clinical practice. A wide array of patient presentations benefit from

upper limb externally powered myoelectric orthoses, such as spinal cord, brain and peripheral nerve injuries, progressive degenerative neurological diseases and cerebrovascular accidents. As these patient populations continue to grow, it is becoming increasingly important as a profession to approach care in a more standardized and comprehensive manner. In keeping with the theme of redefining the norm, it is essential to understand where this technology has come from before discussing how it can be improved, and so Part I of this presentation reviews a brief history of myoelectrics in upper orthoses and the particular challenges that are faced when providing for this patient population. This review will provide insight into the continuum of orthotic care, and the similarities between upper limb orthotics and prosthetics when it comes to the application of externally powered and hybrid systems and the application of counterbalance mechanisms. Part II of this presentation will address the next steps being taken and highlight specific clinical cases.

### TIMELINE OF EXTERNAL POWER IN UPPER LIMB ORTHOTICS

Since the introduction of myoelectric technology with regards to orthoses, there have been numerous advances in what prosthetists and orthotists have been able to supply their patients. The most readily apparent advances have been the materials. Woods and metals used to be the dominant components for both prosthetics and orthotics, but newer materials such as carbon fiber, titanium and thermoformable plastics now provide lighter, cheaper and stronger alternatives. In addition, methods by which these materials can be shaped to accommodate patients have also advanced significantly with vacuum forming and computer aided manufacturing.

Figure 1 below is a sketch of an early myoelectric orthosis designed by Case Institute of Technology in 1963. [1] Clearly shown are the surface electrodes and the battery they are connected to which is too large to be realistically harnessed to a patient comfortably.

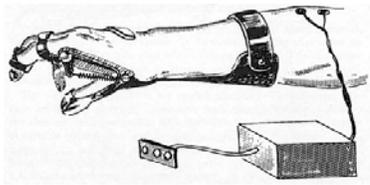


Figure 1: Case Institute of Technology, myoelectrically controlled orthosis, 1963

Rancho Los Amigos hospital in California developed a wheelchair mounted powered upper limb orthosis in the early 1960s. While mounting upper limb orthoses to wheelchairs alleviates issues such as weight restrictions and can eliminate some of the complex harnessing, they are not without issues of their own. The Rancho Los Amigos design did not use a myoelectric signal, but instead relied on utilizing switches activated by the tongue to input fourteen commands in controlling the seven degrees of freedom of the device. This was ultimately too complex to become widely accepted. [2]



Figure 2: Hugh MacMillan Rehabilitation Centre myoelectrically controlled & powered orthosis, 1989

In the 1980s the Hugh MacMillan Rehabilitation Centre designed a myoelectrically controlled hand and elbow orthosis. The device, pictured above in Figure 2, did not provide assistance to shoulder movements or humeral rotation, meaning the patients who could realistically use it were limited to those still capable of independent movement to some extent. Expanding on their work, the Hugh MacMillan Rehabilitation Centre designed a myoelectrically controlled wheelchair mounted object manipulator for quadriplegics. [2] The system was designed to address both the size issue of harnessed orthoses and the control issues of traditional robotic wheelchair orthoses.



Figure 3: Ege University Hospital Shoulder-Elbow Orthosis

In 2000, Ege University Hospital harnessed a myoelectrically controlled shoulder-elbow orthosis for brachial plexus injuries. Specifically this was designed to provide an alternative to amputation, and geared towards patient rehabilitation. [3] The device required a training time of three months for effective use of the muscle groups utilized as electrode sites.

Figures 4 and 5 below are two current designs of upper limb myoelectric orthoses. The first is an elbow orthosis designed by MyoMo called the MyoPro. The second is a WHO by Broadened Horizons called the Power Grip.



Figure 4: "MyoPro" by MyoMo



Figure 5: "Power Grip" by Broadened Horizons

As can be seen above, the more current designs are able to utilize myoelectrics, lightweight metals and plastics, lithium batteries as well as smaller motors and actuators in order to integrate the entire system into more manageable devices that can be easily harnessed to patients with minimal discomfort. The market however currently lacks any standardized system of care for patients who require orthotic care for the shoulder, wrist and hand all at once. This is the necessary next step in order to allow for activities vital to everyday life.

## BARRIERS TO UPPER LIMB ORTHOTICS

Initial forays into externally powered upper limb orthoses were met with various, readily apparent limitations due to the technology of the time. In the following years, many of the limitations have been addressed while new ones have arisen. Below are listed the related concerns and how they have been addressed, if applicable.

### Excessive Size and Weight

Excessive size and weight have always been a concern for upper limb prosthetics and orthotics. The more distal on a limb that weight is added, the heavier it feels for the patient. Batteries, sensors, and all the harnessing for the electrical equipment add up. This must be kept in mind since for patients with impaired muscles a few grams can be the difference in comfortably using the equipment or inability to use it at all. In the case of prosthetics a limb is absent, either partially or in whole. To a degree this gives a margin of error for adding weight as it is replacing what was lost. In addition prosthetic patients don't generally have as large a strength deficit at their proximal joints as their orthotic counterparts. Because orthotists are working around an existing limb and not a void, and their patients exhibit severely reduced strength, more time must be taken by these clinicians to fine tune and reduce weight and bulk for proper fittings.

In some cases, clinicians will find patients with impaired motor control who are unable to lift the weight of their arm against gravity, but if gravity could be eliminated are able to pronate, supinate, flex, and extend their arm with reasonable freedom. Devices that provide this counterbalance mechanism are currently used in the market for quadriplegics and other patients who are wheelchair-bound. Figure 6, the JAECO WREX is one such device.



Figure 6: Wilmington Robotic Exoskeleton (WREX)

Appropriately modified to a harnessing system (evenly distributed load, stable in normal positions of use, and easy to don and doff) these devices are lightweight and very simple for patients to use. Orthotists need to take these existing technologies into account when designing new myoelectric orthoses, even though they weren't produced

with myoelectrics in mind. Utilizing a system such as the WREX as a counterbalance for the arm against gravity can open up more possibilities for WHO myoelectric systems that can be used without overloading patients.

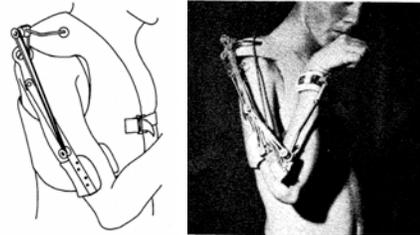


Figure 7: Locking Joint Functional Arm Brace

Figure 7 shows a shoulder flexion assist, which is a very simple way to enable patients to place hand to mouth and other positions useful to daily living. It is only in those cases where the patient's extensors are too weak to stretch the rubber bands that external powered myoelectrics should be applied. [4]

### Difficulty Finding Adequate Control Sites

Reduced signal to muscles is to be expected with stroke, degenerative muscle diseases and peripheral nerve injuries to name a few of the pertinent patient populations. With prosthetics, myoelectric sites tend to be limited but signal strength also tends to be relatively strong. With orthotics, not only are signals generally weak, but in some cases they are confused as well; a good example being brachial plexus injuries. At one point the electrodes were considered the weakest link in a myoelectric system, but as myoelectric sensors efficacy has improved we have begun to see adequate signals for orthoses. Newer sensors are able to detect what previously we could not, increasing the number of control sites available for upper limb orthoses.

### Signal Processing

As discussed above, myoelectric sensors are now able to pick up weaker signals than before. Processing these signals effectively is just as important a step in providing functional devices. Signal amplification is necessary for patients with reduced muscle potential. Since the 1980's, microprocessors clock speeds have increased by an order of magnitude into billions of cycles per second. Advances with microprocessors bandwidth, on-board cache, heat dissipation and ability to complete more instructions per clock cycle continue to improve their efficiency. Cumbersome or remotely placed control units have been an issue with upper limb myoelectric devices [5] but the aforementioned advances are all steps towards rectifying this. More efficient microprocessors allow the devices they are in to be smaller and faster, and to perform more complex actions.

The more complex orthoses become and the more sensors that are added to current and future models, the greater the amount of potential problems that arise. Crosstalk and digital artifacts are paramount among these. It is for this reason that even as microprocessors become more advanced, orthotists need to be aware of potential problems that still exist and not overcomplicate their devices.

#### Extended Therapy Time and Time in Office

Custom devices will always take more time to design and assemble for patients than off the shelf, prefabricated products. In addition, the increased time for custom devices generally means that cost will also be higher, which is the real limitation in developing upper limb orthotics. The sophisticated engineering required is often economically unjustified. [6]

Clinically speaking, upper limb prosthetics by their nature require more patient time in the office for follow up appointments than lower limb prosthetics. Whether this is to perfectly adjust the fit, or to program the system, or to create specialized prosthetics designed for specific tasks, this extra time is the result of attempting to replace something as dextrous as the human hand. By the same token, upper limb orthotics should take practitioners just as much time if not more. The end goal of providing a functional limb that is easy to use for everyday tasks, and the steps to get there are similar as well. To provide the best fits, patients will still need custom sockets and harnessing, the system will need to be fine-tuned or programmed for the patient, and in most cases patients with reduced musculature can benefit from the extra care of specialized tools for specific tasks. The reason orthotists can justify an equal amount of time on upper limb orthoses as prosthetists do on their equivalent prosthetics is simply because there are fewer off the shelf options for orthoses currently available. This barrier is one that will take care of itself in time as the need is recognized and more research and design work is put into good standardized options to cut down on custom fabrication time.

#### Redefining the Norm

Perhaps the greatest threat to the new and innovative ideas that push a profession to improve, are the preconceived notions of what can and cannot be done. In the field of prosthetics and orthotics this is no different, and clinicians should always be ready and willing to challenge what has been provided for patients in the past. It is even more important to challenge these preconceived ideas as healthcare providers, as the quality of work that goes into making prosthetics and orthotics has a direct effect on the quality of day to day life.

“Often a patient will have a better outcome from having a well-fitting, functional prosthesis than a non-functional replanted limb.” [7] This is a direct quote from the Veterans

Health Council, but the idea behind it is pervasive in prosthetics and orthotics. Healthcare professionals will even encourage patients without traumatic injuries to amputate their limb in order to be prosthetic candidates. This approach is a result of an industry that encompasses two areas of interest, both orthotics and prosthetics, but due to limited technology initially consolidated the majority of its efforts into prosthetics. The scope of concerted effort required to address the plethora of types of disability in paralysis or neurological disorders slowed down the development of externally powered orthoses. [6]

The only way this or any pernicious idea can be addressed in the industry is to never follow it blindly. As a group of healthcare professionals, orthotists and prosthetists need to agree to challenge what have been the old normal ways of doing business and instead look to what they can do to provide patients with the best quality of life.

### CONCLUSIONS / NEXT STEPS

If the success of orthotics is measured by a patient’s acceptance of their devices and his or her ability to use them effectively [6], an equal amount of care must be given to both providing systems robust enough to meet complex needs, but simple, reliable, and comfortable enough to not be rejected. Whatever residual muscle power the patient has must be developed to a maximum while eliminating unnecessary or complicated equipment. [8]

Upper limb externally powered orthoses require the same amount of attention as upper limb prosthetics both on the scale of industry time, money, research, and on the scale of patient hours in office. If we are to redefine the norm and provide patients a standardized and comprehensive quality of care, this has to continue to be a profession-wide effort. Part II of this presentation will address the growing patient population and specific clinical cases from various presentations requiring externally powered myoelectric upper limb orthoses.

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## OBJECTIVE OUTCOME MEASURING, DESIGN OF A PORTABLE LOW-BUDGET DATA ACQUISITION SYSTEM

Gerwin Smit\*, Sezen Tosunoğlu\*\*

\*Delft University of Technology, \*\*The Hague University of Applied Sciences

### ABSTRACT

In order to improve the design of prosthetic devices, it is important to have objective quantitative data on the use of prosthetic devices during activities of daily living outside the clinic. Quantitative objective data on the use of upper limb prostheses outside the clinic are very scarce.

In this study a data acquisition system to measure the usage of prosthetic hands has been designed and tested. The system is portable and works for over 8 hours. The sensors and amplifiers are located in the glove of the prosthesis. The total system weighs less than 100 grams. The system is low-cost and easy to build. The data acquisition system can be used to measure and record during activities of daily living outside the clinic

### INTRODUCTION

Rejection rates of upper limb prosthetics are high. User studies show rejection rates varying from 23 to 45 %.<sup>2</sup> From the people who wear a prosthetic device, a large group (~27 %) does not actively use the prosthetic device.<sup>1</sup>

In order to improve the current prosthetic devices, it is necessary to have quantitative objective data on how the prosthetic device is being used during activities of daily living outside the clinic. Currently such data is very scarce. Only a few studies report such data. In a study by Van Lunteren *et al.*<sup>3</sup> the use of the prosthetic device by children was monitored and recorded by an observer. This study is however an exception. Basic quantitative data, like the amount of hand openings and the number of hand grasps during a day, or the level of pinch force, are hardly available in literature. This data would however be very useful in improving upper limb prosthetic devices.

With the current state of technology and the wide availability of small electronic devices, it would be relatively easy to build an electronic device that can monitor and record relevant parameters of prosthetic use, e.g. hand opening, pinch force, wearing time. This study aims for the design of such a data acquisition system (DAS). It would be useful if

such a system would be compatible with all current prosthetic terminal devices. In this study the designed DAS was applied to a body-powered device, as there is an urgent need for improvement in the design of body-powered devices.<sup>4,5</sup> It would however be relatively simple to connect the system to an myo-electric device.

### PROBLEM

In order to improve the design of prosthetic devices, it is important to have objective quantitative data on the use of prosthetic devices during activities of daily living outside the clinic. Although many user studies have been performed in the past decades, quantitative objective data on the use of upper limb prostheses outside the clinic are very scarce. Nearly all available data has been gathered by self-reported user surveys, and is therefore subjective.

### GOAL

The goal of this study was to design a portable, low-cost, easy to build, light-weight data acquisition system (DAS). The device should be able to objectively record quantitative data of prosthetic use for any upper limb prosthetic device, during activities of daily living outside the clinic.

### METHODS

#### Requirements

In order to be suitable for measurements the following requirements were set up for the data acquisition system (DAS). The system should be:

- Portable (Power supply by a battery)
- Light-weight (< 250 grams)
- Small-sized (Not visible from the outside)
- Low power dissipation (Enough power for 8 hours)
- Sufficient data capacity (8 hours recording)
- Low-cost (< €500)
- Accurate data (Within the set measurement error)

- Robust (Should not break during ADL)
- Easy to build (By any researcher)

Parameters of interest are the following:

- Time of wear per day (within 10 minutes of accuracy)
- Position of fingers (Within 5° of accuracy)
- Pinch force on fingers (1-60 N ± 1 N)
- Tension on the harness cable (1-200 N ± 1 N)

Based on the requirements, the most suitable sensors and microcontrollers were chosen, and assembled to a complete working DAS. All selected components were widely available low-cost components. In this study the DAS was applied to an Otto Bock Voluntary Opening hand.

#### Sensors

In order to meet with the before mentioned design metrics, several sensors were investigated, to measure the different parameters of interest. After testing and comparing different options, the most suitable sensors were chosen to be used in the DAS.

**Wearing time** Not only the time of use, but also the time that the prosthesis is being worn is valuable information. The wearing time can easily be measured with a temperature sensor. Various temperature sensors are available for this purpose and most of them are low-cost solutions.

**Hand opening** Not only the frequency of finger movement, but also the position of the fingers or the hand opening provide valuable information about the use of the prosthetic device. Various ways to measure the degree of finger movement were investigated:

- Potentiometer (resistance depends on position)
- String Potentiometer (resistance depends on position)
- Hall-effect Sensor (magnetic fields depends on position)
- Bend Sensors (resistance depends on bending curvature)

**Pinch force** While grasping or pushing objects there is a force exerted to the fingers. This information, along with data about the position of the fingers at the moment the force is being applied, can be useful to understand how the prosthetic hand is functional to the amputee in ADL's. Two types of sensors were investigated:

- Force Sensing Resistors (FSR)

- Strain gauges (resistance changes with elastic elongation of the strain gauge)

**Activation force** In most body-powered hand prostheses a harness is used to close or open the hand. This harness is worn around the contra-lateral shoulder and has a cable to the mechanics of the prostheses. A certain force has to be applied on the harness cable in order to close or open the hand, depending on the kind of hand prostheses; voluntary closing or voluntary opening.

#### Microcontroller

The core of the DAS is the microcontroller, which processes the signals from the sensors and stores this data in an appropriate storage device. The hard requirements for the microcontroller were determined to be the following:

**Size** Small enough to fit preferably in the glove and consequently light-weighted, such that neither the amputee nor someone else can notice the microcontroller

**Analogue pins** Depending on the hand. One to five pins for the force sensors, one for the tension in the shoulder strap. Depending on the type of hand and the degrees of freedom per digit, one to fourteen sensors for the position of the fingers (in case of multi-articulating fingers).

**Data storage** The measured data should be stored somewhere. This could be a SD card connected to the controller. Alternatively the data could be sent wireless to another storage device.

**Price** The system should be low-cost. Among the best options the one with the lowest price will be preferred.

## RESULTS

#### Sensors

The following paragraph briefly describes the tested and selected sensors:

**Wearing time** A digital temperature sensor, DHT11, has been tested with the microcontroller. This particular sensor measures also humidity, but only data about temperature will be stored.

**Hand opening** Of the different sensor to measure the finger position, the Hall-effect Sensor (SS495A1 by Honeywell) was chosen. This sensor is cheap, has a very small build-in space and is easy to mount. Other sensors are not so easy to mount inside a hand or a finger ([string] potentiometers) or have a bad repeatability (bend sensors).

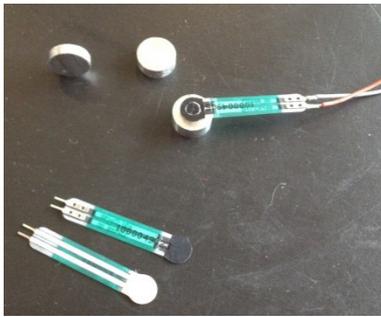


Figure 1 Examples of three FSR's that were tested, for measuring the pinch force. Above the round metal disks that were used to equally distribute the force. Below both sides of the FSR

**Pinch force.** To measure the pinch force, FSR's and strain gauges were tested. Tests were performed by applying the same force several times and measuring the outcome.

- The FSR's (Figure 1) gave different output values, depending on where the force on the active area was applied. To improve the repeatability two thin metal disks were placed at each side of the FSR, in order to distribute the applied pinch force equally over the active area of the FSR. There was however no improvement. Although placing the same weights, the FSR gave different values each time.
- Measurements with the strain gauges (Figure 2) showed a clear linear relation between applied force and output signal.



Figure 2 Two strain gauges applied to the thumb of an Otto Bock VO-hand (front one visible in detail picture). Note the cut-out that was made in the thumb, to enable accurate strain measurement

**Activation force.** To measure the activation force a load cell (LSB200, Futek) was selected, as this sensor was already available at our institute. This sensor is a small

sized s-beam load cell., which is S-shaped from the inside and has a full Wheatstone bridge of strain gauges. Due to its shape this sensor can measure both tension and compression, although only tension is required for measuring the cable force. The sensor can be placed in-between two wires that can be screwed to the sensor easily. These type of sensors are accurate and the change in cable length due to elastic deformation is negligible. The only drawback is the high price of this particular model (~ €400). There are other small sized, low-cost s-beam load cells (~ €100) available on the market that also provide a good solution. The selected load cell sensor has a capacity of 450 Newton, but as it is not very likely that a force larger than 200 Newton will be applied to the harness cable, the sensor was calibrated to measure to a maximum of 200 Newton at 5 Volt output voltage. An op-amp (AD623) with a gain factor of 830 was used to amplify the signal.

Microcontroller

According to the requirements mentioned in the methods section, several microcontrollers were selected for consideration, based on their specifications. Table 1 gives an overview of the microcontrollers which were closest to the requirements. From these selection the Teensy2.0 and Arduino Micro were left for further consideration, as it was desirable to have the largest number of analogue inputs, in case more signals were needed for future applications.

Table 1 An overview of the microcontrollers which were closest to the requirements.

	# Analog pins	# Digital I/O pins	Supply voltage	Dimensions	Compatible standard shields	Price+ shipping
Tiny Duino	6	14	5V	20mm x 20mm	Bluetooth (€37) microSD (€8.95)	€30 + €3
Teensy 2.0	12	13	5V	17,8mm x 30,5mm	microSD (€6)	€12 + €7,60
Arduino mini 05	8	14	7-9V	18mm x 30,5mm	wifi+sd (€20)	€15 + €6,95
Arduino micro	12	20	7-12V	18mm x 48mm	wifi+sd (€20)	€26 + free

Both microcontrollers have 12 analogue input pins. The Teensy has its own micro SD adapter available for just €6,- . Furthermore the Teensy requires a lower supply voltage than the Arduino boards. A disadvantage of Teensy is that there is less documentation and support available compared to the Arduino. Arduino is a very popular microcontroller used by many people. There are many websites and forums where Arduino users share their projects. This knowledge and experience of others helps the programmer to find a quick

solution when encountered with problems. Therefore it was decided to use the Arduino Micro (Figure 3) to build the DAS.

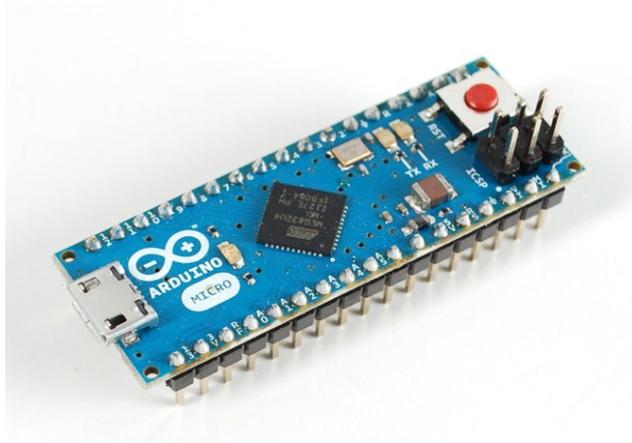


Figure 3 The Arduino Micro was selected to be used in the data acquisition system

After selection of the microcontroller, the sensors were attached to the controller, the controller was programmed and the sensors were calibrated. The DAS was powered by a rechargeable battery pack (Varta, NiMH, 510mAh, 4.8V), which could power the system for 8.5 hours. The total system weights less than 100 grams.

## DISCUSSION

**Portability** The system works for 8 hours on battery. The battery itself is small and weights 60 grams. The system is integrated inside the arm prosthesis and is not visible from the outside. The system does not interfere with the daily activities of the amputee.

**Lightweight** The system weights less than 100 grams which is far within the goal of 250 grams.

**Small sized** The sensors and amplifiers were integrated inside the hand prosthesis (Figure 4). Most other parts can be either integrated into the arm, or be worn inside the sleeve of the clothes.

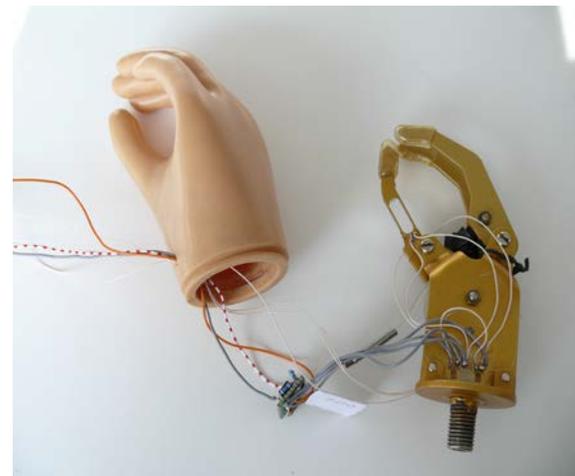


Figure 4 The instrumented Otto Bock VO hand. The strain gauges are attached to the thumb of the hand frame (right). The measurement amplifier (middle) fits inside the inner glove of the hand (right).

**Low power dissipation** Components with a low power current drain were chosen, when possible, in order to maximise the operating time of the system.

**Data capacity** A micro SD card socket has been added to the system, which was a small-sized and easy solution to store large amounts of data. In this study an 8 GB card was used.

**Low-cost** The entire system is assembled from low-cost components. Currently the only expensive part is the load cell. When a cheaper load cell (~ €100) would be used, the total price of the system would be less than €200.

**Reliability** The parameters of interest were measured within the mentioned accuracies. After calibration several tests were performed to ensure reliability. The program has been adjusted according to these tests. Future tests, performed while the system is being used, will show if the reliability is granted during ADL's.

**Robustness** The strain gauges and the circuit inside the hand prosthesis have been coated to protect against friction, static build-up and moisture. The cables are thin and flexible and can move together with the hand without damaging the system. Wires outside are twisted and coated. Further tests during ADL use will show the robustness of the system.

**Easy to build** All components are available from the internet. Documentation about these components is widely available. The documentation together with this article, enables any researcher with basic technical skills build his own DAS.

## CONCLUSION

A data acquisition system to measure the usage of prosthetic hands has been designed and tested. The system is portable and works for approximately 8 hours. Except from the microcontroller and the micro SD card socket, the system is not visible from the outside. The sensors and amplifiers are in the glove of the prosthesis. The total additional mass of the DAS to the prosthetic hand is less than 100 grams. The system is low-cost and easy to build. The DAS can be used to measure and record during ADL's outside the clinic.

## ACKNOWLEDGEMENTS

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## SIMULTANEOUS CONTROL OF A MULTI-DOF WRIST/HAND SYSTEM USING PARALLEL DUAL-SITE CONTROL WITH INTRAMUSCULAR EMG

Lauren H. Smith<sup>1,2</sup>, Todd A. Kuiken<sup>1,2</sup>, and Levi J. Hargrove<sup>1,2</sup>

<sup>1</sup>Northwestern University, Chicago IL

<sup>2</sup>Rehabilitation Institute of Chicago, Chicago, IL

### INTRODUCTION

A limitation of currently available myoelectric prosthesis control strategies is the inability to simultaneously control multiple degrees of freedom (DOFs). Neither conventional dual-site differential control (“direct control”) [1] nor newly commercially available pattern recognition control systems [2] allow for simultaneous control. Previous attempts at simultaneous control using surface electromyography (EMG) signals have included pattern recognition [3], use of neural networks to predict joint kinematics or kinetics [4], and analysis of underlying muscle synergies [5]. While these approaches are promising, most studies have been limited to controlling the wrist without the hand, do not provide independent control of each DOF, or have been confined to offline evaluation and thus have not demonstrated real-time simultaneous control.

Recent advances in implantable recording devices [6, 7] may make intramuscular EMG a clinically feasible option for myoelectric control. Intramuscular signals can be recorded with substantially less cross talk, which may allow for simultaneous control approaches that are not feasible with surface EMG. In particular, an approach termed “parallel dual-site” takes advantage of the decreased cross talk in intramuscular recordings. This extension of conventional dual-site differential control uses the difference in intramuscular EMG amplitudes from an antagonist-agonist muscle pair (i.e., from *dual sites*) to control a physiologically appropriate DOF. Multiple muscle pairs are used in *parallel* to control different DOFs simultaneously. This approach has the potential to allow persons with transradial amputations to control DOFs simultaneously using physiologically appropriate residual musculature.

Parallel dual-site control has been frequently proposed [7-9] but has not been evaluated in depth, although the few previous studies have been promising. Birdwell demonstrated simultaneous finger control using this approach [10]. Parallel dual-site control was also briefly described in two transradial amputees [8, 9], but these studies did not quantitatively evaluate prosthesis control or investigate simultaneous control. The objective of this study was to evaluate parallel

dual-site control using intramuscular EMG for simultaneous control of a multi-DOF wrist/hand system.

### METHODS

The study was approved by the Northwestern University Institutional Review Board. Five able-bodied subjects used fine-wire EMG in a parallel dual-site configuration to control a cursor in a virtual task. Control of three DOFs was evaluated: wrist rotation, wrist flexion/extension, and hand open/close. Each subject had prior experience with sequential pattern recognition control.

Bipolar fine wire electrodes (Natus Neurology Inc.) were inserted into the following six forearm muscles using hypodermic needles: *pronator teres*, *supinator*, *flexor carpi radialis*, *extensor carpi radialis longus*, *flexor digitorum profundus*, and *extensor digitorum*. Insertion sites were guided by palpation and confirmed by electrical stimulation and by appropriate EMG activity during test contractions. After fine wire electrodes were inserted, subjects were restrained using a custom brace to ensure isometric contractions.

EMG signals were collected using a Motion Lab Systems MA300 EMG system, which amplified signals 350x and band-pass filtered them between 10 Hz and 2000 Hz. Signals were sampled at 5 kHz by a National Instruments data acquisition system. Signals were digitally high-pass filtered with a 3<sup>rd</sup> order Butterworth filter with a cutoff frequency of 20 Hz. The presence of cross talk between the EMG channels was evaluated offline using the peak cross correlation of the signals.

A parallel dual-site myoelectric control system was configured to control three DOFs: wrist rotation, wrist flexion/extension, and hand open/close (Fig. 1). Each DOF was controlled by the corresponding physiologically appropriate muscle pair. The mean absolute value (MAV) of each EMG signal was calculated from 250 ms sliding windows with a frame increment of 50 ms. Each EMG channel was digitally amplified and thresholded. The output velocity for each DOF was determined by the difference in the conditioned MAV signals of the corresponding muscle pair. More explicitly:

$$\text{Velocity} \propto [G_1 M_1 - T_1] - [G_2 M_2 - T_2] \quad (1)$$

Where  $M$  is a 2-dimensional vector containing the MAVs of the antagonistic muscle pair,  $G$  and  $T$  are 2-dimensional vectors containing gains and thresholds, and  $[\bullet]$  represents  $\max(0, \bullet)$ . The gains and thresholds were manually set to minimize unintended DOF activity, to maximize the dynamic range of velocities, and to map a comfortable-level contraction to approximately 50% of the maximum possible velocity. A post-processing velocity-dependent ramp that has previously been developed for pattern recognition control [11] was also implemented for each DOF to minimize abrupt changes in DOF activity.

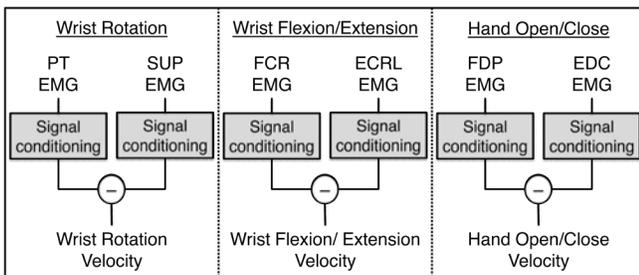


Figure 1: Configuration for parallel dual-site control. A pair of antagonistic muscles was used to control the output velocity for each DOF. The differences in EMG signal amplitude for the two muscles were calculated after each signal was conditioned by a linear gain and threshold. The muscles used included pronator teres (PT), supinator (SUP), flexor carpi radialis (FCR), extensor carpi radialis longus (ECRL), flexor digitorum profundus (FDP), and extensor digitorum communis (EDC).

Subjects used the output of the control system to control a ring-shaped cursor in a pseudo three-dimensional task originally proposed by Scheme [12]. Briefly, subjects were instructed to move the ring-shaped cursor into an annulus-shaped target as quickly as possible. Wrist flexion/extension controlled horizontal ring displacement, supination/pronation controlled vertical ring displacement, and hand open/closed controlled ring radius. Trial success required subjects to dwell within the target for 2 s. If the subject did not acquire the target in 30 s or overshoot (crossed the target boundary) more than 5 times, the trial was unsuccessful. Subjects were not explicitly instructed to either use or not use simultaneous control to complete the task.

Three levels of target complexity were presented with equal frequency: “1-DOF targets” required use of only one DOF, “2-DOF targets” required use of two DOFs, and “3-DOF targets” required use of three DOFs to successfully complete the task. Targets were presented with various combinations of target distances and annulus widths.

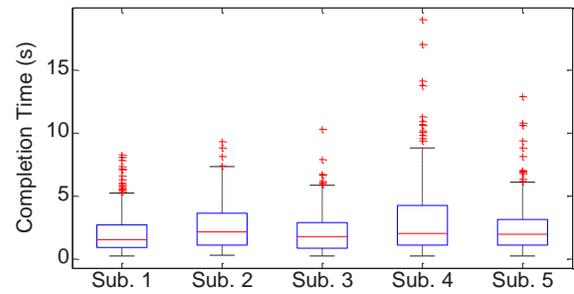


Figure 2: Box plots of completion times (dwelt time not included) for each subject as they completed the ring task. Median completion time was less than 2.2 s for each subject. Distributions were positively skewed, such that most targets were reached in less than 5 s. Timeout penalty was 30 s.

Data recorded included target acquisition success rates, task completion times, and the degree to which simultaneous control was used during a trial. Data collection occurred in six blocks, separated by rest periods. Each block included 54 trials with a new target for each trial.

## RESULTS

Minimal cross talk was present, as the average peak cross-correlation for each combination of EMG signals was  $1.4\% \pm 0.24\%$  (SEM). For each subject, parallel dual-site control was successfully configured within 30 minutes.

On average, subjects were able to successfully complete 99.9% of all trials without timeout or overshoot penalty. The distribution of completion times for each subject is shown in Figure 2. For each subject, the median completion time (dwelt time not included) was less than 2.2 s. The distribution of completion times was positively skewed, such that most targets were reached in less than 5 s. Figure 3 demonstrates a random sample of paths taken from subjects for 1-, 2-, and 3- DOF targets. Off-axis deviations for the 1-DOF targets suggested that subject occasionally experienced unintentional activation of additional DOFs when trying to isolate one DOF. Many of the trajectories used by subjects to acquire 2- and 3-DOF targets are curved (Figure 3), indicating that subjects voluntarily chose to use simultaneous control of the DOFs during the task. Sequential activation of DOFs instead results in blocky, orthogonal trajectories.

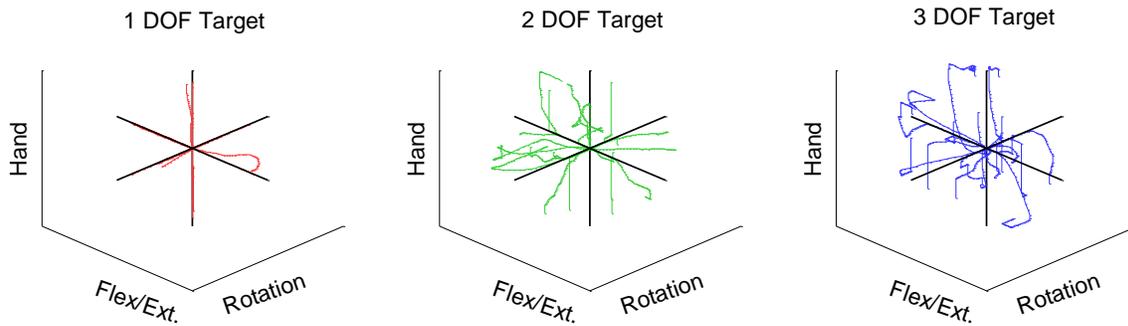


Figure 3: Sample target acquisition paths. Though subjects worked in a two-dimensional environment, the task was three-dimensional. Therefore, the targets and subjects' paths can also be considered in three-dimensional space. Random samples of three-dimensional paths from each subject are shown. The presence of curved trajectories for 2- and 3-DOF targets demonstrates the voluntary simultaneous control of DOFs.

Figure 4 shows in greater detail how subjects used simultaneous control during trials that they completed quickly (Fig. 4A, completion times < 25% percentile) and during trials that took a long time to complete (Fig. 4B, completion times > 75% percentile). In both cases, subjects primarily used one DOF to acquire 1-DOF targets, but also exhibited some simultaneous activation of two DOFs. For 1-DOF targets that took longer to complete, subjects experienced more simultaneous activation of two DOFs (Fig. 4B). For 2-DOF and 3-DOF targets, subjects generally started with higher levels of 1-DOF activity, and gradually introduced simultaneous control of 2- or 3-DOFs during the middle of the trials. Towards the end of the trial, subjects increased the use of sequential 1-DOF control. Of note, subjects used greater levels of 1-DOF control at the end of 2- and 3-DOF target trials with longer completion times (Fig. 4B) than shorter completion times (Fig. 4A).

## DISCUSSION

Multiple previous studies have proposed that intramuscular EMG recorded from implantable myoelectric recording devices be used in a parallel dual-site myoelectric control scheme [7-10]. The results of this study indicate that such a myoelectric control system has the potential to provide simultaneous and proportional control of two wrist DOFs and a hand DOF in subjects with intact forearm musculature (e.g. able-bodied subjects) and thus potentially in transradial amputees. In a virtual reality task, able-bodied subjects were able to quickly acquire targets (Fig. 2) with high success rates. The control provided by this approach differs from most previous investigations of simultaneous control, which either (i) were limited to wrist control without a hand [5] or finger control without a wrist [10], (ii) did not provide independent control of each DOF [3], or (iii) were limited to offline evaluations [4].

When using the parallel dual-site control system, subjects used simultaneous control to acquire targets without prompting to do so. Previous studies regarding the voluntary use of simultaneous control during EMG-based tasks have varied. Williams and Kirsch implemented a system similar to parallel dual-site control with head/neck surface EMG in patients with tetraplegia [13], but found that subjects did not control DOFs simultaneously. In contrast, Birdwell showed that able-bodied subjects using parallel dual-site control with intramuscular EMG from extrinsic finger muscles controlled two of three DOFs simultaneously [10]. Subjects in the current study occasionally controlled all three available DOFs simultaneously (Fig. 4). The subjects of this study performed over 600% more trials than in [10], suggesting that the amount of practice, ease of activation, and/or the physiological relevance of the muscles used may influence the choice to use simultaneous control.

During this study, subjects used simultaneous control in a pattern similar to that reported by Birdwell [10] (Fig. 4). For 2- and 3-DOF targets, subjects were likely to start by activating only one DOF, and later introduced additional DOFs. The use of simultaneous control then decreased towards the end of the trial, as subjects were more likely to be making fine, corrective movements. The difference in simultaneous control profiles in trials that were completed quickly (Fig. 4A) versus trials that took long times to complete (Fig. 4B) suggests that subjects may alter their strategy for using simultaneous control depending on the difficulty of the target. Substantially greater levels of sequential 1-DOF control was present towards the end of trials with >75% percentile completion times, suggesting that subjects may be more likely to revert to sequential control in targets that they find more difficult.

One possible limitation of the parallel dual-site approach might be difficulty in isolating a single DOF when desired. In this study, subjects occasionally produced unintended activation of additional DOFs, seen in the off-axis deviations in Fig. 3 and the presence of simultaneous DOF activity during 1-DOF target trials (Fig. 4). Given the lack of cross talk, this was likely caused by the inability of subjects to isolate individual muscle contractions. Similar difficulty was described in a transradial amputee using parallel dual-site control [8], but was reported to improve with practice over multiple weeks. More extensive studies to examine the process of learning simultaneous control over multiple experimental sessions should be considered in the future. This study was also limited to evaluation in able-bodied subjects, and future studies should extend this evaluation to subjects with transradial amputation.

### ACKNOWLEDGEMENTS

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## SURFACE VERSUS INTRAMUSCULAR EMG FOR LINEAR REGRESSION-BASED MYOELECTRIC CONTROL OF THE WRIST AND HAND

Lauren H. Smith<sup>1,2</sup> and Levi J. Hargrove<sup>1,2</sup>

<sup>1</sup>Northwestern University, Chicago IL, <sup>2</sup>Rehabilitation Institute of Chicago, Chicago, IL

### ABSTRACT

Clinically available methods of myoelectric prosthesis control are limited to sequential control of degrees of freedom (DOFs). Users are unable to simultaneously and independently control multiple DOFs. Linear regression-based myoelectric control systems have the potential to provide simultaneous control without requiring extensive training data sets that include all possible combinations of DOFs. The objective of this preliminary study was to evaluate linear-regression-based myoelectric control systems for the simultaneous control of the wrist and hand, and to compare performance between using surface electromyogram (EMG) and intramuscular EMG signals. Two able-bodied subjects participated in an experiment evaluating real-time controllability in a virtual Fitts' Law task. Completion rates for the Fitts' law task were 100% for each signal source. Throughput and path efficiency were lower using surface EMG than intramuscular EMG for all target types. Offline analysis of prediction accuracy indicated that prediction of supination was substantially lower when using surface EMG than when using intramuscular EMG, possibly contributing to the difference in real-time performance. Accuracy of other movements (pronation, wrist flexion/extension, and hand open/close) was only slightly decreased when using surface EMG.

### INTRODUCTION

Myoelectric control methods that are currently available to patients are not able to provide simultaneous, independent control of multiple degrees of freedom (DOFs). Both the conventional dual-site differential ("direct control") [1] and commercially available pattern recognition [2] methods enforce that users control each DOF sequentially. A large focus of the myoelectric control literature has thus focused on new control methods that would allow users to control each DOF independently and simultaneously. A variety of approaches have been investigated, including pattern recognition [3], neural networks [4], synergy-based blind source separation [5], and linear / nonlinear regression [6].

Many of the control systems listed above have used extensive training data sets, requiring representative examples of all possible combinations of DOF activities [3, 4]. However, both the synergy-based and linear regression approaches assume that the features of EMG activity (typically EMG amplitude) during simultaneous DOF control are linear combinations of the features present during sequential control of single DOFs. Under this assumption, systems can be trained using electromyogram (EMG) signals from single-DOF movements, and are expected to generalize to control multiple DOFs simultaneously. Indeed, such approaches have shown promise for control of two DOFs at the wrist in both offline [5, 6] and online analyses [5]. However, neither approach has previously demonstrated control of a wrist and hand, which, in addition to increasing the number of DOFs, would rely on contraction of more muscles from varying depths in the forearm.

Furthermore, no previous studies have compared use of surface EMG (sEMG) to targeted intramuscular EMG (imEMG) for a linear regression control system. Though previous studies have shown no difference between the two signal sources for sequential DOF control using pattern recognition [7], similar comparisons have not been made for simultaneous control or when EMG amplitude is the primary feature used. Though sEMG is non-invasive and clinically realizable at present, the presence of crosstalk and weaker signal intensity from deeper muscles may affect the accuracy of methods using only EMG amplitude features. Given the current development of chronic wireless implantable recording devices [8, 9], a comparison with intramuscular EMG, which provides access to deeper muscles with little crosstalk, is needed.

The objective of this preliminary study was to investigate the potential for a linear regression control system to provide simultaneous myoelectric control of a wrist and hand, when trained with single-DOF training data. The controllability of this system was evaluated as subjects used either sEMG or fine wire imEMG to contrast performance from the two signal sources.

## METHODS

### Experimental Protocol

The following study was approved by the Northwestern University Institutional Review Board. Two able-bodied subjects participated. Subject 1 had considerable experience in myoelectric control, whereas subject 2 had less experience. The experiment consisted of two sessions. During the first session, subjects used an imEMG-regression based control system in an online virtual task. The second session was similar to the first session, except that sEMG was used. Both systems provided simultaneous control of the following three DOFs: wrist rotation, wrist flexion/extension, and hand open/close.

### EMG Acquisition

For the imEMG session, bipolar fine wire electrodes were used. Electrodes were inserted into the following six forearm muscles using hypodermic needles: *pronator teres (PT)*, *supinator (SU)*, *flexor carpi radialis (FCR)*, *extensor carpi radialis longus (ECRL)*, *flexor digitorum profundus (FDP)*, and *extensor digitorum communis (EDC)*. For the sEMG session, six bipolar adhesive Ag/AgCl electrodes were targeted to each of the six muscles. Electrodes were placed in line with fiber direction of each muscle, guided by [10]. SEMG was targeted, because preliminary findings showed that placing six electrodes equally along the forearm circumference did not provide usable myoelectric control systems. For each session, subjects were restrained in a custom brace to ensure isometric contractions.

ImEMG signals were collected using a Motion Lab Systems MA300 EMG system, which amplified signals 350x and band-pass filtered them between 10 Hz and 2000 Hz. SEMG signals were collected using a Delsys Bagnoli-16 with a band-pass filter of 20-450 Hz. Both signal sources were sampled at 5 kHz by a National Instruments data acquisition system. Signals were digitally high-pass filtered with a 3<sup>rd</sup> order Butterworth filter with a cut-off frequency of 20 Hz. The mean absolute value (MAV) of each EMG channel ( $m_{muscle}$ ) was calculated from 250 ms sliding windows, with a 50 ms frame shift. The vector of MAVs from each of the six muscles,  $m$ , was used as input into the regression models of the myoelectric control system:

$$m = [m_{PT} \quad m_{SU} \quad m_{FCR} \quad m_{ECRL} \quad m_{FDP} \quad m_{EDC}]^T \quad (1)$$

### Myoelectric Control System

Linear regression models were used to predict the user's intended movement and velocity at each DOF from the vector of MAVs,  $m$ . A model was trained for each movement direction in a degree of freedom (e.g. a model for supination and a model for pronation), resulting in six total regression models to control three degrees of freedom:

$$V_i = A_i m + b_i \quad (2)$$

where  $i$  was one of six movement types (supination, pronation, wrist flexion, wrist extension, hand open, hand closed),  $V_i$  was the predicted velocity for the movement as a percentage of maximum,  $A_i$  was the vector of multiplicative coefficients for each MAV, and  $b_i$  was the constant offset.

To train the regression models, subjects provided graded contractions in response to visual prompts. Each 20 s training session used the prompt in Figure 1. Subjects provided two training session for each of the motion types. Training contractions were mapped to velocities ranging from 0-75% of the maximum, where the visual prompt during the contraction was used as the training data label. Training sessions for each of the six motion types was used to train each regression model. Training data from sessions not corresponding to the model's motion type was therefore labelled as 0% velocity. The coefficient vector  $A_i$  and offset  $b_i$  that minimized the sum-squared error were found for each motion type,  $i$ , using QR decomposition [11]. For each model, predicted velocities below 10% of the maximum were forced to zero to decrease small, unwanted movements at rest. The movement direction and velocity at a DOF during real-time testing was given by the difference between the predictions of the two corresponding regression models.

### Offline Evaluation

The prediction accuracy of each of the linear regression models was evaluated using a six-fold cross-validation. Each training session was divided into three parts, so that each fold of the cross-validation was a single training ramp. Prediction accuracy was quantified using  $R^2$ , calculated by:

$$R_i^2 = 1 - \frac{\sum (V_i(t) - A_i m(t) - b_i)^2}{\sum (V_i(t) - \bar{V}_i)^2} \quad (3)$$

where  $V_i(t)$  was motion type  $i$ 's prompted velocity at time  $t$ ,  $m_i(t)$  was the vector of MAVs at time  $t$ , and  $\bar{V}_i$  was the average value of the prompt.

### Online Evaluation

For each signal source, regression-based control was trained for real-time evaluation. Subject 1 trained the systems with two repetitions of each training session, whereas Subject 2 trained the systems with one repetition. Real-time controllability was evaluated using a pseudo-three dimensional Fitts' law task [12]. Subjects controlled the translation and radius of a ring cursor using three DOFs:

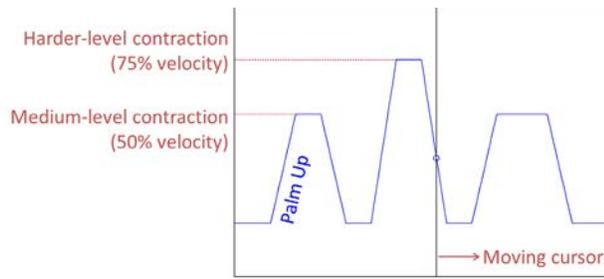


Figure 1. Visual prompt for training. A visual prompt was used to provide training data for each of the six movement types. A cursor moved along a trajectory with three ramps, prompting subjects to increase and decrease the intensity of their contraction accordingly. The first peak represented the level of contraction that would map to 50% of maximum velocity. The second peak mapped to 75% of maximum velocity. The EMG data was regressed onto the visual prompt to find the coefficients and constants of Eq. 2.

wrist flexion/extension controlled horizontal movement, supination/pronation controlled vertical movement, and hand open/close controlled the radius of the cursor.

Three levels of target complexity were presented with equal frequency: “1-DOF targets” required use of only one DOF, “2-DOF targets” required use of two DOFs, and “3-DOF targets” required use of three DOFs to successfully complete the task. Targets were presented with various combinations of target distances and annulus widths, resulting in indices of difficulty, as defined in [13], from 1.22 to 3.17. Each combination of target distance, width, and DOF-complexity were provided to the subjects 3-12 times. Successful acquisition of the target required a dwell time of 2 s. Subjects had 30 s to complete each trial. An overshoot penalty ended trials if subjects entered and exited the target five times. Throughput and path efficiency, as defined in [13] were used to quantify performance.

## RESULTS

### Offline Analysis

For each motion type, average  $R^2$  values were lower when subjects used sEMG signals compared to imEMG (Figure 2). For all motion types except supination, differences between the two signals sources were small. However, there was substantial difference the prediction accuracy for supination (average  $R^2 = 0.18 \pm 0.12$  SEM for sEMG and average  $R^2 = .80 \pm 0.07$  for imEMG). Figure 3a compares the prediction of intended supination between sEMG and imEMG for Subject 2. Each graph represents one of the six predictions averaged in the cross-validated analysis. When sEMG was used, the linear regression model frequently predicted supination

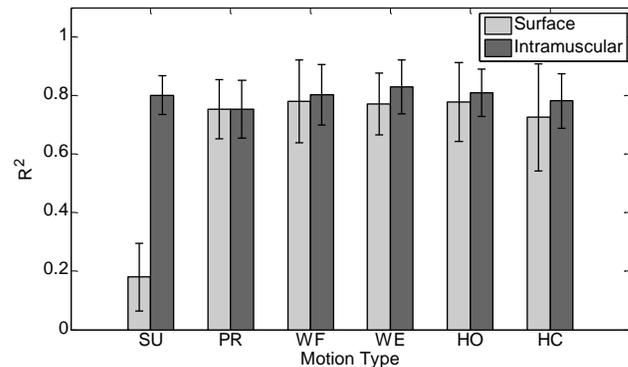


Figure 2. Cross-validated performance of linear regression models by motion type. Error bars represent standard error. SEMG provided substantially worse prediction of intended supination compared to imEMG.

activity when the subject was providing training contractions for other motion types ( $R^2 = -0.21$ ). However, when imEMG was used, the predicted supination activity matched the prompted supination more closely ( $R^2 = 0.88$ ). Figure 3b shows, in contrast, the prediction of wrist flexion between sEMG and imEMG for Subject 1. For this motion type, both sEMG and imEMG were able to predict the prompted wrist flexion activity ( $R^2 = 0.95$  for sEMG,  $R^2 = 0.96$  for imEMG)

### Online Analysis

During the online Fitts' Law task, subjects were able to successfully acquire 100% of targets when using either sEMG or imEMG. Average throughput was lower when using sEMG ( $0.97 \pm 0.31$  bits/s) compared to imEMG ( $1.34 \pm 0.15$  bit/s). Greater differences in throughput were observed for targets requiring more DOFs to successfully finish the task (Figure 4a). Similarly, average path efficiency was lower when using sEMG ( $63.5\% \pm 1.2\%$ ) compared to imEMG ( $77.7\% \pm 4.8$ ). Greater differences in path efficiency were also observed for targets requiring more DOFs to successfully finish the task (Figure 4b). Subject 1's path efficiencies when using imEMG were 98.9% for 1-DOF, 79.0% for 2-DOF, and 70.0% for 3-DOF targets, which was greater than the highest path efficiency possible if the subject had only been allowed sequential movements (which would have been 100%, 73.6%, and 64.1% for 1-, 2-, and 3-DOF movements, respectively).

## DISCUSSION

The results of this study suggest that a linear regression-based myoelectric control system has potential to provide simultaneous control of the wrist and hand in subjects with intact forearm musculature. Able-bodied subjects were able to successfully use linear regression-based control to complete a Fitts' Law style task. Subjects successfully trained such

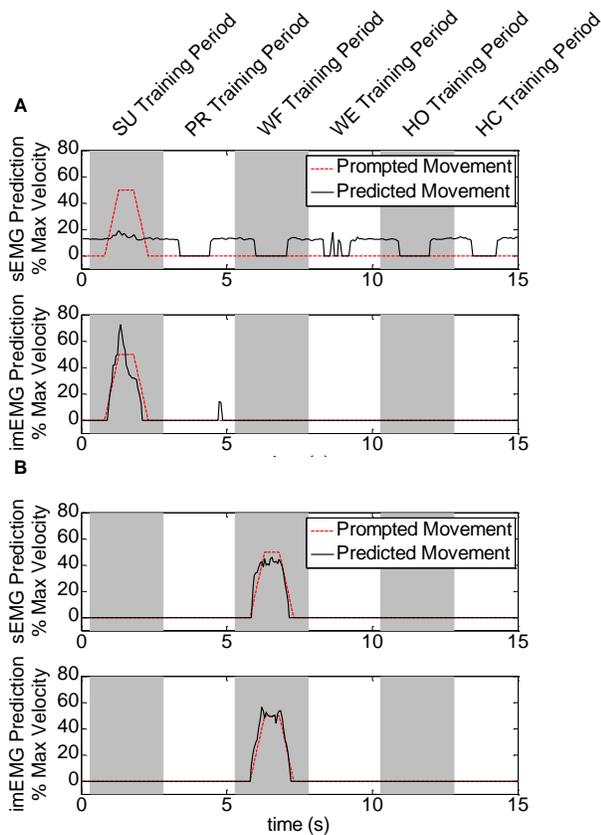


Figure 3. Example estimations of intended velocity from one fold of the cross-validation analysis. (A) Linear regression model predicting supination for Subject 2. SEMG resulted in poor prediction of intended supination. (B) Linear regression predicting wrist flexion for Subject 1. Both sEMG and imEMG predicted intended flexion well.

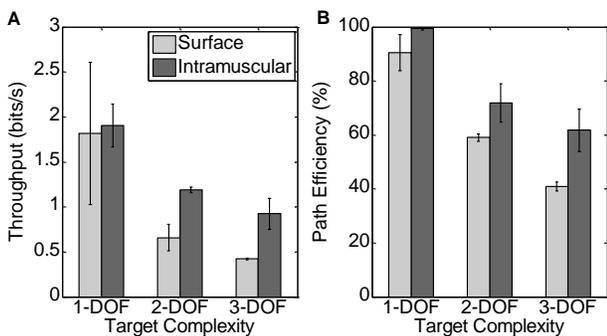


Figure 4. Real-time performance during the Fitts' Law test. (A) Average throughput was less when using sEMG for all three target complexities. (B) Average

systems using data collected from 1-DOF motions only, which were labelled using the visual prompt provided to the subject. The training for this approach required less than 5 min for each subject, and did not require use of additional equipment to measure joint kinematics or torques. The control provided by this system is promising; the more experienced subject using imEMG produced path efficiencies greater than the maximum possible when using sequential control of DOFs.

The results of this pilot study also highlight potential difficulties in using sEMG for such a control system, particularly when targeting deep muscles. Subjects performed worse in the Fitts' Law task when using sEMG than imEMG (Figure 4). Offline analysis suggests that prediction of supination may be largely responsible for this difference in controllability (Figures 2-3). Difficulty in predicting supination from sEMG MAVs may be explained by the relative depth of the supinator compared to other muscles in the forearm, as crosstalk from more superficial muscles may mask sEMG from the supinator. The ability to predict supination may be sensitive to the location and orientation of the sEMG electrodes targeting the supinator, and therefore future work may consider the use of high-density electrode arrays [6] and/or electrode grids [14]. A linear regression system, where each linear model is trained on EMG data from all possible 1-DOF movements, may identify electrode orientations that are highly predictive of supination but not correlated to other motion types, thereby decreasing the possibility of unintended co-activation of DOFs. Control of supination could also benefit from placing electrodes externally over the biceps or intramuscularly.

This work is limited by the small sample size and the use of able-bodied subjects. Future work should also extend these results to persons with transradial amputations.

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## REDEFINING THE NORM: A PROVEN TEAM APPROACH TO FITTING AND TRAINING PROCEDURES FOR LONG DISTANCE AND INTERNATIONAL PATIENTS WITH UPPER LIMB LOSS: A CASE STUDY

Ryan Spill, CP; Chris Bollinger, OTR(L); John Miguelez, CPO, FAAOP; Dan Conyers, CPO, FAAOP  
*Advanced Arm Dynamics, Inc.*

### ABSTRACT

Successful upper limb amputee rehabilitation requires the communication and execution of all members of the allied health team. The physician, nurse case manager, occupational therapist, psychotherapist, and the prosthetist must collaborate in order to maximize the patient's functional potential. Communication is essential between the professional team and the patient/family. As upper limb amputations occur less than lower limb amputations (1:30 ratio), patients with upper limb loss/absence travel to work with specialists. This case study follows the case of such a patient. A 12 year old female and her family traveled from Brazil to the United States for upper limb rehabilitation. Due to the distance traveled, the visit was four weeks in length, and four distinct phases of rehabilitation occurred during this visit. Weekly communication between the occupational therapist, prosthetist, and the vascular surgeon occurred prior to the amputation, and daily communication occurred between the therapist, prosthetist, and the patient/family to set and modify goals during the changes in progress. A well designed and clearly communicated plan allowed all parties to check in on a routine basis and become the cohesive platform for ongoing changes or updates that were and will be required secondary to growth, wear, or mechanical maintenance needs.

### INTRODUCTION

Typically, upper limb prosthetic fittings occur over many days within a window of several weeks of intermittent appointments. More and more in our global economy, an individual with upper limb loss travels for specialized prosthetic and therapeutic training from distances extensive enough to require extended lodging; for example, an international location. The rehabilitation team must address these challenges in an efficient, cost effective manner, while providing quality comprehensive care in an expedited window of patient availability. Furthermore, specific focus must be brought to bear on providing the patient with the tools to successfully continue their rehabilitation at home, and to plan in advance for follow up care and return visits. A successful prosthetic fitting with occupational therapy

requires several phases, and the clinicians must mobilize conscientiously to best assist these patients in reaching their goals prior to their departure. In order for the team to address these needs, the prosthetist and occupational therapist must have all necessary tools to accomplish the expedited fitting, training, fabrication, and planning for the meeting of the patient's short term and long term goals. For the prosthetist, not only are the materials for the prosthesis necessary to have on site, but a detailed initial assessment must be provided to determine the appropriate components for use in the prosthesis. For the therapist, the proper age-appropriate tools and activities must be obtained for therapeutic training. This is a case study of one such patient, a 12 year old female from Brazil, who traveled to the US specifically for such treatment.



### HISTORY

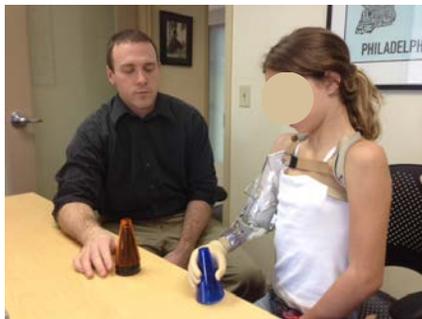
The patient was born with a vascular malformation presenting with several small plane hemangiomas in all segments with the largest one located at the right hand and forearm, with a significantly larger and deeper volume than the left, sound side. Secondary to angio magnetic resonance

imaging exams being performed, the formation of many arteriovenous vascular and capillary malformations were observed with fistula communication within the right segment including the enlargement of the right brachial arterial caliper. During 2011, this young patient presented with ulceration wounds on the dorsal surface of her right hand, second finger, making control difficult and, many times, dramatic bleeding occurred prompting several hospitalizations and surgical interventions. A variety of wound management modalities were attempted without success, including skin grafts and plastic surgeries. The final diagnosis was Parkes-Weber Syndrome, or vascular malformation of the right upper limb (congenital hemangioma), with right forearm and hand open wounds on a 12 year old female.

This young woman's right upper limb was non functional for 12 years and the ulcerations that presented on the limb were not able to be treated successfully. Amputation was elected proximal to the right elbow joint, approximately 1cm proximal of the humeral epicondyles.

### CASE STUDY

The patient is a 12 y/o female from Brazil. Remote communication was initiated with the patient's family prior to their arrival to the Center. These extensive, pre-treatment conversations centered around the patient's current presentation prior to the amputation, limitations of the right upper extremity, and goals for the prosthetic fitting and training. Communication between the team and the surgeon was initiated prior to the amputation to discuss prosthetic components available for various levels of amputation. Given that the patient's mother is a physiotherapist and the father is a physician, it was important to the parents to begin the rehabilitation process immediately following physical healing of the amputation. In this case, the care began six weeks after the amputation was performed.



The initial plan for the four week, expedited fitting and training process was divided into the following four treatment phases: Phase 1: In-person initial assessment with prosthetic component determination and pre-prosthetic therapeutic training [1 day]; Phase 2: Impressions/test socket

fitting/preparatory fabrication with components with initial repetitive prosthetic training [2 days]; Phase 3: Preparatory prosthesis fitting with therapeutic functional task training [10 days]; and Phase 4: Definitive prosthesis fabrication/use with advanced therapeutic training for ADLs, recreational applications and community re-integration [7 days].

During phase 1 of treatment, the initial patient evaluation and assessment were performed by the prosthetist and the occupational therapist. The patient presented with a right long transhumeral residual limb with a well healed scar. Two 5mm scabs were present, but were not hypersensitive. The residual limb was 2cm shorter than the corresponding contralateral arm segment. An aneurysm was removed from the arm segment during the amputation procedure, and this area was sensitive over the medial aspect of the residuum. The left upper limb was intact; range of motion and strength were within normal limits. Phantom pain was present at night, and was medically treated using Amitriptyline. The patient was in the process of weaning from her pain medication. Phantom sensation of the right hand was present, but not noted was any sensation of the absent right forearm.

Goals that were established consisted of bimanual tasks at school and home. The patient and family were educated about activity analysis, prepositioning of the terminal device, and body mechanics to help promote the most natural prosthesis use. The patient enjoys playing soccer, which she did several times per week prior to the amputation surgery. She also had artistic interests, including drawing and painting. Utilizing her mobile phone for text messaging with her friends was of utmost importance. The patient was already connected with outpatient psychotherapy and had a good rapport with her family, according to medical records from Brazil. The patient had grown accustomed to using her left arm for all tasks, as her right hand was unable to provide prehension. Therapeutic training during this phase included education regarding the maintenance of joint range of motion, promoting volumetric control via compression garments, and education on adaptive equipment and other options for increased ease of functional tasks. Also during this phase, expectations were set for proper wear schedule and information was discussed regarding residual limb hygiene and prosthesis care. Regular communication with the family was initiated and maintained due to the knowledge that the patient was going through an extremely challenging process emotionally.

The prosthesis that most clearly accommodated her ADL goals was a hybrid prosthesis incorporating a suction socket with an expandable frame, a Figure of 8 harness, external locking elbow hinges, an Otto Bock size 7 DMC plus hand, an electric wrist rotator with associated controller, and internal batteries. Components were ordered immediately.

Phase 2 included the impression taking, the test socket fitting, and preparatory prosthesis fabrication. This procedure took approximately two days.

During Phase 2, while the prosthesis was in various stages of fabrication, the occupational therapist had discussions with the patient and family about residual limb hygiene, prosthetic care, and wearing schedule. EMG site training was performed with the Otto Bock PAULA hand simulator and training game. The patient demonstrated the ability to isolate two antagonistic sites at the biceps and triceps muscles. The patient was able to incorporate proportional control of the terminal device and the wrist rotator within two days.

Phase 3 involved the fitting of the preparatory prosthesis. Several harnessing techniques were attempted, as well as additional components such as an adjustable device to allow for expansion and contraction of the frame of the prosthesis and a pediatric elbow unit to provide a thinner profile for the prosthesis with friction humeral rotation. Initial programming adjustments were performed using the appropriate user interfaces.

Therapeutic training involved prepositioning of the terminal device and activity analysis with an emphasis on body mechanics and compensatory movements. Many repetitive tasks were performed during this phase such as elbow control and prepositioning, hand prepositioning via the wrist rotator, and hand operation at various speeds and grip forces. During a small number of days, the patient felt excessively tired and emotionally upset as she coped with learning how to use her prosthesis. These emotional and physical fatigue issues were addressed with an occasional partial day off as a break. Activities during this phase consisted of tying standard shoe laces with and without the prosthesis, utilization of elastic laces, buttoning shirts, making bracelets, zipping jackets, folding towels and laundry, preparing and cutting food, and artistic activities such as drawing, painting, and model-building.

Phase 4 began with the delivery of the finished, definitive hybrid prosthesis. The appropriate harness design was incorporated into the prosthesis, and all programming adjustments were finalized.

Therapeutic training involved functional tasks and community based tasks. The patient prepared several different meals, used her prosthesis for drawing and painting, and for sports at a local gymnasium. Also practiced during this phase were donning and doffing techniques, playing card/board games, and school-specific activities such as writing while holding paper and using a ruler.



## RESULTS

At the conclusion of the four week expedited fitting, the team provided a definitive prosthesis that fit appropriately with the patient trained to use it according to her desired activities of daily living. The final prosthesis was fabricated with the external locking elbow hinges instead of a pediatric elbow unit. The patient preferred the symmetry at the elbow axis with the hinges. The device that had been introduced and used for frame adjustability was deemed excessively high in profile in relation to the prosthetic frame, and ultimately was not provided.

In terms of therapeutic rehabilitation, the patient was successfully able to control the prosthetic elbow, to lock and unlock the elbow joint, to rotate the wrist, and to use the hand at various speeds and grip forces. She wore the prosthesis in public several times. Information was sent to the father regarding how to help the patient engage in activities at home in the most natural manner for all tasks

## DISCUSSION

During a four week expedited fitting of a child consisting of daily visits, regular communication between the members of the rehabilitation team as well as with the patient and family is integral. The team had conversations with the family on a daily basis throughout this process. In the beginning of the process, a conversation that took place outlined tentative plans and goals for each member of the rehabilitation team and of the patient. At the beginning of each week, the conversations outlined the tentative weekly plans, and each morning, daily plans were introduced. Conversations took place at the conclusion of each day regarding any modifications to the plans and to make sure that all questions and concerns had been addressed by the team.

It is of utmost importance that the rehabilitation team members communicate with each other, prior to meeting with the patient, to determine the treatment plan and discuss

its introduction to the patient and family. The provision of preparatory and definitive prostheses with adequate therapeutic training within a set timeline requires efficiency between the disciplines.

A clear plan was developed for ongoing therapy and prosthetic follow up prior to the patient's return to her native Brazil. This was communicated verbally and in writing to create shared understanding between the care-givers and the family. This is a crucial part of the patient's ongoing success post-fitting. A well designed and clearly communicated plan allows all parties to check in on a routine basis and become the cohesive platform for ongoing changes or updates that will be required secondary to growth, wear, or mechanical maintenance needs.

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## REDEFINING THE NORM OF THE REHABILITATION HOSPITAL EXPERIENCE FOR A HIGH LEVEL BILATERAL PATIENT

Scott Spring  
*Advanced Arm Dynamics*

### ABSTRACT

#### Introduction:

Regaining functional independence can be a challenging undertaking for the individual who has undergone high level bilateral amputations. In this case study we document how clinical norms are redefined, as the patient is closely monitored and treated by his prosthetic team while undergoing care in different settings; ranging from acute inpatient stay, outpatient prosthetic/therapeutic clinic, and an intensive rehabilitation hospital stay.

#### Methods:

Case Study Patient Information:

Age/gender: 54 y.o./male

Amputation level: Bilateral, Shoulder Disarticulation.

Comorbidities: Vision impairment, Mobility and balance issues, Decreased ROM, Wound healing.

Avocational interests: Fitness training, kayaking, working on race cars.

#### Results:

Description of Treatment

- Acute rehabilitation - Patient's initial treatment consisted of wound and scar management, full body strengthening, balance and gait training. In this setting, the patient was introduced to his outpatient prosthetic team, allowing early prosthetic assessments and education for the family.
- Outpatient Prosthetic/Therapeutic Intervention - Based on detailed evaluations, two prosthetic options were selected: External Power and Activity Specific. Occurring simultaneously with the prosthetic design, fabrication, and fitting; the patient and his spouse were thoroughly trained by the occupational therapy staff in the proper use and care of all prescribed devices [1,2].

- Inpatient stay at Rehabilitation Hospital - Prior to admittance for a comprehensive inpatient stay, the prosthetic team travelled to meet with key rehab staff members. All known aspects of patient's case were relayed via a PowerPoint presentation and Q&A session. Once patient was admitted, he was followed by the same prosthetist who could then interact with patient and hospital OT/PT staff on a daily basis.
- Providing prosthetic fitting and intensive therapeutic training prior to admission into the rehabilitation hospital gave the patient and family a positive head start on the long days and rigorous therapy sessions experienced inside the hospital setting.

#### Conclusions:

It has been our experience that rehab outcomes are greatly enhanced when patients are prepared, and then followed by a dedicated prosthetic and therapy team to provide continuity and to help with advanced problem solving and troubleshooting. These teams should be seen as collaborative partners, as they assist the individual and associated health care workers during the transitions between phases of rehab.

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## UPPER LIMB PROSTHETIC COMPETENCY AND CHARACTERISTICS AMONG SELF-ASSESSED NOVICES-INTERMEDIATES AND EXPERTS-SPECIALISTS

Gerald Stark, MSEM, CPO/L, FAAOP  
Ottobock Healthcare  
Minneapolis, Minnesota

### SUMMARY

The paper compares survey results between self-identified Expert-Specialist (E-S) groups and Novice-Intermediate (N-I) groups as well as practitioners in Privately-Owned prosthetic clinics compared to those in Institutional/Corporate Settings to understand differences in Upper Limb proficiency.

### INTRODUCTION

Due to the decline of self-perceived competence in upper limb in the US, many prosthetists are not providing care personally, but relying on remote visits from experts or specialists. While this service allows the practitioner to concentrate on higher volume prosthetic care, it generally decreases the effectiveness of the average practitioner to provide optimized upper limb care to increase patient acceptance. Although this has created a greater demand for experts and specialists especially for the initial fitting, a majority of prosthetists are increasingly unable to adequately serve the long term needs of the upper limb patient. There appears to be a number of fundamental reasons for this “learned helplessness” or lack of clinical self-efficacy with respect to approaches toward upper limb prosthetic care. These difference between the various experiential and work setting groups included attitudes toward innovation, componentry, interface design, consultation, and patient variability.:

### METHODS

An initial telephone interview of five practitioners from a variety of settings and experiences was conducted to provide a better understanding of the broad areas of concern most pertinent for a broader survey. The main areas identified were low volume, financial risk, access to training, and level of difficulty. A ten question on-line survey was then developed that was posted on a third-party survey administration website for over a month from March 13, 2013 to May 19, 2013. The survey had 152 respondents with 149 who completed the entire 10 question survey. The group self-assessed themselves as 2.0% non-providers, 22.8% Novices, 49.0%

Intermediates, 12.8% Experts, and 13.4% Specialists. These groups were compared as mutually exclusive dichotomous groups of the Novice-Intermediate (N-I) group with 71.8%, and Expert-Specialist (E-S) group with 26.2%. Also the groups were further subdivided into Institutional-Corporate (I-C) of 37.5% and Privately-Owned Clinic (P-O) at 59.7%.

### RESULTS

In the N-I group 62.1% were at private clinics, but only 44.4% in the E-S group. The E-S group saw 24 patients per year while the N-I group saw 3 patients per year. The number of external collaborators was different with the N-I group at 1.76, but the E-S group had 3.42. Those in the E-S and I-C group had 4.85 external linkages. Greater numbers of the E-S group chose “Innovator” statements at 44% with the N-I group with 29%. The N-I group had a higher number of Laggard responses with 28% while the E-S group had only 2.7%. The distribution of the “Reasons for General Lack of Confidence” was “Too few patients” for the N-I group, but “Personal Confidence” and “Materials” was slightly higher for the E-I group. The E-I group was more neutral about asking for help at 2.12 than the N-I group at 1.80. The E-I group was more confident when “Approaching new Upper Limb Projects” at 4.75 while the N-I group indicated they were at 4.01. Also the N-I group indicated agreed they “Were not up to date on External Power” with a 3.36 rating while the E-I group disagreed at 2.36. The E-S group felt that “Socket Design” and “Patient Variation” were more important than the N-I group who felt “Component Design” was critical. The E-S group disagreed that “Body Power is Outdated” at 1.39, while the N-I group indicated less disagreement at 2.03. The E-S group felt slightly “More Innovative” by nature, but the N-I group felt “patient experience” and “expert interaction” were the reason for expertise.

### CONCLUSION

In general the survey did seem to verify the existence of a greater lack of clinical self-efficacy from Novices-Intermediates than Experts-Specialist groups with respect to patient volume, clinical experience, and expert instruction. Differences seemed to be found in numbers of external

heterophilic linkages, attitudes toward innovation, componentry, innate innovativeness, interface design, patient training and variation. Additional differences appear to be present between Privately-owned Corporate-Institutional settings with respect to financial risk and contextual learning. Additional statistical examination is required to examine the various sub-groups and determine the level of correlation between them. The delineating factors for proficiency in Upper Limb seem to be a higher number of external linkages, more clinical experiences, greater confidence with external power, proficiency with interface design, innovative attitude, and ability to address individual needs and training.

### FIGURE AND TABLES

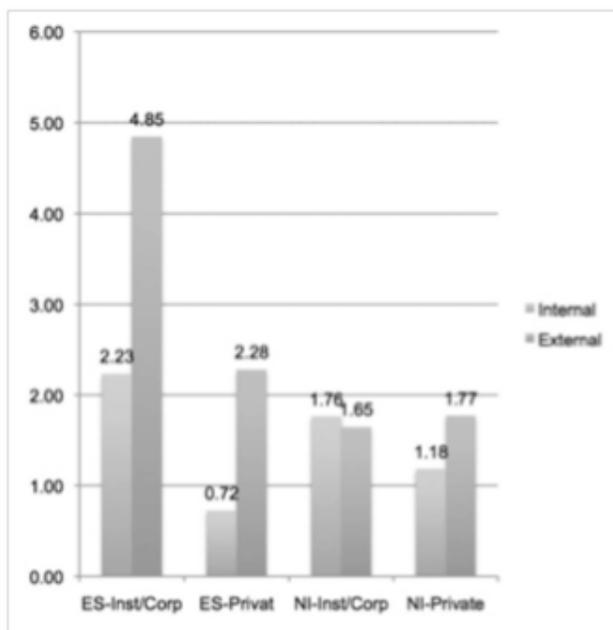


Figure 1: Number of Internal and External Linkages for Novices-Intermediates and Specialists-Experts by Institution

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## DEVELOPMENT OF A HYBRID BODY POWERED TRANSRADIAL PROSTHESIS WITH MYOELECTRIC SWITCHING

Michael Stobbe<sup>1</sup>, Michael Rory Dawson<sup>1</sup>, Jacqueline S. Hebert<sup>2</sup>  
*<sup>1</sup>Glenrose Rehabilitation Hospital, Alberta Health Services; <sup>2</sup>University of Alberta*

### ABSTRACT

This report illustrates a new myoelectric control mechanism for controlling wrist locking in a body powered prosthesis. A 62 year old full time body powered prosthetic user with right transradial and left transhumeral amputations 10 years prior was having difficulty with wrist flexion/extension and pronation/supination on an N-Abler V Type A Wrist (Texas Assistive Devices, Inc.) with right arm extended. Traditional control requires the patient to use bicipital abduction to create tension on the cable, which opens the hook, flexes the wrist or supinates the wrist. Switching between functions is achieved via mechanical bump switches that the patient can bump against the side of their body or exterior surfaces. These switches pull on strings of the wrist unit to lock or unlock the various wrist functions. A disadvantage to this setup is that the patient cannot lock or unlock their wrist functions while their prosthesis is fully extended which is desirable for some tasks such as operating a skidsteer loader. A hybrid myoelectric/body powered system was designed to allow a bilateral upper limb amputee patient to use the muscle signals of their transradial residual limb to myoelectrically lock and unlock the wrist functions while still controlling the degrees of freedom directly with their body powered cable. A Varigrip III programmable controller and battery system (Liberating Technologies, Inc.) were setup such that when one of the patient's EMG signals exceeded threshold a PQ12-63-6-S linear actuator (Firgelli Technologies, Inc.) retracted and pulled on the string to lock or unlock one of the wrist functions. When the patient relaxed their EMG signal below threshold the linear actuator was setup to automatically extend back to its original position. In order to reduce power consumption linear actuators were used that mechanically lock and do not draw power in their fully extended/retracted positions. In order to allow both wrist functions to be locked/unlocked two linear actuators and EMG amplifiers were required. A custom 3d-printed tube mount was designed in order to closely align the strings of wrist unit with the shafts of the linear actuators and allow the entire system to be easily integrated into the forearm of the prosthesis. The hybrid system was successfully fit onto the bilateral upper limb patient and has been in use for 6

months. The patient noted improved function with operating the loader.

## RECOMMENDATIONS FOR THERAPY AFTER TARGETED MUSCLE REINNERVATION

Agnes Sturma<sup>1</sup>, Birgit Bischof<sup>2</sup>, Malvina Herceg<sup>3</sup>, Tanja Stamm<sup>4</sup>, Veronika Fialka-Moser<sup>3</sup>,  
Robert Wakolbinger<sup>1</sup> and Oskar Aszmann<sup>1</sup>

<sup>1</sup>*Christian Doppler Laboratory for Restoration of Extremity Function, Medical University of Vienna, Austria*

<sup>2</sup>*Otto Bock Health Care Company, Vienna, Austria*

<sup>3</sup>*Department of Physical Medicine and Rehabilitation, Medical University of Vienna, Austria*

<sup>4</sup>*Department of Rheumatology, Medical University of Vienna, Austria*

### INTRODUCTION

Basic myoelectric prostheses for upper limb amputees are usually operated by using two myosignals generated from muscles via surface electrodes. Although this allows a fairly good prosthetic function, it is still limited to few degrees of freedom, in speed of combined movements and does not allow intuitive control over the device. In the past few years new possibilities have been found to improve prosthetic control. A special kind of selective nerve transfer 'Targeted Muscle Reinnervation' (TMR) provides the amputee with up to six signals for intuitive prosthetic control. In surgery, the residual nerves from the amputated limb are transferred onto alternative muscle groups that are not functional anymore since they are no longer attached to the missing arm. After nerve regeneration and a long-lasting phase of motor learning the reinnervated muscles serve as biological amplifiers for the motor command to the prosthetic arm. TMR thus provides physiologically appropriate EMG control signals that are related to previous functions of the lost arm. [1,2]

After surgery and nerve healing, the patients have to discover a new neuromuscular interface and learn how to control their prostheses. Because of the complexity of motor learning a long term rehabilitation program is needed. [3] Some important points were described by Stubblefield in 2009 [4] and 2014. [5] In clinical practice in the Christian Doppler Laboratory for Restoration of Extremity Function in Vienna, Austria some of those recommendations were implemented, but also some other suggestions in literature were considered as important. The aim was to have a treatment concept fitting to the present technological possibilities in clinical settings. This did not include the use of pattern recognition or implantable electrodes [6], since they are only used in laboratory settings so far.

To guide the patient through the rehabilitation process the purpose of this study was to develop a protocol for therapy in TMR patients incorporating both current evidence and clinical experience of experts in the field.

### METHODS

To develop recommendations for therapy after TMR involving a maximum number of experts in this topic, a modified Delphi method was used. A review of documents covering clinical practice recommendations and knowledge of TMR, but also general rehabilitation in upper limb prosthetics and selective nerve transfers was conducted. Those findings were discussed in an expert group including two physical therapists (A.S and B.B) and a medical doctor as a specialist in physical medicine and rehabilitation (M.H). A set of recommendations was drafted in face-to-face discussion and distributed via email, to all participants for comments. This included two other medical doctors (O.A and R.W) with experience in the surgical procedure, an occupational therapist, another physical therapist, a prosthetist and a technician, all experts in the field of TMR. Additionally they were asked to rate the different items of the concept according to their importance from 0 (not important at all) to 10 (of utmost importance). The comments received were built into a shorter version of a therapy concept that was again circulated via e-mail. Based on the feedback the final paper of recommendations after TMR was written.

### RESULTS

Nine experts participated in the Delphi exercise. They had a working experience with TMR patients between 2 to 7 years. Within the process a rehabilitation concept for the TechNeuroRehabilitation in TMR patients was developed. The rehabilitation process can be divided into one stage before surgery and 4 stages following surgery as shown in figure 1. Since it takes up to 2 years and involves complex motor relearning it requires both a motivated patient and a multidisciplinary centre with dedicated and skilled team members.

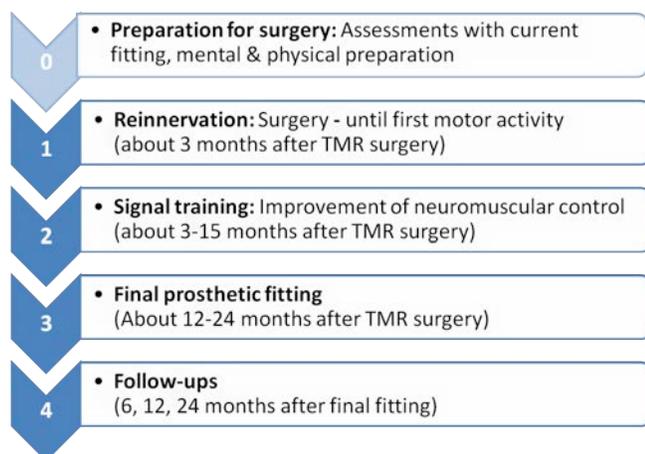


Fig. 1: The 5 stages of TechNeuroRehabilitation

#### Stage 0: Preparation for surgery

Before surgery, there are three important points to be addressed in therapy: patient education concerning the surgical and rehabilitation procedure as well as possible outcomes after TMR prosthetic fitting, assessments of the current level of functional status and physical preparation for surgery.

Patient education is used to inform the patient about the long-term rehabilitation process after TMR surgery which might take up to 2 years and to avoid unrealistic expectations concerning prosthetic function. Also other possibilities for prosthetic fittings (basic myoelectric prostheses, body-powered devices or cosmetic prosthesis) should be presented and discussed with the patient. A TMR fitting is only considered if the patient has a personal desire or need of the improved function in everyday life and is therefore self-motivated and proactive.

The assessment before surgery should include a general anamnesis (e.g. cause of amputation, psycho-social state, aims in rehabilitation,...), a specific anamnesis of phantom limb pain, a physical assessment focussing on the sensomotor status, posture, endurance and range of motion of remaining upper limb joints. If the patient is already using prosthesis, also the function with the current fitting is assessed. [7]

If some deficits occur hereby, they should be addressed in therapy before and after surgery. For some patients training for general fitness and trunk stability might be appropriate to allow them to carry the weight of a prosthesis (about 3 kg) after fitting. Others might need to work on their balance and coordination. [8]

#### Stage 1: Reinnervation

Approximately 3 to 6 months after surgery, the first transferred nerves reach their targets and contractions can be registered either by surface or needle electrodes. Until this time it is important to facilitate the cortical representation of the non-existing arm and its movements, which is later used in the rehabilitation process. This can be done through motor imagery (imagined movements of the arm and hand) [9,10] or mirror therapy (where the patient sees the mirror image of the sound side, where the amputated limb is expected). [4] Since this is mental practice, no movements of the reinnervated muscles are expected at the beginning. At the end of stage 1, the patient might recognize small muscle twitches in response to attempted movement.

Another focus in this stage is alleviation of pain (by medication and mirror therapy/motor imagery) and oedema control as well as trunk stability, body symmetry and restoration of range of movement of the existing joints. Also, first motor activity can be facilitated using bilateral gross movement patterns (like those in PNF).

#### Stage 2: Signal training

As soon as the first EMG signals can be recorded the aim of rehabilitation is the improvement of neuromuscular control. Movement patterns according to the ones typical for the transferred nerves are then being tested and trained to find the movements which elicit the most powerful and best separated signals. Since the precise control of the signals is of utmost importance for later prosthetic use, particular attention should be paid on this stage. To support the patient in learning how to activate the new reinnervated muscles, the use of EMG-biofeedback is advised. This tool provides the patient with visual feedback that cannot be achieved otherwise at this stage and allows training of activating of one single muscle without another. [11,12] Additionally, the application of different contraction forces is trained. Usually, the separation of different muscle signals is easier if the patient is relaxed and only thinks about slight movements.

As soon as the signals can be controlled well using EMG-biofeedback, a training with a prosthesis mounted on a desk is recommended. Thus the patient can learn how to control the different movements of a real prosthesis.

#### Stage 3: Final prosthetic fitting

Once the signals are well established the final prosthesis can be fitted. Up to 6 electrodes have to be embedded in the socket. Thus an experienced prosthetist is needed. For defining the final positions of the electrodes the collaboration of surgeons, PMR specialists, therapists, technicians and prosthetists is recommended. After fitting, the control of the prosthesis needs to be trained again. The weight of the

prosthesis might affect the muscle signals and lead to the necessity of electrode adjustments and further signal training.

During prosthesis training, starting with movements without objects and then going on doing easy grasp/release tasks with different objects is recommended. In the end activities of daily living should be trained. [4] Finally, the patient receives the prosthesis to use it in daily life.

#### Stage 4: Follow-ups

While using the prosthesis at home, patients usually become skilled in manipulation of everyday objects. Nevertheless, some problems may occur as changes in the stump may have led to the necessity for socket changes. Since the muscle reinnervation can still continue after final fitting, the optimal electrode positions can change over time. Thus it is very important to maintain contact with the patients after final fitting and to regularly assess their prosthetic function, pain and quality of life. Follow-ups are recommended 6, 12 and 24 months after final fitting and whenever necessary.

### CONCLUSION

The presented recommendations are a summary of literature reviews and clinical experience. They divide the process of TechNeuroRehabilitation into 1 + 4 stages. After reinnervation of the target muscles, the motor recruitment can be facilitated by using sEMG-biofeedback. This allows the visualisation of motor activity. As soon as good motor control is achieved, the use of the prosthesis can be trained. Finally the prosthesis can be used in daily life and regular follow-ups guarantee good prosthetic function.

We are currently conducting a clinical trial to evaluate the recommendations described in the present study (proof of concept). With advances in technology (as virtual reality training systems or the use of pattern recognition) or changes in the surgical procedure (as using implanted electrodes) also the rehabilitation process needs to be adapted. We hope to improve the therapeutic outcome with our recommendations for therapy after targeted muscle reinnervation.

### ACKNOWLEDGEMENTS

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## RESTORING SENSATION IN AMPUTEES: CHRONIC STABILITY OF IMPLANTED CUFF ELECTRODES

Daniel Tan<sup>1</sup>, Matthew A. Schieffer<sup>1</sup>, Emily Graczyk<sup>1</sup>,  
Michael Keith<sup>1</sup>, J. Robert Anderson<sup>2</sup>, Dustin J. Tyler<sup>1</sup>

<sup>1</sup>Case Western Reserve University

<sup>2</sup>Louis Stokes Cleveland Dept. of Veterans Affairs Medical Center

### ABSTRACT

#### Introduction:

An ongoing challenge in neural interfaces for sensory feedback is to produce multiple locations of sensory restoration with an interface that is stable for chronic, long-term clinical applications. Our approach uses the Flat Interface Nerve Electrode (FINE). The FINE is a multi-channel peripheral nerve cuff electrode that maintains the nerve geometry for selective electrical stimulation. The FINE has demonstrated stability and selectivity in chronic animal studies and selectivity in acute, interoperative human studies. Our hypothesis is that chronic implantation of FINEs will result in stable and selective restoration of sensory feedback in human amputees.

#### Materials and Methods:

Two amputees underwent surgery to implant 2-3 nerve cuffs around the median, radial, and ulnar nerves. S102 received two FINEs (8 channels each) and one spiral cuff (4 channels). S104 received two FINEs on median and radial nerves. Starting 1 month post-implant, stimulus-to-sensation mapping was conducted. Repeated measures of threshold and impedance were taken over time to evaluate stability. Threshold was defined as the minimum amount of charge which elicited reliable perceptual sensation with a multiple-reverses algorithm. After each stimulation presentation, the subject described the sensation and drew the perceptive field location on a schematic of a generic hand. Impedance was measured between pairs of electrode channels on each cuff.

#### Results and Discussion:

For subject S102, 19 of the 20 electrode channels provided a sensory response with approximately 15 unique locations of sensation. For subject S104, 15 of 16 electrode channels provided a sensory response with approximately 9 unique locations. In both subjects, fingertip perceptive fields and natural tactile modalities of pressure and vibration were perceived. Channel-specific perceptive fields and modality responses remained relatively stable after an initial "settling" period post-implant. Individual linear regression of the threshold and impedance on all electrode channels for 22 months and 13 months indicated a slope not significantly

different than zero or that was significantly decreasing, suggesting a stable interface.

#### Conclusions:

Multi-contact, cuff electrodes, such as the FINE and spiral, are stable and selective neural interfaces suitable for providing sensory restoration in amputees. At 22 months and ongoing, the implant is also the longest sensory stimulation system in an amputee to date.

### ACKNOWLEDGEMENTS

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## APPLYING A FITTS' LAW INSPIRED APPROACH TO QUANTIFYING PERFORMANCE IMPROVEMENT IN A TOUCH-FEEDBACK EQUIPPED PROSTHESIS

Zachary C. Thumser<sup>1</sup>, Rock Lim<sup>2</sup>, Paul D. Marasco<sup>2</sup>

<sup>1</sup>Louis Stokes Cleveland Department of Veterans Affairs Medical Center; <sup>2</sup>Cleveland Clinic

### ABSTRACT

One of the challenges in evaluating the practical value of prostheses is quantifying performance. Fitts' Law is an established model for predicting the time required for a rapid pointing task as a function of the distance and size of the target, as well as parameters characterizing the efficacy of the pointer used (MacKenzie, IS 1992). A Fitts' Law inspired model will be used to compare the performance of a prosthesis with and without integrated tactile feedback active. Our model relates the variable difficulty of this task to the time required as:

$$T = a + b \cdot \log_2(1 + D/W \cdot c)$$

While not strictly a Fitts' task, a model in the form of Fitts' Law contains analogous terms, most of which are directly transferrable from Fitts' Law to the new model. Despite the differences, preliminary data suggest that the logarithm of the ratio of the movement size(D) and the precision required(W) is linearly related to the average task time(T). This relation consists of a fixed delay or latency term(a) plus an inverse speed of the system(b). Unlike the standard Fitts' Law formulation, an additional constant(c) is required to relate the different dimensions being measured.

This Fitts' variant can be applied to the ability of subjects to accomplish a task involving fixed delays and difficulty-scaling components. Application of the model can be used to measure changes in task performance, both in able-bodied subjects and in amputees. In our task, a linear slide must be moved a distance(D), maintaining grip force within a target range such that grip force is high enough to move the slide but not exceeding a threshold which activates a brake. Both the resistance and braking threshold are variable and this range provides the precision required(W). Varying both requirements allows for a wide range of D/W and task times(T) to fit parameters a and b.

The task is performed with a prosthetic hand worn by an able-bodied user (a bypass prosthesis) incorporating an Ottobock Myohand with a load cell mounted on the thumb to measure force and a force transducer on the user's hand. This arrangement allows the user to perceive real-time

tactile information and provides instantaneous feedback for the brake controller. By comparing the differences in performance with and without the touch feedback, we can quantify the effect of somatosensory feedback on a user's ability to maintain specific grip force during a task.

## LONG-TERM PERIPHERAL NERVE INTERFACES TO RESTORE SENSATION: PROGRESS AND PROSPECTIVE

Dustin J. Tyler

*Case Western Reserve University*

### ABSTRACT

Present prosthetic devices do not provide tactile or cutaneous feedback to amputees. There have been significant advances in both the mechatronics to provide anthropomorphic hand behavior and in control to provide more degrees of freedom with non-sequential prosthesis control. Providing natural, tactile feedback to individuals with limb loss is as important as the mechatronics and control of prosthetic devices, but has lagged in development. Sensory substitution has shown value in object manipulation, but it is unnatural, requires mental translations and additional external equipment, and does not restore the user with a sense of one's own hand. Targeted sensory reinnervation has shown the value of natural tactile sensation in providing a sense of body self-identification with the prosthetic device. As early as 1970, electrical stimulation of peripheral nerves was shown to restore sensation that is referred to locations on the missing limb, but over a broad region somatotopic region. Recently, intrafascicular peripheral nerve interfaces have shown more localized, single-modal tactile perceptions in clinical trials lasting less than 30 days. In my lab, we have demonstrated extraneural peripheral nerve cuff electrodes with multiple localized sensations over the complete hand with repeatable, stable results for more than 18 months. Novel stimulation approaches provide multiple qualities of sensation at each of the locations of sensation. Sensory feedback during task performance with the prosthesis improves control, accuracy, subject confidence, and incorporation of the prosthesis into body image. Given these recent successes and the accelerating progress towards sensory restoration, a sensory prosthesis within the next four years is a very real possibility. However, despite the laboratory and clinical trial success there are some real practical challenges specific to these devices that need to be addressed before they can be realized in commercial prosthetics. These include regulatory approvals, clinical models of implementation, cost, reimbursement, therapist and user training, device development, and prostheses with sensing technology. In this presentation, I will review the progress and most significant developments in sensory feedback, as well as present some of the most important challenges to making a sensory enable prosthesis a reality. Technical challenges are common to all

countries while regulatory and reimbursement issues will be somewhat specific to issues in the U.S. This should engage the audience in strategic discussions to address critical issues in the drive towards successful translation of sensory restoration from clinical research to clinic distribution

### ACKNOWLEDGMENTS

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## CUSTOM SILICONE SOCKET DESIGN

Jack Uellendahl<sup>1</sup>, Elaine Uellendahl<sup>2</sup>

*<sup>1</sup>Hanger Clinic; <sup>2</sup>New Touch Prosthetics*

### ABSTRACT:

Over the past 10 years the author has gained significant experience in designing upper-limb prosthetic sockets using high consistency rubber (HCR) silicone for various amputation levels. These sockets have been reported by wearers to be much more comfortable than their previous thermoplastic sockets. Techniques have been developed to take full advantage of the unique material qualities afforded by HCR silicones. Use of prepreg composites with HCR silicones has increased the range of socket construction possibilities.

This presentation will demonstrate the design variations that have been developed to take advantage of the material characteristics of HCR silicone.

## USE AND SATISFACTION WITH ADAPTIVE DEVICES IN YOUNGSTERS WITH UPPER LIMB REDUCTION DEFICIENCIES

Corry K. van der Sluis<sup>1</sup>, Ecatarina Vasluian<sup>1</sup>, Iris van Wijk<sup>2</sup>,

Pieter U. Dijkstra<sup>1</sup>, Heleen A. Reinders-Messelink<sup>3</sup>

<sup>1</sup>University of Groningen, University Medical Center Groningen,  
Department of Rehabilitation Medicine, Groningen, the Netherlands;

<sup>2</sup>Rehabilitation Center De Hoogstraat, Utrecht, the Netherlands;

<sup>3</sup>Rehabilitation Center, Revalidatie Frieslandâ, Beetsterzwaag, The Netherlands

### ABSTRACT

#### Background:

Youngsters with upper limb reduction deficiency (ULRD) may encounter limitations in activities of daily living (ADLs) such as using cutlery, lifting heavy objects, doing sports, cycling or driving. Although prostheses can be prescribed to overcome activity limitations, many are rejected due to discomfort or lack of functionality. Children with ULRD may use alternative solutions, such as adaptive devices (ADs). ADs are items used to facilitate ADLs, and are mostly developed by rehabilitation professionals. Devices that can be mounted on a prosthesis are not considered ADs. Information about the use, satisfaction and social adjustment with ADs in comparison to prostheses is lacking. The aim of the study was to evaluate the use, satisfaction and social adjustment with ADs compared to prostheses in youngsters with ULRD.

#### Methods:

A cross-sectional study using questionnaires was performed. Youngsters with ULRD between 2-20 years old responded to questions about personal and ULRD characteristics, difficulties in activities, preferred solutions for activities, usage, satisfaction and social adjustment with ADs versus prostheses. To evaluate satisfaction, the Dutch version of Quebec User Evaluation of Satisfaction with assistive technology questionnaire (D-Quest) was used. Social adjustment was assessed with a subscale of the Trinity Amputation and Prosthesis Experience Scales questionnaire (TAPES).

#### Results:

360 ADs were used by 76% of 218 participants (n=166). Eighty youngsters used or had used prostheses (37%). Participants were mainly boys (58%) with transversal ULRD (87%). ADs were used in 43% for self-care (using cutlery), mobility (cycling, 28%) or leisure activities (sports or playing a musical instrument, 5%). Prostheses were used for self-care (4%), mobility (9%), communication (3%), recreation/leisure

(6%), and work (4%). More than 50% of youngsters had difficulties in performing activities like using cutlery or tying shoelaces, doing sports, handcrafting or household activities. The most preferred solution to overcome these difficulties was using the upper limbs and other body-parts (more than 60%), help from others (more than 50%), using ADs (up to 48%) and prostheses (less than 9%). Satisfaction with ADs was significantly higher than with prostheses.

## TRANSFERRING SKILLS FROM AN EMG CONTROLLED SERIOUS GAME TO PROSTHESIS USE

Ludger van Dijk<sup>1</sup>, Hylke van Dijk<sup>2</sup>, Corry van der Sluis<sup>1</sup>, Raoul Bongers<sup>1</sup>

<sup>1</sup>*University of Groningen, University Medical Center Groningen;*

<sup>2</sup>*NHL University of Applied Sciences*

### ABSTRACT

#### Introduction:

State-of-the-art myo-electric prosthetic hands require generating complex EMG signals for appropriate control. However, current prosthetic rehabilitation training does not train prosthesis users to reach such an advanced level of skill. Employing serious games in rehabilitation may offer a way of doing that - allowing for feedback about EMG signal quality tailored to an individual user while also creating an enjoyable and stimulating learning context. However, serious games often also change the task the generated EMG signals are involved in. As research suggests skill learning may be fundamentally based on the task the actions aim to accomplish, the question that needs to be addressed is whether learning to control the EMG signal in a serious game will transfer to prosthesis use in daily life.

#### Objectives:

To establish whether the control of EMG signals trained through serious gaming transfers to (1) a prosthesis-simulator task, and (2) to different musculature.

#### Methods:

In an experimental pre-test post-test design we trained 15 able-bodied participants to control a video game (Breakout). The goal of the game was to hit bricks by bouncing a ball using a paddle. Participants controlled the movements of the paddle through the EMG signals of the flexors and extensors of the wrist. Another 15 participants, making up the control group, played a regular Mario computer game. Three tests were conducted: (1) one level of the Breakout game was performed and speed, accuracy and EMG signals were measured, (2) the same task was performed but now the paddle was controlled by hand muscles (same task, different musculature), (3) participants grasped objects that varied in size with a prosthesis-simulator (different task, same musculature). Movement time and hand aperture profile were measured.

#### Results:

Preliminary analyses showed strong learning effects within the gaming task — on accuracy and speed as well as on the effectiveness of the generated EMG signals. There was no transfer to the task with the prosthesis-simulator.

#### Discussion:

To employ serious gaming in prosthetic rehabilitation it is required that actions used in prosthetic tasks improve by playing the game. The current research suggests that this is not always the case. The results will be used to provide guidelines for a serious game to train prosthesis use.

#### Conclusion:

Training sophistication of EMG signals through serious gaming leads to improvement of in-game performance but transfer to prosthetic control is limited.

## PROSTHETIC DESIGN CONSIDERATIONS FOR IMES® CONTROLLED PROSTHESES

James Vandersea, CPO<sup>1,3</sup>; John Migulez, CP<sup>1,3</sup>; Dan Conyers, CPO<sup>1,3</sup>;  
Melissa Evangelista, MS<sup>2</sup>; Paul Pasquina, MD<sup>3</sup>; Joseph Calderon, MS<sup>2</sup>; Joseph Lockhart, MS<sup>2</sup>  
*1-Advanced Arm Dynamics, Inc.; 2-Alfred Mann Foundation*  
*3-Walter Reed National Military Medical Center*

### INTRODUCTION

The purpose of this paper is to discuss present and future prosthetic designs and considerations when using implanted myoelectric sensors (IMES®) to control externally powered prostheses<sup>1-5</sup>. IMES® (Figure 1) are implanted directly into subjects' muscles and wirelessly transmit live EMG<sup>2</sup>. This paper reports on prosthetic designs and fabrication techniques required for integration with an IMES® control system. It also suggests new opportunities to address issues of fit and comfort given that the need to maintain positional stability and skin contact, as required by surface electrodes, is eliminated when using IMES®. Examples are provided from a current feasibility study on the use of IMES® in patients with transradial amputation. IMES® prostheses for other amputation levels are also considered.

### METHODS AND MATERIALS

#### Background: issues with using surface mounted electrodes for myoelectric control

Current, conventional myoelectric prosthetic systems are designed with surface mounted electrodes in mind, where most available systems use two EMG electrode pickups over antagonistic muscle groups to control one function at a time (e.g. open/close, pronate/supinate, and flex/extend). Most systems house all the electronics within the structure of the prosthesis. The surface electrodes must be mounted into the inner socket and positioned and secured such that contact with the skin over the intended muscles is maintained. The socket has to be fabricated to allow the patient as much range of motion as possible without losing contact or allowing migration of the electrodes.

Fluctuation in the position of the prosthetic socket or shifting electrodes can result in false EMG signals or diminished EMG output caused by surface electrodes moving on the residual limb to a less than desirable location. Surface electrodes can also lift off or lose contact with the patient's residual limb. Current surface mounted electrodes are susceptible to moisture from perspiration leading to grounding or shorting of electrical contacts and unintended control inputs, which elicit unwanted actions of the prosthetic

terminal device. Additionally, surface mounted electrodes can only detect the gross contractions from a group of superficial muscles, whereas the origin of contraction cannot be discerned among the neighbouring muscles. This limits the number of specific sites that can serve as unique control inputs. When the number of functions offered by the terminal device exceeds the number of control inputs, the control scheme is reduced to the sequential activation of different functions, where the transition between functions is triggered by a specified signal, such as a rapid co contraction. Needless to say, this control scheme is slow and unnatural.

Prosthetic sockets with surface mounted electrodes have to be tight to maintain contact of the surface electrodes. Overly tight sockets can reduce range of motion, limiting the patient's functional envelope. Care must also be taken with surface mounted electrodes to make sure excessive pressure is not applied to the skin. Excessive pressure can cause localized skin irritation, discomfort and even skin breakdown. Special care must be taken when surface electrodes are used over grafted or even insensate skin to reduce the chance of injury. For petite patients, paediatric patients, or patients with very short residual limbs, it can be difficult to find a flat surface that is large enough to effectively mount surface electrodes. In these cases electrodes are often mounted directly adjacent to or over bony prominences, increasing the potential for irritation. In some instances we occasionally see an allergic reaction to the metals in the standard surface electrodes and have to switch to specialty, gold plated electrodes.

#### IMES® overview

IMES® were developed to replace standard, surface-mounted EMG electrodes in myoelectrically controlled prostheses. IMES® can be injected into both superficial and deep muscles where they are encapsulated by the surrounding tissue. The implants are not susceptible to movement or perspiration, offering improved reliability and fidelity of the EMG signals. The signal pickup area is localized to the tissue immediately surrounding the device, which measures only 2.5mm x 16mm<sup>1</sup>. The opportunity to discern signals between neighbouring superficial and deep muscles increases the potential number of control inputs, allowing for simultaneous control of an increased number of prosthetic functions<sup>2,3,4</sup>.



Figure 1: IMES®, implantable myoelectric sensor

IMES® are designed to function within a specific, externally generated electromagnetic field produced by a coil. The sensors receive power, digital addressing, and command signals from this coil. EMG signals are wirelessly transmitted over a shared magnetic link via radiofrequency (RF) communication.

#### IMES® prosthesis considerations

There are several considerations to take into account when fabricating an IMES®-controlled prosthesis (Figure 2). Mainly, the coil that generates the electromagnetic field to power the devices and receive data from them needs to span across all devices implanted. The electronics needed to process this information and direct it to the terminal devices has to be housed on or within the prosthetic frame, or somehow attached to it.



Figure 2: Rendering of an IMES® prosthesis

#### IMES® prosthesis components

The initial prostheses designed for the first-in-man feasibility study of the IMES® System for transradial amputees feature commercially available components that have been customized to interface with the IMES®. These include an iLimb ultra prosthetic hand with a motorized thumb that can be driven separate from the other fingers such that two degrees of freedom (open/close and thumb abduction/adduction) are offered. A Motion Control wrist rotator was adjusted to incorporate a six ring male coaxial plug, and standard Otto Bock quick disconnect wrist, to integrate with the six ring female plug on the iLimb hand. The Motion Control wrist rotator passes control signals from each of three

degrees of freedom (DOFs): 1. wrist, pronation/supination; 2. hand, open/close; 3. thumb, abduction/adduction. Power to the three DOFs of the wrist and hand is supplied by an 1150 mAh Motion Control battery that is internally mounted.

#### IMES® coil

The IMES® coil has two functions: 1. It creates an electromagnetic field to power the implanted IMES® and 2. It acts as an antenna to receive the data being transmitted from each IMES® device.

All the IMES® electrodes have to be contained within the coil's electromagnetic field in order to obtain power and for their transmission signals to be received. The circumference and spacing between the coil windings must be maintained to provide constant impedance or induction level. To meet this requirement, the coil is wound around and laminated within the wall of the rigid frame of the prosthesis and locked into place to minimize distortion (Figures 3). The coil has minimal impact on the weight, form, function and appearance to the definitive prosthesis.



Figure 3: IMES® coil fabricated into frame

The coil is powered by an externally mounted prosthetic control interface, or PCI, that has its own battery and is worn on a belt (Figure 4). The PCI belt pack provides power for the coil, decodes and post processes the EMG signal from each IMES®, and in turn transmits the information, via hard wire, out to the terminal device.

It is recognized that the external PCI belt pack is cumbersome and could be viewed as an inconvenience to the user. There are current efforts to miniaturize the PCI electronics such that they can be located within the prosthetic device.



Figure 4: IMES® System

The coil is something new for prosthetists to incorporate into the prosthetic frame. It has to be carefully wound and tested for impedance during the fabrication process. While fabrication of the coil requires specialized skill, equipment, and tooling, the process was successfully demonstrated in the initial feasibility study.

#### Prosthetic Socket

The socket design for a prostheses controlled by IMES® is restricted in that metallic objects and/or conductive materials cannot occur anywhere on or inside the coil. Metallic objects could interfere with the electromagnetic field, disrupting communication between the implants and the prosthesis. In addition, resistive heating could occur in conductive metals due to the concentration of energy generated by the coil, potentially causing discomfort or even injury to the patient. For this reason, patients with metallic orthopaedic implants or shrapnel in their residual limb are contraindicated for IMES®.

Despite this limitation, the IMES® allow considerable leeway in fabricating the remaining aspects of the prosthesis. Since there is no longer a need for the electrodes to contact the patient's skin, several different socket materials or interface materials can be used to act as a buffer between the patient's stump and the prosthesis. The inner sockets of the IMES® prostheses provided for the feasibility study were made from Proflex with silicone. These sockets utilize direct skin contact with supracondylar suspension.

While the IMES® system can be limiting in some ways, such as the limits for not incorporating metal into the coil area and making certain that the coil covers all implanted IMES®, the design does have other significant advantages. The IMES® prosthetic design does not rely on the electrodes to be in direct contact with the skin, as required by traditional surface electrodes. Silicone or other gel liners have gained popularity for comfort, positive suspension, and potential for more liberal prosthetic trim lines that can increase range of motion and comfort for the patient. The use of silicone liners

for IMES® can readily be fit using a suction suspension and valve, lanyard suspension, or possibly a shuttle lock system (provided any metallic components remain outside the coil field). Another possible fitting technique is to have patients use fitting sock(s) between the skin and socket. Patients often report they prefer wearing prosthetic socks to absorb perspiration in warm climates and to increase insulating properties in cool weather to help prevent the prosthesis from acting like a heat sink and drawing heat from the residual limb. Prosthetic socks can also be used in layers to either tighten or loosen the prosthetic socket. The opportunity to adjust prosthetic fit on the fly as the patient fluctuates in volume throughout the day or over the course of months or years, is a significant advantage of the IMES® system.

## RESULTS

The initial experience with the first two subjects participating in the IMES® feasibility study has been encouraging. Subjects have demonstrated the ability use IMES® to control three DOFs, both independently and simultaneously, offered by a terminal device. The study prosthetist was able to improve prosthetic comfort, fit and range of motion by taking advantage of the liberation from surface mounted EMG electrodes. The first study subject, who has been using an IMES® system for more than eight months now, has demonstrated that as long as the IMES® are contained within the volume of the coil and subsequent electromagnetic field, the IMES® continue to register and transmit EMG from the residual musculature. This subject experienced volume fluctuations that would have caused lost contact between traditional surface electrodes and the skin. Despite these fluctuations, the IMES® continued to operate and transmit continuous, reliable EMG signals. The subject has demonstrated continuous operation of his IMES® prosthesis, even when the prosthesis shifts, such as during strenuous activity or tasks performed overhead or below the knees. This subject has integrated his IMES® prosthesis into Activities of Daily Living (Figure 5).

The fact the IMES® reliably continue to transmit EMG even if the prosthesis rotated on the residual limb demonstrates an advantage over the use of surface-mounted EMG electrodes.



Figure 5: Activities of Daily Living using an IMES® prosthesis

## DISCUSSION

The initial IMES® study for transradial amputees has demonstrated that IMES® can register and transmit EMG and can be used to simultaneously control three degrees of freedom offered by a terminal device. The initial study also demonstrated that a transradial prosthetic device can be fabricated to energize the IMES® electrodes, receive the EMG signal from the IMES® and transfer to the prosthesis and provide functional operation with the control signal received.

Even with the limitations of the coil, there are many advantages including increased functional control of the prosthetic device (3DOF or more). Increase reliability of the EMG signal, not dependant on surface contact with the patient. Additional advantages include increase flexibility for the choice of socket and interface materials.

We expect to have increased comfort and acceptance of the prosthetic device with IMES® controlled prosthesis. We would not expect to have the same discomfort, skin irritation and skin breakdown issues associated with surface mounted electrodes. Reducing socket pressures associated with surface mounted electrodes we would expect to see an increased range of motion and an increase in the user's functional envelope.

Other problems associated with surface mounted electrode and grafted tissue or insensate skin will also not be an issue with IMES®, reducing the chance for skin breakdown. We believe the IMES® provides greater comfort for the patient. Greater comfort combined with increased function and ability of the prosthesis, should lead to a greater acceptance of the prosthetic device. We also believe there are undiscovered fabrication techniques and socket designs which will come about with the incorporation of IMES® technology.

## Future prosthetic designs for other amputation levels

We are hopeful that as we continue to develop the IMES®, this technology will serve a wider variety of patients; including those with transhumeral, shoulder disarticulation, transtibial and transfemoral amputation levels. Future designs include the possibility of incorporating a flat, "pancake style" coil into the prosthesis to energize and receive information for each of the IMES® electrodes. A flat, as opposed to a circumferential, coil design should increase the flexibility of the socket design and socket suspension. The flatter coil design can be more readily incorporated into shoulder disarticulation and inter-scapular thoracic level prosthetics designs.

It is expected that the IMES® technology will also be incorporated into partial hand amputation patients to control partial hand prosthetic devices, representing the single largest upper extremity amputation level. Incorporating IMES® at the patient hand level shows promise to increase the control, dexterity and acceptance of partial hand prosthetic devices.

With each new prosthetic level where IMES® are integrated, new sockets, frames and prosthetic components will have to be developed to optimize the functional capabilities of the IMES®. IMES® hold great promise for future prosthetic designs and operation, which will serve to enhance the patient's incorporation of the prosthesis into their self-identity, activities of daily living and functional capabilities.

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## OVERVIEW AND INITIAL CLINICAL RESULTS OF IMES© IMPLANTED ELECTRODES

Jamie Vandersea  
*Advanced Arm Dynamics*

### ABSTRACT

Poster gives a general overview of IMES© implantable electrodes construction, surgical implantation of IMES© devices, construction of prosthesis, communication between IMES© electrodes and prosthesis, results of the first two transradial subjects using IMES© controlled prostheses. Poster provides a summary of the IMES© and initial clinical study functional assessment results. Discussion of initial components used for the first IMES© prosthesis. Subject operation of the 3 simultaneous degrees of freedom for an IMES© controlled prosthesis. Initial subject functional assessment results for SHAP, ACMC and BBT will be displayed. (Poster will be a place for any presenters of IMES© related material to have more in depth follow up discussions after podium presentations. Or for individuals that may have missed specific podium discussions on IMES© implantable electrodes.)

## CONTROL LOCATIONS FOR HARNESSSES USED IN UPPER LIMB PROSTHESES

Alistair N. Vardy & Dick H. Plettenburg

*Delft Institute of Prosthetics and Orthotics*

*Department of Biomechanical Engineering, Delft University of Technology, The Netherlands*

*[a.n.vardy@tudelft.nl](mailto:a.n.vardy@tudelft.nl); [www.dipo.3me.tudelft.nl](http://www.dipo.3me.tudelft.nl)*

### INTRODUCTION

Traditional body-powered prostheses are still one of the most common cost-effective upper limb prostheses to date. Unfortunately, due to the high control force necessary to operate these devices, users experience discomfort of the harness used to transmit force from the shoulders through a Bowden cable to the terminal device [1]. Current harnesses require that the force exerted by the gripper is generated by the harness. As such, only powerful shoulder movements such as protraction of both shoulders or arm movements such as anteflexion and abduction can be used.

Recent developments in body-powered prosthesis have specifically targeted these high control forces with the aim of increasing user comfort and controllability [2]. However, harness design has remained the same.

This study is part of a larger investigation into novel control interfaces for upper limb prosthetics specifically aimed at low control forces and force feedback. This study aims to facilitate harness/interface design where control forces are substantially lower than in current body-powered systems. More specifically, this study aims to identify locations on the shoulder that exhibit a large relative displacement during shoulder movement. Locations with a larger relative displacement should afford a higher resolution of both control and feedback as they provide a larger proprioceptive response.

This study will focus solely on shoulder movements as a method for controlling the prosthesis as we want to focus controlling grip force and aperture separately from the orientation of the prosthesis. Moreover, shoulder movements can be performed without the need for extra balance requirements, contraction/stabilization of core muscles or a particular body position.

Unfortunately, it is currently unclear which movements yield the largest displacement. This study investigates changes in distance between points on the shoulder during different movements. To this end, 3D motion of 26 points covering the area over the shoulder blade were tracked for

5 motions: elevation, depression, protraction, retraction, and a combination of elevation and protraction. Besides change in distance, the smoothness of those changes are equally important. Moving the shoulder should result in a smooth change in aperture or force of the prosthesis.

We anticipate that elevation and protraction will yield the best results as these are the most powerful movements the shoulder can make. Importantly, we limit this study to movements of only one shoulder (in our case, the right shoulder). This affords the possibility of using the other shoulder as an additional means of controlling other aspects of the prosthesis such as flexion/extension of the wrist or pronation/supination.

### METHODS

A total of 20 healthy male subjects (age: mean=26.7, SD=5.0 years) participated in this study after signing an informed consent form. The study was approved by the local ethics committee.

Subjects were seated with their left arm on their lap and right arm resting on a table in front of them. A 5×5 point grid of reflective markers was attached to the shoulder in a rectangular pattern. The rectangle of markers was placed such that the upper left point coincided with the T1 vertebrae, the upper right corner coincided with the point where the shoulder blade attaches to the acromion. The lower left point coincided with T8 and the lower right with the lowest point of the shoulder blade (see Figure 1).

The grid was projected onto the subject's shoulder and adjusted so that it matched the before-mentioned anatomical landmarks. The projector was set to a fixed distance of 2 m to ensure a constant marker size. Grid points were first marked using an eye-liner pencil, and then reflective markers were attached. The 3D position of the markers was tracked using a Qualisys motion capture system. An additional marker was added to the acromion to track the position of the shoulder.



Figure 1. Marker grid on shoulder and acromion.

Subjects were asked to perform a series of 5 movements: *protraction*, *retraction*, *elevation*, *depression*, and a *combination of protraction and elevation (combi)*. Each movement was performed 10 times yielding a total of 50 trials. Each block of trials started with the arms in front of the mid-line of the body. During each block of trials, the hand remained at this location. Each block of 10 trials was performed in one go with each movement from neutral to maximal excursion and back in 1-2 seconds. The start and movement type were verbally indicated to the subject and the movements counted. The experiment lasted approximately 15 minutes.

The grid of markers applied to each subject differed due to anatomical variation. To afford comparison between subjects, displacement of a pair of markers was defined as the change in Euclidean distance between the initial and final positions. These values were then averaged over the 10 trials for each movement, yielding a single value for each movement and subject. Distance values were analyzed as absolute change in distance (in mm) and relative change (in %) where the values were normalized by the largest observed distance between of each marker pair. This latter affords a way to adjust for differences in movement excursion; subjects were able to elevate and protract their shoulder considerably more than depress or retract. Start and endpoints for each movement were detected using Matlab.

To glean a better insight in the differences between movements, the 10 pairs with the largest displacement were determined and compared between movements. Two repeated-measures ANOVAs were performed to assess if there was an effect of movement on the absolute and relative displacements. Data was assumed to be normally distributed

as assessed by the Shapiro-Wilk test. A significance level of 5% was maintained.

## RESULTS

There was a large difference in displacement between the 300 marker pairs. This was to be expected as the change in distance between the markers along the spine is negligible. Figure 3, top panel illustrates the difference in absolute displacement for all 300 marker pairs averaged over participants.

The 10 pairs with the largest displacement were determined for each movement separately. Figure 3, lower panel illustrates the difference between the movements where depression exhibited the smallest absolute and relative displacement. The combination of elevation and protraction showed the largest displacements. These differences were reflected in the results of both repeated-measures ANOVAs. As sphericity could not be assumed, the Greenhouse-Geisser correction was applied. Absolute:  $F_{1,143,10,203}=880.1$ ,  $p<0.001$ , Normalized:  $F_{1,277,10,187}=888.6$ ,  $p<0.001$ . All Bonferroni-corrected paired Student *t*-tests were significant at the level of  $\alpha = 0.001$  for both the absolute and normalized data.

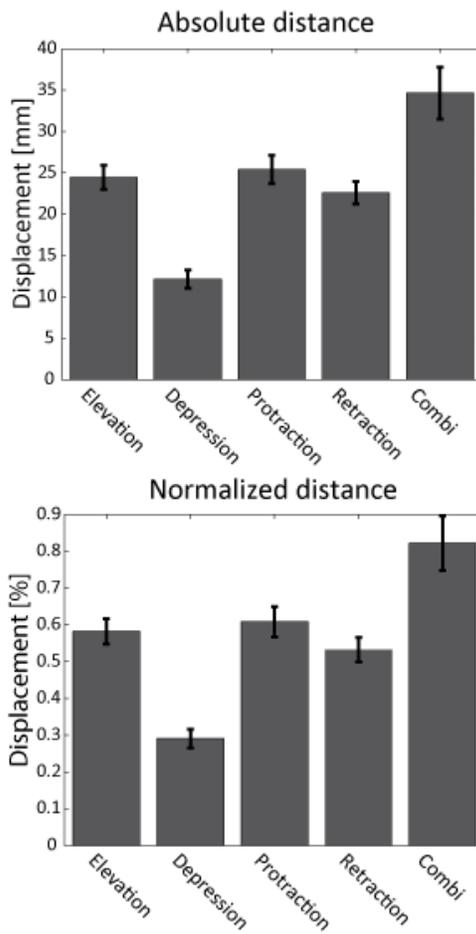


Figure 3. Absolute and relative displacement of the 10 best marker pairs. Error-bars represent the standard error of the mean.

The marker pairs with the largest displacement are illustrated according to their position on the shoulder in Figure 4, left and right panels for the absolute and relative displacement, respectively. The number of pairs is restricted to 5 for the sake of readability. Here again, the combination of elevation and protraction yielded the largest change in distance between markers with a maximum of 40.1 mm. Downward movement showed the smallest change where the maximal change was 13.8 mm.

Changes in distances between markers is not the only factor in choosing a marker pair. For elevation, protraction, and a combination thereof, the distance trajectories are considerably smoother than for down and back. In half of the participants, depression and/or retraction exhibited erratic traces, compared to only 3 where elevation was affected.

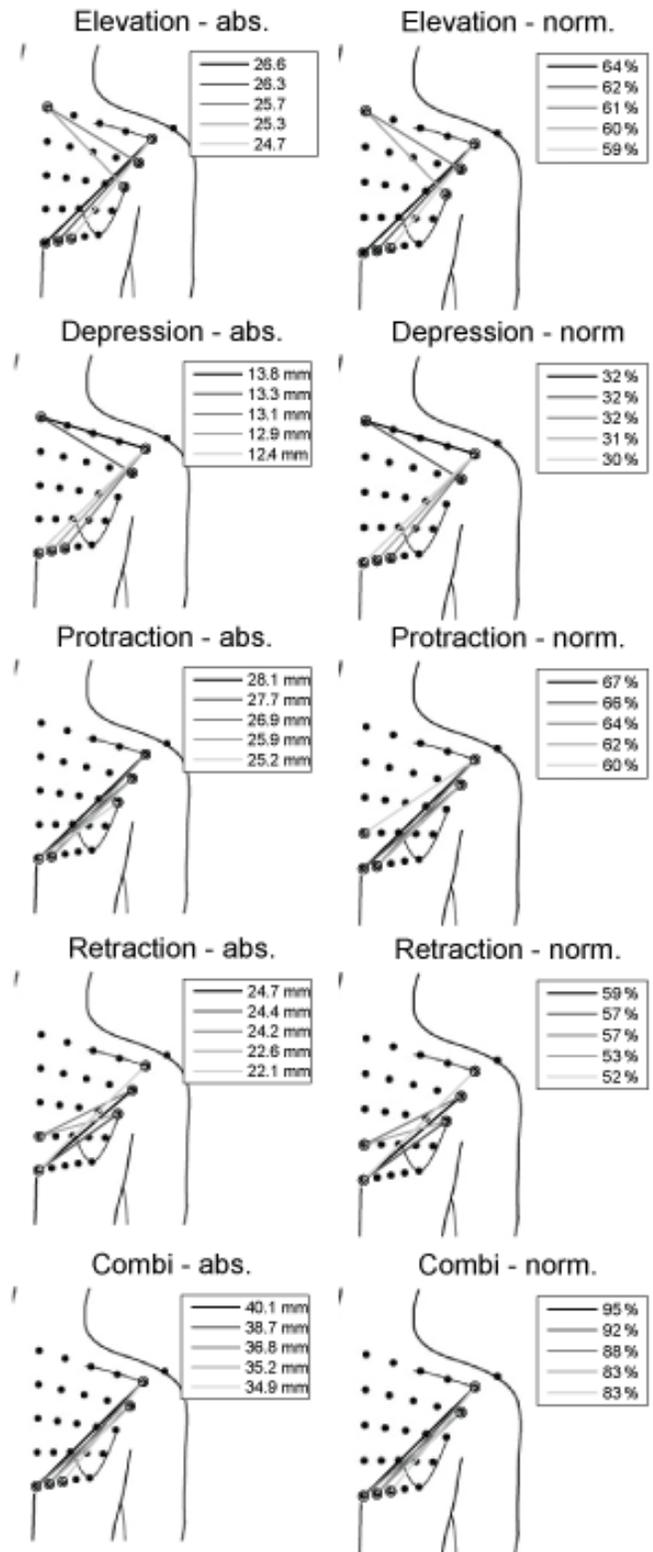


Figure 4. The absolute (left) and relative (right) displacements of the top 5 location pairs. Elevation and the combination of elevation and protraction show the largest displacements.

## DISCUSSION

This study illustrates the differences in skin displacement for the shoulder area which can be used to control an upper limb prosthesis. The results highlight the points on the shoulder that show the largest change in distance for 5 shoulder movements. In particular, elevation, protraction, and the combination thereof show the smoothest results and the largest changes in distance. Within those movements, the points that cross the shoulder blade diagonally from spine to the acromion were found to be better.

Traditional harness design is based on protraction and typically utilizes a figure-of-8 or figure-of-9 design. These designs cross both shoulders as high control forces are required to operate body-powered grippers.

A major benefit of a body-powered prosthesis is the feedback of the force exerted by the terminal device. Unfortunately, opening or closing the terminal device requires high control forces which one of the main reasons for abandonment [3]. Recent developments focus on reducing those control forces thereby improving user comfort and controllability [2]. This study highlights new harness areas and movements that may be used in harness design when lower control force can be used. Some devices use pneumatic, hydraulic or electrical assistance to amplify control forces similar to power steering in cars. Harnesses can then be redesigned without the requirement of high forces. In addition, by focusing on shoulder movements, the position and orientation of the gripper is left unchanged during operation benefiting markedly the use in ADL tasks such as pouring drinks or manipulating containers that should remain upright.

The results of this study opens new avenues for control locations that offer possibilities beyond current harnesses. A case in point is that by using a control location that crosses only one shoulder blade, that the contralateral shoulder may also be used for the control of the prosthesis. This offers an extra degree of freedom (DOF) to control the prosthesis such as flexion/extension or pronation/supination. This added DOF enables the prosthesis to be used in more tasks or the ability to perform tasks with greater satisfaction such as eating or turning a key and could be added to either body-powered or myo-electric prosthetic devices

This form of control can be readily used with externally powered prosthesis. Although the muscles controlling the device do not correspond to the muscle used to move a healthy hand, differences in kinematics and performance do not differ significantly between externally powered and body-powered prosthetic device [4]. Use of multi-DOF in externally powered prosthetic devices may alleviate the need

for switching to different modes and facilitate more natural control.

We conclude that the results of this study can be readily used to develop harnesses for body-powered prosthetic devices most notably in devices with low control force requirements or externally powered devices and afford novel control strategies in both type of movement used for control as well as in the number of DOF that may be employed.

## ACKNOWLEDGEMENTS

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## EVALUATING AN RFID-BASED OBJECT RECOGNITION CONTROL STRATEGY FOR MULTI-ARTICULATED PROSTHETIC HANDS IN A THREE PATIENT CASE STUDY

Martin Vilarino<sup>1</sup>, Rahul R. Kaliki<sup>1</sup>, Tiffany Ryan<sup>2</sup>  
*<sup>1</sup>Infinite Biomedical Technologies; <sup>2</sup>Advanced Arm Dynamics*

### ABSTRACT

Recent advances in mechatronics have led to the commercial launch of multi-articulating myoelectric hands. These devices are capable of multiple grip patterns and modes, offering myoelectric users the potential to expand function at the workplace and at home. However, complex control strategies, involving muscle contraction patterns or use from the contralateral hand, limit the practical and reliable access to the different prosthetic functions. Additionally, the current strategies can be non-intuitive and cumbersome, leaving many users fatigued and frustrated.

To address these issues, Infinite Biomedical Technologies has developed a controller which leverages Radio Frequency Identification (RFID) technology to detect control inputs directly from passively powered RFID tags. These tags can be affixed to objects or on the user's clothes. When the controller's antenna is brought into close proximity of a tag, a unique ID code corresponding to specific functions of the prosthesis is transmitted to the controller. The prosthesis then performs the action as determined by the ID code.

The RFID-based control strategy was compared with traditional dual-site myoelectric control across several metrics in a three patient longitudinal case study. Each of the patients recruited represented unique patient presentations: 1) a bilateral below-elbow user with experience using a multi-articulated myoelectric hand, 2) a unilateral below-elbow with multi-articulated hand experience, and 3) a unilateral below elbow with one day of experience with a multi-articulated hand. Qualitative metrics consisted of the Disabilities of the Arm, Shoulder and Hand (DASH), the Trinity Amputation and Prosthesis Experience Scales (TAPES) and the Orthotics and Prosthetics User Survey — Upper Extremity Functional Status (OPUS-UEFS). Additionally, we utilized quantitative metrics including the functional task portion of the Southampton Hand Assessment Procedure (SHAP) and two novel functional metrics designed by our team.

Qualitative and quantitative data from this study suggest that RFID technology can provide personalized solutions for different patients. For the experienced users, this technology

was determined to be a suitable alternative control method that could be used synergistically with traditional myoelectric control strategies. For the patient with limited experience with multi-articulated hands, the functional and subjective benefit was more pronounced.

## REDEFINING THE NORM IN COMPLIANT HAND CONTROL

Julian Wells  
*Advanced Arm Dynamics*

### ABSTRACT

#### Introduction:

Prosthetic hand technology has advanced greatly as new designs approach the function of the human hand. Current direct control signal strategies are not intuitive and are cognitively taxing. Sequential control may be described as simple to use, allowing users to focus on controlling one motor function at a time. Nevertheless, users commonly experience this control method as cumbersome and slow. [1] Powered wrists have made active pronation and supination a much needed and utilized technological advancement. However, adding a second degree of freedom creates a larger cognitive demand to switch between the intended motor functions. Accessing a heightened number of grasp patterns and positions (currently 24) poses additional challenges in streamlining prosthetic function.

#### Methods:

Two immediate challenges must be addressed to increase the efficiency of multi-articulating hands. The first is providing users efficient access to grasp patterns. The second is developing a more intuitive control mechanism that reduces the cognitive load required to activate multiple grasp patterns and control a powered wrist.

A review of current and emerging control strategies compares and contrasts the advantages and challenges of each.

#### Results:

Haptic interfaces, such as an iPod app to access features on the i-Limb ultra revolution (Touch Bionics™), give users access to 24 grasp patterns. Bluetooth enabled wearable devices may present another option for grasp pattern selection (Touch Bionics™.)

Radio Frequency Identification Tags can increase the number of accessible grasp patterns and reduce cognitive load (Infinite Biomedical Technologies™.) Other potential technologies include digit stalling techniques, voice control and gyroscope-embedded gesture activation (Touch Bionics™.)

Pattern recognition shows promise in making the control of prosthetic devices more intuitive and significantly reducing the cognitive load required to operate multiple degrees of freedom. COAPT™ harnesses the EMG patterns of native anatomical motions, translating them into the intended prosthetic motions.

#### Conclusion:

There are many options for control of hands that feature heightened numbers of grasp patterns. The selection, implementation and therapeutic training of these devices require concerted attention and care among the entire rehabilitation team. Control strategy customization and increased access to compliant hand grasp features is becoming the norm in upper limb prosthetic care.

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## ANALYSIS OF FACTORS INFLUENCING OUTCOMES OF FULL AND PARTIAL HAND MULTI-ARTICULATING PROSTHESES

Lynsay Whelan, MS, OTR/L, Nathan Wagner, LPO, CPO, OTR/L,  
Jeremy Farley, BMED, CPO/L  
*Touch Bionics, Inc.*

### ABSTRACT

Many factors have been identified that influence prosthesis abandonment [1]. These factors should be analyzed to determine how they can positively impact prosthesis use in the upper limb population. Using an outcome measure several of these variables will be compared between two cohorts representing patients affected by upper limb loss or deficiency.

### INTRODUCTION

In an extensive literature review by Biddiss and Chau, rates of rejection for the adult population were averaged at 39% for passive, 26% for body-powered, and 23% for externally-powered prostheses [1]. Some of the factors influencing abandonment, such as the level of limb loss and additional medical complications, are outside of the control of the rehabilitation team [1]. However, other causes of abandonment, such as type of device, poor training, poor fit/comfort, lack of function with the device, unrealistic expectations, and inability to control the device, can be counteracted by a team approach to care [1,2]. In recent research, even late fitting appeared to be counteracted by having an experienced rehabilitation team [3]. In a survey of users conducted prior to multi-articulating hands reaching the market, the users of externally powered systems listed priorities they wanted to see in future prosthetic design [4]. Several of these priorities are now available in hands today including: more motion of digits, thumb able to move to lateral position, improved ability to hold small objects, improved wrist motion, and control of the level of grip force [4]. Given the advancements that have been made and the knowledge of factors influencing prosthesis acceptance/rejection, these areas need further research with regard to multi-articulating full and partial hand prostheses.

### METHOD

#### Subjects

Participants included those individuals being fit with either an i-limb multi-articulating hand terminal device (i-limb ultra revolution or i-limb ultra) or i-limb digits partial

hand multi-articulating device. Individuals were invited to participate through their local prosthetist, and completed both a HIPAA Privacy Notice and consent form prior to participation. A total of 145 participants were included in this study with 55 representing partial hand limb loss/deficiency and the remaining 90 participants representing the wrist disarticulation level or more proximal.

#### Apparatus

The Patient Care Pathway (PCP) is an online tool designed to collect information before and after an individual is fitted with a prosthesis. This tool is currently used internationally by clinics fitting i-limb devices. The PCP collects data using the validated outcome measures of the Disabilities of the Arm, Shoulder and Hand (DASH) and Trinity Amputation and Prosthesis Experience Scales – Revised (TAPES-R). Additional questions regarding therapy involvement and individual goals aim to provide a holistic view while documenting a client-centered approach to the prosthetic rehabilitation experience.

#### Data Analysis

DASH and TAPES-R data were compared pre and post fitting to determine change as a result of prosthetic intervention. Additionally, different variables were identified between different groups that could further influence outcomes.

### RESULTS

DASH scores range from 0-100 with 0 indicating the individual perceives no disability and 100 indicating the most severe level of self-perceived disability. Minimum clinically important difference scores for the DASH have been defined between 3.9 and 15 [5]. Significant differences were found between pre and post fitting scores on the DASH for both the full hand group ( $p=0.004$ ) and the partial hand group ( $p=0.0003$ ).

Strong correlation was found between function (measured on the DASH) and satisfaction with the prosthesis (measured by the TAPES-R) in both groups.

TAPES-R satisfaction scores were only collected in the pre-fitting stage if the individual was currently using a prosthesis. Satisfaction scores range from 8-34 with 34 being the highest level of satisfaction. Significant improvements were found in both users of the i-limb ultra revolution ( $p=0.003$ ) and i-limb digits devices ( $p=0.000015$ ).

Table 1: Results of DASH

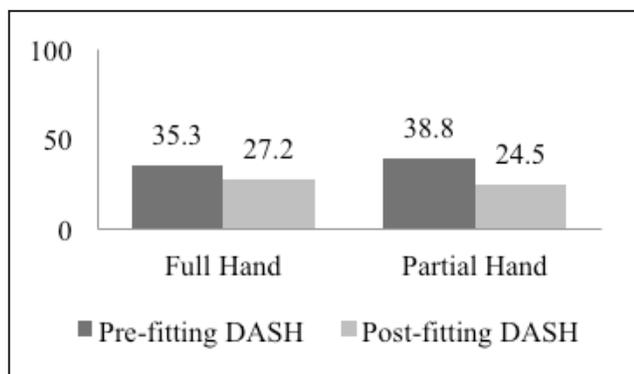
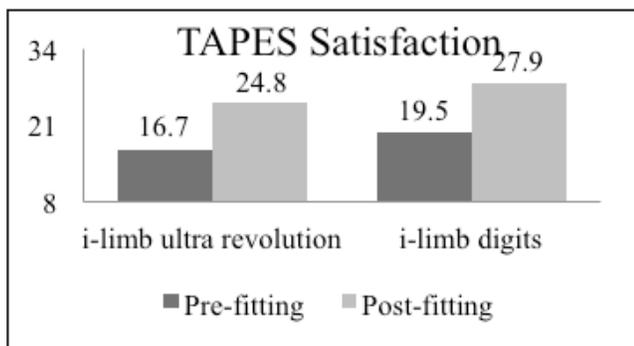


Table 2: Results of TAPES-R

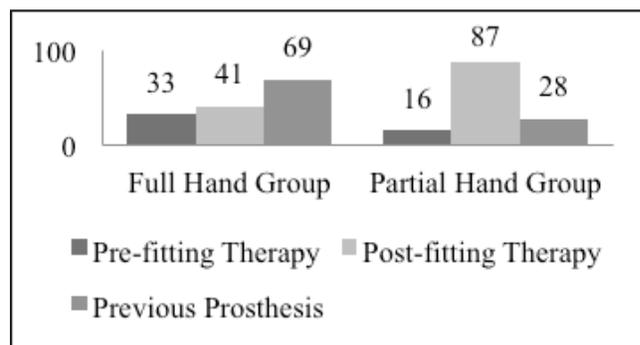


### DISCUSSION

When comparing the partial hand to the full hand group, significantly greater improvement ( $p=0.05$ ) was found in the partial hand group DASH score. Although not at a significant level, it was interesting that the partial hand group pre-fitting DASH scores started slightly higher. This is similar to previous research where the partial hand group reported greater impact on function than the transradial level [2,6]. The greater improvement in the partial hand group led to further questions about the other differences in these groups. The obvious difference identified as an influencing factor would be the full hand group having more proximal levels of amputation. Additionally, the partial hand group often has their own wrist range of motion and potentially additional intact finger motion, strength, and sensation to compliment the digits. Other differences identified include more individuals in the full hand group had a prior prosthesis. More individuals in the full hand group had therapy prior to

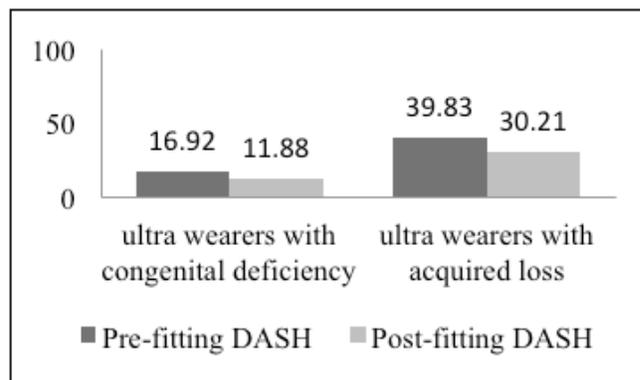
being fit with an i-limb device. This therapy may have been when first learning to use their prior prosthesis. However, in terms of receiving therapy after receiving the multi-articulating hand, the partial hand group included more therapy in the fitting and post-fitting phase.

Table 3: Differences in Groups (percentages)



In the full hand group, additional differences were found between those with acquired limb loss and congenital limb deficiency. DASH scores for those with limb deficiency were found to be significantly lower pre and post than those with limb loss ( $p=0.001$ ). Interestingly, the normative data on the DASH for the general population is 10.1 (SD 14.68) [7]. Comparatively, the group with congenital limb deficiency started at 16.92 pre-fitting and report 11.88 post-fitting.

Table 4: DASH Score Differences in Full Hand Group



### CONCLUSION

Research results revealed improved function for individuals after being fit with multi-articulating full and partial hand prostheses. Several areas are worth further investigation. While significant differences were found between full and partial hand groups in regards to the DASH, and the partial hand group had more therapy intervention, no direct correlation between these two points was found at this time. Additional analysis is necessary to examine the correlations between various influencing factors. The fact

that the group with congenital limb deficiency reached a near normal DASH score, as well as starting with significantly lower scores, may indicate a need for a more sensitive outcome tool with this population.

TAPES-R data revealed significant improvements. This data could be further analyzed to determine differences between prior passive, body-powered, or externally-powered devices.

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## THE INFLUENCE OF ENVIRONMENT – EXPERIENCES FROM USERS OF MYOELECTRIC ARM PROSTHESES

Cathrine Widehammar, Ingvor Pettersson, Liselotte Hermansson  
*School of Health and Medical Science, Örebro University, Örebro, Sweden*

### ABSTRACT

#### Background:

Myoelectric prostheses are used in varying degrees. According to the International Classification of functioning, disability and health (ICF) the environment includes the physical, social and attitudinal environment in which people live and conduct their lives. An environment with barriers, or without facilitators, will restrict the individual's occupational performance and can result in limitations of Quality of Life. Few studies have been made to see the impact of environmental factors on prosthesis use. In this study the ICF- model is the framework to understand the complexity of environmental factors influence on prostheses use. The aim of this study was to describe the experience of how environmental factors affect the use of myoelectric arm prostheses..

#### Method:

A qualitative descriptive approach was used and interviews were conducted with 13 adult prosthesis users at the Prosthetics and Orthotics Outpatient Clinic in Örebro, Sweden. The participants were 9 males and 4 females with age ranging from 20-74 years ; they had acquired (n=5) or congenital (n=8) cause of absence at trans humeral (n=3 ) or trans radial (n=10 ) level. Their experience from prosthesis use was ranging from 2- 30 years. Qualitative content analysis with an inductive approach was used for data analysis.

#### Results:

Participants' experiences of prosthesis use and how environmental factors affect them could be divided into seven categories: Various adaptations to the environment; Other peoples attitudes affect use; Support promotes use; Technical shortcomings affect use; Climate affects comfort and function; Ignorance and legislation complicates; Different approach to usability. Two themes occurred in all the categories and gave an overall perspective of what the participants believe have an important impact on prosthesis use: The prosthesis is/is not a part of my body; and, It is important to be like everyone else. A model was created to clarify the relation between environmental factors and prosthesis use/non-use. It illustrates how a persons coping

strategy interacts with all the different environmental factors it is exposed to and how this leads to usability in different degrees. The prosthesis use can be a barrier or a facilitator for activity, participation and body structure.

#### Conclusions:

Embodiment of prosthesis reduces environmental barriers and facilitates future use in both congenital and acquired upper limb amputees. Support to the persons and their family in prosthetic use, access to prosthesis training close to home, and considerations taken to the prosthesis appearance and function will facilitate future prosthesis use.

## PPP-ARM: THE IMPLEMENTATION OF A NATIONAL PROSTHESIS PRESCRIPTION PROTOCOL

Paula Wijdenes<sup>1</sup>, Michael Brouwers<sup>2</sup>, Corry van der Sluis<sup>1</sup>

<sup>1</sup>*Universital Medical Center Groningen*

<sup>2</sup>*Rehabilitation Center De Hoogstraat*

### ABSTRACT

#### Introduction:

Patients with an acquired or congenital upper limb defect need highly specialized care from multidisciplinary teams. In the Netherlands, various rehabilitation centers had their own method of treatment. Standardized care for these patients was necessary, especially when prescribing prostheses.

**Aim:** To create and implement a national digital protocol, which should be used when prescribing upper limb prostheses.

**Method:** The Prosthesis Prescription Protocol of the upper limb (PPP-Arm) has been developed in the previous 4 years and is a tool to structure, underpin and evaluate the prescription of upper limb prostheses. The protocol is based on WHO's criteria of the International Classification of Functioning (ICF).

The protocol consists of the following layers:

1. Establishing patient's demands
2. Establishing device requirements
3. Preparation of treatment requirements
4. Selection, try-out and final decision
5. Delivery of the device
6. Instructions and training
7. Evaluation

#### Results of Implementation:

The protocol has been created through the collaboration of several patients, rehabilitation teams, orthopedic workshops and insurance companies, collaborating in the working group PPP-Arm.

Implementation started in four rehabilitation teams in the Netherlands. After one year another six rehabilitation teams started using the protocol. In each team a knowledge broker was appointed, who was responsible for the implementation within his own center. A national project coordinator

maintained contacts with all parties involved, collected questions and problems when using the protocol, organized activities and meetings to develop the protocol further and to stimulate using the protocol.

Advantages of the protocol are:

- complete and structured
- user-friendly
- using the same ICF terminology and the same treatment guidelines by all users
- applied nationally
- digital reporting
- workplace independent login possibilities for all team members
- prescription reports are generated for insurance companies
- patients gain more insight in their own treatment process
- the protocol contributes to building a national database for research.

Disadvantages of the protocol are:

- time investment is needed to learn using the protocol
- a computer with internet access is required at the workshop

**Conclusion:** The nationwide implementation project PPP-Arm was successful, since all participating centers use the protocol. By developing PPP-Arm we have managed to create a national uniform and structured method to advise and evaluate the prescription of upper limb prostheses, which might be interesting for other countries as well.

## ABSTRACT

Accurate prediction of grasp type remains perhaps the most elusive target in prosthetics today. Here, we measure the tradeoff between speed of classification and classification accuracy in prediction of hand postures through a subset of data retrieved from the NINAPRO database, a repository of surface EMG collected from healthy individuals performing prescribed grasps. We assess classification error at increments of 5% time, starting with 10% time (i.e. 0.1·T, 0.15·T, 0.20·T, ... 1.0·T) following two prediction approaches: 1) minimized distance within an optimized feature space, and 2) maximized correlation to an ensemble average. We show that classification error decreases non-linearly in a feature-driven approach, versus approximately linearly in the signal morphology-based approach, and that our preliminary analysis indicates a possible enhanced accuracy in the feature-driven approach, versus correlation to ensemble average. Furthermore, we show that the rate of change in grasp prediction error is modest after approximately 0.4·T. We propose that this may prove a fruitful benchmark for development of prediction algorithms, and on-line decision-making in prosthetic devices.

## INTRODUCTION

Reliable decoding of user intention is a key objective of prosthetic control. Surface EMG (sEMG) is the most common paradigm for extracting control signals from the user, but its suitability for incorporation into advanced prosthetic systems is ultimately limited by the speed and accuracy of predictions made from its signals [1]. Naturally, there is a trade-off between speed and accuracy. However, there is little evidence as to the nature of this trade-off in detection of neural signals with application to prosthetic control, and whether there is a point beyond where this relationship becomes weighted in favor of generating a prediction versus waiting to collect more data.

Here, we take advantage of a large database of sEMG signals collected from neurologically intact subjects performing a series of grasp postures. We build simple classifiers for 3 common and relevant grasps, and measure classification error in progressively longer windows of data. Our interest is to identify the stage at which a grasp can be predicted, beyond which further information does not substantially increase predictor accuracy.

## METHODS

### Dataset

The complete Version 1 database of kinematic and sEMG data was downloaded from the NINAPRO database (<http://ninapro.hevs.ch>). This dataset contains kinematic and sEMG data recorded from the upper limbs of 27 intact subjects while performing 52 finger, hand and wrist movements [2]. For reasons related to resource limitations, this dataset was reduced to a sample 3 posture types performed by 16 subjects; these three postures described in previous publication as pertaining to high relevance to activities of daily living (ADLs) [3], and which comprise a subset of the NINAPRO database for which there has already been focused efforts in grasp classification [2]. Specifically, we analysed grasps #39-41 (a power grasp and two precision grasps). For each grasp type, 10 repetitions were performed, yielding a final dataset of 16 (subjects) × 3 (grasps) × 10 (repetitions) = 480 samples. The dataset comprises 8 tracks of sEMG, as well as an “activity stamp” is provided in the NINAPRO data, i.e. a binary index indicating rest period or active movement. These data are available in a raw format (Figure 1A).

### Data Windowing

The activity stamp allows for windowing the data. The primary objective of this analysis was to test classification accuracy across a range of windows, in increments of 5% (i.e. prediction after 10% of data had activity had been completed, 15%, and so on... through 100% of activity). A short window of rest was included with the data on either side of the activity window, equivalent to 5% of the original rest period before and after activity. Data were then length-normalized: While the datasets had very similar durations (total number of samples), this was not uniform across the entire dataset. Whereas one of the analyses (correlation) performed here requires uniformity in sample duration, we length-normalized every dataset to 150 samples: data were interpolated to an integer multiple of the original number of samples great than 10x the final length (i.e. any number of samples >1500), and then down-sampled (Figure 1B).

### Signal Processing

The sEMG data were inspected for artefact due to quantization error, i.e. any track for which less than 5 distinct values occurred throughout the entirety of the track; such tracks were eliminated. Data were then standardized within each trace (subtraction of the mean, division by the standard deviation), and processed using conventional means: data were rectified and filtered bi-directionally with a low-pass Butterworth filter. We note that we use a 10Hz filter (cf. 1 Hz) to allow more nuanced features to remain. The sEMG data were then averaged across all tracks, and this average was smoothed with a 20% loess moving average. Thus, the outcome of all data conditioning steps is a single, averaged and smoothed trace that represents the sEMG record across the entire action, with a small margin of quiescence before and after activation (Figure 1C).

Pre-Classification Feature Extraction

From each trace, 15 features were extracted. Other features were considered, but the feature space was intentionally limited to a small number of parameters in the interest of computational efficiency

Table 1: Features extracted from sEMG traces.

Features extracted (N=15)		
Maximum Value	Path Length	# of Direction Changes
Minimum Value	Area Under Curve	Path Length (Range Norm)
Range (max-min)	Variance	Path Length (Vel. Norm)
Average Velocity	Skewness	Variance (Range Norm)
Avg. Vel. (Unsigned)	Kurtosis	Variance (Vel. Norm)

Classification Approach

Here, we took a within-subjects approach to grasp prediction, which has been shown elsewhere to yield a higher prediction accuracy than inter-subject approaches. Two separate classifications were undertaken: 1) prediction following feature extraction, and 2) prediction based on trace morphology. In both cases, classification accuracy was assessed via a training set of 8 repetitions and a test set of 2 repetitions.

For classification following feature extraction, the centroid of each posture type was created from the training set; this centroid was in N-dimensional space ( $N \leq 15$ ), where the final dimensionality was determined *ad hoc* via an “add-one-in” feature selection routine, where the features were added one-at-a-time in a way that maximizes classification accuracy; this cross-validation was performed 20 times, with the final set of features determined via the Akaike Information Criterion. The resulting optimal feature set was then re-applied to the training set (thus reducing its dimensionality from 15...), and the data in the test set were compared against these targets. To eliminate distortion due to scale, the data set was standardized within each feature (again: subtraction of the mean and division by the standard deviation). The predicted grasp type was then taken as the posture whose centroid was closest to the feature set described by the sample in the test set.

For classification following correlation, the eight training traces for each posture were merged into a single representative trace by standardization followed by ensemble averaging. The predicted grasp type was then taken as the posture whose ensemble average was most highly correlated to the sample in the test set.

Prediction accuracy was computed via the balanced classification error (BCE) averaged over 25 replications. Results reported here reflect an average over all subjects.

**RESULTS**

Classifier Error Over Time: Feature Extraction

Averaged across each subject, the prediction error for the three grasps ranged between approximately 0.2 and 0.4. As expected, the error was greatest for least time analysed, however we note that even very small windows of data, the classification error is substantially better than chance.

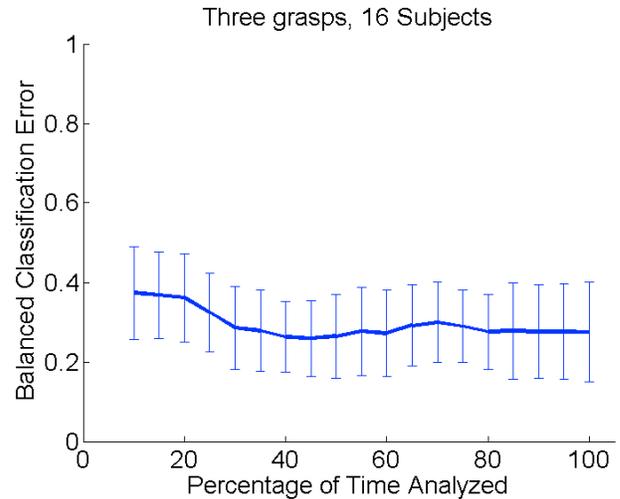


Figure 2: Balanced classification error in within-subjects grasp prediction task following feature extraction. Error appears to decrease steadily until approximately 0.40·T, whereafter the trend is approximately constant.

Importantly, we see that the trend of classification error appears to decrease more across smaller windows of observation, and remain more constant in larger windows. This suggests that there may be a shift in the cost-benefit trade-off of prediction speed versus accuracy. Here we note that the diminishing returns appear to begin near 0.40·T.

Furthermore, we observe that in this preliminary analysis, we did not observe a trend in the number of optimal features across time. That is to say that the dimensionality of the features space did not appear to change as a function of window size.

Classifier Error Over Time: Correlation

In general, the classification error following correlation analysis was higher than the error found in correlation analysis, though for long windows of observation, the error approached that seen in feature extraction.

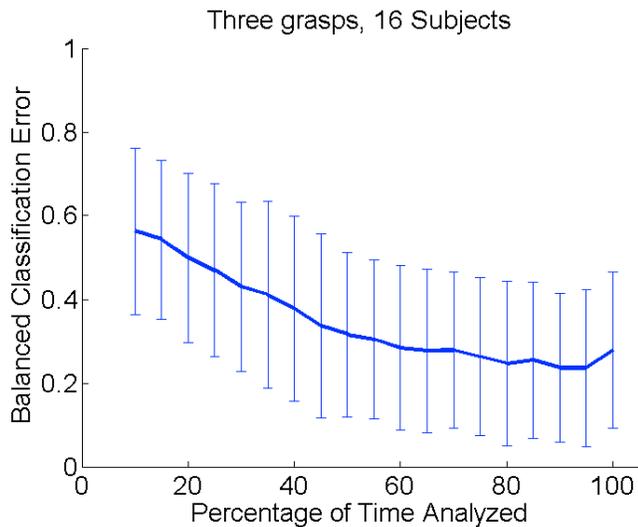


Figure 3: Balanced classification error in within-subjects grasp prediction task following correlation analysis. Trend is approximately linear through  $T=60\%$ .

Here, the decrease in predictor error appeared to decrease more consistently throughout the small windows of observation, and attenuating over larger windows. The trend minimum does not occur until the  $0.90 \cdot T - 0.95 \cdot T$  range.

## DISCUSSION

### Classifier Accuracy

The classifier errors from this study (BCE=0.2-0.4 following feature extraction and 0.2-0.6 following correlation) were in general substantially lower than chance rate (BCE=0.67), but substantially higher than results reported elsewhere for the same grasp types in the NINAPRO dataset [2], or similar discrimination tasks [4-5], (typically BCE<0.10). We believe that the most likely explanation for this is the classifier design: here we use a simplistic linear routine for classification, where the other referenced works incorporate a quadratic programming approach (i.e. support vector machines, SVMs), which have been shown to provide higher performance [6]. Furthermore, within the feature-based classification, the features used here were chosen with an interest in retaining simple, intuitive descriptors of the EMG trace; we make no assertion of the optimality of these features and fully recognize that there may be better features available. Our interest for this work was not *per se* to build the optimal classifier, but rather to quantify classifier performance over different windows of observation.

### Study Limitations

That the simplistic classifier built here was not able to reach the very high success rates of SVM-based classifiers reported elsewhere raises the question of whether the time-

dependent outcomes of this study would not change with a different classification paradigm. While it seems reasonable to speculate that the trend over time (Figures 2 and 3) might look similar (as only the classification routine changes and not the substrate of analysis), this is no direct evidence to support this hypothesis and it would need to be tested.

Another constraint is that the practicability of a study like this (whether SVM-based or not) is not immediate. Perhaps the most essential constraint here is that all data were temporally normalized, and analyses were performed on data truncated as a proportion of total time of movement,  $T$ . In real-world application, i.e. muscular activation being measured in real-time during a movement of undetermined duration, the final time  $T$  is not known until the activity subsides. In this way, a proportion-of- $T$  is not a viable context for decision making. However, it may be possible to leverage data as it's accrued to estimate the final  $T$ , a process which could be continually updated [7-9].

### Outlook

The ultimate goal of this work is to increase accessibility and desirability of prosthetic technology among clinicians and the end user. One pathway by which this work will enhance the user experience in having identified potential target time by which the trade-off between data accrual and prediction error can be optimized. While the results presented here are preliminary, they serve to propose a possible benchmark for optimal classification and rapid, high-fidelity grasp prediction.

In addition to replicating this study with the same or different classifiers, or the same or different signal conditioning methods, future works may be well posed to attempt to identify whether the optimal feature set changes over time, or number of features. Furthermore, this study did not consider inter-subject classifiers, which would add another potentially interesting dimension to this analysis.

## SUMMARY & CONCLUSION

Here, we analysed sEMG data of healthy subjects performing a small number of relevant grasp postures, in order to show 1) feature-extraction may be a potentially more robust pathway to grasp prediction than correlation of waveform morphology in a standard linear classifier, and 2) in a feature extraction paradigm, prediction accuracy increased non-linearly with increasing window observation with an apparent performance plateau near  $0.4 \cdot T$ . We encourage others to access the same dataset and attempt a reconstruction of this work, with their own preferred classifier, in order to refine the understanding of classifier accuracy over time, pursuant to a refined estimate of optimal time of decisioning.

## ACKNOWLEDGEMENTS

This study presented de-identified data obtained from a publicly available database.

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## FINCH: THREE-FINGERED FUNCTIONAL HAND CREATED BY 3D PRINTER

Masahiro Yoshikawa<sup>1</sup>, Yuya Taguchi<sup>1</sup>, Shin Sakamoto<sup>2</sup>, Shunji Yamanaka<sup>3</sup>,  
Tsukasa Ogasawara<sup>1</sup>, Noritaka Kawashima<sup>4</sup>

<sup>1</sup>*Graduate School of Information Science, Nara Institute of Science and Technology, NAIST*

<sup>2</sup>*Graduate School of Media and Governance, Keio University*

<sup>3</sup>*Institute of Industrial Science, The University of Tokyo*

<sup>4</sup>*Dept. Rehabilitation for the Movement Functions, Research Institute,  
National Rehabilitation Center for the Persons with Disabilities, NRCD*

### ABSTRACT

At present, there are body-powered hooks and myoelectric prosthetic hands that trans-radial amputees can use for daily operation. Though the body-powered hook has good workability, the design of the hook spoils its appearance and the harness impairs the feelings of wearing. The myoelectric hand has a natural appearance similar to the human hand and intuitive operability with a myoelectric control system. However, it is high cost and heavyweight. In this presentation, we report newly developed functional hand 'Finch'. It is lightweight, low-cost, three-fingered functional prosthesis created by means of 3D printing technology. It was designed for amputees easier to get-fit-use in their daily living as an alternative to conventional prosthesis. A simple mechanism to control fingers by a linear actuator contributes to workability, lightweight, and low cost. A control system using an inexpensive distance sensor allows intuitive operability as the myoelectric sensor at low cost. A socket is easily removable so that users can wear properly as the situation demands. It has a stylish appearance as a tool and can be produced by a 3D printer. The total weight of the hand and socket is 300 g.

## PRACTICAL IMPLEMENTATION OF ROBUST SENSOR INTERFACE FOR EMG PATTERN RECOGNITION FOR ARTIFICIAL ARM CONTROL

Fan Zhang<sup>1,2</sup> and He (Helen) Huang<sup>1,2</sup>

<sup>1</sup> Joint Department of Biomedical Engineering, North Carolina State University, Raleigh, NC, USA <sup>2</sup> Joint Department of Biomedical Engineering, the University of North Carolina at Chapel Hill, Chapel Hill, NC, USA

### INTRODUCTION

EMG pattern-recognition (PR) has been developed for control of multifunctional prosthetic arms for decades [1-5]. Compared to conventional myoelectric control, EMG PR allows prosthesis users to operate prosthetic devices more intuitively and efficiently. This is because EMG PR recognizes the user's intent by distinguishing different muscle activation patterns and controls the prosthesis function to match the user's intent. More importantly, the PR-based control scheme has great potential to control more degrees of freedom (DOFs) over conventional approaches because it does not require independent electrode sites or a one-to-one mapping between electrode site and DOFs [6].

However, EMG pattern recognition has not yet received widespread clinical application for artificial arms. One of the major challenges is the lack of a robust interface for EMG signal recordings. Since PR-based approach involves learning the muscle activation patterns of intended movements and cannot accommodate the changes in EMG signals, any disturbance in EMG sensors (e.g. severe motion artifacts, environment noise, electrode impedance change, sensor location shift, or loss of sensor contact) may decrease the EMG PR performance and cause the prosthesis to malfunction.

To address this challenge, a robust EMG sensor interface has been proposed and initially implemented in our group [7, 8]. The basic concept of this robust interface is to introduce sensor redundancy to minimize the effects of sensor fault on EMG PR performance. This interface monitors the signal quality of each individual EMG sensor, detects faulty sensor behaviours, and performs a self-recovery strategy to recover EMG PR performance by eliminating the faulty EMG sensors. This concept has been evaluated on EMG PR for locomotion mode recognition [7] and for recognizing arm motions [8]. However, two challenges have been identified in order to make this interface practical. First, the previously designed sensor fault detectors involve building complicated

faulty sensor data models [7] or optimizing a large number of detector parameters [8] customized for individual EMG sensor and person. These approaches complicate the prosthesis calibration procedure and lengthen the calibration time. An efficient and easy-to-apply approach is desired. Second, in the self-recovery strategy, our initial design has been to permanently remove the faulty sensors [7]. Nevertheless, some disturbances, such as motion artifacts, only contaminate EMG signals for a short time period. Such a method in this case can cause loss of information for accurate EMG PR. Therefore, instead of eliminating the faulty sensors permanently, sensor fault detectors should determine whether individual EMG sensors need to be included or excluded for each PR decision in real-time. This requires online activation of fault detectors for all the EMG inputs and frequent online retraining of classifiers. Although our preliminary study has shown the feasibility of real-time implementation of this approach when 4 EMG sensors were used [8], the real-time operation of this algorithm can be significantly intensive when the number of EMG inputs increase and when the disturbances with short durations happen frequently. Whether or not this algorithm can function without sacrificing the response time of EMG PR is unknown, but is critical to determining the practical value of the designed sensor interface.

The goal of this study was to find practical solutions to make the robust sensor interface useful for EMG PR control of artificial arms. A simple sensor fault detector and a fast retraining algorithm for pattern classifiers were implemented and evaluated in real time. The development in this study may significantly improve the robustness of EMG PR for multifunctional prosthesis control in daily practice.

### SYSTEM OVERVIEW

#### Architecture of the robust sensor interface

The architecture of the sensor interface is demonstrated in Figure 1. A sensor fault detection module (SFD) is embedded in an EMG PR system as the robust sensor interface. The SFD closely monitors the status of each individual EMG sensor and automatically recovers the classification performance

in the presence of disturbed EMG signal recordings. First, features that characterize the EMG activation patterns are extracted from each sensor and then sent to a fault detector. The detector determines the status of this sensor (either normal or abnormal) based on the distribution of the extracted

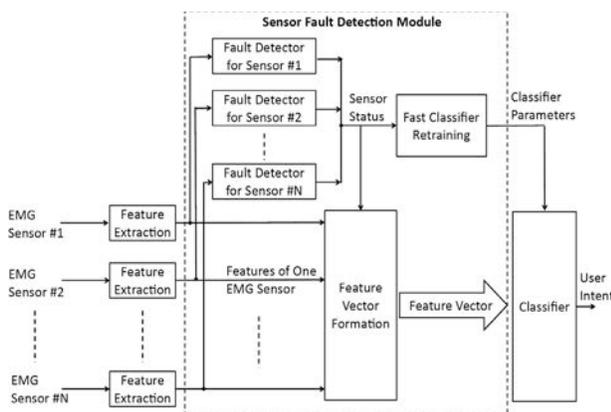


Figure 1: Architecture of the robust sensor interface

EMG features. The features from the abnormal EMG sensors are eliminated; the features from the normal sensors are then formulated into one feature vector for pattern classification. At the same time, based on the sensor status, a fast classifier retraining module updates the parameters in the modified classifier. The newly updated parameters are used in the EMG pattern classifier to identify user intended motion.

#### Fault detector and fast retraining algorithm

A multivariate Mahalanobis-distance based outlier detector was used to design the sensor fault detector. The concept is that the disturbed EMG features can be detected as an outlier, which is away from the multivariate distribution of normal EMG features. This algorithm was selected because it does not require building any faulty data model and only depends on one single threshold that can be computed easily and quickly based on the EMG PR training data in the calibration phase. In this study, the maximum value of Mahalanobis-distance in the PR training data in each EMG sensor was simply used as a threshold, aimed to ensure high fault detection sensitivity. More algorithm details can be found in [9].

A fast classifier retraining algorithm developed in our previous study was used [8]. The technical merit of this algorithm is that the updated classifier parameters can be quickly retrieved from the initial training parameters without accessing large initial training datasets. The algorithm can effectively avoid the intensive numerical matrix calculation, and therefore increase the retraining speed in real-time application.

#### Implementation

The SFD module was implemented in real-time on a PC and integrated into the UNB Acquisition and Control Environment (ACE) software package [10]. The SFD was programmed as a modular function in Matlab and can be easily switched on or off.

### REAL-TIME EVALUATION

#### Subject

This study was conducted with Institutional Review Board (IRB) approval and informed consent of the subject. One able-bodied male subject, free from orthopedic or neurological pathologies, was recruited in this study.

#### EMG signal recording

To demonstrate the practical value of our retraining algorithm, 12 bipolar surface EMG electrodes were used, which was close to the upper-limit of the number of electrodes used for EMG PR for artificial arms in existing literatures. The electrodes were placed on the right forearm of the subject with a center-to-center distance of 2cm. Six out of the twelve electrodes were uniformly placed around the proximal part of the forearm (3cm distal to the elbow crease); three electrodes were placed around the middle portion of the forearm (11cm proximal to the wrist crease); and three electrodes around the distal forearm (3cm proximal to the wrist crease). The electrode placement is illustrated in Figure 2. A ground electrode was placed over the bony area on the elbow of the tested arm. The electrodes contained a preamplifier that band-pass filtered the EMG signals between 10 and 2000 Hz with a pass-band gain of 20. The data were collected by a 16-channel EMG system (Motion Lab System, LA). The anti-aliasing filter in the system had low and high cut-off frequency at 20 and 420 Hz, respectively. The pass-band gain was 1000. The EMG signals were then sampled at 1000 Hz and streamed into the ACE data acquisition system.



Figure 2: EMG electrodes placement on the subject's forearm

#### Pattern classifier and EMG features

A linear discriminant analysis (LDA) classifier [1] was used in this study. Four time-domain (TD) features [2] (mean absolute value, number of zero-crossings, waveform length and number of slope sign changes) were used as EMG features. The analysis window size and window increment were 150ms and 100ms, respectively. The real-time

classification decisions were updated every 100ms and used to control a virtual reality arm.

Experimental protocol

Seven motion classes were included in this study, including wrist flexion, wrist extension, wrist pronation, wrist supination, hand open, hand close, and no movement.

The whole experiment consisted of two sessions: a training session and a testing session. In the training session, the EMG signals used to train the pattern classifier were collected. The subject was instructed to follow graphic demonstrations of each motion displayed on a PC monitor and perform the movements with a comfortable and consistent level of effort. All 7 motions were repeated three times, and each motion was held for 5 seconds. There was a 3-second interval between two consecutive motions.

In the testing session, the subject was asked to follow visual prompts of different motions to control a virtual arm. The virtual arm can respond to the class decisions and allow the subject to observe the real-time results of his movement commands (Figure 3). In each trial, the subject sequentially performed a series of motions in a randomized order. Each of the 6 hand/wrist motions lasted for 5 seconds and was repeated three times. Between two consecutive motions there was a 3-second interval. Ten trials were repeated. A rest period was allowed between trials. To evaluate the robustness of the designed interface against sensor disturbances, motion artifacts were purposely added by tapping on the EMG sensors when the subject performed the motions. We focused on motion artifacts as the studied disturbances in order to evaluate the function of system recovery strategy, which dynamically adjusted the features and retrained the classifiers online. The EMG sensors were randomly disturbed and the number of the sensors that were simultaneously disturbed ranged from 1 to 6. All the EMG signals, fault detection decisions and classification decisions were saved and used to evaluate the real-time performance.



Figure 3: The virtual artificial arm used in testing trials

Evaluation Metrics

The sensor fault detection performance was evaluated by using *detection sensitivity* and *false alarm rate*. The sensitivity was defined as the percentage of correctly detected signal disturbances in the total number of simulated signal disturbances. The false alarm rate was defined as the percentage of the decisions that falsely detected normal signals as disturbances in the total number of decisions without signal disturbance.

To evaluate the real-time PR performance, three clinically relevant metrics were used in this study, including (1) *the motion-selection time*, (2) *the motion-completion time*, and (3) *the motion-completion rate* [4, 5]. The *motion-selection time* was defined as the time taken to correctly select a target movement. The *motion-completion time* measured the time from the onset of movement to the completion of the intended movement, calculated as the time of the tenth correct classification. The *motion-completion rate* was defined as the percentage of successfully completed motions. More detailed information about the definition of these metrics can be found in [4, 5].

**RESULTS**

The performance of sensor fault detection was shown in Table 1. The results were averaged across all 12 EMG sensors. The detection sensitivity was reported as 100%, which meant all the artificially introduced motion artifacts were correctly detected. Even though some false alarm decisions in detectors were observed, we observed that they did not cause erroneous PR decisions due to the information redundancy in EMG recordings.

Table 1: Fault detection sensitivity and false alarm rate

Detection Sensitivity	False Alarm Rate
100%	0.07±0.04%

Figure 4 compared the motion-selection time and motion-completion time of EMG pattern recognition with and without the SFD module when EMG sensors were artificially disturbed. The real-time PR performance without the SFD module (indicated by the grey bars) was calculated offline by using the EMG signals collected during the testing sessions. Using the SFD module, when the EMG sensors were artificially disturbed, both the motion-selection time and motion-completion time were shorter than those when the SFD was not used. This indicated that the SFD module can effectively improve the real-time PR performance if the

EMG sensors were disturbed. The motion-completion rate was not shown in Figure 4, because the rate was reported 100% in each of both scenario, i.e. all the tested motions were successfully completed.

In the real-time testing trials, all the PR decisions were completed within 100ms, which indicated that the additional delays elicited by online retraining in SFD were tolerable since the response time of EMR PR was still within the acceptable range of prosthesis controller delays [11].

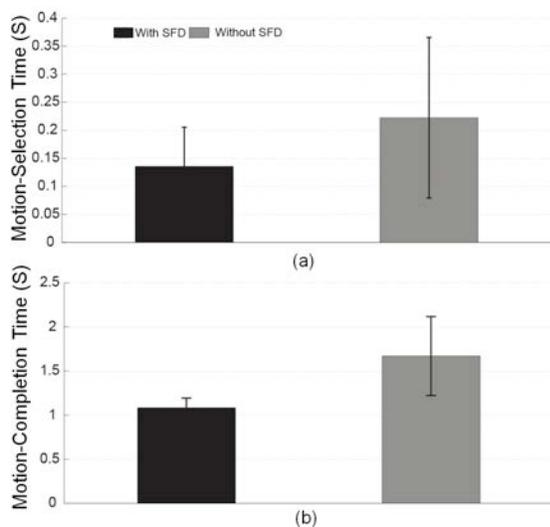


Figure 4: Comparison of (a) motion-selection time and (b) motion-completion time of EMG pattern recognition with and without SFD module when EMG sensors were artificially disturbed.

## DISCUSSION

To make the robust sensor interface practical and useful for EMG PR control of artificial arms, it is crucial to satisfy three criteria: (1) the interface can enhance the performance of EMG PR for daily use; (2) the use of robust interface should not complicate the prosthesis calibration procedure or increase the calibration time; (3) interface implementation should not sacrifice the response time of EMG PR for real-time prosthesis control. Our previous study has proved that the concept of SFD can enhance the robustness of EMG PR against various disturbances [7]. In this study we proposed practical approaches to implement this sensor interface to satisfy the latter two criteria. The results of this study demonstrate the feasibility of our developed robust interface that can further advance the practical value of EMG PR for multifunctional prosthesis control.

Sensor fault detectors usually depend on data models for abnormal recordings, which in this application are difficult and impractical to build. In this study, a simple detector design was developed and implemented for detecting disturbances in

EMG recordings. The newly designed detector only depended on the normal EMG PR training data collected during the initial calibration procedure. In addition, the threshold can be easily calculated from the initial training data. Therefore, the detector initialization can be done automatically in PR training without any extra procedure or increased calibration times. In order to ensure high detection sensitivity, this study selected the maximum distance value in the training data as the threshold for each EMG recording. Since the threshold was not optimized, the false alarm decisions wrongly caused the drop of the “normal” recordings from EMG PR. However, this action did not further elicited additional PR errors based on our observation in this preliminary study because the EMG inputs were redundant. Dropping one or more EMG recordings did not cause significant PR accuracy changes.

A fast PR retraining algorithm was successfully implemented in real-time to efficiently handle the intensive numerical computation caused by frequent online retraining. To test the algorithm efficiency and feasibility, a large number of EMG sensor inputs, i.e. twelve sensors, were used in this study. In addition, frequent motion artifacts were simultaneously added into one or more EMG sensors during the testing. The results showed that all the real-time PR decisions were able to update every 100ms. It is noteworthy that 100ms is still within the range of acceptable prosthesis controller delays [11]. This indicated that the fast retraining algorithm was an efficient solution for real-time sensor interface implementation.

The presented preliminary evaluation has limitations. Only motion artifacts were simulated as the disturbances. Our future work will focus on evaluating the robust sensor interface and EMG PR during real prosthesis control in daily practice.

## CONCLUSION

This study aimed to find practical solutions that make the robust sensor interface useful for EMG PR control of artificial arms. A simple sensor fault detector and a fast classifier retraining algorithm were implemented. The promising results showed that our designed robust sensor interface did not cause additional burdens for the prosthesis users and can improve the real-time EMG PR performance without scarifying the PR response time. Additional efforts were still needed to evaluate the sensor interface in daily prosthesis use.

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