MEC '05

Integrating Prosthetics and Medicine

University of New Brunswick's
MyoElectric Controls/Powered Prosthetics
Symposium

August 17-19, 2005
Fredericton NB CANADA

Conference Proceedings

Hosted by:
Institute of Biomedical Engineering

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Welcome from the **MEC '05** Organizing Committee

Welcome to **MEC '05**. This three-day international symposium and the related exhibits and posters will offer something for everyone in the powered upper limb prosthetics field. The organizing committee is thrilled with the response from delegates this year and **MEC '05** promises to be the biggest and best-attended yet.

The symposium features three keynote speakers and three invited speakers in addition to the many scientific papers, posters and commercial exhibits. The keynote speakers are: **Dr. Rickard Brånemark, Dr. Todd Kuiken, and Professor Malcolm MacLachlan.**

Dr. Rickard Brånemark is the Managing Director of Integrum in Göteborg Sweden. His company develops and markets direct bone anchorage technology for prosthetics.

Dr. Todd Kuiken, of Northwestern University and Director of Amputee Services at the Rehabilitation Institute of Chicago, is involved with clinical trials using the transfer of residual nerves in amputated limbs to spare muscle to develop additional myoelectric signals for intuitive, multifunction prosthesis control.

Professor Malcolm MacLachlan is Founder and Co-Director of the Dublin Psychoprosthetics Group , and is involved with the application of many aspects of psychology to prosthetic use, especially in regard to the rehabilitation of people with amputations.

This program offers plenty of time to meet other people who are attending and to visit with the exhibitors. We hope to see everyone at the ***Wine and Cheese*** reception at Old Government House on Tuesday, August 16. Old Government House is a Fredericton landmark and National Historic Site located next to the Delta Fredericton Hotel. We also hope you will enjoy the ***Maritime Kitchen Party*** on Thursday, August 18, featuring local Maritime food and entertainment.

We know you will enjoy **MEC '05** and your visit to Fredericton. Please don't hesitate to ask if we can help in any way during your stay.

MEC '05 ORGANIZING COMMITTEE

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Wednesday, August 17, 2005

8:15 a.m.	**Welcome to MEC '05**	
8:30 a.m.	**T.A. Kuiken**	**Keynote Address**
9:30 a.m.	R.D. Lipschutz	Transhumeral Level Fitting and Outcomes Following Targeted Hyper-Reinnervation Nerve Transfer Surgery
9:45 a.m.	K.A. Stubblefield	OT Training and Outcomes Before and After Successful Targeted Hyper-Reinnervation Nerve Transfer Surgery: Two Case Studies
10:00 a.m.	***Refreshment Break/ Vendor Displays***	
10:30 a.m.	L.A. Miller	Fitting and Outcomes of a Bilateral Shoulder Disartic Amputee Following Targeted Hyper-Reinnervation Nerve Transfer Surgery
10:45 a.m.	Richard F. ff Weir	Powered Humeral Rotator For Persons with Shoulder Disarticulation Amputations
11:00 a.m.	T. W. Williams, III	A Seven-Degree-Of-Freedom Arm with Unique Shoulder Function
11:15 a.m.	L.A. Miller	Shoulder Disarticulation Fitting with 6 Independently Controlled Motors After Targeted Hyper-Reinnervation Nerve Transfer Surgery
11:30 a.m.	J.W. Sensinger	Phantom Limb Sensory Feedback Through Nerve Transfer Surgery
11:45 a.m.	**Live Demonstration**	**with User of Targeted Reinnervation**
12:00 p.m.	***Lunch Break***	
1:00 p.m.	J.M. Miguelez	Acute Prosthetic Upper Extremity Rehabilitation of Blast Wound Injuries: A 21 Month Review
1:30 p.m.	K. Gulick	A Training Progression for the Upper Limb Myoelectric Prosthetic User
1:45 p.m.	J.E. Uellendahl	Custom Silicone Sockets for Myoelectric Prostheses
2:00 p.m.	J.R. Ronald	Individual Silicon Interface for Upper Limb Myoelectric Prostheses
2:15 p.m.	M.J. Mikosz	Silicone Bladder Suspension for the Wrist Disarticulation Level Amputee Using a Mini Pump System to Alter Volumetric Pressure Inside the Socket
2:30 p.m.	S. Ramdial	Integrating Electrodes in Silicone and a New Application for Myoelectric Suction Sockets
2:45 p.m.	P. McGahey	The Difference a Terminal Device Can Make
3:00 p.m.	***Refreshment Break/ Vendor Displays***	
3:30 p.m.	T. W. Williams, III	Invited Speaker

4:00 p.m.	M.C. Carrozza	An Adaptive Prosthetic Hand with Compliant Joints and EMG-based Control
4:15 p.m.	D. Russell	A Bench-Top Prototype of a Variable Stiffness Prosthesis
4:30 p.m.	J.W. Sensinger	Non-Backdrivable Series Elastic Actuator for Use in a Prosthetic Elbow
4:45 p.m.	D. Cotton	Thick-Film Piezoceramic "Slip Sensors" for a Multifunctional Prosthetic Hand
5:00 p.m.	Stewart Scotland Hill	Electromechanical Analysis of a Complete Arm Prosthesis (Emas)
5:15 p.m.	***End of Day Comments***	

Thursday, August 18, 2005

8:15 a.m.	**Daily Notices**	
8:30 AM	**Rickard Brånemark**	**Keynote Address**
9:30 AM	Trond P. Schonhowd	Implant Supported Suspension Of Trans-Humeral Prostheses
9:45 AM	Richard F. *ff.* Weir	Implantable Myoelectric Sensors (IMES)
10:00 AM	***Refreshment Break/ Poster Session***	
10:30 AM	Todd R. Farrell	Comparison Of Surface Vs. Implanted EMG For Multifunctional Prosthesis Control
10:45 AM	Levi Hargrove	A Comparison Of Surface And Internally Measured Myoelectric Signals For Use In Prosthetic Control
11:00 AM	Isamu Kajitani	Developments Of Myoelectric Controllers For Hand Prostheses
11:15 AM	Ben Jones	Improved Control For An Artificial Arm
11:30 AM	Stewart Scotland Hill	A Minimal Jerk Prosthesis Control System
11:45 AM	Blair Lock	Real Time Myoelectric Control in a Virtual Environment to Relate Usability vs Accuracy
12:00 PM	***Lunch Break***	
1:00 PM	Craig W. Heckathorne	Invited Speaker
1:30 PM	David Wells	An Electromechanical Quick-Connect Mechanism For Myoelectric Prostheses Using Silicone Sleeves
1:45 PM	Oyvind Stavdahl	The Ntnu Revolute Wrist Device (Nrwd): A Kinematically Opitmized Externally Powered Wrist Prosthesis
2:00 PM	Craig W. Heckathorne	Evaluation Of A Prototype Electric-Powered Partial-Hand Prosthesis
2:15 PM	Dick H. Plettenburg	The Wilmer Appealing Prehensor

2:30 PM	Christian Pylatiuk	Preliminary Experience With Hydraulically Driven Hand Prostheses
2:45 PM	Robert (Bob) Radocy	Hybrid And Alternative Prosthetic Designs For Sports And Recreation
3:00 PM	***Refreshment Break/ Poster Session***	
3:30 PM	Stephen Mandacina	Electronic Options & Socket Design For Partial Hand Patients
3:45 PM	Christopher Lake	Experience With Electric Prostheses For Partial Hand Presentation With The Thumb Intact
4:00 PM	Jack E. Uellendahl	Management Of The Very Short/Humeral Neck Transhumeral Amputee
4:15 PM	Stephen Mandacina	Multiple Impairments Overcome by Socket Design and Appropriate Components
4:30 PM	JW 'Bill' Limehouse	Assessments, Conciderations And Fitting Of A High Trans Humeral High Level Brachial Plexus Injured
4:45 PM	John R. Zenie	Externally Powered Management Of The Quadramembral Amputee Using A Modified Thoracic Suspension Orthoses As A Platform
5:00 PM	Kevin Towers	Voice Recognition For Prosthetic Control: A Case Study
5:15 p.m.	***End of Day Comments***	

Friday, August 19, 2005

8:15 a.m.	**Daily Notices**	
8:30 AM	**Malcolm McLaughlin**	**Keynote Address**
9:30 AM	Denis Sennett	Biofeedback: An Enhancement of Amputee Rehab
9:45 AM	Judith Davidson	A Comparison of Upper Limb Amputees and Patients with Upper Limb Injuries Using the DASH
10:00 AM	***Refreshment Break/ Vendor Displays***	
10:30 AM	Liselotte Hermannson	Invited Speaker
11:00 AM	Josie Pesch-Batenberg	Functional Status of Children with a Congenital ULRD
11:15 AM	Laurien Buffart	Reliability and Validity of Functional Tests and Questionnaires for Children with ULRD
11:30 AM	S. Lambregts	Functional Outcome of Adolescents and Young Adults with Congenital Upper Limb Deficiency
11:45 AM	Bill Limehouse	A Preliminary Study of Useage of 40+ Upper Extremity Patients Using the Animated Control System
12:00 PM	Andrea Cutti	Performance Evaluation of the new Ottobock "Bionic Arm" by Means of Biomechanical Modelling

12:15 PM	***Lunch Break***	
1:15 PM	Harold H. Sears	Evaluation Studies Of Flexion/Extension In Electric Terminal Devices
1:30 PM	Harold H. Sears	A New Electric System With Simultaneous Elbow & Hand Control
1:45 PM	Christian Hell	Advances In Responsiveness And Grip Speed Of Terminal Devices Boost The Development Of New Unique Control Options
2:00 PM	William J. Hanson	New Myoelectrode Options
2:15 PM	Thomas Bertels	Biomechanic Aspects And Patient Needs Open Up The Path To A Unique Wrist Joint For Myoelectric Prostheses
2:30 PM	Patrick Prigge	New Concepts In External Powered Arm Components
2:45 PM	David Gow	TouchEMAS Prosthetic Ideas
3:00 PM	***Tour of IBME***	

TRANSHUMERAL LEVEL FITTING AND OUTCOMES FOLLOWING TARGETED HYPER-REINNERVATION NERVE TRANSFER SURGERY

Lipschutz, R.D., [1,2] Miller, L.A, [1,2] Stubblefield, K.A., [1] Dumanian, G.,[3] Phillips, M.E., [1] and Kuiken, T.A. [1,2]

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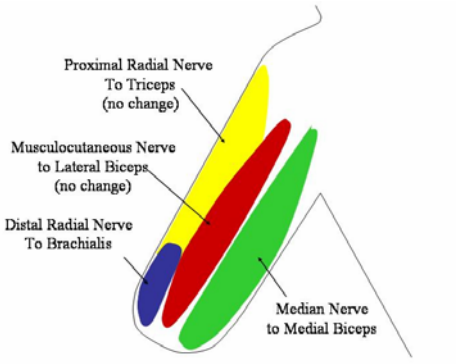
INTRODUCTION

In a typical transhumeral myoelectric system, biceps and triceps control both elbow and hand. Mode selection (frequently co-contraction) is used to switch between these two functions. In addition to requiring that these movements be performed sequentially, use of the biceps and triceps is not physiological for control of the hand. A novel approach for simultaneous control of multiple myoelectric functions was developed. This was made possible by 'Targeted Reinnervation'; a surgical intervention, which involves the transfer of the peripheral nerves that used to provide signals to the forearm for hand function, to remaining muscles on the transhumeral limb.

SURGERY

Surgery was performed on a 43 year old individual who sustained a transhumeral amputation secondary to a motor vehicle accident in August of 2002. The goal of the surgery was to create 2 new myoelectric signals that were more physiologically associated with hand open and hand close. During surgery first the biceps muscle was separated into its two heads, medial and lateral. Then the brachialis muscle and medial head of the biceps were denervated from the musculocutaneous nerve and the nerve ligated. The amputated distal radial nerve and the median nerve were then cut back to remove the neuroma and sutured onto the denervated brachialis and medial biceps muscles, respectively, so that the transfer nerve abutted the distal nerve segments of the denervated muscle. Subcutaneous fat was then removed in order to increase the transmission of the myoelectric signals to the surface of the skin. With these procedures being performed, it was anticipated that the distal radial nerve signal to the brachialis muscle would elicit a muscular contraction that is consistent with hand open while the medial nerve signal to the medial biceps would elicit a muscular contraction consistent with hand close.



Transhumeral Limb Following Muscular Reinnervation

RECOVERY

Initially, following the surgery, the individual was unable to use his prosthesis secondary to post-surgical swelling. Shortly after this period, it was noted that the transhumeral socket was not longer fitting appropriately due to the significant reduction in size of the limb secondary to the removal of subcutaneous fat; therefore, he was fit with a new socket using the original two

myoelectric sites. Approximately six months after the surgery, the individual was able to independently contract his triceps, brachialis, medial and lateral heads of his biceps muscles.

PROSTHETIC FITTING

Pre-surgical fitting of a conventional prosthesis

The initial prosthesis for this individual consisted of a transhumeral myoelectric prosthesis with chest strap, flexible suction socket within a laminated frame, two site myoelectric control, linear transducer, Liberating Technologies, Inc. Boston Digital Arm, Otto Bock wrist rotator and Otto Bock hand. The individual chose a powered hand over a powered hook for aesthetic reasons. For a short period of time, the control scheme was designed to use the two myoelectric sites (over biceps and triceps) to control both the elbow and hand. This was performed with a "time out" from elbow to switch to hand and a co-contraction of lateral biceps and triceps as a means of mode selection back to elbow. The linear transducer was activated with scapular or biscapular protraction and was used to control wrist rotation only. Although the individual could operate the prosthesis with this control scheme, the clinic team felt that use of his biceps and triceps, for hand close and open respectively, would prove confusing following the surgery. This is due to the fact that the individual would get use to using these muscles for hand operations and may later interfere with his attempt to have isolated, simultaneous control of hand and elbow.

It was then decided to use the existing components in a different control scheme. The two-site myoelectric control was then used exclusively for control of elbow flexion and extension. Hand and wrist were controlled via the linear transducer in a "slow vs. fast" pull scheme. For instance, a slow pull initiation of the linear transducer would begin to close the hand while a fast pull initiation of the linear transducer would open the hand. Once the type of pull selected the direction, the motion could then be controlled proportionally by the amount of pull on the linear transducer. As expected, the ability to pull on the linear transducer slowly proved challenging. Because of this; we incorporated a length of elastic webbing parallel to the line of pull of the transducer in order to provide a reminder to the user that he was initiating the hand or wrist motion. Wrist control was performed in a similar fashion. Mode selection between hand and wrist was now accomplished via a bump switch that was mounted on the medial aspect of the humeral section.

Post-surgical fitting of the experimental prosthesis

Approximately 4 months after surgery there were noticeable contractions of the brachialis and medial biceps muscles. The individual underwent extensive myoelectric testing in order to isolate signals to his four different myo-sites. This isolation proved difficult due to the proximity of the medial biceps and triceps sites. Although the triceps muscle covers a fairly broad area on the posterior arm, the best location for a consistent myoelectric signal was on the superior, medial aspect of his arm just distal to the axilla. The brachialis muscle had a detectible contraction through palpation although it was weaker than the other three sites. Additionally, the synergistic control of muscle groups, i.e. elbow flexion/hand close and elbow extension/hand open proved challenging. When strong contractions of the lateral biceps were initiated for rapid elbow flexion, the hand would often close as well. Similar results were seen for triceps and hand open. Six months after the surgery, the individual was fit with his first 4-site myoelectric prosthesis for simultaneous, bi-directional control of the hand and elbow. The components being used were identical to the pre-surgical set-up with the exception of two additional electrodes. Also, the programming scheme within the microprocessor had to be altered to recognize these two additional electrodes and their respective functions. Wrist rotation continued to be controlled via the linear transducer in a "slow vs. fast" scheme as previously described. A chest strap and suction socket continued to be used for

the experimental design with particular attention paid to the location of the electrodes. The site for the brachialis muscle was located distally in the postero-lateral quadrant. This proved challenging as any slipping of the socket would create and inadvertent "hand opening" signal. The chest strap was modified with an additional, adjustable strap that originated from the posterior part of the chest strap and ran across the shoulder through the delto-pectoral groove. This strap provided increased suspension of the prosthesis necessary to reduce any socket pistoning.

OUTCOMES MEASURES

A series of tests showed significant improvement in function for this individual using his experimental prosthesis vs. the conventional myoelectric fitting. Subjectively, this individual preferred the operation of the experimental prosthesis, as the movements appeared smoother and more natural. The patient had voluntary, simultaneous control of the hand and elbow while controlling his wrist rotator with biscapular protraction. Objectively, the individual was able to increase the speed of his movements 3-fold in a Box and blocks test. This is a validated test where individuals are required to move blocks from one side of a box, over a divider, to the other side. (Table 1) The Box and blocks test primarily required the use of only the powered elbow and hand. A Clothespins test was also performed that required the subject to use the powered wrist as well. This was a timed test with requirements relocating three clothespins. There was a marked increased speed of 54% (Table 2). The difference in speed was greater for tests where bimanual manipulation was of greater need as might be expected. On the AMPs testing his motor score increased from 0.51 with the conventional myoelectric prosthesis to 1.05 on the experimental prosthesis and process scores increased from 0.35 to 1.10. The AMPs test is a validated experiment assessing an individual's performance with Activities of Daily Living (ADL's).

Table 1: Comparison of Box and blocks test between conventional and experimental prosthesis (number of blocks moved in 2 minutes)

	Conventional	Experimental
Trial 1	9	20
Trial 2	5	22
Trial 3	8	26
Average	7.3	22.7
% difference		+322%

Table 2: Comparison of clothespin test between conventional and experimental prosthesis (time in seconds to move 3 clothespins)

	Conventional	Experimental
Trial 1	52	45
Trial 2	75	27
Trial 3	61	30
Average	62.7	34.0
% difference		-54.2%

DISCUSSION

This was the first attempt to perform targeted muscle reinnervation on an individual with a transhumeral amputation. The surgery proved successful with reinnervation of both the brachialis and medial head of the biceps muscles. Although the technical fitting of the prosthesis was somewhat challenging, independent myoelectric sites were obtained for use of simultaneous control of an electronic elbow and hand. Physiological control was achieved by transferring the peripheral nerves that use to control the hand to denervated muscles in the transhumeral limb. These targeted muscles were then used as biological amplifiers to send the signal to surface electrodes within the

transhumeral socket and amplified to control a myoelectric hand in a more natural manner than conventional myoelectric, transhumeral fittings. This "natural physiological control" proved effective in the individual's ability to control the prosthesis. Evidence of this is seen in the measured outcomes of various tests.

Several challenges were encountered during the fitting and adjustment of the prosthesis. Accuracy in donning the socket so that the electrode placement was optimal over the intended myo-sites was the primary challenge. The movement of the soft tissue around a singular bone made it possible for the myo-sites to shift depending upon the technique in which the socket was applied. A donning aid was utilized with this suction socket in order to approximate a "hydrostatic" fit. Electrode gain and threshold adjustments were frequently performed during the fitting and training processes. Fortunately, the visual display of these parameters, provided in the software by Liberating Technologies, Inc., proved invaluable in the understanding of which muscles were contracting, at what strength they were contracting and with which other muscles were they co-contracting. Lastly, as is the case in many externally powered fittings, the individual complained that both the pre- and post-surgical prostheses were "heavy".

There was no sensory reinnervation of the skin, which had been seen in a previous surgery involving an individual with bilateral shoulder disarticulation amputations. This is likely due to two things. First, the nerve anastomosis was underneath a thicker muscle so that the sensory fiber had more difficulty regenerating to the skin. Second, the skin was not denervated so that it was not receptive to reinnervation. In the future, we hope to provide targeted sensory reinnervation to transhumeral amputees by identifying discreet skin nerve segments and mobilizing the cut nerve down to the nerve transfer anastomosis so that the sensory nerve fibers can reinnervate skin on the residual limb and provide a pathway to give the amputee the 'feel' of the objects he is touching.

CONCLUSION

It is apparent from both the subjective and objective data that this targeted muscle reinnervation proved effective for this individual's ability to control his transhumeral prosthesis. Although any muscle could have been used to provide input for the hand and potentially provide simultaneous, myoelectric motor movements, the physiological basis for using the peripheral nerves that are consistent with hand open and close allowed the user to control the device in a natural manner without having to re-train his mind and body to substitute an artificial movement in order to control the myoelectric hand. Additional subjects have subsequently undergone similar surgical procedures with anticipated results similar to this first subject.

OCCUPATIONAL THERAPY OUTCOMES WITH TARGETED HYPER-REINNERVATION NERVE TRANSFER SURGERY : TWO CASE STUDIES

Stubblefield, K.A.,[1,2] Milller, L.A.,[1,3] Lipschutz, R.D.,[1,3] Phillips, M.E.,[1,2] Heckathorne, C.W.,[1,4] Kuiken, T.A.[1,3] : [1]Neural Engineering Center for Artificial Limbs, Rehabilitation Institute of Chicago [2]Sensory Motor Performance Program, Rehabilitation Institute of Chicago[3]Dept.of PM&R, Feinberg School of Medicine, Northwestern University[4]Rehabilitation Engineering Research Center, Northwestern University , Chicago, IL

INTRODUCTION

The control of prostheses, both externally powered and body powered, increases in complexity with higher levels of amputation. The externally powered prosthesis has a limited number of options for controlling multiple joints myo-electrically. Some method is necessary to switch control between functions (ie: elbow and hand). Targeted hyper-reinnervation nerve transfer surgery has the potential to greatly improve control of the electric prosthesis for the above elbow and shoulder disarticulation subjects by increasing the number of control options available.

When the limb is lost the Brachial Plexus typically remains intact. The nerve supply to the missing limb is viable and connected to the motor cortex, but the motor end points served are gone. In nerve transfer surgery, the peripheral nerve is relocated to an area of denervated muscle tissue in the residual limb –a muscle that no longer moves the missing limb. Hyper-reinnervation occurs resulting in an area of say the Biceps, being controlled by the Median Nerve (in the intact limb, the Median Nerve supplied finger and wrist flexors). A muscle contraction occurs in the graft area of the Biceps when the subject attempts to close his hand. A myo-control site is added if the subject can isolate the contraction from that of the Biceps muscle served by the Musculocutaneous Nerve distribution.

SUBJECT #1

The first experimental surgical procedure was preformed on a gentleman with bilateral shoulder disarticulation. The successful nerve transfers on his non-dominant side included Musculocutaneous, Radial and Median Nerves transferred to denervated areas of the Pectoralis Major muscle. The Ulnar Nerve transfer to the relocated Pectoralis Minor muscle was unsuccessful. The Median Nerve transfer ultimately resulted in two distinct areas of re-innervation which corresponded to "hand close" and "hand open" (probably thumb abduction). The area of Musculocutaneous re-innervation corresponded to elbow flexion and the Radial Nerve distribution corresponded to wrist extension/elbow extension.

The subject had previously been fitted and trained with a body powered prosthesis on his dominant right side and an externally powered hybrid prosthesis using FSR's to control the elbow and hand on the left. Training post-operatively consisted of isolating contractions corresponding to the nerve transfers. Myoelectric testing ultimately revealed the following control sites: elbow flexion (Musculocutaneous Nerve); hand close (Median Nerve 1); hand open (Median Nerve 2); elbow extension (Radial Nerve). The subject quickly learned to control the prosthesis once electrodes were installed in the socket and necessary adjustments were made. He managed simultaneous myoelectric control of his elbow and hand that physiologically corresponded to his intention.

OUTCOMES

The outcome measures used were chosen for their ability to detect the differences between the hybrid control system and the experimental prosthesis.

The Box and Blocks[1]

The Box and Blocks is a grasp and release test requiring lifting 1" wooden blocks from a well on the side of the arm being tested, over a divider, to deposit into another well on the contralateral side. The sequence is: position the terminal device in the same side box (extended elbow), grasp a block, flex elbow, extend elbow, release block, flex elbow, extend elbow, grasp etc. The standardized test measures the number of blocks moved over the barrier in one minute. For our purposes, the subject

performed the task 3 times and 2 minutes was allowed for each trial. It was anticipated that simultaneous, intuitive control of the elbow and hand would be reflected in the number of blocks moved.

Table 1. Comparison of **Box and Blocks** test for touch pad controlled prosthesis and nerve-muscle graft controlled prosthesis.

	Touch pad control Number of blocks	Myoelectric control Number of blocks
Trial 1	5	10
Trial 2	5	14
Trial 3	7	18
Average	5.7	14

The Clothes Pin Relocaton Task

The Clothes Pin Relocation Task used a standard clothes pin tree commonly found in an occupational therapy clinic for strength and coordination of pinch, forearm rotation and reach. The task was to move three clothes pins from a horizontal bar to a vertical bar. This task required more precise pre-positioning of the terminal device and incorporated wrist rotation, elbow flexion and extension. The task was timed and, as with the Box and Blocks, was repeated three times. Again, it was anticipated that the simultaneous intuitive control of the elbow and hand would enhance pre-positioning, improve the smooth appearance of reach and decrease the amount of time required to perform the task.

Table 2. Comparison of **Clothes Pin Relocation Task** for touch pad controlled prosthesis and nerve-muscle graft controlled prosthesis.

	Touch pad control Time (sec)	Myoelectric control Time (sec)
Trial 1	153	83
Trial 2	137	122
Trial 3	121	99
Average	137	101

Functional Bimanual Tasks

The subject required set up, adaptive equipment, and frequent physical assistance for feeding, upper body dressing, oral/facial hygiene, and telephone communication. He required total assistance for toileting hygiene, lower body dressing (except shoes and boxers) and bathing. He was evaluated and recommended for a foot steering motor vehicle modification. While his ADL status remained unchanged using the experimental prostheses, there was a visible improvement in the quality of movement. He reported that the new experimental arm was much easier to use. He used his prostheses for yard work and during household tasks regularly. The new experimental prosthesis in fact, added some functional advantage. Being easier to use, the subject was more likely to wear it and be engaged in purposeful occupations in and around his home.

SUBJECT #2

The second subject was a gentleman with a unilateral transhumeral amputation on his dominant side. His terminal device of choice was a hand. Prior to the surgery, the subject was independent in all basic and instrumental ADLs using one-handed techniques. The successful nerve transfers

included the Median Nerve to the denervated Medial Head of the Biceps, and the Distal Radial Nerve to the denervated Brachialis.

The subject had previously been fitted and trained with a myoelectric prosthesis using the Musculocutaneous Nerve (Biceps for elbow flexion and hand close) and the Radial Nerve (Triceps for elbow extension and hand open). Co-contraction of the Biceps and Triceps switched function between the elbow and the terminal device. During the recovery period after surgery, a bump switch was installed to change functions in order to discourage co-contraction and encourage isolating of signals. Myoelectric testing ultimately revealed four control sites that physiologically corresponded to a natural limb for elbow flexion, extension and hand open and close. He was able to simultaneously control both hand and elbow positions. Wrist rotation was via a pull switch mounted in the harness

OUTCOMES

The Box and Blocks and Clothes Pin Relocation Task

The outcomes measures used with this subject included those used with the first subject and predictably demonstrated increased speed of performance and improved quality of movement.

Table 3. Comparison of **Box and Block** test for the original myoelectric control prosthesis and nerve-muscle graft controlled prosthesis.

	Original 2 Site Myoelectric Number of blocks	4 Site Myoelectric Control Number of blocks
Trial 1	2	20
Trial 2	4	22
Trial 3	5	35
Average	3.6	25.6

Table 4. Comparison of **Clothes Pin Relocation Task** for the original myoelectric controlled prosthesis and nerve-muscle graft controlled prosthesis

	Original 2 Site Myoelectric Time (sec)	4 Site Myoelectric Control Time (sec)
Trial 1	103	45
Trial 2	110	25
Trial 3	52	32
Average	88.3	34

Bimanual ADL Tasks

In the interest of evaluating the subjects ability to integrate the prosthesis into functional activities, nine bimanual activities were chosen that varied in complexity and allowed the subject to choose how best to use the prosthesis. The tasks were timed, although the subject was not instructed to perform them "as quickly as possible". Each task was performed once during the evaluation.

Table 5. Comparison of the time required to perform bimanual activities

TASK	**Time (sec) (2 Sites)**	**Time (sec) (4 Sites)**
1) Cut food using knife & fork (toast)	90	11

2) 3 items onto a try & carry (roll of tape, small plastic bowl and a plate)	114	37
3) Put 3 soup cans into a grocery bag with handles	109	35
4) Open a jar of peanut butter (plastic)	14	8
5) Stir with a spoon in a mixing bowl	32	3
6) Open a letter using a tool (letter opener)	21	17
7) Pull on socks	28	28
8) Don and doff and button down LS shirt	116	94
9) Wrap a package	356	97

The first evaluation was administered 7/03 and the second 4/04. It is difficult to assess whether the faster performance is a reflection of nine months experience using an artificial limb or attributable to the experimental procedure and prosthesis.

The Assessment of Motor Process Skills (AMPS)[2,3]

The AMPS is an observational assessment that is used to measure the quality of ADL. Performance is assessed by rating the effort, efficiency, safety and independence of 16 ADL motor and 20 ADL process skill items.

The transhumeral subject scores indicated that there were problems with motor skills that affected the quality or effectiveness of task performance. Though improved at the second evaluation, there continued to be problems. Notable improvement was evidenced in Process Skill scores. At the second evaluation, scoring was above the cutoff, indicating that there were no longer problems in the area of process skills. This was anticipated because of the more direct connection between the motor cortex through the peripheral nerve to produce movement responses in the experimental prosthesis..

The UNB Test of Prosthetic Function[4]

The UNB will be the assessment tool for future research subjects with unilateral amputation. It will replace the ADL Tasks above as a functional assessment. While it is not a timed test, it yields scores for "skill" using the prosthesis and "spontaneity" using the prosthesis. It is anticipated that it will capture the differences between the pre-operative and experimental prosthetic solutions.

CONCLUSION

The targeted hyper-reinnervation nerve transfer surgery has been successful for the 2 subjects discussed here. The functional outcomes for both these experimental prostheses are favorable using the measurements described above, however functional ADL status remains unchanged. Both subjects report that the experimental prosthesis is easier to use than the pre-operative prescription. Further experiments with sensory feedback resulting from cutaneous nerve re-generation is promising. In addition, the development of a six motor artificial arm for the shoulder disarticulation subject has the potential to improve functional ADL status by adding shoulder flexion and humeral rotation giving the subject access to a far larger workspace.

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FITTING AND OUTCOMES OF A BILATERAL SHOULDER DISARTICULATION AMPUTEE FOLLOWING TARGETED HYPER-REINNERVATION NERVE TRANSFER SURGERY

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INTRODUCTION

With higher levels of amputation, there are more movements and prosthetic joints that need to be controlled; yet there are fewer control signals remaining to control these multiple degrees-of-freedom. Traditional fitting of a shoulder disarticulation amputee with a myoelectric system uses 2-sites and sequential control. This can be tedious and slow. When an amputation has occurred, the musculature is gone, however, the nerves that controlled the arm remain.

The goal of the targeted hyper-reinnervation nerve transfer surgery was to create additional sites using these nerves to allow simultaneous control of multiple movements using more natural control schemes [1,2,3]. Following an experimental nerve transfer procedure, 4 new myoelectric signals were created on the left pectoralis muscle for a single bilateral shoulder disarticulation (BSD) amputee using nerves that previously controlled hand and elbow function. Subsequent prosthetic fitting found that the user was able to operate the elbow and hand in a coordinated fashion and various outcome measurements showed and improvement in prosthetic function.

SURGERY & RECOVERY

The first patient to undergo targeted hyper-reinnervation nerve transfer surgery was a 54 year-old man working as a high-power lineman, who suffered severe electrical burns in May, 2002, requiring BSD amputations. Seven months after the initial amputations, surgery was scheduled to excise a painful split thickness skin graft. It was also proposed to transfer the residual brachial plexus nerves on to the chest wall. The neurovascular innervation of the pectoral muscles were identified, the nerve branches to the muscles were cut and the proximal nerve segments were ligated. The musculocutaneous nerve was anastomosed to the clavicular head of the pectoralis (p) major muscle. The median nerve was transferred to the superior segment of the sternal head of the p major muscle. The radial nerve was sewn on to the inferior segment of the p major. Finally the ulnar nerve was transfer to the p minor which was moved out from under the p major onto the lateral chest wall to prevent the p minor EMG from contaminating that of the p major and to

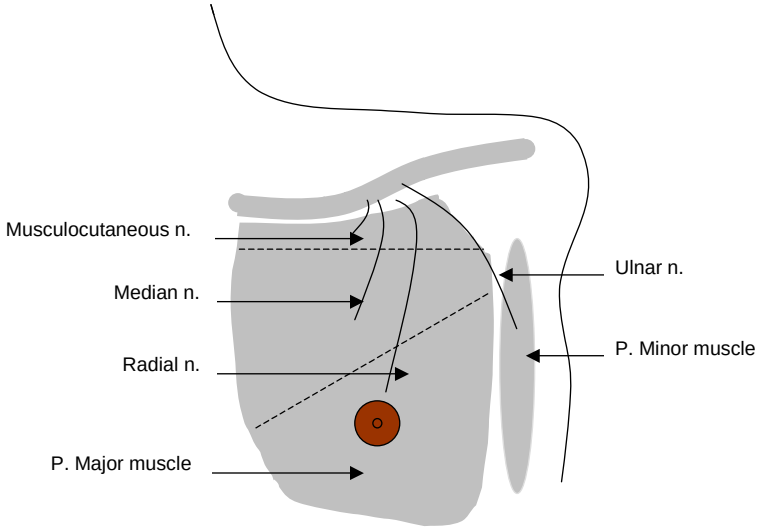

Figure 1: Surgery performed on Shoulder Disarticulation patient

serve a fourth muscle target. All subcutaneous fat over the target muscles was excised so that the skin was as close to the muscle as possible in order to provide for the strongest surface EMG signals with the least cross-talk between reinnervated muscle segments.

The patient was able to use his previous FSR controlled prostheses within a few weeks of surgery. After 3-4 months the first twitches of reinnervation were apparent. After 6 months, he could be fit with a myoelectric prosthesis. The key difference between the conventional prostheses and the myoelectric prostheses was the control systems.

PROSTHETIC FITTING

Conventional Prosthesis

Initially, the right side was fit with a body-powered system, including a positive locking shoulder, elbow, wrist rotator and body-powered hook. The left side was fit with an externally powered system with a positive locking shoulder, Boston Digital Arm, Otto Bock wrist rotator and Greiffer. The prosthesis was controlled via 4 FSRs mounted in the socket. Two anterior placed FSRs controlled hand open and close. A superior mounted FSR controlled elbow flexion in a FSR-servo manner. With this control, as pressure is applied to the FSR, the prosthetic elbow flexes. When held in place, the elbow would hold in that position. To lower the elbow, pressure is applied to the FSR up to the level at which it was held to reactivate motion and as the FSR pressure is "eased-off," the prosthetic elbow extends proportionally.

Experimental Prosthesis

This left side was initially fit with a definitive 3-site prosthesis, with 2 signals used for the hand and one for the elbow. Due to poor signal isolation and interference from the cardiac signal, the radial nerve transfer was not used for this first fitting. Unexpectedly, two independent sites from the median nerve transfer, corresponding to hand open and close, were isolated to allow proportional control of the terminal device. The ulnar nerve transfer did not reinnervate muscle in that no EMG signal could be detected. Therefore, a single independent site, innervated by the musculocutaneous nerve, was used for elbow function, with myoelectric control of the elbow in a "myo-servo" fashion. As the muscle is contracted concentrically, the prosthetic elbow flexes and as the muscle contraction is "eased-off" as in an eccentric contraction, the prosthetic elbow extends proportionally. After approximately one year, when a new definitive device was needed, improvement in electrode placement and signal processing allowed the addition of another electrode, using the radial nerve signal, for bi-directional elbow control.

Wrist function was controlled with the same myoelectric signals that were used for the hand with switching between the two functions performed through either co-contraction or the bump of an external switch (FSR) mounted in the socket. Shoulder lock and unlock for both the body-powered prosthesis and the externally powered prosthesis (LTI-Collier shoulder joints) were controlled through the Boston Digital Arm microprocessor via FSRs mounted in the socket. Other than the shoulder lock control, the contralateral side was fit with a duplicate body-powered prosthesis.

During contraction of the muscle groups, there was substantial movement of the superficial tissue. The magnitude of the skin movement greater than the distance between the electrode sites; therefore if the skin moved within the socket, the electrodes would no longer be over the proper location. In order to maintain electrode contact in the appropriate area, considerable pressure over the electrode sites was required.

Multiple socket designs attempts were created to try and maintain the electrodes in the proper position. Initially a solid "vest" style socket was created. This did not allow adequate pressure to be applied to the socket and still allow donning. The following design used a thermoplastic socket with the electrode area cut out as a flange that could be tightened down on the skin. However the rigid plastic did not allow the socket to be tightened adequately and still be comfortable. To allow higher pressure with maximum comfort and ease of donning, the three electrodes and ground reference were mounted in a custom silicone pad connected to the socket frame. This custom silicone pad allowed a tight fit and provided friction on the skin to minimize movement (though after extended wear and perspiration, there is still movement), while still providing a comfortable interface. A protective shell was fabricated over the electrode cables and gel pad for durability.

OUTCOME MEASURES

Following fitting and training, tests were performed to compare the efficiency of the FSR controlled system to the experimental myo-controlled system. The first test performed was the box-and-blocks test. With this test, the number of blocks moved from one side of a divided box to the other side is recorded for a given time [4]. For the second test, the clothes-pin test, the time required to move 3 clothes pins from a lower horizontal bar to a higher vertical bar is recorded. For both evaluation tools, performance improved significantly in the myo-controlled setup. Additionally, the subject preferred the myoelectric design as he felt that he could perform tasks more easily and quickly and could perform additional tasks with the new set-up compared with the old.

Table 1: Comparison of Box-and-blocks test between FSR controlled prosthesis and myoelectric-controlled prosthesis (number of blocks moved in 2 minutes).

	FSR-controlled	Myo-controlled
Trial 1	5	10
Trial 2	5	14
Trial 3	7	18
Average	5.7	14
% difference		+246%

Table 2: Comparison of clothes-pin test between FSR controlled prosthesis and myoelectric-controlled prosthesis (time, in seconds, required to move 3 pins).

	FSR-controlled	Myo-controlled
Trial 1	153	83
Trial 2	137	122
Trial 3	121	99
Average	137	101
% difference		-26.3%

It is noteworthy that although the individual performed best on a box-and-blocks test with the body-powered prosthesis, the effort required was great. The performance comparing an original external powered prosthesis controlled with FSRs and the myoelectric designs following the nerve graft procedure showed significant improvement due to the fact that multiple joints could be controlled simultaneously. In addition, there were multiple activities that the user said he could do with the myo-controlled prosthesis that he was unable to do with the FSR-controlled prosthesis; these include feeding himself, donning socks, shaving and watering the yard.

DISCUSSION

The first application of the targeted muscle reinnervation technique for improved myoelectric prosthesis control is presented. The idea is fairly simple; muscle is used as a biological amplifier of the nerve signal to obtain additional independent control signals for operation of a multifunction prosthesis. Since the terminal nerve branches of the brachial plexus have discreetly different functions, independent functional controls can be recorded from each nerve-muscle unit. The signals for the prosthesis were obtained without the need for implantable wires, through commercially available electrodes.

The median nerve to p major nerve transfer had an unexpected and very useful result. Although only one nerve was sewn on to the muscle segment, two distinctly different regions of muscle activity could be identified and were used to operate the terminal device. Independent motions of his index finger and the other fingers could also be seen on the chest, indicating that the other nerves may have separated into different regions of the muscle during the reinnervation. Although only 2 independent myoelectric signals could be recorded from the area for this patient, this visual assessment of voluntary multiple degree-of-freedom movements demonstrates another exciting potential of the technique—dividing a nerve into multiple fasicles to reinnervate different regions of muscle, potentially providing even more independent myoelectric control signals.

CONCLUSION

Evaluating the success of upper limb prosthetic function is a very difficult task—especially at the level of the shoulder disarticulation amputation. For objective testing, the only standardized test that we felt was applicable was the box-and-blocks test [4]. This is a fairly simple, validated, and widely-used test. There are few other validated evaluation tools that would assess the improvement of simultaneous versus sequential control. Such tools are needed to accurately assess the improvement in performance that this procedure has provided for the subject.

The performance comparing an original externally powered prosthesis controlled with FSRs and the myoelectric design following the nerve graft procedure showed significant improvement in the box-and-blocks test, the clothes pin test and other activities of daily living due to the ability to control the multiple joints simultaneously with a more physiological direct input.

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Powered Humeral Rotator for Persons with Shoulder Disarticulation Amputations

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Abstract: There are no commercially available externally-powered humeral rotators. This paper describes the development of a new powered-humeral rotator based on an LTI Boston Elbow II drive train suitable for use on persons with short transhumeral amputations or shoulder disarticulation amputations. An initial fitting of the device as part of a six motor arm demonstrated the efficacy of the device and anecdotal evidence suggested the patient found the device to be of benefit.

Introduction: It is estimated that out of a total of about 100,000 upper-limb amputees in the USA that about 18% have trans-humeral amputations or higher [Laplante & Carlson, 1997; Millstein et al., 1985; 1986]. For the upper arm the normal range-of-motion for humeral rotation is 90° of medial (inward-towards the midline) humeral rotation and 40° of lateral (outward-away from the midline) humeral rotation. However, persons with high-level arm amputations (trans-humeral amputation or above) often do not have the mechanical advantage to rotate their prosthesis through a normal range of inward and outward humeral rotation. Without an effective means to transmit rotation from the shoulder joint to the prosthesis, users are impaired in their ability to properly orient and position the hand in space. Humeral rotation is particularly important in activities-of-daily living for those tasks that need to take place at the midline of the body such as eating, toileting and dressing.

Surgical solutions exit in the form of the Marquardt angle osteotomy which places a piece of bone at an angle of 70°-110° to the humerus creating an angled piece of bone that can be used to capture humeral rotation [Marquardt, 1992]. A variation on this idea was to use an implant in the distal end of the bone to again provide a means to mechanically capture humeral rotation.

Non-surgical prosthetic options for humeral rotation are typically achieved with manually positioned friction joints, or turntables located on the top of the prosthetic elbow. Body-powered systems such as the Rimjet body-powered humeral rotator (Rimjet Corp, Sarasota, FL) [Uellendahl & Heckathorne, 1999] or the rotation unit built into the Automatic elbow from RSL Steeper Rotator also exist. There are no externally-powered humeral rotation devices currently available.

The Northwestern University Prosthetic's Research Laboratory (NUPRL) has developed a number of humeral rotators over the years. A working prototype of a cable-actuated locking humeral rotator was built [Ruberté, 2004]. This design had 180 locking positions spaced 2° apart along a full 360° of humeral rotation. It required a cable pull force of about 2 lbs (9 N) and a cable excursion of 0.43" (11 mm). The final prototype weighed 0.6 lbs (270g). The device also had a large central opening (0.875") to allow electrical connections to pass through, thereby increasing its capability of mating with a variety of commercially available elbow joints. This device evolved from an earlier NUPRL Multi-disk Rotator prototype [McCall, 1996]. The goal

of these projects was to develop a prosthetic component that would allow the user to easily control the inward and outward rotation of the forearm about the humeral axis using cable operation.

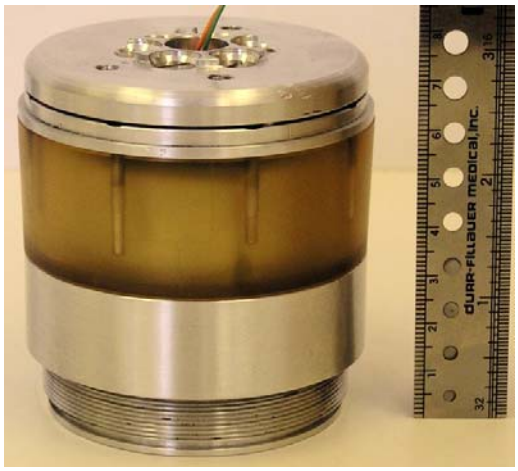

Fig. 1: Photograph of our current humeral rotator prototype.

Current Project: The goal of this project was to develop an externally-powered rotator that used myoelectric signals or electro-mechanical switches for control. This humeral rotator evolved from an old Boston I elbow drive train. It was noticed that the drive for this elbow was very compact and the authors believed it would lend itself to use as a humeral rotator simply by turning it on its side. The Boston I drivetrain consists of a brushed DC electric motor connected to the elliptical wave generator of a harmonic gear transmission. Both the motor and elliptical wave generator are housed within the flexible spline of the harmonic transmission. A harmonic gear transmission allows for high gear ratios in a compact space. The elbow case provides the annulus with which the flexible spline engages. We took an impression of the annulus using RTV rubber and then made an epoxy resin duplicate of this gear. This epoxy housing was then machined to center it and to fit it into aluminium pieces designed to provide standard interfaces to a Boston Elbow and a standard laminating ring. The laminating ring was later replaced with an interface for EMAS Shoulder joint so that the humeral rotator could be used in a 6 DOF arm fit to a subject who undergone targeted Reinnervation [see other papers in this proceedings for more information](**Fig. 1**). EMG signals from Latissimus dorsi and the deltoids were used to control inward and outward rotation on this subject. Current limiting provided by the Boston Elbow III motherboard was used to detect stall at either end of the range.

We plan to build a second generation humeral rotator based the drive system of an elbow we are currently developing. This drive system has a hollow shaft, which is an important feature for a humeral rotator as it allows wires to be easily and cosmetically routed from the prosthetic interface down to the distal components of the prosthesis.

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A SEVEN-DEGREE-OF-FREEDOM ARM WITH UNIQUE SHOULDER FUNCTION

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INTRODUCTION

The most advanced powered prosthetic arm in 2005 is the experimental seven-degree-of-freedom (DOF) arm prepared for a bilateral shoulder disarticulation patient at the Rehabilitation Institute of Chicago. The major components are an LTI Boston Digital Arm supplying the elbow drive and control electronics, a hand with wrist flexion-extension from China, a Bock wrist rotator, a humeral rotator from Northwestern University (NWU), a shoulder flex-extension module from TouchEMAS in the UK, an LTI Locking Shoulder Joint, and an LTI VariGrip III Controller.

THE PHYSICAL CONFIGURATION

To use the powered shoulder for both flexion and abduction, an LTI locking shoulder joint was mounted with its rotation axis vertical and its fixed element attached to the user's socket by a right angle bracket. The TouchEMAS (TE) joint has its motor mounted axially inside the humeral section of the upper arm, and it can forward flex the arm from –15º to +170º with respect to the vertical. The proximal end of the TE joint is mounted to the horizontal rotating element of the LTI joint. To avoid having a second motor for abducting the upper arm away from the body, the LTI joint can be unlocked to allow the shoulder drive and upper arm to rotate to any azimuth angle from –15º inward to +160º outward. This rotation takes the forearm with it leaving it positioned at an inconvenient angle that the user must subsequently correct by using humeral rotation. To allow the powered shoulder to move both up and down at the same speed, bungee-cord springs were added to counteract gravitational torque. The two shoulder components working together and combined with the powered humeral rotator duplicate the ROM of an intact shoulder. This mounting strategy permits the TE unit to provide any away-from-body shoulder motion from pure forward flexion to abduction with only one motor. The LTI joint is locked by alternately pulling on a cable. The lock will later be activated electrically.

Between them, the LTI and TE units provide two of the three shoulder degrees of freedom. The third DOF is supplied by the NWU powered humeral rotator placed between the upper arm and the elbow. It has a greater-than-human ROM between 120° internally and 120° externally.

Distal to the humeral Rotator an LTI Digital Boston Arm provides 135° of elbow flexion. In addition, it supplies a forearm structure, a battery, and the system control electronics.

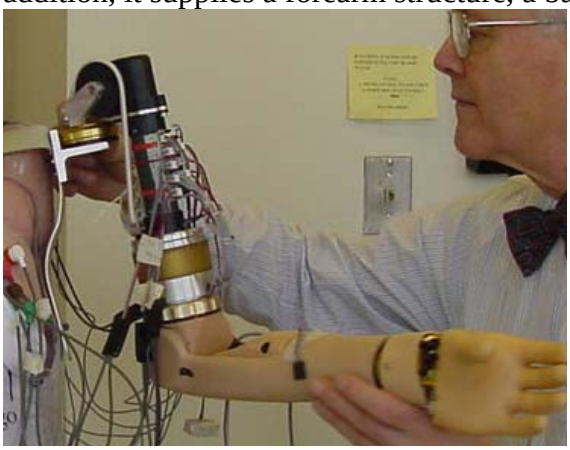

Figure 1. TouchEMAS powered shoulder mounted on an LTI locking shoulder mounted horizontally. Note the elastic cord that provides gravitational compensation.

The distal components consist of a Shanghai Keshen hand from China with an Otto Bock Quick Disconnect added locally. The hand provides both grasp using a thumb and two fingers moving together on parallel axes and wrist flexion-extension with a range of 30º each way. The Quick Disconnect allows the addition of an Otto Bock Wrist Rotator between the forearm and hand using standard Bock components. The Rotator's slip rings permit continuous rotation of the distal components in either direction. In addition they provide two pairs of conductors to supply current to the hand and wrist-flexion motors.

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CONTROLLING SEVEN DEGREES OF FREEDOM

Four DOF of the experimental arm are controlled by the Boston on-board controller. Two additional DOF use an add-on LTI VariGrip III Controller. The add-on receives its power from the main board, and its control inputs pass around the elbow joint using the digital arm's built-in wiring. One motor output on the digital arm is not used in this configuration, but it is available should the clinical team need to configure the system differently.

Input Channels Available To the Clinician

The ideal control has one pair of antagonist-muscle myoelectric signals available for each DOF. The Boston Digital Arm provides four pairs of suitable analog input channels. One pair is reserved for Boston electrode preamplifiers. These signals receive high-level of digital filtering. The other three pairs accept proportional voltages from LTI remote-amplifier electrodes, Bock socket-mounted electrode-amplifiers, Force Sensing Resistors (FSR's) or LTI linear transducers. To go from four to six DOF more channels are needed. These are supplied by using the extra pair of cross-elbow wires available on the Digital Arm to go to the LTI VariGrip III Two-Motor Controller. Note that with just two wires only single-input control algorithms can be used. Here the choice is two FSR inputs using quick-rise, slow-rise to select direction for the two motors.

LESSONS FROM THE EXPERIMENTAL ARM

The author described a shoulder mechanism at the MEC conference in 1997 [2] where the conventional DOF's called *flexion* and *abduction* are replaced by the motions *away-from-vertical* and *around-a-vertical-axis*. This is a mechanism that conserves energy. The *away* motion can be partially or fully compensated for gravitational torque, and there is no gravitational interaction with the *around* motion as long as the axis is held vertical. The experimental arm was an opportunity to try this configuration in a clinical setting. It does conserve energy, but it does not respond the way an intact arm might be expected to. In particular, motion around the vertical axis carries the forearm with it requiring an internal rotation correction afterward. The lesson is that a future energy-conserving powered shoulder will need servo controlled motors so that the controller will reposition the forearm as the plane of action of the heavy-duty shoulder motor is changed by a much weaker motor. The author is currently working on such a design.

A second lesson is that more cross-elbow lines are needed. However, there are already problems accommodating the large number of wires that have to cross both the elbow and internal rotation axes. Future complex systems should replace multiple wires to individual components with an information bus and a pair of power wires. For this, every input device will need a miniature microprocessor board to interact with the bus. Such an input board might contain preamplifiers for multiple myoelectric inputs as well as inputs from other variable voltage devices. A second type of bus board can control remote motors. In this case the board should be software configurable to operate one three-phase brushless motor or two brush-type motors and have a provision to report servo position voltages or other position data back to the main controller via the bus.

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SHOULDER DISARTICULATION FITTING WITH 6 INDEPENDENTLY CONTROLLED MOTORS AFTER TARGETED HYPER-REINNERVATION NERVE TRANSFER SURGERY

Miller, L.A, [1,2] Lipschutz, R.D, [1,2] Weir, R.W., [1,2,3] Williams, T.W., [4] Stubblefield, K.A., [1] Heckathorne, C.W., [2,3] and Kuiken, T.A. [1,2]

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INTRODUCTION

In 2002, targeted hyper-reinnervation nerve transfer surgery was performed unilaterally on a bilateral shoulder disarticulation amputee. The goal of this surgery was to create additional sites using the remaining unused brachial plexus nerves to allow simultaneous control of multiple movements using more natural control schemes [1,2,3].

As a result of the nerve transfer procedure, 4 new myoelectric control sites were created on the left pectoralis muscle. Subsequent prosthetic fitting found that the user was able to operate the elbow and hand in a coordinated fashion using three electrodes. Various outcome measurements showed an improvement in prosthetic function.

However, with the increase in the number of input signals, a goal was set to build a prosthesis with the maximum number of controlled motors available. Six motorized components were identified: three were commercially available in the USA, one was commercially available in other countries and two were a research prototype.

COMPONENTS

An experimental prosthesis was built for control by a person with shoulder disarticulation amputation with the maximum number of motorized joints currently available. The six powered functions, integrated by LTI, included a prototype shoulder, an experimental humeral rotator, a Boston Digital arm, a wrist rotator, and a hand with powered wrist flexion/extension.

A Scottish company, TouchEMAS, was the supplier of the powered shoulder joint. In a traditional mounting, this shoulder has flexion and extension movement only in the sagittal plane. However, in addition to the powered joint, an LTI manual locking shoulder was mounted horizontally to allow rotation of the powered shoulder in various planes. That is, the powered shoulder could be position to flex and extend in the sagittal plane or rotated laterally to move in the coronal plane (abduction-adduction).

The next component distally was the humeral rotator. This mechanism was developed by Weir and Grahn and is based on a Boston 2 elbow drive unit [4]. Because of the mounting technique used to connect the humeral rotator to the Boston Digital arm, access to the electrode and other input connections (located at the top of the elbow turntable) was difficult. A jumper for all of the input connectors was fabricated to facilitate the change of various input cables. This jumper board was then mounted to the humeral shaft connecting the shoulder to the humeral rotator. To prevent over-extension of the cables as they crossed over the humeral rotator, a mechanical stop was fabricated into the rotator mounting plate.

The elbow unit/controller was the Boston Digital Arm. This system allows for nine analog (myoelectric or other proportional) inputs, five digital inputs and direct control of five motors. By passing input signals through the elbow to an additional LTI VariGrip controller

located in the forearm, we were able to increase the number of motor control signals to the required six.

An Otto Bock electric wrist rotator was mounted distal to the elbow. The input cables were modified to allow the coaxial plug to contain the two control signals necessary for hand control. The Shanghai Keshen Hand (Model M21) was used. This hand has powered open and close and powered wrist flexion and extension. It was modified to fit the Otto Bock quick disconnect wrist. The wrist rotation and the wrist flexion and extension were controlled by signals from the extra VariGrip controller. The remaining four motors were controlled directly by motor signals from the Boston Digital Arm.

In addition to the motors, other components required modification. During previous myotesting, the subject had difficulties with co-contraction of radial nerve functions (elbow/wrist/hand extension) and it was not possible to separate the elbow extension signal (radial nerve transfer) from the cardiac signal (ECG). Because the subject had practiced this movement (arm extension) over the past year, he had increased the EMG signal in magnitude but it was still not great enough to isolate for two-site elbow control.

Models were developed of the cardiac interference and algorithms created to remove this interference in the input [5]. However, implementation of these algorithms proved difficult due to the processor setup. Therefore, additional models were created to investigate the use of simpler filtering techniques. Eventually, the LTI DC electrode amplifiers were modified to add a 60Hz low-pass filter and the LTI AC electrode amplifiers were modified to a narrow band pass centered at 120 Hz. This proved to be effective in reducing a majority of the ECG interference and allowed the use of the fourth control site. Improved function was achieved as this modification was done to all of the electrode amplifiers located on the chest wall.

CONTROL

The subject controlled hand opening and closing, elbow flexion and extension and humeral internal and external rotation using myoelectric signals. Unexpectedly, two independent myoelectric signals, for hand open and hand close, could be reliably recorded over the median nerve-muscle unit. Though the median nerve innervates mainly hand-close muscles, the subject imagined thumb abduction to "open" the hand. This is also a median nerve function. These two areas were used for hand open and close. The musculocutaneous nerve transfer was used for elbow flexion and the radial nerve transfer was used for elbow extension.

The subject had some remaining deltoid musculature that could be used for a myoelectric site. This signal was used to control internal humeral rotation while a latissimus dorsi site was used to control external humeral rotation. This is counter-intuitive since the latissimus dorsi muscle is an internal rotator of the humerus, however, this felt easier and more comfortable to the subject and thus was implemented.

Switches could not be placed in the harness for operation of the other functions because the harness was already in use to anchor the straps for the right body-powered system. We also preferred to avoid chin switches. To maximize the number of controls mounted in the socket, a rocker switch was positioned superiorly within the socket to control the shoulder. This rocker was modified by adding two FSR's to make the output signals proportional to the force applied to the rocker. Forward movement of the rocker flexed the shoulder joint (or lifted the arm) and backward movement of the rocker brought the arm down. A force-sensitive resistor (FSR) touch pad was mounted anterior within the shoulder cap to control wrist flexion and a second was mounted posterior to control wrist rotation. Each of these two FSRs controlled movement via a

single-site control where a soft/slow touch moved the motor in one direction and a hard/fast touch moved the motor in the opposite direction.

RESULTS

Although the subject only wore the prosthesis for about 15 hours is his first two week session, he was able to control multiple joints simultaneously, and he could perform tasks that he could not do before. Cleary, his functional workspace was greatly enhanced by the shoulder (allowing active reach up) and the humeral rotator (allowing him to bring the arm into and past mid-line. He was more efficient in doing specific tasks, such as donning a hat and shaking hands. He was also better at pre-positioning the terminal device in space; for example, by reaching forward and moving his hand directly in front of himself in a smooth coordinated movement, by reaching above his head, and also by positioning the terminal device near the mid-line of the body. The subject was able to demonstrate simultaneous control of at least 3 degrees-of-freedom by reaching up (shoulder flexion), out (elbow extension) and opening or closing the hand.

The subject found the nerve transfer EMG controlled functions to be the easiest to use. Operation of the humeral rotator with EMG control was also relatively easy. Control functions with shoulder motion were clearly more difficult, although progress was made even in the short time he had to work with the arm. Shoulder flexion/extension improved quickly. Operation of the wrist rotator and the wrist flexion/extension motors with just single site FSR touch pads using soft/slow for one direction and hard/fast for the other continues to be the most challenging, least reliable and the heaviest cognitive burden.

Figure 1: Photographs showing subject using six-motor prosthesis.

DISCUSSION AND FUTURE WORK

There was a marked increase in functional range of motion for our subject using the six-motor prosthesis. With little training, he showed an ability to control a prosthesis using the four additional control signals added through the targeted reinnervation of the pectoralis musculature. The targeted hyper-reinnervation technique makes possible the creation of new control signals for even more complex prosthetic systems.

A number of different control schemes need to be considered and tested. For example, the subject prefers to use myo switch control of his terminal device and wrist rotator with proportional control on his home set of prostheses, even though the option is available to use FSR touch pads to allow simultaneous control of the wrist with the hand. During future visits,

we plan to implement and test a control scheme where the hand and wrist are sequentially controlled with his hand open and close nerve transfers.

Although the current procedure only used the four major brachial plexus nerves, it opens the possibility to divide the nerves into multiple fascicles reinnervating smaller areas of muscle, and creating even more signals.

In the future, to record the small, tightly-spaced signals created by divided hyper-reinnervation, myoelectric signals will need to recorded by intramuscular electrodes with a transducer or through implantable myoelectric sensors [6,7].

This case demonstrates the need for more research. A larger clinical trial in high-level amputees with appropriate objective and subjective testing is warranted to see if these results can be repeated or even improved upon.

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PHANTOM LIMB SENSORY FEEDBACK THROUGH NERVE TRANSFER SURGERY

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INTRODUCTION

Commercially used prostheses are essentially open-loop devices and provide little or no feedback to the amputee as to how much force they exerting with the terminal device, despite numerous attempts to provide closed loop control in prosthetics [1]. Providing pressure feedback clearly has great value for function of the prosthesis—the goal is for the amputee to 'feel' what they are holding. Indirect pressure feedback has been attempted by methods including vibration [2] and functional electrical stimulation [3, 4]. The information contained in these forms of feedback is not in the same modality as that which they sense. As a result, while providing information to the user, it is likely that it comes at the cost of increased mental load and low level of information transfer [5]. Patterson and Katz [6] have obtained better qualitative feedback with pressure to pressure feedback than with pressure to vibratory or electrical stimulation feedback, offering support for this suggestion. An adaptive process is still involved since the subject must learn to associate pressure in one area with pressure in another area. Ideally, the sensory nerve endings of the amputated area need to be stimulated in direct correlation to the function of the prosthesis. The recent work of Kuiken [7] has made this concept feasible by the use of targeted reinnervation. The potential thus exists for the subject to feel as if touch, pressure and even hot or cold temperatures are being exerted on their hand. This study will examine the potential of intuitive pressure feedback.

SETUP

Nerve Transfers

Using targeted hyper-reinnervation to transfer nerves from a lost limb to denervated muscle as shown in Figure 1a, sensation of the lost limb was achieved on the chest of a subject [8]. Four independently controlled nerve-muscle units were created by surgically anastomosing residual brachial plexus nerves to dissected and divided aspects of the pectoralis major and minor muscles. Sensory reinnervation also occurred on the chest in an area where the subcutaneous fat was removed.

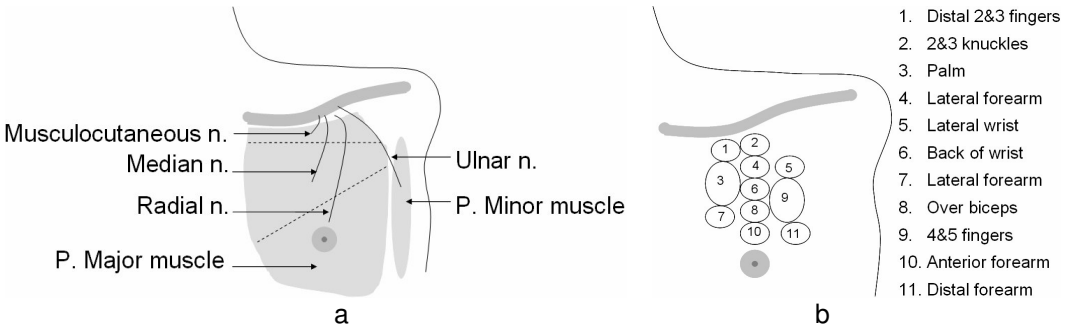


Figure 1: Somatic Representation of Nerve Transfer

a) Diagram of nerve-muscle graft procedure
b) Diagram of sensory reinnervation of anterior chest wall indication where touching the skin surface produced sensation in his phantom arm.

As a result of this surgery, the subject perceived touch, sharp/dull and temperature sensation that he felt in his phantom limb when pin pricks or thermal changes were applied to the chest, as shown in Figure 1b. A representation was acknowledged: pushing in one area elicited perceived pressure in the palm of the hand, in another area on the back of the hand, and so on. In some areas the patient had low sensory thresholds (2 g/mm^2) that he felt in his phantom arm. In other areas, while the subject perceived light touch on his chest, with greater pressure he only felt sensation in his phantom arm. We believe in these cases that the skin was not reinnervated, but added pressure stimulated nerves directly under the skin.

Actuator Selection

For the shoulder disarticulation subject tested in this experiment, nerves were transferred to the pectoral region of the chest for large signal amplification. In order to achieve physiologically appropriate force feedback, an accurate force must be exerted against the chest. Because the chest moves with breathing, this matter becomes more complicated: an accurate force is desired, but the force must track the changing position of the chest. A traditional motor would be able to accomplish this goal to some extent, but as the subject breathed, they would feel the inertia of the load on their chest. A linear backdrivable Series Elastic Actuator (SEA) [9-11] was created to decouple the inertia of the actuator from the force of the actuator. SEA are force controllable actuators with low impedance, high fidelity, and moderate bandwidth. They convert the accurate position control of traditional DC motors to accurate force control through the use of a spring as shown in Figure 2a. They have several advantageous properties, including reliable force output, simplicity, robustness of design, and the use of traditional robotic actuators. Most importantly for this experiment, the compliant spring greatly decouples inertia from force, especially at higher frequencies where inertia dominates the response. The authors significantly modified the traditional concept of linear SEA by using a 0-10 lb load cell in series with the spring, rather than a linear potentiometer in parallel with the spring, to measure output force, as illustrated in Figure 2b. Although linear potentiometers have historically been used in SEA, they cannot compensate for the inherent and substantial friction present in linear actuators. As a result, placing the pressure sensor in series with the spring provided more responsive and accurate control of linear SEA. The actuator applied pressure over an area of 103 mm^2.

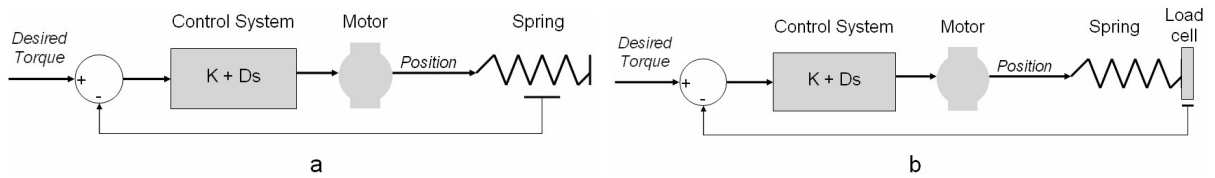


Figure 2: Series Elastic Actuator Schematics

a) shows a traditional linear Series Elastic Actuator, where a linear potentiometer measures compression of the spring.
b) shows the linear Series Elastic Actuator used in this experiment, where a load cell was placed in series with the spring to correctly discriminate between stiction and applied forces.
In either method, the motor generates accurate position. This position is fed through a compression spring, which converts the accurate position into an accurate force. The force recorded by either method is compared to the desired force. The error between the two signals is sent to a control block. In this example, the control block multiplies the error signal by a gain (K) and the derivative of the error by another gain (D) and sends this signal to the motor to correct the output force.

A load cell in the tip of an Otto Bock Greifer was instrumented with a 0-50 lb load cell. The maximum force of the Greifer was scaled to 1. This range was nonlinearly scaled to increase force resolution for small pressures, as illustrated in Figure 3. The scaled force was then linearly scaled between the subject's threshold of perception and threshold of discomfort.

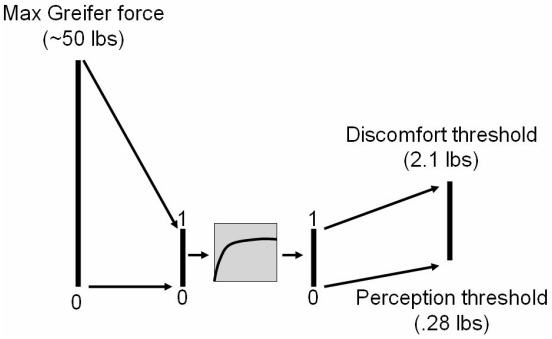
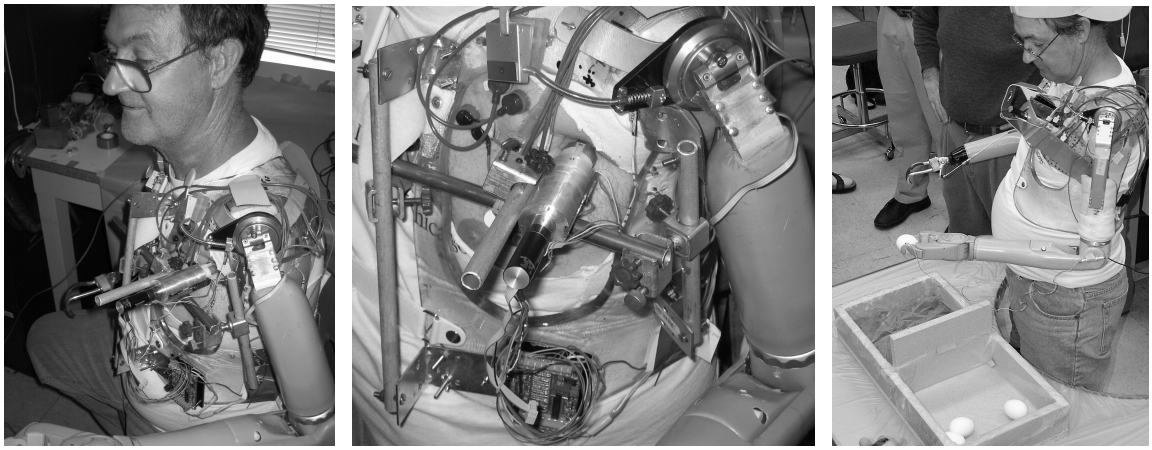


Figure 3: Hand Pressure to Chest Pressure Scaling

Terminal Device pressure was scaled from 0:1, and then nonlinearly amplified to amplify small pressure differences and attenuate large pressure differences. Finally, the pressure was scaled between the perception threshold of the subject and the discomfort threshold of the subject.

In preliminary testing the actuator was located in area 1 of Figure 1b. However, the close proximity of the actuator to EMG sensors prevented the subject from perceiving pressure gradation when he tensed his muscles to move the prosthetic arm. The SEA was moved to a more lateral area adjacent to areas 2 and 5 in Figure 1b that corresponded to a localized area the size of a pen cap between the 4^{th} and 4^{th} metacarpals of the subject. This allowed the subject to perceive pressure gradation while actuating his prosthesis.



a **b** **c**

Figure 4: Series Elastic Actuator

a) Linear Series Elastic Actuator pushes against subject's chest
b) Close up view of the actuator.
c) Box and Egg test setup: Subject moves eggs from one partition to the other

PRESSURE GRADATION RESULTS

The subject was asked to discriminate between a series of pressure ranges. For each range, the pressure was set at the threshold of perception, then increased to one level, reduced back to the threshold of perception, and then increased to the second level. This was repeated 8 times for each pressure range, as the probability of correctly guessing 7/8 trials of a binomial test are less than 5%. Selection of the pressure range was an iterative procedure in an attempt to accurately find the discrimination threshold.

This same test was done while the subject's EMG exertion determined the feedback pressure. A threshold would appear on the screen, and the subject would be required to exert enough force

on his terminal device to reach that level of pressure. After relaxing his grip, another force level would be shown, and the subject would then have to determine which force level had been higher.

Slightly above the threshold of perception, at 1.7 g/mm^2 pressure (0.4 lb force), the subject could discriminate 0.65 g/mm^2. In the middle of the force range at 4.4 g/mm^2 pressure, the subject's pressure discrimination improved to 0.48 g/mm^2. Near the discomfort threshold at 6.6 g/mm^2 pressure, the subject could discriminate 0.52 g/mm^2.

When the subject controlled the force, at 3.6 g/mm^2 pressure, the subject could discriminate 1.5 g/mm^2. The subject-generated force test was not ideal, in that the subject was able to relax once he had achieved the correct force in order to sense it. The subject did not appear to exploit this potential advantage. Using proportional force control instead of proportional velocity control will solve that problem in the future.

IN PURSUIT OF AN OBJECTIVE METRIC: THE BOX & EGG TEST

An objective performance metric was desired to evaluate the usefulness of this feedback paradigm. To accomplish this objective, a clinically accepted performance metric, the Box and Block test [12], was modified. Instead of moving blocks, eggs were moved. If the subject applied too much force, the egg would crack. If the subject did not apply sufficient force, the egg would drop and crack. The subject was directed to transfer as many intact eggs as possible from one partition to the other in 2 minutes. The setup is shown in Figure 4c.

Three control paradigms were tested. The first paradigm, while maintaining a constant force at the level of perception while the subject breathed, did not provide force feedback. The second control provided traditional pressure feedback on the side of the chest that had not been reinervated. Finally, the third paradigm provided force feedback in the area of the chest that corresponded to the phantom hand.

Several mechanical limitations of the feedback system were identified through the Box and Egg test. While these limitations prevented objective conclusions from being made regarding the performance of various forms of feedback, the limitations did provide insight into improvement of the overall feedback system, including:

- The subject does not appear to have proportional control of his terminal device. He uses short bursts, rather than low levels, to move his hand small distances. As a result, he was unable to lightly grip the egg. In the future, he will be given a proportional position control paradigm to train on at home, which requires the same type of control from the user as proportional force control.
- The terminal device used velocity control, as opposed to force control. This makes light grasp of an egg very difficult. Future experiments will use the Otto Bock Sensor hand to provide proportional force control once an object is grasped.
- The single location of the load cell in the terminal device prohibited accurate force representation if eggs were grasped with a different portion of the terminal device. Depending on the placement of the egg, recorded forces ranged from 0.1 lbs to 10 lbs for the same level of exertion. The Sensor Hand, which uses a strain gauge in a parallel linkage, should alleviate this problem as well, providing accurate force feedback no matter where on the terminal device the egg is grasped.

The box & egg test itself, however, appeared to be a good test, in that the subject usually dropped and cracked an egg for no feedback. The authors anticipate that this test will provide a useful metric once the above mentioned problems are corrected. The subject did not feel that breathing disrupted his perception of force for these tasks. When the subject was given various scale weights, he subjectively assessed the inertia of the actuator as equivalent to that of a 50g mass.

DISCUSSION & FUTURE WORK

Nerve transfer surgery, foundational to the work described in this paper, has provided the unique potential of various forms of feedback that accurately correspond to the phantom limb of the user. The authors have demonstrated that a series elastic actuator provides adequate force resolution with low inertia for pressure feedback. Initial results indicate that adequate sensory feedback exists to provide physiologically appropriate feedback.

In future designs, the subject would appreciate less pinpoint accuracy feedback to his perceived phantom limb. In order to achieve this with pressure gradation (ie, not hitting a nerve), it may be necessary to pinpoint several areas on his chest in order to give a more global reading on his phantom hand. Future experiments will take advantage of improved functional characteristics of the terminal device in an attempt to objectively evaluate potential advantages of this novel form of phantom limb feedback.

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ACUTE PROSTHETIC UPPER EXTREMITY REHABILITATION OF BLAST WOUND INJURIES: A 21 MONTH REVIEW

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ABSTRACT

Providing prosthetic rehabilitative services for upper extremity amputees is a challenging endeavor. Providing acute rehabilitative services to multiple individuals who have suffered upper extremity blast wounds adds a further complexity. Perhaps not since the Vietnam War has an upper extremity traumatic amputee population of this size treated in one center utilizing consistent procedures and protocols. The authors have had the opportunity to provide upper extremity prosthetic services to over 90 U.S. soldiers injured in Iraq and Afghanistan over the last 21 months at Walter Reed Army Medical Center.

This paper will review the unique protocols implemented to provide care in a military hospital to a young, athletic, and highly motivated amputee population. A review by amputation level includes surgical considerations, edema control, timeline for prosthetic management, prescription criteria, integration of occupational therapy and lessons learned through aggressive prosthetic intervention.

A TRAINING PROGRESSION FOR THE UPPER LIMB PROSTHETIC USER

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Advanced Arm Dynamics

As we all know, each patient that we see in our practices is an individual and should be approached with a client-centered philosophy. During the early phases of rehabilitation, my role as an occupational therapist on the amputee rehabilitation team is to work with the patient to determine his or her view of their interaction with their environment and their occupation. This is an all encompassing view of their life. A person's environment includes physical, social, cultural and institutional elements. A person's occupation is any aspect of self-care, productivity, and leisure that is part of their life. Once we have a sense of the areas of importance to this person, an occupational therapist will analyze the key components and skills required for the patient to perform in these occupations. This analysis will lead to the development of an individualized rehabilitation program.

My introduction was focused on the individual because the protocol that I will present is generic and requires the previously described approach to individualize the program in order to ensure success for that patient. It is critical for the patient to participate as an equal team member in order for the patient to fully engage in the process. This process is just that, a process. It requires frequent adjustments and fine tuning as skills and goals evolve. There are a few excellent measures that foster this approach to rehabilitation.

Once an individual's needs are established, it can be helpful to use a framework to structure the approach to the rehabilitation program. The following framework is the product of the collaboration of therapists who are experienced in rehabilitation of the upper limb amputee. The significant numbers of amputees who are returning from Operation Iraqi Freedom and Operation Enduring Freedom have necessitated further development and formalization of the protocol. The protocol presented is as it is used at Walter Reed Army Medical Center and it contains sections that will consistently apply to the population there due to the common training and occupation among the patients. This protocol can be easily adapted to respond to civilians whose circumstances are much more varied.

Upper Limb Amputee Prosthetic Training Protocol

Phase 1: Healing

- Evaluation
- Wound care
- Edema control
- Desensitization/scar management
- Pain control
- ROM
- Conditioning
- Psychological support

Phase 2: Pre-Prosthetic Training

- Change of dominance
- ADL training
- Strengthening
- Myosite testing
  - Basics
  - Accurate recruitment – palpation, bilateral contraction, common postures
  - MyoLab, MyoBoy
- Myosite training
  - Myoboy – isolated contraction, contract/relax, co-contraction, quick/slow
    - EMG type screen
    - Hand
    - Cars
- Training with electrodes and TD
- Continue edema control and scar management prn

Phase 3: Basic Prosthetic Training

- Donning/doffing
- Component terminology and operation
- Prosthetic care and limb tolerance
- Controls training
- Basic ADL's

Phase 4: Advanced Prosthetic Training

- IADL checklist: advanced and individualized categories

Phase 5: Discharge Planning

- Community resources and vocational planning

I believe that the early focus on myotraining and use of electrodes attached to an operational terminal device prior to fitting have increased the initial success rate and motivation that patients in this rehabilitation program have had. The goal of this rehabilitation progression is to build basic control skills into individualized function and ultimately a patient who feels confident with their prosthetic limb.

CUSTOM SILICONE SOCKETS FOR MYOELECTRIC PROSTHESES

Jack E. Uellendahl, CPO Hanger Prosthetics and Orthotics
Sandra Ramdial CP(c) Otto Bock

History of Transradial Interface Design

Sockets for myoelectric prostheses have not changed significantly over the past 30 years. The Otto Bock Muenster style socket (MyoBock) was developed in the late 1960's by Fruzinsky based on the original Muenster design of Hepp and Kuhn[3]. This socket was designed to compliment the newly available myoelectric components of the time that allowed for self-contained, self-suspending transradial prostheses. The Northwestern University socket was first introduced by Billock[1] in 1972. Elements of these two socket designs represent the critical design elements of state-of-the-art transradial interface designs even today. Flexible thermoplastics have helped to improve the dynamics of these socket systems, however, the socket design did not change significantly as a result of the more flexible materials. It should be noted that these flexible materials are non-elastic.

Influence of Silicone Sockets and High Definition Silicone Hand Prostheses

Over the past ten years roll-on silicone sockets gained favor in fitting some patients with myoelectric sockets.[2,4] One benefit of 3S socket technology is that a true suction suspension can easily be achieved. Suction suspension is particularly advantageous for fitting the long transradial limb because forearm rotation can be preserved due to the low and flexible trimlines that silicone suction allows, obviating the need for supracondylar trimlines which block all physiological forearm rotation. With the introduction of snap electrodes, electrode contact with the skin was ensured due to the elastic properties of the liner to which the snaps are attached. Two drawbacks of myoelectric fittings using roll-on designs are the need for some type of locking mechanism, and the nuisance of having to attach each of the electrodes separately as part of the donning process.

Concurrently there has been a proliferation of manufacturers of high definition silicone hands that offer excellent appearance and are also suspended by suction. These high definition silicone hands, when fitted to long transradial residual limbs, provided a suction silicone socket as a structural component of the hand prosthesis. Because the silicone socket is a structural component of the prosthesis it does not require a means of locking it to the prosthesis as does the roll-on-silicone socket design. These silicone hand prostheses are donned by lubricating the skin and then sliding into the prosthesis while working any air out by pressing the flexible walls in such a way to direct the air to the proximal trimline.

Custom Silicone Technology

With the introduction and commercial availability of custom silicone sockets (available through Otto Bock Custom Silicone Services, Toronto), the materials and

fabrication methods are now available to produce sockets that combine many of the desirable features of the above outlined silicone designs. Material thickness, stiffness, and color can be precisely controlled. It is also possible to incorporate hardware such as electrode mounts, screw attachments, zippers, and wrist mounts into the silicone during fabrication. To date patients with transcarpal, wrist disarticulation, long transradial, elbow disarticualation, and long transhumeral amputees have been fitted by Uellendahl, however, this paper will focus on below-elbow applications.

CONSTRUCTION METHODS

Total Silicone Myoelectric Prosthesis

Features:

- very flexible, easy for donning
- all-in-one (electrodes, wires, battery, charge plug are all encased)
- 2 wrist plates and 3 spacers that create a negative space (used to hold battery, 4-in-1 connector, charge plug) were used.
- Liner was made with most proximal wrist plate embedded within the silicone

Technical Issues:

- Inner hand was hard to attach
- Very time consuming fabrication
- Although the all-in-one silicone myoelectric prosthesis had great results, it was the most time consuming and expensive method of fabrication.

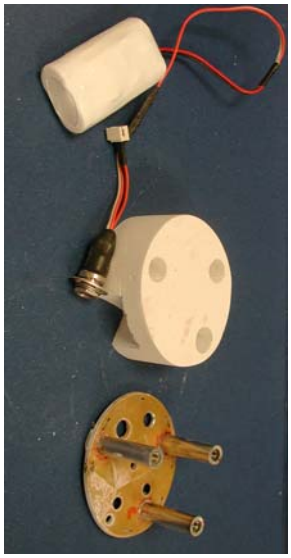

Figure 1

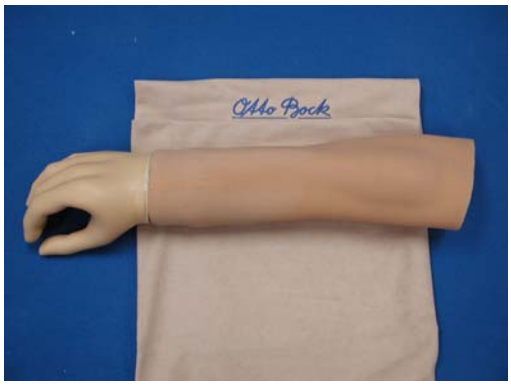

Figure 2

Silicone and Plastic Lamination Combinations (Hybrid):

1st Hybrid Design

- liner was made with embedded electrodes
- stockinette was embedded to attach initial lamination
- within the initial lamination, electrode wire channels were formed
- wax was used to build up a space for battery
- 2nd lamination w/ 4 star washers over distal end for attachment of wrist plate
- wax was melted out through a large hole (large enough for battery to fit through)
- electrode wires fed through initial lamination

Technical problems:

- very difficult to slide battery in due to the inflexibility of the resin
- due to repeated heating to try to get the battery in, the top layer of lamination was not flat and very messy.
- Very difficult to put in the screws to attach to the hand through the wrist plate
- Screws used to attach wrist plate to lamination had to be very small and located where access was difficult

Figure 3

Figure 4

2nd Hybrid Design

- internal battery, enclosed charge plug with zipper opening
- embedded electrodes
- dummies were made for battery and charge plug (charge plug placed proximally)

- electrode wire channels were incorporated in the silicone liner with exit hole at the distal end of liner.
- 1st lamination embedded in silicone
- 4 star washers were embedded in the lamination for attachment of the outer shell
- wrist unit was attached via 2nd lamination

Fabrication drawbacks:

- hard to get silicone liner off the cast
- hard to remove the plaster dummy (battery)

Figure 5

Figure 6

3rd Hybrid Design

- 1st lamination is attached to the silicone
- removable battery and wrist are mounted to second lamination
- this process is the same as a regular myo and definitely is the least labor intensive with equal cosmesis to the other methods described.

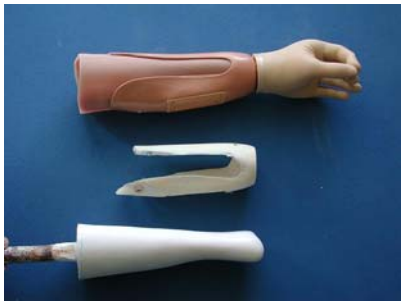

Figure 7

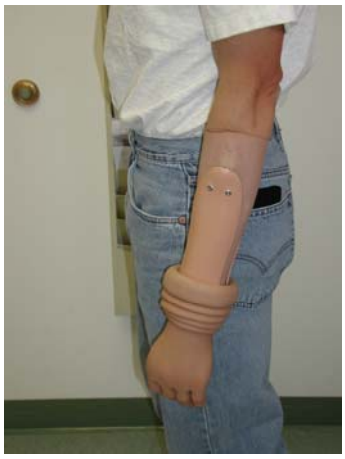

Figure 8

Donning and Doffing

In order to provide a durable and simple suction system, valves have not been installed in the distal sockets. A simple and effective method of releasing air upon insertion of the limb has been developed where a thick nylon cord is draped down the socket wall and allowed to curl around the distal end of the socket. This creates an air channel allowing for evacuation of air from the socket as the limb displaces the air. Once the limb is fully inserted the cord is easily pulled out and an air-tight suction suspension achieved. To remove the prosthesis many of the patients have been able to slide a finger into the socket and then force the air pocket created by the finger to the distal end thereby breaking the suction seal. Some newer amputees who have more sensitive limbs are unable to tolerate the pulling associated with this first method and have preferred to slide a thin corset stay into the arm thereby creating an air channel to the end of the socket allowing easy removal.

Results

To date, all of the fittings using custom silicone socket technology have been successful. There have been no rejections, no skin issues, and only minor adjustments required. The socket adjustments have been required due to shrinkage of the residual limb. In the cases where the residual limb reduced in size, the patients complained of discomfort and/or pain, red to purple skin color upon removal of the prosthesis and swelling. These problems would be expected in a total suction socket whenever residual limb volume reduces. The problem has been corrected by adding a TEC spot (a urethane disc produced by Otto Bock) in the area of swelling.

The longest follow-up is two years. Material durability has proven to be acceptable for all patients fitted, with no significant damage reported.

Battery systems that can be placed inside of the hand geometry are needed to optimize the appearance of these prostheses. Also a charging plug that can be placed at the proximal trim line that would allow charging without rolling down the glove or cutting a hole in it would be very beneficial.

Figure 9

Figure 10

Figure captions:

Figure 1 – use of internal battery positioned between two wrist plates using the transcarpal hand achieves appropriate length for this wrist disarticulation amputee. (This method was recommended by Liberating Technologies, Holliston, Maine).

Figure 2 - the completed all silicone prosthesis.

Figure 3 - a hybrid silicone/plastic laminate construction using internal battery with transcarpal hand.

Figure 4 shows the completed prosthesis on this young man with congenital absence of his hand.

Figure 5 - a hybrid constructions with internal battery. A zipper allows access to the pocket housing the battery. Note the proximal position of the charge plug for easy access without rolling down the glove.

Figure 6 - the unrestricted elbow range of motion using custom silicone designs.

Figure 7 - a hybrid construction for wrist disarticulation with removable battery.

Figure 8 - a finished silicone socket with laminated struts providing hand stability. The patient has a wrist disarticulation amputation. The hand is a Sensor Speed with wrist disarticulation wrist and removable LiIon battery. All components are provided by Otto Bock

Figure 9 - a clear silicone test socket fitting and electrode site selection. Here the electrodes are being calibrated for proper control of an Otto Bock DMC transcarpal hand using the MyoBoy tester.

Figure10 - the excellent cosmetic result that can be obtained using custom silicone techniques. In this case acrylic nails have been attached to the PVC production glove to improve appearance.

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Individual Silicon Interface for Myoelectric upper limbs

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Introduction

Silicon as an interface is a very good material, one that gives us the ability to solve problems not easily solved before. The long Trans-radial has never been easy to fit. When we first started to manufacture bespoke Silicon interfaces we used silicon gel that was painted onto the cast until we achieved the thickness required.
This system has worked well but it was not able to cope with a myoelectric arm easily. We needed to develop a system that allowed the patient to don and doff the silicon interface when fitted with electrodes and cables and did not cause the patient to sweat.

Methods

We originally looked at a technique developed at Strathclyde University – the roll on and off individually made silicon sleeve with a lanyard fitted at the distal end. A polyester resin frame was made to hold the sleeve in place. The lanyard, which is attached to the distal end of the silicon sleeve, is threaded through the frame so the patient can pull the sleeve and residual arm into the frame.
The system worked well but it was not able to cope with a myoelectric arm. We needed to develop a system that would allow the patient to don and doff easily a silicon interface that had been fitted with cables and electrodes.

Result

We have developed a new technique using the Otto Bock Chlorosil Silicon, with a zip, electrodes and cables built into the silicon. The Silicon is put through the mixing rollers to a thickness of 2 cm. It can be thicker dependent on the strength required. Different area's can be fitted with hard or soft Silicon if you have a bony area. (shore rating 20, 35 and 60). Because you put the silicon through rollers it is very easy to control the thickness prior to laying it on the cast. A first layer of silicon is put onto the cast over the electrode blanks. A polythene strip is then laid where the electrode

Cables are to be run. A second layer of silicon is then placed over the polythene strips. (When the silicon has cured the polythene strips are pulled out leaving a channel for the cables to be inserted) The zip is laid into place at the same time ready to be covered up to the teeth with a second layer of silicon. It is very important to put a series of holes in the zip so the Silicon can get a firm grip of the zip.
Part of this development is a new way of attaching the silicon to the Polyester or Acrylic frame that secures the hand or wrist mechanism in place. A shaped resin distal cap is made to fit the distal end of residual limb, the cast is then drilled with a series of holes, this to allow the silicon to get a good hold of the resin end cap. Finally we have added a lattice system which is a series of slots in a lattice formation that allows the silicon to breathe. It will also allow the silicon to stretch if required.

Conclusion.

We are now able to fit individual Silicon sockets to elbow disarticulation, Trans-carpal and Trans-radial residual limbs. Patients with Trans-carpal gain the greatest improvement flexion/extension and pronation/supination of the wrist, which could not be achieved with a conventional socket technique. The new lattice system allows the silicon to breath and stretch; this gives patients better comfort and movement of the prosthesis.

Acknowledgements

The original work was done with the help of Mr W.G.Dykes. National Centre for Training and Education Prosthetics and Orthotics, University of Strathclyde, Glasgow G4 0LS, Scotland. I would also like to acknowledge the work done by Mr K. Dowler.Technical dept. RSLSteeper Nottingham.

SILICONE BLADDER SUSPENSION FOR THE WRIST DISARTICULATION LEVEL AMPUTEE USING A MINI PUMP SYSTEM TO ALTER VOLUMETRIC PRESSURE INSIDE THE SOCKET

Matthew J. Mikosz, CP

ABSTRACT

The purpose of this article is to describe an alternative fitting technique for the wrist disarticulation amputee. The socket design utilizes a silicone bladder contained in a sealed chamber with a one-way expulsion valve and built in mini pump. This design allows the volumetric pressure inside the socket to be controlled by the amount of air inside the chamber. Once the limb is positioned inside of the socket, the pump can be used to fill the chamber with air and apply controlled volumetric pressure throughout the entire length of the flexible bladder. The pump can also be configured to expand the bladder, and then the one-way release valve can be depressed to the appropriate level of compression. The two patients that have been fit with this system stated improved comfort and optimized suspension. This system provides the patient with the ability to alter the socket fit according to the desired activity.

INTRODUCTION

The primary goal in achieving a successful prosthetic outcome is providing the patient with a well-fit socket that is comfortable, functional and provides optimal suspension. As stated by Farnsworth, without a properly designed and fitted socket that is both comfortable and functional, your ability to benefit from your other prosthetic components will be limited [1]. The purpose for utilizing a flexible bladder design with volumetric adjustability is to provide the patient with these very important socket characteristics. The socket design is also beneficial for bulbous distal ends or irregular shapes due to the expandable properties of the silicone bladder. The silicone bladder can offer the patient a self-suspending socket design with improved cosmesis due to the absence of straps or windows, which other designs require for adequate suspension.

The bladder system can also be used in myoelectric fittings. Standard or remote electrodes can be used depending on the specific needs for each patient. Standard electrodes can be used if they can be placed high enough on the forearm to allow for optimal placement of the bladder. If the ideal electrode placement is located more distal on the forearm then remote electrodes may be your best option since they can be placed inside of the silicone bladder. To determine the appropriate location of the silicone bladder a measurement is taken at the widest point of the distal end using an ML gauge. Moving the ML gauge proximal on the forearm until you reach the dimension of the distal end will determine the proximal location of the silicone bladder. Generally, the silicone bladder will end at the apex of the distal end of the limb.

METHOD

The fabrication process for the silicone bladder system for wrist disarticulations is very similar to that used in fabricating silicone Syme's prostheses. When modifying the mold for a silicone bladder system the ideal bladder location should be identified and marked on the mold. Once marked, a circumferential reduction in volume is made in the proposed bladder location to insure that suction will be maintained in the socket. The most ideal method for fitting the test socket is to fabricate the inner socket, as it would be finished in the final prosthesis. The use of flexible plastics for diagnostic purposes typically will not give you an accurate assessment of the socket fit when using a silicone bladder for the finished product. To begin the fabrication process, a PVA bag is applied to the mold then approximately 6-8 layers of elastic stockinette. The stockinette at the distal end can either be saturated with resin prior to laminating in order to pinpoint the area needed to be rigid or laminated into the first lay up. If laminating into the first lay up apply the PVA bag then tie off the areas with string where the bladder will begin and end. Apply tape over the string to secure the string to the bag. This will prevent the bag from slipping and also apply compression to the stockinette to prevent resin from leaking into the bladder area. Cut away the proximal and distal ends of the bag at the location of the string leaving just the middle section PVA bag to protect the material in this area from being saturated by resin. The middle section will be laminated with silicone resin at a later stage. Apply the outer PVA bag to the mold and laminate the proximal and distal segments. Laminating should be done at low vacuum to minimize resin leaking into the bladder location. Once cured, remove the bag and apply pressure sensitive tape over the laminated sections and apply a new PVA bag to the mold. The tape will keep the lamination clean from silicone when laminating the bladder section. Laminate the middle section using silicone resin. The PVA bag can be removed once the silicone has completely cured. A thin polyethylene sheet or X-ray paper is then used to create the void for the expandable wall between the inner and outer sockets. The polyethylene sheet or X–ray paper should be placed on the mold to completely cover the area of the flexible bladder. Make sure to build up enough to accommodate the necessary expansion required to allow the limb to easily pass through without hitting the outer socket wall. Fill the polyethylene with bee's wax and let harden. Once hardened, shape the bee's wax to the appropriate shape and then prepare for the second lamination. Prior to laminating the second lay up, you will need to rough up the proximal and distal lamination to allow for proper adhesion of the inner and outer laminations. Apply your lay up directly over the bee's wax (no inner bag required) and laminate. Once cured, drill a small hole in the outer lamination in the location of the bladder and heat in oven at low temperature to remove the bee's wax. Prepare the socket for fitting by installing the mini pump (pneu-fit) and tubing into the hole drilled to remove the bee's wax. Creating an air tight seal around the tube is necessary when evaluating the silicone bladder for any leaks. A small hole should be drilled at the distal end to temporarily install a peewee valve for the fitting. Once the socket has been donned, the pressure inside of the socket can be controlled by depressing the mini pump. Pressing on the release valve of the pump will reduce the pressure inside the socket. Remove all of the air from the chamber and press the peewee valve to remove the limb from the socket.

DISCUSSION

The first patient that was fit with the system was previously wearing a flexible inner liner made with Proflex with silicone and a two-inch elastic tension band just proximal to the styloids. His main concern with this design was that in order to achieve optimal suspension the tension band had to fit snug and was uncomfortable over extended periods of time. Also, he had to put lotion on his limb and force his limb into the socket. The silicone bladder system that was designed for him had a peewee valve located at the distal end of the socket to maintain and release suction when needed, and a pneu-fit mini pump with an expulsion valve to alter the amount of pressure inside of the socket. The release valve that is attached to the pump would release air inside the chamber and allow easy doffing of the prosthesis.

The second patient that was fit with the design was a nine-year-old congenital amputee with Poland's Syndrome. Poland's syndrome is a congenital deformity consisting of ipsilateral syndactyly and pectoral girdle muscle deficiency [2]. The patient presented with a shortened radius, ulna and humerus with syndactyly. The patient was fit with a silicone bladder design, single site myoelectric control using remote electrodes and an Otto Bock system 2000 hand. The silicone bladder allowed for easy donning even with the syndactyly associated with his condition. He was able to maintain suction within the socket by the intimate fit of the silicone to his limb and the peewee valve located at the distal end. The electrodes were placed inside the silicone bladder located in the palmar area of his limb as shown in Fig. 1. Due to the shape of his limb, the electrodes needed to be able to move as he donned the prosthesis, otherwise he would not be able to pass by the electrodes and fully donn the socket. By placing the electrodes inside the silicone bladder, he was able to donn the socket with ease and achieve total contact with the electrodes at all times.

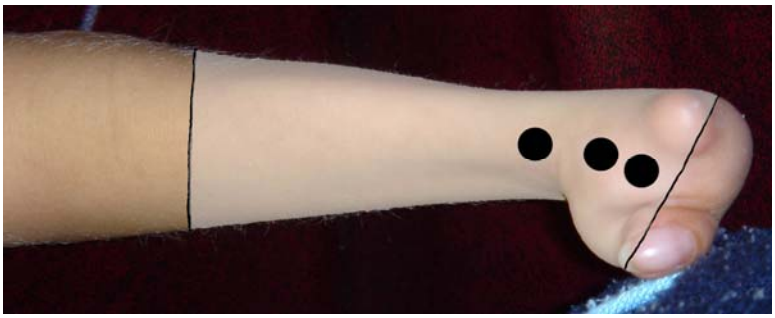

Fig. 1

RESULTS

The two patients that were fit with the silicone bladder system were different in many ways but had similar prosthetic requirements. They were both looking for adjustability within the socket to not only allow for easy donning and doffing of the prosthesis but to be able to adjust the socket fit as needed throughout the day. The silicone bladder system provided them with a streamline self-suspending socket design that could be self adjusted according to their specific activities. The first patient that was fit with the silicone bladder system was looking for improved comfort throughout a full day of use. He also

stated that his limb volume changed throughout the day, and he would like the ability to adjust the socket fit if necessary. The silicone bladder system provided him with the necessary adjustability to accommodate his limb volume change throughout the day. He also stated improved comfort over an extended period of wear time. The nine year old had a very slender build and a streamline design was important to him and his family. The silicone bladder system required no external straps or removable windows to provide optimal suspension. Other self-suspending designs may have been too bulky, complicated and less cosmetically appealing to him and his family. He was able to donn and doff the prosthesis with ease while maintaining suction and total contact with the electrodes at all times.

CONCLUSION

Wrist disarticulation amputees, as well as all other levels, require a prosthesis that is not only comfortable but provides optimal suspension. Combining the expandable properties of silicone with the ability to adjust the pressure inside of the socket can offer patients new levels of comfort and fitting options. Other designs that are being considered could have silicone bladders in specified locations that apply pressure to isolated areas. For instance, silicone bladders could be placed along the antecubital depressions in an ACCI socket design for a trans-radial amputee to assist in lifting power and protect the radius during heavy lifting. Also, placing silicone bladders over the stabilizing wings of a trans-humeral socket can increase stability and rotational control when needed. Miniaturized electro-pneumatic motors can also be placed inside of the socket, which can activate the pump and fill the bladders with air. Available microprocessors or remote power switches can be used to activate the motors. When used in a myoelectric system, the motor can be activated by various control strategies such as a high/low configuration or co-contraction to switch between hand function and pump activation. In conclusion, silicone has many beneficial characteristics that can be implemented with our current technologies to offer patients alternatives to traditional fitting techniques and optimize their functional ability.

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MEC 2005, Canada, August 2005

Integrating Electrodes in Silicone and a New Application for Myoelectric Suction Sockets

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ABSTRACT

There have been numerous attempts to integrate electrodes into liners to operate externally powered devices. Various methods and techniques have been tried from RTV laminated sockets to off-the-shelf liners as well as a number of thermoplastic techniques. All of these techniques, while providing improved suspension and comfort have limitations and drawbacks. However these limitations have not deterred prosthetists from pursuing these options in an effort to provide better control, better suspension and ultimately a better outcome. Through the experience of Otto Bock's Custom Silicone Services, we will present a number of solutions. These include challenging fittings such as the transmetacarpal as well as more standard levels.

Suction Sockets have been utilized in upper extremity fittings for many years now, but it has always been challenging to maintain suction in myoelectric fittings due to the difficulty of sealing around electrodes. A new solution developed by Otto Bock will also be presented.

INTRODUCTION

As clinicians, we are constantly looking for ways to optimize prosthetic fittings. For externally powered devices, suction sockets have become a favourable suspension method. Many new techniques and materials are now available that are making suction sockets possible.

In this paper we will describe the progression of achieving full suction suspension using custom silicone liners. A silicone liner with electrode cut outs was first developed, followed by creating a liner with integrated electrodes. Recently, a new electrode developed by Otto Bock has made full suction suspension possible.

ELECTRODE CUT-OUTS IN LINERS

Silicone liners are fabricated with rectangular or circular openings at the electrode locations. This allows a more comfortable socket with partial suction suspension.

Applications:

- For fitting myoelectric prostheses to patients who are using silicone liners and still fabricate the devices in the standard way
- For longer residual limbs, the cut-outs are close to the proximal trimlines to maintain the suction fit
- For short residual limbs, the liner can be extended beyond the proximal brim of the rigid socket to maintain the suction fit

Liner Advantages:

- easily replaced if needed
- more thorough cleaning as liner is removable
- can be inverted for easier donning
- partial suction achieved

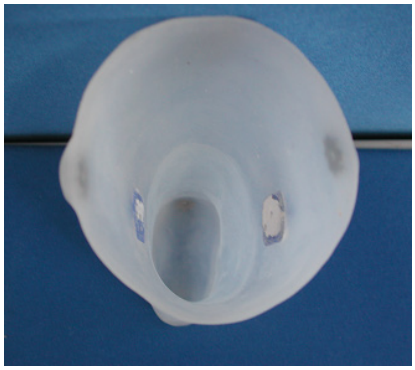
Disadvantages:

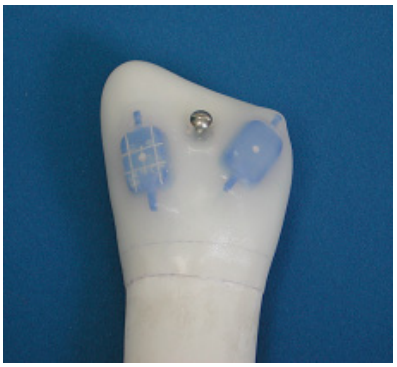
- tissue bulging out of the electrode cut-out areas could cause pinching when donning the device
- suction may be lost at the cut-out areas

Partial suction is achieved when silicone liners with cut-outs are used. The cut-outs could allow air to enter the liner and as a result suction may be lost. Along with this type of liner, a socket using standard fabrication techniques is required for mounting the electrodes. Since our goal was to achieve full suction with a liner for a myoelectric device, we developed another technique using integrated electrodes.

INTEGRATED ELECTRODES LINERS

Integrated electrode liners are fabricated to include a negative space in the silicone. The electrodes are mounted within the liner from the inside and the cable exits from a slit in silicone. These liners are always attached to the rest of the myoelectric device and can be used either as a "socket" on its own or with an outer socket.





Applications:

- For fittings where patients are able to insert their residual limb into the device without having to invert the liner
- For patients who can remove their residual limbs from the liner, without having to remove the liner from prosthesis
- For fitting myoelectric prostheses to patients who are able to use the liner as a “socket” without any lamination support

Advantages:

- Suction suspension can be achieved
- More comfortable for the patient since suction is achieved with full contact of the silicone on skin (no tissue bulging)
- Easier for patient – less parts, components which eases donning
- For increased stability, can be fabricated with lamination struts which in turn eliminate the need for a separate socket
- Channels can be made in the silicone for the electrode cables

Disadvantages:

- sometimes loose suction - when full contact is lost
- cleaning difficult as the liner cannot be removed – especially for the distal end of the socket in the case of a long residual limb
- bulky as a result of encapsulating the electrodes
- more time consuming

Although total suction is achieved, the silicone build-up around the electrodes creates a bulky prosthesis. Maintenance of the device is also a concern as the entire prosthesis must be disassembled. To correct these concerns, a new application for myoelectric suction sockets was designed using a different mounting system for electrodes.

SUCTION SOCKET ELECTRODES

Otto Bock has developed some new electrodes that snap into a rectangular cut-out in the socket. These electrodes are identical to the 200 electrodes but do not have any mounting extensions. They simply snap into the rectangular cut-outs of laminated, thermoplastic or even custom silicone sockets.

Applications:
(same as with integrated electrodes)

- Can also be used in custom silicone sockets or prostheses

Advantages:
(same as with integrated electrodes with additional benefits below)

- Can be used for traditional thermoplastic and laminated socket designs
- Improved suction as the electrode forms a seal around the cut-out
- The seal will prevent perspiration from damaging the electronics
- Easy to install
- Fabrication is simplified
- Low profile

Disadvantage:

- cleaning is difficult as the electrodes are connected by cables to the rest of the prosthesis, therefore, the liner cannot be removed

CONCLUSION

The suction socket electrodes have addressed the concerns of achieving total suction by the seal around the electrode cut-outs. The bulkiness of having to encapsulate the electrode in silicone has been eliminated.

One area still requiring research is in the development of a wireless electrode. This would be beneficial as it would allow the liner to be removed from the prosthesis with no cables attached. This would make the liner easier to clean and maintenance is facilitated as the entire prosthesis does not have to be disassembled.

As research and development continues in this area we can use existing techniques in different ways and adapt materials with new components to offer better fitting options to our patients.

ACKNOWLEDGEMENTS:

I would like to thank the technical staff of Custom Silicone Services at Otto Bock Canada. This dedicated team is constantly making improvements that benefit our patients. Special thanks to Sara Creighton, Wilson Cisneros and Kyla Lau.

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THE DIFFERENCE A TERMINAL DEVICE CAN MAKE

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Introduction

The manufacturers of electronic components have answered the requests of patients in recent years. Advancements were made with microprocessor-based controllers that allow for easy computerized adjustments, answering the need for increased control of the terminal device. Additionally, batteries have moved to Li-Ion, NiMH, and soon Li-Polymer, answering the request to provide a longer life to the charge, as well as the ability to top-off the charge. Another patient request was to increase the speed of the terminal device. In actuality, what patient wanted was a quicker response from the terminal device as they provided an input signal. An earlier response by the field was to increase the voltage to the terminal device. The higher voltage wasn't the perfect answer because it was also difficult to operate at slow speeds. The Sensor Hand Speed answered these requests to have a more responsive hand that operates quickly, but also is able to operate very slowly according to the user's input.

Field Trials

To date, our organization has fit over 160 of the Sensor Hand Speed in the past couple years. The evaluation of candidacy of this device is very simple because the versatility of the control options makes it suitable for almost every electric wearer needing a hand. This option has been favored by our patients who want the quick response of the hand closing from a full open position in 1/3 second, but also the control of slowly closing—all determined by the intensity of signal produced.

Oftentimes we have patients who have only one good EMG signal and still desire the speed and control offered by the hand. Through usage of coding plugs or a hand held programming device, different strategies of control are selected. (This hand held device can also slow the top speed of the hand down for initial training.) One of these strategies is a proportionally controlled single site that opens with an input signal and closes upon relaxation of that input. This single site usage can also be used on the weaker muscle site to provide strengthening, and then changed to a dual site once the weaker muscle is ready.

In occurrences where this weaker muscle doesn't strengthen to the intensity of the stronger site, we often do one of three things. One being leave at a single site; although not maximizing the capabilities of the hand, it is still a functional solution to the patient. The second option is making adjustments on a computer. The customizing process on the computer not only adjusts the hand, but adapts to other devices in the Otto Bock line as well. These adjustments are very practitioner friendly because the prosthesis can attach to the computer and we are able to see the input signals

graphically on the screen. Gain amplification, or "boost", can be done so the weaker muscle mirrors the intensity of the stronger muscle, all easily on the computer. Additionally, on and off thresholds can be adjusted to minimize unwanted signal from interfering with operation.

A third option for this user is to change to a VarioDual strategy with the yellow coding plug. This control gives the patient the same single site proportional control as the blue plug, but now will also provide gripping control with the second weaker signal. This strategy works very well with many users, even with excellent dual site control, because it mimics the natural opening & closing and gripping of a normal hand. This control strategy on this hand is the most physiologically normal way of grasping, gripping, and releasing onto an object than any other device available.

These strategies of control are not the only advantage our patients have enjoyed. Another feature is the autograsp feature it provides. First seen in the field by the original Sensor hand, the Sensor Hand Speed will increase the grip force on an object if the sensor picks up a change in weight distributed on the thumb, or, if the object is slipping or getting heavier. This puts the patient at ease that they will not drop the object they are holding. This is a true advantage for the single site user who does not have control of the grip force.

Other Options

The three strategies of control listed above (standard dual site proportional control, single site proportional control, and VarioDual control) are what we find our patient accepting most often. Although rarely needed or accepted in the field, the hand does have other control options. One option is a low input control that allows a much lower signal to control the opening and a very low pulse to close the hand. This is not used often because the adjustments on the computer can amplify the signals. Another strategy is the single site without proportional control. However, patients prefer adjustable speed so they can slowly grasp onto an object giving them a better sense of control. A final option is a plug that allows the patient to turn the sensor on or off. Of the patients who reported back to us, none desired to turn the autograsp on and off. There were some who didn't want the autograsp, in which case we provided a different hand. Although good options for this hand, the patient does not accept these controls as much as the first three discussed.

A problem is encountered when needing a second terminal device and using any of the options other than the standard dual site. No other terminal device (e.g. ETD, Greifer, Power Gripper) can mirror each of the control strategies as that of the Sensor Hand Speed. Due to varied vocation & avocations, many patients require the use of multiple terminal devices. The single site operation, a low signal to close and a high signal to open, is different in the Greifer than the "cookie crusher" of the hand. This change of control is confusing for patients and oftentimes they don't use the second device as much as they should. However, prototype Greifers are currently being tested to mimic these control strategies of the Sensor Hand Speed and we hope will be available soon.

Conclusion

There is a large acceptance of the Sensor Hand Speed for it’s functional benefits. Not only does the autograsp feature give the patient a security of grasp, but also the responsiveness through a high speed is well liked. Yet, patients also like the ability to control it very slowly if needed. The versatility of control mechanisms allows this hand to be used on many different patients who may not have been able to control myoelectrics in the past. Finally, the computerized adjustments are very easy for the prosthetist to fine tune the hand for optimum control.

CONTROLLING POWERED UPPER EXTREMITY PROSTHESES NOW AND IN THE FUTURE

T. Walley Williams, III, Liberating Technologies, Inc. (LTI), Holliston, MA, U.S.A.

HISTORY

Powered prostheses have come a long way in the last twenty five years. A good way to track this activity is simply to review the proceedings of the Myo-Electric Controls (MEC) conferences that have been held during that time. Some of the highlights are listed here especially those that led to further developments. Two-electrode threshold control, three-state control, quick-slow control, cookie crusher voluntary open control, proportional control, shifting of control to a second device, control using a force servo, the positional servo control, simultaneous control of two or more devices, circuits with plugs or switches to select an optimal control scheme for a particular user, the interfacing of a computer to let the prosthetist tune the control system to the user, the use of a computer to store complex control programs that can be downloaded to the control system to test their suitability for the user, voice activated control, the use of RF to isolate the setup computer from the prosthesis, and much more.

TECHNOLOGIES THAT ARE CHANGING THE RULES

During the last decade a number of new technologies have become available to prosthetics researchers. Listed below are some that seem particularly important. A few are almost available, but are close enough to reality that they will drive our future plans. For the most part the prosthetics field has to wait until a mass market develops a technology that can be adapted.

Batteries with Improved Power Density

Since 1975 the capacity of a nickel based rechargeable AA cell has gone from 550 mAHr to over 1100 mAHr and environmentally friendly Ni-MH cells have for the most part replaced Ni-Cd. Where high current is not an issue, Li-Ion and Li-polymer cells are replacing the nickel types with two cells replacing five. The lithium cells solve the memory problem, but they create a new problem, since their voltage falls substantially during use. Ten years ago that seemed like a problem, but now it is easy to put a microprocessor into the device using the batteries. This processor not only enables proportional control of speed and force, it also supplies current to the motor with pulse width modulation (PWM) so that the system responds as if the battery voltage were constant.

Better Motors, But Not Better Gear Reduction

As magnetic materials have improved, motors have become much more powerful for the weight. However, electric motors are only efficient when they run at high speed. A great deal a weight is still used to transform high-speed, low-torque rotation into low-speed, high-torque motion, and energy is wasted in the transformation. Further, to truly mimic the human body one needs to move joints quickly at low speed and then instantaneously shift to high torque.

Distributed Control Systems

It is no longer necessary to put all of the control circuitry in one place. Soon we will send commands from a central processor to local controllers at each device. The need for in industry-wide protocol for this type of control is discussed below.

Configurable Control Systems

Almost every component manufacturer now provides a way to configure its system using a computer interface. At first this only meant that gains and thresholds could be set for myoelectric signals. Now users can quickly switch between setups as different as all myoelectric control of three devices by two muscles and positional servo control of one device with myoelectric control of two devices.

Implantable Myosignal Transducers

Simultaneous acquisition of signals from surface muscles and the muscles buried underneath has always been a problem. While several groups have contrived ways to extract some of the information present in the underlying muscle using multiple features of the myoelectric signal, [4, 5] nothing works as well as direct acquisition of the myosignal. You will read about one implantable myosignal transducer in these proceedings. A second group is also working to advance this technology. One or both will probably be approved for use in amputees before the next MEC conference. These units are about the size of a large grain of rice, and they acquire their power from outside the body using RF energy. Either RF or infrared light is used to deliver their outputs which may be partially or fully processed myosignals.

Improved Materials and Techniques for Making Sockets

To support the new control systems better light-weight prostheses are needed. The wide availability of high-modulus materials combined with the training of prosthetists in their use has been essential to the development of better powered prostheses. The need is for automating the production of outstanding patient interfaces so that more prosthetists can achieve the results that now come only from a small cadre of specialists.

INTUITIVE PROSTHETIC CONTROL

The most intuitive prosthesis control uses the original muscles to control the same functions in the prosthesis. This is already done when biceps and triceps are used to control the motion of a powered elbow joint. What are the problems with this approach, and how will they be solved?

Full Control Needs Two Muscles per Degree of Freedom

Relatively few muscles are lost during a transradial amputation, but after higher level amputations, only a few large muscles remain. Elsewhere in these proceedings Kuiken et al will report on the current state of research on subdividing muscles and on connecting severed nerves to them. The goal is to create enough myoelectric signal sources to replace all of the muscles that have been lost. The challenge is to perfect techniques for subdividing muscles and nerve fascicles. Identifying which fascicle does what may be particularly challenging. To date this research has relied on surface myoelectric pickups.

Coping with Muscles Too Deep for Surface Myoelectrodes

In many cases good muscle signals are already available for controlling prostheses intuitively. These are often the signals present in muscles too deep for surface pickups. The implantable transducers discussed above will make these muscles usable. Implantable signal pickups will also improve the signals from nerve-implanted muscles because the re-enervated muscle bands can then be made smaller without excessive crosstalk between adjacent bands.

Creating a Mechanism That Responds Like a Human Arm

At present, electric elbows need to be locked when not moving to conserve power. Some engage lock pins and others use reverse locking clutches. Neither approach permits the mechanism to exhibit the compliance of a natural joint. In contrast the muscles of the intact arm act like variable stiffness springs. To gain compliance while still using electric motors, one can place nonlinear springs between the motors and the two joint directions. [1] With a lock, only one motor is needed to move the joint in both directions. With nonlinear springs, two motors are needed. Some weight can be saved by using a powerful direction motor with a slower weaker motor being used to change the stiffness. Adding motors increases weight. To avoid this some workers are looking at pneumatic control with electronics relegated to controlling the gas flow. Such a scheme needs a source of compressed gas. Michael Goldfarb has developed such a system, and it will be interesting to see if it can be made to fit in an arm prosthesis. [2]

The Human Brain Commands a Position Not a Speed

A few years ago Matt Smits at the Liberty Mutual Research Center studied the triphasic myoelectric signal generated when a traumatic amputee suddenly tried to reposition his prosthetic arm. [3] What was curious was that a long-term amputee could still generate such a signal. In the intact arm, it is normal, because the central nervous system (CNS) first generates a large impulsive muscle contraction to accelerate the forearm, then an impulse to stop it in the new position and finally a small third impulse to damp out the second. Experiments of this sort are best comprehended by using a set point model for muscle control. The CNS operates by commanding new set points for the levels of muscle activity. The CNS is certainly not sending speed control signals like our present control systems.

The Two-Step Control System

The goal for future control systems will be to create an accurate model of the human joint system. Myoelectric signals will reset the parameters in the model on a continuing basis, and the model in turn will control joint motion using improved mechanisms. The mechanisms will have settable levels of compliance and stiffness, and will respond much as a normal human arm. The ideal mechanism will only need power to reset the mechanism state. For instance a short burst of motor power will increase or decrease the tension in a spring. Inherent in such a system is the use of spring elements to compensate for gravitational loads.

Proprioceptive Feedback – the Missing Element

Amputees have to rely on eyesight for most of their location feedback. In addition motion of the elbow or shoulder shifts sufficient mass that socket pressures alter supplying additional feedback. However, small changes in wrist orientation or finger position cannot be detected this way. Feedback is so important that many amputees control their elbow position with a positional servo rather than with the more intuitive myoelectric signals from the biceps and triceps. The servo is a popular scheme for controlling the elbow because the user gets position information from the position and spring tension in the servo transducer without watching the prosthesis.

THE HUMERAL ROTATION PROBLEM

High level amputees lose the ability to position the terminal device accurately in space because they have lost internal and external rotation. There are three cases that must be considered. Case one covers almost all transhumeral amputees at present. These persons lose

their epicondyles, and even though they can still rotate the humerus, this rotation cannot transmit force to the prosthesis and thus accomplishes nothing. Case two covers the small number of persons who have received a Marquardt angulation osteotomy or who have implanted metal pseudo-epicondyles. [6, 7] For these persons a socket can be built that transfers humeral rotation into rotation of the prosthesis. This motion is under complete natural control. A third case is those persons who have received the benefits of osseointegration. Here the end of the humerus is rigidly attached to the prosthesis keeping the control completely natural with the added advantage that with osseointegration the user senses forces on the bone and which supplies some proprioceptive feedback.

Help for case-1 amputees

With a socket that stabilizes against inadvertent rotation**,** a simple rotation lock is the first component to add. This scheme is offered in the RSLSteeper Mark 14 Elbow and by a rotation lock made by Rim Jet [8] to go between the conventional elbow and the upper arm socket.

Richard Weir is reporting on a powered humeral rotator at this conference. There are no "left over" myoelectric sites to control this motion. However, the control muscles are almost always intact and only await the appearance of implantable myoelectric pickups to function.

The author has suggested another approach – to implant a powerful bar magnet in the end of the humerus and to use giant magneto-resistive sensors to detect humeral motion. This approach requires a drive like the one Weir will discuss.

RESEARCH SPONSORED BY THE U.S. DEFENSE DEPARTMENT

Early in 2005 the U.S. Department of Defense initiated two projects to improve the performance of upper extremity prostheses. The sponsoring agency is DARPA (Defense Advanced Research Projects Agency.) One initiative has a two-year time line and the other is for four years. The emphasis is on the transhumeral amputee with some attention being paid to shoulder disarticulation.

Goals of the DARPA Prosthesis 2007 Two-Year Project

The project [9] is designed to "improve the capabilities of upper-extremity prosthetic limbs beyond those currently available commercially by increasing range of motion, strength, endurance, and dexterity." The short term goal is to deliver a single prosthetic arm system suitable for transhumeral and shoulder disarticulation amputations leveraging recent advances in neural sensing & control, control systems, actuation, power storage and distribution, freeform manufacturing, micro fabrication, sensory feedback, flexure and transmission design, and signal processing. Some of the quantitative goals are arm free swing, local control, state sensing & task-based mode shifting within the device, simultaneous control of 3-5 joints, fingertip force sensing, local hand grasp slip control, elbow lift capability up to 20 ft-lb (27 Nm), grip strength up to 25 lbf (111 N), wrist flexion strength up to 1.67 ft-lb (2.26 Nm), robust intuitive control, 24 hour endurance until refuel or recharge, a minimum of 3 grasp patterns: fine pinch, lateral pinch, and power grip, wrist 2 degrees of freedom, humeral rotation, effective cosmetic cover, noise level below 50dB(A) at 1 meter, water & grit resistant, automated system for fitting to residual limb, modular to accommodate various stump lengths

Goals of the DARPA Four-Year Project

The following goals are excerpted from the DARPA website: [10] chronically implantable neural interfaces, sensory (afferent) feedback to the nervous system, able to differentiate afferent and efferent signals, simultaneous control of joints, internal hardware viable in vivo more than 1 year, biomimetic kinematics meaning a hand capable of a complete grasp set and an arm with free swing like an intact limb during walking, inertial properties that match the lost limb, weight, shape and appearance similar to an intact limb, robust in typical indoor and outdoor environments, amenable to direct skeletal attachment, wearable for 18 hours per day, modular to accommodate various stump lengths, effective cosmetic matching.

One can get an idea as to how functional this prosthesis is to be by looking at the following list of tasks that an amputee is expected to be able to accomplish: writing sample sentences, turning over 3" x 5" cards, picking up small common objects (e.g., paper clips, bottle caps), simulated feeding, stacking checkers, picking up large light objects, and picking up large heavy objects.

THE CHANGING ARCHITECTURE OF PROSTHETIC CONTROL SYSTEMS

Life was simple when all that was available was a hand with a bidirectional bridge and threshold electrode amplifiers for triggering open and close motion at fixed speed. Or was it? Even a simple system needed many unique polarized plugs with wires of many lengths.

The seven degree-of-freedom arm being discussed at this conference has 21 wires in 3 shielded cables crossing the elbow, and all but 4 are used. There are only four conductors passing through the most popular wrist rotator, and yet some people want to feed back grip force information while others want absolute position for true servo control of grip. Using a separate wire for every function already gives a lot of problems. Surely more complex systems will have to be put together differently.

The Need for a Common Industry-Wide Information Bus Structure

With Otto Bock, Motion Control, Liberating Technologies, and two DARPA contractors all meeting the same challenge of controlling more devices simultaneously, now is the time to create a standard bus structure for the industry. By the end of MEC'05 we should have a working group to begin the required discussions. Within four months there should be a white paper defining the problem. Some of the issues to be addressed are contained in the following list.

1. What will the power bus look like? Will there be a standard voltage, for instance the output of two Li-Ion cells between full charge and discharge. Can we find a connector that will fill the needs of most companies?
2. Will we need a chip-to-chip bus like the SPI bus in the Boston Digital Arm and in many similar systems?
3. Will we use a high-speed serial bus? If so what will the highest frequency be, and will special shielding be needed to prevent interference?
4. How many lines are needed in the information bus? Will there be an agreed voltage for supplying the information bus electronics?
5. What cables and connection schemes will be used to daisy chain the bus?
6. The scheme used by Bock to connect the rest of the system to the electrode amplifiers solves the problem of stocking cables of many lengths. Can this idea be carried into the standard bus structure?

7. Can we agree on a standard protocol for doing computer interfaces? Do we use cables and optoisolation, or is it better to standardize on a wireless technology?
8. Implantable myosignal detectors will be in use soon. Can we agree now how the information will be coded for transmission by RF or by visible or infrared light?
9. What conditions will be dictated by US and EU regulations?

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An Adaptive Prosthetic Hand with Compliant Joints and EMG-based Control

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***Abstract* – In this paper some recent results about the experimental trials we are performing on a functional prosthetic hand characterized by an EMG-control and by a simple and low cost fabrication technology are shown. A compliant under-actuated prosthetic hand has been designed and fabricated. The five-fingered hand (both palm and fingers) is moulded as a soft polymeric single part with compliant joints and embedded tendon driven under-actuated mechanism for providing adaptive grasp. The maximum measured cylindrical grasping force is 30 N. The one DoF prosthetic hand is controlled using two pre-amplified EMG electrodes. The proposed EMG-based control is a Finite State Machine (FSM). A particular attention has been given to the calibration phase. In order to identify the end of the grasp, the intensity of the current is monitored. Moreover, the microcontroller stops the motor when the average current overcomes the value imposed. Compared to other EMG based controllers, the approach proposed is very simple but it presents a good robustness and needs a minimum computational cost.**

Index Terms – Biomechatronics, Prosthetics, Compliant joints, Under-actuation, Rehabilitation.

I. Introduction

In the last 30 years very innovative prosthetic hands have been developed. Nevertheless, from epidemiological analyses, it emerges that about 35% of the upper extremity amputees do not use their prosthetic hand regularly [1].

The reason can be found not only in the poor functionality of presently available prosthetic hands, but also to psychological problems, and is overstated by the modification of cosmetic appearance of the upper extremity.

The research described in this paper results from the experience achieved with a number of projects running at ARTS Lab, that have produced different hand designs, as described in previous papers (e.g., [2-4]).

The ultimate aim of the research in this field is to connect the artificial hand to the brain, but at present research efforts are still focused on the development of components, like neural interfaces, that represent real breakthroughs. Along with this research, there is plenty of space for improving the grasping ability and cosmetics of present hands in order to provide better prostheses for short term clinical application. To this aim, the research described in this paper is addressed at fulfilling cosmetic appearance requirements while keeping as low as possible the production cost, and in this framework a novel and simple fabrication process for prosthetic hands is proposed. A hand prototype has been designed, fabricated and tested in laboratory, and is almost ready for user trials. The basic design and fabrication technology have been partly described in [4].

The hand has only one active DoF (Degree of Freedom) obtained with a commercial motor and a tendon driven under-actuated mechanism that allows distributing the force among the fingers, by adapting the grasp to the object shape. All mechanisms are embedded in a "soft" silicone structure that provides cosmetic appearance and compliance. Finally, the hand prosthesis is controlled by EMG (Electromyographic) signals generated by voluntary user's muscles contractions.

This paper summarizes the basic design approach, describes production process, it is then focused on the prosthetic hand control based on the processing of EMG signals and on experimental analysis.

II. The Prototype

A. The Design Guidelines

The classic design approach in myoelectric prostheses is the *intrinsic* actuation, where all the actuators are embedded in the hand structure. The outcome of this approach is producing artificial hands with one (or a maximum of two) DOFs, which are able to provide a pinch force of about 100 N; in this case, the motion of the phalanges is determined at the design stage and therefore no grasping shape adaptation is possible, also because

of the rigid transmission [5]. On the contrary, if smaller actuators are used, the prosthetic flexibility can be raised and the DoFs can be increased [2]. Anyway, miniature actuators are quite far to offer a real alternative to standard electro-magnetic actuators. This is mainly due to the lack of suitable high torque micro-actuators and the difficulty to implement complex control scheme with a natural interface able to control all the DoFs.

In order to enhance prosthesis flexibility by keeping the *intrinsic* actuation solution and implementing simple control algorithm, we presented in [3] a design approach based on under-actuated mechanisms [6-7].

A mechanism is called under-actuated when it has less actuators than DoFs; traditional actuators (i.e. electro-magnetic motors) are replaced with passive elastic elements and mechanical stops. These elements can be considered as passive actuators, which cannot be controlled. In order to increase the number of DoFs without increasing the user's cognitive burden, we exploited the under-actuation concept as in the Hirose's paper [6]: each finger has 3 DoFs (3 phalanges). The cable is fixed on the fingertip and runs around the idler pulleys in the joints. A DC Motor pulls the cables which flex the fingers, as the *flexor digitorum profundus*. When the motor releases the cables, torsion springs in the joints extend the finger.

Inspired by the human hand, these prostheses have metallic phalanges and pulleys acting as the human skeleton, an actuation system (DC motor(s)) in the palm or in the forearm as muscles and a transmission cable system as the flexor tendons. Both the sensory and the control systems have also been implemented.

Compliant joints, on the other hand, are made up of one continuous part which deforms properly in certain areas in order to achieve motion. As a compliant flexible member bends, the external energy is stored in the form of strain energy. This stored energy is similar to the strain energy in a deflected spring, and the effect of the spring may be integrated into a compliant mechanism design [8].

The main drawback of compliant joints is the relative difficulty in analyzing and designing compliant mechanisms. The main advantages of the compliant joints are cost reduction (part-count reduction, reduced assembly time and simplified manufacturing process) and better performance. Weight and maintenance are decreased; nevertheless reliability and precision in small joints are higher. Also wear and lubrification are reduced; as the mechanism needs fewer mobile joints to carry out the same task. Some of them may be manufactured from a mouldable material and be constructed only of the same piece. Some innovative and promising examples of artificial hands based on compliant joints have been developed in the recent past [9-12] but they still have several assembled components, in general made out of different materials, driving their synergy in order to get the motion thanks to different stiffness.

The hand presented in this paper is moulded as a single piece of the same soft material, in order to allow a simple and low cost production process, thus resulting in a novel methodology to get compliant joints in a unique compliant structure. At this time, the analysis and synthesis processes are so complex and only experimental analysis of the solution adopted validate our works. The results obtained so far are encouraging and led us to fill a patent for further industrial development.

B. The soft hand

The prosthesis has got a really anthropomorphic appearance and kinematics, even if its structure is quite simple. The hand has four articulated fingers, each finger has three joints, but the opposable thumb has two joints. It is possible also to make the palm adaptive to the object, hollowing it between the index and the middle and the thumb.

A DC motor, with its gearbox and a power-off brake, provides the winding of a cable on a reel and its releasing, which in turn moves a slider pulling the five tendons, one for each finger. Both the actuation and the transmission systems are placed inside the palm. There is more than a simple connection between the tendons and the fingers. The phalanxes are flexed when the cable is pulled but, after the first finger touches the object to grasp, a differential mechanism, based on the same cable, allows a further stroke of the slider. When the cable is released the elasticity of the material extends the finger.

There are 10 DoFs and one actuator. Pinch grasp and cylindrical grasps are allowed; under-actuated grasping mechanism does not provide any manipulation task.

The specific force distribution to each finger changes depending on the grasp type. So every phalanx joint has to be shaped properly in order to exploit the output power share as similar as possible to the natural hand. Moreover, by adapting each finger impedance, it is possible to drive their order of movement. This makes the grasping action definitely more anthropomorphic and stable.

When the hand closes, a further balancing differential mechanism distributes the force between the different fingers and makes the grasp further *adaptive.* As a finger touches the object first, the others keep on moving for a little, making the grasp involving all the fingers . At the same time, the actuation gear system increases the torque and decreases the closing speed, in order to augment the stability. So this under-actuated prosthetic device can perform an automatic finger wrapping around the object without any intervention of the user and above all without the need for dedicated sensors embedded in the phalanxes.

Fig. 1: The silicone prototype and the polyurethane prototype performing a grasp

C The EMG-based control

The prosthetic hands can be controlled by the user thanks to an EMG-based discrimination algorithm. Using EMG signals recorded from the upper limb muscles is a common and simple approach for controlling active prosthetic hands [13]. In fact, surface electrodes are easy to use and manage, do not require any surgery and do not impair forearm movements.

However, it is important to point out that EMG signals permit the control of few degrees of freedom (generally, no more than one or two active DoFs) and, above all, the user's interface must be very friendly and handy in order to enable practical long-term use of the device. In fact, the user cannot be productive if she/he must spend a large portion of her/his energy and concentration controlling the artificial hand.

Therefore, a very simple and robust EMG-based algorithm able to control opening and closure of the hand has been developed.

This control algorithm is focused on the idea of using surface EMG signal detected from voluntarily activated muscles in order to generate the input signals for the simple Finite State Machine (FSM).

For this reason, a microcontroller of Microchip (PIC16F876) equipped with a 10-bit Analog-to-Digital (A/D) Converter module has been used in order to acquire and digitalize EMG signals extracted by using a couple of surface commercial electrodes (Otto Bock, 13E125 Myobock Electrode) placed on two antagonist muscles (e.g., biceps and triceps brachialis, or flexor and extensor of the forearm, depending on the level of the amputation).

In order to generate the two digital inputs for the FSM, a simple algorithm compares the EMG signals with two different voltage levels (threshold VTH1 e VTH2).

In particular, during the calibration phase, the microcontroller measures the EMG signals without muscular activity (VBL) and calculates the activation thresholds (VTH1 e VTH2) for the two antagonist muscles selected with the following formulas:

$VTH1 = VBL + \Delta VTH1$

$VTH2 = VBL + \Delta VTH2$

in which $\Delta VTH1$ e $\Delta VTH2$ are fixed values set in empiric way. Nevertheless, these two parameters are stored in an electrically erasable programmable read-only memory (EEPROM) of microcontroller and can be changed via software in order to allow an on-line tuning.

Obviously, the choice of $\Delta VTH1$ e $\Delta VTH2$ affects the signal processing: a low threshold is more noise sensitive, a high one could be less precise in detecting fine movements.

Moreover, a software hysteresis has been developed in order to avoid the generation of false input signals. The use of hysteresis greatly improves noise immunity. In fact, without hysteresis, slow rising or noisy inputs may cause oscillations occurring while the slow rising input signal crosses through the input threshold. Such oscillations can cause false triggering leading to FSM reliability problems.

By the means of an added hysteresis, the digital input will not trip high until the input signal crosses an upper voltage threshold (VTH-Rise) and will not trip low until the input signal crosses lower voltage threshold (VTH-Fall) as showed in Fig. 2.

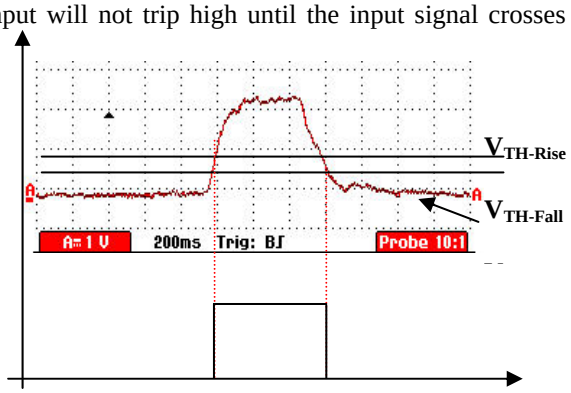

Fig. 2: Generation of input signal for FSM

The EMG-based control proposed is a simple Finite State Machine (FSM) showed in Fig. 3.

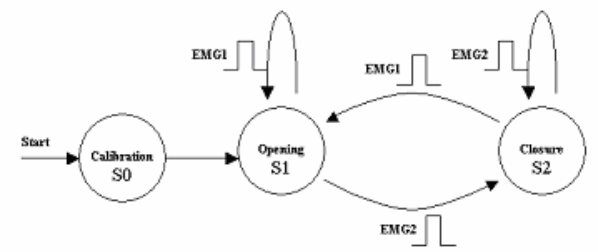

Fig. 3: State Diagram of FSM for the EMG-based control algorithm

The starting state (S0) of this algorithm is dedicated to the calibration phase (initialisation of FSM). After ΔT ms, the FSM goes in the state S1. In this state the opening of the prosthetic hand begins: the microcontroller sends the enable signal and directions signals to a high voltage, high current dual full-bridge driver (L298). A Hall-effect switch is used to detect motor stop during the opening of the hand. The FSM remains in the state S1 until the EMG signal produced by flexor contraction of the forearm does not overcome and drop under its relative threshold V_{TH1} (generation of the EMG logic pulse shown in Fig. 2). When it happens, the FSM advances in the state S2. In this state, the closure of the hand begins. In order to identify the end of the grasp, the intensity of the current is monitored. The current flowing through motor comes out from the bridge at a resistor. This acts as a sense output for detecting the intensity of this current. This sense output voltage is filtered with a 2nd order Butterworth low-pass filter, amplified and sent to a microcontroller (in Fig. 4 is showed the conditional circuit and in Fig. 5 the waveform of the current during a repetitions of closure and opening of the hand).

The microcontroller stops the motor when the average current overcomes 270 mA.

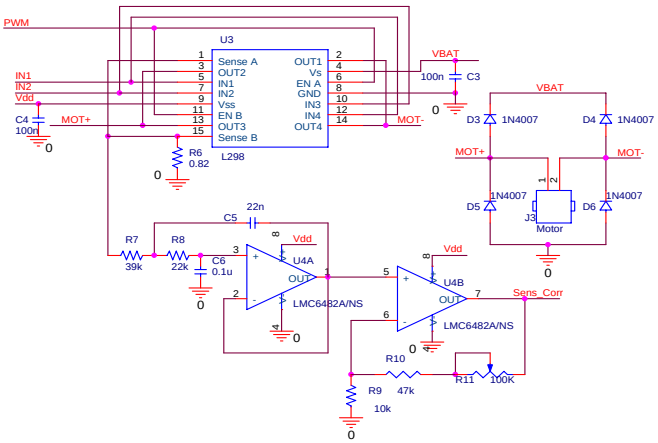

Fig. 4: Conditional circuit for measurement of current.

The FSM remains in the state S2 until the EMG signal produced by extensor contraction of the forearm does not overcome and drop under its relative threshold V_{TH2} (generation of the EMG logic pulse shown in Fig. 2). Compared to other EMG based controllers, the approach proposed is very simple but it presents a good robustness and needs a minimum computational cost.

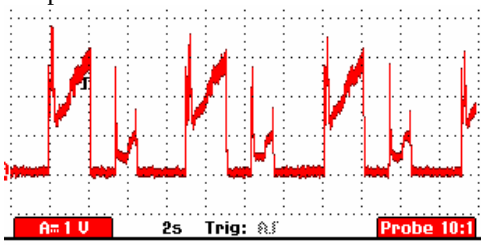

Fig. 5: The waveform of the current during repetitions of closure and opening of the hand

In order to allow “a control of the force” during the closure, it is possible modified the FSM in the following way: the closure of the hand begins when the EMG signal produced by flexor contraction overcomes the threshold VTH1 and stops when the EMG signal falls under . In this way the patient can control the grasp force changing the duration of digital input. Of course, the controller stops the motor when the average current overcomes the value imposed also in this working mode.

This method has been tested with four different able-bodied subjects proving very good results in terms of controllability of the prosthesis (more than 98% of correct classification during the test phase).

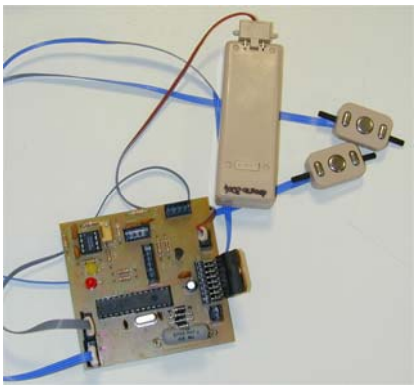

Fig. 4: The control unit with battery and EMG electrodes

III. The Experimental Results

The innovative prosthetic hand has been evaluated with an appropriate experimental protocol. The whole structure is obtained by a single casted part of soft material and the mechanical components are reduced for obtaining an anthropomorphic articulated prosthesis based on a single compliant joint structure. Because of this non conventional technology, the hand has some unpredictable behavior, and thus the experimental phase and fatigue/duration tests are a mandatory step before the user trials.

The experimental tests are aimed at assessing grasping ability of the hand in the two fundamental configurations: pinch grasp for 20 mm diameter objects and cylindrical grasp of object of diameter up to 80 mm.

In addition, as it is shown in Fig. 5, the resulting grasping force has been measured with a dedicated experimental set-up based on a commercial load cell (Vishay Tedea Huntleigh, Basingstoke UK), in order to evaluate the force exerted during a palmar grasp.

The EMG control, as it has been implemented for prosthetic purposes, set the force at 13N with a current threshold value of 270 mA. By-passing this limit, we pushed the prosthesis up to 30N, with a threshold of 750mA. The mechanical limit has not reached yet, because the motor current limit is 900mA and the cable bears up to 50N; both the materials used have not shown any stress problem. The critical point seems to be the wear (particularly for cable) rather than the maximum load.

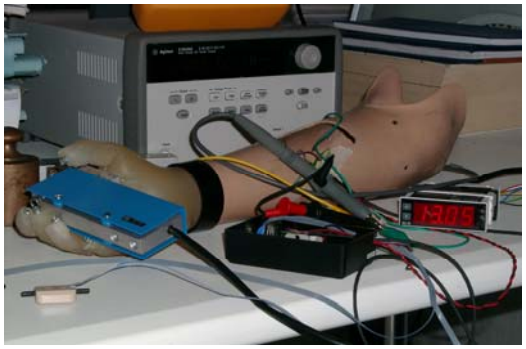

Fig. 5: The experimental set up for maximum force exerted

In order to prevent failures during the product operation, fatigue tests have been planned and an appropriate experimental test-bench has been fabricated. A grasp cycle with four different thicknesses, carried on 1000 grasp, will simulate prosthesis operation; this set-up has been designed with the Prosthetic Centre advice according to the objects normally grasped during activities of daily living. To this aim, the hand grasped a rotating cross with four different thicknesses on each arm: a counter takes out the number of grasps occurring before a failure event. Setting the grasping force at 15 N, the prosthesis performs 1668 grasp tasks in the first test and 1589 grasp tasks in the second.

Furthermore, we can compare some grasping tasks with the Otto Bock Prosthesis. The latter one is able to exert 89N during a cylindrical grasp. However, thanks to the increase contact area, the Soft hand is able to achieve similar results in terms possible grasping tasks even if it can exert a lower grasp force. As shown in Fig. 6, the Soft Hand performs an *enveloping* grasp wrapping all the phalanges around the conical object. On the

contrary, the Otto Bock Prosthetic hand has few reduced contact areas. The large amount of power grasp is wasted because of the little contact areas. The articulated compliant hand can perform the same stable grasp with a lower force level.

Fig. 6: The enveloping grasping of the Soft and compared with the grasp of a commercial prosthesis with rigid transmission

IV. CONCLUSION

This paper presented the Soft Hand, a prosthetic hand made out of a single production process based on soft material casting. It provides adaptive grasp functionalities thanks to an under-actuated tendon driven mechanism controlled by a single actuator embedded in the hand palm. The prosthesis is endowed with an EMG control. The finger dynamics are driven by the means of a cable tendon system, and purposely shaped joints providing a cosmetic appearance and motion. An extensive experimental testing activity is in progress for assessing the design and the engineering solutions before going to user trials. The first results of the grasping force and fatigue trials are encouraging and have showed us new guidelines for redesigning the prototype. Eventually, the obtained results will be exploited in the field of body powered hand prosthesis for obtaining low cost devices for humanitarian applications.

ACKNOWLEDGMENT

The useful work of our colleagues Roberto Lazzarini and Stefano Roccella is acknowledged. The work described in this paper has also been supported by the RTR4 Project funded by INAIL Prosthetic Centre, Vigorso di Budrio, Bologna, Italy and by the CYBERHAND Project (IST-FET Project #2001-35094).

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A BENCH-TOP PROTOTYPE OF A VARIABLE STIFFNESS PROSTHESIS

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ABSTRACT

A prototype of a variable stiffness prosthetic joint has been constructed and tested. The joint is based on two actuator subsystems arranged in an antagonistic configuration. Each actuator subsystem is composed of a small, high-speed electric motor, a single stage, worm-gear based transmission and a nonlinear stiffness element. Each nonlinear stiffness element is composed of a set of short sections of elastic tube that was chosen based on its stiffness characteristics. The system is powered by batteries and can be controlled in a number of ways. The paper will present the basis used for selecting the nonlinear stiffness elements, the details of the design as constructed and a comparison of the actual performance of the prototype to predictions based on design calculations and simulations. The prototype has performance that is comparable to commercially available prostheses and showed good correspondence between simulation and prototype. The prototype was able to lift a 2 kg load through 135 degrees in 1.42 seconds and to vary its stiffness from 14 to 24 Nm/rad.

INTRODUCTION

A variable stiffness elbow prosthesis should look and feel more natural and allow the amputee to perform constrained or contact movements with the arm at a low stiffness, and fast, precision movements at a high stiffness. Previous research suggests that it should simplify control issues, provide increased dexterity and allow greatly improved energy efficiency when interacting with the environment [1, 2, 3, 4, 5]. Antagonistic nonlinear compliant actuators acting about the joint provide stiffness variation capabilities. A nonbackdrivable transmission allows external forces to be statically supported without using the motors.

A variable stiffness prosthetic arm bench top prototype was designed, constructed, mathematically modelled and tested. The configuration consists of two actuators arranged antagonistically around the elbow joint. Each actuator is made up of a dc motor, gearhead, nonbackdrivable transmission, nonlinear spring and two pulleys. Figures 1 and 2 show the resulting prototype.

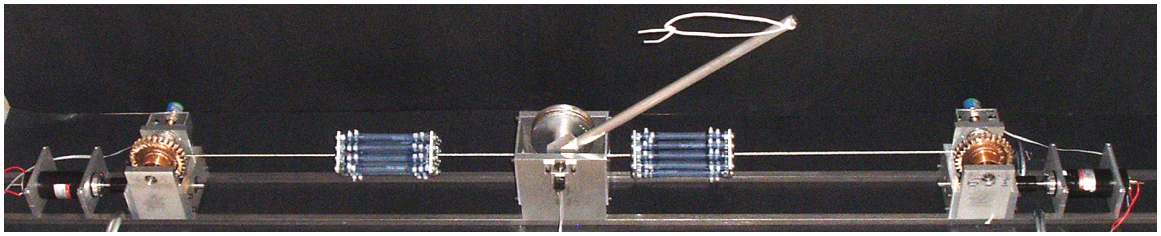
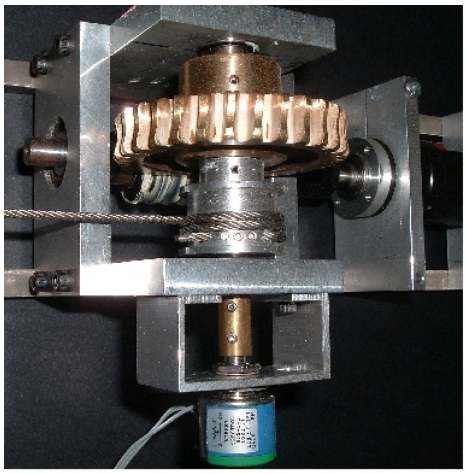
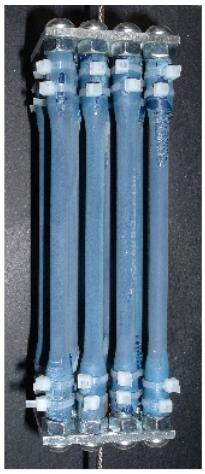


Figure 1: Bench top prototype



a) Motor and transmission



b) Thera-band spring

Figure 2: Bench top prototype - motor, transmission and spring

DESIGN DETAILS

Using design criteria based on currently available upper arm prostheses [6, 7], the components were chosen. A Maxon RE35 dc brush motor with a 4.8:1 gearhead was chosen for its torque and speed capabilities. A worm and worm gear set from Boston Gear makes up the transmission. A single-start, hardened steel worm and 30 tooth, bronze worm gear with a diametral pitch of 12 met the strength and wear criteria for the chosen motor running at its thermal torque limit.

There are two types of nonlinear springs. A hardening spring is a spring in which the stiffness (derivative of spring force with respect to displacement) increases with displacement and a softening spring has a stiffness which decreases with displacement [3, 8, 9]. It has been shown [10] that an antagonistic pair of hardening springs can create either a linear, hardening or softening joint depending on the spring’s curvature. This is also true for a softening spring. Therefore, either a hardening spring or a softening spring can be used in this design.

The two main springs studied were the belleville washer and rubber tubing. Both of these act as softening springs that produce a hardening joint. It was determined that although the belleville washers have ideal spring force and stiffness characteristics, there is too much friction between the washers, creating a much higher spring force than originally calculated. This causes problems with the motor’s ability to deflect the spring. The rubber tubing was chosen and rather than one very thick tube, ten 5 cm lengths of blue Thera-Band tubing are used to create an arm stiffness within the required range.

The radii of the two remaining pulleys were determined to create the desired total gear ratio required to move the arm through 135° in 1.2 s. The elbow pulley radius is 0.035 m while the worm gear pulley radius is 0.0144 m. The system is powered by a 14.4 V, 4500 mAh NiMH rechargable battery.

PROTOTYPE PERFORMANCE

Testing has shown that the prototype works as expected and agrees with the Matlab simulations. Notable differences are higher current and friction which caused the simulation to be optimistic with arm speeds. The differences can also be attributed to inconsistencies in spring length, motor parameters, unmodelled electrical connections and drive friction.

The arm's performance when lifting the 2 kg load was 1.42 s for the full range of flexion. This is very close to the design goal of 1.2 s. The test speed was slower than the simulation for the stiffness change. The motors could not handle the load of the springs at the fully stretched length and gradually slowed to a stop. A spring with a lower overall force would solve this problem.

Both the simulation and the tests showed a relatively constant stiffness can be maintained during a position change. The stiffness increased a small amount during the task due to the spring force and a speed difference was evident between the two uncontrolled motors. This happened because a higher spring force (which occurs at a lower stiffness) puts more torque on the motor. In the agonist motor's case, this causes it to slow down. For the antagonist motor, however, the force pulls the same direction as the motor is turning, which decreases the work required by the motor, decreases friction in the transmission and allows the motor to turn faster. Upon reaching a new equilibrium, the arm stiffness is slightly higher because, overall, both springs have a decreased deflection value. This increase is higher for a lower initial arm stiffness, which gives a higher internal spring force.

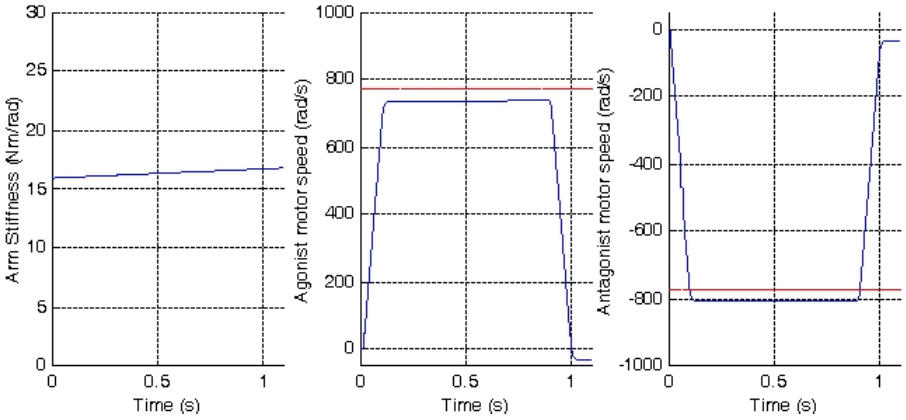


Figure 3: Position change simulation results

When the arm stiffness was decreased from maximum to minimum stiffness, the arm position lowered slightly in both the simulation and tests. This is due to the mass of the arm creating a change in equilibrium position. As the arm stiffness decreases, the same arm mass causes greater arm deflections from the no load equilibrium position. A second factor is that this arm deflection causes the agonist spring to stretch, increasing the internal spring force, and the antagonist spring to compress, decreasing the spring force. This causes the agonist motor to run slightly slower and the antagonist motor to run slightly faster. This also results in a small decrease in arm position. The total decrease in arm position is less in the test than the simulation due to a lighter arm mass.

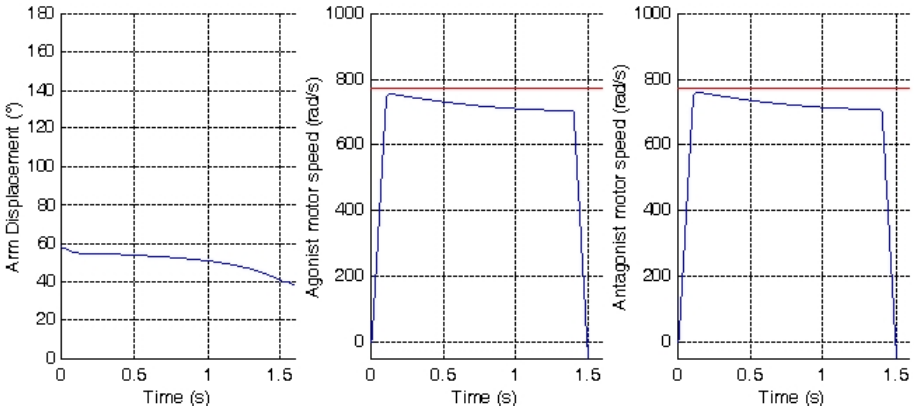


Figure 4: Stiffness change simulation results

CONCLUSIONS AND RECOMMENDATIONS

The performance of the resulting prototype agrees with a mathematically modeled simulation and also is comparable to currently available upper arm prostheses. This prototype will be used as a bench model to test new actuators and springs as well as different control methods. The size and weight must also be reduced to create a wearable prosthesis.

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NON-BACKDRIVABLE SERIES ELASTIC ACTUATOR FOR USE IN A PROSTHETIC ELBOW

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Abstract: Commercially available prosthetic elbows have stiff actuators (motors) that are only capable of motion (position or velocity) control. In an attempt to mimic human physiology while accommodating prosthetic demands, a non-backdrivable motor has been created that is less stiff and capable of impedance control. Impedance control responds well to different environments and the presence of perturbations. Results have shown that this motor, though non-backdrivable to ensure sufficient battery supply, is still capable of exerting sufficient torque, speed, and frequency bandwidth to be useful in prosthetics. In the future, patients will be fit with this type of motor to examine if they objectively and subjectively perform better using this more physiologically appropriate prosthesis.

INTRODUCTION

Commercially available electrically powered prosthetic elbows are stiff and unyielding. Making these artificial limb replacements more closely mimic human elbows by reducing their stiffness may be beneficial. Several other potential advantages of reducing the stiffness of prosthetic elbows include creating elbows that are less likely to break in the event of a fall, improving movement cosmetics, and improving physical interaction with the environment.

In addition to having reduced stiffness compared with current electrically powered prosthetic elbows, humans modulate the stiffness of their joints depending on the task. Different tasks require different levels of interaction with the environment. Able-bodied persons modulate the stiffness of their limbs in accordance with the task by co-contracting their muscles.

English and Russell [1] have suggested using two motors coupled to two quadratic springs in order to mimic the modulation of stiffness found in humans. The size and weight requirements of a second motor, however, are not conducive to application in prostheses. An alternative approach that English and Russell acknowledged was to have a microcontroller control the prosthesis impedance. English and Russell rightly dismissed this option at the time for two reasons:

1) microcontrollers were not capable of doing the additional calculations required to use impedance control
2) Motor-generated torques require a constant power drain, which is unreasonable in prosthetics.

The recent use of microcontrollers in prosthetics [2] has made the limited computational complexity of impedance control viable in prosthetics. In order to generate torque without an energy drain, the authors propose nonbackdrivable series elastic actuators (SEA). Aside from side-stepping the need for complex and precise quadratic springs required by the two motor approach, non-backdrivable SEA only require one motor and take no additional space. Perhaps most importantly, this method of control may easily be incorporated into currently existing prosthetic elbow designs.

SEA are force controllable actuators with low impedance and high fidelity. They translate the accurate position control of traditional DC motors to accurate force control through the use of a spring. They have several advantageous properties, including reliable force output, simplicity, robustness of design, and the use of traditional robotic actuators. Pratt et al. cite greater shock tolerance of gears due to low pass filtering of torques, lower reflected inertia, more accurate and stable torque control, less damage during inadvertent contact, and the potential for energy storage and return as reasons to use series elastic actuators [3]. SEA have been used in walking robots and their underlying concept of reducing stiffness has been applied to such robots as the

Honda P2 [4], though the P2 does not modulate the stiffness. The ability to modulate the effective stiffness of the actuator is appealing in areas such as impedance control and admittance control, which have application in human machine interfaces and prosthetics [5, 6].

The reduced stiffness in SEA reduces their ability to fluctuate rapidly between high amplitudes of positive and negative torque, known as large-force bandwidth. In addition, their stable bandwidth is reduced due to the interaction between compliance, stiction, and backlash [7]. To alleviate this latter problem, SEA have relied on backdrivable transmissions to minimize backlash and stiction. Unfortunately, these previous systems would not be practical in applications that require non-backdrivable actuators to minimize power consumption. For example, an upper limb prosthesis user will often pick up an object and carry it with them. The actuator should be turned off after the proper position has been achieved to conserve power. A backdrivable system would consume power during the entire time that the object is held, whereas a non-backdrivable system could maintain the desired position without power. Because portable power sources have a limited power capacity, backdrivable actuators are not practical for use in prosthetics.

DESIGN

Non-backdrivable actuators have potential application in a wide range of prostheses, including wrists, elbows, shoulders, and even possibly knees and ankles. For an initial prototype, however, it was determined that the requirements for an elbow prosthesis were most easily realized. Prosthetic wrists, which require smaller size, and prosthetic shoulders, which require larger torques, are being investigated based on the work of this paper.

We have designed an actuator that uses a harmonic drive gear transmission to prevent backlash while at the same time creating a non-backdrivable transmission. A torsional spring with a stiffness of 182 Nm/rad is used. A customized Emoteq[i] HT02500 frameless brushless motor capable of producing 2.8 Nm stall torque and a 160:1 gear ratio Harmonic Drive Systems CSD 20 are used, controlled by a Composite Modules Inc.[ii] 5015-28 controller capable of handling large currents. Instrumenting the Harmonic Drive itself was investigated and the authors were able to decrease the inherent torque ripple associated with that technique. We concluded, however, that the torque resolution was inadequate for low magnitude torque control with this approach [8, 9].

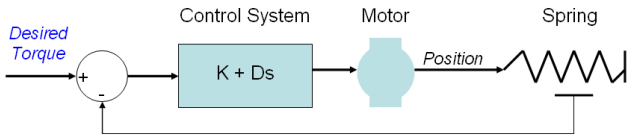
Torsional Spring Design

The authors have found that a modified shape termed a *spandrel* offers improved geometric resiliency, and propose to use this geometric cross section to increase the robustness of future designs. The cross section of the *spandrel* is illustrated in Fig. 1.

Fig. 1. Cross section of a spandrel that provides improved resiliency

Actuator Design & Results

A bench-top actuator has been designed and fabricated to illustrate the feasibility of the concept. A prosthesis prototype is being developed. A model of the prosthesis prototype is shown in Figure 2. Motor characteristics are presented in Table 1.



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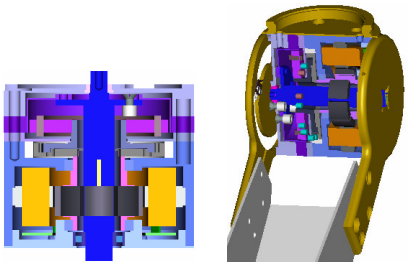

b

Figure 2. SEA design

a) Control schematic: the motor generates an accurate position. This position is fed through a compression spring, which converts accurate position into accurate force. A force transducer measures the compression and converts it into a force reading, which is then compared to the desired force. The error between the two signals is sent to a control block. In this example, the control block multiplies the error signal by a gain (K) and the derivative of the error by another gain (D) and sends this signal to the motor to correct the output force.

b) Prosthesis integrated design

Table 1
Motor Characteristics

Williamson data from [10, 11]

Increased stiffness of author's spring increases bandwidth.

Nonbackdrivable transmission lowers author's bandwidth.

Parameter	Units	Authors	Williamson
Sensor Stiffness *ks*	*Nm/rad*	182	45
Natural Frequency ω_n	*Hz*	15.7	38
Cuttoff Frequency ω_c	*Hz*	19.4	N/A
Damping coefficient ζ	*no units*	0.725	0.103
Stiction Torque	*Nm*	0.1	N/A
Max Power	Watts	42	35
Max Torque	*Nm*	26.2	13
Max Speed	*rad/sec*	5.1	4.2
Efficiency	*%*	10.8	N/A

DISCUSSION

A non-backdrivable *Series Elastic Actuator* has been created that has an adequate bandwidth at +/- 2Nm in spite of its non-backdrivable nature. The maximum torque is larger than any prosthetic elbow commercially available and speed is adequate. The bandwidth has been achievable in large part due to the selection of the transmission system used: a Harmonic Drive with no backlash. Harmonic Drives are only non-backdrivable for small torques (<6 Nm). To address this issue a proposed prosthesis design will use a non-backdrivable octagon roller clutch with minimal backlash on the input stage of the gear. The backlash of the roller clutch will be attenuated by the 160:1 gear transmission, making output backlash negligible.

The actuator has larger maximum torque, speed, and power than the rotary *Series Elastic Actuator* created by Williamson [11], but the cutoff frequency is not as high as that of Robinson's linear SEA [12]. Williamson did not report a single cutoff frequency. More detailed descriptions of the results presented in this paper are available [8, 9, 13-16].

A biomimetic actuator capable of transmitting torque has been fabricated with adequate bandwidth and resolution. The characteristics of this motor, similar to those of physiologic muscle in that it is compliant and able to transmit torque, lend themselves to a subset of position control termed impedance control, as defined by Hogan [5, 17]. In impedance control, the user specifies a desired position and impedance, and then a torque is commanded in light of the discrepancy between the desired and actual position. Abul-Haj and Hogan [18] have shown that this concept may be used in a prosthetic elbow. Future work will examine if users do indeed modulate the impedance of their prosthesis when given the option, and to what extent this ability increases their performance. Popat et al. [19] offered support for this theory in a study that examined four men with above-elbow prostheses, where it was concluded that reduced stiffness in the elbow facilitated rotation of a crank; a bench test used to assess dynamic movement of prostheses. English and Russell [1] extend this advantage to any event in which tasks are done that interact with the environment.

Related work by the authors [20] has shown that the parallel nature of the sensor in Series Elastic Actuators provides incorrect force representation in the presence of static friction. While this is not a problem for rotary motors, which have insignificant friction in parallel with the elastic component, it is a significant problem in linear actuators. As a result, the authors recommend in linear Series Elastic actuators that a sensor be placed in series with the spring rather than in parallel with the spring to provide improved results.

CONCLUSIONS

This paper has demonstrated that non-backdrivable *Series Elastic Actuators* can be realized with adequate frequency resolution to have potential in prostheses. Non-backdrivability is critical in the design of prostheses and many other areas of robotics for the preservation of power. The practical implementation of impedance control will open new avenues of control of prosthetic devices and potentially make prostheses more biomimetic in the future.

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[i] Emoteq, Inc. a division of Hathaway, Tulsa Oklahoma, [ii] Composite Modules, Inc. Attleboro, Massachusetts

THICK-FILM PIEZOCERAMIC "SLIP SENSORS" FOR A MULTIFUNCTIONAL PROSTHETIC HAND

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ABSTRACT

The majority of prosthetic hands lack an intelligent feedback control system meaning that the user has to rely on visual feedback to detect whether an object is being crushed or if it is slipping out of the hand. Recently, a new type of fingertip for the Southampton Hand has been developed encompassing an array of thick-film sensors to measure grip force and the onset of object slip. There are three types of sensor used: piezoresistive thick-film sensors to detect the force on a finger, a piezoelectric thick-film sensor to detect the onset of slip and a thick-film thermistor to monitor temperature.

Some initial results are presented for the "slip" signals produced from the thick-film piezoelectric sensor. The sensor has already shown its ability to differentiate between the initial contact with an object and the object sliding past the fingertip. It may also be able to determine several variables and parameters such as: object acceleration, the coefficient of friction between the sliding surfaces and the force applied by the fingertip to the sliding object. There characteristics could then be used in a closed loop control system.

INTRODUCTION

Conventional prosthetic hands generally have low functionality, with their movement limited to just one degree of freedom. This single movement is normally implemented by combining the first and second fingers with the thumb to form a tri-digital grip. These devices also have little or no tactile sensing capabilities, which means the user has to visually assess the stability of the object in the hand. To compensate for this most conventional devices have adopted a high grip force strategy [1]. However, this does not allow the user to easily handle or manipulate delicate objects. The Southampton Hand is a prototype myoelectrically driven prosthesis, which has been developed over a number of years to investigate a range of potential improvements to prosthetics. These improvements include the control system, mechanical functionality and the integration of sensors for automatic control. Previous work has investigated the potential of optical and capacitive based force sensors, with microphones to detect slip at the fingertips [2,3,4]. However there were a number of disadvantages using these types of sensors including erroneous hand closure when an external sound was detected and relatively high power consumption from the optical force sensors.

Recent work has concentrated on integrating an array of thick-film piezoresistive, piezoelectric and thermistor sensors into a fingertip to monitor the static force, dynamic force and temperature of the hand. Two prototype fingertips have been successfully fabricated and the piezoresistive and thermistor sensors have been characterised [5,6].

Initial experiments and test results from a piezoelectric thick-film slip sensor have focused on the different signals produced by varying applied fingertip force, change in coefficient of friction with the contacting materials and object acceleration.

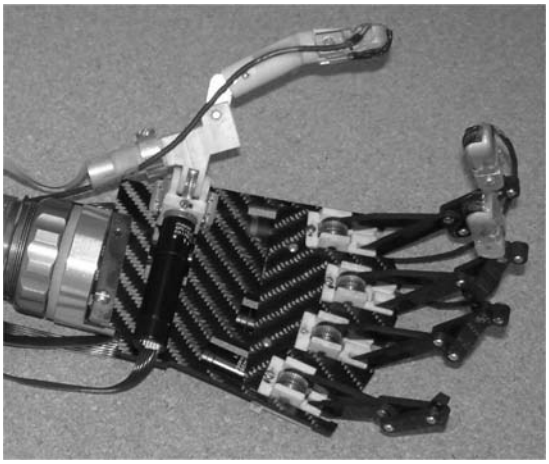

Figure 1: Southampton REMEDI Hand.

THICK-FILM SENSOR TECHNOLOGY

The whole fingertip sensor array is fabricated using a screen printing process, which means that the sensor material is initially in a paste format. This paste is placed on a wire mesh screen covered in a thin layer of UV sensitive emulsion with the sensor patterns photolithographicaly reproduced onto the emulsion, allowing the paste to be deposited onto a substrate (e.g. the fingertip) in the required design [7].

The thick-film piezoelectric sensor paste has been developed in house and is constructed from the three main components common to all thick-film ceramic pastes. These components are an active material, which provides the sensor properties, a glass binder, which when fired at a high enough temperature forms a solid matrix and adheres the sensor to the substrate and a solvent thinner, which added in the correct quantities allows the paste to be screen printed. The active material used in this sensor is a lead zirconate titanate type 5H powder (PZT-5H, supplied by Morgan Electro Ceramics) mixed with the glass binder (CF7575, supplied by Corning), and the solvent (ESL 410, supplied by Electro-Science Laboratories) [8]. The thick-film piezoelectric sensor utilised on this prototype fingertip is of a capacitive design, which means the piezoelectric layer is sandwiched in between two electrodes.

After the paste has been deposited it undergoes a drying process to remove any solvent remaining in the print and is then fired through a belt driven furnace for approximately 1 hour reaching temperatures up to 1000ºC. This high temperature is required to sinter the piezoelectric powder and melt the glass binder, thus adhering the sensor to the substrate.

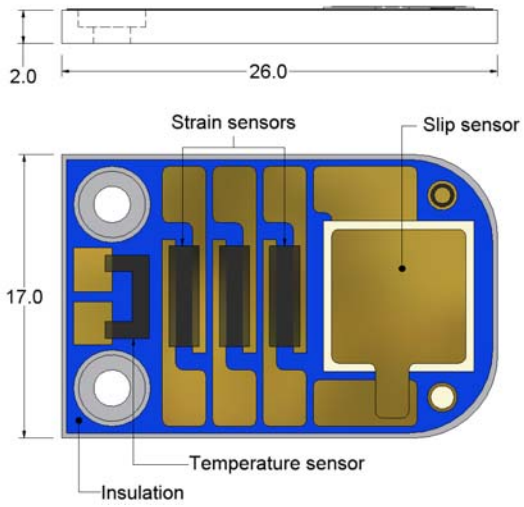

Figure 2: Second prototype fingertip with dimensions in mm.

Figure 2 shows the design of a prototype fingertip and the array of sensors which have been printed onto the surface of the fingertip. The substrate used to make the fingertip is 2mm thick stainless steel type BS 430S17. The two holes on either side of the temperature sensor are to allow the fingertip to be bolted to the end of the finger and provide a means of readily changing the fingertip if any of the sensors become damaged.

SLIP TESTING METHOD

To reproduce slip a test rig was designed and is illustrated in Figure 3. It consists of a large aluminium block, which slides horizontally on a section of aluminium angle. The block is attached to a basket containing weights, via a cable, which runs over a pulley. The basket can be loaded with different weights to produce a range of forces, which leads to a change in the slip acceleration between the sliding block and the fingertip. To alter the coefficient of friction between the sliding block and fingertip a range of different grades of sandpaper were attached to the sliding block's face. Figure 3 shows the fingertip being held against the sliding block by two compression springs; The normal force applied to the fingertip and sliding block is varied by turning two nuts which compress the two springs (spring constant 0.61 N/mm).

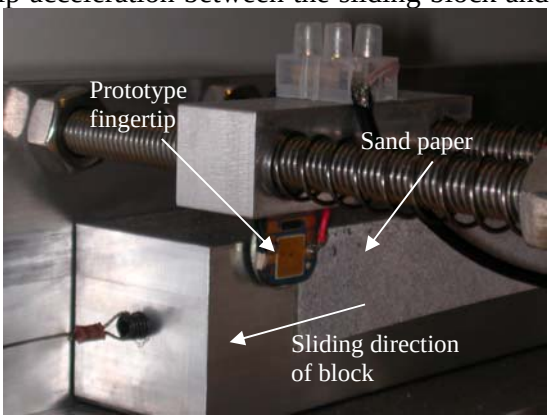

Figure 3: Slip test rig.

Method of operation

Figure 4: Signal acquisition method.

Figure 4 shows the data acquisition technique used. To capture the data from the piezoelectric slip sensor the signal is initially amplified with a charge amplifier. This is because although the piezoelectric thick-film produces a charge directly proportional to the applied force, the latter is typically a small amplitude vibration and the sensitivity of the thick-film is only of the order of a few pC/N. The signal is then sent through an anti-aliasing filter, which cuts out any frequencies above half of the sampling rate of the data acquisition system and allows an accurate digital picture of the signal to be generated [9]. To convert the signals from an analogue to a digital, signal a National Instruments 6036E data acquisition card was used. This is a 16-bit card with a maximum sampling rate of 200kS/s. To allow the collection and manipulation of signals a control program was written using Labview™ 7.1.

RESULTS

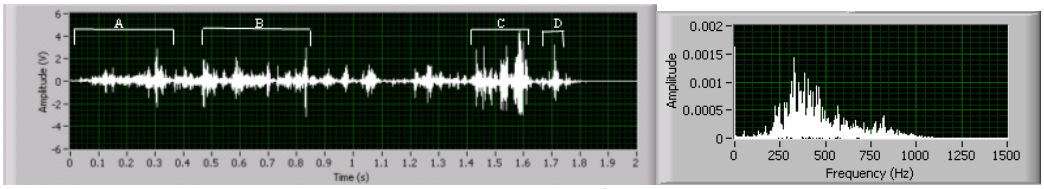

Figure 5: Sliding block acceleration of 0.173 m/s^2

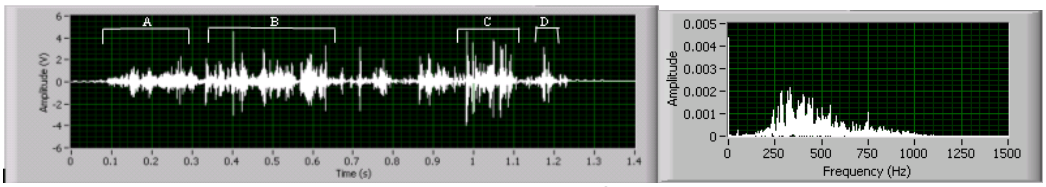

Figure 6: Sliding block acceleration of 0.389 m/s^2.

DISSCUSION

Figure 5 and Figure 6 show the slip signals obtained from the sensor when the fingertip is applying approximately 1.2N of force to the sliding block with the graphs on the left illustrating the slip signal and the graphs on the right being the linear power spectrum FFT of the signal. From these two graphs it can be seen that the signals produced from a single measurement have some very similar characteristics (A, B, C and D). These characteristics relate to the surface profile of the block, with the faster acceleration producing a larger signal amplitude as indicated in the power spectrum FFT. This larger signal output is expected due to the nature of the piezoelectric dynamic force sensor since a faster change in force produces a larger charge output.

CONCLUSIONS

The initial test results of this novel fingertip mounted thick-film slip sensor have shown its ability to produce a signal related to the surface profile of the object slipping past the fingertip at different accelerations, with faster accelerations producing larger output signals. The results show a useful frequency content of the slip signal for this particular material combination over the approximate range of 200-1000 Hz.

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ELECTROMECHANICAL ANALYSIS OF A COMPLETE ARM PROSTHESIS (EMAS)

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Abstract: A characterisation of a complete arm prosthesis is necessary to develop effective control. This is a description of the use of Lagrange methodology to describe the system and to optimise for motion control.

The Lagrange equations of motion are derived from the Newtonian equations of motion. Lagrange analysis describes the system in terms of Kinetic (T) and Potential energies (V). The Kinetic energy (T) is found through a generalised co-ordinate system, where T is a function of the co-ordinates and time derivates. In the non-conservative prosthetic arm, potential energy (V) is found from the generalised forces. These descriptions encompass both electrical and mechanical energies, which are then used to provide the optimum control settings.

This analysis method allows multiple terminal analysis points to be combined, allowing an electrical network with losses, and a mechanical network with losses, combined by a coupling network. Thus the analysis allows for *n* mechanical and electrical terminals in the network. This network approach lends itself to a complete prosthetic arms system, where terminals in the network can range from individual fingers to shoulder joints.

Introduction:

An alternative to the Newton-Euler formulation is the Lagrangian formulation. Whereas the Newton-Euler formulation might be said to be a "force balance" approach to dynamics, the Lagrangian formulation is an "energy-based" approach to dynamics [1].

Generalised Co-ordinates:

The generalised co-ordinates $q_1 ... q_n$, completely locates the dynamic system. Let T and V be the total kinetic energy and potential energy stored in the dynamic system. The Lagrangian L, can be defined by

$$L(q_i, \dot{q}_i) = T - V \tag{1}$$

Since the Potential and Kinetic energies are functions of q_i and $\dot{q}_i, (i = 1,.....,n)$, so therefore is the Lagrangian L.

Mechanical:

The Lagrangian equations of motion of dynamic systems are given by

$$\frac{d}{dt}\frac{\partial L}{\partial q_i} - \frac{\partial L}{\partial q_i} = Q_i \qquad i = 1,...,n \tag{2}$$

Where Q_i is the generalized force corresponding to the generalised co-ordinate q_i. Considering the virtual work done by non-conservative forces acting on the system can identify the generalised force.

The kinetic energy of a link is given by

$$T_i = \frac{1}{2} m_i \vec{v}_{ci}{}^T \vec{v}_{ci} + \frac{1}{2} \vec{\omega}_i{}^T I_i \vec{\omega}_i \tag{3}$$

Where v_i is the velocity vector of the centre of the linkage and ω_i is the angular velocity vector with reference to the base co-ordinate frame. m_i is the mass of the link and I_i is the inertia tensor at the centre expressed in the base co-ordinates.

The total kinetic energy stored in the whole arm linkage is then given by

$$T = \sum_{i=1}^{n} T_i \tag{4}$$

The expression for the kinetic energy is written in terms of the velocity and angular velocity of each link member, which are not independent variables. The above equation can be re-written in terms of an independent and complete set of generalised co-ordinates, specifically joint displacements $q = [q_1, q_2, ..., q_n]^T$.

The Jacobian matrix of nDOF serial manipulator is given by

$$J = \begin{matrix} J_{L1} & J_{L2} \ldots & J_{Ln} \\ J_{A1} & J_{A2} .... & J_{An} \end{matrix} \tag{5}$$

J_{Li} and J_{Ai} are column vectors of the Jacobian matrix respectively associated with the linear and angular velocities. Using vector J_{Li}, the linear velocity of the end-effector can be written as:

$$v_e = J_{L1}\dot{q}_1 + ... + J_{Ln}\dot{q}_n \tag{6}$$

Similarly, the angular velocity of the end-effector was expressed as a linear combination of the column vectors J_{Ai}

$$\omega_e = J_{A1}\dot{q}_1 + ... + J_{An}\dot{q}_n \tag{7}$$

The same method can be applied to v_{ci} and ω_i by regarding the link as an end-effector.

$$v_{ci} = J_{L1}^{(i)}\dot{q}_1 + ... + J_{Li}^{(i)}\dot{q}i = J_L^{(i)}\dot{q} \tag{8}$$

$$\omega_{(i)} = J_{A1}^{(i)}\dot{q}_1 + ... + J_{Ai}^{(i)}\dot{q}_i = J_A^{(i)}\dot{q} \tag{9}$$

The motion of link i depends on only joints 1 through i. Therefore,

$$J_L^{(i)} = [J_{Li}^{(i)} ....J_{Li}^{(i)} 0...0] \quad (10)$$

$$J_A^{(i)} = [J_{A1}^{(i)} ....J_{Ai}^{(i)} 0....0] \quad (11)$$

Substituting expressions for v_{ci} and ω_i into the equation for kinetic energy gives,

$$T = \frac{1}{2}\sum_{i=1}^{n}(m_i \dot{q}^T J_L^{(i)T} J_L^{(i)} \dot{q} + \dot{q}^T J_A^{(i)T} I_i J_A^{(i)} \dot{q}) = \frac{1}{2}\dot{q}^T H \dot{q} \quad (12)$$

H is given as an nXn matrix.

$$H = \sum_{i=l}^{n}(m_i J_L^{(i)T} J_L^{(i)} + J_A^{(i)T} I_i J_A^{(i)}) \quad (13)$$

The matrix H incorporates all the mass properties of the whole arm linkage, as reflected to the joint axes, and is referred to as the manipulator inertia tensor. The manipulator inertia tensor has properties similar to those of individual inertia tensors.

The quadratic form associated with the manipulator inertia tensor represents kinetic energy, and the manipulator inertia tensor is positive definite since kinetic energy is always strictly positive unless the system is at rest.

However, the manipulator inertia tensor involves Jacobian matrices that vary with arm configuration. Hence the manipulator inertia tensor is configuration-dependent. Let H_{ij} be the [i, j] components of the manipulator inertia tensor H, then the total kinetic energy can be rewritten in a scalar form so that

$$T = \frac{1}{2}\sum_{i=1}^{n}\sum_{j=1}^{n} H_{ij}\dot{q}_i\dot{q}_j \quad (14)$$

Where H_{ij} is a function of $q_1,....q_n$.

In addition to the computation of the kinetic energy, the potential energy V and generalized forces are required to be found in order to derive Lagrange's equation of motion.

Let g be the vector representing the acceleration of gravity with reference to the base co-ordinate frame. Then the potential stored in the whole arm linkage is given by

$$V = \sum_{i=1}^{n} m_i \vec{g}^T {}_0^{ci}\vec{r} \quad (15)$$

Where ${}_0^{ci}\vec{r}$ is the position vector of the centre of the link, which is dependent on the arm configuration. Thus the potential function is a function of $q1,...., qn$ [2].

By substituting gravity torque, inertia torques, and coriolis and centrifuge effects into the original Lagrange equations gives the mechanical definition for a complete arm system.

$$\sum_{j=1}^{n} H_{ij}\ddot{q}_j + \sum_{j=1}^{n}\sum_{k=1}^{n} h_{ijk}\dot{q}j\dot{q}_k + G_i = q_i \qquad i = 1,2,....,n \qquad (16)$$

Where:

$$G_i = \sum_{j=1}^{n} m_j g^T J_{Li}^{(j)} \qquad (17) \quad \text{and} \quad h_{ijk} = \frac{\partial H_{ij}}{\partial q_k} - \frac{1}{2}\frac{\partial H_{jk}}{\partial q_i} \qquad (18)$$

Electrical:

The first step in analysing a complicated electromechanical system by a conservation of energy approach is to reduce the system containing electromechanical coupling terms to a minimum. To do this, separate out all purely electrical parts and all purely mechanical parts of the system including losses. This separation procedure is carried out to the extent that each electrical terminal pair is coupled to one energy store, either magnetic or electrical. Any internal interconnections between circuits that are coupled to different energy storages are included in the external electrical network. The mechanical variables represented by the mechanical terminal pairs are those, which affect energy storage in the electric and magnetic fields. The separation procedure results in the general conservative electromechanical coupling network in Figure 1 in which there are n electrical terminals and m mechanical terminals pairs. Each electrical terminal pair will be coupled to either magnetic field energy storage or electric energy field storage. The total stored energy W in the coupling network is

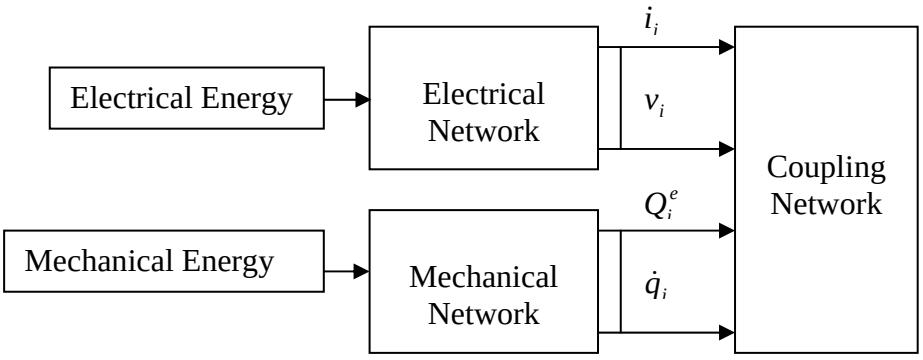

Figure 1. Electromechanical System with Coupling Network

Represented by $W^{el} = W^{\bar{q}} + W^{\psi}$ (19)

Where $W^{\bar{q}}$ is the energy stored in electric fields and W^{Ψ} is energy stored in magnetic fields. It is assumed that W is an instantaneous configuration of the system. Consider an electrical terminal pair coupled to the electrical field storage. When the $\bar{q}_i$ and q_i are specified independently, the current in the ith terminal is $i_i = d\bar{q}_i / dt$ and the voltage v_i at the ith terminal is given by the internal constraints. Next, consider an

electrical terminal pair that is coupled to magnetic field storage. When the ψ_i and x_i are specified independently, the voltage in the ith terminal is $v_i = d\psi_i / dt$ and the current i_i at the ith terminal is given by the internal constraints. It should be mentioned that instead of specifying the $\overline{q}_i$ and ψ_i the voltages v_i and the currents i_i could have been considered as independent.

The next problem is to find the generalized force due to the electromechanical coupling. Since the m mechanical terminal pairs are characterised by independent variables, it is possible to consider each mechanical terminal pair individually to find the force. Let us define the generalized force Q_k^e as the force applied to the kth mechanical co-ordinate by the coupling network. Q_k^e can be found by considering that an arbitrary placement dq_k of the kth mechanical co-ordinate during the time dt takes place. All other mechanical co-ordinates are fixed and the electrical variables may change in accordance to the internal constraints due to the electrical network. This means that only one electrical variable at each electrical terminal can be changed arbitrarily. All losses are said to be either part of the mechanical network or the electrical network.

Step by Step implementation and Conclusions:

Mechanical Network:

Select a suitable co-ordinate system to represent the mechanical configuration if the system. Obtain the Kinetic energy (T) Obtain the potential energy (V), as a function of the co-ordinates, if a conservative system. If the system is non-conservative, find the generalised forces in the form of Q_i .

Electrical Network:

Use the currents to form the generalised co-ordinates. Obtain the total electric energy co-efficient, as a function of the mechanical and electrical co-ordinates. Calculate the power quantities. Define the extended Lagrangian.

$$L^{el} = T + W^{el} - V \tag{20}$$

The Lagrangian method of modelling allows for a complete analysis method in terms of energy. Understanding the energies and losses of a system allows for a more complete implementation for control system design. Lagrange offers one method for modelling a system such as a prosthetic arm. By combining the electrical and mechanical segments for analysis, allows for a more in-depth understanding of a prosthetic arm, than has previously been the case.

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NEW TECHNOLOGY FOR THE SUSPENSION OF TRANS-HUMERAL PROSTHESES – SISA (SUBFASCIAL IMPLANT SUPPORTED ATTACHMENT)

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INTRODUCTION

Trans-humeral amputees have for a long time had suspension methods that limit the functionality and use of the prosthesis. Due to the cone shape of the amputated stump, suspension of the prosthesis has mainly been done with harness securing the prosthesis to the body. The harness crosses the back and goes around the axilla of the contralateral shoulder. This may cause back and neck problems together with pain in the contralateral axilla.

The range of movement for positioning the prosthesis is also limited with the harnessed prosthesis. For the trans-humeral amputee, positioning the artificial arm is crucial to obtain the benefits the prosthesis can give. Since the elbow joint is absent, the only way of positioning the prosthesis is by using the shoulder movements. If the prosthesis and its suspension limit the effective range of movement, the functionality of the prosthesis will be reduced. The prosthesis will in most cases hardly respond to internal - or external rotation of humerus due to the circular shape of the stump and no condyles at the distal end to effectuate the movement.

By trying to recreate the lost condyles of the amputated humerus, we aimed to overcome the suspension problems, including improving the range of movement and control of the prosthesis.

The aim of this study was to create artificial humeral condyles, which would allow the CPO to fit a prosthesis that did not interfere with the shoulder movement on the amputated side. This should increase the functionality of the prosthesis due to increased positioning possibilities, while the pain and problems of the back and neck would be reduced.

Initially, three patients took part in this pilot study, where a titanium implant shaped like a T was surgically cemented into the humerus. At present time, a total of six patients have had their humerus modified in this way.

METHOD

All three patients in the pilot study used harness suspended prostheses, controlled their prostheses myoelectrically, two of them with an electric elbow joint. In common, they all had problems with pain from the neck and discomfort in the axilla of the contralateral shoulder.

Patient 1

Male, age 67 years. Amputated right arm at the distal half of the humerus due to trauma 15 years prior to the study. This patient had earlier tried a prosthesis with a silicon liner without success. Just before the surgery, he was using a hybrid prosthesis with a mechanical elbow joint and a myoelectrically controlled hand.

The titanium implant this patient received had an intercondylar diameter of 55 mm and his amputation stump was elongated by 5mm.

Patient 2

Male, age 58 years. Amputated left arm proximal to the condyles due to a trauma 33 years prior to the study. The prosthesis this patient used was a myoelectrically controlled prosthesis with a Utah elbow. It was harness suspended and pulled on by a stockinette. This allowed, together with the length of the stump, a fairly good range of motion with the prosthesis on.

This patient’s stump was shortened by 40mm and had an intercondylar width of 60mm.

Patient 3

Male, age 22 years. Left arm amputated through the proximal half of the humerus due to a trauma 3 years prior to the study. He used a prosthesis where the hand- and the elbow functions were controlled myoelectrically. By using a push button on the outside of the prosthesis, pro- and supination was also controlled with the electrodes.

Due to the short length of patient 3’s stump, this was elongated by 12,5 mm. The intercondylar diameter of the titanium implant was 65mm.

After a presurgical examination with a CT scan of the patient’s amputated stump, a titanium implant was made with an intercondylar diameter fitting the patient’s humerus. The diameter of the stem was also decided by the diameter of the medullar cavity of the humerus. This diameter should allow a cement mantel of no less than 3mm on each side. The titanium implant was cemented into the medullar cavity of the humerus. The skin surrounded the titanium implant and created a closed environment for the new condyles.

Early prosthetic fitting was made after 3-4 weeks, and different types of sockets were made. Though the oedema was still present, different casting techniques were tried out to control the new ”condyles“ within the socket. The socket shape should also not interfere with the shoulder movement, and no harness should be needed. It was important that the socket fit would withstand torsional forces without the condyles ”slipping” out of their position. The weight of the prosthesis acting as a tractional force should also be acceptable to the patient without creating pain or discomfort.

RESULTS

The check sockets were initially made as a laminated socket in Siegelhartz; where we made a posterior lid to allow for donning and doffing. A cast was taken where the CPO used the first and second finger to maintain a superior pressure on the condyles, similar to casting techniques used in transradial prostheses.

The patient put the check socket on by first extending the shoulder and putting the stump into the socket. Then by flexion of the shoulder, the new condyles were placed into the anterior part of the socket. The lid was closed thereafter by using Velcro and the socket was supposed to stay firmly on the patient's stump. A second Velcro secured the check socket at the proximal end, close to the axilla. After testing the socket fit by pulling and twisting the socket on to the patient’s stump, we soon realised that this technique did not give sufficient control and support needed for a good prosthetic fit. At

high torsional force, the stump had a tendency of slipping out of its position on the lateral condyle, which caused pain and discomfort.

We thereafter discussed how to control the condyles more efficiently, and came up with a solution where ”doughnuts” were made in hard polyurethane to fit around the condyles. These ”doughnuts” were mounted in a jig to maintain the control of the condyles throughout the casting session. The jig stayed on the patient during casting, and was only removed from the plaster negative after it was filled with plaster of paris.

The check socket was laminated in Siegelhartz and for donning and doffing purposes the lid was now moved to the lateral condyle, where it was slid on without any hinges. It was again secured with a Velcro strap over the lid.

This procedure seemed to work better when the check socket was tried on the patient. Positioning the condyles correctly into the socket was easier, and they maintained their position. Displacement of the condyles within the prosthesis was initially not present. After attaching the hand, forearm and elbow to the socket, the patient was able to position the prosthesis in a far better way than he could with his old prosthesis prior to the surgery. (See table 1)

	Patient 1		Patient 2		Patient 3	
	Preop	1 year po	Preop	1 year po	Preop	1 year po
Shoulder abduction	90°	150°	100°	130°	65°	130°
Shoulder external rotation	0°	50°	0°	45°	0°	15°
Shoulder internal rotation	0°	60°	45°	45°	0°	20°
Shoulder flexion	125°	155°	120°	135°	70°	80°
Shoulder extension	30°	60°	60°	45°	25°	30°
Phantom pain (VAS)	4,6	4,6	4,8	4,8	0,2	0,2
Neck/shoulder pain (VAS)	4,5	1,0	8,8	0,2	8,6	1,0

Table 1: VAS (Visual Analogue Scale): Worst imaginable: 10; Best imaginable: 0

However, after wearing the prosthesis for a few hours, the shape of the stump changed due to compression of the soft tissue. We therefore had to make adjustment possibilities in the M-L direction, so the pressure could be adjusted to maintain a good fit.

The socket was made like a framework of metal and the customized ”doughnuts” were mounted into the metal frame, where the lateral lid was made adjustable. This allowed the patient to adjust the M-L pressure according to the stump condition and a snug fit around the condyles was maintained to secure the control of the prosthesis.

To minimise skin pressure further development of the socket was done to increase the surface of contact. Shape and material of the pads that encompass the implant changed from doughnut-like pads made of Pedilin to tear drop shaped pads out of silicone.

After using the new SISA prosthesis, all three patients reported disappearance of neck and shoulder pain.

DISCUSSION

The basic idea of this project was to create an amputation stump similar to that in elbow disarticulation amputees, where the humeral condyles allow a secure fit of the arm prosthesis, without any harness.

All patients still use their SISA-system and there was no loosening of an implant until now. The first three patients had been fitted with different intercondylar widths of the implants and this showed to affect the fitting capability of the prosthetic socket. Patient 1 had an intercondylar width of 55 mm, and due to the shape of the humerus, the lateral condyle became small.

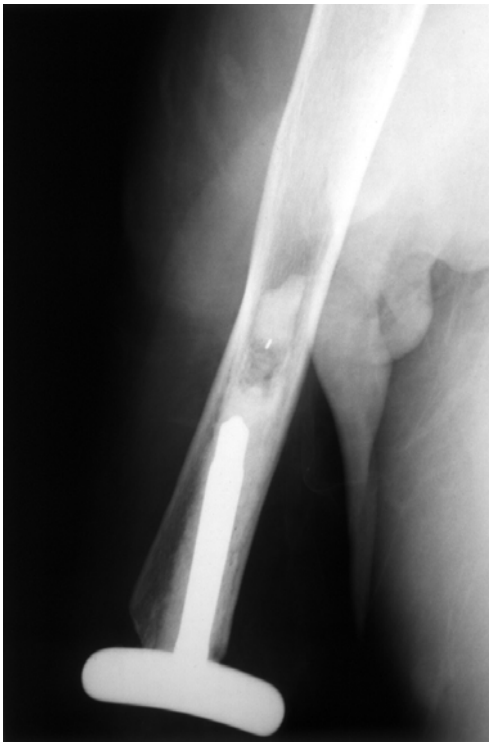

Fig. 1: X-ray of SISA-implant

This gave us a very small weight bearing area on the lateral side, and the prosthetic socket had a tendency to slip over the condyle when torsional forces were added.

On patient 2 and 3, the width was increased and control of the prosthetic socket was made easier. However, there is still a high load that should be carried on the condyles and this could easily cause skin damage and pressure sores. The width and size of the artificial condyles are of great importance to the quality of the prosthetic fit. The soft tissue coverage over the condyles also affects the control of the socket. Patient 2 had a very bony stump with little soft tissue coverage. Here it was very easy to control the stump in the socket, but due to shortness of soft tissue coverage, the pressure area became intolerable, and had to be increased to the maximum.

When the patients put their stump into the socket, it is crucial to have the correct position within the socket. Pressure marks and discomfort will arrive if this is not done correctly. Due to the continuous pressure on the skin around the condyles, it is important to have sufficient space on the tip of the condyle. This will allow the skin be more free and prevent excessive stretching of the skin.

A second generation of the titanium implant has been made, where the shape has been altered. Three patients have been fitted with this new implant, and rehabilitation time for prosthetic fitting has decreased compared to the patients with the first generation of the implant. The weight bearing area has been increased and the angulation of the condyles has also been altered. This made it easier for the CPO to make a prosthetic socket that was more comfortable for the patient.

CONCLUSION

This study demonstrates that by means of non-percutaneous implants the function and control of the exoprosthesis can be significantly improved. In total, we managed to make an amputation stump that could support the weight of the trans-humeral prosthesis without any harness. Pain in the neck and contralateral shoulder area was eliminated and this was of great importance to the patients. When socket shape and fit were optimized, the patients had an increased control of the prostheses and positioning for function became easy.

The shape and width of the artificial titanium implant is of great importance to maintain a good prosthetic fit. The second generation of the implant, proved easier fitting of the prosthetic socket. Minimization of the stress of the skin in the area of the implants is of great importance to the quality of the fitting. Further development of this system will be done in collaboration with Otto Bock Healthcare focusing on an improvement of the implant to reduce the skin pressure by increasing the stressed area and decreasing the required pretension. The fixation of the implant shall be as well non-cemented so that younger patients can also benefit from this new technology. A modular design and a two stage surgery will make the procedure easier to handle. To increase the wearing comfort of the system the load transmission of the prosthetic socket will be optimized. All this development will be scientifically proven by means of FEM-calculation, extensive measurements and a clinical trial.

Implantable Myoelectric Sensors (IMES)

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ABSTRACT: We are developing a multi-channel/multifunction prosthetic hand/arm controller system capable of receiving and processing signals from up to sixteen Implanted MyoElectric Sensors (IMES). The appeal of implanted sensors for myoelectric control is that EMG signals can be measured at their source providing relatively cross-talk free signals that can be treated as independent control sites. Therefore the number of degrees-of-freedom that can be simultaneously controlled and coordinated in an externally-powered prosthesis will be greater than with surface EMG or mechanical control sites. To explore the issue of intra-muscular signal independence and the ability to control them, human subject experiments have been performed in which intra-muscular EMGs were obtained. Choice of muscles was based on a desire to be able to independently control a two degree-of-freedom (DOF) wrist, and 3 DOF prosthetic hand. This paper provide our result so far.

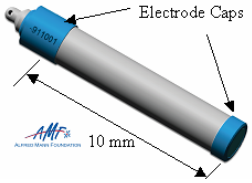

Fig 1: Alfred E. Mann Foundation RFB BION® Package. The inter-electrode spacing is appropriate for whole muscle activity recordings.

INTRODUCTION: The limitation of current prostheses is not the devices themselves but rather the lack of sufficient independent control sources. A system capable of reading intra muscular EMG signals would greatly increase the number control sources available for prosthesis control. We are developing a chronically implantable sensor system to create multiple control sites to detect commanded movements. We are developing myoelectric sensor capsules **(Fig. 1)** that can be chronically implanted into the residual muscles of an amputee's arm. By localizing the points at which myoelectric activity is detected, these points can be treated as independent control sites with minimal cross-talk. Consequently, the number of degrees-of-freedom that can be simultaneously controlled and coordinated in an externally powered prosthesis will be greatly increased in comparison with surface EMG sites, while obviating the problems of tapping into cut motor control nerves. Our Implantable MyoElectric Sensor (IMES) system[1] will be capable of reading EMG signals from up to 16 inductively coupled, implanted bipolar differential electromyographic (EMG) sensors. These sensors receive their power, digital addressing, and command signals from an external transmitter/receiver coil worn by the patient. The external coil required for the inductive link is laminated into the prosthetic socket such that this coil will encircle the implanted electrodes (**Fig. 2**). Each implanted sensor acts as an intramuscular electrode to detect the electrical activity generated as a by-product of normal muscle contraction. The implants transmit these muscle signals, or myoelectric (EMG) signals, over a shared transcutaneous magnetic link. Each sensor's electronics and associated circuitry will be housed in a *previously developed* RFB BION® hermetically sealed package provided by the Alfred Mann Foundation

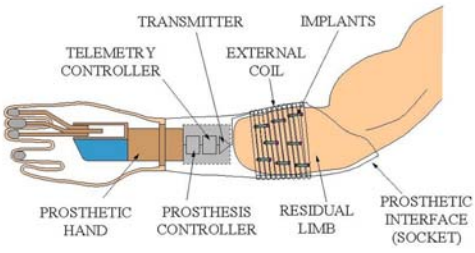

Fig 2: Schematic of planned Implanted Myoelectric Sensor (IMES) system. The external coil is laminated directly into the prosthetic interface. Signals from the implants in the arm, linked through the external coil, control the prosthesis via reverse telemetry. Implant power is supplied through the

(AEMF)[1]. *A major attraction of the BION® technology is that the hermetically sealed ceramic capsule and electrodes necessary for long-term survival in the body have previously been granted FDA IDE approval for use in Functional Electrical Stimulation applications.* Furthermore, no wires are required to be surgically threaded down the arm. No wires are required to penetrate the skin. implant and telemetry design. An external prosthesis controller will decipher user intent from telemetry sent over a transcutaneous magnetic link by the implanted electrodes. The same link will provide power for the implanted electrodes.

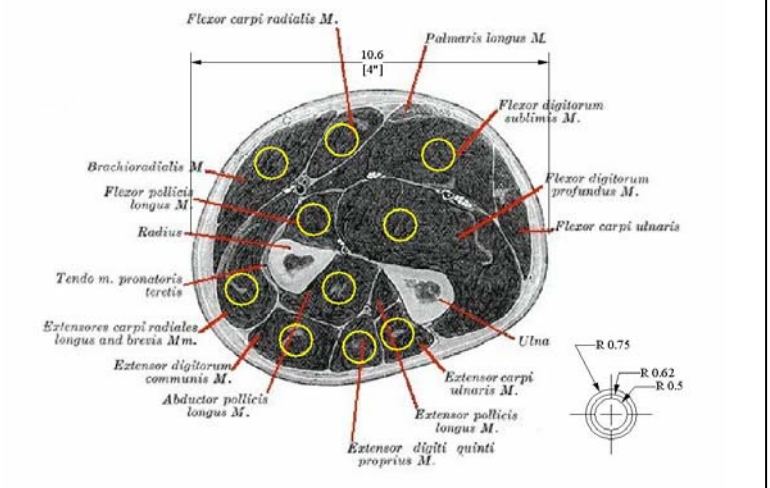

Fig. 3: Projected pick-up area for our Implanted Myoelectric Sensors (IMES) superimposed on an appropriately scaled section of through the proximal forearm. Image shows that pick-up area should not be an issue in the final device with each device having a pick-up area confined to an individual muscle.

We have simulated the potential pickup area for our implants, using electromagnetic finite element modeling techniques[2]. The simulation results suggest that the presence of a thin layer of encapsulation tissue around the IMES should not impede the detection of EMG signals from the surrounding muscle fibers and may in fact cause the amplitude of the EMG signal to increase modestly. The orientation of the bipolar electrode with respect to the fiber direction is an important factor in determining the selectivity of the implanted electrode. Alignment of the electrode along the fiber direction will be particularly important in smaller muscles. We found that for an implant placed along the fibers of the muscle in which it is inserted the pickup area for the sensor will be a cylinder about 5mm in radius about the implant (**See Fig. 3**).

CONTROLLER ALGORITHM DEVELOPMENT

We have conducted extensive human subject experiments in an effort to elucidate the best method of control to use to integrate the contributions from the 16 different implanted sensors. The multi-channel/multifunction prosthetic hand/arm controller must be capable of receiving and processing signals from the implant telemetry system. This same controller must then decipher user intent from the telemetry sent by the IMES to decide which actuators in the prosthesis to drive. From a control perspective there are a number of levels of sophistication that a prosthesis controller can implement.

The simplest control paradigm is to use one muscle to control one function in the prosthesis [*one muscle – one function]*. The implicit assumption underlying this approach is that a contracted muscle EMG signal power is much greater than the relaxed muscles EMG signal powers i.e. high SNR. This system will work with an identifiable signal controlling a single function. For example, using a biceps EMG for elbow flexion and a triceps EMG for elbow extension works quite well. When multiple discrete signals can be identified that directly relate to a given arm function, simultaneous control of multiple degrees of freedom can be obtained resulting in very natural, easy control for the amputee. From a controller design standpoint this is the simplest form of control to implement, and our initial experiments were targeted towards demonstrating this type of control. We anticipate this type of

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control will be possible for the shoulder and the elbow. This type of control is problematic for the wrist and hand because multiple muscles control the same functions—the distal arm is an indeterminate system. Even with all of the control signals available for the arm (such as recording EMG from every muscle in an able bodied person) there is not an exact solution defining movement and different people use their muscles in different patterns to achieve the same function. To use this type of control, without an internal model of the hand, the user needs to relearn how they use their muscles and to think/remember which muscle is tied to which function. This can cause excessive mental loading on the part of the user and yield less-intuitive control.

A more sophisticated approach is to recognize patterns/features of EMG activity associated with different training movements and/or functions and have the controller drive the appropriate prosthesis actuators *[i.e. one pattern of EMG activity - one function]*. The control is intuitive and easy to remember since users execute the movement the prosthetic limb is to perform with their "phantom limb." Recognition is achieved through feature extraction algorithms, artificial neural nets, or some other similar high level classification method. We have been recording the patterns of EMG activity associated with different training movements and/or functions (**Fig. 4**) and then training the controller to recognize these patterns. We have explored recognition through automated classification techniques using neural networks, linear discriminant anslysis, fuzzy clustering[3], and multivariate linear regression. These systems "learn the patterns produced by each user and are optimized to map the EMG activity from multiple channels to the desired limb motion. This approach requires a pattern to be stored for **every** desired movement. *One EMG pattern - one function* does not provide true parallel/simultaneous control of multiple degrees-of-freedom but could enable seamless-sequential control to be executed. The problem with these approaches is that they are still task-based control systems. The user is still compelled to think in terms of what grip they need to do rather than subconsciously reaching out and picking up an object. We have yet to draw any definitive conclusion as to which classification technique we prefer, but the fuzzy clustering method does lend itself to a default "safe" [fail safe] approach when considering driving motors that are inherently non-backdrivable (as are commonly used in prosthetics to preserve power and to hold commanded position and force in the absence of power).

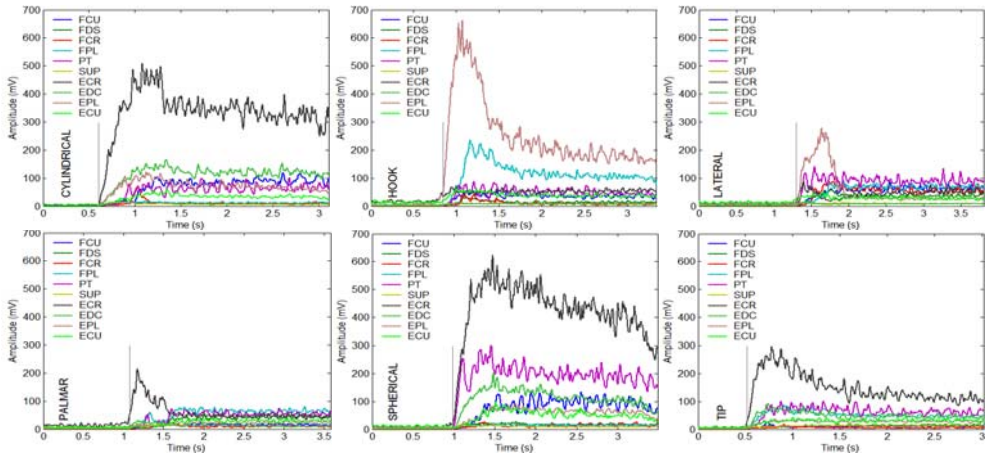

Fig. 4: We have been performing fine wire intramuscular EMG recordings from up to 10 muscles simultaneously in the forearm of human subjects while these subjects perform various grasp patterns and joint functions. The traces above show a typical set of recordings for one subject. Notice how the pattern of EMG activation is distinct for each grasp type.

SYSTEM ARCHITECTURE DEVELOPMENT:

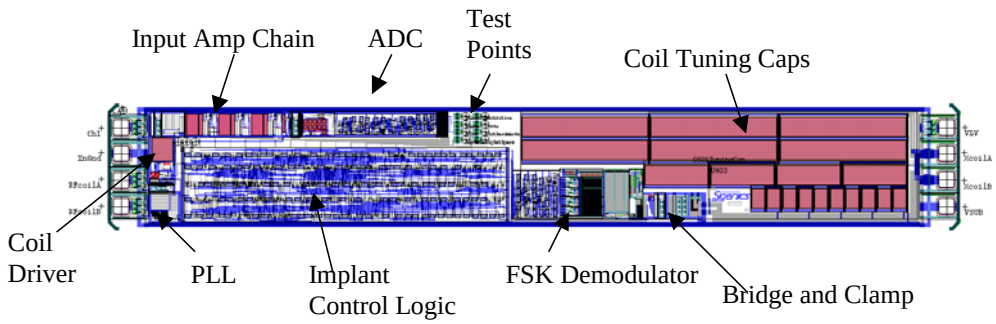

Fig. 5: The Current Implant IC.

A large portion of our efforts have been dedicated to the development of an overall system architecture that maximizes the number of implant devices, and the telemetry bandwidth of those implants, that can be serviced by a single external controller. The Texas Instruments MSP430 family of low-power microprocessors has been selected to perform the high-level Telemetry controller functionality, such as power and implant control as well as outward data decommutation. Most of the higher-level programming will be done in C, with lower-level data streaming routines programmed in assembler. A communications protocol and command set for the telemetry controller has been defined. The prosthesis controller decides which implants require monitoring, and sets up the telemetry controller using a command language to tell the telemetry controller how to configure a serial stream of telemetry output to prosthesis controller input data.

Development of the system is well under way. We are assembling new Class-E exciter modules to test the new implants[4]. PC boards are in hand waiting for the new Class-E Controller chip. We will continue the magnetics design to verify reception of outward telemetry. Next we need to Release the official Interface Control Document (ICD), which completely specifies the interface between the telemetry controller (MSP430) and the prosthesis controller. So that the prosthesis controller can take the data sent from the implant system and use it to control the prosthesis. To date, we have submitted three fabrication runs on the XFab CX08 process. Devices from the first and second runs are being tested. We have chips back from the third run and these are currently in test. These chips are in their near final form factor to fit in the AEMF capsules (**Fig. 5**), however, initial tests have revealed some non-insurmountable defects that are being corrected in a number of up coming wafer runs. The near future goal for the hardware development is an end-to-end demonstration of the system.

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SURFACE VS. IMPLANTED EMG FOR MULTIFUNCTIONAL PROSTHESIS CONTROL: PILOT RESULTS

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INTRODUCTION

It has been hypothesized that, due to the potential to both provide a larger number of independent control sites and selectively record from forearm muscles (in particular the deep muscles), intramuscular EMG should be advantageous for multifunctional prosthesis control [1].

The use of surface electromyograms (EMG) to control a multiple degree-of-freedom prosthesis has been investigated for several decades. A variety of approaches have been employed with groups using different numbers of input channels [2-3], feature extraction methods [3-6] and pattern recognition algorithms [3,7-8]. While much work has been done, all of these efforts have used surface EMG as the control signal. Only a single preliminary study was found that acquired intramuscular EMG for prosthesis control [9].

Admittedly, the technology has not existed for chronic intramuscular recordings to be clinically feasible for prosthetic use. The Implantable Myoelectric Sensor (IMES) that is being developed at the Northwestern University Prosthetic Research Laboratory will make chronic intramuscular recordings clinically feasible [10].

We hypothesize and hope to demonstrate that by utilizing intramuscular EMG it will be possible to substantially increase classification accuracies of multifunctional prosthesis controllers (i.e., increase the percentage of the time that the controller can correctly predict the intended movement of the user). If a substantial increase in classification accuracy is demonstrated, this will justify the invasiveness of using these devices. However, if similar accuracies can be obtained from surface recordings then there will be little justification for pursuing these devices for transradial prosthesis control purposes.

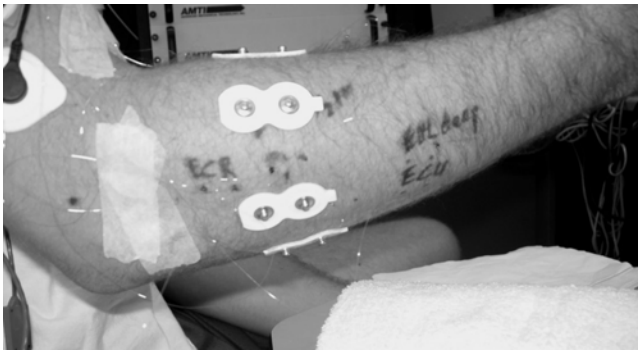

Figure 1 - A photograph of the location of the surface electrodes and fine wire insertion sites. The writing on the forearm indicates preliminary markings to assist in locating the fine-wire sites.

METHODS

Protocol

Both surface and intramuscular EMG data were collected from the forearm of four subjects. Six surface electrodes were placed in

an equally spaced array around the circumference of the forearm (Fig. 1). Additionally, ten pairs of fine wire bipolar EMGs were recorded from 10 muscles of the forearm: extensor carpi radialis (ECR), extensor carpi ulnaris (ECU), extensor digitorum communis (EDC), extensor pollicis longus (EPL), flexor carpi radialis (FCR), flexor carpi ulnaris (FCU), flexor digitorum superficialis (FDS), flexor pollicis longus (FPL), pronator teres (PRON), and supinator (SUP). Verification of the location of the intramuscular electrodes was accomplished by instructing the user to perform a test movement that was indicated by a standard electromyography text [11]. Additionally, the test movements of the neighboring muscles were performed to ensure that EMG was being collected from the correct muscle. The intramuscular electrodes were separated by approximately 13 mm to mimic the recordings we would expect from the IMES sensor.

EMG was collected as the subjects performed a series of contractions corresponding to the six movements that can be produced using commercially available prosthetic components: hand close, hand open, pronation, supination, wrist extension, and wrist flexion (figure 2). For each trial the subject would produce four five-second contractions of the same movement, each time starting from and returning to rest. Four of these trials were collected for each movement with two trials used as training data for the pattern recognition system and the other two used to determine classification accuracies.

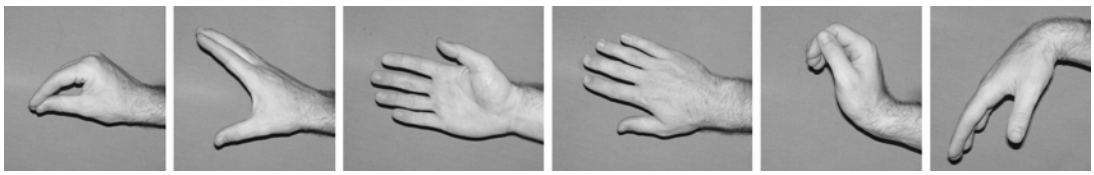

Figure 2 - Photographs of the six movement classes used in the pilot experiments: Hand Close, Hand Open, Pronation, Supination, Wrist Flexionand Wrist Extension.

Analysis

It has been demonstrated that extracting signal features in addition to EMG amplitude from the recorded signals substantially increases the classification accuracy of prosthesis controllers [3-6,8,12-13]. Therefore, an autoregressive (AR) model was created and the root-mean-square (RMS) of each channel was calculated for each 50 ms bin of data. AR parameters have been used repeatedly in previous research [6,12-13] and have been found by the group at the University of New Brunswick to outperform other signal features for myoelectric control. In this study we used a 3^{rd} order model.

Six sets of input data were created for analysis:

1. 6 surface channels: RMS only
2. 6 surface channels: RMS + 3^{rd} order AR
3. 10 intramuscular channels: RMS only
4. 10 intramuscular channels: RMS + 3^{rd} order AR
5. 6 intramuscular channels: RMS only
6. 6 intramuscular channels: RMS + 3^{rd} order AR

Note: The subsets of six intramuscular channels used in input sets #5 and #6 were selected using multinomial logistic regression.

The classification accuracy was determined by comparing the intended movement of the user to the output of a linear discriminant analysis (LDA) classifier.

RESULTS

The classification accuracies that resulted from the six input sets described above are shown in Figure 3. When utilizing only EMG amplitude the surface data produced the largest amount of error of any data set (21.1%). Adding AR coefficients to the input data set decreased the classification error by more than half (10.3%).

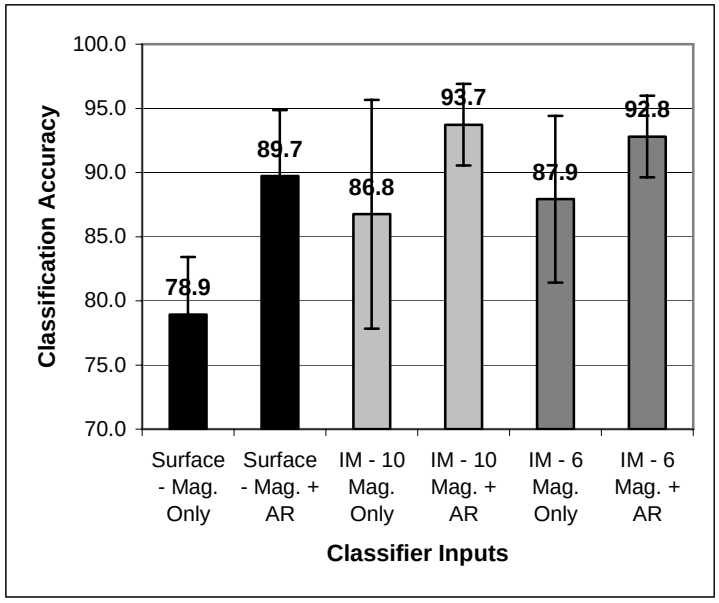

Figure 3 – Classification accuracies resulting from surface and intramuscular EMG. The surface recordings are shown in black, the full set of 10 intramuscular recordings are shown in light gray and a subset of six of the ten intramuscular channels chosen via multinomial logistic regression are shown in dark gray. The left column of each pair shows classification accuracies when only EMG amplitude is used and the right column show classification accuracies from using autoregressive coefficients in addition to EMG amplitude.

When compared to the surface with AR data set the intramuscular data had slightly larger error rates when using only signal amplitude with all 10 channels (13.3%) and 6 channels (12.1%). It was also demonstrated that the use of the AR parameters again reduced the error substantially when applied to the intramuscular input sets of 10 (6.3%) and 6 (7.2%) channels.

DISCUSSION

The increased accuracy that is seen by implanting the electrodes is encouraging. However, these results are based on a comparison of non-targeted surface channels to targeted intramuscular channels and to achieve a truly fair comparison of surface vs. intramuscular EMG we feel we need to target both recordings. In the near future we will be able to compare targeted surface recordings with the targeted intramuscular data.

It was also interesting to note the considerable improvement that is achieved by adding the auto-regressive parameters to each input set. The classifier error was reduced by more than half in two instances with the error being decreased by 51.7% for the surface inputs and 52.8% for the 10 channel intramuscular inputs.

The final observation is the ability of a smaller number of intramuscular channels to

perform as well as the 10-channel set for this set of tasks. This indicates that it is only necessary to record from a subset of forearm muscles to maximize classification accuracy for this six-class problem. The muscles that were contained in these six-channel subsets were not consistent for each subject however when a fixed subset of six channels was used for all subjects it performed comparably (91.0 %) to the sets that were customized for each subject (92.8%).

FUTURE WORK

We are in the process of performing additional experiments in which surface and intramuscular EMG are collected from both targeted and untargeted sites on the forearm. We also plan to investigate the use of multinomial logistic regression for pattern recognition purposes as well as to increase the number of classifier output classes to make the classification problem more difficult.

ACKNOWLEDGEMENT

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A Comparison of Surface and Internally Measured Myoelectric Signals for use in Prosthetic Control

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INTRODUCTION

The surface myoelectric signal (MES) has proven to be an effective control input for powered prostheses. Pattern recognition based controllers use multi-channel surface MES as inputs to discriminate between the desired classes of limb activation. There are two major methods which may be pursued to increase the accuracy of the controller: 1) use signal processing to extract more information from the input signals; or 2) provide more informative raw signals to the controller. As a result of recent technological advances, it is reasonable to assume that there will soon be implantable myoelectric sensors which will enable the internal MES to be used as inputs to controllers [1]. An internal MES measurement should have less muscular crosstalk allowing for more independent control sites. However, it remains unclear if this benefit outweighs the loss of the more global information contained in the surface MES. This study compares the classification accuracy of several pattern based myoelectric controllers which use information extracted from surface MES to the same controllers which use information extracted from fine-wire intramuscular MES.

PATTERN BASED MYOELECTRIC CONTROL

Pattern recognition based myoelectric control systems operate on the assumption that at a given electrode location, the set of features describing the myoelectric signal over an analysis window will be repeatable for a given state of muscle activation [2]. Furthermore, it is assumed that the features will be different from one state of muscle activation to another. The feature sets are then sent to a classifier which determines the output based on training from previous data. Figure 1 illustrates a block diagram of the pattern recognition based myoelectric control process.

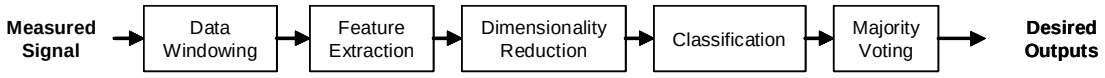

Figure 1: Block Diagram of Pattern Based Myoelectric Control Process

In recent years much effort has been devoted to investigating the effects of feature sets and classifiers on the classification accuracy of MES pattern recognition. Feature sets formulated from time domain (TD) statistics [3,4], autoregressive (AR) coefficients [5], and time-frequency information [6], have been shown to provide accurate signal representation when combined with dimensionality reduction in the form of principal components analysis (PCA) [4]. ANN's [3,7,8], genetic algorithms [4], linear discriminant analysis (LDA) classifiers [6], Gaussian mixture models (GMM) [5], and fuzzy logic [9] have all been shown to be acceptable classifiers for pattern-based myoelectric control. The size of the data analysis window and the number of majority votes are determined by processing speed and the acceptable delay perceived by a prosthetic user and have been shown to affect classification accuracy [4].

METHOD

Experiment I- Simultaneous Collection of Surface and Intramuscular MES

Surface and intramuscular MES were collected simultaneously from six normally limbed male subjects for ten medium force isometric contractions of five seconds duration, corresponding to the motions: wrist flexion\extension, wrist abduction\adduction, forearm pronation\supination, key grip, chuck grip, open hand, and gently move fingers. An arm brace was constructed to hold the arm stationary during the contractions and provide strain relief for the electrode lead wires. The brace supports the arm at the elbow and wrist while the hand is inserted into a padded slot. During contractions, the padded slot provides some resistance for the hand to push against while keeping the forearm immobilized.

A 16 electrode array with an inter-electrode spacing of 2 cm was wrapped around the circumference of the upper forearm to measure the surface MES. No specific placement strategy is required of the surface array as its global pickup provides complete spatial coverage of the circumference of the limb. All surface signals were configured for differential measurements with a gain of 2000 and were sampled at 1024 Hz. For intramuscular measurements, 44 gauge fine-wire electrode pairs were inserted into six control sites in the forearm: pronator/supinator teres, flexor/extensor carpi ulnaris, flexor digitorum sublimas and extensor digitorum communis. The inter-electrode spacing the intramuscular electrode was 1 mm which resulted in very local signal detection. All internal channels were configured with a gain of 2000 and were sampled at 1024 Hz. Software guided each subject through data acquisition sessions which consisted of two trials of two repetitions of each motion (corresponding to 40 contractions in all).

Experiment II – Effect of Electrode Displacement on Surface MES Classification

Surface MES were collected from one normally limbed male subject for the same ten, medium force isometric contractions of five seconds duration listed in Experiment I. The 16 electrode array was wrapped around the upper forearm and the surface signals were configured for differential measurements with a gain of 2000 and sampled at 1024 Hz. Two trials of two repetitions were collected for each of the ten motions. The 15 channel array was reapplied at various locations close to the original position; 1 cm above, 1 cm below, rotated 1 cm right, and rotated 1 cm left. Two more trials were collected at each displacement location to simulate the maximum residual limb/socket misalignment which a user could expect while using a prosthetic limb.

RESULTS

Experiment I- Simultaneous Collection of Surface and Intramuscular MES

Six different feature set/classifier combinations were used to make the comparison between the surface and intramuscular inputs. TD statistics, a 6^{th} order autoregressive (AR) model plus the RMS signal value, and concatenated TD and 6^{th} order AR (TDAR) feature sets were used. The classifiers investigated were a LDA and a multilayer perceptron ANN with 12 hidden layer nodes which was trained using the backpropagation algorithm. For each combination, data analysis windows of 256 ms were used to extract features. A conservative estimate of 32 ms was used as a processing delay which allowed for 17 decisions to be used in a majority vote while keeping the user perceived delay less than 300 ms [4]. The first 40 principal components of the feature sets were used in processing except when the feature vector had a dimension less than 40, in which case all principal components were used. Trial one of the data set was used to train each of the classifiers and trial two was tested to determine the classification accuracy of the

control scheme. Table 1 displays the average classification accuracy and standard deviation across subjects of the six different feature set/classifier combinations.

	Fifteen Channel Surface						Six Channel Intramuscular					
	TD		AR		TDAR		TD		AR		TDAR	
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD
LDA	98.0	2.5	97.2	1.4	98.3	1.3	96.8	2.6	97.1	1.9	97.3	1.8
MLP	97.6	1.2	96.1	2.6	97.1	2.0	95.4	5.5	95.3	3.2	97.4	2.3

Table 1: Average Classification Accuracy

These results suggest that either surface or intramuscular MES inputs yield excellent classification accuracy. Extreme care must be taken to ensure proper localization of intramuscular electrode sites because the measured signal is dominated by the motor units located very close to the electrodes.

The processing time it takes to form a decision is dependent on a number of factors including the number of input channels. Six intramuscular channels would therefore require less processing time compared to the 15 channel surface electrode array. However, the global pickup and closely spaced channels of the surface electrode array results in some adjacent channels containing redundant information.

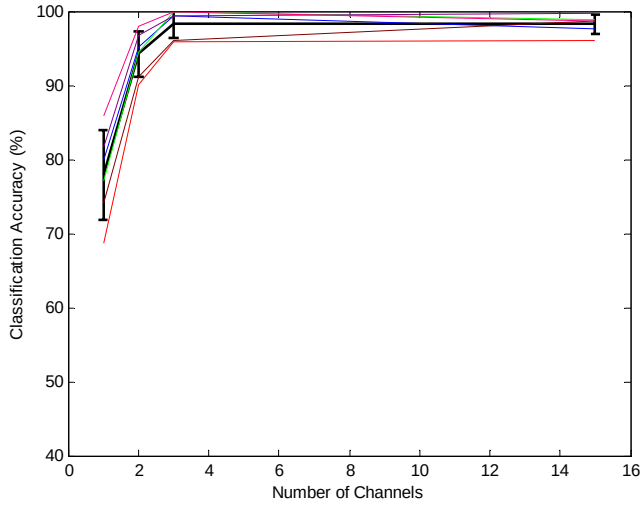

Figure 2: Classification Accuracy of Optimum Channel Subsets

*The thin lines correspond to the classification accuracy at selected channels for each subject. The thicker line with error bars displays the average classification accuracy over the six subjects

A subset of n channels of the original surface data can be chosen while still maintaining high classification accuracy. Figure 2 displays optimum channel subsets of various sizes for the TDAR feature set and the LDA classifier. The optimum channel subset of size n was selected by simply trying all the combinations of channels. The classification of a three channel subset of the surface MES performs as well (app. 99%) as all 15 channels providing the subset is selected properly.

Experiment II – Effect of Electrode Displacement on Surface MES Classification

The TDAR feature set and LDA classifier (256 ms analysis window, 32 ms processing delay and 17 majority votes) was used to investigate the effect of electrode displacement on

classification accuracy. The control scheme was trained with the first trial of the data set collected from the original electrode position and was tested with the second trial of the data set taken from the displacement positions. Table 2 summarizes the classification accuracy of the control system at different displacement locations.

	Original Location	Displacement Location				Average
		1 cm Above	1 cm Below	1 cm Left	1 cm Right	
Classification Accuracy	99.1%	96.2%	89.2%	81.4%	95.1%	92.2%

Table 2: Affect of Electrode Displacement on Classification Accuracy

The control scheme was next trained with a training set consisting of data taken from all locations, and tested with the second trial at each location. Using this training method average classification rate over all the categories increased to 97.5%. This preliminary work suggests that electrode displacement degrades classification accuracy; however, by training the control system to recognize patterns from the displacement locations, this degradation can be mitigated.

CONCLUSIONS

It was shown that both surface and intramuscular MES inputs yield excellent pattern recognition based myoelectric classification accuracy for various combinations of feature sets and classifiers. There was no obvious advantage in using intramuscular MES inputs for pattern based myoelectric control in this particular experiment. The number of input surface channels can be reduced from fifteen to three by carefully selecting a channel subset. This would indicate an optimal electrode placement strategy for the ten motions considered in this study. Displacement of the surface electrodes degrades classification accuracy, however this can be mitigated by training the system to recognize patterns generated under plausible displacement conditions.

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Development of myoelectric controllers for hand prostheses

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1. INTRODUCTION

This paper describes a research project at the National Institute of Advanced Industrial Science and Technology (AIST) to develop a myoelectric controller. The myoelectric controller interprets control intentions from the operator by recognizing myoelectric signals. This kind of controller has typically been applied to control electric-powered prostheses. The most notable advantage of using the myoelectric controller is its capacity to utilize the residual muscular functions of physically-impaired persons. For example, in the case of a hand prosthesis, the myoelectric controller enables the amputee to utilize the residual functions of remnant muscles at their stump.

Within the project, we initially designed a pattern classification LSI (Large Scale Integration) in 1998 [1], and as one central application of the LSI, we have subsequently been developing compact controllers for multi-functional prosthetic-hands. Employing this pattern classification LSI, the controller can adapt itself to the unique characteristics of a myoelectric signal distribution for a given individual user [1].

Moreover, in order to realize hand-prostheses that could become widely accepted, we started developing a basic functional hand prosthesis in 2002. This prosthesis has undergone some clinical evaluations, and the technology has already been transferred to a private company for commercialization.

This paper outlines the development of the multi-function and basic function controller, as well as a basic functional mechanical hand.

2. THE MULTI-FUNCTIONAL CONTROLLER

While single functional hand prosthesis are already accepted by many amputees, it is also true that there have been calls to develop multi-function systems, capable of carrying out more than one function [2]. In response to the demand for greater prosthetic-hand functionality, recently, much research has been conducted on multi-function forearm prosthesis, applying pattern-classification methods, such as neural networks [3,4], in order to determine the hand actions. However, in this application, compact implementation is one of the most important issues, therefore we adopted a logic circuit pattern classifier with the pattern classification LSI.

In the case of myoelectric pattern classification with a logic circuit, it is necessary to quantize myoelectric signals into discrete numbers, which must then be coded as binary bit patterns. An efficient quantization method is, therefore, essential to the realization of high-accuracy myoelectric pattern classification.

Figure 1 shows a typical example of myoelectric pattern distributions and their quantization with linear quantization [5], which is the most basic quantization method. Myoelectric patterns that distribute within the same cell of the grid in Figure 1 are quantized as the same integer numbers. For example, in the worst case of quantizing for the patterns in Figure 1, three different actions--forearm supination (19 patterns), forearm pronation (20 patterns) and hand opening (2 patterns)--would all be are quantized as (0,0). This would result in the distinct

patterns being classified as the same pattern, i.e., as being generated from the same action. This kind of quantization error is an obstacle for high-precision pattern classification.

In order to overcome this quantization problem, we have proposed employing μ-LAW quantization [6,7], where the transformation characteristics can be adapted to the distribution characteristics in terms of a μ-value. The effectiveness of μ-LAW quantization has been confirmed by the pattern classification of myoelectric signals, which were sampled from five subjects, including one experienced person, who has repeatedly participated in our experiments, and four new users joining our experiments for the first time.

By applying the μ-LAW quantization, the pattern classification rate increased by 11.1% (averaged for the five subjects) and by 15.5% (maximum). Furthermore, the classification rate for the experienced subject was 97.8% (averaged over ten trials), demonstrating that skilled individuals are able to operate a multi-functional myoelectric hand with high-accuracy.

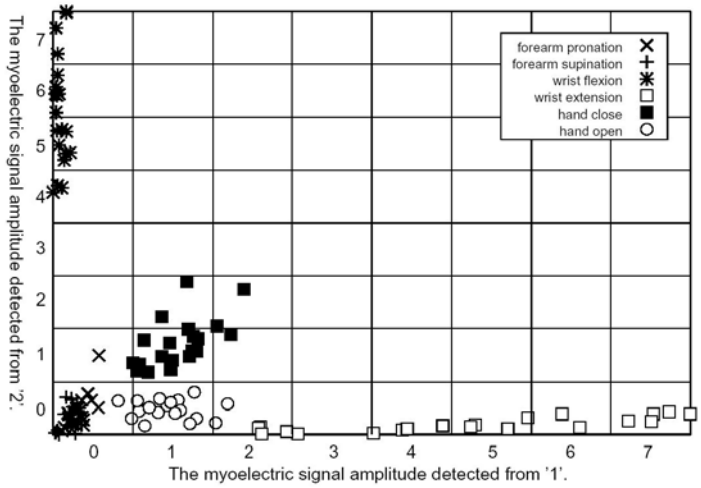

Figure 1 - An example of myoelectric pattern distributions and their quantization with linear-quantization

3. THE BASIC FUNCTIONAL PROSTHESIS

Given the current situation of myoelectric hand prostheses in Japan, where even single function hand-prostheses are not widely accepted, we began developing both of a basic function (less than two functions) controller and a mechanical hand.

3.1. The controller

Control for one or two function systems has already realized in a number of commercialized hand prostheses, such as an *OTTOBOCK* or a *Motion Control* hand. Accordingly, we have adopted the control methods used for such commercial hand prosthesis.

These methods can be divided in two modes. The first is a switch mode, where each myoelectric signal channel has a threshold, and if the signal intensity exceeds the threshold, then the corresponding function, such as hand-open or hand-close, will be activated. Each threshold can be easily adjusted by using a graphical user interface (GUI) on a personal computer (PC).

The second mode is a proportional mode. Although the motor for the activated function rotates at a constant speed in the switch mode, in the proportional mode, the motor rotates at a speed which is proportional to signal intensity.

Hardware specifications of the developed controller are as listed below.

1. Microprocessor: H8/3664F (Renesas technology corporation)
2. Size of the controller board: 6cm*3cm
3. Battery: 7.2V (Lithium-ion)
4. Number of input myoelectric channels: 2
5. Number of controllable motors: 2

In addition to these specifications, the controller can also execute a software program for the logic circuit pattern classifier. This means that the controller handle more than three functions for the prosthetic hand, although the controller has not yet been clinically evaluated at such levels of functionality.

3.2. The mechanical hand

We are developing two basic function mechanical hands, which utilize two different technologies. The first technology is the "leadscrew", which is a fundamental decelerator and converts a rotation motion into a linear motion. Figure 2 shows the mechanical hand with the leadscrew technology. The specifications of this hand are as follows.

1. Length: 141mm (from the finger tip to the wrist)
2. Weight: 290g
3. Number of functions: 2 (hand open-close and wrist flex-extend)
4. Power of finger (three-point) pinch: 2.5kg (measured by *North Coast^{TM} Hydraulic Pinch Gauge NC70141)*

The prosthesis, which incorporates this mechanical hand and the basic function controller, has undergone some clinical evaluations, as shown in Figure 3.

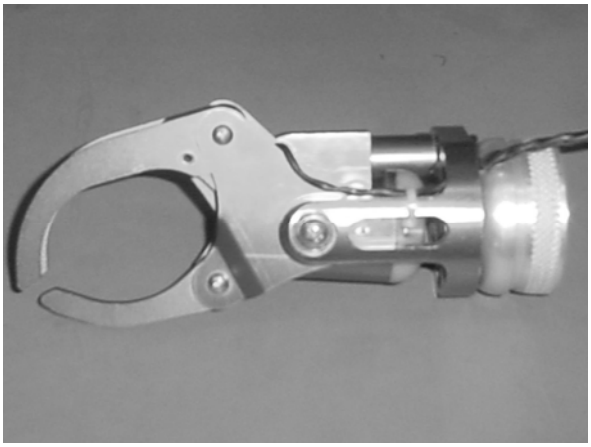

Figure 2 - **The mechanical hand with the leadscrew decelerator**

Figure 3 - **Clinical evaluation using STEF (Simple Test for Evaluation hand Function)**

The second technology is the "harmonic drive", which is a state-of-the-art decelerator that provides a high speed-reduction ratio with a single harmonic drive component. Figure 4 shows a harmonic drive embedded mechanical hand, with the following hardware specifications.

1. Length: 140mm (from the finger tip to the wrist)
2. Weight: 420g
3. Number of functions: 1 (hand open-close)

4. Power of finger (three-point) pinch: 4.5kg (measured by *North Coast^{TM} Hydraulic Pinch Gauge NC70141)*

These specifications indicate that the harmonic hand is heavier than the leadscrew hand, however, this harmonic hand has greater finger power for three-point pinching, making for a stable pinch function.

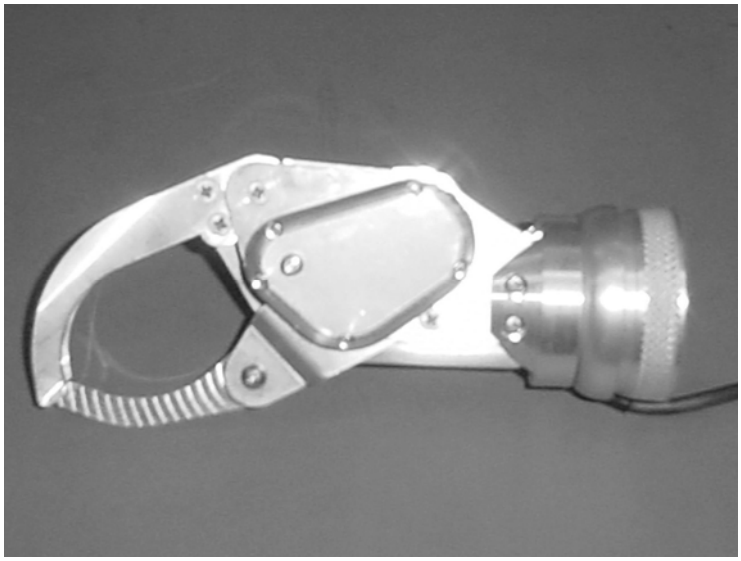

Figure 4 - The mechanical hand with the harmonic drive decelerator

4. CONCLUSION

This paper briefly discussed the multi-function and the basic function controller, as well as the basic function mechanical hand. The basic function controller and the mechanical hand with the leadscrew have already undergone some clinical evaluations. Although clinical evaluations for the multi-functional controller and the harmonic drive embedded mechanical hand have not begun yet, they are due to commence this year. In addition to these developments, we are also carrying out other projects to broaden the range of applications for myoelectric controllers, and reports concerning those projects will appear in the future.

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IMPROVED CONTROL FOR AN ARTIFICIAL ARM

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INTRODUCTION

Successful control of multi-degree of freedom upper limb prostheses generally uses some form of sequential instruction. This is because simultaneous control of multiple inputs requires a considerable concentration to be operated effectively. In contrast, the natural arm is controlled in a parallel manner with a high level of subconscious control. Such control uses feedback, the person is rarely conscious of the feedback information, and most of the control is automatic. Attempts to achieve similar control with a prosthesis would requires a wide bandwidth feedback channel to the controller. This is currently impractical if the controller of a multiple degree of freedom arm is the wearer, because only very low frequency feedback is achievable. The Southampton Arm control philosophy avoids this bottleneck by keeping the low level control within the prosthesis and leaves low bandwidth and strategic control to the operator [1].

This study describes the development of a control system for an artificial arm based on this concept. A joystick was used to input control signals via a harness at the shoulder, allowing a user to manipulate the arm with small movements of their acromium. A co-ordinate control strategy (inverse kinematics or IK) was implemented allowing both the shoulder and elbow joints of the arm to move simultaneously giving a smooth, more life like, motion. A trial was developed to asses the cognitive load required to operate the arm using the Dual Task paradigm. Tests were carried out on 5 subjects, using two different control strategies, IK and a direct control strategy (DC) using two degrees of freedom separately.

BACKGROUND

The Southampton Arm project complemented the work on hierarchical control of the Southampton Hand system, which selected different grip forces and postures on a multi-degree of freedom hand mechanism [2], by use of touch and slip feedback to a microprocessor [3]. For arm control, it was observed that while arm motion could be described in terms of polar co-ordinates [4], this is more likely to be useful for an operator that has to be concerned with the control of the individual joints. It is probable that the operator is more concerned with the absolute position of the wrist in space in order for them to be able to target the hand for manipulation of the object of interest. Thus the Southampton Arm was developed to control the position of the wrist from a series of inputs from the shoulder (pro-retraction, elevation and lateral movement of the shoulder) [1]. A six degree of freedom arm was designed and built and an analogue controller developed. Naturally at the time (1977), the controller and arm were too large and heavy for further clinical development.

In 1997 the ToMPAW project was formed to investigate a fully modular prosthetic arm system controlled via a bus communications system with microprocessor controllers at each joint [5,6]. The advantage of such a system is that it can be extended simply, and new control formats can be changed easily. Thus for this study a ToMPAW hybrid arm, consisting of shoulder and elbow joints from the Edinburgh Modular Arm System (EMAS), hand from Leverhulme/Oxford Southampton Hand project (LO/SH), was extended with an input node that contained an

instrumented joystick, and programmed with a series of different control formats to test how usable such schemes are.

CONTROL INPUT

This joystick was connected to the shoulder of able bodied volunteers via a flexible shaft so that two motions of the shoulder could be mapped to the motions of the joystick (pro/retraction to X and elevation/depression mapped to Y) - Figure 1. With these two motions as input they could then be used to control different actions of the arm. For the experiment described here the actions were either the position of the wrist in a vertical plane extending from the shoulder, of mapping the angle of each joint to its own input.

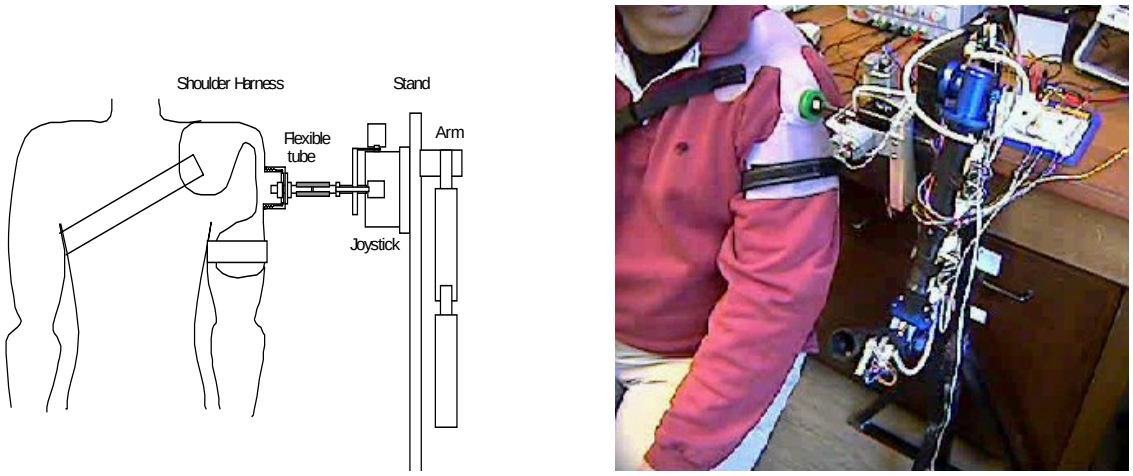

Figure 1 User interface harness and layout of subject relative to the prosthetic arm

The fully modular bus structure of the arm system allows for easy addition of input modalities. The master input processor then takes the signals from the node on the joystick and calculates required positions or motions of the individual joints depending on the control format employed. The two input formats discussed here were:

Direct control (DC) - Each of the two inputs at the joystick were mapped to the control of individual joints, so that the prosthetic shoulder was controlled with shoulder elevation and elbow flexion by shoulder pro-retraction,

Co-ordinated Control (IK) - The position of the wrist was mapped to a position in the plane of the arm motion. For any given position of wrist within a reachable workspace (Figure 2) there were a unique pair of angles for the shoulder and elbow. These values could be accessed from a look up table in the microprocessor's memory and then sent on the bus to the controllers of the individual joints, which controlled the detailed motions. The envelope within which the arm moved was positioned relative to the centre of rotation of the shoulder and for the experiments this centre was placed 15cm laterally to the centre of the subjects own shoulder and three targets were placed within the envelope close to the edge of the area, (Figures 1 and 2).

EXPERIMENT

Dual Task Paradigm

The aim of the experiment was to assess how easy it was to use one form of control, or another. One measure of ease of use is the cognitive burden that controlling the device places on the operator. In this case to measure this load the Dual Task Paradigm was employed [7.8]. This asks a subject to perform the task of interest (ie using the control strategy under test). The person's ability at this task is then measured. Then a second task is introduced that will occupy the person at the same time and their effect on the performance of each of the tasks is measured by the drop in performance of each of the tasks operated alone. The combined effect of one on another is a measure of the mental load the primary task requires. This can then be used to compare with an alternate control method also used as a primary task and measured relative the same secondary task.

Experimental Protocol

Five subjects (male, able bodied volunteers, ages 21 or 22, without any experience of prosthetic arm use) were asked to control the arm via the joystick input in two ways, either by direct control (DC) of the prosthetic shoulder with shoulder elevation and elbow flexion by pro-retraction, or by co-ordinated control (IK) with the XY position of the prosthetic wrist controlled by the XY position of the natural shoulder.

The primary task which used the two control formats was to ask the subject for five minutes to place the wrist through a sequence of target rings, placed close to the edge of the motion envelope. The secondary task was to perform multiplication tables for five minutes.

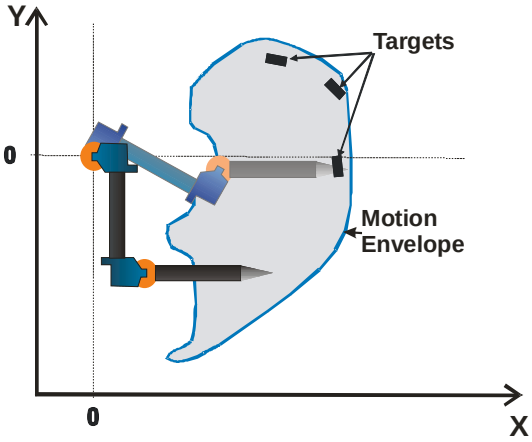

Figure 2 Arm motion envelope and experimental set up, showing the reach envelope and the position of the targets.

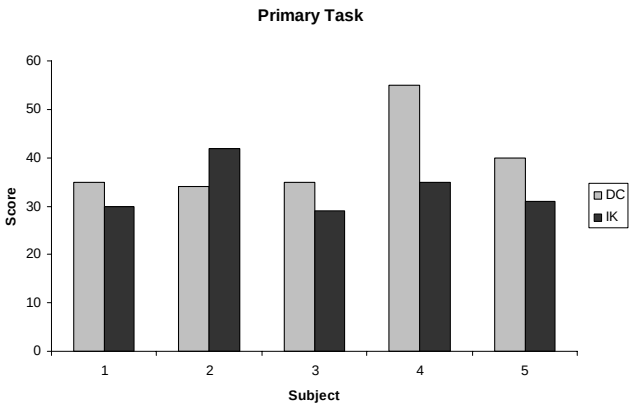

Figure 3 Individual scores for subjects performing the primary task alone.

Sequence

- Subjects were asked to perform the secondary task alone for five minutes and every multiplication completed was recorded as a score of one.

- They were then asked to perform the primary task with one of the control formats for five minutes, each successful wrist through the ring was recorded as a score of one.

- They were then asked to undertake both tasks at once and both scores recorded.

- These last two steps were then repeated for the alternate primary task.

RESULTS

It was observed that attempts to perform two tasks simultaneously detrimentally effected the performance of both tasks.

Primary task - Subjects gained lower scores in Co-ordinated control (IK) task alone compared with the Direct Control (DC), (Figure 3), mean IK = 33±5 (standard deviation) and mean DC = 40±8.

Once the secondary task was added the effect on the alternate primary tasks was very different (Means: IK = 31±5 and DC = 34±4). Essentially the Direct Control task was more effected by the secondary task.

Secondary task - The secondary task was also effected by the primary task. Alone the mean score for the secondary task was: 133± 30. Both tasks had a similar effect on the secondary task (Means: IK = 88±30 and DC = 81±28), Figure 4.

Analysis

One factor that is observed is that the individuals scores depend on their ability to perform the task at all. To enable meaningful comparisons *between subjects*, the effect of each individual's performance needs to be accounted for. Thus a better measure is the *percentage change* in the performance for each individual.

A second factor that needs to be accounted for is the effect of the primary and secondary tasks on the performance of each other. Thus an overall performance score was derived as being the sum of the percentage changes of the primary task on the secondary and the secondary task on the primary. This gave an overall rating for Co-ordinated Control (IK) of 34%±12% and DC of 41%±15%, (Figure 5). Thus the Direct Control was more effected by the secondary task.

DISCUSSION

The subject group was restricted to males as there is an observed difference in the ability of females to multi-task, relative to males and it would have been harder to separate out the influence of this out from the results. The direct control task was chosen, rather than a serial switched direct control task, as this proved to be *much* slower than the Co-ordinate control task. While the volunteers were naive prosthesis users they were of an age and gender with facility for performing such tasks so that it was as equal a test between the different control paradigms as possible to arrange.

The aim of using the second task was to occupy some of the attention (and so by inference the neural capacity) of the subject so lessening the performance in both. Thus if the data is

shown with the percentage of effect the tasks have on one another, the smaller the percentage effect recorded the more the person was able to process enough information to perform both tasks and so the less either occupied the attention of the subject. Thus the smaller the change the "easier" the task was to perform.

In this case the co-ordinated control (IK) task was easier to perform that the direct control (DC).

This study has shown that it is possible to measure the cognitive load imposed by operating a multi-joint arm prosthesis and so objectively compare means of assessing the ease of use of upper limb prostheses.

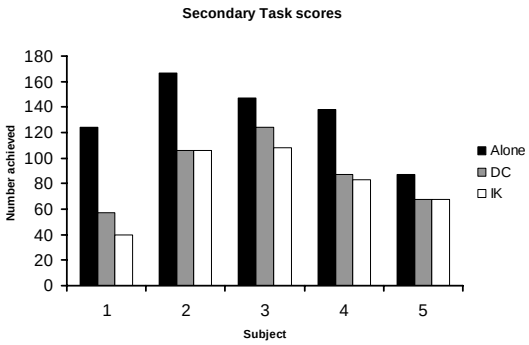

Figure 4 Result of the change in performance of the two primary tasks.

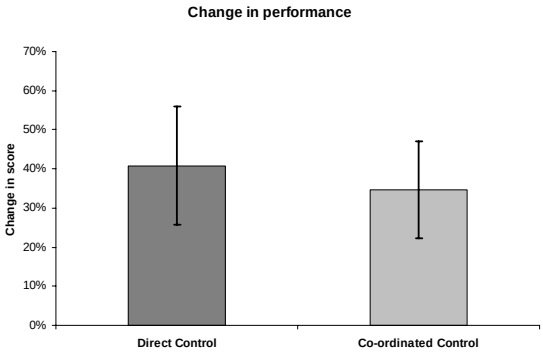

Figure 5 Overall percent change in the performance of both primary tasks.

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A MINIMAL JERK PROSTHESIS CONTROL SYSTEM

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Abstract:

Efficient prosthesis control is dependant on the user's ability to control the desired movements of the prosthesis. Observed manifestations of jerk on a complete arm prosthesis can lead to difficulty in performing controlled movements, especially under load on gravity assisted downward movements.

The application of so called soft start and soft stop routines for controlling the velocity profile of the prosthesis joint through its rotational movement can go some way to reducing this effect. It is proposed that an adaptive velocity control system can be applied to the same prosthesis under the same test conditions and reduce the discernible jerk considerably.

This adaptive system monitors the change is angular velocity thus controlling the second and third derivatives of position. The implication of actively controlled angular velocity lends itself to minimize jerk, combined with reduced power consumption, and an increase in parts life and reliability. This control is applicable to all externally powered prosthetic limbs, regardless of user interface.

Introduction to Jerk

It is well known that the first derivative of position (θ) with respect to time is velocity $(\dot{\theta})$, and the second is acceleration $(\ddot{\theta})$. The third derivative is the rate of change of acceleration $(\dddot{\theta})$, known technically as jerk (jolt in English). Jerk is a vector but may also be used loosely as a scalar quantity because there is not a separate term for the magnitude of jerk, equivalent to speed as the magnitude of velocity. The unit of jerk is metres per second cubed (m/s^3), although there is no universally agreement on a symbol for jerk.

In the field of prosthetics, the control of jerk can be seen in another dimension. That is, through controlling the rate of change of acceleration it is possible to smooth the profile of the arm as it moves trough the spectrum of its revolution.

Smoothed Velocity Profile Motor Control

A system with a feedback controller will drive the system to a state described by the input value. This desired value can be in term of position or of velocity. The combination of a motion controller and a drive actuator is referred to as an axis. A system which has more than one of these axes can be referred to as a complex motion control system. An example of such a system would be that of a complete prosthetic arm, where the shoulder, elbow, wrist and hand are all axes. Where position is the desired output of the control system, a movement from point A to point B dictates that

at Point A the velocity should be 0, and at point B the velocity should be 0. Between point A and Point B there should be a velocity.

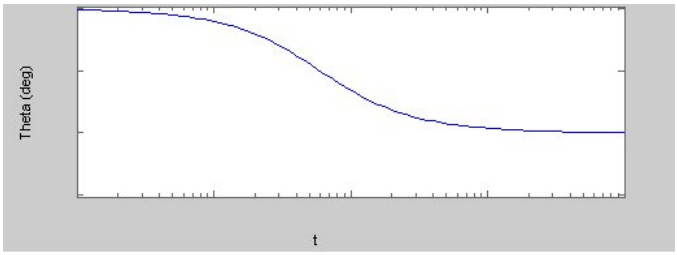

Figure 1. Change in Positon (θ) with respect to time (t).

A condition of moving between point A and Point B (Figure 1) will be that the velocity and the change in velocity be controlled. In this instance a velocity profile graph would produce a trapezoidal profile (Figure 2). The top level of the trapezoid is the maximum velocity. The total area under the profile is the distance moved, and the slopes are the maximum acceleration/deceleration allowed.

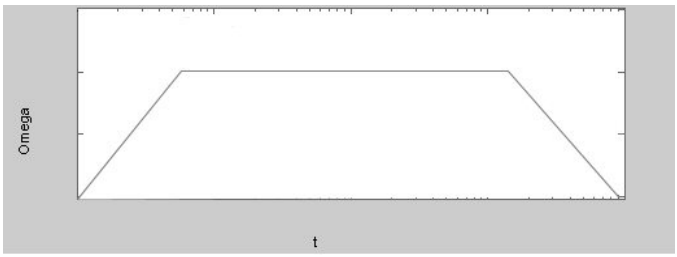

Figure 2. Trapezoidal Velocity (ω) with respect to time (t)

Jerk (m/s^3), described as the rate of change in acceleration with respect to time, can be seen to manifest itself in the trapezoidal profile where changes in acceleration can be seen as sharp corners. To minimise the effects of jerk on the prosthesis, a method that generates a smoothed profile is presented.

Where:

$\omega_{\max}$ = maximum velocity

$\alpha_{\max}$ = maximum acceleration

t_{acc} = acceleration time

t_{dec} = deceleration time

$t_{\max}$ = time at maximum velocity

t_{total} = the total time in motion

It can be seen that, the initial acceleration from standstill can be generated using:

$$\omega(t) = At^2 \tag{1}$$

Where A is defined as:

$$A = \frac{\alpha_{\max}}{2t_{acc}} \tag{2}$$

Where t_{acc} is defined as:

$$tacc = \frac{2\omega_{\max}}{\alpha_{\max}} \tag{3}$$

By substitution

$$A = \frac{\alpha_{\max}}{2(\frac{2\omega_{\max}}{\alpha_{\max}})} \tag{4}$$

This gives the equation,

$$A = \frac{\alpha^2{}_{\max}}{4\omega_{\max}} \tag{5}$$

And therefore,

$$\omega(t) = \frac{\alpha^2{}_{\max}}{4\omega_{\max}}.t^2 \tag{6}$$

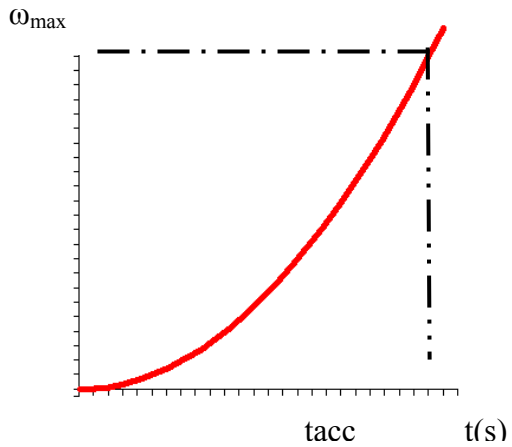

Figure 3. Acceleration from rest towards ω_{max} with respect to time (t)

This equation provides a exponential growth function from rest, but does not provide for smoothing toward the ω_{max} value. In order to perform the smoothing function towards ω_{max} , then a second equation must be applied. So this first equation is applied to the first half of the acceleration (t_{acc}). The conditional statement is made that if:

$$t \le \frac{t_{acc}}{2} \tag{7}$$

then

$$\omega(t) = \frac{\alpha^2{}_{\max}}{4\omega_{\max}}.t^2 \tag{8}$$

For t values greater than $t_{acc}/2$ the first equation can be used to find the second equation.

$$\omega(t) = \omega_{\max} - \frac{\alpha_{\max}^{\ 2}}{4\omega_{\max}}(t_{acc} - t)^2 \qquad (9)$$

This equation now gives the following response.

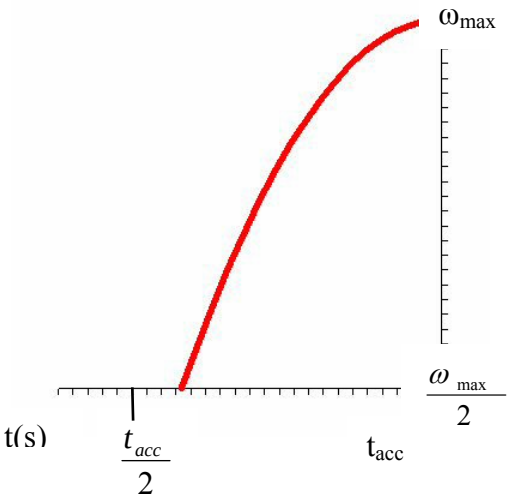

Figure 4. Acceleration from $t_{acc}/2$ towards ω_{max} with respect to time.

If t = t_{acc} then ω(t) would be equal to ω_{max}. Given that ω_{max} is an indefinite period of time controlled by the user, then the deceleration will come immediately after the suspension of the users desired motion.

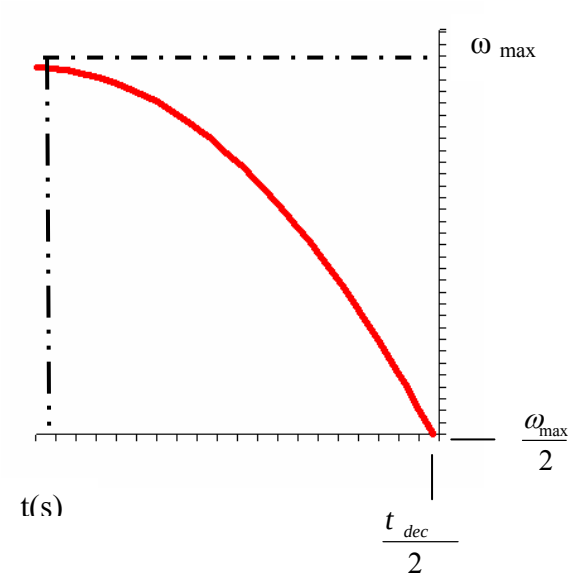

Figure 5. Deceleration from ω_{max} towards $t_{acc}/2$

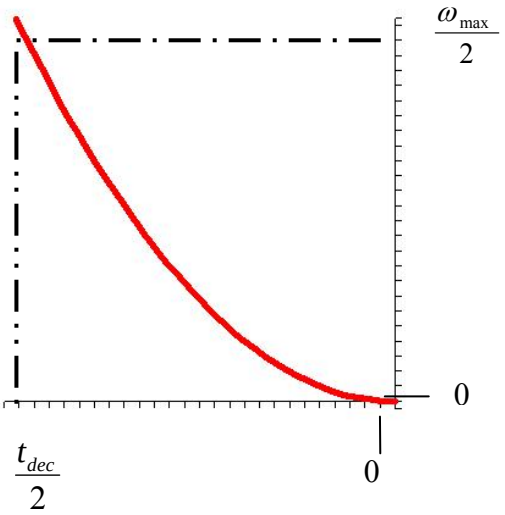

Figure 6. Deceleration from $t_{acc}/2$ towards ω = 0.

The deceleration t_{dec} factor can be related to the acceleration factor. However a totally different value for deceleration can be used, thus the time for t_{acc} could be different in value to t_{dec}. This factor can be tailored to each user's requirement, to minimise the amount of perceived over-run. Given that the human system is able to predict and adapt to the fact that there is a perceivable overrun, it is hoped that the

human can accommodate for this constant factor. Each of the elements can be implemented as a separate function. The initial condition of zero velocity to $t_{acc}/2$ is assigned the tseg1 designation, similarly the element from $t_{acc}/2$ to $\omega_{\max}$ is designated as function: tseg2. It can be seen that these functions can only be used for a positive or forward velocity. An additional 4 tsegs allocations are made to describe the negative or reverse velocity.

By combining these segments with the appropriate logic and modelled in mathworks matlab. The armature resistance, rotor inertia and rotor inductance, and other characteristics of the motor can be encompassed in the simulation.

Hardware Implementation

The hardware takes the form of an embedded bus based communication system. The bus chosen was controller area network (CAN) bus [1,2]. This communication system takes user input signals from the prosthesis/user interface, and distributes these signals along the complete prosthetic arm. The positional transducer chosen was an absolute encoder, giving 128 positional states, equating to 2.851°/step. Acceleration is measured by monitoring the rate of change of these positional states. Given the modelling of the motor response to voltage, it is possible to build a set of rules that describe the acceleration of the motor without having to actively monitor the step changes of the positional transducer.

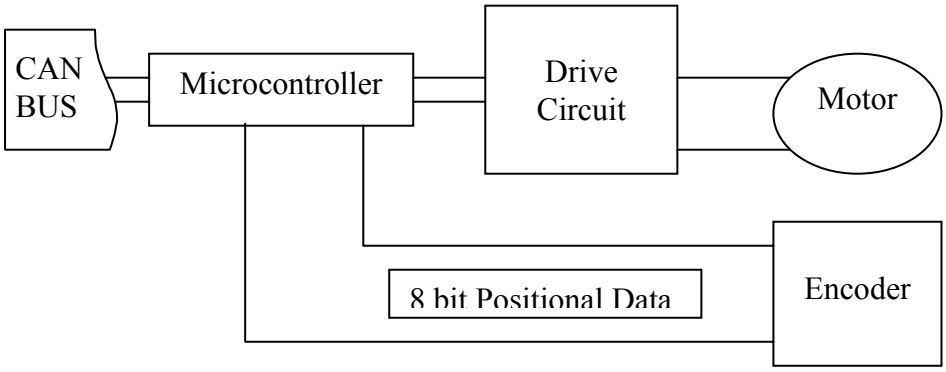

Figure 7. Hardware implementation of Minimal Jerk Control

Conclusions:

It can be seen on a practical level that this velocity profiling approach can reduce the effects of jerk. Moreover this represents a significant improvement for controlling the amount of momentum artefact that exhibits itself on the user. Not only does this approach increase the parts-life of the prosthesis, but it also simplifies control of the prosthesis by making movements more reliable and controlled.

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REAL-TIME MYOELECTRIC CONTROL IN A VIRTUAL ENVIRONMENT TO RELATE USABILITY VS. ACCURACY

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INTRODUCTION

Many clinically available, upper-extremity prosthetic limbs provide myoelectric control of a single device, such as a hand, elbow, or wrist. Most commonly, these systems yield control information from myoelectric signal (MES) amplitude [1] or rate of change of MES [2]. Such systems have been beneficial; however, prosthetic users would no doubt find enhanced functionality and usability if they could reliably control more than a single function (or device). Seeking to address this issue, extensive work has gone into developing schemes that provide multifunction myoelectric classification with very high accuracy [3]. However, for all continuous multifunction MES classifiers, no matter how accurate and repeatable, there exists no defined threshold (classification accuracy) of acceptability. This is due, in large part, to the limited availability of prosthetic devices housing multiple electromechanical functions.

Described in this paper is a recently developed MES control software tool that incorporates state-of-the-art multifunction control and a multifunction, real-time virtual limb. Twelve subjects have performed a virtual clothes pin functional test [4] yielding results that relate classification accuracy and multifunction device usability. Preliminary results indicate no strong relationship between accuracy and usability scores as determined by this tool.

SOFTWARE SUITE

A Matlab[1] based software suite, named CEVEN (Classifier Evaluation in a Virtual ENvironment), has been developed as a MES clinical and research tool. Specifically, the integrated system encompasses continuous, multifunction MES control and is highlighted by a real-time virtual arm environment. The 3D graphical representation of a human arm is able to perform all physiologically accurate motions from shoulder to wrist and includes two hand grips: a chuck (power) grip and a key grip. The virtual limb is controlled in real-time as amplified MES signals are fed to the host PC via A/D hardware and classified using one of many available multifunction control schemes. The control schemes and the associated training processes are further described below.

CEVEN is a graphical user interface (GUI) based system, meaning users require only clinical knowledge of MES rather than technical details to operate in either a clinical or research setting. The main GUI window presents users with three screen choices: "Acquisition", "Control", and "Virtual Environment". Data are obtained during an acquisition session, various control systems are trained on the acquired data, and then the virtual limb can be maneuvered as one of the processed controllers classifies real-time signals.

Acquisition

The "Acquisition" interface allows specification of signal input parameters (choice of A/D hardware, channel selection, sample rate) and customization of experimental protocol. Users compile a list of motion classes (elbow flexion, wrist supination, hand key grip, etc.) which a

[1] The MathWorks, Inc., 3 Apple Hill Drive, Natick, MA, USA 01760-2098

subject will be visually prompted to mimic and maintain during the acquisition of training data. Order of the presented classes can be randomized or arranged, duration of contraction and resting interval can be adjusted, and data recording parameters are established; all before the automated data acquisition session is started.

Additionally, an active display of raw input signals and their spectral content can be called from this "Acquisition" window for diagnostic purposes (activity level, electrode contact, noisy channels, etc.).

Control

When data acquisition is complete, users can select the "Control" tab on the CEVEN GUI to set up processing of the classification schemes. Alternative data to be processed can be loaded from any earlier recorded session. The "Control" interface allows settings and options of continuous multifunction MES control components (depicted in figure 1) to be tailored. CEVEN has been developed to incorporate myoelectric control refinements that have been made in recent years allowing various state-of-the-art systems to be tested. For feature extraction, Hudgins' time domain (TD) statistics [5], autoregressive (AR) features [6], and a combination of both (TDAR) are all available and complemented by dimensionality reduction in the form of principal components analysis (PCA) [7]. Three choices of classifier are also implemented: an efficient linear discriminant analysis (LDA) classifier [8]; artificial neural networks (ANN) for pattern classification [5, 9]; and an experimental Gaussian mixture model (GMM) algorithm [6]. All combinations of feature sets and classifier can be chosen in an investigation. In processing, any subset of available data channels can be used, data window size and overlap can be specified, the number of principal components can be chosen or omitted, post-processing (in the form of majority voting [3, 9]) can be set up, and simple proportional speed control is available. All selected control configurations are automatically trained and tested giving a result (% classification error) presented in a color-coded table. The set of control schemes can be stored for later use or any one can be selected and set to control the virtual arm in real-time.

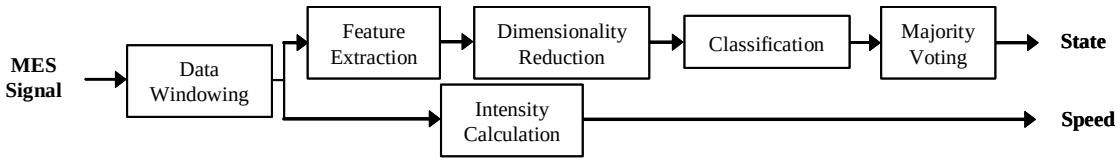


Figure 1: *Continuous multifunction MES control stages for CEVEN*

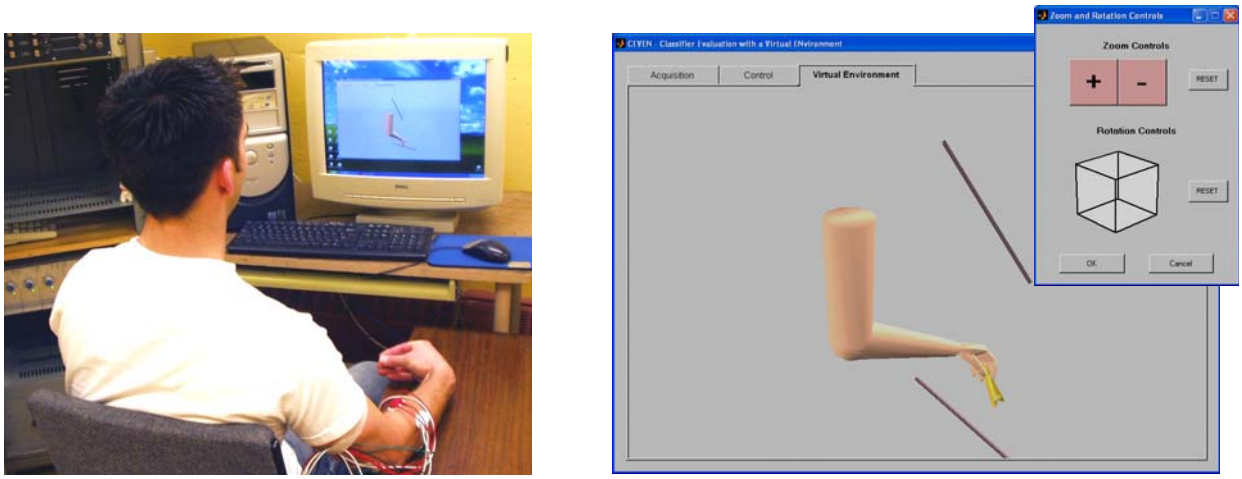
Virtual Environment

The "Virtual Environment" window of CEVEN houses the virtual arm. Aesthetically, this GUI is customizable, allowing choice of skin color, right or left arm, viewing orientation and size. Manual controls for all motions can be called (hidden by default) and limits can be modified from their physiologically accurate presets. If these manual controls are present, they appear beside the arm image and become inactive during real-time control. Finally, the speed of each arm component can be GUI-tailored.

To enhance the desired 3D effect, the virtual arm is complete with directional lighting and shading and is projected with perspective. The virtual environment screen is intentionally uncluttered; hidden menus and options are accessible with a mouse right-click.

As a test of function and usability, the virtual environment has been furnished with an emulation of a clothes pin test used in a real world assessment of multifunction MES control [4].

The test involves picking up a clothes pin from a horizontal bar and placing it on a vertical bar, requiring the use of elbow flexion/extension, wrist pronation/supination, and terminal grip and release. Two modes of testing are available: 1) counting number of pins successfully placed in a set time; or 2) timing how long it takes the subject to place a chosen number of pins. Test timing, pin counting, and result logging are all automatic since the task exists in virtual space on a PC. The virtual environment clothes pin test has visual aids not necessary in the real world task; the clothes pin changes to a red color when the virtual arm is gripping or moving the pin and turns green in color when position and orientation are acceptable for placement. A clothes pin released prematurely or in an unacceptable position constitutes a "drop" and a new pin appears (wooden color) clamped on the horizontal bar ready for retrieval.



(a) **(b)**

Figure 2: *(a) Subject performing the virtual clothespin task; (b) screen capture of the virtual environment tab (showing the virtual limb clothespin task) and the viewing controls window*

METHODS

The twelve normally limbed male subjects who participated in this preliminary study completed five discrete sessions where data for the six required classes of motion (listed above) were obtained, all control configurations were trained, and the virtual clothes pin test was performed for three different multifunction controllers (the most accurate, the least accurate, and a moderate – for each session) in random order. Each clothes pin test timed the subjects to place three clothes pins [4]. Eight differential channels of MES training data for were acquired from four sites equally spaced around the forearm and four equally spaced transhumeral sites.

This minor experiment serves dual purpose: 1) to promote and highlight one of the uses of this developed tool; and, 2) to initiate investigations of accuracy and usability relationships.

RESULTS

The scatterplot of Figure 3 shows the mean clothes pin placement time for each three-pin test plotted against the classification accuracy of control scheme used; 162 data points in total. The high degree of scatter suggests minimal correlation between classification accuracy and usability; much lower than that expected. A linear "best fit" line illustrates the loose relationship uncovered in this study.

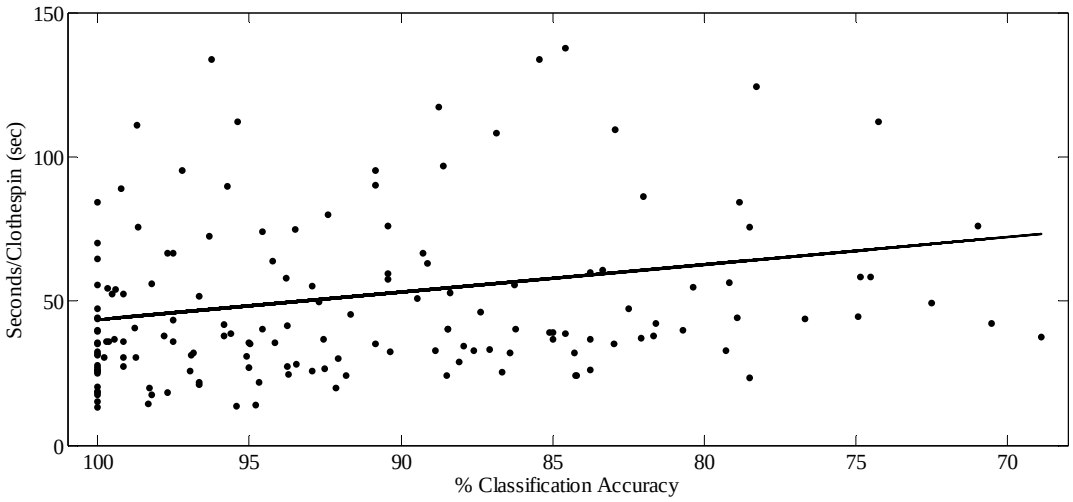


Figure 3: *Usability scores versus classification accuracies*

DISCUSSION

Observations made during the virtual testing support combination of additional measures and accuracy to better predict usability. It has been noted that having the majority of classification errors occur in one class of motion can yield relatively high overall accuracy, but the usability test becomes nearly impossible to complete. Conversely, a larger number of errors, more evenly distributed among classes, will compute lower overall accuracies but can still provide good functionality. Other key observations include: faster than anticipated user-acceptance of virtual control; and a possible need for contractions of variable intensity in the training data. Finally, the need to implement and test additional functional tasks into the existing software is evident.

CONCLUSIONS

This paper details a multifunction control software suite which has been developed for clinical and academic use, for assessment and refinement of multifunction myoelectric control systems. A clothespin virtual test has been carried out which highlights one use of the software and uncovers the need for further functional testing to relate usability and classification accuracy.

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AN ELECTROMECHANICAL QUICK-CONNECT MECHANISM FOR MYOELECTRIC PROSTHESES USING SILICONE SLEEVES

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INTRODUCTION

A unique children's upper-limb prosthesis prototype with an improved suspension and electronic interconnect has been developed. Silicone liners provide effective suspension for prostheses. However, they have been problematic for the upper extremity myoelectric application. Recently, we fit a child using a silicone liner and hardwired, remote electrodes. During this fitting, a number of challenges were encountered, including wire breakage, difficulty of electrode and wire attachment, and difficulty donning/doffing the prosthesis due to mechanical hindrance from the wires and the modifications made to accommodate them. Consequently, we developed an **electromechanical quick-connect** attachment. It enables the user to easily connect/disconnect the prosthetic shell and hand system to/from the liner, yet still provides for the mechanical suspension/connection between liner and prosthesis and provides for the electrical connection of sensor electrodes with control electronics. For adults and children with upper-extremity limb loss, this development makes the use of silicone sleeve systems more practical.

BACKGROUND

The use of silicone suspensions for upper extremity prostheses is becoming more prevalent, in many cases replacing the use of traditional 'hard' shell systems [1-6] and in some cases enabling solutions that weren't previously possible [7]. However, silicone suspensions require a mechanical and electrical connection between the silicone liner and the distal shell and hand system. Previous silicone suspension systems have not integrated the mechanical and electrical connections required, generally hardwiring the proximal signal electrodes with the distal electronics in the shell [3-5,8-10]. This results in several significant challenges for the clinician and the user, which include wire breakage, difficulty of electrode attachment, and difficulty of donning/doffing of the prosthesis [7]. There are no "off-the-shelf" solutions to these issues. Several types of upper extremity connectors exist for myoelectric prosthetics, but they do not address the unique challenges that come with the silicone suspensions. A combined electromechanical system that addresses these challenges and which is easy to put together for the clinician in the field was developed.

DISCUSSION

The clinical group in our Myoelectrics department was faced with a client they could not fit with a standard suspension. As a result, they fit her with a silicone sleeve suspension [7] (Refer to Figures 1 and 2 below). This worked well, but the wiring from the electrodes to the electronics in the prosthesis posed several problems, including wire breakage, cosmesis, difficulty of electrode and wire attachment, and difficulty donning/doffing the prosthesis due to mechanical hindrance from the wires and the modifications made to accommodate them. These problems were addressed by our powered upper extremity prosthetics program group (PUEPP), which is comprised of clinical staff, rehabilitation engineering staff, and industry partner staff. The result was the design and development of the electromechanical quick-connect. The quick-connect design enabled the prosthesis to have two completely separate pieces that could easily be connected, forming both the mechanical connection and electrical connection between the two pieces at the time of donning (Refer to Figures 3 and 4 below). The quick-connect also enabled the prosthesis to be more cosmetically appealing. In addition to the initial client who was fitted with this system, we have successfully fitted a new client with this system.

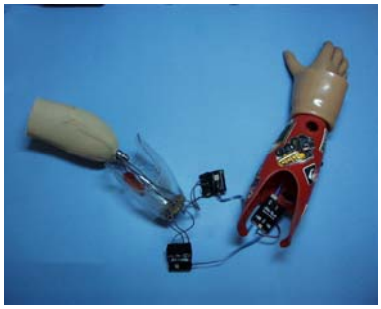

Figure 1. Hardwired Prosthesis

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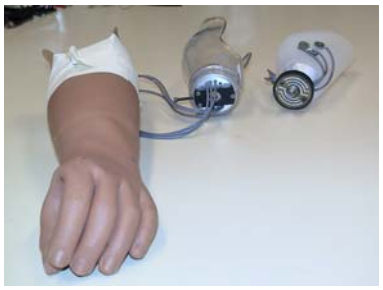

Figure 3. Prosthesis with Quick-connect

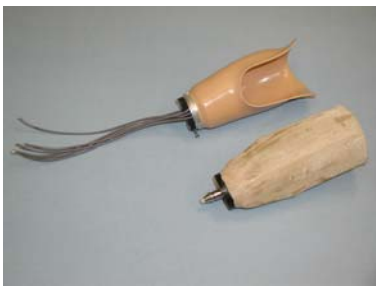

Figure 2. Prosthesis with Quick-connect

This development takes advantage of components developed in other industries for electrical and mechanical connection. Integrating the electrical connection and the mechanical connection in one location alleviates the problems with wire breakage and difficult donning/doffing due to wires external to the mechanical connection. The two connector sides (proximal and distal) are guided into contact with each other and then locked into place with a slight rotation. This design allows the user to establish both the

electrical and mechanical connections quickly with a simple one-handed movement, which is beneficial for those with unilateral limb loss. The quick-connect has a thin profile so that it minimizes the length that is added to the prosthesis, which is important for both functionality and cosmesis. Furthermore, since a component of the connector is a printed circuit board, electronic functions such as gain and filtering are easily incorporated. This also allows the quick-connect to be more flexible for other input signal types, such as switch input and mechanomyograms, since the electronics to accommodate them are easily incorporated. The silicone suspension itself is most advantageous in the cases of long and short residuum. With a long residuum, the suspension does not need to go over the proximal joint, giving greater range of motion. With a short residuum, suspension can be maintained without the need for a more complex, range-of-motion-limiting system (such as a transhumeral traditional suspension in the case of a short below elbow residuum).

Potential drawbacks of the development are increased electrical complexity and additional potential sources for noise injection, leading potentially to decreased signal-to-noise ratios. However, we tested the prototype during the operation of a VASI artificial hand, and the tests showed the noise level to be acceptable compared to a version wired directly with no electrical connector. The increased complexity also has the potential to affect reliability and robustness. However, during the tests and client use these potential disadvantages have not been seen. Also, if we do encounter noise problems, the design allows us to incorporate additional electronics in the quick-connect or to change the connector metal (e.g., gold coating) so that the performance is less affected by noise.

SUMMARY

Our PUEPP team has developed an electromechanical quick-connect which has improved the functionality of our prostheses which use silicone sleeve suspensions. This development has resolved difficult issues related to the wiring of silicone sleeve prostheses, such as wire breakage, difficulty of electrode and wire attachment, and difficulty donning/doffing. It has resulted in functional silicone sleeve prostheses which are easier to use and more cosmetically appealing and which should have improved reliability.

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THE NTNU REVOLUTE WRIST DEVICE (NRWD): A KINEMATICALLY OPITMIZED EXTERNALLY POWERED WRIST PROSTHESIS

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INTRODUCTION

Transradial amputation implies the loss of the wrist. It therefore also implies the loss of the ability to do flexion/extension and radioulnar deviation as well as the forearm's ability to rotate the wrist (prosupination). Prosupination is offered by some prostheses, but many users choose not to use a powered wrist because the benefits are literally outweighed by its weight, poor cosmesis and the complexity of controlling multiple degrees of freedom. This is also why a wrist with more than one motorized function is considered by many not to be a viable concept.

This paper reports a study that aims at producing a lightweight, single-degree-of-freedom, externally powered wrist whose kinematic properties are optimized to match the movements of a healthy wrist in conjunction with certain Activities of Daily Living (ADL). The rationale for this is the assumption that reducing the need for compensatory movements whilst keeping the weight and control complexity unchanged will improve user acceptance and utilization of the device. The prosthesis provides for digital communication and coordinated motion.

OPTIMAL KINEMATICS

Data Acquisition and Processing

Eight healthy subjects were asked to perform 15 different ADL while their forearm and hand movements were measured with a MotionStar electromagnetic tracker (Ascension Technology Corp., Burlington, VT, USA). Subsequently a one-degree-of-freedom kinematic model was fitted to the entire data set as well as to the data subsets for each activity, age group and sex. The kinematic model comprised an axis of rotation and a rotational offset that allowed the hand to be statically tilted perpendicularly to the axis of rotation [1]. Orientation statistics was employed in order to decouple the results from the choice of reference coordinate frames [2].

Results

The grand axis of rotation (based on all the data) was found to go through the wrist from a dorsal-lateral position proximally to a ventral-medial (palmar) position distally, forming an angle of 10.7° with the forearm and 6.9° with the sagittal plane. Furthermore the axis was displaced 18.9° with respect to the 3rd metacarpal and 13.2° with respect to the palmar plane of the hand. Figure 1 shows two wrist postures corresponding to this grand axis of rotation; note the presence of a slight wrist extension (i.e. a rotational offset) throughout the wrist's workspace.

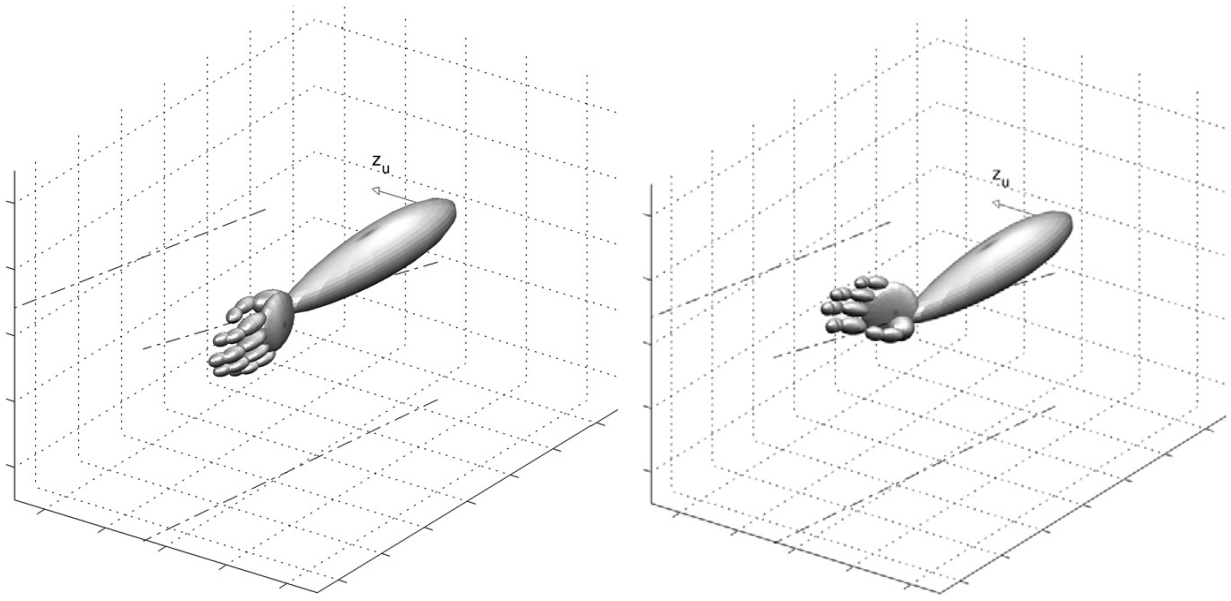

Figure 1 Grand axis of rotation. The axis is indicated with a dash-dotted line, which is projected onto a horizontal and a vertical plane in order to help in the three-dimensional interpretation of the figure. The arrow labelled Z_u marks the medial direction. The left figure shows the wrist in a somewhat “neutral” posture, while to the right the wrist is rotated 180º about the grand axis.

There were only small differences in the dominating axes of rotation for various groups of subjects, but significant deviations were found between the different ADL [1].

THE PROSTHESIS

The NTNU Revolute Wrist Device is an experimental design adapted to research on prosthesis kinematics, coordinated joint control and digital communication. The following sections outline the features of the first version of the wrist, referred to as NRWD-1.

Kinematics

Obliquity of the axis of rotation with respect to the forearm causes the wrist’s cross-section to be elliptic, cf. Figure **2**. This is desirable because a healthy human wrist is in fact elliptic rather than circular. While the grand axis of rotation identified implies an almost circular design, the axis for the individual ADL with the largest component in the dorsal-ventral direction yielded a wrist aspect ratio of more than 3:1.

The orientation of the axis of rotation and the static rotational offset between forearm and hand heavily influence the cosmetic appearance of the device, for example some male users consider a permanent wrist extension to be feminine. Since the different ADL also suggested quite different kinematics it was decided to make the rotational offset adjustable, with the grand kinematic parameters as the preferred configuration.

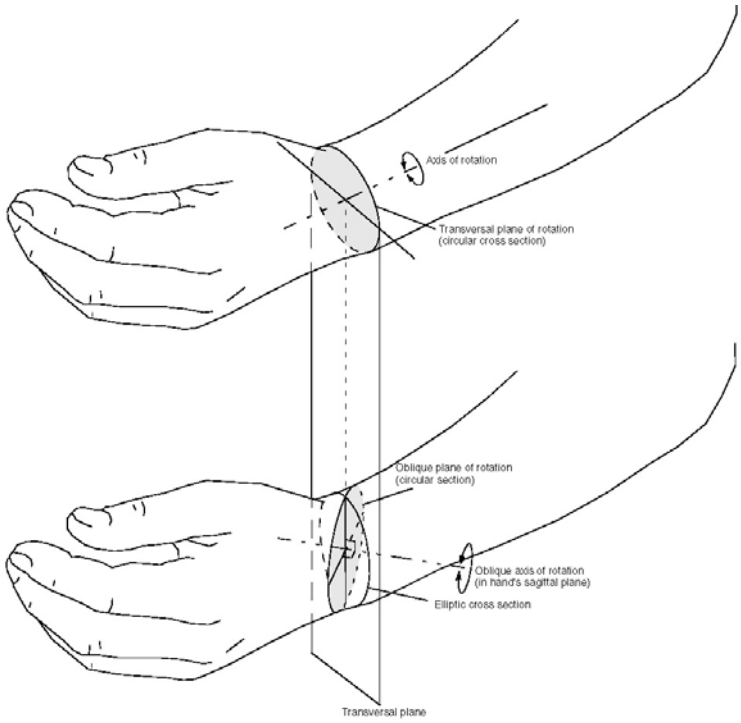

Figure 2 An oblique axis of rotation gives the wrist a more human-like elliptic cross-section.

Interfaces

Being an experimental device, the NRWD-1 should be able to operate with a multitude of other prosthetic components. Therefore it will not have a definite mechanical interface but rather be adapted to each situation.

The power interface tolerates supply voltages in the range 6-12 V, which covers most commercially relevant voltages. The device includes a microcontroller with two proximal and two distal communication lines. These lines are configurable as a CAN or I^2C bus or as dual analog inputs and outputs, respectively, for compatibility with commercial terminal devices as well as other experimental systems. Digital communication is believed to dominate the prostheses of the future, as it is nearly a prerequisite for more advanced coordinated multi-joint movement and it easily gives room for future growth in the amount of data transferred between i.e. the terminal device, wrist, elbow and control input electronics. The initial prototype circuit (Figure 3) employs DIP-switches for switching between the communication modes, but a future version will be based on solid-state components only. Furthermore, future versions will most likely be limited to a single data bus standard, depending on the success and direction of relevant developments and standardization activities [3]. Provisions are made for the NRWD-1 to read digital commands proximally and simulate analog EMG signals distally so that traditional terminal devices can be combined with novel arm components, using the NRWD-1 as a “digital-to-analog gateway”.

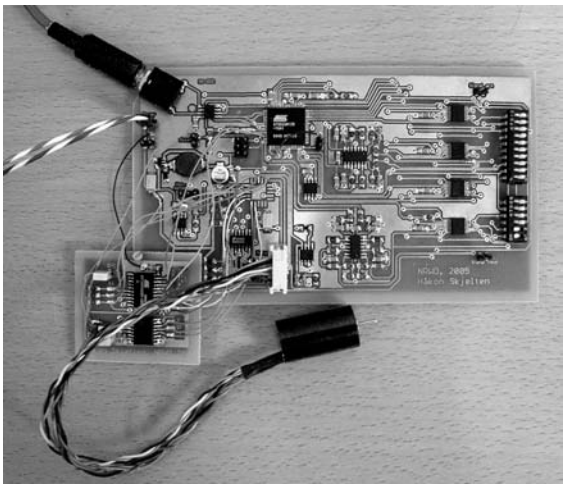

Figure 3 First prototype of the NRWD-1 control electronics. Below the large circuit board one can see the minute but powerful brushless motor that drives the wrist.

Modes of Operation

The NRWD-1 control software is modular, transparent to the communication mode. The control modes available include on/off, proportional torque, velocity or position and mechanical impedance control. Electromechanical properties are comparable to those of commercial devices. Emphasis is put on creating a fast dynamic response to allow continuous active use of the wrist as opposed to using it occasionally for re-orienting the hand.

DISCUSSION

The NRWD-1 is a novel experimental wrist prosthesis adapted to research on prosthesis kinematics, coordinated joint control and digital communication. A number of challenges are yet to be met, including miniaturization of the control electronics to obtain a single, self-contained unit. An expected outcome is to document the benefits, if any, of a kinematically optimized wrist design in the presence of traditional as well as advanced and novel control schemes.

ACKNOWLEDGEMENTS

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EVALUATION OF A PROTOTYPE ELECTRIC-POWERED PARTIAL-HAND PROSTHESIS

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ABSTRACT

A prototype partial hand mechanism has been developed for fittings at the trans-metacarpal level. Subjects selected for evaluation of the prototype device have absence of all fingers of the affected hand, all or some portion of the metacarpals present, at least one active intrinsic hand muscle, and relatively unimpaired wrist movement. The presence of one or two intrinsic hand muscles allows for proportional myoelectric control without resorting to forearm muscles that would be recruited during wrist positioning. The mechanism is suspended from the hand structure with a custom silicone socket. The socket extends no more proximal than the styloids so that the physiological wrist motion is unencumbered, allowing for orientation of the hand in a near physiological manner.

BACKGROUND

The hand mechanism is shown in Figure 1. It uses a single motor with a gear transmission and bi-directional backlock [1]. The motor and transmission are placed in the volume corresponding to the knuckles of the ring and little finger. The finger assembly is made up of the index and middle fingers machined as a unit and a separate thumb. The fingers are driven by the gear transmission and a linkage transfers movement of the finger unit to movement of the thumb. Using a 9-volt rechargeable battery, the fingers have a maximum speed of about 2 radians/sec (≈ 105°/sec) with a maximum grip force at the tips of about 53 N (≈ 12 lb_f).

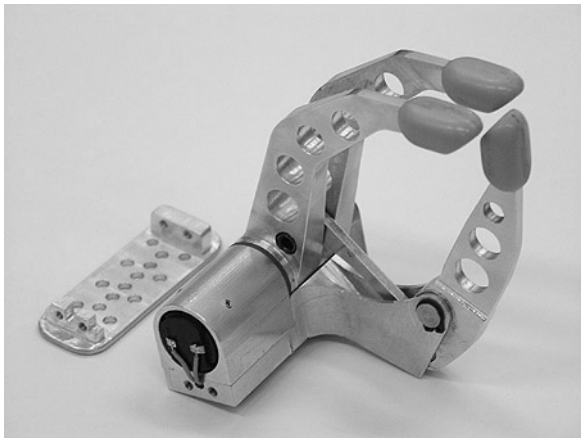

Figure 1. Prototype hand mechanism with socket mounting plate.

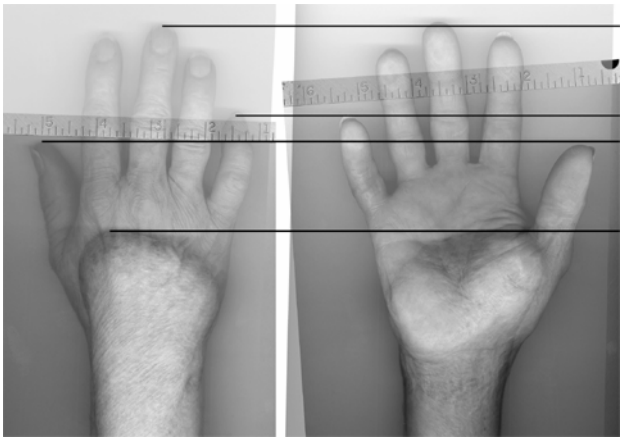

Figure 2. Subject's partial right hand superimposed on a reversed image of her left hand.

Our initial evaluation subject is a 64 year old female with a unilateral traumatic amputation of all fingers and thumb of the right hand through the metacarpals (see Figure 2). The amputation resulted from a burn injury in 1996 and both the residual hand and distal forearm are scarred. However, the skin is in good condition and the wrist is intact and mobile. The left hand and arm are unaffected by the trauma.

The subject has phantom finger sensation and palpation of the hand revealed several intrinsic muscles responding to her efforts to move the phantom fingers. Specifically, the Abductor Pollicis Brevis, the Flexor Pollicis Brevis, and the Abductor Digiti Minimi appear to be functional. Myotesting showed consistently moderate to strong signals from the Flexor Pollicis Brevis (elicited by imagining the thumb flexing across the palm) and the Abductor Digiti Minimi (elicited by imagining the fingers extended and spreading apart).

SUSPENSION

With functional intrinsic hand muscles, it is possible to have myoelectric control without resorting to signals from forearm muscles. Consequently, the physiological wrist can be used for positioning of the hand prosthesis without compromising control. To achieve this, the suspension system must be secure enough to hold the mechanism to the residual hand but without encumbering the wrist.

A surlyn check socket was constructed having a proximal trimline distal to the styloids. The check socket was used to evaluate placement of electrodes over the intrinsic muscles and control interference during wrist movements. Using Otto Bock 13E125 electrodes and a custom VASI Single Programmable Module (SPM) as a controller, the subject demonstrated the ability to proportionally open and close the hand at will and to rotate the forearm, flex and extend the wrist, and move the wrist in radial and ulnar deviation without inadvertent operation of the hand.

Confident of the subject's control capability, two custom silicone sockets were fabricated by the Custom Silicone Service of Otto Bock Canada. One socket had the hand mechanism mounting plate formed into the silicone. The second socket interfaced with a laminated shell that incorporated the mechanism mounting plate. The silicone socket with laminated shell was an alternative in the event that the silicone only socket did not offer enough rigidity to maintain the position of the hand mechanism when acted upon by external forces.

The subject was evaluated with both sockets and both provided excellent suspension. The sockets were able to remain on the subject's residual hand up to a maximum test loading of 11.4 kg_f (25 lb_f) with minimal displacement.

RESULTS

Our initial subject has been able to demonstrate good proportional myoelectric control of the prototype hand mechanism using intrinsic muscles of her residual hand. A custom silicone socket with myoelectrodes and mounting of the hand mechanism has been constructed. The next phase of the project is to complete the cosmetic covering of the mechanism and begin the field trial.

ACKNOWLEDGEMENT

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THE WILMER APPEALING PREHENSOR

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Abstract

A new type prosthetic prehensor for children aged 4 – 9 has been designed, constructed and build in an attempt to improve the appearance of the split-hook prosthesis. The new prehensor is a mechanically operated voluntary opening device. All mechanical parts are within the frame, and covered by a polyurethane cosmetic cover, which can be made in almost any color desired.

Clinical testing of the prehensor by 4 children with a total time of use of over 170 months, revealed the mechanism to be very robust and reliable. The colorful hook cover is highly appreciated by the children.

Encouraged by this success, a second larger size has been made intended for children aged 7 - 14. This medium size prehensor is now in clinical use by 1 child for over 12 months. Again, the appearance of the hook is highly appreciated, and again, the mechanism proves to be very reliable.

Introduction

The standard split hook prosthesis is, despite its functionality, most often rejected by parents of a child with an upper limb defect because of the very poor and deterring outward appearance. These parents prefer the more cosmetic, but less functional, hand prosthesis. The WILMER group started a project to develop a new prosthetic prehensor for these children. The objective was to preserve the functionality of the standard split hook prosthesis, while improving on the outward appearance.

Method

A shape study was performed to determine the outline of the new prehensor [1]. The resulting shape, preferred by almost all, is a hook-like prehensor, Figure 1. Its volume and outline are derived from the contour of a hand of a 4-6 year old child. The length of the fingertips and the position of the rotating finger are approximately similar to a healthy hand. The connection to the forearm is harmonic and smooth. All mechanical parts, including the operating cable, can be placed out of sight in the interior of the prehensor.

The mechanism design has gone through several iteration steps [1 – 3]. After each step a successful laboratory test was succeeded with a clinical trial with 4 – 5 children. As the initial mechanisms caused many repairs, a third redesign was made, Figure 2 [4]. A four bar linkage mechanism is used to reduce the input forces. A slightly inclining input characteristic ensures the controllability, for both shoulder harness and elbow-control. The resulting grip force is approximately constant over the full range of opening.

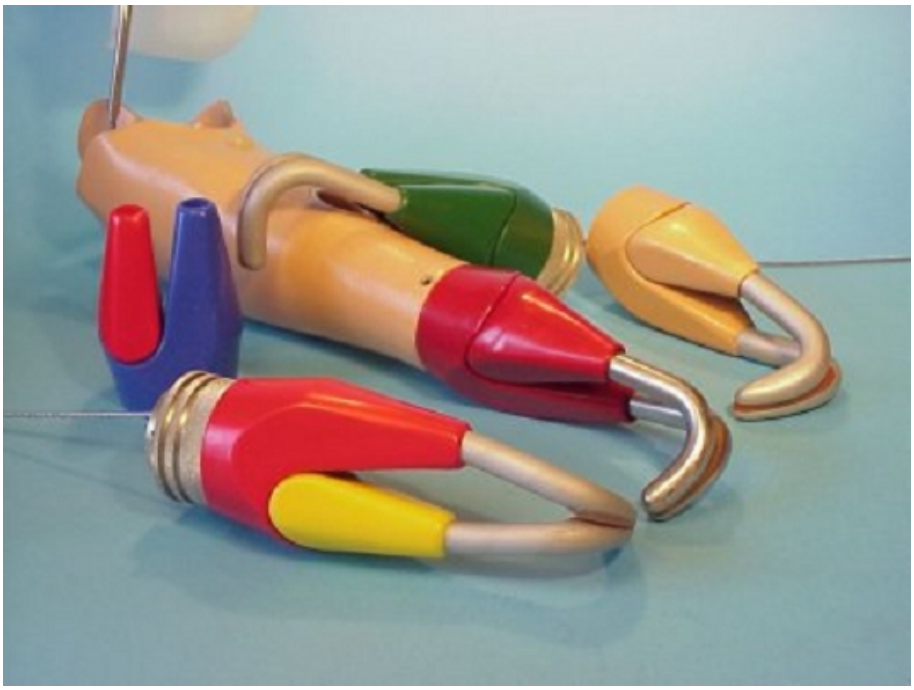

Figure 1. The WILMER appealing prehensor.

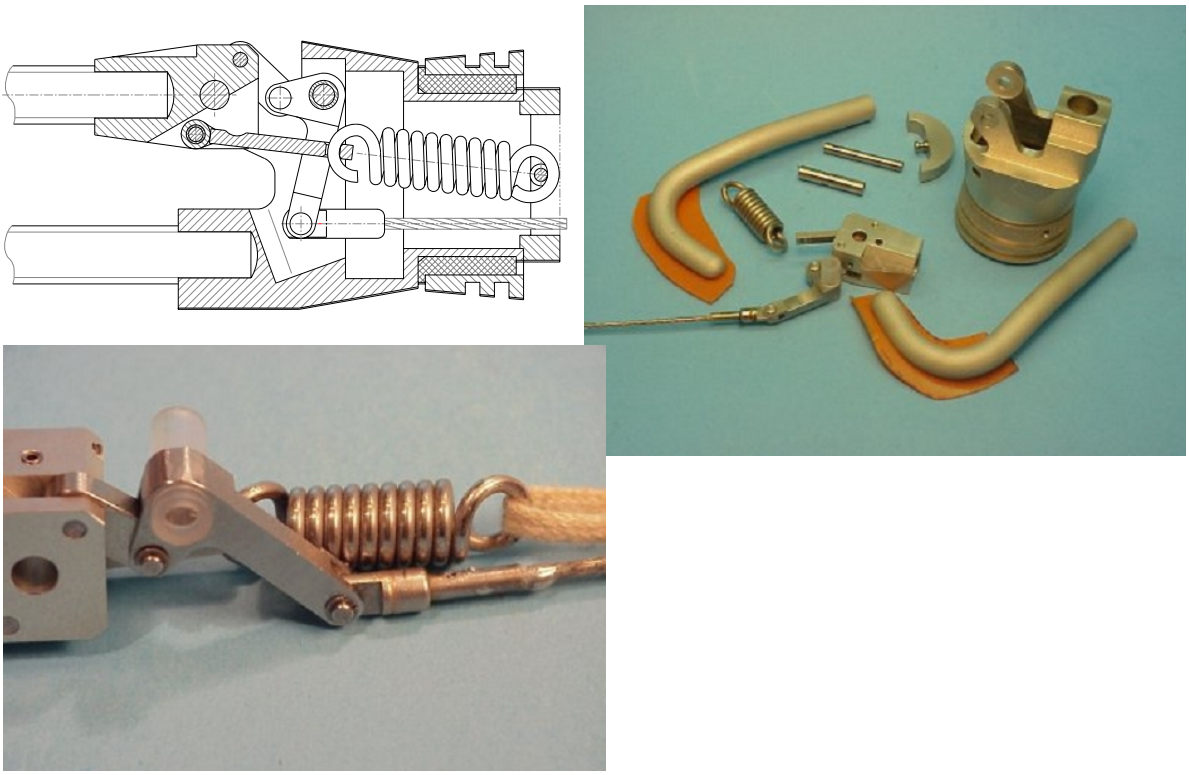

Figure 2. The WILMER appealing prehensor. Top left: cross sectional drawing of the mechanism; top right: several parts, partly assembled; bottom: assembled four bar mechanism.

Integrated into the frame of the prehensor is a lightweight friction wrist prosthesis. The frame is enclosed by a cosmetic cover made out of flexible polyurethane resin. This way several unique features were obtained: the outside of the prehensor is rugged and easy to maintain; the cover can be easily removed to access the mechanism; and the cover can be coloured. Giving the cover a bright primary colour emphasises the toy-like nature of the prehensor, thus advancing the acceptance and use of the prehensor by the child. It is even possible to supply several covers in different colours, which can be exchanged by the child according to its daily moods.

Results

After successful laboratory testing, the latest version of the WILMER appealing prehensor has been issued for clinical trials. It has been in daily use by four children for over 170 months of total testing time, Table 1.

Table 1. Users of clinical prototype WILMER appealing prehensor

Name	Gender	Born	Affected side	Period of use	No of fittings	Colour
AS	V	199208	L	199912 – to date	1	Yellow Red [200102] Blue [200305]
NB	V	199411	L	199904 – to date	2	Yellow & Red [1st fitting] Red & Yellow [2nd fitting]
JG	V	199508	R	199907 - 200002	1	Yellow
YK	M	199612	L	200111 – to date	2	Blue [1st & 2nd fitting]

User JG unfortunately suffered from several medical problems, not related to the arm defect, a few months after the start of the clinical trial. Her parents therefore decided to stop the use of any arm prosthesis. The other three users still use the WILMER appealing prehensor. User AS changed the colour of the hook on two occasions, but still uses the original mechanism. Users NB and YK on the other hand, each received a second prototype due to excessive wear of the fingers of the hook after respectively 27 and 40 months of use.

All children highly appreciated their device. It has not caused any negative reactions or strange associations, with, for instance, science fiction. Because of the smooth outline of the prehensor and the integration of the control cable, wear of clothing is reduced considerably. The children are delighted by the bright coloured appearance of the prehensor. All children chose a colour out of our standard palette, one [NB] of them mixed colours between the housing and the thumb.

Over the overall testing period only a few repairs were necessary. On two occasions an axle of the four bar mechanism was lost due to corrosion of the stainless steel [!!] retaining rings. After replacement of the axle and the retaining rings from a second delivery batch the problem did not occur again. On three occasions the operation cable broke at its connection to the mechanism. This could be repaired by renewal of the

soldered connection. Finally, on two occasions the rubber grip surface on the fingers of the hook partly detached from the fingers.

Discussion and conclusion

The overall results of the clinical testing have been very satisfying. The children and their parents highly appreciate the prehensor, because of its looks and because of its functionality and reliability. The reliability was also highly appreciated by the designers and by the prosthetists.

Encouraged by this success, a second larger size has been made intended for children aged 7 - 14. This medium size prehensor is now in clinical use by user NB since April 2004. She chose the hook cover purple and blue. Again, the appearance of the hook is highly appreciated, and again, the mechanism proves to be very reliable; no repairs have been necessary despite continuous daily use.

Furthermore, the appealing prehensor has drawn the attention of philosophers who deal with body image and personal identity in relation to bodily differences [5, 6]. They see the bodily difference as something that positively adds to the person's identity. In this respect wearing a brightly coloured artificial prehensor can be noted as a clear statement of personal uniqueness.

With the development of this new appealing prehensor we hope to have provided a way out of the classic prehensor dilemma: either cosmetics or function. This prehensor we believe combines both cosmetics and function.

Acknowledgements

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PRELIMINARY EXPERIENCE WITH HYDRAULICALLY DRIVEN HAND PROSTHESES

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INTRODUCTION

The need for further development of prosthetic hands with enhanced functionalities and better cosmetic appearance than conventional prosthetic hands became evident in many investigations of this topic [1-3]. Consequently, a new generation of multi-articulated hands for prosthetic application were designed in the past decade. Some of these hands are characterized by a multitude of miniature DC gear motors integrated into the hand [4-8], or by an underactuated mechanism driven by a single DC motor [4,9]. However, the transition from an experimental hand to a clinically viable hand is a crucial test for any new development. Different approaches using fluidic actuators were chosen by [10] and by our research group [11]. Unlike [10], we do not use a pneumatic drive system powered by pressurized CO_2 from disposable cartridges, but a compact electro-hydraulic system. Its components are micropump(s), microvalve(s), a reservoir, a controller, and small flexible fluidic actuators integrated into the finger joints. The flexible fluidic actuators expand during inflation, generating the flexion movement of the digits, whereas the extension movement is achieved by elastic elements [11]. Publications on standards for prosthetic hands and criteria to meet user requirements [1-3, 12] were analyzed and taken into consideration in designing new experimental hands. The results obtained in the first year with three hydraulically driven experimental hands are displayed and test experiences are presented.

FIRST PROTOTYPE

Designing the first hydraulically driven experimental prosthetic hand served these purposes: (1) Demonstrate the feasibility of a hydraulic mechanism as an alternative to DC motors. (2) Move the center of mass from the hand towards the residual limb by placing heavy components (valve and DC motor-driven micropump) on an external unit. (3) Design a multi-articulated hand able to grasp adaptively. Finally, the hand should have an anthropomorphic design and fit into a cosmetic glove.

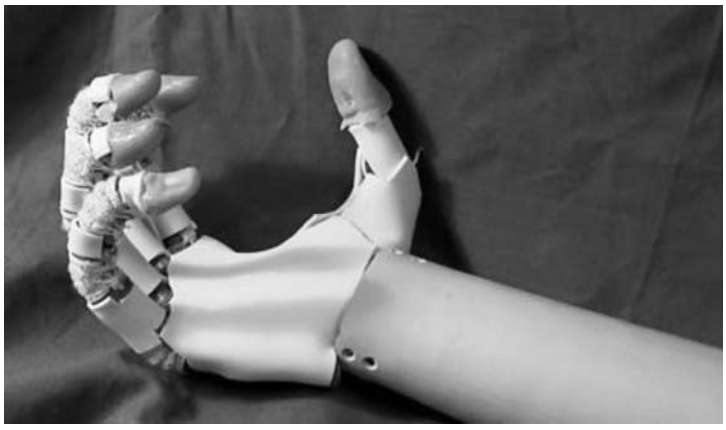

Fig. 1: The first prototype with 14 dependant DOF´s.

The hand is divided into two parts connected by a pressure line and a conductor. The external power supply contains the micropump (Speed 300 from Behotec, Bergkirchen, Germany) of 69 mm x 32 mm, a valve (A321-OC2 from Camozzi, Italy), the microcontroller (Infineon C164CI), and high-current NiMH batteries (Varta VH4000 4/3A). Measurements of power consumption indicated the need for a battery capacity of 4000 mAh to allow the hand to be operated for 12 hours with one charge. The digits include a framework of carbon fiber reinforced plastic "bones," and the joints are made of aluminium of high tensile strength. A total of 14 small flexible fluidic actuators are integrated in the digits at the metacarpophalangeal (mp) joint and at the interphalangeal (ip) joints which can be flexed up to 90° each. The thumb has only two joints, i.e. one base joint allowing a palmar abduction movement of the thumb, and one ip joint rotated by 30° in relation to the base joint, resulting in a thumb plane of motion of 60° in relation to the plane of the palm, which proved to be beneficial [9]. A 34 year old test person with congenital transradial limb deficiency appreciated the low weight of the hand mechanism (140 g) and the flexibility of the digits, but also said that he would not accept the lower weight if it meant wearing a separate pressure supply unit. All actuators were connected to each other in such a way that they were inflated at the same time, resulting in a tangential force of 4 N at each finger tip. Despite the low force, cylindrical objects with a maximum diameter of 120 mm and a weight of approx. 2 kg can be held, as the contact area between the object and the hand is increased by conforming to the shape of an object [13]. However, the maximum grasping force was rated by the test person to be too low. It was restricted largely by the maximum operating pressure of 4 bar. Also, the grasping speed of 3 s was found to be too slow, which can be explained by the quotient of the maximum pump flow rate of 400 ml/s and the volume needed to fill 14 actuators. Finally, the hand was covered with a latex glove which offered reasonable functionality. The cosmetic appearance was rated as unsatisfactory as that of PVC gloves.

SECOND PROTOTYPE

A new prototype design was to fulfill these criteria:

- Integration of the pressure supply unit into the socket of the hand.
- Five different prehension types possible: tripod prehension, cylindrical grasp, lateral prehension, hooks grasp, and index position.
- Increased grasping speed.
- Modularity of the skeletal framework and the joints to allow different hand sizes to de designed and reduce production costs.
- Reduced noise level of the micropump.
- Hardware and software for multifunctional control.

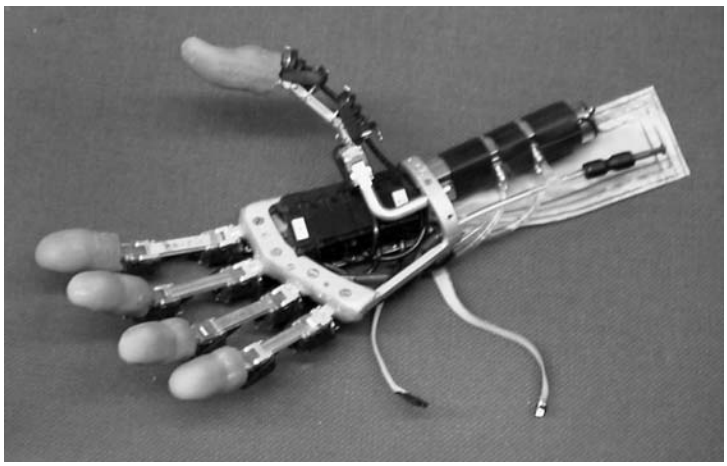

Fig. 2: The second prototype with six valves.

To increase grasping speed, the volume necessary to move the fingers was decreased by reducing the number of joints actively actuated to one ip joint and one mp joint in each finger, and another micropump was added. The metacarpus of the hand was redesigned such that it houses six customized microvalves. They are connected to one to three actuators so that different types of prehension can be performed. As the hand was only designed for lab testing, it does not have a cosmetic glove but elastic finger tips, as proposed by [12]. A simple, rugged two-step control scheme is used to operate the hand. First, depending on the first control signal, the digits move from a neutral state into a predefined preshape state, each representing one prehension pattern. The user receives feedback when the hand performs the desired prehension pattern. Grasping speed was between 1 and 2 seconds [11]. It turned out that the stability of this prototype had to be improved. Moreover, the power consumption of the system was too high. Furthermore, users with a distal transradial amputation cannot be fitted this prosthesis.

THIRD PROTOTYPE

A third prototype was designed to solve the drawbacks of earlier prototypes. One major challenge was the need to fit all requisite components into the small volume of the metacarpus by miniaturizing the components, thus allowing users with long transradial amputation to wear the device. Another consequence was the reduction of the power consumption of the micropump. Moreover, the maximum pressure of the micropump was increased to 6 bar by using a different DC motor and pump design, which resulted in higher holding forces of up to 110 N [14]. The noise level of 60-70 dB of the initial outer gear pump was reduced by 20 dB by using an optimized pump design and adding acoustic insulation. The mass of this hand complete with battery, socket and cosmetic glove is 860 grams.

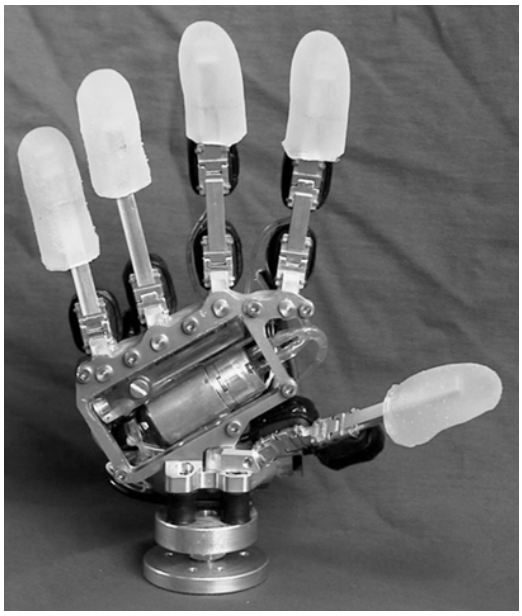

Fig. 3: The third prototype contains all components within the hand.

Kinematic analysis showed that active ip joints can be replaced by passive ones in the ring finger and the little finger without significantly diminishing dexterity [11]. Control of the motor speed was also improved by the integration of a pulse width-modulated signal. The first custom-made cosmetic silicone rubber glove gave the hand a very realistic appearance. Initially, the first glove restricted grasping to objects with a diameter of ≤ 45 mm; a redesigned glove model then allowed an active opening range of 90 mm and a passive range of 120 mm to be achieved. To increase the robustness of the new hand, test rigs for all components were developed, and mechanical loads were simulated by using CATIA-FEM simulation software. For software updating, the microcontroller can be connected to a PC via Bluetooth™ interface for wireless communication. A new real time control concept for grasp-type classification using on-line feature extraction from EMG-signals was developed [15].

The teaching process is based on statistical classifiers, fuzzy rulebases, and artificial neural networks. It also incorporates a routine automatically generating source code for microcontroller implementation. The results obtained from seven upper limb deficient test persons showed that a switch signal can be generated after five minutes of training with a signal classification rate of > 91 percent. An optional sensory feedback system was developed for the prosthesis, which is based on mechanical vibration. It consists of a tactile sensor integrated into the fingertip of the thumb, a controller, and a coin-type DC motor with an integrated eccentric (VM14B-S1, JinLong Machinery Co. Ltd., Yueqing, China). First clinical trials with the feedback system indicated high acceptance and showed the force necessary to hold an object securely to be reduced significantly.

ACKNOWLEDGMENT

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HYBRID AND ALTERNATIVE PROSTHETIC DESIGNS FOR SPORTS AND RECREATION

Robert (Bob) Radocy M.S., TRS Inc.

The pursuit of sports and recreation activities continues to expand in both able bodied and physically challenged populations. Persons with a hand absence(s) may in certain instances have more leisure time than their two handed peers allowing them to participate more frequently in activities like golf, tennis and water sports.

Many individuals with a physical challenge, such as a hand absence, wish to perform their leisure time or sports activities competitively. These individuals require prosthetic technology that allows them to achieve those goals. The type of prosthesis that they utilize for their ADL, work environment and domestic life most likely will not provide the level of technology required for participation in high performance sports and recreation activities.

Sophisticated, externally powered prostheses, although highly functional, are not necessarily the logical choice for these activities. In fact the application of this type of prosthetic technology to rigorous sports and recreation can prove detrimental to the life and function of these prostheses. Similarly, most basic body powered prostheses, although simple, rugged and dependable most likely do not provide the level of function required to perform at optimal levels in sports and recreation activities.

The bio-mechanical complexity and physical demands that many sports and recreation activities place upon an upper limb prostheses are significant. Engineering difficulties arise in the ability of a standard prosthesis to transfer kinetic energy and duplicate the degrees of motion and freedom that exist in natural human anatomy. Two dimensional prosthetic joints simply cannot provide for the smooth transfer of energy that is required for high performance, bi-manual power and control. A clear illustration of these bio-mechanical demands exists in the simple swing of a baseball bat. The transfer of energy from the torso through the arms into the bat requires multiple movements within all the upper extremity joints as well as forearm pronation and supination. The action that is commonly referred to as “wrist break” is actually a complex series of motions involving many degrees of freedom within the forearm and wrist. No standard prosthetic forearm or mechanical wrist system provides the range of motion and degrees of freedom required to duplicate the dynamic “wrist break” required to swing a bat bilaterally with power and control. Attempting to swing a bat with a prosthesis that has limited degrees of freedom typically results in either damage to the prosthetic components and or damage to the tissues of the residual limb.

Solutions to these prosthetic dilemmas exist but they represent a departure from traditional prosthetic technology design in both function and appearance.

The word “hybrid” as currently used in prosthetics refers to a prosthesis constructed with multiple components that represent combinations of both traditional body powered and externally powered technology. Hybrid prostheses as well as unique alternative designs can provide viable solutions in achieving performance in sports and recreation if the prosthetic user is willing to sacrifice certain cosmetic elements in exchange for function and competitive high performance.

The basic myo-electric forearm can be easily modified to receive a more traditionally designed body powered component or custom recreational and sports terminal device equipped with a standard one half inch diameter threaded mounting stud. Both Texas Assistive Devices® and Otto Bock® produce myo to body power converters or couplers.

Three companies manufacture activity-specific terminal devices for sports and recreation. Texas Assistive Devices®, Hosmer Dorrance® and TRS Inc®. TRS Inc. manufactures the broadest array of sports and recreation adapters but these products are complimented by those additional and or alternative products produced by Texas Assistive Devices® and Hosmer®. The variety of sports and recreation adapters has blossomed over the last ten years. Activities from archery to wind surfing are now accessible using either specialized prosthetic accessories or by modifying sports equipment to function for a person with a hand absence.

Proper prosthetic alignment is an important consideration in specific sports to ensure that safety and success are achieved. In sports such as weight lifting or archery, the socket-to-arm alignment and wrist-angle-to- prosthesis centerline are critical in achieving performance. Pre-flexion in a trans-radial prosthesis can be counter productive and inhibit an athlete from handling the heavy loads generated in activities like the "bench press". Similarly pre-flexion makes it difficult to handle the loads and torque generated while drawing a bow to shoot an arrow. A neutrally aligned prosthesis allows for more direct axial load onto the residual limb improving stability without inducing torque that could impact control and accuracy.

The key emphasis in the development of most prosthetic sports accessories has been to encourage bi-manual function and capability rather than to emphasize uni-lateral performance. Bi-manual involvement and performance are valuable to the athlete from a therapeutic perspective, especially over a lifetime of activity.

Radical departures from conventional designs clearly illustrate this attempt to increase bi-manual performance. The arrival and acceptance of the shorty sports prosthesis and its application for both trans-radial and trans-humeral athletes points the way to future developments that can incorporate unique materials and components designed to achieve specific bio-mechanical and energy assistive properties for persons with an upper limb absence. The profession has seen these departures in lower extremity prostheses especially in the realm of custom sprinting and running legs. In upper extremity we are only scratching the surface as to what may be possible.

The shorty sports arm typically utilizes a roll-on silicone type suspension to help eliminate harnessing thereby improving range of motion. It may contain carbon fiber reinforcing for improved strength coupled with weight reduction or it may be constructed of co-polymers for water sports applications. The shorty's primary goal however is to eliminate unnecessary joints, mass and weight. The prosthesis is designed to terminate as closely as possible to the end of the residual limb. The interior lock system of the roll-on liner may dictate some additional length. Either a friction or disconnect style wrist is attached. On the trans-humeral athlete the prosthesis's length without a terminal device should attempt to duplicate the length of the normal humerus. For the short trans-radial athlete, depending upon the activity, the "wrist component" should be positioned not longer than mid-forearm. The prosthesis may also be slightly pre-flexed for particular activities like swimming to enhance ease of use and performance.

Designed in this way the shorty sports arm provides more direct control over the sports accessory and improved proprioception for the user. During an activity like swimming the resistance of a swimming accessory like the TRS Freestyle® swim device creates high torque and stress loads on the residual limb. The shorty prosthesis helps minimize these forces by bringing the loads closer to the end of the residual limb.

Eliminating the prosthetic elbow joint and forearm for the trans-humeral athlete in an activity like golf is proving to be very productive. The entire lower end of the prosthesis is replaced by a long, energy- storing polymer coupling that terminates in a golf grip attachment component.

Several different golf adapters are applicable to his concept including the TRS Golf Grip® and Golf Pro® technologies. The flexible coupling provides numerous degrees of freedom duplicating the necessary movements required to swing a golf club bi-manually with power and control. The shorty sports arm with its silicone suspension and light weight allows for a normal range of unrestricted, gross motor movement throughout the arms and torso. This frees kinetic energy to be transferred through to the club when striking the ball. The end result is improved bi-manual performance, ie. longer drives and a swing that exhibits better control with less effort and more proprioceptive response.

In summary, the design and construction of a hybrid or custom sports arm requires that the prosthetist and athlete communicate closely to determine the capabilities and needs involved. The environmental conditions, stresses and loads that the prosthesis will encounter must be defined. Materials and suspension design will vary depending upon the primary application. An activity-specific prosthesis needs to be constructed to enable the user to duplicate the bio-mechanical elements required by the sport or recreation. Prosthetic alignment can play a valuable role in achieving performance in specific sports activities like weight lifting and archery.

In certain cases an externally powered arm can be modified to perform sports and recreation activities with the addition of a body powered wrist coupler accessory. The prosthetic sports arm may in fact be built for a single activity like golf and incorporate more radical design elements that sacrifice appearance for function and performance, such as the shorty sports arm. Placing the sports accessory as close to the end of the residual limb as possible can typically enhance performance and control of the activity. A wide range of prosthetic sports and recreation accessories exist to help the athlete be functionally bi-manual and competitive with two handed peers. Future prosthetic designs for the upper limb may incorporate technological innovations that have been successful in lower extremity sports prostheses.

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Otto Bock Inc. USA
TRS Inc.

ELECTRIC OPTIONS & SOCKET DESIGN FOR PARTIAL HAND PATIENTS

Stephen Mandacina, CP, FAAOP
Hanger Prosthetics & Orthotics

Introduction

Recent improvements in prosthetics has allowed for a marked increase in success and functional rehabilitation of the partial hand patient. Manufacturers are designing much smaller and lighter components such as electrodes, switches, batteries, and programmable microprocessors that allow a much smaller and simple prosthesis. In most partial hand cases, this allows the socket to not extend proximal to the wrist for full wrist ROM, simplified socket design, and lighter weight prostheses, all leading to greater acceptance.

Improvements are being made with two separate designs for partial hand patients that use an intact thumb to assist with prehension, or a smaller drive unit decreasing space necessary in the hand. Although both designs are progressing, the components are not readily available for most prosthetists fitting a Transmetacarpal/ Transcarpal level amputation. The focus of the paper and presentation is to educate on options currently available and easily used by a majority of the prosthetic field.

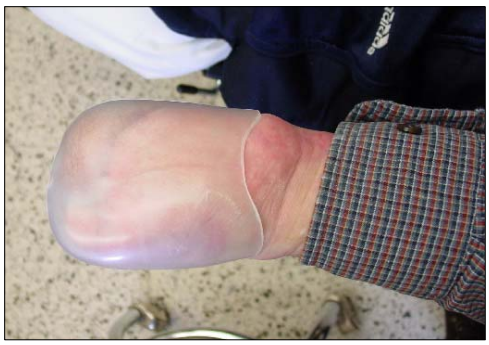
History

In the past, partial hand levels incorporating electronic control raised a complication with two main issues: component space & placement, and limiting range of motion at the wrist. For an adequate suspension with laminate plastic, the socket needed to come proximal to the wrist joint. Coming proximal to the wrist warranted external batteries to be placed alongside the forearm for cosmetics. By locking the wrist with this design and extending the socket, wrist flexors and extensors were excellent placement for myoelectric control of the terminal device.

Advancements

Socket Design

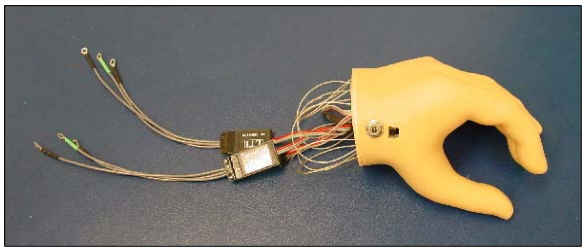
Sockets have changed in years to a more aggressive fit allowed by flexible sockets permitting an increased range of motion and improved suspension without going beyond the wrist joint. Flexible plastics such as Northvane, Bioelastic, and Proflex allow the patient to easily don the device and maintain necessary suspension. Oftentimes, a lubricant such as silicone gel eases the donning of the device. For heavy-duty tasks or to increase the suspension, a small Velcro strap can be attached

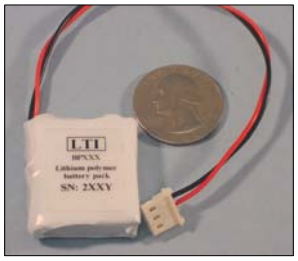


just proximal to the wrist. This is not necessary in most situations, but does allow for a stable skeletal lock of the prosthesis on the hand.

Electronics

Improvements in electronics have also increased functionality of partial hand electric systems. Internal batteries have allowed a much smaller frame built in the prosthesis, thus improving cosmetics. Some batteries can be placed inside the hand shell of the terminal device completely eliminating any bulge in the frame. Although not for an active adult user, Li-Polymer batteries by Liberating Technologies, Inc provide the smallest dimensions and lightest option readily available and are recommended for light to moderately active users.





Smaller size electrodes coupled with smaller preamplifiers have also allowed for a much smaller device at the residual hand. Because electrodes must maintain contact on the skin, there's a greater acceptance of the remote preamp-electrode versus the standard electrode. Depending on the density of the soft tissue, the remote electrode is replaceable and easily maintains skin contact on the hand. However, if there's a considerable amount of movement of the skin in the socket, as there is sometimes in Transcarpal levels depending on the weight of the object they are lifting, a switch or touch pad is recommended. All of these components are readily available to the prosthetist.

Terminal Devices

Electric hands for partial hand patients have also improved and are readily available for this clientele. The hands available for this level by Otto Bock and Motion Control are much shorter and lighter than their standard counterpart. These hands weigh in about 1/3 as much as the larger version and save approximately 1 ¼". With hybridization of manufacturers, using the Animated Control System increases the speed of the hand up to 380mm/sec and the grip force up to 90-100N. Both of these are adjustable if this is too much for the patient.

The hands have the ability to be laminated directly to the frame of the prosthesis, or can be attached to a quick disconnect unit to interchange TDs if the patient needs other tools than just the hand.

Conclusion

All of these components allow the Transmetacarpal/Transcarpal patient to be successfully fit with electric systems without compromising the cosmesis of the device.

Improvements in socket design and socket material using these new components provide a comfortable device without limiting the necessary range of motion that patients find valuable.

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EXPERIENCE WITH ELECTRIC PROSTHESES FOR PARTIAL HAND PRESENTATION WITH THE THUMB INTACT

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ABSTRACT

Limb deficiency distal to the wrist represents a common and difficult level to treat with a functional prosthesis. Challenges include functional limitations of prosthetic technology, prosthetic interface discomfort, cosmetics and absence of tactile sensation. Until the late 1990s, the lack of acceptable electric prosthetic options as well as concise treatment parameters have limited prosthetic treatment. A new electric partial hand prosthesis design has been developed expanding on the earlier work of Biden and Bush at the University of New Brunswick in the late 1990s.

An externally-powered prosthesis was designed for the individual presenting with partial hand deficiency with the thumb intact. The electric partial hand was preferred as a more functional option. Since the electric partial hand did not require contralateral hand involvement to preposition the fingers and allows active movement of the first and second digits against the thumb, it was observed to allow more spontaneous function.

MANAGEMENT OF THE VERY SHORT/HUMERAL NECK TRANSHUMERAL AMPUTEE

Jack E. Uellendahl, CPO Hanger Upper Extremity Prosthetics Program

Prosthesis control for persons with very short (axilla level) transhumeral (VSTH) amputation can be challenging. Due to the short lever arm, fully body-powered options are often not possible due to the lack of available excursion. In some cases it is possible to use ballistic control of an elbow with Automatic Forearm Balance yet terminal device control is still problematic. When an electric hand is desired, myoelectric control is preferred.[4] However, consistent contact of the electrode(s) using a current state-of-the-art TH socket design may not be possible.

Roll-on Silicone Suction Sockets (3S) using snap electrodes have been employed for the VSTH interface with limited success. Due to the absence of a limb projection, the 3S socket often cannot maintain suspension and electrode contact is lost. When the liner is mechanically connected to the prosthesis though a locking liner, motion of the prosthesis tends to work the liner off of the limb. Even when a non-locking cushion liner is used, sweat and residual limb motion can cause the liner to loose position on the limb and thereby compromise electrode position.

Another alternative is to use a linear transducer using scapular motion for control with a TH socket. Because this type of control is body position sensitive, it can be difficult to separate the control motion from other body motions such as reaching forward, which causes inadvertent hand function.

Shoulder disarticulation socket designs provide an alternative but also present problems. A thoracic frame type interface can be designed to provide good electrode contact. However, treating the VSTH as a shoulder disarticulation complicates the prosthesis due to the need for a prosthetic shoulder joint. Location of the shoulder joint is problematic because the physiological shoulder joint is present and there is no space for a prosthetic joint to be positioned in a natural and cosmetic location. Also patients fitted in this way often feel overly encumbered by the more extensive socket and tend to prefer a less inclusive socket if appropriate function and normal appearance can be attained.

It is the author's opinion that this patient group stands to benefit greatly from emerging technology and surgical interventions using implantable electrodes[1] with or without neuromuscular reorganization as proposed by Kuiken.[2,3] If myoelectric signal acquisition can be accomplished with implanted electrodes and control signals are wirelessly sent to the microprocessor controller, requirements for prosthetic socket fixation may be reduced. Neuromuscular reorganization would allow for more natural hand control using signals originating at the physiologically appropriate nerves. Normally innervated biceps and triceps can then be used for myoelectric elbow control or the elbow can be cable controlled.

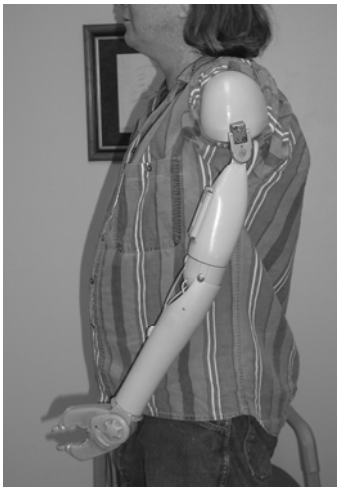
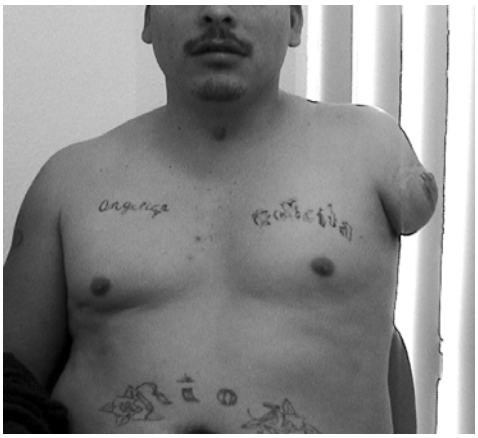


Figure 1



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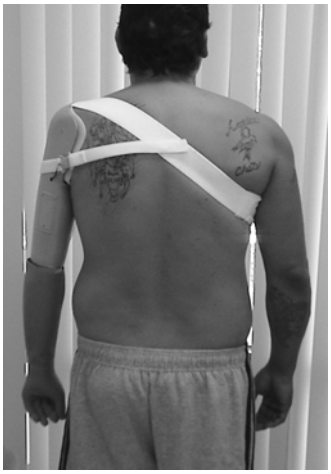

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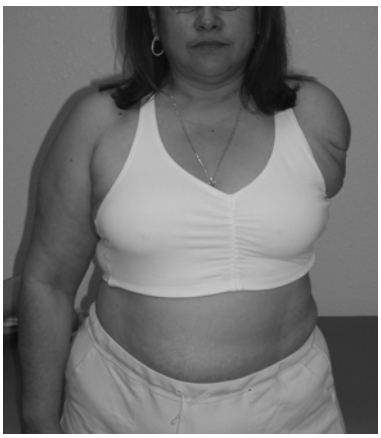

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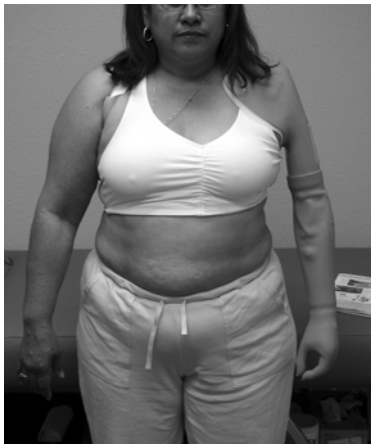

Figure 5

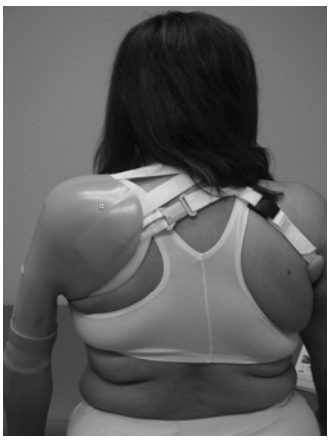

Figure 6

Figure 1 shows a VSTH amputee (DD) fitted with a thoracic frame socket using 2 site myoelectric control of his terminal device and a passively positioned elbow.

Figure 2 shows a VSTH amputee (RR) without his prosthesis.

Figure 3 shows RR fitted with his prosthesis using linear transducer for control of his Sensor hand and ballistic control of his AFB elbow. Chest expansion operates the elbow lock using this thoracic harness system.

Figure 4 shows a VSTH amputee (JS) without her prosthesis.

Figure 5 shows JS with her prosthesis. In order to provide a more cosmetic result a figure 8 harness is used for suspension avoiding visible straps when wearing open neck blouses.

Figure 6 shows JS from the back. A linear transducer provides control of the Sensor Speed hand. The elbow is passively positioned.

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MULTIPLE IMPAIRMENTS OVERCOME BY SOCKET DESIGN & APPROPRIATE COMPONENTS

Stephen Mandacina, CP, FAAOP
Gary Lenzini, CPO

A Case Study

Introduction

The success of prosthetic rehabilitation involves a number of facets including optimum therapy, appropriate components, interdisciplinary network, patient motivation, and a good fitting socket. All of these facets get multiplied in severity when more than one ailment is presented in the patient.

This case study will discuss the socket design principles and components that allow for vocational return for a hip disarticulation-shoulder disarticulation patient who is also blind. The design of the myoelectric shoulder disarticulation prosthesis had to be lightweight and provide as much feedback as possible to control the elbow, terminal device, and wrist rotator—as well as aid in balance while ambulating with a sight cane.

History

At the time of fitting, the patient was a 41-year-old male who has a right humeral-neck level shoulder disarticulation amputation secondary to a motorcycle accident. He is a healthy individual who is involved in a variety of activities where superior function and a positive self-image are very important. Prior to his amputation, he was employed as an Aircraft Mechanic. Since then, his job has changed to a Telephone Operator/Scheduler in order to accommodate his lack of dexterity and bimanual function. He is employed at a local company, where his job duties include answering phones, taking messages and scheduling appointments using a computer. Aside from his independence in ADLs and maintaining his home, his hobbies include customizing show cars and attending the car shows, as well as fishing. For all of his daily activities, he utilizes an upper extremity prosthesis for his missing right arm. The utilization is important for balance while walking, as well as carrying any object for he uses a sight cane in his sound hand.

As a result of the motorcycle accident, he lost his vision do to an optic nerve injury. He also sustained a right hip disarticulation and has been fitted with a state of the art hip disarticulation prosthesis with a microprocessor knee unit. The blindness causes unsteadiness in his ability to walk. Due to his intense motivation to progress, he has successfully learned how to use the microprocessor knee prosthesis and has been able to discontinue the use of his wheelchair and return to ambulating, and an active and productive lifestyle. This is typically not seen in hip disarticulation patients alone, especially when adding ailments such as blindness and a shoulder disarticulation amputation. Because of his vision complications, his left hand gets overcompensated assisting in sensory feedback. Without an upper limb prosthesis, his balance is sacrificed and he won't be able to carry anything.

He currently has a cable-operated prosthesis that was over three years old. Although this prosthesis is worn consistently, the functional outcome is not sufficient and this prosthetic design is creating further problems. Evaluation of his current cable operated prosthesis revealed a poor socket fit resulting in discomfort as well as poor control. Control is also compromised due to a limited Gleno-Humeral motion. All control is obtained by Scapular-Thoracic motion. The harness is causing axilla pressure and discomfort. His residual shoulder has no lever arm, which decreases his range of motion and limits his ability to control a cable-operated device.

Fitting Process

The first part of the fitting process was a thorough evaluation of the patient, patient lifestyle, and his goals. An electric system was indicated due to the lack of G-H motion and small amount of S-T motion. EMG signals were sub par on deltoids, but more than sufficient EMG was found with good separation on Pectoralis and inferior Trapezius. Oftentimes, the inferior Trapezius can cause unwanted signal while ambulating or movement while seated, in which case superior Trapezius with shoulder elevation is an excellent option. With the EMG results and the need for better control of the device and increased grip force, the patient decided to forgo the kinesthesia and proprioceptive feedback obtained with the cable-operated system and opt for a myoelectric system.

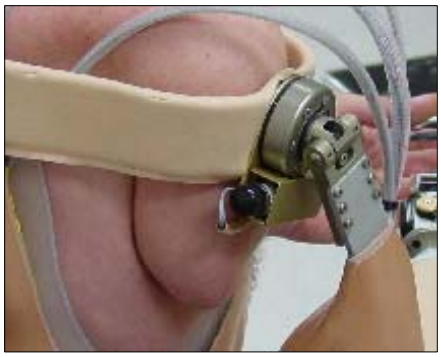
A Utah3 electric system was indicated for two reason: 1) Usage of a wrist rotator without incorporating an additional switch. The rate of contracting a single input muscle will operate either the hand or the wrist and allow cocontraction for unlocking the elbow. An advantage of this setup is that it does not incorporate any external switch. 2) Because of the vision complication, the patient and rehab team considered that the autograsp feature of the Sensor Speed hand would be advantageous. Also, the patient wanted the ETD. The Utah3's Auto-Detect feature permits the patient to use either of these devices by a detection circuit that allows an in-hand controller to be used on the elbow system.

The patient was also given the option of a third terminal device. A Motion Control hand with a flexion unit was chosen to assist with midline activities. This hand was eventually the hand of choice for the patient for two reasons. The first reason was the ability to hold onto objects closer to midline by using the flexion unit. The second reason was that the sound and speed of Motion Control hand were more similar to the ETD; the patient had a more consistent knowledge of opening and closing between the two devices.

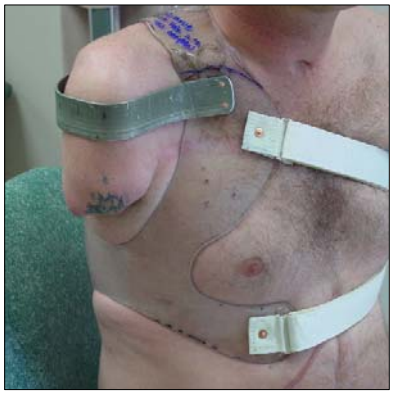
An electric wrist was chosen using Fast Access to eliminate the need of additional input or toggle switches. Cocontraction, therefore, would be specific to elbow unlocking. All of these controls are easy for the prosthetist to adjust, and visually friendly via a computer.

An older version of the LTI shoulder joint was used because the desire to incorporate electronic locking and unlocking. Although the electronic shoulder lock is typically used for bilateral applications, it was deemed worthy in this case. Many

sensory inputs are performed with the sound left hand; therefore, the team didn't want to remove the hand from any task to lock/unlock the shoulder.

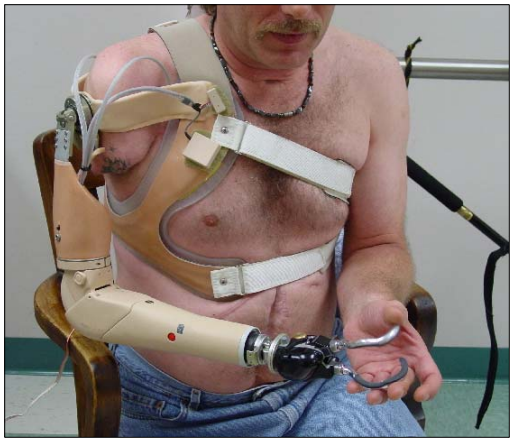


The socket design was critical to achieve a good anatomical purchase to transmit as much proprioception and prosthesis knowledge as possible to the patient. To ensure an accurate and identical mold was achieved, a laser scanning process with Insignia was performed. With very little modifying to the carved mold received from the CAD/CAM method, a snug fit was quickly received. Eventually, a thin, corrugated frame with Carbon over a thin flexible inner liner socket was made that gave the patient the necessary feedback provided by an electric system. The design incorporates a good Delto-pectoral purchase countered with posterior support on the Scapula. Adequate relief on the scapular spine allows the very slight motion for control of the system to not cause discomfort. A flexible strap was placed over the shoulder for comfort providing minimal suspension assistance. Although minimal, the socket was able to provide feedback of a real-time spatial recognition of the entire system.



Conclusion

The shoulder disarticulation socket design coupled with the new Utah3 elbow with Auto-Detect and Fast Access wrist control has allowed the patient to return to work and able to carry boxes and papers while he walks. He is able to interchange easily between the ETD, Motion Control Hand, and Sensor Speed Hand, while the onboard computer can detect which type of TD is attached and makes the necessary microprocessor control it. Additionally, the Fast Access to wrist allows him the best control of the electric wrist without complicating the device with switches.



This case study exemplifies the need for good OT, appropriate component selection, as well as the need for a good fitting socket on such a high level, multi-limb complex situation.

ASSESSEMENTS, CONCIDERATIONS AND FITTING OF A TRANS HUMERAL HIGH LEVEL BRACHIAL PLEXUS INJURED INDIVIDUAL WITH HO

JW 'Bill' Limehouse, CP FAAOP
Hanger Upper Extremity Prosthetics Program

This talk will discuss the assessment and prosthetic problems associated with fitting an individual with a mid level Trans Humeral amputation.

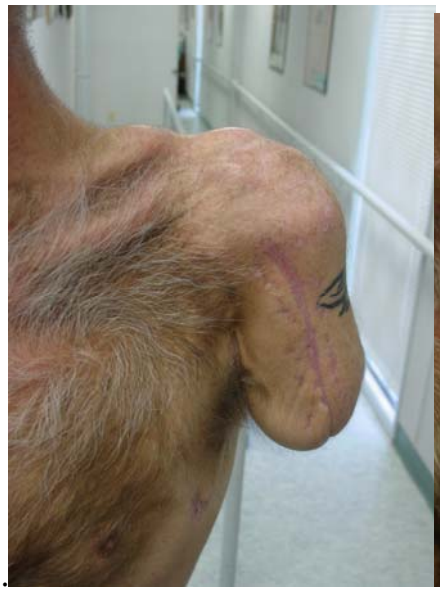
TM was a very active healthy 40 yo male, RN Critical care nurse and amateur Triathlete. He does free diving, fishing, canoeing, kayaking and lots of other activities in and outside. He was riding to work one morning on his motorcycle and as a result of someone running a light he 't' boned a car. The result was a fractured femur, dislocated knee, shoulder separation with Brachial plexus avulsion, crushed forearm and slight head injury and retina detachment. He also had several ribs that were broken and a punctured lung.

His hospital stay involved a left mid level Trans Humeral amputation. He also had a rod inserted into the fractured femur. During the course of stay at the hospital he started developing Hyper Ossification of the shoulder, neck, ribs and chest. This HO had encompassed the ribcage and would not allow him to take a deep breath. He was being restricted on his movement even to help get in and out of bed. Approximately 3 months post injury he was discharged with a wheel chair to home. He had contacted our local office to see what could possibly be done prosthetically. He was very disappointed in not being able to do much of anything.

The evaluation included what were his expectations and goals. He would like to do some of his previous activities and would like to have at least some elbow and 'hand' grip. Since he was so limited with his upper movement he would like to have an assist to do things around home.
Physical examination revealed he had a severe involvement of HO of the upper torso, neck and left shoulder. The patient brought in a copy of his 3D CT scan. This showed all of the HO involvement along the chest, spine, shoulder and neck. There was a lot of associated pain in and around the shoulder, neck and torso.

Since there was essentially no functional ROM of the humerus and scapula, myoelectric control was the first consideration. The only functional myo sites were from Rhomboids and Trapezius muscles. An SD type prosthesis was fabricated designed to accommodate the remnant humerus with cutouts to allow the humerus to side in during donning and doffing. A shoulder joint and Utah 3 elbow with powered wrist and terminal devices were provided.

Successful operation of the prosthesis was noted early on and the pt was capable of fine control of the system. After approximately 3 weeks, his pain had been ongoing and a visit to his pain specialist resulted in a couple of Botox injections around the medial spine. This resulted in complete loss of the Rhomboid muscle contractions for a period of time. He later had an implantable electric stim to reduce pain. This was suggested to be at a different location as to not interfere with the myo sites



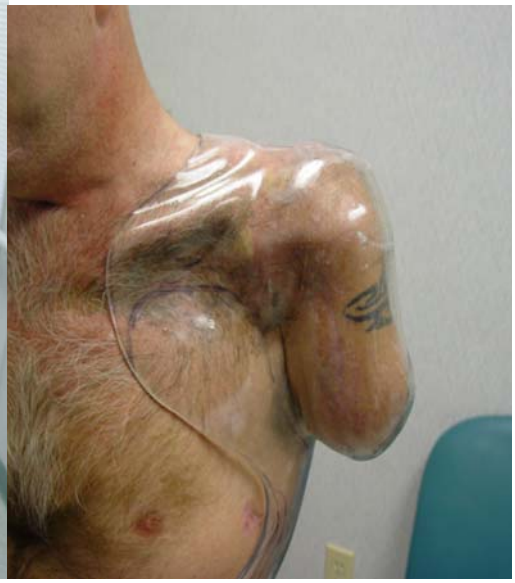
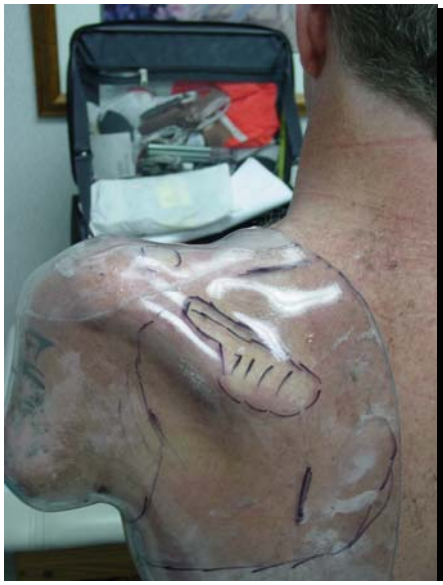


Figure 1 Figure 2



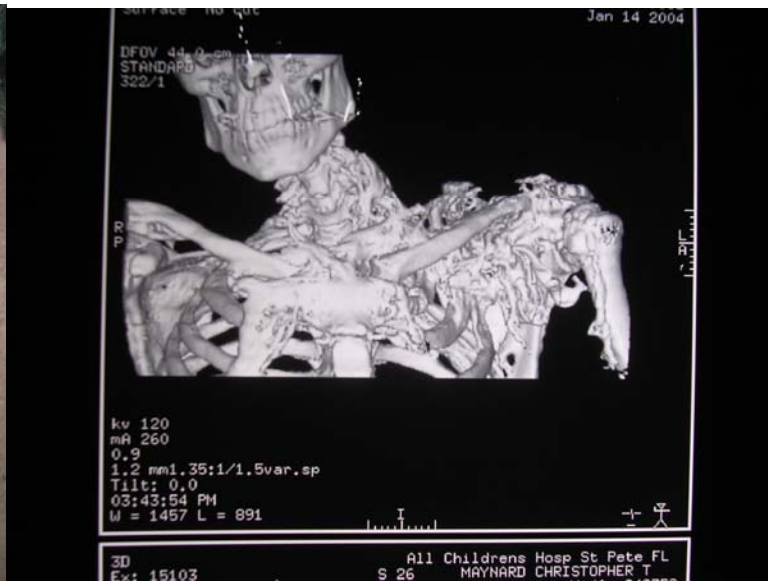


Figure 3

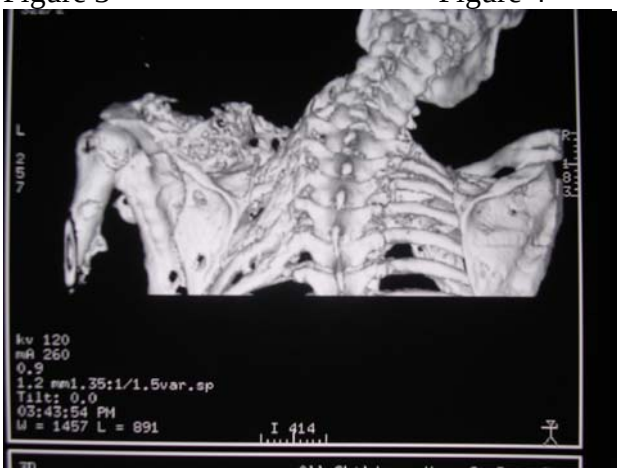


Figure 4

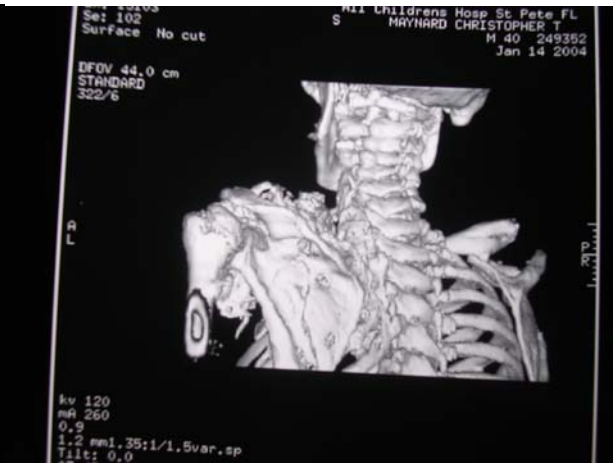


Figure5

Figure 6

Figure 1 left anterior anatomical view

Figure 2 same view test socket

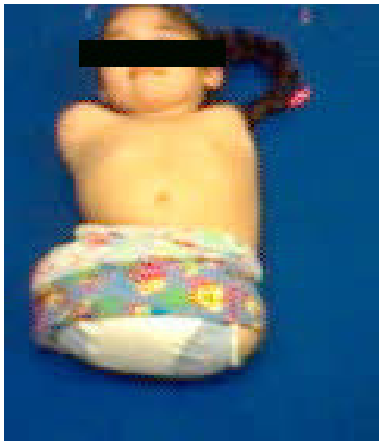
Figure 4 same view 3D CT scan

Figure 3 left posterior view test socket

Figure 5-6 posterior 3D CT scans

Externally Powered Management of the Quadramembral Amputee Using a Modified Thoracic Suspension Orthoses as a Platform

John R Zenie MBA, CPO



Client @ initial valuation

The female client was initially evaluated, at a regional rehabilitation center, specializing in early development and adolescent reintegration of children to maximum mainstream capacity. The initial presentation of this seven-year-old female multilevel congenital amputee was in the presence of the therapeutic rehabilitation team, the child presented with independent mobility in an externally powered wheelchair. Direction and speed of the wheelchair are controlled, utilizing a proportional multidirectional chin switch mounted on the right upper corner of the powered chair. The clients' body position is maintained using custom fabricated seating interface system complete with bolsters. At the time of initial evaluation a platform based thoracic suspension orthoses was being fabricated. The design was to incorporate an erect posture when outside the chair. Removal of the bolsters from the seating system allowed the orthoses to be utilized in the chair. The objective of prosthetic intervention was to provide functional grasp and release as well as positioning of the terminal device in space. In an attempt to simplify and increase the utilization of a prosthetic device the decision was made to utilize the exterior wall of the suspension orthoses as the platform to incorporate an externally powered arm. Physical evaluation finds the patient's anatomy compatible with a frame type, shoulder or scapular thoracic prosthetic socket. This alternative was rejected on the basis that it would require removal of the positioning orthoses for utilization. The positioning orthoses is modified proximally on the left upper margins so as to allow increased scapular and shoulder girdle movement. The left quadrant of the body was selected for prosthetic management so as not to interfere with the existing chin switch control of the powered wheelchair. It was believed that the ability to have independent mobility and grasp /release was a prerequisite for the patient to achieve maximal independence.

Evaluation of the patient's physical anatomy found the range of motion about the shoulder girdle to be within normal limits. Electromyographic testing finds major muscle groups that are intact produced sufficient signal to control an externally powered prosthesis. However, signal consistency, and differentiation of individual muscle groups provided cause to believe that myoelectric control may not prove viable with this particular client. An evaluation of the clients' ability to target specific areas on a frame superstructure provided encouraging results and prompted further exploration of force sensitive resistors orientated in a highly specific array to gain control of multiple output devices. The devices considered initially for the system included a proportionally controlled terminal hand device, a wrist rotator, and an electric elbow. The electric wrist rotator is also fitted with a constant friction multidirectional wrist flexion unit. Proximal orientation of the prosthetic extremity was to be achieved utilizing a constant friction shoulder joint that permitted flexion and extension as well as controlled abduction. Particular attention is paid to the orientation of the shoulder joint so as to maximize functional tasks near the face with the elbow in maximal flexion and the wrist in normal pronation.

The initial production of the frame type superstructure began with a direct casting of the client's residual anatomy in neutral orientation in an erect position within the modified suspension jacket. This casting technique allowed for the trimlines of the suspension orthoses to be captured in the model. The initial fitting was conducted using a low temperature clear thermoplastic to permit precise target location for the proposed force sensitive resistors. The anterior F. S. R. was conceived as the input to close the terminal device, initiate pronation and provide elbow flexion. The posterior FSR was conceived as the input device to open the terminal device, supinate the wrist and extend the elbow. A small rocker switch was mounted on the superior anterior margin of the suspension jacket as a mode selector between hand and wrist. Through initial trials and experimentation precise placement of the rocker switch for hand/ wrist mode selection was determined and consistently operated via a brief momentary contact with the chin. Additional experience with the system found that the number of output devices became difficult for the client to operate consistently. As a result of this experience in additional F.S.R. type switch was mounted directly superior to the residual anatomy. The client had difficulty targeting the relatively ambiguous input device. As a result the input was replaced with a second rocker style switch. Utilizing only one pole on this switch momentary contact was used for elbow flexion/extension control. The microprocessor controllers utilized allowed for refinement of this strategy so that contact with the active pole produced elbow movement for as long as contact was maintained. Any interruption in this contact resulted in reversing previous elbow drive pattern and direction. This methodology of control in the elbow sacrifices proportional control of the elbow speed in favor of consistent operation. It is believed that this compromise is favorable in the contexts of functional activities. Further, it is believed that proportional output at the terminal device is significantly more important. Therefore utilizing two FSR inputs and two rocker type contact switches three output devices are controlled effectively. Once this strategy was programmed the client was able to demonstrate within a very short period of time prepositioning of the components in space in order to favorably approach

small objects of varying shape, size and durometer. These functional activities were demonstrated both while erect utilizing the standing base of the thoracic suspension orthoses, and while positioned in her electric wheelchair.

Internal lithium cells provide adequate power on a single charge for greater than one full day of use. This cell configuration allows for easy charging when the client retires to bed in the evening. The therapeutic staff has increased the complexity of tasks, progressively as the client has mastered each series of objectives. The single most exciting, and probably most significant task has been for the client to extend the elbow, rotate the wrist to an appropriate position and grasp a food product. Then reversing the sequence and consuming the food unassisted has been demonstrated. Upon achieving this goal the principal objective of independent mobility utilizing chin control of a powered wheelchair and the ability to manipulate objects in space unassisted has been achieved. The client to this day continues to utilize the system in excess of eight hours each day without significant incident or difficulty. Of particular interest is the increasing effectiveness client demonstrates in preplanning and executing ADL tasks. The clinical staffs unfounded concerns that the “novelty” of the system would wear off, resulting in an underutilized device has not been realized. To this end it was very encouraging that the child is now in fact requesting an upper extremity device for the contralateral side. As of this date, this request is being considered on its merits, and the implications and possible negative impacts of compromising the existing control strategy for the powered wheelchair are being evaluated.

Device for preparatory fitting

VOICE RECOGNITION FOR PROSTHETIC CONTROL CASE STUDY

Kevin Towers, CPO – POSI
Kevin Barnes, CP – POSI
Craig Wallace -- LTI

Client - 48-year-old make, height is 5'7" and is 135lbs, shoulder disarticulation patient. Employed as a heavy equipment operator until August 2000. Client sustained traumatic brain injury and subsequent loss of motor control on left upper extremity (flail arm), with effect on left lower extremity through quad weakness and drop foot. Client is posturally effected with incomplete hemiparisis and also exhibits minor speech impairment.

The client stated that he had numerous falls that lead to multiple dislocations and chronic pain in left upper extremity. The client elected to amputate the left upper extremity in April 2004 to assist with pain management and postural consideration.


Initial Consult

The client expressed desires to regain some of his independence in his personal living. His expressed needs ranged from independent donning of his AFO, dressing, meal preparation, to some minor home improvement projects. His current daily arrangements included extensive assistance from his wife and family members for his activities of daily living (ADL).

The client contacted our clinic looking for upper extremity prosthetic information. He was interested in his prosthetic options to help with his ADLs and limited functional envelope. He has received prior speech, physical, and occupational therapies to promote a restored independent life and exhibits normal cognitive abilities.

Myo signal testing on the effected side did not provide a consistent usable signal source due to the paralysis of the thoracic region. Current voluntary myo or switch control strategies would depend on inputs from the effected side which are limited or non-existent. Involving inputs from the contralateral side could potentially reduce the current level of function on that side and diminish ability for consistent prosthetic control. With these challenges, a call was made to Liberating Technologies to seek out ideas for possible control strategies. Craig Wallace with LTI made mention of a prototype mode selection device that was voice activated. The voice recognition device was a new direction that had not been field tested thus far. This concept offered possibilities for ease of inputs and deliberate mode selection without relying on contralateral inputs to meet the goal of bi-manual assistance.

INITIAL SETUP

The concept of control strategies in the first stage include the currently available Boston Elbow 3, mode selection program, a miniature microphone, voice recognition module, and 2 touch pads for device directional control. The touch pads and microphone were mounted as a triangulated keyboard to allow control inputs from the patients chin. The close proximity of the microphone to the mouth would help keep inputs more direct and controlled. (Figure 1)

Figure 1

The voice commands were initially set up as a direct device command system. The voice recognition module measured 2.5" wide, 4.75"long, and 2.0 deep. (Figure 2) This module was trainable to recognize the patients' voice patterns for direct mode selection. Each device was assigned an input number and a corresponding command word. The device command word has the flexibility of being a word that can be given by the patient consistently. These included; hand, wrist, elbow, shoulder, and included other outputs as needed. These inputs could be further developed for pre-positioning of the terminal device, elbow or wrist as desired by client.

Giving a voice command for the device required would initiate prosthetic control. The command module, upon acceptance of the command, will verbally state confirmation of the selected device with a mounted speaker. (Initially the default was set to the previous device, due to limited module/elbow interface development.) Once the selected device was engaged, the patient would use his chin to nudge the proportional touch pads, and control its' desired movements.

In the clinic this control appeared to work well. The commands were received, the desired device would be activated, and control of the device worked reasonably well considering the amount of further practice needed and the effect of the client's hemi pareses.

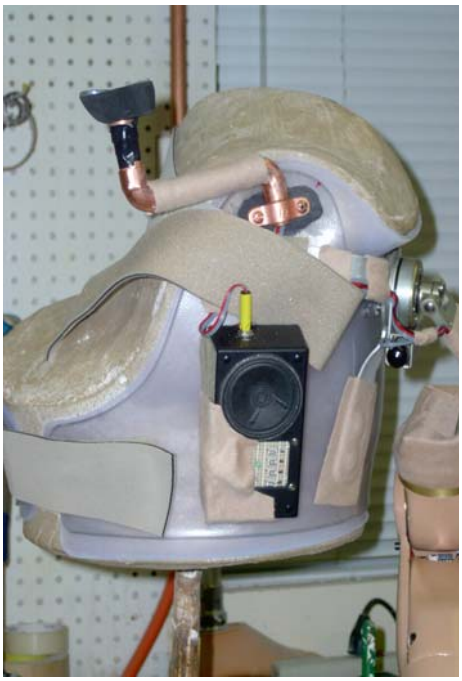


Figure 2

In the real outdoor world, the voice command module showed how receptive it was to "outside" commands. Although only a passenger, the client's radio was able to provide the needed frequency patterns to erroneously give active commands to the clients' prosthesis, which proved frustrating and unacceptable for long-term assistance.

SECOND SETUP

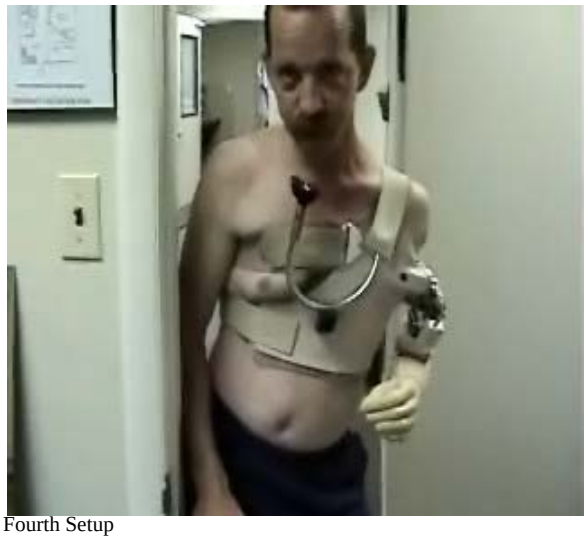
The second level of development in control strategy included the voice command module using two words to activate it. The first word acts as a key to "unlock" or wake up the system, and then is followed by the device command word. This added level of command input has appeared to reduce erroneous operation, but has become slightly more cumbersome and time consuming for the client. After a couple of weeks of use in this configuration the system stopped working consistently, and more development time in the interface between the voice command and BE3 software became evident.

THIRD SETUP

The third change to the control strategy has been the temporary elimination of the voice recognition module while further communication development is pursued. The current control strategy has a proven track record which will allow for longer field trials to allow the patient to build on successes. The hand is now the constant default device, and this arrangement allows the first response of inputs to be simple, and of immediate assistance using the myo hand or ETD. If further positioning is needed, mode selection is obtained through a third touch pad by chin input using rotation mode selection.

Further development goals include using direct device mode selection, pre-selected positioning, with default to a particular group of motors, and a comprehensive visual interface program for clinical facility fine-tuning.

FOURTH SETUP



Fourth Setup

The newest (and current) fourth control strategy now includes the previous "unlock" control with direct mode control to each function with the hand currently set to default. Through the use of the new computer interface, the motor functions can be altered to suit the desired control speeds and directions, and allows the system more flexibility in the field.

The client currently uses the prosthesis approximately 4 hours daily at home practicing integration into his ADLs and is pleased with the potential assistance it can provide. Ken has been provided a donning tree for independent donning and doffing, but is not independent as he is still using his wifes' assistance for donning.

The client has been encouraged to work with an Occupational Therapist for further training in prosthetic use and techniques. To date our feedback and interaction with the clients' OT has been minimal as experience with upper extremity prosthetics is limited.

Current Configuration

CONCLUSION

The clients' feed back is positive to its potential. Our clinical assessment is a work in progress. We will continue to evolve the current socket and frame design to minimize the socket trim lines and improve independent donning and doffing. The client would benefit from an experienced therapist to integrate his usage into his ADL's and provide us with additional feedback to refine design parameters.

The practical application of voice recognition mode select could after practitioners alternative control strategies to unique clients. It has great potential to preposition and select default devices for 3-4 device control designs. We look forward to continued development in the following months.

Biofeedback:
An Enhancement of Amputee Rehabilitation

Dennis Sennett (PT)[1], Peter Ferris (CPO)[2], Kathy Calvin (OT)[2]

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Introduction

The drive for continuous improvement led the ADP to recognize the need to develop a more in-depth myoelectric training program.

A physiotherapist with specialized training in biofeedback was consulted on two cases.

Case 1 – was a short above elbow amputee

- The client had reported having a cold limb, leading to phantom pain, which he set as a priority for our intervention.

Intervention

- Problem of reported cold amputated limb confirmed through objective assessment,
- The client was introduced to autogenic relaxation enhanced by use of biofeedback.
- Successfully learned to control the physiology of the sound limb re- warming and relaxation, feedback was transferred to the affected side.
- After several sessions working on the affected side he gained control and was able to achieve his goal of warming and relaxing the amputated limb.
- Expanded the use of biofeedback to simulate use of his myoelectric limb

Case 2 – above elbow amputation just proximal to the epicondyles

- Primary problems related to determining optimum electrode placement and poor signal control due to co-contractions.

- Our program consulted with Dennis to determine the benefit of using biofeedback to select electrode sites, develop and train the appropriate muscle groups.

Intervention

- The triceps and biceps were evaluated and optimum electrode placements were determined.
- The client was then taught the technique of contracting one muscle group while minimizing the contraction of the opposing muscle group.
- Initially he was asked to visualize opening and closing of the hand, then a target SEMG level was chosen alternating from biceps to triceps.
- Progressed to practicing with residual limb held in various functional positions- simulating reaching above the shoulders, to the side, and working in midline at waist height

Conclusions

- Benefits came from focusing on the condition of the residual limb; problems such as cold limb, muscle tension and phantom limb pain were resolved through the combination of biofeedback and autogenic relaxation.

- Benefits came from focusing on issues related to preparation for myoelectric prosthesis provision such as, electrode site location, reduction of co-contraction, and improved muscle endurance.

Discussion

- Changes were made to our approach based on these two cases
    - Assessment equipment options improved
    - The use of current technology was enhanced
    - The prosthetic hardware options have improved
    - Maximized options for clients on an individual basis through extensive training

A COMPARISON OF UPPER LIMB AMPUTEES AND PATIENTS WITH UPPER LIMB INJURIES USING THE DISABILITY OF THE ARM, SHOULDER AND HAND (DASH).

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Introduction

Amputation of a limb represents a catastrophe for the adult amputee and their family. However data regarding the prosthetic, functional and psychological outcomes achieved by these patients is limited.

There is no well recognised outcome measure used throughout Australia, UK or America especially for aquired adult amputations of the upper limb. Most scales currently in use fail to identify the psychological adjustment problems which many of these amputees demonstrate. Many adult amputees continue to report significant behavioural limitations and discomfort, associated with low self-esteem, anxiety and depression when compared with an able-bodied control sample.[1]

Phantom limb phenomena is also a well recognised problem for many upper limb amputees . Scales used to measure prosthetic use rarely investigate the impact pain has on the amputee and his or her well being.

The Disability of the Arm Shoulder and Hand Scale (DASH) is an evaluative outcome measure for patients with upper extremity musculoskeletal conditions designed in the mid 1990s by the American Academy of Orthopaedic Surgeons and Toronto Institute for Work and Health. [2,3,4] It is a region specific questionnaire as opposed to diagnosis specific questionnaire. It measures function & symptoms of musculo-skeletal disorders in upper limb.

The DASH was able to demonstrate change in all situations in which change was presumed to have occurred. The DASH was found to have comparable responsiveness to the joint specific measures. It demonstrated suitable levels of sensitivity and specificity.

Method:

In 2004 I had a paper published by Disability and Rehabilitation reporting the outcomes of a cohort of 275 patients between 1998 and 2003.[5]
.

Results:

The diagnoses, number of respondents, average age and the compensability of the patients is recorded in Table 1 along with the average DASH score.

Table 1: The total cohort of patients completing the DASH

Diagnosis	No. patients	Ave. age	Mean Dash Score	Standard Deviation	No. compensable	No. non compensable
Quadruple amputations	4	33	67	18	0	4
Bilateral upper limb amputations	2	42	68	5	2	0
Major unilateral upper limb amputation	39	42	39	20	19	20
Partial hand amputation	26	38	49	22	16	10
Amputation following brachial plexus injury	8	35	37	16	5	3
Brachial plexus	10	33	65	15	8	2
CRPS	42	38	68	19	40	2
Arm pain	21	41	50	22	15	6
Carpal tunnel	13	56	40	21	4	9
Fracture	4	43	30	23	3	1
Arthritis	10	47	37	14	5	5
Hand Injury	6	44	53	19	5	1
Radiating from neck	16	40	58	16	16	1
Shoulder	37	46	51	17	33	4
Tendonitis	25	42	51	22	18	7
Wrist	14	40	49	17	10	4
Total	**274**	**41**	**51**	**21**	**195**	**79**

There are 75 upper limb amputees in the cohort and 26 of these have had partial hand or digit amputations. The remaining 200 patients have suffered a variety of diagnoses including Complex Regional Pain Syndrome, Shoulder Capsulitis, Carpal Tunnell Syndrome, and Brachial Plexus injuries.

The average DASH score for the whole cohort was 51/100. The highest average DASH scores were for bilateral amputations (68/100) and quadruple amputations (67/100) and patients with Complex Regional Pain Syndrome (68/100). The lowest average score was 30/100 for fractures. These diagnostic differences are significant ($f = 4.809$ $p = .000$).

There are major differences within the scores by both diagnosis and compensability. Of the total cohort, 75% of injuries were covered by workers compensation or motor vehicle accident insurance. Within the highest 50

scores 96% (47) of patients were compensable. However in the lowest 50 scores 56% of injuries were compensable.

Even the diagnoses of the patients in cohort do not appear to be evenly distributed between the top and lower end of the scale. In the highest 50 scores there are 19 patients with complex regional pain syndrome but only 7 amputees (and 4 of these were partial hand amputees). Within the lowest 50 scores there are 22 unilateral major upper limb amputees and only 2 patients with complex regional pain syndrome.

The mean DASH score for different groups of patients by their compensability is listed in Table 2. When all the 274 patients are compared by whether or not the injury was compensable there are significant differences. The non compensable patients have an average DASH score 15 units lower (f = 29.15 p =.000). The difference between compensable and non compensable amputees is 12 units lower (f = 5.612 p=.02).

Table 2 : Comparison of DASH Scores by compensability of the patients:

Compensability	**No**	**Mean Dash Score**	**SD**	**Anova**
All Compensable participants	196	56	21	F 29.15 P = .000
All non-compensable participants	76	41	18	
Compensable amputees	42	50	23	F 5.612 P = .02
Non compensable amputees	33	38	19	

Table 3: Comparison of DASH Scores and their standard deviation between amputees and non-amputees

Category	No.	Mean Dash Score	SD	Anova
Amputee	75	45	22	F =10.18 P=.002
Non amputee	199	54	21	

Table 3 shows the mean DASH scores for amputees compared to non-amputees. There is a 9 unit difference which is significant (f = 10.18 p = .002)

Table 4: Comparison of Dash Score and their standard deviation by level of amputation

Amputation level	No.	Mean Dash Score	SD	Anova
Bilateral and quadruple	6	67	14	F = 3.95 P = .012
Brachial plexus proceeding to amputation	8	37	16	
Major upper limb amputation	35	39	21	
Partial hand	26	49	22	

Table 4 shows the mean DASH scores for amputees by level of amputation. The mean score for brachial plexus injuries not proceeding to amputation was 65 compared to 37 for those brachial plexus injuries who have proceeded to amputation. The mean score for major unilateral upper limb amputation was 39/100 compared to partial hand amputation which was 49/100. This difference was significant . (F = 3.95 . and p = .012)

The DASH score includes components for symptomatology (pain) and psychosocial status (esteem). Table 5 shows the mean esteem and pain scores for amputees compared to non-amputees and there is no significant difference.

Table 5: Comparison of Pain and Self-esteem Scores and their standard deviation between amputees and non-amputees

Category	No.	Mean Pain Score	Pain SD	Mean Esteem Score	Esteem SD
Amputee	75	19	16	10	3
Non amputee	199	20	5	12	6
Anova		P = .515 F = .425		P = .017 F = 5.73	

The symptomatology and psycho-social scores for different levels of amputation are listed in Table 6. The psycho-social status between the groups of amputees was significant (f = 4.46 with p .002) . There was no significant difference in the pain scores.

Table 6 : Mean Pain and Self Esteem Scores and their standard deviations by level of amputation

Amputation	No.	Mean Pain Score	Pain SD	Mean Esteem Score	Esteem SD
Bilateral and quadruple	6	16	6	13	1
Brachial plexus proceeding to amputation	8	20	8	9	3
Major upper limb amputation	35	19	23	9	3
Partial hand	26	19	16	11	2
Anova - significance between levels		F = .069 P = .96		F = 4.465 P = .002	

Table 7 indicates the mean DASH scores for the 48 patients recorded on assessment and after treatment. The numbers are small in each of the categories.

Table 7: Comparison of Assessment and post Treatment Scores using the DASH.

Diagnosis	No. patients	Mean total Dash score	Mean total Dash score post treatment	Change in scores
Quadruple amputations	1	82	47	Down 35
Major upper limb amputation	9	60	45	Down 15
Partial hand amputation	3	60	47	Down 13
Brachial plexus proceeding to amputation	1	37	20	Down 17
Brachial Plexus	3	72	47	Down 25
CRPS	10	72	52	Down 20
Arm pain	2	69	51	Down 18
Arthritis	2	51	46	Down 5
Hand injury	2	65	63	Down 2
Radiating from neck	1	61	68	Up 7
Shoulder	8	61	36	Down 25
Tendonitis	3	63	43	Down 20
Wrist	3	55	19	Down 26
Total	48	64	44	Down 20

However, all of the categories of patients except those with pain radiating from the neck showed an improved health status in their DASH scores. Using the paired samples T test the result is significant (t = 4.24 p = .000). Of the 14 amputees who were reassessed, 7 (50%) showed improved health status, 3 (21%) were unchanged and 4 (29%) had a diminished health status. Of the remaining 34 patients 26 (78%) showed improved health status, 4 (11%) were unchanged and 4 (11%) showed a diminished health status.

Table 8 lists the mean DASH scores by diagnosis of the patients who completed the Work and/or Leisure module. Patients with Complex Regional Pain Syndrome and Brachial Plexus injuries saw themselves as just as disabled with work and leisure tasks as they did with their normal tasks. However, unilateral major upper limb amputees, amputees from brachial plexus injuries and partial hand amputees saw themselves as having a greater disability affecting work tasks. Interestingly multiple limb amputees and patients with shoulder injuries and fractures saw themselves as being more affected in leisure tasks than work tasks.

Table 8: Mean scores for the Work and Leisure module by diagnosis.

Diagnosis	**Mean Dash Score**	**No. Leisure module patients**	**Mean Leisure score**	**No. Work module patients**	**Mean Work score**
Multiple upper limb amputations	67	4	83	3	77
Major upper limb amputation	39	19	63	23	87
Partial hand amputation	49	17	74	14	80
Brachial plexus proceeding to amputation	37	6	81	5	82
Brachial plexus	65	5	87	7	80
Carpal Tunnell	40	3	33	8	43
Arthritis	50	Nil	NA	5	41
Hand injury	53	2	50	3	41
Wrist	49	7	93	7	80
CRPS	68	20	91	29	90
Arm pain	50	7	66	12	64
Radiating from neck	58	8	79	12	79
Fracture	43	2	81	4	76
Shoulder	51	12	98	28	77
Tendonitis	51	5	45	15	60
Total	**51**	**118**	**77**	**176**	**73**
Anova	**F = 4.809 P = .000**		**F = 2.753 P= .001**		**F = 2.990 P = .000**

Conclusions

Health professionals have widely recognised that the level of amputation has no relationship to the psychological effects of the injury. The results of using the DASH in this study supports this view.

Many health professionals in Australia also believe that patients with compensable injuries have a higher perceived level of disability than those with non-compensable injuries. This research also supports this belief. It may be that those with compensation develop more anger because they can "blame" another person or organisation for the injury or that the financial reward from litigation also encourages an increased perceived level of disability. The social security system in Australia provides a relatively poor standard of living and it is therefore likely that patients with non-compensable injuries have a financial need to improve their functional status and return to independence and productive employment. Recognition and acceptance of these issues may assist health professionals in simply understanding that compensable patients do have a higher degree of perceived disability and to provide psychological services to assist them adjust to their disability and return to functional independence.

The marked difference in level of disability between brachial plexus injuries which proceed to amputation and those which do not proceed to amputation was surprising. Perhaps it is those patients who accept the long term dysfunction who accept amputation. Acceptance of the injury can aid in reducing the level of disability. It may be that those patients with the brachial plexus amputations are in a more acute phase of their disability whereas the patients who have proceeded to amputation would nearly always be at least 12 months post injury and maybe this time span and treatment during this period results in the differing level of disability. Although all of the brachial plexus injuries who proceeded to amputation, were provided with the chance to use a body-powered prosthesis, most were not successful and most eventually rejected the prosthesis. The difference cannot be attributed to one group of patients having to operate in a one handed fashion while the other have more bilaterality.

Forty-eight patients in 13 diagnostic groups completed the post-treatment DASH. Twelve diagnostic groups, showed a better health status after treatment. The only group that showed a reduced health status were those with pain radiating from the neck. The time between initial completion of the DASH and the post treatment completion varied enormously. The relatively low response level after treatment make it difficult to draw good conclusions from those scores. It may be that those patients who are not responding to treatment ceased attending and did not complete the post – treatment DASH. Other patients lived too far away and were referred elsewhere for treatment.

In general there was a relationship between the DASH score and the DASH work and leisure scores with those with the greatest level of perceived disability on the DASH score having the highest perceived level of disability on the work and leisure scores. The work scores are clearly affected by the type of work that the patient normally performs and those whose work involves a lot of manual handling may well perceive themselves as more disabled than those in clerical or professional roles. It may also be that high

technology equipment provided to upper limb amputees assists them more in the performance of their daily tasks than in the performance of manual handling work tasks and leisure tasks. The compensation system in Australia includes a focus on encouraging employers to offer suitable duties and this may affect a patient's ability to work.

Further research is required to better understand the factors that affect a patient's perception of their disability and assist them to adjust to that disability.

The DASH is a useful scale for measuring the health status of a patient with upper limb dysfunction. It is purported to quantify disability (predominantly physical function). It is a subjective measure designed to measure the patient's perception of their disease. The scale is easy, simple to administer and relevant to a patient's needs. It appears to encompass all cultures. It provides an excellent guide to O.T. treatment. It gives an indication of the needs that the patient sees that they have. We have found it useful for outpatients but for inpatients with a new injury they have not had the chance to attempt the vast majority of the activities and they find it very difficult to complete. The work and leisure modules are phrased differently to the bulk of the DASH items. The bulk of the items are phrased so that the patient rates their level of difficulty in performing the task. However the work and leisure module items compare their difficulty in performing the tasks in a their normal way. This different phraseology may account for some of the results seen in this research.

The measure is very useful for guiding Occupational Therapy treatment and especially focusing on the physical functioning since their injury. If a patient scores the maximum score on the 3 psycho-social items, I use that as indication to discuss referring them to a Psychiatrist. Likewise, severe symptomatology might necessitate a referral to an appropriate pain medicine specialist.

The reason for the differences in the patient's perceived level of disability needs more research especially since their perceived level of disability can have such a marked effect on their performance in tasks.

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FUNCTIONAL STATUS OF CHILDREN WITH A CONGENITAL UPPER LIMB REDUCTION DEFICIENCY

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INTRODUCTION

Congenital reduction deficiencies of the upper limb are rare. Nearly 50% of children with unilateral transverse upper limb reduction deficiencies (ULRD) wear a prosthetic device to enhance the ability to perform functional activities. Unfortunately, empirical evidence as to whether prostheses yield improved functional outcomes in these children is lacking. Assessment of arm and prosthetic functioning mostly relies on clinical observation of task performance. The purpose of this study was to assess the functional status of children with ULRD as measured by these standardized instruments.

To adequately measure arm and prosthetic functioning in children with ULRD, standardized measures at activity level are required. The difference between what a child "can do" in a clinical setting and "does do" in daily life is well known, also recognized as capacity and performance of activities [1]. Therefore both aspects should be measured. Capacity can be measured with functional tests and to measure performance of activities, assessment of spontaneous arm use or self-reported or parent-reported questionnaires are options.

Several instruments have been developed to measure arm function in children with various kinds of diagnoses. For adequate measurements of arm or prosthetic function functional tests and questionnaires have to be attractive for children, they have to assess bimanual activities of daily living and measure quality of movement and/or difficulty in performing activities [2]. The Assisting Hand Assessment (AHA) [3], the Unilateral Below Elbow Test (UBET) [4, 5], the Prosthetic Upper extremity Functional Index (PUFI) [6, 7] and the ABILHAND-Kids [8] questionnaires met these criteria.

METHOD

20 children with ULRD between 4 and 12 years old (mean: 8.7 ± 2.9 years) participated in the study. One child had below shoulder reduction deficiency, 16 below elbow and 3 had a partial hand. Eight children used a myoelectric device, one child used a passive device and eleven children did not use a prosthetic device (non-users).
Each child performed the UBET and an explorative assessment using the AHA. Parents and all children of 8 years and older were asked to fill out the PUFI and ABILHAND-Kids. For better comparison between instruments, results of sum scores on tests and questionnaires are presented as percentage of the total possible score. Sum scores of users with prostheses were compared to sum scores of non-users (without prostheses) with an independent t-test.

Instruments

Assisting Hand Assessment (AHA)
The AHA [5] was developed and tested for children who have one well functioning but one dysfunctioning hand (cerebral palsy and obstetric brachial plexus palsy). The test evaluates performance of arm/hand function as an assisting hand, since the affected hand primarily serves as an assisting hand rather than a non-dominant hand. Scoring of the quality of movement of 22 items is completed from the video using a 4-point scale from "effective (=

4)" to "does not do (= 1)". In the AHA version for children with ULRD, the item "moves fingers" is left out. Sum scores range from 0 to 100.

Unilateral Below Elbow Test (UBET)
The UBET [6, 7] was developed to measure arm or prosthetic functioning in children with ULRD 2 to 21 years old. The test has 4 versions for different ages, each consisting of 9 bimanual activities. Method of arm or prosthetic use is scored on a nominal scale describing 4 methods of grasp and stabilisation. The completion of task is scored on a 5-point scale from "no difficulty (= 4)" to "unable to complete the task (= 0)". Sum scores range from 0 to 36.

Prosthetic Upper extremity Functional Index (PUFI)
The PUFI [8, 9] evaluates the extent to which a child actually uses the prosthetic limb for daily activities, the comparative ease of task performance with and without the prosthesis and its perceived usefulness, which are respectively scored on a 6-point nominal scale, 5-point ordinal scale and 3-point ordinal scale. Higher scores represent less difficulty in performance and higher usefulness of the prostheses. Sum scores range from 0 to 100.

ABILHAND-Kids
The ABILHAND-Kids [10] was developed to measure the ability of children with cerebral palsy to user their hands in daily living. This questionnaire consists of 21 bimanual activities of daily life. The perceived difficulty in performing activities irrespective of the limb(s) actually used to perform the activity is rated on a 3-point scale ranging from "impossible (= 0)" to "easy (= 2)". Raw sum scores range from 0 to 42.

RESULTS

All 20 children and their parents completed the tests and questionnaires, except for one prosthetic user who did not complete the AHA without prostheses and one prosthetic user did not complete the AHA with prostheses. Since no differences were found between parent reported and child reported versions of the PUFI and ABILHAND-Kids, only data of parent-reported questionnaires are presented.

Method of performance

The way children can and do use their prosthesis or residual limb in performing activities is respectively scored in the UBET and PUFI (table 1).

Can do (UBET)			**Does do (PUFI)**		
Method of use prosthesis/arm	Users	Non-users	Method of performance	Users	Non-users
Actively	30 ± 28	55 ± 30	Prosthesis actively	15 ± 18	0 ± 0
Passively	38 ± 25	17 ± 13	Prosthesis passively	15 ± 21	0 ± 0
Elbow/trunk	11 ± 14	27 ± 27	Residual limb	41 ± 34	85 ± 12
One-handed	21 ± 13 *	1 ± 3	One-handed	23 ± 18 *	5 ± 6
			Some help	2 ± 4	3 ± 3
			Cannot do	4 ± 4	7 ± 9

Table 1. Method of use of prosthesis or residual limb of UBET and PUFI. Mean percentage of activities ± standard deviation. (* = Significant differences between users and non-users)

Results of the PUFI show that users and non-users performed respectively 94% and 90% of the activities independently. Results show a difference between how many activities a child *can* perform with the prosthesis (68%) and how many activities a child actually *does* with the

prostheses (30%). Both the UBET and the PUFI show that users perform more activities one-handed compared to nonusers (p=0.01 and p=0.00 respectively).

Difficulty and effectiveness of performance

Sum-scores of tests and questionnaires regarding effectiveness or ease of performance are presented in figure 1. Results show that children with ULRD perform well on daily activities.

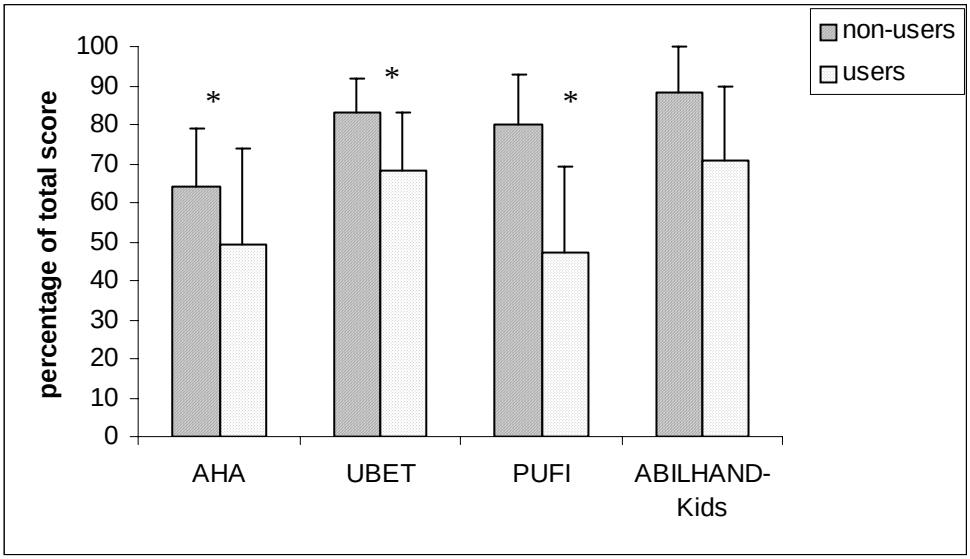

Figure 1. Sum scores of tests and questionnaires expressed as percentage of total score.

AHA results show that non-users used their arm more effectively compared to users (p=0.03). Also, non-users had less difficulty in performance of activities compared to users performing activities with prosthesis (p=0.02 UBET; p=0.00 PUFI). However, when we only take the activities into account in which the prosthesis is actually used (30% of the activities, see table 1), children perform these activities easily (sum score: 86.4 ± 9.8). Results of the ABILHAND-Kids did not show differences between users and non-users.

Usefulness of prosthesis

On average, users found the prosthesis useful in 34% of all activities as reported by the PUFI. Most activities of daily living, such as dressing tasks, are usually performed without prosthesis. Children use their prosthesis for specific activities such as riding a bicycle, playing with a beach ball and using scissors. For these activities, children find their prosthesis very useful (sum score: 79 ± 11).

CONCLUSIONS

Children with ULRD perform well on daily activities. 45% of the children wore prostheses and only 3 children wore the prosthesis more than 6 hours a day during the week. Children with ULRD find their prosthesis very useful in specific activities and can perform these activities very easily with prosthesis. Thus prosthetic devices have additional value in children with ULRD in specific activities rather than in activities of daily living in general.

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VALIDITY AND RELIABILITY OF FUNCTIONAL TESTS AND QUESTIONNAIRES FOR CHILDREN WITH A CONGENITAL UPPER LIMB REDUCTION DEFICIENCY

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INTRODUCTION

Standardized measures are required to adequately measure arm and prosthetic functioning in children with unilateral upper limb reduction deficiency (ULRD). A prerequisite for a test or questionnaire is that the instrument has to actually measure what it is supposed to measure, also referred to as validity.
With respect to validity, functional tests and questionnaires that are able to measure arm or prosthetic functioning were selected according to the following criteria: (i) they had to be attractive for children aged 4 to 12, (ii) they had to represent bimanual activities and (iii) they had to score quality of movement (functional tests) and/or difficulty (questionnaires) [1].
According to these criteria, we selected two functional tests, the Assisting Hand Assessment (AHA) [2] and the Unilateral Below Elbow Test (UBET) [3, 4] and two questionnaires, the Prosthetic Upper extremity Functional Index (PUFI) [5, 6] and ABILHAND-Kids [7].
In the present study we make a head-to-head comparison of these instruments to judge their clinical usefulness and to identify the best functional test and the best questionnaire for children with ULRD.

METHOD

Twenty children with ULRD between 4 and 12 years old (mean: 8.7 ± 2.9 years) participated in the study. Every child had three visits to the Erasmus Medical Centre in Rotterdam. Each visit, one test and one questionnaire was assessed. Repeated measurements for UBET and PUFI were performed in 10 children and for the AHA and ABILHAND-Kids in the other 10 children. Time interval between test and retest was two weeks (range: 11-18 days).

Clinical usefulness of the instruments

The usefulness of instruments is dependent on the validity, reliability and on the ability to detect important clinical changes. Regarding validity, we focused on convergent validity since a gold standard is lacking. Convergent validity refers to the consistency with the results of another measure that is believed to be assessing the same attribute. Convergent validity was investigated according to the hypotheses that sum scores of tests and questionnaires show significant ($\alpha < 0.05$) positive correlations with (1) the therapists global assessment of arm functioning with and without prostheses (as measured on a 10-point scale) and with (2) the other tests and questionnaires in case of difficulty and effectiveness of performance. To test the hypotheses, Spearman correlation (one-tailed) was used.

Test-retest reliability provides information about the stability of persons' responses over time in persons who truly remain unchanged. The Intraclass Correlation Coefficient (ICC) was calculated to quantify test-retest reliability. Reliability was considered excellent when values of

ICC exceed 0.75; values of ICC from 0.60 to 0.74 were evaluated as good, from 0.40 to 0.59 as moderate and less than 0.40 as poor.
In addition, standard error of measurement (SEM) was calculated to quantify the amount of measurement error. SEMs were calculated in the instruments' unit of measurement. For clinical use, the most relevant information is the magnitude of change between tests that is required to detect a real change, which is specified in the smallest detectable difference (SDD). SDDs at 90% confidence level were calculated. For comparison between tests, SDDs were expressed as percentage of the total possible measurement range of the instrument.

RESULTS

Convergent validity

For arm functioning without prostheses, no correlations were found between the therapist's global assessment of arm functioning without prostheses and sum scores of instruments (table 1). Regarding arm functioning with prosthesis, the therapist's global assessment correlated with the AHA (r_s = .84; p = .01) and PUFI (r_s = .66; p = .01).
We found several inter-relationships between sum scores on different instruments (table 3). When assessed without the prostheses, the AHA correlated nearly significant with the PUFI (r_s= .33; p = .08) and significant with the ABILHAND-Kids (r_s= .44; p = .03). The ABILHAND-Kids also correlated with the UBET (r_s= .53; p = .01) and PUFI (r_s= .52; p = .01) assessed without prosthesis. With prosthesis, the AHA correlated significantly with the PUFI (r_s= .79; p = .01) and there was a trend that the ABILHAND-Kids correlated with the UBET (r_s= .51; p = .08).

	Without prosthesis				**With prosthesis**			
	AHA	UBET	PUFI	ABILHAND	AHA	UBET	PUFI	ABILHAND
1. Therapist's global assessment	n.s.	n.s.	n.s.	n.s.	.84*	n.s.	.66*	.n.s.
2. other measures								
- AHA	-	n.s.	.33#	.44*	-	n.s.	.79*	n.s.
- UBET	n.s.	-	n.s.	.53*	n.s.	-	n.s.	.51#
- PUFI	.33#	n.s.	-	.52*	.79*	n.s.	-	n.s.
- ABILHAND	.44*	.54*	.52*	-	n.s.	.51#	n.s.	-

Table 1. Convergent validity of tests and questionnaires
* $p < 0.05$
# $0.05 < p < 0.08$

Test-retest reliability

Ten children performed test and retest without prosthesis and 5 children performed test and retest with prosthesis. For the AHA, only 8 children performed tests and retests without prosthesis and 4 with prosthesis. Excellent ICCs were found for each instrument, except for the PUFI with prosthesis, which had a good ICC (table 2). SEMs ranged from 1.5 to 12.9 and corresponding SDD_{90} ranged from 3.5 to 20.6. The SDD_{90}/range ratio ranged from 0.10 to 0.21 (table 2).

Instrument		Test	Retest	ICC	SEM	SDD_{90}	SDD/range
AHA	without prosthesis	62.6 ± 19.1	57.3 ± 15.5	0.70	7.4	17.4	0.17
	with prosthesis	36.3 ± 10.9	34.0 ± 23.3	0.94	5.4	12.5	0.13
UBET	without prosthesis	29.6 ± 2.9	29.6 ± 3.7	0.80	1.5	3.5	0.10
	with prosthesis	23.4 ± 5.6	26.2 ± 6.3	0.79	2.7	6.8	0.19
PUFI	without prosthesis	78.8 ± 11.8	78.3 ± 15.7	0.88	4.9	11.4	0.11
	with prosthesis	47.4 ± 22.0	50.8 ± 27.4	0.65	12.9	20.6	0.21
ABILHAND-Kids		35.2 ± 6.1	36.3 ± 4.3	0.79	2.4	5.7	0.13

Table 2. Test-retest reliability of functional tests and questionnaires. Mean values ± SD of test and retest, ICCs, SEMs, SDD_{90}s and SDD/range ratios are presented of 10 children (5 users and 5 non-users).

DISCUSSION

The therapists' global assessment of functioning with prostheses correlated with the sum scores of the AHA and PUFI. Without prostheses no correlations were found. Probably the therapist is experienced to evaluate arm functioning of children wearing prostheses and was therefore more focused in scoring with prostheses. The AHA and PUFI correlated to each other both with and without prostheses. Overall, these results suggest that the AHA and PUFI seem most valid for assessing arm and prosthetic functioning in children with ULRD.

Test-retest reliability was good to excellent for all instruments. Although all instruments were reliable, measurement errors were relatively high. To determine if measurement errors were small enough to be able to detect real clinical changes, SDDs were expressed as percentage of the total measurement range of the instrument. SDD_{90}/range ratios varied between 0.10 and 0.21 Instruments with SDD/range ratio between 0.10 and 0.13, i.e. the ABILHAND-Kids, PUFI without prosthesis, UBET without prosthesis and AHA with prosthesis, will be able to distinguish at least 8 steps on the total range of the instruments. Instruments with SDD/range ratio of 14 to 20% can distinguish 5 to 7 steps in the total measurement range.

CONCLUSION

All instruments are reliable to measure arm or prosthetic functioning. However, clinicians must be careful with conclusion about functional improvements since measurement errors of instruments are high. Preliminary judgments of our data suggest that all instruments are sensitive enough to detect real changes.

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FUNCTIONAL OUTCOME OF ADOLESCENTS AND YOUNG ADULTS WITH CONGENITAL UPPER LIMB REDUCTION DEFICIENCY

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BACKGROUND

In the Netherlands approximately seventy children with a congenital upper limb reduction deficiency will be born each year. One third of these children will be eligible for a prosthesis.
Little is known about the functional outcome of adolescents and young adults with congenital upper limb deficiency. This study is the second part of a project to investigate the functional outcome of adolescents and young adults with an unilateral congenital upper limb deficiency in the south-west region of the Netherlands. The first part consisted of a description of functional status and level of participation of this group by using postal questionnaires. In this part the adolescents and young adults have undergone a functional assessment.

PURPOSE

This study is performed to gain insight into the functional outcome of adolescents and young adults with an unilateral congenital upper limb reduction deficiency, with or without using a prosthesis. Functional outcome addresses performance of activities and capacity of the upper limbs, the prosthetic skills and the benefit of the prosthesis.

METHODS

Subjects

A cohort of 39 adolescents and young adults aged between 12-35 years with a unilateral congenital upper limb deficiency participated in the first part of the study. All of them were patients at the departments of rehabilitation medicine in the south-west region of the Netherlands. Twenty-seven people agreed to participate in the second part of the study.
Twenty-one people with a below-elbow deficiency were selected. One person, who does not use a prosthesis, was lost due of moving to another country between the first and the second part of the study. The mean age was 22.7 ± 6.7 years and there were 11 women and 9 men in the sample. Nine of them used a prosthesis, six used a myoelectric device and three used a passive device. Wearing pattern of the prosthesis, the Skill Index Ranking Scale (SIRS) and other characteristics of the subjects are shown in table 1. The SIRS provide an ordinal scale of a person's ability with prosthesis (0-14). For example ranking 4 means: spontaneously place the terminal device in position and use it for support; ranking 14: control the grip while moving the arm, throw things from above the shoulder.

Subject	Gender	Age	Prosthesis	Wearing (hrs/ day)	SIRS
1	M	29.0	No	0	
2	V	20.3	Yes: myo	11-15	14
3	V	19.6	No	0	
4	V	22.3	No	0	
5	M	15.6	Yes: passive	11-15	4
6	V	16.0	No	0	
7	V	26.8	Yes: myo	6-10	14
8	M	20.0	No	0	
9	M	14.7	No	0	
10	V	33.5	No	0	
11	V	18.4	Yes: passive	11-15	4
12	V	23.2	Yes*: myo	0	*
13	M	27.5	Yes: myo	11-15	13
14	M	34.8	Yes: passive	6-10	4
15	V	29.3	No	0	
16	M	15.3	No	0	
17	V	22.5	Yes: myo	6-10	12
18	M	15.8	Yes: myo	6-10	12
19	M	33.4	No	0	
20	V	16.6	No	0	

Table 1. Characteristics of subjects
*Does not wear prosthesis because of oedema

Procedures

Each participant had one visit to one of the departments of the rehabilitation medicine in the south-west region of the Netherlands or were visited at home. Each participant performed the Unilateral Below Elbow Test (UBET) and people using a prosthesis performed the Southampton Hand Assessment Procedure (SHAP). All functional tests were administered by the same investigator (SAML). For people using a prosthesis, the UBET was first performed with the prosthesis followed by test performance without using a prosthesis.
In addition, each participant filled out the Child Amputee Prosthetics Project-Functional Status Inventory (CAPP-FSI) and the Prosthetic Upper extremity Functional Index (PUFI).
Performance of users with prostheses were compared to non-users (without prostheses) by means of a t-test for independent samples; comparison of users with and without prosthesis by means of a t-test for dependent samples.

Instruments

Capacity and prosthetic skills were measured by functional tests scored by a professional: the Unilateral Below Elbow Test (UBET) and the Southampton Hand Assessment Procedure (SHAP). The UBET consists of 9 bimanual activities (the adolescent version, 11- 21 years) which have to be performed with and without the prosthesis. The completion of task and the method of use with the prosthesis or residual limb is scored for each task. The completion of task is scored on a 5-point scale from "no difficulty (= 4)" to "unable to complete the task (= 0)". Sum scores range from 0 to 36. Method of prosthetic use or residual limb is scored on a nominal scale describing 4 methods of grasp and stabilisation.
The SHAP consists of manipulating a series of lightweight and heavyweight objects of different shapes with the prosthesis based on specific grip patterns and in addition 14 daily activities performed with the prosthesis. The kind of grip and the time of performance of each task or activity is measured.

Performance of activities was assessed by self-administered questionnaires: the Child Amputee Prosthetics Project-Functional Status Inventory (CAPP-FSI) and the Prosthetic Upper extremity Functional Index (PUFI).
The CAPP-FSI consists of 34 upper-extremity daily activities and is focused on the level of independence of performing each activity (range 0-4; 0 = dependent, 4 = independent) and the frequency these activities are completed with the prosthesis versus without.
The PUFI consists of 38 daily activities (the older-child version) and evaluates the extent to which a person actually uses the prosthetic limb for these activities, the comparative ease of task performance with and without the prosthesis and its perceived usefulness. This is scored respectively on a 6-point nominal scale, 5-point ordinal scale and 3-point ordinal scale. Higher scores represent less difficulty in performance and higher usefulness of the prostheses. Sum scores range from 0 to 100.

RESULTS

All twenty adolescents and young adults completed the questionnaires and tests. Due to oedema, one prosthetic user was unable to perform the UBET and SHAP with prosthesis.
The way people *can* use their prosthesis or residual limb in performing activities is scored in the UBET and SHAP. The way people *do* use their prosthesis or residual limb in performing daily activities is scored in the CAPP-FSI and PUFI.

Prosthetic skills

Users manipulate objects with a tripod or spherical grip. Most of the daily activities of the SHAP were difficult to perform with the prosthesis. The most difficult tasks were pick up coins, to undo buttons, turning a page and pouring from a jug.

Method of performance

Results of the PUFI showed that 93 % of the activities were performed independently; in users this percentage is 90 %, in non-users 95 %. According to the CAPP-FSI and PUFI independent performance was found in 71% of functional activities. Activities that were most often performed with a person's assistance according the PUFI were: to put on a necklace, cut meat with knife and fork, chop fruit, use a can opener, hammer a nail and skipping rope, and according to the CAPP-FSI: peel an apple, sew a button, blow-dry hair and cut meat with knife and fork.

Method of performance is scored by the UBET and PUFI as shown in table 2. Subjects *can* use their prostheses actively or passively in 84 % of the activities as shown by the UBET. However, in daily live persons *do* use their prostheses in 30 % of the activities as reported by the PUFI.

Can do (UBET)				**Does do (PUFI)**		
Method of use prosthesis/ arm	**Users with (n = 9)**	**Users without (n = 9)**	**Non-users (n = 11)**	**Method of performance**	**Users (n = 9)**	**Non-users (n = 11)**
Actively	35 ± 26	19 ± 28	18 ± 21	Prosthesis actively	18 ± 28	0 ± 0
Passively	49 ± 33	54 ± 34	56 ± 32	Prosthesis passively	12 ± 15	0 ± 0
Elbow/trunk	0 ± 0	9 ± 15	9 ± 15	Residual limb	38 ± 28	78 ± 12
One-handed	14 ± 33	17 ± 33	17 ± 32	One-handed	26 ± 15	18 ± 11
Cannot do	2 ± 6	1 ± 4	0 ± 0	Some help	2 ± 2	3 ± 5
				Cannot do	4 ± 4	1 ± 2

Table 2. Method of use of prosthesis or residual limb in UBET and PUFI. Mean percentage of activities ± standard deviation

Difficulty and effectiveness of performance
Sum-scores of tests and questionnaires regarding effectiveness or ease of performance are presented in table 3.

Instrument		Mean	SD	Min	Max	P
UBET						
Competion of task score	Non-users	34.0	2.1	29	36	] p =
(Range 0 – 36)	Users - with prosthesis	32.9	2.3	28	35	] 0.27] p =
	Users - without prosthesis	34.0	1.7	31	36	] 0.12
PUFI						
Ease of performance	Non-users	93.2	6.0	82.8	98.6	] p =
(Range 0 – 100)	Users - with prosthesis	60.3	23.3	11.2	91.4	] 0.000] p =
	Users – without prosthesis	87.5	4.9	78.4	94.0	] 0.007
Usefulness of prosthesis	Users – with prosthesis	46.7	21.6	9.7	81.0	
(Range 0 – 100)						

Table 3. Mean sum scores and standard deviation (SD) and minimum (Min) and maximum (Max) sum scores of UBET and PUFI of all subjects, non-users (n = 11) and users (n = 9) of prosthesis.

Non-users compared to users with prosthesis performed activities equally well as measured by the UBET. However in daily life, measured by the PUFI, non-users had less difficulty in performance of activities compared to users performing with prosthesis (p= 0.000).
However, when we only take the activities into account in which the prosthesis is actually used (30% of the activities, see table 1) subjects perform these activities easily with prosthesis (sum-score: 94.6 % ± 5.0).

Usefulness of prosthesis
Wearers used the prosthesis on average 10 hours a day and its usefulness was reported to be 46.7 % (SD 21.6) assessed by the PUFI. Usefulness scored by users of a myoelectric prostheses was 55.5 % (SD 18.4). Most activities of daily living, such as dressing tasks, are usually performed without prosthesis. However, persons find the prostheses very useful in specific activities (sum-score: 85.5 % ± 9.7) such as riding a bicycle, pushing a lawnmower and twist the lid of a bottle.

CONCLUSIONS

Adolescents and young adults with a congenital below elbow reduction deficiency perform well in functional activities with or without using a prosthesis. Subjects are very independent in performing daily activities. Nine of the twenty subjects wore a prosthesis on average 10 hours a day. They find their prostheses very useful in specific activities and can perform these activities very easily with prosthesis.
These results suggest that prosthetic devices have additional value in persons with a congenital below elbow reduction deficiency in specific activities rather than in overall performance of activities of daily living.

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A PRELIMINARY STUDY OF 40+ UPPER EXTREMITY PATIENTS USING THE ANIMATED CONTROL SYSTEM

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This talk will discuss the findings of 40+ who have been fit with the Animated control system (ACS). All individuals have been fit with the ACS-1100 or ACS-1105 controllers. The information being obtained is from the controller.

The information revealed is an assimilation of cases with history from new fittings to over 4 ½ years. There have been over 450 patients fit over the last 4 ½ years. The types of fittingvary from Trans Carpal, Wrist Disarticulation, Trans Radial, Elbow Disarticulation,Trans Humeral, Shoulder Disarticulation and Interscapular Thoracic. Fittings of all age ranges was done from pediatric to the elderly adult.

All individuals have agreed to the information being obtained. The types of information is amount of hand cycles (open cycles only), wrist rotations (either supination or pronation), elbow cycles ,version of software, mfg date, serial number, which type of TASCS (control options). Other information being gathered is ratios of hand to wrist cycles (if applicable), age of user, types of devices (TD's),, hobbies, occupation, etc. The type of useage work, hobbies,adl's are also being collected.

A questionnaire for the prosthetist and patient was formed and with a face to face meeting the electronic gathering of information was started.

The statistics will be presented on the useage, levels, and time of use.

PERFORMANCE EVALUATION OF THE NEW OTTO BOCK "DynamicArm" BY MEANS OF BIOMECHANICAL MODELLING

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1. INTRODUCTION

Through infra-red motion analysis systems it is possible to acquire the 3D joint kinematics of a patient while performing every day activities. These data, combined with a biomechanical model of the anatomical structures under investigation and clinical rating scales, can form the basis for an objective assessment of the patient motor ability. When the subject acquired is an amputee fitted with a new prosthetic arm, the information provided can be useful not only for the practitioner but also for the prosthesis designer. The aim of this work is to give an example of this kind of clinical/technology assessment, presenting the results obtained for a young trans-humeral amputee fitted with a prototype of the new Otto Bock DynamicArm. In particular, the analysis intended to quantitatively evaluate: 1) the performances of the Otto Bock arm, and in particular of the electromechanic elbow, when controlled in-vivo by the patient EMG signals; 2) how the patient controls the prosthesis, in order to identify critical movements and prevent possible disorders; 3) if the new prosthesis increases the patient abilities.

2. MATERIAL AND METHODS

2.1 The new Otto Bock DynamicArm

The 12K100 DynamicArm is a myoelectrically controlled and electromotively powered elbow joint for the fitting of upper arm amputees up to very distal trans-humeral amputation level (theoretically up to 4cm less the natural length of the arm). The main advantages of the DynamicArm are: 1) high lifting and holding force (with lift arm set to 305 mm: lifting force= 60N; holding force=230N); 2) minimal lift time 0.5s depending on forearm length and terminal device used; 3) natural free swing characteristics; 4) low noise level and no internal noise in free swing; 5) integrated lithium-ion battery with capacity for one day; 6) appealing exterior. The DynamicArm and hand prosthesis is controlled by electrodes, linear control elements, switches or a combination thereof. The CPO can adjust to the individual patient via Bluetooth® technology. The electric elbow can be locked or unlocked even with a switched-off DynamicArm or with an empty battery by operating the pull-cable from any position, even under load. An AFB (Automatic Forearm Balance) flexion tool stores the gravitational energy released when extending the arm and uses it for flexion. The battery operating time is thereby considerably increased, hoist time shortened. The vario drive allows for harmonic movements and high lifting forces.

2.2 Subject description and prosthesis set-up

The subject involved in the present tests (after giving his informed consent), was a young adult (initials GF) who underwent a first proximal trans-humeral amputation in 2003 due to a work-related trauma. GF received a first myoelectric prosthesis with threshold control in 2004, and after few months he was given the first version of the Otto Bock arm with proportional control, which he was immediately able to manage. For both he received proper

training. GF controlled the elbow flexion and extension with contractions of the trapezius and deltoid, respectively. The length from the Otto Bock elbow axis of rotation to the load carrying position of the hand was 365mm. Since different patients will have different performances depending on their lever arm and hand type, the performances of the Otto Bock elbow were set to a medium level (in terms of maximal excursion and velocity) to reasonably report a mean performance.

2.3 Clinical rating scale assessment

GF ability in controlling the threshold and the Otto Bock prostheses was clinically assessed at the end of the respective training periods through an interview and using the ABILHAND rating scale [1].

2.4 Motion analysis protocol

GF was acquired with a VICON 460 stereophotogrammetric system (Vicon Motion Capture, Oxford, UK) with 6 video cameras, while performing 5 elbow flexion-extension trials with different loads on the hand: 0Kg, 0.52Kg, 1Kg, 2Kg, 3Kg. For every trial, at least two consecutive flexion-extension cycles were repeated. GF was instructed to always perform the movement as fast as possible.

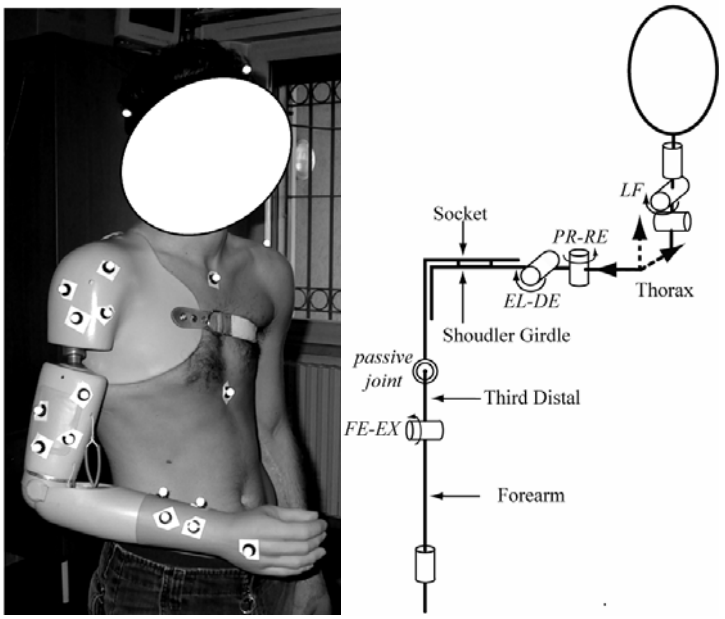

Figure 1a,b a) Marker-set used: markers on the left acromion and the hand are used for visualization purposes only. Not visible in the figure: markers on the 7th cervical and 8th thoracic vertebra and on the left and posterior part of the head; b) kinematic model of the amputee with the prosthesis; abbreviations are only given for the joint angles analysed in the paper (see text for details).

To analyze the electromechanic elbow performances, the relative motion of the forearm with respect to the third distal had to be tracked. Since GF engaged the elbow with contractions of trapezius and deltoid, possible critical movements could arise from excessive motion of the head and shoulder girdle, leading to early deterioration of the sterno-clavicular, acromio-clavicular and scapulo-thoracic joints (hereinafter referred together as the "shoulder"), and cervical rachis problems. Given these considerations, twenty retro-reflective markers were attached on the head, thorax, socket, third-distal and forearm, thus defining an open kinematic chain with 7 active degrees of freedom (Figure 1a,b), associated to the neck, shoulder and elbow joints. To define the mobility of these joints, a system of reference (SoR) had to be defined for each segment. For the head, thorax and shoulder girdle these were obtained through the "calibration" of relevant anatomical/prosthetic landmarks with respect to the correspondent cluster of markers [2-5]. For the definition of the third-distal and the forearm SoR, we combined the use of well-identifiable landmarks, with a functional, optimisation-based method [6], which enables to compute the real axis of rotation of the elbow (flexion-extension) and of the hand (prono-supination) [7]. Joint angles were then obtained decomposing the relative orientation of adjacent segments using appropriate sequences of Euler angles: flexion-extension (FL-EX)

and prono-supination for the elbow, flexion-extension, lateral flexion (LF) and axial rotation for the neck and protraction-retraction (PR-RE) and elevation-depression (EL-DE) for the shoulder.

2.5 Data analysis

Using the shoulder angles and their first derivative, we identified in each trial the following phases: elbow flexion, transition from flexion to extension, elbow extension. In each phase the head, shoulder and elbow patterns were analyzed in terms of excursion and velocity.

3. RESULTS AND DISCUSSION

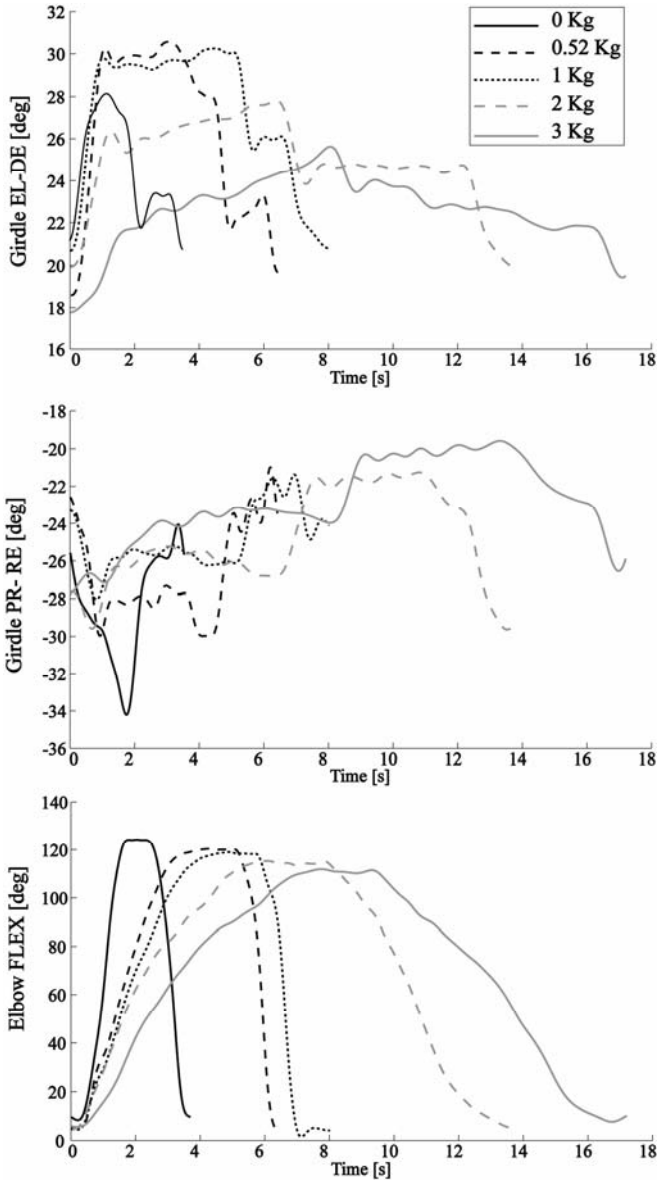

Figure 2 Shoulder girdle elevation-depression (top), protraction-retraction (middle), elbow flexion-extension (bottom) for the different loads tested.

3.1 Clinical rating scale assessment

GF reported a generalized satisfaction in the use of the new prosthesis. In particular he stressed the possibility to lift heavy loads with the prosthetic arm. The comparison of the ABILHAND post-training results with the threshold and with the Otto Bock prosthesis, showed with this latter an improvement in the feeding and dressing issues, which changed from difficult to easy. These results appear to be related to the Otto Bock proportional control.

3.2 Elbow performance

Elbow flexion and extension patterns for the different loads tested are shown together in Figure 2. Given the high level of repeatability found in the gesture, only one representative curve among the repetitions is reported for every load. The mean range of elbow flexion-extension observed was 115° (coefficient of variation: 4%), which decreased of about 11° changing the load from 0Kg to 3Kg. The minimum elbow extension ranged from 2° (1Kg) to 8° (0Kg), while the maximum flexion from 112° (3Kg) to 126° (0Kg). The range measured is thus smaller than that of an able-bodied subject (140°-150°) but, for light loads, appears to be adequate for most feeding activities [8].

The maximal and mean velocities reached in flexion were 145°/s 77.4°/s, respectively, when no load was applied on the hand; these velocities decreased to 32°/s and 14.8°/s for 3Kg. The maximal and mean velocities in extension were 220°/s and 88°/s respectively, for 0Kg; these velocities decreased to 31°/s and 19°/s for 3Kg. From the patterns reported it appears that the elbow behavior can be roughly subdivided in two

groups: from 0Kg to 1Kg, in which the elbow gave its best performances with limited degradation, and 2Kg-3Kg, in which a noticeable lengthening of motion duration could be observed, both in flexion and extension.

3.3 Subject movements for prosthesis control

The macroscopic effects of the trapezius (for flexion) and deltoid (for extension) contractions are reported in Figure 2: the elbow flexion appears to be activated by a shoulder elevation and retraction, while the extension by a shoulder depression and protraction. An exception was the slight girdle protraction for elbow flexion with 2Kg and 3Kg, which was however followed by a higher protraction for extension. Shoulder motion usually ranged from 11° to 8°, both for elevation-depression and protraction-retraction, generally tending to decrease with the increase of the load. An explanation for this result can be found observing that heavier the load, higher the muscle force required to move the shoulder girdle against gravity. Since the EMG level for maximal performance is fixed, this means that with heavier loads the motion required to reach this level is smaller. While the shoulder motion tended to decreases, with increasing loads the duration of the contraction tended to increase, almost doubling from 0.52Kg to 2Kg. The elbow flexion-extension was also followed by 8°-12° of lateral flexion of the head toward the shoulder during flexion. These frequent, incongruous, asymmetric and repetitive gestures appears to be all risk factors for potential shoulder and cervical rachis cumulative trauma disorder syndromes, with mechanical irritation of the tendinous and peritendinous structure and chronic muscular fatigue. This latter problem may also result in a decrease in the quality of the EMG signals acquired by the prosthesis sensors. A focused rehabilitation (e.g. relaxant massotherapy, physical exercises for a correct articulation and stretching) may help in prevention. Further analyses are required to draw definitive conclusions.

4. CONCLUSIONS

The aim of this paper was to quantitatively assess the performances of a prototype of the new Otto Bock DynamicArm in-vivo when directly controlled by an amputee. The Otto Bock arm proved state-of-the art performances in terms of liftable loads, angular range and velocity, with general satisfaction of the patient. Considering the patient into the measurement loop, gave the possibility for a kinematic analysis of his movements, bringing in evidence potentially critical motor strategies. Future efforts will be intended to confirm these observations on the patient examined and on a representative population of amputees.

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EVALUATION STUDIES OF NEW ELECTRIC TERMINAL DEVICES

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INTRODUCTION:

Two new electric terminal devices (TDs) have been developed for use with myoelectric prostheses of the upper limb, at all levels. They each feature a quick disconnect wrist connection, so they may be interchanged with each other, and also with existing electric TDs from other manufacturers. The two TDs studied are:

1. The Motion Control (MC) Hand with Flexion Wrist.

Figure 1. The Motion Control Hand with 30^o flexion and extension

- The wrist will lock in three positions- 30 degrees of flexion & extension, as well as neutral position.
- The lock/unlock of flexion is performed by the wearer pushing a simple spring-loaded button on the TD, which allows repositioning of the wrist, which then will relock into a new position. The button is intended to be easy even for bilateral arm amputees to push.
- The flexion lock has high strength – it will support leaning on the TD, with loads up to 25 kg.
- No additional length is required for the Flexion Wrist in the Hand, compared with standard adult electric hands.

Figure 2 – A mechanical lever on the Motion Control Hand locks the wrist into flexion or extension.

2. The Electric Terminal Device (ETD), intended for rugged work and hobby situations.

- The ETD utilizes the same motorized drive as the MC Hand, which typically produces a maximum of 10 kg of pinch force.
- Classic hook-shaped fingers are used, specifically the APRL design manufactured by Hosmer, Inc. These hook fingers were used on the earlier Hosmer product, the Synergetic Prehensor, which was successful with many wearers.
- The housing for the ETD is water-resistant, so that water or dirt will not enter the gear and drive area. The quick disconnect contacts could be susceptible to water or dirt, and a protective sleeve is available optionally to cover the wrist and forearm of the prosthesis.
- Flexion Wrist is available in the ETD, although it adds approximately 2.5 cm. of length, additional to the hand. Only one of these ETD subjects has the Flexion Wrist in the ETD.
- Quick disconnect attachment at the wrist will allow the TDs to interchange.

Figure 3 – The Electric Terminal Device (ETD), pictured with the flexion wrist (right), and without (left). Because of the longer length of the hook fingers, compared to the hand, flexion increases the length of the ETD by about 2.5 cm.

DESIGN OF SHORT-TERM EVALUATION STUDIES:

The initial surveys were conducted of both TDs, with six wearers in each group. No subjects participated in both surveys.

Composition of MC Hand with Flexion Wrist study:

- 6 wearers: 5 male, 1 female
- 2 Bilateral BE's, 4 Unilateral (1 AE, 3BE)
- All were regular wearers of electric hands previous to the study; the only change to their prosthesis was the addition of the new hand with Flexion Wrist.

Composition of Electric Terminal Device Study:

- 6 wearers: all male
- 2 bilateral BE's, 4 Unilateral (1 SD, 3 BE)

The purposes of the study may be summarized as:

- To document the use of the TDs, in terms of:
  Hours per day, Usage during wearing, Use of Flexion/Extension

- To learn how typical tasks are performed with each TD. Categories included: Grooming, Daily Activities, Kitchen, Work, Hobbies

- Ratings of TD Performance were performed for relevant functions of each survey. Ratings are made in comparison with the previous TD.

Discussion of a "Small n" survey.

Though the number of subjects is small and statistical significance is low, a great deal of useful data is collected from these subjects. In a development context, important results are, one, to guide development, and two, to validate that design goals are met.

SUMMARY OF SURVEY RESULTS:

MC Hand with Flexion Wrist:

The most valuable results for our use were to learn from these active wearers about the actual *performance* of activities assisted by the Flexion Wrist. Activities improved by the Flexion Wrist included:

- Dressing - buttons, zippers, pulling socks, tying necktie, etc.
- Hanging up & folding clothes
- Opening/closing doors
- Holding books or papers
- Using hand tools - tweezers, camera, sewing, pens, etc.
- Handling briefcase and bags
- Steering - cars, shopping carts, etc.
- Handling children
- Caring for livestock- milking cows, holding collars, etc.
- Opening bottles, holding a cup, using utensils, etc.

Estimating from memory, the subjects guessed that they used the Flexion Wrist in performance of one-fourth to two-thirds of the tasks they performed. The Flexion Wrist often gave better positionability near the subject's mid-line, e.g., for dressing and eating tasks. Also mentioned frequently was the ability to hold flat objects, like trays and books, at a convenient angle without requiring unnatural shoulder or torso positions.

Fig. 4. The use of the Flexion Wrist allows the wearer to position the Hand closer to the midline.

All ratings obtained from the subjects cannot be mentioned here, because of space, but the overall rating of "Contribution of the Flexion Wrist to Function of the TD", achieved a rating of "much better" from four subjects, and "better" from the remaining two.

Results of the Electric Terminal Device (ETD) Survey

The range of activities which utilized the ETD was extremely broad. As above, the actual activities of the subjects were the biggest lesson for the development team. Examples of activities improved by the ETD:

- Holding utensils: Knife, Fork & Spoon; Pen & Pencil
- Grasp of a variety of Tools : Nuts & Bolts, Tool Handles
- Reaching into pockets
- Turning pages
- Holding large objects: mugs, briefcase, etc.
- Driving
- Change diapers
- Keyboarding
- Scratching an itch
- Carry dishes
- Handle papers, money, etc.

Fig. 5 Some of the advantages mentioned by ETD wearers were the fine tips, visibility of the curved shapes, high gripping pressure possible with the narrow finger shapes, and wide opening.

Advantages mentioned were the fine tips, visibility of the curved shapes, high gripping pressure possible with the narrow finger shapes, and wide opening. Important to many was the similarity of the hook shapes to earlier body-powered TDs they had used, or still used for part of their daily activities. Several subjects, with both bilateral and unilateral loss, continue to use both their body-powered and their electric prostheses for significant parts of the day. The weight was an advantage for most – the ETD is lighter than other electric TDs, and although heavier than a body-powered TD, was not found to be awkward.

All of the ratings which were collected will not be discussed, except the rating of "Overall TD performance". Five of the six subjects rated the ETD "much better" than their previous TD, and the sixth subject rated it "better" overall. The TD of comparison was in four cases the Greifer, in one case the Synergetic Prehensor, and in one case the body-powered hook TD (#5XA).

Fig. 6. The ETD compared to the Otto Bock Greifer and Hosmer Synergetic Prehensor.

A NEW ELECTRIC SYSTEM WITH SIMULTANEOUS ELBOW & HAND CONTROL

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Motion Control, Inc., a subsidiary of Fillauer, L.L.C.
Salt Lake City, Utah

The evolution of electric prosthetic systems has resulted in the widespread transition to digital controllers, i.e., microprocessors. In fact, it may be true that every available electronic controller in the field of prosthetics will soon be a digital controller. In our own experience, the ProControl 2 was introduced seven years ago, which provided digital control of hand and wrist. The last seven years have brought new generations of improved software (the newest version is 5.1.2, so five generations have evolved), as well as development of new sensors. Also, continued evolution of commercial microprocessors has resulted in more powerful controllers, in even smaller sizes than were available previously.

In December of 2002, initial units of the Utah Arm 3 with electric elbow, hand, and wrist functions were prepared, including completely new digital controllers for elbow, hand, and wrist. Five units were placed into field trial usage on every-day wearers of a transhumeral (or higher level) electric arm prosthesis.

The number has been increased as improvements have been implemented, although until January 2005 all were considered "Beta Units," and the early ones upgraded with the improvements to the circuits, or other components.

The goals we have tried to accomplish in the digital Utah Arm 3 include:

- Simultaneous elbow and hand function (when desired)
- Wide variety of input choices:
  single/dual site EMG, harness-mounted force sensor, slider(Linear Potentiometer), touch pads, as well as other manufacturer's sensors.
- Computer Interface for set up of the system and adjustment, also allowing for improvements in training of the patient, and additional options and features.
- Auto calibration of the sensitivity of the hand control, so that the patient can readjust these values at will.
- No sacrifice in the performance relative to the Utah Arm 2, e.g., battery life, controllability, features, configurations, etc.

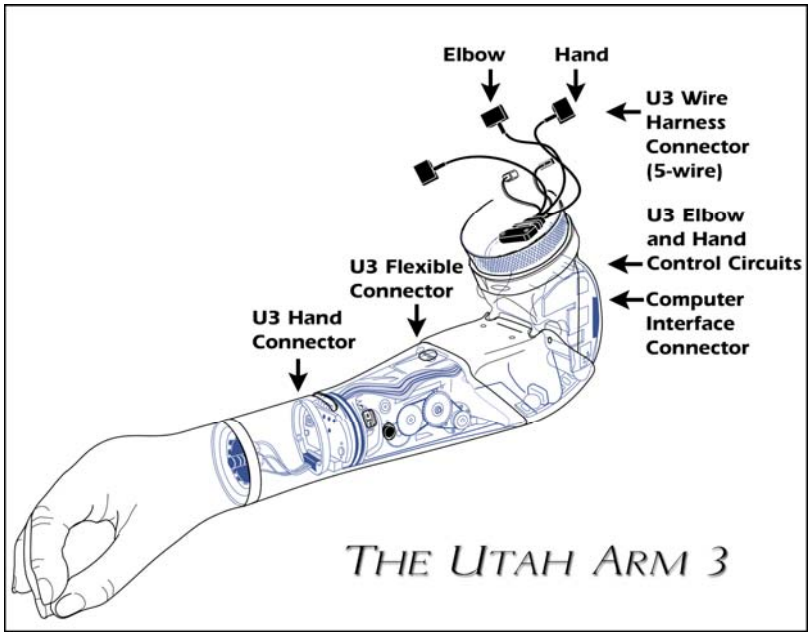


Figure 1: The Utah Arm 3 features entirely new internal electronics, as well as new connector options for separate elbow and hand inputs.

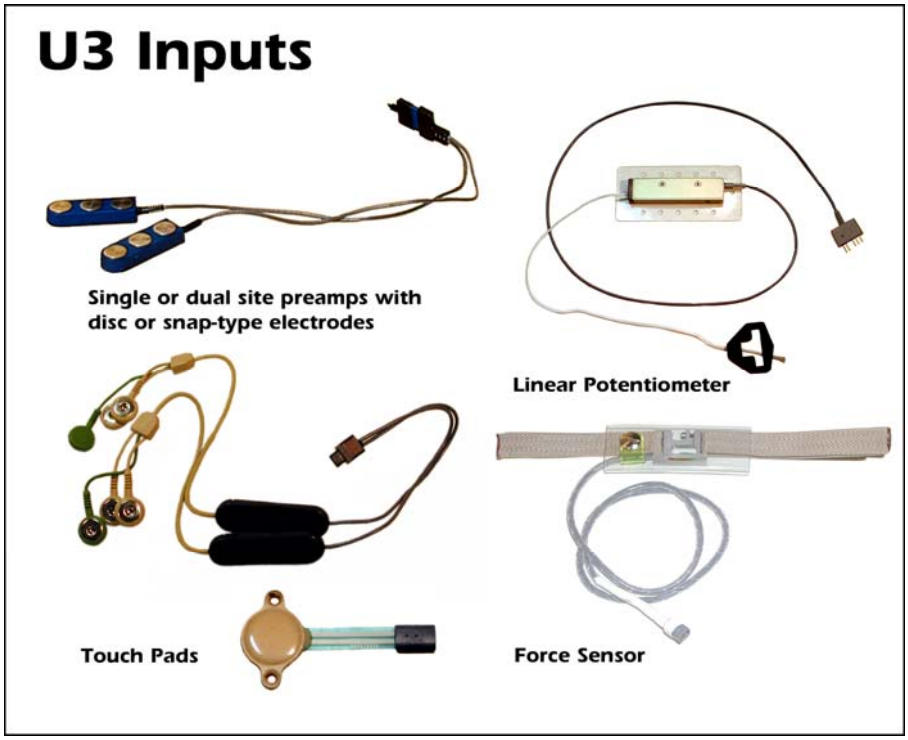


Figure 2: A variety of control inputs allow a wider range of candidates to be considered for electric prosthesis fitting, including EMG (dual or single site), harness-mounted force sensor, linear potentiometer, touch pads, and other manufacturer's sensors in some cases.

Technical issues included the transition from a complex analog controller to a digital-based system, which was not without its challenges. In our "digital world" we might take for granted the sophisticated controls that we find in our automobiles, personal digital assistants, music recorders and players, and many more examples. However, consider the steps involved in making this transition:

- Inputs from a variety of analog sensors must be digitized and signal conditioning must alter the signal for processing in the digital controller.
- Complex processing must be implemented in a digital logic, such as, EMG filtering, thresholds, differencing of signals, transition of elbow states (locking, unlocking, freeswing, etc.)
- Processes previously developed for hand control must be adapted for elbow control, which in some cases requires significant modification.
- Performance of every function must be validated with every small change in the design, e.g., functions of elbow flexion/extension, lock, unlock, freeswing, hand open/close, hand sleep and wake-up, and wrist pronation/supination (7 functions) must be carefully evaluated for each of the 42 control configurations – *with every change in the controller or hardware design.*
- Hardware must be newly designed, to fit into the same spaces in the Arm prosthesis, such as, connectors, circuits, interface with an external computer,
- Battery life must be monitored and evaluated with each change in design. Early versions of the U3 circuits were shown to have unacceptably short battery life, due to the high current draw of the microprocessors. A redesign of the control circuits was required, consuming several months of development time.
- Manufacturing processes must be developed, and refined, to make the new hardware, and software. In this case, these processes will *not replace* the U2 processes, since both versions will continue to be offered, so the same staff must learn to make a second version simultaneously.

Advantages for the patient:

- A more natural combination of elbow and hand function, while still eliminating control cables potentially. The force sensor and linear potentiometer have actually been harnessed to use scapular abduction in most cases (to control elbow flexion), but the force required is very much lower than the forces required for a body-powered control cable.
- The linear potentiometer has been used successfully, usually as the control input for elbow flexion. It seems to provide better proprioception to the wearer, compared to the force sensor.
- In cases where EMG signals are inadequate for dual-site control, single-site control has been conveniently used in some cases. Touch pads are available, but few fittings to date have required touch pad input.
- Although most wearers use simultaneous elbow and hand set up, the battery life has been acceptable- lasting the wearer the entire day.

Advantages for the prosthetist:

- A very wide range of system configurations will be available to the prosthetist, which will allow many patients to utilize as many as three functions, who would not be candidates for this level of function previously.
- The ability to provide two functions simultaneously will allow the prosthetist to overcome what previously could hold back some patients from more natural and rapid use of their prosthesis.
- Training the patient with the system will be much more easily accomplished, using a more visible User Interface program. Adjustments will also be easier, due to the improved visibility and intuitive nature of the adjustments.

Future plans: The Utah Arm 2 will continue to be offered as an alternative, for those not requiring the advanced features of the U3. Other features will continue to be developed, which will be more easily implemented because of the digital control's programmability and its computing power.

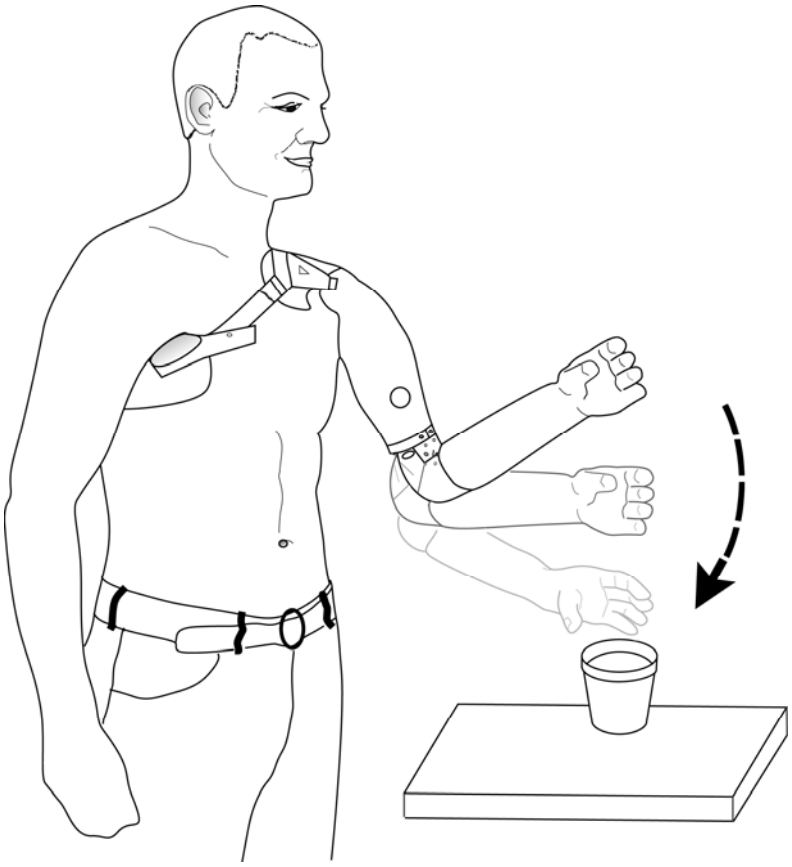
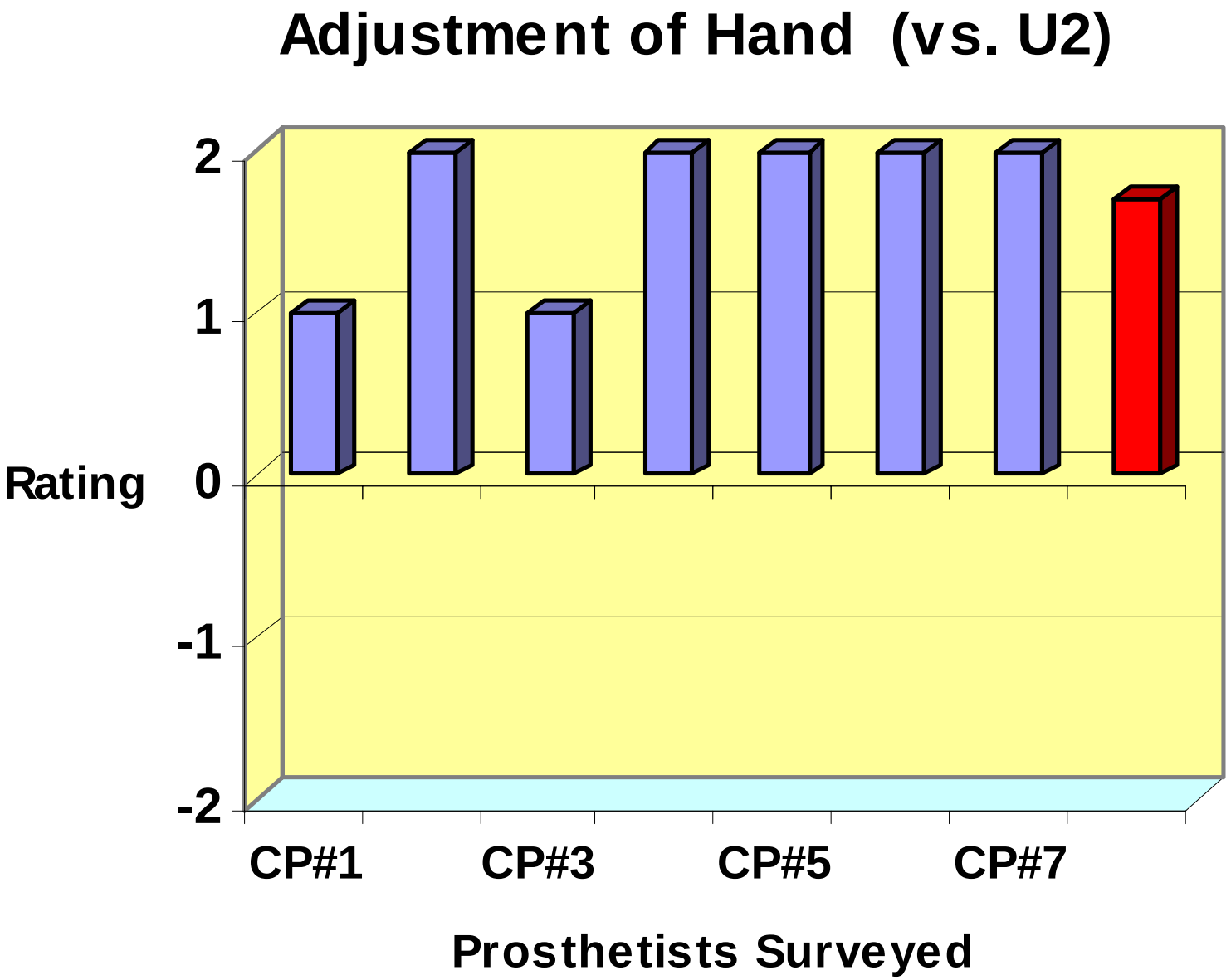
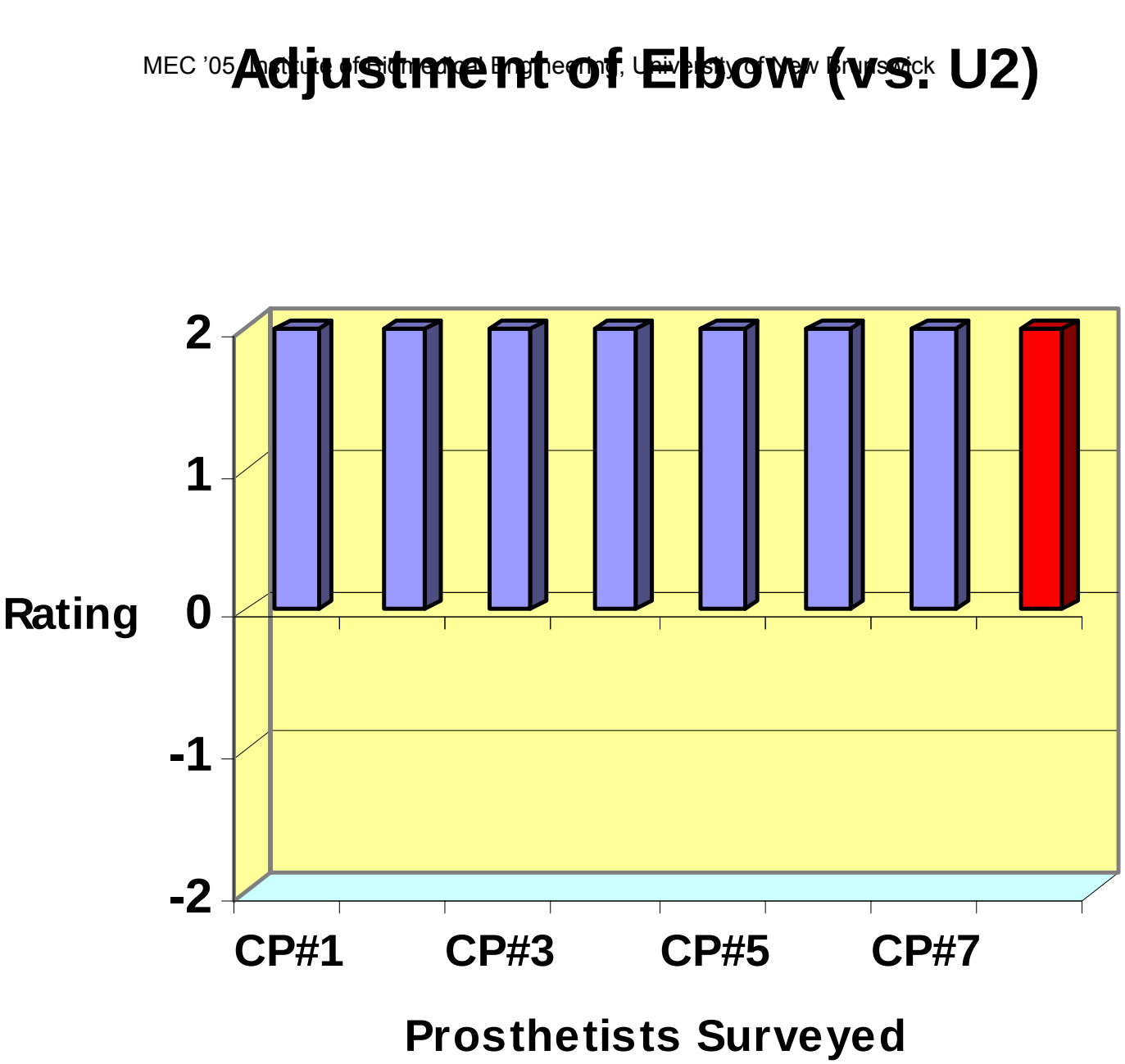


Figure 3: Patient using the U3 Arm learning to perform hand functions without locking the elbow

Results of Clinical Field Trials

A total of seven of the prosthetists involved in the clinical field trials of the Beta units were asked to rate the U3 compared to their previous experience with the U2. Overall, the U3 was rated as much easier to fit and adjust. Survey results follow. (The red bar at the end of each graph represents the average rating).


Adjustment of Hand (vs. U2)
Rating
2
1
0
-1
-2
CP#1
CP#3
CP#5
CP#7
Prosthetists Surveyed



Figures 4 & 5-Adjustment of Hand and Adjustment of Elbow graphs - The U3 arm changes the method of adjustment from physically turning a small potentiometer to adjusting a clearly visible parameter on a computer screen. Both Hand and Elbow adjustments were judged to be "Better (+1) or Much Better (+2), by all surveyed prosthetists.

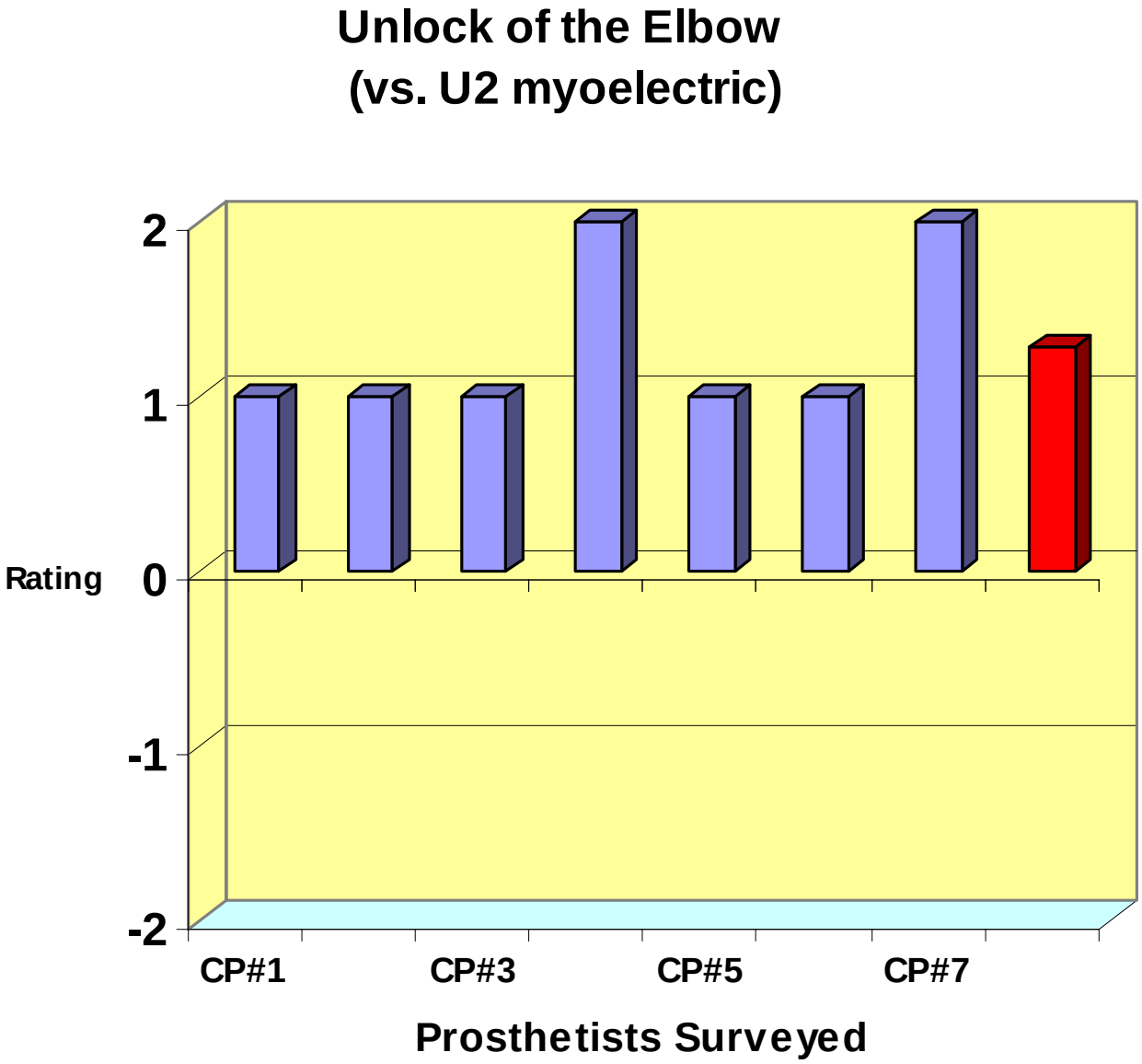


Figure 6- Unlock of the elbow graph - Using simultaneous control, the elbow has nearly always been controlled by a harness-mounted input. This allows unlock of the elbow by a easy-to-learn quick pull on the elbow sensor. This has been judged much easier for the patient to control, compared to muscle co-contraction, formerly the usual method of unlocking.

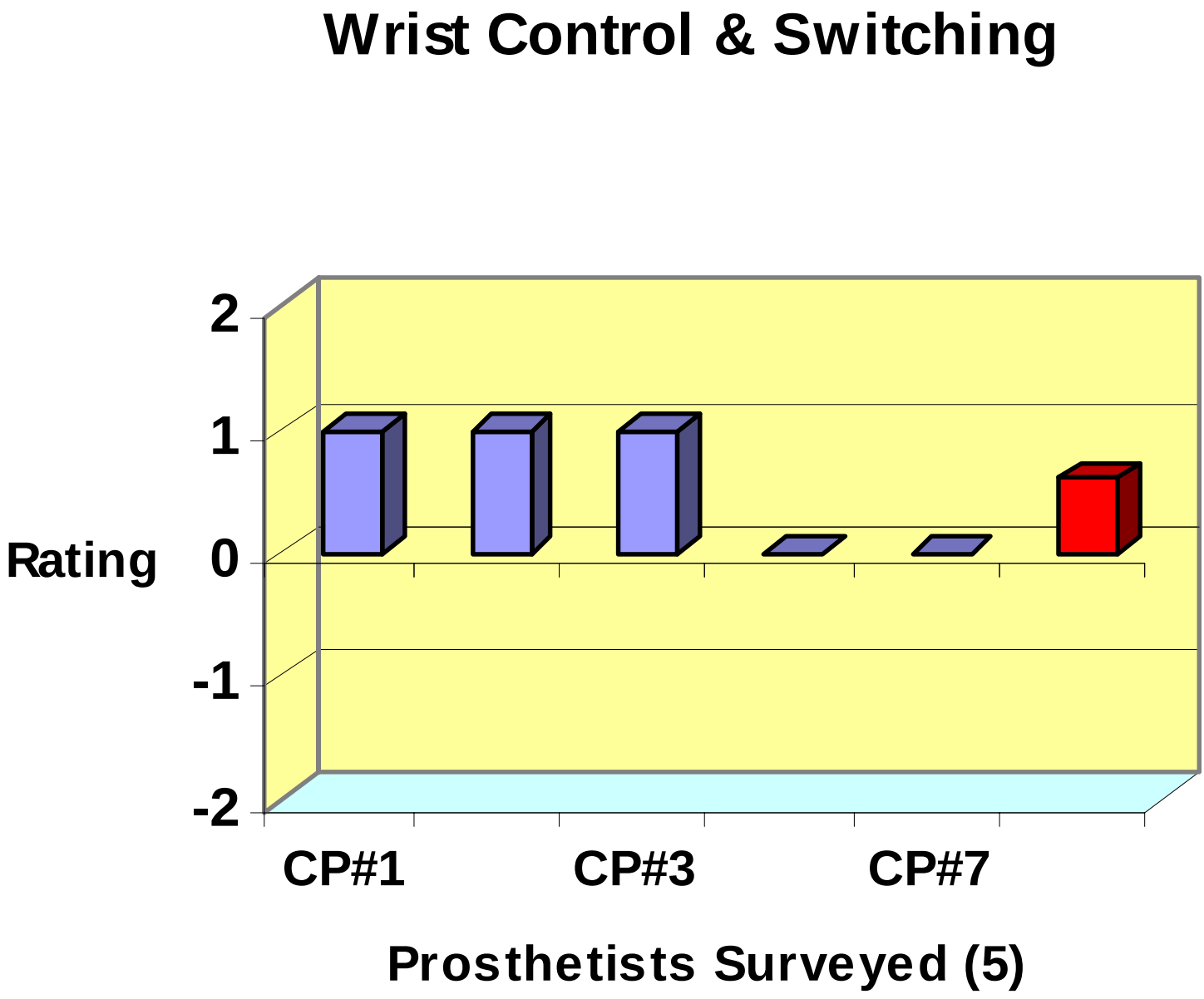


Figure 7 -Wrist Control and Switching - The U3 is capable of providing three powered functions, without requiring any external switches. The push or pull state switch formerly required to transfer control from hand to wrist may be replaced by muscle co-contraction, or Fast Access switching by each individual muscle. At the time not every prosthetist had experience with hand/wrist muscle switching, so the sample size was small.

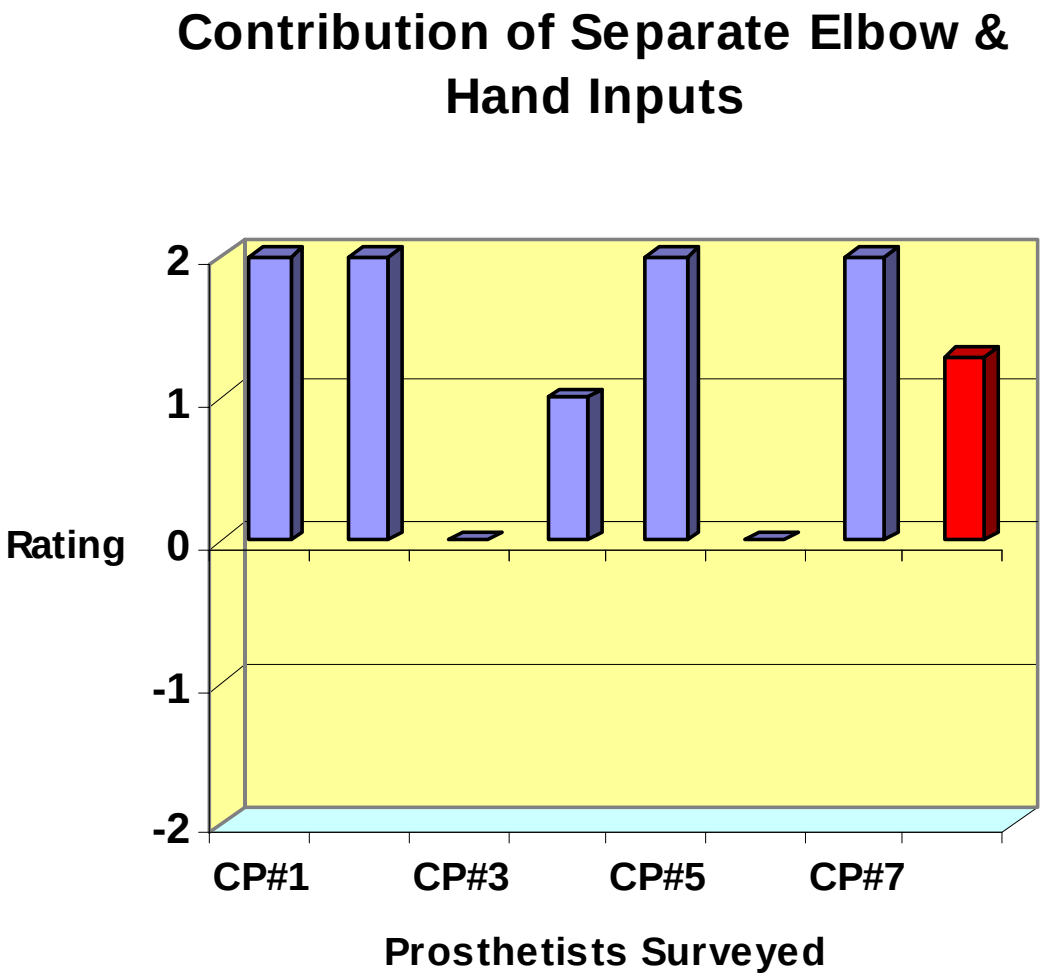


Figure 8. Overall, prosthetists found the contribution of separate Elbow and Hand inputs to be a significant improvement over the Utah Arm 2.

MEC 2005, Canada, August 2005

ADVANCES IN RESPONSIVENESS AND GRIP SPEED OF TERMINAL DEVICES BOOST THE DEVELOPMENT OF NEW UNIQUE CONTROL OPTIONS

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ABSTRACT

The SensorHand Speed, which was introduced last year, has set a new standard in grip speed and responsiveness. To achieve this goal, the hand received a new high performance drive unit, reinforced mechanical parts and new electronics.

The biggest task within the development process was the adaptation of current control strategies and the development of new control options, thus enabling the patient to handle the high speed and the unique acceleration of the new hand. Besides the DMC plus®, AutoControl and VarioControl options the SensorHand Speed offers a new control called VarioDual. VarioDual uses two signal sites and mimics the physiological activities used to control the sound hand. VarioDual combines the two classic control strategies EVO (Electronic Voluntary Opening) and DMC (Dynamic Mode Control).

The overwhelming echo from patients all over the world led to the development of a new Electric Greifer with control options similar to the SensorHand Speed.

INTRODUCTION

Until the end of the 1980s, all commercially available controls for electric hands were based on a simple digital control mode:

Once the myoelectric signal reaches a defined "On"-level the terminal device will start to open or close until an end position is reached or the signal drops below the "On"-level. This kind of control is still used for patients with very low signals that can only reach the "On"-level. Therefore, and only therefore, a digital control, such as the Digital Twin, is still one of the best options.

The problem with digital control is that it does not allow controlling the speed of a terminal device. The patient can overcome this to a certain extent by pulsing the control signals over the "ON" threshold but this creates jerky movements. Clinical studies have proven that the number of mistakes when trying to grasp significantly increases if the speed of a terminal device with digital control exceeds a certain limit.

The Introduction of proportional controls, which allow for control of the terminal device’s speed by the strength of muscle contraction, this time has been overcome. The door was opened for the development of faster components.

TECHNICAL LIMITS FOR THE DESIGN OF HIGH SPEED PROSTHETIC COMPONENTS

But, a new problem came up which once again limited the speed of the devices: The inertia of complex mechanical systems. In the case of myoelectric grasping devices due to the short distance between open and close the acceleration, both positive and negative, becomes much more important than the sheer maximum speed.

It is extremely important that the patient is able to control this high acceleration. Therefore, the whole system (mechanics, electronics, software, sensory) has to be fine-tuned in a way that the patient gets the feeling that he is directly controlling it with his muscles. Any delay of any of the system’s components would make a high-speed system such as the SensorHand Speed uncontrollable and therefore would lead to frustration of the patient. The faster an active prosthetic system is, the more important controllability becomes.

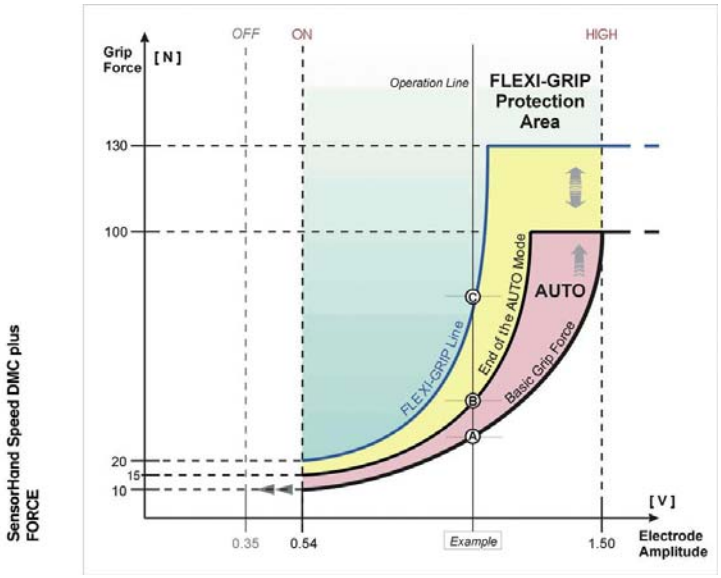
CONTROLS

Several different control schemes were developed which allow optimizing the interaction between patient and prosthetic system depending on the patient’s ability.

DMC plus

The classic one (and in most cases probably the best option) is the Dynamic Mode Control plus (DMC plus). This control, which uses 2 electrodes to open and close, independently allows for proportional control of speed and grip force.

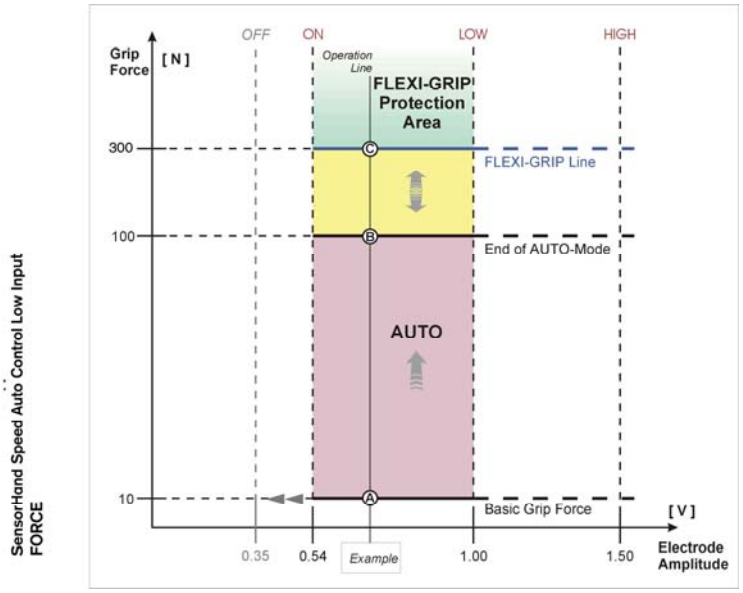
After gripping with maximum grip force the virtual “hand switch” is activated which requires a slightly higher open signal to open the terminal device. This improves gripping security and facilitates the holding of such things as eating utensils, because small signals, which are often fired of unintended, do not lead to a slackening. The DMC plus control requires two normal signals. Electrodes as well as Otto Bock Linear Transducers can be used for control.



AutoControl LowInput

The second option is called AutoControl LowInput. This is a control especially designed for patients with low myoelectric signals. The opening signal is used for proportionally controlling the opening of the device. The signal level for maximum speed has been lowered which decreases the window of speed control for opening. With Auto Control LowInput the closing of the TD only requires a digital level signal which closes the hand full speed. The control of the grip force is depending on the type of TD: Either the AutoGrasp functionality of a sensor system does the necessary adjustments or the patient himself controls it digitally.

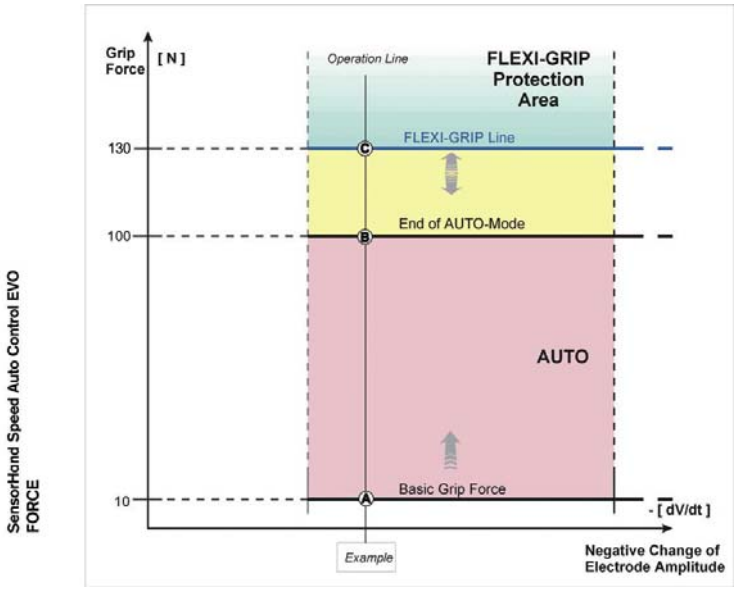
Alternatively, open or close could also be controlled with a switch.



AutoControl

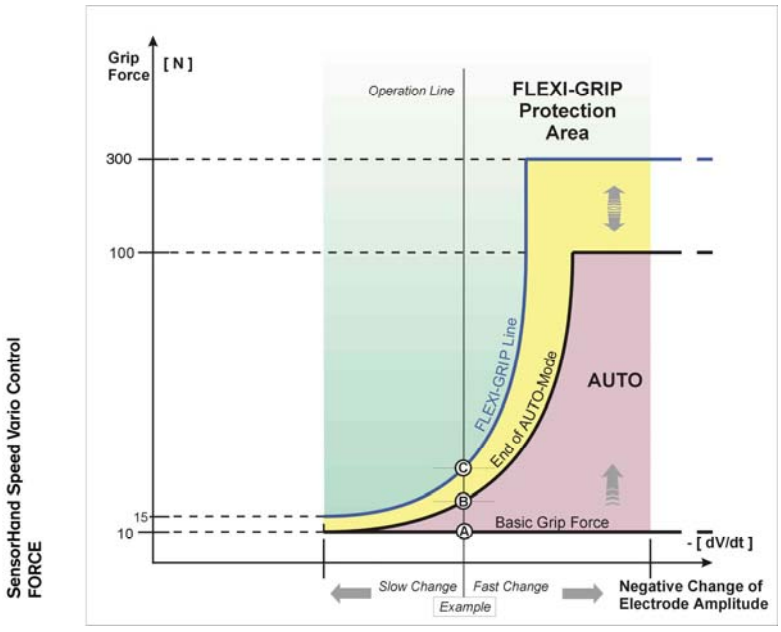
AutoControl is a control option exclusively available in SensorHand and SensorHand Speed. It basically works like the so-called “Cookie Crusher Control” and requires only 1 very low myoelectric signal. Once the opening signal reaches a certain level

the hand starts to open until the signal drops below the level. Then the hand closes automatically down to a minimum initial grip force. After that the Sensor System of the SensorHand kicks in automatically and increases the grip force to a level just high enough to securely hold the object in the hand.



VarioControl

VarioControl is the preferred control option for patients with just one good signal. Opening is controlled by a rising signal and the speed is dependant on the speed of the rising signal i.e. the faster the rising signal the faster the opening. Closing speed and initial grip force are controlled by speed and duration of the muscle relaxation. If the patient relaxes fast the TD will close fast and start with a high initial grip force. Whereas a slow relaxation will let the TD close slow and start with a gentle grip.

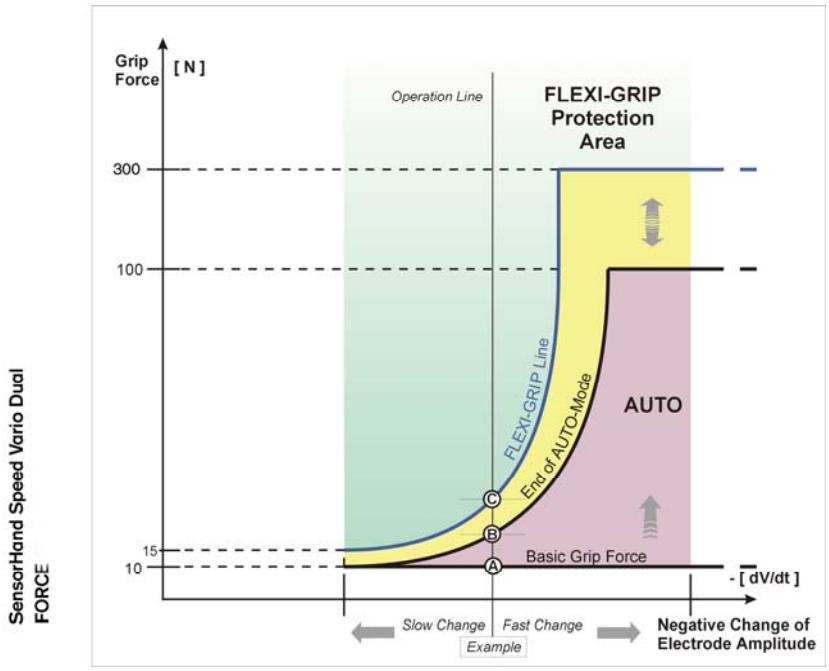


VarioDual

Last but not least a new unique control option has been implemented for the first time in the SensorHand Speed: VarioDual.

VarioDual has been initiated by the demands of several Northern American customers that asked us to develop a physiologically optimized control. VarioDual uses two normal myoelectric signals and like a sound hand it utilizes the signals from the extensor group to open the TD proportionally (similar to VarioControl). Contracting the extensor muscles will open the hand and depending on the speed of the muscle relaxation the patient can control the closing speed, just like a sound hand would close after relaxing the muscles which are needed to open.

To further increase grip force the patient contracts the flexor muscles. This signal is picked up by the second electrode and controls the TD´s speed and grip force proportionally.



SYSTEM ELECTRIC GREIFER DMC VariPlus

The overwhelming echo from patients and CPs from all over the world led to the development of a new Electric Greifer with control options similar to the SensorHand Speed: System Electric Greifer DMC VariPlus.

Although due to technical reasons the new Electric Greifer does not have a Sensor and its advantages, the new control options make it easier to switch back and forth between a SensorHand Speed and an Electric Greifer.

MyoSelect

To make choosing the right control option easier in the future, a new tool has been released: MyoSelect.

MyoSelect is plugged in directly to the prosthetic component. On a screen it gives you feedback regarding the component, the software version in the component and the currently activated program. By means of a small jog-dial on the side the CP may choose the desired program and depending on the program change some settings: E.g. with some of the available control options the SensorHand Speed allows to reduce the maximum speed.

The new System Electric DMC VariPlus has been set to DMC plus control in the factory. Any other control program can only be chosen by means of MyoSelect.

CONCLUSION

The progress in microprocessor technology allows new possibilities in the design of prosthetic controls. Like in all complex systems there are close interdependencies between all components. Progress in the mechanical part of the system often require further developments in the field of sensory, electronics and software and vice versa. A significant acceleration in the development of new prosthetic components is the result, bringing us closer to our goal: The improvement of the quality of life for people with disabilities!

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NEW MYOELECTRODE OPTIONS

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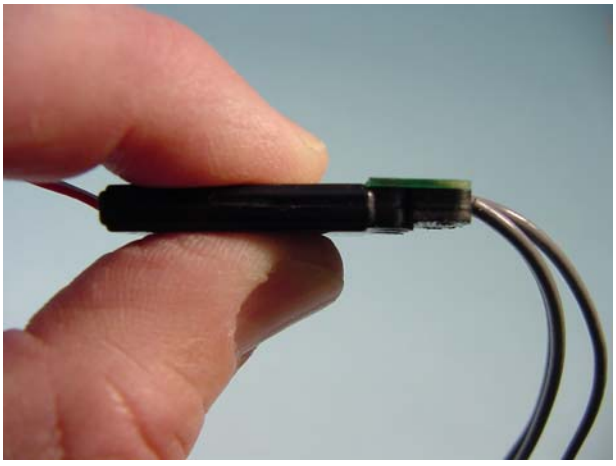
Practitioners have been requesting improvements in myoelectrodes to facilitate better client fittings. Greater sensitivity, better immunity to external interference, more and better (linear) gain adjustment and good common mode rejection are all desirable features. Mechanical changes were also needed to create more cosmetically appealing sockets. Preamplifier cases had to be small and thin and provided with electrode cables of various lengths. When LTI designed their Remote Electrode System, they addressed all of these issues. This new remote electrode system has improved performance and low-profile packaging. The CavityBack™ Electrodes are available in three sizes/shapes, so the prosthetists can choose the best for their client. In addition to the cables for test sockets and soft socket liners, cables can be adapted for snap-on attachment to roll-on silicone sleeves.

LTI Remote Electrode System:
Traditional cased electrodes have been used for years and until recently were the only choice for myoelectric signal detection. Often however, there is inadequate space for these electrodes and as a result they are placed in the socket over the muscle. Because of their thickness (") they often result in an undesirable bulge in the outer socket. This is cosmetically unappealing and sometimes rejected by users. The remote electrode-amplifiers (which are only 0.2" thick) can be placed anywhere in the vicinity of the muscle site and therefore can easily be concealed resulting in better cosmeses.

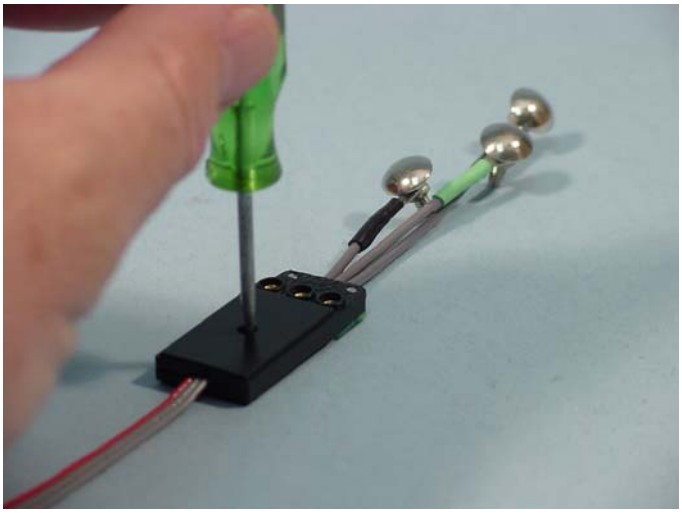
Test sockets are formed to the client's residual limb or torso and then electrodes are mounted to this socket for set-up and testing. Cased electrodes require the clinician to mold the electrode into the test socket surface, thus fixing their location. Remote Electrodes are mounted by drilling holes in the test socket and placing the metal electrode contacts over the intended muscle site. Since one of the purposes of a test socket is to locate the optimal muscle site, the remote electrodes facilitate this. They can easily be moved to a different/better location once the test socket is fitted to the client. After the best electrode location is established, these remote electrodes can be removed from the test socket and re-mounted on the definitive socket in the same location, thus assuring the best performance.

Electrode-Amplifier:



The electrode amplifier is the largest component in a myoelectric sensor system. With the Remote Electrode System, this component is separate from the metal electrodes and can be placed in a convenient location in the socket where it will not adversely affect the cosmetics of the prosthesis. Additionally, this case has been compressed to a thickness of just 0.2 inch (5mm), enabling it to fit in a cavity between the inner and outer socket if necessary.

The Remote Electrode Amplifier has a gain adjustment potentiometer similar to traditional cased electrodes. This can be turned with a flat jeweler's screw driver to increase or decrease the gain. For applications where a microprocessor-based controller is used, this electrode gain can be set at mid-range and further gain adjustments made through the controller software. This method allows minor "field adjustments" to be made by the clinician without a computer.



The LTI Remote Electrode Amplifiers have excellent linearity as shown here.

CavityBack™ Electrodes & Cables:

The unique design of the CavityBack™ metal electrodes allows them to draw the soft socket liner into the back, thus reducing the overall thickness and These are available in three sizes/profiles. The two adult sizes measure ½" in diameter with two different profiles – standard and deep dome. Standard dome electrodes are used for most applications or where bony protrusions are present. Deep dome electrodes are used when there is soft tissue or a deep muscle. These electrodes have significantly greater surface area for better signal detection and the deep dome penetrates the tissue better for acquiring good signals from difficult muscle sites. A third "pediatric" electrode is available for use in children's prostheses and sometimes for partial hands where space is limited. These measure ⅜" in diameter.

BIOMECHANIC ASPECTS AND PATIENT NEEDS LEAD THE PATH TO A UNIQUE WRISTJOINT FOR MYOELECTRIC PROSTHESIS

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ABSTRACT

The wrist plays an important role for the function of the human hand. The wrist joint is a complicated anatomical unit, consisting of various bones, ligaments and muscles. The distal and proximal joints combined with the rotation of the forearm excite movements at various levels that bring the hand into different positions.

Apart from the rotation, the newly developed wrist with low construction height for myoelectric prostheses allows for volar and dorsal flexion at various locking positions. Besides those typical functions which allow positioning of the hand, the joint must meet further technical requirements. Therefore the prosthetic wrist joint clearly differs from its natural model. The most important difference is the exchangeability of the terminal device. This flexion joint with its additional degrees of freedom in motion considerably enhances the radius of action and thus increases the practical value of the myoelectric system to the patient.

INTRODUCTION

The function of an arm prosthesis is primarily determined by the quality of the prosthetic hand. After all, the patient uses it to grasp objects and carry out different activities. To be able to grasp these objects the prosthetic hand must optimally be positioned. In the human hand these movements are mainly and often unconsciously carried out with the wrist joint. Even hand gestures are actively supported by the movements of the wrist joint. The patient, however, often has to integrate shoulder movements into the whole process of movement. To simplify the entire process of movement for the patient, a new flexion wrist unit for myoelectric prostheses has been developed.

ANATOMY

Anatomically, the human wrist divides into a proximal and distal joint (Fig 1). Both joints are responsible for volar and dorsal flexion. From the neutral 0-position, dorsal flexion, mainly produced by the distal wrist joint, normally is within the range of 35° to 60 ° and maximally possible up to 80°. The proximal wrist joint mainly supports volar flexion, which normally is in the range from 50°to 60° and maximally possible up to 85° [4]. The flexion movements are divided into four sectors (Fig. 2). The sector up to 20° (I) is the one mostly used. The ligaments remain relaxed with minor pressure forces being generated. In the sector up to 40° (II) movements are free. The ligaments are slightly tensed. In the sector up to 80° (III) the wrist joint reaches its physiological limit of movement. Movements in sector IV are always pathological and due to an overstretching or rupture of the ligaments [2].

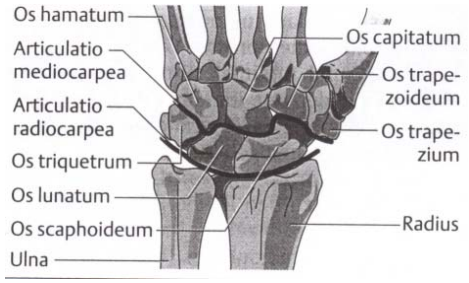

Fig. 1: *Proximal and distal wrist joint [1]*

Figure 3 shows the results of a study and depicts the frequency of the flexion angles of the wrist joint. It clearly shows that the movements assigned to sector II are the most often ones.

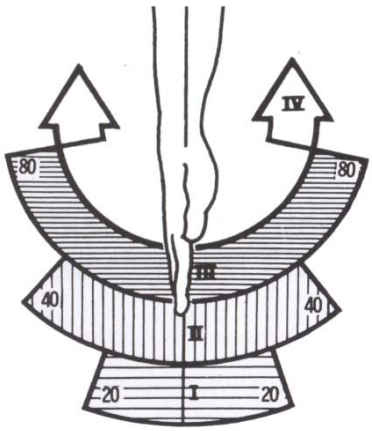

Fig. 2: *The 4 sectors of volar and dorsal flexion [2]*

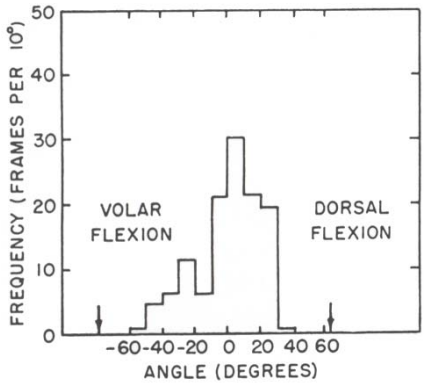

Fig. 3: *Frequency of extreme wrist-joint positions. Histograms obtained from analysis of 128 frames taken from kinematic studies of one average nonamputee subject performing 51 daily-living activities [3]*

Through shifting and rotation of the carpal bone, the hand can additionally be abducted radially by 25°-30° and 30° to 40° towards ulna [4].
The rotation movement of the hand does not have its origin in the wrist joint. But this movement is important for positioning the hand. Pronation and supination movements of the hand, for example, are produced by the proximal and distal radioulnar joint. From neutral 0-position the forearm can be pronated and supinated by 80° to 90°. With amputated forearm the rotation movement is severely limited (Fig. 4). Depending on the amputation level this movement in the forearm is still existent but cannot be adequately used with the prosthetic socket (Fig. 5).

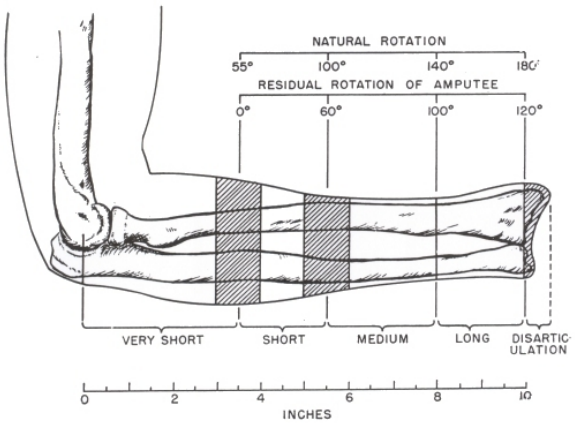

Fig. 4: *Below-elbow amputee types, based on the forearm length, epicondyle to styloid [3]*

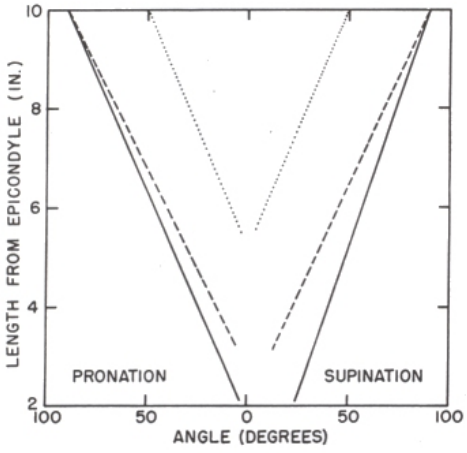

Fig. 5: *Forearm rotation. Solid lines, radioulnar rotation range of a skeleton; dashed lines, mean torsion of flesh in five none amputee subjects; dotted lines, estimated socket rotation, based on six amputees [3]*

With increasing amputation level and a prosthetic fitting the natural freedom of movement becomes more and more restricted for the patient. The aim is to restore this freedom of movement to the greatest possible extent by means of artificial components. With additional degrees of freedom a wrist unit can contribute to reduce this restriction.

WRIST UNIT

The artificial wrist joint of a prosthetic fitting offers several functions. On the one hand it connects the artificial hand to the prosthesis. Furthermore, a wrist unit allows different hands to be exchanged. To make it easier for the patient to grasp an object, the wrist unit brings the hand into the desired position. In this connection the rotation is a very important function for the patient. For a natural positioning of the hand a combination movement is required. Dorsal and volar flexion of the hand are helpful additional movements to rotation. Only these movements make it possible to grasp and hold objects with an expedient, overseeable position. Especially for above-elbow amputees and bilateral amputees this is a very important prerequisite for performing simple, daily activities. With artificial wrist units a radial and ulnar abduction of the hand can only be achieved with ball joints or, for example, with a Greifer.

RESULTS

To offer these advantages also to all other patients, a new passive flexion wrist unit for myoelectric prostheses has been developed. Using this flexion wrist unit in combination with the Transcarpal Hand allows achieving a length that does not exceed the structural length of a normal myoelectric hand. Based on the model of the natural wrist joint, this flexion wrist unit allows for volar and dorsal flexion of 40°. For this range of motion there are five locking positions in increments of 20°. In these locking positions, a static torque of 32 Nm / 283 lbf-in can be transferred. This corresponds to a weight to be carried in the hand (distance to the wrist unit approx. 12 cm / 4.7 in) of approx. 27 kg / 60 lb. A manual pressure on the switch located on the medial side unlocks the wrist unit. An integrate ratchet system ensures that the joint is held in the current position when it is unlocked and that the hand does not automatically flex due to its own weight. Moreover, the defined ratchet points help to easily find the locking positions.

Due to the low weight of the wrist unit and of the Transcarpal Hand, the weight as well is similar to that of a normal myoelectric hand. This combination can therefore easily replace the myoelectric hand of an existent prosthetic fitting.

In case of short below-elbow residual limbs this wrist unit can be used in combination with the endoskeletal system. Especially in the case of such fittings as well as for above-elbow amputees, a flexion wrist unit is advantageous. The range of motion of these patients is largely restricted and only with unnatural shoulder movements they are able to grasp some objects with their prosthesis. With a flexion wrist unit many objects are more overseeable and easier to grasp, for example when getting on clothes or doing the dishes.

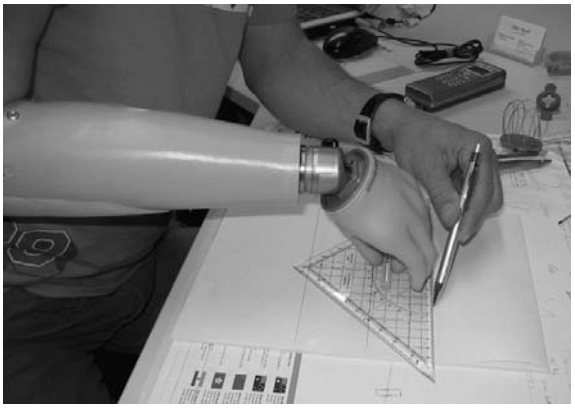

Fig. 6: *Patient with a new flexion wrist*

Up to now, a few patients have been fitted with such a flexion wrist unit. All patients immediately managed to cope with this new wrist unit and spontaneously found out new possibilities (Fig. 6). The switch is easy to manipulate due to its size. Even bilateral amputees were immediately able to manipulate the switch with their chin. The fact that the switch is located within the socket plane prevents unintentional manipulation of the switch.

CONCLUSION

A lower extremity amputation always leads to a restriction of the patient's range of motion. This range of motion should be restored to the largest possible extent by means of artificial components. The artificial wrist joint is an important component of an arm prosthesis. Among other things, the wrist unit provides for the optimal gripping position of the hand and thus for a harmonious movement of the whole system. The flexion wrist unit for the Transcarpal Hand enlarges the possibilities for the patient. Due to its low height and weight, this wrist unit can be adapted to a normal, existent system. It thereby completes the entire prosthetic arm fitting and makes it easier for the patient to use their prosthesis.

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NEW CONCEPTS IN EXTERNAL POWERED ARM COMPONENTS

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MEC 2005

INTRODUCTION:

An electrically powered joint can be of considerable support for many unilateral or bilateral above elbow amputees in every day activities. The DynamicArm (Photo 1) represents the next evolutionary step, a milestone in powered elbow prosthetics. It has been designed around previously proven components and also adds new developments that will change the way the amputee interacts with the prosthesis. Multiple control strategies and inputs running in parallel as well as hardware that allows for a more natural operation with much

Photo 1: DynamicArm system

of expectation from every patient wearing this design. This is already the case based on the response of the patients that have been fitted so far. Patients who have had many years experience with powered systems or those who have had experience with hybrid systems report the decrease in mental and physical energy within moments of being fit. Repositioning exercises, switching between components and actively controlling each component seem to be done with ease. Some of the advantages for the amputee are rapid movement and pre-positioning. This allows the patient to lift higher weights and position the elbow without complex control sequences. Because pre-positioning is more easily facilitated the patient will chose to move the component rather than make up for some controllability deficit with a compensation movement of the trunk. High loads can also be held in position without fear that something will be damaged if it is overloaded thanks to the already known locking mechanism of the 12K50 ErgoArm. Built in force sensors protect the system from unlocking in undesirable conditions. The control characteristics follow those of previously designed components like the DMC+ and the SensorHand Speed. Otto Bock electronics follow the logic of „First Signal Wins" control strategy but a new improvement allows for fast changing between opening and closing the hand or changing from flexing to extending or from supination into pronation. This gives the advantage of being able to fit the patient who co-contracts but still allows for high performance controls. Signal inputs are now digitized by the locking microprocessor and from there are not as susceptible to Electromagnetic Interference. Power consumption

has been reduced because when the system is in free swing, the mechanics are completely de-coupled from the movement of the Automatic Forearm Balance mechanism. New possibilities of control include being able to raise and lower loads at various speeds and repositioning without needing to produce an "unlock" signal. The wrist rotator is controlled by the main electronics which allows for an ultra sensitive control and the possibility to run the wrist motor proportionally and at incredibly low speed allowing for very precise positioning. The terminal devices as before can be controlled with proportional speed and grip force based on the strength of the muscle signal which is supported by having the electronics combined with the hand mechanics. Simultaneous control of the hand and the elbow with using a variety of input devices is now possible. And now for the first time new input devices and new switching methods allow for more flexibility for the amputee to control the prosthesis.

TECHNICAL DATA:

LiIon battery 1900 mAh internal
268°/sec full ROM in 0.5sec
960g overall weight of the elbow without hand
Maximum live lift of 12lbs
ROM 15° - 145°
4 Microprocessors with Axon Bus® technology
Max load = 50ft/lbs

The mechanics are built around the known AFB mechanism which is at its core. The frame of the AFB is constructed of Aluminum to aid in dissipating heat generated by the lifting motor.

LIFTING MOTOR:

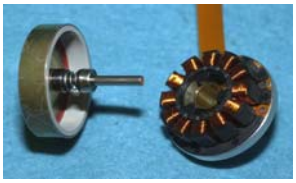

Photo 2: Lifting motor

The lifting motor is a flat brushless disk motor that was designed by Otto Bock Vienna. It has a high torque moment with a flat construction which makes it light weight and takes up little space. Without brushes means it has a very high life cycle and with the synthetic rotor it allows for a very low inertia for a quick start of the motor. A specially designed cover is designed with cooling fins since the spool creates a lot of heat. Overheating is protected by monitoring the temperature with a sensor which shuts down the system if the temperature rises above 80 degrees Celsius.

VARIO-GEAR

The DynamicArm also introduces a new concept in transmission controls for powered elbows with the Vario-Gear (Illustration 1) continuously variable transmission. It is a miniaturized design based on those that were used in large printing presses. The gear ratio can be adjusted between 2:1 to 18:1 based on the input from the force sensor built into the forearm and the control input signal generated

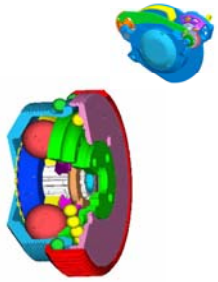

Illustration 1: Vario-Gear

by the patient. When the DynamicArm senses a load the Vario-Gear seamlessly adapts to avoid any abrupt movement for precise controlled lifting. With this Vario-Gear the drive and the output can be completely decoupled which allows for free swing with natural movement that does not consume precious battery power and is very quiet. The Vario-Gear is driven into position by a Servomotor which is controlled by the motion processor in the main electronics. Again based input from the force sensor and the control inputs from the patient, the servo-motor positions the Vario-Gear for correct operation and is adjusted 100x per second. Central to the Vario-Gear and the lifting motor is the Automatic Forearm Balance mechanism. This well known concept has various levels of compensation based on the exact flexion angle. In complete extension it compensates very little which is another reason that free swing functions the way it does and as the arm is flexed the compensation increases until approximately 90 degrees when the compensation again decreases to prevent uncontrollable acceleration to full flexion. To engage free-swing the patient must bring the elbow within 15 degrees of full extension and the locking unit will not lock the elbow. In order to relock the elbow at this point the patient only needs to give a small flexion signal and it will lock automatically when the signal stops.

LOCKING UNIT:

The locking mechanism is known from the ErgoArm family of elbows. Its infinite position lock and easy operation make it a natural selection for the DynamicArm. The movement of the forearm and the locking action are harmoniously coupled so that the patient does not need to produce an unlock signal only a control signal to initiate either flexion or extension.

AXON-BUS®

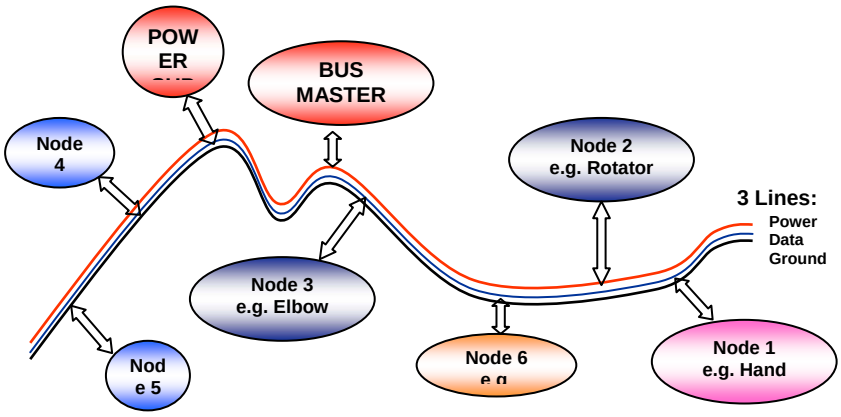

Illustration 2: AXON-BUS®

For the first time in prosthetics is the Adaptive eXchange Of Neuroplacement data AXON-BUS® (ill. 2) has been implemented. In this application all components can be communicated with and programmed independently and si's for minimized EMI, automatic recognition of the connected components and for automatic error recognition.

ELECTRONICS:

The electronics include a combination of 4 microprocessors; the lock processor with the analog to digital signal converter, the signal processor which controls the AXON-BUS®,

gets input from the sensors and controls the Bluetooth interface, the motion processor which controls the lifting motor, servo-motor and the wrist rotator, and finally the hand processor which is located in the terminal device.
The input devices are connected to the Easy Plug connector at the top of the electronics for the elbow lock. This signal is digitized and sent to the main electronics through the AXON-BUS® system and can then simultaneously go to the hand or wrist or lifting motor depending on the chosen control. It does not however need time to be analyzed by each component thus slowing down the signal. There are five sensors in the system including the angle sensor, servo sensor, force sensor, battery monitor and temperature monitor.

CONCLUSION:

The progress in microprocessor and mechanical technology allows new possibilities in the design of prosthetic systems. Like in all complex systems there are close interdependencies between all components. Progress in the mechanical part of the system often require further developments in the field of sensory, electronics and software. A significant acceleration in the development of new prosthetic components is the result, bringing us closer to our goal: The improvement of the quality of life for people with disabilities.

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TouchEMAS will demonstrate their prosthetic ideas in relation to a range of modular components which the company plan to launch over the coming year. These will include a multifunctional hand, electric elbow, wrist flexor and electrical shoulder

USING GROSS MOTOR ACTIVITIES TO ASSESS UPPER LIMB PROSTHETIC FUNCTION

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BACKGROUND AND RATIONALE:

For children with unilateral below elbow limb loss, many of the activities for which a prosthesis is of value involve gross motor function during play. To assess function in these activities we studied normally limbed children and children using below elbow powered prostheses while they performed various tasks. This paper is focused on two tasks, swinging on a swing, and zipping up a vest. The study had three primary objectives:

1. To collect data for normally limbed children to provide baseline data,
2. To collect data for children who use either a single degree of freedom friction wrist or an "omniwrist", and
3. To compare the results for users of the two types of wrist to their normally limbed peers.

The two tasks chosen represent two different types of activity. The swing has a cycle time or rhythm which is determined principally by the length of the ropes. As a result the child who is swinging must react to the activity on a time scale which is out of their control. By comparison the zipper task is one where the child can simply slow down, to whatever extent needed, in order to be able to perform it. The strategy of taking more time is a common one if activities are difficult to accomplish.

METHODS:

The populations reported here include a convenience sample of 20 normally limbed children with an age range from 4 to 13 years. The ages are not uniformly distributed and 9 of the 20 children were 7 years old. The prosthesis wearers include 6 children as described below.

Subject ID	Gender Male/ Female	Age (yrs)	Height (cm)	Weight (kg)	Affected side	Type of wrist	Daily prosthesis wear time (hrs/day)	Length of time wearing omni wrist
PFR04a	Female	4				Omni		
PFR06a	Female	6	120.0	24.8	Left	Omni	8 (school only)	12 months
PMR08a	Male	8	132.1	29.5	Left	Omni	12	8 months
PFL12a	Female	12	178.4	70.0	Right	Friction	0-4	N/A
PFR13a	Female	13	160.0	58.6	Left	Friction	8 (school only)	N/A
PML09a	Male	9				Omni		

Children in the study were dressed in a tank top or similar garb which exposed the arms and shoulders. Retro-reflective markers were applied over bony landmarks as indicated in the table below. Marker position data was collected using a VICON M-Cam system with 8 cameras, and a video recording to assist in visualization.

Acronym	Anatomical Landmark	Marker Diameter (mm)
Frhd	front of head	25
Lfth	left side of head	25
Rth	right side of head	25
c7	Vertebra c7 in spine	25
Lcla	left sternoclavicular joint	14
Rcla	right sternoclavicular joint	14
Lsho	left shoulder	25
Llel	left lateral epicondyle	14
Lrad	near left styloid process of radial	25
Lulna	near left styloid process of ulna	25
L2mc	left 2nd metacarpal head	16

Marker locations and sizes (22 markers in total)

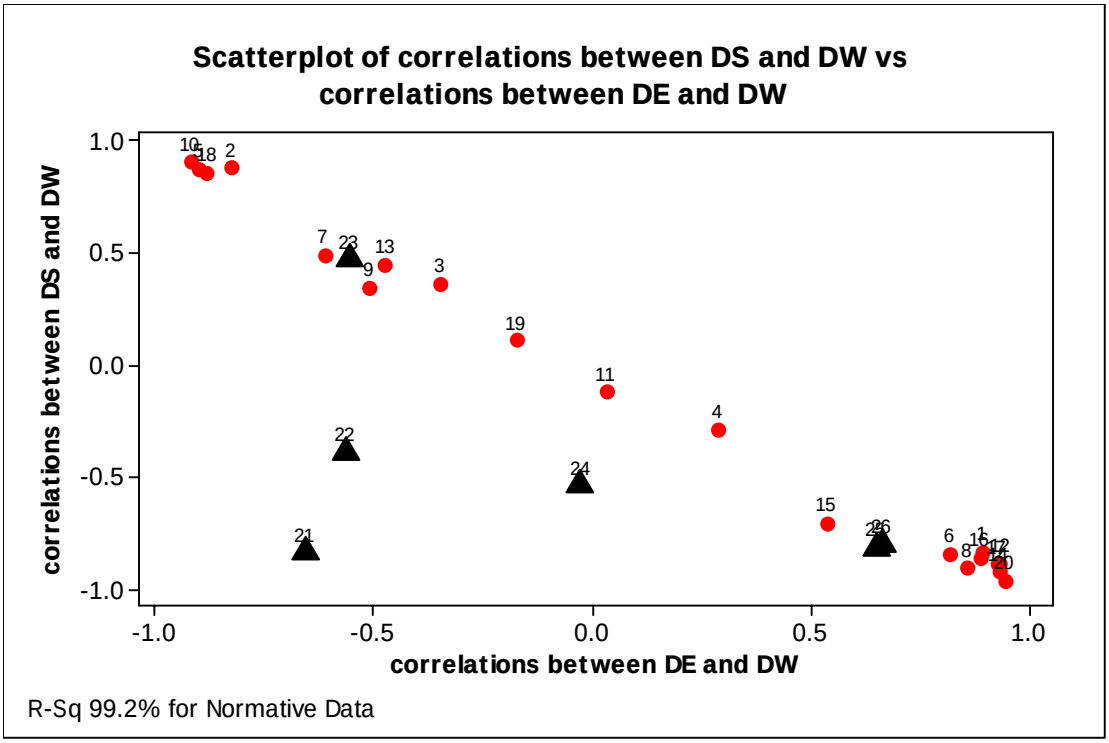
Children participating in the study were familiarized with the laboratory and then performed four different gross motor tasks while data were collected. These included riding a cooter, swinging on a swing, zipping up a vest and batting a T-ball. We report here the results of the "swing" activity and "zipper" activities. For the swing, the children were seated in a playground type swing attached to the ceiling of the laboratory. The children were instructed to pump until they were swinging actively at which point data were collected. A cycle was defined as the time from the swing being at the maximum "back" point, through the downswing and back to the original "back" position. The zipper activity involved putting on a vest and then beginning with the hands at the child's side, moving to insert the zipper ends together and zipping it up closed and then open with the ends disconnected. For this task, the time from initiation to the beginning of the zipping motion was considered separately from the actual zipping motion since there was a very substantial difference in the time taken by the normally limbed children to put the ends of the zipper together as compared to how quickly the children using prostheses could do the task.

The data were manipulated using MatLab codes to obtain joint angles and other measures such as difference in shoulder height. Statistical analysis was done using Minitab. Children using the "omniwrist" or single degree of freedom wrist were compared to the group of normally limbed children. The dominant side for children using prostheses was taken to be their sound side.

RESULTS:

Due to limits in space we present only two sets of results. For the swing task we found the children using prostheses have different and inconsistent motions as compared to normally limbed children. To address this, we have examined correlation between dominant shoulder and wrist, and dominant elbow and wrist motion. The figure below shows the correlations between the dominant shoulder and wrist on the vertical axis and the dominant elbow and wrist on the horizontal axis during the swing activity. A value near 1.0 means that both joints are flexing in

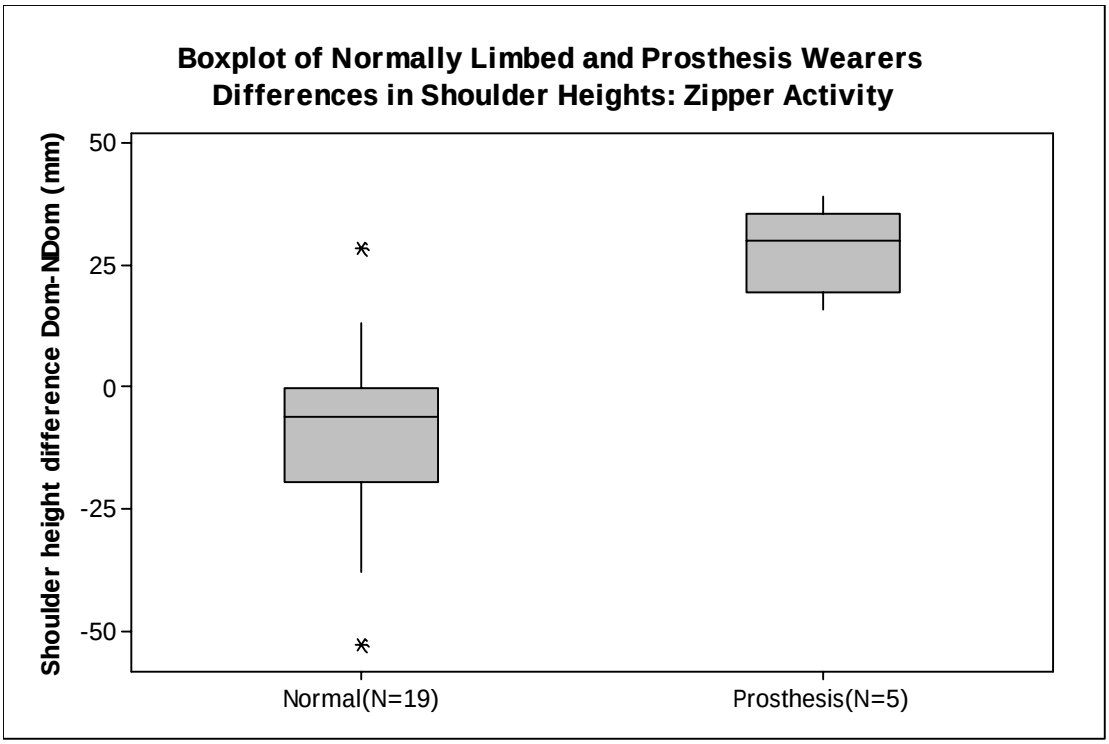
unison, a value near 0 means that they are not moving in a coordinated way and a value near -1 means that as one joint flexes, the other extends. We compare the sound limbs of the prosthesis users since this is a measure of whether the coordination of motion is affected by the compensations required by the prosthesis.



Correlations for the swing activity. "dots" represent normally limbed children, "triangles" represent the sound side for prosthesis users.

The data for the normally limbed children fall in a tightly bounded line which is either very correlated in both motions or not. None of the 20 normally limbed children had a result which fell away from this axis. The OmniWrist users (#25 and #26) fall within the normal band as does #23 a single axis wrist user. However, three of the single axis wrist users (#21, #22 and #24) demonstrate patterns which are not coordinated at all like the normally limbed children. While our numbers of subjects are small at this point (we are continuing to enroll both additional normally limbed children as well as prosthesis users) these results suggest that the Omniwrists, at least in these children, lead to a more normal pattern of motion and less overt compensation.

For the zipper activity, there were no strong differences between the two types of wrist, but the results did indicate that the children using prostheses took nearly 3 times as long to connect the ends of the zipper as did their normally limbed peers. They also assumed a rather different posture as shown in the figure below. Normally limbed children tended to have their non-dominant shoulder higher than their other shoulder. The prosthesis users were completely opposite with the dominant shoulder higher resulting in a more awkward posture held for a longer period of time. Such awkward postures and their related non-normal muscle loading may contribute to overuse injures or other problems over time.



DISCUSSION:

This work has demonstrated that patterns of gross motor activities can be quantified in children, providing useful information about differing strategies of motion. The activities which we have reported here can be difficult for some unilateral prosthesis users to do without their prosthesis. As such they provide a fun and challenging way to assess how effectively the prosthesis compensates for the loss of the limb.

The swing activity shows that the interactions between shoulder, elbow and wrist motion in normally limbed children produce different patterns but all of them share certain, quite consistent, interactions. The sound side motions of the prosthesis users frequently demonstrate interactions which are distinctly different from the normally limbed. These "non standard" patterns involve awkward postures and may leave these children at risk for repetitive strain injuries over their lifespan. The Omniwrist users, while not without some awkward postures, followed a much more normal pattern of interaction in the motions of their sound arms.

The zipper activity did not discriminate between the two wrists but did demonstrate that the prosthesis wearers adopted a significantly different posture in order to accomplish the task.

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DESIGNING A SHOULDER SOCKET FOR USE OF FIVE FORCE SENSING RESISTORS

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A patient combining transhumeral amputation and brachial plexus injury presented an opportunity to revisit the design of a frame sockets for control using multiple FSR's. The superior socket wings bear on the user more medially than with a myo frame leaving all the anatomy at the acromion free to move in two directions. Both the initial socket and the definitive used adjustable-height mounts for the five FSR's. Key to the initial fitting was a separate clear-plastic arc for the FSR's. The ends of the arc and the attachment angle could be varied along with the location and depth of the individual FSR's. The fitting was a success and the user returned to work.

A SIMPLIFIED CAD MODEL OF TRANS-RADIAL SOCKETS SUITABLE FOR SUBJECT ENERGY EXPENDITURE ASSESSMENT

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1. INTRODUCTION

The measurement of the mechanical energy expenditure of a subject in accomplishing a given motor task has been reported as a valuable index to quantitatively assess his/her motor ability or pathology stage [1-3]. In order to compute this parameter applying inverse dynamics techniques [1], the subject's joint kinematics during the motor task and the inertial parameters (i.e. the mass, centre of mass and inertia matrix) of his/her moving limbs must be known. While the joint kinematics can be measured *in-vivo* using a motion analysis system (e.g. an optoelectronic system), the inertial parameters of *human* limbs are usually retrieved from anthropometric tables [1]. When the subject acquired is an amputee fitted with an *artificial* limb the problem of determining the prosthesis inertial parameters therefore arises. Considering a trans-radial amputation levels, a prosthesis is composed by standardized parts (hand, battery, lamination ring and actuators), and the subject-specific inner and outer sockets: since the CAD models of these parts are usually unavailable their single and cumulative inertia parameters remain unknown. The aim of this work was therefore to propose two possible simplified CAD models for trans-radial prostheses and to identify among them which one leads to the best estimation of the mechanical energy expenditure during a flexion-extension of the elbow in the sagittal and horizontal plane and during a shoulder internal-external rotation. The models, of increasing complexity, take into account the sockets, battery and lamination ring.

2. MATERIALS AND METHODS

2.1 Gold standard

As a gold standard and basis to develop the two simplified CAD models, the inner and outer sockets of three amputees where digitalized using a high-accuracy laser scanner (Minolta VIVID 900, resolution: 25pts/mm^2, accuracy: 0.25mm). Since the scanner was unable to fully digitize the inside of the sockets, only the external surfaces were retained from the scans. For all the three prostheses, measurements on the assessable parts showed a 3mm thickness for the inner and 2mm for the outer sockets. These offsets were then applied to the external surfaces to obtain the solids shown in Figure 1. Using the CAD software Rhino 3.0 (Robert McNeel & Associates), the inner parts were visually fitted into the corresponding outer parts, inspecting parallel sections of the solids to exclude interpenetrations. The obtained inner-outer sockets

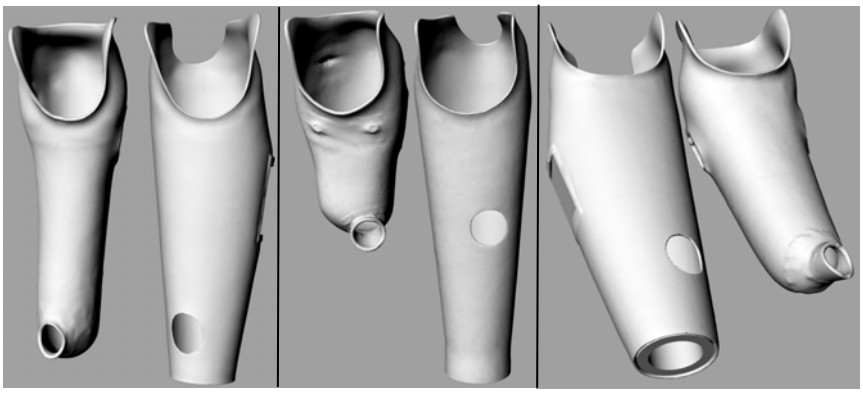

Figure 1 Gold standard models GS1 (left), GS2 (middle) and GS3 (right), this last shown with lamination ring and battery.

couples will be referred hereinafter as DS1, DS2 and DS3. For each couple, a hollow cylinder (d_i= 3.6cm, d_e= 5.4cm, h=2cm) was generated distally to model the lamination ring Otto Bock 10S1. For models DS1 and DS3, a parallelepiped of 1.5cm height was generated into the battery housing digitized in the scans to approximate the battery Otto Bock 757B21; this was done by "virtual palpating" on the CAD models the three corners of the battery housing to define the base of the parallelepiped and then imposing the 1.5cm height. No battery model was added to DS2 since for this type of sockets no fixed battery position is prearranged. The sets formed by a DS, a lamination ring and a battery (where applicable) will be referred hereinafter as GS1, GS2 and GS3.

2.2 Simplified CAD models design steps

Given the high complexity of the shapes of the sockets, we developed two simplified models characterized by increasing details: a first order approximation based on truncated cones (TC) and a second order based on the combination of truncated cones and tubular shapes (TS).
The conceptual design steps were the same for both: 1) definition of a set of easily identifiable geometric landmarks on the outer socket only; 2) "virtual palpation" of these landmarks on GS1-GS3 in Rhino; 3) creation, for every GS, of TC/TS design curves through these points; 4) application of solid modelling techniques to obtain TC/TS, generating both an outer and an inner socket model; 5) generation of the lamination ring into the models; 6) generation of the battery. Since the two models are were generated directly on every GS, there was no need to actually perform steps 4) and 5) since the lamination ring and battery models would have been the same generated for the corresponding GS. In the following two sections the basic features of TC and TS are given. Further details can be found in [4].

2.3 Generation of the TC model

The 7 landmarks selected for the construction of TC are shown in Figure 2a: A-B-C on the wrist edge forming a circle, D at the most distal point of the slot between the outer and inner socket, E-F where the lateral and medial elbow epicondyles are located, G at the point of maximal curvature on the proximal left or right crest of the socket, depending on which of them is the furthest from the wrist. The outer socket was obtained by designing an hollow truncated cone (2mm thick) using points A-B-C-E-F-G (Figure 2b). The inner socket was derived from the outer by offsets (3mm think), adjusting its length to have D as the most distal point. Only the distal bottom of the inner socket was closed. Taking into account the construction of the battery, TC required the identification of 10 landmarks. The construction of TC was realized for all the GSs, obtaining TC1-TC2-TC3.

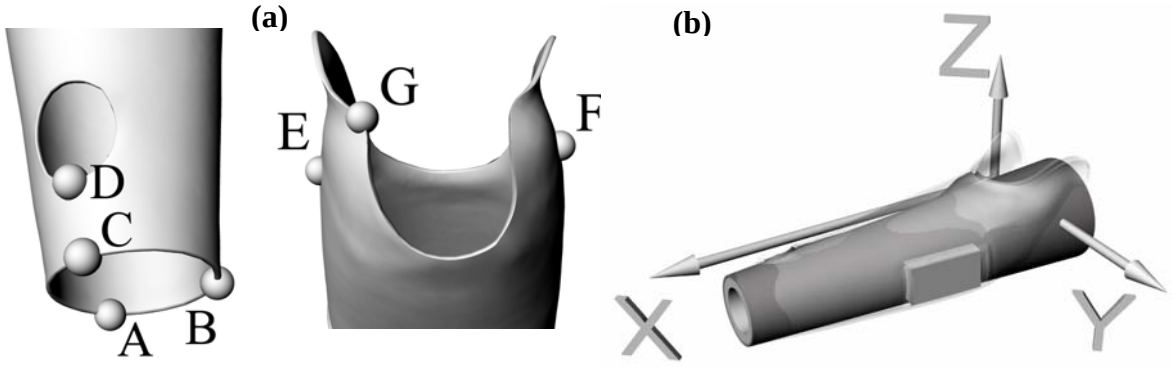

Figure 2 Selected landmarks for TC construction (a) and TC outer socket for GS1 (b)

2.4 Generation of the TS model

The 15 landmarks selected for the construction of TS are shown in Figure 2a and 3a: A-B-C-D are the same used for TC; E-F-G form a circle approximately parallel to the A-B-C cycle but at the edge of the stump insertion hope; M-N-O-P are locate at the top of the corners of the proximal aspect of the socket; H-I lie at the end of the straight edge forming the slot for the olecranon; L lies midway on the socket edge between E and M. The A-B-C and E-F-G cycles were used to build the hollow truncated cone of the outer socket. The E-F-G, E-H-I and L-M-N cycles were used to build the tubular shapes forming the proximal part of the outer socket. The same points were used to define 3 cutting planes to design the olecranon slot. After imposing a 2mm thickness, the solid of the outer socket shown in Figure 3b was obtained. The inner socket was derived from the outer, imposing a 3mm thickness, and adjusting its length to have D as the most distal point. Only the distal bottom of the inner socket was closed. Taking into account the construction of the battery, TS required the identification of 18 landmarks. The construction of TS was realized for all the GSs, obtaining TS1-TS2-TS3.

Figure 3 Proximal landmarks for TS construction (a) and TS outer socket for GS1 (b)

2.5 Outcome measures

For the comparison, each triplet $(GS\text{-}TC\text{-}TS)_{i=1;2;3}$, was set in the Rhino global reference frame (G) to resemble a forearm when the elbow is flexed 90°, with the palpated lateral and medial epicondyles laying on the ${}^{G}y$ axis, and the normal to the A-B-C circle parallel to the ${}^{G}(xy)$ plane (Figure 2b). In this configuration, ${}^{G}y$ replicated the elbow flexion-extension axis [5]. For each triplet, the mass, centre of mass and inertia matrix with respect to G of GS_i, TS_i and TC_i were computed using the Rhino "Mass property" function and custom-made MATLAB (The Mathworks) functions. A mean density of 1109 Kg/m^3, 900 Kg/m^3 and 1897 Kg/m^3 was assumed (based on measured weights and volumes) for the sockets, lamination ring and battery, respectively. Using these data and applying the torques law of dynamics, the energy mechanical expenditure (MEE) [1] during an elbow complete flexion-extension in the sagittal (FLEX) and horizontal (FLEXh) plane was computed and that of TS_i and TC_i was compared to the one of GS_i. MEE was obtained by integrating the absolute value of the power consumption over the simulation time. In particular, FLEX and FLEXh were simulated by imposing a cosinusoidal rotation (amplitude: from -70° to 70°, frequency: 280°/s) for 1 second, around ${}^{G}y$ considering, for FLEX, and not considering, for FLEXh, the gravitation force along $-{}^{G}z$. Finally, the MEE was also computed for a shoulder internal-external rotation movement (INEX), assuming the elbow fixed at 90°. This movement was simulated by applying the same motion described above around ${}^{G}z$.

3. RESULTS AND DISCUSSION

Results for the different MEEs are reported in Table 1. As can be observed, TS errors in MEE computation ranged from 2% to 8.5%, with best performances for FLEXh and INEX, i.e. in the trials where the inertial forces, and thus the estimation of the inertia matrix, were the most important factors. On the contrary, TC gave the best results where the gravitational force was considered, due to its systematic underestimation of the gravity lever arm (-0.5%– -2%) which

compensated the systematic overestimation of the mass (2.5%–12%). TS always overestimated the lever arm (0.7%–1.3%) and the mass (5%, 7.5%), with the exception of the second triplet (mass:-2.5% and lever arm: +7.8%). The MEE errors of TC ranged, on the whole, from 2.5% to 12%. TS thus ensured the overall best performances. However, the use of TC appears to be the best practical choice, considering 1) the time required to digitize the 18 landmarks of TS, 2) that the differences from TC and TS are limited to 4% in the worst case, 3) the absence from the model of a prosthetic hand, which - being about 500g located distally - will reasonably reduce the percentage MEE error.

The generation of every TS_i and TC_i directly on the respective GS excluded the problem of the spatial registration of the models on the gold standard. It is important here to notice that the procedure used for the models generation is applicable even if no high-accuracy models are available. The "palpation" of the landmarks can in fact be easily done using a stereophotogrammetric system directly on the real prosthesis, either applying micro-markers on the landmarks, or using the CAST technique [6]. The procedure thus appears to be applicable in routinely motion analysis.

	GS1 [J]	TC1 % error	TS1 % error	GS2 [J]	TC2 % error	TS2 % error	GS3 [J]	TC3 % error	TS3 % error
FLEX	1,71	2,51	6,01	1,19	10,23	4,79	1,55	6,08	8,37
FLEXh	0,87	4,07	5,39	0,64	9,18	3,94	0,75	9,03	8,48
INEX	1,02	4,45	4,64	0,68	10,94	2,09	0,83	11,89	8,45

Table 1 Results for the MEE computation for the three prostheses. Results for TCs and TSs are expressed as percentage errors of the values reported for the corresponding GSs.

4. CONCLUSIONS

The aim of this work was to identify the level of detail required in modelling a trans-radial prosthesis (formed by the socket, the lamination ring and the battery) to properly compute the mechanical energy expenditure of a subject during a flexion-extension of the elbow in the sagittal and horizontal plane and during a shoulder internal-external rotation. The two simplified model proposed, when compared to gold standard models, proved to reproduce the real mechanical energy expenditure within a 12% error for the less detailed model and 8% for the most detailed, in the worst case. Given these limited difference, the simpler model appears to be most practical choice for energy expenditure assessments. Future developments will be intended to evaluate the effect of a prosthetic hand on the models, and to measure the differences in the estimated mechanical energy expenditure at the shoulder level during activities of the daily living.

5. ACKNOLEDGMENT

The authors are grateful to ENEA and Ing. Sergio Petronilli for having managed the laser scanner acquisitions.

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THE EFFECT OF CONTROLLER DELAY ON BOX AND BLOCK TEST PERFORMANCE

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INTRODUCTION

Multifunctional prosthesis controllers have shown higher classification accuracies when EMG feature extraction and pattern recognition are performed on time windows of longer duration [1] (see figure 1). However, there is a limit to the time over which EMG data can be collected and analyzed before the delay causes the control of the prosthesis to become cumbersome.

While no one has objectively examined the impact of controller delays on performance, the controller delay that can be present before prosthesis control degrades has been debated. Childress and Weir [2] believe that controller delays should be kept below 50 ms to ensure that these delays are imperceptible to the user. However, another group has stated that delays as large as 300 ms are not perceivable by the user [3-4] and, while they have not commented on the effect of these delays on performance, they have stated that a 300 ms delay is acceptable for control of a prosthesis.

The desired controller delay may affect the choice of signal processing and pattern recognition algorithms that can be utilized. Thus it would be beneficial to establish this value for future investigations. Experiments were designed to find the longest period of delay that does not significantly degrade prosthesis performance and can thus be dedicated to EMG collection and processing.

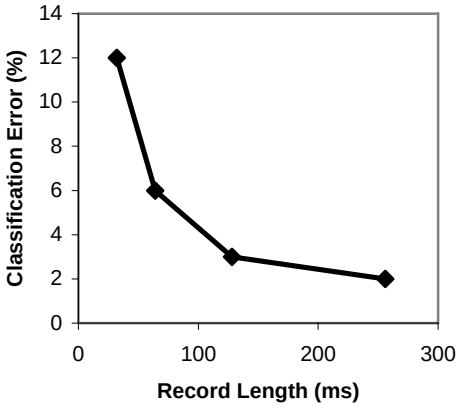
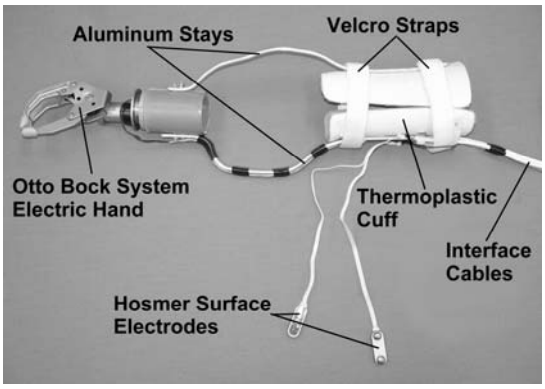


Figure 1 – Classification Error vs. 'Record Length', i.e. the length of recorded EMG on which feature extraction and classification was performed. As the record length increases the classification error decreases. Data from Englehart, 2001.

METHODS

'PHABS' (**P**rosthetic **H**and for **A**ble-**B**odied **S**ubjects) has been created to allow able-bodied subjects to operate the terminal device of an upper-limb prosthesis (figure 2).

While it is infrequently used in clinical practice, myo-pulse control [5] was chosen as the control algorithm for PHABS to minimize the delay inherently contained in the controller. Myo-pulse control is a variation of pulse width modulation in that it provides proportional motor control by varying the pulse width of a digital control signal. The lack of low pass filtering used by myo-pulse control minimizes the time constant of the controller, reducing its inherent delay to the sampling period.



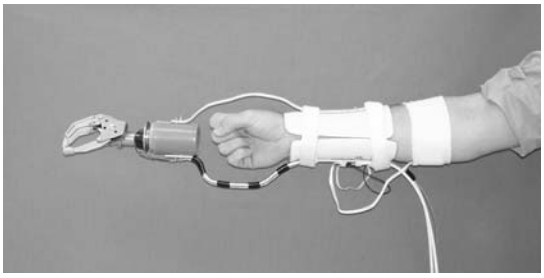


Figure 2

A. A photograph of PHABS (Prosthetic Hand for Able-Bodied Subjects) detailing the components of the device.

B. A photograph of a subject wearing the PHABS.

The Box and Block test was used to quantify prosthesis performance of ten subjects. This test was chosen because it is quantitative, quick and easy to administer, sensitive [6] and normative data have been collected [7-8]. In addition, it has been used to quantify the effect of treatment on upper limb function for disorders such as cerebral palsy [9], multiple schlerosis [10] and stroke [11].

The testing apparatus for the Box and Block (Sammons Preston Inc., Bolingbrook, IL) consists of two adjacent compartments separated by a six-inch barrier (figure 3). The Box and Block is a sixty-second timed test in which subjects are instructed to pick up blocks from one compartment, transport them across the barrier and release them in the opposite compartment as quickly as possible.

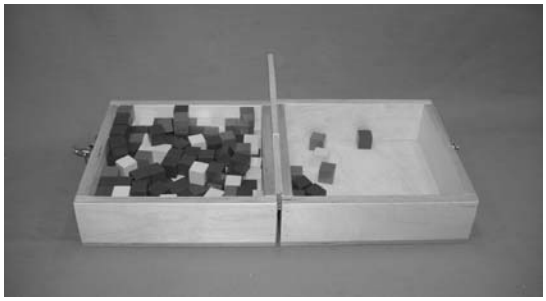


Figure 3 – A photograph of a Box and Block apparatus. Subjects pick up blocks from one side of the box, transport them over the barrier, and then release them on the other side. The number of blocks transported over the barrier in one minute determines the test score.

A prosthetist in our laboratory located two EMG sites on the forearm of each subject according to standard clinical practice. Threshold levels in the myo-pulse controller were altered so that they were as low as possible without allowing the noise contained in the EMG signals to elicit movement of the prosthesis. The amount of pronation/supination was altered to a

position preferred by each subject. After becoming accustomed to the PHABS, the subject would complete two one-minute practice trials of the Box and Block test.

A significant improvement in the Box and Block scores was observed in preliminary studies during the first several trials as users developed strategies for completion of the task and became more comfortable controlling the PHABS. As a result of this learning effect, each subject performed a total of twenty-eight trials with the data from the first seven trials being ignored. Seven different controller delays were introduced into the system with the level of delay ranging from 0 to 300 ms in 50 ms increments. Each of the seven delay levels was introduced once within the first seven trials. The delay level was randomized for the final 21 trials with each of the seven delay levels being selected three times. A repeated measures design was used to compare intra-subject differences in the average Box and Block scores for all controller delays for the ten subjects. Post-hoc analyses were performed with the use of Bonferoni adjustments for multiple comparisons and significance was set at 0.05.

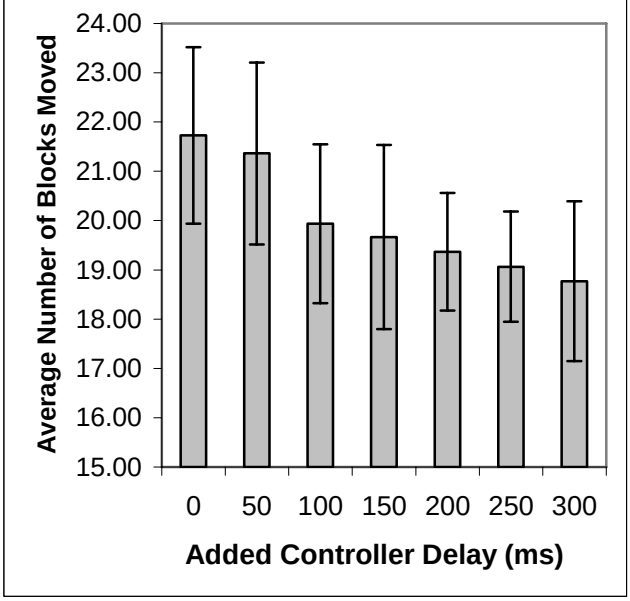


Figure 4 – Average Box and Block scores for all four subjects.

RESULTS

The average box and block scores for all subjects is shown in figure 4 and the comparison results of the repeated measures analysis comparing the different delay levels is shown in Table 1. The differences that are significant to the 0.05 level are highlighted in light gray. Statistical differences exist between 0 and 100,150, 250 and 300 ms as well as between 50 and 300 ms.

Delay (ms)	**50**	**100**	**150**	**200**	**250**	**300**
0	1	0.003	0.001	0.102	0.049	0.007
50		0.533	0.151	0.272	0.193	0.004
100			1	1	1	1
150				1	1	1
200					1	1
250						1

Table 1 – Results of the repeated measures analysis on the Box and Block test scores for all combinations of delay lengths (Bonferroni adjusted p-values for multiple comparisons). The statistically significant results ($p<0.05$) are indicated in light gray.

DISCUSSION

A statistical power analysis indicated that 16 subjects would be necessary to accurately detect a difference of two blocks in the Box and Block test scores with a power of 90%. Since only ten subjects were tested there may be differences between other combinations of delay levels that were not detected due to the number of subjects. For example, a difference may exist between 0 and 200 ms even though no difference is indicated here.

While all of the data have not been collected and therefore no conclusive statements can be made, it appears, based upon the data from the ten subjects that have been collected thus far, that a statistically significant difference will exist between delay increments of 0 and 100 ms. These pilot results suggest that the controller delay and analysis time should be kept below 100 ms when using the Otto Bock System Hand. Additionally, the lack of a difference between the 0 ms and 50 ms delays indicates that Childress and Weir's limit of 50 ms might be too strict.

FUTURE WORK

In the future we aim to investigate the effect of prehensor bandwidth on the 'optimal controller delay' value. We hypothesize that this value will become larger as the bandwidth of the prehensor decreases. In this proposed study, we will conduct the analysis on 16 subjects and for prehensors with two different bandwidths, allowing us to determine the optimal controller delay for each device.

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CLINICAL EXPERIENCE WITH THE OTTO BOCK ELECTRIC ELBOW

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ABSTRACT

This paper discusses prosthetic and therapeutic experience with the Otto Bock Electric Elbow. A beta site patient was chosen for this evaluation based on several factors. First, this patient has been a long time wearer of electric elbow systems and was familiar with the functionality, benefits and characteristics of this particular type of prosthetic management. This aspect is important so that an accurate evaluation of this elbow in comparison to existing technology could take place. Furthermore, this patient represented a challenging level that indicated electric prosthetic management. This gentleman presented with a glenohumeral level deficiency with associated neurologic deficits which created a demand for highly programmable electronics to assist in prosthetic function. This report will chronicle the patient's prosthetic experience from initial fitting, electronic followups in Vienna with our upper extremity prosthetic team and experiences to date the different versions of electric elbow hardware and programmability.

RECENT IMPROVEMENTS IN UPPER LIMB PEDIATRICS

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Introduction

Not only are children fit with myoelectric arms at a much earlier age than years before, but also they are now much more functional and successful with these devices. Advancements such as microprocessor-based controls, longer lasting batteries, improved socket design, and flexible socket materials have improved the functionality of children, especially those under the age of 5, with their prosthesis. Because the pediatric population is so small in our field, it is infrequent that the practitioner knows these advancements. Understanding these improvements and changes to fitting protocols not only helps the child with the device, but also strengthens the rehab team, including the O&P practitioner and OT, as well as strengthens the relationship with referrals and payors. Yet, most importantly, the entire rehab team becomes aware of these advances and can provide optimum care not seen even just a few years ago.

Having a child born without a limb is an emotional trying time for parents. Although many children develop a one-handed independence, parents do not want their child to struggle physically or psychologically. Our organization has found an increasing acceptance of myoelectric control for this young population, primarily from the results received in improved socket design and components. However, the three most important criteria for functionality at this age comes from 1) a team approach, 2) continual follow up, and 3) discussions by the parents, Certified Prosthetists, Occupational Therapists, manufacturers, and other referrals such as Case Managers.

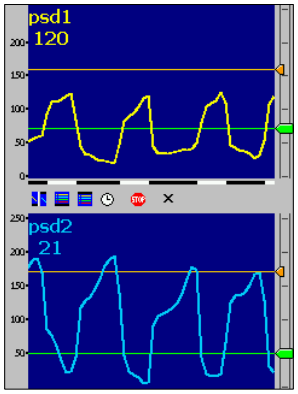
Advancements

The most critical factor in any prosthesis is the socket. The socket must be comfortable, easy to don, provide adequate suspension with maximum ROM, and allow functional control of the terminal device. If there is any concern regarding function of a prosthesis, a full assessment of the socket must be performed.

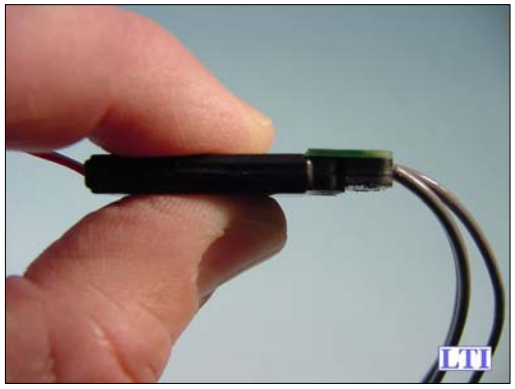
As with any prosthesis, appropriate fit is the single most critical factor towards appropriate prosthetic usage. No matter the cosmetics nor advanced electronics, a true functional assessment can not be performed with a poor fitting prosthesis. An easy guideline is an anatomical shaped flexible socket that doesn't allow for limb movement within the socket. As a side note, any true functional assessment must take into consideration that the child is wearing an appropriate device for the specified activity. Not only should we should be training the wearers appropriate activities, but also we should be providing appropriate devices for those activities. Just as it is illogical to assess a snow skier's ability to snow ski while wearing water skis, it does not make sense to measure a child's ability to hold onto an object requiring an increased grip force, such as a toy bow & arrow, with a TD that has an

inferior grip force. Different terminal devices are made for specific activities just as skis are made for specific types of skiing. Although they are both skis, they are meant for different activities.

The anatomically designed socket is not only heat moldable to accommodate for growth, but also made of plastics which are flexible allowing for an increased ROM and terminal device control. The socket must be anatomically correct and growth adjustable. In recent years the microprocessors controlling the terminal device have drastically improved, providing optimum control by the child. Not only can the gains be adjusted, but also thresholds, rate of contraction, and strategy of control. All adjustments are visually friendly on a computer or hand held PDA.



Our team of specialized clinicians have found an increase in functionality with children of very young age with myoelectrics using devices such as VASI's SPM circuit that uses a standard computer to communicate with the prosthesis, and Animated Prosthetics Inc, which uses a wireless link with a PDA to perform similar communications, or OttoBock's coding plug setup, which uses color specified plugs to change the strategy of control.



Manufacturers have switched to Li-Ion technology in the batteries, which allow for a longer lasting and quicker charging battery that can be placed inside the forearm, thus improving cosmetics. Additionally, electrodes are much smaller and can be placed inside the socket without much cosmetic concern.

Newer smaller pediatric electrodes can be moved without the need of a new socket when adding a second site electrode.

Conclusion

These improvements in socket material and design, computer adjusted controls, smaller electrodes, parental involvement, and longer lasting batteries have provided a larger acceptance of myoelectric devices for young children.

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MULTI-FUNCTION SWITCHING OPTION FOR THE HIGHER LEVEL UPPER LIMB AMPUTEE

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ABSTRACT

The purpose of this article is to describe an alternative switching option for the higher-level upper limb amputee. Currently available body powered switches allow for only one function which can become quite cumbersome when multiple control methods are used. A multi-functional switch that can activate many different controls either independently or simultaneously could greatly reduce the amount of hardware required to operate all of these devices. When working with bilateral shoulder or interscapulothoracic amputees one of the main goals is to provide as much function as possible while maintaining simplicity in the design. The first patient that was fit with the actuator was currently wearing bilateral shoulder prostheses with seven switches. The result was a reduction in the amount of switches necessary to operate the different controls from seven to two. The switch can also be configured to operate Force Sensing Resistors (FSR's) or linear transducers to provide proportional control.

INTRODUCTION

Higher-level amputees generally require additional components to help restore their functional ability. The shoulder disarticulation and interscapulothoracic level amputees require the addition of a prosthetic shoulder joint to help preposition the prosthesis to assist them with their activities of daily living. The primary objective for the practitioner and rehabilitation team is to decide on what components and control options are most suitable for the patient while maintaining simplicity in the design. Uellendahl states, when fitting the bilateral shoulder disarticulation amputee, it is advisable to start with as simple a prosthetic system as possible and control complexity should be kept to a minimum [1]. Taking into consideration the patients interests, hobbies and vocational goals is a good place to start when beginning the component selection process. Also, the cognitive and motivation levels of the patient will play a critical role in their outcome. When deciding on what components to utilize for a particular patient you need to consider how they will operate them. Will they be using myoelectric control to operate the prosthesis or do they have enough range of motion to utilize some body-powered components? If weight is a concern, would a passive semi prehensile device be more appropriate or a hybrid design to capture the benefits of both myoelectric and body powered controls? Regardless of the type of prosthetic design used, the higher-level amputee requires the use of multiple components to fully operate the prosthesis at this level. Many of these components work in conjunction with switches to lock or unlock the particular component. Commercially available body powered switches only allow for a single function. The multi function actuator was designed to allow the patient the ability to operate multiple functions with only one switch. The following will describe the progress of a bilateral shoulder disarticulation amputee and how the multi function actuators have improved his efficiency and functionality with his present prostheses.

METHOD

The multi function actuator was fabricated using stainless steel bar stock. The switch dimensions are 4¼ X 3 inches as shown in Fig. 1. The nudge pad is made from pelite and is 1 7/8" in diameter (Fig. 1A). The stainless steel bar is contoured to allow for 5/8" of cable excursion, which is sufficient to operate most locking devices. If more cable excursion is necessary the switch can be contoured to allow for more excursion. The center axis is located 1 1/8" from the center cable attachment. The center cable attachment (Fig. 1B) is slotted to allow the outside cables to travel without activating the center cable. Rotating the switch to the left or right will activate the outside cables. Triple swivels were used to attach the outside cables to the switch. A ball terminal was used to attach the center cable. A ½" rubber bumper (Fig. 1C) was used along with a #10 copper rivet to attach the switch to the socket and allow the switch to depress, rotate and return to a neutral position. The hole drilled into the switch was slotted to allow the switch to pivot on its axis. Pressing down on the pad will activate the center cable. Simultaneous activation can be achieved by rotating and depressing the switch at the same time.

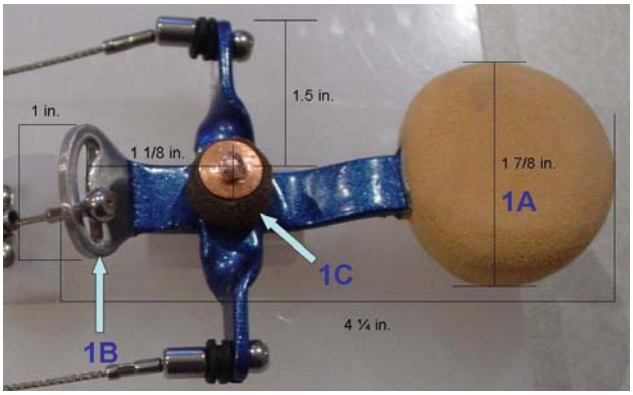

Fig. 1

The first patient that was fit with the multi function actuator was presently wearing bilateral shoulder disarticulation prostheses consisting of the following components: The left prosthesis consisted of an X-Frame socket design with a flexible inner socket, an LTI Collier locking shoulder joint, Otto Bock AFB elbow, Rim Jet humeral rotation unit, electric wrist rotator, and a myoelectrically controlled electric terminal device (ETD). Prior to the addition of the multi function actuator the patient was using three separate switches to operate the shoulder lock, humeral rotation lock and elbow flexion lock. The switches were aligned in such a way to allow the patient access to each one with his chin. This configuration will generally place one of the switches in an ideal position and the other switches will be positioned along side. Custom fabricated switch extensions were used to place the switches in a more functional position.

The right prosthesis consisted of an LTI Collier locking shoulder joint, RimJet humeral rotation unit, Otto Bock AFB elbow, 4-function wrist, and a model SS-555 hook. The patient was using four separate switches to operate the shoulder joint lock, RimJet humeral rotation lock, elbow flexion lock and 4-function wrist activation. Custom

fabricated switch extensions were also used on this prosthesis to position the switches in the ideal position.

DISCUSSION

Advancements in upper extremity technology have allowed patients to improve their functional ability and increase their level of independence. Higher-level amputees require additional measures to assure their optimal functional ability. These measures include a socket designed to allow for maximum comfort, range of motion and stability, a lightweight simple design and the proper components to meet their needs and goals. The components necessary to provide the most functionality at higher levels can often times be heavy, cumbersome and difficult to master. Reliability and durability are also very important and necessary for the bilateral amputee. Functional bilateral amputees who rely on their prostheses for independence become dependant upon others when their prostheses are in need of repair. If the overall amount of switches can be minimized by using a multi function actuator we can reduce the tendency for component failure by reducing the overall mechanical components on the prosthesis. The multi function actuator can offer patients increased freedom to manipulate their prosthesis and provide them greater control by being in contact with one switch to perform multiple tasks as apposed to alternating between switches to perform the same tasks.

RESULTS

The addition of the multi function actuator had resulted in an overall reduction of the weight of the prosthesis. We were also able to simplify the design by reducing the amount of switches necessary to operate all of the components. Maintenance requirements were also reduced by the reduction in hardware necessary to perform the desired tasks. The prostheses were more cosmetically appealing under clothing due to the reduction in switches that protruded from the socket, and the patient required less range of motion to activate the switches. His overall control with the locking devices and functional ability with the prostheses had also improved.

CONCLUSION

Higher-level amputees face many challenges with regards to returning to a functional level of independence. Currently available technology is not capable of replicating the intricate and finite control that nature has provided for us. Providing the higher level amputee even with the latest in technology is far from replacing what was lost, but we can provide them with a functional tool to assist them in achieving independence and improve their functional ability. As stated by Uellendahl, whatever control arrangement is used, it should impose the minimum amount of mental loading on the user. In other words, the control of the prosthesis should not be so complicated as to make it the primary object of the user's attention. Rather, the task the user is attempting to perform should be primary, and the mental effort devoted to the prosthesis should be secondary (1). The multi function actuator was designed to help minimize the complexity of the

system which can allow the patient to focus on the task being performed and less effort on the steps necessary to perform that task.

1. Uellendahl J, Heckathorne CW, Creative Prosthetic Solutions for Bilateral Upper Extremity Amputation, *Functional Restoration of Adult and Children with Upper Extremity Amputation*, Meier and Adkins, 2004, pg. 225-237.

USING THE INTERNET FOR AN ANONYMOUS SURVEY OF MYOELECTRICAL PROSTHESIS WEARERS

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INTRODUCTION

In the last decades, a multitude of surveys were carried out to investigate broadly different fields associated with prosthesis acceptance, reasons for non-use or user preferences for artificial hand improvements [1-9]. Typically, interviews were made by telephone [1,2] or during maintenance work on the prosthesis [3-5], or questionnaires were sent to potential users [6-9]. In most surveys, the persons to be questioned were pre-selected by the interviewer using different inclusion criteria, e.g. level of amputation, age, and contact with the limb fitting centre in a certain period or residence in a certain area [6]. Therefore, little is known about non-users that broke tie with the facility that supplied the device. Return rates of completed postal questionnaires typically range from 38% [7], to 49% [9] to 69% [6]. For the first time now, an anonymous internet survey has been performed using a questionnaire that can be answered on a homepage. By this method, it is intended to enable prosthetic hand users and non-users as well to provide information on the acceptance, usage, and wishes for the development of new and improved hand prostheses.

METHODS

An internet survey with 32 questions was designed to anonymously gain information from persons fitted with a myoelectric hand. Prior to making the homepage accessible, a different questionnaire was distributed in a pilot study to patients who routinely visited the prosthetic fitting centre at the University Hospital of Heidelberg. Questionnaires were comprised of all persons meeting the following criteria: Upper limb deficient persons of 14 years of age and older at the time of assessment that have been fitted with an electrically driven prosthetic hand. These questionnaires had been answered between 1st of April 2002 and 2005. The present evaluation is based on the completion of a five-page questionnaire on the non-commercial homepage http://www.handprothese.de/umfrage.htm.
The questionnaire consists of different parts: In the beginning, facts about the age and gender of the user and date, reason, and level of amputation are requested. The second part concerns the kind of prosthesis and its use in recreation and working hours. Part 3 covers the subjectively perceived noise, weight, and cosmetic appearance of the prosthesis. Finally, questions about activities a prosthetic hand should be used for and open questions for comments and wishes are given.

RESULTS

Age and gender:

35 persons answered the questionnaire, 27 males and 8 females. The ages of the limb-deficient persons at the time of assessment ranged from 15 to 65 with a mean of 32.9 years. 4 women and 18 men had a traumatic amputation (mean age 36 years), and 4 females and 9 males had a congenital upper limb deficiency (mean age 27.9 years). The period between the accident and this survey ranged from 0 to 51 years with a mean of 18.2 years.

Level of loss:

There were 19 males and 7 females with below-elbow amputations, 6 males and 1 female with above-elbow amputations. 2 persons did not provide any information on the level of loss.

Use:

The question "For how many hours do you employ your prosthetic hand at work" was answered by 69% (24/35) choosing "more than 8 h", by 11% (4/35) with "4 to 6 hours", and 20% (10/35) answered "I don't know". A recreational use of "more than 8 h" was correct for 66% (23/35), "4 to 6 hours" for 6% (2/35), "occasionally" for 26% (9/35), while 3% (1/35) did not know for how long. 26% (9/35) can operate the prosthesis with one battery charge, 34% (12/35) just one day, 16% (4/35) less than one day, and 28% (10/35) answered "I don't know".

Subjective evaluation:

80% (20/25) consider their electrically driven hand to be too slow. The cosmetic appearance is rated "good" or "very well" by 40% (10/25) and "bad" or "very bad" by 60% (15/25). 72% (18/25) are "little" or "not at all" personally irritated by the noise of the prosthesis, whereas 28% (7/25) do not feel acoustically disturbed. 20% (5/25) complain about the prosthesis weight as being much too high, 56% (14/25) consider it a little too high, and for 24% (6/25) it is acceptable.

Desired activities and functions:

The respondents were asked "which of the following activities would you like to perform with your prosthesis?" and the answers given are:
Personal hygiene (washing, cleaning teeth): 76% (26/34)
Using cutlery: 79% (27/34)
Dressing and undressing 68% (23/34)
Writing 56% (19/34)
Opening and closing a door 74% (25/34)
Operation of electronic and domestic devices and switches 71% (24/34)
Bricolage 82% (28/34)
A difference was found between persons with a traumatic amputation, who selected an average of 5.4 of the 7 possible activities, and persons with a congenital limb deficiency, who marked only 4.2 activities with a cross. In particular, the prosthesis is wanted to be usable for personal hygiene by 86% (19/22) of the persons with a traumatic amputation compared to 58% (7/12) of the other group.
86% (18/21) would like to have a force feedback system integrated in the hand, whereas 65% (11/17) want additional functions like a temperature sensor. 33% (9/27) would accept a new prosthesis that has less grasping force, but a lower weight by

30% compared to conventional myoelectrical hands. 33% (9/27) would not accept such a new hand and 33% (9/27) do not know whether they would accept it.

DISCUSSION AND CONCLUSIONS

Limits of the survey:

As the survey was designed to be answered anonymously, there is no certainty of limb deficient persons only having completed the questions. However, there is no evidence of manipulation indicated by a certain repeated pattern of answering the questions. Another restraint of the results can be attributed to the number of questions and pages of the survey. Whereas almost all users answered all questions of the first two pages, the rate decreased to 63% (22/35) on the last page. Although there are an increasing number of persons in Germany fitted with a speedy myoelectric hand, all six responses received in the last six months rated the grasping speed of their artificial hand as being too slow. One explanation is that the respondents have became aware of the fact that there are faster hands in the meantime. Consequently, a question for the exact type of myoelectrical prosthesis should be included in a revised survey.

Comparison of the results

It was found that the majority of users utilise their prosthetic hand 8 hours or more on an average working day, which corresponds to the results found by [3] and [6]. In line with the results of [5-7], it was also found that light weight, speed, and the cosmetic appearance are important issues to be raised when designing a new prosthetic hand. With the present study, it was demonstrated that information on whether prosthetic hands are used in daily life and for what kind of activities they should be applicable can be gained by an anonymous internet interview.

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A BUS PROTOCOL FOR INTERCOMPONENT COMMUNICATION IN ADVANCED UPPER-LIMB PROSTHESES

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INTRODUCTION

Microcontrollers are ubiquitous in modern electronic products, powered prosthetic components being no exception. Likewise, digital communication buses are the key technology for interconnecting such smart devices for reasons such as reduced wiring requirements, high information throughput, robustness in the presence of noise and flexibility. This flexibility contains the potential of interoperability, which means that similar components from different manufacturers may communicate in the same way so that one can easily be replaced by another or components from different manufacturers can be combined in one and the same system. However desirable this situation is, it requires an open, sandardised communication protocol that is adhered to by the majority of the manufacturers and research organisations. Presently no such standard exists in the prosthetics industry, while it has existed in related fields such as wheelchairs and environmental controllers for a number of years.

In this paper we propose such a bus standard, and outline the potential benefits for end users, prosthetists and technicians, healthcare providers, manufacturers and researchers. We then list several important aspects of a prosthesis bus that must be carefully considered, and invite interested parties to engage in the completion of a draft specification.

WHY A STANDARDISED BUS PROTOCOL?

A standardised digital prosthesis communication channel enables the following, as is demonstrated in conjunction with the introduction of so-called "field buses" in industrial process control:

- Reduced wiring and thus production cost and hardware failure rate.

- Advanced coordinated control schemes with a large number of sensor signals and control variables.

- Remote adjustment, fault diagnosis and software upgrades.

- Interoperability and thus improved interchangeability of different devices.

This latter point is clinically important because it simplifies the integration of hybrid systems and increases the freedom of choice of components for the prosthetists and users. For the same reason, a standardised interface improves the possibilities for joint projects in advanced prosthesis control research, in that adherence to a standard simplifies the experimentation with novel components and control schemes (an example of such a component is the NTNU Revolute Wrist Device, which is described elsewhere in these proceedings [1]). The potential benefits for the end user and research community are apparent. For prosthetists and technicians the simplified wiring implies a somewhat simpler manufacturing process, and the benefits related to parameter adjustments and fault diagnosis are already demonstrated by several commercial systems. Interchangeability allows simpler and faster experimentation with different components, so that the prosthesis can be optimised for each patient more efficiently. As the number of powered joints now used in a prosthesis increases the addition of the bus and the simplification of the wiring become essential.

The situation for the prosthetics industry is slightly different. The willingness of each company or cluster to take part in a standardisation process and (more importantly) to adhere to the standard in the future, is necessarily coloured by the degree to which it is considered compatible with current company strategy. This issue is not considered any further here, except to point out that digital communication is inevitable in conjunction with most advanced prosthesis control schemes. One must therefore hope that the industry will find it beneficial to join forces in order to have an impact on this emerging and enabling technology. Other areas of technological rehabilitation have already adopted this strategy despite the small markets and margins associated with the field.

A structured thinking about control options, coordination and information flow aspects, which is required in conjunction with a stadardisation activity, may in itself induce new and improved prosthesis control concepts. This may therefore be an incitement to take part in the work.

FUNCTIONAL META-SPECIFICATION

The format of the present paper does not allow the presentation of a full protocol draft, nor is it natural to do so because such a draft should be the result of a workinggroup process. We therefore present a brief meta-specification, indicating some of the issues that must be addressed by a protocol specification. The meta-specification is functional rather than technical in that it relates to functional properties rather than implementation issues. The reader should keep in mind that it is the *bus protocol* (intercomponent communication channel) that is presently in focus, not the individual prosthesis components nor the entire prosthesis as a system, although it is the component or system level functions that are to be served by the protocol. The list below does not constitute a complete overview of the properties that need to be specified or taken into account, but may hopefully serve as a starting point for discussions. Each issue is given a reference code. It is noted that many of the issues are connected or interrelated.

MS-01 **What functionality is required** in order to realise current and future prosthesis control schemes?
- MS-01-01 **Functions** (motor control functions, feedback loops, coordination, other)
- MS-01-02 **Commands** (state change commands, other)
- MS-01-03 **Status information** (state machine states, error states, other)
- MS-01-04 **Service commands** (diagnosis, configuration, parameter setting)

MS-02 **Device Profiles** (standardised descriptors of communication participants)
- MS-02-01 **Typical configurations** (known current device combinations)
- MS-02-02 **Growth potential** (to accommodate future developments)
- MS-02-03 **Hiearchical profiles** (for flexibility wrt. control schemes)

MS-03 **Data transfer rates** (information throughput requirements)
- MS-03-01 **Number of communicating nodes** (physical or logical)
- MS-03-02 **Cyclic communication** (transfer of continuous signals or status)
- MS-03-03 **Sporadic communication** (state changes, errors and exceptions, configuration and parameter settings)
- MS-03-04 **Grouping of data** (data with similar transmission rates or contextual aspects transmitted together).

MS-04 **Exception handling**
- MS-04-01 **Fail-safe mechanisms** (what to do if things do not work as usual)
- MS-04-02 **Power-up behaviour** (communication sequence, default behaviour, other)
  - MS-04-02-01 **System level** (coordiated power-up)
  - MS-04-02-02 **Node level** (different kinds of nodes to behave differently?)

MS-05 **User friendliness**
- MS-05-01 **For the amputee**
  - MS-05-01-01 **Safety** (robust design to avoid rapid unsolicited movements, other)
  - MS-05-01-02 **Reliability** (graceful degradation, other)
  - MS-05-01-03 **Availability** (prioritised communication, graceful degradation, other)
  - MS-05-01-04 **Responsiveness** to user commands
    - MS-05-01-04-01 **Admissible delay** from input to response
    - MS-05-01-04-02 **Minimum ranges and resolutions** for control and sensor variables

- MS-05-01-05 **Space requirements** (physical attributes of electronics, plugs, wires)
- MS-05-02 **For the prosthetist/technician**
  - MS-05-02-01 **System configuration** (parameter setting, other)
  - MS-05-02-02 **Intraoperability** (flexibility with respect to the choice of components)
  - MS-05-02-03 **Diagnostics** (auto-diagnosis, remote access)
  - MS-05-02-04 **Software upgrading** possibillities
  - MS-05-02-05 Ease of **mounting and maintenance**
- MS-05-03 **For the system designer/manufacturer**
  - MS-05-03-01 **Space requirements** (identical to MS-05-01-05)
  - MS-05-03-02 **Implementation cost**
  - MS-05-03-03 **Availability of relevant technology** (components, firmware, compilers and more)
- MS-06 **Robustness issues**
  - MS-06-01 **Electromagnetic compatibility** (noise immunity, low emissions)
  - MS-06-02 **Hot-swapping** of system components (replace components while system is running)

DISCUSSION AND FURTHER WORK

The present state of prosthetics and data transfer technology indicates that this is a good time for working towards a common bus standard. This paper lists a number of essential aspects that need to be considered in order to define the problem at hand.

The authors hereby invite interested parties to take part in the formation of a workgroup that will thus define the problem and continue working towards a complete bus specification. Ideally the final result of this process will be a specification that covers everything from the physical bus connectors up to the higher protocol levels. The protocol should be based on a Commercial Off-the-Shelf low-level protocol such as I^2C, CAN or ZigBee, which implies the immediate availability of hardware and firmware components.

ACKNOWLEDGEMENTS

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MANAGEMENT OF HIGH LEVEL BILATERAL ARM AMPUTEES WHO USE WHEELCHAIRS FOR MOBILITY

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Management of the high-level bilateral arm amputee poses many challenges to the prosthetist and rehabilitation team. When the amputee is also in a wheelchair, use of arm prostheses becomes more difficult. Positioning of the prosthesis without the use of the lower extremities is remarkably difficult. The work envelop is further reduced when trunk motion and stability is compromised by paralysis.

Heckathorne and Uellendahl have recommended a framework for component and control strategy selection when designing prostheses for high level bilateral arm amputees.[1-3] This strategy calls for use of dissimilar components on each side to enhance prosthesis usefulness and control schemes that provide dedicated control of as many components as possible allowing simultaneous control when functionally desired. Uellendahl has used this approach for over 15 years for clinical fittings. The long-term success of this fitting philosophy demonstrates the clinical efficacy of this approach. In fact, one of the amputees (MM) reported on in this paper has used prostheses of the same original design for more than a dozen years.[3]

The authors have experience fitting bilateral SD, TH, and unilateral TH with quadriplegia clients who are confined to wheelchairs for mobility. Case presentation of these fittings will illustrate the benefits of implementing the above-mentioned fitting philosophy. One on-going fitting will be discussed that uses an innovative control strategy using a mini joystick to control two components simultaneously on an x/y coordinate system. The joystick is from a Play Station™ control unit. The technical modification to prosthesis control were performed by Liberating Technologies of Holliston Maine.

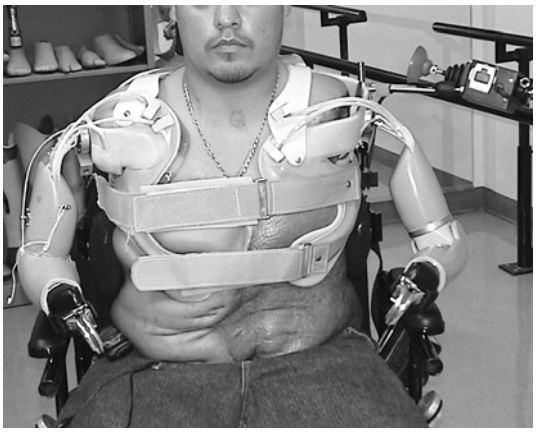


Figure 1

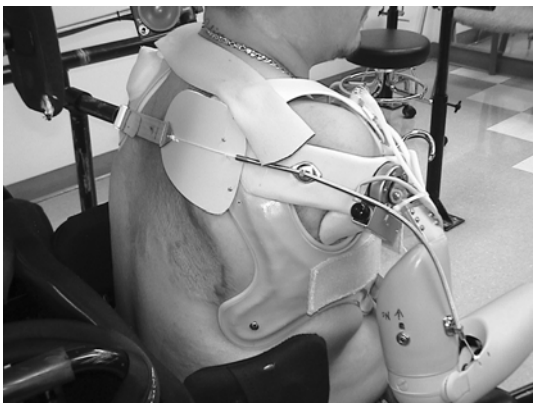
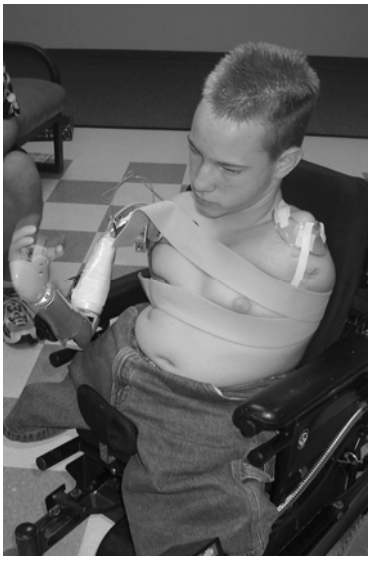


Figure 2



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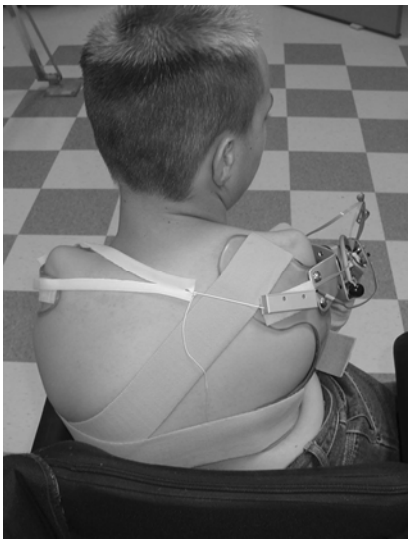
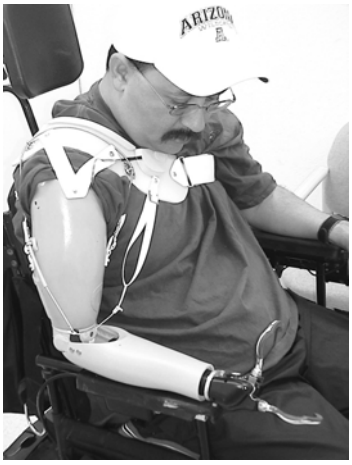


Figure 4



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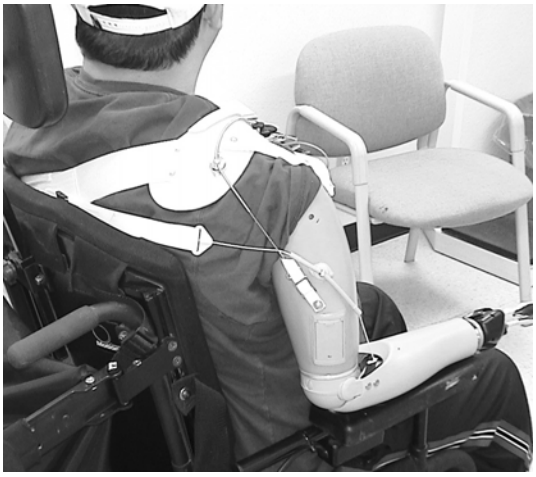

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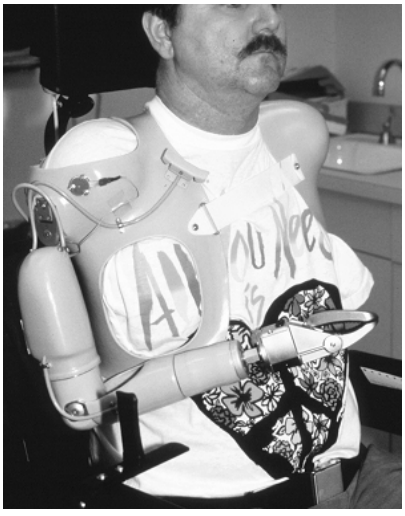

Figure 7

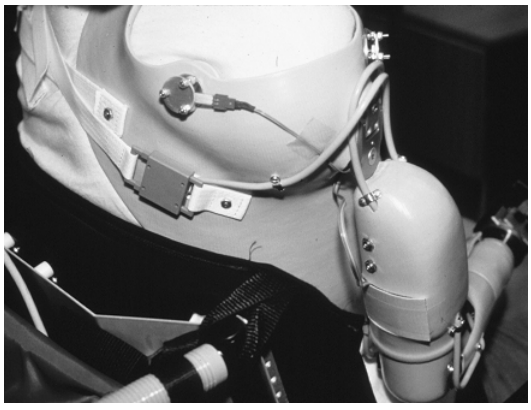

Figure 8

Figure 1 shows a bilateral humeral neck amputee (MS) who also has bilateral transfemoral amputations. He uses an electric wheelchair controlled with chin activated joystick. The left prosthesis is fully electric using myoelectric 2 site control of wrist, TD, and elbow using a chin nudge for mode selection. The right dominant side uses FSR's under the humeral remnant for control of TD and wrist with a chin operated mode select between these two devices. Both shoulders use electric locks operated by chin nudge.

Figure 2 shows the cable control of the elbow on the right prosthesis for patient MS.

Figure 3 shows a quadrimemberal congenitally limb deficient boy (CG). In this case only one prosthesis was fitted to avoid gadget overload. The hand is myoelectrically controlled using 2 electrodes. The wrist is controlled by FSR's located superior to the acromion.

Figure 4 shows the elbow control using linear transducer for patient CG.

Figure 5 shows a unilateral TH amputee (DB) with C7 quadriplegia. His left hand is minimally functional, it is used to control his electric wheelchair. The right prosthesis uses a single FSR for terminal device control using chin operation.

Figure 6 shows the harness used for cable operated elbow control for patient DB

Figure 7 shows patient MM. He has bilateral humeral neck amputations, bilateral TT amputations, and paraplegia. He is fitted unilaterally. He uses FSR's placed anterior and posterior to his shoulder for elbow control. The TD is controlled with a chin operated rocker switch. The T mount on the wheelchair is used to position humeral rotation as well as to activate the lock lever for wrist flexion.

Figure 8 shows a posterior view of MM. Forearm rotation is controlled by switch using scapula abduction

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Removal of ECG Artifacts from Myoelectric Prosthesis Control Signals

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Abstract - We investigated removal of electrocardiogram (ECG) artifacts from the myoelectric prosthesis control signals, taken from the reinnervated pectoralis muscles of a patient with bilateral amputations at shoulder disarticulation level. The performance of various ECG artifact removal methods including high pass filtering, spike clipping, template subtracting, wavelet thresholding and adaptive filtering was presented. In particular, considering the clinical requirements and memory limitation of commercial prosthesis controllers, we further explored suitable means of ECG artifact removal for clinical application.

I. INTRODUCTION

Electrical artifacts (electrocardiogram, ECG) produced by the heart may contaminate electromyogram (EMG) signals when recording surface EMG from torso muscles. In particular, this source of noise is a big concern for myoelectric prosthesis control in a patient with bilateral amputations at shoulder disarticulation level, where the control signals are taken from the reinnervated pectoralis muscles of the patient [1]. This noise will result in a compromise of accuracy for the prosthesis control and therefore has to be removed. The aim of this work is to investigate various ECG artifact removal methods for myoelectric prosthesis control. The performance of these methods was evaluated with regard to the efficiency of ECG artifact removal, the possible distortion of EMG, and the feasibility for potential clinical application.

II. METHODS

Subject and Data Collection

Our subject (patient), a high power lineman, suffered severe electrical burns resulting in the amputation of both arms at the shoulder disarticulation level. To develop additional myoelectric control sites for improved prosthesis control, targeted muscle reinnervation was performed in this patient. Four independently controlled nerve-muscle units were created by surgically anastomosing residual brachial plexus nerves to dissected and divided aspects of the pectoralis major and minor muscles (left side of body, Figure 1). After surgery and recovery, independent surface EMG signals (corrupted by ECG artifacts) were obtained from the reinnervated muscles to control the myoelectric prosthesis [1].

Various algorithms were investigated in this study to remove the ECG artifacts from the independent myoelectric control signals developed by targeted muscle reinnervation. The patient was seated upright in a chair and positioned in a standardized fashion, and was asked to actuate various movements (Hand Open; Hand Close; Elbow Flexion and Elbow Extension). For each movement, the patient was asked to repeat 10 times by comfortably exerting a level of contraction at a medium force. A rest time was allowed between each repetition. As indicated in Figure 1, four bipolar surface EMG channels (represented by EF, HO, HC, EE, with each channel corresponding to the intended movement it controls) were recorded from the myoelectric control sites, located at upper, middle and lower pectoralis major regions where the

musculocutaneous nerve, the median nerve and the radial nerve were anastomosed surgically. An additional channel of signal was recorded from the pectoralis minor muscle, and was used as ECG reference in this study. Surface EMG was recorded with a Noraxon (Phoenix, AZ) Telemyo 2400R System™, with a bandwidth of 10 – 1000 Hz, a gain of 2000, and an internal sampling rate of 2000 Hz.

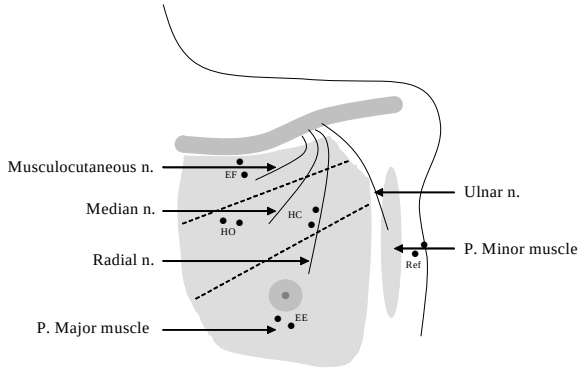

Figure 1. Diagram of nerve-muscle graft procedure and electrode locations

Various ECG Artifact Removal Methods

(1) High pass filtering (HPF): Compared with surface EMG, ECG signal contains relatively low frequency components. High pass filtering was therefore implemented for ECG artifact removal from the myoelectric control signals.

(2) Spike clipping (SC): ECG spike clipping is a threshold based suppression method, which requires an ECG reference signal to identify the occurrence of ECG artifacts. The surface EMG signal and the ECG reference were read simultaneously. Once the ECG reference was detected to reach the preset threshold, a concurrent window of the EMG signal (spanning the main QRS complex) was replaced by the uncontaminated signal immediately before this segment.

(3) Template subtraction (TS): ECG template subtraction takes advantages of the quasi-periodic characteristics of the ECG signals. First, ECG template contaminating the EMG signal was formed from an ensemble average of 'clean' ECG complexes with no muscle activity or of all the ECG complexes based on the assumption that the EMG has a zero-mean Gaussian distribution. Then the occurrence of ECG spikes was detected by calculating the correlation between the processed signal and the ECG template. A threshold was set in a way that a value greater than the threshold indicated the occurrence of an ECG spike (heart rate information was also considered in this process). Finally, the ECG template was subtracted from the corrupted signal aligned by peak location of the cross correlation. The subtracted ECG templates were adapted according to the amplitude of the detected ECG spikes.

(4) Wavelet transform (WT): Nonlinear thresholding in the wavelet domain was also used for ECG artifact removal. Signal corrupted by ECG artifacts was first decomposed by wavelet transform to sequences representing different frequency components of the signal. The wavelet coefficients in the low frequency scales then underwent a nonlinear thresholding process, where the coefficients greater than the threshold were set to zero. For each scale, the threshold varied with the background coefficients in a way that the conspicuous coefficients (representing the

ECG artifacts) or the coefficients much greater than the neighboring ones were adjusted. The inverse wavelet transform was then implemented using the new coefficients to obtain the "clean" EMG signal.

(5) Adaptive filtering (AF): Adaptive filtering for the cancellation of ECG artifacts was also investigated, which requires adjustable filter weights and two input signals. The primary input signal was the EMG signal (containing ECG artifacts) from the myoelectric control sites and the reference input was the ECG signal measured simultaneously. The filter weights were initially set at zero and iteratively adjusted according to the least mean square (LMS) or recursive least square (RLS) algorithm.

III. RESULTS

The performance of different ECG artifact removal methods is briefly presented in this section. Figure 2 (a) shows an example of the raw signal detected from one of the myoelectric control sites (channel EE) when the patient actuated the "Hand Open" movement from the relaxed condition. Figure 2 (b-f) presents the signals after ECG artifact removal using various algorithms. A second order butterworth filter with cutoff frequency setting at 60 Hz was implemented for high pass filtering. For wavelet denoising, seven-level decomposition was used and the four low frequency components underwent the coefficient thresholding process. Table 1 presents the elimination of ECG artifacts at different myoelectric control sites. The average rectified value (ARV) of the ECG artifacts before and after the denoising was compared. The distortion that different algorithms may impose on the EMG signal was investigated by comparing the ARV of the EMG segments between consecutive ECG spikes. Table 2 shows an example of this comparison, where the signals were detected from the HO channel during different movements. Finally, the time delay imposed by different algorithms and their average implementation time is presented in Table 3 (Pentium PC, 3.4 GHz CPU, 3GB RAM).

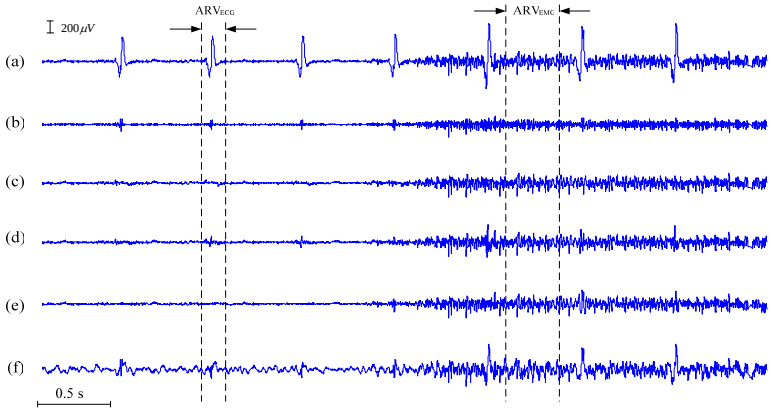

Figure 2. An example of ECG artifact removal: (a) raw signal; (b) high pass filtering; (c) spike clipping; (d) template subtracting; (e) wavelet denoising; (f) adaptive filtering

Table 1. Amplitude (mean±std, μV) of ECG artifacts before and after the denoising process

	ECG Artifact	HPF	SC	TS	WT	AF
HC Channel	196.94 ± 9.34	22.98 ± 3.68	10.17 ± 3.09	18.10 ± 9.26	16.45 ± 9.94	92.60 ± 11.24
HO Channel	168.14 ± 3.52	8.83 ± 0.75	9.48 ± 3.30	11.67 ± 2.61	16.82 ± 7.61	75.15 ± 14.39
EF Channel	104.84 ± 3.66	7.22 ± 0.69	7.92 ± 1.65	11.83 ± 1.80	17.87 ± 9.68	46.19 ± 8.60
EE Channel	133.12 ± 4.97	17.10 ± 2.03	6.43 ± 2.47	11.38 ± 4.25	13.51 ± 5.25	57.64 ± 10.64

Table 2. Amplitude (mean±std, μV) of uncontaminated EMG signals before and after the denoising process

	EMG	HPF	SC	TS	WT	AF
Hand Open	183.98 ± 39.74	143.00 ± 27.69	183.98 ± 39.74	183.98 ± 39.74	182.10 ± 39.95	187.75 ± 39.77
Hand Close	122.39 ± 26.69	91.18 ± 20.86	122.39 ± 26.69	122.75 ± 26.53	121.30 ± 26.60	126.15 ± 25.48
Elbow Flexion	28.82 ± 3.88	21.58 ± 3.07	28.82 ± 3.88	28.82 ± 3.88	28.46 ± 4.30	33.52 ± 4.21
Elbow Extension	37.74 ± 8.61	23.75 ± 4.87	37.74 ± 8.61	37.74 ± 8.61	37.30 ± 8.60	41.15 ± 9.15

Table 3. Algorithm delay and implementation time for various ECG artifact removal methods

	HPF	SC	TS	WT	AF
Delay caused by the algorithm	0	QRS duration	QRS duration	Inter-ECG spike-interval	0
Calculation time per sampling point (ms)	0.000041	0.000109	1.4	0.000841	2.1

IV. DISCUSSION

We investigated various ECG artifact removal methods for myoelectric prosthesis control in a bilateral shoulder disarticulation amputee. We found that high pass filtering removed most of the ECG artifacts while some useful EMG signals were inevitably lost. Due to the local resolution of the wavelet transform, wavelet denoising was able to remove the ECG artifacts while still keeping the useful EMG information, even in the case of overlapping spectra. Taking advantage of the quasi-periodic characteristics of the ECG signal, template subtraction proved to be an alternative way to remove the ECG artifacts without sacrificing sections of the EMG signal. After adaptive filtering we found that significant ECG contamination can still be recognized. This implied that the linearity assumption underlying the classical adaptive noise canceller was not satisfied due to some nonlinear and time-variant distortions of the ECG signals. It is worth noting that in clinical the maximum delay for controlling the myoelectric prosthesis is required to be no more than 50 ms. In addition, the memory of the commercial prosthesis controller is limited. In this regard, although performing effectively in ECG artifact removal, some methods such as template subtraction and wavelet denoising are not suitable for clinical application due to the long time delay caused by the algorithm and the heavy computation burden (Table 3). On the contrary, our study indicates that, despite of the loss of EMG signals, high pass filtering can remove most of the ECG artifacts while still keeping sufficient EMG power. This suggests that high pass filtering (with optimal cutoff frequency and filter order) may be an effective approach to remove ECG artifacts in real time for myoelectric prosthesis control. Practically, high pass filtering can be implemented by selectively modifying the hardware or software configurations of the commercial prosthesis controller. With an ECG reference channel to detect the occurrence of ECG artifacts, ECG spike clipping is another method characterized with simple implementation, which may be an alternative suitable means for clinical application. Finally, if the ECG reference channel can be optimized to satisfy the assumption for adaptive ECG canceller, with development of the more powerful commercial prosthesis controllers, adaptive filtering represents another potential approach for real time ECG artifact removal from the myoelectric control signals.

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