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Grasping the Future

Advances in Powered Upper Limb Prosthetics

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Grasping the Future: Advances in Powered Upper Limb Prosthetics

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FOREWORD

This eBook is published at an opportune time in the history of prosthetics. Particularly, recent technological advances in actuation, microelectronics, batteries, and fabrication methods have fueled the emergence of upper extremity prostheses with far greater movement capability than was previously possible. With the ability to provide a large number of possible movements, such prostheses offer great promise for enhancing the ability of amputees to better perform the activities of daily living. Use of this enhanced capability, however, requires in most cases a user interface that enables efficient and intuitive access to the multiple movements offered by these prostheses. Thus, leveraging advances in motor functionality in upper extremity prostheses is fundamentally dependent on corresponding advances in user interface and control.

The appropriate availability of possible movements and the nature and capability of the control interface are strongly coupled. Introducing additional movement capability will in many cases impose a greater control burden on the user. Although neural interfacing has the potential to supply a rich set of control information, the amount of control information is likely (for the foreseeable future) to be far less than that employed within the native limb. A single-degree-of-freedom hand, for example, is limited in movement capability, but is relatively easy for an amputee to control. A twenty-degree-of-freedom hand, conversely, has a great deal of movement capability, but may be difficult for an amputee to dexterously control. Thus, the extent of appropriate movement capability of the prosthesis is highly dependent on the control interface approach. Understanding the balance of movement capability and control burden requires knowledge of advances in both areas, and additionally requires knowledge of appropriate assessment tools with which to measure functional efficacy.

This eBook provides an excellent overview of all these areas of research. Specifically, the eBook describes a number of recent advances in the design of prosthetic fingers, hands and arms, which provide considerably greater movement capability than previously possible. The text additionally describes recent work in which peripheral neural implants are used in an effort to provide motor control of a multigrasp hand prosthesis, and to provide some degree of sensory information to the user. Finally, the development of dexterous prostheses and associated control methodologies must be conducted in the context of functional assessment. As such, this eBook also includes important and expert treatment of methods for the functional assessment of upper extremity prostheses, and describes such assessment as applied to multigrasp prosthetic hands.

This is an exciting and promising time in the history of upper extremity prostheses. The functionality of such prostheses has changed relatively little over the past several decades. The technological advances presented in this eBook promise to change substantially the capability and functionality of upper extremity prosthetics, and by extension, promise to substantially improve quality of life for upper extremity amputees.

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PREFACE

The loss or the congenital deficiency of a human upper limb part represents a serious physical and psychological trauma, apart from having an evident and considerable restriction on personal autonomy in everyday living. Rehabilitating an amputee with a proper device allows the patient to recover (part of) the lost autonomy and the sense of psychophysical integrity, and thus to enable his/her reintegration in domestic, working and social environments. The prosthetic intervention is a complex process which involves technical aspects and clinical issues strictly dependent on the amputee to be treated. Prosthetic rehabilitation is therefore carried out by a multidisciplinary team including physicians, technicians, therapists and psychologists which operates with the aim of providing the amputee with the device and the services that best match his/her different requirements. Prosthesis developers study different solutions with the aim of optimizing the prosthetic system performances, its usability, wearability, and maintenance. Therefore, also in the design process many key factors of different nature are of primary importance.

Due to many reasons, the technological level of the powered artificial arms for upper limb amputees has always been fairly poor so far if compared with that of other analogous systems (e.g. lower limb prostheses, assistive robots, human-robot interfaces, biomedical robots…). However, there is no doubt that in the recent years the upper limb research stimulated the most exciting developments in prosthetic technology. Indeed, new terminal devices (articulated hands, sensorized hands, partial hands for finger amputation…) and novel articulations for the artificial arm (wrist and shoulder with one or more degrees of freedom, elbow joint) have been recently proposed (and some of them are also commercially available); a new concept of the socket has been developed; the control hardware, software and firmware are in continuous progress for the implementation of more and more effective control options for the wearers as well as for an easier management of the electronic boards. Finally, also the clinical treatment of the patients is improving: the most important novelties of the last years are the surgical technique known as Targeted Muscle Reinnervation for an enhanced myoelectric control of the artificial arm, the implantation of nano-sensors on nerves for the development of neuroprosthetic systems, and the prosthesis osteointegration for the direct suspension of the prosthesis to the residual limb.

A number of internationally renowned public agencies and industries are investing a great amount of financial resources. For instance DARPA (Defense Advanced Research Projects Agency of the USA government) that recently invested tens of million dollars in two ambitious programs (involving many research teams) intended to face the increased incidence of amputation injuries being seen in the ongoing conflicts in Afghanistan and Iraq. In particular the prosthetics industry is trying to exploit the fruits of the Revolutionizing Prosthetics Program RP2009, which is probably the most outstanding project in upper limb prosthetics so far. On the other side, the wave of renewal of upper limb prosthetics systems is dragging also many other manufacturers and research institutes, which recently presented interesting enhancements of their own products and prototypes.

The proposed eBook aims at illustrating the most significant milestones provided by the scientists in this new prosthetic research era and also sheds lights on new trends, future developments, and on the most challenging issues of the fascinating field of rehabilitation robotics. In Chapter 1 the most recent technology innovations are reviewed with the analysis on the critical issues involved in the design of upper limb systems. Chapter 2 focuses on clinical applications for the treatment of partial hand amputations with the new prosthetic solutions commercially available (Pro-Digits by Touch Bionics, SC). Chapter 3 presents the basic principles at the basis of

the neural control of a prosthetic device, with the results of an *in vivo* test performed in Rome (Italy). In Chapter 4 the milestones of prosthetics outcome measurements are presented, aiming at defining standardized protocols to evaluate prosthetic systems and prosthesis wearers. Functional and psychological evaluations are matters of Chapter 5, where the analysis of a patient wearing a new multi-grip commercial hand (Michelangelo by Otto Bock, GE). The last two chapters of the eBook are devoted to the presentation of technologies that can have a major impact in the next future for the upper limb prosthetic filed. In Chapter 6 the design guidelines for the development of anthropomorphic robotic hands are outlined, with the presentation of a bio-inspired hand with soft pads and compliant joints. Chapter 7 finally presents the framework for the development of virtual reality systems useful for the simulation of upper limb functioning, both for the development of new high-level prostheses and the training of patients in the first stages of their prosthetic rehabilitation.

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CHAPTER 1

The Design of Advanced Prosthetic Limb Systems

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Abstract: In the USA there have been many initiatives to develop advanced arm/hand prostheses in the light of the casualties seen in the wars in Iraq and Afghanistan. In this chapter the issues involved in some of these designs is presented as well as an overview of some of the more high profile prosthetic system development efforts.

Keywords: Prosthetics, Upper-extremity, Myoelectric, Amputation, Neural Interface, Decoders

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From the legends of The Early Mythological Cycle of the Oral Tradition of Irish Literature. The chief characters in this cycle belong to the Tuatha Dé Danann (The Tribes of the Goddess Danu), a supposedly divine race which inhabited Éireann (Ireland) before the coming of the Celts. "It was the law and custom of the Tuatha Dé Danann that no man who was not whole could be King. Nuada, King of the Tuatha Dé Danann, had led his people to Éireann and victory at the First Battle of Moy Tura but he also lost his arm in the battle. Thus was Breas the Beautiful made King by the Tuatha Dé Danann. The rule of Breas, son of Eolatan the Immortal of the Fomar, was an oppressive one and the proud Dé Danann were forced to pay tribute to Balor of the Evil Eye, King of the Fomar. Dian-Cecht, the Healer, fitted Nuada with an arm of silver, miraculously made, such that each joint answered his will as though it was his own flesh and blood. *Once again considered whole, Nuada was reinstated in the sovereignty by his people and from that time forth was known as Nuada Airgitlámh or Nuada of the Silver Arm."*
[From: "The Book of Conquests" by Jim Fitzpatrick, 1978. Paper Tiger Press.]

INTRODUCTION

Today's prostheses and interface techniques are still a long way from realizing Nuada's Silver Arm - *such that each joint answered his will as though it was his own flesh and blood.* To today's users every advancement in limb prosthetics is compared against re-creation of the physiological limb and the experience of the artificial limb. Although many people use current prostheses well and, in this way, accept the state-of-the-art, they are generally not satisfied with it. It is the nature of the work that prosthetics research is driven by dissatisfaction.

The ongoing conflicts in Afghanistan and Iraq have focused much media attention on prosthetics in the last few years. Watching this media, one is led to believe that there are a multitude of revolutionary new "bionic" arms available to our wounded troops. This is not the case. Soldiers needing upper-limb prostheses are being fit with similar upper-limb prosthetic systems to what was standard-of-care at the beginning of these conflicts.

What did change was that, due to the perceived lack of function of current commercially available upper-limb prostheses, a large number of major research programs were initiated to develop advanced replacement arm/hand systems, for persons with arm amputations. In particular, the Defense Advanced Research Projects Agency (DARPA) had two "Revolutionizing Prosthetics" initiatives. 1) A two year initiative, "Revolutionizing prosthetics 2007 (RP2007)", that was awarded to DEKA - the goal of which was to develop a multifunction prosthesis for above-elbow amputees using commercially available off-the-shelf (COTS) components and was to be controlled using conventional non-invasive techniques, and 2) a four year initiative, "Revolutionizing prosthetics 2009 (RP2009)", that was awarded to Johns Hopkins University Applied Physics Laboratory (APL) - the goal of which was to develop a fully functional biomechatronic analog of the human hand and arm. The arm was to be capable of duplicating the function of the original limb, including sensation, and withstanding the rigors of daily living and was to be controlled using advanced techniques such as brain machine interfaces or peripheral nerve interfaces or targeted muscle reinnervation with implantable myoelectric sensors.

The arms seen in the media were the European Cyberhand/SmartHand Project from Italy, the Rehabilitation Institute of Chicago's (RIC) Targeted Muscle Reinnervation (TMR) Arm, the DARPA RP2007 DEKA Arm, and various DARPA RP2009 APL Arms. These projects represent "what might be" rather than "what is".

The demographics of the upper-limb population make the design and development of arm/hand replacements a particularly challenging problem. In the USA lower-limb amputees outnumber upper-limb amputees by

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about 10:1. LaPlante and Carson [1] and Ziegler-Graham *et al*. [2] estimated that absence or loss of an upper-extremity limits 102,000 people in the USA. The most common is loss of one or more fingers (61,000): next most common is loss of one arm (25,000). While Dillingham *et al*. [3] in a review of hospital discharge records between 1988 and 1996 found that approximately 18,496 individuals were reported to have upper-limb amputations or congenital limb deficiencies annually. Ninety two percent of these were distal to the wrist i.e. partial hand amputations. Of the ~100,000 people with upper limb loss in the USA ~57%, are below-elbow amputees and 23% [4, 5] are above-elbow amputees. Most upper-limb amputations are traumatic and work related and therefore the individuals involved are typically 20-60 years old and still engage in active lifestyles. Atkins *et al*. [6] conducted a survey of 1,575 persons with upper-limb amputation and found 1,020 (65%) were body-powered users, 438 (28%) were electric users and 117 (7%) were bilateral users of prostheses. About ~80% all people with amputations use a prosthesis [7] which for ~50% of below-elbow amputees is an myoelectric prostheses [8] the remainder wear body-powered prostheses. However, it should be noted that while these studies tend to capture the dominant prosthesis used by the people queried, many active prosthesis wearers wear several different prostheses, crossing the entire spectrum of devices – the more active the person, and varied the lifestyle, the more different prostheses might be utilized – of both body-powered and electric.

A few of things arise from these observations: 1) There are not many people with upper-limb amputations - so the potential market is small. 2) The vast majority (93%) of people have a unilateral amputation therefore most of these individuals can live quite well with only their remaining intact arm. This means that the barrier to acceptance for arm/hand replacements is high. 3) The higher the level of amputation the greater the impairment and so too is the rejection rate.

Rejection of a UE prosthesis has been attributed to a multitude of factors: lack of cosmesis, lack of function, length of amputation, skin condition, post-amputation pain and discomfort, prosthesis control system discomfort, time of fitting (i.e., time since amputation), cultural and family attitudes, presence or lack of training, fitting techniques, quality of amputation surgery, absence of "team concept" of rehabilitation, ignorance of 3rd party funding sources, ignorance of doctors etc. Prime among these issues are interface comfort issues such as weight and ability to dissipate heat. The highest rejection rate is among individuals with unilateral trans-humeral or higher amputations because these individuals can live adequately with their sound limb and wearing a prosthetic arm is often more trouble than the perceived functional advantages. It is interesting to note however that just wearing an arm of some form - passive cosmetic device or active - can help to balance an individual and thereby help reduce future issues related to living with an unbalanced spine.

Current research on advanced arm prostheses can be divided into the following categories:

- Interface Techniques
- Decipher User Intent – what is the user trying to command the prosthesis to do
- Prosthetics technology/Mechatronic Systems.

Prosthetic arm technologies have historically offered patients limited functionality because there were insufficient control sources with which to control a mechanical arm with an equivalent number of degrees-of-freedom (DOF) to the natural arm. The function/performance of standard-of-care prosthetic systems is optimized to the control interface (type and number of control sources) available.

Standard-of-care prosthetic systems are either body-powered or electric-powered. In a body-powered system a body motion is captured and used to transfer force and excursion to an attached prosthetic component. Body-powered systems consist of a cable driven split-hook as a terminal device. It is the cable that is the interface between the user and the prosthesis that gives these systems their advantages. Because of the cable the user retains a sense of feedback about the terminal device through forces exerted on the body through the cable as the terminal device interacts with the world. Body-powered systems, while highly functional, simple, rugged and inexpensive, are also limited by their control interface - the control harness - to control 1 DOF at a time with switches used to cycle control between DOFs. In reality, there are only so many ways to harness someone to capture body motions before this becomes burdensome [9]. This technology dates to the end WW2 and has remained largely unchanged: a testament to its good design and simplicity? Perhaps it is also true that in the USA the reimbursement is such that prosthetists are not motivated to fit these devices which reduces the incentive for R&D in cable–driven technology. It is also true that many users, and society, expect more hand-like devices which tends to drive much of the research.

Electric-powered systems typically use myoelectric control to control at least some of the DOFs. In myoelectric control, (myo-)electrodes on the skin surface sense electric fields, the electromyogram (EMG), generated by muscles that lie beneath them. The EMG seen at the skin is a composite of all the active muscles in the residual limb [10]. Three, maybe four, independent (cross-talk free) surface EMG sites can be located on a residual limb [11]. A prosthetist needs skill to locate them and typically only two EMG sites are used in an agonist-antagonist pair over the flexor and extensor muscles of the forearm or the biceps and triceps for an above elbow amputee or the pectoralis and deltoids for a patient with a shoulder disarticulation amputation.

After amputation, there are not enough surface control sources to simultaneously control the many DOFs required to replace a physiological hand and arm. This lack of control sources combined with a lack of simultaneous control ability and a lack of robust/stable control signals reduce both body-powered and electric-powered prostheses to sequential control. Sequential control as currently implemented is slow and unintuitive. Consequently many amputees do not use a prosthesis because they can perform tasks more easily with their sound hand. The ability to simultaneously control 2 or 3 DOFs in a standard myoelectric arm prosthesis would represent a significant advance [9].

In the author's opinion, significant advances in the control interface will not occur without further surgical intervention to either revise the limb, implant some device, or some combination of both. But somebody who has undergone traumatic limb loss and much surgery is often highly resistant to further surgery.

INTERFACE TECHNIQUES

While advanced prosthetic arms can be designed with more joints for increased function, unless better control interfaces are developed these arm systems will not be able to be controlled. Interfaces that will provide the user with the increased number of independent control sources are needed to take advantage of the recently developed advanced multi-DOF arm systems. There are a number of interface techniques under development at this time to increase the number of control sites. All involve some form of surgical intervention and some involve permanent implantation of various devices. In the body, motor commands come from the motor cortex in the brain, descend through the spinal column, and go through peripheral nerves from the spinal column to the muscles of the arm. In an amputee some of the muscles of the arm are lost along with some peripheral nerves but much the command pathway remains. Research into interface techniques that use residual muscles, peripheral nerves, the dorsal root ganglia of the spinal column, and motor cortex in the brain are all ongoing. Our group, at the Biomechatronics Development Laboratory (BD Lab.), likes residual muscles since any technique must out last the life of an active individual (>80 years) and muscle signals are stable and large and relatively easy to interface with. All the other approaches still have lifetimes measured in 2-3 year range. Our approach is based on implantable myoelectric sensors, as described in the following section.

Implantable Myoelectric Sensors (IMES) for Prosthesis Control

Under NIH funding the BD Lab. is developing Implantable Myoelectric Sensor (IMES) devices [12] that can be chronically implanted into the residual muscles of an amputee's arm using minimally invasive surgical techniques (Fig. 1). These sensors receive power, digital addressing, and commands from an extracorporeal telemetry controller (TC) that will ultimately be built into an amputee's prosthetic socket. Each IMES acts as an individually addressable, intramuscular bipolar differential electrode, or myo-electrode, to detect the electrical activity (EMG) generated as a by-product of normal muscle contraction. The IMES transmit these EMG signals, over a transcutaneous magnetic link to the TC. With IMES no wires penetrate the skin. The TC "strips" the EMG data from the telemetry, and sends it to the prosthesis controller. This technological development addresses an important limitation in the development of sophisticated hand prostheses: obtaining sufficient physiological signals to control the many DOFs that compose complex hand functions.

By implanting an IMES sensor in small independently innervated muscles and sensing these muscle's EMG signal at its source the muscle acts as a biological amplifier of the neural command. There are some 18 muscles in the forearm related to the control of the hand. At present we pickup 2 independent sites using surface techniques. Our goal is to someday capture as many as 26 independent sites in the forearm (there are 3 muscles that each 4 independent slips running to each finger). Both experimental and modelling work suggest that the pickup area for an IMES is an ovoid about 5mm in radius about the long axis of the IMES [13]. This means we can implant in muscles with a diameter of 10mm and still expect crosstalk-independent signals.

The IMES system is currently capable of reading EMG signals from up to 32 sensors. The hermetically sealed package used for IMES is the same package used for the AMF Microstimulator which has been designed to survive in the body for 80 years [14]. Furthermore, the sensors can be implanted using minimally invasive surgical techniques in a relatively benign outpatient procedure [15] and can use the devices within a couple of weeks post-implantation. IMES has application anywhere myoelectric signals need to be sensed - be it upper or lower limb prosthetics or controlling powered orthoses for patients with ALS. We are at present working with Alfred Mann Foundation (Valencia, CA, USA) and Sigenics (Chicago, IL, USA) to get these devices FDA approved for use in People.

Targeted Muscle Reinnervation

Targeted muscle reinnervation (TMR) is a relatively new surgical procedure developed by Todd Kuiken and his Laboratory, Neural Engineering Center for Artificial Limbs (NECAL), at the Rehabilitation Institute of Chicago [16]. In TMR, muscles that are not functionally critical are deinervated and then reinnervated at multiple locations with nerves that formerly went to the hand and forearm. The muscle,

Fig. (1) – Holding an IMES in an Alfred E. Mann Foundation RFB BION® Package. The inter-electrode spacing (10 mm) is appropriate for whole muscle activity recordings.

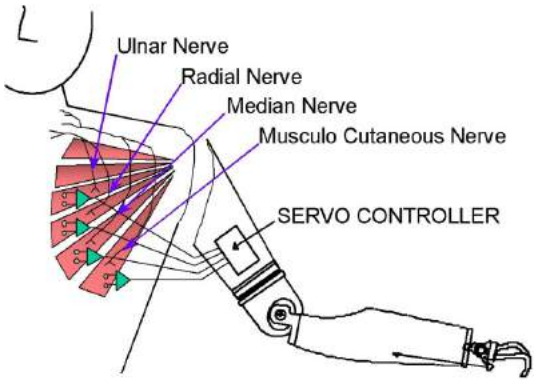

Fig. (2) – Targeted Muscle Reinnervaton. Schematic of how reinervation of the pectoralis muscle with the major nerves of the arm might be used to create physiologically correct myoelectric control sites for prosthetic arm control.

after reinnervation, should then be a good source of myoelectric sites for prosthesis control. The main advantage of this nerve-graft system is that there are potentially many discrete control signals available and that these control signals relate directly to the original function of the nerve.

In the case of the high level amputee (Fig. (2)) the median (M), ulnar (U), radial (R) and musclo-cutaneous (MC) nerves are usually still present. The musclo-cutaneous nerve controls elbow flexion while the radial nerve controls elbow extension. Pronation of the forearm is directed by the median nerve and supination by the radial and musclo-cutaneous nerve. Extension of the hand is governed by the radial nerve and flexion of the hand by the median and ulnar nerves. Since each of these nerves innervate muscles that control motion about different DOFs, they should supply at least four independent control signals. The nerves are controlling functions in the prosthesis that are directly related to their normal anatomical function. In addition, the shoulder is free to control other functions.

This approach has been successfully applied to number of people - the first being an individual who received bilateral shoulder disarticulation amputations following an electrical accident. In the case of the individual with the bilateral involvement, portions of the left pectoralis muscle were deinervated then reinnervated with the median, radial, ulnar, and musculocutaneous nerves. Three of the four targeted nerve/muscle grafts succeeded, enabling this individual to myoelectrically control his prosthetic elbow, wrist rotator, and gripper. The BD Lab. helped to develop the RIC TMR arm [17] that was subsequently fit to this individual to allow him to push the envelope on what he might be able to functionally control.

IMES and TMR can be viewed as complementary technologies. IMES can be used to read the EMG signals needed for TMR. TMR can be used to create hand sites in individuals who have lost all their hand musculature such as in the case of a trans-humeral amputee. Both groups work in the muscle because the signal amplitudes are much greater and the tissue is much more robust than nerve tissue. If a focal recording can be taken from a muscle then it is a good representation of the descending neural command. The down side of the TMR procedure is that it requires additional surgery which many amputees are resistant to and it involves a long rehabilitation time before any prosthesis can be worn. The patient needs to wait six months while the nerves regenerate before any device can be controlled.

Peripheral Nerve Interfaces

Peripheral nerve interfaces have the advantage that in the periphery the nerves are decoded by their geography i.e. if one finds a nerve in the left arm then one knows it is a left arm nerve as opposed to recording from neurons in the brain. Unfortunately for amputees one is also dealing with cut nerves which complicates matters when interfacing with nerves. Nerve tissue is sensitive to being penetrated so lifetime is an issue. Also all the current penetrating electrode technologies are too stiff, being made out of silicon, so there is a stiffness mismatch between the nerve tissue and the electrode causing irritation and necrosis of the nerve tissue. Typical nerve signals are in the micro-volt range, EMG signals are in the mV range. This means high gain amplifiers are needed to sense nerve signals.

Current approaches are exploring Longitudinal Intrafascicular Electrodes (LIFE). Dhillon *et al*. [18, 19] were able to demonstrate both command and sensation through a LIFE interface in human amputee subjects controlling a Utah arm. The SMART hand project (See below) in Europe used LIFE electrodes for their demonstration of peripheral nerve control.

Under the DARPA RP2009 project Utah Slant arrays

were being explored for peripheral nerve interfaces by Greg Clark and Dick Norman at University of Utah [20]. The Utah Slant array is a bed of nails approach similar to the Utah Array [21] used for Brain machine interface (BMI) work but with sloping electrodes so the cross sectional area of the peripheral nerve can be sampled. This work is still in the animal study phase.

Dustin Tyler at Cleveland Clinic [22] is using a cuff electrode array that fits around the outside of nerve to stimulate and sense from encircled fibres. This is the least invasive of the peripheral nerve approaches and therefore has the best chance of success - however being on the outside of the nerve it reads a composite signal from the neurons inside of the nerve - so it lacks the specificity of a penetrating electrode. Also explored are sieve electrodes where neurons are encouraged to grow through an array of electrodes [23]. Nerve interfaces are being worked on but they are still a long way from being practical. In my opinion, the Cleveland stimulating cuff electrode array holds the most appeal for use in sensory feedback back into the body since it is the least invasive of the approaches mentioned. Sensory feedback, or a sense of touch, is important if one wishes to perform true dexterous manipulation [24].

Brain-Machine Interfaces

Another human-machine interface technology that is under intensive development is brain-machine interfaces (BMI). A BMI involves placing electrode arrays on, or under, skull to detect EEGs, or under the Dura and into the brain tissue itself to detect individual neuron trains. Most commonly a "Utah Array like" [21] electrode array is inserted into the motor cortex region of the brain and individual neuron signals are passed out of the brain to an external computer that runs various complex algorithms to map the received neuron signals to an intended motion of a monkey arm or computer cursor or wheelchair controller or mechanical arm.

Prosthesis control is frequently given as a justification for BMI research. However, in the author's opinion, it is not reasonable to ask someone who still has the use of an intact arm to undergo brain surgery and have the immunity of their brain compromised by penetrating their Dura. For someone with high level quadriplegia whose choices are more limited this level of invasiveness may be warranted.

One of the leaders in this area is John Donoghue's group at Brown University and part of the VA COE on Restorative and Regenerative Medicine - they are the developers of the BrainGate2 Neural Interface System and are the only people to have implanted humans with a BMI [25, 26]. Working with individuals with high level tetraplegia the BrainGate research consortium has implanted a number of their BrainGate2 Neural Interface Systems and used them to control external devices such as a communication device, desktop computer, wheelchair, or prosthetic arm, or working with VA Cleveland COE on functional Electrical Stimulation (FES) to use FES to control a paralyzed limb. There are many other teams working in the area of BMI using EEGs [27] or invasive approaches with monkeys [28, 29]. Miller [30, 31] at Northwestern is notable because he is looking at sensory feedback as well as motor commands. Greg Clark [20] at Utah was part of the DARPA RP2009 BMI team as was Richard Anderson [32] of Caltech, and Marc Schieber at Rochester. A good review of neural interfaces can be found in Horch and Dhillon [33]. The BMI aspect of the RP2009 initiative has come to dominate phase III of this project.

DECIPHER USER INTENT

Once control signals have been transduced the prosthesis controller must determine from these signals which motors to drive in the prosthesis. In the BMI world this is known as the Decoder. The various control approaches for EMG signals fall broadly into one of the following categories.

Direct control: This is the most primitive, simple and robust approach, and represents current commercially available prosthetic devices. In Direct control the user voluntarily contracts one muscle in the absence (independently) of all others. If intuitive control motions can be found Direct control can provide intuitive simultaneous control of multiple DOFs within a prosthesis. Generally, the prosthetic component is driven at a speed that is proportional to the difference in the amplitude of the two EMG signals. TMR patients use this approach for their take home prostheses. The BD Lab. is exploring use of this approach with IMES to simultaneously control multiple DOFs in a dexterous hand prosthesis [34].

Control using Pattern Recognition: Developed over the last 20 years by University of New Brunswick Canada, a Pattern recognition based controller recognizes a specific pattern of EMG signals and executes an associated function. Pattern recognition approaches requires a pattern to be stored for every movement. From the user's perspective the control is intuitive and easy to remember since users execute the movement they want the prosthetic limb to perform with their "phantom limb." These approaches do not provide true parallel/simultaneous control of multiple DOFs. This approach has been dogged by clinical issues such as having to retrain the system every time the system is donned, or electrode lift-off or muscle motion problems.

Control by Internal Model: An internal model approach uses EMG signals measured in the user's residual muscles to predict muscle activation in an anatomically correct biomechanical muscle model of the intact limb [35, 36]. These activations drive muscles in a simulation of the intact hand causing joints in the hand and wrist of the simulation to move. The simulation predicts which joints should move.

Based on these predictions the prosthesis controller commands the motors of the prosthesis to match the joint positions of the internal model. This approach holds the promise of true simultaneous/dexterous control of an artificial hand.

Endpoint control: In endpoint control the user commands the end position of the hand rather than thinking about the position of each joint. The user focuses on the hand position and commands up, down, left, right, forward or back and the controller computes what the joint position for the shoulder, elbow and wrist should be to place the hand where the user wants it. This approach was first used by Simpson in the 70s [37] and is now being revisited by DEKA for the DARPA RP2007 arm.

PROSTHETICS TECHNOLOGY/ MECHATRONIC SYSTEMS

This is the physical prosthesis hardware. There have been a number of initiatives to develop multi-function arms and hands in the past few years as well as the introduction of a number of new commercially available multi-articulated hands. Recent developments in this area have been dominated by the DARPA "Revolutionizing Prosthetics" programs which were in turn driven by the injuries being seen as a result of conflicts in Afghanistan and Iraq. The recent arm systems are as follows:

- Cyber Hand/SMART hand Project in Europe.
- RIC Six Motor Arm
- DARPA RP 2007 DEKA Arm
- DARPA RP 2009 APL Proto 1 Arm
- DARPA RP 2009 APL Proto 2 Arm
    - Intrinsic Hand
    - Extrinsic Hand
- DARPA RP 2009 APL Final Limb System
- The UNB Hand

Only the CyberHand project (now the SMART hand project) in Europe and RIC Six Motor Arm pre date the conflicts in Afghanistan and Iraq. The BD Lab. has been involved in all these initiatives with the exception of the European Cyber Hand project.

CyberHand/SMART Hand Project

The Cyber Hand/SMART hand project is made up of a consortium of institutions led by Paolo Dario at the Advanced Robotics Technology and Systems (ARTS Lab), Scuola Superiore Sant'Anna, Pisa, Italy (Fig. 3). The SMART hand mechanical design consists of five under-actuated fingers [38]. All fingers have 3 phalanxes and have angular and grasping sensors. The thumb has a two DOFs. The hand architecture is a 4 DOFs hand with a 2 DOFs thumb, an index finger and a combined Middle-Ring-Little finger drive. The mechanism uses cables to articulate the fingers. Grasping capabilities, robustness, cosmetics, small weight and human size as design philosophies were

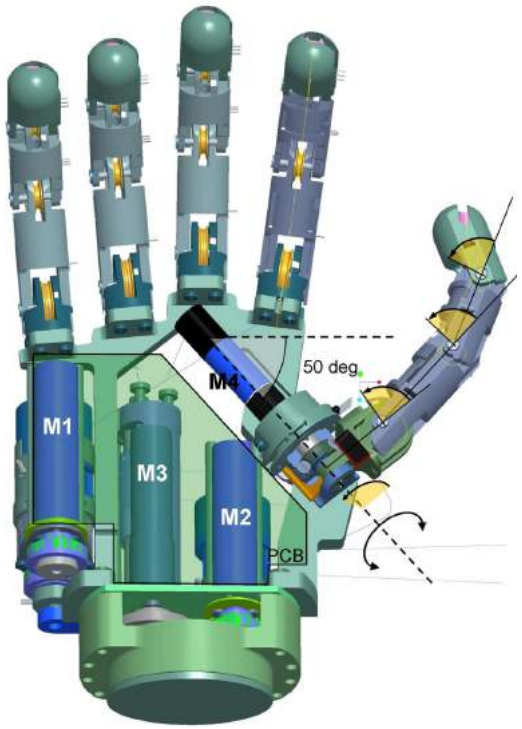

Fig. (3) – SMART hand CAD model showing how motors sit in the hand to pull fingers and thumb closed.

emphasized over high dexterity, high power and manipulation capabilities. This hand is capable of grasping function and not object manipulation within the hand. The project has recently garnered a lot of media attention because it implanted LIFE electrodes into the peripheral nerves in the arm of a subject in Sweden [39]. This subject was able to elicit sensations and commands with their temporary implant similar to those reported by Dhillon *et al*. [18, 19].

The RIC Six Motor Arm

The RIC Arm (Fig. 4), was developed just before the conflicts in Afghanistan and Iraq started as a research system for TMR [17]. It was built to push the envelope in terms of how many DOFs might reasonably be driven by someone with bilateral shoulder disarticulation amputations and TMR. Built around a Liberating Technologies Inc., (LTI) Boston Elbow, the arm was an international affair consisting of a hand with built-in wrist flexion unit from Kesheng Prosthetics CO Ltd., Shanghai China, a wrist rotator from Otto Bock, Vienna, the Boston Elbow, LTI, USA, a humeral rotator [40] and an Edinburgh Modular Arm system, (EMAS) shoulder, from Touch Bionics, Scotland [41]. The BD Lab., in addition to designing the humeral rotator, helped Dr. Kuiken to integrate the components into a system and patch into the electronics to make EMG sensors capable of reading chest EMG without picking up heart beat [42]. This arm was built around a centralized controller in

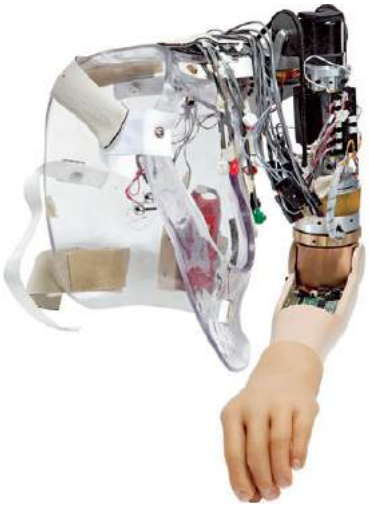

Fig. (4) – RIC Six Motor Arm. Developed for use with TMR patients.

the Boston Elbow - it did not use a bus system making it easy to work with. It is still used by NECAL at RIC today because of the straightforward way the drives are interfaced to the central controller.

The DARPA's RP2007 and RP2009 initiatives saw huge investment in the development of advanced upper extremity prosthetic arm systems. It has had a huge impact on all areas of upper-limb system development as can be seen by the control interface work in the area of BMI and peripheral nerve interfaces. While this has been a good thing for the field of P&O it should be realized that the specifications for all the arm systems developed were based on a unilateral trans-humeral amputee or higher - because the highest rate of non-prosthesis use is among this population. This decision initially drove the design discussion away from making systems suitable for the largest population segment of upper-limb amputees i.e. trans radial amputees. In spite of this both the RP2007 and RP2009 have developed trans-humeral solutions.

DARPA RP2007 DEKA Arm

This is a modular powered arm system with 10 powered DOFs (Fig. 5). The hand has independent control of a 2 DOFs thumb, a 1 DOF index finger, and a Middle, Ring, and Little (MRL) finger unit. All fingers are kinematically linked so they can wrap around objects as they are flexed. There is a 2 DOFs wrist with rotation and flexion/extension, powered elbow, humeral rotator and a 2 DOFs shoulder with flexion/extension and abduction. DEKA Research (Manchester, NH) also explored dynamic socket concepts [43] and with Randell Alley of BioDesigns, Inc., developed the “High-Fidelity Interface” [44]. The High-Fidelity Interface uses an alternating series of compressions and tissue release areas working in concert with the dynamic elements involved. Allegedly, this allows a much heavier weight to be

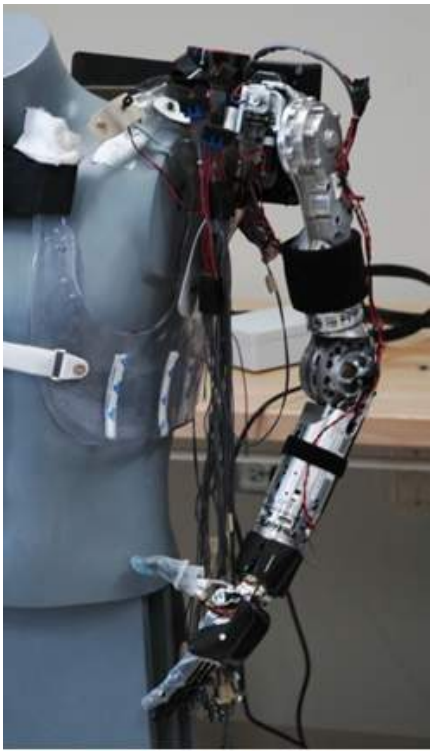

Fig. (5) – Generation 1 DEKA Arm.

borne by the user while offering significant improvements in range of motion and heat dissipation. The DEKA arm is designed to be controlled with conventional control approaches and with a foot based inertial controller. Because of the trans-humeral amputee specification the first version of the hand is too long to be used on all but the shortest trans-radial patient. The DEKA arm is currently undergoing an optimization study run by the VA [45]. The goal of this study is to inform the design of the Generation 3 version of the DEKA arm by providing real use data to the engineers.

DARPA RP 2009 APL Prototype 1 Arm

The DARPA RP2009 initiative was led by APL. This program was structured around the development of three arm systems. Prototype 1 was a 7 DOFs arm which was to be provided as a take home prosthesis for those subjects undergoing TMR as part of the program. This arm was designed by the BD Lab. in collaboration with Otto Bock, Vienna, with electronics and control software provide by APL. It was built in one year with supposedly off-the-shelf-components but in reality everything had to be designed from scratch (the wrist flexion unit, the humeral rotator, and shoulder, all electronics and software) or modified (Michelangelo Hand, wrist rotator, Dynamic arm). At the time the Michelangelo Hand existed only as a first prototype and the Dynamic arm had to be modified so it could be run using an Axon bus for control (the Axon bus is the new proprietary communications bus developed by Otto Bock to control the joints in the

arm). The humeral rotator, shoulder and wrist were designed the BD Lab. (Fig (6)).

The Michelangelo Hand is the new hand from Otto Bock. It has a speed of 4 rad/sec and a pinch in excess of 28 lbs_f. It consists of 2 drives. A main drive that pulls all fingers and thumb closed together and a thumb switch drive that moves the thumb into lateral prehension or palmar prehension. The main drive is based on the EC45 drive from Maxon and has all electronics for motor control and bus communication integrated into it. We used the Michelangelo main drive for both the humeral rotator and shoulder. This arm worked surprisingly well and was fitted to a number of TMR patients at the RIC (Fig (7)). This arm, without the BD Lab. components, forms the basis for Otto Bock's TMR arm announced at ISPO 2010.

DARPA RP 2009 APL Prototype 2 Arm

The Prototype 2 arm (Fig (8)) was a built as a technology demonstrator to prove that conventional DC electric motors could used for the actuators and still meet the DARPA specifications. The arm had to be able to lift 50 lbs at the elbow and 90 lbs at each DOF at the shoulder and have no-load speeds of at least 120 degs/sec at each joint. The arm was built by New World Associates (now Hunter Technology). Because this arm was specified for persons with trans-humeral or higher amputations the design team assumed that the forearm volume was available to them for hand actuators. This was in spite of it being pointed out that this would preclude persons with trans-radial amputations. In light of this and the availability to Otto Bock of some new small 10 mm motors, the BD Lab. and Otto Bock were tasked with designing a hand that met the DARPA hand specifications but that would be suitable for trans-radial amputees. Thus two hand systems were built for prototype 2. A 15 DOFs Extrinsic hand with actuators in the forearm built by Hunter Technology and Kinea (Evanston IL), and an 18 DOF Intrinsic hand built by the BD Lab. and Otto Bock. APL provided electronics and software for both hands and arm. Both hands were designed to do 311N for cylindrical prehension, 89N for tip prehension and a speed 360º/sec per finger.

Fig. (6) – Prototype 1 arm based on early version of the Michelangelo Hand and modified Dynamic Arm.

Fig. (7) – Trans humeral TMR Subject wearing Prototype 1 arm.

Fig. (8) – Prototype 2 Arm capable of curling 50lbs at the elbow.

Extrinsic Hand

This hand was a beautiful piece of engineering in which a Cobot approach [46] was used to pull cables that crossed the wrist and went into the hand. Springs were used to provide finger extension - similar to the Cyberhand. A Cobot was attractive because it has a continuously variable transmission (CVT). With a CVT it is possible to adjust the effective gear ratio and keep the motor operating at its most efficient point. The basic architecture was to have a single large modified Maxon EC45 motor rotate a central shaft at a constant speed and then to have a series of 5 'pucks', each with 3 steering drives, sit along this central rotating axis. Each puck controls 3 DOFs (Fig. (9)). The pucks essentially act as a power "take off" - flowing power from the large motor through the small steering motor to drive a winch which winds in or releases the cable to actuate a DOF. The angle of attack of a steering motor on a roller pressed against

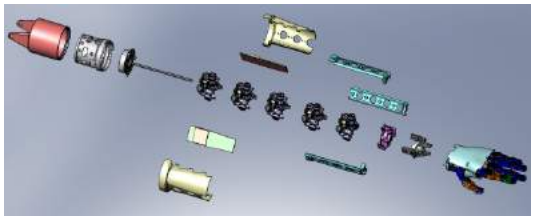

Fig. (9) – Extrinsic hand showing Cobot Module of 5 pucks each with 3 CVTs.

the central shaft dictates the amount of power to flow in that DOF. The Cobot idea is very elegant but complex. This hand used all custom motors and drives. In addition the cables were very hard to maintain and controlling the hand to get a desired motion was very complex since it involved having to control multiple cables simultaneously. The electronics also turned out to be very difficult due to the number of wires that had to be accommodated for each drive.

Intrinsic Hand

To develop a fully functional hand suitable for persons with trans-radial amputations all the actuators necessary to drive 18 DOFs had to be intrinsic to the body of the hand (some 15 motors in all) with another 3 motors housed in the wrist. For this hand everything from sensors, to high level processors, low-level control electronics, wires, motors, prosthetic socket, cosmetic covering, and power source had to fit into a volume that was broadly anthropomorphic and weigh no more than ¾ kg [47].

Custom brushless DC electric motors, developed by Maxon (Switzerland), capable of providing high torque at physiological hand speeds were used for the 15 drives in the hand and Axi motors where used for the wrist. These motors were coupled with what were supposed to be high torque capacity Wolfrom transmissions to form the basic actuator for the hand. Each finger had 3 articulations with 2 motors. The distal and medial phalanges were driven by 1 motor coupled by a differential drive mechanism - proximal interphalangeal/distal interphalangeal (PIP/DIP) drive. The proximal phalange had its own drive - the metacarpophalangeal (MCP) drive. In the palm, the index, ring and little fingers had ab/adduction (Ab/Ad) drives. The thumb had 4 DOFs each with its own drive. The wrist was supposed to do flexion-extension, radial-ulnar deviation and wrist rotation. Low-level electronics to handle basic motor operation such as commutation and transduction of local sensors were developed by APL (Fig (10)). In addition, force and position sensors were integrated into each finger tip and joint axis. All these sensors and motor controllers were coupled to a single large hand control module that was ultimately to be mounted in the back of the hand.

During testing, the robustness of the electronics suite

Fig. (10) – The finished Proto 2 Intrinsic Hand.

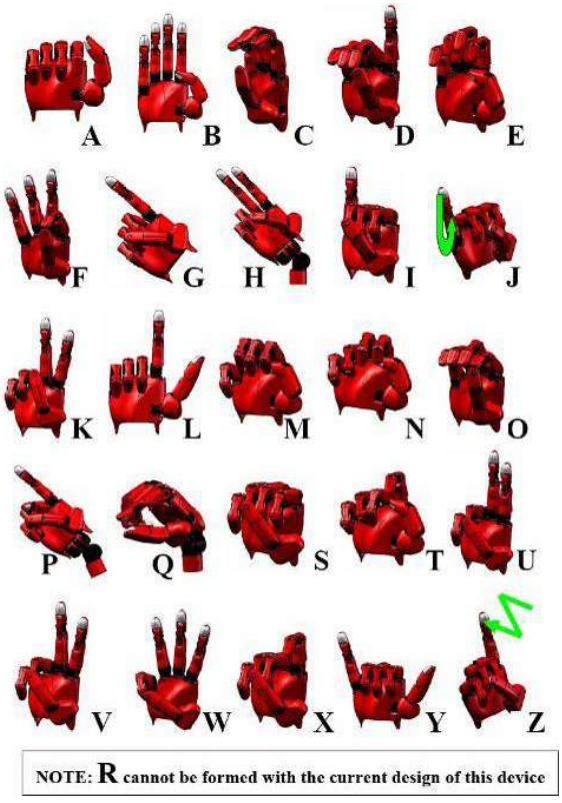

Fig. (11) – CAD model of the Intrinsic Hand recreating the American Sign Language Alphabet.

became a source of concern with the largest problems stemming from wires pulling free of the motors, connectors and motor control boards as hand grasps were performed. Moreover, Wolfrom drive efficiency problems resulted in lower than predicted grips. The wrist electronics were never fully integrated and therefore never ran. Although not all of the design goals were met, the time between design concept to a functional prototype donned by a test subject was less

than one year.
The Intrinsic hand was mechanically capable of generating all but one of the American Sign Language (ASL) postures (Fig (11)). The Intrinsic Hand did not have the capability to do "R" which needs to cross the middle finger over the index finger – this could be achieved by increasing the range of MCP extension for the fingers and the range of adduction of the fingers. This demonstrates the Intrinsic hands capability of complex posture generation, a necessary prerequisite for dexterous manipulation.
This hand, of all the hands mentioned in this chapter is the only one capable of true dexterous manipulation because it had independent control of the MCP and PIP joints in each finger as well as control of finger and 4 DOFs for the thumb (See Mason and Salisbury [48] for theoretical requirements for dexterous manipulation). Ultimately an Intrinsic hand architecture was chosen for the Modular Prosthetic Limb (MPL) hand because it could meet the DARPA specifications in terms of speed and grip strength and could do so in a form suitable for persons with trans radial amputations.

DARPA RP2009 APL Modular Limb

The Modular Prosthetic Limb (MPL) (Fig (12)) arm is an iterated version of the Prototype 2 arm. The MPL hand maintained the intrinsic architecture but the palm and fingers were redesigned and the drives where completely redesigned. Motors were second sourced from Emoteq rather than Maxon. The Wolfrom transmission was eliminated in favour of a planetary transmission for the hand, thumb, Ab/Ad drives, and one of the wrist drives. The BD Lab. and Otto Bock were tasked with developing a 2 motor finger (2MF) variant to enable dexterous manipulation for this limb but by the end of the second phase of the DARPA RP2009 project the priorities for whole RP2009 initiative were changed and the focus was shifted to obtaining BMI control of the arm. To execute this program shift, resources were diverted and the 2 motor finger variant shelved, settling for a simpler hand with single motor fingers and a single Ab/Ad drive for all fingers.
The resulting hand has fewer DOFs then the Intrinsic hand of prototype 2, it no longer has independently controlled PIP and MCP joints and has a single Ab/Adduction drive and a 4 motor thumb. The end result is a hand that is capable of sophisticated grasping function only.

Fig. (12) – Photograph of the DARPA RP2009 Modular Prosthetic Limb as it is being assembled. The hand is shown with both 2 motor fingers (index and middle fingers) as well as one motor fingers (ring and little fingers) and a 4 motor thumb.

The UNB Hand Project

The UNB hand [49] (Fig. 13) aims to create a hand solution that optimizes function with respect to the number of drives needed [50, 51]. In order to create a hand capable of executing the six basic grip posture of Keller *et al*. [52] it was determined that 3 degrees of motion were required. Additionally, the limitations posed by using two or three articulations per finger along with experience gained from other multi-DOF hands was incorporated into the overall design specification. In this hand the index finger and the thumb are individually actuated at the meta-carpal phalangeal (MCP) joint and carpometacarpal (CMC) joint respectively. The middle, ring, and little (MRL) fingers are actuated by a single drive followed by two differential gearing systems. The differentials, located at the MCP joints of the MRL fingers, are configured similar to a whiffle tree, and balance the torque between the fingers while providing independent movement of each finger.
All digits have a kinematic linkage system within the proximal phalange which creates an extra degree of motion about the proximal interphalangeal (PIP) joint. Enabling the fingers to conform (wrap around) to the shape of the object grasped. To execute both lateral

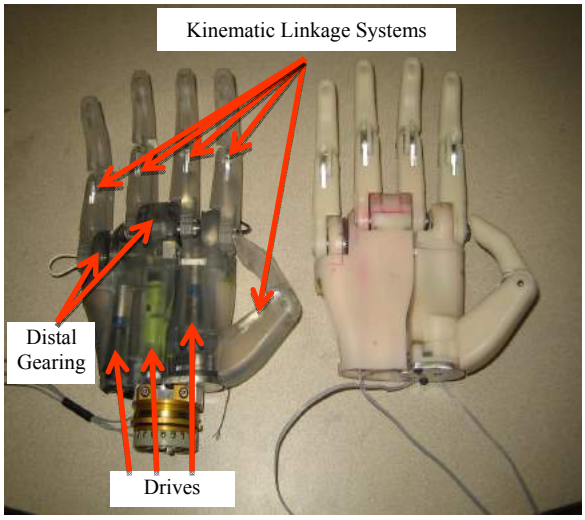

Fig. (13): Photograph of first prototypes of the UNB Hand. The current iteration is much narrower and is appropriate for an adolescent or a small woman.

prehension and palmar prehension with a single DOF thumb a novel cam profile was developed to enable the thumb to execute lateral prehension at full thumb open, maximum hand opening at mid-range of its travel and palmar prehension at the full close end of the thumb travel. Compliant elements in the MCP and PIP joints allow for compliance in the flexion-extension and abduction/adduction directions respectively. Where possible, the mechanical design uses off-the-shelf components in order to minimize production costs. The drives use coreless brushed DC motors and associated 4-stage planetary gearheads. The distal gearing and differential systems are modified parts designed to withstand maximum loading conditions. The majority of the purchased components are also modified to fit within the hand envelope. The UNB hand is a multi-axis adaptive hand prosthesis currently in its third iteration that can form all six basic hand grasps through the novel thumb architecture. The mechanism is able to both passively and actively adapt to the shape of a grasped object and is short enough to be fitted as a trans-carpal hand. The mechanism includes the local microprocessor controller.

CONCLUSION

A lot of arms and hands have been described here and for all of them their true function cannot be taken advantage of unless better control interfaces are developed. As mentioned, there is much research going on in the realm of neural interfaces but for the most part it is still research and a long way from being of clinical use. In the author's opinion further surgical intervention to create additional control sources or to implant devices will be necessary to create these advanced interfaces. The downside is that individuals who have undergone traumatic amputation are often resistant to further surgery. Also, those individuals with unilateral involvement can function adequatly with just one arm, meaning they do necessarily need what we can offer them – so the barrier for acceptance is high. If a device is too heavy, a hassle, to uncomfortable it will not be worn. It is also my opinion that when designing hands you should build to the control interface available if you want a clinically viable device. Many of the devices mentioned do not have an adequete control interface. In the case of the DARPA RP2009 project we were told to ".. *to assume the control would be available*.." and given that what could we build. So given carte blanc we built the Intrinsic hand - a hand with 18 DOFs. Although the control does not yet exist to take advantage of this hand. I view this hand as a technology demonstrator and an existance proof to say that "yes" we can build such a hand and it worked. We can say to the neural interaface people we are capable of making these devices if you can provide the control. These are all excititng projects that push the envelope of the available systems and now the media is gone we need to settle in for the other 95% of the effort needed to make these devices reliable and ready for use in a clinical enviroment.

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CHAPTER 2

Experience Fitting Partial Hand Prostheses with Externally Powered Fingers

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Abstract: Prosthetic management of partial hand amputation poses many challenges to prosthetists and other treating professionals. Partial hand amputations have been challenging to fit with externally powered devices due to the limited space available for prosthetic mechanisms. It has long been the goal of prosthesis designers to mimic as many of the six commonly referenced grasp patterns as possible. With the commercial introduction of individually powered fingers exciting possibilities for fitting externally powered finger prostheses that can replicate various hand postures is now feasible. Powered fingers have allowed individuals with partial hand absence to regain some of the dynamic and conformable grasp functions they lost. This chapter will present a general overview of prosthetic options available for partial hand prostheses with specific focus on externally powered fingers.

Keywords: Partial Hand, Prosthetics, Powered Fingers, Multifunctional, ProDigits, Vincent Systems.

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INTRODUCTION

Prosthetic management of partial hand amputation poses many challenges to prosthetists and other treating professionals. Loss of a hand causes obvious functional disability but also has substantial psychological and social impact. The hand is an incredibly complex instrument with its many joints providing an almost unlimited numbers of possible hand postures. Sensory feedback provides a wealth of information of both hand position and information about the environment the hand encounters. Prosthetic replacement of these features represents a monumental challenge which is presently unattainable. Loss of sensation alone in an otherwise functional hand causes significant disability, this reality underscores the difficulty in replacing an amputated hand.

It has long been the goal of prosthesis designers to mimic as many of the six commonly referenced grasp patterns as possible [1]. To date there have been four multifunctional hands introduced for commercial distribution, the iLimb hand from TouchBionics (www.touchbionics.com), the BeBionic hand from RSL Steeper (www.rslsteeper.com), the Michelangelo hand from Otto Bock (www.ottobockus.com), the Vincent hand from Vincent Systems (www.vincentsystems.de). These hands are primarily designed for persons with amputations above the wrist. For the majority of partial hand amputees there are only two commercially available electrically powered options presently available; ProDigits from TouchBionics and Vincent fingers from Vincent Systems. With the introduction of these individually powered fingers exciting possibilities for fitting externally powered finger prostheses are now feasible. This chapter will present a general overview of prosthetic options available for partial hand prostheses with specific focus on externally powered fingers.

INCIDENCE

Partial hand amputation is the most common upper limb amputation level in the United States. In a review of hospital discharge records between 1988 and 1996, Dillingham *et al.* [2] found that approximately 18,496 individuals were reported to have upper-limb amputations or congenital limb deficiencies annually. Ninety two percent of these were below the wrist. Trauma is the predominant cause of partial hand amputations and it is important to note that often the remaining portions of the hand are also damaged. Limited joint range of motion, hypersensitivity, scarring and a lack of strength in the remaining portions of the hand may be complicating factors.

PROSTHETIC OPTIONS

Until recently prosthetic management of partial hand amputation has relied on three basic categories of prostheses: aesthetic passive hand restorations, opposition posts, and body powered designs. When deciding on the type of prosthesis to be prescribed, variables include age, sex, occupation, degree of physical activity, gadget tolerance, type of amputation and unilateral *versus* bilateral involvement. As with other upper limb amputations two or more prostheses may be required to meet the multitude of patient

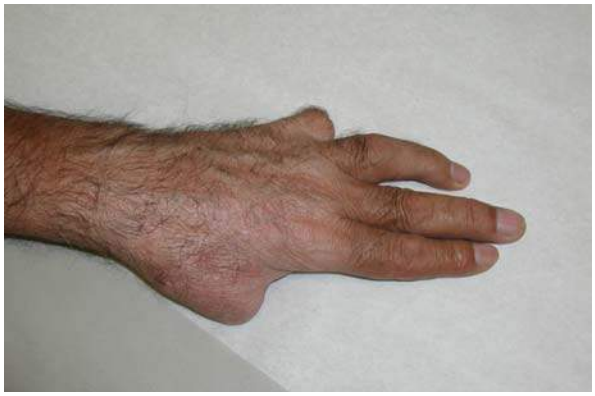
Vincenzo Parenti Castelli and Marco Troncossi (Eds)

needs. Use of a trial prosthesis can be invaluable in determining if a particular type of prosthesis will meet the users needs and expectations.

Silicone Hand Restorations

High definition custom silicone prostheses have been, and continue to be, the best option for reproducing the natural appearance of the hand [3]. Pillet pioneered the use of custom silicone hand prostheses that very closely match the color and form of the missing hand allowing the user's hand absence to go unnoticed by the casual observer. This is an important point because if the individual with a hand difference resorts to hiding his hand in a pocket to avoid being noticed the extremity is functionally more disabled than necessary. Silicone prostheses have a long history of use and are generally well accepted [3, 4]. These prostheses provide static replacement of the missing parts of the hand and are often referred to as passive prostheses since the prosthetic components do not provide active motion. However, these prostheses do offer function in addition to the extremely important psychological benefit of restored body-image [3]. The amount of function they offer is dependent on the active features of the remaining hand. For example if the thumb is the only finger remaining the silicone fingers provide opposition for the thumb allowing the hand to grasp objects, a function that would not be possible without the prosthesis.

Silicone prostheses also can serve to protect sensitive areas of the amputated hand. Custom silicone hands provide good stain resistance however the material can be damaged if subjected to use for manual labor. Since silicone prostheses do not provide active finger function they do not provide the grasping functions required by many partial hand amputees.



(a)

(b)

Fig. (1) – (a) Hand requiring opposition post to allow grasp using the intact index, middle and ring fingers; (b) Prosthesis made of a rigid plastic with compliant foam on the gripping surface.

Opposition Posts

Opposition posts are best utilized by persons with either their thumb remaining and fingers missing or thumb missing and fingers remaining. As the name implies an opposition post consists of a rigid post attached to the remnant hand with a prosthetic socket in order to oppose any remaining fingers. Opposition posts are generally simple and durable (Fig. 1).

Body Powered Options

Historically there have been attempts to provide body-powered prostheses for partial hand amputations such as the Robin Aids hand which could be configured in various ways to replace complete loss of the fingers. This hand used a shoulder control harness. However, the Robin Aids hand is no longer commercially available. More recent body-powered systems include the X-Fingers (http://www.didrickmedical.com) and M-Fingers (http://partialhandsolutions.com).

The X-Fingers use a multiple linkage design, typically used to power the distal interphalangeal (DIP) and proximal interphalangeal (PIP) joints with the force and excursion of the metacarpophalangeal (MCP) joint of the affected finger (Fig. 2). In the authors' opinion the X-Finger has limited indications for use since it is primarily a finger driven design and requires enough lever distal to the MCP to produce useable force. Cosmetic covering is problematic.

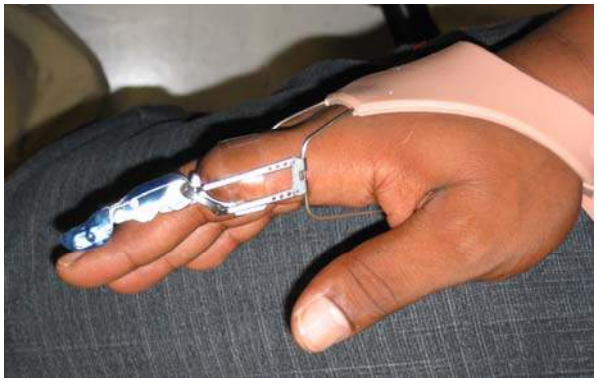
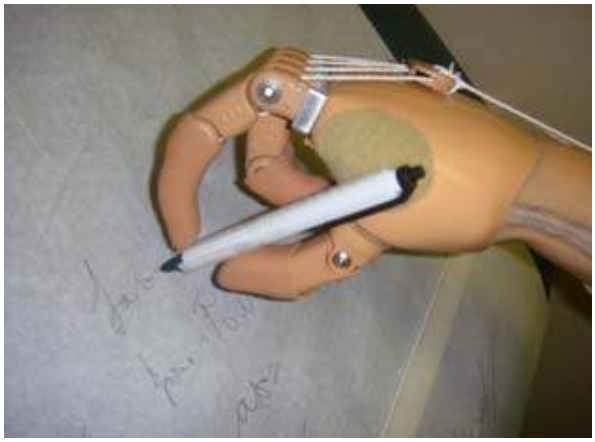


Fig. (2) – X-Finger prosthesis fitted for amputation of the index finger at the PIP joint.

The M-Fingers use a cable actuated wrist driven design where wrist flexion causes finger flexion at the MCP and PIP joints of the prosthetic fingers, the DIP joint is fixed [5] (Fig. 3). The M-Finger prosthesis uses a Whiffletree design to allow conformable grasp where finger motion continues until blocked by the object being grasped. Grip force is low due to the mechanical design, however due to the conformable grasp prehension is adequate in many cases. A disadvantage of the wrist driven design is that wrist motion is linked to finger flexion and neither can be positioned independently. Also sustained grip force requires maintenance of wrist position. Cosmetic finishing is difficult. The M-Fingers are also available in a finger driven design for partial finger amputees. These fingers use MCP flexion to drive PIP flexion by routing a cable from the dorsal side of the intact MCP joint to the palmar side of the prosthetic PIP joint (Fig. 3b). Use of the M-Fingers for finger driven control requires sufficient length of the involved finger distal to the MCP joint in order to produce adequate force and excursion.



(a)

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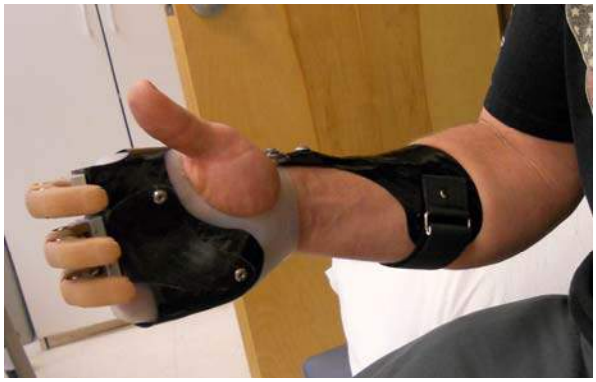


(c)

Fig. (3) – (a) M-Finger prosthesis fitted to a trans-metacarpal amputee; (b) bilateral M-Finger fitting with wrist driven design on the left and finger driven design on the right;(c) three finger M-Finger prosthesis with wrist level silicone socket and prepreg carbon frame.

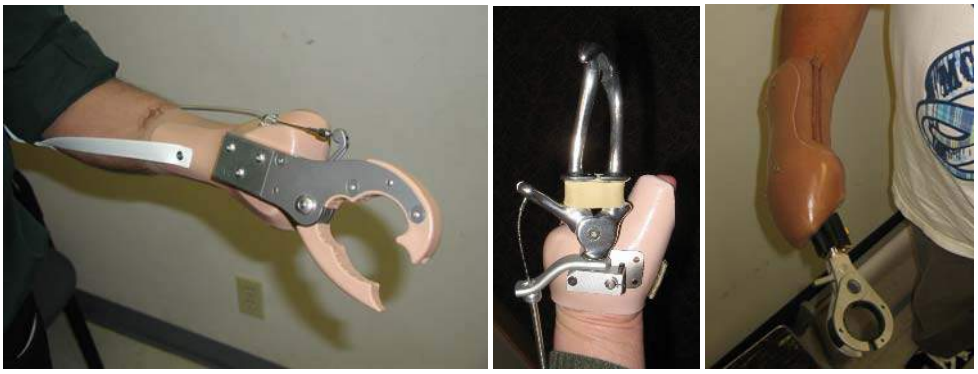
Another body-powered partial hand option is the so called “handi-hook” (Fig. 4). This device uses a conventional hook prehensor attached in the palm of the partial hand prosthesis using a rigid socket, flexible hinges and a figure of 8 harness. Control is achieved with combinations of elbow, shoulder, and scapular motions. This type of control produces force and excursion for good terminal device function. The disadvantages of using proximal joints to control the terminal device are that the terminal device is linked to these proximal motions and cannot be controlled independently and these motions may appear unnatural. Body-powered cable actuated control has the inherent advantage of providing proprioceptive feedback to the user regarding force, position and speed of movement through the linkage of the controlled component to the proximal physiological joints [6].

Externally Powered Options

Partial hand amputations have been challenging to fit with externally powered devices due to the limited space available for prosthetic mechanisms. Early powered prostheses for partial hands such as those described by Weir [7], Gow [8], Putzi [9], Biden [10], and Lake [11] were not commercially available. The commercially available Otto Bock transcarpal hand is not configurable for different finger absences and is best suited, as the name implies, for the most proximal of partial hand amputations.

A distinction between the attempts at providing externally powered solutions for partial hand amputees is whether the drive mechanism is in the body of the hand or contained in the fingers. For the mechanism to be applied to the largest number of partial hand amputees the drive mechanism should be contained in the fingers. Only the designs originally proposed by Weir [7] (Fig. 5), Gow [8], and more recently Schulz [12] satisfy this requirement.

Electrically powered partial hand prostheses offer advantages over the body-powered systems discussed above. No force or excursion is needed for operation and finger activation can be independent of proximal



(a) (b) (c)

Fig. (4) – (a) a voluntary closing hook (www.oandp.com/products/trs) with palmar attachment; (b) a partial hand fitted with voluntary opening hook (http://www.hosmer.com) [photo courtesy of Wayne Daly]; (c) a quick disconnect mechanism shown with passively positioned gripper (www.n-abler.org).

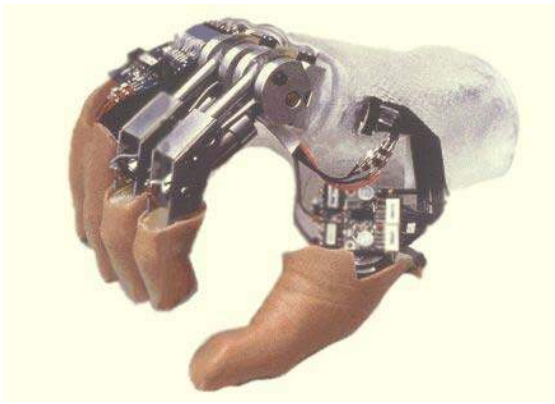


Fig. (5) – Prototype partial hand prosthesis with individually powered fingers (printed with permission of Richard Weir).

joint position. Without the need for user produced force these systems may be better suited for individuals with sensitive residual limbs as may result from damage caused by a traumatic injury. Also grip is maintained without prolonged control input. The electric device is not back drivable and the prehension force is generally greater than that obtained with wrist driven devices.

Electric prostheses for partial hand prostheses also have disadvantages compared to body-powered designs. There is no direct feedback from the control system regarding finger position, speed of movement or force generated. The space needed for hardware to control and power the fingers is difficult to house within the normal confines of the natural hand space making cosmetic finishing difficult. It should be pointed out that in cases of more proximal amputations such as transcarpal, wrist disarticulation and long transradial, powered finger systems can allow for improved cosmesis compared to other hand designs by affording room in the body of the hand for the control system and power supply (Fig. 6). With the commercial introduction of ProDigits by Touch Bionics Inc. in 2007, exciting possibilities for fitting externally powered finger prostheses became feasible.

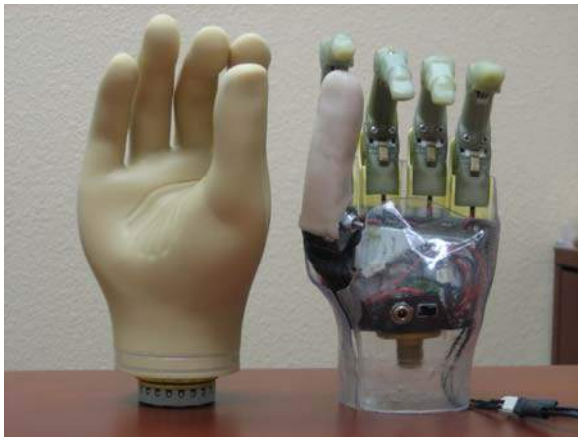


Fig. (6) – Prototype hand designed by the lead author using four ProDigit fingers and an M-Thumb suitable for amputations at and proximal to the transcarpal level. The controller, 1300 mAh battery, charge port, and on/off switch are housed within the hand body. An Otto Bock system electric hand with quick disconnect is shown for size reference.

DESCRIPTION OF PRODIGITS

ProDigits grew out of the Edinburgh Modular Arm System (EMAS) project headed by Gow at the Rehabilitation Engineering services, Princess Margaret Rose Orthopedic Hospital, Edinburgh, Scotland. The EMAS project began in 1988 and led to the development of an electrically powered full arm prosthesis fitted to a shoulder level amputee in 1998 [8]. The first clinical trials of ProDigits for partial hand prostheses was reported by Ronald in 2001 [13] (Fig. 7).

The present ProDigits system consists of fingers, a control unit, power supply and signal input sensors. Each finger consists of a motor and drive train which articulates the MCP joint (Fig. 8). A cable is routed so that MCP flexion causes PIP flexion on the index, middle, ring, and pinky fingers, the thumb IP does not articulate. There is no articulation at the DIP. There are two motor sizes allowing for four different

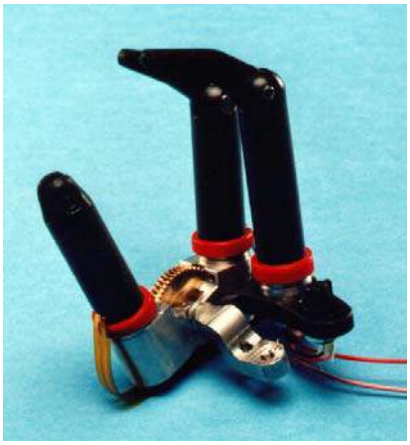


Fig. (7) – Example of the first ProDigit system fitted clinically (copyright NHSLothian).

proximal finger lengths. Additionally there are interchangeable finger tip lengths. The present design requires finger absence 34 mm proximal to the MCP joint in order to achieve normal finger length. Since each finger is self-contained a variety of partial hand amputations can be accommodated.

The microprocessor control unit accepts two control inputs and has five motor drivers. The control unit is configurable using a wireless Bluetooth interface. Control strategies employing one or two inputs can be selected from drop-down menus according to the particular needs and preferences of the amputee. Signal gains and thresholds can be adjusted in the software.

Power supply is provided by a 7.4 volt Lithium power pack. There are presently two battery sizes, an 800 mAh and a 1300 mAh. Selection of the battery depends on the space available to house the battery, the number of fingers being driven, and the anticipated daily duration of use. Ideally a battery should provide enough power to operate the prosthesis for one full day. Requisite with the battery is a charge port and on/off switch. These two components require space and must be accessible to the user. The present charge port is much larger than would be ideal, however, at this time no suitable replacement is offered.

There are currently two types of control inputs recommended for use with ProDigits, myoelectrodes and force sensitive resistors (FSR's). Both types of inputs can be used for single site or dual site control strategies.

DESCRIPTION OF VINCENT FINGERS

The Vincent© Systems GmbH - Medical Technics Group from Germany offers the latest powered finger system to become commercially available. The Vincent System grew out of the work by Schulz who has been involved in development of prosthetic and robotic hands at the Karlsruhle Institute of Technology for over a decade [14]. The Vincent powered finger system consists of individually powered fingers with articulation at the MCP joint and the PIP joint (Fig. 9). The MCP articulation is driven by the motor and the PIP joint is linked to MCP motion with flexible metal struts.

This design allows force to be exerted by the distal finger segment when the finger is moving in extension as well as flexion, a feature not available in designs that employ a unidirectional link to drive the PIP joint. Two motor sizes and several finger tip lengths are offered in order to match the finger lengths of a variety of hand sizes. In contrast to the mostly plastic construction of the ProDigits, the Vincent fingers are made of an aluminum alloy (AlmgCu) and are fitted with a soft and compliant finger tip. The size of the Vincent finger is significantly smaller than the ProDigits, especially the build height proximal to the MCP joint which is 22mm shorter when fitted with spring loaded electrical contacts and 25mm shorter when fitted with flying leads for motor connection. This short build height allows fitting of many more persons with partial hand absence while maintaining natural hand proportions [15]. The fingers are also much smaller in the anterior/posterior dimension providing a more slender hand (Fig. 10).

Control of the Vincent fingers is provided by a microprocessor that accepts one or two control inputs. Two controller designs are available, a four finger controller or a 5 finger controller. The five finger controller offers the option of vibration feedback to the user regarding grip force. Signal gains and thresholds can be adjusted with the computer interface.

Power supply is provided by a 7.4 volt Lithium power pack. The most notable feature of the power supply system is the charge port/on-off switch module which is attached to the charger with a magnetic link. This provides a charge system that is low profile allowing a better appearance to the finished prosthesis compared with the charge port/on-off module supplied with the ProDigit system.

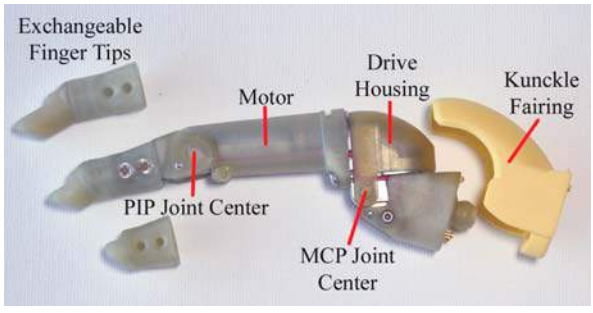


Fig. (8) – ProDigit component parts.

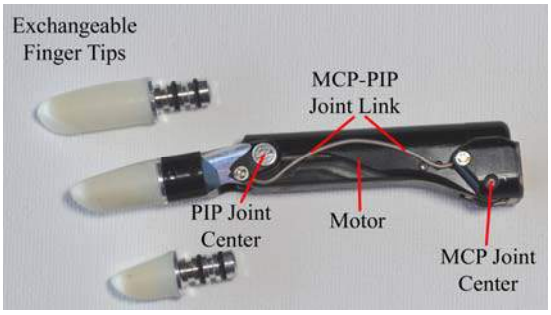
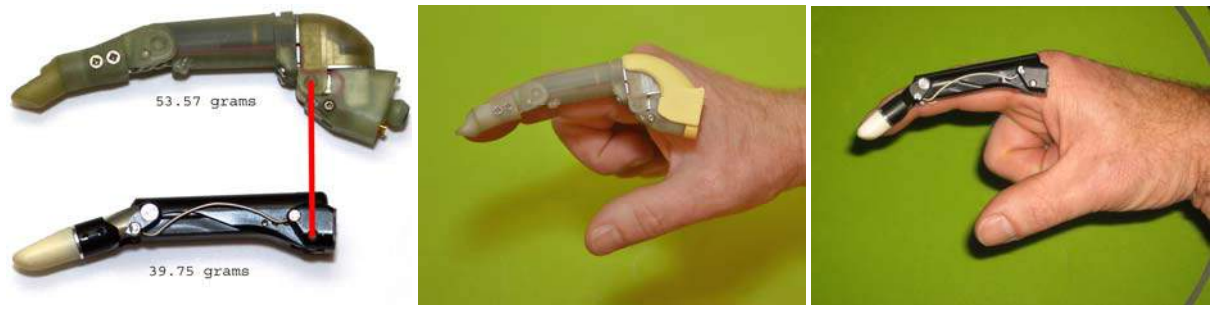


Fig. (9) – Vincent Systems finger component parts.



(a) (b) (c)

Fig. (10) – (a) Vincent Finger (black finger) shown with a ProDigit. Both fingers are shown with the larger motor that is available from both manufacturers. As shown the ProDigit weighs 53.57 grams and the Vincent Finger weighs 39.75 grams. MCP joint centers are indicated by the red line; (b) ProDigit shown superimposed on adult male hand; (c) Vincent Finger shown superimposed on same adult male hand. Both fingers are the same length from MCP joint center to finger tip.

Finger tests comparing the speed and torque of the ProDigits and the Vincent System fingers reveals comparable performance. Testing was performed by an independent laboratory under the direction of Richard Weir. Both systems were purchased in the fall of 2010 and were tested as delivered by the manufacturer with the supplied control unit, power supply and FSR's using the large motor fingers. The maximum speed for the ProDigits was 1.48 radians/second and 2.04 radians/second for the Vincent System fingers. Torque at the MCP joint measured 0.60Nm (0.44 ft-lbs) for the ProDigit and 0.49Nm (0.36 ft-lbs) for the Vincent System finger. The Vincent System controller offered a "Power Boost" mode that increases grip force by rapidly pulsing the finger motor. With "Power Boost" enabled the torque at the MCP was 0.58 ft-lbs. Although TouchBioinics offers a similar "Pulse" feature in the full iLimb hand, it was not an option in the controller supplied with the ProDigits.

CONTROL

Typically control inputs for powered fingers are provided by either myoelectrodes or force sensitive resistors (FSR). Other inputs are possible such as linear transducers and switches. Schulz has also suggested use of strain gauges for control of powered fingers, for example a strain gauge might be positioned at the MCP joint of an intact finger to provide control of one or more missing fingers on that hand [12]. Control schemes that have been used clinically include single site with fast/slow selection of open/close or with alternate direction control where one signal opens and the next signal closes and so on. Present controllers for powered fingers also can be set for voluntary open/automatic close or voluntary close/automatic open when using one input. Dual site proportional control is also possible and is generally the preferred method of control. In the primary prehension mode all fingers are driven simultaneously, each finger continues to move until an object blocks its motion or it reaches its mechanical stop thus providing a conformable grasp. A second prehension mode can be selected using a variety of mode selection commands such as maintaining a hold-open signal after the fingers have reached their extension limits, cocontraction, or rapid repeat contractions in the same direction. The typical second prehension pattern is precision pinch where the index finger and thumb become the only active fingers with the other fingers driving to a fully flexed position or remaining extended depending on the program selected or user input. If a powered thumb is not utilized then only the index finger would be active in this mode. This second mode is useful for fine grasp using tip prehension between the index finger and thumb or to allow the index finger to be extended for activities such as keyboarding or pressing buttons on a phone.

When myoelectric control is used the controlling muscles can either be in the hand or forearm. Using the muscles within the hand has the advantage of providing finger control independent of wrist motion and avoiding the need to cross the wrist joint with electrode wires. The thenar and/or hypothenar muscles have been used for control. When some fingers remain it is advisable to use muscles that do not control the remaining fingers, i.e. if the thumb remains the thenar muscles should not be used to control the prosthetic fingers. Use of intrinsic muscles for myoelectric control also has disadvantages. The size of the electrodes can make the hand bulky and it is difficult to house the electrode package within the hand space. Often when using intrinsic muscles only one usable muscle site is possible and experience and user feedback has suggested that two site control is preferable to single site control. Since the majority of muscles controlling the physiological hand are in the forearm it is natural to use these muscles for control of prosthetic fingers. Experience has shown that the users can isolate myoelectric activity sufficiently to control the prosthetic fingers without producing significant wrist motion when desired.

New technologies for prosthesis control are on the horizon. These include pattern recognition [16, 17, 18] as well as implantable myoelectric sensors (IMES)

[19, 20]. With IMES it may be possible to control a greater number of finger motions directly or through pattern recognition. Already there have been a limited number of clinical trials involving control of ProDigits using more than two control inputs. One such example was fitted to an individual who retained slight motion at the MCP joints of his index, middle, ring, and pinky fingers. Each of these mobile remnants controlled a single FSR with a dedicated microprocessor for each finger. This configuration allowed direct control of each finger independent of the other fingers. Another approach for control of multiple fingers is to assign one of the two control inputs to specific fingers. One way to do this is to partially overlap two FSR's so that a mobile limb segment can press in one of three locations i.e. only on one or the other FSR or on both together. Using this control method it is possible to have direct access to a variety of finger motion combinations. Other control methods that can produce interesting results are the use of fast/slow contractions, co-contractions, timed mode changes, and use of component position for mode changes. Using fast/slow (rate) control with two electrodes can produce four movement patterns from two electrodes. Co-contraction or one of the other mode change commands adds another dimension of control using the same two electrodes. Mode changes initiated by the component state have been used for many years such as the elbow hold steady used on the Utah arm (http://utaharm.com) where once the elbow is held steady the elbow locks and power is transferred from elbow to hand or the RSL Steeper mechanical elbow where cycling of the elbow lock also turns the electric hand on and off. A more recent example is the RSL Steeper BeBionic hand where the passively positioned thumb selects different grasp patterns depending on whether the thumb is in an opposed position or non-opposed.

PROSTHESIS DESIGN

Sockets

The role of the prosthetic socket is to provide a stable connection of the prosthetic fingers to the residual limb, to securely suspend the prosthesis and serve as a place to mount the prosthetic components such as battery and myoelectrodes. Traditional upper limb prosthesis construction utilizes plastics for both the interface material and structural components. With the introduction of high consistency rubber (HCR) silicones it is possible to design and fabricate silicone interfaces in new ways that take advantage of the unique attributes of this material and means of manufacture [15, 21].

Material thickness, stiffness, and elasticity can be selectively controlled. It is possible to incorporate hardware such as electrode mounts, screw attachments, zippers, battery compartments and wrist mounts into the silicone during fabrication (Fig. 11).

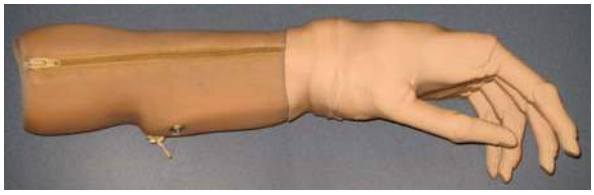


Fig. (11) – Finished ProDigit prosthesis with HCR silicone socket. Electrodes and wires are housed within the silicone construction. A battery compartment is accessible through a zipper.

Custom silicone sockets have been reported to provide better comfort as well as the ability to protect fragile skin from breakdowns when compared to other materials used for prosthetic sockets [22]. Therefore, silicone has proven to be the socket material of choice in the authors' experience [22, 23]. Since most partial hand amputations present with a hand remnant that is larger in circumference than the wrist it is useful to incorporate a zipper to allow entry of the remnant hand and when zipped provide excellent suspension. Also the socket should not restrict useful limb motions such as those that may be available at the wrist, thumb, or other remaining joints. In cases where the distal limb is only slightly larger than the wrist the elasticity of the silicone material may expand sufficiently to allow entry of the limb without a zipper. The material can be made thin and flexible in select areas to allow joint mobility. HCR silicones are manufactured using a high pressure two-roll mill. These materials are clay-like in their forming state. The stiffness of these materials using the Shore A durometer scale is generally between 15 and 70. These different durometer materials can be precisely positioned to control regions of stiffness and elasticity as desired. The material can be blended to modify the durometer and other physical properties, has a high tear strength and produces no volatile by-products.

A critical attribute of the physiological hand is the exquisite sensation it provides. When possible it is advantageous to leave normally sensate areas of the hand exposed such as in cases where the thumb remains and the majority of the thenar eminence is left exposed [11, 22]. However, in some cases when dealing with an injured hand hypersensitivity may be experienced in which case the amputee may benefit from the protection provided by covering these sensitive areas with the silicone socket. Ultimately a balance must be achieved between exposing sensate skin, protecting hypersensitive areas, providing secure fixation of the prosthesis, and providing sufficient area to house and connect the required hardware.

Prosthesis design related to thumb type

In light of the vast variety of possible hand amputation configurations, one key feature that will categorize the type of prosthesis is the condition of the thumb (Fig. 12). Since the thumb is the most important finger representing 40% of hand function [24] the level of

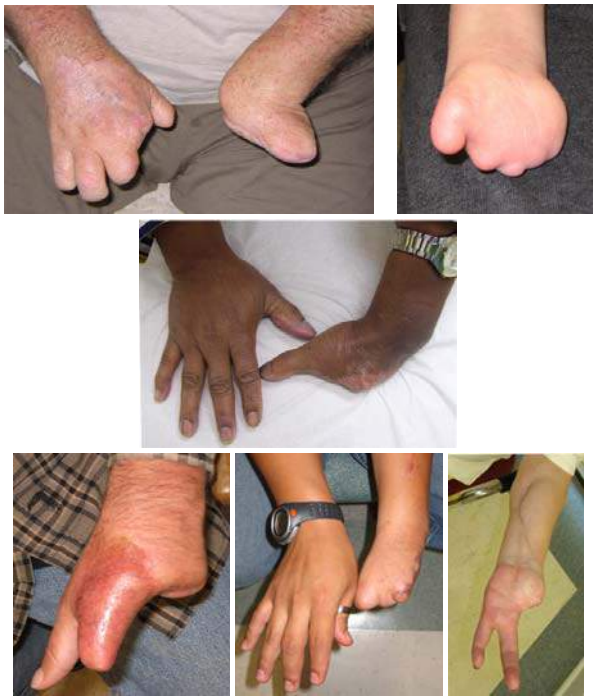


Fig. (12) – Examples of partial hand amputations that are candidates for fitting with powered fingers.

thumb function retained deserves further consideration. In order to provide good function the thumb should afford sensibility, stability, opposition and length [7]. When the thumb is partially amputated it may be treated surgically with lengthening, web space deepening, toe transfer, pollicization or osseointegration [25, 26]. In many cases where the thumb is partially amputated other fingers are also damaged or missing. Communication and planning between the surgeon and prosthetist is important for optimization of hand function. Through early interaction using a team approach outcomes can be optimized.

Surgeons should consider not only the cosmetic and functional results but also how a particular surgical intervention may affect the use of a prosthesis [11]. For example, in cases where all fingers have been amputated and the thumb is missing at the MCP joint it may be advisable to provide an osseointegrated thumb prosthesis in combination with powered fingers for the index, middle, ring, and pinky fingers. This would allow the prosthetic thumb to be positioned in opposition or for lateral prehension using intact mobility of the metacarpal. The powered fingers would provide force and a conformable grasp. The osseointegrated thumb provides direct proprioceptive feedback to the user regarding the force exerted on objects grasped.

In cases where the thumb is missing, and surgical options are not pursued, the options include provision of a powered thumb or a passively positioned thumb. When there is enough space a powered thumb can be provided. The powered thumb allows for a wider range of object sizes that can be grasped without manually repositioning the thumb compared to the passive thumb options. One disadvantage noted when using the powered thumb is difficulty targeting small objects that would be grasped using tip prehension. When both sides of the grasping unit are moving the users need to estimate the point in space that the fingers will meet. Users of a fixed thumb hand will place the object to be grasped against the thumb and allow the index finger to come against the object using the thumb as a fixed point of reference. Fixing the position of the thumb in space is a natural mechanism of grasp as reported by Wing who observed a pattern of natural hand usage in which the index finger rather than the thumb was responsible for reduction of grasp aperture as the hand approached an object [27]. This was observed in able bodied individuals as well as individuals using a prosthetic hand where compensatory proximal motions were used in order to keep the moving thumb in a relatively fixed location [27]. Compared to the full iLimb hand the current ProDigit design does not generally allow rotation of the thumb to a lateral prehension position. Due to the volume of

(a) (b) (c)

Fig. 13 – (a) the APRL thumb in the narrow opening positions; (b) the APRL thumb in the wide open position; (c) the M-Finger thumb rotated for precision pinch prehension.

the ProDigit thumb proximal to the articulation many individuals who are missing their thumb do not have sufficient space for mounting of the powered option.
The Vincent System thumb attachment point is much closer to the MCP joint articulation allowing greater opportunity to fit a powered thumb within the normal volume envelop of the natural hand. In cases where the powered thumb is not possible either a passively positioned friction thumb or a passive locking two position thumb can be provided.
The friction thumb from the M-Finger system (www.partialhandsolutions.com) offers rotation where the locking thumb from the APRL (www.hosmer.com) hand does not. The locking thumb is better suited to activities where higher loads are experienced since the friction thumb would move as a result of these high forces (Fig. 13).

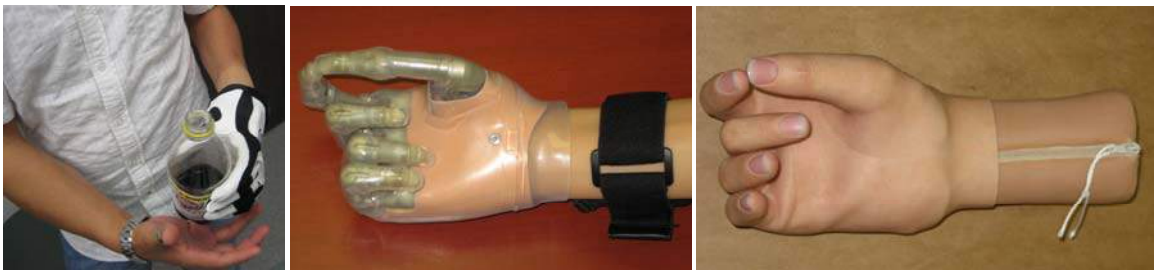
Coverings

Covering of the prosthesis is desirable for aesthetic reasons, to keep dirt out of the finger mechanisms, and to improve the functional grasp of the hand with the addition of a soft compliant covering. A variety of coverings have been utilized including very thin leather athletic gloves, textile gloves, and silicone gloves (Fig. 14). The silicone iLimb skins have been used in some cases where the finished hand was not excessively large. In cases where some fingers are remaining off-the-shelf silicone gloves have been cut to avoid covering the intact fingers. Custom silicone gloves have also been provided. Certainly finding a suitable covering is difficult and remains one of the goals of future development. The external covering is a critical component providing not only the final aesthetic appearance but also must provide increased compliance and friction for better stabilization of objects. Ideally the covering material should have high elasticity under low deformation forces, be tough, and capable of being shaped and colored to match the user's hand characteristics. Since every hand has a unique shape, provision of a custom glove would be ideal. However, manufacturing a custom glove is both time consuming and expensive. One solution for custom glove manufacturing may be found with computer aided design and manufacturing [28].

CASE EXAMPLES

Carpometacarpal Level: Powered Thumb

Case one presents the prosthetic considerations for an individual who suffered a traumatic amputation at the carpometacarpal level (Fig. 15). This 67 years old male had a traumatic amputation of his non-dominant hand as a result of a water-skiing accident. His residual hand is scarred and there is very little motion at the wrist. The initial prosthetic recommendation was to consider revision surgery to the wrist disarticulation level. This change in amputation level would allow fitting of a wider variety of terminal devices. Without physiological wrist motion the benefit of partial hand amputation versus wrist disarticulation is mostly lost. This patient refused revision surgery and therefore fitting proceeded using five powered ProDigit fingers. The finished prosthesis was provided eight months after the injury. The finished prosthesis provided several grasp patterns due to the conformable grasp provided by five independently powered fingers. A second mode of operation was programmed into the controller. This second mode was accessed myoelectrically by a sustained hold-open signal. When second mode is selected the middle, ring, and pinky fingers fully flex and the thumb and index finger become the only active fingers in a precision pinch grasp pattern. All fingers resume function with another sustained hold-open signal. The socket was made of silicone and incorporated a zipper for easy donning and doffing. Full forearm rotation was retained in the prosthesis. The battery and charge plug assembly were housed within the silicone construction on the medial forearm surface. Control was achieved with two electrodes, one over the forearm flexors and the other over the extensors. At one-year follow-up this patient reports wearing his prosthesis "a few times" a week. On days he wears the prosthesis he tends to wear it all day, about 8-10 hours. He wears the prosthesis on days when he anticipates activities that the prosthesis will facilitate. He finds the prosthesis particularly useful when he needs to hold objects while manipulating them with his sound-side hand which is as one would expect for a unilateral amputee especially when the amputated hand is the non-dominant hand. His 1300 mAh battery is sufficient to operate the five powered



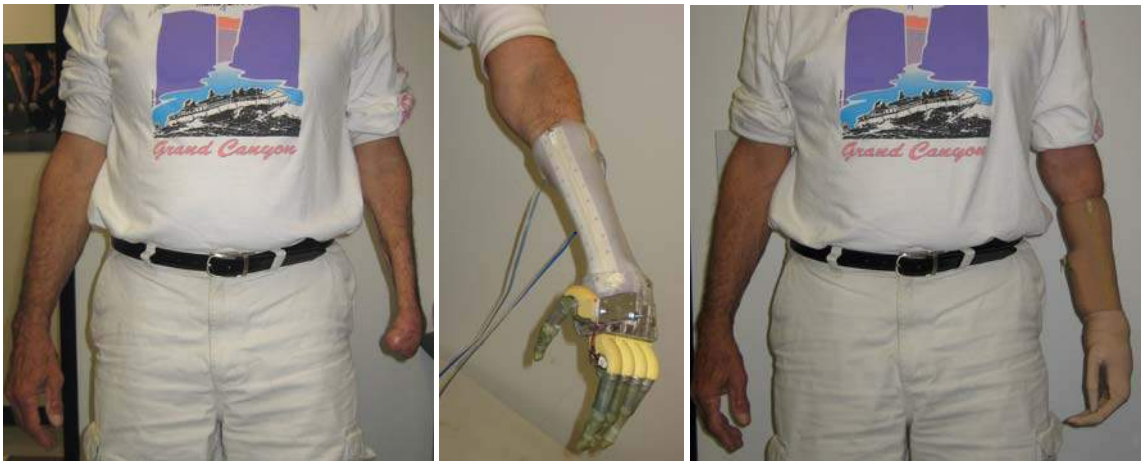
(a) (b) (c)

Fig. (14) – (a) a leather athletic glove; (b) TouchBionics iSkin silicone glove; (c) TouchBionics LivingSkin custom silicone glove.

fingers all day for his typical usage pattern. When not wearing his ProDigit prosthesis this patient wears a custom silicone socket to protect his scarred residual limb which is prone to abrasions.

Transmetacarpal Level: Friction Thumb

Case two reviews the fitting of a transmetacarpal level amputation for an individual who sustained a traumatic amputation 18 years before this episode of care. This 45 years old man presents with good wrist motion. He has previously been fitted with body-powered hook prosthesis as well as a myoelectric transcarpal hand. These previous prostheses were rejected due to the excessive length and bulk of the finished prostheses. Since too much of the hand remained to fit a powered thumb within the normal geometry of the hand a friction thumb and four powered fingers were provided (Fig. 16). The friction thumb acts as a static back stop for the powered fingers during grasping. Since the thumb does not move, small objects are easier to grasp then generally is the case for a moving thumb design. This user demonstrated good ability to grasp small objects by placing the thumb against the object and then closing the powered fingers to achieve grasp. The socket design used was the same as described in case one. Due to the flexibility of the silicone wrist motion is hampered only slightly. The zippered silicone socket provides excellent fixation on the residual limb. Initially, during the test socket phase of fitting, intrinsic hand muscles were evaluated for control using the hypothenar muscle signal to alternate direction of the fingers. Also during the fitting two site control using forearm musculature was evaluated. This patient felt that use of the forearm muscles was preferable due to the ease of control and the direct access to the open or close functions. At eight months follow-up this patient reports wearing the prosthesis all day (12 hours) almost every day. He uses the prosthesis to carry his computer bag, hold his cell phone, and other daily activities such as cooking, cleaning and home maintenance. With the 1300 mAh battery the prosthesis operates for a full day with few exceptions.



(a) (b) (c)

Fig. (15) – (a) appearance of residual limb; (b) prosthesis shown during trial fitting; (c) finished prosthesis.

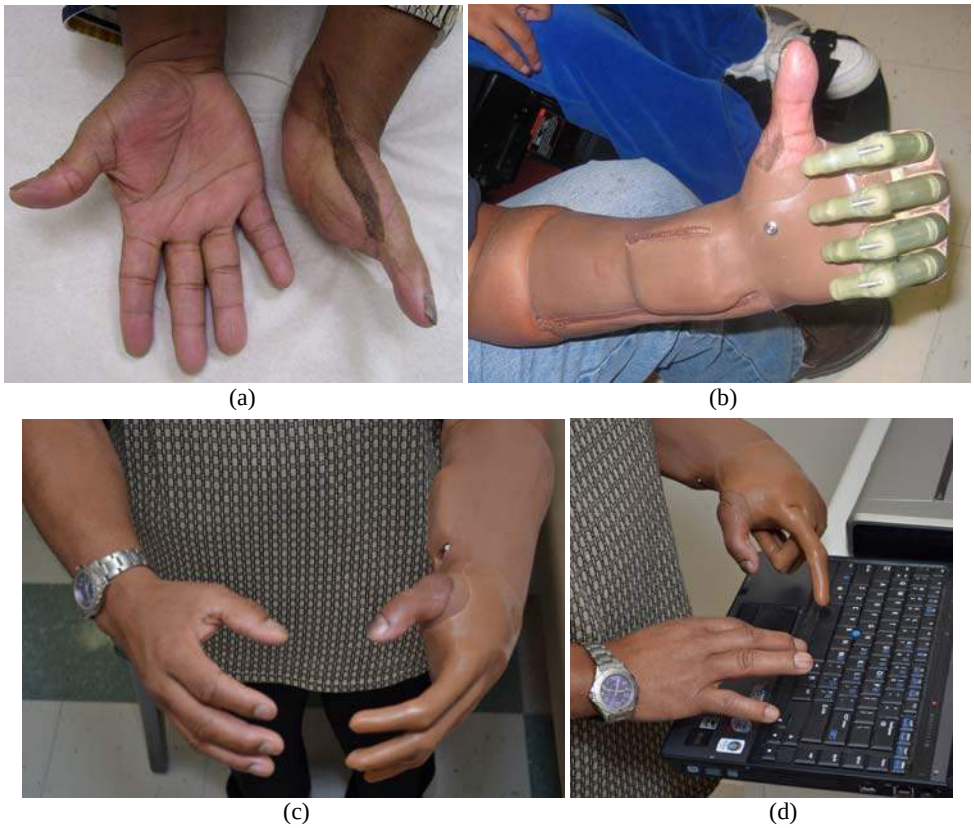
(a) (b) (c)

Fig. (16) – (a) residual limb; (b) prosthesis during test fitting; (c) final prosthesis (photo courtesy of TouchBionics)

Thumb Remaining

Case three is a 50 years old man who presents with a diagonal amputation of his index, middle, ring and pinky fingers starting at the transmetacarpal level of the index finger and approaching the carpal level of the pinky. The thumb is intact (Fig. 17). The amputated hand was his non-dominant hand. The amputation resulted from trauma sustained by a hydraulic press. Since the thumb provides normal motion and sensation, it can be expected that function with four powered fingers will be superior to the first two case examples. The intact thumb provides for a variety of grasp patterns where the thumb is in opposition as well as for lateral prehension. The thumb provides excellent proprioceptive feedback regarding objects grasped and can modulate grip force. Of primary concern for ProDigit fitting is the length of the fingers. The ProDigits require approximately 34 mm proximal to the MCP articulation to achieve physiologically natural length. In order to provide a prosthesis that is as close as possible to the normal length the shortest fingers were provided. The disadvantage of using the short fingers is that they provide less force than the longer fingers due to the smaller motors. In this case that trade-off was acceptable because the thumb was capable of producing enough force as well as detecting any slippage of objects held. The silicone socket was designed similar to the previous cases with attention to maximizing thumb range of motion as well as exposure of normal skin of the thenar eminence for tactile sensation. Since the thumb remained the thenar muscles were not considered for control due to their involvement in thumb function. Single site control using hypothenar muscles was evaluated as was two site control using forearm muscles. Two site control using forearm muscles was preferred. At an 18 month follow-up this patient reports full-time use of his prosthesis. He is employed full-time and performs shipping and receiving functions that require him to lift and manipulate packages using both hands.

After two years of daily use the ProDigit prosthesis was replaced due to normal wear and tear accelerated by exposure to the corrosive environment encountered working at a salt mine. At this time the Vincent System fingers were provided. Two forearm located myoelectrodes again provide control of the four powered fingers. Due to the much shorter build height of the Vincent System fingers compared to the ProDigits, the large motor fingers were provided for all four digits while maintaining anatomical overall length of the hand. The longer fingers provided a noticeable improvement is grasp due to the greater encapsulation of objects grasped. Most noticeable to this user was the lighter weight and more natural and slender appearance of the hand.



(a) (b)

(c) (d)

Fig. (17) – (a) appearance of partial hand amputation without prosthesis; (b) ProDigit prosthesis fitted allowing restoration of a variety of prehension patterns; (c) Vincent System showing anatomically correct finger lengths; (d) Vincent System shown utilizing index point feature for keyboarding.

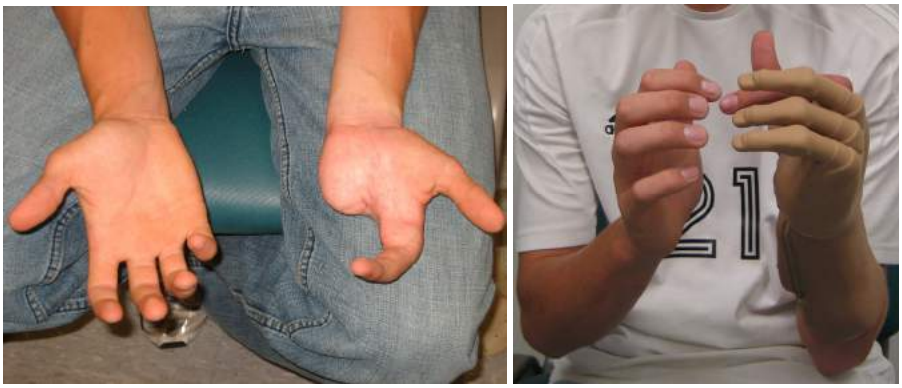


Fig. (18) – (a) appearance of hand without prosthesis; (b) three finger ProDigit prosthesis fitted.

Thumb and Index Finger Remaining

Case four demonstrates ProDigit fitting when multiple fingers remain. This 16 years old male suffered a traumatic farming accident that resulted in amputation of his middle, ring, and pinky fingers. The index finger remained but was limited in function with a 75 degree flexion contracture at the MCP joint. The thumb function was normal (Fig. 18). Without a prosthesis, grasp between the thumb and index finger was weak and instability of objects grasped was noted. Powered fingers were provided for the missing fingers. This fitting achieved greater stability of objects grasped due to the larger contact area and conformability of the powered fingers. The thumb, including the majority of the thenar eminence, was exposed from the silicone socket as was the complete index finger. Control was achieved with forearm flexors and extensors. At initial follow-up this patient reported that the prosthesis was useful for holding objects that he could not hold without it. He wore the prosthesis "most days", usually for 4-5 hours at a time. At later follow-up, one year after delivery, this patient had outgrown the prosthetic socket and refitting of the prosthesis using the existing components is being pursued.

CONCLUSIONS

Powered finger prosthesis for partial hand amputation is in its infancy. Powered finger prostheses have allowed individuals with partial hand absence to regain some of the dynamic and conformable grasp functions they lost. This experience as well as new technologies for fitting, manufacturing, and controlling powered fingers will undoubtedly lead to improvements upon these early designs in the near future. New prototype ProDigits are in the design phase to significantly reduce the size and build height of the finger proximal to the MCP joint [29]. The Vincent fingers already provide a minimal build height proximal to the MCP joint. A short build height allows many more persons to be fitted with powered fingers that achieve a physiologically appropriate length. Shorter build height also allows for provision of a rotating thumb mechanism that can be fitted in some partial hand prostheses to allow lateral positioning of the thumb. Rapid manufacturing techniques may allow customized structural components to be made in a time and cost efficient way [30]. Computer aided manufacturing may also offer a way to create molds for custom silicone coverings that match the unique shape and size of any individual powered finger partial hand prosthesis [28]. With the many hand postures possible with individually powered fingers comes the task of controlling these motions. In recent years much progress has been made in development of real-time pattern recognition that could allow direct access to various grasp patterns. In this regard IMES also offer interesting possibilities. Whatever control method is employed it should provide consistent and reliable control and with experience use should become subconscious.

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CHAPTER 3

In Human Implant of Intraneural Multielectrodes for Controlling a 5-Fingered Hand Prosthesis and Delivering Sensorial Feedback

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Abstract: Recent findings in clinical neurophysiology show that the cortical representation of an amputated hand is not so largely affected, as once thought, by critical rearrangements, and that central and peripheral neural connections somehow maintain their functions. These findings paved the way towards the exploitation of cortical and peripheral residual functions by neural interfaces for hand prosthesis control. In the present study, a young male amputee has been implanted with four intraneural multielectrodes, two in the median and two in the ulnar stump nerves. During the next four weeks, these electrodes were used with the double purpose of recording neural signals (for the extraction of subject's motor intentions to be performed by the robotic hand prosthesis) and eliciting sensory feedback through proper electrical stimulation (pulses frequency and duty cycle). Recorded neural signals were mapped in real-time onto three actions of the robotic hand through an amplitude on the best matching channel threshold method, while, thanks to an AI classifier trained offline, was achieved an 85% accuracy. Recorded peripheral nerves activity was then compared with the cortical activity over the missing hand motor area. By performing the classification of the motor intention over the peripheral signals solely during a time window compatible with the transmission delay of the motor command from the cortex, identified as an event related desynchronization of the EEG rhythm, the classification rate approached 100% of success. The study also aimed at investigating possible neurorehabilitative effects of the re-acquired stream of data to/from the missing limb and the continuative use of a high-interactive hand prosthesis. Results show that training for robotic hand control and for sensory perception produced a normalization in the electroencephalographic activation pattern and a reorganization of the motor cortical maps as evaluated *via* TMS, with restriction of the cortical overrepresentation of muscles proximal to the stump. In parallel, a clinical improvement of phantom limb pain has been observed, that recognizes in the correction of the aberrant plasticity in its anatomical substrate.

Keywords: Robotic Hand Electrodes, Multi-channel Interfaces, Proprioceptive and Tactile Feedback, Clinical Neurophysiology, Brain Plasticity, Phantom Limb Pain.

INTRODUCTION

Since thousands of years, Humans suffering for post-traumatic limb amputation try to compensate cosmetic and functional deficiencies with the use of prostheses. Cosmetic prostheses were already present in Egyptians portraits, whereas devices specifically conceived for functional restoration appear in the early 1500's [2]. Despite the huge innovations in science and

Vincenzo Parenti Castelli and Marco Troncossi (Eds)

technology which characterized the last five centuries, there are still several sensible limitations which have to be overcome before high-tech prosthetic devices will be really able to substitute the amputated limb. The most advanced hand prostheses commercially available are little more than pincers with one degree of freedom, with no or a few embedded sensors which do not provide any proprioceptive feedback, hence requesting the user to continuously visually monitor the prosthesis operations. This can explain why the vast majority of amputees still use cosmetic prostheses only, if any at all. It is quite obvious that the performances theoretically required to an upper-limb/hand/fingers prosthesis are remarkably more complex than those expected from lower limbs prostheses. This problem has not been solved by upper limb transplantation, which does not seem –so far– to represent an acceptable and viable solution, at least for large scale use [3]. The overall user acceptability of a prosthetic device depends from multiple factors, such as dexterity, anthropomorphism, controllability, autonomy and dependability. The current advanced research in robotics is trying to answer to these challenges by studying novel sensing and actuation miniature devices, possibly based on MEMS technologies for allowing the monolithic integration of motion control and sensory data acquisition and processing capabilities. Also advanced kinematic solutions are under investigation to allow natural movements with under-actuated mechanisms [4], in order to keep the mechanical complexity –and the weight– within acceptable limits. Such efforts are enabling the first developments of compact, light-weighted, multi-fingered hand prostheses which embed low-level controllers for active grasping control and could be prone to perform dexterous manipulative tasks, if properly interfaced to the user [5].

The bottleneck of current research on novel robotic prostheses has been moved to the interfacing level. Prosthesis motor performances and acceptability are now influenced mainly by the quality and quantity of sensory information fed-back to the user [5, 6]. Under this regard, information *transfer rate* and *latency* -i.e. the time delay between command and action enabled by the interface- play a major role [7].

Ideally, a Brain-Machine Interface (BMI) should be able to implement a closed-loop control of the mechatronic prosthesis while exchanging bidirectional (efferent and afferent) information with the nervous system of the amputee. In fact, one of the most intriguing and disruptive innovation that might have a dramatic impact on the application of such a new generation of hand prostheses is the enhancement of the exteroceptive and proprioceptive channels that the device is able to establish with the patient in a physiological fashion, in order to partially recover natural sensations and re-obtain full consciousness of the missing limb, by embedding it again into the personal body scheme. From a design perspective, the development of highly performing hardware (in terms of mechanics, sensing capabilities and actuation means) should be intimately interwoven with the development of suitable control techniques, and both such aspects should build upon a proper natural and intuitive bidirectional interface with the user, that, at present, still represents the main actual functional and technological limitation of the whole system.

Persistence of Neural Pathway

Amputation of a limb involves the complete and sudden truncation of all the afferent and efferent nerves that innervate the lost part, triggering anterograde and retrograde changes to and from the stump, which affect both peripheral and central nervous systems. Cortical reorganization that follows limb amputation is accompanied by an invasion from the adjacent territories of the cortical areas devoted to the representation of the lost limb [8-10], but several recent evidences show how the deafferented hand-controlling cortex still remains responsive to stimuli which "seem" to be coming from the missing hand [11-13]. Since the amputation does not completely erase relays and connections, these areas can be exploited as potential targets for user-prosthesis interface by implanting a bidirectional intraneural electrode in the stump nerves for controlling the artificial robotic device [14].

Amputation of a limb is, for nearly all amputees, followed by the sensation that the lost body part is still present and kinesthetically perceived. These phenomena are grouped in the frame of the so called *phantom awareness* and *phantom sensation* respectively [15]. In 50-80% of amputees a painful dysesthesic perception in the lost limb named *phantom limb pain* (PLP) or *syndrome* (PLS), is observed, which is an additional cause of disability [16]. Cortical reorganization in both motor and somatosensory systems, assessed with several methods, primarily manifests in patient with PLP, and probably represents its neuro-anatomical substrate [17-21].

Interfacing with Peripheral Nervous System

New approaches in developing artificial architectures, that compensate for the loss of motor and sensitive functions, involve the capability of directly interfacing to the nervous system at a peripheral level.

This can be done *via* implanted microelectrodes capable of exchanging information with the nerves. Peripheral nerve interfaces aim at detecting electrical activity of the nerve fibres and/or to excite them as selectively as possible [22-25].

From a neuroprosthetic standpoint, invasive peripheral neural interfaces offer a very good compromise

between the extremely high invasiveness of cortical implants and the low selectivity of electroencephalographic (EEG) or electromyographic (EMG) interfaces, which are also limited by the unnatural control strategies and lack of proper sensory feedback.

Neural interfaces mostly use an electrical coupling method both for detecting the bioelectrical activity of the nerve fibers (recording) and/or to induce their excitation (stimulating). Thus, the neural electrodes have to be implanted in the proximity of a peripheral nerve, even within it, to reduce tissue impedance and the current intensity needed to trigger stimulation. Unfortunately, the selectivity for stimulation or recording from individual nerve fibers increases with the invasiveness of the electrode. Implanting an electrode inside the nerve, instead of placing it around the nerve, increases the selectivity of the contact and the signal to noise ratio [26, 27].

Given the same fibre-stimulating contact distance, large myelinated fibres are picked up more effectively than small myelinated and unmyelinated fibres. Therefore, tactile or position sensations can be selectively and focally elicited without concomitant pain, while motor signals to extrafusal fibres are recorded much more easily than those to intrafusal and vegetative fibres. If the number and spatial density of recording sites within the nerve are higher, then the probability of recording signals from the fascicles, originally innervating the missing limb and conveying the information relevant to the desired movement, is significantly higher [22].

Recently, thanks to peripheral intrafascicular electrodes implanted in the stump nerves of amputees, two different research groups were able to record volitional motor nerve activity that has been used by the patient to control the grip of a hand prosthesis [28] and the flexion-extension of an artificial finger [29]. Horch and colleagues were the first to stimulate nerve fascicles using thin insulated conducting wires LIFEs, implanted into median and ulnar nerves. Half of eight amputees discriminated tactile and proprioceptive unimodal sensations with stable topography [30] which improved with training [31]. Three subjects were able to rate the level of force from a strain gauge placed on a robotic hand thumb and to match with the intact arm the elbow angle of the robotic arm. The other three volunteers succeeded in matching with the robotic arm different levels of force and elbow angles through one motor channel [28].

OBJECTIVES

The goal of this study was to evaluate the reliability and biocompatibility of a novel peripheral intraneural multielectrode in recording neural signals from motor fibres of a human subject. The study lasted four weeks. Recorded signals were classified into rest and three voluntary emitted commands dispatched to the missing hand/fingers, while sensory feed-back was electrically delivered as a surrogate of action-driven perception. Byproduct of the study is the development of a neural classifier that allows to improve the amount of correct classified intentions of the user. In parallel, this work evaluates the redirection of previous aberrant cortical neuroplasticity and the concomitant amelioration in phantom limb symptoms due to the reacquired bidirectional stream of data.

METHODS

Four intrafascicular multielectrodes have been implanted in the peripheral nerves of the upper limb in a human volunteer.

Subject

A 26 year old right-handed male with left arm trans-radial amputation due to a car accident happened two years before the experiment (Fig 1), who had previously tried, without satisfaction, aesthetic and myoelectric prostheses, was selected.

Previous medical history was unremarkable. Full neurological and neurographic/electromyographic exams were normal. Neuropsychological and neuro-

Fig. (1) – Different views of the transradial amputation of left upper limb of the subject who took part in the experiment after the car accident. (From [1])

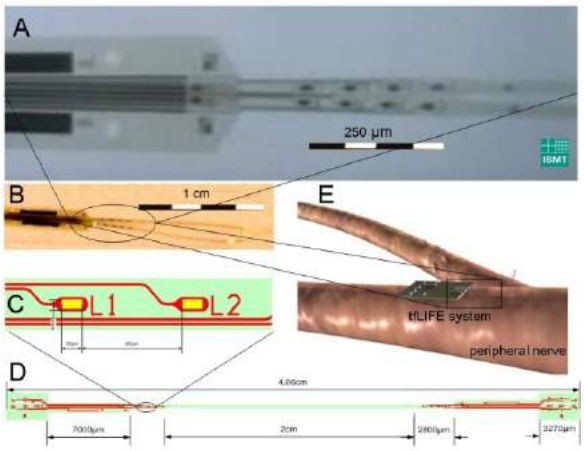

Fig. (2) – A and B: optical microscope views of tf-LIFE4 at different magnifications; C and D: electrodes footprint with dimensions. Note that only the encircled part remains within the nerve and contains the recording contacts (i.e., yellow rectangular L1 and L2 of C), as schematised in E. (From [1])

psychiatric tests (MMPI-2, WAIS) demonstrated normal comprehension and intellectual capacity, and excluded personality disorders.
The study was approved by the local Ethics Committee and by the assigned office of the Italian Ministry of Health. An informed consent was signed by the patient in the presence of a witness from his family.

Electrodes

Longitudinal intrafascicular electrodes (LIFEs) are inserted longitudinally into the nerve to lay in-between and parallel to the nerve fibers [26]. Thin-film LIFE4s (Fig 2) represent the last generation of such electrodes and present several contacts deposited over a polyimide substrate. Their contacts can be used to perform invasive multi-unit peripheral nerve recording and stimulation, thus allowing the combination of neural signals from multiple sites to better reconstruct the patterns of input or output information. Each single electrode is fabricated from a thin insulated conducting Platinum (Pt) stripe embedded in a polyimide, with eight recording pads left exposed. Both thin Pt layer and polyimide assure biocompatibility and flexibility [32, 33]. tfLIFEs allow more selective interfacing than extraneural electrodes, which wrap the whole nerve, because they acquire signals from few axons only. Electroneurographic (ENG) signals from peripheral nerves have been recorded for long periods (months) in animals using LIFEs, demonstrating their biocompatibility [34].

Surgery

After general anaesthesia, skin was incised along the medial edge of the biceps muscle for 10 cm to expose ulnar and median nerves in the distal upper arm; following epineural microdissection, two tf-LIFE4s [33], 3 cm apart, were inserted in each nerve under surgical microscope (Opmi Vario/NC33, Zeiss). A tungsten needle allowed electrode filament introduction into the nerve fascicle. tf-LIFE4s were introduced 45° obliquely to assure stability and to increase the probability of intercepting nerve fibres. The distal handle of the electrode was anchored to the epineurium by a 8.0 nylon suture. Four through holes, lateral to the incision, allowed the positioning of the tf-LIFE4 wires. Four weeks later, tf-LIFE4s were removed.

Prosthesis

A stand-alone version of the CyberHand prototype (Fig. 3), which approximates dimensions and grasping capabilities of the human hand with five underactuated fingers actuated by six motors (5 for the independent flexion/extension of the fingers, 1 for the opposition of the thumb), was employed [35].

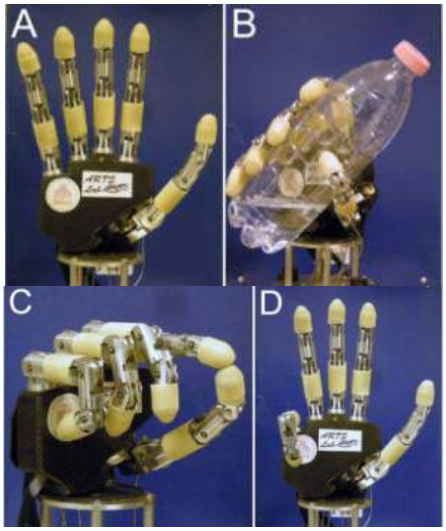

Fig. (3) – Robotic hand experimental movements (A-Rest, B-Power grip, C-Pinch grip, D-Flexion of little finger).

It was endowed with 6 position sensors and 5 tensiometers able to measure tension of the cables controlling finger flexion, similar to Golgi tendon organs. However, validation of real time artificial-sensor feed-back was not part of the present experiment; sensory stimulation and feed-back were delivered by the experimenters and not by the robotic hand sensors.

Stimulation/Recording System

The Stimulation and Recording System is devoted to the amplification, filtering, acquisition and the storage of the neural (ENG) and myoelectric (EMG) signals, as well as to the electrical modulated stimulation of the neural fibres in contact with the tf-LIFEs.
Four integrated 4-channel amplifiers (Grass QP511 Quad AC Amplifier System for Evoked Potentials, EEG, EMG) simultaneously amplified ENG signals (by a factor of 10.000, and band-pass filtered between 100-10k Hz), and EMG signals (by a factor of 5.000 and band-pass filtered between 30-3kHz).
A 16 channels, 16 bit, 1 Ms/s analogue-to-digital converter (ADC), connected to a PC, was used.
A two channels stimulator (Grass S88X Dual Output Square Pulse Stimulator) was programmed for releasing trains of rectangular biphasic cathodic current pulses. Trains duration ranged between 300msec and 500msec, releasing between 3 to 250 pulses per train (with pulse frequency ranging between 10Hz to 500Hz). Pulse parameters such as current intensity (10-100 μA) and duration (20-300 μsec) were set in accordance with tf-LIFEs intrinsic limitations on charge density.

For safety reasons, the whole system was powered by an isolation transformer for medical applications.

Motor Output Recording

The protocol included the following phases: 1) Pre-implant training with a virtual hand for standardizing movement imagination; 2) Post-implant training to control output of tf-LIFE4s during motor imagination and stimulation for hand/fingers sensory feed-back ; 3) On-line prosthesis control designed to train the subject to control and standardise tf-LIFE4 output induced by movement imagination; 4) Off-line development of a classifier-algorithm for optimal prosthesis control.

During phase 1), the patient practiced by imagining three single movements performed with the missing hand as shown in dedicated videos: (i) fist grip; (ii) pinch grip; (iii) flexion of the little finger (Fig. 3). These three actions were considered representative of the variety of movements controlled by the nerves under investigation: mostly median fibres for the pinch, ulnar for little finger flexion, and both for the power grip.

Following tf-LIFE4s implant, phase 2) began, in which the same videos were used to trigger user motor intents while recording his neural signals. Videos showed alternating open-relaxed hand movements and were synchronised with the recording system. Signals from tf-LIFE4s, biceps and triceps EMG electrodes were simultaneously recorded using a 48 kHz sampling rate, and were data-windowed in 1000 samples for mean rectified value calculation.

In phase 3) ENG channels with the best signal-to-noise ratio were selected while analyzing the recordings from the previous phase. Their online activities, together with EMG activity, were shown to the subject, who was asked to modulate them while keeping the EMG silent.

Once a stable level of training was achieved, LIFE signals were translated into robotic hand actions and the subject had direct visual feed-back on the correct execution of the intended movement. Each movement type was triggered by the signal level of a proper single channel. In order to exclude activities caused by unwanted muscle contraction or environmental noise, only rectified values greater than 3÷8 μV in a time window ranging from 5 to 20 ms were used. Channels were chosen depending from their signal-to-noise ratio and anatomo-functional location (i.e., channels from the median nerve for power or pinch grip, channels from the ulnar for little finger flexion).

For phase 4), off-line examination of the original ENG signals and their processing was carried out to optimise the prosthesis control in order to avoid 'false' positive (unwanted movement) and negative classifications (no movement performed despite the intention). For efferent signals processing, selected features were extracted as input to an artificial neural network, after wavelet denoising and spike-sorting by using a template creation and matching approach [32, 36, 37]. Support vector machines were trained to use waveforms of the identified spikes to infer the type of imagined movement. The analysis was applied to a progressively higher number of active sites in order to test whether correct classification improves.

The reliability of classification was evaluated as the accuracy of the system in avoiding ''false" positives (unwanted movements) and negative classifications (no movement performed despite command dispatch).

To further improve the performance of the motor control system, only the features of the signals falling in the time window compatible with the transmission delay of the motor command from the cortex, identified as an event related desyncronization of the EEG rhythm, have been taken into consideration.

Whenever one type of action was classified, the robotic hand started and completed a movement after a time lag appropriate to a natural condition.

Sensory Stimulation

To identify afferent fibres eliciting sensations, full mapping of all 32 contacts within the tested nerves was carried out. Rectangular cathodal pulses with 10-300 μs duration and 10-100 μA current intensity were employed. To avoid electrode damage, all stimulation trials were below 75% of the maximum charge (~4 nC), in the form of 300-500ms 10-500Hz pulse trains. The best active sites for sensation were characterised, beginning with short and low-current stimuli (10 μA, 10 μs), which were progressively increased in order to elicit different sensations; either the electrode's safety limits or subjective discomfort determined maximal stimulus intensities. A psychometric staircase method was used to quantify sensation, ranging from the minimal perceived threshold (score=1) to discomfort (score=5).

Stimulation was also tested the 7th and the 8th day after implant as feedback during a control motor task, in which the patient was asked to produce a power grip every 5 seconds. In a set of trials, an operator triggered a stimulus train (0.3s train of 70Hz, 10 μ A, 10 μs pulses) after each burst of efferent activity recorded by tf-LIFE4s; success rates with or without sensory feed-back were then measured.

Neuroplasticity and PLS

Cortical reorganization has been monitored through Electroencephalography (EEG) and Transcranial Magnetic Stimulation (TMS) before the implant and repeated the day before the explantation of the electrodes.

Phantom awareness and phantom-limb syndrome (PLS) were evaluated pre-surgically and were followed up at the end of the training period and 3 months after implant removal, using an abbreviated version of the McGill Pain Questionnaire (sfMcGill), the present pain intensity scale (PPI), the pain visual analogue scale (VAS), and an open section for description of phantom awareness.

Electroencephalography

Voluntary movements are accompanied by a definite pattern of changes in oscillatory firing of cortical neurons. While the presence of an ERD (Event-Related Desynchronization or a power decrease in a frequency range timely related to a given event) has been linked to the activation of cortical areas related to preparation of movement, the ERS (Event-Related Synchronization) has been associated to inhibited or idling areas.

Topographical maps of ERD/ERS in alpha-1 (8-10 Hz), alpha-2 (11-14 Hz) and beta (15-25Hz) bands were obtained by averaging for each band the time-frequency representation in the period from 500 to 1500 ms. The topographical maps of the different sessions were compared.

EEG signals were recorded from the scalp in different sessions: i) before surgery (PRE) during voluntary command to perform left hand grasping, ii) after LIFEs implant (POST) and intensive training for motor commands control simultaneously to ENG acquisition, iii) during right hand movement. Thirty-two electrodes (scalp sites defined according to the international 10-20 system), mounted on an elastic cap and binaural reference, were used. Skin/electrode impedances were kept below 5 k Ω. Recordings were carried out using a time constant of 0.1 s. EEG data were sampled at 1024 Hz (pre-sampling analogical filter 0.48–256 Hz, BrainAmp System).

Transcranial Magnetic Stimulation

Cortical motor output was mapped via TMS (Magstim200; eight-shaped coil with an inner wing diameter of 70mm; stimulus rate 0.1-0.2c/sec; intensity 10% above standardised excitability motor threshold [38]) for each hemisphere. To check for interhemispheric differences, Motor Evoked Potentials (MEPs) were recorded from proximal muscles of both limbs (biceps and deltoid) during separate mapping of right and left hemispheres.

The patient wore an elastic cap with a 99-square grid over the sensorimotor cortex. Square 1 corresponded to the point where the minimal intensity triggered MEPs of largest amplitude and shortest latency (*hot spot*); this was coincident for biceps and deltoid on both hemispheres.

RESULTS

Motor Control

tf-LIFE4s recorded a progressive improvement of signal-to-noise ratio which stabilised within about 10 days following surgery. All the contacts in all electrodes recorded properly during the entire 4-week experimental period. Contacts belonging to one electrode stopped working on the last day; the electrode was found dislodged from the nerve during LIFEs removal. Off-line signal processing using the algorithms previously described allowed up to 85% of grip type correctly identified in the phase of mental activity immediately following the resting period. Evaluation of signals extracted from several contacts of different electrodes from both nerves improved discrimination performance in comparison to information obtained from only one contact. Moreover correct classifications improved with time (75% on day 26 to >85% on day 28), indicating a learning effect. When only the time windows compatible with the transmission delay of the motor command from the cortex, identified as an ERD of the EEG rhythm, were analyzed, this value reached almost 100%.

Optimal control can be achieved by combining information extracted from grip-related neural signals, together with fixed parameters embedded in the robotic controller and selected by a trial-and-error procedure. The level of "shared control" between the user's brain and the robotic controller can be modified according to the performance of the prosthesis [39].

Sensory Stimulation

Three of the four electrodes appropriately stimulated for 10 days. Discrete tactile sensations were elicited from different stimulating sites of three electrodes (i.e., from 4 sites of EL1 and EL 2 in the median nerve and from 5 sites of EL3 in the ulnar nerve). In all cases, sensations were referred in the fascicular projection territories of the corresponding nerves. As an example, sensations related to median nerve stimulation through L1 and L2 sites of EL1 electrode were referred by the subject both as touch and tingling in the middle of the palm and near the base of the index and middle finger, while sensations related to ulnar nerve stimulation through R1 site of EL3 electrode was referred as touch sensation on the wrist that irradiated towards 4th and 5th fingers.

In agreement with the results of previous studies [30], stimulus frequency concurs to modulate the sensation intensity on a log scale.

When stimulation was added as sensory feedback for the motor control task, the success rate increased significantly and rapidly.

The electrical charge necessary to elicit sensations (minimal threshold) increased during the first ten days

from 0.1 to 1 nC, until no sensation was elicited through any of the three electrodes despite the maximum allowed charge (~ 4 nC) was delivered. In order to avoid Pt electrochemical irreversible damage, with possible contamination of motor signal recordings, stimulation procedures were halted. Several, not mutually exclusive, explanations for this failure can be proposed: a) progressive 'habituation' of the patient, moving from an initial 'hypersensitivity' due to long-lasting sensory deprivation (subjective sensory threshold below maximum of stimulation), which then decreased and stabilised at a more physiological level (subjective sensory threshold above maximum stimulation); b) surface of the miniaturised contacts limiting the maximum applicable current charge, i.e., ~ 4 nC, well below parameters reported in [30]; c) fibrotic tissue reaction which, however, was only suspected on visual inspection during LIFE removal but was not histologically demonstrated due to ethical restrictions.

Electroencephalography

In the first session (PRE) before surgery a slight power decrease in both alpha 1 (8-10Hz) and beta bands in the 'event' period after trigger onset was observed. Interestingly, in the second session three weeks after LIFEs implant (POST) and intensive training for motor control of the missing hand/fingers, a clear pattern consisting in an abrupt and intense power decrease (ERD) over the central sensorimotor areas contralateral to the missing hand in the time preceding voluntary movements was reliably observed in alpha 2 and beta bands (Fig 4), as widely documented by previous studies in healthy people. In alpha 1 band a diffuse desynchronization in bilateral central, frontal and parietal regions was also evidenced. In both sessions, in alpha 1 band a higher ipsi- than contralateral desynchronization was found in sensorimotor regions. The time preceding the real right hand grasping was associated with an ERD maximal to the contralateral central areas in all frequency bands. In alpha-1 band, a great involvement of the bilateral parietal areas was also observed.

Transcranial Magnetic Stimulation

Stimulating left hemisphere while recording contralaterally in the distal (ADM e ECD) and proximal muscles of the healthy limb both the number of responding sites and the stimulating threshold do not significantly change before and 4 weeks after the implant (distal muscles PRE: responding sites 22, threshold 37%; POST: responding sites 20, threshold

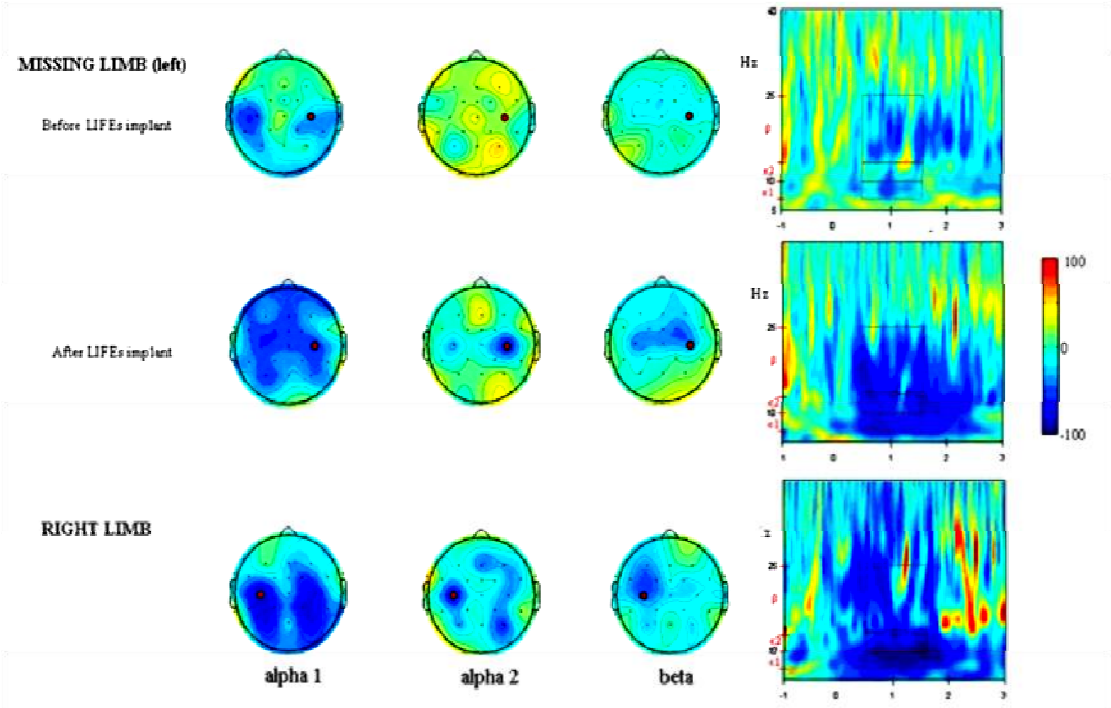

Fig. (4) – In the first three columns Scalp topography of ERD (blu)/ ERS (red) in alpha-1, alpha-2 and beta bands in the time period of 500-1500 ms after that the subject received motor trigger is shown. C3 and C4 location are evidenced by red circles. In the last column the time-frequency representations of EEG power modulation in primary motor cortex (C4 on right and C3 on left) is shown. Boxes indicate the frequency bands and time period considered for scalp topography. Three lines correspond respectively to voluntary motor command of the missing limb before and after LIFEs implant (up) and for voluntary movement of right hand (bottom). (From [1])

36%; Proximal muscles PRE: responding sites 30, threshold 52%; POST: responding sites 35, threshold 53%). On the contrary as regard as right hemisphere stimulation, recording from the proximal muscles of the severed limb, the number of responding sites significatively ($p<0.05$) decreased (PRE 36 *vs.* POST 16). Stimulating threshold did not change (Fig 5).
Pre-surgical TMS motor maps showed an abnormal interhemispheric asymmetry of motor cortex excitability in the hemisphere governing the stump. Following training, post-surgical maps, in parallel with reduction of PLS, showed a clear reduction of this asymmetry and a trend toward balanced muscle representation in the two hemispheres, as in control subjects.

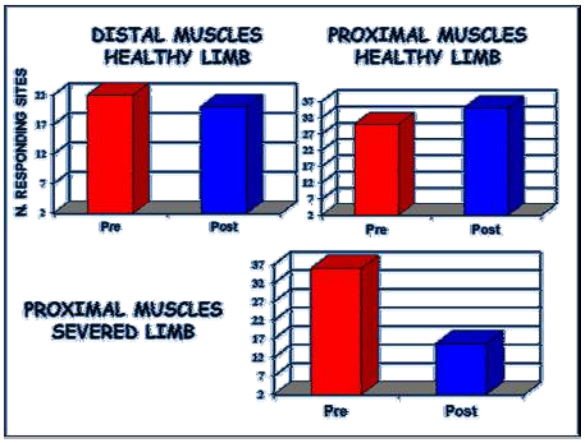

Fig. (5) – Istogramms of responding sites before (red) and four weeks after (blu) the implantation recording respectively from distal and proximal muscles of the healthy limb and from the proximal muscles of the stump. (From [1])

Phantom Limb Syndrome

Before implant, the patient referred a moderate PLS and perceived the phantom of his left upper limb as if '*…the missing hand is directly attached to the stump, without any forearm, but blocked by an heavy load or tightly fastened by a belt that makes any movement impossible'*. (Fig 6).
This weird perception underwent to a normalization when investigated the fourth week after surgery, but returned by the 3 months follow-up. A similar behaviour was evidenced by the improvement of PLP, as evidenced by all the three clinical scales adopted in the questionnaire (Table 1).

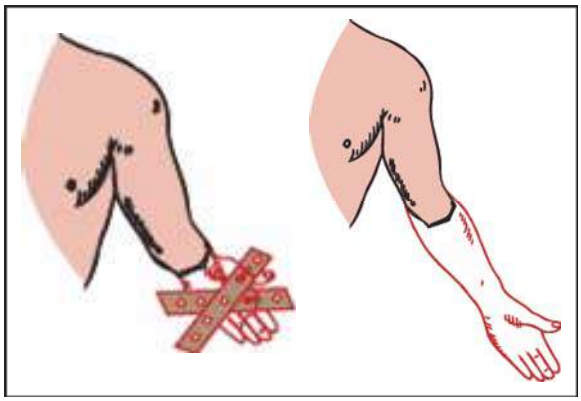

Fig. (6) – Modification in subjective awareness of phantom limb before (left) and after four weeks of training in active prosthesis control and feedback perception. (From [1])

DISCUSSION AND CONCLUSIONS

A 4-weeks duration of tf-LIFE4s implant was dictated by the European Authorities as the maximum test period for an experimental medical device still under scrutiny for human biocompatibility. The robotic hand was not directly worn by the subject: the hand, connected to tf-LIFE4s, was on a table in front of the subject. Despite such limitations, several new findings emerged from the study, as reported here. First, this generation of LIFEs can be implanted and used in humans for several weeks with a high rate of success in picking up signals with a good signal-to-noise ratio. Electrode positioning remained stable *in situ* even when carrying out everyday life activities (the patient lived at home except for 3 days for surgical procedures). Multiple electrodes in different nerves with several contacts guaranteed a reliable flow of signals. Second, simultaneous recordings from several sites of 3 electrodes from two nerves improved the rate of correct classification for movement control with

Table 1 – Modification of phantom limb syndrome during the experimental period and after 3 months. (from [40])

	Sf McGill (0→45)	**PPI (0→5)**	**VAS (0→100%)**	**Subjective description of PLS**
Pre-implant	18	3	38	No forearm, blocked hand
Post-explant	11	2	23	Regain forearm, movable hand
3 months Follow-Up	17	3	36	No forearm, blocked hand

higher sensitivity and specificity (i.e., less false positive/negative) compared to a single-contact classification, particularly when discriminating independent movements. The amount of correct classification can be further improved by combining simultaneous acquisition of signals from the sensorimotor areas dispatching the motor command (*via* EEG) and from the nerve fibres as final output to the target muscles (*via* tfLIFE). Quality and selectivity of efferent signals dispatched during movement imagination was augmented by concomitant sensory feed-back.

Tactile sensations were elicited and modulated by afferent stimulation during the first ten days following the implant.

Moreover, training for robotic hand control and for sensory perception produced a normalization in the electroencephalographic activation pattern and a reorganization of the motor cortical maps distinguished by TMS with restriction of the cortical overrepresentation of muscles proximal to the stump. In parallel, a clinical improvement of PLS, with a progressive return to perception of full-length forearm and of the hand free of motion, has been described by the subject. Such clinical improvement was no longer present at 3 months follow-up.

When making an analytic comparison between the present study and previous reports some important differences are found: 4-weeks from a maximum of 2 weeks, a significantly larger number of electrodes and contacts, the ability to control up to 3 individual movements with a high success rate; the demonstration of plastic rearrangement of motor areas with parallel changes of clinical condition. One main weakness of this study is the short-lasting ability to deliver stimuli for sensory feed-back (Tab. 2)

The adopted prosthetic device, thanks to the bidirectional interface established at the neural level, showed a high interactivity with its user, thus achieving those neurorehabilitative qualities that could address the cortical reorganization toward a curative plasticity and that can be considered as beneficial "side effects" of the use of such device.

The work presented here opens a number of research avenues centered on the concept of effectively and bidirectionally interfacing a hand prosthesis to the human brain. Although noteworthy improvements have been made over the past decade in terms of BMI and mechatronic technologies, for future direction of this study, advances are expected in a number of areas, including: i) analysis of tissue-electrode biomechanical interface (for solving issues related to electrodes mechanical stability against nerves physiological deformation and inflammatory response); ii) improvement of wearability of the prosthesis (for solving issues related to energetic autonomy; embedded control; wireless data transmission; weight; number of actuated and controllable degrees of freedom); iii) achievement of overall system biomimicry.

The latter aspect does not refer to mere cosmetics, but to the whole set of anatomo-functional dynamic aspects, that play a major role in the effective interplay between the neural system and the upper limbs. Under this regard, emerging trends in robot system design, such as embodied intelligence and neuro-inspired control, may drive the development of novel prostheses, featuring highly biomimetic cosmetic, behavioural and interaction solutions.

Table 2 – Analytic comparison between the present study and previous two reports [28, 29]

	Dhillon *et al.*	**Jia *et al.***	**Present work.**
N. of subjects	6-8	case report	case report
Implant duration	2 days (first study) 14 days (last study)	Acute study	4 weeks
Surgery	Implant		implant/explant
Pre-training	NO	2 weeks	2 weeks
Interfacing protocol	recording/stimulation	Recording	recording/stimulation
Type of electrodes	Single filament LIFEs	spring-like single filament	tfLIFE 4
Amount of electrodes	4-8 mono-electrode each subject	better 4 among 6 monoelectrodes	4 multielectrodes 8 channels each one
Implanted nerves	Median and ulnar	median, ulnar and radial	median and ulnar
Stimulating electric charge	Min 4,7nC max 12,7 nC Increase in the time of the needed charge to evoke the same sensation	-	min 0,1nC max 3 nC Increase in the time of the needed charge to evoke the same sensation

Table 2 (continued) – Analytic comparison between the present study and previous two reports [28, 29]

	Dhillon *et al.*	**Jia *et al.***	**Present work**
Stimulation parameter	P. width 250 μs (300) P. ampl 200 μA(1-200) P. freq (randomly assigned by PC!) Train duration 500 ms	-	P. width 20 μs (300) P. ampl 10 μA(1-200) P. freq 10-100 Hz Train duration 500 ms Train frequency 0,2 Hz
Stimulation duration	Whole study	-	Ten days
Motor control	Attempt to make a movement that results in maximum audible activity with loudspeaker		Artificial Intelligence Classifier applied in a time window compatible with the presence of ERD seen in EEG
Controlled movements	1 D-o-F (graded control)	Only finger extension with one electrode implanted in radial nerve.No achieved control with ulnar and median nerve electrodes	3 whole hand complex movements and rest
Feedback	Loudspeaker noise	-	Operator driven intraneural stimulation
Clinical correlated features			brain plasticity and phantom limb phenomena monitoring

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DISCLOSURE

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CHAPTER 4

Introduction to Assessment

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Abstract: Assessment tools are vital in measuring the outcomes of any practice or procedure. In the development and use of a prosthetic limb, this can be divided into three areas; the basic functions of the design, activities the limb is used for, and the amount the user actually employs the hand in everyday life. Each area is distinct and different and it needs different tools designed specifically for each area in order to reliably measure these outcomes. The development of these tools must include means to make sure the tool measures what the tester thinks it measures and makes sure that such measurements are consistent across time and between testers. Once a consistent set of tools is developed it allows clinicians to discuss and compare devices, training methods and solutions. It also allows investigation of different designs.

Currently, the emphasis is on the basic practical measurements of function, activity and participation. This uses simple methods based on observation, timing or questionnaires to measure the use of simple prostheses. With newer designs of multifunction hands and microprocessor controllers being introduced, there are more varied control methods for the different hands. This requires more sophisticated methods to measure the impact of the new designs. These new methods include the measurement of the motions of the body and upper limbs with optical methods, and looking at measuring the cognitive load that controlling such hands impose on the user. To allow simple comparisons between users, the tasks and methods have to be constrained. This creates more artificial activities which may themselves be too artificial to tell the observer what they need to know, so the choice of activity is a balance between realistic tasks and reliable results.

Keywords: Upper Limb Prosthetics, Assessment, Outcome Measures, Occupational Therapy, Prosthetic Hands.

WHY WE NEED ASSESSMENT

In all aspects of endeavour it is important to be able to measure the impact of our efforts. In medicine this is especially true, as the effectiveness of a treatment or technique has an impact on how we might direct our efforts or resources to ameliorate or fix a problem. Assessment is the act of measuring change in a complex system. In the physical sciences it generally is quite easy to measure the impact of a procedure through the direct measurement of the change created, such as changes in strength, length, or mass. In the biological sciences it is much harder to find something that is unique and meaningful to measure. Instead, what commonly happens is we must find something we can measure that changes at the same time, and in a similar way, to the change we want to measure, and use this as our window into the changes. Of course this measurement only works as long as the relationship works. In any complex activity (such as employing a prosthesis) where there are many different aspects to consider. A single measure will only capture some small part of a total picture.

For example; in a team sport, it might seem fair to pay all team players in proportion to how good a player they were, but, one must be able to say what a 'good' player is. We might assess their ability, give that a value and then perhaps divide the salary pool in proportion to that value. The bigger the value, the larger the pay cheque, so the better the player the better the salary. This all comes down to the definition of what "a good player" means. So the important question is what indicators can we use that can tell us their level of ability.

A simple measure of the value of a player might be the number of goals they have scored. Clearly a high goal scorer in any sport *has* to be a good player. But would this be fair? Of course not. We all know of excellent players whose role is more to set up their team mates to score, which is why in team sports such as ice hockey, goals *and* assists are counted. This is fairer, but even this cannot cover the matter completely, as the

Vincenzo Parenti Castelli and Marco Troncossi (Eds)

goalkeeper rarely gets to score or assist scoring. So the idea of using the number of *shots on goal* or *saves* too could be used. Hence, these measures cover much of what a player can do, but not all of it. Finally, when one has considered all these mechanical aspects of the game, there remain an intangible factor, some teams can do better simply because someone special is playing, irrespective of the number of goals, assists or saves they make. How could you capture their impact on the team? So one can see that even something as straightforward as a team sport has many factors that cannot be easily measured, and no one factor sums up the complex idea of a good player. So when it comes to the use or function of a prosthetic arm, the majority of things that are important to the use of the device cannot be easily captured by a single, simple, measurement and what you need to know will depend on who you are or what you are doing; from designing to fitting a device, and training or using the prosthesis. Each will dictate what your priorities and interests are and so what you want to measure. It is no more complete to rely on the number of things one can pick up in a set time as a measure of function for a prosthesis, than it would be to count only goals. How often does the user employ the arm to assist in a task? Or how much of the time is an object picked up by the prosthesis and passed to the natural hand, or is passed across the body (if it is too large or awkward). Finally, how does the person feel about their hand? This can be the intangible, but important, aspect of the fitting.

LIFE CYCLE OF A PROSTHESIS

A prosthetic hand or arm is like other manufactured devices. It has to be designed, built, and tested, before it goes to the users to see what they think. A difference in the process, as compared to other products is that a clinical team will have a perspective on the product. Can they fit it? Is it compatible with other components? Furthermore, the user has her/his own feelings. Will it perform the functions the user wants to do with it? Is it easy to use? What does it look like? Can I afford it? Each person is part of the process and has a different outlook and aim. The process can be divided into four phases [1]:

Research In this phase, it is the engineer who wants to make a device to undertake some task. S/he is interested in the basic ideas of the mechanical form and control of the hand. What the prosthesis needs to do may come from a specification, which is based on the input from other people in the field. The design process is always iterative; design, test, adjust, retest and so on. At this stage assessment simply involves measuring the physical properties of the device; its mass, size, speed, grip/lift strength. This phase is the one that most resembles the physical sciences where the measurements can be simple (in the sport analogy like the ability to control the ball in isolation). Ideas are tested and accepted or rejected based on the physical properties of the device. At this stage it is not a practical device and cannot be used by a limb deficient operator for any length of time. It is the simplest measurement, like the goal tally in the sporting example.

Development When the concept is shown to have some merit, the ideas can then be tested performing activities that the final device might undertake. Here the design must evolve into one that can be used by a person in the field. They can then perform tasks to test if the design is capable of the functional range set for it. It is like recording the number of goals scored. The results of this may be fed back to the original designer so that the research can continue, or used simply to adjust the current design to allow it to work in the field. These are assessment that use the hand's performance and are not an evaluation of the hands themselves.

Clinic The next phase is to take the device to the clinic and see volunteers try the device out for the first time. This requires simple tasks to be undertaken, where the results are unambiguous and the team can see if it is working or needs changes to the design. In this phase assessments cover both performance and social aspects such as size and shape of the device. Hence, different and more complex measures are needed, thus in sport it would be like measuring the number of assists.

Home The final arena is that of everyday life. Does the prosthesis do the things the users want from it? Many of these things are the intangible. The way persons feel about their prosthesis may be dictated by factors that have very little to do with the design of the device, or what the designer was thinking about when s/he created the device. Perhaps the person's feelings are more to do with how recently they underwent their amputation, or how people at work reacted to them, or what they believed the device would do for them when they ordered it. All of these factors are useful to measure so that the research and/or clinical teams can determine why the person accepted or rejected the device and subsequently can change how they approach the design or the individual next time to ensure that they can provide the correct service for that person at that time in his/her life. None of these are closely related to the way the device behaves or moves. Used in this phase are methods that resemble the social sciences where many measurements are qualitative by their nature, such as questionnaires and interviews. On the field, this is that intangible extra that a player brings to the game, the morale or the feeling of success or failure, which clearly cannot be read on a simple scale.

From the above, it is clear that the different people involved in the evolution of a new prosthetic limb have different interests and will ask different questions. The requirements for an assessment for each of them will be different and so it is unlikely that one and the same tool will suffice, hence the need for a range of assessment tools. As part of this process it is important to ask the different questions clearly and unambiguously.

International Classification of Functioning

One way that gives a perspective to ask the right questions is the International Classification of Functioning, Disability and Health (ICF), developed by the World Health Organisation (WHO) [2, 3]. It aims to create a single unified language for different professionals to describe health and health related status in the hope that everyone can adopt and use it and thus make communication between professionals, groups and countries, simpler, removing ambiguities. With this model it is possible to make the specification of goals (whether engineering, clinical or personal) easier and clearer. The ICF divides human functioning into four components: "'Body Functions and Structures"', "'Activities and Participation"', "'Environmental Factors"' and "'Personal Factors"'. The first two components are sub-classified into *domains* and *categories*. Each component is part of the process of creating, supplying and using interventions for persons with impairments.

The ICF relates the interactions between the health condition and the context of the person, see (Fig.(1)). The context can be the *environmental* or the internal *personal* factors, which are specific to that person (age, gender, education, experience). The domains are the components describing how the condition impacts on the person. The classification can then be used to identify how much function an individual has and will help to identify their limits and what sorts of assistive devices might overcome these limits. Each domain can then have qualifiers, for example if a body structure is missing or impaired, the first qualifier will be that it is missing and the second relates that it is a partial or entire absence. These are each given a code number. The advantage of this is that it is a systematic way of looking at, and recording, the condition so that comparisons can be made between stages of a condition, or persons with the same condition. They can be made by different clinicians at different times and the results will be standardized and reliable. From this, many different sorts of analysis can be made from demographics to healthcare policy, from research to development. It is this last two categories that are focused upon here.

The ICF system built on an earlier methods [4], and while it filled many of the gaps inherent in the earlier approach it still contains too little on the measurement of the quality of life of the persons assessed. In this current example, the ICF has been used as a starting point by professionals in the upper limb prosthetics

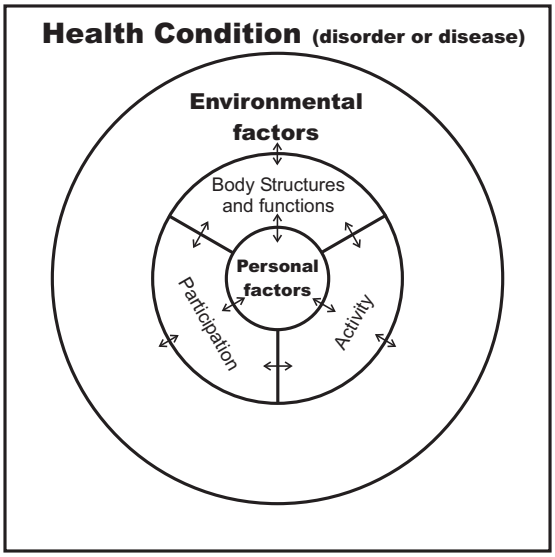

Fig. (1) — Representation of the entire WHO-ICF model, showing the different contextual areas (Personal and Environmental) and the three domains (Body structures and functions, Activity and Participation) and their interconnections to each other.

field. They have built on the approach and attempted to address some of the flaws in the method. For example: In the ICF prosthetic limbs are considered assistive devices and are thus, covered by the component "'Environmental Factors"'. However, as with other assistive devices such as, e.g., electric wheelchairs, when frequently used they become part of the users' body image. This is not acknowledged by the ICF. Instead, the ICF significantly ignores the experience of the profession and reduces the chances of creating a unified language and approach.

A working group of professionals in upper limb prosthetics has made the suggestion that, as the prosthesis is a replacement of a bodily part and considered as such by most of its users, when studying the outcome of upper limb prosthetics it should be regarded as part of the "'Body Functions and Structures"' domain [1]. For prosthetics, this component captures how well the person has the basic functions to operate the device (number of possible EMG channels and the length of the residual limb) as well as the performance of the device itself, (the speeds, strength and power of the device). It is this interpretation that will be used here. The "'Activities"' component looks at how the device can be used. One aspect is the person's *capacity* to operate the hand (grasp, release, hold) and a second quality is the *performance* of the hand, which tasks a person can do with the prosthetic design, and "'Participation"' is how the prosthesis is *really* used, by the wearing public. It is concerned with how much the person engages in real life situations with the prosthesis. From this perspective it can be seen that the ICF captures most of the areas described for the development of the prosthetic

hand within its three domains, shown in (Fig.(1)).
The full ICF model then looks at the relationship between external and internal factors and the different domains and allows coding of those factors and whether the impairment is mild, medium or severe. For the purposes of this exercise most prosthesis users would have very similar codes, but it is the discipline of how the interactions are considered that is useful for this study of assessment.
Each domain has a different method to measure the quality within the domain:

Body Structures and Functions are the technical aspects of the device or the physical aspects of the person. How big it is, how fast, what is the size of its grasp, how fast can it grasp an object, which objects? These can often be measured with tools such as rulers and scales or timers.

Activity is about the operation of the device and what the device can be used for. So the measures are based on performing activities with domestic equipment such as jugs or cups or other items that allow Activities of Daily Living to be simulated. For a prosthesis, it is easy to imagine this form of assessment taking place in the clinic, or a simulated domestic environment. Clinicians would then observe the device being used.

Participation is what the users really do with their devices. To capture this generally requires asking users, so this is measured using questionnaires. This captures the opinions of the users, which is a complex and multi-factorial process.

What is clear from this is that each domain is distinct from the others and the chances that one tool designed for one domain can measure appropriately any other domain is slight, (Fig.(2)). So what is needed is a series of tools, one or more for each domain, which together can cover the range from the design office to the home.
The development of a prosthesis becomes a process with different assessment tools used at different times:
First, the designer wishes to find out if a particular design can open wide enough to grasp everyday objects or flex fast enough to be useful. So they use *Functional/technical* tests, such as measuring the gape of the hand or timing it as it opens and closes.
Later on, as the device has advanced, assessment shifts to *Activity* based measures. Can the hand pick up household objects? Can it retain them to move them about? While this is important to the engineer, now input from the clinical team becomes important as their insight into its long-term use becomes relevant. Early fittings in the clinic use observation-based measures of prosthetic use.
Finally, the device transfers to the home/community and the activities must reflect tasks specific to the needs of that user. Now the outcome measure may tell the clinician about the functional capabilities of the hand or of

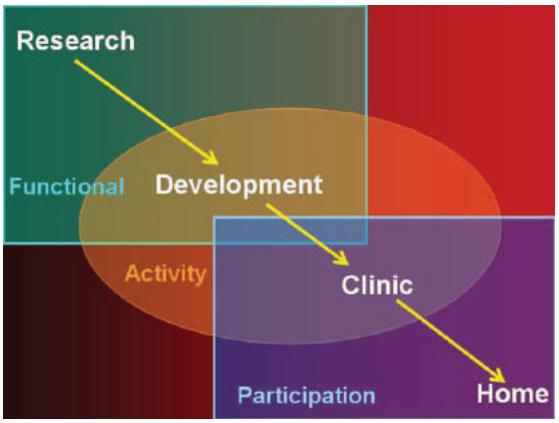

Fig. (2) — The development process of a prosthesis mapped onto the WHO-ICF domains. The overlapping areas show the potential competences of different tools to assess different parts of the process.

the person's ability and something of their motivations. If the clinicians want to know how the user feels about their device and how it integrates into their lives (i.e. participation), how it changes their quality of life, then the information is more likely to be obtained through questionnaires.
From the above it can be seen that at each stage a different tool is used to obtain the information, and that each provides a different insight. Some of the measures will overlap with adjacent domains, but only with multiple assessments will the full picture be clear. To cover the entire range will require the identification of a sufficiently broad set of tools.
Now that we understand what it is we need to measure, we must now understand how complex the simple act of measuring something is.

MEASUREMENT

When we wish to measure something we use a means that is understood by everyone, because it is an agreed standard. For example; if we wish to measure length we use a ruler that is marked in meters. The length of all rulers are based on a single standard. So any meter is the same as any other.
Measuring mass is more like assessment. We measure mass by inferring it from the weight of an object. The weight is the force experienced by a mass in a gravitational field. So if we measure the weight on the earth and the moon we will get very different results. The moon's gravity is one third of that of the earth, so an object's weight will seem to drop to a third of what it was on earth. For a correct measurement we would need to use a method that is known to be the same on both planets, or know that one method is not appropriate (or *valid*) for one particular place.
The important matter for any measurement of a property is that; is it the same if one person makes the measurement or a second person does? Also, is it the same

measured today as it will be tomorrow? If two identical subjects are tested will they get the same score? There are a number of things we must be able to control for such a thing to be possible and it is not always appreciated how many of these things are common knowledge. For example; in measuring length, it is likely to be true that if two people perform the same measure on the same object days or months apart they will come up with the same value. But this is true, *only if they both know how to use a ruler*. While this is probably the case, other measures are harder to make as consistent as this.

Validity

The *validity* of a measure is those qualities that make it mean what you think it means and *reliability* is that it will do so irrespective of who or when the measurement is made. Validity and Reliability are ways of demonstrating that any measure can be trusted to give an answer that reflects the underlying truth. They are referred to as the *psychometric* properties of a test.

There are a number of different qualities that a test can have show that it is valid:

If a test has *Face validity* it means that, on the face if it, the instrument appears to assess the right qualities. This is a subjective assessment of whether it does it do the job it is supposed to do?: "Do the best players really get paid the most?"

A second consideration is *Content validity*. This is a judgment as to whether the instrument samples all of the relevant content, or domains. For example; does our test of a player use goals, assists, shots on goal and saves, to make its determination? Also, are these the only qualities we know that indicate a good player?

If there is an existing test that will reliably make the assessment, but if it is too complex, or time consuming to be used routinely, you may want to devise a new test. In this case you may assess the new test's *Criterion validity*. For this you make measurements with both tests and see if they both come up with the same results. For example if the scores go up and down together and by the same proportions (but not necessarily the same values), then the new test has Criterion Validity. Additionally, if these sets of measurements are made at the same time, then the new test has *Concurrent validity*.

When there is no existing scale to compare a new test, the originator constructs a hypothesis to explain the results of the assessment and compares those results to those genuinely obtained. The hypothesis is that the something of interest (say manipulative function) is tied to a variable, (such as number of blocks moved in a minute). The originator of the test would then measure subjects with that test and see if the results vary in a way that matches her/his expectations. If yes, the test has *Construct validity*.

Finally if the test can predict an outcome, then it has *Predictive validity*. This is like testing a person before fitting them with a prosthesis and being able to tell how functional a user they will become, or testing a student before they start their university career and being able to say what grades they will get at their final exams. For prosthetics this sort of prediction is extremely hard to do.

Reliability

If the test has validity then another important feature is that it must be *reliable*. This has already been touched on. We need to know that the result does not depend on the person who measures it, the person who is being measured or the day the measurement is being made on.

So if the same thing is measured two successive times, seconds or days apart and it hasn't changed then the score must be the same (*inter-sample or test-retest, reliability*). There are two forms of this. The first is if one person measures the same thing twice, then it is *repeatable*, (intra-rater reliable). The second is if two *different* people measure it in succession, this may prove the *inter-rater* reliability. The third factor is if two objects have the same property then they must score the same if measured under the same conditions (irrespective of who measures it). This is *inter-subject* reliability.

A final aspect of interest to the person making a measurement is how to interpret the result. One aspect of interest might be to measure the change brought about by a new treatment or technique. Ideally, an objective measure would be to run a test before and after the treatment, and see the difference in the scores. The question is then; is the difference due to the normal difference between measurements (the reliability) or due to some change in the subject? Which it might be clearly must depend on how much difference is normal between tests of the same person who hasn't changed. The more reliable the test, the smaller the differences between measures of the same state, and so the smaller the measured change can be that is due to a real change in the subject. This is the test's *Sensitivity* to change. However, just because it is possible to measure a difference this might not have any real meaning. For example; if we add the number of assists to that of the number of goals scored, and the same person still has the highest score, then the salaries will not change. Or if a person's new hand scores five points higher than their old one, but they still cannot tie their shoe laces, then the difference will be one that makes no real difference. This is the *Responsiveness* of the test; its ability to measure a clinically relevant change.

Functional Measurement

While for physical objects we can see if we are measuring the same thing, with biological measures it is much harder to be sure we are measuring what we think we are. So in any measure, it is important to establish what our tool is actually measuring.

This is similar with functional assessment in prosthetics. The idea of functional performance is something that an experienced practitioner can recognize, but it is much harder to describe or quantify. Some tests are observational and require a skilled practitioner who is trained to conduct the test. If an observer sees a different functional performance with two prostheses and the same user, what is different? Is it that one of the two devices is a better manipulator, or is the user more skilled at using one device over the other? This is *capacity versus performance*. Both are interesting questions, but it is important to establish which it is we are measuring and if the tool can differentiate between these two circumstances. Otherwise we may come to the wrong conclusion; that one device is better than another, when the subject has just had more practice with the first than the second.

As we have seen, for every new measure the validity needs to be demonstrated, so that any new measure needs to be compared with an older tried and validated one (if one exists). If the older measure records a drop in value or performance, then if the new measure is sensitive or valid, it too must show a similar proportionate change. So when a new measure is devised it must have its various properties measured and compared to older ones so that we can be sure of the results. This means the development of a new measure is a long and complex one, estimates say up to ten years for full validation [5]. For more information on this topic: [6].

Development of a New Assessment Tool

Once a particular aspect of the development process of a new prosthesis is identified, then the appropriate means to measure it must be chosen. For example; if the test is activity based, then how the activity is scored is important. One way would be to record the time it takes to complete the test, or the number of times the subject can perform the test in a fixed time. This is the person's capacity to operate the prosthesis. Scoring can be simple and can make the measurement objective as it does not depend on the opinion of an observer. It may be repeatable as the task can be performed well and often, but this captures only a part of the entire process of undertaking an activity. The alternative is if you are interested in the actual use of the prosthesis in daily life, then one needs an observer to watch the person perform an activity and score how good the subject is at a task or how easily or often they perform all the task or important aspects of it, such as picking up objects with the prosthesis, or only grabbing them with their intact side then passing it over to the prosthesis. To score the tasks reliably requires the observer to be trained to perform the test and skill and practice to observe the task consistently. A skilled observer, someone who is knowledgeable in a subject or activity will be able to see more of what is happening and interpret the motions of the user more accurately, in the same way as a keen sports fan will get more information about how good a player is by watching them score a goal. A different method to look at how the prosthesis is used is to employ a questionnaire, which allows the subject to score how well or often s/he uses the hand and for what tasks.

Once we have chosen what we will measure and the way it is done, we must test the face validity. Does it measure a hand's ability to hold an object? Or is it really the operator's ability to use the hand in an effective manner. These are two different things and if it measured one when the intention was the other then the face validity would be poor. Similarly, the content validity of a hand test might be that it measures how well the hand picks up, holds on to and releases a range of objects of different sizes and shapes. If it fails to measure the hand's ability to hold large smooth round objects, but only small light rough objects, then its content validity would be suspect.

After this the reliability of the tool needs to be assessed. We can see this means many separate measurements of the same people on different days with different observers to see if the same ratings are obtained.

If we know the above then we need to work out how much it can measure, and what its sensitivity is. For example, we know that we do not use a meter rule to measure the width of a piece of paper, or micrometer to measure the size of a room. Each tool's sensitivity is appropriate to the job to which it is put. Another matter of the sensitivity is what it can differentiate. What is the smallest difference between samples that we can measure and know that the two are really different? If one item is 3.5 and the other 3.6, is one really different to the other, or simply is it the normal variation between two measures of the same object that causes this difference in values? Above all, these statistical concerns must be considered when designing the measure.

Beyond this, how well is it likely to be received by practitioners? If it is to be used generally, the tool must be user friendly, i.e. easy to administer, inexpensive and the results must be easily understood and easily interpreted. If it does not fulfil these specifics, then no matter how valid or repeatable it is, it is unlikely to get wide use.

THE FUTURE OF ASSESSMENT

From the above description it can be seen that there are a number of barriers to making an assessment and understanding any results generated. The first would be that the problem is poorly defined. Without a framework within which two people in different places can talk together it is hard to know what someone means or if there is agreement between them, like if the definition of the word 'length' in two different countries is understood to mean the same thing.

Historically, there has been little consensus over any forms or methods of assessment. A group may develop a tool and use it, perhaps with only limited validation.

The group may continue to use it and promote it over a number of years, and so others may adopt it without criticism. The idea and theories of assessment have developed over the years and those who have adopted a particular test may not understand fully the implications of their choices or methods. One result of this is that there have been local variations created, based on a particular test. A group of therapists in a busy clinic may find that a specific test takes too long, so they choose to use only some of the test. They may pick the parts of the test that make sense to them, or seem to tell them about things that they are interested in. They may substitute one test for another, or incorporate some elements from another test. The drawback of this is that any of the validation work conducted by the originators of the test is now completely irrelevant. The test may not be repeatable any longer, and it may not measure what they think it does. The choices made are based on the practitioners experience and so they may get subjective information from one of these modified assessments. More importantly, they would not be able to generalize these findings, or compare the results with others, as their interpretation would not mean the same thing as the original unmodified test results.

In the example above, the second group may use the same terms as the originators and mean something quite different, hence no common terminology exists. When such terms are used in relation to outcomes, there is ample opportunity for misinterpretation of the meaning if the terminology is not clarified. For example, the question: Is this device reliable? This question seems simple and straightforward, but the words mean different things to different people within the process. *Reliability* to an engineer would be the average time between failures, but to a user it is likely to be if the hand opens when s/he wants it to, or if the battery lasts all day irrespective of the amount of use.

A further problem is that any work addressing the validity of any particular measure requires large numbers of subjects in order for the statistics to reach significant levels. The overwhelming majority of limb fitting centres rarely see sufficient users in a year to reach this goal. Only with the combined results from multiple centres might the numbers of people tested climb to usable levels. Similarly, if a consortium of different centres wished to work together it is hard to pool the data without an agreement on what is meant by any measure.

In recent years there has been a push for treatments to be more based on clinical evidence. This means that it must be possible to measure the impact of an intervention or treatment. Validated measures of assessment therefore become an important part of this process. When Marshall conducted a survey in 2000, of the 300 randomized trials of schizophrenia treatments, they found that a test was 40% more likely to be positive in their outcomes if they used unpublished (and so non-validated) scales [7].

In the early years of the 21st century individuals and groups in the prosthetic field were becoming convinced that the only way forward was to develop validated tools and one such group reasoned that this state of affairs could only exist if a consensus was reached within the professions of therapists, prosthetists and engineers. Given that it can take a decade to develop a brand new tool, the group did not believe it would be effective to create new tools when some may already exist and knowledge of their strengths and weaknesses were already known and understood. This group became the *Upper Limb Prosthetics Outcome Measures* group (ULPOM), which drew from professionals in Europe and North America [8]. They set out to create a consensus by analyzing the state of the science at the time (roughly 2005) and to make recommendations concerning which tools were useful. Around the same time, the World Health Organisation published the International Classification of Functioning, Disability and Health model and it was very simple to incorporate this idea into the thinking of the group. The ICF was a revision of the earlier, International Classification of Impairments, Disabilities, and Handicaps (ICIDH), which was first published by the World Health Organization in 1980 [4]. After field trials and consultation, the modified system was endorsed by the World Health Assembly in 2001. The ICF defines terms and makes it clear when two persons are talking about the same concerns or not.

The ULPOM Process

The group investigated the assessment tools for upper limbs on record. They located any literature and reviewed if these tools had been tested for validity (and what sorts of validity) and what area of upper limb function they had been tested for [1]. Based on these results they were then able to draw up a list of the different tools with a recommendation for their domain of competence and if they were usable, and if not, what would need to be done to make them usable. Additionally, the recommendations would allow groups with an interest in furthering the idea to fill in the gaps identified, carrying out the validity testing identified by the group.

Thirty five assessments in the areas of; hand function, activity, goal-setting, quality of life, and user satisfaction were identified. The group determined one of three recommendations for each test:

Accept if the test had psychometric merit and was clinically useful for upper limb prosthesis users.

Consider if with modifications or psychometric validation the instrument could be useful for upper limb prosthesis users.

Reject if the test did not have psychometric merit and/or was not clinically useful for the target group.

Table (1) contains the important results of this analysis.

Table (1) — Upper limb prosthesis outcome measures recommended by the ULPOM group, categorized according to the ICF a - primarily intended for adult population p - primarily intended for paediatric population.

ICF Domain	Accept	Consider
Function	SHAP[9] (a)	Box and Blocks [10] (a,p) Sollerman [11] (a,p) Michigan Hand Questionnaire [12] (a)
Activity	ACMC [13] (a,p) PUFI [14, 15] (p) PODCI/POSNA [17] CHQ [18] (p) SHAP (a)	UNB Test (p) UBET [16](p) Assisting Hand Assessment [19] (p) TAPES [20] (a) OPUS [21] (a,p) CAPP-PSI [22] (p)
Participation	COPM [23] (a,p) PEDSQL [24] (p) CHQ (p) Disabkids [25] (p) PUFI (p) WHOQOL [26] (a) Goal Attainment Scaling [27](a,p)	TAPES (a) OPUS (a,p) CAPP-PSI (p)

The ULPOM ultimately joined forces with members of the American Academy of Prosthetists and Orthotists at a State of the Science Meeting on the topic of Upper Limb Outcome Measures to extend and deepen this review of the state of the profession and broaden the recommendations. As a part of the meeting an evidence based review was performed with recommendations based on an in-depth review of 640 peer-reviewed journal articles. The recommendations from the EBR paired with the recommendations from the ULPOM and were combined into a document as a supplement to the Journal of Prosthetics and Orthotics [28].

From this analysis, the next stage for the profession is to adopt the tools that are already valid, adapt those tools that are in the 'consider' category to make them valid and reject others (not listed). If anyone wished to add a test to the list they would need to become involved in the process of validation of the tools.

The critical consideration concerning the ULPOM process is that the members cannot dictate which tools must be used by the profession, they can only point in the direction of those tools that are the most valid. Beyond this it is up to the individuals and clinics to decide which tools to use. It is for this reason that the "accept" category has more than one tool in each domain. Many centres will have experience in a particular measure, or may not be able to afford a new tool or the training to use it, thus they have a choice between several different ones and it is hoped that they can find one to suit their needs and tastes. Others may wish to conduct research based investigations and others will be more clinically focused, which will influence their choices.

NEWER METHODS

The ULPOM method has so far concentrated on the measurement instruments that were already in existence and partly validated. At the same time a revolution in the application of technology to prosthetic arms has been taking place. After many years of stagnation with minimal changes to devices (outside specialist centres which continued to innovate alone), an increasing interest from the manufacturers in improved designs of prosthetic hands was observed. Many of these designs are multifunction hands with multiple independent fingers. Similarly, advanced controllers, which are able to use more information taken from the user and the prosthesis were becoming practical and new surgical techniques that could provide more control channels were being perfected. This has created a need to measure more than the person and the hand's simple ability to grasp and move an object. Fortunately, at the same time, new measurement tools and techniques were being developed by other industries. These could then be used to measure more advanced concepts about how a prostheses is used and how it might be controlled. With all these measures it is important to design the right tests and be sure they measure what we want to measure. The following describes the new tools and some of the pitfalls encountered in trying to design a new assessment method for prosthetic arm users.

Motion analysis

We are all able to spot when a person or animal is injured from the way it moves. So, the idea of using visual tracking of motion, especially for gait, is clearly an

appealing one. Marey in 1873 was the first to record the motion of people and animals walking and then performing functional activities [29], but this analysis was restricted to simple measures. It required the development of electronic cameras and high speed computers to be able to perform detailed analysis of the motion of humans [30]. Initially, this was limited to walking (gait) studies. This was for several reasons, not the least being that the resolution of the cameras did not allow the easy recording of any other motions. A second factor was that walking is a very well controlled activity. Normal healthy gait is cyclic and repeats closely. This allows a single motion (or stride) to be recorded and for it to be regarded as representative of the person's motion in general. Motion data from different strides can be overlaid one cycle on another to gain average values. Speed and stride length variation can be removed when the graphs of a person's gait is normalized to 100% of the gait cycle (that the same point in the stride is taken as 0% and that point in the next stride as 100%, and the motion scaled so that the entire stride matches this). This makes the gait of two persons of different heights and walking speeds easily comparable. Upper limb use is very different. Every reach and grasp tends to be different to every other, both by the same person and between two different persons. This means that if measurement is not designed carefully it cannot be compared easily. If one was to repeat many different tasks, many different times, with many subjects, it would be possible to get statistical measures of the tasks as the person performs them. With this one would get an indication of the repeatability of the action. However, this would require extremely large numbers of subjects. Even then, to record a representative number of movements might prove impractical.

A second way is to control which tasks are performed and how they are executed to ensure that the subjects all perform them sufficiently similarly, so that the same types of comparisons as undertaken with gait analysis, can be made. This latter solution can result in reasonable results but it can be criticized as making the tasks so abstract as to be artificial. The person may not perform the task in a natural manner and the motion may not actually reflect how they would use the hand or arm in a real situation. However, in a similar manner to the other assessment tools discussed previously, some valuable information about the way a device can be used, or is used, can be measured. As with the other instruments it is important to establish the aim of the measurement. One aspect may be whether it is for assessment of the individual or the gathering of population data.

Examples of the Role of Motion Analysis

Over the past decade the adoption of different tasks to drive motion analysis of the use of upper limb has grown. One solution proposed by Rau was to use repetitive tasks so they could observe differences in elbow and wrist angular patterns in populations of unimpaired individuals and subjects' affected by a plexus lesion [31] and shoulder impingement syndrome [32]. The tasks used did not include daily activities, however the authors indicated suitable parameters (such as the smoothness of execution of the task, and the angular range of motion during critical events within a task) to identify the differences between healthy and non-healthy individuals.

Light and others used a tool developed for hand function assessment; the Southampton Hand Assessment Procedure (SHAP) as the driver for tasks of daily living [9]. The design of SHAP was particularly effective at creating a closely controlled task set. The tasks revolve around a form board and are self timed. Each task is commenced and concluded by the pressing of the timer button. This creates the same start and stop posture, similar to heel strike or toe off in gait. Each SHAP task (twelve abstract objects and 14 simulated Activities of Daily Living), has a prescribed position on the form board, so each motion is controlled and the trajectories of each body segment is similar between many different subjects. This ensures that the motions of the segments in three dimensions can be plotted together against the standard deviations of the population means and hence the differences between the subject and the population can be observed. This means that, by using SHAP in combination with Motion analysis we are able to study different prosthetic devices impact on motion during task performance, such as the construction of handgrip and how it influences the position of the elbow when grasping the key in the Key turning task, or if different prosthetic devices create different body positions for the same task. If this is the case, would any of the devices indicate a risk for over-use problems in the elbow?

Murgia advanced these ideas creating a nominative group and then compared the results with subjects with a distal radial fracture [33, 34, 35, 36]. The normative results are shown in (Fig.(3) and (4)) for one task. It is turning a key in a lock. (Fig.(3)) shows the actions, with the subject pressing the button, reaching to turn the key, then after letting go pressing the button to stop the timer. In (Fig.(4)) the times are scaled from 0% when the timer is started and 100% when it is stopped. In a similar manner to gait studies, the axes of the arm are show left to right and joint levels from top to bottom. The standard deviation of the normal range is shown as a shaded area and the mean as a bold central line. Note: Pro/supination is properly a function of the forearm, but is grouped with the wrist as it is visually more appropriate.

Currently, only the kinematic data of upper limb motion can be recorded easily. In human gait it is easy to measure the distribution of ground reaction forces to build into the model. To be representative, upper limb motion needs to be rather broader in scope, with multiple motions recorded. For a complete study it would be

Fig. (3) — SHAP Key turning task. Typical events as cycle percentages, left to right, top row: Timer started at 0%, key reached about 40%, key turned at about 60%, bottom row: Timer stopped at 100%. Kinematic data is recorded using motion tracking equipment (for clarity, not shown here)

necessary to be able to record the forces generated for each finger and across six axes (three translations and three rotations). This is a complex process and some progress is being made towards this goal [37]. In the meantime the recording of the kinematic data alone is proving valuable [33, 36, 38].

One criticism of the SHAP style of assessment is that it was aimed to assess the function of one hand, thus most of the tasks were mono-manual, with the hand under test dominating proceedings. For a person with a single sided loss the prosthesis is generally used in a support role, rather than as the primary manipulator. Additionally SHAP restricts the way someone can use their hands to a more limited range. This shows the conflicting requirements for the design of a means of assessment. If a task is restricted then it reflects reality less well, but as the next example shows, the less constrained the test the more variable the result and the harder it is to compare them.

The Use of the Wrist in Manipulation

The single biggest impact that a prosthesis has on the motion of the user is the number of axes in the prosthesis that are easily driven. If there are fewer motions possible than the natural arm (which is generally true) the operator has to use other bodily motions to place the prosthesis in the correct place to manipulate the object. These motions could be through greater ranges or used more frequently than a non impaired person would or they may have to use greater forces through the remaining joints. All of these responses are associated with over-use injuries (such as Repetitive Strain Injury, RSI). There is evidence that having a unilateral loss is associated with greater risk of over-use injuries [39]. It is very hard to prove this directly, and the most direct way would be to record those injuries reported by many users over very long periods of time. This is something that is complex and hard to complete successfully when the numbers of patients are so small. If the greater ranges of motion are required to perform tasks with a prosthesis, these can be observed using motion analysis and this might improve the ability to study this question. Additionally, if the adoption of newer prostheses changed these results, perhaps towards the more natural ranges, it might suggest their use had long term advantages.

Table (2) — Simulated ADL tasks of Zinck et al. [40]

Task number	**Task**
1	Hanging clothes on a clothesline
2	Slicing bread
3	Simulated eating with a knife and a fork
4	Sweeping the floor with a broom
5	Stirring in a pot
6	Cutting a circle in paper with scissors

One such experiment was conducted by Zinck [40]. He recorded the motion of both limbs while 20 able bodied subjects and 4 prostheses users performed six simulated ADLs (Table (2)) in order to observe the impact of the wrist on the rest of the body. Each able bodied subject performed the task with their arms unrestricted and then wearing a custom splint on their elbow and hand to limit the range of motion of the joints as if they were wearing a condylar suspension socket and a single degree of freedom prosthetic hand, (Fig.(6)a), but with an unrestricted wrist. The subjects ability to grasp was significantly impeded by the splint, but they did have a little more motion in their fingers than with a genuine single degree of freedom prosthesis.

The tasks were designed to address some of the limitations of the SHAP style of assessment. SHAP can be simply the use of the apparatus and timed tasks, the motion analysis being an addition. Zinck's work was designed to use motion tracking to follow the person as they performed everyday bimanual tasks. The expectation would be that the prosthetic side would be used in the supporting role, but to find this out, the motions of impaired and unimpaired subjects were measured, along with the motions of a small number of prosthetic arm users.

Some experimental data for the cutting task is presented in (Fig.(5)), the subjects are shown (Fig.(6)a) and (Fig.(6)b). Each graph is for a different axis of the upper body. The angular displacement of each axis is shown, progressing down the body from the head, with the X, Y and Z axes shown side by side. If a particular joint does not possess that particular degree of freedom then the space is left blank so that the position of each axis is

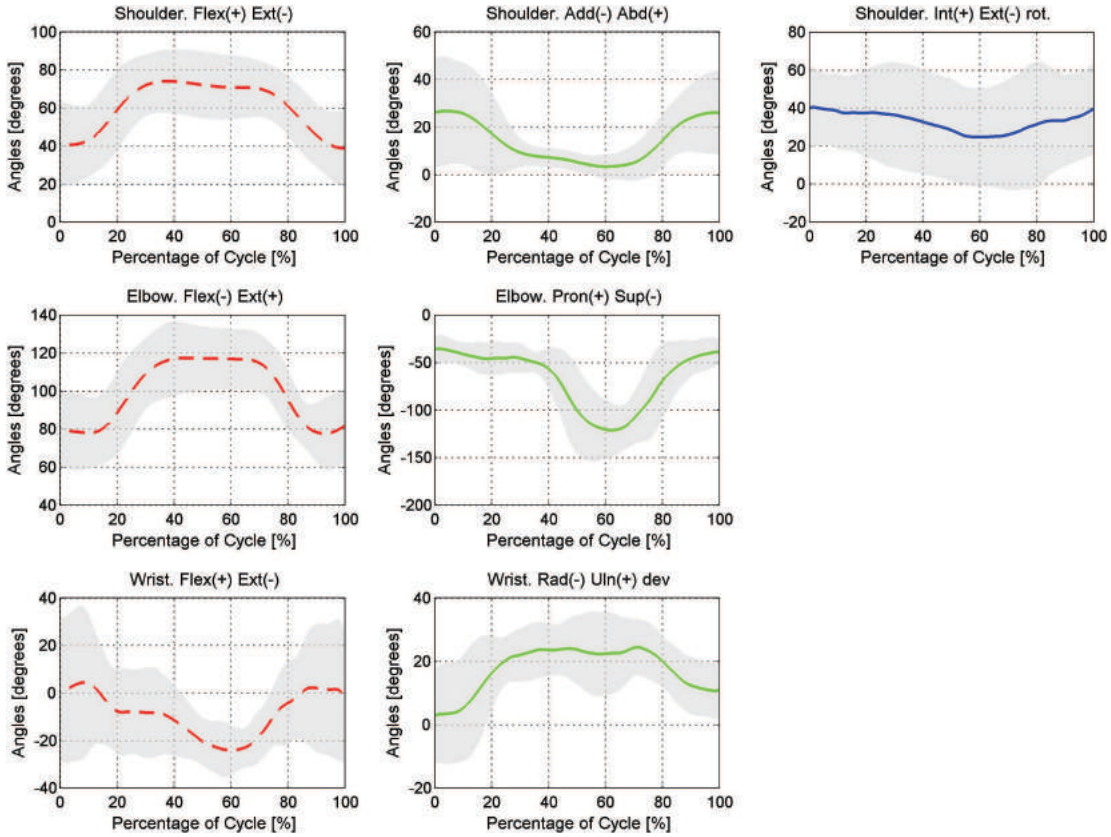

Fig. (4) — Plots for the different angles of the arm during the SHAP task to turn a key. The columns are left to right; flex/extension, add/abduction and internal/external rotation. Reading down the rows are the joints going down the arm, shoulder elbow wrist. The grey areas represent the 95% confidence interval and the bold line is the mean of five normal subjects. Time is scaled to the start of the test as defined by pressing the timer button, to stopping it similarly.

always in the same relative position on the page. When both limbs are involved each chart shows both the sound and prosthetic side. The red lines are the mean paths of the unconstrained able bodied subjects, the green the same with splints and the blue is a single user of a myoelectric single degree of freedom hand, with a passive wrist.

What was observed was a large variability in the way that different prosthesis users performed the same task. The activity of the able bodied users with the limitations of motion resemble those of the same people when unimpaired. Generally, the splinting has a limited impact on the motion of the arms. Only the shoulder and humeral elevation on the "prosthetic" side and the pronation on the unimpaired side resemble the differences that the prosthesis user employs. This suggests that a fully mobile wrist can position the hand in the correct place and the rest of the arm is used to compensate for the limitations in the socket attachments.

Taking the cutting task as an example and the results of a single subject compared with the unimpaired population (red): What can be seen is that the subject chooses a different strategy to perform the cutting manoeuvre. Instead of cutting roughly fifty degrees then regrasping the paper fifty degrees round and cutting the next segment, she holds the paper close to her lap with the prosthesis and cuts round with her sound hand. As she gets to the end of her cut, she chooses to cut from the same starting point in the *opposite* direction. To undertake this she holds her head on one side and her sound arm abducts away from her body, as can be seen in the charts. This reduced the number of times she needs to open and regrasp her prosthesis, something which will take longer and increase uncertainty in the success of the task. While the splinted, unimpaired users continue to cut the paper in small stages and move the paper around.

Similarly, compensation was seen with the other tasks. From this it is clear that prosthesis users employ wider ranges of motion in the trunk and the prosthetic side than the unimpaired group. So there is a higher risk that the person will suffer from overuse injuries in later life compared with the general population. Additionally, these injuries may not be isolated to the prosthetic side, but may affect any joint of the body. Since people with limb absences depend even more on their ability to perform tasks one handed than most non affected individuals, it is a concern that the wrong prosthesis, or one

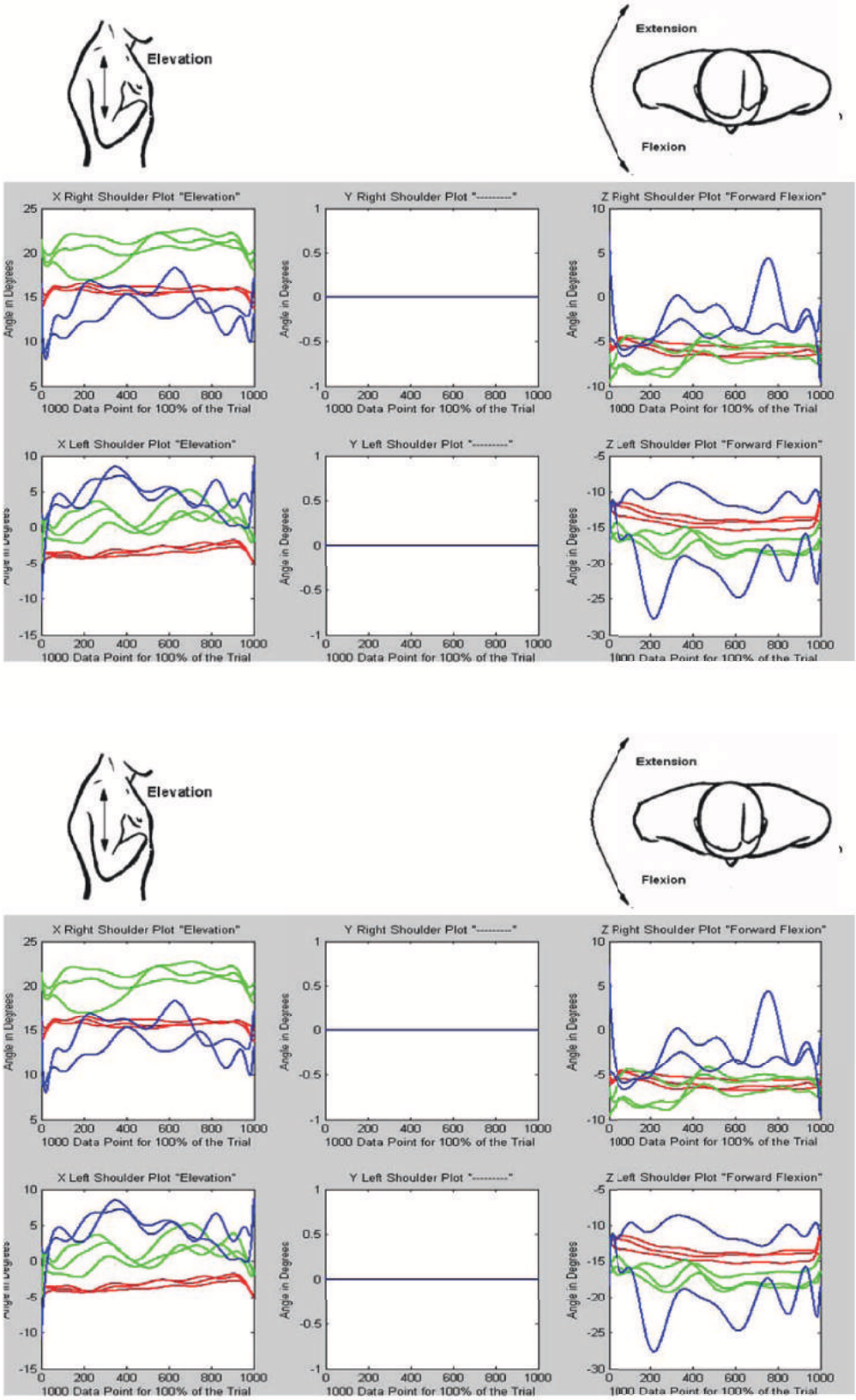

Fig. (5) — Cutting task (Figure caption at page 52)

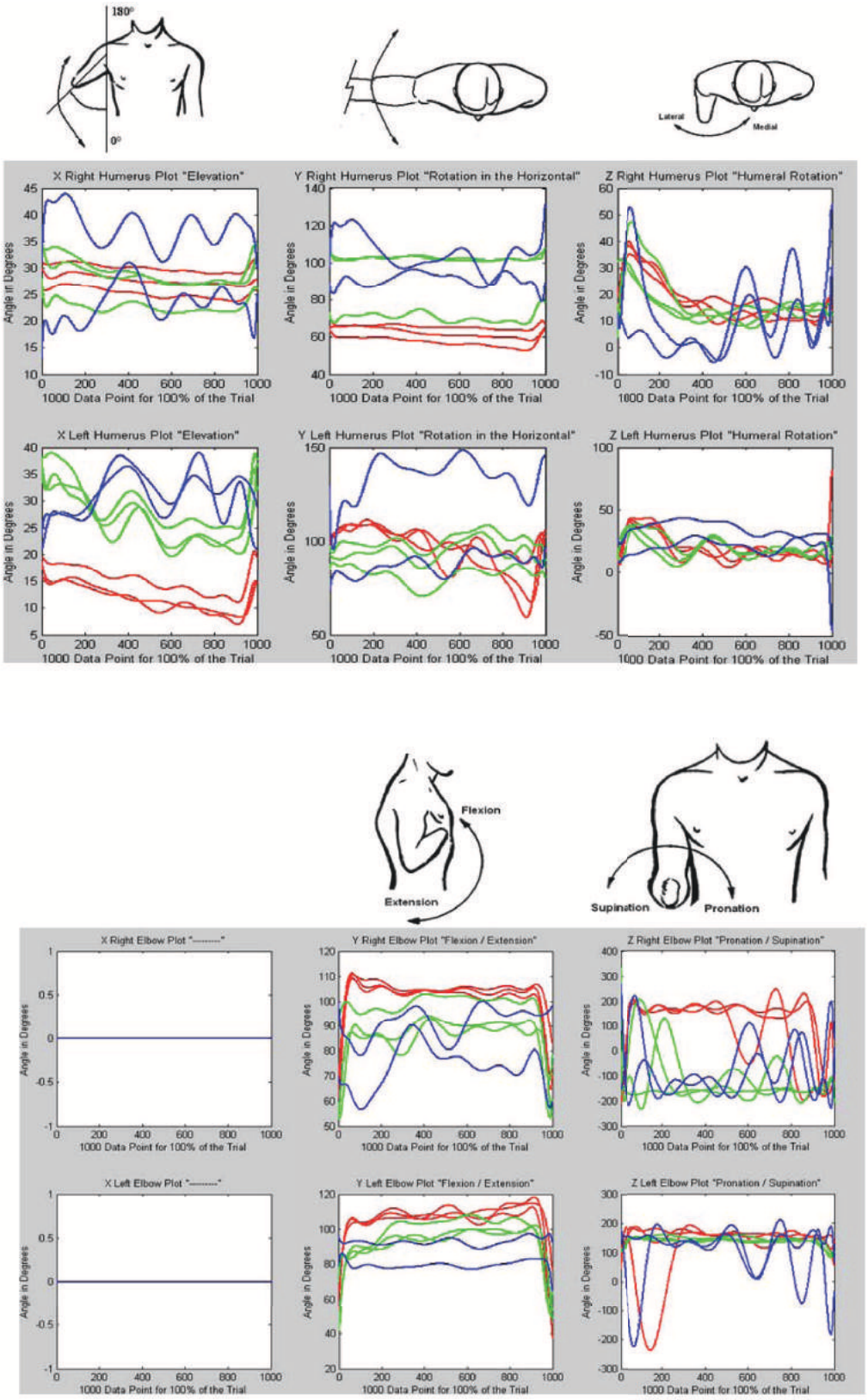

Fig. (5) — Cutting task (Figure caption at page 52)

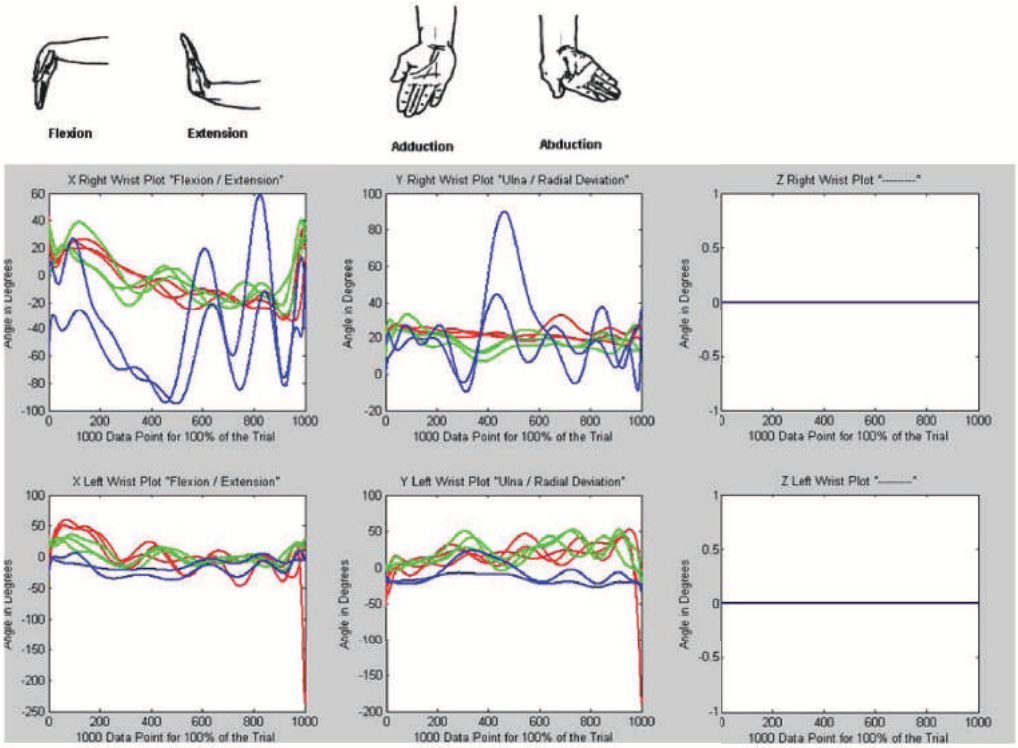

Fig. (5) — Kinematic data of the cutting task. Able bodied population are green curves and the two blue lines are the motions of a single prosthesis user. Each joint is plotted with X, Y and Z axes, with the full execution of the task being resampled to 1000 points and this is plotted as the abscissa. If the joint does not flex in that axis the plot is blank, but maintaining all three plots ensures that the relative position of the different plots remains the same. The task is to cut a circle in a piece of paper with a pair of scissors. The prosthetic subject keeps the paper in her lap and rotates the paper more rarely than the two handed subjects, cutting from on top after 450 points. This is most clearly visible as the peak in the head tilt, humeral rotation and wrist abduction. Red = Unimpaired population mean, Green = splinted population mean, Blue = User of a single degree of freedom myoelectric hand. Upper set are the right, unimpaired arm, lower plots are the left impaired or prosthetic arm.

(a) (b)

Fig. (6) — (a) The cutting task performed by an unimpaired user and the same user wearing splints to impede motion to simulate the use of an extra-condylar suspension socket and a single degree of freedom hand, but a fully articulated wrist. Note: Unlike the experienced user of a prosthetic hand, the subject continues to cut the paper but makes small cuts and moves the paper in his hand; (b) the cutting task performed by a user of a single axis myoelectric hand with passive pro/supination. Notice that instead of repositioning the paper in the prosthesis the user changes the approach of the good hand and wrist. This change in the wrist flexion is visible in the traces in (Fig.(6)a) roughly 50% of the way through the task.

with too few selectable motions may decrease the person's chances to be able to act independently when they grow older. Thus the ability to objectively measure the impact of a different design on the motion of the user is a useful method in determining the value of a design.

Assessment of Cognitive Load

Introduction

As the focus of study shifts towards the activity based measurements, what becomes important is to know how effective the person is with their prosthesis. There are a number of ways to measure this, and most centre on the activity itself, such as ADLs etcetera, but what is missing is; *how easy it is for the wearer to use the prosthesis?* This is clearly becoming more important as the prosthesis designs get more complex and potentially harder to operate. If a hand is harder to use, then there is a high probability it will not be used fully, or not worn at all. With existing designs of assessment, it is difficult to separate the design of the prosthesis, the controller and the ease by which a user can operate the device. Without this, it is hard to know if a person is slow or awkward in their usage of a device because of the mechanical constraints a particular design or the mental load the operation of the hand places on the person. While manipulation is not as stereotypical as walking, many parts of the control of grasp are still subconscious. Therefore, if the use of a prosthetic hand requires more attentional load on the part of the operator, knowing the extent of this effect on the prosthesis use is critical. To explore this difference we may use a technique known as *Dual Task measurement*.

Dual Task measures ask a person to perform an activity under test and then uses a second task to partly occupy the attention of the subject while performing both tasks. If the subject cannot perform both tasks at once to the same ability as they do either task, then the second task is having an impact on the first. The level of this impact can then be used to compare the result with a second primary task of interest. Basically, if it is hard to do two things at once, one tends to slow down one or both of them until one can. For Dual Task measures, the assumption is that this difference will result in a drop in performance in proportion to how hard it is. In the case of prosthetic control, one can imagine comparing the control of a prosthetic hand using pattern recognition with the more conventional, dual site myoelectric channels, to see which is harder to perform.

It is important to study this as the consequences of trying to use a prosthesis that needs too much effort to complete a task that might otherwise require little or no attention when performed using a natural limb, is that the person may understandably reject the limb. If it is harder to perform the task with a prosthesis, the user will tend not to undertake the task or will use a solution that is easier to achieve instead of the harder one. This may be awkward or encourage the sort of motions that are associated with over-use injuries, which should be discouraged.

There is no single protocol to test any situation. Different types of tasks (walking or grasping) need different methods to measure the cognitive load in each situation. The basic form of a dual task test is to have a subject perform two tasks, both separately and together, and see how much of each task can be achieved in each situation. Each task uses some of the subject's cognitive capacity. If the tasks are well chosen, it will be impossible to conduct both tasks to their full capacity simultaneously, but it is possible to perform them both, at a lower rate or skill. It is important for the tasks not to be so difficult that the subjects stop one task altogether and concentrate on the other. So it can be seen that the selection of the correct task is paramount to the success of the experiment.

The second task may impact on the first in one of two ways: The first possibility is if the same processing circuitry in the brain is used to perform both of the tasks. This would be achieved through a division of processing resources. In the prosthetic case, we are talking about the motor cortex and so, we need a second motor task. The important problem is that it is quite hard to find two independent motor tasks that can be performed together. For example; one pair of tasks might be to use ones hands to manipulate something, while walking. This is clearly two motor tasks, but it is tricky to work out how to do this practically. Cognitive psychologists use these tools to measure the way the brain instructs the body to move. If instead, two tasks use different processes (say motor and speech), but need to employ the same input/output resources there is still going to be compromises in the allocation of those resources. In fact in Psychology there is still debate as to which occurs in Dual Task and which is important. While this uncertainty may confound the interests of the cognitive psychologists, who wish to know more about the way the human central nervous system works, it is less of a concern to someone in prosthetics research. What we want to measure is simply how hard it is to perform the task and so the drop in performance between two controllers of interest will give us a gauge even if we weren't quite sure what was affected. So any task that exploits the limits of resource sharing is a useful task.

An example of a task might be for the subject to perform simple mental arithmetic while they move objects around with the prosthesis.

Scoring the Task

Scoring of the task is important. It is possible to imagine that different persons may use different strategies to get the task completed, and scoring calculation needs to be able to account for this. For example; if a person performed both tasks at once just slowing down each, this

would look like a different result to the alternative strategy where they perform each in turn and swap quickly between them. The amount each is degraded may well depend on how the subject chooses to divide the task up. Each subject is likely to be different.

Let us assume that each subject performs each task separately so that their full capacity of the task is known. Such tasks might be performing reaching and grasping for the primary task, so the count of the number of successful completions are then the primary score, *Ps*. The secondary task could be the number of successive subtractions of seven from a starting number of 93. This would be the secondary score Ss. Finally the two tasks would be performed together and the dual scores are measured *Pd* and *Sd*.

The cognitive load is then the impact of the primary on the secondary and the secondary on the primary:

$$Cg = ((Ps - Pd) + (Ss - Sd)/2) \quad (1)$$

Looking at only one measure (say the drop in primary task) will only capture part of the impact, hence the mean of the two impacts are taken.

To test the difficulty of one control task compared to another, the primary task would now change to the second different control method (such as the *same* manipulative task but using a different myoelectric controller format), then if it is easier to do the new task, then the reduction in score will be less because it will take less effort to operate and so will have less impact on the secondary. With a smaller score it would be possible to see that it is harder to use the first controller compared to the second.

Choice of Task

The choice of task is crucial to the success of the measure. If it is too easy and the subject can perform both without saturating their capabilities, it will not give a useable measure. If it is too hard, the task operation may stop altogether and so that too will not furnish us with a usable result. For example; if the second task was juggling balls while walking, then some people would be unable to juggle at all, while others would walk and juggle with little impact on either.

When walking is tested, it is possible to use a visual task, as one does not need to use much visual attention to walk. A popular task used in these circumstances is called the *modified Stroop test* (named after its designer) [41]. The subject has to identify the *colour* of the letters on a screen. The test being that the words that are *written* are the names of different colours. So the word *reads* a different colour to that it is written in. This exploits the fact that once one can read, one can't help reading, so one tends to read the word, rather than see the colour. It is possible to overcome this, but only with practice and leisure. If one is pressed to perform the primary task as well, it becomes harder.

The visual Stroop is hard to use for manipulative tasks. One needs to be able to see what one is doing with ones hands, so that the reading becomes too much a distraction. The alternative is an audio version of the Stroop. In this method, clearly male and female voices read the words; "'Man"' and "'Girl"' interchangeably and the listener has to determine and state the gender of the speaker, not the sense of the word. These forms of the words are used, as "'woman"' has more syllables than "'man"' while "'girl"' does not, this difference in length would give the listener a different cue to work from other than the sense of the spoken word. The trouble is that although some literature claims its usefulness [42], the authors' experience is that one can divorce ones hearing from listening and react entirely to *frequency* of the signal and not content, (something much harder do with vision). Thus in most situations it is too easy to perform and it does not impact on either test, so the scores do not change and the test tells us nothing. Of course, a person who is illiterate would be able to perform the visual Stroop perfectly and would only respond to the colour of the words.

There are other word games that one can perform to saturate ones cognitive processes. For example; one can listen to a sentence and report if it makes sense. The sentence is constructed from real words that are used in a standard and meaningful way, but don't, of themselves, makes sense[43].

For example:
The man crosses the road - makes sense, grammatically and logically

but
The man runs the wall - only makes sense grammatically

so it follows all the rules of English grammar and is nonsense otherwise. Thus, the listener has to comprehend the meaning and analyze the words as well as their greater meaning.

With many of these tasks, learning is a real problem. As with the other tests, if the tests are to be objective, then each time a person performs a test they should achieve the same score. For a dual task the reduction in score should be from the impact of the second test. If instead, each time the subject takes the secondary test s/he is getting better, then the order of the tests will make an enormous impact on the final score. So if both tasks are taken separately first and then together then the secondary score may be *higher* when done in combination. Again, experience with these tests does not show they are reliable enough to be used in conjunction with dual task measures.

One technique employed in the psychology community is mathematical tests. These are really effective as they are hard to learn to do better and they are very objective. For example, counting backwards by sevens is quite hard to do (while backwards in twos is much easier). With practice one will only improve slowly, so

the results across a morning experimental session won't change due to the repetition. Most people can't remember the answers, so a second test will have quite similar results to the first, which is important for repeatability of the test. If the person is good at this task, they are even less likely to get better between runs of a dual task experiment, as they may already may be as good as they can get. This sort of test is often used in cognitive psychology experiments to great success. However, the population of subjects that undertake these tests is seldom typical of the general public. They are often psychology undergraduate students, whose schooling is likely to have been better and more recent than the general population. This is not a problem for these experiments which aim to unpick some aspect of the human cognitive processes, but this test has a basic flaw for a test on the general population, including real users of prostheses. For although it is a good test, requiring effort and concentration and will allow the person to occupy much of their mind, it is mathematical. This is a skill fewer members of the general public possess than a similar sized population of numerate undergraduates. Indeed, the mere mention of numbers may terrify some people. Of course, if ones congregation is engineering or psychology students, one can assume a comfort and familiarity with mathematics that makes such tests possible. If we chose to use this on a more general population, then we may find that some of our potential subjects would not wish to participate in the experiment, being uncomfortable or even scared by numbers. The result of this is that we sub-select our population to the numerically confident members of the population, this reduces the validity of the experiment as we do not know if there is anything special about such a group of users that would change the overall result of the test. For example; such people may well be far happier with new technology and more ready to try new ideas and less likely to give up on a difficult prosthesis. If what we want to know is how easy the average user finds the prosthesis, this method will not get to them. Indeed, the population of users of prosthetic hands at any one centre can be so small, that to lose those who will not do the test (rather than try the hand) may drop the potential pool of users considerably.

An Example of Its Use in Upper Limb Control

An example of the development of dual task in use in manipulation tasks, demonstrates some of the pitfalls in this process.

A test was designed to measure the impact of using coordinated control of a shoulder/elbow prosthesis. The idea here was that the operator only needs to control the position of the wrist in space, they do not need to attend to the relative positions of the shoulder and elbow, this is taken care of by the computer controller. This was first tried in the 1970's at Southampton University [44, 45]. It required a heavy arm mounted on a post and large electronics to control it. By the turn of the century it was possible using the first ever bus-based prosthetic arm system; ToMPAW (Totally Modular Prosthetic Arm with Workability) [46]. This allowed a modified Edinburgh arm [47] to be controlled using the microprocessors and instructed by a joystick attached to the shoulder of the subject.

The protocol had two control formats:

1. Direct Control (DC) The X position of the joystick corresponded to shoulder elevation The Y position corresponded to elbow flexion
2. Coordinated Control (CC) The X position of the joystick corresponded to X position of the wrist The Y position corresponded to the Y position of the wrist

Subjects then drove the arm using their acromion and they were asked to place the wrist in one of three positions as quickly as possible (Fig.(7) and (8)). This was the primary task. One can imagine that there is a range over which the arm can move and targets were placed at the limits of this envelope. The two alternate primaries being the Coordinated Control or the Direct Control of each axis by an axis of the input joystick. The secondary task was to perform the multiplication tables. The subjects were final year engineering undergraduates.

Fig. (7) — Assessment of the control of a prosthetic shoulder. The movement of the acromion is mapped to the motion of the end of the arm, or the motion in the horizontal is mapped to the shoulder elevation and vertical motion to elbow flexion. The three spatial targets are shown ahead of the subject.

For an ideal test, the scores of each of the two tests should always be lower when both tasks are performed together, each having an impact on the subjects' ability. However some subjects were actually getting better at the task with practice, so that despite the distraction, their overall score did not change between tasks. The mean percentage change of the primary task is shown in Table (3). The Coordinated Control task is impacted less by the mathematics task than the Direct Control.

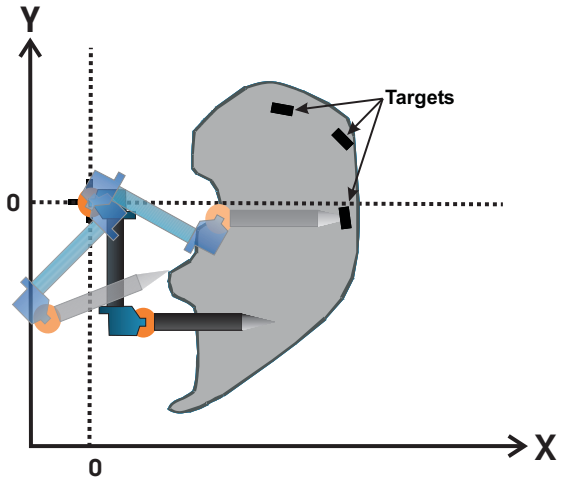

Fig. (8) — Envelope of the motion of the arm prosthesis.

The secondary task was more similarly affected, however it is necessary to consider the context of the two tasks. A mathematical task, even on this population was potentially harder for 21 year old engineers than those of a generation earlier who went to school before calculators became the norm and learning the multiplication tables had a greater emphasis. Naturally, it was still appropriate to expect these students to be comfortable with a mathematical task. Perhaps if their lecturers had been set the same task they may have found the multiplication easier. If this was the case then the result would be more dependent on testing their education than their ability to control the arm. Conversely, the younger subjects after years of computer games may have found the control of the arm simpler, therefore again testing age rather than ability. So it is clear that a simple test like this still has many dimensions to be considered.

Relative changes in scores for two tasks would remove some of this dependence on other factors. Thus Equation 1 is applied to the individual scores, which records the effect of the primary on the secondary and the secondary on the primary.

In all three areas the Coordinated Control had a smaller drop in performance. Application of the effect of the two tasks on each other brought a small, barely significant, difference in favour of the Coordinated Control. The smaller drop in performance means it was less demanding as a task. Also the standard deviations were smaller, therefore it was more uniformly achieved, compared with the Direct Control.

There were many reasons for the weak result:

Learning of the task is not uniform; it is hard to judge and it cannot be easily determined without many repeated measures of the task with each subject. This would make the experiment much longer.

It is not clear that the task really saturated the users' abilities. If the tasks used less than 50% of the cognitive resources, then they could service both tasks without a drop in performance, hence the impact of each is imperfectly measured.

The test measures the subjects kinaesthetic skills and mathematical skills which will be different for each subject, giving wildly different scores. Taking changes in scores does remove some of this effect.

Finally, current prosthetic control tends to have a single continuous motion input and one switch input so a real prosthesis controller would be one myoelectric channel and one bump switch, which would have been much longer in execution than the DC control. If this had been one of the control paradigms the results should have been much more distinct.

The role these techniques will play in assessment in the future, depends on the interests of the persons applying the techniques. In the examples shown here motion analysis was used to record aspects of the Activity domain. For all three examples the tests looked at the performance aspect of Activity. Had a new wrist design been added to the second tests, (page 50), then this would have been able to measure the capability of the device. Such tools are hard to employ to measure Participation, not only would it restrict the activity to a single room, but the presence of markers and cameras may inhibit natural behaviour, the very aspect that is supposed to be measured. It is possible that with lighter and more compact cameras, that activities may be measured at home, but with all such aspects of monitoring there are important concerns over invasion of privacy of the users.

CONCLUSIONS

Assessment is becoming a more important part of the process of developing and applying a prosthetic limb. Each part of the process requires different tools to measure the impact of choices in the design or application of the technology to prosthetic users. The tools have to be designed to supply reliable information that is easily comprehended by others in the same field, allowing the easy exchange of ideas and information about the devices and controllers that are in use or being developed. Older, established tools are simple measures using observations, timed activities or questionnaires of the users of limbs. With the new devices now on offer more sophisticated tools are possible. They use more computational techniques to observe the action electronically, or newer theories of the mind to attempt to measure the cognitive load required to operate the devices.

Important in all this is the need to identify a set of tools that would make comparing results between clinicians and centres possible. With a degree of understanding of the principles of design of the tools and the standardization of approach this will allow a broader knowledge and greater transparency between centres, techniques and programs.

Table (3) Mean of the changes in performance of the shoulder control experiment. DC = Direct Control, CC = Coordinated Control. The changes are percentage changes from when each task is performed separately to when the primary and secondary task are tried together. The combined score is the mean of the individual scores, showing the impact of one on the other, and *vice versa.*

	Movement Task		Mathematical Task		Combined score	
Primary Task	**Mean Change**	**Standard Deviation**	**Mean Change**	**Standard Deviation**	**Mean Change**	**Standard Deviation**
DC	15	20	67	15	41	15
CC	8	10	62	18	35	12

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CHAPTER 5

The Psychosocial and Biomechanical Assessment of Amputees Fitted with Commercial Multi-grip Prosthetic Hands

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Abstract: The scenario of upper-limb prosthetics is rapidly changing: innovative solutions are "moving out" from laboratories to be used by patients in the every-day-life. In particular, prosthetic hands are facing major changes, with the availability of multi-grip options. While these new technologies are *potentially* effective for patients, they are surely more expensive and complex in terms of mechanics, electronics and cosmetic covering, i.e. aspects that also determine an increase of maintenance costs. Since it is important to provide patients with effective components while keeping costs under control, technology assessment is crucial. In this framework, the aim of this Chapter is to provide an overview of some evaluation tools that were set-up at Centro Protesi INAIL to gain insight into the psychosocial and biomechanical aspects of upper-limb amputees using high-tech prostheses. A case study reporting the application of these tools is also presented, regarding a patient using the Otto-Bock Michelangelo hand. Results highlighted an increased satisfaction with the new multi-grip hand and, remarkably, the new prosthesis triggered a higher level of embodiment, with a mind-changing in the use the previous hand as well. Thanks to pleasant appearance and functional features of Michelangelo, the patient started to assume more natural gestures and postures also with the traditional myoelectric hand, reporting this different way of thinking the prosthesis as "*a fundamental step for an amputee*". Regarding the biomechanical assessment, the shoulder biomechanics was positively influenced by the availability of the lateral grip and by the overall hand shape, which allowed the patient to approach cylindrical and coin-shaped objects in a more natural way, limiting the shoulder compensatory movements. Overall, the assessment tools that we set-up provided a valid contribution for the systematic analysis of the changes taking place in the amputee due to the use of new technologies. The broad on-the-field experimentation will ultimately prove the validity of the approach.

Keywords: Amputation, Upper Limb, Prosthetic Arm, Hand, Myoelectric Control.

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INTRODUCTION

Upper-limb amputations in Italy are about 4.000 per year, i.e. about 25% of the total number of amputations reported by the National Healthcare Service [1]. The most frequent cause of amputation is traumatic, with percentages ranging from 53% to 69% depending on the source [2, 3]. Upper-limb amputations taking place during work are about the 75% of the total work-related amputations and are mostly fitted with a prosthetic solution by Centro Protesi INAIL, which is a specialized division of the Italian Workers' Compensation Authority for prosthetic/orthotic treatments and assistive technologies provision. Among the anatomic segments of the upper-limb, the hand is the most affected, i.e. about the 95% of cases at national level [1]. Statistics regarding work-related upper-limb amputations confirm this trend, with an 85% of the total.

From the prosthetic perspective, patient satisfaction remains a challenge, with statistics reporting a rejection rate or non-wear of about 25% [4, 5]. Functional limitations and ease of use can discourage amputee from the active use of the artificial arm, especially for more proximal levels. After a few decades of relative stagnation, innovations in mechatronics and science of materials are leading to a technological leap: new prosthetic components are becoming available *on the market* which might provide amputees with substantial benefits in terms of function and positive implications at the activity, participation and, more generally, quality of life level (see Chapter 4 for an introduction to ICF domains).

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These components are, in particular, myoelectric controlled, multi-articulated hands.

While these new technologies are *potentially* effective, they are for sure more expensive and complex in terms of mechanics, electronics and cosmetic covering, i.e. aspects that also determine an increase of maintenance costs. Since it is important to provide patients with effective components while keeping costs under control, technology assessment is crucial. Since prosthetic fitting requires the cooperation of a team of professionals with the patient and his/her family, the assessment must be multi-factorial. In the framework of the methodologies introduced in Chapter 4, the aim of this Chapter is to provide an overview of the evaluation tools that were set-up at Centro Protesi INAIL to gain insight into the psychosocial and biomechanical aspects of the amputees using high-tech prostheses, with specific reference to new prosthetic hands. A case study reporting the application of these instruments is also presented, regarding an amputee using the Otto-Bock multi-grip hand named Michelangelo.

PSYCHOSOCIAL ASSESSMENT

Psychosocial aspects play an important role in acceptation and proficient use of upper-limb prostheses, as highlighted in recent studies on the topic [6]. The question arises, therefore, if new high-tech components can have an impact on the acceptation of the amputation, social integration and, ultimately, perceived quality of life. The need for such kind of an assessment was further supported by the first experimentations with commercial multi-articulated hands carried out at Centro Protesi INAIL since 2007, from which we had the perception that amputees were experiencing a feeling of appeasement just operating the hand and seeing the fingers opening and closing in a "human-like" fashion.

For a systematic collection and analysis of psychosocial reactions and adjustments to new advanced prostheses, in comparison with standard myoelectric solution providing cylindrical tri-digital grasp only, we developed an assessment protocol. The protocol consists of a specifically developed semi-structured interview and a battery of self-report questionnaires, aimed at investigating the following variables:

1. prosthesis usage and satisfaction;
2. psychological reactions (anxiety, depression);
3. presence of psychological distress before the amputation and preview treatments;
4. psychosocial interventions after amputation;
5. expectations concerning the prosthetic device;
6. social integration and perceived social support;
7. social role of the prosthesis, with particular attention to its role in the presentation of self and in the relationship with others;
8. personality characteristics.

In particular, the protocol comprises 4 parts:

1. personal form that collects patient's personal information (age, gender, educational level, marital status, work, etc.);
2. semi-structured interview that consists of 3 sections, with multiple choice and open-ended questions: *Part I* collects information about variables related to amputation, use/satisfaction with prosthetic device, and motivation for technology use; *Part II* investigates the presence of pre-amputation psychological distress and previous treatments, along with post-amputation psychological emotional sufferance and possible interventions; *Part III* consists of questions related to other important psychological and social factors: social integration, perception of vulnerability, activity restriction and social avoidance, perception of others' reactions to the prosthesis, relationship with family and friends;
3. paper and pencil questionnaire on patients' expectations related to prosthetic device and believes about its impact on their life;
4. battery of standardized self-report questionnaires.

For what concerns this last part, the following six questionnaires were included, based on recent reviews that identify common psychosocial post-amputation challenges [6-8]:

1. State-Trait Anxiety Inventory – Forms Y1 and Y2 (STAI-Y) [9]
   The STAI-Y consists of two 20-item scales for measuring the intensity of anxiety as an emotional state (State Anxiety, STAI-Y1) and individual differences in anxiety proneness as a personality trait (Trait Anxiety, STAI-Y2). In responding to the STAI-Y1 items, subjects report the intensity of their feelings of anxiety "right now, at this moment", whereas responses to the STAI-Y2 items require subjects to indicate how they generally feel, reporting how often they have experienced anxiety-related feelings and cognitions.
2. Beck Depression Inventory-II (BDI-II) [10]
   The BDI-II consists of 21 items to assess the intensity of depression in clinical and normal patients. Each item is a list of four statements arranged in increasing severity about a particular symptom of depression, referring to criteria of the *Diagnostic and Statistical Manual of Mental Disorders* IV (DSM-IV): sadness, pessimism, fatigue, loss of interest, self-criticalness, changes in sleeping pattern and appetite, etc. The time reference for the response set is two weeks.
3. Multidimensional Scale Perceived Social Support (MSPSS) [11, 12]
   The MSPSS is a 12-item scale for measuring perceived social support of 3 specific sources: family, friends and significant other. MSPSS evaluates an important aspect of the broader construct of social support: perceived quality of relationships. Social support is believed to be a

relatively stable construct and to contribute a moderating influence between stressful life events and depression-anxiety symptomatology.

4. EuroQoL Questionnaire (EQ-5D-3L) [13]
   EQ-5D-3L is a standardised instrument for use as a measure of health-related quality of life. Applicable to a wide range of health conditions and treatments, it consists of a descriptive system that comprises 5 dimensions (mobility, self-care, usual activities, pain/discomfort and anxiety/depression) of 3 levels (1- no problems, 2- some problems, 3- severe problems) and provides a simple descriptive profile and a single index value for health status. It include also a visuo-analogue scale (EQ-VAS) on which patient indicate the perceived level of health status. The use of this instrument is recommended in studies evaluating the cost/effectiveness interventions.
5. Trinity Amputation and Prosthesis Experience Scales (TAPES-R) [14, 15]
   TAPES is a multidimensional measure theoretically and empirically derived to enable examination of the psychosocial process involved in adjusting to amputation and the experience of wearing a prosthesis. Recently, authors have suggested a revisited version of the TAPES, using both classical test theory and Rasch analysis [15]. TAPES-R comprises: three psychosocial adjustment subscales with a 4-point rating scale (General Adjustment, Social Adjustment, and Adjustment to Limitation); an activity restriction scale with a 3-point rating scale; two satisfaction subscales with a 3-point rating scale (Functional Satisfaction, Aesthetic Satisfaction) and a single item for overall satisfaction with the prosthesis. The questionnaire includes a final section formed by multiple choice and open-ended questions that investigate the experience of *phantom limb pain* (PLP) and *residual limb pain* (RLP), as well as other medical conditions not related to the amputation. Authors have highlighted the potential applicability and clinical relevance of TAPES with upper limb amputees [16].
6. Big Five Adjectives (BFA) [17]
   BFA is an instrument that belong to the tradition of the Big Five model, based on psycho-lexical approach to the study of personality. BFA consists of 175 adjectives, measuring five dimensions of personality (*E – Energy, A – Friendships, C – Conscientiousness, S – Emotional Stability, M – Open-mindedness*) and ten sub-dimensions (*Di – Dynamism* and *Do – Dominance*; *Cp – Cooperativeness* and *Co – Cordiality*; *Sc – Scrupulousness* and *Pe – Persistence*; *Ce – Emotions control* and *Ci – Impulses control*; *Ac – Openness to culture* and *Ae – Openness to experiences*). BFA includes a *Social Desirability Scale* (DS) in order to evaluate patient's tendency to provide socially desirable profile.

The protocol is intended for application in longitudinal studies. In our intentions, a baseline must be established before the application of the new prosthesis. After provision of the new system, intensive device-specific training (from 5 to 7 days full-time) and home-use for 3 months, the protocol is reapplied. Thus, pre-post treatment comparisons are possible. A 6-months and 12-months follow-up are then carried out to evaluate long-term patients' adjustment.

It should be noted that stable variables or trait measures (e.g. perceived social support, personality characteristics) are administered only during initial assessment because they are thought to be unchanged over time. The baseline measures of these variables may predict psychosocial outcomes at follow-up evaluations. Thus, follow-up assessments include an abbreviate version of the semi-structured interview, along with the designed battery of self-report questionnaires.

BIOMECHANICAL ASSESSMENT

In most daily activities, the kinematics of the upper-limb and trunk has the goal to position and orient the hand in such a way that the hand can effectively and efficiently reach/grasp an object and transport it to the desired target. When either the end-effector or one of the joints in the upper-limb kinematic chain is replaced by a prosthetic joint with intrinsic motor or sensorial limitations compared to the original body-part, compensatory movements are performed by the remaining joints. At present, the typical prosthetic solution for a transradial amputee does suffer from such limitations: the wrist rarely features the flexion-extension and it does not present the radio-ulnar deviation; the hand just features a tridigital grip pattern; none of the joints or part of the prosthesis offers a direct sensory feedback. Consequently, transradial amputees need to compensate these limitations and a possible reduction in elbow flexion due to the socket, with compensatory movements of the shoulder, trunk and neck. Despite the clinical observations, quantitative evidences about the actual compensatory movements and strategies adopted by trans-radial amputees to complete activities of the daily living (ADLs) are quite limited [18-21], and they generally suggest that compensatory movements depend on the specific task and that the whole body is involved.

In addition to being limited in number, four methodological issues exist in the literature. The first and most evident, is the lack of information about scapulo-thoracic motion, which is instead of great interest for the existing connection with shoulder pathologies. In particular, Ludewig and Reynolds [22] reported that alterations in the scapulo-humeral

rhythm, i.e. the coordinated movements of scapula and humerus when this latter is elevated, exist in pathologies such as shoulder impingement, rotator cuff tendinopathy, rotator cuff tears, glenohumeral instability, adhesive capsulitis and stiff shoulders. Both shoulders of an amputee are at risk: the controlateral side for overuse and the affected side for muscle weakness [23], which has been previously documented [24]. For these reasons, we think that scapulo-thoracic motion should always be analyzed, for instance with the acromion tracker described in [25]. Moreover, we think that the upper-limb assessment of an amputee should always include the measure of the scapulo-humeral rhythm during basic flexion-extension movements in the sagittal and scapular plane. With regard to the effect of new prosthetic hands, it would be relevant to understand if the availability of different gripping patters has an effect on the scapula and humerus kinematics, allowing the amputee to avoid positions known to be related to subacromial or internal impingement, i.e. related to rotator cuff tendinopathy; these positions are, in particular, humerus flexion & abduction with intra-rotation or abduction with external rotation.

The second limitation is the lack of information about head and neck kinematics: these motions can be essential to complete a task when upper-limb joints have a limitation in the range of motion (ROM) [18]. The existence of joint ROM restriction should be always assessed through basic, single plane, single joint motions. In particular, it seems important to identify if the socket is limiting the elbow maximum flexion and ROM.

The third limitation is the lack of information and standardization about the motion analysis protocol used, as opposed to recent recommendations [26]. Specific indications about marker placement on the prosthesis are not generally provided as well as the definition of the anatomical Coordinate Systems (CSs), i.e. the CS associated with each body segment whose relative orientation describe the segment/joint kinematics [26]. In particular, the elbow epicondyles, which are commonly used as anatomical landmarks to define the humerus anatomical coordinate system [27], cannot be easily identified, as the lay underneath the socket.

Finally, the forth limitation is the long-standing discussion about the lack of standardization of the activities to be tested, both in terms of *tasks selection* and *constraints in the execution of a specific task*, e.g. regarding body posture, objects to manipulate, starting position, ending position, timing. As discuss in [26] and also in Chapter 4, the answer lies in the hypothesis to be tested. First of all, it is important to define the ICF domain of interest: impairment or activity? We think that laboratory motion analysis is best suited for impairment assessment, while quantitative information about activity should be acquired by wearable technologies [28-30], that enable measurement outdoor, in the "real" every-day life.

Table 1 – Anatomical landmarks used for the definition of the anatomical/functional coordinate systems.

Abbreviation	Name
NB	Proximal aspect of the nasal Bone
CH	Mental Protuberance
C0	External Occipital Protuberance
IJ	Incisura Jugularis
PX	Xiphoid Process
C7	7th Cervical Vertebra
T8	8th Thoracic Vertebra
AA	Angulus Acromialis
TS	Trigonum Spinae
AI	Angulus Inferior
GH	Centre of Glenohumeral Head
EL	Lateral Epicondyle
EM	Medial Epicondyle
RS	Radial Styloid; *for the prosthetic side*: identified during the static calibration trial to replicate controlateral side.
US	Ulnar Styloid; *for the prosthetic side*: opposite to the RS.
M3	3rd Metacarpus; *for the prosthetic side*: just proximal to the 3rd finger knuckle.
V_{flex}	Direction of the elbow flexion-extension axis
V_{ps}	Direction of the forearm prono-supination axis
V_{flexW}	Direction of the prosthetic wrist flexion-extension axis

Replication of ADLs in the laboratory with the aim of gathering information about the *typical behavior* of a patient is, we think, unrealistic, since the fact that the subject is intensively monitored, unavoidably conditions the individual, who will tend to perform at his/her best. Consequently, the lack of standardization in the execution of a specific task does not bring additional information, but introduces variability in the data which make difficult to compare a subject over time (longitudinal studies) or a sample of subjects. With this framework, we support the use of tasks from the SHAP scale, which is a proposal brought forward in Chapter 4 as well. Additional tasks might be necessary, depending of the responses that the investigator wants to elicit in the subject, but these additional tasks should be equally clearly defined and should involve standard or easily replicable objects whenever possible, if manipulation is required. In the following sections, we describe the motion analysis protocol implemented at Centro Protesi INAIL for the

assessment of new prosthetic hands in transradial amputees. The description is based on the recommendations provided in [26].

Motion Analysis Protocol for Trans-radial Amputees

Segments and joints of interest

Both sides of the amputees are measured, to allow a within-subject comparison. The segments of interest are the head, thorax and, for both sides, scapula, humerus, forearm and hand. Neck kinematics comes from the relative orientation of head and thorax. Scapula and humerus attitude are referred to the thorax, following current standards.

Elbow flexion-extension is computed from the relative orientation of humerus and forearm, while prono-supination differs between the sound and the prosthetic side: for the sound side it is computed between forearm and humerus, while for the prosthetic side between hand and forearm.

Anatomical coordinate systems

The anatomical landmarks of interest are reported in Table 1 and the anatomical/functional CSs are defined in Table 2.

Table 2 – Definition of the anatomical and functional coordinate systems.

Segment	**Axes**
Head (HD)	$Y_{HD} = (NB - CH)/\| NB - CH \|$: longitudinal $X_{HD} = Y_{HD} \wedge (CO - NB)/\| Y_{HD} \wedge (CO - NB) \|$: medio-lateral $Z_{HD} = (X_{HD} \wedge Y_{HD})/\|\cdot\|$:antero-posterior Origin: *NB*
Thorax (THX)	$Y_{THX} = ((IJ + C7)/2 - (PX + T8)/2)/ \| (IJ + C7)/2 - (PX + T8)/2 \|$: longitudinal $X_{THX} = Y_{THX} \wedge (T8 - PX)/\| Y_{THX} \wedge (T8 - PX) \|$: medio-lateral $Z_{THX} = X_{THX} \wedge Y_{THX}/\|\cdot\|$: antero-posterior Origin: *IJ*
Scapula (SC)	$X_{SC} = (AA - TS)/\| AA - TS \|$: medio-lateral $Z_{SC} = (X_{SC} \wedge (AA - AI)/\| X_{SC} \wedge (AA - AI) \|$: antero-posterior $Y_{SC} = (Z_{SC} \wedge X_{SC})/\|\cdot\|$: longitudinal Origin: *AA*
Proximal humerus (H1)	$Y_{H1} = (GH - E)/\| GH - E \|$: longitudinal $Z_{H1} = \big(Y_{H1} \wedge (EM - EL)\big)/\| Y_{H1} \wedge (EM - EL) \|$: antero-posterior $X_{H1} = Y_{H1} \wedge Z_{H1}/\|\cdot\|$: medio-lateral $E = (EL + EM)/2$ Origin: *GH*
Proximal humerus (H2) (for internal-external rotation assessments only)	$Y_{H2} = (GH - E)/\| GH - E \|$: longitudinal $X_{H2} = (Y_{H1} \wedge Y_{PS})/\| Y_{H1} \wedge V_{PS} \|$: antero-posterior $Z_{H2} = X_{H2} \wedge Y_{H2}/\|\cdot\|$: medio-lateral Origin: *GH*
Distal humerus (H3)	$X_{H3} = V_{FLEX}/\| V_{FLEX} \|$: medio-lateral $Z_{H3} = X_{H3} \wedge (GH - E)/\| X_{H3} \wedge (GH - E)$: antero-posterior $Y_{H3} = (Z_{H3} \wedge X_{H3})/\| Z_{H3} \wedge X_{H3} \|$: longitudinal $E = (EL + EM)/2$ Origin: *E*
Forearm (F)	$Y_F = V_{PS}/\| V_{PS} \|$: longitudinal $Z_F = ((RS - US) \wedge Y_F/\| (RS - SU) \wedge Y_F \|$: antero-posterior $X_F = (Y_F \wedge Z_F)/\|\cdot\|$: medio-lateral $S = (US + RS)/2$ Origin: *S*
Hand – sound side and prosthesis without wrist flexion (HN)	$Y_{HN} = (S - M3)/\| S - M3 \|$: longitudinal $Z_{HN} = (Y_{HN} \wedge (US - RS))/\| Y_{HN} \wedge (US - RS) \|$: anterior-posterior $X_{HN} = (Y_{HN} \wedge Z_{HN})/\|\cdot\|$: medio-lateral Origin: *M3*
Hand – prosthesis with wrist flexion (HW)	$X_{HW} = V_{FLEXW}/\| V_{FLEXW} \|$: medio-lateral $Z_{HW} = (X_{HW} \wedge (S - M3))/\| X_{Hw} \wedge (S - M3) \|$: anterior-posterior $Y_{HW} = (Z_{HW} \wedge X_{HW})/\|\cdot\|$: longitudinal Origin: *M3*

Table 3 – Sequence of Euler angles for each joint/segment kinematics of interest

Joint /Segments	Euler Sequence (positive sign)
Head *relative to* Thorax	XZ'Y'' (flexion-abduction-internal rotation)
Thorax *relative to* Global CS	XZ'Y'' (flexion-left limping-internal rotation)
Scapula *relative to* Thorax	YZ'X'' (protraction-lateral rotation-posterior tilt)
Humerus *relative to* Thorax	*Mostly sagittal plane movements:* XZ'Y'' (flexion-abduction-internal rotation) *Mostly frontal plane movements:* ZX'Y'' (abduction-flexion-internal rotation)
Forearm *relative to* Humerus	*Sound side:* XZ'Y'' (flexion-carrying angle-pronation) *Prosthetic side:* XZ'Y'' (flexion-carrying angle-*dummy*)
Hand *relative to* Forearm	*Sound side:* YZ'X'' (*dummy* – radial deviation-flexion) *Prosthetic side:* YZ'X'' (pronation – radial deviation-flexion)

For thorax and scapula, definitions follow the ISB-ISG recommendations [27].

As proposed in [26], for the humerus two coordinate systems are defined: 1) a proximal CS (normally H1, H2 only for the assessment of the internal-external ROM only) to describe the attitude of the humerus relative to the thorax, i.e. for the computation of the scapulo-humeral rhythm, and 2) a distal CS (H3) to describe the elbow kinematics. The proximal CS follows the ISB-ISG recommendations, with the epicondyles "calibrated" when the amputee is not wearing the prosthesis, as further detailed in the Section "Marker-set and landmark palpation". The distal CS, instead, is based on the estimation of the elbow functional flexion-extension axis of rotation [26], which allows to minimize the kinematic cross-talk with the prono-supination.

For the forearm, the CS is based on the estimation of the functional prono-supination axis, obtained, for the sound side, from the relative motion of humerus and forearm during a pure prono-supination task. For the prosthetic side, instead, the functional axis is obtained from the relative movement of hand and forearm. In both cases, the functional axis is referred to the forearm segment. For the definition of the forearm CS, two anatomical landmarks are also required, namely RS and US. For the prosthetic side, it is proposed to ask the amputee to adjust the prono-supination so that the hands can join and the forearms can touch each other (named hereinafter "styloid calibration posture"). RS on the prosthetic side is then identified by replicating the position of the sound side RS on the prosthesis. The prosthetic side US is obtained as opposed to RS on the wrist circumference.

For the sound side hand, the CS (named HN) is based on anatomical landmarks. HN is used for the prosthetic side when the wrist does not feature the flexion-extension. When the wrist does feature the flexion-extension, instead, a functional axis describing this hinge joint is estimated using a functional method [26] and becomes the basis for the construction of the prosthetic hand CS (named HW). An exception to this approach is when the comparison of the radio-ulnar deviation is of interest between different hand models. In this case it is suggested to measure a static trial and perform the radio-ulnar deviation analysis on this trial using HN for both hands.

Joint or segment angles

Joint or segment angles are obtained following [26], as reported in Table 3. In particular, for the elbow joint the Euler sequence XZ'Y'' is applied to decompose the relative orientation of the forearm and humerus orientation matrices. The first rotation provides the elbow flexion-extension angle. The third rotation provides the prono-supination angle for the sound side and a constant value for the prosthetic side, since the prosthetic forearm does not feature the prono-supination (which is between hand and forearm). By applying the Euler sequence YZ'X'' to decompose the hand to forearm orientation matrices of the prosthetic side, the first rotation provides the prono-supination, the second the radio-ulnar deviation and the third the flexion-extension. The same sequence is applied to the sound side, but the first rotation reports a theoretically constant value (in the ideal case of absence of soft-tissue artefact [26]).

Marker-set and landmark palpation

Assuming that the system of measurement is an optoeletronic system, the marker-set is based on the CAST approach [31], with a few exceptions. Four

markers are positioned over an elastic band around the head to form a cluster of markers, and the anatomical landmarks of the head are calibrated with respect to the cluster during a static trial. Similarly for scapula and humerus, for which the clusters are positioned, respectively, on an acromion cluster and in the central part of the bony segment, slightly posterior. For the humerus, the cluster can be based on 5 makers, in case of visibility issues. The humerus epicondyles must be calibrated relative to the humerus cluster when the amputee is not wearing the prosthesis. For the estimation of the centre of the humerus head, a functional task as suggested in [32] is performed, and the method by Gamage et al [33] is applied [34]. For the sound side forearm, the cluster must be as close as possible to the wrist, while for the prosthetic side it must be proximal, outside of the prosthetic glove. For the hand, a three marker cluster is used, with markers positioned over M3 (Table 1) and on the middle of the 1st and 5th metacarpal bones. Two additional, small markers are place on the index and thumb tip, to have information about hand opening. All the object to manipulate should bring a marker, to allow the temporal segmentation of the activity (Fig. (1)).

Tasks

Firstly, a set of static trials are measured:

1. styloid calibration trial, as described previously
2. upright, elbow flexed 90°, humerus alongside the body perpendicular to the ground: this trial is useful to check the overall scapula posture and for the analysis of the radio-ulnar deviation of the prosthetic side.

Secondly, a set of functional tasks are required, to complete the definition of the CSs. These are:

1. a start arc task [32] with the humerus, for the estimation of the glenohumeral centre of rotation;
2. pure elbow flexion-extensions, to estimate the relative functional axes;
3. pure prono-supinations, to estimate the relative functional axes;
4. pure wrist flexion-extensions, to estimate the relative functional axes.

Functional movements 2)-4) should be repeated at least 5 times and the estimation of the axis is through the Woltring algorithm [35].

Then, a set of tasks to assess the condition of the elbow ROM and shoulder scapulo-humeral rhythm are performed, with the subject standing still in upright position:

1. a full ROM elbow flexion-extension;
2. a pure flexion-extension of the shoulder in the sagittal and scapular planes;
3. a pure humerus internal-external rotation with the elbow flexed 90°; the humerus anatomical frame named H2 is used for an accurate measure [36].

Also these movements should be repeated 5 times, to consider at least the central 3 repetitions.

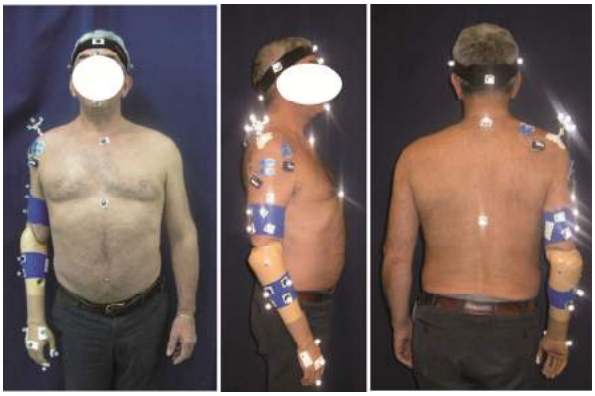


Fig. (1) – Marker set for the motion analysis based on an optoelectronic system. The acromion cluster was realized in Centro Protesi through rapid prototyping.

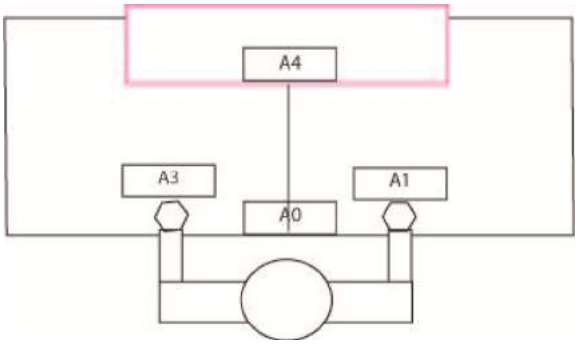


Fig. (2) – Marker set for the motion analysis based on an optoelectronic system. The acromion cluster was realized in Centro Protesi through rapid prototyping.

The subject is then asked to sit-down on a stood in front of a table, in the following reference position (RP):

- joint knee flexed at 90°;
- mid-line of the subject aligned with the mid-line of the table, placed in front of him/her;
- the distance of the thorax from the table fixed such that the subject can stand with the elbows flexed at 90°, neural rotation of the shoulders and the wrists aligned with the edge of the table.

Five technical areas (10x15 cm) are then marked on the table, considering the subject in the RP (Fig. (2)):

- A0 – frontal area (aligned with the midline), horizontally aligned with the table edge closest to the subject;
- A1 – lateral area, aligned with the right hand of the subject, just placed in front of the hand;
- A2 – in front of the subject, aligned with the body's midline at the maximum reachable distance of the arm under investigation (without moving the thorax);
- A3 – lateral area, aligned with the left hand placed just in front of the hand;
- A4 – frontal area, placed on a shelf, above A2; the height of the shelf must be chosen to aligning it with the mouth of the subject.

Once the RP is defined and the technical areas marked, the subject completes the following activities at least three times, at self-selected speed:

1. Jar task – Unilateral task
   Setup: Position the jar (SHAP object) over A2 and place a marker on the cap of the jar;
   Task: Starting from the RP, the subject reaches the jar placed over A2, carries it to A0, returns to RP, brings back the jar from A0 to A2 and returns to RP;
   Motivation: observe the reach-to-grasp, transport and release sequence of a cylindrical object (power-grip) of large diameter; shoulder kinematics and inter-joint coordination can be analyzed.
2. Carton pouring task – Unilateral task
   Setup: Position the SHAP board on the table, aligned with the midline, at a distance of 8 cm from the proximal edge. Position the carton (filled with 200 ml of water) and the jar (without the cap) in the designed areas (SHAP guidelines);
   Task: Starting from the RP, the subject first pours the water from the carton into the jar, brings the carton back on the board and returns to RP;
   Motivation: observe the reach-to-grasp & hold of a squared squeezable object. Shoulder compensatory movements can be analyzed.
3. Drinking task – Unilateral task
   Setup: Position a plastic glass (filled with water) over an area located midway between A0 and A2;
   Task: Starting from the RP, the subject brings the glass to the mouth, drinks the water, brings back the glass on the starting position and returns to RP;
   Motivation: observe the reach-to-grasp & hold of a small conic squeezable object. Since the amputee is asked to drink the water, neck and trunk compensatory movements can be analyzed, as they might be essential to complete the task while not spilling the water.
4. Tray task – Bilateral task
   Setup: Position the SHAP tray and the SHAP case over the table, following SHAP guidelines. Place 3 markers on the board making a triangular shape;
   Task: Starting from the RP, the subject reaches the board and moves the tray from one side of the case to the other, using both hands;
   Motivation: observe the strategy adopted to grasping a flat object lying on the plane, which would be better performed with a lateral pinch.
5. Disk tasks – Unilateral task
   Setup: The disks used in the standard Minnesota test are used for this task. Place a disk (with a marker on top) over A1 if the right hand is assessed (A3, if the left hand is assessed);
   Task: Starting from RP the subject completes three exercises:
   1) the subject carries the disk from A1 (A3) to A2, returns to RP, brings the disk back to A1 (A3) and returns to RP;
   2) the subject carries the disk from A1 (A3) to A3 (A1), returns to RP, brings the disk back to A1 (A3) and returns to RP;
   3) the subject carries the disk from A1 (A3) to A4, returns to RP, brings the disk back to A1 (A3) and returns to RP.

   Motivation: observe the strategy adopted to reach-grasp-hold-transport a flat small object, which requires a fine pinch.

Report

Range of motions are reported as max-min scalar values. Scapulo-humeral rhythm is represented as 3 angle-angle plots for each movement, reporting the scapula protraction-retraction, medio-lateral rotation and anterior-posterior tilting relative to the humerus-thoracic flexion-extention or ab-adduction. Functional tasks kinematics is reported as a percentage of motion, possibly with additional time normalizations for sub-activities (reaching, grasping, transport, return to starting position).

CASE STUDY: OTTO-BOCK "MICHELANGELO"

Centro Protesi INAIL is active in the clinical assessment of all commercial multi-articulated hands. In this context, the psychosocial and biomechanics protocol presented in the previous sections are currently applied for the assessment of the prosthetic hand named Michelangelo, developed by Otto-Bock (D). After an overview of this new technology, we present the results from a case study involving an amputee who had the opportunity to use Michelangelo for a period of 3 months.

Technical Features: Pros and Cons

Michelangelo, also referred to as "M hand" hereinafter, is an innovative multi-articulated hand proposed by Otto-Bock [37] (Fig.(3)).

Compared to the TouchBionics [38] iLimb (Chapter 2), the single digit is not the basic element of the hand, and the hand does not feature the interphalangeal joints (PIP), which are instead fixed in slight flexion to resemble the human hand when in relaxed position alongside the body. The long fingers articulate at the level of the metacarpophalangeal (MCP) joints and move at the same time, but only the index and middle fingers are responsible for the generation of force.

The thumb articulates at the level of the carpometacarpal joint (CMC), with active movements both for flexion-extension and ab-adduction (differently from iLimb, in which thumb ab-adduction

is passive, so far). The palm hosts the electronic control unit and the motor responsible for the flexion-extension of all fingers i.e. responsible for the hand grip force. A second motor, which actuates the thumb ab-adduction, is located inside the phalanx of the thumb itself. This second motor, being responsible only for the orientation of the thumb, can be of limited power. Due to the orientation of the MCP joints' axes of rotation (not in line), the long fingers feature an abduction movement during extension (in contrast to iLimb, where fingers move parallel to each other).

Michelangelo can operate in three modes, depending on the action of the thumb. At power-up, the hand assumes a "neutral" position, similar to the attitude of the hand when relaxed alongside the body. This is also the mode in which the hand returns automatically after a few seconds of inactivity (which can be set or disabled), provided that some safety conditions are met, including that no objects are hold. The second mode is named "opposition", since the thumb moves in addunction and realizes a grip in opposition with index and middle-finger. The third mode is named "lateral", since the thumb moves into abduction and performs a lateral grip with respect to the index. Since the maximum hand opening is 12cm, the opposition and lateral modes allow to perform both power grips (over wide objects, like and apple or a glass), and fine grips, e.g. to hold a fork or a pen. The hand can be stopped in a fully open position, which is useful to hold a tray or a plate. Finally, during opening and closing, fingers perform an ab-adduction movement, which allows to grasp or hold objects between them. Overall, therefore, Michelangelo performs 7 grasping patterns (Fig. (4)).

For what concerns the control, Michelangelo as iLimb takes advantage of standard myoelectric control: it is thus required at least a single EMG site. However, two are preferable to use, for instance, the co-contraction to switch between opposition and lateral modes.

So far we have been referring to Michelangelo as a multi-articulated hand. However, Michelangelo is more properly defined as a "hand-wrist system" (Figs. (3) and (5)). The hand is proximally connected to a passive joint, which implements the flexion-extension and prono-supination of the wrist. In prono-supination, the joint behaves as a standard passive prosthetic wrist. On the contrary, in flexion-extension it is either free or locked in predefined angles. When unlocked, a spring-loaded mechanism ensure its increased mobility around the neutral flexion while making it stiffer moving to the extreme of the range of motion, as well as the return to the neutral flexion when unloaded. The base of the wrist is not circular but elliptic, to resemble the human wrist and the system must be built in slight ulnar deviation, to simulate the human hand attitude when relaxed alongside the body.

The prosthetic system (hand plus wrist) has an overall weight of 600g. The opening and closing velocity allows a complete movement in 0.4 seconds. The gripping force is 100N in opposition and 70N in lateral, while it is 15N in neutral mode.

Given the dimension of the batteries, the best aesthetic results can be obtained with stump ranging between short and medium. With long stumps batteries should be placed laterally. It is important to notice that all Michelangelo components are not compatible with previous products by the same company, due to a substantial change in the communication system between components (named AxonBus).

The glove is in PVC and its durability will be a matter of attention since PVC is not as flexible as silicon and it might tear due to repetitive movements of the thumb.

Fig. (3) – The Michelangelo hand and wrist in a schematic drawing (courtesy of Otto-Bock) and the actual prosthesis, covered with a cosmetic PVC glove.

Fig. (4) – Gripping patterns of Michelangelo (courtesy of Otto-Bock).

Fig. (5) – The Michelangelo hand and wrist when donned by the patient. The wrist flexion-extension allows some natural static postures, including holding a sheet of paper and putting the hands on the waist.

Similarly, the choice of Otto-Bock not to implement the digits as single units imposes restrictions over the amount of gripping patterns available and this limitation also needs a specific evaluation.

Patient and Study Design

The patient we report about is G.S., a 50 year-old male, who gave his informed consent to participate in this study. G.S. underwent a work-related traumatic amputation of the right, dominant side at mid-transradial level when he was 20. He has been using a myoelectric prosthesis for more than 15 years. At the time the study initiated, the most frequently used prosthetic hand was an Otto-Bock Digital Twin, that will be refer hereinafter as the "standard" prosthesis or DT hand. G.S. was firstly assessed with the psychosocial and biomechanical protocol before starting the fabrication of the prosthesis for Michelangelo. Then, a temporarily prosthesis was developed, allowing G.S. to experience all the functions of Michelangelo during a 3 days occupational therapy. After the training, the definitive socket was realized, and G.S. used Michelangelo during his daily life as exclusive prosthesis for a period of 3 months. This included his usual job as office manager, house-keeping and hobbies. During this period he reported to prefer the wrist unlocked and the later grip. After the 3 months, the psychosocial and biomechanical assessment were repeated. The psychosocial protocol was also repeated after other 3 month, during which G.S. was asked to temporarily return Michelangelo to Centro Protesi and use the DT hand exclusively. Results for the psychosocial and biomechanical assessments are reported in the next two Sections.

Psychosocial Assessment

Results

G.S. was identified as an ideal candidate for the application of Michelangelo. His medical history does not include other relevant conditions except amputation. G.S. is a 'successful' prosthesis user: he wears his device all day doing a wide range of activities (e.g. sports, gardening, etc.). He has a broad social support system; he lives with his wife, close to his sons. He has an high school educational level and works as office manager in a company.

This section illustrates the data emerging from the evaluative sessions conducted by a psychologist at different time points. We start by describing the elicited data from interviews, then reporting scores of self-report questionnaires. In regard to the application of standard published instruments in single-case research, changes in response to treatment must be interpreted considering the sensitivity of the instrument and other sources of error. The integration between different sources of information allows us to draw a comprehensive picture of the patient.

During the initial semi-structured interview, G.S. appeared motivated and collaborative. He related in a calm and sociable manner, speaking openly about his limb loss. G.S. reported that he did not experience past episodes of significant psychological distress and he never sought psychological or psychiatric treatment after amputation. Due to the long-time interval and his

young age at the time of accident, he claimed not to be able to clearly distinguish the "before" and "after" amputation period. He seemed to deny or attenuate the sufferance faced in the early post-amputation phase. In effect, the whole interview reflected the more general tendency of G.S. to mainly focus on objective aspects, neglecting more intimate themes. He evinced a focus-problem oriented approach to life-threatening experiences (e.g. amputation it-self). He made several references to his perceived control over difficulties faced in life, increased by the adoption of active and practical coping strategies: "*I try to deal with aversive situations by finding external stimuli...*". G.S. described himself as an active person; he reported to be curious, and to love to meet new people and to have new experiences. In fact, patient showed high level of social integration, not merely limited to family and colleagues. He claimed not to have difficulties in relating with others due to being an amputee. Even when some people show offensive reactions to his amputation/artificial limb, he reported to feel himself self-confident and capable to deal with these situations. With respect to prosthesis, G.S. appeared globally satisfied with his traditional myoelectric hand, identifying the robustness and functionality as the main positive aspects of his device. However, at the same time, he highlighted the importance of the pleasant appearance of the artificial limb, comparing it to a "*beautiful article of clothing*". The fact that sometimes other people did not notice that he had a prosthesis emerged somehow as gratifying for him.

At three months follow-up, the interview evidenced stable level of emotional well-being and social integration. The application of Michelangelo seemed to had further improved the patient's everyday life. He claimed to use the prosthesis "*in a different way*". Especially, he highlighted three main aspects of the new device: a) the pleasant appearance of Michelangelo and its "natural movement" in social interaction; b) the functional aspect and, in particular, the involvement of the prosthetic hand in various activities that he previously executed exclusively with the controlateral limb; c) a feeling of increased reliability in grasping, holding and realising objects. As evinced also by scores of the *Satisfaction with the Prosthesis Subscale* of TAPES-R (see Table 4), G.S. appeared very satisfied with all aspects of the new prosthesis: the aesthetic as well as the functional ones.

After three-months testing period with Michelangelo, G.S. returned to use exclusively the DT hand. The psychosocial assessment protocol was then re-applied after three months, confirming findings of previous evaluative sessions. Besides, interview evidenced a relevant change in the patient's relation with the prosthesis: G.S. reported a "*mind-changing*" in the way to use the old device. In particular, he claims to no longer use the device as a "*support*", but to fully exploit its functionality. At the beginning, he felt the old prosthesis like a "*stick*" and he returned to the "old mentality". Then he "*suddenly*" realised that he could use DT hand as he used Michelangelo. Moreover, he seemed somehow to become aware of his previous attitudes in concealing the limb loss. He claimed that thanks to the pleasant appearance and functional features of Michelangelo, he started to assume more "natural" gestures and postures also with the traditional myoelectric hand. He stated that "*this is a fundamental step for an amputee*". This different way to use (and "*think*") the prosthesis appeared to be maintained regardless the type of technology used.

Self-report questionnaires scores were in line with findings emerged from interviews.

The patient obtained the results reported in Table 5 at the different time points.

Specifically, from the psychodiagnostic testing the following information emerged:

STAI – Scores showed low level of trait anxiety, with high levels of state anxiety possibly related to evaluation setting or prosthetic fitting phases.

BDI – Score revealed an absence of significant distress at the three evaluation time points. Scores below 85° percentile are considered in the average and do not have a particular clinical significance.

MSPSS – Patient showed elevated level of perceived social support, with high score in all support sources (*family*, *friend*, and *significant other*).

Table 4 – Results for the TAPES-R

Patient G.S.	**Initial Assessment (Digital Twin)**	**3-Months Follow-Up (Michelangelo)**	**6- Months Follow-Up (Digital Twin)**
Psychosocial Adjustment: General, Social, to Limitation	4.00, 3.25, 4.00	4.00, 4.00, 3.80	4.00, 4.00, 3.80
Activity Restriction	0.00	0.00	0.00
Satisfaction with Prosthesis: Functional, Aesthetic	2.60, 2.00	3.00, 3.00	2.60, 2.00
Overall Index of Satisfaction	*missing value*	10/10	10/10
PLP and RLP	Absence PLP/RLP	Absence PLP/RLP	Absence PLP/RLP

* The sum of the items for each subscale is divided by the number of the items that were deemed applicable or answered. *Psychosocial Adjustment*: high scores on these subscales are indicative of positive adjustment. *Activity Restriction*: high score on this scale is indicative of activity restriction (it not includes upper-limb tasks). *Satisfaction with Prosthesis*: high scores on these subscales are indicative of satisfaction with prosthesis. PLP = phantom limb pain; RLP = residual limb pain.

Table 5 – Results for the set of serf-report questionnaires

Tools	Initial Assessment		3-Months Follow-Up		6-Months Follow-Up	
	Scores	*Percentiles*	*Scores*	*Percentiles*	*Scores*	*Percentiles*
STAI-Y1	53	89	51	84	53	89
STAI-Y2	31	25	–	–	–	–
BDI-II	1/63	30	0/63	20	0/63	20
MSPSS	Total = 76/84 Family = 25/28 Friends = 24/28 Significant other = 27/28		–		–	
EQ-5D	Descriptive System 11121 Index = 0,796 EQ-VAS = 90		Descriptive System 11111 Index = 0,919 EQ-VAS = 80		Descriptive System 11111 Index = 0,919 EQ-VAS = 90	
BFA	*Principal Dimensions of Personality* *Energy* [55 ≤ T < 65]: He loves being with other people; he is glib in his actions; he is fairly loquacious and communicative; he has good leadership skills; he is an active, vigorous and energetic person. *Friendships* [45 ≤ T < 55]: He usually pursues its own interests, but he also knows how to meet other's needs and to be cooperative; generally, he tends to relate with others in a kind and courteous manner, even though he can adopt sometimes abrupt attitudes. *Conscientiousness* [55 ≤ T < 65]: He follows the rules and respect the deadlines, acting with order and method; he loves to work hard proving tenacious and persevering. *Emotional stability* [55 ≤ T < 65]: He is able to efficiently control his anxiety; he is rarely worried and he is able to remain calm and to be in balance. *Open-mindedness*[55 ≤ T < 65]: He shows good originality and creativity; he is generally informed and has a good cultural level. *Social Desirability* [45 ≤ T < 54]: Profile does not show falsification in a positive or negative way.					

EQ-5D – Patient reported only moderate difficulties in the pain-discomfort dimension at the initial assessment, with a total absence of impairment in the five EQ domains at the 3-months and 6-months follow-up. G.S. showed high weighted index of quality of life at the three evaluation time points, partially conform to visual-analogue self-rated health status.

BFA - Patient showed to be an active person, engaged in various activities. He feels comfortable with others; he shows friendly and courteous manners. He carries out his activities with sense of responsibility, commitment and accuracy. He has an optimistic vision of life; he is able to control emotional reactions. He has many and different interests; he is curious and creative.

Discussion

This case illustrate the critical role of psychosocial factors in post-amputation adaptation process. The availability of internal personal resources, such as adaptive coping strategies and personality traits (e.g. perceived internal control), and the presence of stable sources of support, allowed G.S. to well adjust with limb loss and to face different challenges related to being an amputee at different stages of his life.

In addition, this case evidence relevant aspects related to the embodiment of artificial limb. We extend the successful prosthetic fitting behind the acceptance of prosthetic device, considering the extent to which prosthesis is experienced as a part of the body: the degree to which the patient transforms the prosthesis from a *tool* to a *corporeal structure* over time [39]. With the application of Michelangelo, G.S. seemed to have further integrated the artificial limb in his self-image and self-concept. Patient reached through insight a new way to use and think the prosthesis regardless the type of technology used: the prosthesis became integral part of the user.

Finally, this phenomenological experience of prosthetic were associated to a more natural movement of the artificial limb.

Biomechanical Assessment

Results

Regarding the elbow pure movement assessment, the maximum flexion with the sound side was 134°, while it was limited to 108° and 115° with the socket for DT and Michelangelo hands, respectively. The range of motion was 135° for the sound side, while it was the same for the prosthetic side, namely 112°. The socket, therefore, limits the maximum flexion of the elbow, in a similar way between prostheses.

The graphical representation of the scapulo-humeral rhythm is reported in Fig. (6) for the sound and affected side (using the M hand). Substantial differences exist both in terms of shapes and range of

motion, for the same range of humerus elevation. In particular, the affected side features an increased retraction, lateral rotation and tilting, which are indicative of compensatory movements, that might be due to muscle weakness or a degeneration of the rotator cuff, compared to the sound side. This suggests the need for a dynamic ecographic screening.

Regarding the Jar task, scapula and humerus kinematic patterns are reported in Fig. (7). The whole movement was divided into 6 phases, which were time-normalized: 1) reaching, 2) holding and transport, 3) back to starting position, 4) reaching, 5) holding and transport, and 6) back to starting position. The percentages of movements were based on the sound side timing. The most noticeable differences are that with the DT hand the patient approaches the object in adduction and with a relevant posterior tilting Fig. (8a). On the contrary, with the sound and the M hand, the patient approaches the object in abduction and without almost relying on scapula tilting. Regarding the overall time taken to perform the activity, with the sound side the mean duration was 7s, while it was about 15s with both DT and M hands.

Regarding the Drinking task, Fig.(9a) reports the neck and thorax kinematics, Fig. (9b) scapula, humerus and elbow kinematics, while Fig (9c) the movement of the chin marker (CH) over the percentage of movement in the anterior-posterior and cranio-caudal direction.

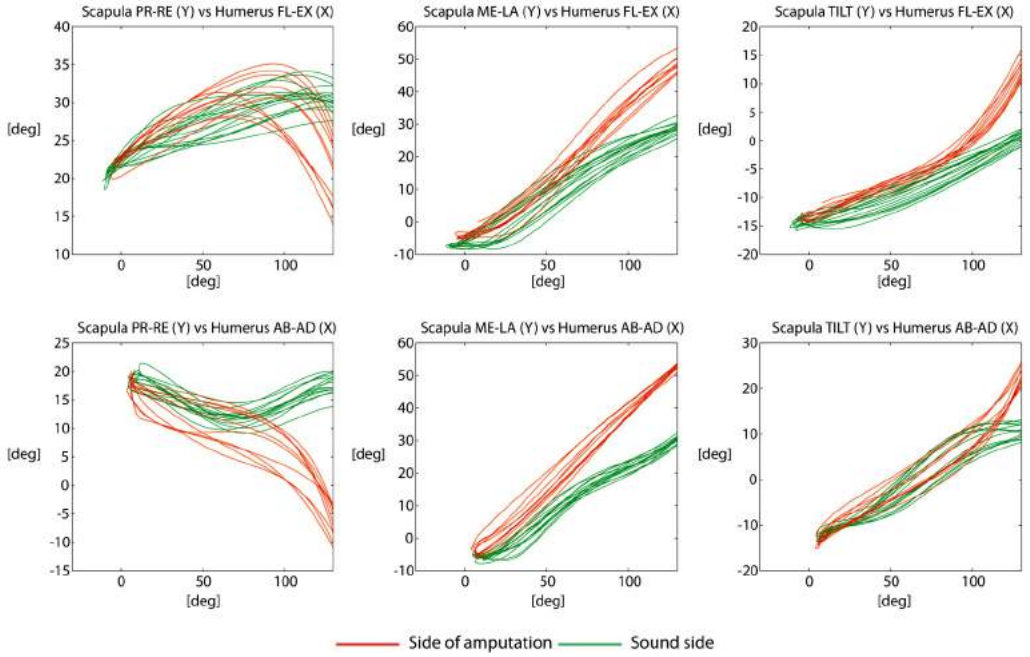


Fig. (6) – Graphical representation of the scapulo-humeral rhythm for the side with (red) and without (green) amputation.

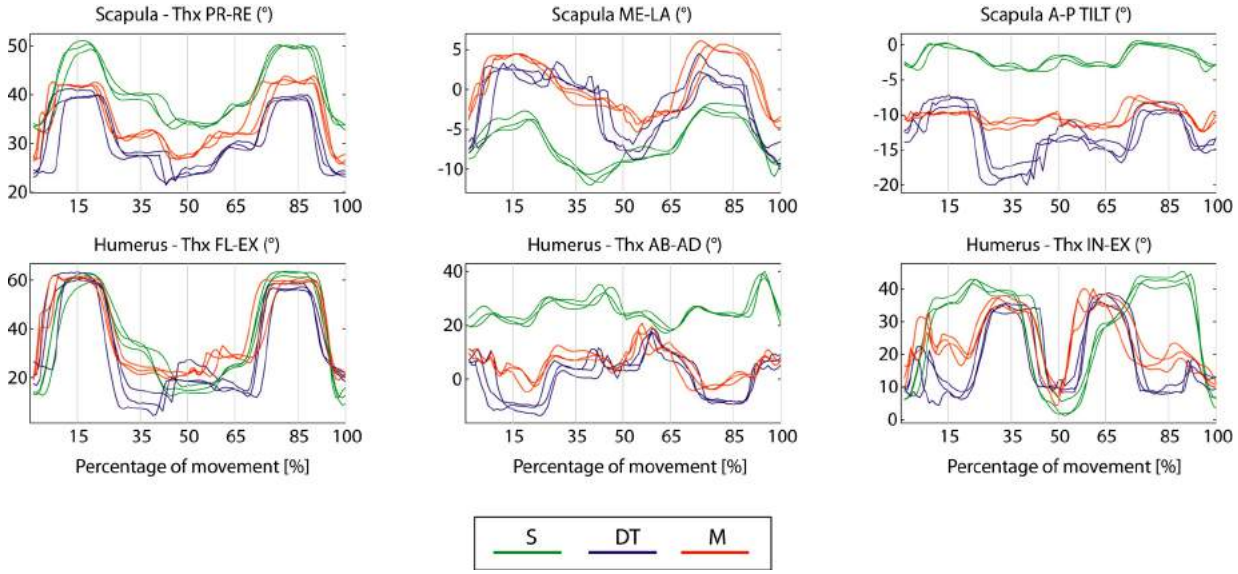


Fig. (7) – Kinematic patterns for the Jar task.

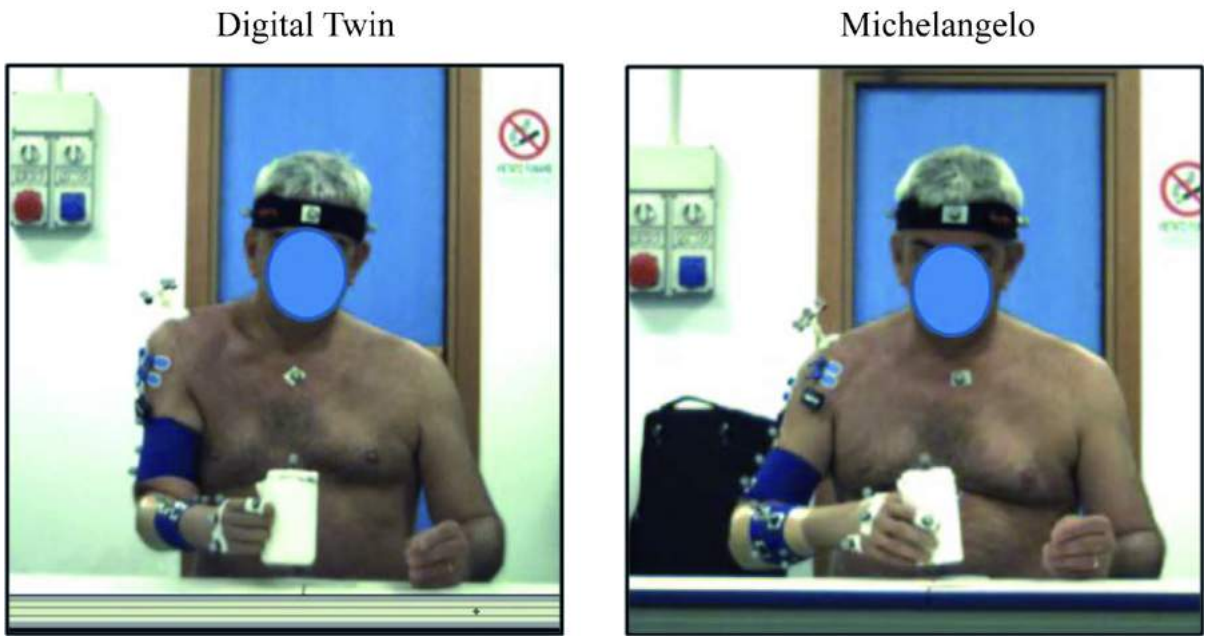


Fig. (8) Snapshots from the sagittal video tape showing G.S. during the Jar task, with the DT and Michelangelo hands. A relevant change in scapula kinematics and humerus adduction can be noticed.

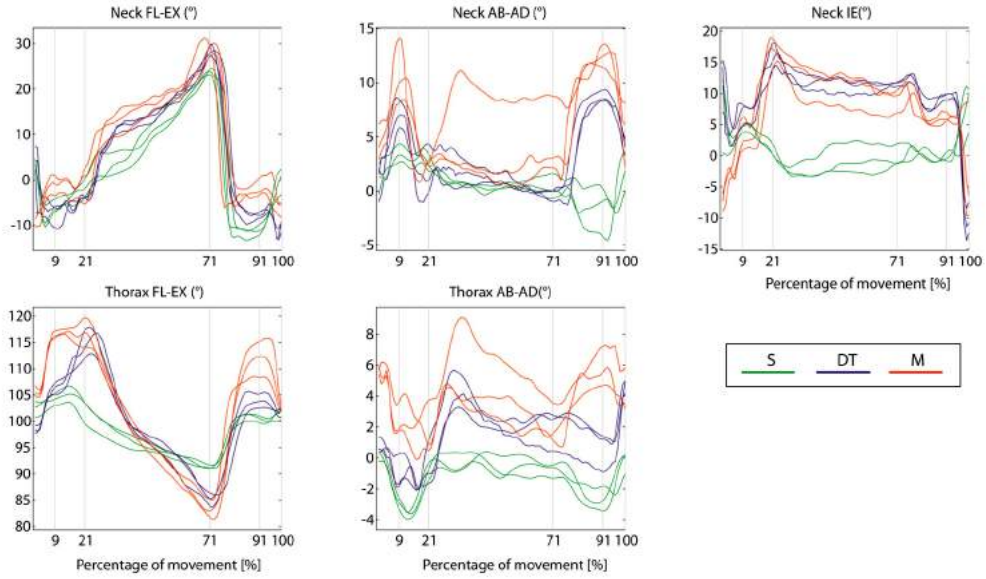
The task was divided into 5 phases, namely: 1) reaching the glass of water, 2) bringing it to the mouth, 3) drinking the water, 4) bringing back the glass to its original position, and 5) returning to the starting position. The percentages of movements were based on the sound side timing. The first relevant set of differences takes place when the glass touches the mouth (Fig.(10)). Due to limitations in elbow flexion and radio-ulna deviation in the prosthetic side (both with DT and M hands), the patient uses a different strategy to reach the target and not to spill water, namely moving into humerus adduction with scapula retraction, lateral rotation, anterior tilting and, most of all, head anteposition and thorax flexion. It is interesting to notice that neck flexion alone is not indicative of the strategy adopted by the patient. All other differences follow from this initial set.

Regarding the Carton pouring task, Fig.(11) reports the scapula and humerus kinematics, with the movement divided into 4 phases, namely: 1) reaching the carton, 2) pouring the water, 3) bringing back the carton to its original position, and 4) returning to the starting position. Percentages of movements were based on the sound side timing. Due to the lack of the active wrist prono-supination, the patient is forced to complete the pouring activity relying on increased scapula lateral rotation and humerus internal-external rotation compared to the sound side, both with the M and DT hand. However, with the M hand the amputee is able to reach the carton in abduction, similarity to the sound side, as seen with the Jar task. For what concerns the timing, the activity required about 11s for the sound side and about 20s with the prosthetic side.

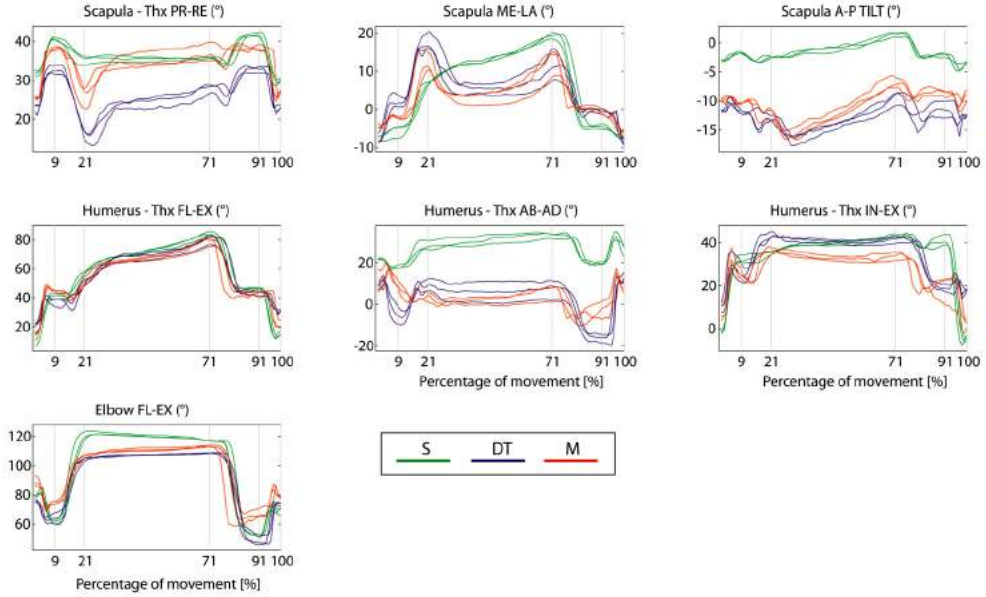
Finally, results from the Disk task are similar between the different variations of the task, and therefore we will present only the case of the movement between position 1 and 3, i.e. from one side to the body to the other, on the same level. Fig. (12) reports the kinematics for scapula and humerus. The movement was divided into 6 phases, namely: 1) reaching the disk in position 1, 2) grasping and moving the disk to position 3, 3) returning to the starting position, 4) reaching the disk in position 3, 5) grasping and moving the disk to position 1, 6) returning to the starting position. Percentages of movements were based on the sound side timing. For this task, the patient used the lateral grasping with the M hand. Due to this reason, major difference can be observed compared to the movement while using the DT hand (Fig.(13)). With this latter, the amputee is forced to perform the movement relaying on humerus flexion, abduction, internal rotation, and scapula lateral rotation. With the M hand and the sound side, the amputee avoids the subacromial impingement position. Timing is equally informative of the change between DT and M hands. With the DT hand the whole movement took11s on average, while it took 7s with both the M hand and the sound side.

Discussion

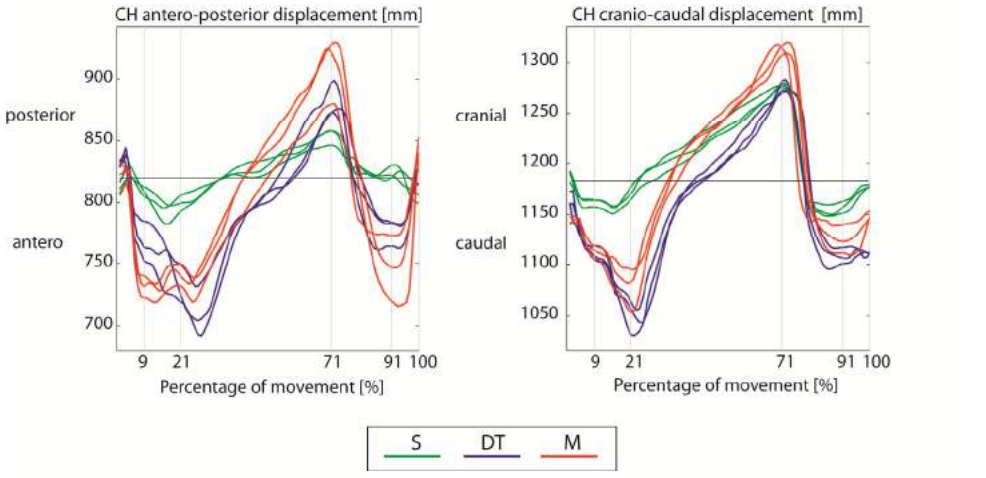
The biomechanical analysis brought in evidence differences between the upper-limb motion strategies while the patient uses DT, M hands and the sound side. Firstly, the amputee has to face an elbow flexion restriction, which most noticeably affected the drinking, with relevant compensatory movements. Secondly, the analysis of the scapulo-humeral rhythm revealed that the shoulder on the side of the amputation is probably affected as well. Further



(a)



(b)



(c)

Fig. (9) – Kinematic patterns of thorax, neck (a), scapula, humerus, elbow (b) during the Drink task. The trajectory of the CH landmark (c) is also reported.

Fig. (10) – Snapshots from the sagittal video tape showing G.S. while drinking, with the sound and prosthetic side. A very relevant anteposition of the head can be observed, as quantified by the trajectory of CH in Fig. (8).

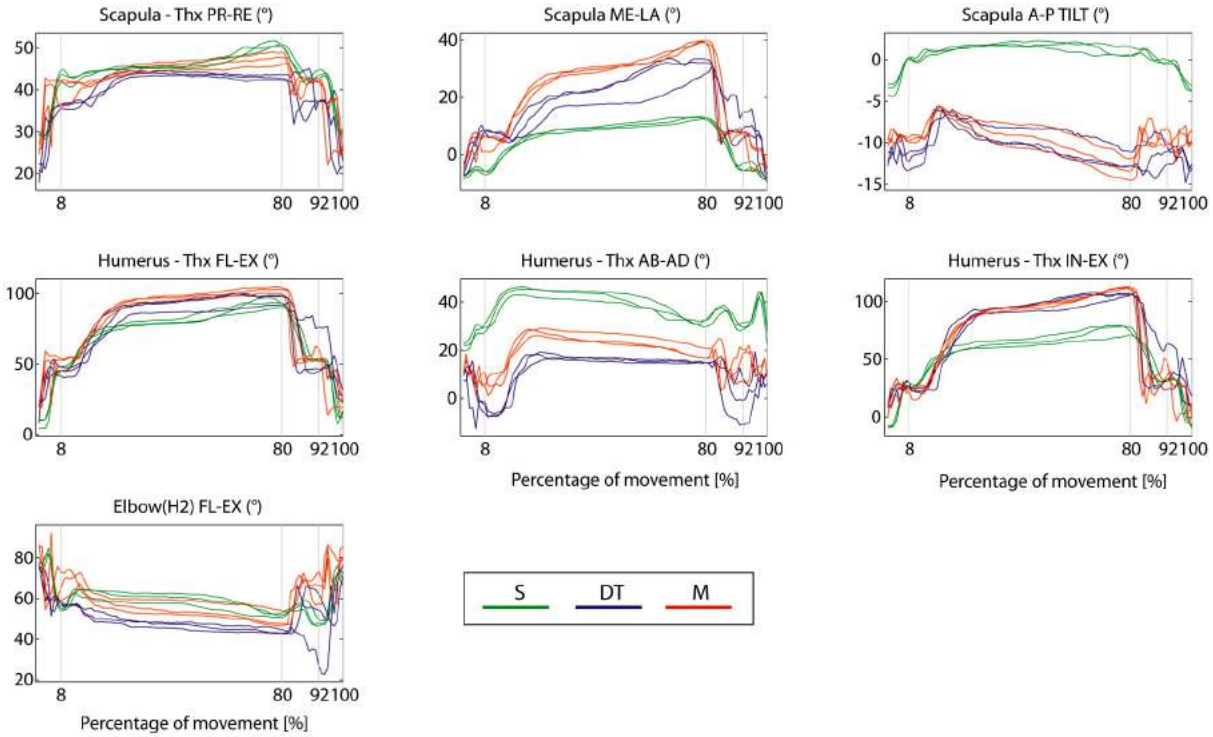


Fig. (11) – Kinematic patterns from the Pouring task.

image-based analysis might lead to the set-up of specific rehabilitation exercises and treatments. Similarly to the Drinking task, the Pouring task highlighted differences between the prosthetic side and the sound side, with advantages for M hand in the more natural approach to the object, as also highlighted by the Jar task. The Jar task also showed that M hand allows the amputee to reduce the scapula tilting motion, with the scapula working in a more physiological way. Among the tasks, the Disk showed the most relevant improvements of M hand over DT hand in comparison with the sound side. The Disk task reproduces a very common working-place activity. M hand can thus lead to a major advantage over DT hand in terms of preservation of the shoulder joint by avoiding subacromial impingement positions.

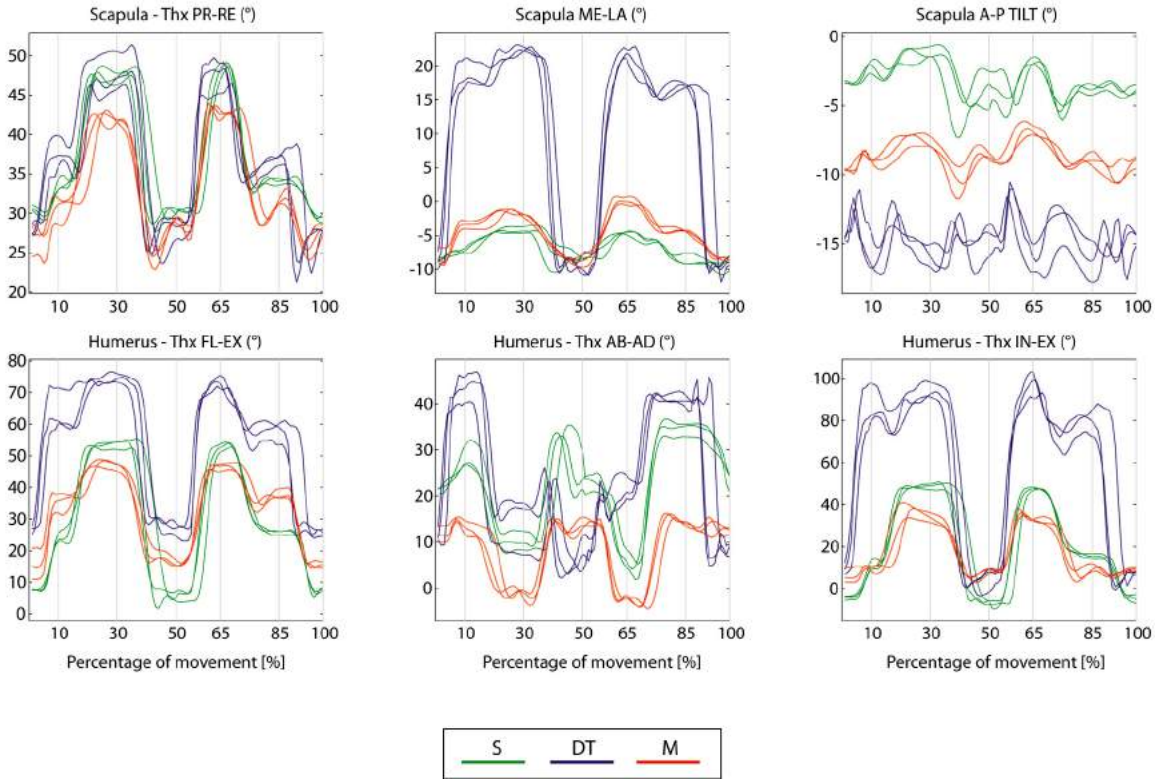


Fig. (12) – Kinematic patterns from the Disk task (from A1 to A3).

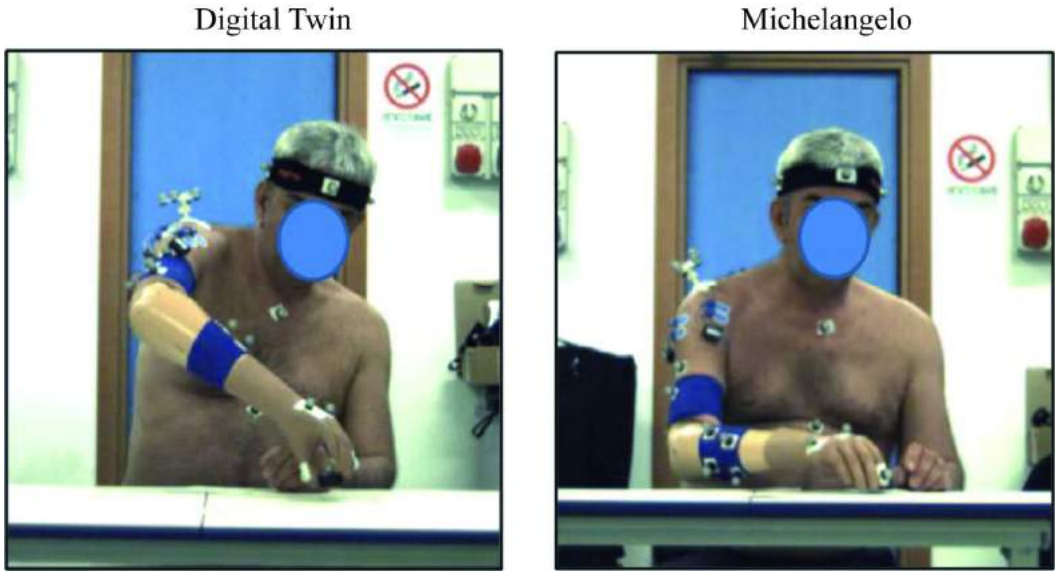


Fig. (13) – Snapshots from the frontal video tape showing the different approach following by G.S. to complete the Disk 13 task.

CONCLUSIONS

Great changes are occurring in the upper-arm prosthetic field: innovative solutions are coming out from the laboratories to be used by patients in everyday life.

In order to offer successful solutions to patients and justify higher costs, it is necessary to intensify the activities of technology assessment, based on scientific evidences, which is possible only through the planning and execution of appropriate clinical trials. INAIL is moving in this direction by means of the research projects in progress at Centro Protesi INAIL. In this perspective, we think that the psychosocial and the biomechanical assessments presented here will serve as a valid support to collect results systematically, in

different although crossing domains. Further activities will be focused toward an extensive application of the protocols to gain further insight about their completeness and applicability. The presented proposals are intended as a base for discussion within the community of professionals involved in prosthetics.

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CHAPTER 6

Design Solutions and Methods for Robotic Hands that Can Help Prosthetic Hands Development

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Abstract: After introductory considerations on the main functional and design differences between anthropomorphic hands conceived as robotic end effectors or as prostheses, this chapter presents two topics related to advances in robotic hand design that seem transferable to prosthetic hands, in order to increase their functional capability yet coping with specific constraints like simplicity, lightweight, cost effectiveness, robustness, etc. The development of a bio-inspired robotic hand, called UB Hand IV, based on an endoskeletal articulated structure, actuated by tendons and covered by a soft dermal-epidermal layer is briefly illustrated, in order to show the potential of its design solutions to be transferred into prosthetic hands. The first part of the chapter presents alternative design approaches for articulated joints and finger structures based on purposely designed compliant hinges. The basic problem of compliant hinges adoption in robotic structures, that is the limitation of secondary compliance effects, is analyzed and considerations about comparative metrics are proposed. Two hinge morphologies which show promising features are critically compared and pros and cons the production of fully integral fingers with compliant joints are discussed. The second part reports on the development of thin soft covers for robotic (and prosthetic) hands capable of strictly mimicking the actual compliance of human finger pulps. A design method, called by the authors Differentiated Layer Design (DLD), is reviewed and its potential for application on both robotic and prosthetic devices is underlined. Conclusions summarize the main aspects that encourage the transfer of the described results from the world of robots to that of human portable devices.

Keywords: Robotic/Prosthetic Hands, Large Displacement Compliant Joints, Soft Fingertips, Design Methods, Performance Indexes

INTRODUCTION

Defining the "optimal functionality" of robotic end effectors can be quite a challenging task as far as the possible applications of this devices span industrial robotics, humanoid robotics, rehabilitation medicine and prosthetics to name a few. Therefore, it is reasonable to think that solutions which are well suited to a single domain might not be readily taken as general guidelines.

As an instance, industrial manipulators are often equipped with basic grippers which are conceived so as to increase throughput and reliability [1] and are assumed to operate in structured environments. In this case, the system compliance, being a source of uncertainties and oscillations, is introduced when strictly necessary and the gripper is usually designed to be very fast and stiff in order to achieve highly precise position control. Similarly, enhanced manipulation and sensing capabilities, which carry a subsequent cost increase, must be carefully motivated by the application requirements. At last, anthropomorphism cannot be considered neither a plus nor a design goal but simply one possible solution whenever the gripper has to interact with "made-for-human" tools.

A different scenario arises when dealing with the fields of humanoid robotics and prosthetics. In such a case, the robotic end-effector is expected to provide high flexibility and adaptability to unstructured environment, ideally replicating the overall functionality of the human hand. Therefore, issues such as *dexterity*, *anthropomorphism*, *sensing capability* and *human-like motion* become fundamental. In fact, the human hand is not only a complex tool used to grasp and manipulate objects but also a sensory system which provides useful information about the physical properties of the surrounding (e.g. roughness, temperature, shape, etc [2]). Since the mid 80s, anthropomorphic robotic hands represent a mechatronic system which has been employed as a benchmark for non-trivial design solutions both at the mechanical level (kinematical structure, transmission and actuation system, joint design, mechanical properties at the contact interface) and in terms of the control architecture (sensory sub-

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system, sensory-motor coordination, control strategies for dexterous manipulation).

It is rather obvious to think of upper-limb prosthetics as a natural application of anthropomorphic robotic end-effectors. Nevertheless, some specific but fundamental problems arise concerning the technology transfer from the field of humanoid robotics to the field of prosthetic. First of all, it should be kept in mind that present technology is still far from being capable of fully replicating the human hand functions. Secondly, a set of psychological and environmental boundary conditions often convey the end-users of prosthesis (i.e. the amputees) to choose very simplistic devices (such as hooks or one degree of freedom (d.o.f.) mechanisms), switching to less functional but cosmetically appealing tools for social activities [3, 4, 5]. Therefore, the target of fully replicating the human hand functionality (as it happens in the field of humanoid robotics) might lead to unsuccessful products within the prosthetic industry. In the same manner, the use of design solutions which comes from the world of industrial robotic could provide enhanced reliability and cost effectiveness but dramatically lead to products which are not appealing in terms of aesthetics and acceptance.

Within this scenario, the objective of the present chapter is to discuss topics related to advances in robotic hands design that seem transferable to prosthetic hands, in order to increase their functional capability yet coping with their specific constraints. After a brief review of the state of the art concerning prosthetic devices, the discussion focuses on specific design solutions which have been adopted by the authors during the development of a robotic hand prototype called UBHand IV [6, 7, 8]. Solutions which are readily transferrable to the world of prosthetics are identified along with solutions which seem promising but in the need of further investigation. In particular, assembly simplification and the inherent compliance of the hand mechanical structure (both within its articulated structure and at the contact interface) represent the key features which are envisaged as a winning concept to be shared between the world of robotics and the world of prosthetics.

DESIGN APPROACH

The optimal prosthetic hand, ideally replicating the human hand, would possess full anthropomorphism and full dexterity. Note that, citing [9], the term anthropomorphism is related to external perceivable properties, and it is not a measure of what the hand can do. On the contrary, dexterity is related to actual functionality and not to shape or aesthetic factors.

At present, a variety of devices raging from purely cosmetic anthropomorphic tools to purely functional devices (e.g. hooks) with varying level of dexterity (e.g. *Otto Bock SensorHand*, http://www.ottobockus.com or *i-Limb*, http://www.touchbionics.com) can be found on the market. Concerning actuated prosthesis, body-powered or externally-powered prosthesis can be used. In the first case, the hook or the hand is attached to a harness and to a cable transmission which is controlled by the movement of the patient residual limb. In the second case, myoelectrically controlled actuators are introduced (most of the time small electric motors) in order to increase the dexterity of the device. Still, it is obvious that the number of actuators is limited by: 1) the number of independent myoelectric signals which can be extracted from the operator, 2) the increase in weight and size of the prosthesis, 3) the need of a suitable power source.

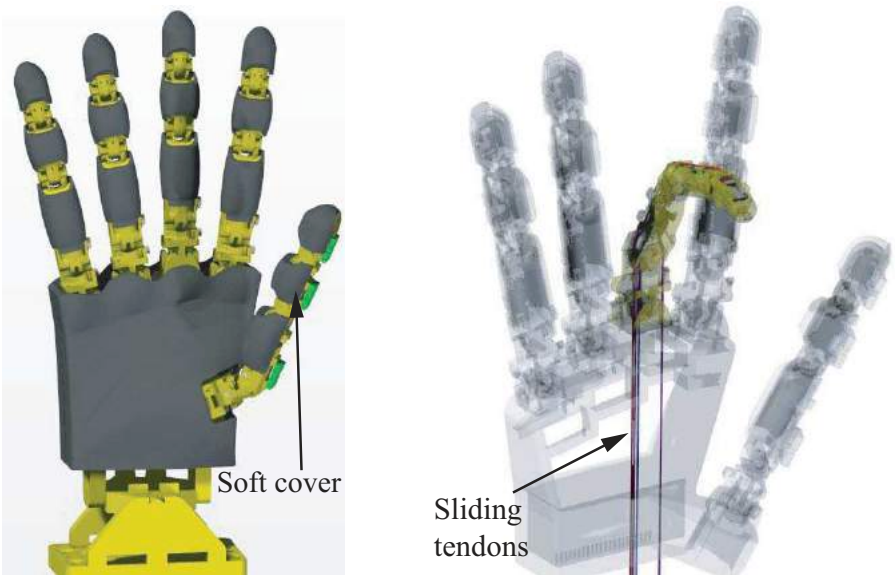
Even if the most successful prosthetic hand is maybe the aforementioned *Otto Bock SensorHand* (according to [5]) having one d.o.f and one actuator, late research has been mainly directed in the development and optimization of multifunctional underactuated mechanisms [10] possessing one or two actuators but several d.o.f.. The prototypes developed within the academic community (e.g. [3, 5, 11, 10]) and the recently commercialized *i-Limb* from *Touch Bionic* are clear examples of this philosophy. Naturally, beside the kinematical structure and the number of actuated/passive d.o.f., the level of dexterity of an artificial hand depends on the quality and type of sensori-motor system, embedded control system and signal processing at the interface between the device and its operator.

Clearly, the requirements of an optimal prosthesis would be many and mostly dependent on the user preference which are primarily dictated by his social, psychological and cultural background. Still, a survey taken in 1996 [12] shows that functionality is more important to the amputees than aesthetic, the level of dexterity being determined by the desire to autonomously performing “normal” activity of daily living. On the basis of previous design attempts [13], the following list of generic requirements seems a reasonable choice in the development of novel, better-behaved and more successful commercial devices:

- Anthropomorphism (including kinematics, contact interface properties, size) and cosmetic appeal.
- Reduced weight.
- Robustness and reliability.
- Cost effectiveness.
- Sufficient level of dexterity.

It should be noted that a “sufficient” level of dexterity is hardly quantifiable (see [9]). In fact, the trade-off between a fully dexterous device and a simple, reliable one is what truly differentiates one prosthesis from the other.

Taking a look into the wide field of humanoid robotics, several multi-fingered anthropomorphic hands have been developed in a number of research institutes all



(a) (b)

Fig. (1) — (a) Hand and palm soft cover. (b) Sliding tendons network

over the world (e.g. JPL Hand, MIT/Utah Hand, DLR Hand, UB Hand, Shadow Hand, see [14] for a critical review).
Some of these hand prototypes possess rigid and hard structures and complicated sensori-motor systems, design solutions being mainly based on non-biologically inspired mechanics, with abundance of gears, pulleys, bearings and similar hardware. This "classical" approach leads to efficient devices yet very complex, expensive and often not sufficiently reliable.
An alternative design philosophy is to aim at simplification and, but not necessarily, at the investigation and reproduction of some biological features. In particular, the hand that is under development at the University of Bologna, the UB-Hand IV, has been inspired by the following driving issues [1]:

- To adopt an endoskeletal structure articulated by means of non conventional joints, sliding or compliant;
- To actuate the joints by means of remotely located actuators with tendon-based transmissions routed by sliding paths (*sliding tendons*);
- To exhibit surface compliance through a purposely designed soft cover mimicking the human dermal-epidermal layers;
- To reduce manufacturing and assembly complexity by systematic parts integration adopting proper advanced materials and technologies (e.g. polymers and additive manufacturing technologies like Fused Deposition Manufacturing or stereolithography [17]);
- To reduce weight and cost of the overall hand system, increasing its "affordability".

A general view of the present hand prototype is shown in Fig. (1). The main results that seem transferable to the design of a prosthetic fully anthropomorphic hand are:

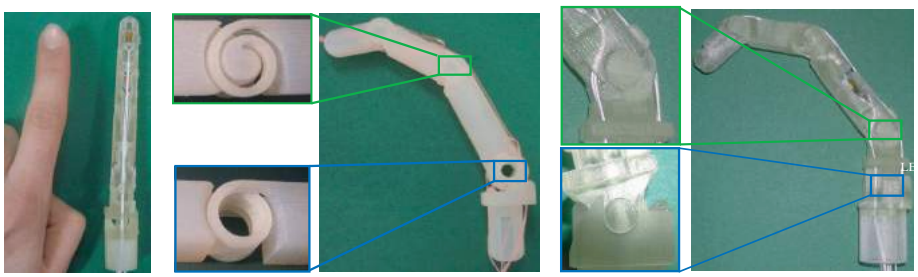
- A simplified endoskeletal finger structure, based on adoption of integral joints actuated by means of tendons;
- A distributed soft cover capable of replicating the human finger behavior without the need to integrate a cosmetic glove.

Note that tendon transmissions are easily integrable with either body-powered or externally-powered actuation methods (in the latter case, the actuators being located in the palm). However, in case of prosthetic application, the tendon network will be necessarily simplified, e.g. coupling the tendons of each finger and/or of the different fingers according to one of the many underactuation patterns well discussed in the literature [10].

TRANSFERABLE DESIGN SOLUTIONS: ARTICULATED FINGERS BASED ON COMPLIANT JOINTS

As previously stated, the main goal in the design of the finger joints is to search for the maximum achievable integration between the various components in the perspective of a structural simplification, allowing one-step

[1] Issues related to the actuation and sensing technologies and to the modeling and control of the hand will not be addressed. The interested reader can refer to [15, 16, 6].



(a) (b) (c)

Fig. (2) — (a) Fully integral finger: dimensions when compared to human finger, (b) finger with integrated compliant joint, (c) finger with pin joints integrated into the phalanx

monolithic manufacturing and consequent reduction of the assembly complexity.

In general, joint design is heavily dependent on the inseparable binomial material-technology. At present, different technologies and a wide range of materials (including lightweight metal alloys) can be used in order to produce the articulated finger structure in a single production step (*fully integral finger*). Such technologies include CNC machining, plastic molding (such as Shape Deposition Manufacturing (SDM)), Selective Laser Sintering (SLS), Fused Deposition Modeling (FDM), Stereo-Lithography (SLA) and Electron Beam Melting (EBM). Nevertheless, recent advances in the plastic materials technology suggest that the use of polymers might be well suited for the production of artificial hands once a lightweight, relatively economical solution is sought. For instance, plastic materials recently developed for SLA and FDM are beginning to offer acceptable performance and costs allowing the production of complex joint shapes (as it is remarked by the introduction of plastic grippers obtained though FDM within the robotic industry).

In detail, two concepts have been explored for the development of fully integral fingers, similar in size to the human finger, Fig. (2)a:

- Monolithic fingers with integral Compliant Joints (CJs) made of the same material of the phalanx structure (Fig. (2)b). It can be recalled that a CJ consists in a flexible region that provides displacement (rotational and/or translational) between two rigid parts through material deformation;
- Fingers with pin joints integrated into the phalanx body (Fig. (2)c) simply consisting in a plastic shaft which slides on a cylindrical surface.

In both cases, tendons are routed through a series of sliding paths which are obtained directly within the finger structure (sliding tendons, Fig. (1)b, (2)b, (2)c). A complete analysis of the tendon transmission modeling, control and material selection is reported in [18].

As for the integrated pin joints, in spite of the sliding contacts, the joint shows very good reliability. On the other hand, stiction and dynamic friction deteriorate the open-loop position control of the finger and can lead to mechanism locking as the contact pressure between the shaft and the hub increases (due to increased tendon traction). Because the authors think that monolithic finger structures best cope with the specifications of prosthesis design, in the following results will be presented concerning development of CJs. The benefits of CJs when compared to traditional kinematic pairs (like bearing couplings) include the absence of wear, backlash and friction still ensuring size and weight reduction.

Clearly, the use of *large-displacement* CJs [19, 20] within the hand endoskeleton is very attractive as long as it can allow the generation of very slender and light mechanisms that are more safe, robust to impact and better respect the goal of reproducing biological structures. In addition, the CJ can store energy during the actuation phase, restituting it during the return stroke. In such a way, as demonstrated in [2, 21], the joint can be actuated by means of one single-acting backdrivable actuator instead of a couple of agonistic-antagonistic ones (with obvious advantage in terms of weight and cost).

Basic Definitions About Compliant Joints

In practice, most applications still require CJs which are designed to provide one d.o.f. only. In such a case, the CJ is conceived in order to allow a *principal displacement* along a desired reference direction when subjected to a *principal load* (torque or force) acting along the same direction. For instance, Fig. (3) shows a corner filleted flexural hinge [22], i.e. an elastic structure which can act as a revolute joint by generating a rotational principal displacement θ_y under the principal load M_y. The y axis is called *compliant* (*sensitive*) *axis* [23, 24] and the ratio between θ_y and M_y is called *principal compliance*.

Secondary displacements (also referred to as parasitic effects in [22]) along the other reference directions may occur in real applications for two reasons: 1) the presence of *secondary loads* (also referred to as parasitic loads in [22]) acting along those directions 2) the presence of compliant *axis drift* (also referred to as parasitic motion in [25]). As for axis-drift, even in the absence of secondary loads, the point which models the center

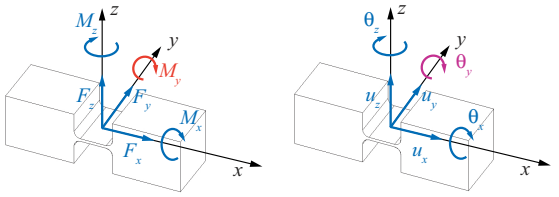
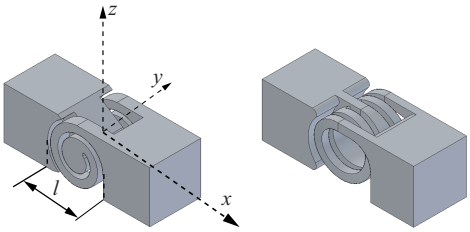


Fig. (3) — Principal and secondary loads (left), principal and secondary displacements (right).



(a) *SPIR*, Spiral joint. (b) *HEL*, Helical joint.

Fig. (4) — Large displacement rotational compliant joints

of rotation for rotational CJs or the axis which identifies the straight line motion of translational CJs can be subject to a spatial motion during CJ deformation.
In general, being undesired motions, secondary displacements should be avoided or at least minimized. In particular, it is possible to add rigid guides that would reduce parasitic effects. For instance, parasitic bending might be limited by adding lateral walls to each phalanx. Nonetheless, in such a case, sliding contact would occur at the wall contact interface, hence reducing the benefits of CJs when compared to traditional kinematic pairs. On the other hand, for a given load, the amount of secondary displacements is inherently dependent on the CJ morphology such that joints that behave similarly as to the principal displacement can exhibit quite different sensitivity to secondary displacements. Therefore, the design challenge is to determine the best CJ morphology which allows the desired principal displacement while minimizing parasitic effects.

Compliant Joints for Robotic Fingers

Commonly adopted CJs usually work in the small displacements range, while in robotic fingers the required range of displacement is much larger (revolute joints may be required to move up to 90°, that is more than ten times the maximum range required for normal applications). Generally, joints capable of large displacements can be easily obtained increasing the extension of the region occupied by elastic material subjected to an imposed deformation (e.g. very long slender beams). Still, such solution inevitably increases the sensitivity to undesired secondary displacements and represents a very critical factor against the application of CJs to robotic structures. In addition, small secondary displacements at joint level can be dramatically amplified at the end of serial articulated chains, especially in case of high number or relevant link lengths. Both these aspects (difficulty in design, modeling and evaluation of joint behavior together with high sensitivity to secondary movements) advice against the use of CJs as it seems they create more problems than they can solve.
This point of view can change whenever affordability is considered a value (as in the field of prosthetics) and task compatibility is demonstrated. As to task-compatibility, it means that the presence of un-desired displacements can be limited and, in any case, it does not compromise the capability to satisfactorily performing the task. Considering robotic and prosthetic hands with articulated fingers, during the approach-to-object trajectory, fingers behave as serial independent chains with very limited loads and the trajectory errors due to secondary displacements are quite acceptable; after the application of the contacts, the whole fingers-object system can be conceived as a parallel structure [21] where fingers contribute to the overall grasp stability and robustness according to their different placement with respect to the external load, thus mitigating the effects of the single joint secondary displacements.
As for the choice of CJ morphology,various types of CJs have been proposed in the literature and are classifiable in terms of amount of displacement (small [22] or large [19]), number and type (rotational or translational) of d.o.f. [26], adopted materials (multi- [27, 28, 3] or mono-material [29, 30]) and morphology (for instance, notch type CJ or leaf spring type CJ [31]).
In the robotics literature several examples of use of CJs in robotic structures (especially hands) can be found [13, 5, 3]. The applications were presented mainly under the point of view of conceptual design, without detailed analysis of the joint behavior, problems of secondary displacements being most of the times ignored. Within the UBHand project, two large deflections CJs were purposely designed and tested, namely a spiral joint (SPIR, Fig. (4)a) and a helical joint (HEL, Fig. (4)b).
As a proof-of-concept prototype, both joints have been employed in the realization of the compliant finger depicted in Fig. (2)b (in the present implementation HEL joint connects the proximal phalange to the palm whereas SPIR joint connects medial and distal phalanges). For given overall dimensions, the CJ morphologies have been designed on the basis of intuition, in order to maximize the extension of elastic material subjected to bending. Then, each CJ have been sized in order to achieve:

1. The same range of motion (i.e. $\pm 45°$ rotation) before limit stress;
2. The same overall dimensions. It is assumed that the available space to host the joint is a 10 mm side cube;

3. The same principal displacement under the application of the same principal load (i.e. the same principal compliance, $C_{\theta yMy}$);

In addition, note that the same material (with the same E/Y ratio) and production technology [2] have been used. Owing the aforementioned similarities, it can be hard for designers to select between joint SPIR and joint HEL without a deeper study of their behavior for what concerns secondary displacements. In order to overcome such difficulties, a comparison metric becomes necessary. In particular, the comparative evaluation of different CJ morphologies can be envisaged as the last step of an iterative design procedure once multiple conceptual solutions are available.

Evaluation of the CJ Kinematical Behavior

Once defined concurrent joint morphologies, with the same nominal performance in terms of principal compliance, the big challenge is to identify that morphology that is more suitable for application on a robotic structure. Several criteria can be defined, but many authors [22, 27, 32] agree stating that joints with reduced secondary compliance should be preferred. Therefore, it is important to identify and compare revolute joint configurations suitable for *large* principal displacement but with reduced sensitivity to secondary displacements. A possible method is suggested in the following and is finally tested on the two proposed configurations SPIR and HEL.

In the field of validity of the effects superposition principle (which assumes linear elastic material and small deflections) the kinematical behavior of a CJ in 3D space can be deduced by analysis of its compliance matrix **C**: given the vector of loads acting on a reference point of a joint, the displacement vector along the reference directions **x**, **y**, **z** is expressed by Eq. 1.

$$\Delta\mathbf{u} = \mathbf{C} \cdot \Delta\mathbf{f} \tag{1}$$

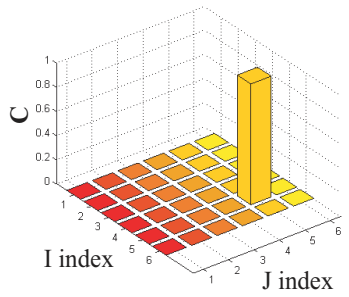
$$\Delta\mathbf{u} = \begin{bmatrix} \Delta u_x & \Delta u_y & \Delta u_z & \Delta\theta_x & \Delta\theta_y & \Delta\theta_z \end{bmatrix}^t$$

$$\Delta\mathbf{f} = \begin{bmatrix} \Delta f_x & \Delta f_y & \Delta f_z & \Delta m_x & \Delta m_y & \Delta m_z \end{bmatrix}^t$$

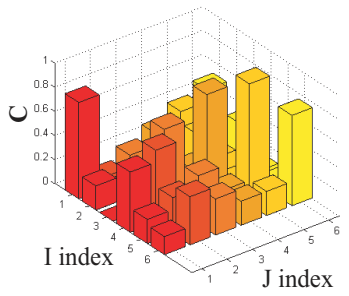
$$\mathbf{C} = \begin{bmatrix} C_{11} & C_{12} & C_{13} & C_{14} & C_{15} & C_{16} \\ C_{21} & C_{22} & C_{23} & C_{24} & C_{25} & C_{26} \\ C_{31} & C_{32} & C_{33} & C_{34} & C_{35} & C_{36} \\ C_{41} & C_{42} & C_{43} & C_{44} & C_{45} & C_{46} \\ C_{51} & C_{52} & C_{53} & C_{54} & C_{55} & C_{56} \\ C_{61} & C_{62} & C_{63} & C_{64} & C_{65} & C_{66} \end{bmatrix}$$

where $\Delta\mathbf{u}$ is an incremental displacement vector composed by 3 incremental translations (Δu_x, Δu_y, Δu_z) and as many incremental rotations ($\Delta\theta_x$, $\Delta\theta_y$, $\Delta\theta_z$), $\Delta\mathbf{f}$ is a perturbation wrench composed by 3 incremental forces (Δf_x, Δf_y, Δf_z) and as many incremental torques (Δm_x, Δm_y, Δm_z). The ratio between any secondary load variation and any secondary displacement variation can be defined as *secondary compliance*.

[2] E indicates the material Young modulus, Y indicates the material yield strength.



(a) ideal joint



(b) real joint

Fig. (5) — Graphical representation of compliance matrix of ideal (a) and real (b) revolute joint. Index i for rows and index j for columns

An ideal revolute CJ, which is expected of pure rotation along the principal direction even in presence of secondary loads, will present a compliance matrix **C** where only the term $C_{55} = C_{\theta yMy}$, corresponding to the rotation around the principal axis y under the action of the principal load M_y, is finite, being all the others null. On the contrary, a real joint will exhibit finite values of compliance coefficients also along the other directions. A graphical 3D bar representation of compliance matrix **C** for ideal and real revolute joints is shown in Fig. (5).

The properties of a compliant structure to maintain an high compliance in one direction and an high stiffness in every other directions is defined as *selective compliance* [22]. This property, well described by the entries of the compliance matrix defined in Eq. 1, can be used as a comparison criterion.

As for the actual **C** calculation, the analytic form of its entries is known concerning simple beam-like structures [33]. In any other case (i.e. general CJ morphologies) the calculation might be carried out by means of Finite Element Analysis (FEA). In addition, it should be noted that the compliance matrix is not frame invariant [34, 35, 23, 24] and its derivation should be performed

at a reference frame having one axis coincident with the compliant axis and located on the point envisaged as the CJ center of rotation (see [35] for details).
Still, being a differential operator, the compliance matrix measures a local property which depends on the joint's configuration. Therefore, it is clear that **C** cannot be directly used as a comparing metric. First of all, because a unique compliance matrix is not enough to describe the behavior of a large displacement joint in its whole workspace, W_s. Secondly, because the elevate number of the elements composing **C** do not allow to easily compare different CJs. Hence, with the purpose of defining a global CJ comparison method, let define:

- The CJ workspace, W_s, as the principal displacement range which can be achieved by the joint before limit stress when subjected to principal load of adequate intensity;
- An equivalent CJ moment of inertia as

$$I^* = l/EC_{\theta yMy} \tag{2}$$

  where $C_{\theta yMy}$ indicates the CJ principal compliance.

In order to reason on dimensionless quantities, displacements and lengths are normalized by the overall joint length l (refer to Fig. (4)), forces by EI^*/l^2, and moments by EI^*/l such that the normalized compliance matrix $\widetilde{\mathbf{C}}$ has principal compliance $\widetilde{\mathbf{C_y}} = 1$.
The CJ comparison is then carried out *via* the following steps:

1. Discretization of the CJ workspace into a finite number, N, of CJ configurations. For instance, assuming SPIR and HEL joints are capable of 90° rotation (±45°) and $N = 5$, the workspace of the joint can be divided as in Fig.(6), 0° representing the joint undeformed configuration.
2. Evaluation, be means of FEA, of a normalized, dimensionless local compliance matrices, $\widetilde{\mathbf{C}}_\mathbf{k}$, in each joint configuration. The matrix $\widetilde{\mathbf{C}}_\mathbf{k}$ is calculated *via* the following steps:
   - A fraction of the maximum principal load m_y is applied to reach the k-th configuration where $k = 1,..,N$.
   - A small variation of one secondary load component is applied while maintaining the principal load previously set. Then the generated displacements $\Delta\mathbf{u}$ (3 translations and 3 rotations) can be measured and used for the calculation of the six components of $\widetilde{\mathbf{C}}_\mathbf{k}$ which are related to this particular loading condition (i.e. k-th CJ configuration). A load variation is said to be small if it generates a small displacement [36].

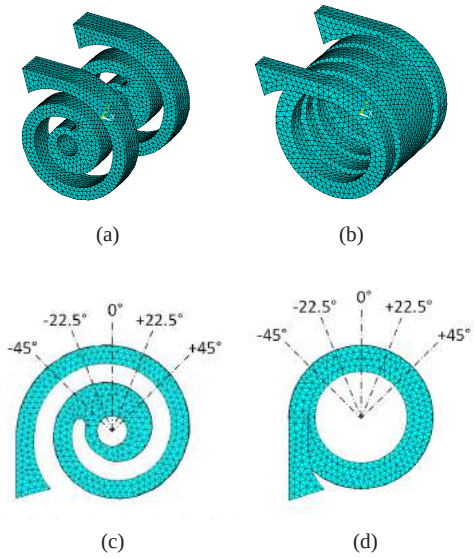


Fig. (6) — FEA models and workspace discretization: *SPIR* (a), (c) and *HEL* (b), (d) respectively

- The procedure is repeated for each secondary load component allowing to calculate the whole joint's compliance matrix in the k-th configuration.
  The normalized local compliance matrix is then split into two sub-matrices $\widetilde{\mathbf{C}}_\mathbf{Rk}$ and $\widetilde{\mathbf{C}}_\mathbf{Tk}$ containing respectively the coefficients relative to angular and linear displacements along the reference directions.

$$[\widetilde{C}]_k = \begin{bmatrix} \widetilde{C}_{Tk} \\ \widetilde{C}_{Rk} \end{bmatrix} \tag{3}$$

$$[\widetilde{C}]_{Tk} = \widetilde{C}_{i,j}, \qquad i = 1,..,3; \;\; j = 1,..,6.$$

$$[\widetilde{C}]_{Rk} = \widetilde{C}_{i,j}, \qquad i = 3,..,6; \;\; j = 1,..,6.$$

- Evaluation of two Local Performance Indexes (LPIs) which characterize the joint performance in each configuration. The proposed indexes are defined by means of the weighted Frobenius norm [36] that is

$$||\mathbf{A}||_F \equiv \sqrt{\frac{1}{n}\sum_{i=1}^{m}\sum_{j=1}^{n}|a_{ij}|^2} \tag{4}$$

  which intuitively indicates *how large* the entries of a generic $m \times n$ matrix **A** are.
  Specifically, a rotational LPI, I_R and a trans-

lation LPI, I_T are defined as:

$$I_{Rk} = \|\widetilde{\mathbf{C}}_{\mathbf{Rk}}\|_F \quad (5)$$

$$I_{Tk} = \|\widetilde{\mathbf{C}}_{\mathbf{Tk}}\|_F \quad (6)$$

A smaller LPI indicates a better local CJ behavior.

3. Definition and evaluation of Global Performance Indexes (GPIs) which summarize the overall joint performance over the whole workspace W_s:

$$I_{Rg} = \frac{\sum_{i=1}^{N} I_{Rk}}{N} \quad (7)$$

$$I_{Tg} = \frac{\sum_{i=1}^{N} I_{Tk}}{N} \quad (8)$$

A smaller GPI indicates a better global CJ behavior. Note that, small secondary rotations at joint level can be dramatically amplified at the end of serial articulated chains, hence the evaluation of I_{Rg} is usually more significant.

As for SPIR and HEL joints, due to the complex morphology, the calculation of the compliance matrices at each joint configuration has been developed by means of FEA, using the software *ANSYS Release 12.0*. The material adopted in these simulations, named *Fullcure 720*, is suitable for SLA and exhibits a Young module $E = 2870MPa$ and a Poisson ratio $\nu = 0.33$. The GPIs obtained from the analysis are reported in Tab. (1).

Table (1) — Global indexes for *SPIR* and *HEL*

	I_{Tg}	I_{Rg}
SPIR	$9.88 \cdot 10^{-3}$	$1.00 \cdot 10^{-2}$
HEL	$1.59 \cdot 10^{-2}$	$1.60 \cdot 10^{-2}$

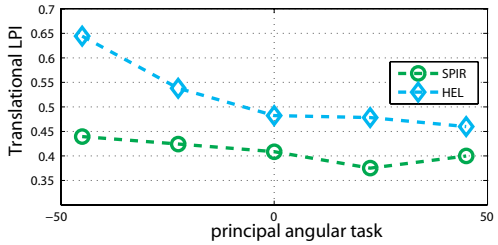
The values of all the GPIs show the better performances of joint SPIR. This result is confirmed by the analysis of the LPI's trend reported in Fig. (7). The analysis has been limited in a $\pm 45^\circ$ range of rotation in order to limit the joint stress. Nevertheless, as depicted in Fig. (8) both joints are capable of a positive 90° rotation and a negative 45°. In terms of system reliability, fatigue failure is still an issue. However, it is strongly believed that future improvement in terms of available materials will allow to overcome the problem without any further modification of the joint morphology.

TRANSFERABLE DESIGN: SOFT COVERS BASED ON DIFFERENTIATED LAYER DESIGN

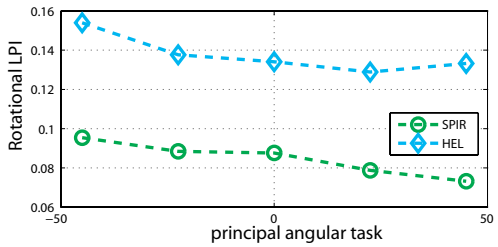
The adoption of soft covers (pads) for artificial hands and fingers is important primarily for three reasons: *functionality* in some specific tasks, *safety*, and *acceptance* by the users.

Concerning functionality, the presence of a surface compliance can highly influence the performance of the hand when contacting the environment during force/position controlled task, similarly to what happens in human fingers or feet which are covered by pulpy tissues [37, 38, 39]. First of all, the presence of a passive compliant surface is beneficiary in terms of contact effectiveness. In fact, an increased pad compliance (or, inversely, a low pad stiffness) means larger contact areas for a given load and therefore reduced contact pressure, reduced material stress and better contact stability [40, 41]. Furthermore, the soft pad allows local shape adaption in case of contact with sharp edges or objects with morphological irregularities and can contribute to vibration damping [42]. At last, a compliant covering surface helps protecting both the mechanical structure of the hand (including the transmission system) and the delicate sensory apparatus (if present).
Concerning safety, as demonstrated and quantified in [14], the soft pad can be seen as a passive device which reduces possible injuries in case of accidental impacts with humans.
Finally, concerning acceptance, a soft-touch feeling can be important in the case of human-machine interaction. This issue is particularly evident in the prosthetic industry where hand-like gloves providing enhanced functionality and increased cosmetic appeal are usually chosen at the expense of efficiency, cost and weight of the overall prosthesis.

In terms of design requirements, the properties of an ideal soft cover are hardly definable. For instance, the overall stiffness of a robotic fingertip, which is designed for manipulation purposes, can be different if compared to the stiffness of an arm soft cover, whose main func-

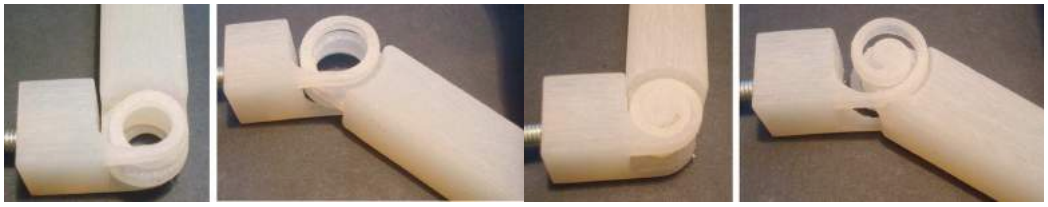


(a) trend of LPI I_T for *SPIR* and *HEL*



(b) trend of LPI I_R for *SPIR* and *HEL*

Fig. (7) Local performance index (LPI) trend for joints *SPIR* and *HEL* (Fig. (4)a and (4)b)



(a) (b)

Fig. (8) — Deformed compliant joints: (a) SPIR, (b) HEL

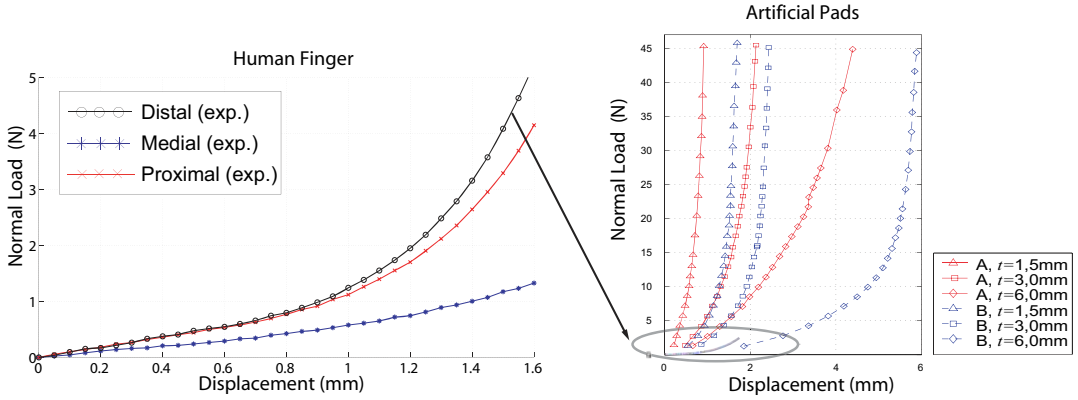


Fig. (9) — Experimental results [42]. Displacement (mm) *versus* normal load (N) for the human fingertip and for soft pads with different hardness (materials A and B) and different thickness, t. Material A = soft silicon rubber (hardness 18 shore a). Material B = very soft silicon rubber (hardness 20 shore 00)

tionality is limited to safety issues. In addition, a complete characterization of a robotic pad must include investigation on many properties and behavioral aspects [43]. However, a primary role is played by the behavior of the pad under normal contact load, in interaction with a rigid planar object. Therefore, in the following, the investigation will regard the contact behavior of soft fingerpads pressed against a rigid flat surface.

Concerning robotic hands, the majority of soft pads studied so far were made by viscoelastic polymers homogenously shaped over an internal rigid core mimicking the bone or the robotic finger inner rigid structure [44, 45, 7]. In such a case [42], the parameters that mainly contribute to the pad compliance, for a given external geometry, are:

- The material hardness. A higher material hardness, which is beneficiary in terms of surface reliability, signifies lower compliance.
- The layer thickness. A higher thickness signifies higher compliance which is beneficiary in terms of safety and increases grasp stability/sustainability. On the other hand, high pad thickness signifies high overall limb dimensions. As a matter of fact, thickness reduction is a significant goal for the robotic limb designer, that cannot easily reduce the overall size of the internal rigid core (hosting actuators, transmissions, sensors, etc.) but wants to obtain slender bio-mimetic limbs at the same time.

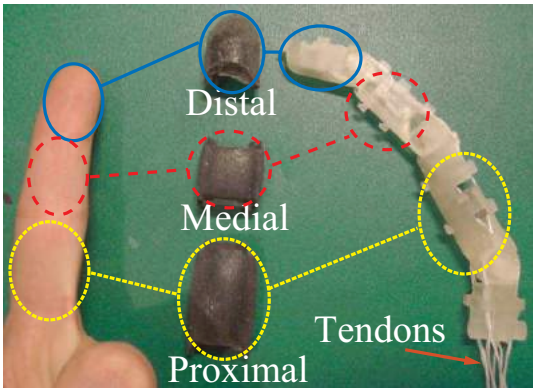


Fig. (10) — Robotic finger endoskeletal structure and soft layer (proximal, medial, distal phalanx cover)

As an example, let us consider the behavior of the human fingertip [46, 47] (distal phalange, Fig. (10)) which is shown on the left diagram in Fig. (9). In order to replicate the compression behavior of the human fingertip, it is necessary to employ a very soft silicon rubber (hardness 20 Shore 00) with very high thickness (6,0 mm).

Usually the adopted pad design is a trade-off between the need of slender robotic limbs and good material properties. Still, it is sometimes impossible to tailor the pad properties to the specific application by simply using an homogenous viscoelastic layer.

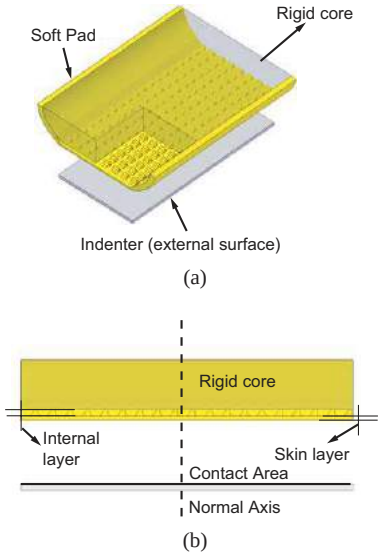


Fig. (11) — Differentiated layer design concept

Looking for alternative solutions to homogenous soft covers, the authors have previously proposed the concept of Differentiated Layer Design (DLD) [7] which allows to increase the pad compliance while minimizing its thickness. The concept of DLD consists in the adoption of a single solid material dividing the overall thickness of the pad into layers with different structural design (i.e. an external continuous skin layer coupled with an internal layer with voids). Fig. (11) shows a DLD soft pad.
In particular, a methodology have been proposed in [8] which allows to minimize the designer effort when trying to replicate the non-linear relationship $F = f(\delta)$ between the applied normal Load, F and the contact Deformation, δ (LD curve), which is representative of endoskeletal structures covered by pulpy tissues.
Given the allowable pad thickness and the overall contact area, the purpose is to tailor the pad properties to the specific application by:

- Selecting a skin material characterized by proper tribological features (hardness);
- Designing an internal layer geometry (Fig. (11)) so as to obtain a specific static compliance (increased with respect to a non strucutured pad).

The methodology adopted for designing the internal layer is composed of two steps:

- Firstly, the cover surface (overall contact area) is conceptually split into finite elementary triangular sub-regions;
- Secondly, the internal layer of each Triangular Element (TE) is designed in order to replicate the shape of the given non-linear relationship $f(\delta)$. A series of symmetrically disposed inclined microbeams is used for the purpose.

Once the compression law of each triangular element is known, the overall pad compliance can be modulated by correctly choosing the number and, consequently, the size of the elements composing the pad.

Selection of a Skin Material with Proper Hardness

Following the conceptual procedure outlined in the previous section, the design of the pad starts with the selection of a suitable polymer with proper hardness. Two solutions have been considered:

- Silicone rubber Wacker ELASTOSIL RT 623 A/B: two component silicone that vulcanizes at room temperature whose hardness can be varied in a very wide range by adding a third component (silicone fluid AK). Various pad geometries can be obtained through injection moulding.
- Tango Plus Fullcure 930 (hardness 27 Shore A): polymeric resin used for stereo-litography. This stereo-litographic technique allows to get complex shape in a short producing time.

By using both Rapid Prototyping or injection moulding, a multi-layered DLD pad can be obtained as a single piece object, the intermediate layer being realized with various geometries with exception of closed-cell structures. In fact, concerning rapid prototyping, a removable wax must be deposited as a sustaining additional material in case of negative slope of the lateral surfaces. Concerning injection moulding the possible geometries are limited by the extraction of the mould. In addition, silicon pad must be carefully cured in order to avoid the presence of air within the mould. Nonetheless, beside the technological limitation and production difficulties, different DLD pads are realizable with both materials.

Design of the Structured Pad Inner Layer

The basic idea concerning the choice of the inner layer geometry is that it is simpler to design and analyze a simple shape element and then to replicate it as many times as needed. Therefore, it is suggested to conceptually divide the overall contact planar area into finite elementary regions. Once the element LD curve is known (by means of numerical analysis or experiments), the number of elements, N, can be chosen such that:

$$F = N \cdot F_t \tag{9}$$

where $F_t = f_t(\delta)$ is the non-linear LD curve of each element.
Hence, it is proposed to divide the pad contact area into finite TEs by using a *triangular grid* [48].
A triangular grid is defined as an isometric grid formed by tiling the plane regularly with equilateral triangles. The grid cells that fall outside the object are removed. The result is a mesh with equal interior TEs. If needed,

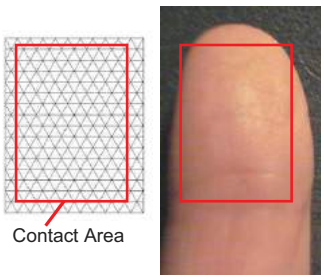
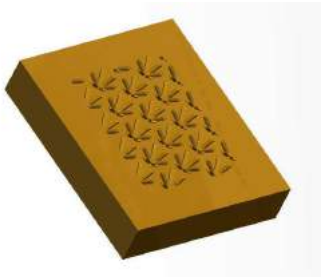


Fig. (12) — Triangular grid for a fingertip contact area.

the grid cells that intersect the object boundary can be adjusted or trimmed so that they fit into the object. Nevertheless, deformed TEs which are located on the boundary might present an LD curve which slightly differs from the LD curve of interior TEs. An example of a triangular grid for meshing a fingertip contact area is shown in Fig. (12).
Obviously, smaller TEs are beneficiary for two reasons:

- The object boundaries can be better captured by a fine mesh than by a coarse mesh.
- Whatever will be the inner layer geometry of each TE, the contact pressure will tend to be a uniform function (i.e. a continuous function) as $N \rightarrow \infty$.

Note that: 1) the procedure outlined in the following regards the definition of an overall pad contact force (i.e. overall pad compliance) which is an integral (rather than a local) property of the Pad; 2) N elements are involved in the contact simultaneously and the contact area is displacement independent.
The smallest size, A_{TE}^{min}, of feasible TEs is determined by the technological feasibility of the pads. On the other hand, Eq. 9 constraints the number of elements which must be contained within a given contact area and therefore the element size $\overline{A}_{TE}$. If the size $\overline{A}_{TE}$ required by Eq. 9 is lower than A_{TE}^{min}, a practical solution cannot be achieved for the given TE inner-layer geometry. On the other hand, if $\overline{A}_{TE} > A_{TE}^{min}$, the TE outer-layer (skin) can be enlarged without altering the TE LD curve.
As for the TE internal layer, it is designed in order to replicate the qualitative shape of the non-linear compression law which is typical of endoskeletal structures covered by pulpy tissues. This behavior is well exemplified by the LD curve of the human finger shown in Fig. (9): it is possible to note an initial, quasi-linear LD curve for small displacement followed by a rapid load increase. In order to replicate this particular compression law, it is proposed to use a series of micro-beams inclined of $\vartheta = 45^\circ$ with respect to the normal to the external surface (normal axis, Fig. (11)b), thus transforming normal loads acting on the contact into bending actions applied on each beam. The micro-beams are placed on the edge of the TE as depicted in Fig. (13) (artificial pad internal layer surface and associated TE). This peculiar geometry presents a quasi-linear LD



(a)

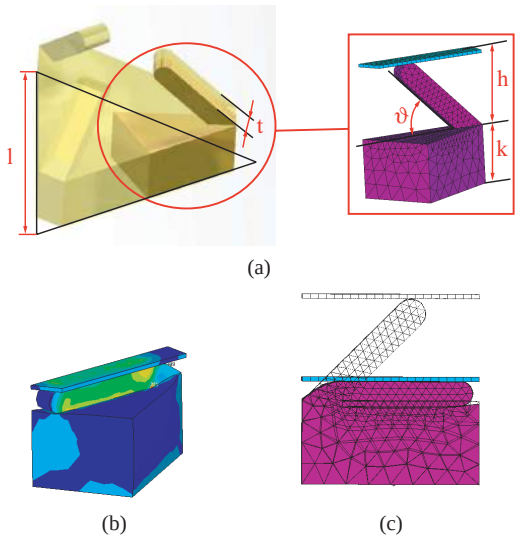
(b)

Fig. (13) — Soft pad based on pattern with equally spaced micro-beams (a) and associated TE (b).

curve for small displacements which is characterized by a very low stiffness. On the other hand, the load rapidly increases once the micro-beams collapse on the outer skin. In such situation, the TE behave similarly to a pad made of a uniform soft material. A finite element model of the TE is shown in Fig. (14). In particular, Figs. (14)b and (14)c depicts one collapsed micro-beam.

Soft Pad Realization

The smallest TE which is considered realizable by means of SLA is an equilateral triangle having surface area of 6.9 mm^2 (i.e. 4mm side). The pad thickness is chosen to be 3.0 mm (i.e. half the thickness of previously published solutions, see Fig. (9)). The TE design is exactly the one depicted in Fig. (13), (15) ($t = 0.5mm, h = 2mm, k = 1mm, \theta = 45^\circ$, l to be designed) and numerical relationship between the applied normal force F and the consequent displacement δ is shown in Fig. (15)a (FEA results).
Let us consider first the distal phalange. The contact area to be meshed is a 20mm x 15mm rectangle which is meshed by means of 36 TEs in order to obtain the desired compliance. Such TE presents a surface area of 8.3333 mm^2. Figure (15)b shows the numerical relationship between the normal load (N) and the resulting displacement (mm) for: 1) the structured pad depicted in Fig. (13)a; 2) a uniform PAD of the same thickness (3mm) made of a softer material (refer to Fig. (9)); 3) the human finger. It can be seen that a 3mm thick structured pad represents a substantial step forward in human finger mimicry in terms of stiffness, when com-



(a) (b) (c)

Fig. (14) — Numerical analysis of TE based on a series of inclined micro-beams. model mesh (a), collapsed micro-beam (b,c). material: *tango plus* rubber [8].

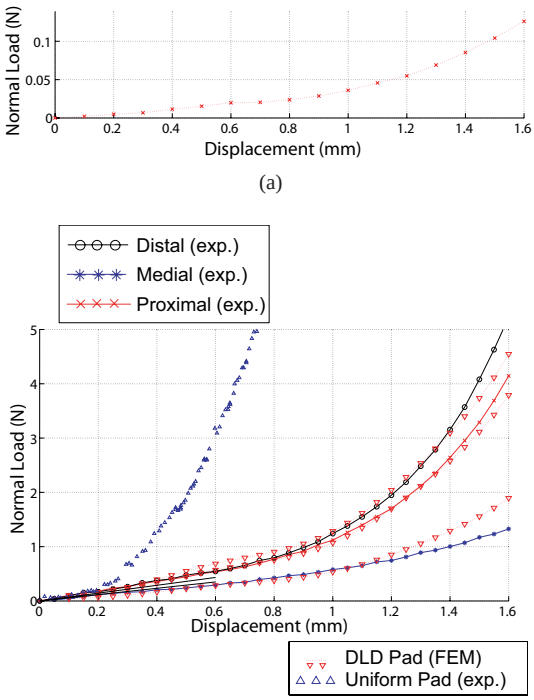
pared to previously published solutions where different materials and higher pad thickness are used. Finally, the first pad prototype is shown in Fig. (15)c. Concerning the medial phalange, the number of TEs is reduced to 30. Concerning the distal phalanges, the overall contact area is spilt into two 20mm x 15mm rectangles and the number of TEs for each rectangle is 15.

The potentialities of the DLD concept have been experimentally evaluated on hemispherical soft pads shaped over a rigid core [7]. The pad to be mounted on the hand prototype are shaped as in Fig. (15)c and their physical implementation is shown in Fig. (16).

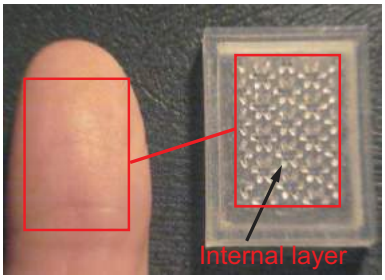
DISCUSSION AND CONCLUSIONS

This chapter has reported design solutions and methods concerning the adoption of sliding tendons, integrated joint and soft covers which have been proven achievable with current technology. Nonetheless, the adoption of the aforementioned solutions arises serious issues which must be addressed. In particular, the implementation of robust, low cost artificial hands is a target of primary importance within the prosthetic industry. To this respect, any approach that enhances simplification, reliability and at the same time reduces manufacturing and assembly costs can help.

In light of these considerations, transfer of design concepts and technological solutions from the world of robotic hands can be feasible under the assumption that they are compatible with the peculiar features of prosthetic devices, that is a limited availability of energy for actuation and a poor capacity of control. This greatly reduces the capability to manage complex multi-degrees of freedom structures: joints that can be independent in robotic hands must be necessarily coupled within a prosthetic device, reducing the number of actuated de-



(a) (b)



(c)

Fig. (15) — Displacement (mm) *versus* normal load (n) for TE made of inclined micro-beams for uniform pad (a), DLD pad and human fingertip (b), Artificial soft pad (c). Experimental (exp.) and FEM results.

grees of freedom.

To this respect, actuation by sliding tendons shows controversial aspects. On one side, it is relatively easy to achieve fully-integral finger structures with tendon couplings placed inside the hand palm; on the other side, the mechanical efficiency of this kind of transmission is rather poor and no control strategies can be provided, as it happens in the case of robotic hands [18], in order to compensate for the many negative effects due to friction and tendon compliance; furthermore tendon transmission is intrinsically reversible, while most of prosthetic hands still adopt non-reversible mechanisms in order to guarantee grasp holding in case of power-off. As for joint design, experimental activity [6] has proven that joint friction is more crucial than tendon friction and can cause undesired phalanx locking. To this respect, the adoption of large displacement CJs should be preferred when compared to simple pin joints. Nonethe-

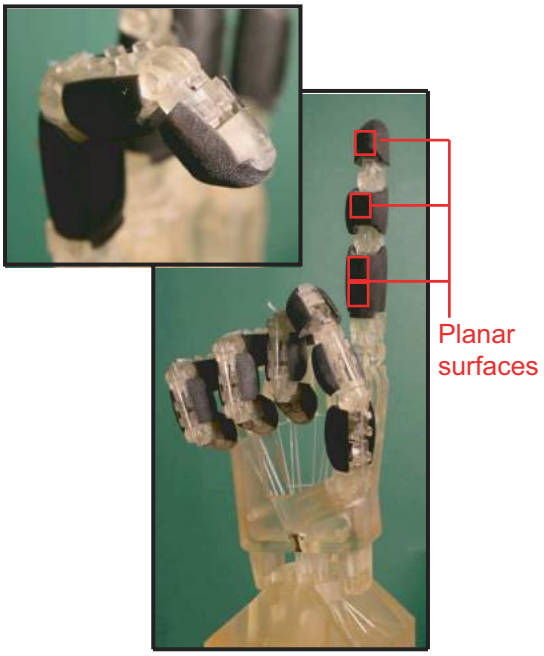


Fig. (16) — Anthropomorphic robotic with enhanced surface compliance

less, it should be noted that fatigue failure can be an issue whenever compliant mechanisms are considered, especially in the case of prosthetic devices where external forces acting on the structure are largely unknown. In any case, as said, CJ design is heavily dependent on the inseparable binomial material-technology. For instance, techno-polymers, shaped by means of pressure moulding or CNC machining, can provide good reliability but present shape complexity limitations and are cost ineffective. Per contra, rapid prototyping allows a simple, fast and inexpensive production of items but is rather poor in terms of available materials. Still, due to the high market demand, it is reasonable to envisage an high improvement of the material properties in a near future. Owing the aforementioned limitations, a comparison method has been proposed which provides practical indications on how to choose the optimum CJ for prosthetic hand. The method is applicable to novel CJ designs (such as SPIR and HEL joints) or to any known CJ morphology.

As for the soft cover, a more direct transferability can be envisaged: in particular, the development of DLD pads, as compared to non-structured layers or to simple foam layers, allows to tailor the local surface compliance according to specific needs. Furthermore, this concept permits the use of one material only, to be chosen according to its tribological properties (e.g. resistance to wear) or manufacturabilty (e.g. compatibility with additive technology processes that allow easy implementation of complex shape items). In practice, the DLD methodology enables obtaining thin compliant layers, thus limiting the overall size of fingers not compromising their slenderness, but at the same time to achieve a global compliance comparable with that of human finger pulps. Therefore, the application of this concept can increase design flexibility and easily integrates the need of acceptable aesthetics with the functional requirement of thin, highly compliant external covers.

A final consideration is that neither the proposed CJ design nor the proposed DLD pads seem to go into the direction of morphological simplification: part integration on one side means reduction of separate items, on the other dramatic growth of morphological complexity. This drawback can be overcome thanks to the availability of the so-called computer-driven Additive Manufacturing Technologies that allow to imagine, for an imminent future, the full feasibility of integrated multi-material complex structures, like robotic fingers or hands, in a single production step. These parts, being fabricated with an additive approach, eventually represent a final product instead of a mere prototype. In fact, the term Additive Manufacturing is slowly substituting the term Rapid Prototyping [17] in order to underline a closer link to the end-use component.

In conclusion, the effort of the authors was in the direction of exploring new design solutions and methods which largely relies on the possibilities, before unknown, deriving from this technological progress that gives more freedom to design creativity.

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CHAPTER 7

Haptic devices for the simulation of upper limb in Virtual Reality

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Abstract: This contribution deals with a particular type of robotic systems, i.e. exoskeletons or wearable systems. With respect to conventional robots, exoskeletons present the main feature of being wearable and, consequently, always in contact with the human operator during operative conditions. The design and control of exoskeletons must then necessarily take into account this condition, not only for safety issues, but also in terms of transparency for user's movement and fidelity in the generation of torques/forces to the operator. The experience of the PERCeptual RObotics laboratory of Scuola Superiore Sant'Anna in the design of exoskeletons is presented, by addressing the description of developed robotic exoskeletons. Implications for the usage of exoskeleton systems in the simulation of grasping in Virtual Environments are discussed, with an analysis of the issues associated to the test of myoelectric control of prostheses in Virtual Environments.

Keywords: Human-Robot Interaction, Exoskeleton, Haptic Devices.

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INTRODUCTION

The idea of building artificial machines mimicking the anthropomorphic aspect and functionality of the human arm has been strongly tied with all, even preliminary, developments of robotic arms.

For a specific type of mechanical structures, named robotic exoskeletons, the above property represents a fundamental and intrinsic requirement, since they rely on an anthropomorphic structure. In general terms, exoskeletons are robotic devices which can be worn on the body, with a kinematics congruent to the limbs (arm, hand, torso, leg or foot) they are connected to at specific points (usually more than one), in order to be able to follow and/or constrain their motion.

The analysis of this particular type of robotic interfaces is presented here in the framework of haptic interaction, since the first examples of exoskeleton systems date back to the early developments of teleoperation, when robotic systems were designed to generate force feedback remotely to astronauts in space [1, 2, 3]. The development of force feedback devices (recently named haptic interfaces) finds its roots in the stream of research dealing with teleoperation systems, developed since the mid 1940s, when the first master-slave teleoperator was built by Goertz in Argonne National Laboraotry, then followed the work of Vertut in France [2]. In mid 1980s, robotic master systems such as the Salisbury's Force-Reflecting Hand Controller [4] were purposely designed with a different kinematics from the slave systems they intended to control. Only in the late 1980s, a new attempt to introduce force feedback for teleoperation tasks brought to the design of innovative robotic structures specifically intended for generating force/torque components to the hand of the human operator: such systems were also applied to the control of manipulative operations during the interaction with Virtual Environments, and from the end of 1980s were named "haptic interfaces". An interesting sign of the definitive merging of the parallel developments of force feedback devices and haptic interfaces into a single area of research is testified by the creation in 1990 of the Journal "Presence: Teleoperators and Virtual Environments" by MIT Press. Here the topic of telepresence and virtual presence were considered as identical in terms of the interface system of the human operator with the remote or synthetic spaces. In general terms, the common functionality of a haptic interface or of a force feedback device is that of being able to generate forces at the level of the human limb where these forces are required or simulated [5]. Usually these places are the palmar surface of the human hand when manipulative operations must be controlled, or other parts of the human body, such as the anterior or posterior parts of the trunk in the case of whole body motion haptic interfaces. To exert forces on the human limbs, when the user intends to manipulate objects in the remote or virtual operational space, as far as the interaction with the Haptic Interface (HI in the following) is concerned, two possible operative conditions are possible:

- the HI is always maintained in contact with the user's limb during the control of the operation

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- the HI enters in contact with the user's limb only at the instant of time in which the generation of forces is required

The first category, that we will call "Always in Contact" devices [6], comprehends the large majority of present haptic interfaces, i.e. robotic devices that are external with respect to the user's body and that are usually grasped at their end-effector during the whole duration of the interaction task. The second category, usually called "Encountered Haptics" [7, 8], represents a completely different concept of haptic interface: here the robotic system is controlled to follow the user's hand or limb at a certain distance except when a contact force is required given the specific hand position and orientation at the specific instant of time. At that exact instant of time, the "encountered" type haptic interface is controlled in order to enter in contact with the human limb and generating the required force vector.
In this chapter, we will analyze only the first class of "Always in Contact" Haptic Interfaces, that are characterized by the fact that the parts of their structures where forces are transmitted to the human operator's body remains always in contact with the limb during the execution of the interaction task. "Always in Contact" devices can be classified according to two different types of systems, as External devices and Wearable devices.
As above introduced, in the definition of the Always in Contact External Devices, the term "external" refers to the fact that

a) the kinematic structure presents a geometry (joint axes and their relative positioning and orientation in space) that differs from that of the human limb to which they are attached, e.g. of the hand-arm complex;

b) the workspace of the device differs from that of the human limb; these two workspaces possess an intersecting volume, usually located around the reference position, which is only a fraction of the whole volume spanned by the arm;

c) the base frame of the haptic device is usually grounded on a fixed location or to a mobile location that can extend its effective workspace; in this last case, however, the complexity of the whole haptic system increases a lot, due to the need of allowing the mobile platform to track the user's movements

Exoskeletons are a particular type of wearable devices [9], that can implement a larger number of functionalities, i.e:

- to record the movement of the limb to which they are attached in terms of specific joint motion; such a functionality cannot be realized with external always in contact devices since they are able to record only the movement of the human point of attachment in the cartesian space;
- to intrinsically possess a workspace very close to that of the human limb to which they are attached;
- to generate wrenches to the human limb in all the points of attachment with the human body; also in this case, while for an external always in contact haptic interface the number of attachment points is limited to one (usually the hand), exoskeleton systems can be designed to have several points of attachment (usually one per limb member) with the human body;
- to implement gravity compensation on the limb they are covering; in general terms, exoskeleton systems can apply a defined force field to the limb;
- to generate or constrain torques/movements at the level of the joints of the limb they are covering; this feature is extremely interesting in the case the exoskeleton system is used for rehabilitative purposes;
- to implement joint trajectories that map specific limb movements;
- to scale torques at each specific joint in order to implement an effect of amplification of human capabilities;
- the wearability of the exoskeleton system allows the human operator to move in the control space; such an increased mobility feature is suitable for applications in which the user is requested to cover a large workspace. Moreover, this is an effective feature to be exploited in the cases the human operator must interact not only with remote/virtual objects but also, at the same time, with real objects in the control space, e.g. control panels or real tools.

In the following, a description of present recent developments of exoskeletons systems at the Perceptual Robotics Laboratory PERCRO is reported. Issues of design and control of exoskeleton structures constitute the fundamental part of the presentation. Moreover implications for the usage of exoskeleton systems in the simulation of grasping in Virtual Environments will be discussed, with an analysis of the issues associated to the test of myoelectric control of prostheses in Virtual Environments.

EXOSKELETONS FOR THE HUMAN ARM

Since the beginning of 1990s the Perceptual Robotics Laboratory of the Scuola Superiore Sant'Anna, Pisa, Italy, addressed a large amount of research activities to the design and development of wearable robotics systems as haptic interfaces for Virtual Environments applications and Rehabilitation.

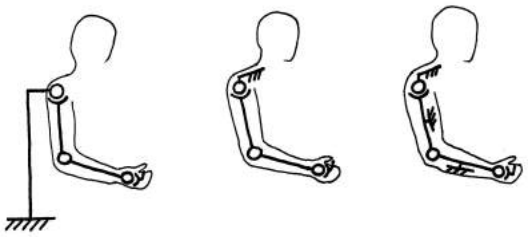

Fig. (1) — Different schemes of attachment of an exoskeleton device to the human arm

In 1991, being the research at PERCRO mainly focused on robotics manipulation and robot hands, the design of exoskeleton systems immediately followed the same area and the first versions of exoskeletons were obtained in 1992 both for hands and arms.
The design functionalities that drove exoskeleton developments were described in [10] in which the issues of transparency and fidelity were outlined as fundamental for achieving an effective solution.
Different exoskeleton structures might be devised, as shown in Fig. (1), according to the number and location of attachment points between the exoskeleton structure and the operator's arm. If attachments are considered also in correspondence of the medium part of the forearm and arm, it is possible to (locally) generate forces also in these regions.
Four versions of the first design exoskeleton (one point of attachment) were designed and realized at PERCRO respectively in 1992, 1995, 2002 and 2008.

Arm exoskeleton version I (1992)

The arm exoskeleton designed at the beginning of 1990s was one of the first example of wearable robotic systems ever realized for teleoperation and Virtual Environment interaction [11]. It consisted of a 7 DoFs (Degrees of Fredom) robotic structure designed to completely wrap up the user's arm and supported, by means of a purposely conceived frame, by the shoulders and trunk of the user itself (see Fig. (2)).
The Degrees of Freedom imitated the joints of the human upper limb: 2 DoFs at the level of the shoulder (flexion-extension and abduction-adduction movements); 1 DoF in correspondence of the arm (rotation of the arm); 1 DoF at the elbow (flexion-extension); 1 DoF in correspondence of the forearm (prono-supination movement); 2 DoFs at the wrist level (flexion-extension and abduction-adduction). The scheme of the kinematic chain representing the complete 7 DoFs system is reported in Fig. (3).
In principle, the arm exoskeleton had the capability to follow a significant percentage of movements of the human arm, but constraints due to mechanical interference did not allow to span the complete workspace of the human arm. However, very good mobility was achieved

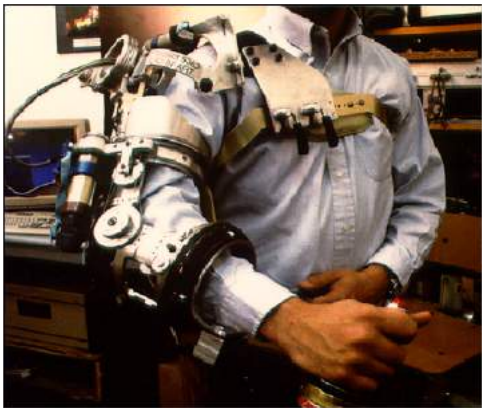

Fig. (2) — Arm Exoskeleton Version I (1992)

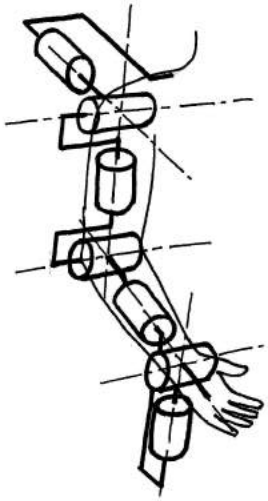

Fig. (3) — Kinematic scheme of an exoskeleton for the right arm

around a significant reference position for manipulative operations, consisting of vertical arm, flexed forearm in the horizontal plane, wrist in intermediate position between pronation and supination.
Each joint of the exoskeleton was actuated by means of DC servomotors integrated in the structure: such an arrangement of the motors, although presenting a high degree of complexity in terms of design layout and packaging of mechanical components, allowed the great advantage that the resulting robotic system was completely portable and no mechanical or actuation component was remotely located.
The motion at each joint was obtained through a tension tendon-based transmission system: while the 2 DoF of the shoulder and the arm rotation movements were implemented without the use of idle pulleys, the transmission system for the elbow and forearm joints presented complex cable routing. The correct arrangement of cables and pulleys was one of the most critical aspects of the whole design and testing phase. Joint rota-

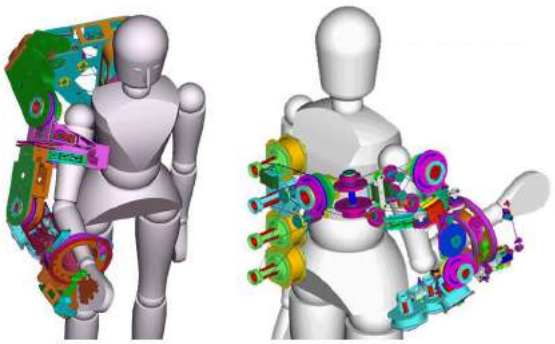

Fig. (4) — CAD model of the Arm exoskeleton II, with a view of the innovative transmission system

tion sensors were integrated for each joint. The following values of generated torques were obtained at different joints: a) shoulder joint abduction-adduction and flexion-extension: 20 Nm; b) arm rotation and elbow flexion-extension: 10 Nm; c) forearm prono-supination: 2 Nm.

Version I of the arm exoskeleton, being all the structure made in aluminum, presented a total weight of 10 kg including motors and transmission units. Gravity compensation techniques were implemented in order to allow a transparent behavior of the user's arm when the exoskeleton system was worn. In fact, experimental tests indicated that when the system was worn by the operator, and supported by his/her shoulder and trunk, the weight of the structure affected the maneuverability performance only when the system was not controlled [11].

Arm exoskeleton II (1995)

A second version of the Arm Exoskeleton was designed in 1995 [12] in order to better take into account the other two functional requirements of adjustability and wearability.

Adjustability refers to the introduction of variability in the size of the exoskeleton geometry, in order to accommodate different arm sizes (different arm and forearm lengths and circumferences), while wearability refers to the easiness of wearing, which represents a key issue for the acceptance of the device among users who are not specifically trained or have specific upper-limb motor disabilities. These two requirements were translated into innovative design solutions introduced in the Arm Exoskeleton Version II, respectively the design of tendon transmissions and a a mechanical design solution allowing the lateral wearing of the device, as it can be seen from the CAD model in Fig. (4).

The tendon-based driving system allowed to place the motors away from the joints. Our design goals were to achieve reduced link weight (especially for the distal joints), increased joint compactness, reduced encumbrance in the workspace of the moving parts of the device. Moreover, the devised solution allowed the usage of grounded actuators with high peak torque and consequent reduction or elimination of gearboxes (with the associated friction and torque transmission problems). The main drawbacks of tendon transmissions are associated to the tendon elasticity and routing. The elasticity between the motor and the link, which is inevitably introduced by tendons, tends to lower the stiffness and the mechanical bandwidth of the device. Therefore these values were kept under control during the design. The routing of the tendons from the motor to the joint become complex in multi-DoFs systems. If the tendons are guided by sheaths the routing is largely simplified but severe problems arise in force control due to dry friction. In all the designs of Arm Exoskeletons at PERCRO, the tendons are routed over idlers mounted on ball bearings. Such a method has been used in other robotic structures [13, 14, 15, 16]. Usually such a type of transmission is planar, i.e. all idlers lay in a common plane. A variant to this approach has been proposed in [13] allowing the presence of a small skew angle between the an idler and the following. In other designs, such as the WAM from MIT and the Arm Exoskeleton Version I from PERCRO, the axes of two successive idlers can be perpendicular in order to route the tendon between two orthogonal planes. In the design of Version II, an innovative design solution was introduced for arbitrarily placing in space the axes of two consecutive idlers, subjected to the constitutive condition that the two successive idlers share a common tangent line. This solution was a key factor for allowing the proper routing of tendons around the human arm and forearm.

Another important design solution was addressed in terms of wearability of the device. With the aim of achieving a comfortable wearability, "open" links for the arm and forearm were designed and patented. In this way, the links of the exoskeleton do not wrap completely the user limb, but instead just adhere to the external aspect of the arm. This solution, implemented also in successive designs, allows easiness of wearing, since the user arm enters laterally into the device (while an arm exoskeleton has usually to be worn just like a sleeve) and it adapts to a broader range of user arm circumferences. Fast lateral wearing/unwearing gives also a greater intrinsic safety and a broader acceptance among users. Moreover such a solution allows the user to bring his arm very close laterally to his trunk. The design of open links around the user's arm and forearm, while meeting the constraints of coincident anthropomorphic kinematics, was possible thanks to an on purposely developed partial (semicircular) rolling ball bearing. Such a mechanical component possesses the same performance in terms of stiffness, weight and friction of a precision ball bearing of the same diameter.

Arm exoskeleton version III (2002)

A third version of the Arm Exoskeleton design was ad-

Fig. (5) — Side view of the L-EXOS

dressed in 2000 and brought to the realization of a new prototype in 2002, shown in Fig. (5). This third version was named L-EXOS, i.e. Light Exoskeleton, due to the higher payload/weight ratio of the final design, approximately equal to 1.
The L-EXOS is a 5 DoFs robotic device with a serial kinematics, isomorphic to the human arm. Only 4 DoFs are actuated, being the non-actuated DoF the last one, aligned along the anatomical prono-supination axis of the forearm [17]. Also for the L-EXOS, one of the main design specifications was achieving high transparency of use of the device, i.e. the system exhibits low mechanical impedance when moved by the subject (low friction, low inertia). To fulfill such a goal, a set of design guidelines were adopted:

- **Remote placement of actuation:** This solution allowed to drastically reduce the perceived inertia and joint torques required for gravitational compensation, and consequently also the actuator size and the transmission tensions. Moreover, the remote placement of motors allowed a better weight balancing of the structure. All the motors of the exoskeleton were located on the fixed frame (Link 0), as shown in Fig. (8). As a consequence of grounding the motors, long transmissions were required [18, 19].

- **Use of tendon transmissions:** Tendon transmissions can easily transmit torque to joints, placed far apart from motors, with zero backlash, low friction and low weight; For each actuated DoF, the torque is delivered from the motor to the corresponding joint. Transmissions were implemented through steel cables that can guarantee low weight and zero backlash, (see in Fig. (6) the routing for the transmission of the third joint), that are also more efficient than gear transmissions, ensuring a better backdrivability of the system.

- **Integration of speed reducers at the joints**: A reduction gear was integrated at the joint axis, as depicted in the scheme of the transmission of axis 2 in Fig. (7). Such an arrangement allowed to reduce the masses of the moving parts, by reducing the mass of the motors (near 40% of the overall mass of the exoskeleton) and the additional mass of the structural parts, to be reinforced in order to sustain the weight of heavier motors. The inertia perceived by the user at the palm was also consequently reduced.

- **Selection of motors with high torque to weight ratio**: Electric actuators offering the best torque to weight and torque to volume ratios were selected [17]. To achieve a higher stiffness of the device at the end-effector, reduction gears with low reduction ratio were located at the joint axes, thus allowing to reduce the tendon tension, their elongation and their diameter. The reduction of the tendon diameter led to a consequent saving of mass and volume of all the mechanical parts of the transmission system (pulleys, axes, etc.). Thanks to this solution and to the development of expressly conceived speed reducers, a mass reduction of about 35% for the transmission system and of 40% for the structural parts was achieved.

- **Low transmission ratio:** this allowed to reduce the contribution of the motors to the perceived inertia at the end-effector and so to lower the perceived transmission friction;

- **Use of light materials for the construction of the moving parts** The structural components (links) were designed as thin-wall parts, that can house the mechanical parts of the transmission (pulleys, tendons, axles, spacers, etc.) within the links, protecting the inner parts from external interferences and the user from potential harm deriving by the tensed steel cables. Hollow sections, presenting the larger moment of inertia than bulk sections with the same area, were used to enhance the stiffness of the thin-wall parts. In order to further improve lightness and stiffness, the structural components were made of carbon fiber. Also Aluminum parts were bonded on the carbon fiber structure to realize the connections with other mechanical components. Carbon fiber parts were manufactured with the vacuum-bag technique, and with dies made of carbon fiber too, due to the low number of manufactured prototypes.

As far as the kinematics, as depicted in Fig. (8), the first three rotational axes are incident and mutually orthogonal (two by two) in order to emulate the kinematics of a spherical joint with the same center of rotation of the human shoulder. The target workspace of the shoulder joint was assumed to be the quadrant of a sphere, as shown in Fig. (9). The orientation of the first axis was optimized to maximize the workspace of the shoulder joint, by avoiding singularities and interferences be-

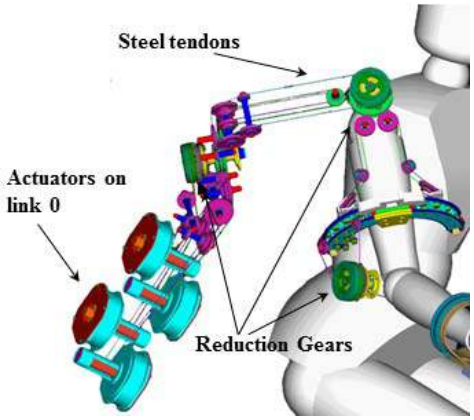

Fig. (6) — CAD model of the L-EXOS transmission system

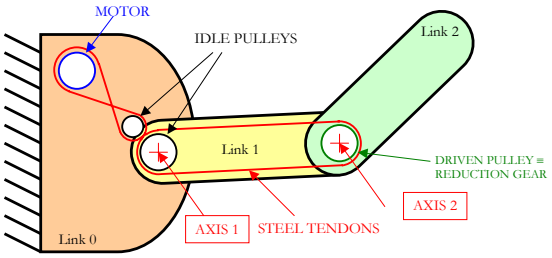

Fig. (7) — L-EXOS: Scheme of the actuation and tendon transmission system

tween the mechanical links and the operator. The optimization process provided also indications for the definition of the shapes of the links. As a result from the kinematic analysis, the final orientation of the first axis (the fixed one) was chosen to be skewed with respect to the horizontal and vertical planes, while the third axis was assumed to be coincident with the ideal axis of the upper arm. This implied that the third joint had to be implemented with a rotational pair aligned with the forearm. The fourth axis was assumed coincident with the elbow joint and the fifth axis with the forearm, in order to allow the prono-supination of the wrist.
The L-EXOS, shown in Fig. (5), can attain very remarkable performance, as summarized as follows:

Force: 50 N continuous, 100 N peak force;

Backlash: 10 mm at the end-effector;

Stiffness: estimated 3 N/mm, measured 2 N/mm;

Workspace: approximately 70% of the human arm's workspace.

The L-EXOS has a weight of 11 kg, of which approximately 6 kg distributed on link 0, i.e. the fixed part, and mostly due to the mass of the 4 motor-groups. This means that the L-EXOS achieves the desirable low

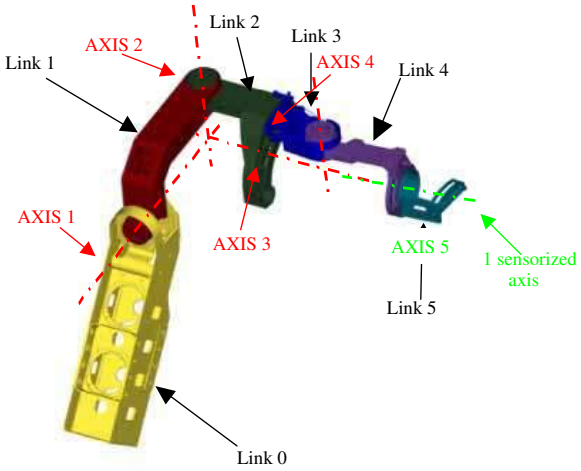

Fig. (8) — General kinematics of the L-Exos

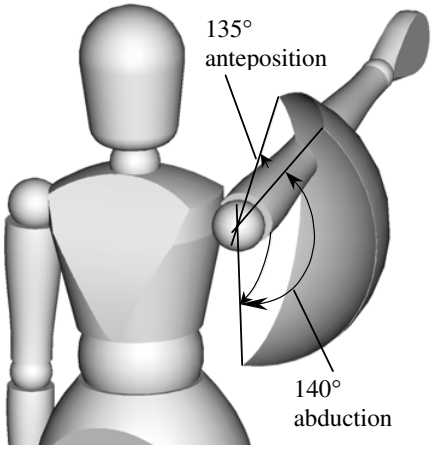

Fig. (9) — L-Exos Workspace at the shoulder joint

value of weight/payload ratio close to 1 (100 N vs. 11 kg). The reported value of stiffness of 3 N/mm represents the theoretical worst-case condition.

Arm exoskeleton version IV (2008)

We have recently assisted to recent progress of upper limb exoskeleton robots for rehabilitation treatment of patients with neuromuscular disorders, see reference [20] for an up to date review on the topic. Virtual reality (VR) applications have been widely used as schemes for rehabilitation in stroke survivors, due to the feasibility and capability of being accepted as videogame-like exercises. Recently, in alternative to lightweight designs with joint-delocated motors and back-drivable transmissions without any force/torque sensors, an innovative design for arm exoskeletons was addressed at PERCRO for rehabilitation purposes [21].
Based on the experience gained at PERCRO laboratory during the clinical evaluation of the L-EXOS [22] in the

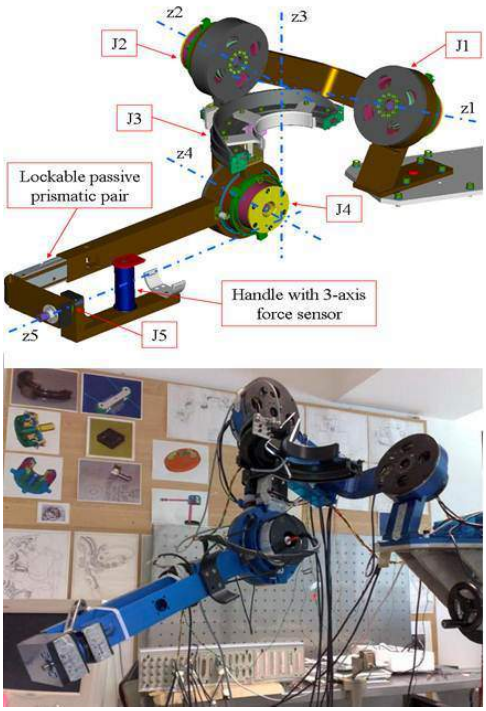

Fig. (10) — The CAD model and the prototype of the Rehab-Exos (2009)

rehabilitation of stroke patients, a novel exoskeleton, called RehabExos (Fig. (10)), was developed for the rehabilitation of upper limb. The RehabExos is aimed at generating controlled contact forces/torques not only at the exoskeleton end-link handle, but also at some intermediate link, so that the patient can be constrained to the RehabExos links at the level of his/her hand, arm and forearm according to the therapy requirements and can be easily rearranged for both left and right arm use.

As depicted in Fig. (10), the RehabExos adopts a serial architecture that is isomorphic with the human kinematics and inclues: a shoulder joint which is fixed in space and composed by three active joints J1, J2 and J3; an active elbow joint J4; a passive revolute joint J5 allowing for wrist prono-supination. Actuation of the RehabExos is provided at each joint *via* identical custom made Actuation Groups (hereafter indicated with AGi) for joints J1, J2 and J4, and *via* a different custom made actuation group for joint J3. Both AG1, AG2 and AG4 comprise an electric motor, a geared transmission with rather large reduction ratio (which optimizes the actuation group torque-to-weight performance) and unavoidable compliance, a motor-side rotary encoder and a joint-torque sensor. For a more detailed description of both RehabExos and actuation groups please refer to [21].

Figure (11) shows an example of a VR scenario for the rehabilitation of reaching movement in hemiplegic patients after stroke [23]. There are seven glasses located into a virtual book shelf. The starting position of each

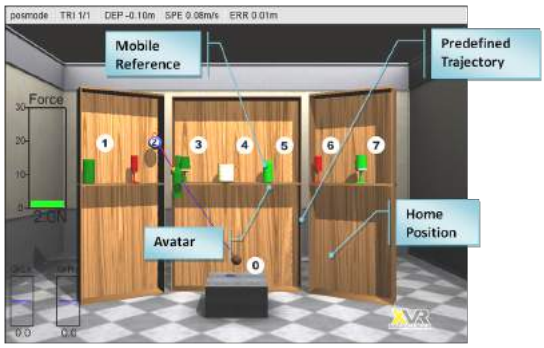

Fig. (11) — Example of a virtual rehabilitation scenario

trial is labelled as 0. The goal of the task begins when the therapist chooses the desired object to be reached. Then the patient is asked to perform a movement to move the bottle from the start position to the glass and pour the water into the glass with a prono-supination of the wrist. During the movement, the exoskeleton drives the patient arm to the target, providing the assistance as needed to help the patient to execute the movement. This class of exercises in Virtual Reality is suitable for the training not only of neurological patients, but also of amputees that need to learn the procedures for the myoelectric control of new powered prostheses.

Control of arm exoskeletons

The problem of force replication (generation) by means of an exoskeleton system can be considered identical to the problem of force generation by an ordinary robotic system but with the further constraint of considering that the number of contact points with the human operator can vary according to the number of attachment points between the exoskeleton structure and the human limb.

In the case that the exoskeleton structure is connected to the human arm only at the level of the hand, the system can be considered as an external manipulator with its base link attached to the shoulders and trunk (if dressed) or to the ground. The system can exert applied vector forces, that can be reduced to a resultant wrench, only at the level of the hand. Under this assumption it is possible to replicate external forces by controlling the wrench applied by the exoskeleton to the operator's hand. By considering as an example the kinematic scheme depicted in Fig. (3), the variables that can be used to achieve such a control objective are the joint torques.

In quasi-static conditions and with no gravity, the mapping between the applied wrench and the joint torques can be derived by means of the principle of virtual works and is given by the transpose jacobian of the manipulator:

$$\tau = J^T(q)F, \qquad (1)$$

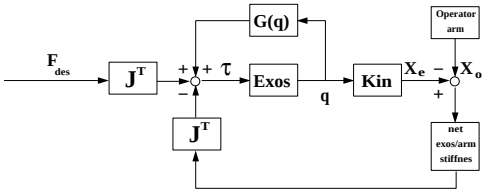

Fig. (12) — Scheme of the open loop control procedure

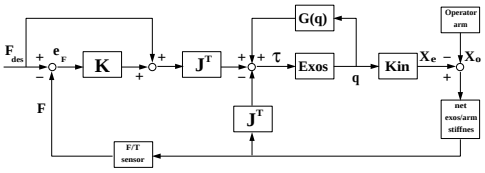

Fig. (13) — Scheme of the closed loop control procedure

where J is the jacobian of the exoskeleton depending on the joint position vector q, F is the wrench applied on the operator's hand, and τ is the vector of the joint torques. In dynamic situation and in presence of gravity, the joint torques that must be applied to give a desired wrench F_{des} must contain additional terms devoted to compensate the exoskeleton inertia, Coriolis and centrifugal effects, friction and gravity. A complete force mapping will thus be dependent on the exoskeleton configuration, joint velocities and accelerations:

$$\tau = M(q)\ddot{q} + C(q,\dot{q}) + D(q,\dot{q}) + G(q) + J^T(q)F_{des} \quad (2)$$

where M is the inertia matrix of the manipulator, C is the vector of Coriolis and centrifugal terms, D is the vector friction terms, and G is the vector of gravity effects. Notice that the effect of compliance (for instance of the transmission system) is not included in (2). The model described by equation (2) can be used to build a controller that regulates the wrench F to a desired reference value. If good performance of the force replication system is required not only in quasi-static conditions, the control must include the compensation of dynamic effects on the exoskeleton. If the requirements are not so strict in terms of bandwidth (say $\omega_0 \leq 3rad/s$), some of the terms of the full dynamic model can be neglected. Quasi-static operation ensures that the effects of Coriolis and centrifugal terms are small and, in the case good backdrivability of joint actuators is ensured by the mechanical design in order to have low values of friction, only the gravity compensation term can be used. The feasibility of the control law will depend on the availability of external wrench and/or joint torque measurements. For the sake of simplicity, in the following the dependency of J and G on joint positions will be omitted. If no force/torque sensors are present, an open loop control law can be devised as follows, as shown in Fig. (12):

$$\tau = \hat{G} + J^T F_{des} \quad (3)$$

where $\hat{G}$ indicates an estimate of G. The open loop scheme cannot be used if non backdrivable drives are present. Further, poor performance is expected due to friction and modeling errors in the estimation of G. If torque sensors are present, the following control law can be devised:

$$\tau_{des} = \hat{G} + J^T F_{des} \quad (4)$$

where τ_{des} is a reference value for the joint torque vector. A servo term at the hand level based on the wrench error may be added to F_{des} in order to improve the tracking of the reference value. A low level joint torque control loop is then used. Joint torque sensing allows to overcome the problems of friction, although the measure of the wrench F is affected by many errors due to errors in the kinematic model of the exoskeleton, and rough or no modeling at all of the exoskeleton dynamics. If a 6 component force/torque sensor is present, the loop can be closed at the hand level and the following control law can be used, as shown in Fig. (13):

$$\tau = \hat{G} + J^T \left(F_{des} + K(F_{des} - F)\right) \quad (5)$$

where K is a 6x6 diagonal matrix of constant gains. Closing the loop at the hand level allows a better measurement of F, although the rejection of friction torques cannot be effective and is strongly dependent on the arm configuration q.

Summary on arm exoskeletons

We are assisting nowadays to an increasing interest and number of applications for upper limb exoskeleton systems, in particular in the area of rehabilitation of motor disorders and human power augmentation. The frontiers of research in the field of exoskeletons are now moving towards new human-robot interfaces that can intuitively couple human intention to robot behavior for assistance to the human, exploiting biometric signals, ranging to mention just the most important one from EEG, through Brain Computer Interfaces [24], eye-tracking [25] and last, but not least EMG [26].

EXOSKELETONS FOR THE HAND

Hand exoskeleton version I (1994)

Hand Exoskeletons found a relative success in 1990s, being conceived as specific hand masters for teleoperation tasks and haptic interfaces for controlling the interaction with VE [27].

The first hand exoskeleton prototype developed at PERCRO in 1994 consisted of four finger exoskeletons each one exerting forces to the phalanges of the hand's fingers (little finger excluded).

A single finger exoskeleton, which kinematic structure is represented in Fig. (14), consisted of four links con-

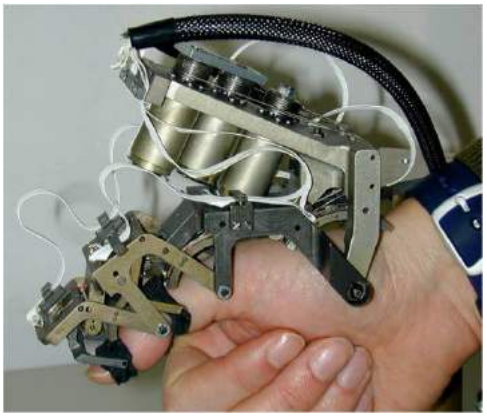

Fig. (14) — Finger Exoskeleton Version I (1994)

nected by revolute joints disposed as the joints of each finger. Each joint axis of the finger exoskeleton was designed in order to approximate the instantaneous position of the flexion-extension axis for each phalanx. At the metacarpo-phalangeal joint a passive abduction-adduction movement has been also integrated.

The actuation system for one finger exoskeleton was based on three DC servomotors and associated tension tendon-based transmission systems. Each tendon was intended to pull on the middle point of each phalanx of the finger in order to execute the extension movement; on the contrary, at each joint, the flexion movement was obtained by a passive torsion spring integrated on the joint axis. The three motors were located on a cantilever structure fixed with the base frame of each finger exoskeleton. Rotation sensors, based on plastic conductive technologies, were integrated at each joint while force sensors, capable of recording the interaction force between the exoskeleton structure and each phalanx, were located directly on the dorsal surface of each phalanx link.

Particular attention was devoted to design a specific kinematic structure for the thumb exoskeleton. One of the critical factors encountered during the design of the system was the constraints in terms of weight and volumes needed to allow good maneuverability of the hand. Despite these constraints, the ranges of motion of the fingers with the worn exoskeleton can be considered very close to those of a free human hand.

In terms of mechanical performance, the hand exoskeleton system obtained a maximum extension force of 0.3 N, being the force sensor range of -0.5 N to 3.0 N. Force resolution was 0.0025 N, while the force feedback bandwidth was 0.5 Hz with an angular displacement of 90 Degrees for all the 3 DoFs.

A new version of the hand exoskeleton was addressed in 1997 by considering a different transmission system including an agonistic-antagonistic tension tendon-based transmission system for each joint of the different finger exoskeletons, a recent prototype was completed in 2009 ([28], see Fig. (15)). .

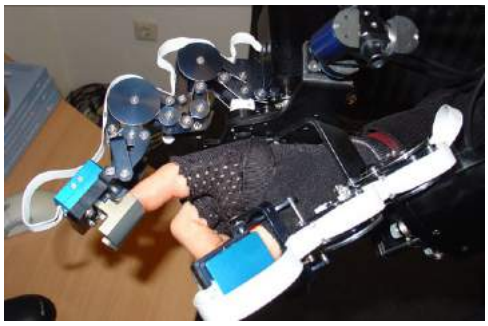

Fig. (15) — Hand Exoskeleton Version III

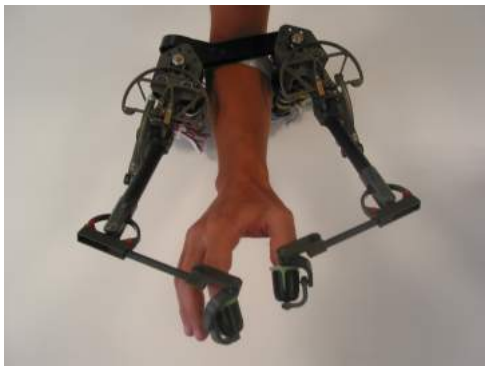

Fig. (16) — Hand Exoskeleton Version II

Hand exoskeleton version II (2001)

Psycho-perceptual studies performed by [29] demonstrated the importance of providing multiple points of contact on the operator's hand during haptic interaction with Virtual Environments. Based on these assumptions, the design of an innovative hand exoskeleton system was conducted at PERCRO by considering a different approach with respect the one utilized for the design of Version I (1994) considered expensive both in terms of complexity and cost.

The new design approach consisted in devising a mechanism that, despite a reduced number of DoFs, could be capable of providing force feedback components on thumb and index fingertips of the hand [30].

The design of such an innovative, wearable haptic interface consisted of two serial mechanical limbs, called finger mechanisms, each one with 3 actuated rotational joints. The base frame of each limb was attached to the user's forearm, while their end-tips were coupled through spherical passive joints to the thumb and index fingertips respectively.

The user, by inserting his/her fingertip in a thimble, was able to haptically perceive the generated forces. The spherical joint placed at the end-tip allowed the mechanism to reach the user's fingertip with any orientation.

As an ergonomic requirements of such a wearable system, the motors were located on the forearm of the user, as represented in Fig. (16).

In order to decrease the weight of the moving masses

and enhance the backdrivability of the system, a tendon driven bidirectional transmission was adopted, which enabled to locate the brushed DC motors at the base.

The kinematics of each finger mechanism was composed by three mutual orthogonal revolute joints, actuated with a tendon drive by three actuators placed at the base. The design of the system has considered an optimization strategy for the kinematic dimensioning based on a desired performance to be achieved over the entire hand workspace. The influence of singularities was analyzed functionally to the task [31].

APPLICATIONS TO GRASPING IN VIRTUAL ENVIRONMENTS AND TO VIRTUAL PROSTHETICS

Grasping represents a fundamental aspect of the simulation of physical interaction in virtual environments. The anatomical and functional design of the human hand is based on grasping.

Grasping possibilities can be mainly divided into the two group of power grips and precision grips, depending on the task to be performed, i.e. according to the size and weight of the object to handle. Large and heavy objects are grasped with a power grip, since objects get enclosed securely between palm and the closed fingers and the thumb, and this grasp is especially useful for positioning, turning and moving of an object. Small and light objects are grasped with a precision grip, so that objects are grasped with the fingers in opposition with the thumb without involvment of the plam. With the precision grip, objects can be grasped in a more sensitive way, and be positioned, turned and moved more precisely than with the power grip [32].

The number of grasping fingers influences the grasping stability: the grip used for grasping an object is normally decided automatically by the subject according to the size, weight, trajectory and direction of approach to the object.

Different models of virtual hands have been developed to allow the implementation of the grasping affordance in virtual environments. It is clear that modeling a virtual hand is a complex issue, and that some simplifications are needed to achieve a real-time physical simulation of grasping.

The development of a virtual hand model is developed observing the anatomical and biomechanical parameters of the human hand and simplifying it to achieve an approximation of real hand movements, fulfilling the needs of the application and the implementation hardware as well as software constraints. The physical model adopted at PERCRO is a reduced model to 18 limbs instead of the 29 carpal and metacarpal bones of the human hand. This is mainly a result of merging the metacarpal bones to one single rigid object. From the software side, rigid body dynamics are used instead of soft body dynamics.

The computational approaches based on a physical modeling of the scene and of the hand/object interaction are of particular interest. One common implementation of a physically-enabled virtual hand model [33, 34, 35] is based on the implementation of three separated and interconnected hand models corresponding to the three main functionalities required to the virtual model, respectively called the visual, the physical and the tracked hand model.

Fig. (17) — Example of simulation of a virtual hand

The visual model of the hand is used for the purpose of displaying a visual representation of the scene to the subject, see for instance Fig. (17), with the implementation of shapes, textures and local joint deformation providing a realistic visual feedback to the user.

In order to interact with the environment, a physical model of the hand with some internal compliances should be associated to the visual model. The physical model can interact with the objects and resolve the interaction force between the fingers and the virtual objects during grasping, associated to the solution of a hyperstatic problem, thanks to its own internal compliances. The physical model usually presents a simplified geometry of masses ad volumes to allow the real-time resolution of the associated dynamic equations and the determination of collision detection with external objects, as shown in Fig. (18) where both models are represented. The movements of the visual models are driven by the physical model, whose configuration is in turn determined by the tracked hand model under the direct control of the user.

The control input of the user can be generated by a dataglove in the most general case and is directly mapped to a posture of the tracked hand model. The connection between the physical and the tracked hand model is made by means of a virtual coupling [36], consisting in a series of damping and elastic connections between the limbs of the two models.

Recent studies [37, 38, 39] demonstrated that a few input control variables, named postural synergies, can account for most of the variance in the patterns of hand configurations during manipulation and grasping tasks. This equivalently mean that not all the joint angles of the hand are controlled independently from each other

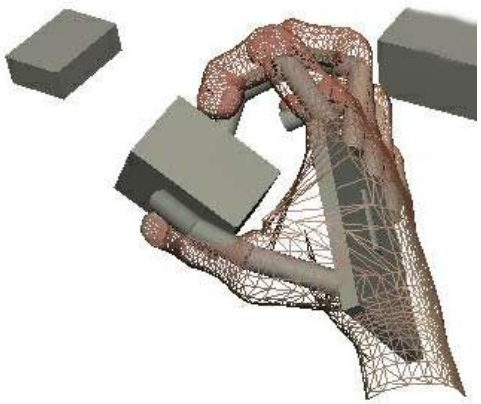

Fig. (18) — Example of interconnection of the physical and virtual hand models

in shaping the hand to grasp different objects. Postural synergies are expressed as a linear combination of the hand joint angles and form a reduced set of independent variables that can be used to suitably approximate the hand postures in object manipulation [40].

When the grasping posture of the hand is determined by a control input of reduced dimension, e.g. surface EMG or finger tips position, the mapping of control inputs to the grasping postures can be implemented by means of the grasping synergies, expressing the joint angles of the tracked model of the hand as a linear combination of the available control inputs. This technique has already been successfully employed to simplify the control of virtual hands [40] in the case of reduced control input deriving from the finger tips position only.

An interesting extension of the above concept is the case of the development of virtual reality scenarios for testing the capability of controlling new advanced prosthetic hands, where only EMG signals are available as potential control inputs. In the case of most powered transradial prostheses, typical set-ups for the myoelectric control of opening/closure of the prosthetic hand are based on the placement of single unipolar electrodes on the forearm of the subject, exploiting the amplitudes of surface electromyography (EMG) signals from the forearm flexors and extensors to control the opening and closing of the prosthesis.

It is however still unclear whether the residual muscles of the forearm following amputation can provide stable EMG information for accurate real-time control of multifunctional transradial prostheses. The control of myoelectric prosthetic hand using EMG signals from residual muscles of the amputee requires that the conditions of the muscle of amputation are good and the training of skills required to control and use the prosthesis by EMG. As this is not usually, a possible approach is that of carrying out the training in virtual environments. For instance, grasping of rigid object requires just large EMG signals, but for fragile objects EMG signals should be within a certain region to produce appropriate grasping force.

The outputs from the surface EMG electrodes are usually employed to train a classifier of a network for pattern recognition and to drive the joints of fingers of a virtual prosthetic limb [41]. A recent study by [42] has shown how real-time pattern recognition can be used with good results for the control of a virtual prosthesis. In [43] a successful application of a Virtual Environment is shown for testing the ability of controlling a myoelectric prosthesis.

In combination with force-feedback wearable devices, Virtual Environments represent an ideal tool for testing myoelectric control of prostheses. In particular, we consider that the technologies presented above can be suitably used to achieve realistic simulation of grasping procedures with the purpose not only of training the ability of controlling hand prostheses, but also of testing new control strategies of more elaborate prostheses.

CONCLUSIONS

The different design solutions for arm and hand exoskeleton systems developed at PERCRO since early 1990s have been presented. The design of robotic exoskeleton systems can be considered as one of the main lines of research carried out at PERCRO so far. While 20 years ago only few attempts were present, nowadays the term wearable robotics identifies a concrete area of development in Robotics and we believe that for specific application domains there will be an increasing request for future developments.

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