UNIVERSITY OF SOUTHAMPTON

AN INTELLIGENT HAND PROSTHESIS AND EVALUATION OF PATHOLOGICAL AND PROSTHETIC HAND FUNCTION

Colin Michael Light

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UNIVERSITY OF SOUTHAMPTON ABSTRACT
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Commercial hand prostheses provide the user with insufficient dexterity and functionality due to the highly restricted number of prehensile patterns that may be achieved. Demographic studies show that the potential market for upper limb prostheses is largely stable, and with the identification of the functional differences and inadequacies of existing prostheses, users are increasingly dissatisfied with the status quo.

The six degree of freedom Southampton-Remedi hand has been developed to address this need. The mechanical adaptability of the lightweight prosthesis provides a wide range of grip types with a greater degree of stability than produced by any conventional device. This is due to the ability of each digit to independently contribute to the integrity of the grip.

The command and coordination of more than a single device or function is difficult, and frequently increases the cognitive burden on the user. The optimal use of multifunction prostheses lies in the synergistic control of several actuators without increasing the number of inputs that a user must independently initiate. This has been achieved by the development of a hybrid SAMS-UNB controller that enables the user to directly implement prehensile patterns from their myo-signal whilst the process of maintaining a secure grasp remains automated.

The effectiveness of the new prosthesis and controller must be quantified in terms of its functionality. However there is little or no conformity to a standardised and objective procedure for the assessment of either pathological or prosthetic hand function. The Southampton Hand Assessment Procedure (SHAP) has been designed to account for these shortcomings and therefore allow the evaluation of hand function in the clinical setting. The outcome measure is a contextual rating of functionality (relative to that of 'normal' hand function), which enables the clinician to initially determine the subject's disability, and subsequently monitor their performance throughout a course of treatment or rehabilitation.

Despite the notable advances made by SHAP, clinical assessment techniques of pathological hand function are crude by comparison to that of gait analysis. A pilot study of upper limb motion analysis suggests that the principles of assessment, although more complex than that of the lower limb, may be applied in a similar manner, which would ultimately result in stark change to the current methods of clinical diagnosis and assessment of upper limb disorders.

In memory of my father –
I'll see you on the city hall steps

Contents

Glossary

ADC	Analogue to Digital Converter
ADL	Activities of Daily Living
ANN	Artificial Neural Network
BP	Body-Powered
CMC	Carpometacarpal (joint)
DSP	Digital Signal Processor
EMG	Electromyogram
EPP	Extended Physiological Proprioception
FES	Functional Electrical Stimulation
FIFO	First In First Out
IP	Interphalangeal (joint)
ISR	Interrupt Service Routine
MAV	Mean Absolute Value
MCP	Metacarpophalangeal (joint)
MES	MyoElectric Signal
MOSFET	Metal Oxide Silicon Field Effect Transistor
MUAP	Motor Unit Action Potential
PCB	Printed Circuit Board
PWM	Pulse Width Modulation
PZT	Piezoelectric ceramic consisting of mainly Lead (Pb), Zirconium (Zr) and Titanium (Ti)
RMS	Root Mean Square
ROM	Range of Motion
SMA	Shape Memory Alloy
USM	Ultrasonic Motor

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Chapter 1

Upper Limb Prosthetics

1.1 Introduction

Prosthetics is an emotive issue. The absence of a limb, either by amputation or by congenital defect, is deemed a highly visible 'disability'. Despite this prejudice, many of those with congenital defects do not consider themselves 'disabled', or feel in need of a prosthesis due to their ability to adapt to the surrounding environment or task. The potential use of functional prostheses involve an interface between man and machine that is sometimes viewed as unnatural or unappealing. However others often wish to use a prosthesis in order to regain at least some of the functionality, or more simply, the appearance lost with the limb. Hence there are many issues surrounding the use of a prosthetic device, regardless of its design or indeed function, that predetermine the potential market for any artificial limb.

The desire to regain additional function through the use of a prosthesis is nevertheless fundamental to many users. Given the increasing cost of assistive devices (such as prostheses and orthoses), it is understandable for health policy makers to expect a return on the rehabilitative care investment. According to American studies [1], $1 spent on assistive device technology returns $11 in benefit to society through the individual leading a more productive and independent life.

Yet the technical difficulties involved in the replication of natural hand function are immense. The human hand has 27 bones, over 30 muscles and is arguably the most complex part of the human body [2]. An artificial hand must conform to many

constraints (anthropomorphism, weight, size etc.) whilst simultaneously realise a range of conditions such as natural movement, optimum grip forces, and reliability.

This chapter details the current status of upper limb prosthetic development, and the aforementioned constraints and consumer trends and opinions that affect successful prosthesis design. Commercial devices fail to meet these consumer requirements, which are addressed by the development of the novel lightweight multiple degree of freedom Southampton-Remedi hand prosthesis. The subsequent chapters detail the mechanical design of the hand and implementation of the intelligent control system that provides automated and adaptive prehension. This enables the user to maintain only supervisory control with little cognitive effort. In order to provide a method of evaluating the efficacy of the prosthesis an extensive review of existing techniques was carried out (see Chapter 4). This highlights little adherence to medical outcome measurement standards by the majority of existing natural or prosthetic hand function assessments – a shortfall addressed by the development of a new objective and standardised hand assessment procedure (see Chapter 5). Clinical case studies using this "Southampton Hand Assessment Procedure" (SHAP) are detailed (in Chapter 6) along with a pilot study into the application of motion analysis as a more precise clinically-effective tool for upper limb assessment. Although the Southampton-Remedi hand cannot be fully assessed using SHAP due to the limitations of the development hardware, a partial assessment reveals the prosthesis to be capable of multiple adaptive prehensile patterns.

1.2 Historical Development of Upper Limb Prostheses

The American Civil War (1861–1865) marks the first example of modern warfare and subsequent rehabilitation of amputee veterans. Revolutionary changes in prosthesis design were made during the post-war years but all remained mechanical in nature. The modern era of technological upper limb prosthetic development commenced at the end of the World War I with the number of amputees (upper and lower extremity) exceeding 42,000 [3] in the UK alone. Prosthetic development was then dominated

by European researchers until World War II when the USA took an increasingly active role in establishing research programmes.

Despite the advances made, externally powered prostheses were still rare over three decades later. In the early 1970s Childress evaluated the use of both body powered and externally powered prostheses [4], concluding at that time, that powered prostheses were not widely available enough in the USA on clinical trial to be significant to potential users. However he also noted that these devices could be used at a subconscious level to remove the conscious supervision of the prosthesis from the user, and therefore emphasised their future potential and significance.

It was apparent that powered prostheses required high speed digits to curl around an object, and then act as low excursion, high force generators to ensure a secure grip. This "synergistic" approach was later to be adopted in externally powered devices [5]. The concept resulted in a "synergetic prehensor" using two motors (one low-speed high-force drive and the other a high-speed low-force drive) working independently towards the common prehensile goal [6].

Towards the end of the decade Sörbye in Örebro, Sweden began a progressive approach to prosthesis fitment when a three year old with a trans-radial congenital absence became the first child to be fitted with a myoelectrically controlled powered prosthesis [7].

This milestone typified an era of substantial development in 'myo-prostheses' that spanned several decades. The tendon driven Belgrade prosthetic hand [8], the Utah arm [9], and the Stanford/JPL hand [10] were all advanced multiple degree of freedom, electrically powered hands. Much of the prosthetic research and development of the time later saw application in anthropomorphic robot end effectors. However multiple degree of freedom hand prostheses saw popularity amongst researchers, and resulted in devices such as the Waseda [11], and Vaduz (or "French Electric") [12] hands but with little clinical success. The multifunctional SVEN hand [13] showed a significant advance in prosthesis control theory based on EMG pattern recognition from six active surface electrodes. Similarly, research at the

University of Southampton concentrated on the development of a prehensile control scheme [14] that would result in an intelligent multiple-axis device [15, 16]. Rather than requiring the user to maintain multiple degree of freedom control as with the SVEN device, the Southampton Hand philosophy centres on transferring the low level reflexes of prehensile control from the user to the prosthesis [17, 18]. This hierarchical control scheme was the focus of long term research, and has been implemented more recently in the European collaboration MARCUS [19] and in the two degree of freedom Leverhulme-Oxford/Southampton Hand (see Figure 1.1) that is currently undergoing clinical trials [20, 21].

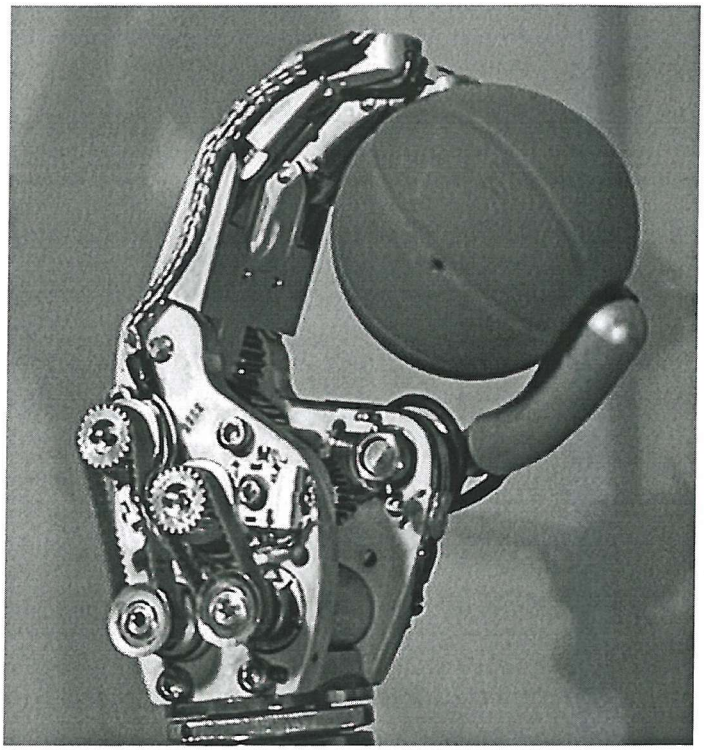


Figure 1.1: The Leverhulme-Oxford/Southampton Hand MyoProsthesis

1.2.1 Drive Systems

The majority of these devices used electrical drive systems, although attempts have been made at employing pneumatic power [22, 23], often through adaptations of

McKibben muscles[1] [25, 26, 27]. These were designed to overcome the poor power to weight ratio obtained with traditional DC motors and the excessively large battery packs that often accompanied them. However, a pneumatic drive requires a power source (usually a CO_2 canister) that is of limited availability, and the device itself is often restricted either due to legislation or due to the operating environment. Hence, a small potential market, inaccessibility to a reusable power source, and the disadvantage of notable operating noise, has precluded the widespread use of these actuators.

1.2.2 Control Suspension

Irrespective of drive systems, externally powered devices have tended to dominate prosthetics research over the last three decades. Nevertheless body-powered (BP) prostheses have repeatedly been shown to allow better positioning control than the use of multiple velocity-control sites (as is common in myoelectrically controlled devices). Trans-radial BP prostheses possess a harness that is usually worn across the back, so that gleno-humeral flexion of the opposing shoulder causes a cable to actuate the hook or hand (see Figure 1.2). This motion enables the user to feel device actuation through cable tension and harness position, thereby giving the subject direct feedback and potential control of the position, velocity, and prehensile force of the device. This form of mechanical feedback is known as extended physiological proprioception (EPP), and was originally proposed and adopted in prosthesis design by Simpson at the Princess Margaret Rose Hospital, Edinburgh [28]. Further investigation of this control technique by Childress [29] demonstrated the superior performance of cable-linked, force-actuated position servo control, over the velocity control of a conventional powered prosthesis. It is notable that the implementation of force feedback from an externally powered device to the user is difficult to achieve as part of an integral control system, and accounts for its absence in all commer-

[1]Developed in the 1950's, these actuators consist of an internal rubber bladder surrounded by a braided mesh shell (constructed from fibres of high longitudinal stiffness). When the bladder is pressurised, the mesh (attached at either end to tendon-like fittings) causes the actuator to shorten in length according to the increase in volume, and thereby exert tension through the tendons [24].

cial prostheses. One study suggested that 23% of body-powered users experienced feedback through tension in the cable and the position of the harness, but perhaps of more interest, is that 33% of myo-prosthesis wearers also claimed feedback from the device [30]. However, this was attributed to the more intimate socket fit for myo-prostheses and a more natural use of muscle control, rather than due to EPP.

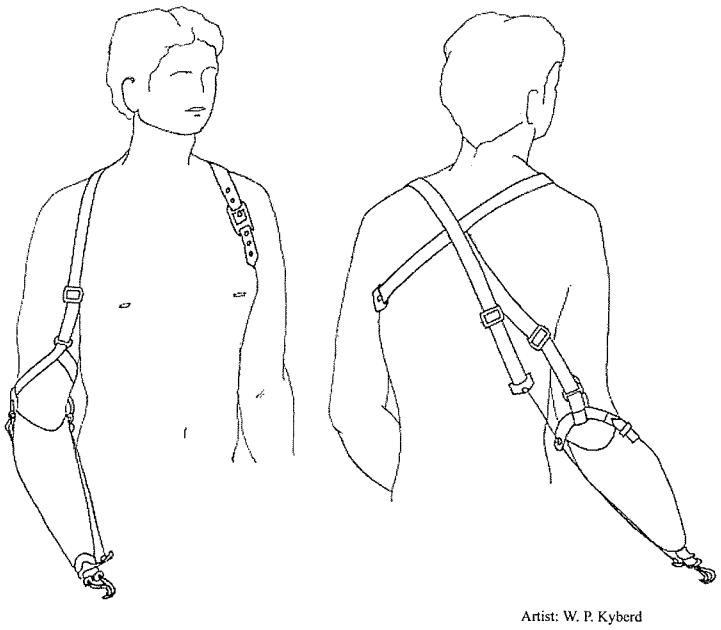
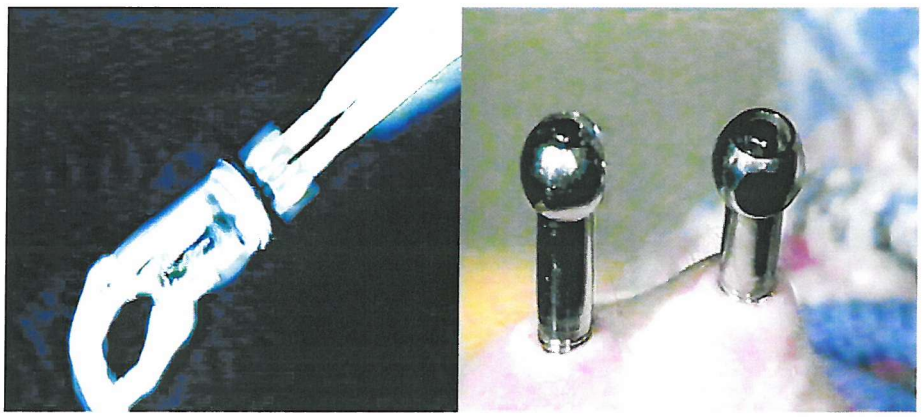


Figure 1.2: Body Powered Split Hook Configuration

Both body-powered and myoelectrically controlled prostheses are devices worn by the user in a non-intrusive manner. However Childress has also investigated the use of tunnel cineplasties [31] to allow direct prosthesis connection to a muscle via a controller, thereby potentially allowing a form of integral EPP. There remains a question about employing intrusive surgical procedures to provide enhanced control of a prosthesis, as many users may consider this an unacceptable compromise, with a move away from the utilisation of the prosthesis as a tool, toward a physical and biological integration of the device. The additional risk of infection at the skin interface requires that such procedures would have to be demonstrably advantageous, resulting in significantly improved functionality over current methods.

Nevertheless titanium implants have been used in oral and craniofacial recon-

structive surgery in Sweden since 1965 [32]. Pioneered by Brånemark, this osseointegration procedure has been employed in amputation prostheses for the last decade. This technique has notable advantages over existing socket methods by improving the stability and dynamic motion of the prosthesis, as well as eliminating soft tissue problems caused by changes in stump volume [33]. A high level of osseoperception[2] is described, indicating a level of sensory feedback through the implant that may improve functionality in a similar manner to EPP. The disadvantages are the frequent complication of superficial infection at the implant/skin interface. In one study, loosening of the fixture or deep infection also has been encountered in around half of the patients [32], but results have seen a 76% success rate following further treatment. The technique has mainly been used for lower limb amputees (at the transfemoral level), although upper extremity procedures have been undertaken (see Figure 1.3). Osseointegration represents a new approach to the fitment of prostheses and has distinct advantages over existing practice, but does not overcome the fundamental needs of the client for a terminal device of greater function.



(a) **(b)**

Figure 1.3: Osseointegration: (a) X-Ray of Titanium Implants and Prosthesis, (b) Titanium Implants at the Skin Surface

Although prosthetics research during this modern era has made notable steps

[2]A term used by the Brånemark team to describe the feedback, through the skeletal frame, from the terminal device.

in improving the weight, sound, appearance and reliability of the devices, little has been achieved in a clinical setting to enhance their function.

1.3 Current Status of the Industry

1.3.1 Identification of the consumer

Ensuring the efficacy of prosthesis design can only occur by attempting to fulfil consumer requirements. The market for upper limb prostheses is small, but the demographics of this group highlight important differences that reflect the need for consumer-driven design. In addition, factors such as the type of prosthesis selected, functional performance and duration of use affect design specifications.

Demographic Trends

Due to the small population groups affected, consensus opinion concerning the demographic trends are difficult to achieve. However, studies have shown that the number of subjects warranting upper limb prostheses has been declining over the last 40 years [34, 35], and in the case of individuals with amputations, may fall to zero in the early parts of the 21st Century (ranging from 2003–2024 depending on statistical predictions). Realistically this scenario is improbable and the trend is likely to stabilise at a low level prior to this time. The reasons for this decline may be attributed to the reduction in industrial accidents due to factory automation, and the improvements in automotive safety, which have traditionally been the highest causes of amputation [34]. In addition, advances in medicine have produced a more successful approach to the treatment of trauma cases that would previously have resulted in amputation. However, the declining trends noted in these studies have been compiled from developed countries during peacetime, and therefore are not necessarily reflective of a global trend. In the mid-1990s one study from a central European country has shown that of the 89.5% of amputations resulting from accidents, 47.7% were due to war injuries [36]. The large effect of all traumatic injuries is also reflected in the gender distribution that shows males with three times the

frequency of amputation than females [34].

Congenital deficiencies typically represents 30%-50% of upper limb prosthesis users [37]. UK government statistics and other clinical studies suggest an estimated UK population with congenital absence of between 4300 and 7000 at current population levels [35]. As this group has remained largely consistent in number over the last half-century (ignoring the effect of the Thalidomide drug), then it is reasonable to assume that the global market for upper limb prostheses will continue to maintain its current rate.

Principal Type of Prosthesis

The users' choice of prosthesis type appears to follow national trends. One European study shows that the UK and Italian populations predominantly select mechanical devices (such as the body-powered split hook), whereas the Swedish prefer to use electrically powered hands [37]. It is likely that these biases are exhibited in most developed countries where the choice of device is frequently dependent upon socio-economic factors such as public exposition to technology, the preference of local clinics, and insurance or government policies. Despite this, studies show an approximately equal proportion of users select cosmetic (passive) hands as functional ones [35].

Prosthetic Use

Most users wear their prosthesis (irrespective of type) for over 8 hours a day, although identifying the amount of actual 'use' during this period is difficult and entirely dependent upon the subject's opinion [36, 37, 38]. Technically it is feasible to obtain a more accurate estimate by direct measurement within the prosthesis controller, however this raises moral questions concerning the monitoring of a subject.

The extent of use, and indeed the type of prosthesis worn, depends on the user's lifestyle. A subject whose occupation requires heavy lifting, or a dirty work environment often uses a cable operated device, whereas electrically powered prostheses are most often worn during light activities in relatively clean surroundings [38]. Powered

devices are frequently cited as a social preference due to better cosmesis, although cable operated hands or cosmetic prostheses are both preferred to a hook terminal device.

1.3.2 Achieving Stable Prehension with a Prosthesis

Despite the advantages and disadvantages that exist between body-powered devices and myoelectrically controlled prostheses, the prehensile function remains virtually common to all. Consequently it is appropriate to examine the method of achieving a secure grip with a prosthesis. As prehension can be defined as "the application of functionally effective forces by the hand to an object for a task" [39], the number of grip scenarios (and therefore the distribution of force) that a prosthesis may achieve is fundamental to its prehensile ability.

In order to achieve multiple grip patterns, the artificial hand must possess more than a single degree of freedom. Although multiple digit prostheses have been developed [8, 9, 10, 40], few have ever reached clinical trial. Moreover, the development of a device with mechanically enhanced prehensile function, such as the "Prodigits" hand prosthesis [41], is not sufficient in itself. The user must be able to access the device's grip potential without any additional psychological effort. This is typified by the development of the SVEN hand prosthesis [13, 42] in the 1970s that utilised the perception of the phantom-limb in amputees to allow the user to perform six independent movements (grasp, release, pronation, supination, wrist flexion and extension). Pattern recognition techniques were used to isolate the relevant muscle group and control the multifunction prosthesis. Although this device saw commercial development in the form of the ES hand, it was not widely accepted. The requirement for a socket of perfect fit (to ensure accurate siting of the six electrodes), the difficulty in isolating unique movements from the user, and the need for a large electronic controller highlight the obstacles to conscious control of a multifunctional prosthesis.

Essentially there are three hand characteristics that are necessary to ensure sta-

ble prehension for a range of tasks [39]: pad opposition, palm opposition, and side opposition. These characteristics are less specific than the hand's functional prehensile patterns (as discussed in section 4.2.2), however these general forms explain the ability for multiple digit opposition that ultimately provides secure grip.

Pad opposition requires small forces, fine motor control and precise sensory information. Its form is based upon the object's shape, the number of fingers used, the opening size of the hand, which digits are in contact with the object, and the flexed or extended state of each digit. Palm opposition allows the hand to equalise, or indeed exceed forces from an object whilst ensuring stable grasping. Side opposition is a compromise scenario between the power exhibited in a palmar grip, and the precision of pad opposition. It utilises some sensory feedback and a medium range of forces to ensure that the thumb pad remains in contact with the object [39]. These are the characteristics that a truly functional multiple degree of freedom prosthesis must achieve in order to provide stable prehension for the widest range of everyday activities. They are also the functional properties not shown by existing devices.

1.3.3 Functional Differences in Prostheses

Commercially available devices consist of cosmetic (or passive) devices, body-powered mechanical hooks (and hands), and powered prostheses (usually myoelectrically controlled). There are only a handful of myo-prosthesis manufacturers supplying a global market and although Otto Bock dominates this arena (see Figure 1.4), RSL Steeper maintain a near monopoly in the UK. At this time no commercial manufacturer of hand prostheses provides a device capable of multiple prehensile patterns. Although devices with such potential do exist [41], without the necessary intelligent control systems (as discussed in section 1.3.2) and more importantly, commercial backing, they are unlikely to see widespread clinical use.

Prostheses (irrespective of type) are most commonly used to perform a stabilising action during normal tasks. Hence there is a clear indication that even cosmetic devices are used functionally when evaluated outside of the clinical setting. It has

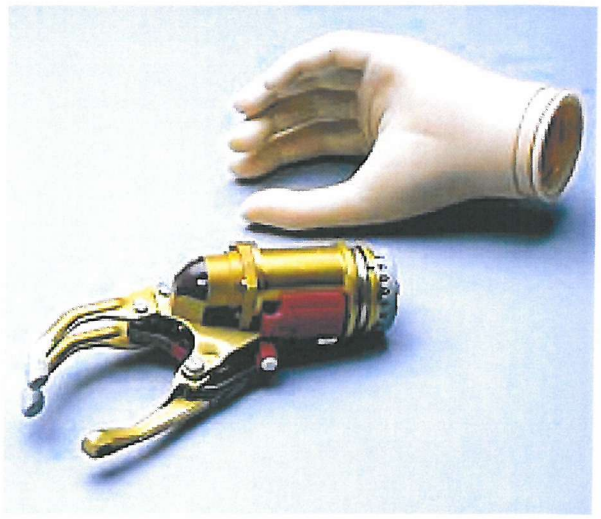


Figure 1.4: Otto Bock Electrically Powered Prosthesis

also been noted that unilateral amputees may demonstrate high levels of skill in performing tasks with the prosthesis when under assessment, but are more likely to use the natural hand to carry out everyday activities [43]. Despite this supplementary functional role of the device, consumers who wish to regain some form of upper limb function must choose either body-powered hooks, hands or powered prostheses. The implications of this choice are detailed in Table 1.1 [44, 38].

Child Prostheses

Clinicians in the UK fit prostheses to children with congenital absence according to key development stages in their growth. Typically when a child is able to sit upright then a passive prosthesis may be fitted, whilst when able to crawl (from around 14 months) a myoelectrically controlled device may be considered [46]. Sweden maintains a long and progressive approach to prosthetic research and development. Passive prostheses are fitted there at 3 to 6 months to achieve body symmetry from an early age and ensure that the child becomes accustomed to wearing a socket. The additional benefits are that equal limb length enables a more normal crawling pattern in the child and the parents also gain significant psychological benefit from

Hooks	**Myoelectric Hands**
Offer precision prehension	Provide crude 3 point prehension
Digit moves in an transverse plane (poor cosmesis)	Thumb operates in opposition to first two digits in sagittal plane (better cosmesis)
Table top object grasping usually easy to accomplish	Table top object grasping is difficult and requires gross upper limb movements
Good visibility of objects in all positions	Visibility at an optimum when hand is supinated
Not very stable when grasping large contoured objects	Medium sized round objects are grasped with relative stability
Light	Heavy (over 3–4 times that of hooks)
Easy to operate	Conscious user effort is higher (perhaps as much as 60% greater [45])
Requires unnatural movements to operate but provides mechanical feedback	Relatively intuitive operation but no feedback
Requires high operating force, and grip strength is dependent on user	High grip force (30–120N) that is device dependent.
Low cost, rugged and reliable	High cost at fitting, requires specialist maintenance, and not as durable.

Table 1.1: Functional Comparison of Prosthetic Systems

the care their child receives [7]. In general, users overestimate the length of the residual limb whilst wearing the device [47], and this effect is considerably reduced when the prosthesis is removed. As a sense of limb length is essential for motor co-ordination, the perceptual adaptation of the user over time is crucial to successful artificial limb control, and therefore verifies the efficacy of early fitment.

Young children wearing body-powered prostheses frequently are hindered in achieving stable prehension [48]. Voluntary opening devices often cause children difficulty in overcoming the resistance of the spring or elastic band (that acts in opposition to user actuation and applies the grip force). Voluntary closing devices require the wearer to maintain a strong contraction over an extended period of time

[49]. Terminal devices have been designed to reduce this operating force, whilst maintaining a firm grip [45, 50] as the myoelectrically controlled hands have been viewed as too heavy, unreliable, and expensive (both to purchase and to maintain). However there is strong evidence that myo-prostheses should be fitted from a young age [7] as the introduction of such devices to (congenital) subjects after 20 to 25 years results in a high risk of rejection. The continued advancement of the technology incorporated into these prostheses has also sought to improve the cosmesis, reduce the weight, and allow reliable myoelectric control of these devices [26].

1.3.4 Attitudes and Trends within the Field

Given the limited function of the commercially-available devices, it is perhaps unsurprising that new research into the field has attempted to determine current attitudes of the consumer [37] and the rest of the industry [51].

In 1992 the sole UK manufacturer of upper limb prostheses suggested that "myo-prostheses are as well developed as the market warrants" [51], and given minimal change in product design it is apparent that this opinion continues to date. Perhaps a more alarming statement is that the manufacturer noted that "there is a higher profit made in the UK from the servicing of myo-electric prostheses than there is from the sales of the devices". Despite consumers and clinicians alike noting the potentially unreliable nature of these devices [50], there appears little motivation for change within the current UK system.

Clinicians are aware of the value of myoelectric prostheses, especially in the treatment of younger amputees and congenitally deficient children [51]. However these clinicians have highlighted the limited functionality produced by the single degree of freedom devices, and expressed a desire for greater dexterity. Both consultants and suppliers have also noted that the weight of the prosthesis (specifically at the distal end) is a substantial factor in the fatigue of the user and a potential reason for subsequent disuse. As the device is worn on the end of a closely fitting external socket, the lever-arm created by its weight can obstruct blood flow in the underlying skin

and thereby propagate symptoms that range from discomfort to skin breakdown.

1.4 Consumer Driven Development

1.4.1 Establishing Consumer Requirements

The restricted choice and function of existing prostheses has been cited as one of the reasons for user dissatisfaction [52]. The socket interface is fundamental to the successful fitting and use of the prosthesis. Hence other design deficiencies include sweating of a subject's remaining stump within the socket [36], loss of sensation [53], and the excessive weight of the device [37, 51], although the time between amputation and fitting may impact on their significance [36]. Education and employment status are also noted as factors affecting prosthesis use, yet pre-amputation hand dominance appears to have no significant effect [43, 53]. The appearance and sound of the prosthesis remain high among users' concerns. Hence the functional limitations of the device, and current fitment methods result in the poor acceptance of myo-electrically controlled prostheses and may account for an approximately equal proportion of users selecting cosmetic passive hands as any form of functional device (either body- or externally-powered) [35].

The development of a prosthesis that may address the issues raised by consumers must therefore focus on achieving a balance between optimum engineering input[3] and client satisfaction. The application of industrial design techniques, the Quality Function Deployment (QFD), to upper-limb prosthesis development has provided client-focused design criteria [52].

This multidisciplinary approach utilises customer and engineering input, the relationship between the functional requirements and engineering design parameters, as well as evaluations of existing products. These factors are constructed within a *House of Quality* template that results in a set of goals where engineering effort is focused to suit the needs of the consumer.

This study [52] resulted in a series of optimum design parameters in order of

[3]This must include factors such as cost per degree of complexity and functionality achieved.

priority:

1. Number of possible grasping patterns
2. Visual feedback
3. Grasping surface compliance
4. Maximum opening width

As discussed previously, there clearly is a demand for greater function from the prosthesis, and as validated by the QFD study, increasing the number of prehensile patterns is considered to be of paramount importance.

1.4.2 Conclusion

Although prosthetic research and development has progressed markedly over the past few decades, current commercial hand prostheses are clearly providing insufficient dexterity and functionality[4] due to the highly restricted number of prehensile patterns that may be achieved. Demographic studies show the potential market for upper limb prostheses to remain largely stable, and with the clear identification of functional differences and inadequacies of existing prostheses, users are increasingly dissatisfied with the status quo.

Hence consumer requirements show a key demand for the development of a lightweight multiple degree of freedom hand prosthesis without increasing the current physical or psychological burden on the user.

The Southampton-Remedi hand demonstrates the ability to produce a prosthesis that is mechanically capable of implementing several grip scenarios. The original use of a myoelectric signal classifier (the UNB system) in conjunction with an intelligent control system (the Southampton Adaptive Manipulation Scheme) further illustrates how the prosthesis can be used without excessive control demands being placed upon the user. The need to evaluate the efficacy of prostheses (and indeed dysfunctional

[4] A term that may be defined as a hand's 'suitability to the task', see section 4.1.1

natural hands) has been addressed through the development of the Southampton Hand Assessment Procedure, and has led to an innovative pilot study that applies gait analysis techniques to that of functional hand assessment.

Chapter 2

Adaptive Prosthesis Design

2.1 Introduction

The limited functionality exhibited by commercial devices remains rooted in their single degree of freedom format. Users continue to request an increase in the number of possible grasping patterns and an improvement in the visual feedback of the object in the hand [52].

The hypothesis for the development of a new Southampton hand is that a multiple degree of freedom prosthesis should provide greater stability in prehension with minimal grip pressure than a single axis device [54, 55]. Hence given all other criteria being equal, a prosthesis with maximum mechanical adaptability should result in maximum dexterity (providing that no additional control burden is placed upon the user).

The complexity of the natural hand cannot be replicated in a prosthesis due to the prohibitive size and weight of the required drive systems. Nevertheless, it does provide key functional design cues. The index and middle fingers of the human hand are used to carry out precision tasks with the opposing thumb, whilst the ring and little fingers provide strength during prehension [2]. The thumb itself is the crucial component of a secure grip, as noted by Sir Charles Bell in 1833, *"on the length, strength, free lateral motion, and perfect mobility of the thumb, depends the power of the human hand"*[56]. The palm creates the final form for the prehensile range of the hand through an adaptable and compliant structure. Consequently

the multiple degree of freedom prosthesis[1] must focus on three key design aspects, namely independent digits, a mobile thumb, and their integration to a palm.

In addition to these factors, it is imperative that the design of any new prosthesis adheres to a number of constraints in order to meet client objectives:

1. Anthropomorphism (both in static and dynamic appearance).
2. Low weight (less than 500 grams by current standards for a prototype device).
3. Low power consumption (to make efficient use of the limited battery energy).
4. Modularity (to ensure the design is not handed, thereby minimising components and aiding maintenance).
5. Appropriate size (to match that of an adult hand and fit within a prosthetic glove).
6. Appropriate speed (full digit curl should occur within approximately 1.5s by current standards).

Based on these design philosophies and constraints, a six degree of freedom hand prosthesis was developed. The lightweight Southampton-Remedi hand possesses four independent finger digits, and a dual axis thumb.

2.2 Drive Systems

2.2.1 Conventional Methods

The method of actuating an externally powered prosthesis bears influence over the majority of the design criteria. The drive system of a hand prosthesis must be passively stationary (i.e. it cannot be backdriven by a load when power is removed from the actuator). This criterion is necessary to ensure minimal power consumption, and becomes imperative when prehension is to be achieved with multiple drives. The brushed DC motor, coupled with a mechanically-locking gear train is by far

[1]Named the Southampton-Remedi Hand based on the financial support of the Rehabilitation and Medical Research Trust, REMEDI.

the most common form of drive due to high reliability and efficiency, whilst using a power source that is easily rechargeable. However, alternative systems exist and in some cases have been implemented within prosthesis designs.

Pneumatic actuators have been applied to robotic manipulators, usually in the form of McKibben muscles [25] (see section 1.2), whilst similar work has also seen their use in prostheses [27] to provide a lightweight and fast alternative to electromechanical drives [23]. However the pressurised power source is restricted. For example, several states within the USA limit the transportation of bottled gas, as does the airline industry. Hence factors such as a small potential market, inaccessibility to a reusable power source, and the disadvantage of notable operating noise, have tended to preclude the widespread use of these actuators. Hydraulic systems overcome many of these disadvantages, most notably in the reduction of operating noise. However the significant weight penalty that accompany these devices prohibits their use in multiple degree of freedom prostheses.

2.2.2 Alternative Methods

Alternative actuators that overcome the size, weight and power consumption difficulties of the electrical, pneumatic and hydraulic systems have been investigated [57]. Recent developments have seen devices such as piezoelectric motors, polymer gels (or 'artificial muscles'), and shape memory alloys presented as potential solutions.

Piezoelectric Motors

Piezoelectric motors possess high torque, low speed characteristics, that ideally suit applications within prosthetics or robotics [58]. Polarised piezoelectric (PZT) ceramics (consisting mainly of lead, zirconium and titanium compounds) change dimension when subjected to an electric field. A disc of PZT ceramic with a thickness of 1mm will vibrate with an amplitude of 1μm along the axis of its thickness when excited by an AC voltage. This principle is used within a piezoelectric motor to generate rotary or linear motion by frictional contact with a rotor [59] (see Figure 2.1). An application of this principle can be found in the ultrasonic 'travelling-wave' mo-

tor (USM) where a number of disc-shaped PZT elements are bonded together to form a resonator. When excited, these elements create a travelling wave around the circumference of the disc and move the rotor by intermittent friction contact [57].

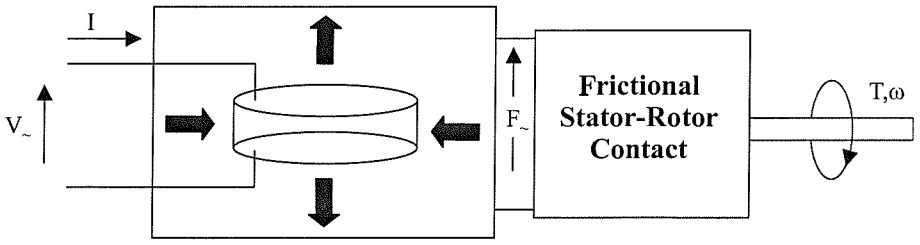


Figure 2.1: Principle of Piezoelectric Motor Operation

The high torque at low speed characteristic enables PZT motors to be used as direct drives without the need for a gearbox thereby reducing the size and weight of the actuator. In addition, when the electrical power source is removed, very high frictional forces hold the rotor stationary (unlike electromagnetic DC motors that require energy or coupling with a mechanical system to maintain position).

However, piezoelectric motors are relatively inefficient (typically η=20-35%) and have a limited lifetime due to frictional wear of the rotor. The overriding constraint precluding their use at present is the lack of commercial development that belies their obscurity and anonymity within the drives industry.

Artificial Muscles

'Artificial muscles' consist of polymer hydrogels [60] that exhibit 'muscular' reflexes (with high changes in the elastic modulus of the gel) under stimulation from a chemical or electrical trigger [61]. Although these microscale actuators possess interesting characteristics, they remain laboratory based experiments and warrant significant development before any commercial application will occur.

Shape Memory Alloys

Shape memory alloys (SMA) are materials that exhibit a two-phase characteristic, and possess different thermal, electrical and mechanical properties in each phase. In

the low (room) temperature martensitic phase (see Figure 2.2), the alloy (usually a nickel-titanium compound) is easily deformable and electrically conductive. Once heated to the austenitic phase the SMA will recover its shape until it is undeformed (set initially by heat treatment methods) [62]. This effect permits the use of the alloy as an actuator by allowing deformation in the low temperature phase, and then applying sufficient energy to heat the SMA element, thereby causing the alloy to return to its original size and shape.

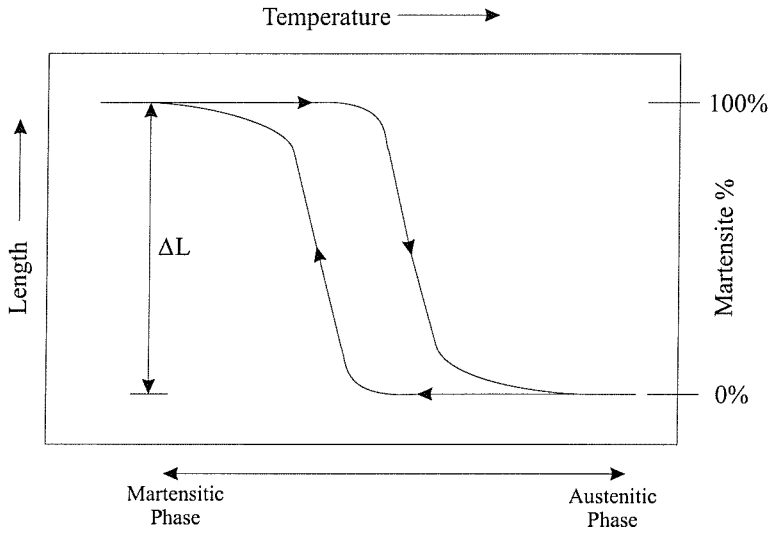


Figure 2.2: SMA Property Characteristic

The high power to weight ratio, and ability to produce a structurally integral actuator, suggest SMAs to be an ideal solution to the size and weight constraints of a hand prosthesis. Indeed, both robotic [63, 64] and prosthetic devices [65, 66] have been developed using these actuators. Additional benefits also include an inherent position feedback method (due to a near linear relationship between ohmic resistance and extension), silent operation, and the lack of requirement for force or motion transmission devices [62].

However, the cooling rates of the alloy are dependent on environmental factors, and the high temperature of the actuators (a problem in itself) does not dissipate at an adequate rate to allow the grip/release actions of a prosthesis to be performed

with sufficient speed [67].

These actuators are capable of recovering from strains of up to 8% but only for a few cycles, whereas strains of $\frac{1}{2}$-1% can be sustained for around 10^7 cycles. Hence it is difficult to achieve an optimum force/extension/reliability configuration, as the stability of the alloy is highly dependent on the strain induced throughout its useful life [68].

SMA devices also produce a continuous power requirement in order to maintain any grip or release scenario. Hence there is a large cumulative power consumption that is difficult to reconcile with a portable, battery-powered hand prosthesis.

2.2.3 Conclusion

The selection of an appropriate drive system (see Table 2.1) relies on an optimum balance of issues such as size, weight, power consumption, cost and availability.

Actuator	**Advantages**	**Disadvantages**
Pneumatic	Fast and light	Noisy, Inconvenient Power Source
Hydraulic	High force, Low Noise	Heavy
Electromagnetic (DC) Motors	Extensive Range Available, High Efficiency	Limited Miniaturisation, Gear Train Required
Piezoelectric Motors	High Torque, Low speed, Passively Static	No Commercial Availability
Artifical Muscles	Microscale Actuators	Laboratory Based
SMA	High Power to Weight Ratio, Silent Operation, Inherent Position Feedback	Poor Reliability to Force Ratio, Cooling Difficulties

Table 2.1: Comparison of Actuator Systems

Although the application of ultrasonic motors appears the most suitable choice based upon engineering design constraints, the lack of commercial availability (even within the primary development area of Japan) precludes their use within the multiple axis prosthesis. Therefore precision miniature brushed DC motors were selected

to be used in combination with a mechanically-locking gear train. This forms the most crucial aspect of the prosthesis design, and governs the subsequent design of independent digits and thumb.

2.3 Digit Design

Multiple digit prostheses have frequently used a mechanical linkage design, either with a tendon [9, 10, 17] or direct drive system [69, 70, 71], to replicate the dynamics of the natural finger. The linkage (see Figure 2.3) usually consists of three pivot points (representing the proximal, middle and distal interphalangeal joints of the natural finger) curling around two main links in a dual four bar linkage format. These designs historically suffered from high backlash and poor mechanical efficiency, such that a high input force (or torque) at the base of the linkage only resulted in a small active grip force at the distal tip.

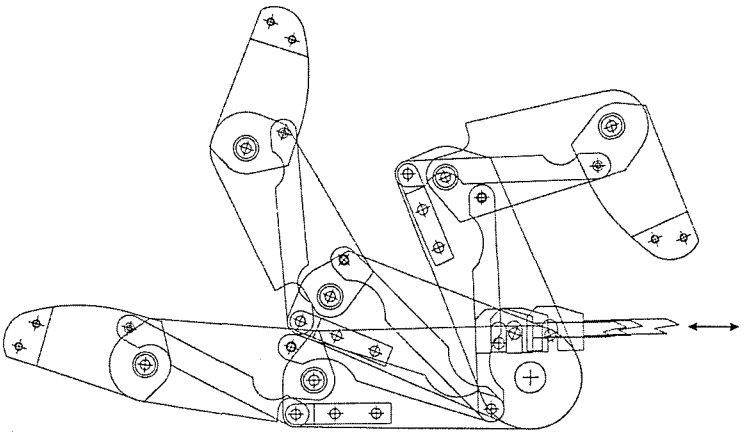


Figure 2.3: Finger Linkage Curling Action from the Mk V Southampton Hand

2.3.1 Design Process

Guo et al. addressed the disadvantages of these systems with the 5-bar linkage [72] that was designed to replicate the trajectory of the human finger during 'natural curling'. A subsequent single degree of freedom design, with a 6-bar configuration,

was optimised for mechanical efficiency [73].

To investigate the kinematics of the system, and subsequently evaluate its suitability for the multiple degree of freedom prosthesis (see section 2.3.2), the design presented by Guo was simulated in Matlab . Data from this simulation provided criteria for the sizing and selection of the motor and gear train units (chosen from the Maxon RE series of DC motors – see section A.1 in Appendix A). Design changes were made to the linkage to enable the integration of the drive train through a separate housing unit.

A linear motion transmission (e.g. a leadscrew) applied to a linkage (as in the case of the Mark V Southampton Hand) can cause the system to cease operating cohesively, thereby disrupting the dynamic operation and appearance of the prosthesis [70]. Consequently the rotary input from a worm-wheel gear unit was selected to ensure stable passive prehension of the hand. Although known to be inefficient, this method provides an optimum solution to minimising size and complexity (unlike active braking systems), whilst adhering to the design constraints previously imposed (see section 2.2.3). Commercial stock items of this nature include excessive material, and therefore weight, as part of the gear train, hence readily modified versions were commissioned (see section A.3 in Appendix A).

The integration of motor, gearbox and worm-wheel transmission to the input of the linkage mechanism occurs through the design of the 'knuckle block' housing unit. The prototype was machined from an aluminium billet and contained bearings along the central drive and perpendicular linkage axes. At 105 grams, the finger unit exceeded acceptable size and weight limitations.

Hence material selection proved to be an essential component in providing a lightweight prosthesis that was able to fulfill the design criteria. A polymer thermoplastic (Vesconite Hilube[2]) was found to provide a weight reduction of over 30% due to the material's low density and low coefficient of friction ($\mu \simeq 0.1$) that eliminated the need for bearings (see Figure 2.4). In addition, the prototype aluminium finger

[2]Vesconite Sales, 338 Billing Rd East, Northampton, NN3 3LJ

linkage was replaced with a planar carbon-fibre epoxy composite design further reducing weight and frictional power consumption. All manufacturing drawings are given in Appendix B.

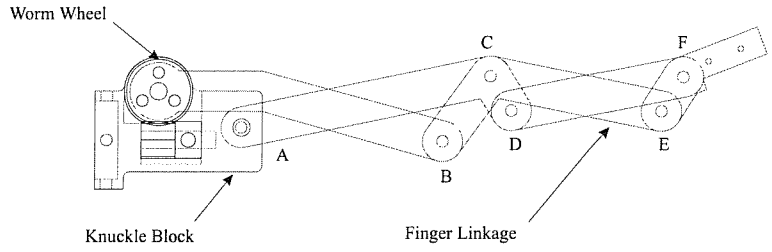


Figure 2.4: Schematic of the Digit Design for the Southampton-Remedi Hand

2.3.2 Evaluation

The functional assessment of the design can only be made in the context of a complete hand prosthesis, however adherence to engineering constraints such as weight, anthropomorphism, grip force and efficiency can be evaluated independently.

Weight

The digit weighs a total of 70 grams, of which 69% is attributable to the mass of the motor[3] and worm-wheel gears (see Figure 2.5). The knuckle block and finger linkage comprise 16% of the total mass, and yet these represent the only apparent areas for additional weight saving by removal of structurally-excess material.

Anthropomorphism

The Matlab simulation of the modified finger linkage was compared to that of a natural finger curl[4] to evaluate anthropomorphism. The trajectory plot (see Figure 2.6) shows the fingertip loci of the linkage and natural finger from the joint rotation centre, or the metacarpophalangeal (MCP) joint in the case of the natural hand. Based on model data [73], the distal tip of the linkage has an RMS error of 10mm from the

[3]Includes the weight of the motor, planetary gearhead, and digital magnetic position encoder.
[4]Obtained from an Exos Dextrous Glove [73]

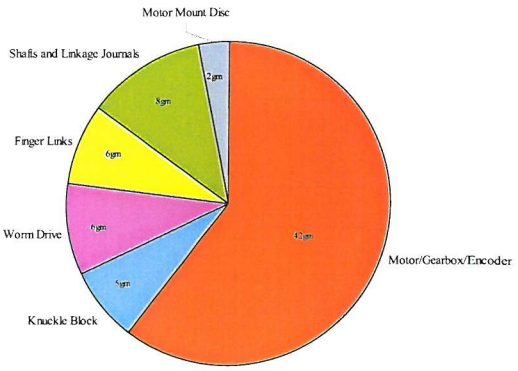


Figure 2.5: Weight Distribution of Digit(s)

natural fingertip trajectory (see Figure 2.7), which notably increases near full curl but clearly shows an anthropomorphic movement.

Full flexion of the digit occurs at approximately 81° (1.41 rads) rotation of the base link. When the finger is horizontal and supinated, a fast ramp input demand[5] causes an average curl time (from fully extended to fully flexed) in 0.84 seconds. The finger runs at 97% of the motor's geared no load speed irrespective of orientation, which is attributable to the low weight and inertia of the linkage.

Force Generation

The force capabilities of the finger directly influence the grip strength of the prosthesis. Performance evaluation is achieved by measuring the maximum tangential force at the fingertip with the motor at stall. A calibrated and balanced strain gauge bridge was used to measure the tension in a steel wire attached to the distal tip of the linkage. The data suggest a maximum active grip force of approximately 9.2N (±0.5N) at the fingertip, at 74° (1.29 rads) flexion (see the experimental data plot in Figure 2.8).

[5]A step input was found to cause over-current spikes in the MOSFET power electronics. Consequently a fast ramp input of 90V/sec was used to eliminate this problem and still retain a high speed response.

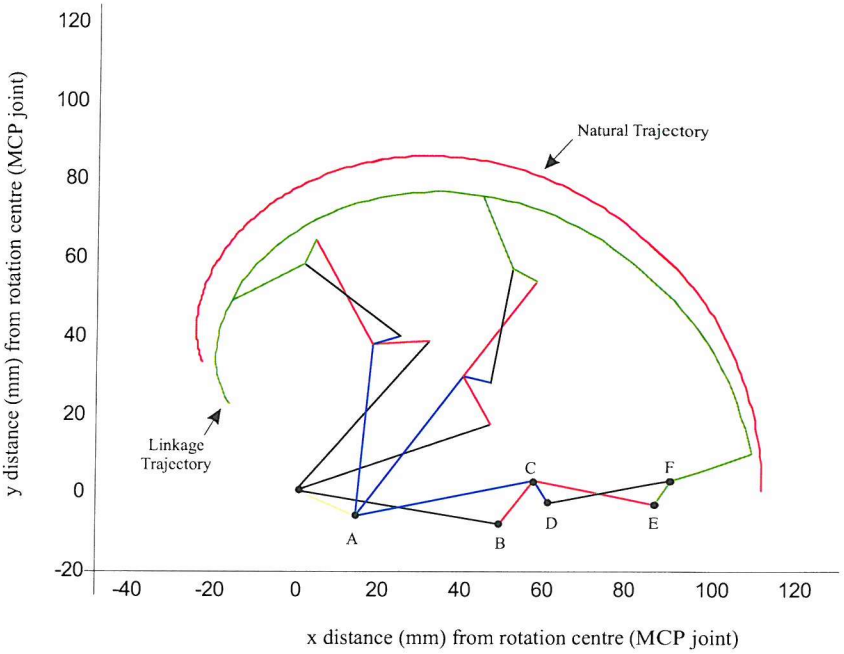


Figure 2.6: Fingertip Stick Diagram and Trajectory Plot

Efficiency

The powered digits form approximately 68% of the main sources of power consumption within the hand, and therefore must be as efficient as possible whilst being able to sustain a secure grip. The measured torque at the worm-wheel output is 0.56Nm, suggesting an efficiency from motor to linkage of only 11.4%[6], which can be attributed to the frictional losses in the knuckle housing as well as the inefficiencies of the worm drive (often quoted as ranging from 20-50%). Although performance may be improved by the use of bearings, there is a notable weight penalty that accompanies such a design change. Moreover, the maximum tip force suggests that the prosthesis should be capable of stable prehension irrespective of the system's inefficiency.

A Computer Aided Engineering tool (Working Model^{TM}) was used to simulate the dynamics of the linkage by adapting the model characteristics to produce an

[6]The quoted motor stall torque is 8.5mNm, the motor gearbox has a reduction ratio of 16.58:1 with η_{max}=0.83, and the worm drive has a ratio of 42:1 with η unknown.

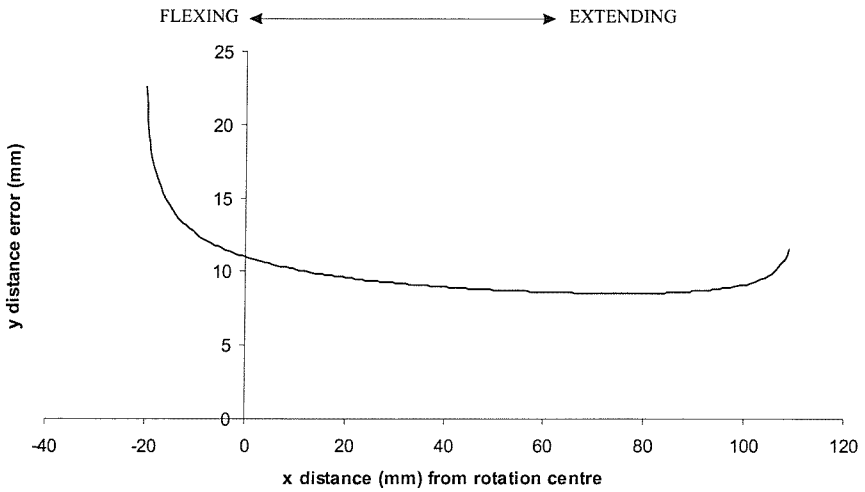


Figure 2.7: Trajectory Difference between Linkage and Natural Fingertip

accurate representation of the empirical fingertip force data (see the simulation plot in Figure 2.8). The general trend indicates that the model is valid by an RMS error of only 1.17N between simulation and experimental results.

The model subsequently provided estimates of frictional power loss within the system (an effect that could not be measured directly). The efficiency of the linkage (η) is defined as:

$$\eta = 1 - \frac{N_f}{N_i} \tag{2.1}$$

where N_i is the input power to the linkage, and N_f is the cumulative power consumed by joint friction.

The simulation results (shown in Figure 2.9) were obtained for the finger in horizontally supinated and pronated positions. The marked decline in efficiency after approximately 45° (0.79 rads) flexion can be attributed to an increase in frictional power consumption at linkage pins B–F (see Figure 2.10). This is caused by a proportional increase in pin velocities with finger rotation, and is accentuated by a steep rise in torque acting around each pin as the linkage approaches full curl. Further tests revealed that although pin C sustains the highest peak loading (at full curl), pin A is subjected to the greatest consistent load and is therefore sized

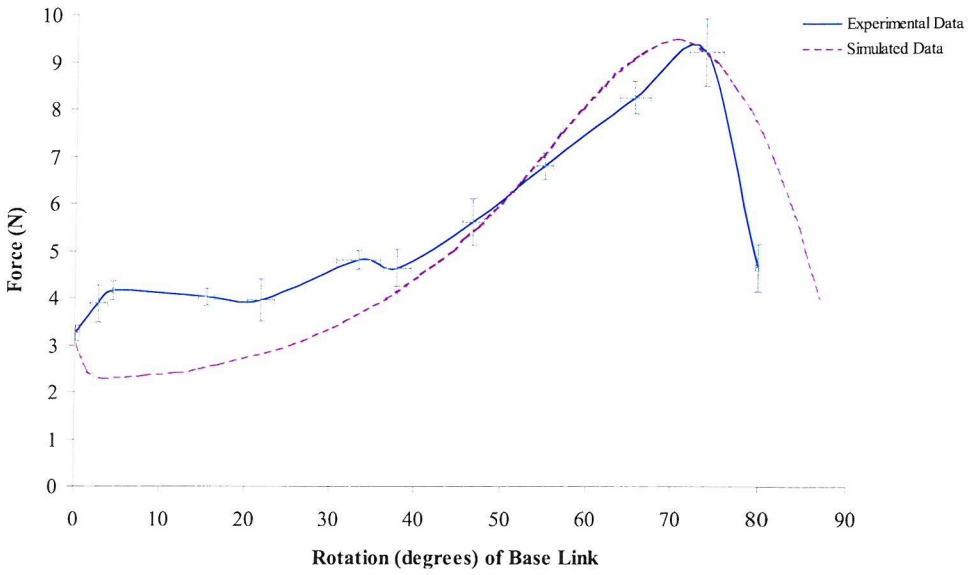


Figure 2.8: Fingertip Force Characteristic

accordingly.

2.3.3 Conclusion

The prehensile strength of the hand prosthesis stems mainly from the powered digits. As demonstrated from previous designs, these components can be highly inefficient and unreliable. Consequently it is necessary to ensure that sufficient grip force can be provided by the digits to maintain stable prehension, whilst also adhering to weight and anthropomorphic constraints.

The maximum fingertip force implies that the multiple degree of freedom prosthesis should be capable of stable active prehension given a cumulative grip force of 38N (omitting the active capabilities of the thumb digit). Moreover, the mechanical adaptability of the device, resulting from independent digits, enables a low fingertip pressure without compromising prehensile integrity. This factor is particular important in ensuring that the multiple drives do not raise power consumption above

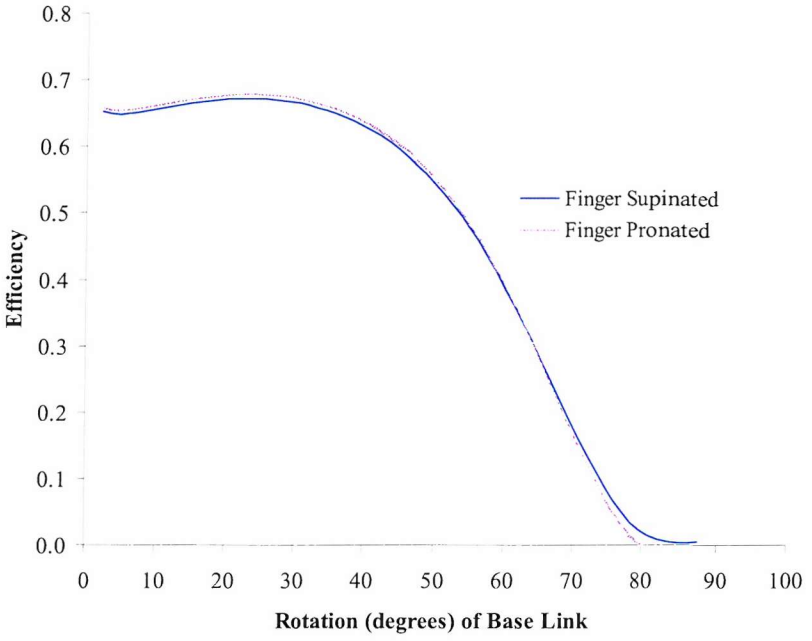


Figure 2.9: Efficiency Plot of Simulated Finger Linkage

acceptable limits.

The kinetic analysis of the linkage suggests high inefficiencies towards full flexion, which also coincide with peak force generation at the fingertip, hence confirming that exerting maximum grip pressure has a significantly detrimental effect on power consumption. The high frictional losses within the drive train can be lessened, but at the expense of additional size and weight, which cannot be compensated for in other component changes.

The kinematics of the system have been shown to be comparable with that of the natural finger, thereby indicating an anthropomorphic design. The weight of the digit is within acceptable boundaries, and suggests that a prosthesis with four independent digits (excluding the thumb) may still ensure a total mass of less than 500 grams. Consequently the prosthetic digit adheres to the design requirements detailed in section 2.1. Although there are notable detrimental aspects to the design such as the high inefficiencies of the system, these compromises appear to have a

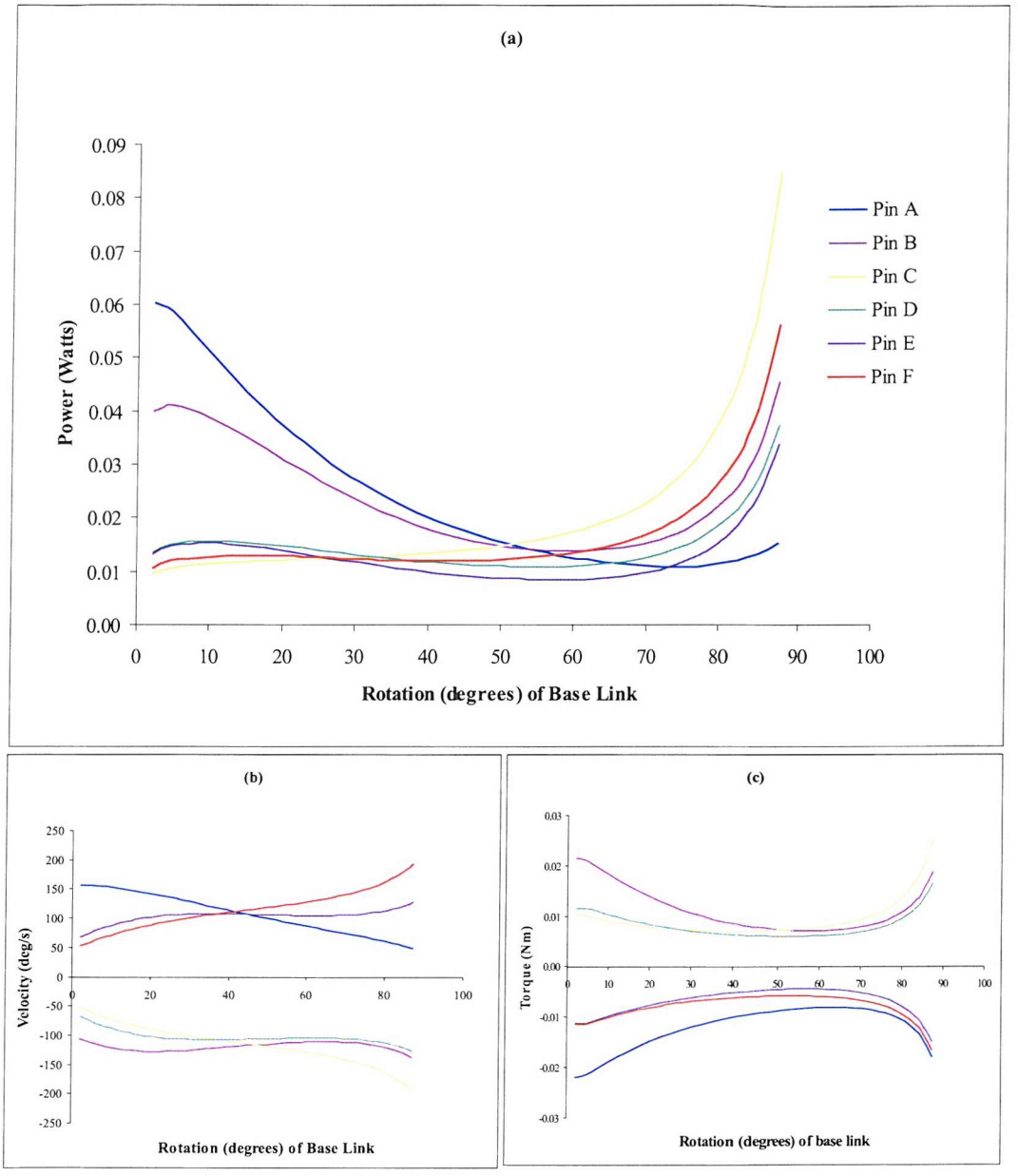


Figure 2.10: (a) Frictional Power Consumption at each link pin, (b) Rotational Velocities at each link pin, (c) Link Torques acting at each pin

negligible effect on the adaptability of a multiple degree of freedom prosthesis.

2.4 Thumb Design

The human hand possesses dexterity unsurpassed anywhere else in nature, and its functionality is centred on the ability of the thumb to oppose the fingers. It is clinical consensus that the loss of thumb function causes a minimum of 50% of the hand's subsequent disability [74]. During restoration, or rehabilitation of the thumb following traumatic injury, surgeons and therapists strive to implement a digit of maximum mobility due to its influence on virtually every prehensile task.

Consequently the single axis pincer movement of the thumb and fingers in conventional prosthetic hands must severely impinge on functionality. Hence there is a need to develop an artificial thumb within the hand that is capable of more than a single degree of freedom, with the consequence of improving the visual feedback, cosmesis, and functionality of the device [75].

2.4.1 Design Process

As the natural thumb possesses five degrees of freedom, detailing the movements afforded by each of the four joints allows identification of the dynamics that the hand prosthesis should strive to encompass.

The scapho-trapezian joint has limited mobility, and although aids in the movement of the carpometacarpal joint (CMC), its contribution to the dexterity of the thumb can be considered minor in comparison to the other joints.

The carpometacarpal joint is saddle shaped (sellar), and is central to the functionality of the thumb. It is capable of two orthogonal axes of motion (see Figure 2.11), flex-extend indicated by y-y' and abduction-adduction indicated by x-x' [2]. It also enables axial rotation of the thumb, thereby maximising the surface area of pulp contact when in opposition to the finger(s). Although critical to the stability of grip, this is a passive motion, and hence is not considered to be an additional independent degree of freedom.

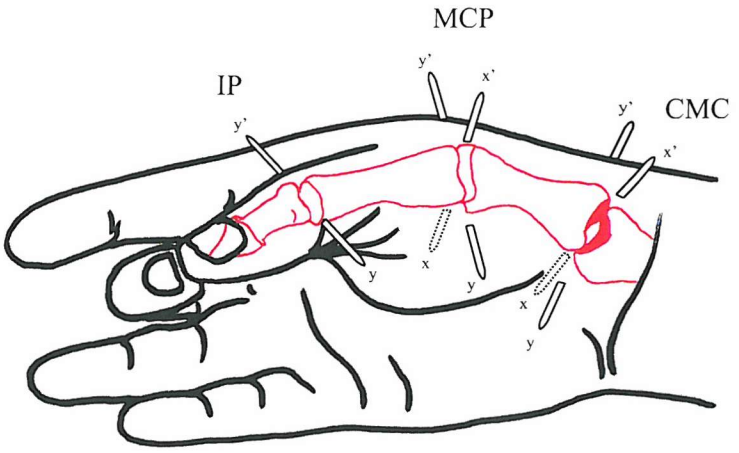


Figure 2.11: Representation of thumb axes

The metacarpophalangeal (MCP) joint has two orthogonal axes of motion which are variable during thumb movement. It is notable that these motions do not occur about axes that are parallel, or perpendicular to the plane of the hand [76]. Both the carpometacarpal and metacarpophalangeal joints can be mechanically represented as universal joints, thereby including the two active degrees of freedom, as well as axial rotation. The interphalangeal (IP) joint is considered analogous to a hinge, and is capable of only a single degree of freedom.

The primary function of the MCP is exhibited during flex-extend movements. Its fundamental role during thumb opposition is the stability of the joint [77] rather than the functionality afforded by its range of motion. Consequently, it is the CMC joint that provides the thumb with its full mobility, by providing movement in the flex-extend, abduction-adduction, and axial rotation axes.

The natural system enables the thumb to possess the characteristics of intrinsic strength and stability to oppose the fingers as well as dexterity for handling fine objects. This mobility originates not only from the skeletal joints, but also from the nine motor muscles used to actuate it, which would be impossible to duplicate in the artificial hand.

Conclusions drawn from the biomechanical analysis, suggest two independent degrees of freedom to be the minimum necessary to perform a range of prehensile

tasks, by replicating the combined mobility of the CMC and MCP joints in the orthogonal flex-extend and abduction-adduction axes. Although study of the natural axial positions infer an optimum solution for the artificial hand, the limitation of only two degrees of freedom raises concern over the efficacy of such a design in a multiple degree of freedom device. It is not a reasonable assumption for the location of the axes in the artificial thumb to be identical to those in the natural hand.

Consequently it is necessary to identify the primary characteristics of the natural thumb that afford versatility. Flexion is frequently used during grasping, and it is apparent that this motion must be preserved as an independent degree of freedom in the prosthesis. Pure abduction or adduction movements are used very rarely during any form of prehension, whereas passive axial rotation is a critical component to generating a stable grip in the natural hand. Consequently, the second degree of freedom was selected to be a combination of ab/adduction and axial rotation, known as circumduction of the thumb.

The location of these 'artificial' axes with respect to the rest of the hand, is determined from the importance of each joint in the skeletal system. The carpometacarpal joint is noted to source the functionality of the thumb, and indeed those with restricted movement of the CMC, for example osteoarthritis patients, frequently have highly limited mobility with no axial rotation [78]. Consequently the two artificial axes (flexion and circumduction) should be located as close as possible to the natural carpometacarpal joint.

To ensure passive stability, and maximise modularity, the worm-wheel drive employed in the development of the finger, was applied to the design of the two degree of freedom thumb. Although the flexion and circumduction axes were to be implemented, no clear criteria existed for locating one axis proximally to the other. The natural thumb causes rotation of ab/adduction (x–x') around flex-extend (y–y') at the CMC joint [79] (see Figure 2.11). However the flexion axis maintains a distal juxtaposition to circumduction in the artificial device in order to maintain anthropomorphic movement.

For the prosthesis to adopt a 'natural' position at rest, the thumb resides at the hand's neutral position upon startup. This is defined as the position of 'myoelectric silence' [2] in the natural hand, where the thumb and index finger metacarpals lie at 30° (0.52 rads) frontally, and 40° (0.70 rads) sagittally.

2.4.2 Implementation and Evaluation

The two degree of freedom thumb must also adhere to the constraints imposed upon the finger design, although mobility is of greater importance than grip strength in order to improve the range of grasping patterns and visual feedback.

A prototype device, machined from aluminium, produced an active grip force of 0.87N and achieved full flexion in 1.4 seconds [75]. However, a more compact and lightweight design was fabricated from Vesconite, with the Maxon motor-gearbox combination (used in the finger design) powering the circumduction axis. A Faulhaber Minimotor and gearbox (see section A.2 in Appendix A) driving the flexion of the thumb provides a peak active grip force of 3.7N and achieves full flexion in 2.5 seconds.

The slow flexing speed of the thumb is compensated for by the primary role of thumb as a static opposer during prehension. Hence, the digits are designed to be the primary active force generators within the hand, whilst the dexterity of the thumb ensures a stable grasp through opposition. The poor efficiency of this system (approximately 7%) may be attributed to the lack of bearings within the unit and the poor performance of the worm-wheel drive.

The modular two degree of freedom thumb unit is reversible in design, so that it may be used for either a left or right handed prosthesis. The thumb circumduction and flexion units (including the thumb stump) weigh 56 grams and 49 grams respectively.

The thumb provides the dexterity necessary to complement the active independent digits within the prosthesis. The role of the circumduction axis is primarily one of mobility, whilst the flexion of the thumb provides an anthropomorphic move-

ment and contributes to active grip pressure during prehension. The device is also passively stable, and although inefficient, adheres to the size and weight criteria necessary for implementation within a six degree of freedom hand prosthesis.

2.5 Palm Design

The design of the palm must fulfil several criteria for integrating the artificial digits and thumb, as well as adhering to specifications inherent in the development of the prosthesis as a whole. Consequently, factors such as modularity, ease of component integration, and anthropomorphism of the prosthesis are key design aspects to the palm. Moreover, it is the palm that facilitates digit opposition and dexterity during grasping.

2.5.1 Design Process

The natural hand maintains three distinct arches [2] to achieve adaptability during prehension: (1) Transverse arch - corresponding to the concavity at the base of the hand and wrist; (2) Longitudinal arch - running from the base of the palm to the fingertip (see Figure 2.12); (3) Oblique arch - formed by the thumb in opposition to the fingers (see Figure 2.12).

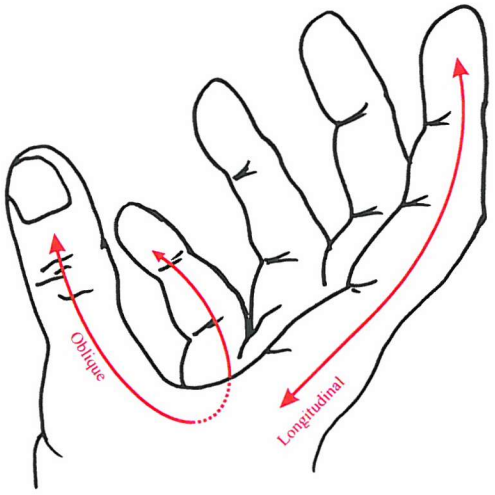


Figure 2.12: Biomechanical Arches of the Hand

A 'palmar gutter' is formed when the hand becomes hollow, where the thumb, index and little fingers represent the extreme points of this oblique arch (that approximately corresponds to the palmar crease, or 'line of life'). It is also the natural direction taken by a cylindrical objects when held in a power grip (see section 4.2.2). The significance of oblique flexion is due to its ability to allow the more medial fingers to oppose the thumb (with increasing obliquity), thereby providing a more secure and adaptable grip [2].

Although these arches afford the hand structural mobility, it is the soft tissue and intrinsic muscle groups that enable sufficient compliance for adaptive prehension. Implementing active, or indeed passive degrees of freedom within the prosthesis palm proves highly complex, and warrants exact three-dimensional modelling of digit interference. The associated control overhead may be minimised given an homology between palmar shape and prehensile pattern, however the additional size and weight necessary to actuate the device prohibits such a design.

The dexterity of the hand stems from a highly mobile and compliant structure, that cannot be divided into elemental components for selective application to the prosthesis. Hence rather than attempting to replicate each of the palmar arches, the design simply includes a 15° (0.26 rads) axis from the middle 'MCP joint', to the fifth 'MCP joint', in order to aid digit opposition.

In the interests of modularity, the first three digits are of the same design (with the little finger linkage being at 75% scale whilst utilising the same motor and drive system). Consequently the digit length discrepancies from the natural hand that influence prehensile ability are compensated for within the palm by varying the proximal location of the knuckle block.

The palm, machined from carbon fibre epoxy composite, weighs 25 grams and is small enough to fit within a standard prosthetic glove (size $7\frac{3}{4}$"). It remains the only component within the prosthesis that is 'handed' (i.e. it requires a mirrored design for left and right hands), and enables the quick-access replacement of individual digits.

2.6 Prosthesis Evaluation and Conclusions

The weight distribution of the prosthesis (see Figure 2.13) shows an approximately equal 17% share for each of the four finger digits, with the thumb units accounting for an additional 26% of the overall mass of 407g (compared to a target weight of 500g for a prototype device, or 350g for a commercial prosthesis). Over 55% of the prosthesis weight is attributable to motor-gearbox drives (excluding worm-wheel transmissions), and represent an intangible area of weight saving. Hence, this suggests that six independent actuators are the current maximum number of degrees of freedom obtainable within a device of this type whilst adhering to prosthesis design constraints.

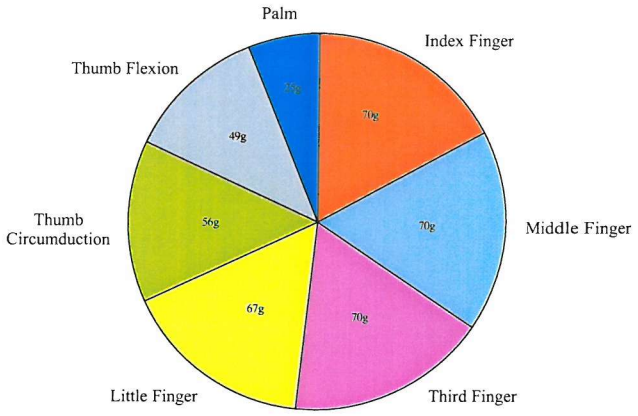


Figure 2.13: Weight Distribution of the Prosthesis

Maximum power consumption occurs during a full squeeze motion involving all digits. This would produce a power of 10.46W[7], however the SAMS hierarchical control scheme (discussed in chapter 3) ensures that minimal power consumption and optimal grip pressure is maintained during prehension.

Nevertheless, power consumption remains a critical design issue. A standard lithium-ion battery used in a prosthesis is rated at 1000mAh at 7.2V. Therefore it

[7]Each of the four finger motors produces a maximum of 2.5W, with the thumb flexion motor (also contributing to grip strength) rated at 0.46W.

is possible to estimate the number of grips achievable with the multiple degree of freedom hand on a single battery charge. The execution of a grip is classified in this case as the movement of the device from the 'natural' position, to grasping and releasing an object, and then returning to the original position. The charge required to perform this manipulation is dependent upon the prehensile pattern, and has been estimated to require 0.472mAh for a power grip, 0.446mAh for a precision grip, and 0.256mAh for a lateral grip when ungloved (see section A.4 in Appendix A). Everyday activities require an individual to implement a power grip for 30% of the tasks, a precision grip for 50%, and a lateral grip for 20% (see chapter 5). Consequently accounting for this balance, the average grip will require 0.416mAh, ensuring that the prosthesis should be capable of over 2400 grips per battery charge. There is notable conjecture over the number of grips a user implements in a day, but informal estimates range from 500–3000, suggesting that despite six degrees of freedom, the Southampton-Remedi hand should maintain effectiveness during average everyday use.

However the effect of long-term use of the prosthesis remains unknown. Forty-eight hour cyclic testing revealed no reliability difficulties, but without extensive clinical trials it is not possible to determine particular areas of design weakness. Further assessments of functionality must be conducted in conjunction with the prosthesis controller (see Chapter 3) in order to be representative of the hand's efficacy.

The mechanical adaptability of the prosthesis provides the prehensile range to achieve tip, tripod, power, lateral and spherical grip types (see Figure 2.14), which is a range of dexterity that is not achievable with any commercial prosthesis. Moreover, each of these prehensile patterns (see section 4.2.2) should be executed with a greater degree of stability than produced by any conventional device. This is due to the ability of each digit to independently contribute to the integrity of the grip that also enables the grasping of objects far larger than that achievable with a single axis commercial device. However, ultimately the functionality of the hand is governed

by the intelligence and adaptability of the control system. It is clearly unacceptable to generate a multiple axis prosthesis at the expense of imposing a significant psychological burden upon the user. This issue has been addressed by intelligent myoelectric prosthesis control.

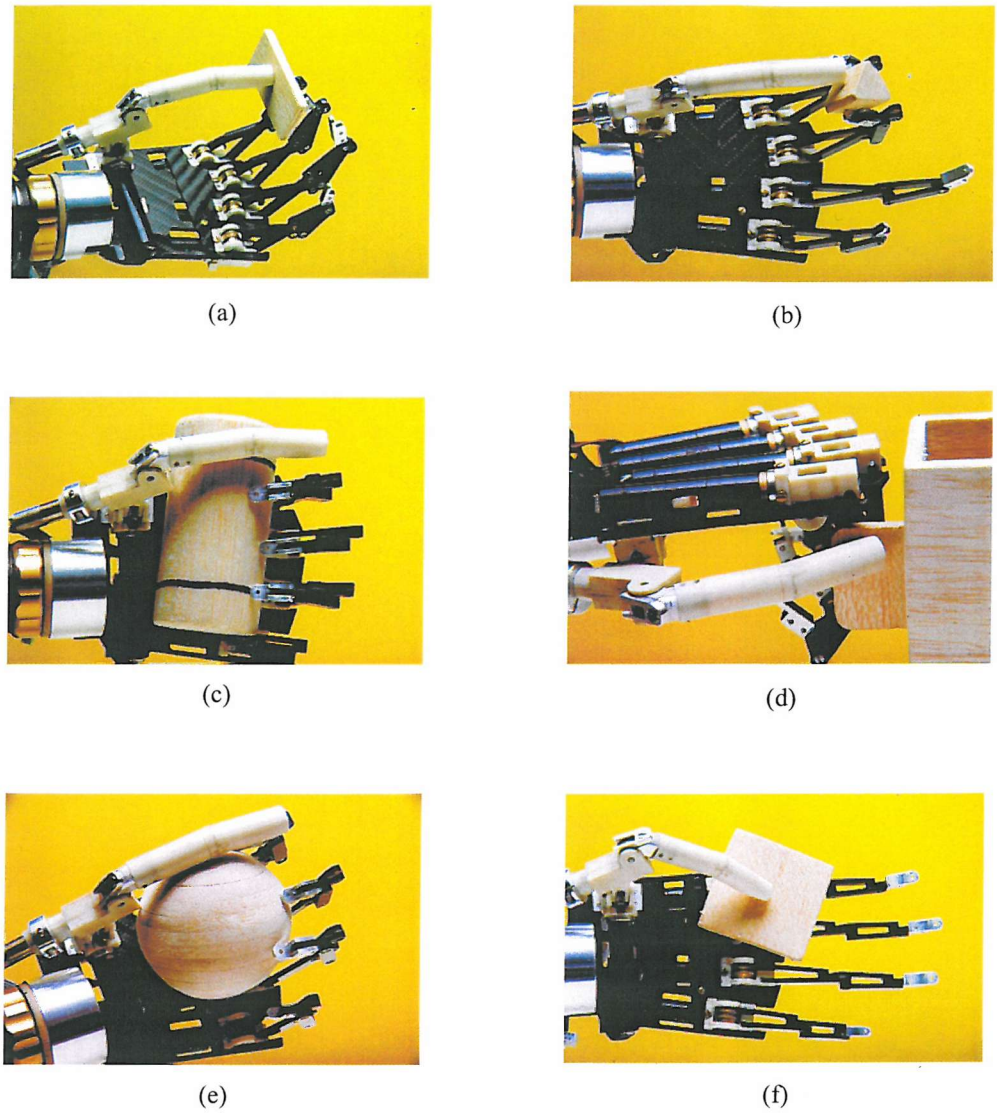


Figure 2.14: Southampton-Remedi Hand in Prehensile Forms (a) Tip, (b) Tripod, (c) Power, (d) Lateral, (e) Spherical, (f) Extension

Chapter 3

Intelligent Myoelectric Control

3.1 Myoelectric Control

3.1.1 Application of the Myoelectric Signal

A potential goal of a prosthesis controller is the emulation of voluntary muscle function for the control of the terminal device to ensure stable prehension. Although unable to provide this degree of proprioceptive feedback, myoelectric prosthesis control is increasingly favoured as the electrical activity of the muscle can be voluntarily initiated, maintained and ended. In addition, there is an approximately constant relationship between muscle tension during isometric contraction and the voltage of a rectified and integrated electromyogram (EMG) that provides an obvious source of control input [80].

The myoelectric signal (MES)[1] occurs due to the depolarisation of the cell membrane in individual muscle fibres during contraction. Groups of fibres are activated by a motor unit, thereby creating a signal called the motor unit action potential (MUAP). In order to maintain a contraction, a large number of these motor units are continually activated and thereby generate the myoelectric signal. The simultaneous activation of fibres of different lengths and motor units produces a useful signal at electrodes on the skin surface with a frequency spectrum of between 30–300Hz [81, 82]. This MES is subject to a number of variations as the tissue between the

[1]MES is typically used to describe the physiological signal produced by muscle contraction, whilst EMG usually describes the trace of such a signal

fibres and the electrodes has a low pass filtering effect, and the electrodes themselves are susceptible to sweat, humidity and temperature at the skin surface. The measured signal resembles that of white noise, and is therefore effectively unpredictable (Figure 3.1 shows a moderate level of isometric contraction of the biceps brachii sampled at 1kHz). Hence signal processing is required to extract useful control data from the MES.

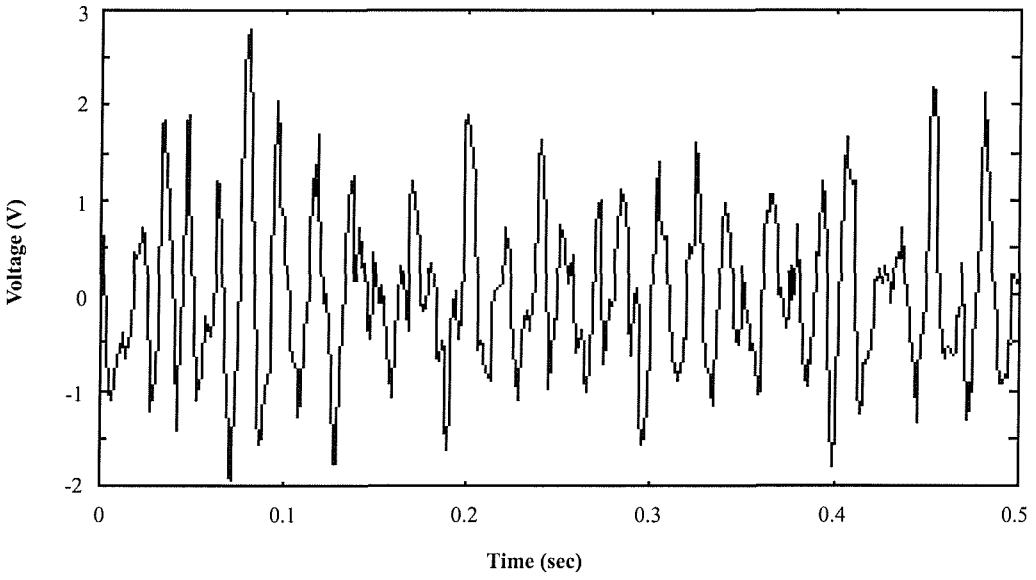


Figure 3.1: The MyoElectric Signal

In myo-prosthesis control, the amplified[2] EMG is usually rectified to form a non-zero mean and filtered to produce an approximation of muscle force as a control input to the prosthesis. Some simple adaptive filters have been applied successfully in real-time to reduce the variation effects in the measured signal, and thereby produce reliable and noise-free representations of muscle force [81, 83]. Other techniques, such as Kalman filters, require estimates of both the signal and the noise effects, and warrant notable processing power for real-time implementation. Methods with less computational overhead, such as low pass filtering or time averaging,

[2]The signal usually has an amplitude range of a few microvolts to several hundred millivolts. Hence differential amplifiers are used to provide common mode noise rejection and signal amplification [82].

are limited without the ability to adapt the filter bandwidth to an EMG that alters with variations in movement type.

3.1.2 Current Myoelectric-Controllers

Current myo-controllers are available in different formats and are usually selected on the basis of user preference and operational success during the period of prosthesis fitment. The single site, two state system requires only a single muscle to operate the normally closed, voluntary opening control (see Figure 3.2a) similar to that of body-powered devices. The two site, two state system uses contraction of one muscle to close the prosthesis, and the other to open the device (see Figure 3.2b). Another option, although less common, is the single site, three state system [82] where the controller monitors the output with respect to two preset levels of muscle tension (see Figure 3.2c). All of these systems may be used as bang-bang controllers (i.e. the speed at which the hand opens or closes is predetermined and the control state merely selects activation or direction), or as proportional controllers where the motor voltage or current (and therefore the speed or force of the hand) is proportional to the amplitude of the signal. It has been shown that experienced myo-prosthesis users prefer the latter method due to the increased control of pinch force [84].

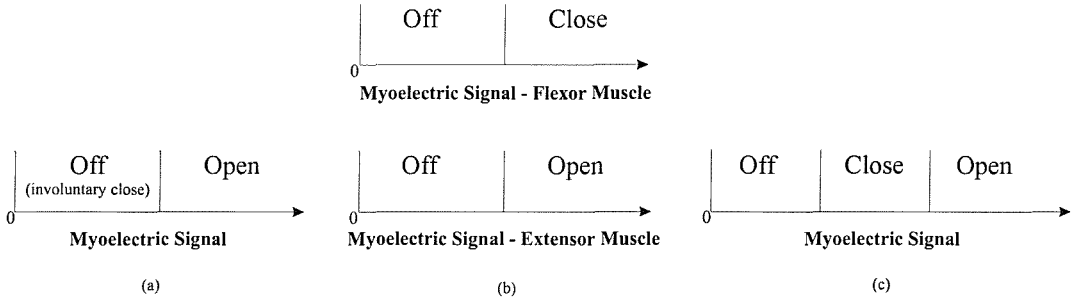


Figure 3.2: Myo-Control Schemes: (a) Single site, 2 state (b) 2 site, 2 state (c) Single site, 3 state [82]

The disadvantage of myoelectric prosthesis control is the lack of proprioceptive feedback that forces the user to rely primarily on visual information. As proprioception is fundamental to the acquisition of motor skills, myo-control that attempts to

emulate the natural system is therefore exceptionally difficult to achieve [80]. Conscious grasping decisions that are based solely on visual feedback require the user to continuously monitor the prosthesis, leading to fatigue and handling errors [85].

3.1.3 Multiple Degree of Freedom Controllers

These difficulties are particularly evident in multifunction prostheses, where the conventional command structure requires the user to sequentially select[3] the function (e.g. a powered hand, wrist or elbow), and then employ standard two channel myocontrol. However, the command and coordination of more than a single device or function is difficult, and is the primary cause for high-level amputees rejecting the prosthesis [86]. Hence the successful use of multifunction devices lies in the synergistic control of several actuators without increasing the number of inputs that a user must independently initiate.

Extracting additional control information from multiple EMG input signals is a long-term goal in the teleoperation of robotic hands. Such controllers have used multiple sensor sites [87, 88] (although not necessarily myoelectric[4]), and neural networks [90] to command several degrees of freedom. This technology can be traced to multifunction devices such as the development of the SVEN hand prosthesis in the 1970s [91], although few ever have reached clinical form.

One of the most significant advances in this field is the multiple degree of freedom controller developed by Hudgins et al. [86] at the University of New Brunswick. This system uses an artificial neural network to derive multifunction control inputs from the MES. The operation of this myo-controller and application to a multiple degree of freedom prosthesis such as the Southampton-Remedi hand is outlined in greater detail in section 3.2.

An alternative method to these multiple command schemes is an intelligent con-

[3] By manual switching, or by a myoelectric command such as co-contraction.

[4] Abboudi et al. [89] have implemented a controller for a multifingered hand prosthesis using inputs from sensors that transduce tendon motion by pneumatic foam sensors at the skin surface. However the potential usage of this system is restricted to transradial amputees and has yet to see implementation in a clinical format.

trol system capable of autonomous and adaptive manipulation without an increase in the number of cognitive input states. The hierarchical Southampton Adaptive Manipulation Scheme (SAMS) has historically demonstrated the efficacy of such a design in controlling multiple degree of freedom hand prostheses [85, 92], and is detailed further in section 3.3.

The requirement for improved prehensile function in prostheses has been demonstrated and addressed by the development of the multiple degree of freedom Southampton-Remedi hand. However, by association, there is a requirement for the device's myo-controller to effect a stable grip without increasing the control burden on the user.

3.2 UNB Myo-Controller

The UNB myo-controller was borne out of the need to improve the operation of multifunction prostheses. The requisite was to extract a greater number of control outputs (or functions) from the one or two myoelectric input channels. Pattern recognition techniques have been used to address this requirement by identifying unique patterns of activity that may provide separate control functions. Historically the biceps and triceps have been used to provide these myoelectric inputs (as the controller was designed for above-elbow prosthetic use), however any antagonistic muscle pair may be used.

Previous myoelectric control schemes have obtained an estimate of contraction level by the use of the steady state signal. Unfortunately, this restricts the amount of time available for signal processing (as the user should perceive no increase in control lag). However Hudgins et al. [86] discovered that the myoelectric signal possesses notable structure at the initial stage of dynamic contraction. Any control system based on this structure would therefore eliminate the delay associated with the processing of a signal that had reached steady-state. This therefore presents the opportunity of additional time for signal analysis without affecting perceived user control.

In addition to this feature, the signal structure of contractions that produce different limb functions (such as humeral rotation, co-contraction, flexion and extension) were found to be distinct. Hence the characteristics of the myoelectric signal can be used to determine specific limb function. This concept was used subsequently in the development of a new multifunction myo-controller with four (or more) potential output functions or states.

The controller uses a two layer artificial neural network (ANN) as a pattern classifier, where each output is essentially a measure of the similarity of the unknown signal pattern to each of the function classes[5]. The requirements of the system are that it must be trainable and allow for variances such as electrode position, changes in body-weight, amputee or congenital deficiencies, and that the user must perceive the system to be quick and reliable in order to ensure continued usage.

Although the transient myoelectric waveforms (at the onset of contraction) were shown to have a significant deterministic nature, random components also exist. Consequently any attempt to use the network to classify the raw signal would produce poor results, yet in contradiction, an average cannot be taken over the transient period without losing structural detail. Hence a set of signal features are determined based on the statistics of the segmented transient waveform. The set of five features describe the magnitude and frequency effects of the signal, and include the mean absolute value (MAV), the MAV of the slope, the segment waveform length, and the number of zero crossings and gradient sign changes.

During training of the network, ten 'feature sets' from each contraction type are collected and input to the network with the corresponding pre-selected class or function output. A back-propagation algorithm is used to adjust the network weights that subsequently are stored.

The main elements of the myo-controller are detailed below and shown in Figure 3.3.

- Signal Acquisition and Feature Extraction – The myoelectric signal is acquired

[5]This is the selected function of the prosthesis, e.g. the hand, wrist or elbow.

with a standard bipolar electrode pair, amplified, sampled at 1kHz and segmented every 0.2 seconds. The five features subsequently are extracted from the signal.

- Pattern Classification – The feature set is input to the network, and the largest (MAX) output of the ANN is chosen. If this is above a preset threshold then the function corresponding to this output class is selected.

- Proportional Control – Once a function is selected, then the input demand to that device (e.g. position or velocity) is driven in proportion to the MAV level of the myoelectric signal. If the signal drops below a threshold value for more than a specified time period then the function is terminated and the system returns to the original state awaiting a new signal input.

- Weight Adaptation – During training the ANN outputs are passed to a weight adaptation algorithm after each contraction, where the errors between the actual and the desired outputs are used to update the network weights using the backpropagation algorithm.

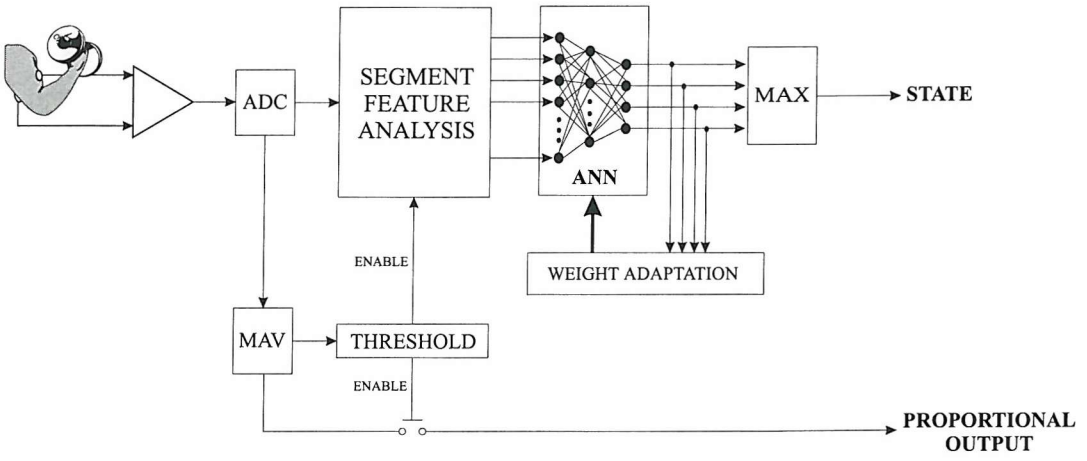


Figure 3.3: UNB Myo-Controller System Design (from Hudgins et al. [86])

The system has been successfully implemented on a fixed point digital signal processor (DSP) and used to control a prosthetic single degree of freedom hand,

wrist and elbow system. However, the ability to initiate multiple functions without increasing the control burden on the user has application beyond that of a single degree of freedom hand as part of an upper limb prosthesis. Artificial hands that are able to initiate a range of prehensile patterns through multiple independent digits suffer from control difficulties due to the restricted number of user inputs available. The UNB myo-controller presents a potential solution to this problem. The disadvantage of this controller is that the user must still use visual feedback to maintain prehensile and kinematic control.

3.3 Southampton Adaptive Manipulation Scheme

Multifunction control of a prosthesis also can be achieved by transposing the low-level control from the user to the device itself. Thus, the user maintains superficial myoelectric control (in the conventional two site manner) whilst a microprocessor and sensor system provide sufficient feedback for the prosthesis to self-regulate prehensile movement and grip force. The form of hierarchical control developed to achieve this goal enables the user to instruct the prosthesis to open, close, hold, squeeze or release an object. This control system is known as the Southampton Adaptive Manipulation Scheme [92].

The concept of the *Southampton Hand*, first developed in the 1960's [14, 18] is that the control of the device involves "only limited conscious command activity from the wearer but with similar coordination between joint kinematics as in normal hand function" [85]. Furthermore it proposed that the success of manipulation is not to be critically dependent on visual feedback given the inherent disadvantages and additional control burden placed upon the user. The efficacy of this design is proven in its continued validity [20].

The control structure resembles that of a simplistic model of motor control in the central nervous system (see Figure 3.4). The lowest-level of the hierarchy manages the position and force reflexes of the fingers. This is governed by the intermediate level of peripheral neural loops to coordinate the hand's shape and grip force in

response to tactile feedback, whilst strategic control resides with the individual [93]. This organisational structure has been replicated in the SAMS prosthesis controller by the user maintaining cognitive input, and the microcontroller implementing force or 'posture' control based on feedback from sensors on the device [85]. For example, if the palm of the prosthesis strikes an object, the force sensors provide feedback to the 'posture' controller to implement a power grip (with full digit curl and the thumb adducted). The 'force logic' controller ensures that optimal grip is applied at each digit, depending on the individual's choice of function. The basic control states are POSITION, TOUCH, HOLD, SQUEEZE and RELEASE (see Figure 3.5).

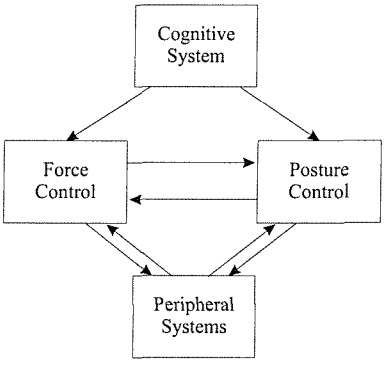


Figure 3.4: Model of Motor Control Hierarchy [85]

The POSITION state enables the hand to adopt the correct prehensile posture. The prosthesis acts in a voluntary opening manner, where extensor muscle activity on the part of the individual will cause the device to open in proportion to the MES amplitude (using position feedback). Hence in the absence of user intervention, the hand will involuntarily close until an object is detected by sensors on each digit, at which point the controller will move to a TOUCH state and terminate movement causing the prosthesis to exert only minimal grip force.

By generating a flexion signal the user will cause the control state to change to HOLD whereby prehensile control is automated using slip sensors on the hand. The controller will maintain optimum grip pressure to ensure that the object does not slip from the grasp.

This state can be overridden by a further period of flexor activity (moving the controller to a SQUEEZE state) where direct control of grip force can be afforded to the user (in proportion to the MES amplitude). During HOLD or SQUEEZE, extensor muscle activity (above a preset threshold) will cause the controller to release the object and return to its original state.

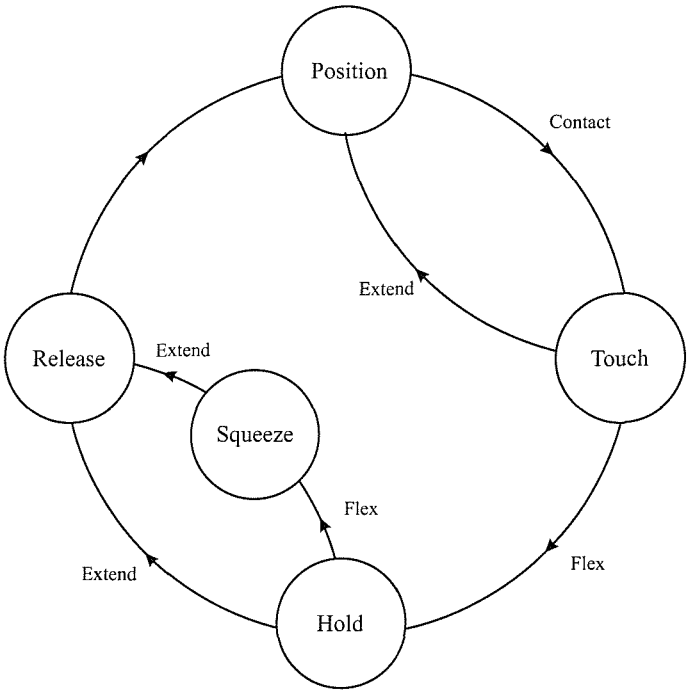


Figure 3.5: SAMS Control Structure [19]

Consequently the user may maintain stable prehension by minimal control input without the need for continuous visual feedback. Although the specific implementation of this control has varied according to sensor and microprocessor technology [14, 15, 16, 17, 18, 70], the hierarchical control philosophy has remained constant. The disadvantage of this system is that the various prehensile patterns afforded by the mechanics of the hand must be originated by specific sensor contact rather than by voluntary muscle function. For example, triggering the lateral sensor on the index finger will initiate a lateral grip posture. This method generates neither a natural or fluid movement.

3.4 The SAMS-UNB Controller

The Southampton Adaptive Manipulation Scheme affords the user the opportunity to maintain prehensile control with minimal conscious effort. The UNB myo-controller provides a more natural and adept method of implementing multifunction control but requires visual feedback during grasping.

Consequently the controller for the new six degree of freedom Southampton-Remedi hand uses a hybrid of these control systems to enable the user to directly implement prehensile patterns from the myo-signal whilst the process of maintaining a secure grasp remains automated (see Figure 3.6). This form of control is applicable to any form of multi-axis hand prosthesis [54].

The radial and ulnar flexor and extensor muscle groups are used to provide the myoelectric inputs for the primarily trans-radial prosthesis controller, although other antagonistic muscle groups may be used. The UNB system is used as a myo-classifier, producing up to four potential state outputs, although additional pattern classes may be achievable depending upon the discriminate muscle function of the individual. The SAMS system then implements a specific prehensile pattern or function (either lateral, precision or power grips, or operation of another device such as an active wrist). Once the grip posture has been selected the hierarchical control system is initiated (and any subsequent state change of the myo-classifier is disregarded). At any point during grasping, a maintained period of extensor activity on the part of the user will cause an object to be released from the grasp and both systems to return to initial conditions.

The prototype system has been implemented in two separate units [54], as the UNB myo-controller already exists in a clinically-ready format. However, it is feasible and logical that future developments will see both controllers integrated to a single microprocessor (thereby reducing power consumption, hardware and communication times).

A Texas Instruments fixed point Digital Signal Processor (TMS320F240 DSP) is used as the main prosthesis controller. This is comprised of the SAMS system and

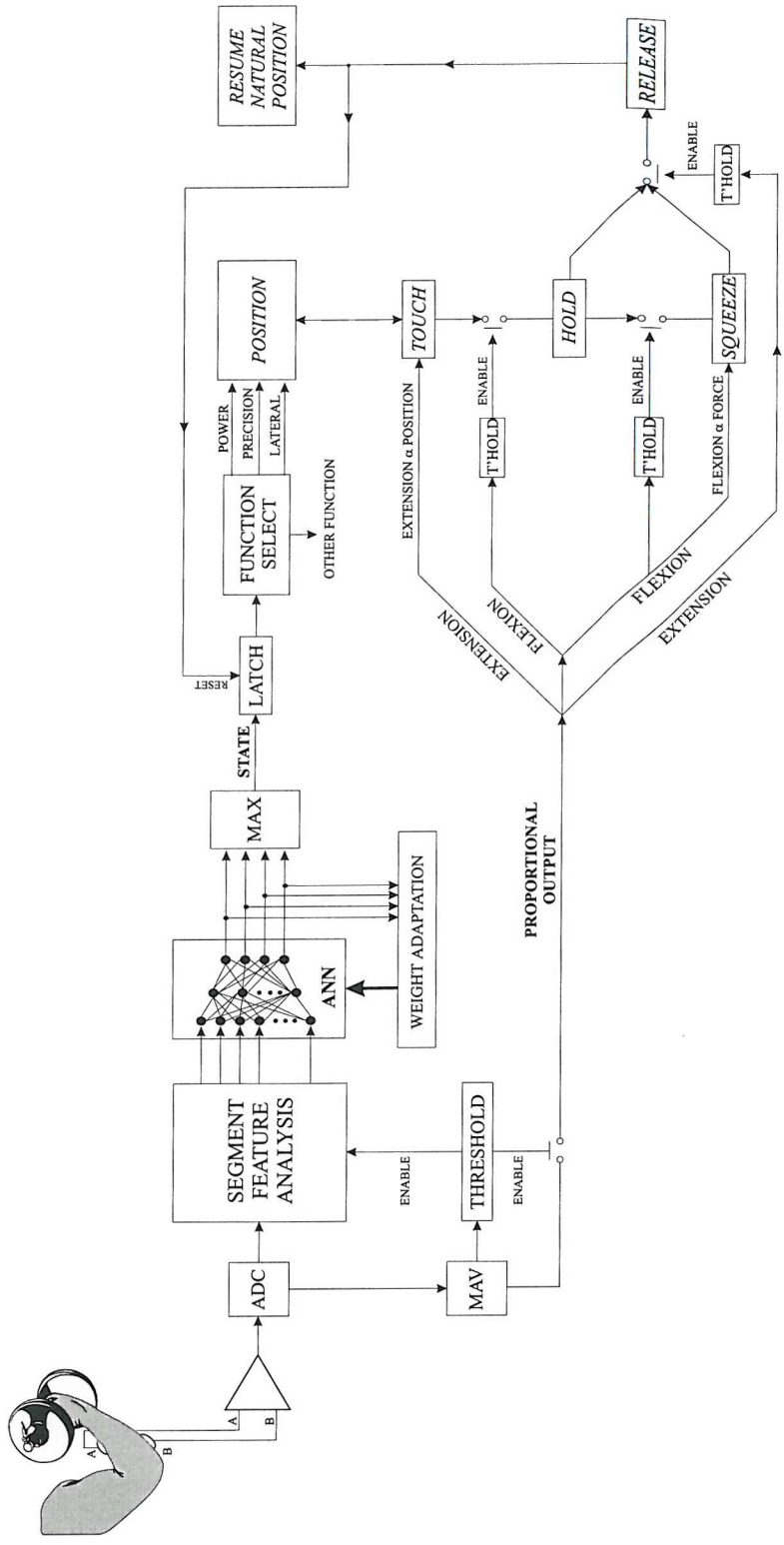


Figure 3.6: The SAMS-UNB Controller

all input/output (I/O) routines to enable communication with the hardware, which includes drives, sensor systems and the UNB myo-classifier. The TMS320F240 is optimised for motor control, and more specifically can be used for multiple drive systems due to the dedicated PWM outputs, digital I/O lines, and analogue inputs. Although many subsystems exist on-board the DSP, external hardware is required to power the drives and interface between the processor and position, force and slip sensor systems (see section 3.6).

3.5 Power Electronics

Pulse Width Modulation is an efficient method of drive control whereby the mark/space ratio of a fixed amplitude rectangular waveform may be varied to control the voltage at the motor terminals. This duty-cycle variation is easy to achieve by microprocessor control, and the use of an H-bridge provides full 4-quadrant motor control. Commercial H-bridge packages were rejected due to excessive power consumption. Instead, the design shown in Figure 3.7 produces a more efficient and controllable[6] drive circuit. The low drain-source resistance of the MOSFETs (0.04Ω for P-channel devices and 0.02Ω for N-channel devices) results in an i^2R power loss that is significantly less than that of the commercial packages. This characteristic is crucial to optimising the use of the hand's battery power supply, and also eliminates the need for heat sinks (thereby reducing size and heat dissipation requirements). Electronic system schematics are supplied in Appendix C.

The PWM signal is used to control the logic-level P-channel MOSFETs by switching in the 6V power supply according to the duty cycle. Forward or reverse digital control signals maintain the N-channel MOSFETs in the relevant on/off configuration (and also minimise the transient current spikes that would arise if both N- and P-channel devices were driven simultaneously by the PWM signal). Inde-

[6]The independent control of each N- and P-channel MOSFET affords the 4-quadrant control that is not available in most commercial packages. Although not intended for implementation in the prototype system, the opportunity for regenerative braking is afforded to a scenario where power consumption is crucial.

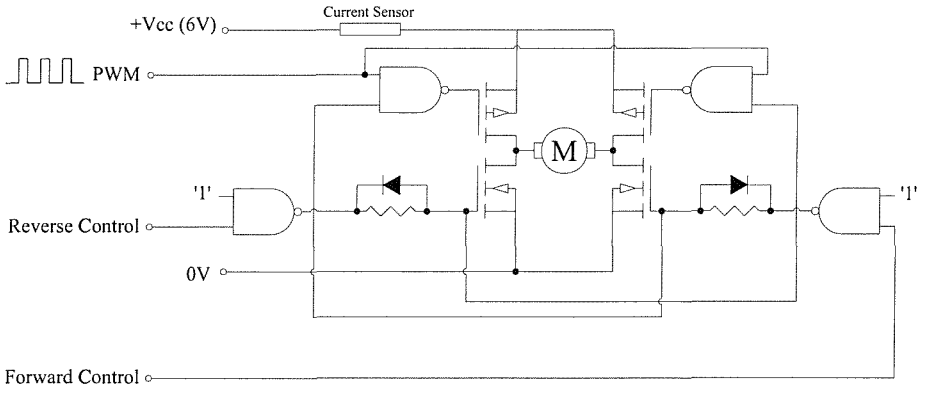


Figure 3.7: H-bridge Circuit Diagram

pendent direction signals afford control opportunities such as dynamic braking and direct deadband control in software, but also increase the number of control lines that the DSP must possess. The H-bridge has been designed with an integral hardware deadband[7] to ensure that no shoot-through can occur during a sudden direction change (i.e. the N-channel devices possess a fast-off, slow-on characteristic thereby ensuring that adjacent N- and P-channel devices cannot conduct simultaneously and create a short circuit).

The design includes a high-side current sensor that can be used to provide information on dynamic grip force (see section 3.6.2), and whether the motor is approaching stall. In addition, given the implied knowledge of motor terminal voltage (from the microprocessor PWM demand), the current sensor may be used to monitor the thermal condition of the drive. Motor performance is directly dependent upon the difference between the ambient and the motor temperatures as well as the duty cycle. Manufacturers quote power and efficiency figures based upon a specific test ambient temperature (usually 25°C) but also reference a range over which the device may be operated safely. Therefore a thermal model of the motor would enable the control parameters to be varied in order to continuously extract optimum performance from the drive and ensure reliability by not exceeding design limits.

[7]The resistor and diode pair are used with the input capacitance of the N-channel MOSFET device to create an increased turn-on delay time of 14μs, whilst maintaining a fast turn-off time of 43ns (see Figure C.1)

Although of academic interest at this stage, if a simplistic model (to minimise processing overhead) could be incorporated into the controller it may provide tangible performance and reliability benefits.

3.6 Sensor Systems

The SAMS control system requires either position, force or slip feedback in order to ensure stable and secure object manipulation (see Figure 3.8). The POSITION and TOUCH states utilise closed-loop position feedback so that the hand's digits proportionally respond to the user's myo-signal demand during digit extension. Force feedback is necessary to determine whether the prosthesis has come into contact with an object, and is also subsequently used during the SQUEEZE state to apply force in proportion to the user's EMG. During object manipulation in the HOLD state, slip feedback ensures that optimum grip force is maintained.

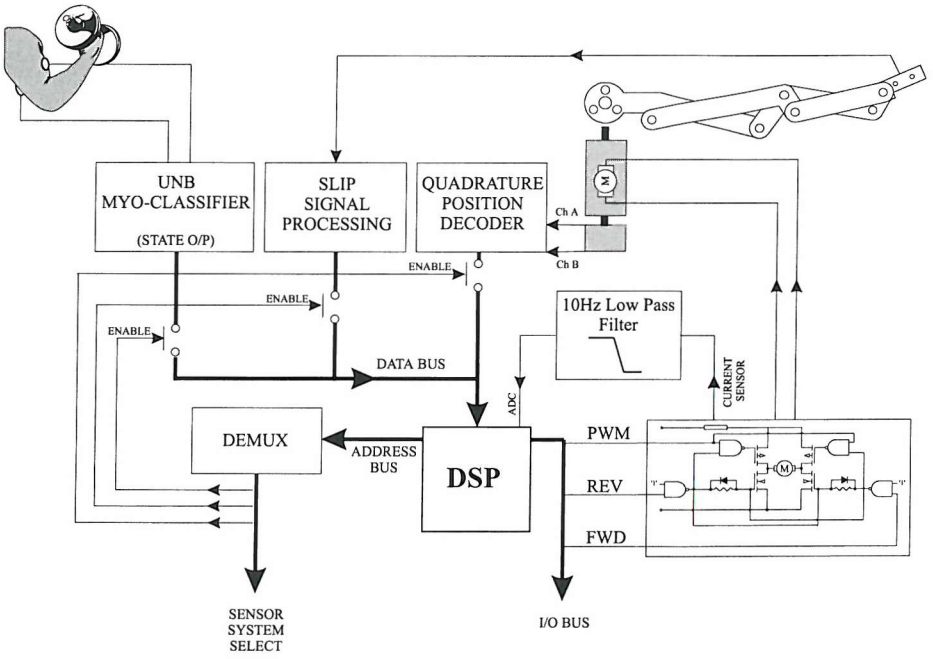


Figure 3.8: Sensor Systems and Controller (with only one of the six motors/H-bridges/signal processing units shown for clarity)

The appropriate sensor system (or state output of the UNB myo-classfier) is se-

lected (or reset) from the microprocessor's address bus via an external demultiplexer. Once selected, the processed signal is transferred to the 16-bit databus, or to the DSP's on-board 10-bit analogue-to-digital converters (ADCs). The mean absolute value of the user's myo-signal (from the UNB controller) is input to the microprocessor's ADC (not shown in Figure 3.8 for clarity) and is sampled at 100Hz[8].

3.6.1 Position Feedback

Position feedback enables the controller to determine the location of the prosthetic fingertip. Previous device mechanics have suffered from notable backlash and mechanical inconsistencies [16]. Position measurement therefore has been unreliable, and often achieved by the use of skeletal potentiometers (to minimise bulk) mounted directly to the digit drive shaft at the 'knuckle' [15, 70]. In addition to the inferior mechanics of the hand contributing to the control problem, these sensors suffer from poor accuracy and repeatability, and thereby inherently limit the reliable operation of the prosthesis.

However the mechanics of the Southampton-Remedi hand possess little backlash, hence digit position can be estimated more accurately without measurement at the base of the finger. This has the benefit of reducing bulk and improving reliability by eliminating sensors from exposed areas of the hand. The motors for the six degree of freedom hand each have a digital magnetic encoder mounted to the drive shaft. Once processed, the resultant quadrature signal provides an accuracy of approximately 0.03° ($0.52\text{x}10^{-3}$rads) of digit rotation.

The encoder output pairs are connected to six dedicated quadrature position decoders (see Figure C.2) that produce a directional 16 bit count of shaft position. The decoder output is loaded onto the microprocessor databus in two 8 bit segments.

[8]The UNB controller converts the digitised myo-signal to analogue form due to current commercial device requirements, however any future SAMS-UNB controller implemented on a single DSP could dispense with this stage and eliminate the need to resample the signal.

3.6.2 Force Feedback

Contact and grip force information is crucial to the success of adaptive manipulation and is often gained through the use of force sensitive resistors [94, 95]. However these analogue sensors must be mounted on the digits of the prosthesis[9] and are notable for output drift over time or due to temperature fluctuations.

However, the motor-current sensors provide sufficient information to determine if the prosthesis has come into contact with an object, and also quantify the force that the digit is applying. The advantage of this system is that the sensors are an integral part of the electronic hardware interface. This is crucial to the minimisation of lead length between the analogue sensor and signal processing components, as well as eliminating the need for externally mounted devices that are susceptible to reliability problems. The disadvantage of this system is that the force feedback is only operational whilst the digits are in a dynamic state (as each drive must be powered to ascertain the current in the H-bridge). However, this is of negligible importance when operating under the current hierarchical control scheme.

The output of the current sensors requires amplification and filtering[10] (see Figure C.3) in order to eliminate high transient effects (that are particularly noticeable at start-up) prior to input to the ADCs of the microprocessor. Although filtering could be implemented in software on the DSP, the additional processing power required (due to higher sampling frequency and real-time filter computation) would be detrimental to overall controller efficiency. Figure 3.9 shows the effect of the filter during digit extension given a fast ramp input (90V/s) demand at time 0.19s. The unfiltered start-up current spike is particularly high and has a rise time of approximately 5ms, during which time the controller would have to determine whether the motor is overcoming stiction, or is about to stall (which actually occurs at time

[9] Direct mounting of sensors onto the hand increases the risk of poor connections due to the cyclic actions of the digits. In addition, analogue sensors mounted remotely from the signal processing electronics also suffer from poor signal-to-noise ratios due to the motor drives in the prosthesis generating notable interference.

[10] A two-pole low pass Bessel filter (f_c=10Hz) is used to provide a non-inverting amplifier (G=3.7) and filter with little ripple in the pass band.

0.67s). However the filtered signal demonstrates a marked difference between these scenarios, thereby enabling a simple threshold to be set to determine stall.

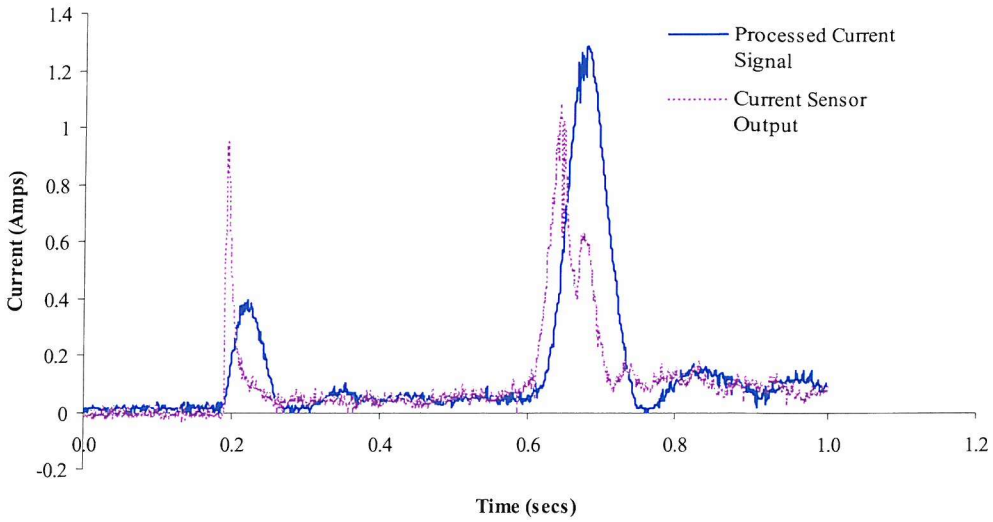


Figure 3.9: Current Sensor Output

Unfortunately the current sensors showed susceptibility to a coupling effect between drives caused by electromagnetic interference generated by the operation of multiple actuators. During start-up the fingers are simultaneously driven to full extension, thereby causing a high current level as each motor approaches stall. This current signal is used subsequently as a control input to reset each digit state (and position). However, there was found to be an effect whereby one motor at stall caused adjacent drive systems to also see a high level on the analogue current sensor causing an erroneous reset of those digits. Hence although individual activation of a digit is readily achievable using this sensor system, the coupling affect ultimately precluded the successful operation of the whole hand.

The prototype system clearly lacks sufficient noise suppression, and could be compensated for by the production of a multi-layer PCB solution with adequate shielding and interference compensation. Integral current sensors remain an effective method of measuring the envelope limits and grip force of the hand. Further

development of the power electronics and sensor systems should result in a fully operational device.

3.6.3 Slip Feedback

The acoustic slip sensor developed by Barkhorder [15] has undergone evolutionary change and modification to result in the device used in the MARCUS collaboration [19] and the LO/SH hand [20, 94]. This consists of a Knowles hearing aid microphone sealed within a rubber tube, and is capable of detecting air movement that is highly coupled to fluctuations at the finger surface. Hence the signal resulting from an object sliding across the surface of the tube is much greater than any extraneous noise. The sensor is particularly effective, however the disadvantage of this design is that any break in the tube or seal to the outside air will act as a point receiver and eliminate the sensor's specificity.

This device has been integrated to the tips of the thumb, index and middle digits of the Southampton-Remedi hand, as only three slip sensors are required to determine object slip in any prehensile configuration.

The slip signal produced by the microphone is broadband in nature with constant production at low frequencies; but it is the high frequency content that is dependent upon the speed of slip and the contact surface [96]. The signal is processed in distinct stages to produce a measure of slip (see Figure 3.10 and in detail in Figure C.4). The primary stage of the circuit removes dc offset from the microphone signal and acts as a high-pass filter with a cut-of frequency of around 550Hz. The signal is then attenuated (with the variable gain nominally set to 0.5), and fed into a band-pass filter with a bandwidth of 2kHz–4.5kHz. These two stages produce an overall characteristic of a notch filter with a peak gain of around 25dB, at a frequency of 550Hz, with the signal referred to a mid-rail voltage of 2.5V [51]. A comparator stage is used to create a pulse train from the processed signal, with a trigger point set by a variable resistor (so that small extraneous signals do not cause slip pulses to be generated). A binary ripple counter is incremented with each slip pulse, and on

demand the microprocessor is provided with a slip count via an octal tristate latch (as shown in Figure 3.8). Each counter can be reset on command from the DSP.

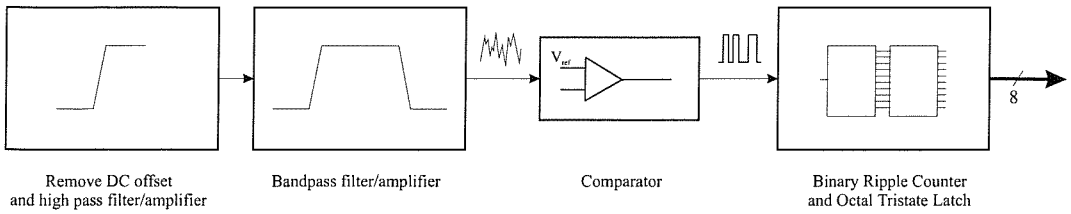


Figure 3.10: Slip Signal Processing Units

3.7 Controller Implementation

The prosthesis control system simulates the autonomous grasping characteristics displayed in the hierarchical human motor control structure. This warrants three separate features: input from the prosthesis (or sensory feedback), calculation of control functions (or cognitive reasoning), and output to the actuators (comparable to muscular contraction).

The TMS320F240 possesses at least 12 independent analogue to digital converter inputs, partially multiplexed with 32 digital input/output (I/O) control lines, a 16-bit databus, and six independent PWM channels suitable for multiple drive control. DSPs can handle multiple axes and more sophisticated control/signal processing algorithms within shorter update times than conventional microprocessors [97]. The TMS320F240 is optimised for digital motor control, and with the six independent actuators and multiple feedback systems of the prosthesis, the I/O capacity is fully utilised.

In order to achieve reliable and accurate control of the mechanical system, the motion controller must be implemented in real-time. Therefore I/O interfacing and control effort calculations must be made within the bounds of interrupt or 'event' driven software, also known as interrupt service routines (ISRs). Thus the main program executes the higher level 'cognitive' control states of the SAMS-UNB system, whilst the ISRs maintain real-time interface control with the prosthesis (see

Figure 3.11).

The ISRs were validated by breakpoint evaluation of the code to assess the impact of each I/O and real-time control module. Interrupts handling I/O access were given priority over control ISRs due to the need for higher sampling frequencies. However, the overall efficacy of the real-time control system could not be evaluated in functional terms due to the limitations in the electronic hardware (as discussed in section 3.6.2).

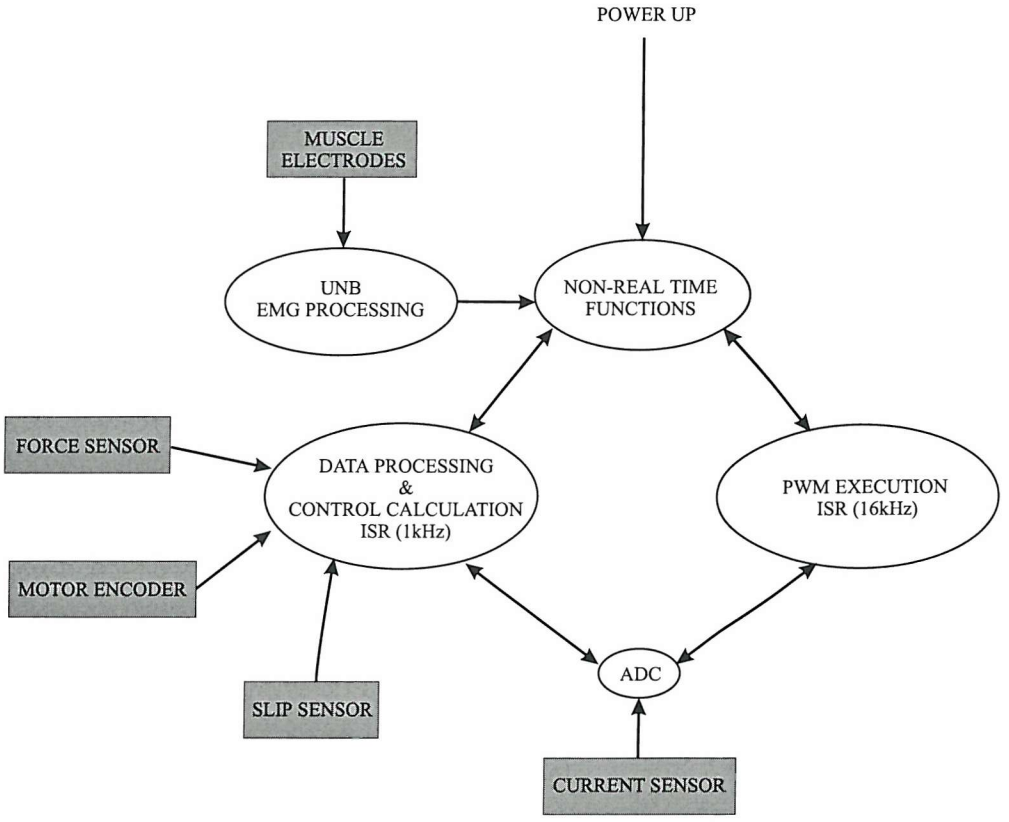


Figure 3.11: Data Flow Diagram

3.7.1 Interrupt Service Routines (ISR)

The prioritised interrupt procedures are predominantly written in assembler to ensure fast execution times. Throughout the interrupt service routines, efficiency of code and minimal I/O access times are of paramount importance. The ISRs rep-

resent the lowest level of the hierarchical structure and handle all data transfer between the controller and the prosthesis.

The TMS320F240 possesses three general purpose (GP) timers/counters that can be used to generate sampling periods in a control system and to provide time bases for the operation of PWM circuits. Key events in the timing sequence, such as overflow (the timer reaching a count of FFFFh), and underflow (at a count value of 0000h) cause an interrupt flag to be set, thereby suspending the main program and executing a subroutine (the ISR).

An interrupt is also generated by the ADCs at the end of conversion. Two channels can be captured simultaneously (taking a maximum of 6.6μs), and the result is stored in a 2-level deep FIFO register. Hence a total of four analogue input channels may be sampled before the data must be stored to memory or lost.

Other interrupts can be flagged by software events or by a range of hardware devices (either on-board or external). However the two main ISRs in use for the prosthesis controller are caused by a timer underflow event, at a frequency of 16kHz, whereupon control signals are output to the motors, and by an ADC end of conversion (EOC) that is used to handle sensor data capture and processing.

Timer Underflow ISR

This high priority interrupt is serviced every 62.5μs and is used to output control instructions to each H-bridge. A counter is decremented within the ISR and triggers the capture of digital sensor data at a frequency of 1kHz, as well as initiating an ADC sampling of motor current (see Figure 3.12).

The control parameters consist of both magnitude and direction, stored in arrays *controllerOP[n]* and *flx_ext[n]* respectively (where n=1...6). The PWM waveform has a fixed period (of 62.5μs) but a variable duty cycle that enables the mean motor terminal voltage to be accurately controlled. This is achieved by writing 16-bit words to specific registers that are subsequently compared to the GP timer value. Thus the values written to the control output range from zero (producing a mark/space

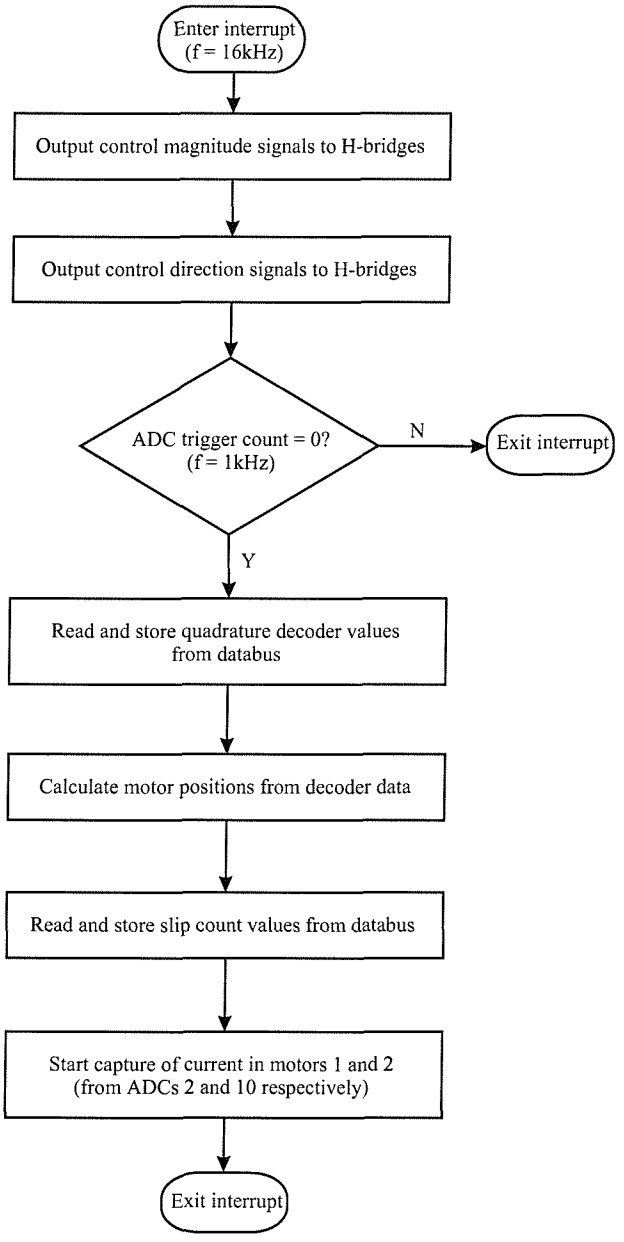


Figure 3.12: Flowchart of the Timer Underflow ISR

Cycle No.	Databus	
	bits 15–8	*bits 7–0*
1	Motor 2 High Byte	Motor 1 High Byte
2	Motor 2 Low Byte	Motor 1 Low Byte
3	Motor 4 High Byte	Motor 3 High Byte
4	Motor 4 Low Byte	Motor 3 Low Byte
5	Motor 6 High Byte	Motor 5 High Byte
6	Motor 6 Low Byte	Motor 5 Low Byte

Table 3.1: Access Order of Quadrature Decoder Pairs

ratio of 0%/100%) to the PWM waveform period (with a duty cycle of 100%/0%). The directional control signals are output to the H-bridges via the digital I/O lines, requiring only a single instruction cycle (50ns) to effect a change in direction control in two H-bridges simultaneously.

The quadrature decoders (used to determine motor position) can only download values in 8-bit segments, and require a control bit to toggle between low and high bytes. Hence in order to minimise I/O access time, bytes from motor pairs are loaded onto the databus at the same time (according to Table 3.1). The positions are subsequently calculated from these data (by bit manipulation) and stored to memory.

Similarly the slip count is also accessed via the databus and stored to memory, and occurs prior to initialising the ADC sampling of two motor current sensors. The end of this analogue-to-digital conversion causes an interrupt to be flagged.

ADC end of conversion ISR

This low priority interrupt is nominally generated at a frequency of 1kHz from the initialisation of ADC data sampling in the timer underflow ISR. Consequently upon first entry to the interrupt the current sensor values from motors 1 and 2 are stored to memory (see Figure 3.13).

The interrupt procedure then triggers the sampling of the remaining current sensors (and the UNB proportional myo-signal at a frequency of 100Hz). The ISR exits immediately following the initialisation of each data capture in order to again service

the end of conversion interrupt generated by the ADC. The slow operational speed of the device's mechanics afford the relatively low sensor sampling frequency of 1kHz (although a higher rate may be achieved without adversely affecting operational control).

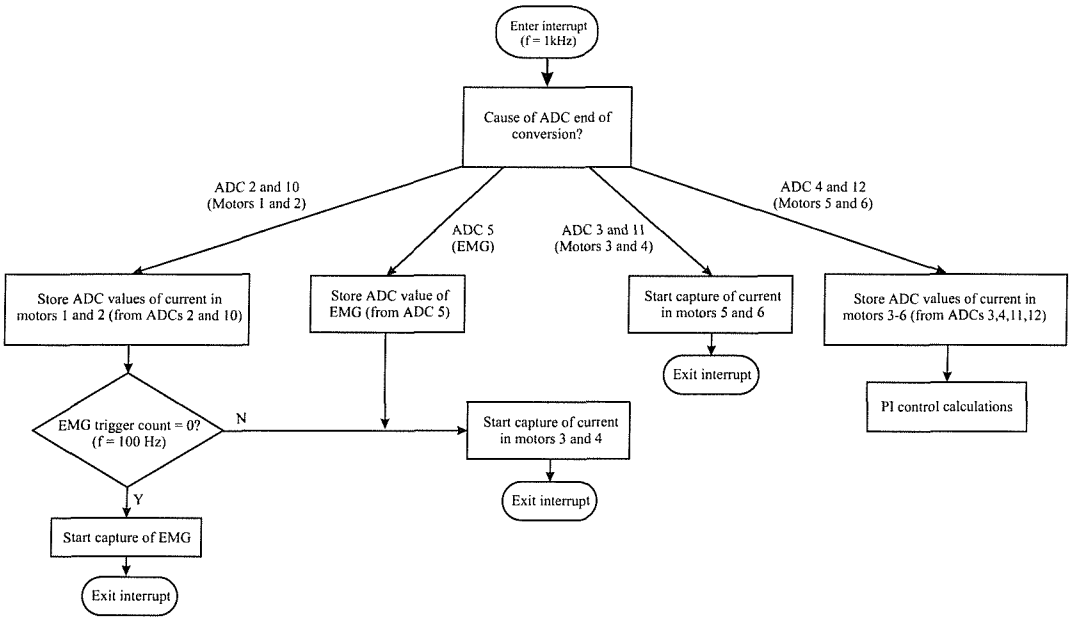


Figure 3.13: Flowchart of the ADC end of conversion ISR - Data Capture

Once all of the sensor data have been stored, closed loop control calculations are made for each motor n (see Figure 3.14). This procedure does not require I/O access, and although contained within the ISR, is coded in 'C' in order to integrate the multiple feedback systems to the control algorithm more readily.

There are two control states in which this structure is bypassed: if the controller is in reset mode, the prosthesis is driven open-loop until initial conditions are met and all sensor systems can be initialised (see section 3.7.2); or if a fast motor-shutdown has been requested. The latter state acts as a safety harness to the control of the mechanics. Hence any requirement to cease activation of the prosthesis, either by user demand or by sensory feedback (e.g. the motor reaching stall current), can be serviced in the shortest time period.

If the system is in normal operation then proportional plus integral[11] closed loop control is used with the appropriate feedback system according to the current SAMS state. A deadband is included in the PI loop due to the inherent backlash in the mechanical drive thereby avoiding controller oscillations, however the second prototype prosthesis possesses significantly less backlash and thereby allows more accurate tuning of the controller. A more sophisticated method, such as model-reference control [98], could be used to compensate for the mechanical inconsistencies of the system, and the friction controlled backlash characteristics of the worm-wheel drive. However the need to minimise processing overhead is of greater importance.

3.7.2 Initialisation

Following reset (or power-up) the configuration of the prosthesis is unknown to the controller. Hence an initialisation state is entered, and a flag set, in order to commence operation with preset conditions. The reset flag causes the closed loop control system (serviced during the ADC end of conversion ISR) to be bypassed.

The digits are driven in open loop mode to full extension, at which point all control parameters and flags[12] are reset. Hence the prosthesis has reached a known condition, whereby all external systems and sensors have been initialised.

A demand characteristic resulting in a 'neutral' hand position is then set prior to leaving this state, and thereby enabling closed loop operation and the hierarchical SAMS control system to be activated. Any power failure, control error, or unknown mode of operation will cause the DSP to reset and commence this initialisation routine.

[11] PI control ensures a fast system response with minimal steady state error. The mechanics of the prosthesis are sufficiently slow during operation to ensure stable prehension (a relatively slow cognitive process) without a derivative control term.

[12] These terms include the sensor arrays (stored in memory), the control parameters (error, demand, and integral sum values) and magnitude and direction outputs. The flags are used to indicate modes of operation to determine flow through the control structure.

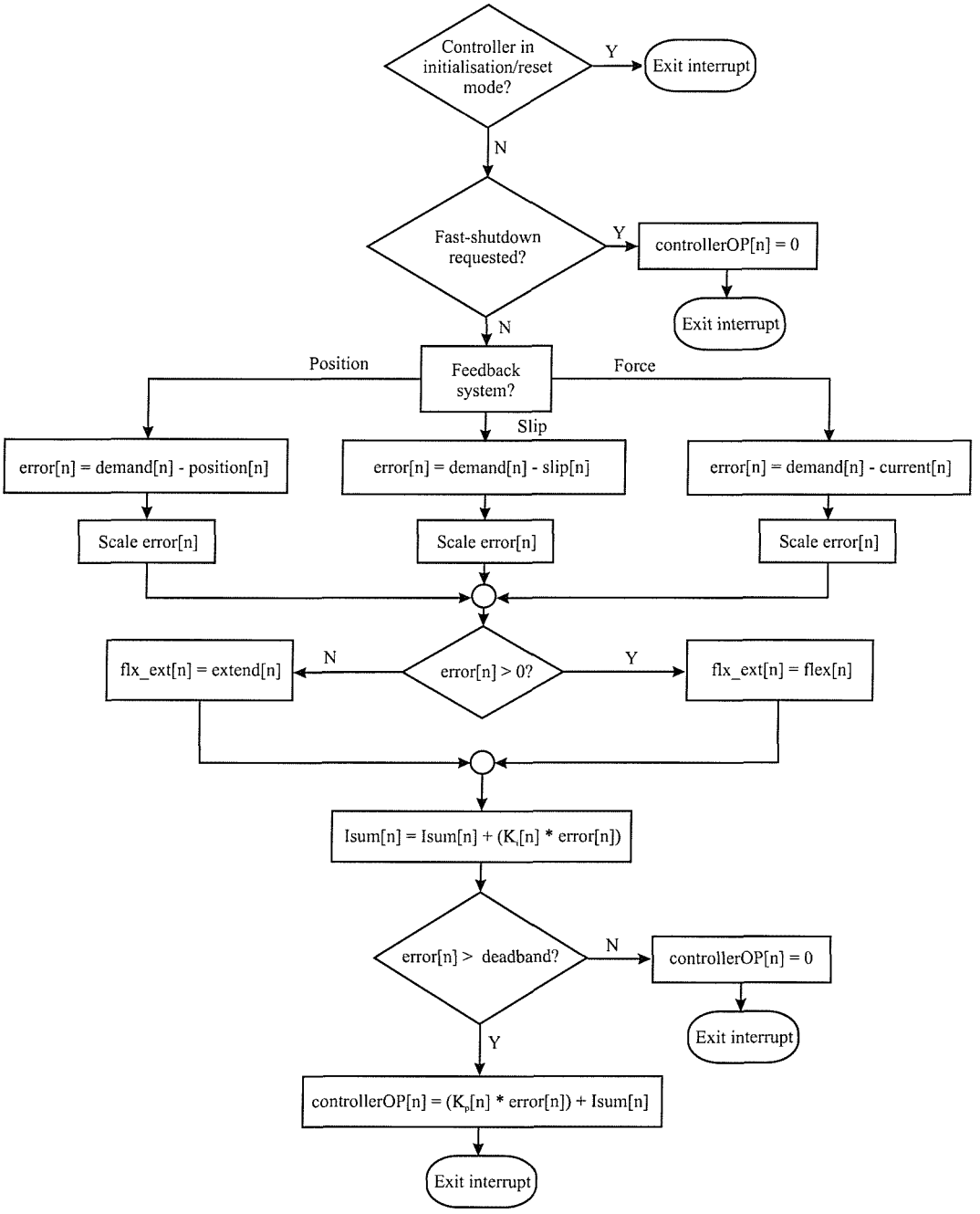


Figure 3.14: Flowchart of the ADC end of conversion ISR - PI control

3.7.3 Hierarchical control

The SAMS hierarchical structure consists of a state decision block contained within an infinite loop (see Figure 3.15). Hence the system is entered upon completion of initialisation, and only exits due to a system error or power down.

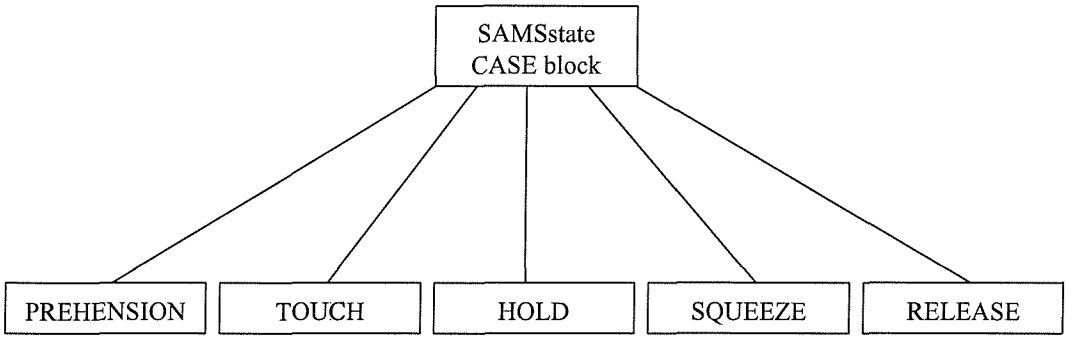


Figure 3.15: The SAMS hierarchical program structure

The PREHENSION software module governs the position control state (as discussed in section 3.3). This mode downloads the UNB myoclassifier state, and subsequently activates the appropriate control function (see Figure 3.16). Additional functions, such as active wrist rotation, are also accessed from the PREHENSION state, but are unique to the configuration of the prosthesis. The *position()* software routine ensures that preset grip postures are adopted by the hand before proceeding. Different independent digit movement is necessary for each prehensile pattern, and the corresponding drive units are identified. The controller then moves to the TOUCH state upon contact with an object.

Position feedback is used in POSITION and TOUCH to allow the hand to open in proportion to the sampled myo-signal (see Figure 3.17). The control structure of this state is shown in the pseudocode program [99] of Figure 3.18. If there is no extensor activity then the digits of the prosthesis will involuntarily close until each current sensor detects object contact, at which point the corresponding motor is shut down. Thus the prosthesis will form an adaptive grip around the object and exert a light touch. Only once each active digit has reached touch (or full flexion),

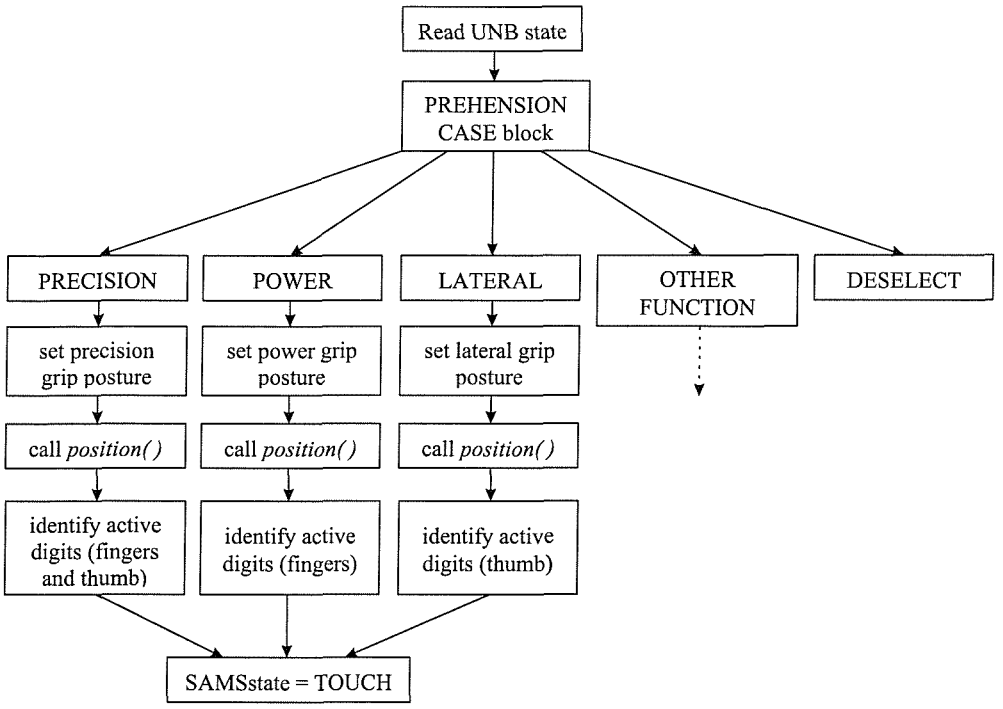


Figure 3.16: Prehensile selection control structure

may the user then move to the HOLD state by exceeding the flexor EMG threshold.

Multifunction devices must adhere to predetermined control rules to ensure successful envelope gripping [40], whereby the operation of the independent mobile thumb cannot interfere with the motion of the fingers and cause the hand to 'lock'. The six degree of freedom hand and controller achieves this mapping readily due to the slow speed of thumb flexion movement, and its use in the primary grip types of power, tip and lateral prehension. Initiation of the power grip causes the adduction (or retroposition) and extension of the thumb prior to the fingers curling. Tip (or tripod) prehension causes the thumb to flex (from an abducted position) to meet the middle of the first and second finger distal tips, unless a TOUCH state has already been achieved. A lateral grip causes the fingers to curl to a predetermined position prior to the flexion of the thumb (which is oriented in a position of abduction).

The automated grasping function uses slip feedback, and is activated in the HOLD state (see Figure 3.19). The control demand is set to zero, and the PI

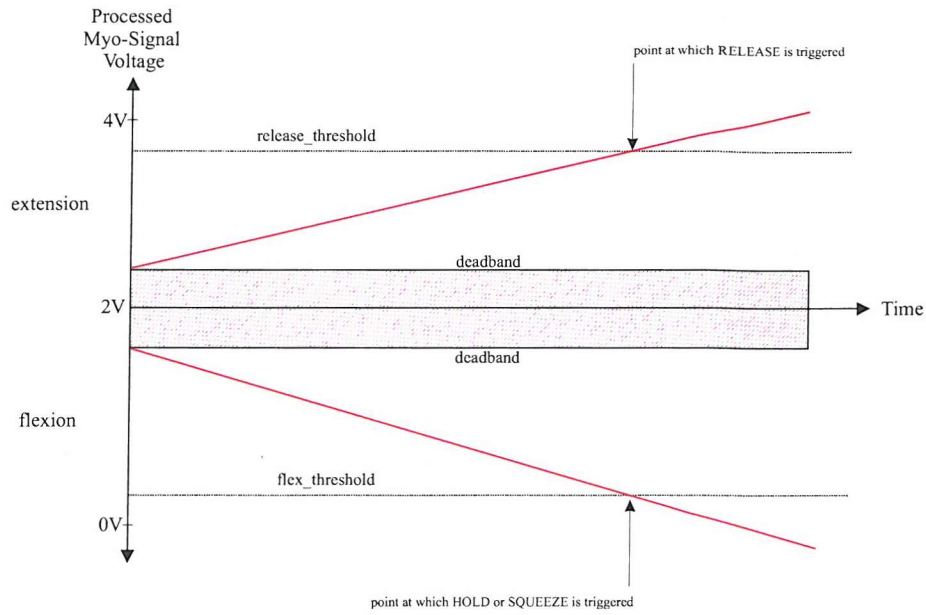


Figure 3.17: Myosignal level and state control thresholds

controller used to increase grip force until slip is arrested. In order to override this function and maintain manual force control, the user may move to the SQUEEZE state by exceeding the flexor myo-signal threshold. To avoid direct transfer from the TOUCH state to SQUEEZE, the user must first relax their myo-signal to within the deadband limits.

The slip sensors are used in the control of the thumb, and index or middle fingers, depending on the prehensile pattern selected. The remaining digits only contribute to object stability during the power grip (see section 4.2.2), in which case the control output is set equal to that of the middle finger (within a subroutine of the ADC end of conversion ISR) to ensure the uniform application of force to the object. The *trigger_release()* routine calculates the average of the last three myo-signal samples, and triggers the RELEASE state if the value exceeds a preset threshold.

The SQUEEZE state uses force feedback (provided by the motor current sensors) to activate grip pressure in proportion to the user's myo-signal when below a flexor myo-signal threshold. The RELEASE state (entered into by maintained extensor activity) causes the active digits to fully extend and thereby release an object from

```
feedback = position
DO WHILE (SAMSstate = TOUCH)

    IF (EMG > deadband)
        FOR (all active digits)
            IF (motor_current > stall_threshold)
                request motor shutdown
            IF (demand < EMG)
                demand = EMG

    IF (flex_threshold < EMG < deadband)
        FOR (all active digits)
            demand = fully flexed position
            IF (motor_current > touch_threshold)
                set touch_flag
                request motor shutdown

    ELSEIF
    (EMG < flex_threshold) AND (all active digit touch_flags are set)
        SAMSstate = HOLD
```

Figure 3.18: Pseudocode of TOUCH state

the grasp. The control system subsequently is returned to the PREHENSION mode in preparation for the next prehensile task.

The efficacy of this hierarchical control structure has been demonstrated by the ready adaptation of users to this unfamiliar myo-control strategy [17, 92, 93]. The integration of the UNB system into the new Southampton controller should provide increased fluidity of movement and grasping. This is due to direct prehensile pattern selection from the user's MES and the controller's adaptive and intelligent manipulation scheme.

3.8 Evaluation

The UNB system has been demonstrated to provide a 91.2% classification accuracy rate in correct state selection (with a standard deviation of 5.6%), based on a sample of nine normal subjects [86]. When the system was evaluated on six amputee sub-

```
DO WHILE (SAMSstate = HOLD)

    IF (EMG > deadband)
        call trigger_release()

    IF (flex_threshold < EMG < deadband)
        FOR (all active digits)
            feedback = slip
            demand = 0
            reset external slip counter
            IF (motor_current > stall_threshold)
                request motor shutdown

    IF (flex_threshold < EMG) AND (previous EMG is within deadband)
        SAMSstate = SQUEEZE
```

Figure 3.19: Pseudocode of HOLD state

jects, the classification rate fell to 85.5%, but nevertheless indicates that the hybrid hierarchical control scheme should allow the user to provide direct (and accurate) myo-control of the prosthesis. The current limitations of the hardware preclude the further evaluation of the controller. Additionally, the development format of the device and controller prohibits assessment by users, however reaction from members of the upper limb prosthetics community has been positive.

Furthermore, this approach to multifunction control by integrating the myoclassifier with an intelligent control system is unique. In doing so the user is provided with the facility to have autonomous grasping capability (thereby relieving them of the conscious burden) as well as being able to achieve a greater functional ability with the prosthesis. Consequently this system also has the flexibility of providing control of an arm prosthesis before switching to the prehensile function of the hand.

Experimental evaluations of the position, force, and slip control algorithms are not possible due to the current DSP development system and require a supplementary emulator. Specific evaluations of the device's performance are necessary to ensure engineering integrity, however these measurements bear little relation to the operational efficacy of the artificial hand as a whole. Prehensile trials involving the

grasping of compliant/lightweight and non-compliant/dense objects have shown that the hand is capable of executing these grip formats (see section 2.6) but do not, in themselves, form a comprehensive evaluation.

It is clear then that as the primary goal of a myoelectric prosthesis is to reduce the impact of the subject's disability, then the effectiveness of this, or any other device must be quantified in terms of its functional implications. More specifically, an assessment of the 'functionality' of the prosthesis is of paramount importance to all concerned parties, however the method of achieving these functional evaluations is open to significant debate.

Chapter 4

Functionality Assessment

4.1 Issues in Medical Outcome Measurement

4.1.1 Introduction

'Functionality' is a broad-based term that has application in a wide variety of disciplines. Its use in relation to hand evaluation techniques encompasses dexterity, gross manipulative ability, and more generally, the ability to perform tasks encountered during everyday living. Due to the scope of this terminology, one might suggest a generic definition of functionality to be 'suitability to the task'. Thus the level of functionality at which the hand or upper limb performs, must be a measure of how adaptable, or suitable it is to performing the required tasks.

Despite the specific nature of quantifying upper limb function, the underlying principles of evaluation stem from the desire to analyse outcome measures in medicine. This form of procedure, embodied by the more general term of a 'medical audit' has been in existence for some time due to the interest of health-care managers, surgeons, and other parties in assessing treatment effectiveness (that also embodies medical devices such as prostheses).

The medical audit was introduced in the mid-70s as a means of identifying a measurable deficiency from the health-care goals, which were, and are, related to the ultimate objectives of care. Consequently the outcome measure can be defined in terms of achieving, or indeed failing to achieve these goals [100]. This may be reflected in quantifying the impact of an injury or disease on an individual's quality

of life. The World Health Organisation has defined this impact as [100]:

- Impairment – is a loss or abnormality of psychological, physiological, or anatomical structure or function.
- Disability – is any restriction or lack of ability (resulting from an impairment) to perform an activity in the manner or range considered normal for a human being.
- Handicap – is a disadvantage for a given individual, resulting from an impairment or disability, that limits or prevents fulfilment of a role that is normal (depending on age, sex and social and cultural factors) for that individual. Thus the term 'handicap' represents the socialisation of impairment or disability.

Therefore the objective of clinical evaluations of function is to provide a metric of the subject's disability with the rehabilitative goal of reducing their handicap.

4.1.2 The Purpose of Outcome Measures

There are several reasons for monitoring treatment effectiveness by clinical measurement. The overall objective is to improve the quality of medical care by assessing the performance of existing techniques [101]. However there are also many more localised objectives that have greater relevance to the patient, and hence to those more immediately involved in their care (including the medical device researcher).

Clinicians, researchers, and indeed policy-makers frequently require outcome measures for evaluating and monitoring longitudinal change in individuals, especially during a course of treatment, rehabilitation process, or for assessing pre/post-operative performance.

However, more generally there is a common demand for discrimination between individuals (and groups) on an underlying health issue, such as functional performance. This is necessary as a means of not only describing the difference in treatments but also as a method of identifying current deficiencies. Moreover, if the

magnitude of these differences can be quantified, researchers may use these figures to establish group or population trends, which may also serve as an identifier of future needs. Specifically in the area of functional assessment, it may provide a means of highlighting target care standards given current shortcomings.

The techniques used to evaluate individuals vary according to the specific disability under assessment, however the outcome measurement process must adhere to common procedural requirements to ensure its efficacy.

4.1.3 Procedural Requirements for Medical Auditing

Standardisation and Measurement Systems

Outcome measures form an integral part of quality improvement and assurance through medical auditing [101]. At each stage of this process, standards in practice must be set as a model or baseline for subsequent comparative assessments. This ethos is also true within the development of the outcome measurement system itself. A standardised procedure is an essential component in any evaluation process to ensure that the study is both repeatable and reliable.

The traditional approach to assessment has focused on the consultation of 'experts' either individually or as a consensus panel [102]. This form of global subjective measure not only is divergent from the goal of standardisation, but also highlights the invalidity of comparative studies, especially when data are taken from sources outside of the original group.

These assessment procedures are frequently composed of several individual evaluations or tasks. The purpose of such multi-item measures is to combine these tasks to create a score. If a number of different tasks (or dimensions) are covered, these can be presented as separate scores known as profiles or combined to provide an aggregate 'index'. The index is an attempt to create a truly multidimensional measure that accounts for several factors in the overall outcome measure [100]. The scale on which these scores are measured produce varying degrees of quantitative results. A nominal rating involves simply the systematic identification of object

classes without an inherent ordering of values (e.g. sex being classed as male or female). The ordinal scale provides a similar description of classes but are ordered along a continuum, however there are no magnitude values associated with these groups, and hence a hierarchical ranking is all that can be achieved. Interval scales not only provide rank ordering but also specify distance between points on the scale, although no zero point is specified. The ratio scale operates in a similar manner to interval ratings but an absolute zero point, or point of origin is defined (e.g. time or distance). This represents the most comprehensive measurement system, and is therefore appropriate for any objective outcome measure warranting discriminatory analysis.

Creating a standardised and objective assessment technique that adheres to these guidelines is insufficient support of its efficacy. The variability, reliability and validity of the procedure also must be evaluated.

Variability

A normative 'control' group is virtually mandatory if a magnitude-effect outcome measure, which uses a quantifiable means such as the ratio scale, is to be employed. The variability of such a group should be tested, and proven to be normally distributed. Even so, if the data are focused about the mean with a notable absence in the spectrum ends, then much of the chosen scale clearly would be redundant [100], thereby compelling a re-evaluation of the procedure.

Reliability

Establishing reliability warrants minimal random and non-random errors to be present within the normative data set. Test-retest reliability can be shown by minimal variance between two or more sets of replicate results. Inter-rater reliability requires consistency between assessors (or observers). These experiments should be carried out with the same subjects and with a very short time interval between evaluations thereby minimising external random effects. Internal consistency is the final measure of reliability, and is designed to assess the extent to which individual

items are correlated with each other, and with the overall scale scores (i.e. it is an estimate of the homogeneity of the assessment procedure).

Validity

Reliability is not a singularly sufficient measure for the effectiveness of the procedure, as validity must also be ensured wherever possible. However, this can prove to be extremely difficult to quantify during the development of any novel procedure, as the validation process is usually by comparison with a criterion (or 'gold') standard. If such a standard does not exist[1], then it is acceptable to defer to the subjective measure of expert consensus.

In addition to this criterion validity, it is also important to demonstrate content validity, whereby the relative importance of each component within the evaluation procedure can be justified. This decision relies upon the opinion of a panel of representative judges, or by reference to the existing literature, thereby demonstrating that the new measure covers all of the topics previously considered to be important. Although far less demonstrable than the quantitative measure of reliability, the validity factors are nonetheless vital to the outcome measurement process.

Conclusions

These procedural requirements are applicable to all aspects of medical outcome measures, ranging from specific assessment procedures, to health care policies, and therefore fall under the remit of the medical audit. Although patient-focused clinicians may be less interested in the wide ranging implications of outcome measures, it is worthy to note that these procedures are fundamental to ensuring the efficacy and quality of the health care the clinicians themselves provide. Moreover, the generalisations of the procedural requirements are equally as applicable to an evaluation of the upper limb. This is never more so apparent than when considering the potential use of existing procedures for the functional assessment of both natural and

[1]It is reasonable to assume that no gold standard will exist for validation of a new procedure – if such a standard were in place, then clearly there would be little recourse for dismissing the new system as redundant unless a significant time or cost saving had been made.

prosthetic limbs [103].

4.2 Hand Evaluation Techniques

4.2.1 Introduction

Natural Hand Assessment

The use of the hand contributes about 90% of the function to the upper limb [104], however the upper extremity is an entire system with coordinated movement creating overall mobility and dexterity. In order to distinguish the 'functionality' of the hand, it must be assessed as an isolated manipulator (i.e. 'decoupled' from the rest of the upper limb) wherever possible. The predicament lies in identifying the specific tasks that comprehensively evaluate the limb's 'useful' range, for example, rolling a coin between fingers is not an everyday requirement, hence it can be deemed irrelevant during the assessment of hand function; however it is no less valid a test of the true potential of a fully functional natural hand.

Prosthetic Hand Assessment

Evaluations of hand prostheses are even harder to assimilate than those procedures relating solely to the natural hand. Existing solutions have employed either superficial techniques (loosely based on hand assessment tests), or have focused on engineering evaluations (such as range of movement, strength, and system reliability). Since the advent of commercial myoelectric prostheses, these tests often have been used to evaluate the comparative benefit, or disadvantage, of these devices over body-powered split hooks [105, 106], rather than striving to produce an independent assessment of functionality [17, 107, 108]. This has arisen due to the historical need to evaluate new prostheses with respect to the well proven benchmark of the split hook device [109, 110].

The assessment of prosthesis users warrants specific criteria. Unilateral prosthesis wearers rarely use the device for reaching and grasping of objects, and it mainly fulfills a stabilising role for the natural hand in bimanual tasks [43]. The functional

ability of the wearer is dependent upon a wide range of factors relating to prosthetic use (such as the time between amputation and fitment, the user's age, and the weight of the device).

However there has been little attempt to objectively quantify the level of functionality achieved by a hand prosthesis whilst adhering to medical outcome measurement standards. Furthermore, fewer still have attempted to compare such a rating with that achievable by the natural hand (whether dysfunctional or healthy). Establishing a context of hand performance in this manner, whether pathological or prosthetic, allows ready identification of an individual's functionality.

Conclusion

Despite the fact that many hand assessment procedures have been devised and implemented, there is little or no uniformity, thereby precluding any form of meta-analysis[2]. Hence the demand exists for the standardisation of a hand assessment procedure that adheres to the procedural requirements of medical outcome measures and is capable of quantifying both pathological and prosthetic functionality (as detailed in section 4.1.3).

The implications of a technique capable of providing this facility are wide ranging. A comparative evaluation of impaired hand function allows contextual results for a vast range of subjects with disabilities ranging from those using functional electrical stimulation (FES), to upper limb prostheses, or even rehabilitation robots [111].

4.2.2 Prehensile Pattern Classification

In addition to the general medical audit requirements previously described, hand assessment procedures also warrant the identification of prehensile patterns as an additional means of validation. The distinct functional positions of the hand are vital to the method of evaluation by ensuring assessment of the complete range of grip postures. Although there is little conformity to the specific classifications

[2]The ability to analyse data from multiple, independent sources

of prehensile patterns [2, 112, 113, 114] the general characteristics remain largely consistent [115], (see Figure 4.1):

1. Tripod pinch – the pulp of the thumb is opposed to the pulp of the index and middle fingers.

2. Tip pinch – the tip of the thumb is opposed to the tip of one or two fingers. Also known as pulp pinch [116].

3. Lateral pinch – the pulp of the thumb is opposed to the lateral aspect of the index finger.

4. Power grip – all fingers are flexed around the object. In general the palm of the hand is used for object opposition rather than active force generation by the thumb. Also categorised as diagonal and trans-volar [116], fist, cylindrical or hook grips [117].

5. Spherical or Flexion grip – all the fingers and thumb are flexed, rotated and abducted to surround and support the object. Also categorised as the 5 fingered grip [116].

6. Extension grip – all fingers are extended and adducted with the thumb in extension and opposition.

Despite the academic differences between the classes, the majority involve commensurate movements within these categories. The primary goal of this pattern identification is to establish the integrity of the evaluation procedure, and is implicated within the content validity criteria of outcome measurement, despite being unique to hand assessment.

4.2.3 Natural Hand Assessment Procedures

Current clinical assessments are often made by the measurement of range of motion (ROM) [118]. A goniometer is used to measure passive and active joint motion

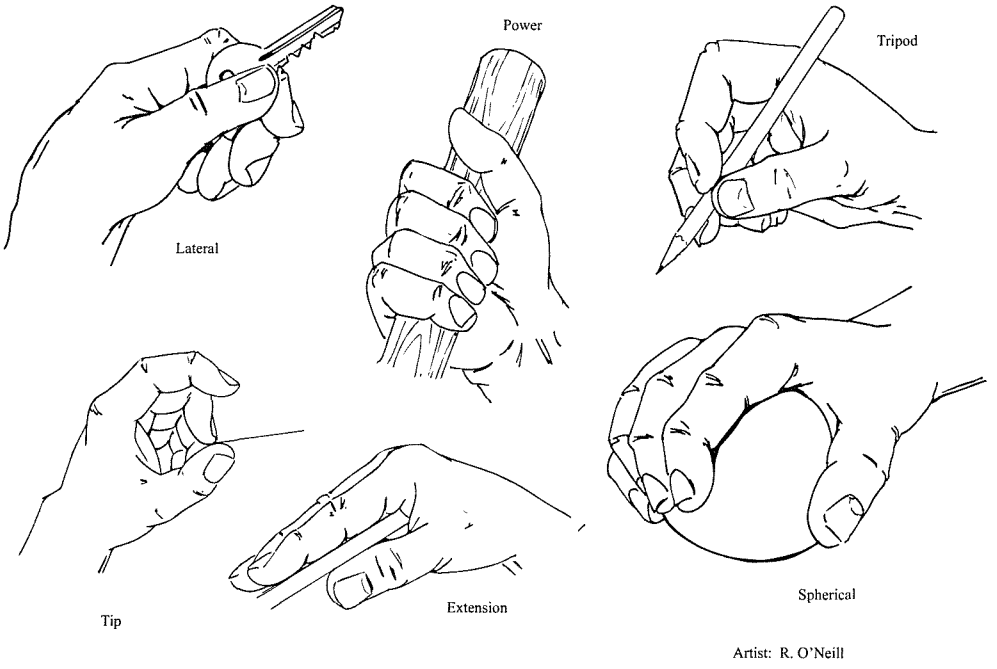


Figure 4.1: Prehensile Patterns

angles (see Figure 4.2), as well as the force/torque exerted around the joint. This evaluation appears to be a valid method of assessing the prehensile range of the hand, and it is obvious that a dysfunctional natural hand or prosthesis is unlikely to achieve the same standards of ROM test results as a healthy hand. However, it remains unclear how these results may be interpreted to provide an unambiguous indication of functionality. No information is provided regarding the capability of the hand whilst carrying out everyday tasks. These tests have also been cited as an ineffectual and unreliable method of assessing hand function [119], as there is need to evaluate the hand whilst accomplishing tasks [120]. Due to the necessity of evaluating the hand in isolation from upper body movement, often the activities of daily living (ADL) used in this context are those tasks that can be performed whilst remaining seated at a table.

Although most existing techniques embody these everyday tasks, other evaluation criteria often are overlooked. Caroll [121] defined a 'quantitative test of upper extremity function' involving 33 sub-tests. The subjective assessment procedure

Figure 4.2: Goniometer for Measuring Range of Motion

used an attributed rating of between zero and three, and was the only measurement criteria employed. Given that no quantitative data was collated, the test can not be evaluated for reliability or variability. These statistical measures were later addressed by Lyle [122] in a development of the procedure called the Action Research Armtest (ARA). However, this method still relies upon a four point subjective scale that precludes a magnitude effect measure.

Based on a set of everyday activities, Jebsen et al. [123] produced a more objective assessment procedure, which was later to be adopted as the classical hand function test. All task assessments centre on determining hand function by speed, with no subjective appraisals (such as assessor's observations or subject's views). The advantage produced by the lack of this subjective component has led to the reliability and longevity of this test. However, as the tasks have no basis in prehensile pattern classification, the functionality rating produced is founded only on a limited scope of the hand's dextrous range. Moreover, repeatability could not be demonstrated given the limited sample size, and the tests were designed solely to be comparative in their use, thereby measuring the *relative* change in the performance of subjects.

Similar in nature is the Moberg test [124] which is frequently used by therapists in a clinical environment to assess the rehabilitation of nerve lesions in the hand. An assortment of objects are manipulated at table top level in a sequence of tests involving both the affected hand (with and without visual feedback), and the un-

affected hand. The timed assessment is used as a measure of hand function. The lack of prehensile pattern identification, as well as little statistical data as evidence of reliability or validity suggest that this test is used in the majority of cases as an instantaneous evaluation of gross dexterity, rather than for use as a quantitative procedure.

Time measurement in these tests precludes the equal assessment of those failing to complete a certain task (due to an imposed limit) but capable of performing in others. In addition to this, the assessment of writing as a functional task (as used in the Jebsen test) is potentially unreliable, given the dubious correlation between speed and dexterity [120].

In order to provide a quantifiable measurement (and thereby reduce comparative evaluations), Clawson et al. [119] attempted to create a reliable index of hand function. This index was based on the results of five subtests measuring muscle power, architectural stability, and gross grasp and co-ordination, by the measurement of grip strength or time to complete tasks. Although quantitative data was collected, the limited range of tests utilised to assess hand function, and the lack of identification of prehensile patterns, provides uncertainties over the reliability of the index rating and its correlation with functionality.

Gross body movement during assessment is a factor that must be minimised in order to focus solely on hand performance, and those procedures that fail to adhere to such criteria [125] invite doubt over their validity. To avoid such criticism, Walker et al. [126] developed a set of apparatus, designed specifically for the metacarpophalangeal joint, to assess arthritic hands. The tests measured active and passive ROM, strength (for various prehensile patterns), and manipulation (holding, placing, twisting of objects) by recording force and task performance times. The apparatus used was specific to the application, and proved unfeasible for widespread hand functionality assessment. Other studies have also led to the production of hand assessment apparatus [127], however these devices are frequently complex, specialist, and difficult to evaluate in conjunction with other clinical testing.

Sollerman [128] addressed many of these issues in the development of a series of hand function tests based on activities of daily living. Scoring is based on the use of appropriate grip patterns, and ranked by an interval time scale (e.g. the task was completed within 20, 40, or 60 seconds). From these results, it was intended to produce a rating that would establish functional levels for patients with hand disabilities (e.g. osteoarthritis, finger amputations etc.). The test is founded on the sound principles of assessing hand function by determining, and implementing, suitable tasks based on prehensile patterns, and their percentage use in everyday living. The disadvantage of this scale can be highlighted by the subsequent analysis, which is limited solely to a hierarchical ranking rather than a measure of the magnitude effect of functional deficiency.

The TEMPA[3] test of upper extremity function for the elderly [129] conforms to many of the criteria necessary for clinical outcome measures. The procedure is a bimanual assessment of the upper limb based on the premise that functionality cannot be isolated to the hand, and instead warrants measurement of the "global function of the entire upper extremity". However the main disadvantage is that the focus on a single cohort group (the elderly) does not accommodate widespread clinical use. Reliability of the procedure has been established [129], and a large normative database has been collated and correlated with sensorimotor parameters such as range of motion, strength, sensibility and dexterity [130]. In addition the validity of the procedure has been evaluated [131] with respect to the Action Research Armtest [122], which was cited as a good measure of functionality but lacking in clinical relevance or commercial availability. The TEMPA activities of daily living are not based upon prehensile patterns, despite the authors highlighting everyday grip use as an important factor in procedural development. Although reliability has been established, the subjective four point scoring scale (used in conjunction with objective measures) provides a restrictive method of classifying functionality, and raises questions over the efficacy of a validity measure with only four potential outcomes. The

[3]Test Évaluant les Membres supérieurs des Personnes Âgées

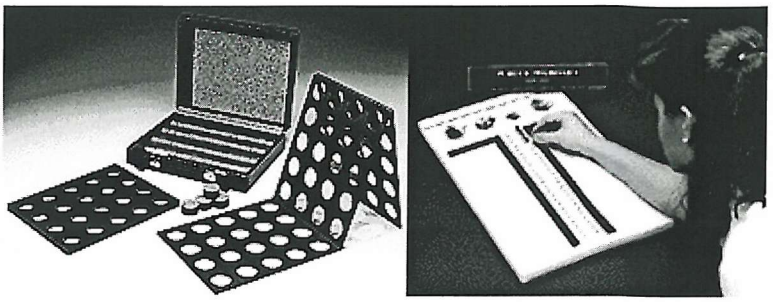
assessment of bimanual upper extremity performance also detracts from the focus of evaluating the subject's disability due to hand function, and instead provides a measure of their adaptability to an impairment.

Several other hand functionality tests have been developed with differing emphasis on assessment requirements or evaluation procedures [132, 133, 134]. These tests suffer from poor statistical validity, often attributed to either subjective assessment, or poor timing accuracy [135], which results in adverse reliability. Even procedures that are founded on sound outcome measurement criteria such as validity and repeatability [136], may not achieve the necessary reliability during trials (possibly due to flaws in the standardisation or execution of the procedure).

Hand function or general impairment assessments used by therapists [104] include the Minnesota rate of manipulation (gross manipulation assessment – see Figure 4.3a), Purdue pegboard test (fine co-ordination – see Figure 4.3b), Crawford small parts dexterity test (fine co-ordination), use of the Jamar dynamometer (a clinical evaluation of grip strength) and the pinch prehension meter (a clinical evaluation of pinch force). Tests such as the box and block test used to measure gross manual dexterity, and the 9 hole peg test (to assess finger dexterity) have undergone extensive testing to produce normative adult data [137]. However, all of these assessments are either highly specific (i.e. cannot provide information on overall hand function), or are subjective in their evaluation procedure. This does not make them redundant in the clinical environment however, as they all provide an indication of functionality, and are particularly useful for a therapist monitoring the rehabilitation of a patient over a course of treatment. Nevertheless these tests cannot be used to provide more quantitative data for widespread comparisons of relative impairment.

4.2.4 Artificial Hand Assessment Procedures

Hand prostheses have been evaluated by numerous methods, but technical assessments (such as grip strength and maximum grip size) have tended to dominate.



(a) (b)

Figure 4.3: (a) Minnesota Rate of Manipulation Test, (b) Purdue Pegboard Test

Given the engineering development involved in the production of the prosthesis, this is perhaps unsurprising. However, there is a notable lack of broad functional assessment for the use of the prosthesis during everyday living, and as mentioned previously, there appears to be little or no comparison between prosthesis functionality and an obvious benchmark, such as the natural hand.

The lack of objectivity in the design of these assessment procedures was addressed by Kay [138] who developed a series of comparative tests for the Belgrade hand, based on functional testing proposals [139]. The tests were to provide an assessment of the device's adaptability, for comparison with two other types of commonly used hand prostheses (voluntary opening and voluntary closing split hooks). The procedure consisted of two abstract tests (the 'form' board, and 'pigeon hole' tasks) and one practical test (involving twenty five bimanual ADL).

The form board test was designed to focus specifically on the grasping functions of the prosthesis, and involved the transfer of objects of differing sizes, weights, and consistencies from a flat surface to a 'form' board (with appropriate slots for the objects). The assessment was measured by the time taken to perform each task, a count of handling errors, and the subjective ratings of a therapist.

The second abstract 'pigeon-hole' test required the subject to move different objects through several levels of shelving (at mouth, chest, waist and knee height). The evaluation was designed to focus on the subject's ability to grasp and release objects at differing elevations. Although arguably an assessment of upper limb

functionality, and absent from any evaluation of natural hands, this test has specific emphasis and validity for prostheses due to some devices relying on upper limb movement. Body-powered hook prostheses require shoulder movement to activate the device, hence the manipulation of objects at different heights to the body may be affected. This also enabled an evaluation of myoelectric prostheses during reaching to ensure electrode contact was not affected by the motion. Scoring was based, as for the previous test, on time, error count, and subjective assessment. The abstract tests have sound basis, however the lack of identification of prehensile patterns and the subjective scoring throughout, raise questions of reliability and validity.

These techniques were also employed by Codd [17], who considered time measurement alone to be an insufficient criteria on which to base the effectiveness of hand function. A subjective rating scale of performance (compared to a Dorrance hook), times taken to execute tasks, the orientation of the device, the type of grip used, and other comments, were compiled to form an assessment of functionality. Although the tests were based on the Belgrade evaluations, a greater identification of grip patterns was used, thereby lending more credibility to the subjective commenting. The abstract test remained unchanged from those previously detailed, whilst a new set of ADL was compiled. Although the grip patterns have been identified during the testing, it is not clear whether each task has been founded on the appropriate prehensile pattern. The results rely on the opinion of the trained assessors [93], which again raises questions over the reliability or validity of this form of scoring protocol.

This form of expert evaluation is also used in the University of New Brunswick (UNB) Test of Prosthetics Function [140], which is designed to provide a method of assessing an individual child's progress during functional training with an upper extremity prosthesis. The UNB procedure represents one of the most comprehensive facilities for monitoring functionality in prosthesis users. A rating scale is used for different age groups to measure spontaneity and skill of prosthesis use during children's activities that are representative of their everyday living. The tasks are

matched to the child's age and experience with the prosthesis, which leads to a series of tasks that are unsuitable as activities for adult subjects. The test is proposed as a set of guidelines, rather than a regimented procedure, in order to allow for differing cultural emphasis in hand function. The subjective measurement (and variability in test items) produces a lack of qualitative data regarding functionality. Consequently, although the test has been designed for the evaluation of upper limb prosthesis function (with extensive testing having taken place), the specialisation of the procedure for children, and the subjective measurement technique employed precludes a more general use.

Myo-prostheses (an ES adaptive hand, and an Otto Bock) have also been evaluated using the natural hand function tests developed by Sollerman [128]. Bergman et al. [107] utilised the 20 tasks, assessing each patient on three separate occasions. Using two independent assessors, a reliable result was achieved allowing direct comparison of the two prostheses, however the disadvantages of the original tests remain.

Other prosthesis testing includes comparisons between myoelectric devices and split hooks [105], which used block testing (similar to the 9 hole peg test), and activities of daily living as part of broader field evaluations [106], including medical examination and psychological testing. Neither of these assessments sought to provide comparative data for the general evaluation of functionality. Mendez [110] adapted occupational therapy ADL tests for the long term assessment of myoprostheses for children. This constitutes one of the most significant evaluations of prosthetic hands, given the duration of the trial (2 years) and the large sample size (87 subjects), but provided information about the general acceptance of prostheses by the users (as well as the technical ability of the devices), rather than the level of functionality achievable.

The selection of everyday activities (in conjunction with identifying grip patterns) is one of the critical factors in the development of hand assessment procedures. As is evident from the review, few have adopted this criteria. Although each assessor has differing priorities for the representation of standardised ADL, the se-

lection of these activities for the majority of assessment procedures appears to have little foundation on prehensile patterns, or even uniformity of 'everyday activities'. McWilliam [141] addressed this problem by an extensive survey and compilation of "a list of everyday tasks for use in prosthesis design and development". The list was formed from the ranking of activities considered to be needed for the personal independence of an adult, into essential, useful and trivial ratings. Activities pertaining to jobs or recreation were excluded on the grounds of specificity (given that functional assessment cannot account for the idiosyncrasies of a wide range of occupations or hobbies). Over six hundred tasks were identified as potentially useful, and following prioritisation, a resultant one hundred and forty two tasks were proposed for assessment purposes. The compiled list was based purely on likely ADL, rather than by identification of specific grip patterns, hence the result is an extensive list that would be impractical to implement for regular assessment purposes. However, the compilation can be used as a source of reference for determining a more realistic set of activities for functionality assessment trials.

4.3 Discussion

Both the natural and prosthetic hand assessments have utilised objective and subjective evaluation techniques. It is important to distinguish between these methods and highlight whether they have an impact on the validity or reliability of the tests. An objective evaluation is usually based on quantifiable measurements (such as the time to perform a procedure), and is inherently likely to produce the most quantitative and reliable data (providing the measurement is unaffected by external factors such as the reaction times of the assessor). Consequently, objective assessments should be advocated wherever possible. Subjective assessments rely on personal opinion, and high correlation between assessors has been demonstrated for different procedures at some centres [107, 137]. However, these results cannot be extrapolated to be the norm for a standardised procedure, especially when introduced to an international forum, where potentially broad cultural differences in hand therapy become

more evident. This in no way minimises the importance of subjective evaluation during the rehabilitation of a subject, but merely aims to quantify the requirements necessary for developing a rigorous, standardised and objective procedure.

The factors that adversely affect the selection of existing assessment procedures are many and varied. Current evaluations have tended to overlook effects such as body movement, the specificity of the procedure, or the requirement to establish magnitude effects when measuring functional deficiency rather than simply allowing ranked comparisons. There is also an obvious expectation to prove the reliability and validity of any assessment procedure, yet there are many that do not attempt to statistically establish these criteria. This remains a common characteristic of virtually all assessment procedures – a notable divergence from medical outcome measurement standards.

4.4 Conclusion

From the review it is apparent that there is little or no conformity to a standardised procedure for the assessment of both natural and prosthetic hands. Conventionally the measurement of hand function during these procedures has been by: time limit (quantity completed in a finite time period); work limit (time to complete task); qualitative scoring (the way in which the object is handled); or assessor's opinion (e.g. level of difficulty for the subject). Time is obviously an easy parameter to measure and manipulate statistically, however it is not necessarily the most valid measure of hand function [109, 123, 124]. Despite this, there appears to be little alternative without the development of more complex yet specialist apparatus [126, 127].

Any assessment requiring subjective ratings cannot easily lead to the production of a standardised and reliable evaluation procedure, as opinion cannot be considered conformist when data collation could potentially be from international sources. This is particularly evident when assessing grip stability, or difficulty of prehension, by subjective means.

Hence, given that the majority of procedures are based upon everyday tasks, a series of conclusions and design criteria can be drawn regarding the compilation of standardised assessments [142]:

Criterion 4.1 *The tests must cover all ranges of grip (and percentage use thereof), with direct relevance between prehensile patterns and the selection of ADL.*

Criterion 4.2 *No subjective opinion should be used during the assessment of hand function as this inherently introduces greater variability in the evaluation procedure.*

Criterion 4.3 *The procedure must establish a standardised protocol, and the normal distribution variability of a control group must be proven.*

Criterion 4.4 *The reliability (test-retest, inter-rater, and internal consistency) of the assessment must be demonstrated.*

Criterion 4.5 *The validity (criterion standard and content validity) of the procedure must be evident.*

Criterion 4.6 *The current technology (for hand prostheses) should not in any way affect the determination of the test procedure.*

Criterion 4.7 *The time to carry out the assessment procedure, and the apparatus required must be suitable for the therapist and the clinical environment, i.e. a lengthy test procedure, or costly apparatus are unacceptable.*

Chapter 5

Development of SHAP

5.1 Objectives

The desire to evaluate the functionality of the Southampton-Remedi hand prosthesis has highlighted a more general and clearly identifiable need to quantify the functionality of both pathological and prosthetic hands in the clinical environment. Current evaluations frequently fail to meet the criteria necessary for a comprehensive outcome measurement system (as outlined in section 4.2). Applicable to the assessment not only of impaired natural hands, but also of prostheses, the Southampton Hand Assessment Procedure (SHAP) is a timed evaluation, comprising of 12 abstract object tasks and 14 activities of daily living [143, 144, 145]. The index and prehensile profiles of functionality that are derived from the timed tasks are metrics of a subject's disability.

SHAP provides the assessor with an objective appraisal of the subject's functional level relative to that of a norm. The advantage of such a rating method is that it may be used both to quantify a subject's impairment in a single appraisal (rather than by relative improvement measures [123]), and also track the rehabilitation of a patient during a course of treatment.

The purpose of the hand assessment procedure is not to establish a subject's overall function, which is actually a measure of their adaptation to their disability[1].

[1]Prosthesis users require coordinated movement of the upper limb and therefore do not exhibit the separable functions of hand shaping and arm movement seen in natural upper limb subjects [146]. Hence conventional assessment of the user does not isolate 'hand' function.

Instead, in the case of prosthesis users, it is to determine the effectiveness of a terminal device and controller by focusing the evaluation on the unilateral performance of the user. This process should highlight functional differences between devices and suitable control schemes [147] for the wearer, whilst still providing a contextual measure of hand function relative to that of a norm.

The procedure also has wide-ranging research implications for the investigation of specific patient cohort groups, such as the evaluation of hand function in osteoarthritis sufferers, stroke patients, subjects using functional electrical stimulation, burns victims, and hand trauma injuries.

5.2 Assessment Methodology

During the development of the assessment procedure a number of tasks were considered and their relative merits evaluated. The review of existing techniques [103] provided a forum for emphasizing potentially useful activities as well as highlighting those that may be inappropriate. The following section outlines the methodology for inclusion or exclusion of these activities, which ultimately result in the 26 tasks that comprise the Southampton Hand Assessment Procedure.

5.2.1 Abstract Object Tasks

The stability of grip, or deviation of prehension from a norm, has consistently been evaluated by subjective assessor opinion. As previously described (see section 4.2), this causes difficulty in reconciling the standardisation of the procedure. Hand assessment techniques are composed conventionally of a series of activities of daily living that are used to evaluate functionality, however other methods do exist. Formboard tasks have been employed in different formats for both natural and prosthetic hand assessment [104, 138]. This 'abstract' test evaluates prehension without the complication of tools or equipment used during activities of daily living (ADL), which often cause intermediate grip patterns or adverse evaluation effects – as the shape and form of an ADL task is likely to be known to the subject, a psychological

prejudice may exist as to their ability to perform the task. To a limited extent, the abstract objects remove such an effect.

The adaptation of the form board test for SHAP has led to a procedure whereby specific grip patterns (determined by object shape) account for the functional range of the hand. The timed tasks involve the movement of each of these objects individually. Any subject unable to perform the natural grip for a specific task is expected to take longer by implementing an abnormal grip pattern in order to move the object with an optimum balance of grip pressure and stability. Subjects with impaired hand function (either pathological or prosthetic) frequently perform compensatory movements that enable them to carry out the task more quickly than if implementing a natural pattern. Although they are encouraged wherever possible to achieve natural prehension, the result is scored nevertheless as it remains a reflection of the subject's functionality.

The form board object designs are produced in two sets for use in the procedure (see Figure 5.1). The first set are manufactured from non-compliant dense materials (denoted as 'heavyweight abstracts'), and the second from marginally compliant, low density materials (denoted as 'lightweight abstracts') to produce a difference in both weight and yield. This provides a means of implicitly evaluating the method of grip implementation and also mitigates for subjects with poor grip pressure.

The 'pigeon hole' test (see section 4.2.4) has been used to evaluate the functionality of prostheses during upper limb movement [108, 138]. However, this test requires coordinated tracking, trajectory and grasping motions (a notably different process in prosthesis users than in normals [148]), and is questionable as a method of assessing pure hand function. This omission will fail to highlight the poor performance of body-powered devices in this area of upper limb movement. However the assessment procedure is designed to allow an evaluation of all forms of hand, natural and prosthetic, and therefore must preclude, or at least minimise, the assessment of arm function wherever possible[2].

[2] All SHAP tasks focus on prehensile ability and involve minimal transport effects thereby limiting the assessment of gross upper limb function.

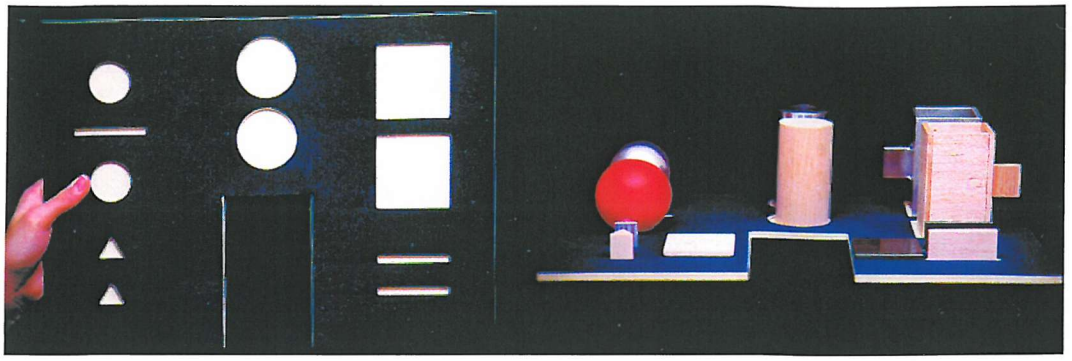


Figure 5.1: SHAP Form-Board and Abstract Objects

5.2.2 Activities of Daily Living

The compilation of Activities of Daily Living (see Table 5.1) was sourced from the most significant and reliable hand or prosthesis assessment procedures. Although not reliability tested, the essential tasks identified by McWilliam [141], were also included within the selection criteria to ensure the broadest range of daily living activities were to be considered.

Any activities requiring subjective assessment or likely to cause a large variability in timing were omitted. For example, the task of writing, although an important everyday activity, was excluded on the basis of large variability in the performance of writing skills with little relevance to hand functionality (i.e. the speed of writing is not necessarily correlated to hand dexterity).

Sollerman [128] estimated the percentage use of their eight types of grip pattern during everyday living. These results were incorporated within the six SHAP prehensile categories described in section 4.2.2, thereby obtaining estimates of grip pattern use during habitual activities.

Each selected activity was assigned to the most appropriate prehensile pattern classification(s). The SHAP ADLs were then compiled in approximate proportion to the Sollerman percentage use values, such that a spherical grip was required for 10% of the SHAP ADLs, a tripod grip for 10%, a power grip for 25%, a lateral grip for 20%, a tip grip for 20% and an extension grip for 10%. This ensures that the

Figure 5.2: SHAP Activities of Daily Living

full range of natural grips has been evaluated. It also enables the measure of overall functionality to be a reflection of everyday hand performance without the need for any weighting or adjustment of the prehensile pattern data.

Based on the above criteria, the 14 activities of daily living were selected as shown in Table 5.1 and Figure 5.2.

No.	Task	'Natural' Grip Classification
1	Pick-up coins	Tip
2	Buttons	Tip/Tripod
3	Food cutting	Tripod/Power
4	Simulated page turning	Extension
5	Remove jar lid	Spherical
6	Pour water from jug	Lateral
7	Pour water from carton	Spherical
8	Move empty tin	Power
9	Move full jar	Power
10	Move tray	Lateral/Extension
11	Rotate key 90°	Tip/Lateral
12	Open/close zip	Tip/Lateral
13	Rotate door handle	Power
14	Rotate screw 90°	Power

Table 5.1: SHAP Activities of Daily Living

5.2.3 Deriving an Index Of Functionality (IOF)

The traditional subjective method of assessment has been avoided by the use of a self-timed technique. Reaction time effects from both assessor and subject have been known to cause adverse effects in the standardisation of other hand assessment procedures [135]. Consequently the individual under assessment always starts the timer at an origin position, and then returns to stop the timer once the task is complete. By asking the subject to commence the test at their own discretion, the disadvantages of assessor timing (that involve verbal start and aural stop commands) are avoided.

In the case of a subject taking an excessive period of time (or being unable) to complete a task, then a boundary condition must be introduced. Other procedures have imposed boundary times and conditions without consideration of the individual nature of the task [44, 123, 128]. For example Jebsen [123] limited subjects to a time of eighty seconds for each task, which ranged from 'moving of large light objects' (with a normal mean time of three seconds) to 'writing' (with a norm mean time of twelve seconds). Hence a single boundary condition seems inappropriate given the large variability in task times.

This boundary limit can be viewed as the point of minimal function. Myo-prostheses users take approximately six times longer to complete a task than a subject with natural hand function, and twice as long as body-powered prosthesis users [149]. Hence as one of the slowest functional groups, myo-prosthesis wearers have little function, but nevertheless require classification on the functional scale. Given consideration of previous assessment procedures, as well as the average expected performance of those with severely impaired hand function, a boundary condition of eight times that of the norm is imposed for each SHAP task individually.

Initially the score attributed to a subject was compiled from the summation of each of the twenty-six task times, which is converted to a functionality index by means of a z-score [145]. This measure quantifies a subject's rating in terms of

standard deviations from the norm (see Equation 5.1).

$$z = \frac{(x - \bar{x})}{s} \tag{5.1}$$

where z is the z-score, x is the subject's task time, $\bar{x}$ is the mean time in the normative sample, and s is the standard deviation of times in the normative sample.

The z-score measure is rescaled to a value of 100 when x is equal to the corresponding $\bar{x}$, diminishing to 0 for a subject who reaches the boundary condition for each task (and hence is deemed to have 'minimal function').

In addition to this generic measure of functionality, a more specific result may be obtained by the summation of task times within the prehensile groups. Hence a functionality profile for each of the six prehensile patterns can be obtained and may be used to supplement the overall analysis of the subject's disability.

This approach creates an aggregate score of overall function and a set of six prehensile pattern profiles based upon the whole procedure. The disadvantage of this method is that it disregards the truly multi-dimensional nature of the assessment. The philosophy for establishing a procedure based upon multiple prehensile patterns (or dimensions) is that these form the underlying basis of hand function. Consequently the resulting index of functionality (IOF) should be founded on the outcome of these groups.

5.2.4 A Multi-variate Approach to an Index Of Functionality

This IOF measure must inherently account for the multiple prehensile groups that are integral to the procedure. Hence it is necessary to take a multi-variate approach to establishing the six prehensile profiles, and subsequently the overall index of functionality.

The Euclidean squared distance is a measure between samples in an i-dimensional problem where i=1,2,...,6 (prehensile patterns) in this case. This can be illustrated by means of a geometric equivalent: for the case when there are two dimensions, say y_1 and y_2 (see Figure 5.3a), the use of this Euclidean metric equates to measuring

distances by circles – points A_1 and A_2 are on the same circle and lie the same distance from the centre C, whereas B_1 and B_2 lie on an outer circle and are therefore further away from C [150]. This principle can be applied equally to any number of dimensions, although the geometric model becomes harder to visualise.

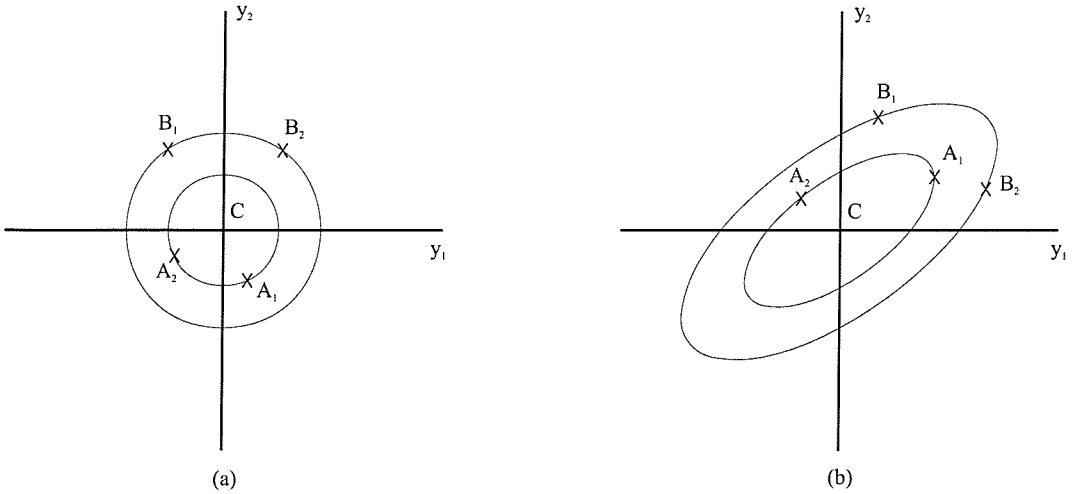


Figure 5.3: (a) Euclidean and (b) Mahalanobis Squared Distance Measures

The Euclidean distance (d) is determined using the z-value (from Equation 5.1) for each of the prehensile patterns (i), thereby giving a multi-variate metric from the norm in each case (as shown in Equation 5.2).

$$d = \sqrt{\sum_{i=1}^{6}(z_i)^2} \tag{5.2}$$

If observations (or tasks in the case of SHAP) have different variances, the unweighted Euclidean measure may be an inappropriate method of determining the extent of a subject's abnormality. Based on the example given previously, if y_1 has a larger variance than y_2, then it may be desirable to weight a deviation in the y_1 direction less than an equivalent deviation in the y_2 direction. This can be achieved using an "elliptical" distance measure, such as the Penrose distance [151]. However, this measure does not take account of any correlation between the i variables. Consequently if two variables are essentially providing the same evaluation (and are therefore highly correlated), they both contribute to the distance in approximately

the same proportion as a third uncorrelated variable. The Mahalanobis squared distance (see Figure 5.3b and Equation 5.3) is an elliptical distance measure that also accounts for the correlation between variables [150] and therefore can be used as a metric for a subject's abnormality.

$$d_M^2 = (x - \bar{x})' S^{-1} (x - \bar{x}) \tag{5.3}$$

where d_M^2 is the squared Mahalanobis distance, x is a vector consisting of $(x_1, \ldots, x_6)'$ that are the six prehensile pattern times of the subject in question, $\bar{x}$ is the mean x vector of the normative group, and S is the sample covariance matrix for the normative group.

In a similar manner to obtaining an index of functionality from the multi-variate prehensile patterns, each of these groups themselves are made up of multiple tasks. Each of these individual indices is based on the Euclidean distance in t dimensional space, where t is the number of tasks in the patterns, ranging from 4 to 7.

Abnormality or Disability?

The purpose of the hand assessment unit is to determine the extent of a subject's disability due specifically to an impairment of the hand. The Mahalanobis distance is a multi-variate measure of the 'abnormality' of a subject relative to the mean norm. This measure can be used to determine the probability of the subject coming from a normative population, by

$$P = 1 - F(d_M^2) \tag{5.4}$$

where P is the probability of a normal subject producing the Mahalanobis distance (d_M), and F is the cumulative distribution function for the index or profiles of functionality[3]. For example, if a subject should take well above the norm time to complete the SHAP test, then the resulting Mahalanobis distance measures will also be high. Therefore the probability of the subject possessing normal hand function

[3]The F functions differ according to the number of tasks, t, that comprise each of the prehensile profile scores, or the overall index of functionality.

is small according to Equation 5.4. Hence a metric of abnormality can be obtained from the Mahalanobis distance.

However, the abnormality of a subject does not necessarily provide an indication of the level of disability. To quantify the extent of functional loss (or distance from the norm) therefore warrants a return to the Euclidean distance metric. If each task contributes to an assessment of functionality, then the correlation between them can be considered irrelevant in determining disability. Consequently, the Euclidean measure is used to calculate the six prehensile pattern profiles according to Equation 5.2, and an overall index of functionality by similar means.

5.3 Adherence to Procedural Requirements

Medical outcome measures must adhere to a set of procedural requirements (as outlined in section 4.1.3) to ensure their efficacy. The Southampton Hand Assessment Procedure has been evaluated and the results used to ensure conformity to these standards.

5.3.1 Standardisation and Measurement Systems

The procedure for executing a SHAP test is documented in detail (see Appendix D) and includes specification of the subject's initial position relative to the assessment board, and the exact layout of each task. It also specifies the prehensile pattern to be used during the demonstration of each activity, and the subsequent instructions that must be given to the subject. Hence a standardised procedure has been established that ensures, to as great an extent as possible, that the process is both repeatable and reliable.

Task times are an absolute measure and therefore fit a ratio scale (with a known zero). The outcome of SHAP avoids both subjective assessment as well as ensuring that results possess a magnitude effect rather than relative ranking as exists with alternative scales.

However, speed is not synonymous with hand function [120, 152] as other factors

such as grip strength, inherent stability of the hand, and proprioception contribute to overall functionality. Nevertheless, essential hand function and the speed of performance of tasks are both determined by the ability of the hand to form a natural and optimal prehensile structure. Hence the time taken to execute a task will be strongly correlated with function. Although this execution time undoubtedly varies among normal subjects (see section 5.3.3), pathological hand function will almost certainly cause that speed to differ from that of the norm to an appreciable degree. The comprehensive assessment of all factors affecting impairment would require a complex and expensive procedure that precludes it from clinical use. The foundation of the SHAP measure on prehensile patterns should result in a more valid evaluation of hand function than is possible from existing timed procedures.

5.3.2 Establishing Normative Data

In order to demonstrate statistically that the procedure meets medical auditing requirements, it is necessary to establish a normative control group whilst adhering to ethical guidelines[4]. This group was comprised of 18-25 year old undergraduates that experienced no adverse hand trauma, neurological condition or disabling effects of the upper limb. These subjects were hypothesized to possess near optimum hand function given that they are within a prime health age group, and have no specialist occupation that may adversely affect performance. A graphical user interface and relational database has been created to aid data collation (carried out in accordance with the structure shown in Figure 5.2).

Initially a group of eighteen subjects (nine males and nine females, designated 'Group A') were assessed, and each evaluation replicated three times. A single assessor was used throughout these control studies, and a minimum period of 24 hours was allowed between replicate assessments (thereby attenuating a direct learning effect by the subjects). Inter-rater reliability (see section 5.3.4) was indicated by the assessment of three further subjects (designated 'Group B').

[4]Ethical approval obtained from the Southampton and South West Hampshire Joint Research Ethics Committee, UK (submission no. 014/98).

Group	No. of Subjects	M/F	No. of Assessors	No. of Replicates
A	9	M	1	3
	9	F		
B	2	M	3	3
	1	F		
C	2	F	1	3
	1	M		

Table 5.2: Normative Data Collection

An additional three normative studies ('Group C') were carried out by a single assessor, thereby producing an overall group of twenty-four samples (each with three replicates). These results form the basis from which all subsequent assessments of pathological or prosthetic hand function can be referenced. Figure 5.4 shows the mean task times obtained from the normative group. Error bars indicate twice the standard error of the mean, thereby depicting a 95% confidence interval.

Although subjects are provided with an index of functionality relative to that of the 'optimum norm', it would be advisable to establish additional normative data for different age groups – hand function is expected to diminish with age [137], hence it is prudent to provide clinicians with an aged-matched benchmark of normal function.

5.3.3 Variability

The control group data must be shown to possess a normal distribution in order to meet outcome measurement criteria (see section 4.1.3). A common method of verifying normality is to form a normal quantile plot [153]. If the data are normally distributed, all points in the plot will map close to a straight line[5].

In order to comprehensively ensure normality, each of the tasks from the normative database were tested as well as the multi-variate index of functionality. A linear

[5]For example, given ten data points arranged in ascending order, the first data value lies at the 0.1 quantile. This is the value z for the normal distribution such that 0.1 of the area under the probability density curve lies to the left. A table of standard normal probabilities reveals a z-value of -1.28 for an area of 0.1. This is repeated for all ten quantiles, and the original data is then plotted against the z-value.

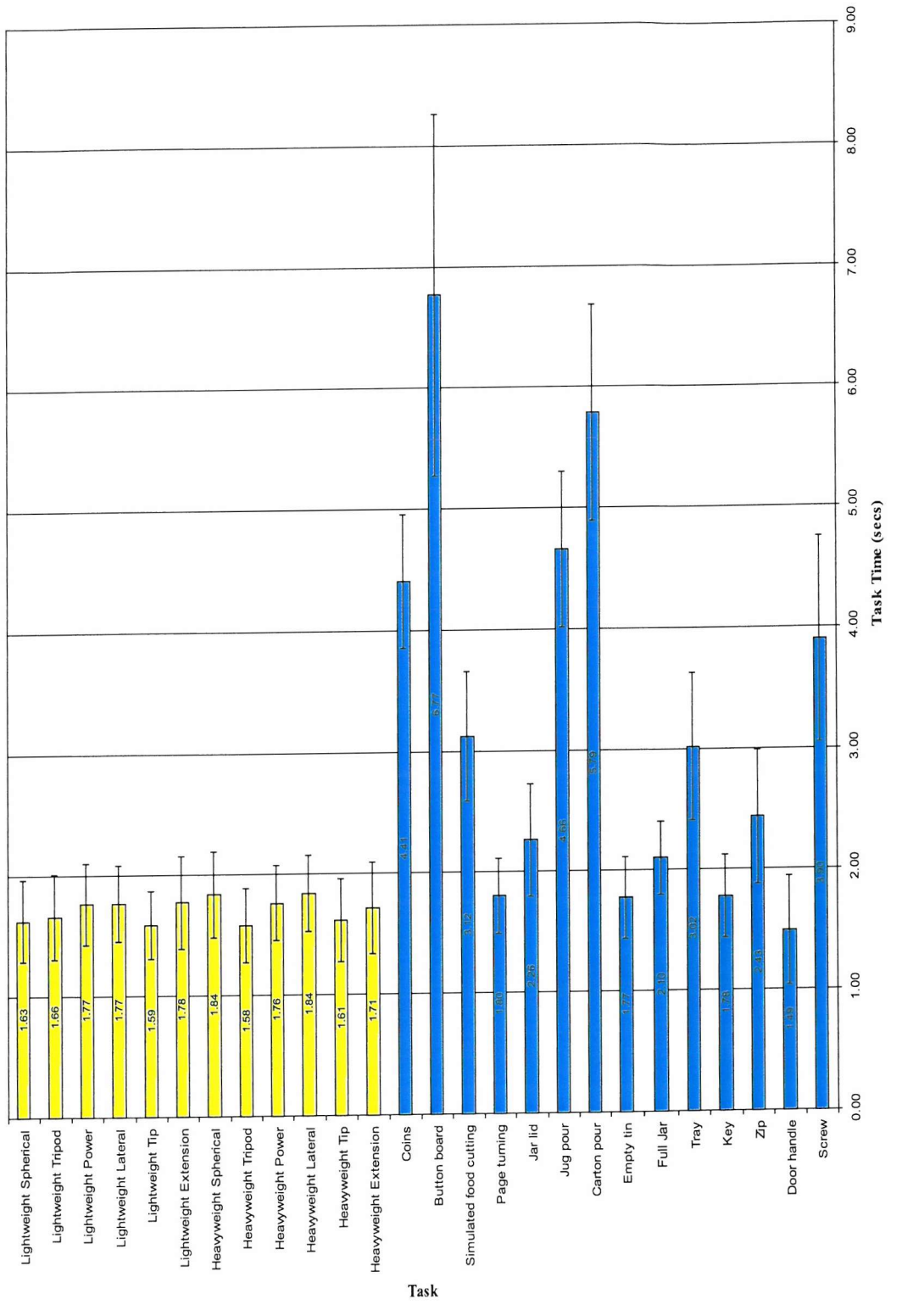


Figure 5.4: Mean Normative Task Times

fit trendline to the normal quantile plots for each task revealed R^2 values ranging from 0.88 (heavyweight extension abstract object) to 0.99 (pouring water from a carton ADL), with an overall average of 0.93. Hence the control data for each task can be deemed to be normally distributed given a near straight-line relationship in each plot.

In order to use a multi-dimensional metric for the index and profiles of functionality, it is necessary to establish the multi-variate normality of the control group. If the data are indeed multi-variate normal then the Mahalanobis distance from an individual will have an F-distribution when re-scaled according to Equation 5.5.

$$M = \left(\frac{n}{(n+1)} \times \frac{(n-i)}{(n-1)i} \times d_M^2\right) \sim F_{i,(n-i)} \tag{5.5}$$

where M is the F-scaled squared Mahalanobis distance (d_M^2). This is proportional to the F-distribution ($F_{i,(n-i)}$), where n is the number of observations from the normative population (24), and i is the number of dimensions (6).

The M values are calculated for each subject in the normative group, subsequently sorted into increasing order, and then plotted against the quantiles of the $F_{(6,18)}$ distribution. The result is plot with an R^2 value of 0.93 to a linear trendline (see Figure 5.5), thereby indicating that the data appear to be near multi-variate Gaussian [150, 154].

5.3.4 Reliability

Reliability is perhaps the most crucial measure of the efficacy of the assessment procedure. The test must produce consistent results among a control group with both a single assessor (test-retest reliability) and multiple assessors (inter-rater reliability).

Test-Retest Reliability

A single rater assessed control Group A with three replicate evaluations for each subject. To establish test-retest reliability, it is necessary to show minimal variance between the interaction of the subject and the tasks (thereby indicating that there is no statistically significant effect in the replicate trials).

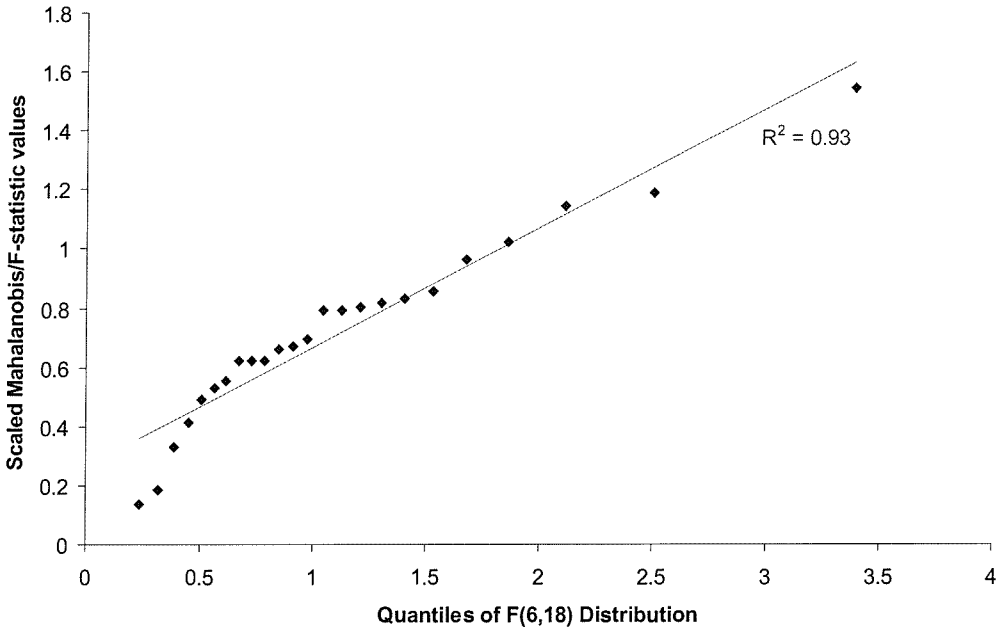


Figure 5.5: Quantile Plot of Multi-variate Normality

The most appropriate method of evaluating whether the data fulfil this criterion is by an analysis of variance (ANOVA), based on the null hypothesis that no significant difference exists between replicates. This hypothesis is tested in this case at an alpha level of 0.05 (the significance level related to the probability of having a type I error, that is, rejecting a true hypothesis). The F-value obtained from the ANOVA must exceed an F_{crit} value, that is based on this 95% confidence interval, in order to prove that the null hypothesis should be rejected, and therefore that there is a statistically significant difference between replicates [155].

The disadvantage of this statistical technique is that the result depends not only on the degree of departure from the null hypothesis, but also on sample size [156]. Consequently the result of performing an ANOVA test on all eighteen subjects, covering all twenty-six tasks and three replicate trials (a total of 1403 degrees of freedom) means that small effects can readily achieve extreme significance levels.

Hence an ANOVA test has been performed for all subjects and replicates on a

task by task basis. If the procedure can be shown to be repeatable at an elementary level, then the overall assessment should also adhere to this level of reliability. The result (shown in Figure 5.6) indicates that the F_{crit} value is not exceeded for any of the 26 tasks, however some tasks do appear to show a departure from the null hypothesis (with F values exceeding unity). This suggests therefore that some tasks are less repeatable than others based on the normative data available, although not to an extent that may be considered statistically significant.

To verify this effect, the Euclidean multi-variate metrics for all subjects and replicates were also tested for reliability. The ANOVA test revealed an F-value of 0.39 ($F_{crit} = 3.28$), and p-value of 0.68, thereby indicating that there is no statistically adverse effect in the repeatability of the assessment procedure. As the subjects were undoubtedly affected by external factors (such as the trials being performed at different times of the day, or in some cases, in different locations) then this result can be considered as a valid indication of the procedure's test-retest reliability.

The F-tests also reveal highly significant differences between subjects ($p < 0.001$), however as the main focus was to establish any differences in replicate performance, this result is merely an interesting aside and has no bearing on the reliability of the procedure.

Inter-Rater Reliability

In order to establish consistency between assessors, an experiment was constructed using normative Group B and three different raters. All evaluations were carried out consecutively for each subject and the experiment was performed in one time period to minimise external effects from influencing subject performance. A Latin Square design approach [156] was adopted that also eliminates the order of assessment (or learning) effects.

The rater was found to be statistically insignificant (F=2.65, F_{crit}=3.09, p=0.075) at the 95% level, as was the interaction between the rater and the subject (F=2.12, F_{crit}=2.46, p=0.084), and the interaction between the rater and the task (F=0.75,

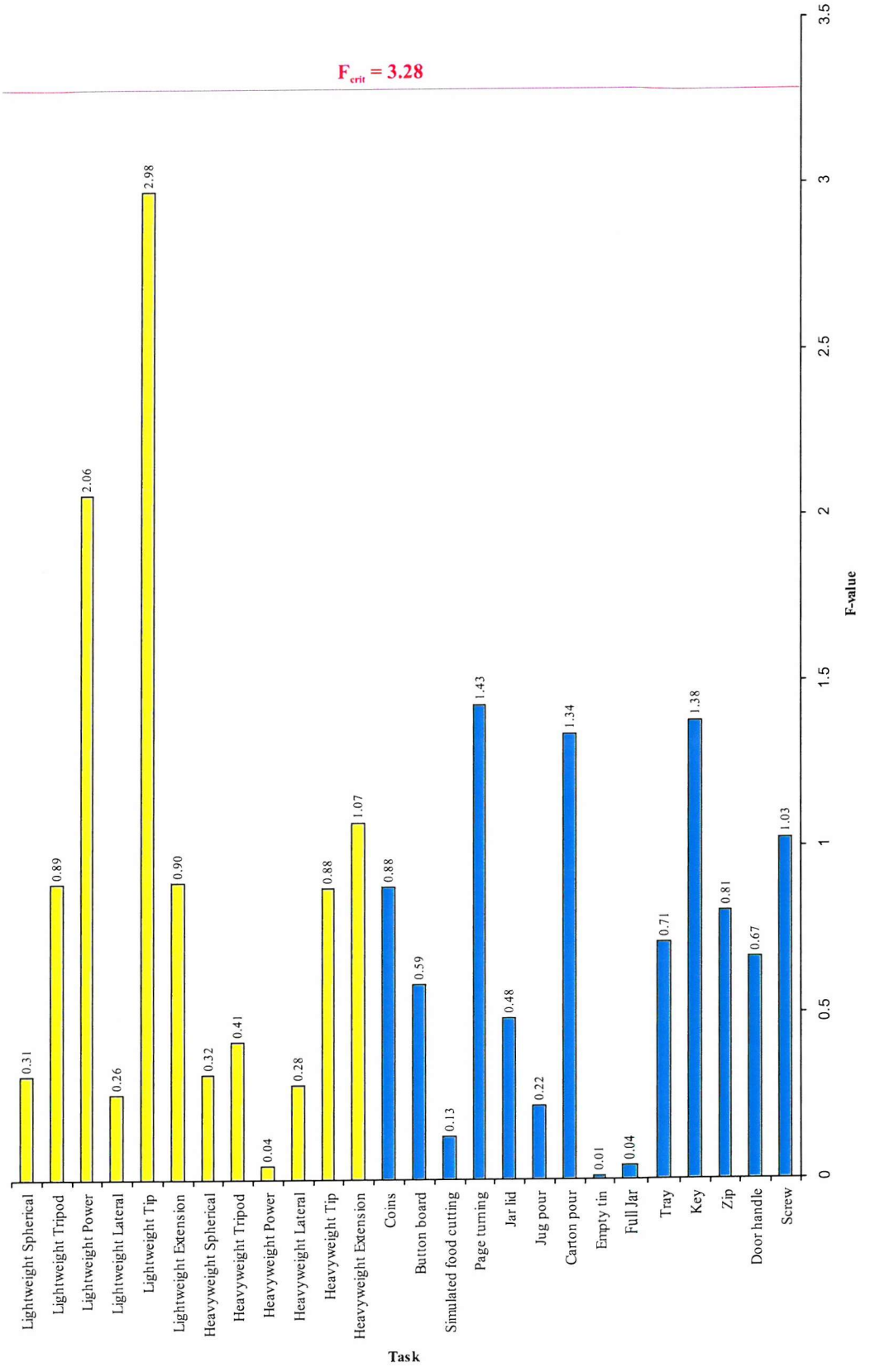


Figure 5.6: ANOVA F-values for Replicate Reliability

F_{crit}=1.48, p=0.87). Consequently the assessor appears to have statistically little effect on either the execution of the assessment procedure, or the subject's performance, thereby indicating inter-rater reliability.

Internal Consistency

This reliability effect is designed to establish consistency between measures. However, in the case of SHAP, these measures are designed to be independent rather than homogeneous assessments of a subject's specific area of disability. Therefore any internal consistency effect is meaningless in this context.

5.3.5 Validity

It was not possible to demonstrate the criterion validity of the hand assessment procedure due to an absence of any existing benchmark technique (see section 4.1.3). The specific objectives of SHAP fail to fall within the remit of other evaluation procedures, as current methods either fail to comprehensively adhere to medical outcome measurement criteria, or evaluate gross upper limb function. If the objectives of criterion standards fail to adhere to that of the procedure under validation, then the result is largely irrelevant. Hence validation of SHAP compared to a clinical 'gold' standard procedure was rejected as unfeasible at this time.

An alternative approach was to attempt to establish consistency between clinicians' subjective ratings of disability, and the SHAP index of functionality scale. However, the validation of an objective ratio scale relative to that of a subjective ordinal scale would be extraneous. As is the case with a number of new clinical outcome measures, it is believed that the traditional criterion validity can not be applied in this instance.

The content validity of the assessment procedure can be illustrated by the methodology of development. The critical review of existing techniques [103] highlighted specific areas of weakness as well as extracting topics considered important by consensus opinion. The results of this study were used to form the Southampton Hand Assessment Procedure, and was subsequently presented to a panel of hand thera-

pists[6]. Recommendations were implemented, and subsequent approval indicates the content validity of the procedure.

5.4 Conclusions

There is little or no conformity to a standardised and objective procedure for the assessment of pathological and prosthetic hand function [103]. Existing procedures frequently fail to adhere to medical outcome measurement design criteria, or are unable to comprehensively cover the prehensile range of the hand.

The Southampton Hand Assessment Procedure (SHAP) has been designed to account for these shortcomings and therefore allow the evaluation of hand function in the clinical setting. The outcome measure is a contextual rating of functionality (relative to that of 'normal' hand function), which enables the clinician to initially determine the subject's disability, and subsequently monitor their performance throughout a course of treatment or rehabilitation.

The procedure has wide-ranging implications for the assessment of hand function ranging from clinical groups (such as burns victims or stroke patients) to the research arena (for example the investigation of impairment in wrist fracture patients). It is also able to quantify, compare and monitor subjects' unilateral functional performance of hand prostheses and controllers.

A normative database (totally twenty-four subjects aged 18-25 years) has been formed as a benchmark of normal hand function. This control group has also been used to indicate the statistical integrity of the hand assessment procedure according to the criteria described in section 4.4. Adherence to these criteria is detailed as follows:

1. The Southampton Hand Assessment Procedure consists of 12 abstract object tasks and 14 activities of daily living, each of which is founded upon one (or

[6]Participants included occupational therapists and physiotherapists from the School of Health Professions and Rehabilitation Sciences at the University of Southampton, and the Wessex Rehabilitation Centre at Salisbury District Hospital, UK.

more) of six prehensile patterns. The expected degree of everyday implementation of these prehensile groups is reflected proportionally in the procedure. *(Criterion 4.1)*

2. The self-timed nature of SHAP eliminates the need for subjective opinion on the part of the assessor. *(Criterion 4.2)*

3. A standardised procedure ensures that assessments are consistent. The data from the control group has been shown statistically to be normally distributed. *(Criterion 4.3)*

4. SHAP has been demonstrated to be reliable by statistically insignificant differences between subjects' performance during replicate assessments, or with various assessors. Internal consistency measures are inappropriate in this instance. *(Criterion 4.4)*

5. Criterion validity cannot be established due to the lack of a benchmark, however the content validity of the procedure is indicated by critical review and expert consensus opinion. *(Criterion 4.5)*

6. The procedure is able to evaluate the functionality of hand prostheses, irrespective of whether they are passive, mechanical or myoelectrically controlled, and is unbiased to the type of terminal device used. *(Criterion 4.6)*

7. SHAP takes approximately 20 minutes to complete, and is a self-contained portable unit ideally suited to use in a clinical environment (see Figure D.4 in Appendix D). *(Criterion 4.7)*

Chapter 6

Clinical Evaluation of Functionality

6.1 Clinical Assessment

The Southampton Hand Assessment Procedure has been used at the Wessex Rehabilitation Centre (Salisbury District Hospital, UK) in the clinical assessment of over thirty five outpatients, and in the assessment of upper limb prosthesis users at the Oxford Orthopaedic Engineering Centre (UK), and the Institute of Biomedical Engineering (New Brunswick, Canada).

Presented below are a series of case studies representing a cross-section of the data collated, including the subject's index (and profiles) of functionality (IOF), and brief clinical notes concerning the reasons for prehensile disability in each case.

6.1.1 Impaired Natural Hand Function

Case Study 1 *24 year old female with right partial hand loss due to traumatic injury. Absence at the proximal IP joints of the third, fourth, and fifth digits, with some injury to the index finger.*

The subject maintains a high level of function (IOF = 93.88), with the prehensile profiles revealing reduced performance during tripod and power grips (see Figure 6.1). The length differential between the index and middle fingers may well account for the difficulty in implementing tri-digit grips. The majority absence of

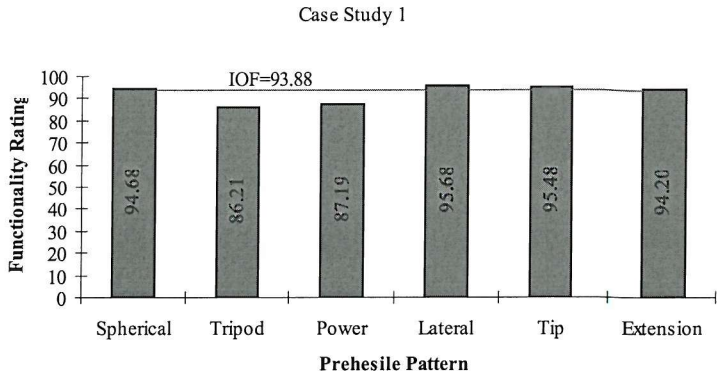


Figure 6.1: SHAP Results for Case Study 1

the fourth and fifth fingers are likely to restrict prehensile strength, which is evident in the impaired performance of the power grip.

Case Study 2 *54 year old male with left partial hand loss due to traumatic injury. Absence of the thumb, and amputation at the proximal IP joints of the third and fourth digits, with additional injuries to the index and little fingers causing reduced range of motion.*

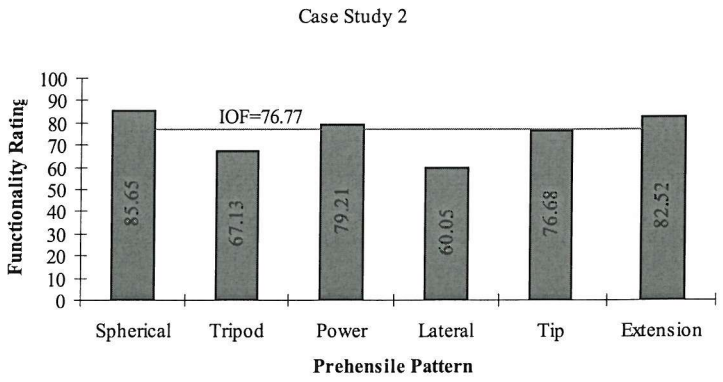


Figure 6.2: SHAP Results for Case Study 2

The subject has notably impaired function (IOF = 76.77), with pronounced disability in the performance of tripod and lateral grips (see Figure 6.2). As illustrated

in the previous case, the discrepancy in finger and thumb length affects tri-digit prehension. The thumb absence may also cause the impeded lateral grip function due to the subject experiencing significant difficulty in opposing the lateral aspect of the index finger.

Although these subjects possess similar pathological hand function, clearly they do not exhibit equivalent disabilities, either overall, or within the prehensile classes. This demonstrates the efficacy of a procedure that is capable of providing additional information concerning the specific nature of the disability. It should be noted that the second subject falls outside the bounds of the normative age group, and thereby illustrates the need for an age-matched database in order to determine expected functional levels.

Case Study 3 *52 year old female with left hand extensor tendon graft to middle, ring and little fingers.*

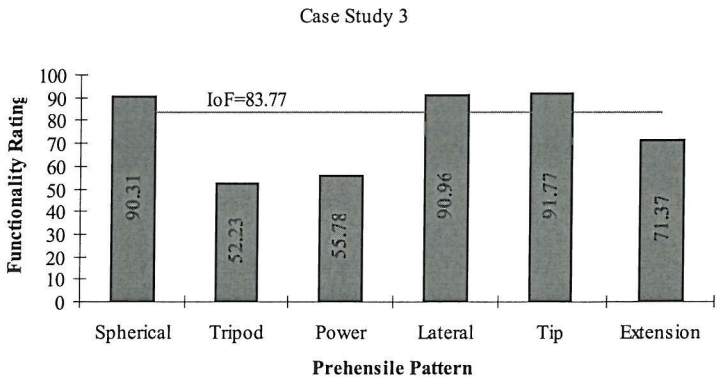


Figure 6.3: SHAP Results for Case Study 3

The subject has diminished function (IOF = 83.77) with particular impediment in tripod, power and extension grips (see Figure 6.3). The limited function in the fourth and fifth digits is likely to reduce strength, and thereby adversely affect a power grip. The extensor grafts are expected to cause restricted stability and range of motion in the wrist, which is a necessary component of most tripod prehensile

tasks. These weakened tendons are also a probable cause of a functional deficit during extension grips that warrant mainly extensor action in order to maintain stable prehension.

Case Study 4 *36 year old female with a fractured left wrist.*

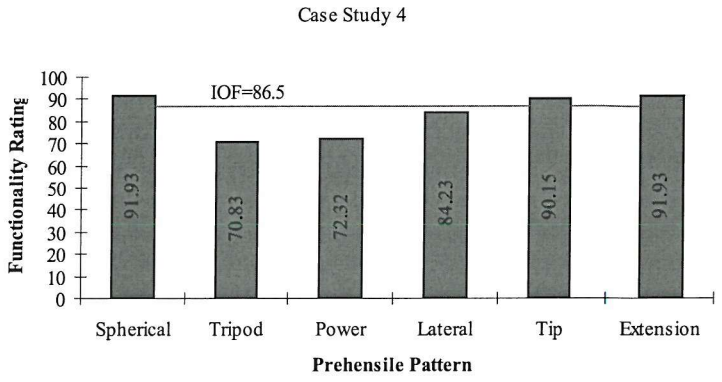


Figure 6.4: SHAP Results for Case Study 4

The subject has reduced function (IOF = 86.5) and is notably impaired whilst executing tripod and power grips (see Figure 6.4). In this instance the profiles of prehensile function fail to reveal an obvious cause of specific disability, hence a closer study of individual task performance is warranted. This highlighted the subject's significant difficulty in pronating or supinating the forearm (due to the fracture). In addition, the application of force during flexion of the wrist also appears to impair function (as seen with some ADLs).

Other subjects with fractures of the wrist also displayed reduced function, although possessed differences in performance of the prehensile patterns. Hence, it is possible to analyse the individual aspects of the assessment to obtain a more detailed understanding of a subject's disability.

During clinical trials, seven subjects were assessed near the beginning of treatment, and again at the end of the rehabilitation process. All showed a marked improvement in their index of functionality, and more specifically, in the prehensile

pattern most affected by the original injury.

6.1.2 Prosthesis Function

Given the small sample size of prosthesis users, it is not possible to draw conclusions concerning the functionality of prostheses or controllers. Instead, the assessment provides an indication of the prehensile difficulties that a subject encounters, as well as providing an overall metric of their disability.

Case Study 5 *16 year old male with right trans-radial amputation at the wrist (with a brachial plexus injury on the same side). The subject is fitted with an Otto Bock hand prosthesis in conjunction with a two muscle myo-control system, and passive pro/supination at the wrist*[1]*. First fitted with a passive prosthesis for a period of six months, before using the current device for three days intensive training prior to assessment.*

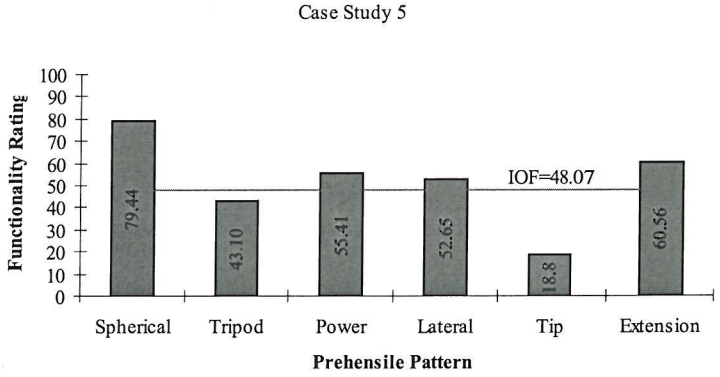


Figure 6.5: SHAP Results for Case Study 5

The subject has notably restricted function (IOF = 48.07), and is severely impaired in the performance of tripod and tip grips (see Figure 6.5). This suggests a difficulty in precision manipulations, which is not displayed in the other prehensile groups, and is commonly seen in the use of commercial prostheses due to the single degree of freedom format.

[1] Additional pro/supination is afforded to the subject due to the socket style.

Case Study 6 *18 year old female with a congenital right transverse carpal partial deficiency. The subject is fitted with a Centri UltraLite hand prosthesis in conjunction with a two muscle myo-control system, and passive pro/supination at the wrist*[2]*. The subject has worn a variety of prostheses in the past, but stopped use for five years preceding the current fitment.*

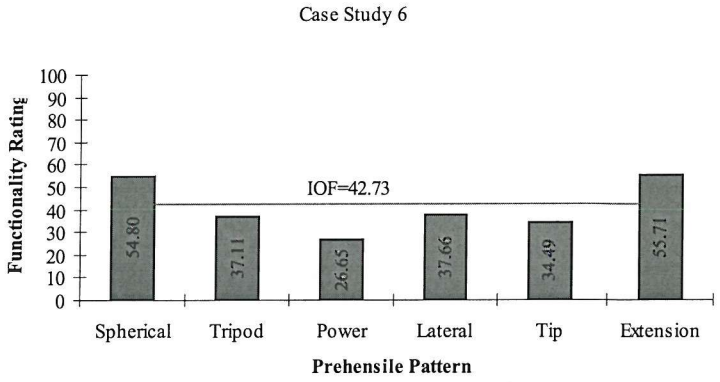


Figure 6.6: SHAP Results for Case Study 6

The subject has severely impaired function (IOF = 42.73), with a greater ability to effect spherical and extension grips than other prehensile groups (see Figure 6.6). The potential reason for improved function in these grip structures is the indiscriminate nature of grasping that is required, whereas the other tasks warrant an exacting grip in order to ensure stable prehension.

Case Study 7 *17 year old female with congenital trans-radial absence. The subject is fitted with an Otto Bock and active wrist rotator in conjunction with a two muscle myo-control system.*

The subject demonstrates highly impaired function (IOF = 37.02), notably in the areas of tripod and power prehension (see Figure 6.7). The improved functional ability during lateral tasks may be due to the active wrist component of the prosthesis that facilitates pro/supination.

[2]Some limited natural pro/supination remains due to a suction style socket fit.

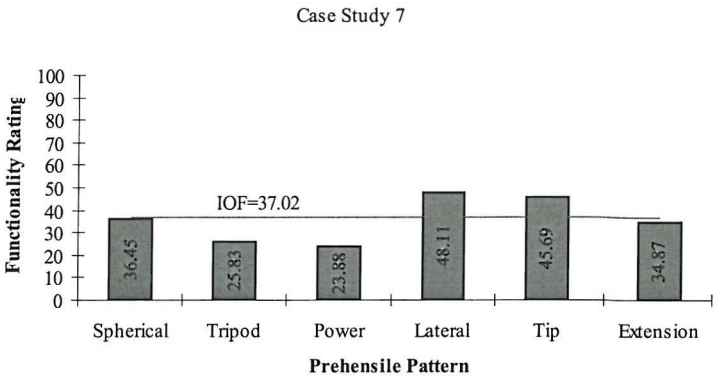


Figure 6.7: SHAP Results for Case Study 7

The data show that it is possible to determine a subject's prehensile difficulty as well as an overall metric of hand disability. In addition, the expected use of grip patterns during everyday living further affords a greater understanding of the subject's handicap. This is of particular importance during legal disability claims, as well as forming an integral monitoring aspect during rehabilitation.

6.2 Enhanced Assessment

6.2.1 Introduction

The assessment of impaired hand function has been addressed by the development of the Southampton Hand Assessment Procedure. The main flaw of this and numerous other assessment tools are their sole reliance upon a direct and valid correlation between timed performance and functionality.

Temporal measures are also used in gait analysis, which has seen extensive development and increasing use over the last two decades. However the effectiveness of this tool has been achieved through the use of motion analysis systems that enable clinicians to monitor the kinematics of pathological gait, and thereby diagnose specific joint or muscle dysfunction. Statistical techniques such as Fourier series and Bootstrapping [157] are used to establish the level of deviation from the norm in the individual under assessment.

Fourier series are matched to kinematic data to act as a low pass filter to the characteristically noisy signal. However there are further additional benefits to this analytical method: all subjects are described by the same number of coefficients irrespective of the gait cycle duration; joint angles at any point in the cycle can be readily interpolated; and calculation of velocity and acceleration profiles are easier to achieve and more accurate with Fourier series than by numerical analysis.

Deviation from normative gait data enables identification of neuromuscular, postural or limb abnormalities in a subject. Any statistical method that is able to identify abnormal kinematic patterns is therefore a useful clinical tool. Bootstrapping has seen application to gait analysis to aid in the performance of this function by boosting the power of a sample (and therefore decreasing the number of subjects required) by a 'sample and replace' method [157]. More importantly, the Bootstrap enables the derivation of a sample mean with a given bandwidth that defines 'normal' kinematic patterns at each point in the gait cycle.

It is therefore feasible that the kinematics of the upper limb can be captured and analysed in a similar manner. If a characteristic set of joint trajectories exists for a range of prehensile activities, then impaired hand and arm function can be quantified relative to a set of normative data.

Hence, upper limb motion analysis has the potential to provide an enhanced form of functionality assessment, as well as highlighting key aspects of the hand's kinematics that may be used in prosthesis design. For example, if wrist movement can be correlated with prehensile patterns, then a multiple degree of freedom prosthesis may incorporate a similar biomechanical link within an intelligent control system.

Prehensile patterns are unique identifiers of a subject's hand position during grip manipulations and thereby infer knowledge of individual digit configuration. This has implications for biomechanical studies of reaching and grasping, as well as improving the accuracy of functional assessment. The additional advantage of this technique is its application to prosthesis design, by enabling a study of useful movements in active wrists. However, few clinical motion analysis systems pos-

sess the necessary resolution to quantify individual finger position and trajectories. Moreover, it would prove difficult to produce a cogent taxonomy from any results in terms of the subject's functional ability due to the high number of degrees of freedom. Therefore it is unlikely that any upper limb kinematic analysis will be able to provide detail beyond that of the wrist, elbow and shoulder function at the present time. Nevertheless, such an assessment tool would represent a significant advance over existing methods.

6.2.2 Upper Limb Motion Analysis

Consequently the objective of the pilot study was to capture the kinematics of the upper limb using motion analysis equipment. The abstract object tasks of the Southampton Hand Assessment Procedure were used to provide known prehensile pattern activities. The following key joint trajectory movements were extracted from the analysis:

- Flexion/extension of the elbow
- Pronation/supination of the forearm
- Flexion/extension of the wrist
- Radial/ulnar deviation of the wrist

Thirty two anatomically positioned markers (see Figure 6.8) were placed on a subject of normative hand function, although only seven are necessary to describe the kinematics of the elbow and wrist[3]. The Vicon motion analysis system was used at the Oxford Orthopaedic Engineering Centre to capture the three-dimensional co-ordinates of each marker.

The markers used to indicate the position of the limb segments are denoted as: F2MCP (middle finger metacarpophalangeal joint), WRI (wrist, ulnar side), WRLT

[3]Thirty two markers were used in an attempt to determine whether the full upper limb kinematics could be captured in a single trial. The complexity of digit movement during prehensile activity appeared to obscure some markers from view (not pertaining to the main experiment). Hence it was deemed unlikely that a full joint analysis could occur with the existing 7-camera system.

(wrist lateral, radial side), ELB (elbow), LHUM, MHUM, and UHUM (lower, middle and upper humerus respectively). Three frames of reference can be constructed from these marker positions to define the orientation of the hand/wrist, forarm/elbow, and upper arm limb segments (see Figure 6.8).

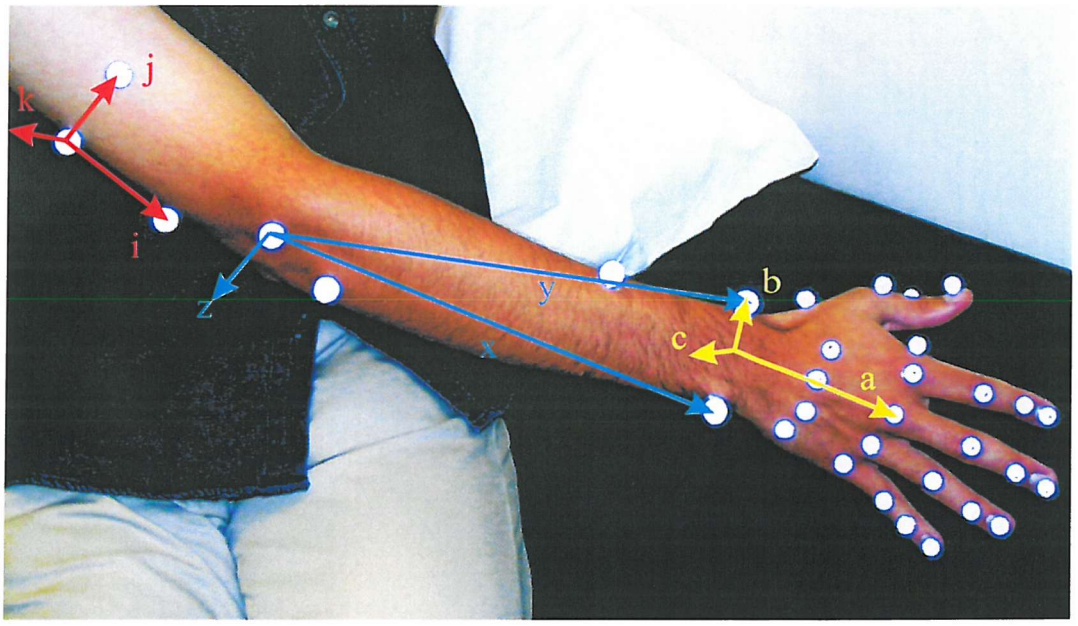


Figure 6.8: Anatomical Marker Set

Vectors describing the plane of each limb segment are created using the three-dimensional marker positions. The product of these vectors are used to generate each frame of reference, which is ultimately comprised of three orthogonal unit vectors. For example, the ELB, WRI, and WRLT markers produce a frame centred at the elbow (according to Equations 6.1-6.3). Similarly, unit vector frames are created at the humerus (using MHUM, LHUM, and UHUM markers), and at the centre of the wrist (using WRI, WRLT and F2MCP markers).

$$\vec{x} = \begin{pmatrix} WRI_1 \\ WRI_2 \\ WRI_3 \end{pmatrix} - \begin{pmatrix} ELB_1 \\ ELB_2 \\ ELB_3 \end{pmatrix} \tag{6.1}$$

$$\vec{z} = \vec{x} \times \left[\begin{pmatrix} WRLT_1 \\ WRLT_2 \\ WRLT_3 \end{pmatrix} - \begin{pmatrix} ELB_1 \\ ELB_2 \\ ELB_3 \end{pmatrix} \right] \tag{6.2}$$

$$\vec{y} = \vec{z} \times \vec{x} \tag{6.3}$$

The ability to determine limb segment trajectories relies on the extraction of the angular displacement between the frames. In order for these angles to provide standardised and interpretable data, an anatomical reference frame must be employed. Consequently the kinematics of the wrist are studied with respect to the elbow frame, which similarly is analysed with respect to the humeral frame.

Angular displacement can be studied by extracting the relative angle between vectors of two frames. For example, the angle α between $\vec{a}$ and $\vec{x}$ (or $\vec{a}$ and $\vec{z}$), may be used to indicate wrist flexion/extension (according to Equations 6.4 and 6.5).

$$area = \vec{a} \times \vec{x} \tag{6.4}$$

$$\alpha = sin^{-1}\left(\frac{|area|}{|\vec{a}|.|\vec{x}|}\right) \tag{6.5}$$

The disadvantage of this technique is that it fails to delineate between the directions of limb motion about the anatomical reference axis (due to the absolute nature of Equation 6.5). Hence it is not possible to distinguish between flexion and extension (or other antagonistic muscle movements) without a knowledge of the initial frame conditions (e.g. that the wrist is flexed at time zero). This can be seen in Figure 6.9 during a cyclic motion of wrist flexion and extension. From a visual interpretation of original data, the wrist commences extension at time t=15, passes through the 'neutral plane' (where target and reference vectors align) at time t=20, and does not commence flexion until t=72. Although a set of initial conditions may be devised to ensure that subjects commence the procedure in a known anatomical position, such criteria may not be a feasible objective in cases of pathological hand function.

As the elbow and wrist joints do not possess three degrees of freedom, there is clearly an redundancy within this form of analysis. This duplicate measurement (as can be seen by α_{ax} and α_{cz} measuring angular displacement about the same joint axis) is advantageous as it provides a means of verifying the correlation of the model with the anatomical joint axis.

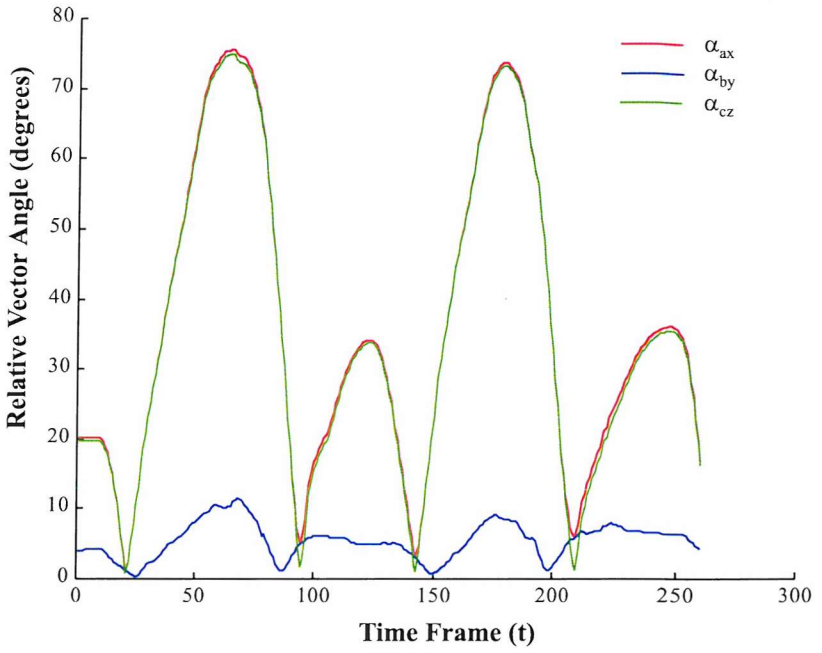


Figure 6.9: Relative Vector Angles during Wrist Flexion/Extension

An alternative method of calculating angular displacement involves the analysis of each axis of the target frame with respect to the three planes (or axes of rotation) of the reference frame. This can be achieved by the sequential mapping of each vector to the reference axes by the use of direction cosines. In the example shown in Figure 6.10, the vector $\vec{a}$ is mapped to the x-y-z frame using the direction cosines l_a, m_a, n_a. The result is three angles, $\phi_{yz}, \phi_{xz}, \phi_{xy}$, of vector $\vec{a}$ in the y-z, x-z, and x-y planes respectively. As seen in the previous method, a duplicate measurement of angular displacement provides a more robust model.

This mapping is achieved by transforming both the target and reference frames to the orientation of the Vicon global coordinate system. The frame at the origin of the global workspace, which is comprised of three unit vectors, can be represented by the Identity matrix. Therefore to transform the anatomical reference frame (denoted as R in Equation 6.6) to align with the Vicon workspace merely involves multiplication by R^{-1}. This same transformation matrix can be applied to the target

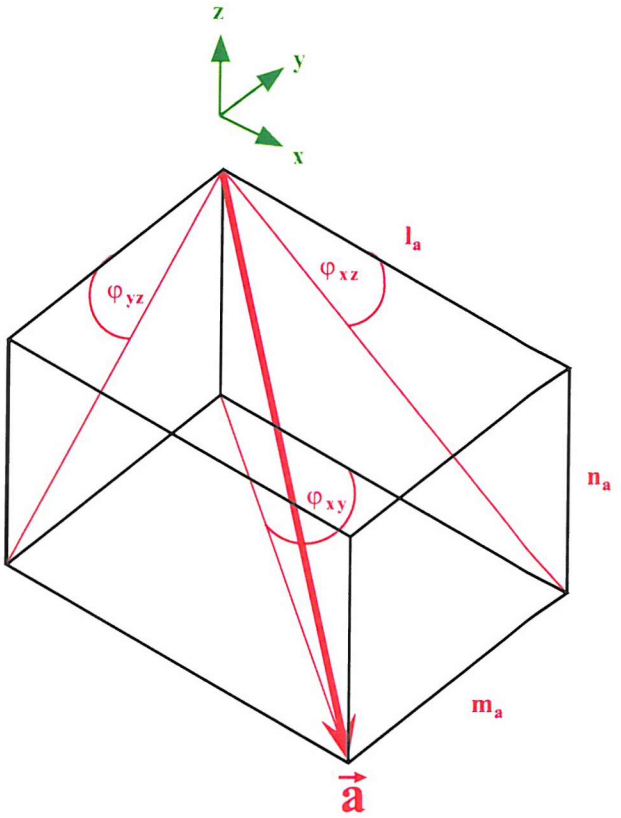


Figure 6.10: Vector Mapping

frame (denoted as T in Equation 6.6 at each sample instance t (with a sampling frequency of 50Hz) to enable the simple calculation of the direction cosines and angles (see Equation 6.7). Due to the aforementioned redundancy inherent in this analysis, only a select number of angles are necessary for description of the segment frame kinematics.

Transformation of the vectors into the Vicon workspace can cause a sign change in the angle (as it changes quadrant), which results in erroneous trajectory plots. Consequently, if the velocity change in the limb segment is in excess of 5000 de-

grees/second[4] then 180° compensation is added to or subtracted from the angle.

$$R(t) = \begin{bmatrix} x_1 & y_1 & z_1 \\ x_2 & y_2 & z_2 \\ x_3 & y_3 & z_3 \end{bmatrix} \qquad T(t) = \begin{bmatrix} a_1 & b_1 & c_1 \\ a_2 & b_2 & c_2 \\ a_3 & b_3 & c_3 \end{bmatrix} \tag{6.6}$$

$$\textit{Direction Cosines} = \begin{bmatrix} l_a & l_b & l_c \\ m_a & m_b & m_c \\ n_a & n_b & n_c \end{bmatrix} = R^{-1}.T \tag{6.7}$$

A Matlab program was written to calculate and plot all pertinent angles. A graphical user interface enables simple selection of trials and plot types prior to initiating the analytical routine detailed in Figure 6.11.

In order to identify the optimum selection of angles, a series of 'pure movements' was carried out by the subject. These trials included cyclic motions of wrist flexion and extension, radial and ulnar deviation of the hand, and pronation and supination of the forearm. SHAP abstract tasks were used for subsequent trials to examine the feasibility of using upper limb motion analysis as a clinical assessment tool.

Visual motion analysis systems are susceptible to missing data when a limb segment obscures the marker from view. For any data loss of less than ten samples (0.2 seconds), the position of the marker was estimated using a cubic spline interpolation routine. Any trials containing a greater period of data loss were disregarded.

6.2.3 Results

As can be seen in Figure 6.12[5], the vector mapping technique can be used to describe the kinematics of the arm. In this case, the trial indicates an approximate 100° cycle of forearm pro/supination. The additional deviations seen in the elbow flexion/extension trace cannot clearly be distinguished from the original motion capture imagery. Further study is required of whether these may be attributable to a skin artefact[6] or may occur naturally.

[4]This represents a realistic boundary condition for human joint velocity. Some baseball athletes in the USA have been recorded with upper limb speeds in excess of 2000 degrees/sec whilst pitching.

[5]The key indicates the anatomical movement associated with a change in *slope* direction of each plot.

[6]Marker placements are subject to variation and inaccuracy due to the movement of the muscular and skeletal structure without corresponding movement at the skin [158].

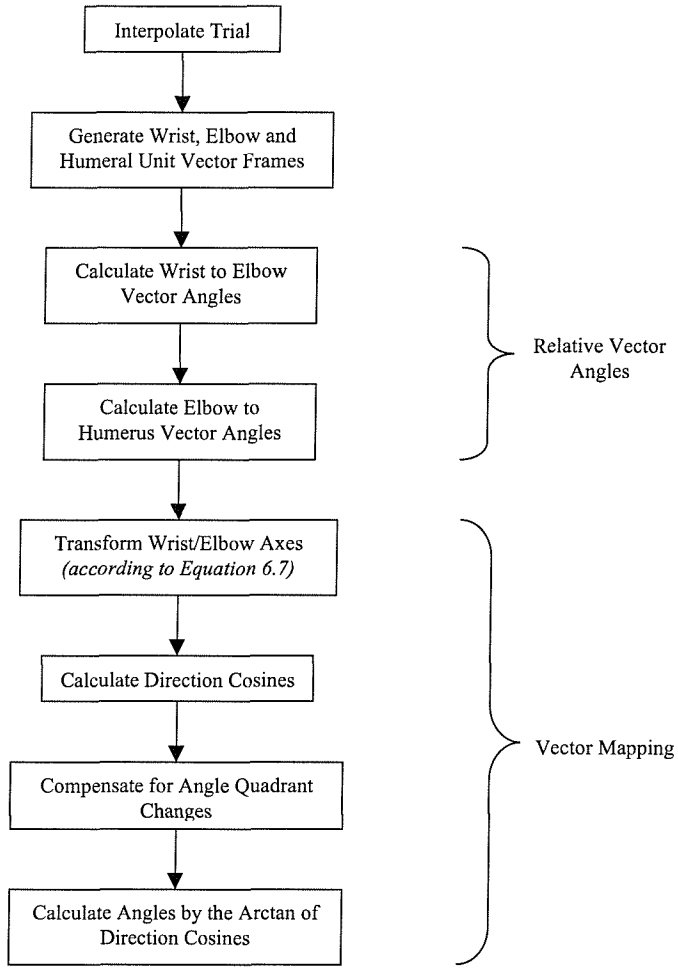


Figure 6.11: Flow Chart of Upper Limb Motion Analysis Routine

In order to interpret vector mapping plots, it is necessary to define the anatomical location of each of the neutral axes (i.e. the position at which target vectors produce a zero angle with respect to the anatomical reference vectors). The wrist flex/extend position lies with the hand slightly flexed so that the vector $\vec{a}$ from the centre of the wrist to the middle finger MCP joint is parallel with vector $\vec{x}$ when viewed sagittally. The radial/ulnar deviation neutral axis lies with a slight radial deviation of the hand so that vectors $\vec{a}$ and $\vec{x}$ are parallel when viewed frontally. The neutral axes of the forearm and elbow are at full pronation and extension respectively.

Figure 6.13 shows the kinematic plot of the SHAP abstract power task (where

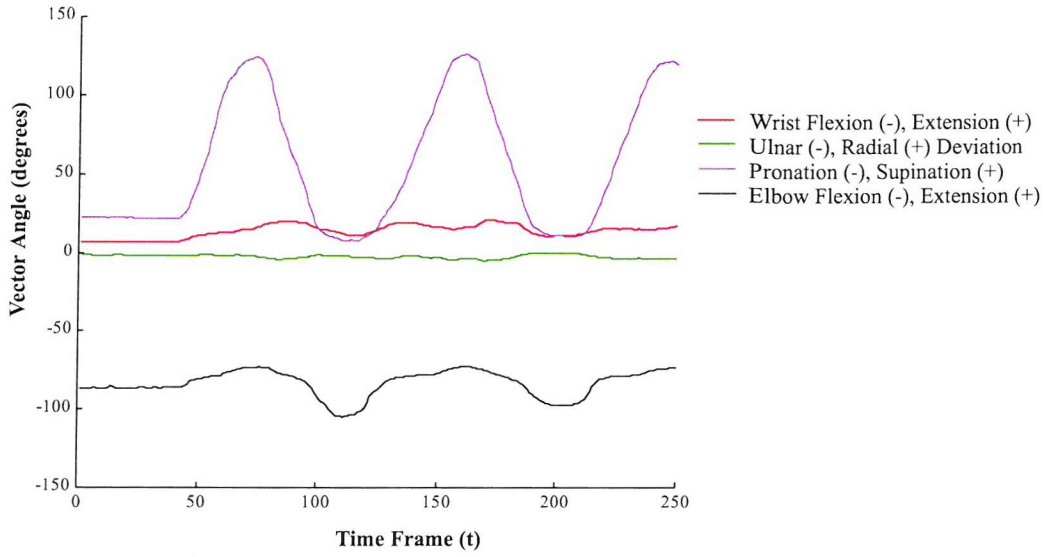


Figure 6.12: Vector Mapping Technique Describing Pronation/Supination of the Forearm

a cylindrical object is moved within the form board – see section 5.2.1). It is clear that a complex range of motions is necessary to carry out a relatively simple activity. There are key events within the plot that warrant explanation, and have been noted from the original motion capture.

- Time Frames t=60-90 – the hand leaves the table and moves to strike the timer button (signifying the start of the SHAP task). At this point the wrist has flexed by 10° and moves into a 40° extended position.
- Time Frames t=90-130 – the hand strikes the timer button. To reach this point, the wrist has flexed from an extended position. The wrist is then extended again whilst the forearm supinates to an approximately 90° position to align with the cylindrical abstract object. To ensure the hand is perpendicular to the shape, the hand is radially deviated and the elbow extended as the subject reaches for the object.
- Time Frames t=130-150 – the object is transferred from the rear slot in the SHAP abstract board to the front slot. This accounts for the increase in elbow

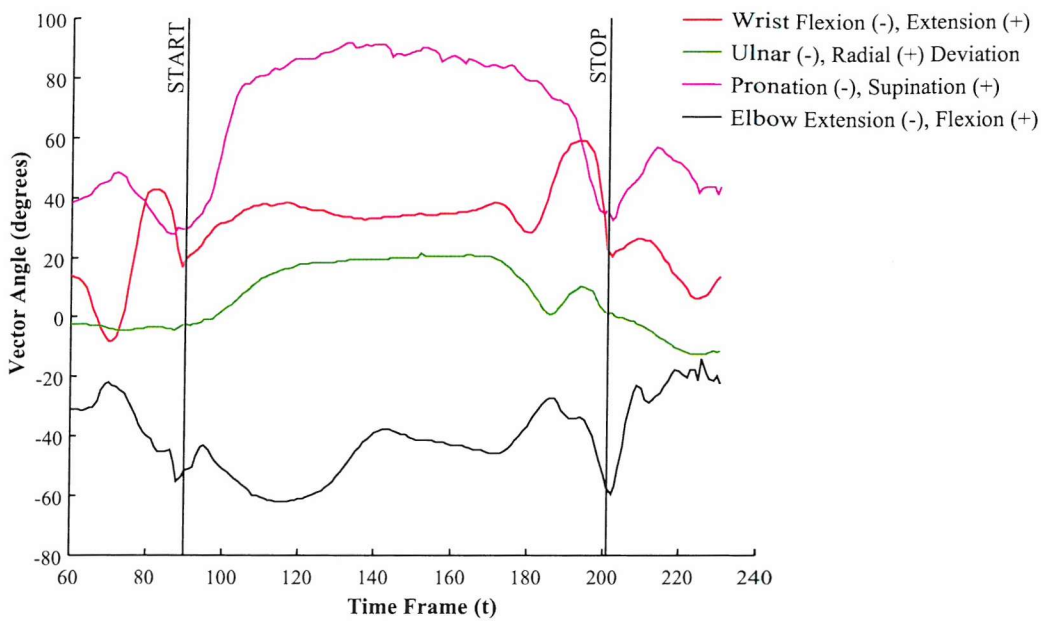


Figure 6.13: Vector Mapping of SHAP Abstract Power Task

flexion.

- Time Frames t=150-201 – the hand retracts and supinates as the subject tracks towards the timer.

- Time Frames t≥201 – The timer is struck to indicate the end of the task. As expected, at the key indices of starting and stopping the timer, the angles of each joint trajectory are approximately the same.

This demonstrates that it is possible to monitor upper limb kinematics during predefined tasks and thereby suggests the potential for subsequent clinical analysis. The joint trajectory plots of the other SHAP tasks reveal unique movements in each activity, further suggesting application as part of the hand assessment procedure. However, these preliminary results must be confirmed by additional studies, and a subsequent normative database constructed in accordance with the clinical outcome measurement criteria discussed in section 4.4.

6.2.4 Conclusion

Motion analysis systems are frequently used to identify abnormalities in pathological gait by description of the dynamics of the trunk and lower limbs. Temporal and spatial parameters (such as swing phase and cadence) can be used in conjunction with plots of joint displacements to provide an assessment of stability. In order to analyse the data, Fourier series are often fitted to the trajectory plots and then compared statistically to a normative database to assess the level of abnormality.

The development of the Southampton Hand Assessment Procedure has addressed many of the issues necessary to provide a comprehensive evaluation of hand function. However, the ability to describe the kinematics of the upper limb during the evaluation would greatly enhance the effectiveness of the procedure (albeit at the expense of having to conduct evaluations within the bounds of a clinical gait lab). Clinical assessment techniques of pathological hand function are crude by comparison to that of gait analysis. This pilot study of upper limb motion analysis suggests that the principles of assessment, although more complex than that of the lower limb, may be applied in a similar manner. The periodic nature of gait enables a more natural method of pattern identification, however the use of specific prehensile tasks shows that pathological hand function may be identified using a similar statistical analysis.

Chapter 7

Conclusions and Recommendations

Current commercial hand prostheses possess a single axis pincer movement that affords the user little functionality for the wide range of prehensile tasks that are part of everyday living. Myoelectric control of these devices enables the user to vary the hand's grip strength by voluntary muscle contractions, however visual feedback must be maintained with the object to ensure stable prehension. Thus, the existing solution affords minimal function whilst warranting continuous and conscious effort on the part of the user.

Hence there exists a key requirement for the development of a lightweight, multiple degree of freedom hand prosthesis, without increasing the current physical or psychological burden on the user. The Southampton-Remedi hand and the SAMS-UNB intelligent multifunction controller address this demand. The efficacy of design was to be established by an evaluation of the hand's functionality, however there is a notable absence of standardised and objective assessment techniques in this area. Hence the Southampton Hand Assessment Procedure has been developed to provide a evaluation of both pathological and prosthetic hand function, with implications that may redirect the focus of clinical upper limb assessment as it exists today.

7.1 Mechanical Design

The lightweight and adaptable six axis Southampton-Remedi hand possesses four independent digits, and a two-axis thumb. It has been designed to adhere to critical prosthesis design constraints such as anthropomorphism, low weight, low power consumption, and a high level of modularity.

Modelling of the finger linkage has assured the static and dynamic anthropomorphic movement of the hand, which weighs 407g in total and is sized to match that of an average adult hand by fitting within a $7\frac{3}{4}$" prosthetic glove. Full digit curl occurs within less than one second, and the hierarchical control scheme ensures that minimal power consumption and optimal grip pressure is maintained during prehension, thereby providing an estimated 2400 grip cycles per battery charge. The prosthesis is fully modular in design – each digit is a self-contained unit that may be removed individually for repair, and therefore also is applicable to both left- and right-handed prostheses. The multiple independent digits and mobile thumb afford the hand a prehensile adaptability and security that is far in excess of commercial devices.

However reliability is equally as important as functionality. User trials are prohibited at this stage due to the development format of the prototype prosthesis (and controller). Hence there is a clear requirement to identify potential areas of weakness and primary failure points in the hand prior to clinical evaluation.

Carbon fibre and lightweight polymer plastics are used in the current design to ensure minimal weight, however other material combinations may produce a prosthesis of better strength and reliability without adversely affecting overall mass. For example, a forked design in the base links of the digit, and manufacture from magnesium alloy[1], may be used to increase lateral stability. Consequently the optimum solution may reside in the use of lightweight alloys for areas of high stress, composite materials for predominantly unilateral loading and structural rigidity, and polymer

[1]The planar linkage design is identifiably weak at the base under lateral loading, however carbon fibre cannot easily be formed or machined into a forked link without creating high stress fractures at the corners.

composites for low friction bearing surfaces.

The functionality of the prosthesis is sourced from the mobility of the digits, however the palm remains a rigid and unyielding platform for the majority of tasks. A more stable grasp may be achieved by modelling the palm's curvature based on the natural hand. This technique requires three-dimensional analysis and interference simulation of the digits to ensure that an optimum configuration may be achieved. Instead of this complex solution, it may simply be sufficient to increase the palm's compliance through the use of mechanical or material means. For example, the use of damped hinge joints along the palmar arch axes (discussed in section 2.5) may create a passive but adaptive grip.

However the issue of mechanical compliance cannot be reserved specifically for the palm. The prosthesis as a whole must exhibit forms of adaptation to a grasped object in order to maintain a secure grip. Hence materials worthy of investigation are those capable of creating fingertip pulps and compliant hand bulk, without compromising the operational efficacy of the device.

Ultimately the functionality of the multi-axis prosthesis is governed by the intelligence of the control system, as the additional mechanical adaptability of the device cannot be achieved at the expense of imposing a significant psychological burden upon the user. This issue has been addressed by the SAMS-UNB control system.

7.2 Sensor and Control Systems

The UNB system has established multiple degree of freedom myo-control of an upper limb prosthesis. Although application of this controller to multi-function artificial hands is obvious, it does not overcome the cognitive burden of maintaining stable prehension through visual feedback alone. The Southampton Adaptive Manipulation Scheme addresses this shortfall, but warrants external trigger inputs to initiate various grip patterns. Hence to improve the user's natural and fluid movement of the prosthesis during grasping, and yet maintain intelligent and adaptive prehen-

sion, the hybrid SAMS-UNB controller has been designed and implemented on a DSP microprocessor. Its use with the multiple degree of freedom Southampton-Remedi Hand illustrates the benefit of automated prehensile control (both in shape and strength) with inputs solely from the user's myo-signal.

This form of intelligent control is achieved through the use of accurate and reliable sensor systems, which form a mechanically integral part of the new prosthesis. This has provided the unique ability to enhance the low level control of the device over previous generations of the Southampton Hand. Digital magnetic encoders mounted to each motor-drive provide an accuracy of 0.03° of digit rotation, whilst force information is derived from current sensors integral to the H-bridge power electronics. External slip sensors are used to provide the microprocessor with information on objects sliding from grasp. The analogue current sensors displayed a notable susceptibility to noise generated by the drives, thereby causing the controller initialisation program to erroneously reset independent digits. Ultimately this effect precluded the successful operation of the device, however this may be overcome by employing more effective production methods such as a printed circuit board format with adequate shielding and noise suppression. Nevertheless, the inclusion of these systems indicates the efficacy of integral and accurate sensors to aid intelligent prehensile control.

Despite the complex feedback systems present in the new hand and controller, further sensor development represents the tangible opportunity for improving the functionality of the prosthesis. The existing motor-current sensors potentially enable environmentally adaptive performance by the use of thermal modelling. When used in conjunction with a fingertip sensor capable of detecting force and slip, the controller may provide self-regulatory operation of the prosthesis by compensating for ambient temperature differences as well as for the long term changes in the mechanics of the device. As the linkage joints and motor drive systems become worn, compensation in the control outputs can be made in order to maintain consistent grip forces.

A sensor array such as this also highlights the potential for compliance control. Regulating the feedback of hand opening and contact force has been explored for use in functional neuromuscular stimulation [159]. The system is designed to allow both position and force to be controlled when grasping compliant objects. Similar algorithms and feedback loops have been used in the stiffness and contact force transient control of robotic devices [160, 161], and could be applied to the prosthesis (given sufficient processor power) to better regulate a compliant yet stable grip. Adaptive control also could be used to compensate for the non-linearities (such as the backlash and variable joint friction) of the linkage and drive system.

The implementation of these recommendations remains governed by the limitation of the processing power currently available. It is clear that the logical progression for the SAMS-UNB controller is integration to a single DSP. This would eliminate much of the existing hardware and additional communication times, thereby freeing some of the processor overhead as well as reducing the overall power consumption of the electronic systems.

The initial premise and concepts of the hand have been established, and the hybrid controller provides the unique potential for direct and intelligent prehensile control of an adaptable prosthesis from a user's myo-signal. However additional development is required to reach an appropriate platform for functional evaluation.

7.3 Southampton Hand Assessment Procedure

Although the assessment of functionality forms a critical component in determining the efficacy of the device, existing natural and prosthetic hand functionality assessments fail to provide a useful evaluation. These procedures often are unreliable, subjective, lacking in statistical evidence or more generally, fail to achieve the standards necessary for medical outcome measurement. The development of SHAP addresses each of these shortfalls in order to evaluate the efficacy of the multi-axis prosthesis and controller. A normative database has been formed, and a metric of functionality formed to provide clinicians with a method of quantifying an indi-

vidual's hand function with respect to a benchmark. The reliability and validity of the procedure have been demonstrated in accordance with medical audit requirements. Clinical trials have been undertaken, and continue to provide a range of data concerning impairment and disability. It is suggested that age-matched normative groups should be established to provide the clinician with a more contextual result than currently exists.

The scope of the Southampton Hand Assessment Procedure is more extensive than the assessment of the new prosthesis and controller. SHAP allows the evaluation of patient groups ranging from stroke victims and burns patients, to those using functional electrical stimulation systems, hand prostheses or rehabilitation robots.

Nevertheless, the lack of an automated database tool remains a barrier to the widespread clinical use of the hand assessment procedure. Although the timed unit provides data that is easily collated, the calculation of the index and profiles of functionality are outside the bounds of a clinician's working expectations. Consequently further development of the relational database systems and graphical user interface are necessary to enable the automated calculation of these measures to provide an instantaneous result.

The disadvantage of a stand-alone database is that the patient data remains localized to the institution, whereas the widespread collection of functional measures would enable clinicians to compare and contrast similar patient cohort groups and treatment methods. The Internet is a primary facilitator to achieving this goal. A single database (on a secure website), or individual on-site databases that download to the main server on a daily basis, are potential solutions to maintaining an effective global clinical assessment tool. Although issues such as patient confidentiality must be addressed, the benefits are clearly demonstrable. The information collated from various centres using the standardised SHAP procedure provides large scope for extensive meta-analysis of a range of diseases, dysfunction and treatments.

7.4 Upper Limb Motion Analysis

The assumption that timed tasks produce an accurate representation of functional performance is the main limitation of SHAP and other timed assessment techniques. Temporal measures are used in gait analysis, but only as part of an overall kinematic and kinetic evaluation. Based on the Southampton Hand Assessment Procedure, a pilot study has demonstrated the feasibility of carrying out similar assessments on the upper limb.

Whilst performing periodic prehensile tasks, unique trajectories of upper limb segments may be extracted from motion analysis data. Further trials are required to validate preliminary results, and refining the representation of limb segments is necessary to ensure that the biomechanical model accurately reflects that of the natural limb. However, the study suggests that motion analysis may allow enhanced clinical evaluation of the upper limb. The foundation of this technique on a standardised and effective tool such as SHAP lends notable credence to its future development, which should be to follow that of gait analysis. The use of Fourier series and statistical techniques such as Bootstrapping provide a clinically effective means of quantifying abnormality from the normal population, and would therefore also aid diagnosis in a wide variety upper limb cases.

7.5 Conclusion

It is clear that in order to afford the user greater function, hand prostheses must possess more than a single degree of freedom. The disadvantage that accompanies the development of multi-axis devices is the increased cognitive burden necessary to ensure accurate control of grip strength through visual feedback. Consequently the six axis Southampton-Remedi hand and the hybrid SAMS-UNB controller have been designed to produce a prosthesis capable of adaptive and intelligent prehension by direct myo-control, without the need for continuous visual monitoring of the object under grasp.

The system is effective in both concept and implementation, and demonstrates the integration of a number of technologies to produce an client-focused device with the caveat of being a prototype hand that has not been evaluated by a patient. Although unlikely to see commercial development in this form due to the current market attitude, it clearly highlights the vast disparity between current practice and feasible engineering solutions. Refinement of the prosthesis and controller design may lead to clinical trials using the standardised and objective SHAP procedure that should emphasize this functional contrast in a more formal and public manner. Indeed there is a clear opportunity to establish SHAP as the international clinical standard in hand assessment, particularly given the expediency of the introduction of an on-line database.

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Appendix A

Data Sheets

A.1 Maxon Motor/Gearbox/Encoder

A.1.1 Maxon Motor RE013-032-06EAB103A

Assigned power rating	W	2.5
Nominal voltage	V	6.00
No load speed	rpm	11400
Stall torque	mNm	8.50
Speed/torque gradient	rpm/mNm	1360
No load current	mA	23.0
Starting current	mA	1720
Terminal resistance	Ω	3.50
Max. permissible speed	rpm	12000
Max. continuous current	mA	590
Max. continuous torque	mNm	2.92
Max. power output at nominal voltage	mW	2520
Max. efficiency	%	78.6
Torque constant	mNm/A	4.95
Speed constant	rpm/V	1930
Mechanical time constant	ms	6.89
Rotor inertia	gcm^2	0.484
Terminal inductance	mH	0.11
Thermal resistance housing-ambient	K/W	33.0
Thermal resistance rotor-housing	K/W	7.00
Ambient temperature range	°C	-20/+65
Weight of motor	g	21–24

A.1.2 Maxon Planetary Gearhead GP013A020-0017 B1A00A

Reduction ratio	16.58:1
No. of stages	2
Max. continuous torque	0.20Nm
Max. intermittent torque	0.30Nm
Max. efficiency	83
Weight	14g
Length (incl. encoder)	59.05mm

A.1.3 Maxon Digital Magnetic Encoder 3425

Supply voltage V_{cc}	3.8–24V
Output signal at V_{cc}=5V	TTL compatible
No. of channels	2
Counts per revolution	16
Phase shift between channels	90°
Power output	max. 8mA
Max. operating frequency	20kHz

A.2 MINIMOTOR DC-Micromotor/Gearhead/Encoder

A.2.1 MINIMOTOR DC-Micromotor 1016 006G

Nominal voltage	V	6.00
Terminal resistance	Ω	20.1
Output Power	W	0.46
Efficiency	%	68
No load speed	rpm	17600
No load current	mA	10.0
Stall torque	mNm	1.06
Max. permissible speed	rpm	13000
Max. continuous current	mA	180
Max. continuous torque	mNm	0.5
Speed/torque gradient	rpm/mNm	15605
Torque constant	mNm/A	3.35
Speed constant	rpm/V	2854
Mechanical time constant	ms	9
Rotor inertia	gcm^2	0.06
Rotor inductance	mH	0.06
Thermal resistance housing-ambient	K/W	65.0
Thermal resistance rotor-housing	K/W	10.0
Ambient temperature range	°C	-30/+85
Weight of motor	g	6.5

A.2.2 MINIMOTOR Planetary Gearhead 10/1

Reduction ratio	64:1
Max. continuous torque	54mNm
Max. intermittent torque	200mNm
Max. efficiency	70
Weight	8g
Length (incl. encoder)	43.1mm

A.2.3 MINIMOTOR Digital Magnetic Encoder 30B

Supply voltage V_{cc}	4.5–5.5V
Current consumption at V_{cc}=5V	5mA
No. of channels	2
Counts per revolution	10
Phase shift between channels	90°
Max. operating frequency	7.2kHz
Signal rise/fall time	5/0.2μs

A.3 Worm-Wheel Design

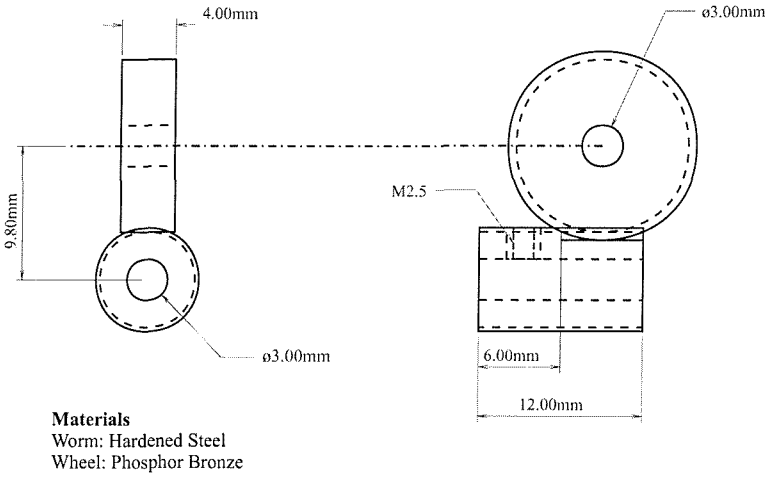

Figure A.1: Custom Worm-Wheel Design

A.4 Estimates of Power Consumption

The estimates of power consumption are based on the following assumptions/data:

- The execution of a single grip consists of the hand moving through the following states: from a natural posture, to the prehensile pattern position (SAMS POSITION), to SAMS TOUCH, to SAMS HOLD, to SAMS RELEASE, and then returning to the original position. This is achieved by flexion (FLEX) or extension (EXT) of the digits, or clockwise (CW) and counter-clockwise (CCW) circumduction of the thumb.

- The current required during the movement of each digit has been recorded at 100mA.

- The current required during the movement of both the flexion and circumduction axes of the thumb has been recorded at 80mA each.

- The digits and thumb circumduction units move at 1.68 rads/s (96.4°/s) with the digits taking 0.84s to reach full curl.

- The thumb flexion unit moves at 0.63 rads/s (36°/s) and takes 2.5s to reach full curl.

- The fingers collectively require 6.88A at full grip force (not normally achieved in a normal HOLD state).

- The thumb flexion unit requires 0.316A at full grip force (not normally achieved in a normal HOLD state).

- Full grip force can be achieved within 50ms.

Natural Posture → SAMS Position						
Grip	Digits	Active Direction	Position (rads)	Time (secs)	Current (mA)	Charge (mAh)
POWER	Fingers	EXT	0.17	0.10	400	0.012
	Thumb	EXT	0.79	1.25	80	0.028
	Thumb Circ.	-	-	-	-	-
PRECISION	Fingers	-	-	-	-	-
	Thumb	-	-	-	-	-
	Thumb Circ.	CCW	0.35	0.21	80	0.005
LATERAL	Fingers	FLEX	0.79	0.47	400	0.052
	Thumb	-	-	-	-	-
	Thumb Circ.	CW	0.35	0.21	80	0.005

SAMS Position → SAMS Touch						
Grip	Digits	Active Direction	Position (rads)	Time (secs)	Current (mA)	Charge (mAh)
POWER	Fingers	FLEX	1.41	0.84	400	0.093
	Thumb	FLEX	1.57	2.50	80	0.056
	Thumb Circ.	-	-	-	-	-
PRECISION	Fingers	FLEX	1.41	0.84	400	0.093
	Thumb	FLEX	1.57	2.50	80	0.056
	Thumb Circ.	-	-	-	-	-
LATERAL	Fingers	-	-	-	-	-
	Thumb	FLEX	1.57	2.50	80	0.056
	Thumb Circ.	-	-	-	-	-

SAMS Touch → SAMS Hold						
Grip	Digits	Active Direction	Position (rads)	Time (secs)	Current (mA)	Charge (mAh)
POWER	Fingers	FLEX	-	0.05	6880	0.096
	Thumb	-	-	-	-	-
	Thumb Circ.	-	-	-	-	-
PRECISION	Fingers	FLEX	-	0.05	6880	0.096
	Thumb	FLEX	-	0.05	316	0.004
	Thumb Circ.	-	-	-	-	-
LATERAL	Fingers	-	-	-	-	-
	Thumb	FLEX	-	0.05	316	0.004
	Thumb Circ.	-	-	-	-	-

SAMS Hold → SAMS Release						
Grip	Digits	Active Direction	Position (rads)	Time (secs)	Current (mA)	Charge (mAh)
POWER	Fingers	EXT	1.41	0.84	400	0.093
	Thumb	EXT	1.57	2.50	80	0.056
	Thumb Circ.	-	-	-	-	-
PRECISION	Fingers	EXT	1.41	0.84	400	0.093
	Thumb	EXT	1.57	2.50	80	0.056
	Thumb Circ.	-	-	-	-	-
LATERAL	Fingers	-	-	-	-	-
	Thumb	EXT	1.57	2.50	80	0.056
	Thumb Circ.	-	-	-	-	-

SAMS Release → Natural Posture						
Grip	Digits	Active Direction	Position (rads)	Time (secs)	Current (mA)	Charge (mAh)
POWER	Fingers	FLEX	0.17	0.10	400	0.012
	Thumb	FLEX	0.79	1.25	80	0.028
	Thumb Circ.	-	-	-	-	-
PRECISION	Fingers	FLEX	0.17	0.10	400	0.012
	Thumb	FLEX	0.79	1.25	80	0.028
	Thumb Circ.	CW	0.35	0.21	80	0.005
LATERAL	Fingers	EXT	0.79	0.47	400	0.052
	Thumb	FLEX	0.79	1.25	80	0.028
	Thumb Circ.	CCW	0.35	0.21	80	0.005

Grip	Total Charge to Execute Grip Cycle (mAh)	Percentage Use of Grip (see section 5.2.2)
POWER	0.472	30%
PRECISION	0.446	50%
LATERAL	0.256	20%

Accounting for the prehensile pattern weighting, the average charge required per grip cycle is 0.416mAh, which results in 2405 grips achievable with a 1000mAh battery.

Appendix B

Technical Drawings

Drawings shown represent the components manufactured to form the second (and latest) Southampton-Remedi prototype hand. All figures shown are NOT TO SCALE.

Figure	Title
B.1	Finger Linkage (links A–C)
B.2	Finger Linkage (links D–G)
B.3	Digit 5 Finger Linkage (links A–C)
B.4	Digit 5 Finger Linkage (links D–G)
B.5	Knuckle Block
B.6	Digit 5 Knuckle Block
B.7	Thumb Circumduction Unit
B.8	Thumb Flexion Unit
B.9	Thumb Stump
B.10	Palm Unit

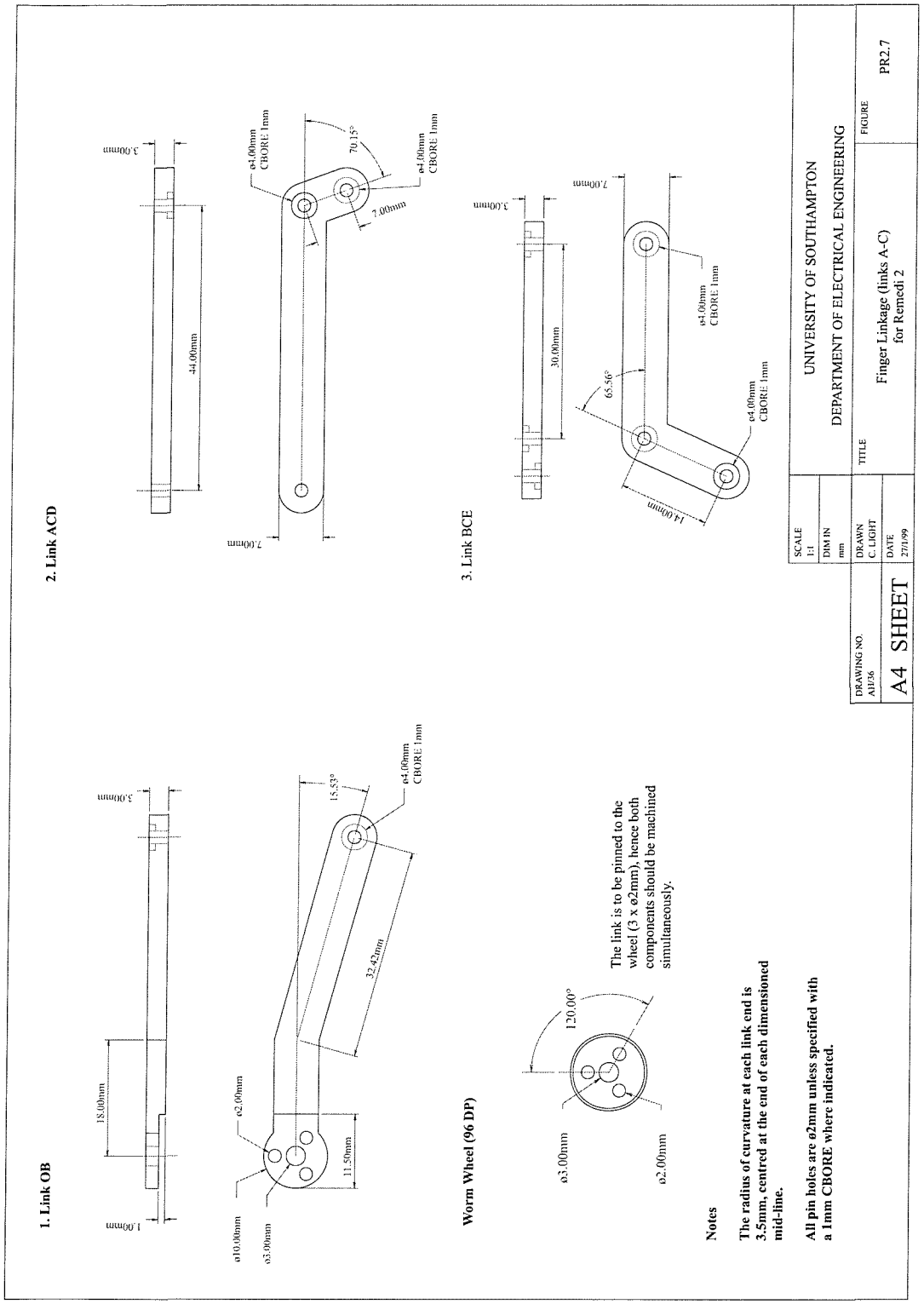

Figure B.1: Finger Linkage (links A–C) for Southampton-Remedi Prototype

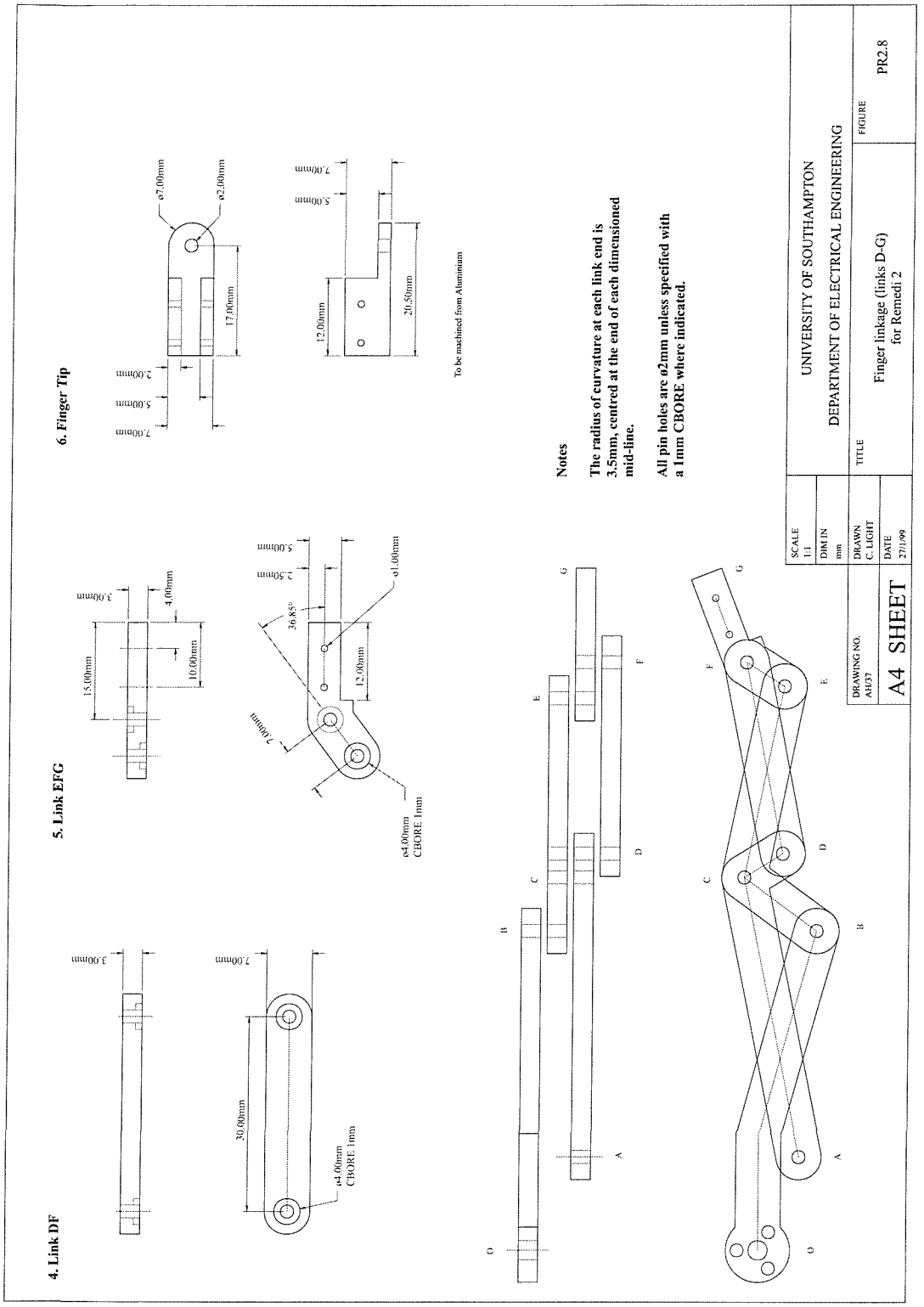

Figure B.2: Finger Linkage (links D–G) for Southampton-Remedi Prototype

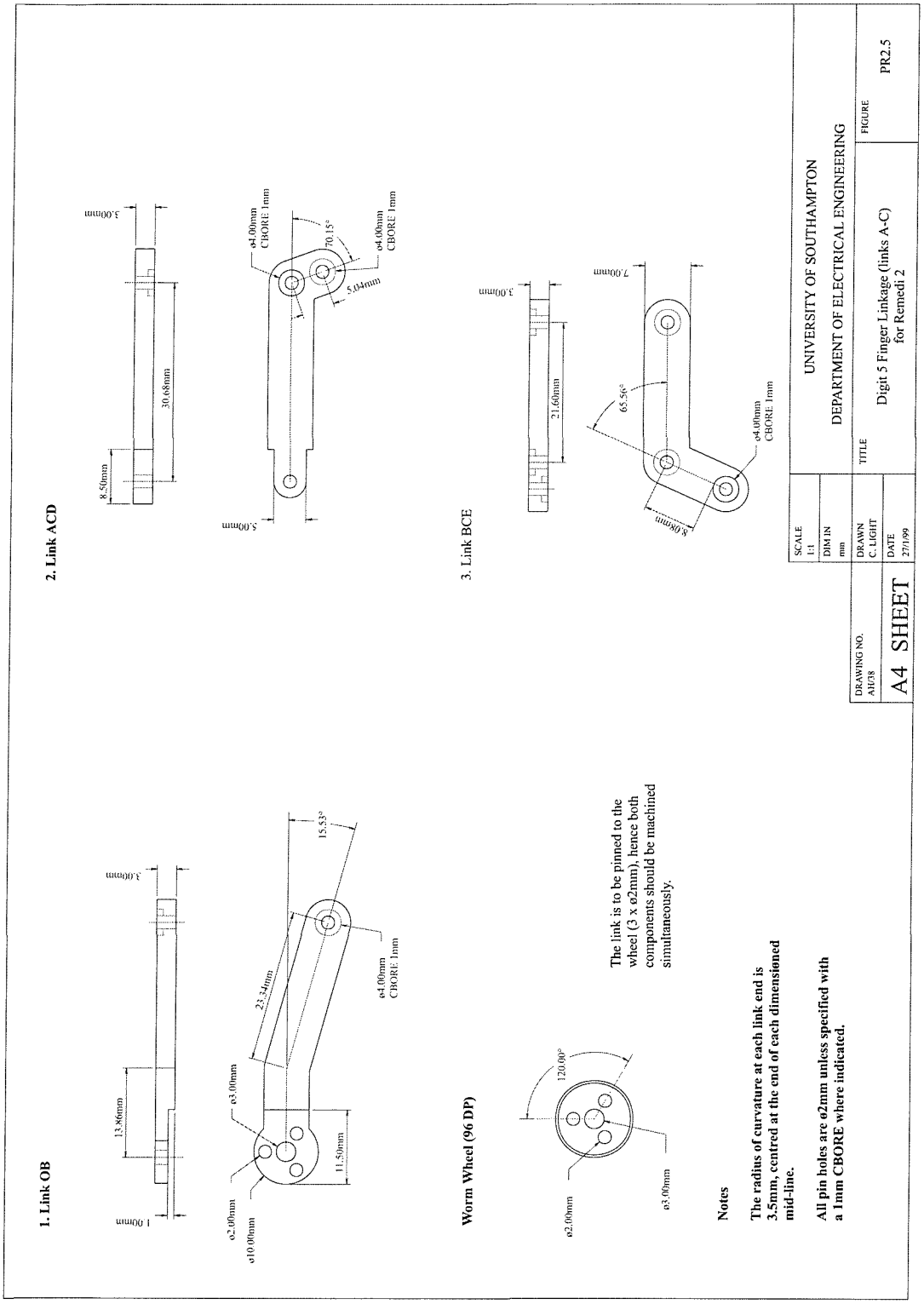

Figure B.3: Digit 5 Finger Linkage (links A–C) for Southampton-Remedi Prototype

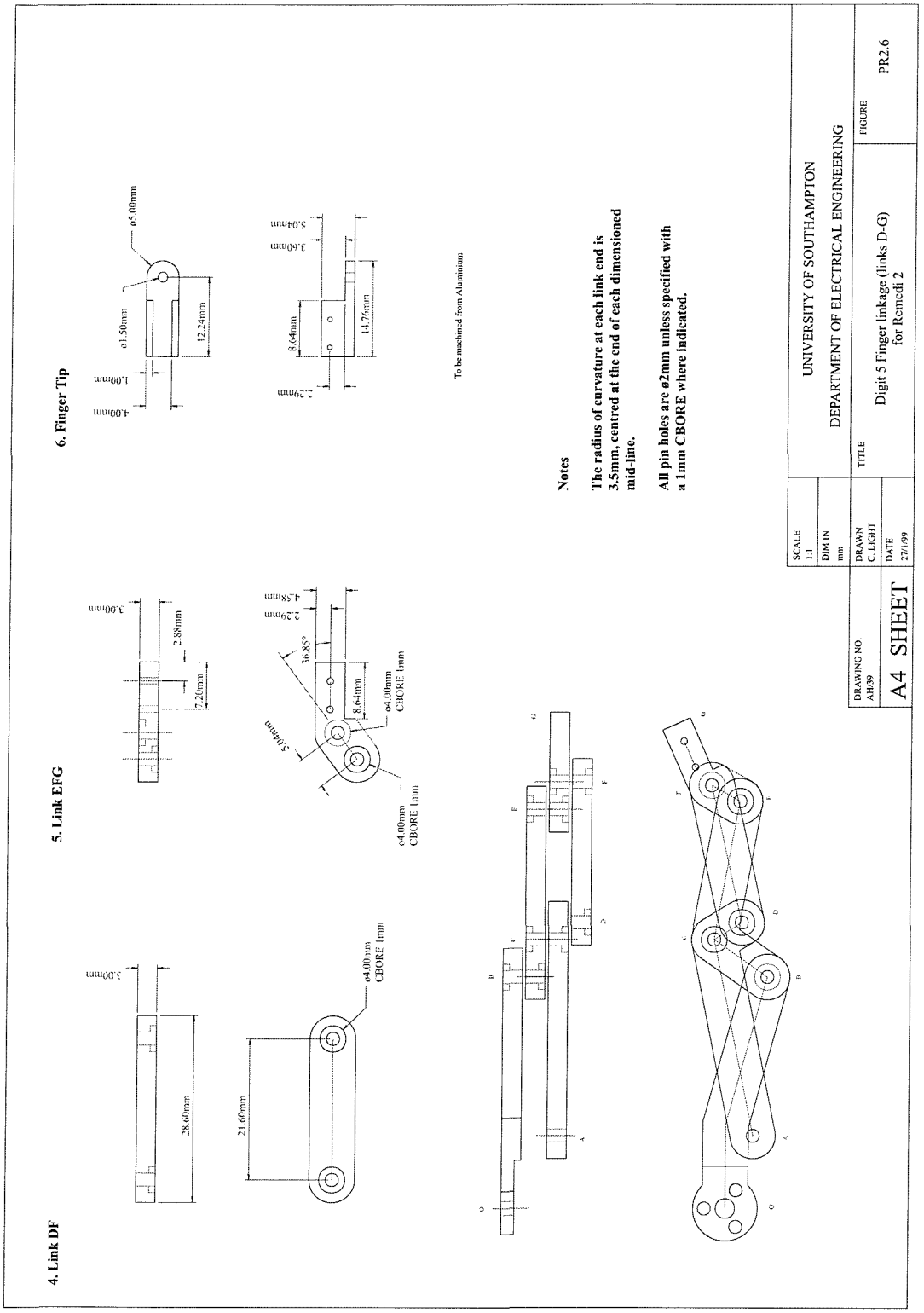

Figure B.4: Digit 5 Finger Linkage (links D–G) for Southampton-Remedi Prototype

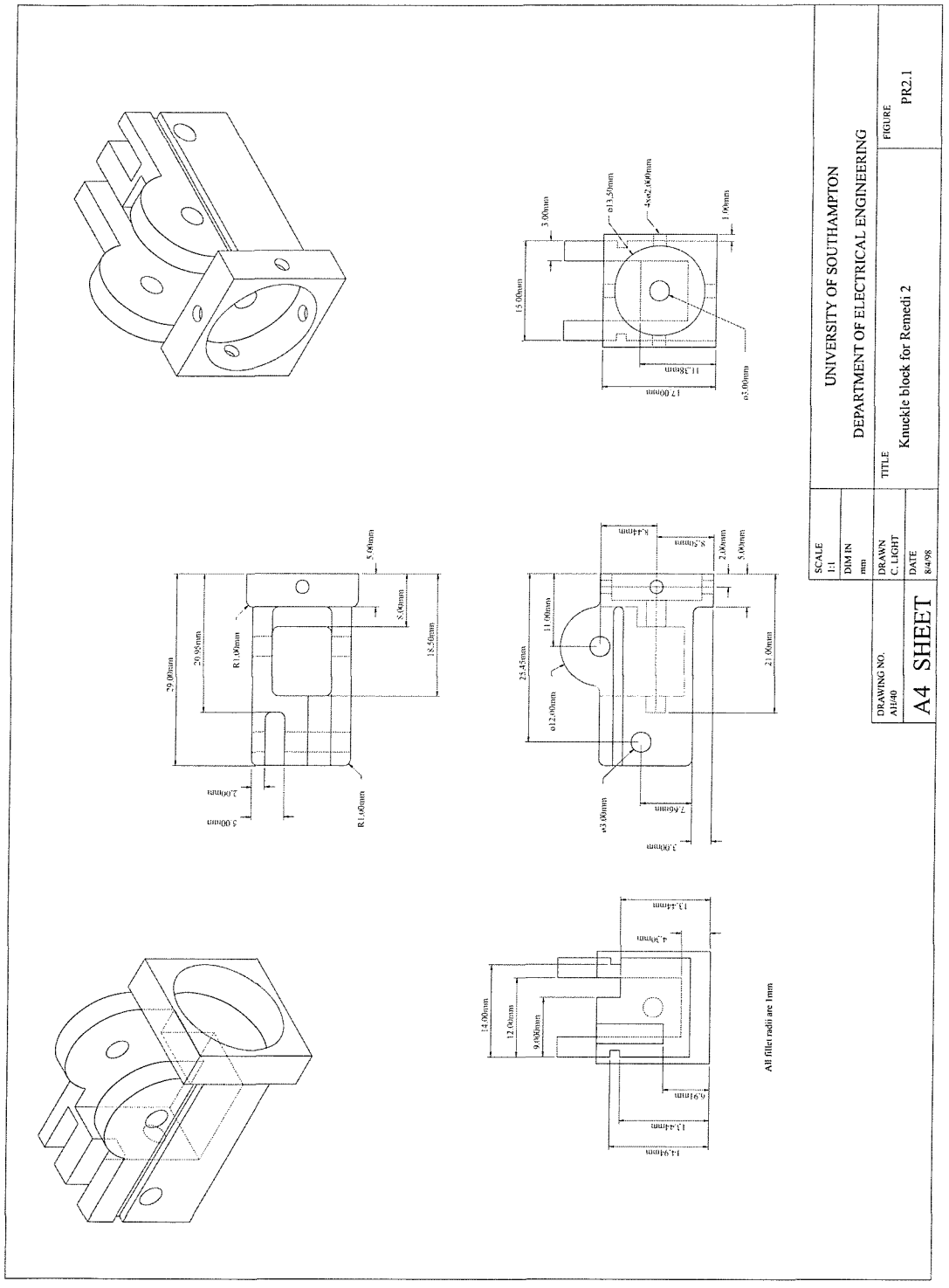

Figure B.5: Knuckle Block for Southampton-Remedi Prototype

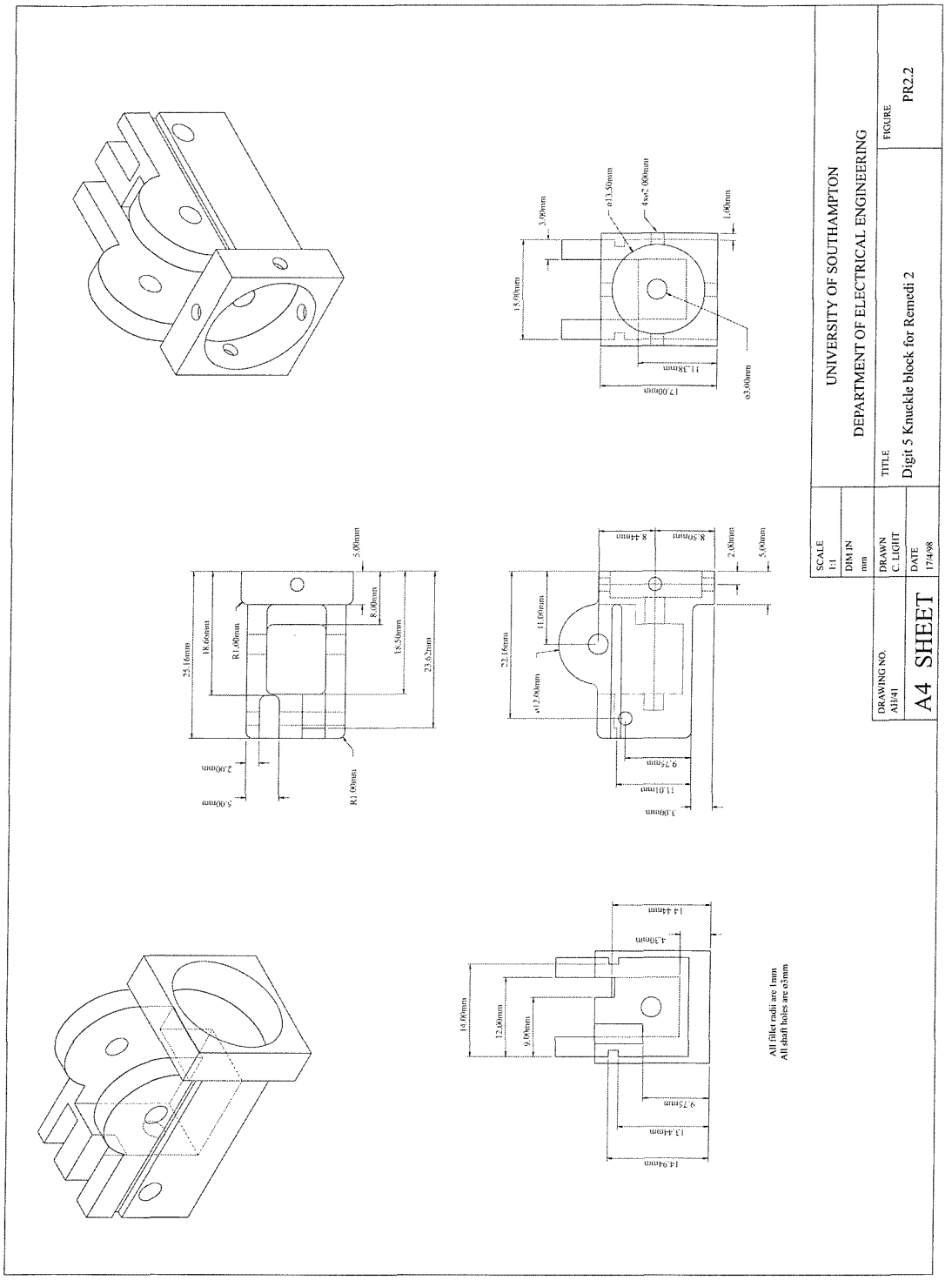

Figure B.6: Digit 5 Knuckle Block for Southampton-Remedi Prototype

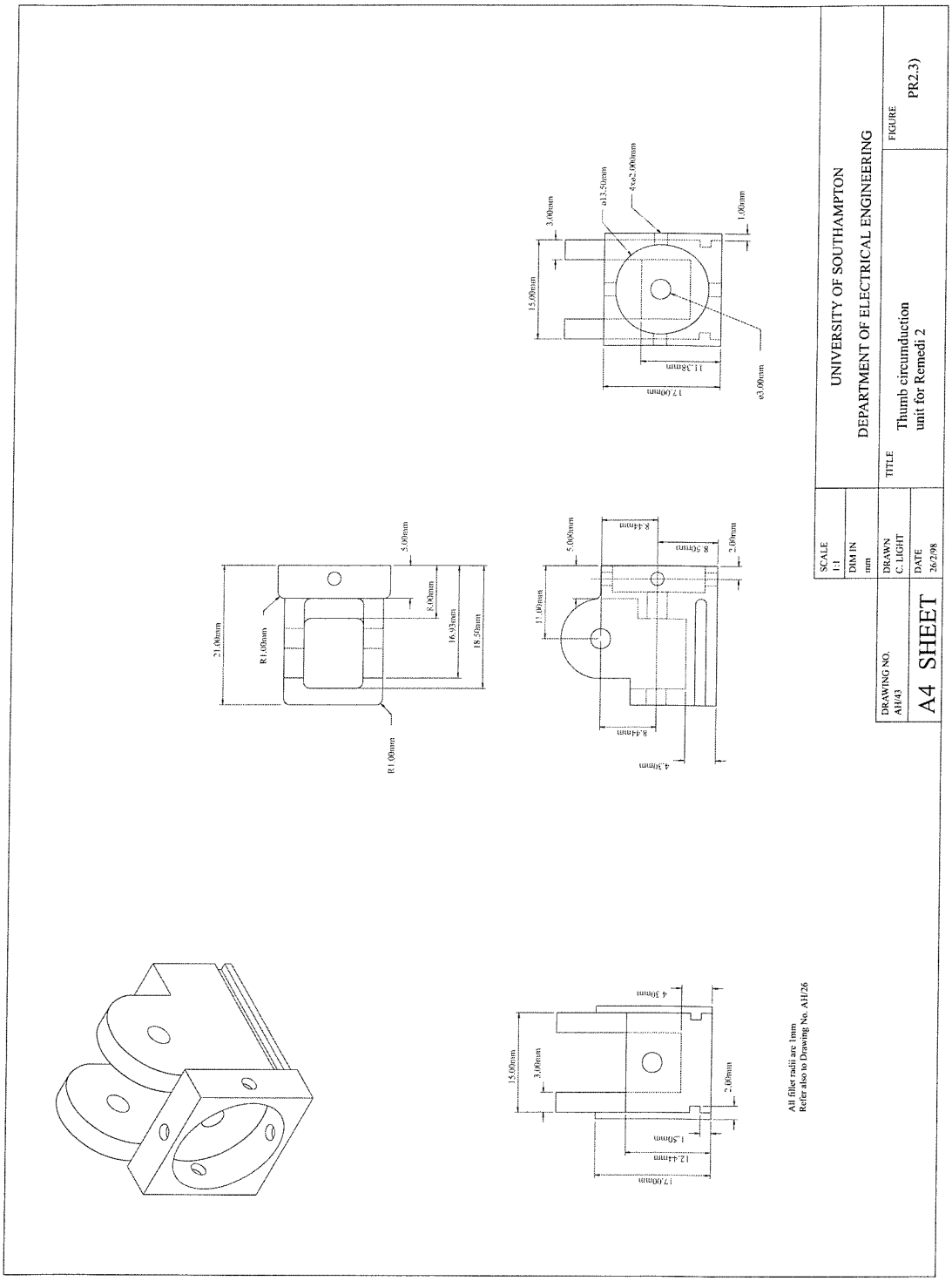

Figure B.7: Thumb Circumduction Unit for Southampton-Remedi Prototype

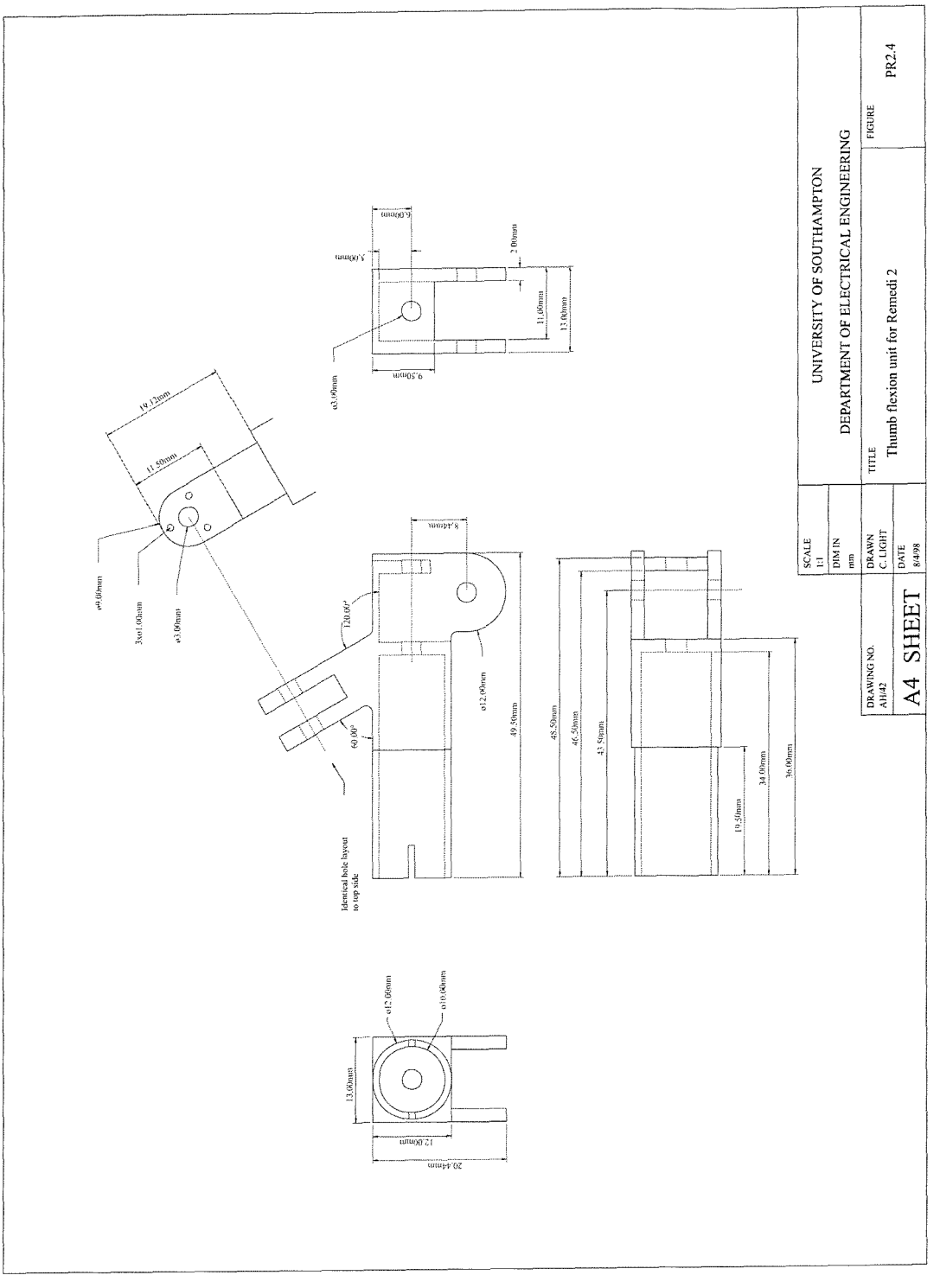

Figure B.8: Thumb Flexion Unit for Southampton-Remedi Prototype

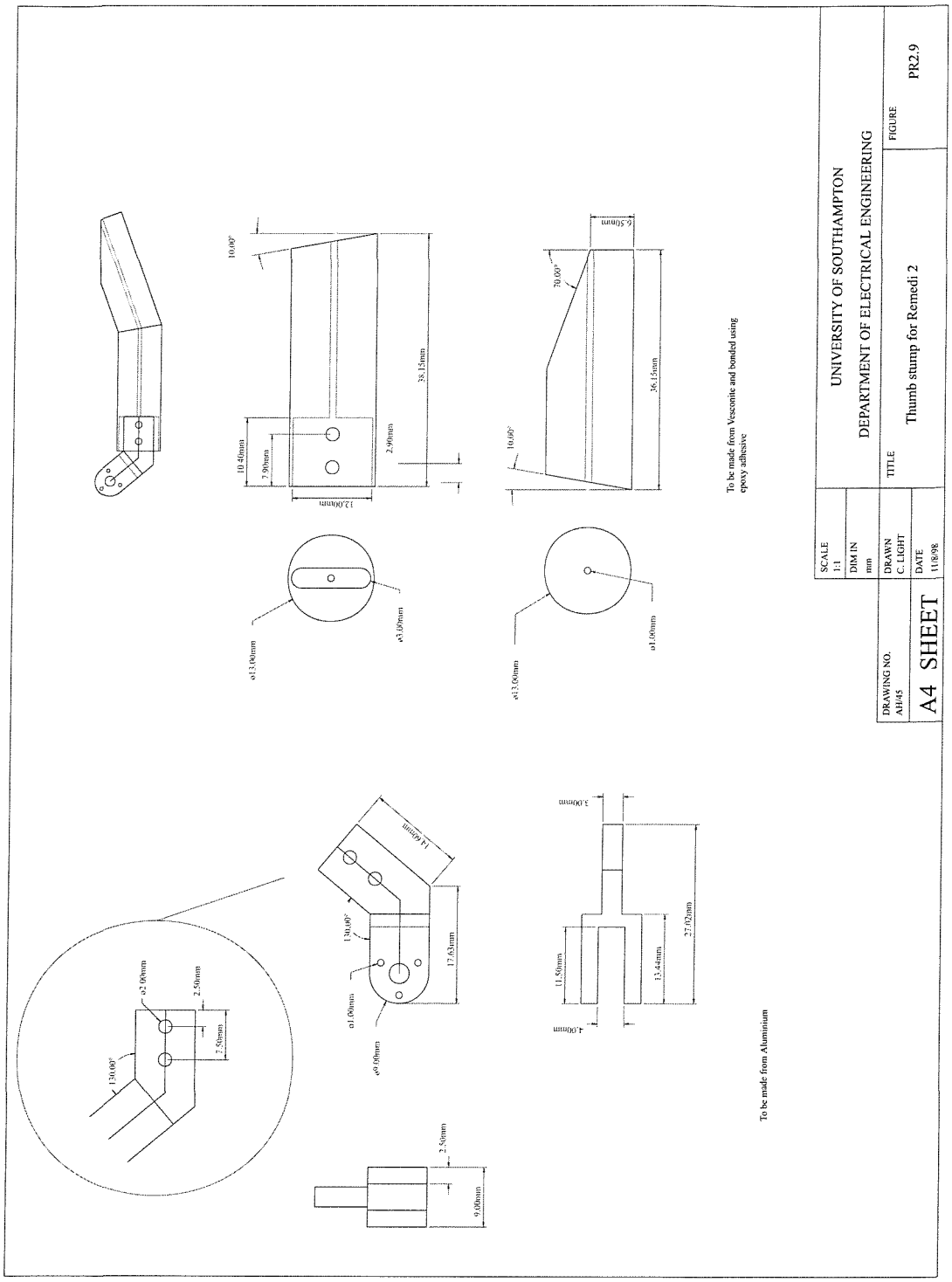

Figure B.9: Thumb Stump for Southampton-Remedi Prototype

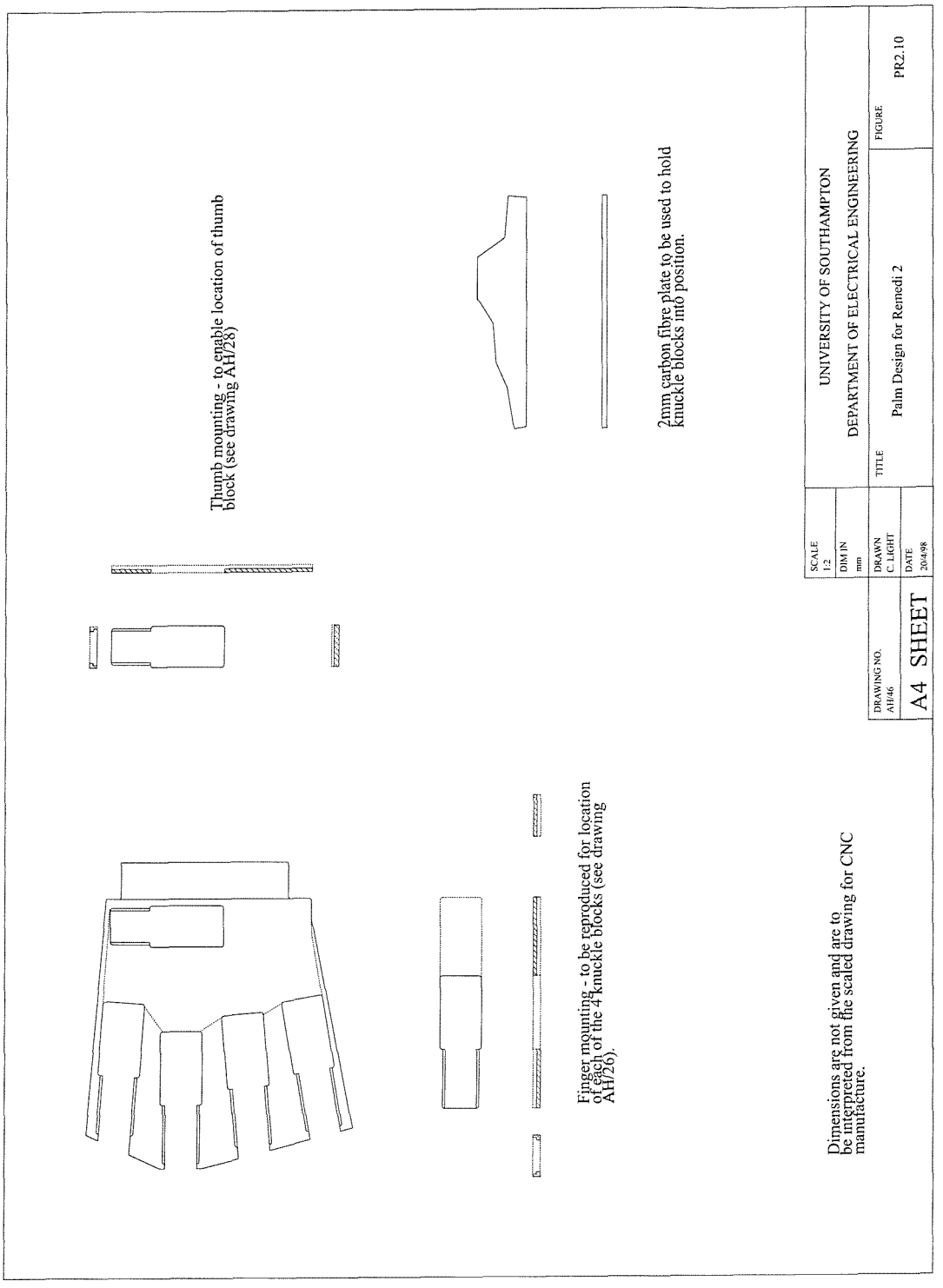

Figure B.10: Palm Design for Southampton-Remedi Prototype

Appendix C

Electronics Design

Schematics shown represent the electronic hardware interface system between the Southampton-Remedi hand and the DSP microprocessor.

Figure	Title
C.1	H-bridge Design
C.2	Hardware Interface and Position Decoders
C.3	Motor Current Filters
C.4	Slip Processors

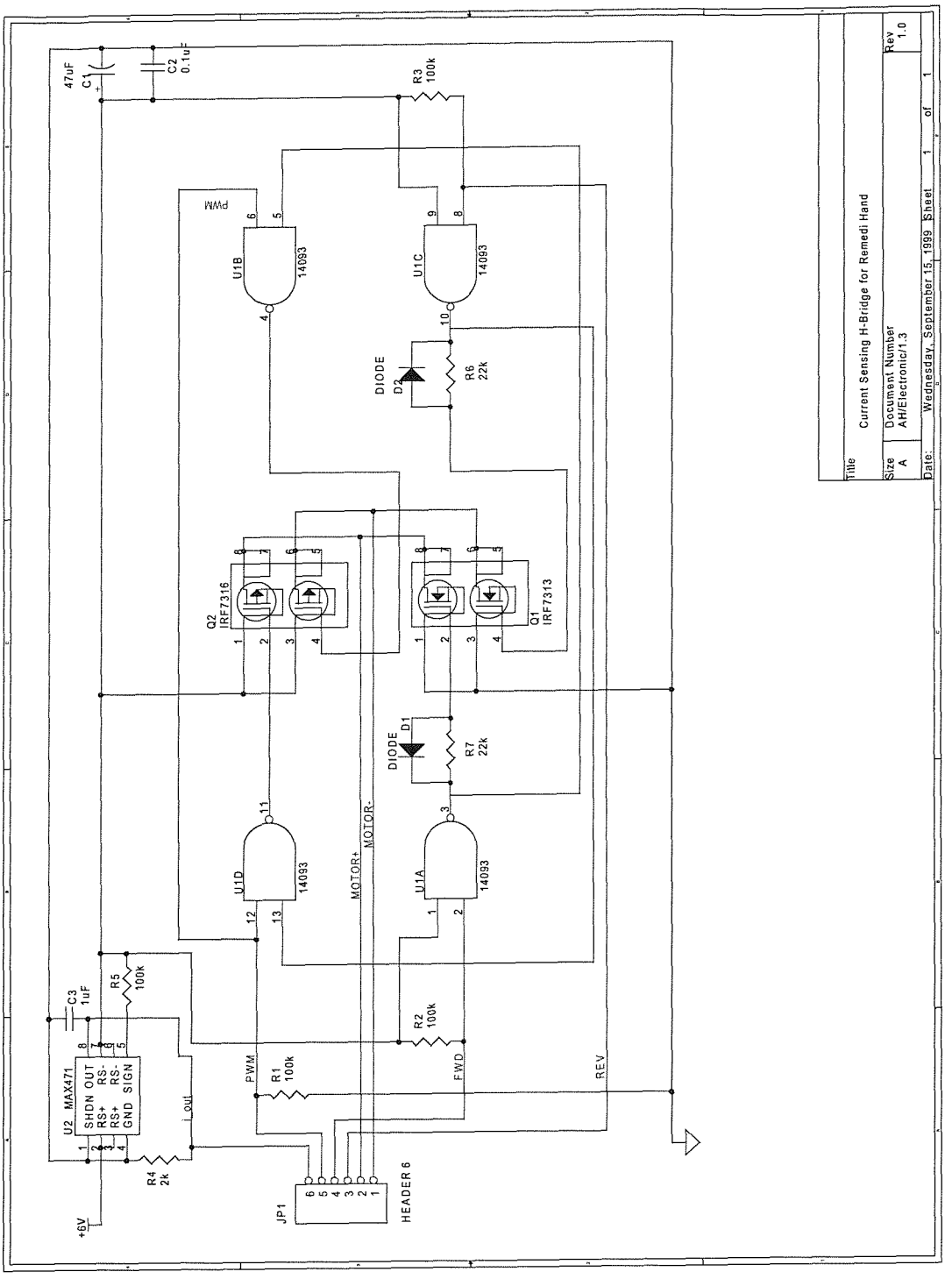

Figure C.1: H-bridge Design

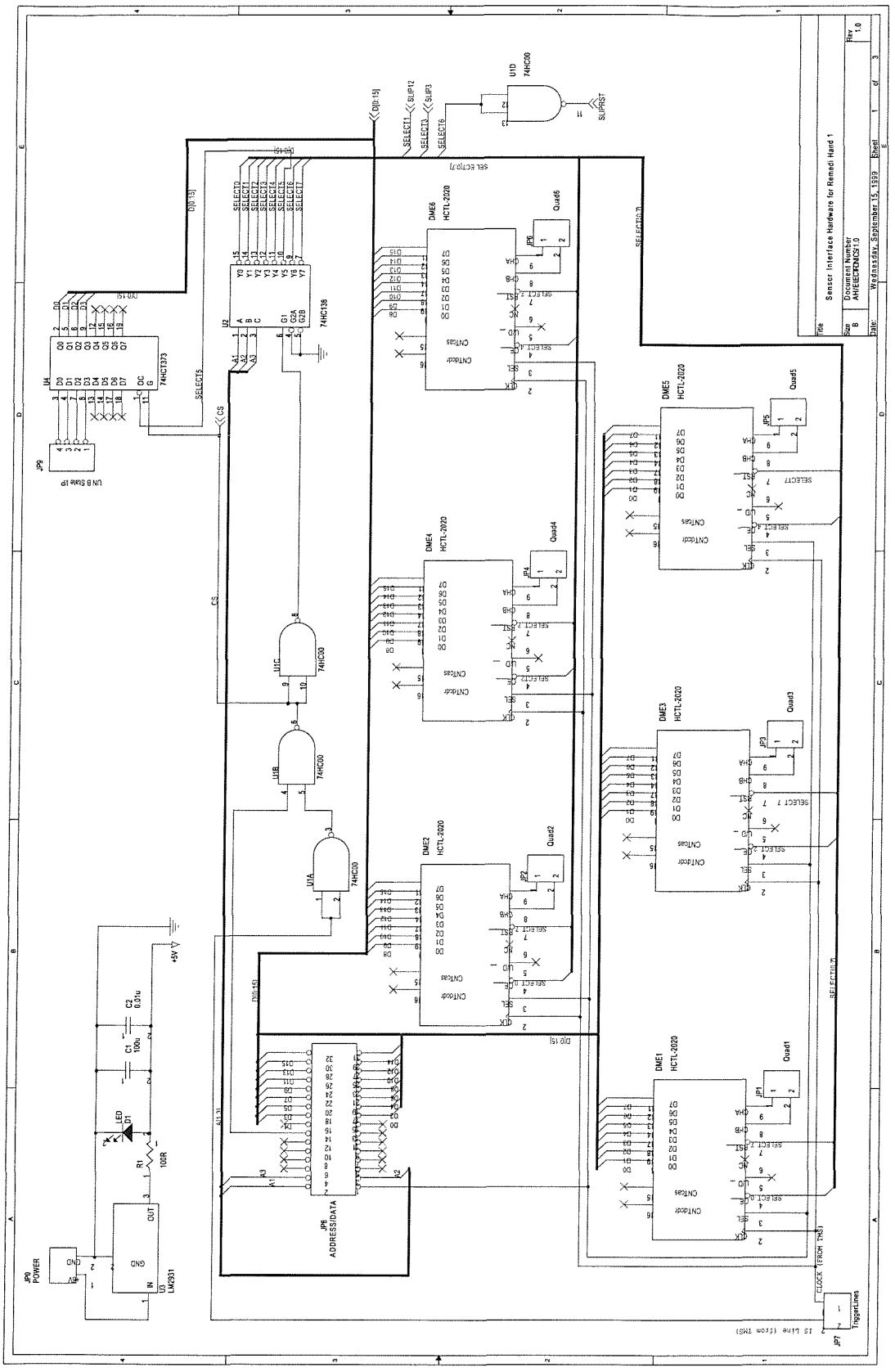

Figure C.2: Hardware Interface and Position Decoders

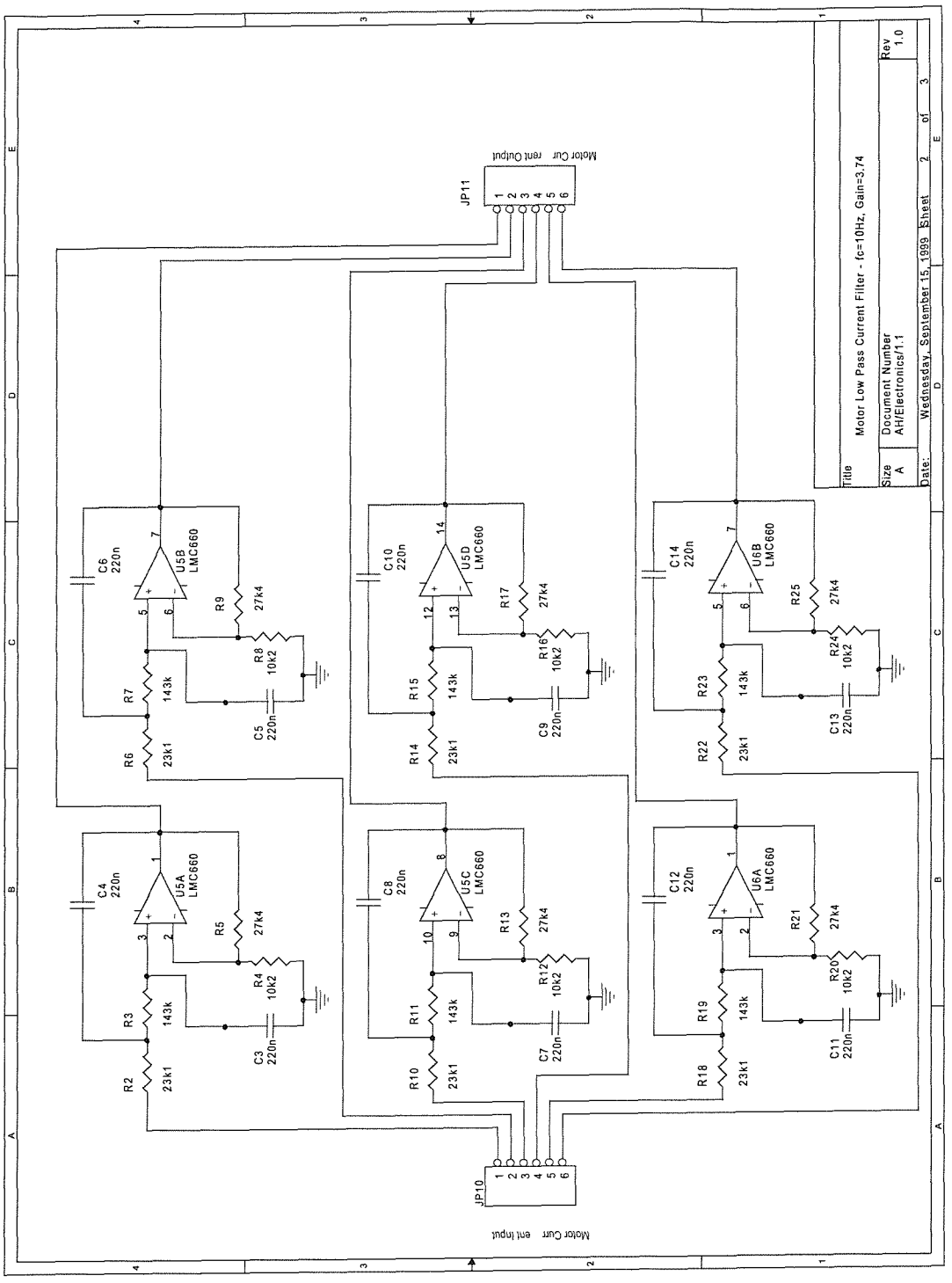

Figure C.3: Motor Current Filters

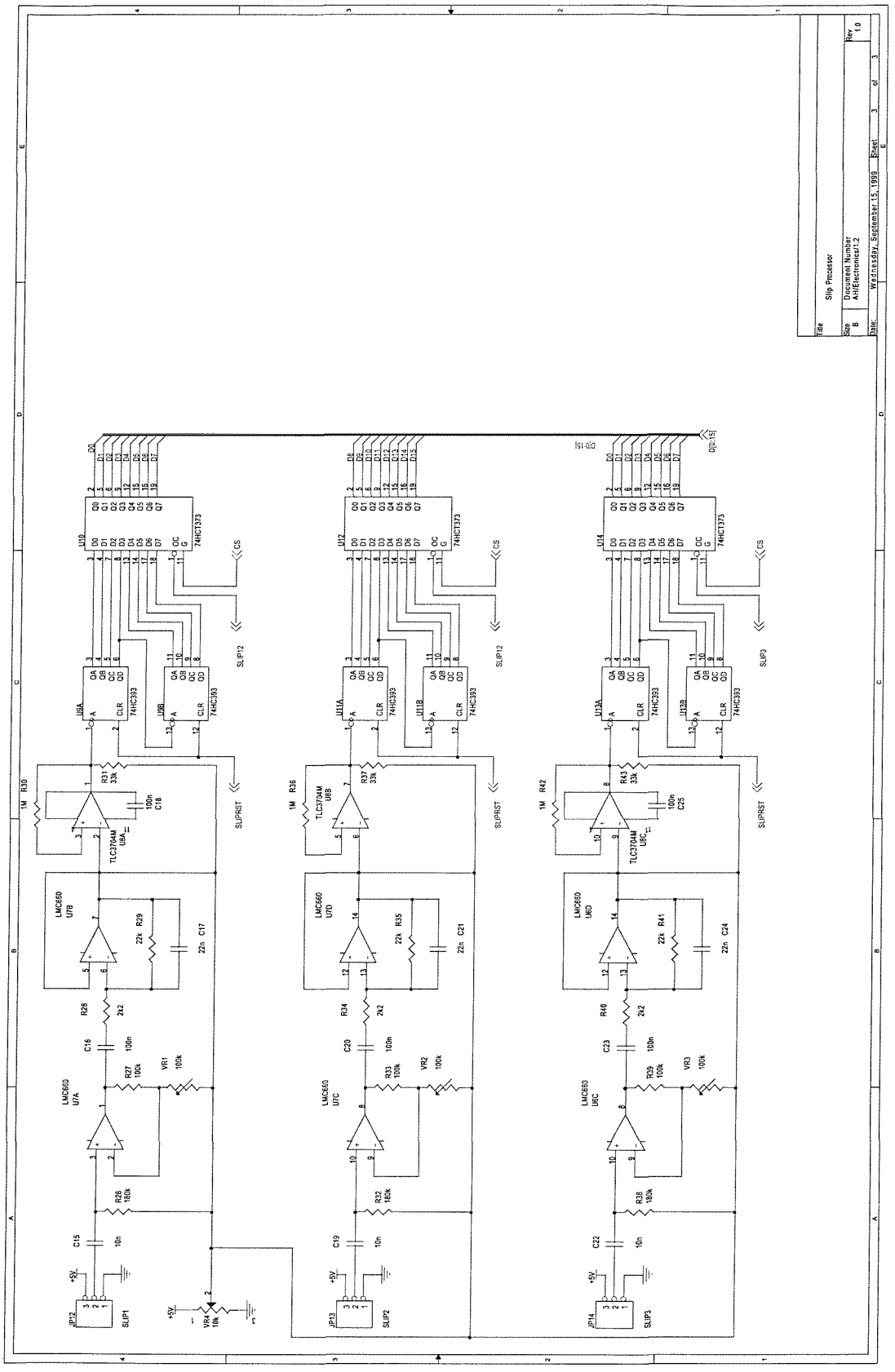

Figure C.4: Slip Processors

Appendix D

Southampton Hand Assessment Procedure

D.1 General Information

The Southampton procedure has been formed based on the analysis of grip patterns, and their frequency of use in Activities of Daily Living (ADL). Therefore it is considered to cover the wide range of prehensile tasks the hand usually undertakes (with the omission of specific occupational or recreational requirements).

The test consists of the manipulation of a series of both lightweight and metal abstract objects. These are intended to directly reflect specific grip patterns, whilst also assessing the strength and compliance of grip. This is followed by 14 ADL tasks. To ensure standardisation, the assessor's test procedure must be followed, whilst objectivity is maintained by subject self-timing. A complete assessment is expected to take around 20-30 mins to complete, (including all of the relevant explanations to the subject).

The procedure is designed to provide a 'functionality rating', hence on completion of the test, a percentage of optimum hand function can be obtained. This figure provides a tangible result describing the level of hand impairment. As the procedure has been designed to be standardised and objective, this figure cannot only be used for comparative assessments of a patient's performance throughout a course of treatment, but also provides information on their level of function (with respect to

the benchmark of a healthy subject).

The protocol outlined in the following pages provides details for the assessor concerning the setup and execution of the assessment procedure. The assessor is required to demonstrate each task according to the descriptions given (the following diagrams may aid identification of the appropriate grip patterns).

Plain text denotes the demonstration instructions to the assessor.

Text in italics denotes instructions to be spoken to the subject

D.2 Contents of Test Unit

Quantity	Item
1	Test unit/case (containing all assessment equipment)
1	Backboard mounted in case (with Yale lock, door handle, and zip)
1	Red/Blue sided test platform
1	Foam containing all objects
1	Timer unit
6	Lightweight Abstract Objects (see Figure D.2)
6	Metal Abstract Objects (see Figure D.2)
1	Yale lock (mounted in backboard), and key
1	Zip (mounted in backboard)
4	Coins (two 2p, two 1p)
1	Button board (containing 4 buttons)
1	Plasticine block
1	Knife
1	4 by 6 inch card
1	Jar (with lid)
1	Small jug
1	1 ltr carton (Tetrapak)
1	Empty tin (approx. 420g size)
1	Door handle (mounted on backboard)
1	Screw (mounted on metal clip)
1	Screwdriver

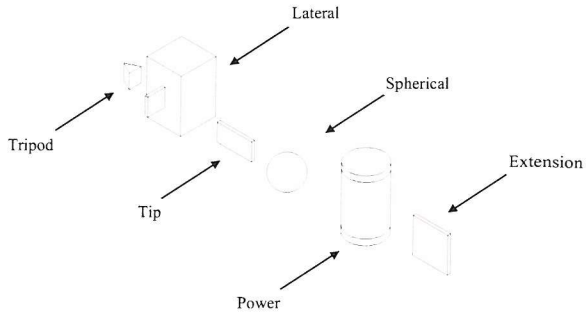

Figure D.1: Abstract Objects Nomenclature

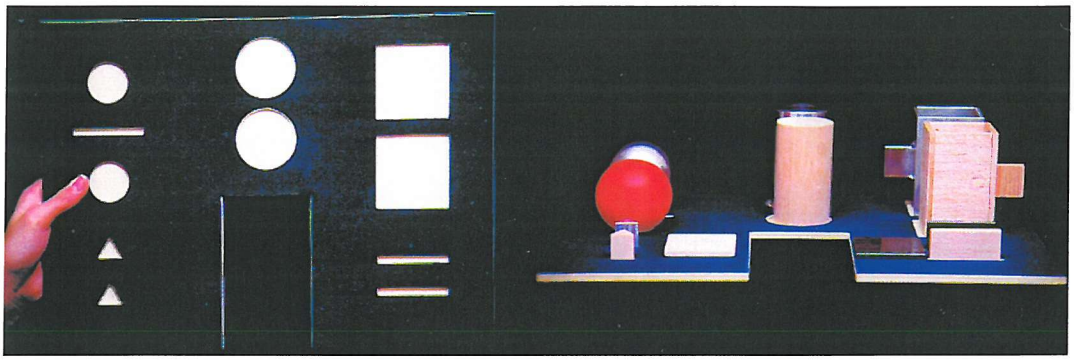

Figure D.2: SHAP Form-Board and Abstract Objects

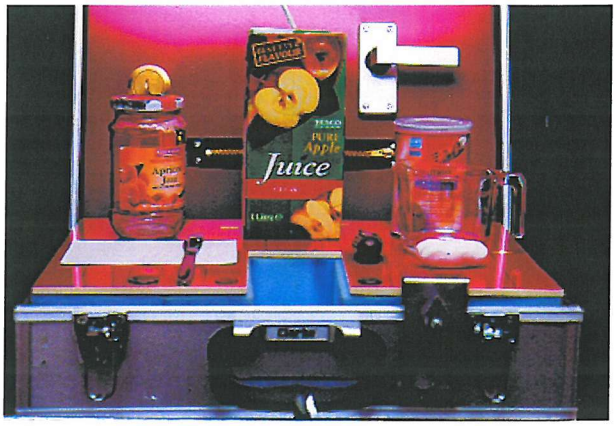

Figure D.3: SHAP Activities of Daily Living

D.3 Assessor's Test Procedure

D.3.1 Setting up the assessment

The subject should be seated at a table. With relaxed shoulders and arms resting on the table, the subject's elbows should be at a 90° angle.

Place the test platform (red/blue sided) directly in front of the subject (blue side facing upwards), approximately 3 inches from the front edge of the table. Fit the timer unit into the space provided in the front of the platform. For each of the following abstract tasks, the board should be moved from left to right so that each task is directly in front of the subject, thereby ensuring no bias towards one hand. The case and all ADL objects may be removed from the table.

D.3.2 Procedural Notes

Each task should be demonstrated to the subject using slow, clear movements, ensuring that the subject is aware of the appropriate grip. The subject should be given the opportunity to ask questions prior to the commencement of each task.

It is important to note that the demonstration should be carried out using the corresponding hand under assessment, to avoid any confusion for the subject.

The **'optional'** instructions should be used only when the assessor feels that the subject would be unable, is uncomfortable, or unnatural in using the demonstrated grip.

Prosthesis users should be encouraged to practice each task, prior to timing the event, in order to determine the most appropriate technique (as many users often carry out tasks with the natural hand alone). Due to the difficulties associated with myoelectric prostheses, if it is apparent that the device has failed to respond to user demand, then a note should be made, and a retest allowed. If the prosthesis is similarly unresponsive, the second task time should be recorded and a note made of the difficulties encountered.

Only one chance to carry out the timed task should be given, unless a serious handling error causes an unrealistic result. The time to complete the task (and the

appropriate grip if readily identifiable) should be recorded, as well as any relevant notes.

When establishing any form of normative data it is imperative that the task is carried out fully. Due to the need to complete in the minimum time, there is frequently a temptation to 'rush' the task without actually fulfilling the exact requirements. Under these circumstances the task should be repeated.

Abstract Objects

The lightweight objects are to be used first. If a subject cannot complete the task, this should be recorded as C/C (Cannot Complete).

"A series of objects will be placed on the board. The task involves moving the object from the rear slot to the front slot. Only the hand under assessment should be used for any of these tasks, including the starting and stopping of the timer."

Spherical – Place the 'spherical object' in the appropriate slot. Place the 'tip object' in the slot between rear and front 'spherical object slots' to create a small barrier. Move the board so that these slots are directly in front of the subject (maintaining the distance from the front of the table). Using a spherical grip, move the ball over the barrier to the front slot.

"Start the timer, pick up and move the object as demonstrated with as few mistakes as possible, and as quickly as possible, to the front slot. Complete the task by depressing the blue button again."

*[**Optional**: "If you feel unable to pick up the object as demonstrated, you may use any method you wish whilst using only one hand"]*

Tripod – Place the 'tripod object' in the appropriate rear slot. Using a tripod grip, move the object to the front slot.

"Start the timer, move the object as demonstrated and as quickly as possible to the front slot, and then stop the timer."

Power – Place the 'power object' in the appropriate rear slot. Move the board so that these slots are directly in front of the subject (maintaining the distance from

the front of the table). Using the power grip, pick up the object by the cylinder (between the two markers), and move to the front slot.

"Start the timer, pick up the object between the two markers as demonstrated, and move it as quickly as possible, to the front slot, and then stop the timer."

Lateral – Place the 'lateral object' in the appropriate slot with the handle facing towards the subject. Move the board so that these slots are directly in front of the subject (maintaining the distance from the front of the table). Using a lateral grip, pick up the object by the handle, and move to the front slot.

"Start the timer, move the object as demonstrated and as quickly as possible to the front slot, and then stop the timer."

Tip – Place the 'tip object' in the appropriate slot. Using a tip (either 2 or 3 point) grip, move the object to the front slot.

"Start the timer, move the object as demonstrated and as quickly as possible to the front slot, and then stop the timer."

Extension – Place the 'extension object' in the appropriate rear slot. Using an extension grip (with the thumb on the front of the object, and fingers extended flat on the rear side), move the object to the front slot.

"Start the timer, move the object as demonstrated and as quickly as possible to the front slot, and then stop the timer."

The procedure should now be repeated, in the same order using the metal objects. If a subject has failed to complete tasks with the lightweight objects, then the appropriate heavier object tasks may be ignored (to avoid undue strain on the subject). In this instance, a 'Cannot Complete (C/C)' should be recorded on the form.

Once completed, place the form board objects in the foam. Turn the test platform over (the red side facing upwards), and position as before, with the timer in the space provided. The platform should remain centred in front of the subject for all ADL tasks.

Activities of Daily Living

As before, each task should be demonstrated to the subject using slow, clear movements, ensuring that the subject is aware of the appropriate grip. The 'optional' instructions should be used when the assessor feels that the subject would be unable, is uncomfortable, or unnatural in using the demonstrated grip.

During instructions to the assessor, references to 'handed' infers the hand under assessment (not necessarily the subject's dominant hand).

"The second stage of this assessment consists of 14 everyday activities, which should be timed in the same manner by depressing the blue button to start and stop the timer. Again tasks should be completed as quickly as possible, with as few mistakes as possible, using only the appropriate hand unless otherwise instructed."

1. Pick up coins – Arrange the two 2p and two 1p coins in the designated areas on the red platform. Place the jar in the designated spot for this test with the lid removed. Pick up each coin in turn (by sliding to the edge of the platform), using a tip or tripod grip, and drop into the jar. Move from right to left. Reset the task.

"Start the timer, lift each coin in turn as quickly as possible, drop in the jar, as demonstrated, and then stop the timer."

*[**Optional:** "If you feel unable to pick up the object as demonstrated, you may use any method you wish whilst using only one hand"]*

2. Button board – Place the button board to the right of the timer unit if assessing the right hand, and to the left if assessing the left hand. The buttons should be farthest from the timer unit. Undo each button in turn, using only the assessed hand (as a test of dexterity) in a tripod grip. The other hand may be used to steady the board, but may not assist in the task. The board should remain on the platform. Reset the task.

"Start the timer, and using only the appropriate hand, undo all four buttons in any order as demonstrated and as quickly as possible. You may steady the board with the your other hand so that it remains on the platform throughout the task. Then

stop the timer using only the appropriate hand. You may now practice this task."

3. Cutting – Place the knife to the side of the timer unit (appropriately arranged for the assessed hand). Place the plasticine 'food item' in the designated area on the red platform. Pick up the knife and using the other hand to steady the object, cut it clearly into two sections. Then replace the knife on the platform, remould the plasticine, and reset the task.

"Start the timer, use the knife provided to cut the plasticine object clearly into two sections, as demonstrated and as quickly as possible. You may use the other hand to steady the object. Return the knife to the platform, and then stop the timer."

4. Simulated page turning – Place the 4 inch by 6 inch card in the designated area on the opposing side of the platform to the hand under assessment. Using an extension, or tripod grip, pick up the card, turn over, and place in the opposite designated area (as if turning the page of a book). Reset the task.

"Start the timer, lift, turn over (as if turning the page of a book), and replace the card on the platform, as demonstrated and as quickly as possible. Then stop the timer."

5. Jar lid – The lid should be placed on the empty jar, and tightened only with sufficient force as would be expected for everyday use/shelf storage. The jar should be placed in the designated area on the red platform. Both hands should be used for this task. Pick up the jar with the non-assessed hand, undo the lid, (using a flexion grip with the lid firmly in the palm to form a combined power/precision grip) using the assessed hand, and return both the jar and lid to the platform. Reset the task.

"Start the timer, pick up the jar, and undo the lid with the hand under assessment as demonstrated and as quickly as possible. Return the jar and lid to the platform and stop the timer."

To avoid repetitive filling/emptying of objects with water during the following 4 tasks, it is advisable to fill a separate container with approximately one pint of water. It may also be advisable to have a towel nearby.

6. Pouring from jug – Fill the jug with 100ml of water (100ml is marked on

the jug). Place the jug in the designated area on the red test platform, with the handle pointing to the right for right handed subjects, and to the left for left handed subjects. Place the jug (without lid) on the designated left area for right handed people, and on the designated right area for left handed people. Lift the jug by the handle (in a lateral grip), and pour the water into the jar. Reset the task.

"Start the timer, and whilst ensuring as little spillage as possible, pour the water from the jug to the jar, as demonstrated and as quickly as possible. Then stop the timer. You should avoid trying to empty the jug of every last drop, and merely ensure the vast majority of the water has been transferred."

7. Pouring from Carton – Fill the carton with 200ml of water. Place in the designated area on the red platform with the spout pointing towards the jar (according to the handedness criteria described for the previous test). Pick up the carton using a flexion grip (similar to a 'flat' spherical grip), and pour the water into the jar. Reset the task.

"Start the timer, and whilst ensuring as little spillage as possible, pour the water from the carton to the jar, as demonstrated and as quickly as possible. Then stop the timer. Again you should avoid trying to empty the jug of every last drop, and merely ensure the vast majority of the water has been transferred."

8. Large Heavy object – Fill the jar with water (to the full mark), and tighten the lid. Place in the designated area on the left side of the red platform (for right handed), or on the right side (for left handed people). Place the empty carton lengthways along the middle of the platform (without obscuring the timer) to create a barrier. Lift the jar over the carton, using a power grip, and place in the opposing marked area.

"Start the timer, move the jar over the carton to the opposing marked area, as demonstrated and as quickly as possible. Then stop the timer."

The water may now be disposed of and will form no further part of the assessment procedures.

9. Large light object – Place the empty tin in the appropriate area on the

left hand side of the red platform (if right handed), or on the right hand side (if left handed). Place the carton to create a barrier as before. Lift the tin over the carton, using a power grip, and place in the opposing marked area.

"Start the timer, move the tin over the carton to the opposing marked area, as demonstrated and as quickly as possible. Then stop the timer."

Place the test unit (with foam inside) on the table, directly in front of the subject, 3 inches from the front. Place the platform on the foam base and timer unit in the appropriate slot. The final 5 tasks will involve the use of the unit.

10. Lift tray – Place the platform (red side upwards), on the table to the left of the test unit (for right handers), or to the right (for left handers), with the board slightly overhanging the front of the table by approx. one inch, with the long edge facing forwards. The timer should remain in the unit. Both hands should be used to pick up the platform, using a lateral (or extension) grip. Assuming a right hander: lift the 'tray' over the test unit base (whilst remaining seated) and place on the table to the right of the unit. Return the platform to the left hand side of the unit.

"Start the timer, move the tray from the left to the right hand side of the test unit, as demonstrated, and as quickly as possible. Then stop the timer."

11. Rotate key – Return the platform to the test unit base (red side upwards). Place the key in the lock so it appears vertical. Turn the key to the white mark using a lateral grip.

"Start the timer, rotate the key as demonstrated and as quickly as possible, at least one quarter turn clockwise, to the white mark, and release (at which time the key will spring back), and then stop the timer."

12. Open/close zip – Ensure the zip is closed and lies flat against the back board. Open and close the zip using a lateral, or 2 point tip grip.

"Start the timer, open and then close the zip in as short a time as possible, as demonstrated, and then stop the timer."

13. Rotate screw – Place the screwdriver in the designated area on the red platform (on the right hand side for a right handed subject, or on the left for a

left handed subject). The screw is mounted on a clip, which should be attached to the front of the case. Use the area directly in front of the screwdriver (between the handle and clasp on the case). Ensure the arrow is vertical. Use two hands to guide the screwdriver to the screw, and rotate it 90° clockwise to the mark using one hand only (in a combined power/precision grip, also known as a diagonal volar grip). Reset the task.

"Start the timer, and use the screwdriver to rotate the screw a quarter turn clockwise to, or beyond, the white mark, as demonstrated and as quickly as possible. Once completed, the screwdriver should be replaced on the platform and the timer stopped. Two hands may be used to guide the screwdriver to the screw, but only the appropriate hand should be used in turning the screwdriver."

14. Door handle – Rotate the door handle (using a hook or power grip) until it is fully open, and then release.

"Start the timer, rotate the door handle until it is fully open, and then release, as demonstrated and as quickly as possible. Then stop the timer."

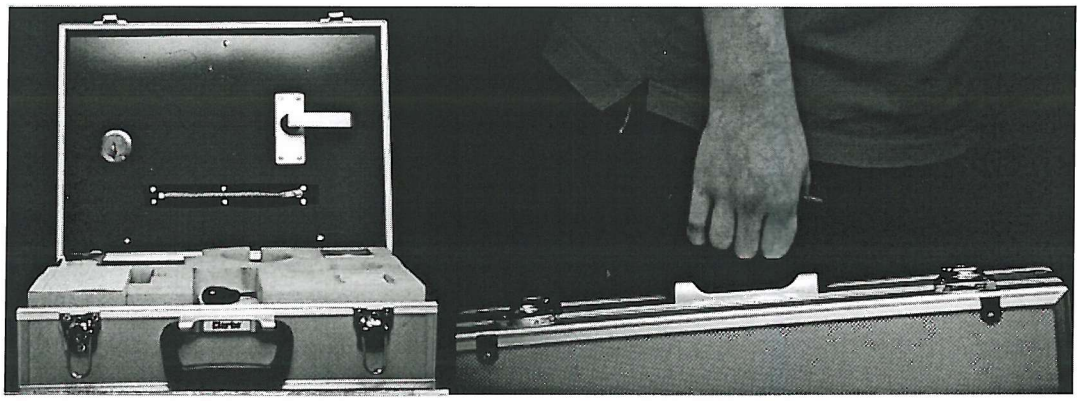

Figure D.4: The SHAP Unit

Southampton Hand Assessment Procedure

Subject Test Data

Subject ID: _____ Date of Assessment: / /

DOB: _____ Hand Under Assessment: L / R (delete as appropriate)

Assessor: _____ Dominant Hand: L / R (delete as appropriate)

Location: _____ Diagnosis:

Other Data Available: ☐ Video

☐ Vicon

Abstract Object Test - Lightweight

		Grips							
Task	Time (secs)	Spherical	Tripod	Power	Lateral	Tip	Extension	Other	Notes
Spherical									
Tripod									
Power									
Lateral									
Tip									
Extension									

Abstract Object Test - Metal

		Grips							
Task	Time (secs)	Spherical	Tripod	Power	Lateral	Tip	Extension	Other	Notes
Spherical									
Tripod									
Power									
Lateral									
Tip									
Extension									

Southampton Hand Assessment Procedure

Activities of Daily Living

		Grips							
Task	Time (secs)	Spherical	Tripod	Power	Lateral	Tip	Extension	Other	Notes
Coins									
Button board									
Cutting									
Page turning									
Jar lid									
Jug pour									
Carton pour									
Full jar									
Empty tin									
Tray									
Key									
Zip									
Screw									
Door handle									

END OF TEST