3D Printed Myoelectric Prosthetic Arm

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Statement of Student Contribution

- I independently carried out background research, design and manufacturing to complete a thesis for the school of Aeronautical, mechanical and mechatronic engineering (AMME). My supervisor was Professor Graham Brooker from the Australian Centre for Field Robotics (ACFR)

- Starting from scratch I designed and manufactured a low cost myoelectric prosthetic arm using computer modelling software and 3D printing facilities provided by the school of AMME

- Using my knowledge of mechanics, electronics and programming I produced several functional prototypes with design improvements between each iteration.

- I wrote an original Thesis

The above represents and accurate summary of the student’s contribution

Mahdi Hussein **Graham Brooker**

Abstract

Throughout the entirety of this thesis it has been approached as a full time endeavour. Manufacturing, assembly and testing was undertaken in the school of AMME during the day, leaving research, design and documentation to the evening.

A 3D printable design for a myoelectric prosthetic arm is presented. The arm is electronically actuated and controlled by a user flexing his/her muscles. The bionic arm presented has the potential to be used by an amputee or person born without a limb. This type of technology does exist although it is expensive and generally not available to people in developing countries.

Rapid growth and advancement of the 3D printing industry allows individuals to become small scale manufacturers. Recent advancements show 3D printed prosthetic arms being attached to victims of war throughout North Africa. Such devices are purely mechanical and significantly less complex than myoelectric devices. Nevertheless, we can see that 3D printed devices have the potential to positively impact people’s lives. 3D printing does have its limitations but growth and development in the field will only lead to improvements over time.

This thesis topic covers a broad range of engineering disciplines. The root of the system is an innovative mechanical design for a 3D printed prosthetic arm. Modern day electronic actuators and circuitry animate the device and allow for sophisticated control schemes. It is hoped that this work will be of value to a diverse audience.

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Definitions

Anthropomorhpic – Human Like

Actuator – Mechanism which converts energy into mechanical motion

Servo – Geared DC electric motor

Degree of Freedom – Unique way in which a feature can move

Acronyms

ACFR – Australian Centre for Field Robotics

AMME – Aeronautical, Mechanical and Mechatronics

EMG – Electro Myography

ABS - Acrylonitrile butadiene styrene

PLA – Polyactic Acid

PIC – Peripheral Interface Controller

SMA – Shape Memory Alloy

ADC – Analogue to Digital Converter

USB – Universal Serial Bus

DOF – Degree of Freedom

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Chapter 1.0 Introduction

It could be argued that the most valuable possession to any human being is their body. Replacing a missing human limb, especially a hand, is a challenging task which makes one truly appreciate the complexity of the human body. For centuries innovators have been trying to replace lost limbs with manmade devices. Several prosthetic devices have been discovered from ancient civilisations around the world demonstrating the ongoing progress of prosthetic technology.

Until recent times the design of prosthetic limbs has progressed relatively slowly. Early innovations such as the wooden leg can be thought of as simple prosthetic devices. History shows that for a long time prostheses have remained passive devices that offer little in terms of control and movement.

Over time materials improved and designs started incorporating hinges and pulley systems. This led to simple mechanical body powered devices such as metal hooks which can open and close as a user bends their elbow for example.

Recent times however have given way to enormous advancements in prosthetic devices. Focus is not only on the physical aspects of a device but also the control and biofeedback systems. Slowly we are approaching an advanced trans-human integration between machine and body. Perhaps sometime in the future prosthetic devices will be faster, stronger and maybe even healthier than our biological limbs.

Throughout the course of this thesis we will explore myoelectric prosthetic arms. It is aimed to design a device which mimics the function of the human arm as best as possible and can be controlled to some extent by muscular contractions.

1.1 Types of Prosthetic Limbs

There are several different categories of prosthetic devices. They are generally grouped by the way in which the device is controlled, including

Passive Prostheses

Passive prosthetics are simple, non-moving devices that aim to restore cosmetic appearance and basic functionality to an amputee. A simple wooden 'pirate' peg leg is an example of a simple passive prosthetic. Prosthetic toes have even been found attached to ancient Egyptian mummy's as show in the image to the right.

Figure 1.1 – **Prosthetic toe made from leather and wood found on an ancient Egyptian Mummy, dating between 950-710 B.C. (courtesy Museum of Egyptian Antiquities in Cairo)**

Mechanical/Body Control Prostheses

Body powered prosthetics are controlled via a harness connected to the user. They are generally a simple device such as a mechanical hook which is linked to elbow/shoulder movement. Although these devices are relatively simple they remain the most popular type of prosthesis today.

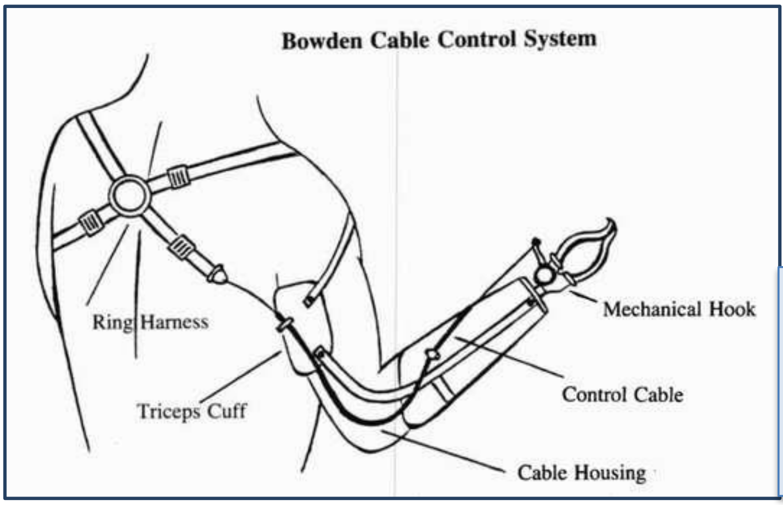


Figure 1.2 – **Body powered hook (courtesy Digital Resource Foundation for the Orthotics and Prosthetics community)**

Myoelectric Controlled Prostheses

Myoelectric prostheses measure electromyography (EMG) signals generated from the contraction of muscles near an amputee's residual limb. These signals are measured through electrodes placed on the surface of the skin or embedded directly into muscles. These signals are then amplified and sent to a microcontroller which analyses this information and controls the internal actuators. Myoelectric devices allow for far greater amounts of control than mechanical devices.

Figure 1.3 – Advanced DARPA myoelectric prosthetic arm

Direct Brain interface

The most cutting edge type of control is a direct brain neural interface. A surgical procedure places electrode arrays on the surface of the brain which are attached to pedestals implanted into the patient's skull. As the patient thinks of motion signals detected on the pedestals are used to control the movement of a robotic arm. This type of technology is still in its infancy but has already demonstrated disabled people controlling bionic devices with their thoughts alone.

Figure 1.4 – Paraplegic woman uses her thoughts to guide a robotic arm

1.2 The Problem

To produce a functional prosthetic arm there are numerous design and manufacturing challenges to overcome. The challenge in this thesis is to create an arm of reasonable complexity and quality which can be further used for research in the field of prosthetics.

It is important for the reader to note that this work encompasses several different engineering fields. Appropriate discussion will be allocated to each field and we shall aim to tie together all areas into a single, robust, functional system.

Below are the major areas that will be addressed throughout this thesis.

Physical Design

The complexity of the mechanical and electrical systems determine how well the device mimics the human arm and the amount of dexterity it is able to offer. The design will aim to be as physically advanced as possible.

Control Scheme

Ideally we would like a prosthesis to be as easy and natural to control as possible. If the user is straining to complete the most basic of tasks, such as grasping an object, then the prosthesis is most likely not beneficial in any practical sense.

Practicality

The device must aim to be useful to an amputee. Whether or not this device is ever actually used by an amputee in uncertain, however the goal is to develop a prosthesis which has the ability to benefit people with missing hands.

Affordable

We shall aim to keep the material cost of the device as low as possible. Modern commercial myoelectric prosthetic arms generally cost about $20 000 - $40 000. The material cost for this design will be under $500.

Each of these challenges specifically the physical design and control system will be discussed thoroughly throughout this thesis.

1.3 Plausible Solutions

The overall goal of a prosthesis is to return as much functionality as possible to a person missing a limb. An ultimate goal would to one day be able to perfectly replace missing limbs.

The aim of this work is to design and build a 3D printed prosthetic arm. A user of this device will be able to control the arm via muscular flexing detected using surface electrodes placed on the skin.

Modern 3D printers allow for detailed mechanical components to be created and assembled relatively fast. Fabricating complex designs using other methods would be far more expensive and would not be possible in such a short period of time.

To actuate the device the first approach is to implement an artificial tendon network. This method is used to actuate various robotic hands which will be discussed in the literature review section. The benefits of this system are that it is a low cost and relatively simple way of controlling fingers.

Another option would be to design a rigid joint linked system to control finger movement. Such a solution is in fact more popular among commercial prosthetic arms; however trying to design a small intricate gear linkage system from weak 3D printed components would not be possible with the available 3D printers.

Details of the systems inner workings shall be discussed appropriately throughout this thesis. There is always more than one solution to any problem and many compromises have been made in design of this device.

1.4 Method of Attack

It is extremely difficult to produce an exceptional design from scratch. Many solutions may seem plausible at first but later lead to unforseen problems. Trying to perfect the first design over several months is not only wasteful of precious time, but also leads to a narrow minded approach.

A far better method of attack is to produce the first prototype as quickly as possible, analyse the system and make improvements. 3D printing allows us to easily manufacture new and improved designs.

Chapter 2 – Literature Review

2.1 Modern Prosthetic Arms

Several arms such as the *Bebionic 3* and *iLimb* are myoelectric controlled robotic arms commercially available to the public. Numerous more prosthetic arms exist in research labs around the world which are usually developed as prototypes to test advanced designs and concepts. Research prosthetics are generally more complex in terms of mechanical design and control and monitoring systems but are inferior to commercial devices in terms of practicality, cost and robustness [1].

2.2 The Human Hand

The human hand comprises of at least 27 bones (depending on the individual) [2], more than 30 individual muscles [3] and over 100 named ligaments, nerves and arteries [4].

Prostheses aim to replicate the functions of the human body and return functionality to persons with missing extremities. No current prosthetics can match the dexterity, flexibility and fluidity of the human hand [1].

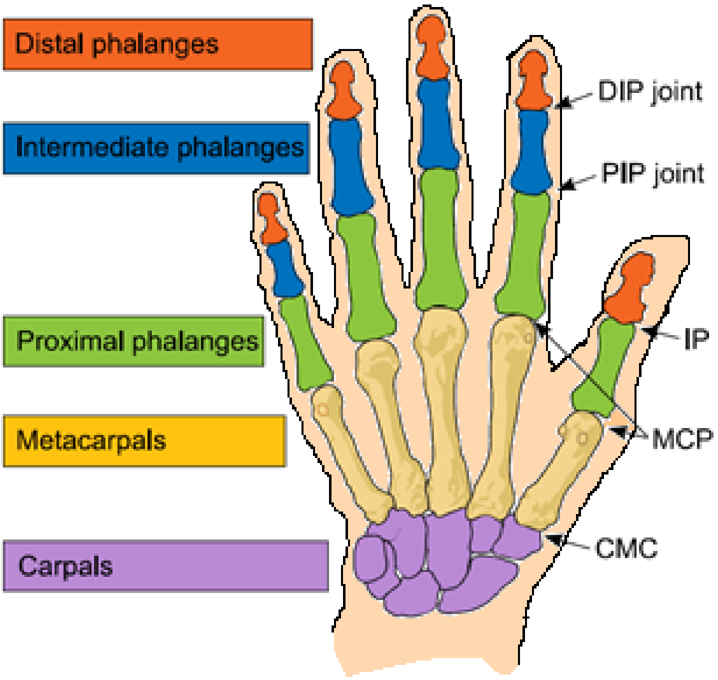


Figure 2.1 – Major Bones in the human hand (image courtesy Dexmart)

Human fingers contain 3 joints, distal, intermediate and proximal (knuckles).

Before any further discussion, let us briefly explain the meaning of a *degree of freedom (DOF)* for a reader with a non-engineering background.

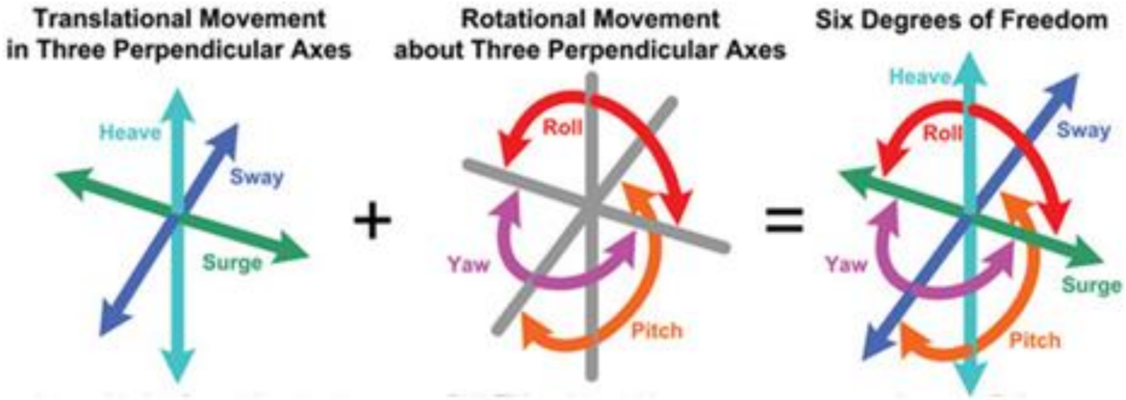


Figure 2.2 – Degrees of freedom at a single point (courtesy Ben Nelson, 2013)

Looking at the above image, imagine a point in space. From this point we can *translate* (move) along 3 different axes, i.e. we can move forward/backward, up/down and left and right. At the same point we can also *rotate* around 3 different axes. The human neck for example has 3 degrees of rotational freedom – we can look left/right, up/down and tilt our head sideways. So in total a single point can have a maximum of 6 degrees of freedom (3 translational, 3 rotational)

The human finger in total has 4 degrees of freedom [7]. Three of these are the rotations of each joint (DIP, PIP, MCP) which combine to control flexion and extension of the finger. The knuckle (MCP joint) also allows for abduction/adduction (wiggling the finger from side to side).

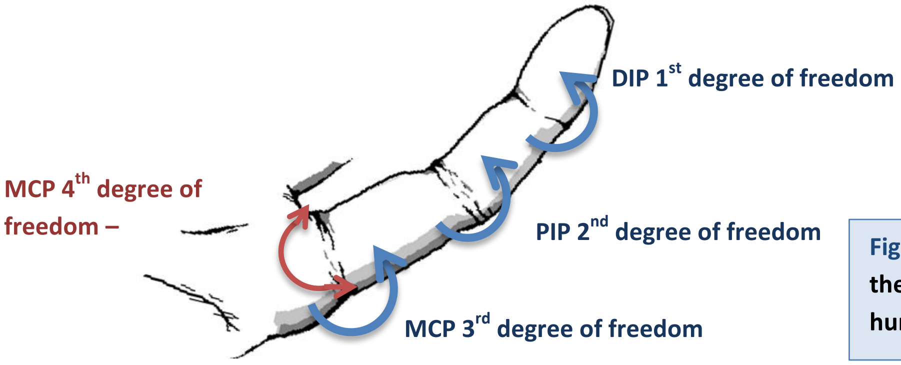


Figure 2.3 – Depiction of the degrees of freedom in a human finger

In the thumb the lower CMC joint also allows for abduction/adduction – which gives 5 DOFs in the thumb [8].

Fingers, and all joints in the human body are actuated (moved) via contraction of muscles and tendons.

2.3 Capabilities of Prosthetic hands

The vast majority of commercial prosthetic fingers are actuated through a *joint linkage* system powered by DC electric motors [1]. Kinematic models for various prosthetic fingers are show below.

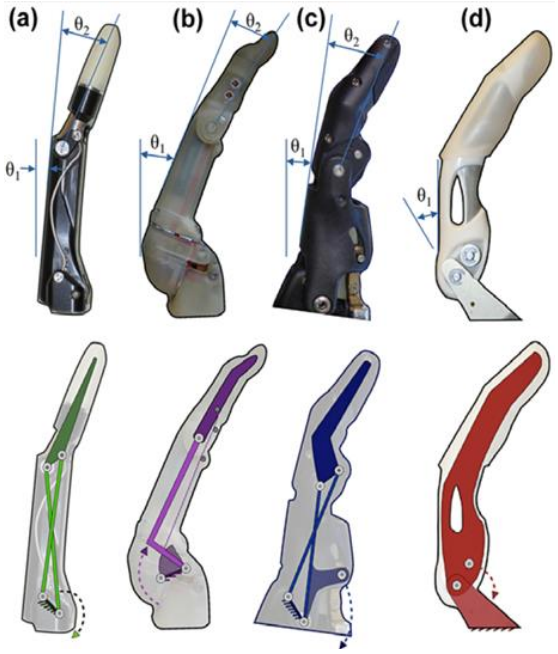


Figure 2.4 – Commercial finger images (top) kinematic models of finger joint coupling mechanism (bottom) (courtesy Yale University)

Each finger incorporates its own mechanism to mechanically couple joints together. Rotating the metacarpal joint (knuckle) simultaneously rotates the higher phalange joint.

The problem with this type of design is that there is no control over individual finger joints. All joints in the finger are controlled through a single actuator which means the entire finger has only a single degree of freedom – these fingers can only open/close in a single way. In reality a human finger has control over individual joints so is capable of flexing in a variety of ways, shown below.

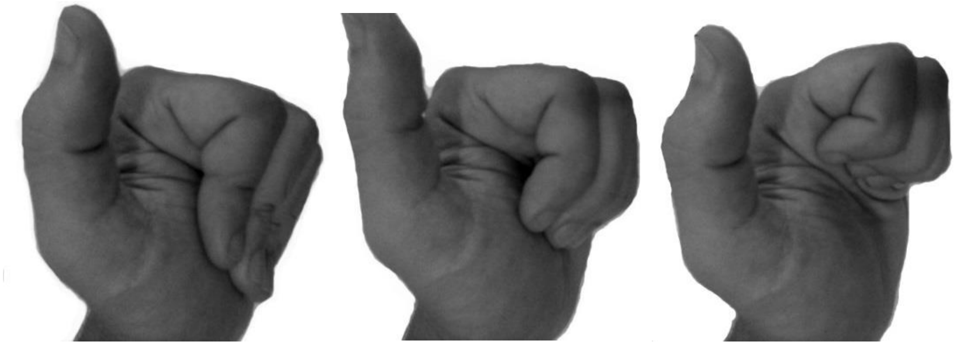


Figure 2.5 – Various flexion arrangements of the human fingers – not possible with modern prosthetic hands

Although commercial prosthetic fingers may have the same number of joints as a human finger, they have fewer degrees of freedom. Usually 1 or 2 compared to 4.

Dexterity arises from the numerous degrees of freedom of the human hand. The fine motor control a person has over their individual finger joints allows for a vast array of intricate tasks to be achieved. In contrast, commercial prostheses are limited to simple tasks partially due to the lack of fine control in the fingers. For example, trying to knit, sew or play a musical instrument like a guitar with a modern commercial prosthetic device would be extremely difficult if not impossible [7].

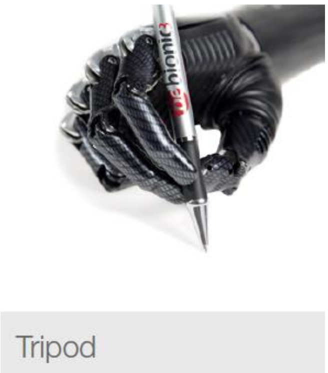
Another critical design point in commercial prosthesis is durability. The average user will wear a myoelectric prosthetic hand in excess of 8 hours per day [9]. Therefore, prosthetic arms for commercial use must be robust, lightweight and packaged into a closed system that can be attached to an amputee. Mechanical complexity determines the degrees of freedom in the system; however, there is usually a trade-off because increasing complexity can lead to an increase of the size of the device and also reduce robustness and durability [1].

The Bebionic 3

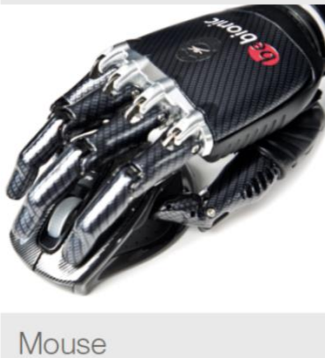
The Bebionic 3 is a world leading commercial myoelectric arm. Like others of its kind, the Bebionic 3 uses a predefined grip system. A user can select from 14 different grip patterns using muscle activity around their upper forearm [5]. The user does not essentially have control of individual finger movements, rather they can select a grip pattern and then use muscle activity to activate the movements of that specific grip. Four of the fourteen grips of the Bebionic 3 are shown below.

Power

Hook



Tripod



Mouse

Figure 2.6 – Various grip patterns of the Bebionic 3 (courtesy RSL Steeper)

The problem with a predefined grip system is that the user cannot finely control finger positions in order to grip a specific object or complete a task. Rather, a user must choose a grip pattern that best suits the job at hand and then actuature that grip pattern.

Furthermore, the user must cycle through a number of grip patterns before they get to their desired choice. For example, unzipping a bag, picking up a heavy object, placing it in that bag and then zipping the bag up could require a number of grip changes. As a result certain simple tasks like this could actually take quite some time to complete and can become tedious and frustrating.

The thumb accounts for arguably 40 percent of human hand use [1]. Thumb design is critical in all prosthetic hands and is more complex than the other fingers.

The Bebionic 3 has an adjustable thumb which can be placed in an opposed or non-opposed position, the difference between these positions can be seen in the images below. This prosthesis cannot directly change the thumbs position. In order to switch between oppopsed and non-opposed positions the user must apply an external force to “click” the thumb into position, e.g. use the other hand to change the prosthetic thumb position.

Figure 2.7 – Opposed and Non-opposed thumb positions of the Bebionic 3 (courtesy RSL Steeper)

Suppose a user of the Bebionic 3 is using a computer mouse. If that user reaches to pick up a bottle of water not only would they have to change the thumb position using their other hand, they would also need to cycle to a new grip state. The user would be better off using their other biological hand to fetch the water bottle in the first place. In such a situation this prosthesis provides no practical benefit.

With all that being said, prosthetic devices are intended to provide more than functional practicality. For many amputees, the loss of an extremity is also accompanied by a significant decrease in confidence and self esteem. A prosthetic arm can help allieviate these issues [18].

> **"When people used to see me they would look at me once, they would see that there was something missing… and the eyebrows would come down. With this, the eyebrows go up, and it's a far more positive thing!"** – Nigel Ackland, Bebionic 3 user. [10]

Physical appearance is an important aspect of prosthetic arms. A survey of myoelectric prosthetic hand users found that the majority of adult users were dissatisfied with their devices cosmetic appearance [9].

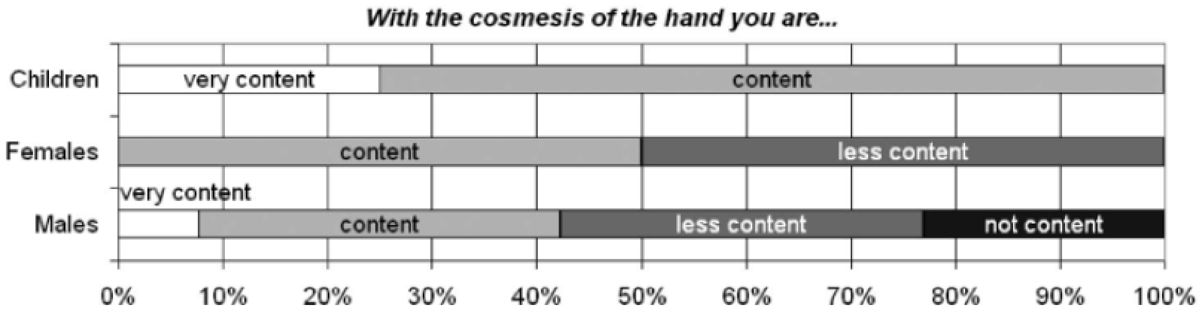


Figure 2.8 – Survey results of myoelectric prostheses users (courtesy International Society of Prosthetic and Orthotics)

The Bebionic 3 is controlled through *electro myography* (EMG) electrodes placed on the surface of the user's skin. The placement of these electrodes depends on the level of amputation but is usually around the upper forearm. Bebalance computer software can be used to adjust a number of settings to enhance the user's control of the device and tune the system to the user's myoelectric signals.

The level of EMG control is dependent on the specific amputation but is generally limited to only a few different commands which cannot be executed simultaneously.

In fact, it is due to this limitation in control that a predefined grip system and a manually adjustable thumb were designed. If a more complex system was designed the user would simply have no way of controlling it.

iLimb digits

One major difficulty with developing prosthetic arms is that amputation can occur at any point along the arm and is unique in every case. The *Bebionic 3* arm previously discussed incorporates electric motors into the palm to actuate the fingers. As a result the *Bebionic* would be of no use to an amputee who has lost several fingers but still has their palm intact.

The *iLimb digits* developed by touch bionics incorporate electric motors directly into the prosthetic fingers [6]. This allows for the palm area to fit into a socket connection attaching the prosthetic fingers to the hand. The image below shows possible amputations which would be suitable for use of the *iLimb digits*.

Figure 2.9 – Examples of amputations suitable for the iLimb Digits (courtesy Touch Bionics)

A custom socket is designed to fit around the remaining area of the users palm and as many digits as necessary can be added to the system.

Like the Bebionic 3, the iLimb digits are controlled through EMG electrodes, which are placed over muscle regions in the palm. A small package must be worn around the user's wrist which contains the battery and controller for the system

A disadvantage of this system is that relatively small motors have to be used to be able to fit inside the fingers. This leads to digits which move slower and are weaker than those in other commercial prostheses.

Figure 2.10 – iLimb Digits attached to an amputee four fingers and the thumb but palm still intact (courtesy Touch Bionics)

Vanderbilt Hand

Research prosthetic devices are designed to test advanced mechanical designs and sophisticated control methods. Most research hands require an external power system, making them non-suitable as an attachment to an amputee [1].

Many research arms like the Vanderbilt and Bologna Universities anthropomorphic arms experiment with artificial tendon designs to drive finger movements [11], [12] – as opposed to a mechanical linkage system in commercial devices.

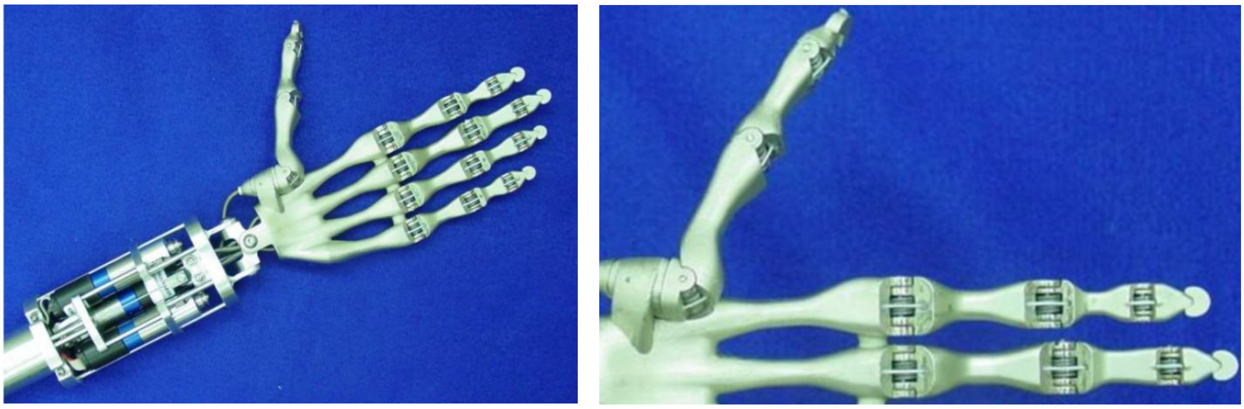


Figure 2.11 – Artificial tendon driven anthropomorphic hand (courtesy Vanderbilt University)

The above images show the tendons (white cables) running through the fingers and thumb. Brushed DC motors drive a pulley system which tensions the tendons. This tension results in all three finger joints closing simultaneously.

In order to open the fingers, springs have been implemented into each joint. When tension is released in the tendons these springs return the finger to its initial open position. [9]

The image to the right shows the coiled steel springs incorporated into each joint.

By incorporating springs, the energy stored in the joints during the closing phase is used to perform the opening phase without the need of further actuation [10].

Figure 2.12 – Integrated springs (courtesy Vanderbilt University)

The Bologna University (UB) hand uses a similar tendon/spring design to actuate fingers [12].

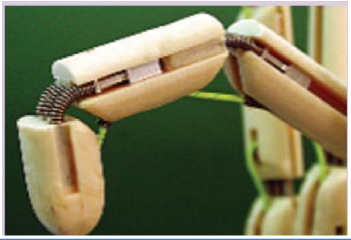


Figure 2.13 – Close wound linear spring (courtesy Bologna University)

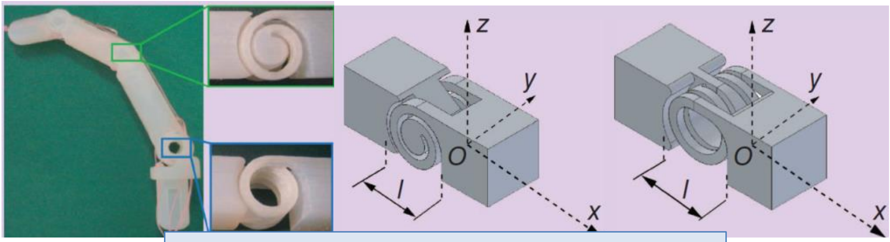


Figure 2.14 – 3D printed integrated spring design (courtesy Bologna University)

We can see above that linear springs can be used to return fingers to an open position or the spring mechanism can be directly incorporated into the structure.

The *UB hand IV* has also experimented with a twisted string actuation system. Essentially, a DC motor is used to twist two tendons together. This shortens the length of the tendons which generates tension and closes the fingers.

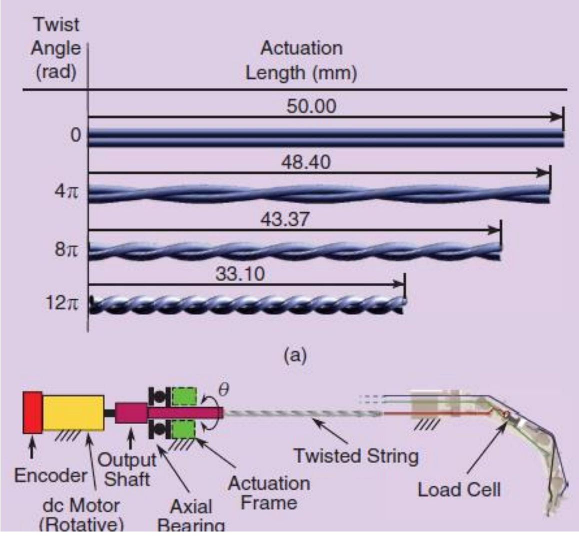


Figure 2.15 – Twisted tendon actuation method (courtesy Bologna University)

The big advantage of this method lies in the fact that there is a direct transformation of rotational motion in the motors to linear motion in the tendons – no intermediate pulley system is required to move the fingers. Another advantage is that large forces can be exerted by the fingers. Essentially, the tendon pair can continue to be twisted, increasing the force exerted by the finger – until of course there is a mechanical failure somewhere.

However, this can be a slow way of actuating fingers since many rotations are required in order to fully close a finger. In their most recent public video it seems the UB hand researchers have switched to using Servo motors to control fingers for improved speed.

The *UB hand IV* achieves a high number of DOF by using a large tendon network. The commercially available *Shadow Hand* also utilises a large tendon network and achieves 20 DOF's [13] [14]. The Shadow Hand however uses pneumatic air muscles to tension its artificial tendons. In both cases a large area is required to drive all the tendons which make the systems too bulky to be attachable to an amputee.

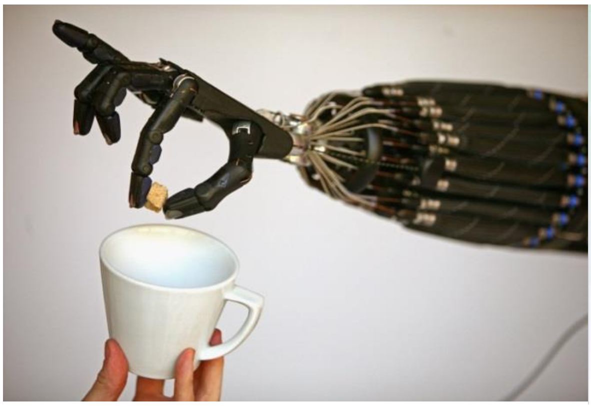


Figure 2.16 – Dextrous Shadow hand (courtesy shadow robotics)

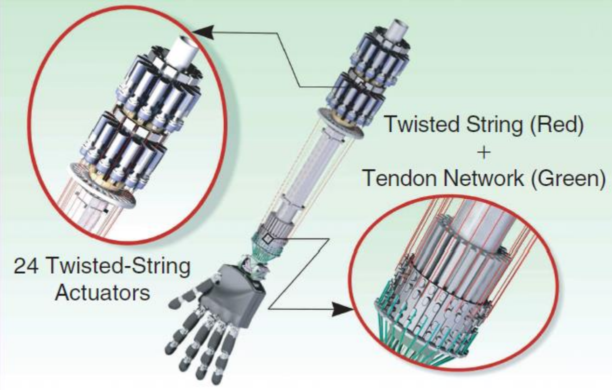


Figure 2.17 – UB hand IV tendon network (courtesy Bologna University)

As can be seen both systems require a large amount of space for all the actuators.

One major advantage that artificial tendon driven prosthetic hands have over earlier discussed mechanically joint linked system is what is known as joint conformity [1]. Joint conformity allows fingers to adapt to the shape of the object they are grasping. Joint linked fingers, like the Bebionic, are stiff and rigid when they close; tendon systems however have some flexibility at the joints, illustrated below.

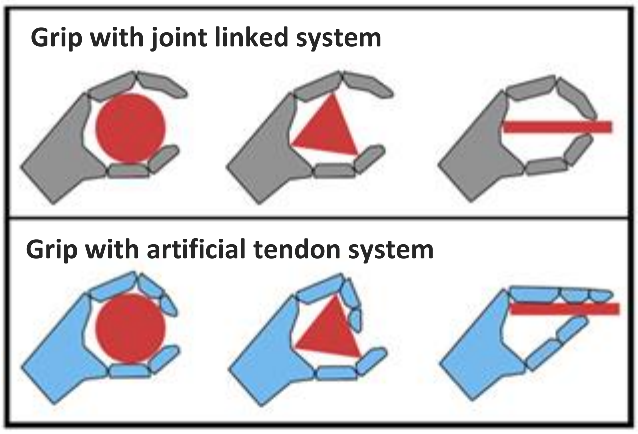


Figure 2.18 – Depiction of joint conformity (courtesy Open Hand Project)

Tendon systems allow for compliant joints which conform to an objects shape

22 Degree of Freedom APL Hand

One of the most advanced modern prosthetic arms is the 22 degree of freedom *Intrinsic Hand* developed at the *John Hopkins Applied Physics Laboratory* [15]. This hand has been developed through DARPA initiative and funding and has unmatched mechanical dexterity. To achieve such fine control designers incorporated a total of 15 miniature DC motors directly in the fingers, palm and wrist.

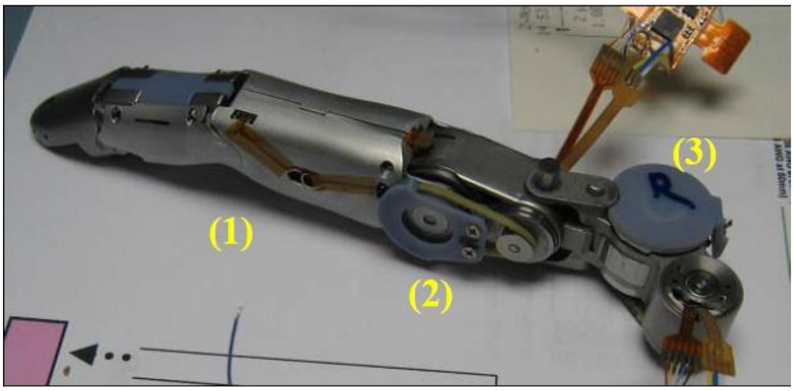


Figure 2.19 – High DOF finger module (courtesy John Hopkins APL)

Furthermore, this device is designed to fit to a 50th percentile female arm making it truly exceptional in terms of its complexity and size. The Intrinsic Hand is able to replicate almost every movement of the biological human hand.

Using standard EMG sensing techniques there is no way of obtaining enough control for a user to practically use all the degrees of freedom of this device. However, DARPA is further funding the development of a prosthesis/brain neural interface to connect the user's nervous system directly to inputs in the arm.

Figure 2.20 – APL intrinsic hand being used by an amputee (courtesy John Hopkins APL)

Shape Memory Alloy Actuation

Another interesting method of actuating prosthetic hands is by using shape memory alloys (SMA's). SMA's return to a predefined shape or size when subject to the appropriate thermal procedure (heating or cooling). The design below from Vanderbilt University uses SMA springs which contract when heated, therefore tensioning tendons which close the fingers [16].

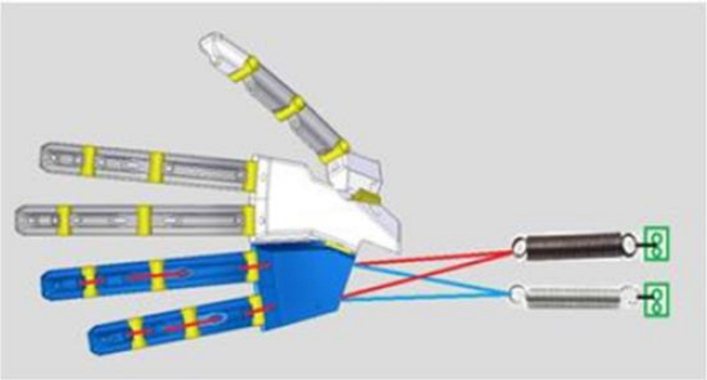


Figure 2.21 – SMA's used to pull tendons which close/open ring and pinky fingers (courtesy Vanderbilt University)

In this design an electrical current is passed through the SMA's to heat them up and initiate contraction. There are two major problems with this method though

- Designing a precise control system for SMA actuators is very complex. Contraction rate of the SMA, spring response and varying weights of objects grasped all contribute to this problem [16]
- Heating the SMA and then waiting for it to cool can take quite some which makes closing and opening a finger a relatively slow process

At the University of Utah researchers have used hot and cold water reservoirs to speed up the SMA actuation process [17]. Fingers in this system flex and extend in a reasonable amount of time but there is a very large area required for heating and cooling apparatus to drive this system and reservoirs of water must be present.

As this technology is developed further SMA's may become a viable actuation option for prosthetic hands. A great benefit it that they are noise free and low weight.

One final note about research prosthetic hands is that there is a vast number of actuation methods not discussed, pneumatic, hydraulic and so on [19]. In almost every situation the apparatus required to drive the systems are bulky and take up far too make space to be able to be fitted to an amputee.

3D Printed Bionic Arms

Over the past couple of years developing 3D printed bionic limbs has become quite popular. InMoov is an independently run project developing a life like humanoid robot from 3D printing technology. The entire project is open source and provides great mechanical design insight into producing 3D printed robotic body parts [20].

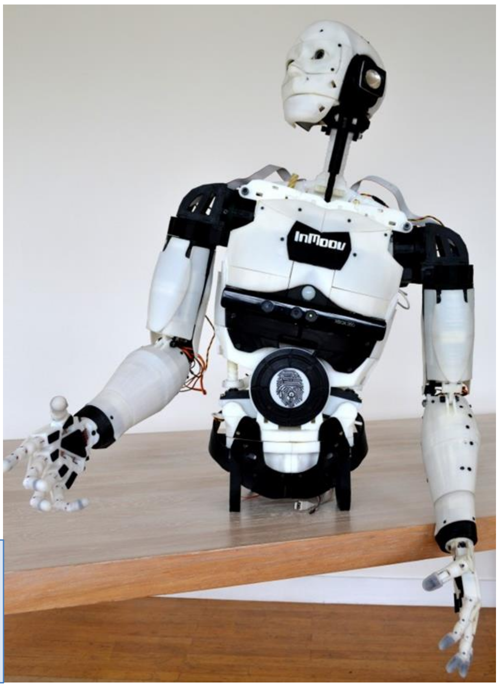


Figure 2.22 – 3D printed humanoid (Courtesy InMoov Project)

The open source nature of this project allows the public to access computer aided designs and follow step by step guides on how to 3D print and assemble this system. The InMoov fingers are controlled by tendons actuated through servo motors placed in the forearm.

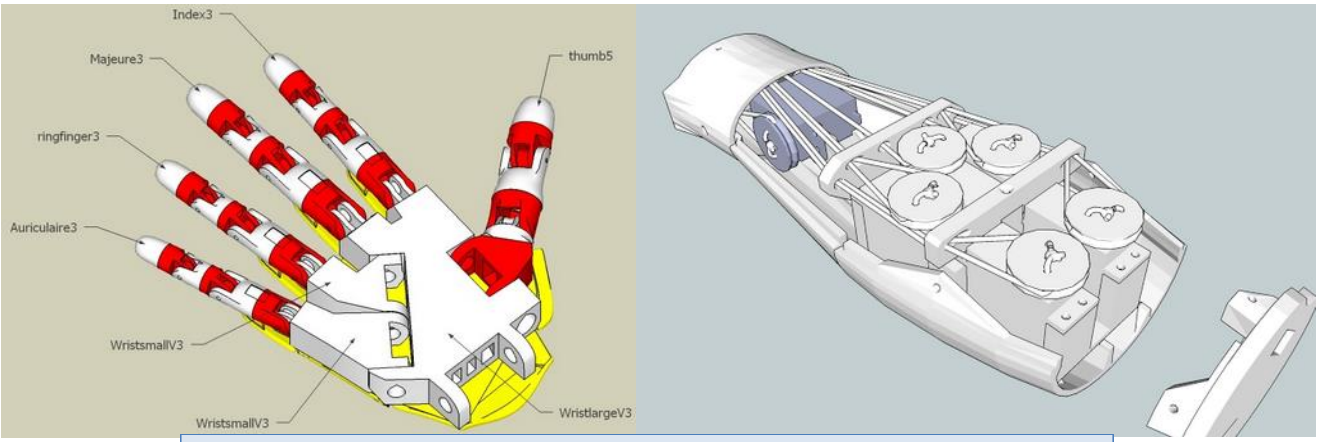


Figure 2.23 – Design of 3D printed hand and forearm (courtesy InMoov Project)

The InMoov fingers (including the thumb) only have a single degree of freedom which limits the dexterity of the hand. However, this is a simple solution which has a great advantage over other anthropomorphic hands which is that this design is low cost and easily manufactured through 3D printing.

The problem with the InMoov hand is that the Servos take up the entire forearm leaving no room for it to be attached to a stump between the elbow and wrist.

2.4 Connection to the Body

Socket Design

The first goal of prosthetic management is protection of the residual limb since 90% of upper limb amputations arise from physical trauma [18]. Prolonged pressure exerted on damaged soft tissue areas can lead to a significant compromise of the remaining appendage. Problems an amputee may experience include pain, swelling, blisters, skin irritations, edema, and a restriction of blood flow [19].

Sockets must be designed in a manner that is safe for the user, comfortable, hygienic and distributes the weight of the prosthesis in an optimal manner.

The most common way of fitting a prosthetic is by creating a custom socket that fits around the amputees stump. This socket can either be self-suspending, suction fitted or secured by harnesses to the user. Comfort and load distribution can be increased by providing some form of padding such as a prosthetic sock, inflatable air pockets or by reducing the density and stiffness at a sensitive region [17, 19].

Several lost cost prosthetic arms use some form of thermo softening plastic to create a custom socket. A plastic sheet is heated and formed around an amputees stump. A prosthetic sock can be worn to create a snug fit between the user and the device.

Ossiointegration

Ossiointegration is the process of permanently integrating a non-biological component with a human bone. In prosthetic devices a titanium stud is screwed into a long bone in the arm or leg at the amputation site. Over time the titanium and bone fuse together to create a firm anchor point for the prosthetic to be attached to [19]. Ossiointegration is not an overly common practise, however it does offer several benefits including

- A strong, sturdy anchor point
- No need for soft tissue to bear the weight of the prosthesis
- No skin/blood flow issues induced by a socket
- No fitting problems due to a gain or loss in weight

2.5 User Control

Electromyography sensing

Myoelectric signals are electrical pulses within the body produced by contracting muscles. Surface electrodes on the user's skin can detect these small signals and in the case of prosthetics be used to control the device [23].

The problem with surface EMG techniques is that there is a lot of cross talk between muscle signals. Because muscles groups, especially in the arm, are physically close together, it is difficult to distinguish exactly which muscle is generating the measured signal via the surface electrodes. One way of alleviating this problem is through target muscle reinnervation (TMR). TMR is a surgical procedure which takes residual nerve endings from an amputation site and spreads them across an alternative intact muscle group [19]. Because the nerves and corresponding muscle contractions are now spread over a larger area it is easier to decipher individual signals.

Surface EMG electrodes require a clean and secure connection to the user's skin which makes measurements susceptible to sweating and electrode displacement [22].

Implanted myoelectric sensors

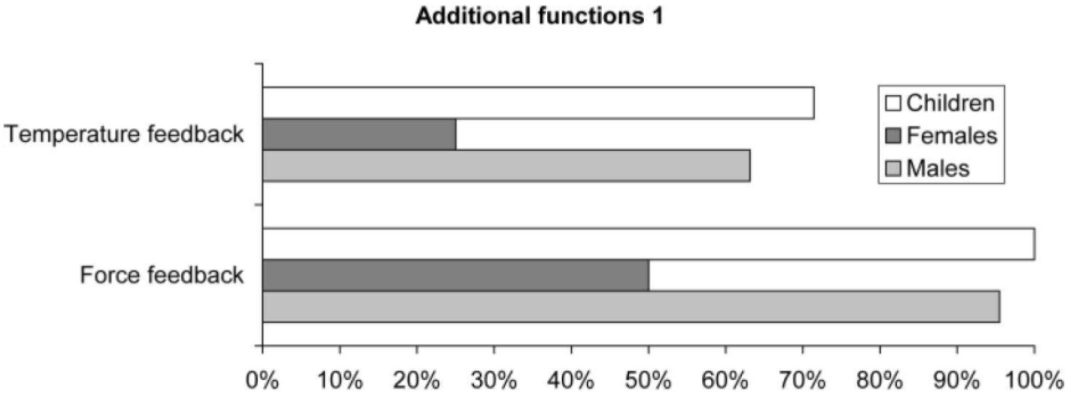
Implanted myoelectric sensors (IMES) are inserted directly into targeted muscles through a surgical procedure. Using IMES for EMG control allows for relatively cross-talk free signals. This means muscle signals are more distinguishable and deeper muscles can also be used for control [20]. As a result this method allows a user to have more control over their prosthetic device. Users of IMES systems have been able to independently control their prosthesis' thumb, fingers and wrist rotation simultaneously [25].

Brain Controlled Interface

The most cutting edge form of control is a direct brain interface be researched and developed by DARPA [27]. Two sensor arrays placed on the surface of the brain through a surgical procedure are wired to two pedestals embedded in the skull. A patient using this technology has been able to control a robotic arm in 3 dimensional space as well as open and close the hand – all through the power of her thoughts alone.

2.6 Sensory Feedback

One major problem of many prosthetic devices is that they lack feedback to the user. The sense of touch is a natural feedback mechanism which allows a person to make physical adjustments both consciously and subconsciously. With no form of feedback a user must rely entirely on vision to determine the position and force of their prostheses. Survey results show almost all users of myoelectric prostheses want some form of modern sensory feedback [9]. Modern prostheses can provide feedback by stimulating senses in some area of the body. Vibration motors and temperature pads placed on the surface of the skin provide rudimentary sensory substitution. The majority of myoelectric prosthesis users found these forms of feedback to be useful.



3. Percentage of individuals that wanted force feedback or temperature feedback in a prosthetic hand.

Figure 2.24 – Results of a survey on myoelectric prosthetic arm users (courtesy International Society of Prosthetic and Orthotics)

Another emerging form of feedback is through electro-tactile arrays. By electrically stimulating arrays of electrodes a false sense of touch can be created [26]. For example, an electro-tactile array on a smart phone screen can create artificial surface textures of wood and stone.

Electro-tactile pads could be fitted around the stump of an amputee generating artificial senses of touch. The stimulating signals could be controlled via pressure sensors connected to fingers tips. Essentially this would shift the sense of fingertip touches to the stump.

Chapter 3 – Design & Manufacturing

3.1 Mechanical Design

To create a useful myoelectric prosthesis it is necessary to have a well-designed mechanical system which mimics the functionality of the human arm as best as possible. Among many other things mechanical design involves how joints are actuated and the types of forces present in the system. The bionic arm design presented in this section can be entirely manufactured with a 3D printer and basic tools.

3.11 Early Ideas & Concept Drawings

After researching several actuation methods for prosthetic arms an artificial tendon design was chosen. As seen in the literature reviews of the *Shadow Hand*, *UB Hand and InMoov*; artificial tendons are a viable way of actuating bionic hands. The tendons can be any high strength line which does not stretch when tensioned. These lines connect to the fingers and are tensioned by motors in the forearm. Pulling on the tendons cause the fingers to open and close.

The electric motors driving these tendons must be completely housed inside the device in order to make it portable and attachable to an amputee. Ideally we would like these motors placed as closely to the fingers as possible, however due to their relatively large size we cannot house the motors used inside the palm section. Instead the motors housed within the forearm.

Shown on the following page are some early concept sketches outlining several key starting ideas. The choice to use standard servo motors to drive the tendons was made very early. Servo Motors are geared DC electric motors which can be controlled to rotate to specific angular positions.

The final model still incorporates several of these early design features however many key design points have changed such as servo positioning and the use of guide pulleys.

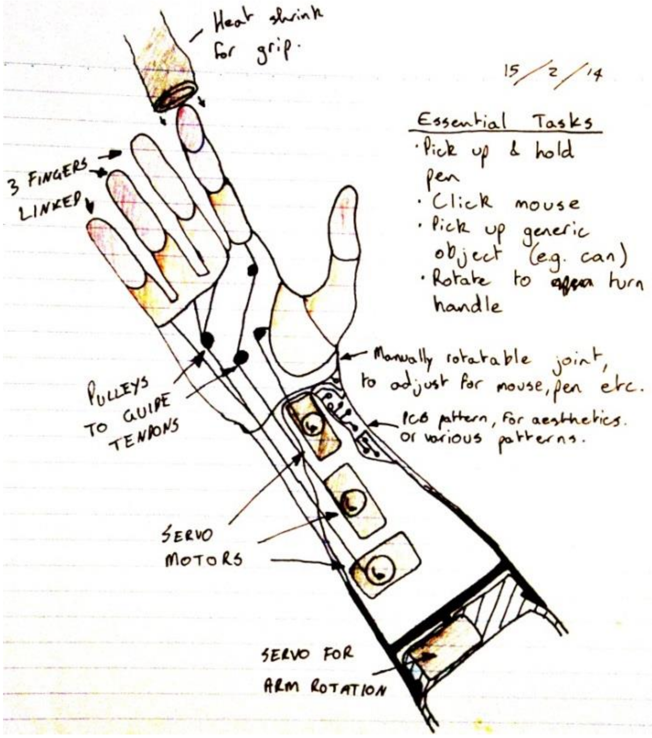


Figure 3.1 – Early design concept drawing

The images below outline some more developed early designs. The image to the lower left shows the assembly of individual mechanical components of the hand. The image to the right shows some rough calculations for a geared wrist rotation design. Rough scketches like these are key to developing a solid design foundation.

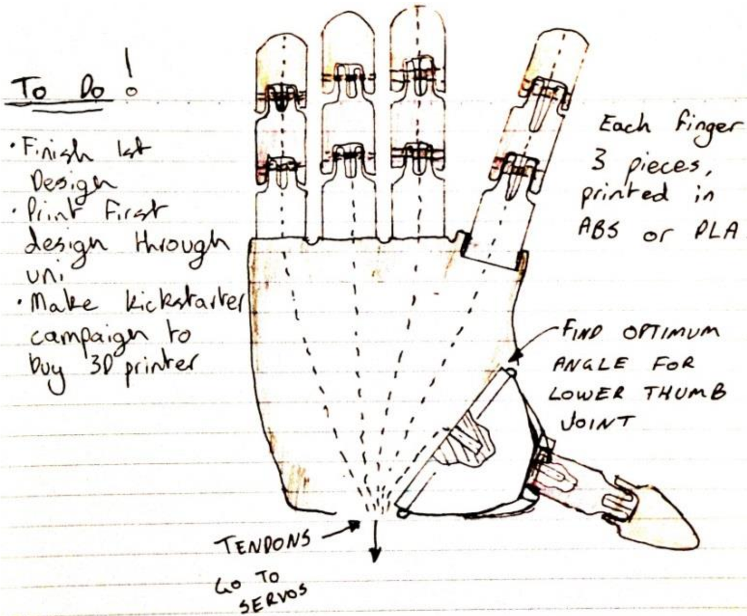


Figure 3.2 – Semi detailed drawing of the hand

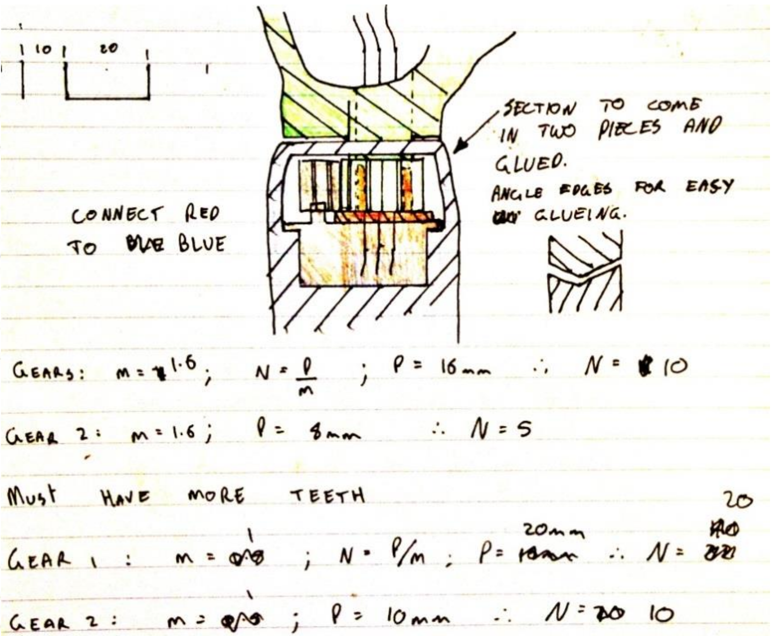


Figure 3.3 – Early design of wrist rotation mechanism

3.12 Ergonomics

Ergonomics is the interaction between humans and machines. The field of prosthetics is interesting as it deals with ergonomics between prosthetics and amputees such – physical attachment to the body and sensory feedback. Ergonomics must also be considered for the interaction between a person’s prosthesis and other people. An ideal prosthesis is physically comfortable for the amputee to wear, easy and natural to control, provides useful sensory feedback and interacts well with its environment.

The dimensions of a large male hand have been used for design proportions. A universal goal in prosthetic design is to achieve shapes and sizes that match an average female physique. It is much easier to scale a design up in size rather than shrink it down to fit a smaller person.

Scalability has been kept in mind throughout the design process. Components can be easily rescaled in computer modelling software and printed relatively fast. This allows for various size prototypes to be developed with ease.

3.13 Computer Assisted Design

Solidworks is a computer design software package made for modelling solid mechanical components and assemblies. Solidworks is a popular tool in the engineering industry and has been used extensively in designing and analysing mechanical components.

Fingers

Each finger consists of three individual printed components linked together with polypropylene pins. The artificial tendon loops around the inside tip of the finger to create a tendon locking point. This tendon runs through channels inside the finger to form an enclosed loop. When the tendon is pulled rotational forces are applied to all the joints and the finger curls up.

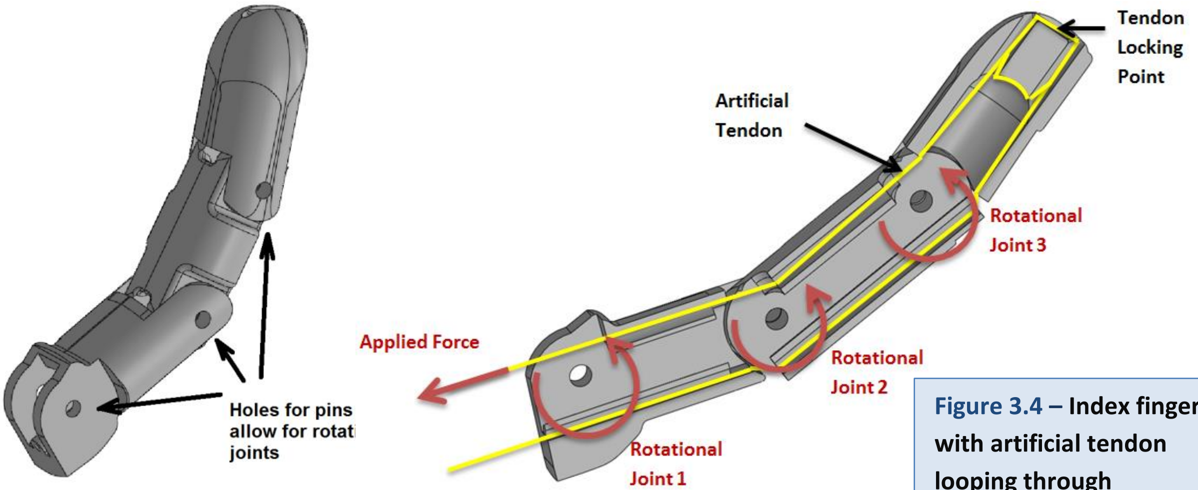


Figure 3.4 – Index finger with artificial tendon looping through

The tendon locking point is essential so that when the tendon is tensioned it pulls the tip of the finger and causes all joints to rotate. If the tendon did not lock it would just slip when tensioned and the finger would not move. To open the finger from a closed position tension is applied to the other end of the tendon.

High quality braided fishing line has been used as it offers minimal stretch when tensioned. Nylon fishing line would stretch over time leading to a loss of tension which would negatively affect finger movements. Tendons in the biological human hand work in a similar way, however there are far more biological tendons attached to different bones – allowing for more precise control of the fingers.

Thumb

The thumb has also been designed in a similar fashion. Most commercial and research prosthetic hands aim to provide at least two degrees of freedom in the thumb. This thumb however only provides a single degree of freedom – it can only open/close in a single way.

Guide holes have been incorporated into the design of the fingers and thumb to optimise tendon orientation and prevent the tendon lines from getting caught on a sharp edge.

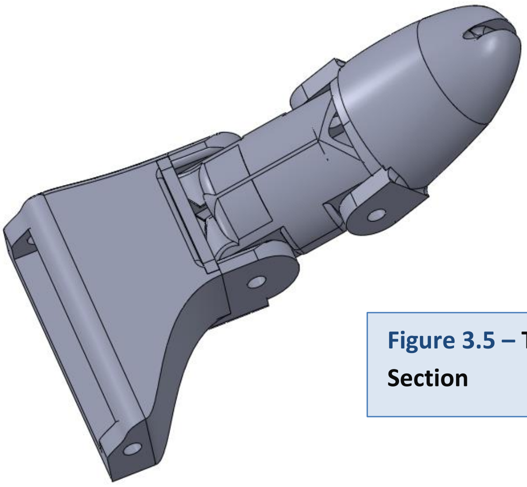


Figure 3.5 – Thumb Section

Palm

Each finger connects to the palm by polypropylene pins. The bottom of the palm incorporates part of the wrist rotation mechanism discussed on the following page.

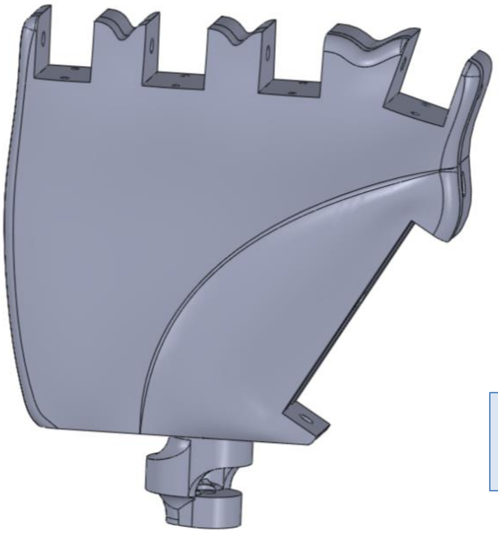
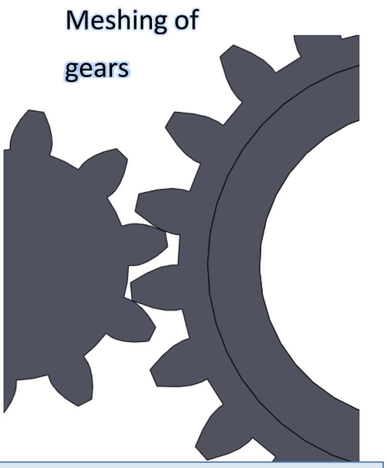


Figure 3.6 – Palm Section

Wrist

Initially a gear driven system was implemented to control wrist rotation. A small gear was 3D printed and pressed onto a servo shaft which would then drive a larger gear connected to the palm section. Solidworks physical dynamics tools were used to test how well the gears meshed together before any components were printed.



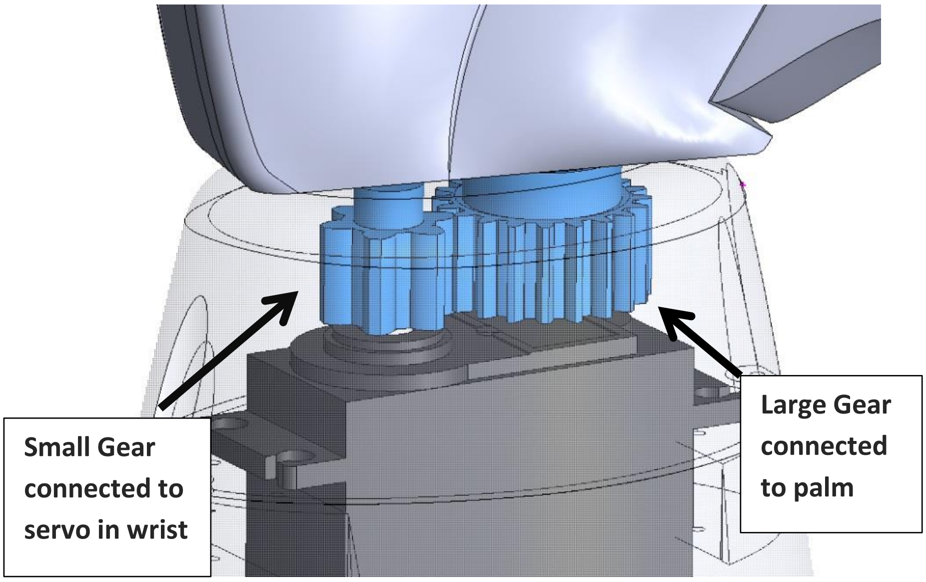


Figure 3.7 – Early wrist rotation mechanism

Unfortunately the gears did not rotate in a smooth manner. This is because 3D printers produce components which are slightly warped from their original design – this is especially noticeable when printing in ABS plastic. The teeth or the gears warped ever so slightly which resulted in poor meshing and rotation of the gears.

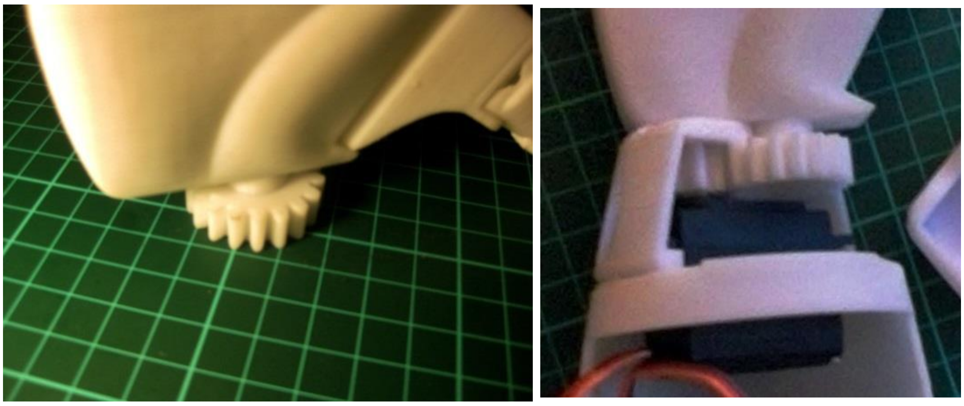
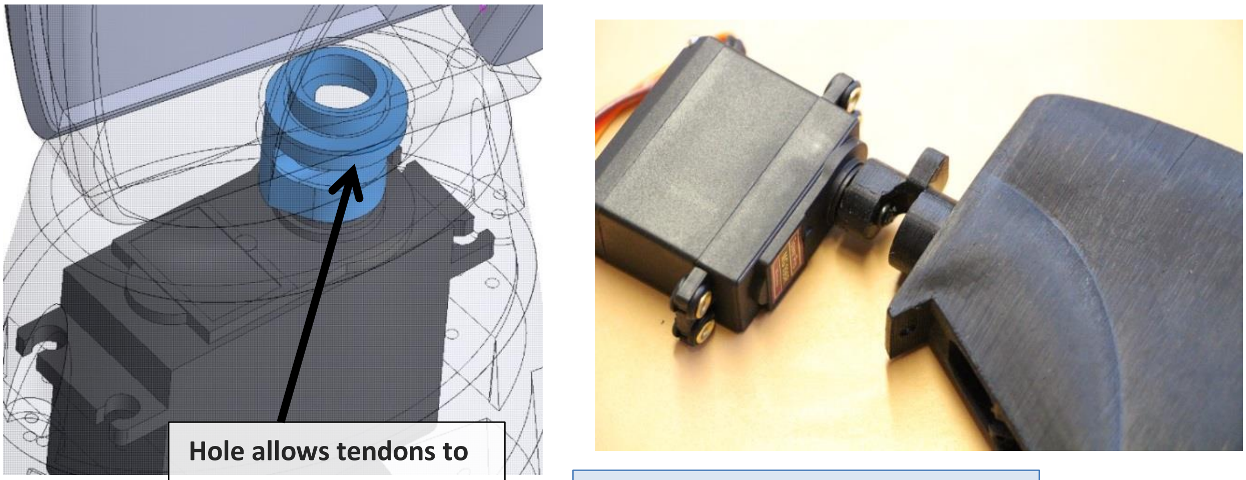


Figure 3.8 – Printed wrist rotation mechanism

The design could have been adjusted for this warping by increasing the mechanical tolerance to allow for bigger gaps between gear teeth. However, for further reason discussed on the next page, the entire gear system was scratched.

The problem with the gear system arose from the fact that standard Servo motors can only rotate to a maximum of ± 90 degrees. The driving gear mounted on the servo was 2.5 times smaller than the driven gear on the hand. This meant that by using a standard servo to drive the small gear only ± 36 degrees of rotation about the wrist would have been achievable. This was going to be addressed by using a continuous rotation servo – which can rotate indefinitely. However, continuous rotation servos have no angular position feedback. In order to rotate the hand to a specific position a mechanical block would have to be designed or less accurate open loop control methods would have needed to be used.

To avoid these problems it was decided to drop the gear system and instead press fit the palm section directly onto the shaft of the servo in the forearm as shown below.



Hole allows tendons to run from fingers down to Servos in the forearm

Figure 3.9 – Improved wrist rotation mechanism

A passage through the pivot point of this joint allows the tendons to pass from the fingers through to the forearm. This allows for ± 90 degrees of rotation about the wrist and eliminates the problem of angular position control.

The challenge with this design is providing enough strength. A large opening around the base cylinder had to be left for the tendons to pass through as the wrist rotates through 180^{O}. This big opening concentrates stress around the small cross sectional area near the base of the palm.

The images on the previous page show a wrist model which snapped during testing as it was too weak. The latest model operates on the exact same principle but includes far more material around the base to increase the component strength and prevent fractures occurring at this point. A small horizontal ridge around the outside of the small cylinder helps transfer lateral loads to the outer forearm shell.

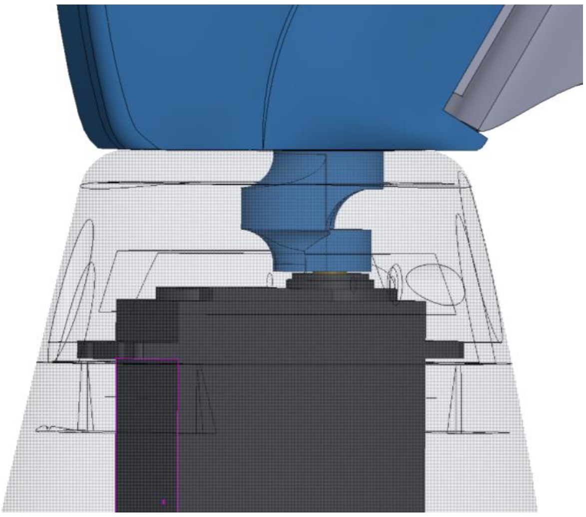


Figure 3.10 – Latest wrist rotation model with improved strength and rigidity

The tendons bottle neck as they pass through the small wrist opening to the motors in the forearm. Wrist rotation of 180^O causes the tendons to twist and overlap which in undesirable but cannot be avoided. This type of design would not allow for the wrist to continuously rotate around as is would cause the tendons to twist around and become entangled.

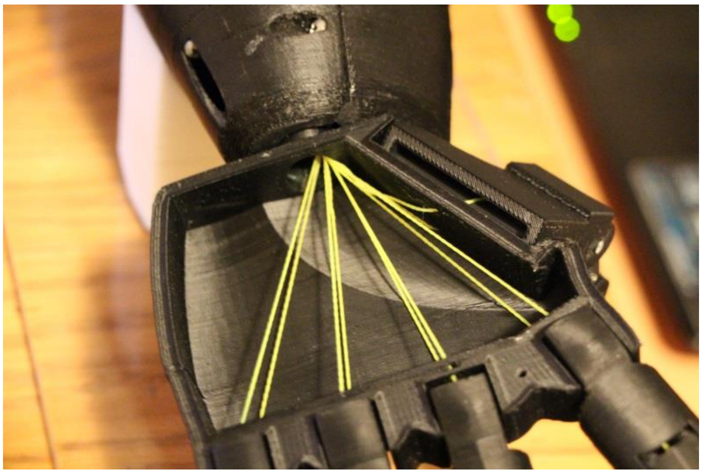


Figure 3.11 – Depiction of tendon bottle necking through wrist section

Drive System

The tendons wrap around custom 3D printed servo horns creating a closed loop shown below. As the servo motor rotates one way it pulls on the tendon and closes the finger. To open the finger the motor is rotated in the opposite direction.

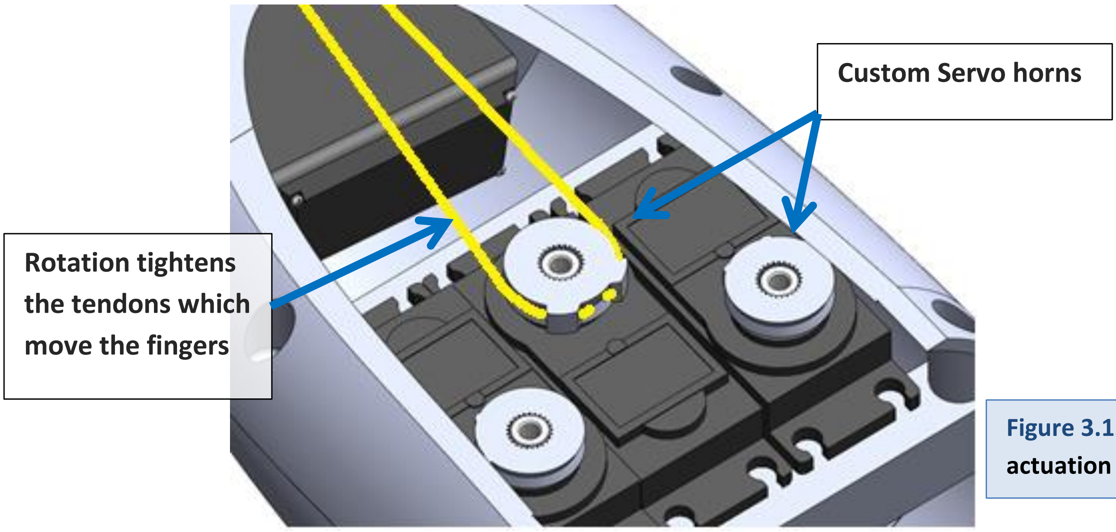


Figure 3.12 – Tendon actuation system

The image below shows the artificial tendon drive for the index finger. All other tendons have been omitted for clarity. The thumb, index and middle fingers are connected to individual servo motors. Because the interior space of the arm is limited the ring and pinky fingers have both been tied to the same servo, meaning they open and close in tandem.

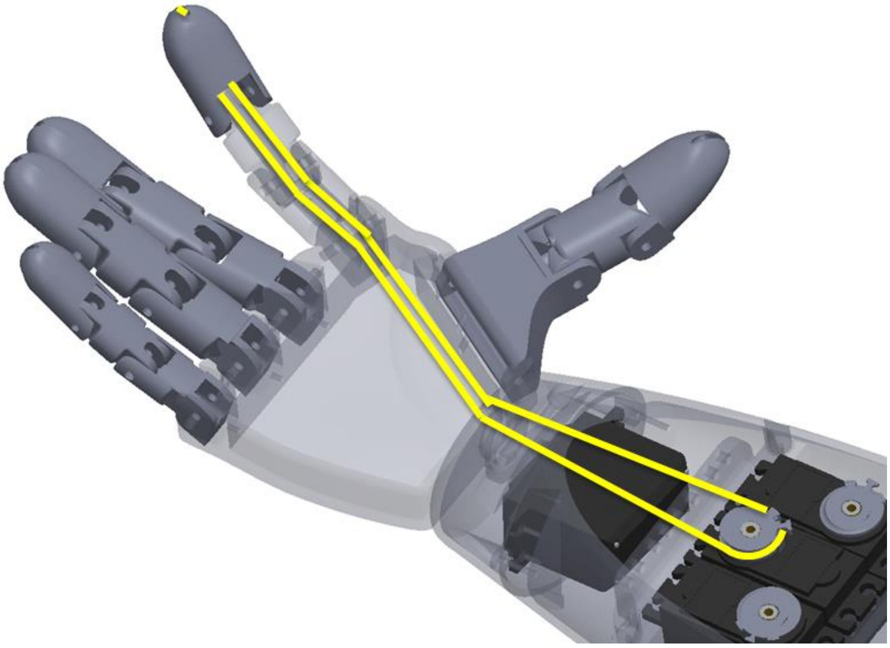


Figure 3.13 – Depiction of the tendon drive for the index finger

Modularity

Amputation can occur anywhere along the arm and is unique in every case. An ideal design facilitates connection to a stump located anywhere along the arm. The hand and wrist section could now be fitted to a person amputated along their forearm.

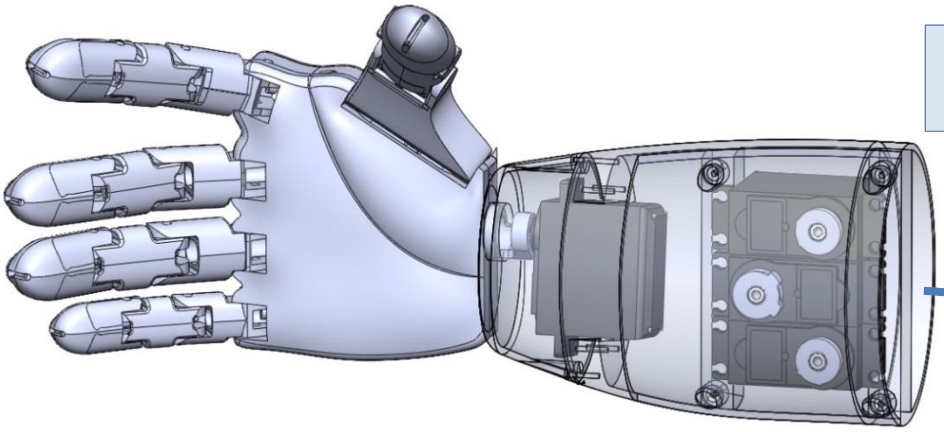


Figure 3.14 – **Assembly of Hand and modular forearm**

Connection to socket or elbow module

Figure 3.15 – **First prototype of the arm, used old wrist rotation method**

This hand/wrist module can now also be connected to an elbow section which provides a solution for an above elbow amputation.

Methods of mechanically fitting this prosthetic hand to an amputee have only briefly been touched on in this thesis and will be discussed further in the future work section. In order to design a socket connection a mould or CAD rendering of an amputee’s stump would be required. Thermoforming plastic could be used to mould around the stump and some form of a harness or straps would most likely be required to produce a stable connection.

Forearm

Although the forearm section contains no moving components its design is still somewhat challenging as this section needs to house five servo motors, lithium polymer (LiPo) battery and allow for assembly.

After the complete forearm section was designed it had to be split into separate components which could then be assembled with screws. If the forearm was 3D printed as a single large component then there would be no way of assembling the motors and tendons inside the arm.

3D printed ABS plastic is relatively weak and can easily be split by the turning a screw. To minimise the chance of a crack occurring guides holes for the screws have been incorporated into the design and care has been taken to ensure there is enough material to firmly support the screw.

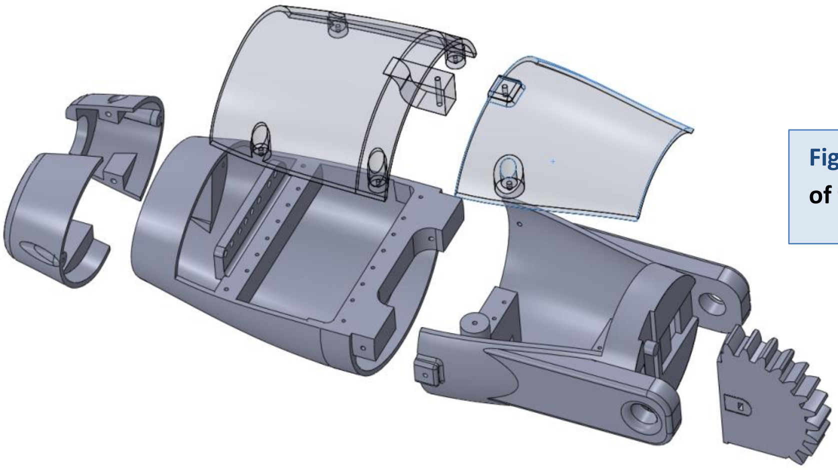


Figure 3.16 – Exploded view of forearm assembly

The two large sections of the forearm could be 3D printed as a single piece without affecting the assembly of the device. However, the UP 2 is simply not large enough to print an object of this size. The two pieces were printed separately and then secured together with super glue.

The Gear section seen in the image above is part of the elbow rotation mechanism. It is crucial that the quality of this gear be as dimensionally accurate as possible. Printing the gear separately produces a printed gear of higher quality. This gear was then pressed into a groove in the forearm and glued into position.

Elbow

The elbow actuator must always move the weight of the forearm on top of any additional load. The minimum required torque to lift the forearm with no load is roughly 13.5kg-cm (see calculations). The TowerPro servo being used provides 10kg-cm of torque. In order to lift the arm using a single servo a gear system had to be implemented. Gears allow us to generate more torque (turning force) at the cost of speed. A small gear pressed onto the bicep servo drives a larger gear section connected to the forearm. The designed gear system increases the torque from the servo by a factor of 2.10.

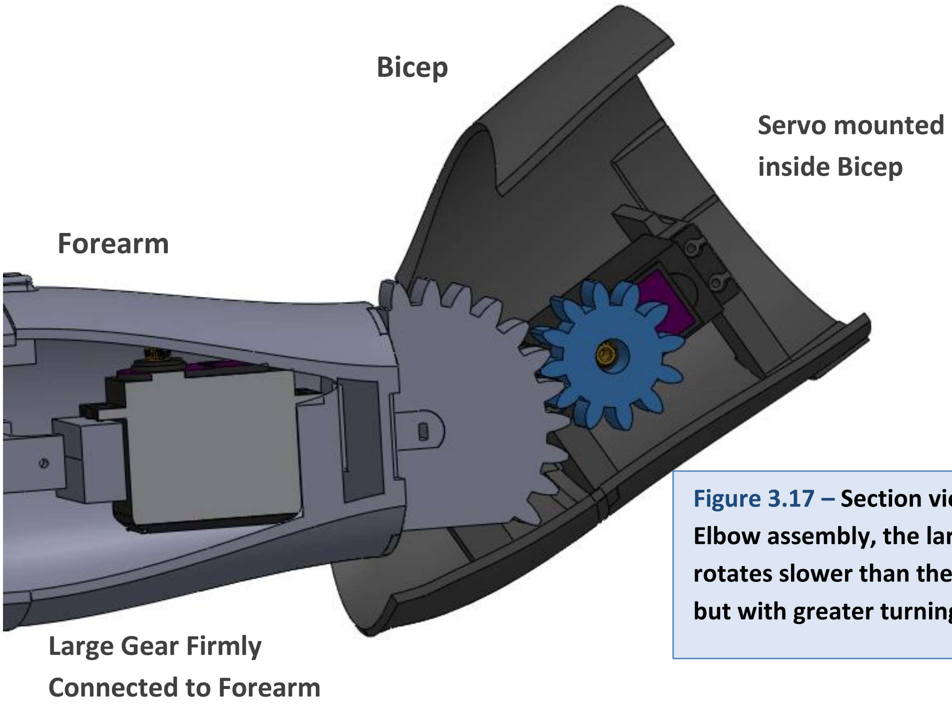
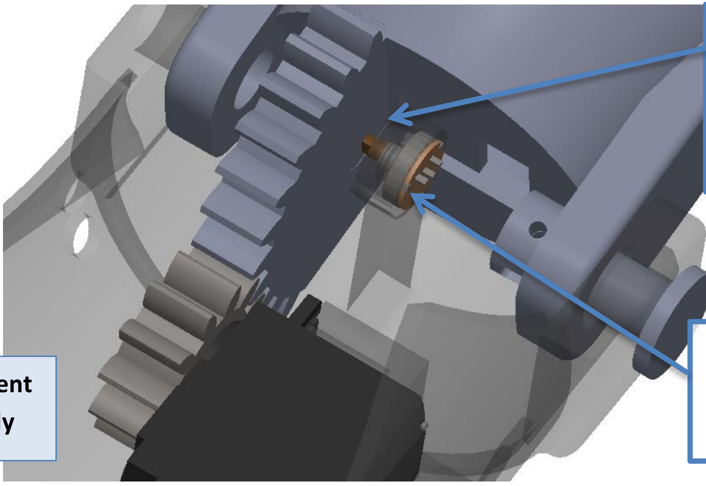


Figure 3.17 – Section view of the Elbow assembly, the large gear rotates slower than the small gear but with greater turning force

This Elbow has been designed to provide 110 degrees of rotation. This allows for a straight orientation and a right angle bend. With the addition of the gears the servo now has to rotate the small gear by 290^{O} to completely bend the elbow. As previously metioned a standard servo can only rotate through 180^{O} so modifications had to be made to increase the servos rotational range.

The elbow servo was opened up and the mechanical stop inside was removed. Inside servo motors there is also a small potentiometer which provides feedback to an internal chip controlling the rotation of the motor. This was removed from inside the servo and slotted into a groove in the large forearm gear.



Shaft of the potentiometer rotates with the forearm gear, allowing for angular control

Base of the potentiometer is held stable by the bicep section

Figure 3.18 – Transparent view of elbow assembly

By mounting the potentiometer in the forearm we have altered the servos angular feedback. The potentiometer now rotates 2.1 times slower than before, resulting in an increased servo range of 380^{O} – more than enough to move the elbow. However, by increasing the rotation range we have decreased the angular resolution of the servo.

The lower left image shows the elbow servo motor with its custom gear meshing with the gear on the forearm (left). The image on the right shows the shaft of the potentiometer slotting into the forearm gear. The motor and the potentiometer are fixed to the bicep section which is not shown below.

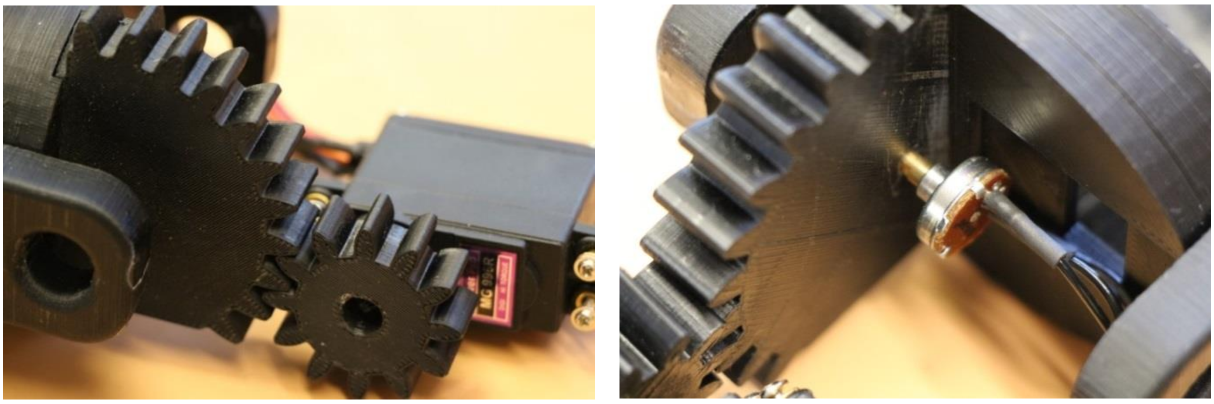


Figure 3.19 – Printed elbow gears with servo potentiometer mounted in the large forearm gear

The final design consists of 35 individual 3D printed components. The large bicep section took the longest amount of time to print at 18 hours.

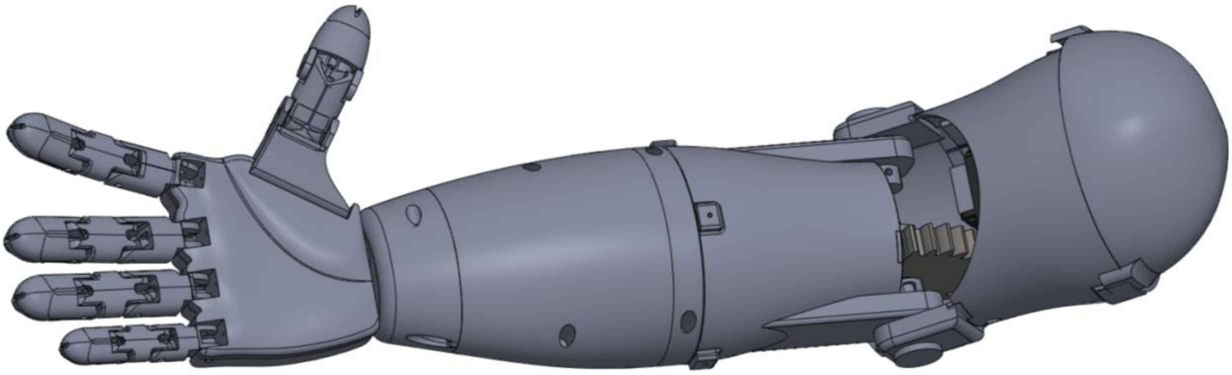


Figure 3.20 – Full mechanical assembly

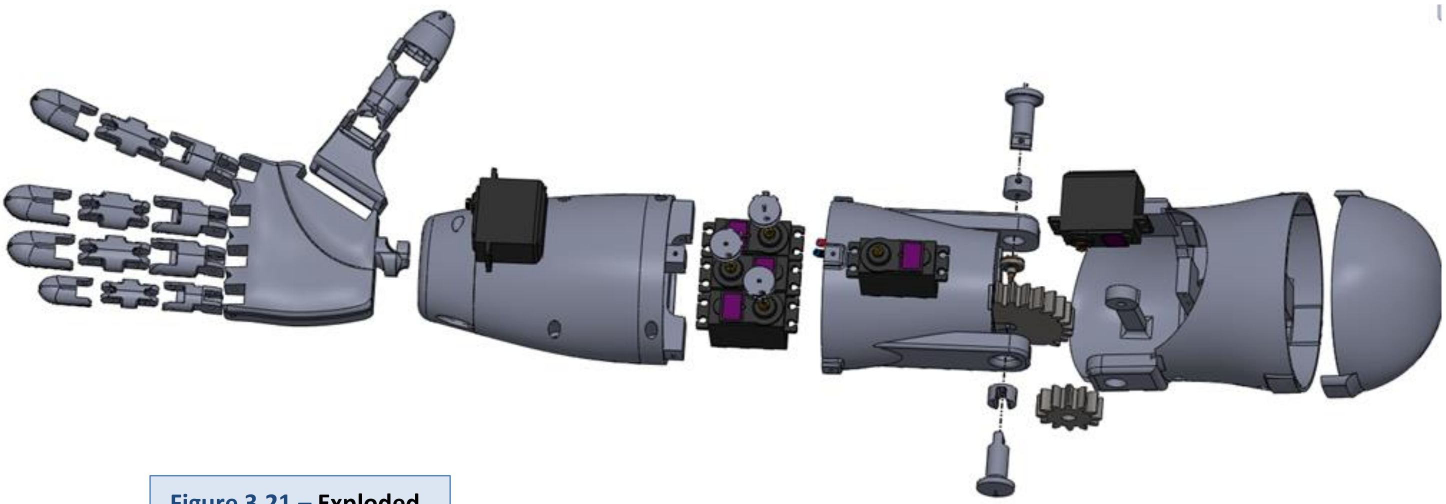


Figure 3.21 – Exploded view of full assembly

All components were printed in ABS plastic using a UP 2 3D printer. The assembly guide in the appendix discussing how the device has been assembled.

3.14 Mechanical Calculations

Required Elbow Torque

Early measurements indicate the final weight of the arm will be around one kilogram. To simplify calculations let us assume that a 1kg point load acts on the arm 13.5cm from the elbow pivot.

$$\tau = Fd$$

$$\tau_{REQUIRED} = 1(9.81)(0.135)$$

$$\boldsymbol{\tau_{REQUIRED} = 1.32\ Nm\ (13.5\ kg\ cm)}$$

This is the torque required at the elbow to lift the arm. The TowerPro servos can only provide a maximum torque of 10kg-cm. An increase in torque by %135 should theoretically be enough to lift the arm. However, it is never good for servo motors to be running at their maximum torque rating, especially for prolonged periods. Ideally we want the servo to only ever run at half its maximum torque.

The key to increasing the torque is to have a large driven gear connected to the forearm. The forearm gear shown was designed to be as large as possible and still fit nicely into the elbow space.

The small gear was made as small as possible while still providing gear teeth strong enough to transfer high torques.

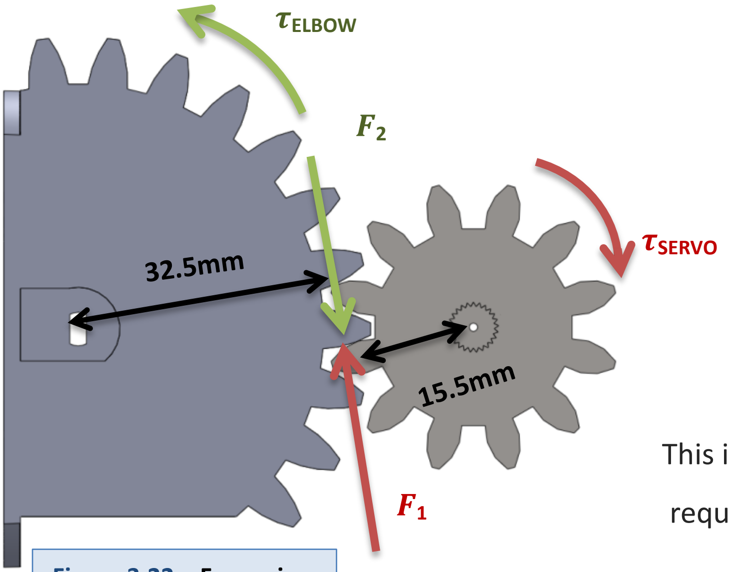


Figure 3.22 – Forces in the elbow gears

$$F_1 = \frac{\tau_{SERVO}}{15.5}$$

$$F_1 = 6.66\ kg$$

$$F_1 = F_2\ (Newton\ 2nd\ law)$$

$$\tau_{ELBOW} = F_2 d_2$$

$$\boldsymbol{\tau_{ELBOW} = 21.6\ kg\ cm}$$

This is significantly greater than the required torque to lift the elbow.

Force in Fingers

To calculate the theoretical finger strength let us first look at the situation in which the index finger is fully extended and apply a force near the tip of the finger.

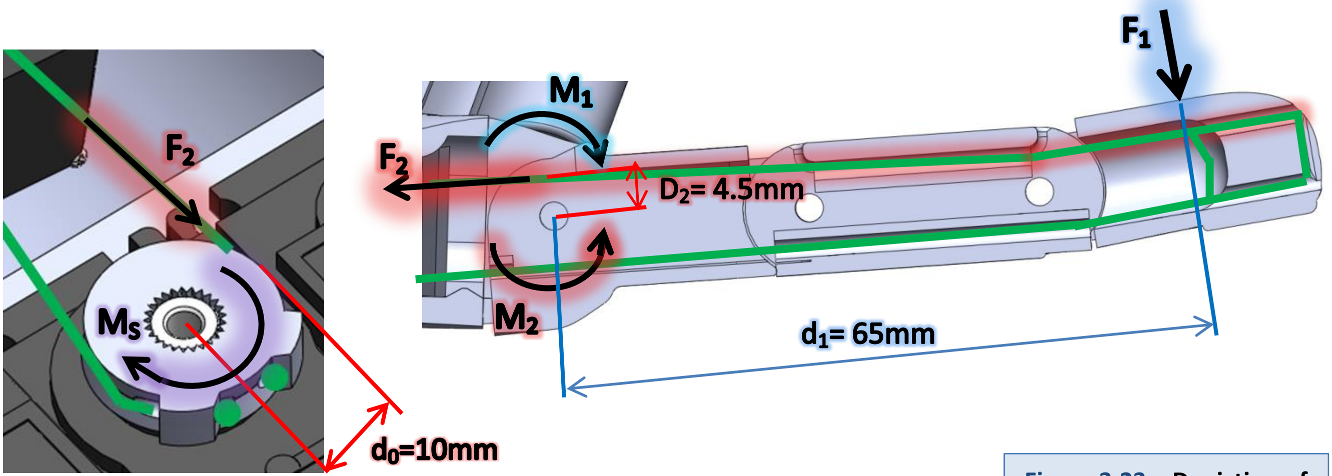


Figure 3.23 – Depiction of forces acting on fingers

$$Moment = Force \cdot Perpendicular\ Distance$$

$$M = Fd$$

In this case the tendon creates a moment about each joint in the finger. The moment about the knuckle joint will be the greatest since it is the furthest away from the applied force. Therefore it is the turning force at the knuckle that limits the load we can lift at the tip of the finger.

At the point where the maximum lift-able load is applied the moments M_1 and M_2 will balance out. To begin our calculations we must determine the tensile force in the tendon. The stall torque (maximum turning force) of the MG996R Servos is 10kg-cm (1 N/m).

$$\tau_{servo} d = F_2$$

$$F_2 = 1(N/m) \cdot 10mm$$

$$\boldsymbol{F_2 = 10N\ \ (tension\ in\ the\ tendon)}$$

$$F_1 D_1 = F_2 D_2$$

$$F_1 = \frac{10(N) \cdot 4.5(mm)}{65(mm)} = 0.70\text{ N}$$

$$Mass_1 \approx 70g$$

So a force of 0.7N can be applied at each fingertip when fully extended – or a 70g mass can be lifted. This may seem quite low but it is important to note that this is not necessarily the maximum force the finger can apply.

As the finger curls the perpendicular distance between the knuckle joint and the applied load decreases – which results in a lesser moment about the knuckle joint. This means the finger tips apply more force as they close further. Suppose the hand is curled around an object, then the applied force to the index finger would be acting in an orientation similar to the depiction below

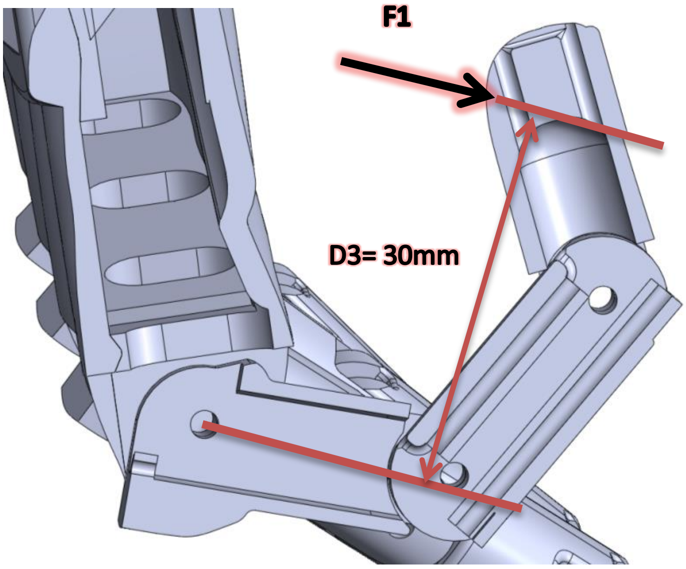


Figure 3.24 – Depiction of applied force when gripping an object

Since the perpendicular distance from the applied for to the knuckle joint is now smaller, the fingertip can apply a greater force.

In this case each finger can support a mass of roughly 150g, which would give the entire hand a lifting/holding capacity of about 600g.

$$\max finger\ tip\ gripping \approx 150g\ per\ finger$$

Finger Actuation Speed

The MG996R servo has an operating speed of 0.15sec/60^{O}. A full wrist rotation from a *palm up* to a *palm down* position (180^{O}) therefore takes (0.15/60)*180 = 0.45s.

$$\frac{0.15s}{60^{O}} \cdot 180^{O} = 0.45s$$

It has been measured that a tendon must move about 2cm to move the finger from fully extended to fully flexed. Using the arc length formula

$$length = \frac{n^{O}}{360^{O}}\ x\ 2\pi r$$

Where length is 2cm, r is the radius of the custom servo horns (7mm)

We find that the Servo must rotate 160^{O} to completely open/close each finger. We find that the maximum time to open/close a finger is (0.15/60)*160 = 0.4s.

3.14 Manufacturing & Assembly

All mechanical components have been produced using an UP 2 – a fused deposition modelling 3D printer. This type of 3D printer produces what is known as support material which provides support to horizontal planes during printing. Care has to be taken when removing this support material as to not damage the component.

All the pins used within the device, such as at the finger joints, have been 3mm diameter polypropylene filament. After After printing these pin holes were drilled with a 3mm bit to improve dimensional accuracy.

The UP 2 provides its own development environment which allows for fine tuning of the printer options. Several features have been experimented with such as print speed, layer resolution and extrusion temperature to produces high quality printed components.

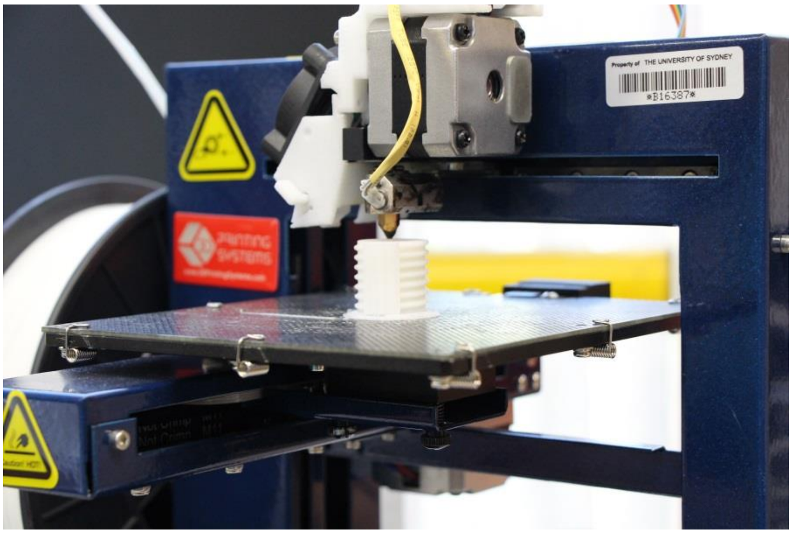


Figure 3.25 – UP 2 printing a component

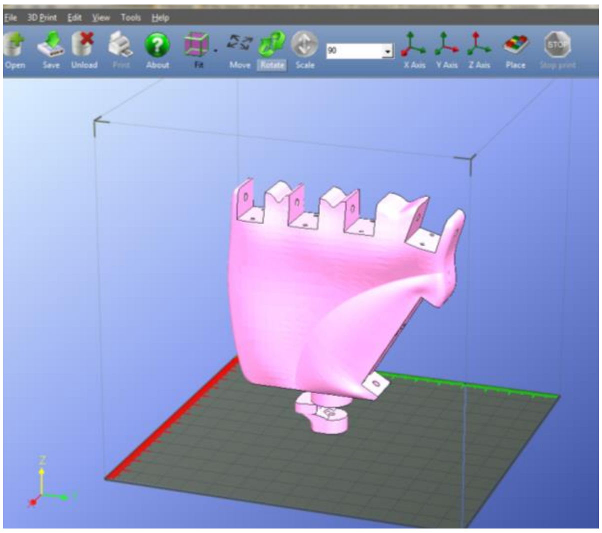


Figure 3.26 – .STL file of the palm model displayed in the printing environment

Assembly of this device proved to be quite challenging and required the use of several tools. Threading the tendon lines through their guide holes, tensioning the tendons and tuning the servo/finger movements required precision and patience. The assembly guide in the appendix provides some basic instruction on how to put the device together.

3.2 Electrical Design

Signal Flow Overview

A user flexing generates an analogue signal which is amplified, rectified and smoothed by the EMG sensor board. The microcontroller uses this analogue signal to generate a pulse width modulated signal. This drives servo motors which tension the tendons causing the fingers to curl up.

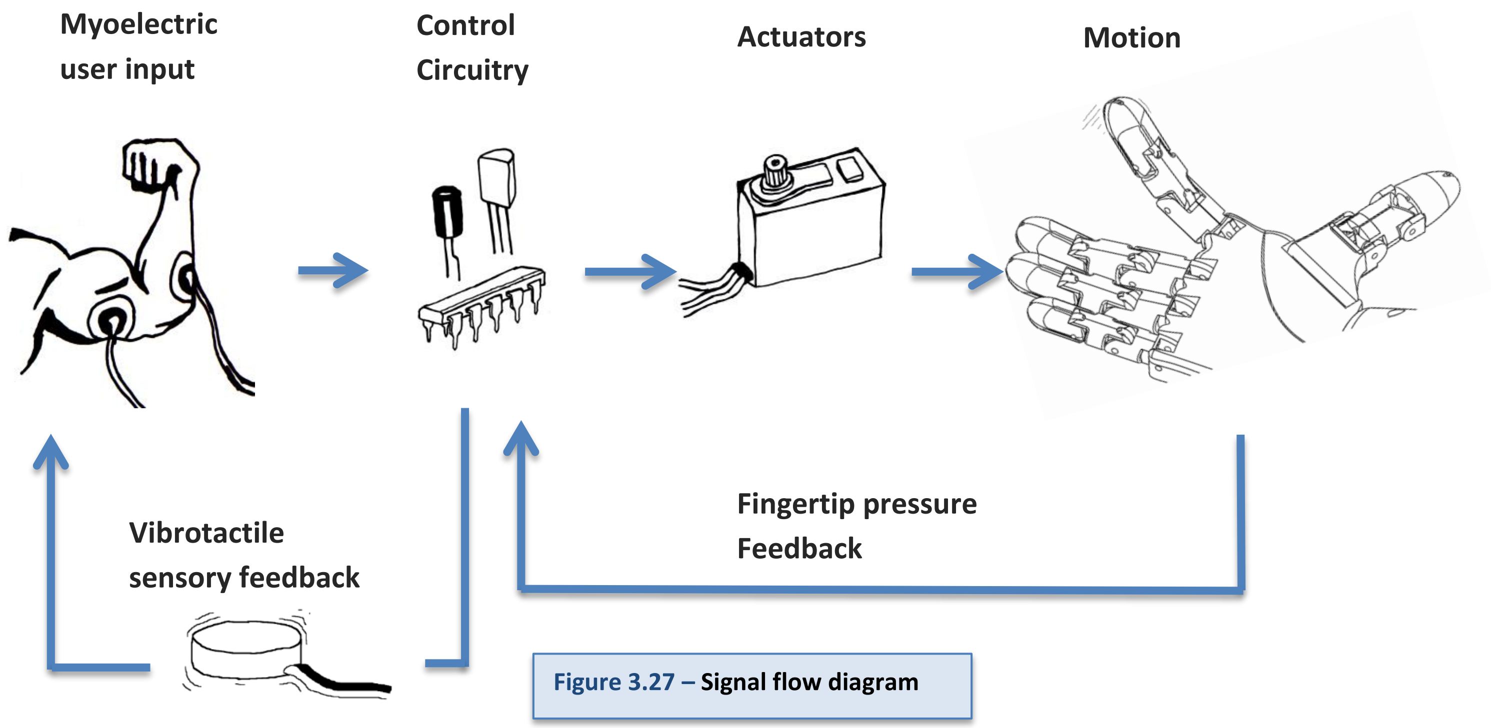


Figure 3.27 – Signal flow diagram

3.21 Actuation

As previously discussed the actuators used in this system are standard servo motors. These motors can be controlled to rotate to angular positions up to ± 90 degrees from rest.

Since the artificial tendons move fairly little in order to open and close each finger, the angular precision of each servo somewhat affects how precisely the fingers can be controlled. Relatively inexpensive servo motors have been used in this system to maintain a low cost. The use of higher quality servos would of course increase finger strength and precision but would cost significantly more.

Figure 3.28 – TowerPro MG996r Servo motor

3.22 Microprocessor

An 8-bit microcontroller from the Microchip PIC18F series has been used as the central computer for this system. This family of processors has limits on computational power but are more than adequate for this design. In order to program the device a pickit3 debugger/programmer attaches to the microcontroller circuitry and is passed data through a usb connection to a PC.

Figure 3.29 – Two 8-bit PIC microcontrollers and a PICkit3 programming device

3.23 Voltage regulators

Power regulators have been used in order to control the voltage and power supplied to the servos and the microcontroller. These regulators prevent a situation in which a servo could be stalling and drawing a large amount of current from the battery. If such a case happened to all six servos simultaneously it could easily damage the electronics or even cause a fire in a worst case scenario.

Out of availability 5V surface mount regulators have been used to supply the servos and a 3.3V regulator used to power the microcontroller. The specific regulators used output a maximum current of 1A.

A fair assumption is that each servo will need access to at least 500mA of current to operate. Therefore each 1A power regulator supplies two servos. These specific regulators hold the output stable at 5V. This results in slightly slower and weaker servo performance as ideal servo power is about 6.5V.

3.24 Electromyography Sensing

Easy to use single channel EMG sensor boards have been used to sense and measure muscle activity. This kit contains a small PCB and three surface electrodes. Two of these electrodes measure the voltage potential across a muscle and the third is a ground reference point placed on a boney feature.

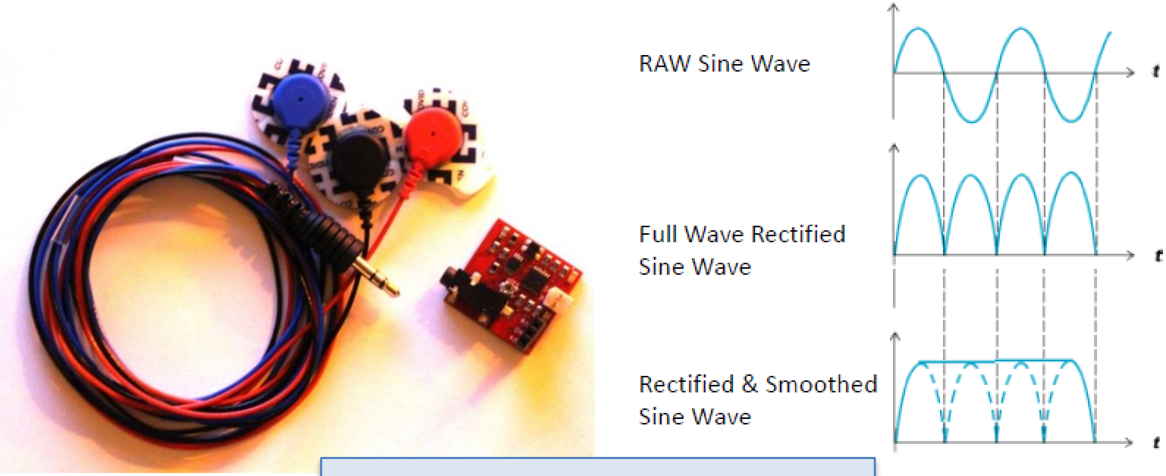


Figure 3.30 – EMG sensor kit & signal output

The muscle sensor kit is designed to be used directly with a microcontroller. As a user flexes, an internal amplification system converts minute electrical pulses into a rectified and smoothed signal that can be used as an input to a microcontroller's analogue to digital converter.

3.25 Power Supply

It is important that this system is portable and completely powered by internal sources. Using a wall power supply is fine for testing and debugging but a prosthetic arm needs to be powered by a source an amputee can easily carry around.

Servo motors use a significant amount of current during operation. Disposable batteries would not be a good solution since the servos would drain power too fast meaning they would have to be replaced quite frequently. Lithium Polymer (LiPo) batteries offer a high energy density and are rechargeable.

There is a trade-off between battery life and battery size. Ideally we would like the arm to be able to run for several hours without needing to be recharged. However, to achieve this, the size of the battery may become too large to be housed within the device.

Figure 3.31 – 2 Cell Lithium Polymer battery powering the device

The muscle sensor kits require very little power whereas the Servos and microcontroller require a significant amount. The muscle sensor kits require two power sources to create a positive and negative voltage reference. These sensors are sensitive to input voltage spikes and require a stable power supply to generate high quality signals.

For these reasons the EMG sensing boards are supplied power by two separate 9V batteries. Two disposable batteries should provide power for a substantially long time. When they do run out they can be easily and cheaply replaced.

A more sophisticated electrical system could link all components to a single power source using techniques to generate a negative voltage for the EMG sensors. This would reduce the size, weight and clutter of the current power supply system.

3.26 Circuit Design

The following schematic outlines the circuitry of the system. Note that six power regulators are incorporated but only one has been shown for clarity.

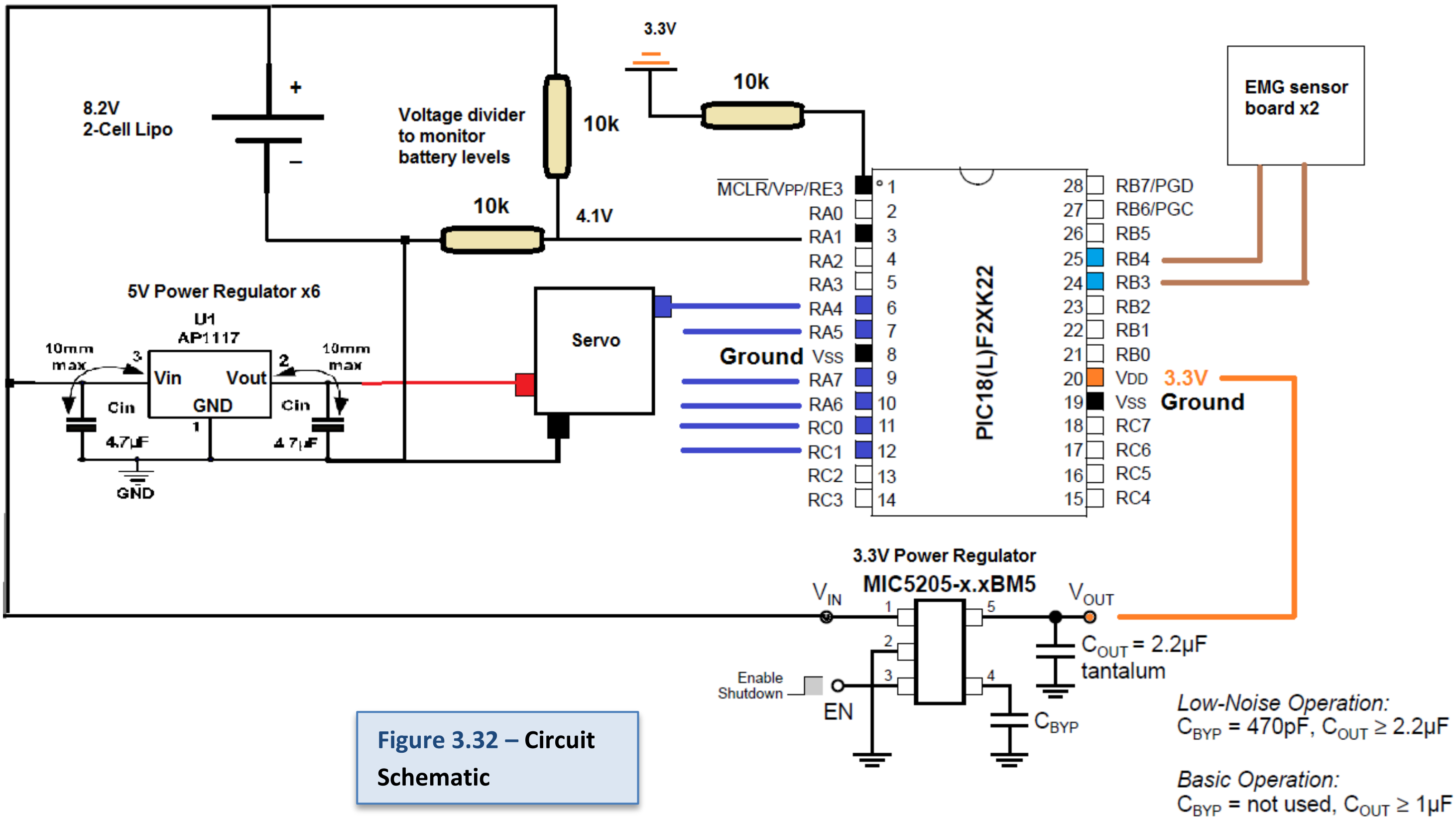


Figure 3.32 – Circuit Schematic

A larger version of this schematic can be seen in the appendix.

Due to time constraints the pressure sensors and vibrotactile motors were not implemented. These components would have been attached to the remaining pins of the microcontroller and powered through their own voltage regulators. Another useful feature would be a potentiometer that can be rotated to control characteristics like EMG sensitivity.

Printed Circuit Board Design

Below is an image of the printed circuit board design made for this prosthetic arm. It was hoped that the university would obtain an electronics 3D printer before the completion of this thesis. This type of printer could be used to print out the circuit traces shown below onto a variety of materials. Unfortunately the circuit printer was not received in time.

Nevertheless a PCB schematic is extremely helpful in designing prototype circuits on perforated Veroboard. The schematic below shows the servo output pins and the power supply input on the left followed by the voltage regulators, microcontroller and the programming pin port for the debugger. This schematic file could be easily sent to manufacturer to get the board professionally made.

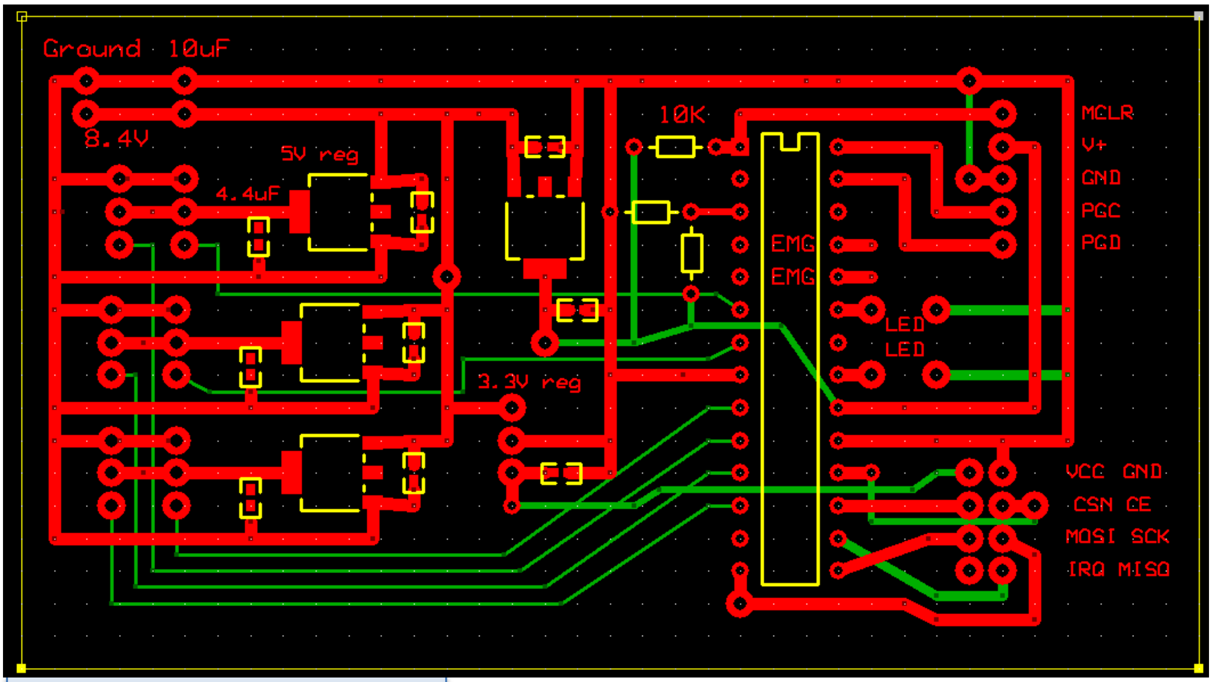


Figure 3.33 – Printed Circuit Board design. Made using ExpressPCB

Note: The lower right section of the PCB design (Vcc, GND etc.) caters for a wireless radio chip which could be used to wirelessly send and receive data. This was going to be used to receive information from a motion sensing glove a user could wear to control the arm. This feature has been excluded from the thesis as it does not directly relate to the field of prosthetics.

3.27 Prototyping

The electronics were prototyped on Veroboard using a combination of through-hole and surface-mount components. The circuit board had to be kept as small as possible so that it could easily fit within the bicep section. The circuit below was made following the PCB design layout on the previous page.

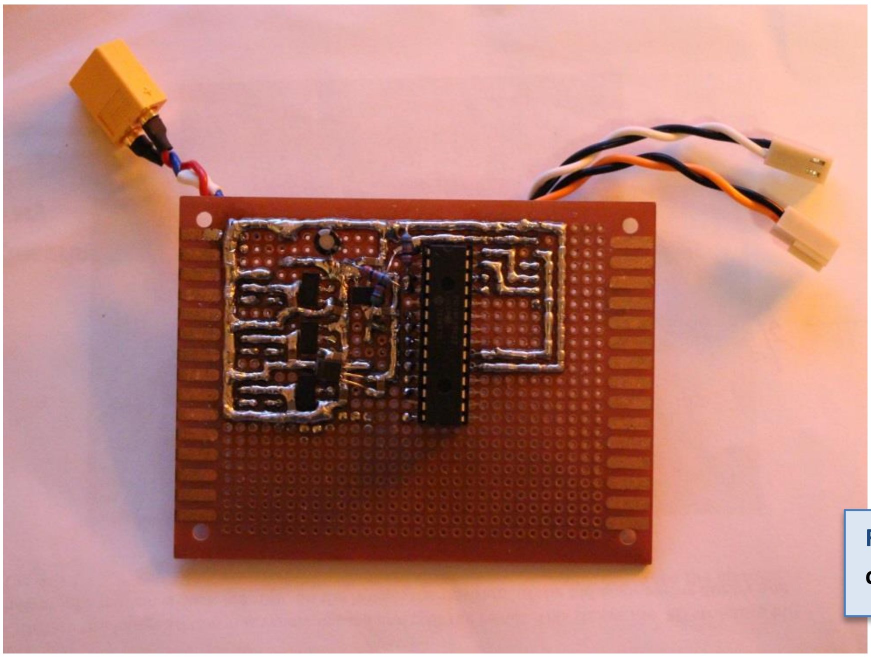


Figure 3.34 – Prototyped circuit board

The image below shows this central circuit board connected to the LiPo battery and the two EMG sensor kits and their power supply. Housing this entire system within the arm proved to be challenging.

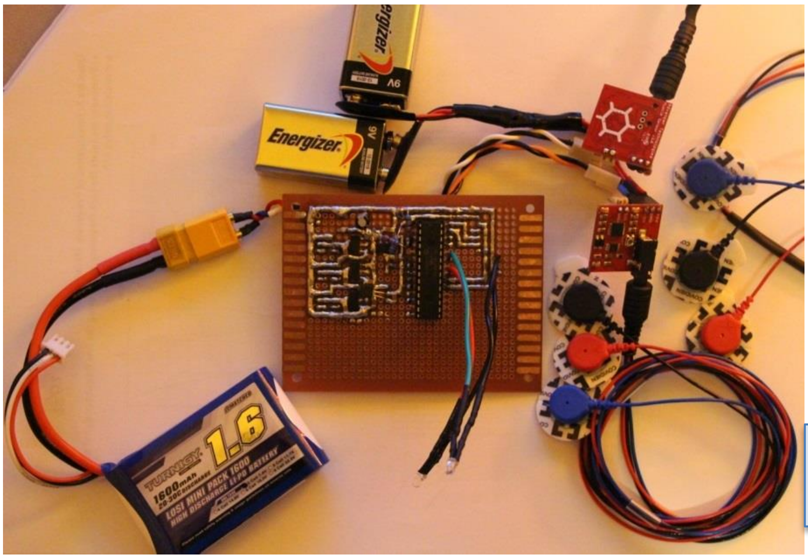


Figure 3.35 – Complete electrical System

3.3 Firmware Implementation

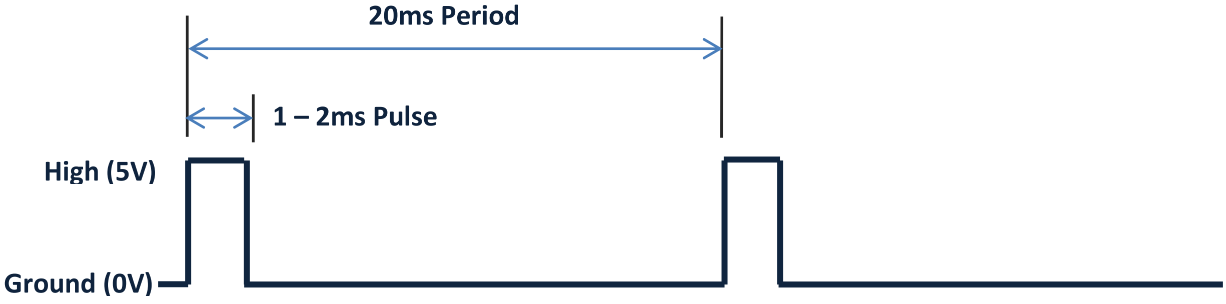
3.31 Programming

A PIC microcontroller understands only its own specific assembly language. In order to program the microcontroller software code must be written in assembly language. Fortunately, compiler software can convert standard C code into assembly code automatically. Coding in the standard C language is generally much easier and faster.   MPLABX is the development environment provided by Microchip. Microprocessors are be programed via a USB connection.

The microprocessor monitors input signals from EMG sensors, makes calculations to determine the required action associated with that input and generates corresponding signals for the motors.

3.32 Servo Signals

A pulse width modulated signal is used to control the servo motors. Every 20ms a pulse between 1ms and 2ms long is sent from the PIC microcontroller to the internal control circuitry of the servo.



A 1.5ms pulse rotates the servo shaft to its central position. Different pulse widths correspond to different motor shaft positions.

To create a PWM we could use a software timer to accurately control the timing and duration of a pulse. Another option would be to use the inbuilt PWM generator feature of the microcontroller. The problem with both these options is that there are not enough software timers to control each servo. Six servos need to be controlled and only four timers are available.

For this bionic arm two 16-bit software timers have been used to accurately control six servos. Six PWM signals need to be generated on individual output pins.

Every 3.3ms the beginning of a new pulse is started on a new output line. After six cycles 20ms have passed a new pulse begins on the first signal line and the cycle repeats. A software timer is used to control the 3.3ms period time. If for example only five servos were used then we would start a new signal every 4ms.

The image below outlines how two timers have been used to generate all six of the PWM signals.

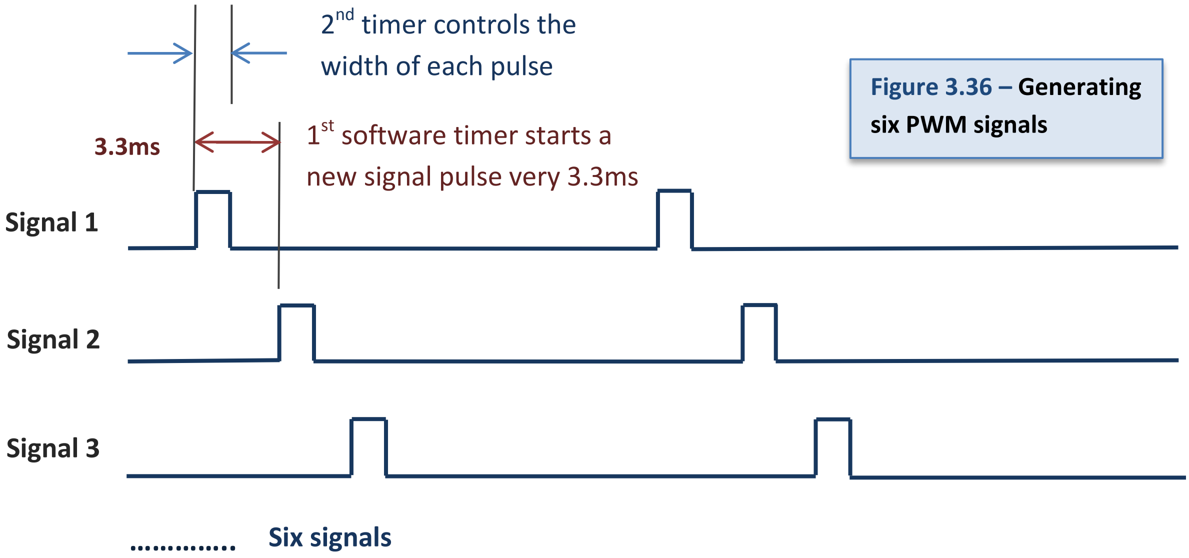


Figure 3.36 – Generating six PWM signals

A maximum of eight individual servos can be controlled using this method. Anymore and the time between the start of each pulse becomes less than the pulse width of each signal. Meaning another timer or method would have to be used.

3.33 EMG Control

As discussed in the electronics design section, each muscle sensor board outputs an analogue signal (0–3V) into an analogue pin on the microcontroller. The microcontroller performs and analogue to digital conversion on this signal storing the result as a 10-bit binary value which is used to control the positioning of the servo motors.

This thesis has only explored relatively basic EMG control algorithms. Two electrode pairs provide two analogue signals which are used to control the actuators of the device. As the magnitude of the EMG signals pass above arbitrary thresholds specific commands are executed.

The basic EMG control is as follows.

- One electrode set measures myoelectric signals from a muscle region such as the bicep. As a user flexes the EMG signal is used to switch between different grip pattern states of the Arm. A state could be a precision grip, power grip or wrist/elbow rotation configuration
- The other EMG sensor monitors another muscle region such as the forearm. Flexing the this muscle region actuates the specific state the device is in – fingers close or joints rotate to pre-set positions.

Using this method it is possible to control the opening and closing of different grips as well as allowing for wrist and elbow rotation states. However, only a single command can be executed at a time and it naturally takes a user some time to cycle between states. This means it is not possible to close individual fingers and rotate the wrist at the same time using this basic control.

Ideally we would like a system which allows the user to control the exact positions and force applied by each digit and also allow for control of several movements simultaneously.

A basic form of proportional control was implemented and tested on this device. This allows the user to close the fingers more by flexing harder. The magnitudes of the EMG signals were used to linearly increase the pulse widths of the PWM servo signals – shown in the equation below

$$\boldsymbol{w_{PWM}(t) = a + k|emg(t)|}$$

W_{PWM} = servo signal pulse width **a = arbitrary offset** (servo PWMs start at 1ms)

k = Scaling factor **emg = EMG signal magnitude**

3.34 Sensor Feedback

Ideally we would like to include pressure sensors on each finger to provide some feedback. These sensors provide information to the microcontroller about how much force is being applied at each fingertip. This information can be used to control vibration motors housed in a flexible band than can be worn around the upper arm. This provides some basic sensory feedback to the user letting them know if they are grasping an object and how much force they are applying.

3.35 Program Flow

The basic structure of the current program is outlined below.

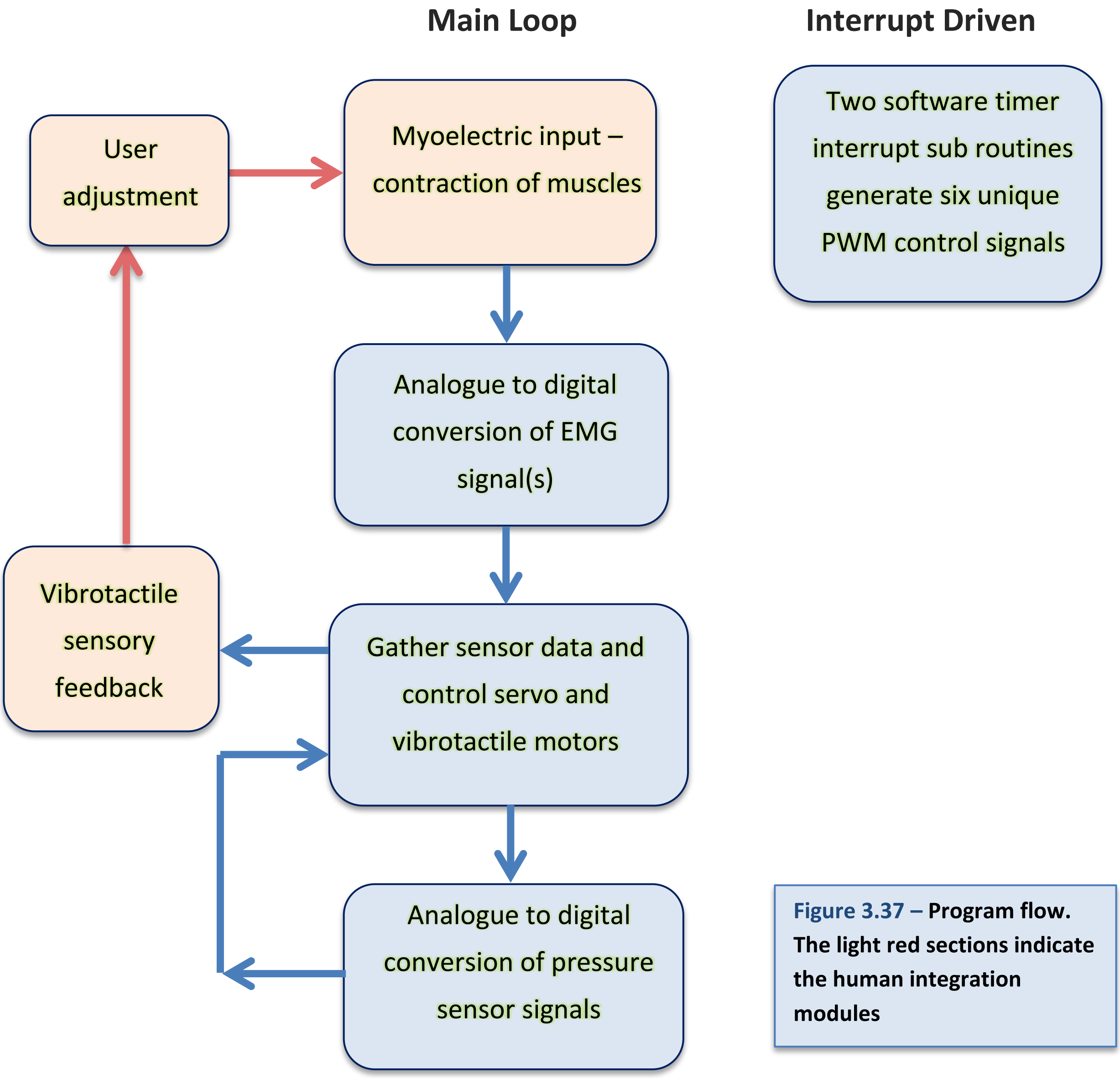


Figure 3.37 – Program flow. The light red sections indicate the human integration modules

Chapter 4.0 – Testing & Results

The final system is a bionic arm offering six degrees of freedom and the ability to be controlled through myoelectric signals. The total design consists of thirty six individual 3D printed components.

It is highly recommended that the user watch the short videos in the digital appendix section on the provided CD. These videos show some preprogramed commands to demonstrate the speed and movement of the device as well as footage of the EMG control.

Figure 4.1 – Final 3D printed prosthetic arm

Several features of the prosthetic arm have been tested and measured to improve the performance and characteristics of the system.

4.1 System Specifications

The thumb, index and middle fingers each move independently and the ring finger and small finger move in tandem. The wrist allows for 180 degrees of rotation and the elbow allows for 110 degrees of bending. A user can control the motion of the arm through a set of EMG electrodes placed on their forearm and/or bicep. At this stage two electrodes sets allow for rudimentary control of the arms range of motions. The device is completely portable and has a battery life of more than 3 hours.

The table below outlines some key features of this design

System Properties	Description
System Weight	950g
Material	ABS plastic
Tendons	Braided Fishing Line
Power Source	2-Cell 7.4V Lithium polymer rechargeable battery
Microcontroller	PIC18f25k22
EMG Sensors	Muscle Sensor Kit V3
Servo Voltage Regulators	AP1117 – 5V output, maximum current 1A

Figure 4.2 – **System specifications table**

4.2 Achievable grasps

The system is physically capable of several various grip pattern arrangements shown below.

Figure 4.3 – **Various Device Grip Patterns**

4.3Component Strength

All components have been 3D printed in ABS plastic. ABS is commonly used for 3D printing due to its low melting point and properties. Another commonly used material is PLA plastic. ABS is tough and impact resistant which makes it ideal for a prosthetic arm – a device which has to absorb significant impact without breaking. PLA components warp less than ABS ones allowing for more dimensionally accurate parts, however PLA is brittle and cannot absorb much shock on top of this the UP 2 3D printer struggles to print with PLA

Unfortunately, 3D printed components are naturally quite weak due to the way they are made. A 3D printed object is slowly built up by a nozzle extruding molten material in cross sectional layers. The boundary between two 3D printed layers is essentially an imperfection in the grain structure of the material. The ultimate tensile strength of 3D printed ABS is significantly lower than ABS formed by injection moulding. For example, LEGO blocks are made from ABS but are injection moulded rather than 3D printed. Injection moulding results in much stronger components and improved surface finishes. 3D printed components offer poor mechanical properties which are dependent on the specific 3D printer being used and the quality of the print.

Several components broke when being handled and assembled. Many components had to be given an acetone coating after they had been printed. Acetone dissolves ABS so by allowing it the flow through cracks and seams the layers bond together better and component strength is improved.

Another method is to suspend components in an evaporating stream of acetone gas. This bonds layers together much better and increases the strength of components. It also creates a good durable surface finish. However, much care has to be taken to not overly dissolve component surfaces which could lead to significant warping of the part.

4.4 Actuation

Fluidity

The fingers move in a relatively smooth and natural manner. The fluidity of finger movements is dependent on several factors including friction between moving plastic components, servo control and also how well the tendons are tensioned.

The middle finger opens and closes exceptionally well. The ring and small finger do not move as smoothly and do not completely close. This is because both ring and small finger tendons are tied to the same servo which has to work to move both fingers in tandem. Since the small finger is a 0.8 scale of the ring finger it means its tendons do not have to move as far to open and close the finger. Neither tendon is actuated in an optimal manner which reduces the fluidity of the ring and small fingers.

Strength

Testing using small kitchen scales indicates the fingers can provide at least 300g of force each. Indeed the servo motors could be rotated more to further increase the tension on the tendons. This would effectively increase the closing force of each finger. The limiting factor in finger force is not the torque of the servo motors but rather the strength of the printed ABS components. If we kept increasing the tension applied by the servos either a finger component would break or a fracture would occur at the wrist.

The only sure way of determining the absolute maximum finger force would be to test to destruction. Unfortunately tests to destruction could not be carried out – otherwise there would be no working system to present.

Unfortunately the servo actuating the elbow joint is simply not strong enough to move the entire forearm reliably and smoothly. The servo used in the elbow is rated at 10kg-cm. A high quality top of the line servo could offer 25kg-cm, coupled with the torque increase from the gear system this would be more than enough to safely rotate the arm about the elbow. These super high torque servos would significantly raise the cost of the system.

4.5 Control

The basic Boolean EMG control allowed for different states to be cycled through and actuated. One state allowed for wrist rotation and another state allowed for finger actuation. It was possible to instruct the hand to close, rotate to a certain position, rotate back and then reopen. Such a movement could be used to grasp and pour a liquid from a bottle. However having to switch between the two states made the task slow and tedious using this basic EMG control.

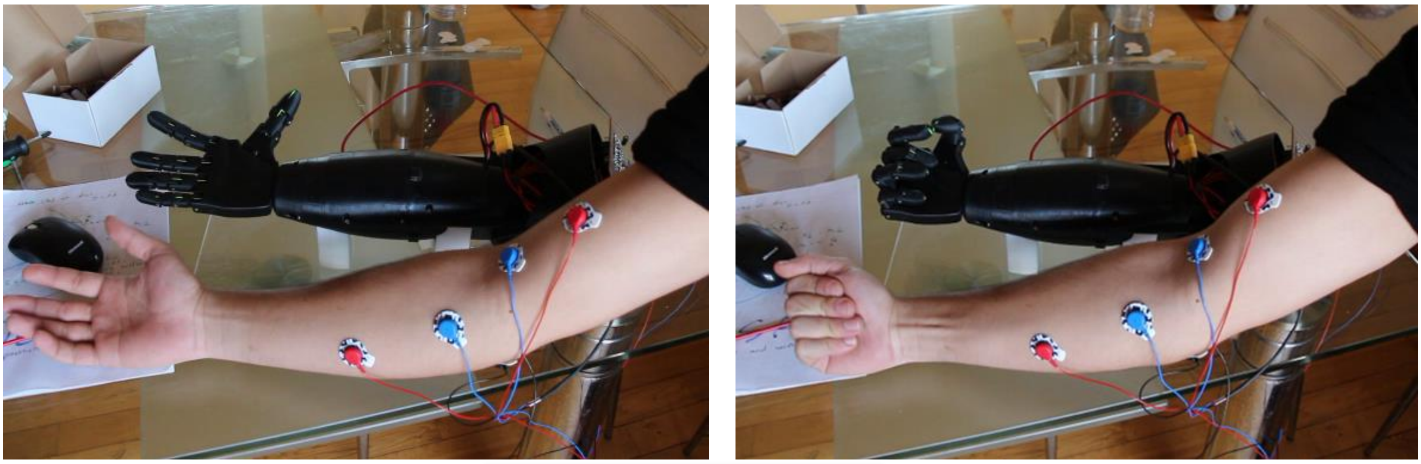
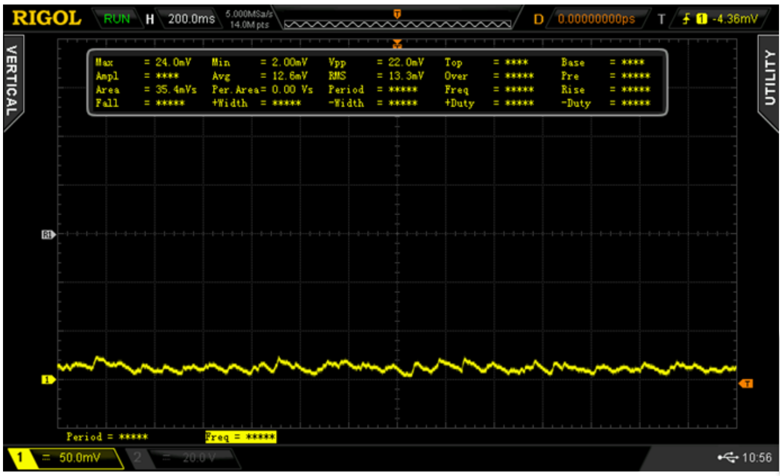


Figure 4.4 – User controlling the opening and closing of the hand through muscle flexing

Proportional control of the fingers worked fairly well – the harder the user flexed the more the fingers would close. However, this proportional control caused the fingers to start shaking when trying to close – which was due to noisy signals being used to control servo positions.

Initially the EMG signals were sampled every 100ms converting the magnitude of the signal to a decimal value. The servos were also updated every 100ms.

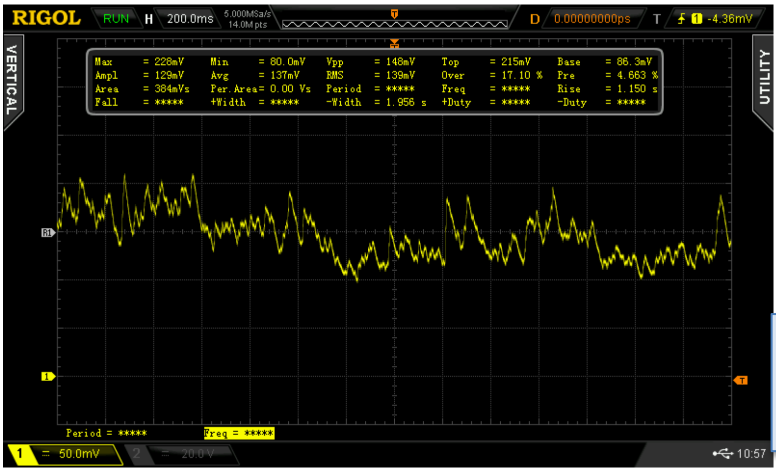
As seen in the oscilloscope images on the next page the EMG signals are quite noisy. This means the signal voltage level can significantly jump or drop in 100ms which results in the servos being instructed to constantly move around to different positions. This is especially bad with high level signals.



```
Max    = 24.0mV   Min       = 2.00mV   Vpp    = 22.0mV   Top    = ****    Base   = ****
Ampl   = ****     Avg       = 12.6mV   RMS    = 13.3mV   Over   = *****   Pre    = *****
Area   = 35.4mVs  Per.Area= 0.00 Vs    Period = *****    Freq   = *****   Rise   = *****
Fall   = *****    +Width    = *****    -Width = *****    +Duty  = *****   -Duty  = *****
```

```
Period = *****          Freq = *****
```

Figure 4.5 – Light flex output (200ms/div, 50mV/div)



```
Max    = 228mV    Min       = 80.0mV   Vpp    = 148mV    Top    = 215mV     Base   = 86.3mV
Ampl   = 129mV    Avg       = 137mV    RMS    = 139mV    Over   = 17.10 %   Pre    = 4.663 %
Area   = 384mVs   Per.Area= 0.00 Vs    Period = *****    Freq   = *****     Rise   = 1.150 s
Fall   = *****    +Width    = *****    -Width = 1.956 s  +Duty  = *****     -Duty  = *****
```

```
Period = *****          Freq = *****
```

Figure 4.6 – Strong flex output (200ms/div, 50mV/div)

As illustrated below the sampled points used for servo control are constantly shifting up and down. A smooth signal is desired which means some filtering should be done to these EMG outputs to improve control characteristics.

Increasing the time between ADC samples and servo updates slows down the shaking of the fingers but does not fix the problem.

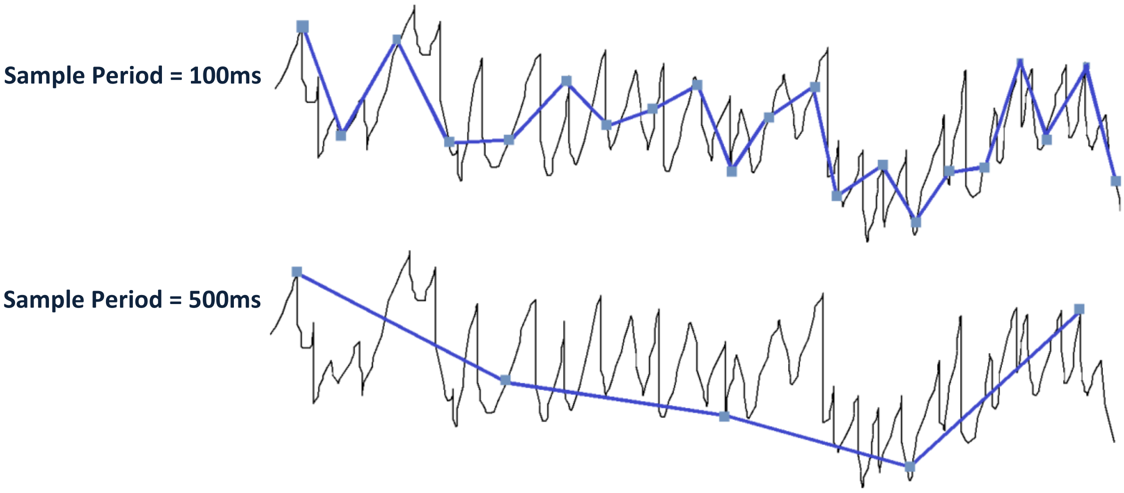


Figure 4.7 – EMG signals (black) with ADC samples (blue). The voltage level of these samples controls servo positions

In order for proportional control to work we need an EMG signal that increases linearly with flex intensity. To generate the screen shot below the user focused on starting at rest and gradually increasing flex intensity to a maximum and then releasing.

```
Max    = 728mV    Min      = -16.0mV  Vpp     = 744mV   Top     = 696mV   Base    = -16.0mV
Ampl   = 712mV    Avg      = 300mV    RMS     = 375mV   Over    = 10.11 % Pre     = 4.494 %
Area   = 839mVs   Per.Area= 0.00 Vs   Period  = *****   Freq    = *****   Rise    = 1.454 s
Fall   = *****    +Width   = 1.308 s  -Width  = *****   +Duty   = *****   -Duty   = *****
Period = *****            Freq = *****
```

Figure 4.8 – Varying EMG output with increasing flex intensity

The magnitude of this signal seems to increase linearly with increasing flex intensity. This is required for accurate proportional control.

4.6 Advanced EMG Integration

The device was integrated with the work done by fellow thesis student Michael Cerbara on researching and developing advanced EMG control for prostheses. A data acquisition device gathered information from three sets of permanent electrodes measuring signals from the bicep, tricep and chest.

Using simple binary levels the three electrode sets allowed for 8 separate commands to control the arm. This allowed for more of the arms motions to be controlled without having to cycle through device states. For example, the wrist and elbow rotation could always be directly controlled through specific flexing patterns. However, this system still only allowed for one actuation command to be carried out at a time. More work would be required to develop a control system which allows the user to control separate actuators simultaneously.

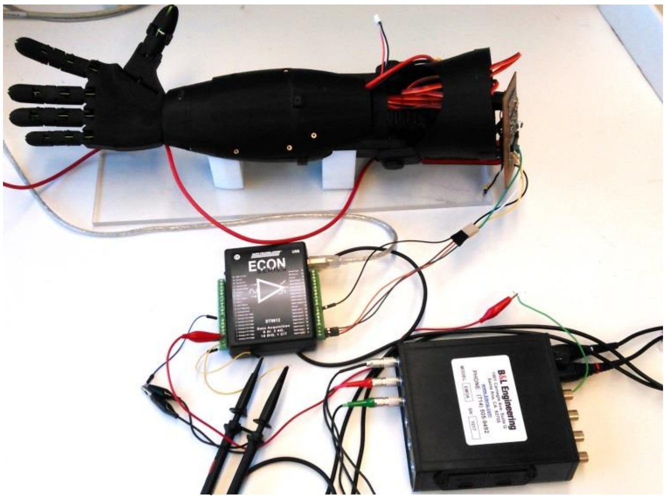


Figure 4.9 – Arm connected to the data acquisition setup for advanced EMG control

4.7 Cost

The table below shows the total material cost to be $250. Ideally higher quality servo motors should be used which could add over $100 to the total cost.

Description	Cost
700gram Spool of ABS	$40
6 x Standard Servo motors	$60
2 Cell LiPo Battery	$20
2 x Muscle Sensor kit v3	$100
Electronics (micro, regulators, Veroboard, solder, wires etc.)	$20
Miscellaneous (Braided fishing line, screws, Acetone etc.)	$10
	Total = $250

Figure 4.10 – Table of material costs

For what the muscle sensor kits offer they are quite expensive at $50 each. Using other EMG sensing devices could significantly reduce the system cost.

4.8 Battery Life

The LiPo battery being used is rated at 1600mAh. The maximum current the system can pull in any circumstance is around 3.1A. This corresponds to an extreme worst case battery life of roughly 30 minutes (1600mAh/3.1A). However, such a scenario would only happen if all the servo motors were continuously pulling at their maximum capacity – which would never happen for an extended amount of time.

A much more realistic estimate is an average current consumption of 75mA by each servo. Incorporating power requirements of the microcontroller and other electronics a reasonable estimate of 550mA required system power gives an estimate of just under 3 hours of battery life.

In practise the battery life is significantly longer lasting up to 6 hours. This is because the majority of the time the joints are at a rest position minimising the power usage of the motors.

Chapter 5.0 – Discussion

5.1 Overall System Quality

The characteristics and build quality of any engineering design greatly affect system performance. A secure, practical and durable prosthesis is incomparable to something of minimal quality.

The nature of 3D printing leads to components of minimal quality. Unfortunately the bionic arm prototype is not of high enough quality and development to be used as a prosthesis or benefit amputees at this point in time.

To truly be a practical myoelectric prosthesis the quality of this system would need to be improved, specifically:

- The strength and rigidity of the structure would need to be improved through design changes and use of better materials
- Future work designing a socket connection to attach the system to an amputees stump is required.
- System control improvements would be necessary before an amputee could successfully operate the device reliably
- Non-reusable, permanent electrodes would have to be used for the myography sensing

5.2 Weight

The final portable system weighs 970g which is relatively low compared to the average human arm (2.5kg) but slightly more than most prosthetic arms. The entire weight of a prosthetic arms acts on a relatively small area of skin at the socket connection. Because of this amputees feel the weight of a prosthetic arm more than their biological arm. Minimising weight is important but is not crucial at this stage and can be addressed if and when a socket connection is designed in the future

5.3 Mechanical Robustness/Strength

Testing with the small kitchen scales has determined low estimates of the grip forces of the prosthetic arm. We aim to minimise weight and increase the arms gripping and lifting strength. The plot below is taken from a research paper that analyses and compares several prosthetic arms.

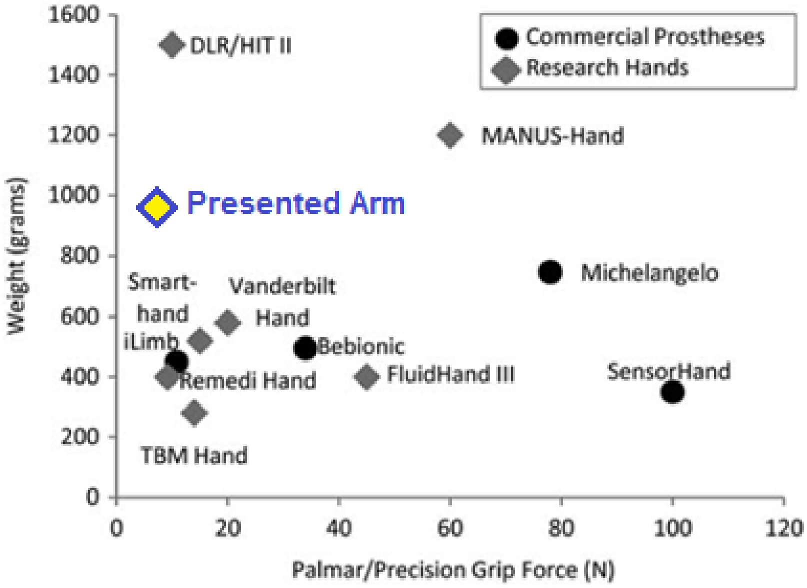


Figure 5.1 – Prostheses Weight vs Grip Force

As illustrated in the plot above, the weight and grip force of the presented device are significant weak points in its characteristics compared to other designs.

The Michelangelo and SensorHand offer low weight and high grip force. This is because these devices use only a single actuator to control all the fingers. This not only reduces overall weight, but can also increase grip strength because it means a single large, powerful actuator can be used to drive the fingers. The obvious down fall of such a system however is that it is not possible to move the fingers independently.

The high weight of the presented prosthesis comes partly from thick 3D printed components – which are essential for strength. The heavy servo motors also contribute slightly more than a third of the overall system weight. Weight distribution is not smooth across the length of the arm. The weight of the palm is very low compared to the forearm. Ideally a prosthetic device should have an even weight distribution across its length.

5.4 Performance Comparison

Compared to other myoelectric prosthetic arms this device offers poor strength but fast actuation speeds. As outlined in the chart below the strength and dexterity of this device are somewhat limited compared to typical commercial and research prosthetic arms.

Comparing to suggested practical values the presented device would need a drastic improvement in grip force before it offers practical strength. However, this suggested grip force value is rather optimistic and is even above average commercial arm values.

The characteristics of the biological human arm are far superior to any prosthesis. A long term goal in the field of prosthetics is to create devices with characteristics comparable to the human arm.

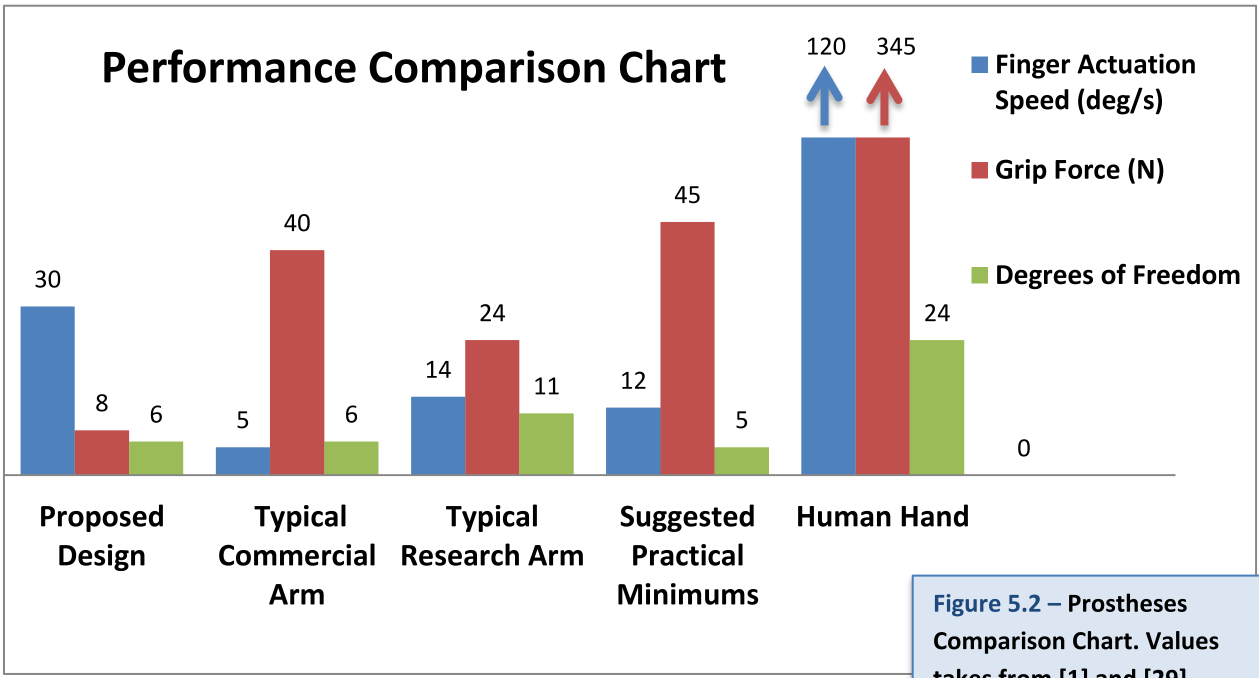


Figure 5.2 – Prostheses Comparison Chart. Values takes from [1] and [29]

5.5 Grasping Objects

The one area this device does excel in is actuation speed. Fast actuation is not absolutely crucial but it does allow the user to control the device more comfortably. A survey of myoelectric prostheses users found that more than 75% complained of their device being too slow [18].

In order to effectively grasp objects of various shapes and sizes pressure sensors are required on the finger tips to provide force feedback. Currently the fingers will try to reach a closed state even if some object is obstructing their path. This causes the servos to pull as hard as hard as possible trying to get the fingers to their closed position. Feedback could signal the servos to hold their position once an object was being touch and allow for controlled pressure to be exerted.

Interestingly most commercial myoelectric prosthetic arms rely on slower finger closing speeds to allow the user to visual decide when to stop the closing the fingers[1]. Such a system would be hard to control using the fast actuating fingers of this design.

5.6 Actuation & Complexity

As discussed in the literature review a degree of freedom is a unique way in which the device can move. The human arm (not including the shoulder) has 23 degrees of freedom, making it far more dextrous than the bionic arm presented which only has 6 degrees of freedom.

Although each finger has three joints these joints all rotate together when the tendons are tensioned. Furthermore, no fingers allow for abduction/adduction movement (wiggling side to side) which greatly limits the dexterity of the hand.

The palm can rotate about the wrist by 180^{O} – which is similar to the movement of the human wrist. However, the lack of ability to flex the wrist forwards and backwards greatly reduces the dexterity of the arm. The ability to flex the wrist forwards and backwards is another feature many users desire [18].

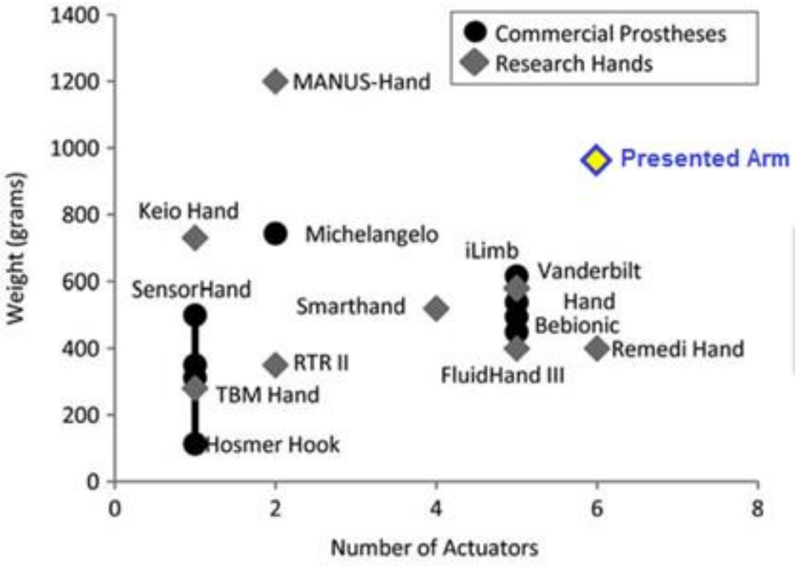


Figure 5.3 – Comparison of prostheses weight and actuation

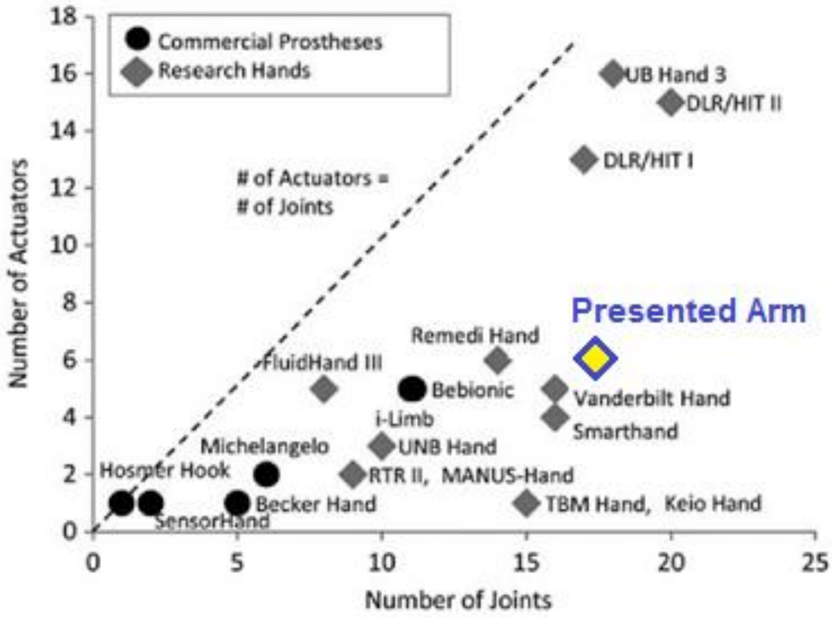


Figure 5.3 – Comparison of prostheses actuation and joints

While the Michelangelo hand is superior in terms of low weight and high grip force we can see that it is quite poor in terms of its number of actuators and joints. The above plot outlines how commercial prostheses are generally simpler devices in terms of the number of movements the device can offer. This is due to the limited forearm and palm space which must contain the entire system as well dealing with requirements on weight and battery life.

It is worth noting that the prosthetic arm presented in this thesis is completely portable and self-contained – unlike the Vanderbilt and UB Hand 3 for example which require large external areas for actuators and drive systems.

Using servo motors to drive artificial tendons is not an ideal solution. Servo motors are too large and too loud to provide the best solution. Also, for tendons to provide practical finger strengths they need to be tensioned by powerful actuators.

Using DC motors and custom gear drives instead of an artificial tendon design can greatly reduce the size of the drive system while maintaining strength.

5.7 Reliability

The developed system is only a prototype so is not expected to maintain its characteristics and performance for a long time. However, it is important to take care to develop a reliable prototype that can be used for future research. If any components of the device ever break they can be easily replaced or reprinted.

5.8 User Control

Rudimentary EMG allows for control of the arms movements at a basic level. The control methods implemented in this thesis illustrate the proof of concept for myoelectric control of 3D printed prostheses. More sophisticated control is necessary to be of any practical use to an amputee.

Proportional control of finger positions with flex intensity was implemented but for reasons discussed in the results section was not implemented in the final version. The EMG signals used by the controller need to be filtered and smoothed before being used to directly control servo positions. Some basic logic would also have to be implemented to optimise servo performance.

Advanced EMG control should allow the user to proportionally control several movements of the arm simultaneously.

5.9 Improvements

Mechanical Design & Manufacturing

This thesis has only experimented with ABS plastic for 3D printing. Nylon can be used for 3D printing as well and offers significantly more strength than ABS. Reviews from 3D printing specialists show clear advantages in strength Nylon has over ABS [31].

Unfortunately the UP 2 printer does not fully support the use of Nylon filament. A future student wishing to alter the mechanical design would be advised to use a more advanced 3D printer available throughout the AMME school. A range of 3D printed materials could be used to optimise the device – such as custom 3D printed rubber fingertip grips.

The joint-linked finger actuation design discussed in the literature review would offer greater strength and reliability compared to an artificial tendon network. However, designing the small intricate gears would be challenging and problematic to print.

A future student wishing to experiment with other actuation systems is advised to print and assemble the open source prosthetic arm available on the InMoov website. This project is pushing 3D printed prostheses to the limits and has even developed a joint-linked actuation system driven by small electric motors in the palm.

Finally, the design of a wrist joint which allowed for two or three degrees of rotational freedom would be a design of significant value. The current wrist design can rotate but cannot flex back and forth. This means there is no joint absorbing and dampening the shock from an impact to the hand. The flexibility that would come from a multiple DOF wrist joint would greatly improve the overall impact resistance of the device.

Pressure sensor feedback

One of the biggest limitations of this device is its lack of feedback. Pressure sensors could be used to provide fingertip pressure feedback, which would allow the fingers to automatically conform to specific object shapes. Feedback would also facilitate even and controlled fingertip pressure and would reduce the strain on the servos when trying to grasping an object.

Pressure sensors would also allow vibration motors to be controlled depending on the measured fingertip pressure. Vibro-tactile feedback to the user is a basic form of sensory substitution which the majority of myoelectric prostheses users find useful [18].

Custom PCB/Servo Controller

A custom printed circuit board offers much greater reliability than prototyped electronics. If the circuitry becomes more complex by incorporating several pressure sensors and vibration motors a custom PCB will be necessary to produce a board small enough to fit inside the device.

The school of AMME is expecting to receive an electronic 3D printer. It is advised that future students experiment with this machine before sending any PCB designs to be made. It may also be useful to use a servo controller board which would reduce the overhead on the microcontroller and remove the need for the voltage regulators.

Advanced EMG Algorithms and Processing

Future students are encouraged to develop more advanced EMG control algorithms which facilitate the control of various movements of the arm simultaneously. Such a control system would have to offer some kind of calibration routine to allow the system to adjust to the myoelectric signals of each individual user.

Chapter 6.0 – Conclusion

The academic goals of this thesis were initially uncertain and certainly change throughout the course of the year. The initial aim was to develop a low cost 3D printed myoelectric prosthetic arm. The goals and expectations for this thesis have been achieved and it is hoped that the presented body of work allows for several new thesis topics to be researched in the future.

6.1 Overall System Performance

The final system provides relatively good performance and characteristics for a prototype 3D printed model. The device is fast and responsive to electro-myography user input but offers limited strength. Over the course of testing the system has proven to be reliable and has required minimal maintenance since being assembled.

The biggest downfall of this design is its lack of toughness. Certain regions such as the wrist are at a high risk of breaking if the device is subject to moderate forces. In the real world a practical prosthetic arm must be able to absorb sudden shocks and support heavy loads without failing. Ways to improve the strength and toughness have been discussed in the previous results section.

6.2 Benefits to an Amputee

At this stage the presented prosthetic arm is not at a state where it can be used by an amputee – it is more so a low cost bionic arm.

With the design of a proper socket connection the possibility exists for the University to arrange collaboration with a medical institute to allow the device to be tested and used by amputees. Such testing would be invaluable in analysing and improving the devices performance.

6.3 Implications & Contribution to the Field

The presented device provides a platform for future research by final year engineering students to develop and test advanced prosthetic designs such as sophisticated EMG control algorithms, integrated pressure feedback and other advanced bio-mechatronic concepts and designs.

With future growth of the 3D printing industry advanced printers and materials will allow students to develop more ‘commercial-like’ prosthetic devices – robust and durable systems that could benefit a wide range of peoples with a missing limb.

With ongoing research improvements will hopefully lead to a system that is more durable and offers improved dexterity and control. Perhaps a future design will someday benefit amputees and improve the quality of people’s lives.

References

[1] Belter, Joseph T,M.S., B.S., J. L. Segil, Dollar, Aaron M, PhD,S.M., B.S. and R. F. Weir PhD. **Mechanical design and performance specifications of anthropomorphic prosthetic hands: A review.** ***Journal of Rehabilitation Research and Development*** *50(5),* pp. 599-618. 2013. Available:

[2] Europen Commission community research, Cognitive Robotic Systems. DEXMART. **DEXterous and autonomous dual-arm/hand robotic manipulation with sMART sensory-motor skills: A bridge from natural to artificial cognition**. February 2, 2009

[3] Tim Taylor. Muscles of the Hand and Wrist. Inner Body org.
http://www.innerbody.com/image_skel13/ligm27.html#full-description

[4] Hands Facts and Tivia. **The electronic textbook of hand surgery**. http://www.eatonhand.com/hw/facts.htm

[5] RSL Steeper. **Bebionic 3 Technical Information**. 2014. http://bebionic.com/

[6] Touch Bionics. **i-limb Digits Clinician User Manual**. 2014. http://www.touchbionics.com/

[7] George ElKoura, Karan Singh. **Handrix: Animating the Human Hand**. Department of Computer Science, University of Toronto, Toronto, Canada Side Effects Software, Inc., Toronto, Canada. Eurographics/SIGGRAPH Symposium on Computer Animation (2003)

[8] Lillian Y. Chang and Yoky Matsuoka. **A Kinematic Thumb Model for the ACT Hand**. The Robotics Institute, Carnegie Mellon University. Proceedings of the 2006 IEEE International Conference on Robotics and Automation.

[9] Christian Pylatiuk, Stefan Schulz and Leonhard Döderlein. **Results of an Internet survey of myoelectric prosthetic hand users**. Prosthetics and Orthotics International, Sage Publications Dec1, 2007

[10] Myoelectric Bebionic 3 bionic hand. CNET, November 2012.
https://www.youtube.com/watch?v=KCIpbRSMfGM

[11] Tuomas E. Wiste, Skyler A. Dalley, Thomas J. Withrow *Member, IEEE*
and Michael Goldfarb, *Member, IEEE.* **Design of a Multifunctional Anthropomorphic Prosthetic Hand with Extrinsic Actuation**. 2009 IEEE 11th International Conference on Rehabilitation Robotics, Kyoto International Conference Center, Japan, June 23-26, 2009.

[12] Claudio Melchiorri, Gianluca Palli, Giovanni Berselli, and Gabriele Vassura. **Development of the UB hand IV**. Bologna University, *Overview of Design Solutions and Enabling Technologies, September 2013.*

[13] Frank R¨othling, Robert Haschke, Jochen J. Steil, and Helge Ritter Neuroinformatics Group, Faculty of Technology, Bielefeld University. **Platform Portable Anthropomorphic Grasping with the Bielefeld 20-DOF Shadow and 9-DOF TUM Hand**. Proceedings of the 2007 IEEE/RSJ International
Conference on Intelligent Robots and Systems San Diego, CA, USA, Oct 29 - Nov 2, 2007.

[14] Shadow Robot Company. Shadow Dextrous Hand, 2014.
http://www.shadowrobot.com/products/dexterous-hand/

[15] Weir, R., Mitchell, M., Clark, S., Puchhammer, G., Haslinger, M., Grausenburger, R., Kumar, N.,

Hofbauer, R., Kushnigg, P., Cornelius, V., Eder, M., Eaton, H3 Wenstrand, D.**THE INTRINSIC HAND – A 22 DEGREE-OF-FREEDOM ARTIFICIAL HAND-WRIST REPLACEMENT**. Measuring Success in Upper Limb Prosthetics," Proceedings of the 2008 MyoElectric Controls/Powered Prosthetics Symposium, held in Fredericton, New Brunswick, Canada, August 13–15, 2008.

[16] Erkan Kaplanoglu. **Design of Shape Memory Alloy-Based and Tendon-Driven Actuated Fingers Towards a Hybrid Anthropomorphic Prosthetic Hand**. International Journal of Advanced Robotic Systems, July 2012.

[17] Patrick Maudsley. **Shape Memory Alloy (SMA) Robotic Hand - University of Utah Mechanical Engineering**. May, 2009. https://www.youtube.com/watch?v=zQih9tLbEzo

[18] *Chris Lake.* ***Chapter 14:*** **Partial Hand Amputation: Prosthetic Management**. *American Academy of Orthopaedic Surgeons*.
[19] Brooker, Graham. **Introduction to Biomechatronics**. Scitech publishing, 2012.

[20] Gael Langevin. **InMoov, Open source 3D printed life size robot**. 2014. http://www.inmoov.fr/

[21] David Senger. **MIT PhD Student Invents New Prosthetics for Amputees**. THNKR, Nov, 2012. https://www.youtube.com/watch?v=2-6anyWQAME

[22] Dr. Scott Day . **Important factors in Surface EMG measurement**. Bortec Biomedical Ltd

[23] D P J Cotton, A Cranny, P H Chappell, N M White and S P Beeby. University of Southampton. **Control Strategies for a Multiple Degree of Freedom Prosthetic Hand**. Electronic Systems Design Group, School ofElectronics and Computer Science, University of Southampton.

[24] R. F. Weir, P. R. Troyk, G. DeMichele, T. Kuiken. **Implantable Myoelectric Sensors (IMES) for Upper-Extremity Prosthesis Control - Preliminary Work**. Proceedings of the *25'* Annual lntemational Conference of the IEEE EMBS Cancun, Mexico * September 17-21.2003.

[25] Rebecca Boyle. A Marine With A Prosthetic Hand Controlled By His Own Muscles. Australian Popular Science, Jan, 2014. http://www.popsci.com.au/tech/video-a-marine-with-a-prosthetic-hand-controlled-by-his-own-muscles,380649

[26] Danielle M. Rager, *Student Member IEEE*, Darren Alvares*, Student Member IEEE*, Ingvars Birznieks, Stephen J. Redmond, *Member IEEE*, John W. Morley, Nigel H. Lovell, *Fellow IEEE*, Richard M. Vickery. **Generating Tactile Afferent Stimulation Patterns for Slip and Touch Feedback in Neural Prosthetics**. 35th Annual International Conference of the IEEE EMBS Osaka, Japan, 3 - 7 July, 2013.

[27] CBS News. Breakthrough: Robotic limbs moved by the mind. Dec, 2012. http://www.cbsnews.com/videos/breakthrough-robotic-limbs-moved-by-the-mind/

[28] i.materialise. **ABS 3D Printing Design Guide**. 2014. http://i.materialise.com/materials/abs/technical-specifications

[29] exrx.net. Body Segment Data. http://www.exrx.net/Kinesiology/Segments.html

[30]Howida A.Shedeed, Mohamed F.Issa, Salah M.El-sayed. **Brain EEG Signal Processing For Controlling a Robotic Arm**. Nov, 2013.

[31] 3D Universe. **Review of Taulman 3D's Bridge Nylon Filament for 3D Printing (3D Universe).** Uploaded to Youtube March, 2014.

[32] Weir RF. **Design of artificial arms and hands for prosthetic applications**. Standard handbook of biomedical engineering and design. New York, McGraw-Hill; 2003. p. 32.1–32.59.

[33] Waryck B. **Comparison of two myoelectric multi-articulating prosthetic hands**. MEC Symposium; 2011 Aug 14–19; New Brunswick, Canada.

[34] Dhillon GS, Horch KW. **Direct neural sensory feedback and control of a prosthetic arm**. IEEE Transactions on Neural Systems and Rehabilitation Engineering 13, 468-472.

[35] Hudgins, B Parker, P and Scott. **A new strategy for multifunctional myoelectric control.** 1993. IEEE Transactions on Biomedical Engineering

Appendix

A.1 Assembly Guide

Removing support material

To remove support material more easily the small finger components were printed with the fan heat vent open. Large components are printed with the fan vent closed to produce stronger layer bonding.

Certain components were given a quick wipe down with acetone. This dissolves the surface of ABS components improving strength and surface finish.

Fingers

Holes for the pins were drilled with a 3mm drill bit to improve diameter accuracy. 3mm polypropylene pins were used as the pins for the finger joints.

A long length of braided line (about 60cm) was looped through the fingertip pieces at least twice. A drop of super glue was applied to the locking point to firmly anchor the tendon. This is necessary to make sure the tendons firmly lock at the tip of each finger and do not slip when tensioned.

Care has to be taken when feeding the tendons through the wrist so that lines do not cross over each other

Wrist

Rotate the servo to a completely counter clockwise position. Carefully press fit the palm section onto the servo shaft – making sure we have a palm up position when the servo is lying flat. Use a small servo screw to better attach the palm to the servo shaft.

Forearm

The two large forearm sections were carefully aligned and glued together.

Servos were screwed down into position using small servo screws. Each servo motor was sent a 0.8ms pulse width. This completely rotates the servo in one direction – note that other servo models may require a short or longer pulse to reach this position.

The custom servo horns were then press fitted onto the servo shafts so that the tendon holes are pointing towards the elbow. The fingers are placed in an open position and the two tendons from each finger are then connected to their corresponding servo horn.

Each of these tendons will pass through the two holes in the servo horn and need to be tied to produce to hold tension in the tendon lines.

Pull on these two tendon lines to find the one which closes the finger. Since the servo is completely rotated to one side, this tendon must be passed through the hole which will pull on the line and cause it to wrap around the servo horn.

Pull on the tendons and using a pen, mark the point where the tendons just pass through the servo horn openings. Remove the servo horn from the shaft. Slide the horn down the tendons and tie thick knots at the pen markings on the tendons. When the servo horn is placed back on the shaft both tendons should be tight.

Elbow

Apply a few drops of glue to the extruding cylinders at the base of the large gear and firmly press it into its forearm slot.

The locking pins in the elbow use the same 3mm polypropylene used in the finger joints.

Removing the internal servo potentiometer and slotting it into the forearm gear is an optional step which will increase the rotational range of the elbow section.

The potentiometer mounting has been designed for TowerProMG996r servo. There is a possibility that potentiometers from other servo motors will not fit inside this housing.

Press fit the small gear onto the elbow servo shaft. Attach this servo to is position in the bicep section using small servo screws. Connect the bicep to the forearm using the designed 3D printed pin locks.

If further assistance with assembly is required contact

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mahdihussein.91@gmail.com
0423 936 106

A.2 Datasheets

- Voltage Regulators – AP1117
  http://diodes.com/datasheets/AP1117.pdf

- Microcontroller – PIC18525k22
  See Microchip website

- Servos – TowerProMG996r
  https://www.hobbyking.com/hobbyking/store/__6221__Towerpro_MG996R_10kg_Servo_55g_10kg_20sec.html

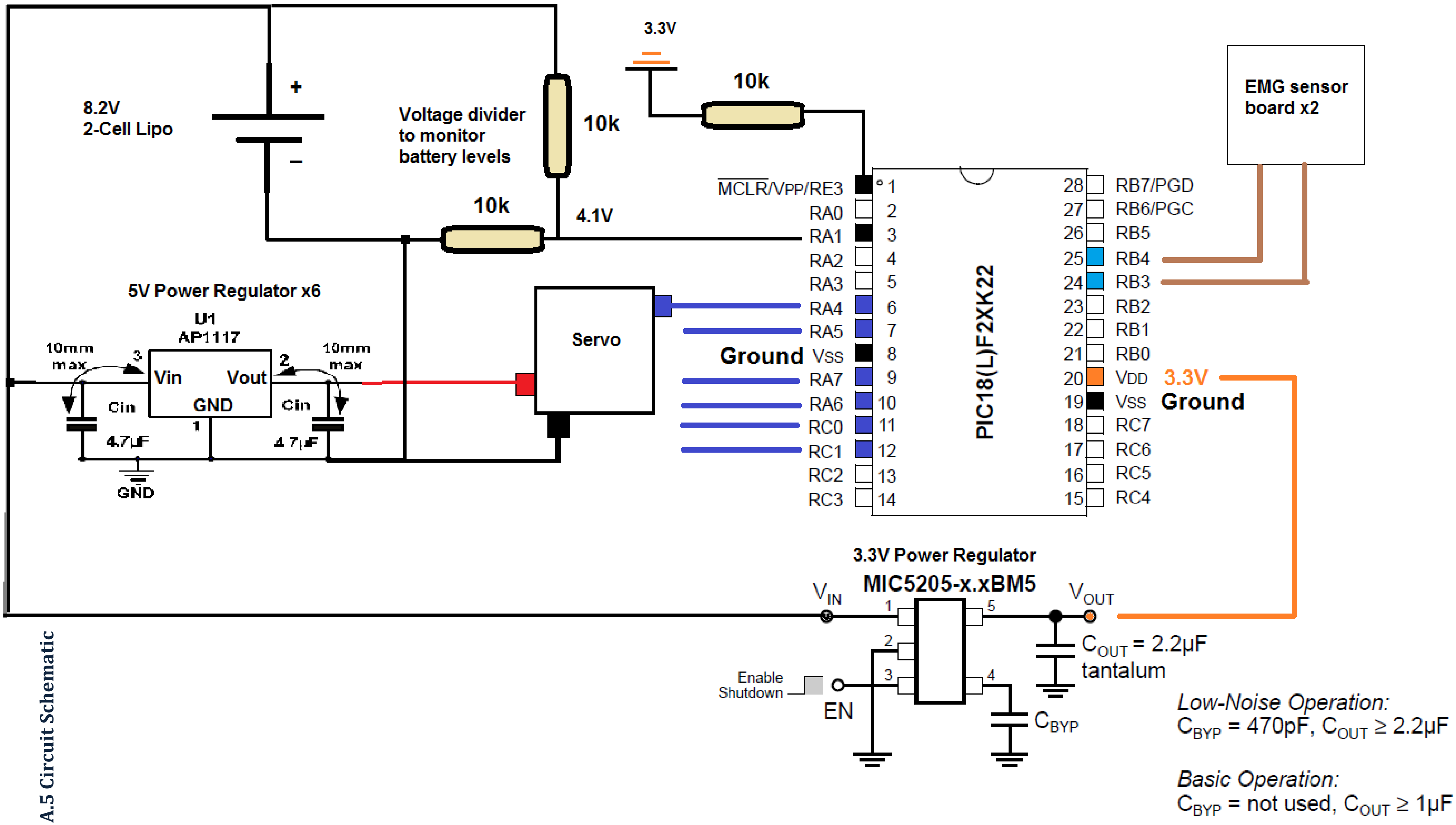
A.3 Digital Appendix

The digital appendix can be found on the disk provided.

- Videos
- Solidworks Files
- Printable STL Files
- Software CODE

A.4 Mechanical Assembly
Exploded View

A.5 Circuit Schematic



Notes