Investigation of Upper Limb Prosthesis Functionality using Quantitative Design Tools

Chung Han Chua

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Abstract

Upper limb prostheses offer those with limb loss a solution to restore some of their lost functionality by allowing them to participate in bilateral tasks, especially those required for daily living. Whilst there is a wide range of upper limb prostheses available, there remain high device rejection rates. Low functionality and discomfort are major factors in prosthesis rejection, which had been identified as challenges more than 60 years ago. These issues have not been effectively addressed due the lack of design tools for engineers and clinicians. Upper limb prostheses have seen greater technological advances than the methods to evaluate them effectively, which has resulted in over-engineered designs which do not meet the needs of their user.

In this thesis, I aim to improve future upper limb prostheses through the development of three design tools. These design tools seek to quantify the functionality of prosthetic devices using motion capture analysis, virtual environments, and joint optimisation. By developing these tools, there is greater opportunity to optimise prostheses earlier in the design cycle which can result in improved functionality. It is anticipated that improvements in functionality will increase user satisfaction and therefore reduce device rejection rates

Motion capture analysis was used to study the compensatory movements that arise from operating an upper limb prosthesis. Using a motion capture suit, the motor strategy of a participant was compared between using their biological hand and using a prosthesis through the use of an able-bodied adaptor. It was found that the shoulder and trunk had to make the most compensatory movements to complete several grasping tasks due to the lack of degrees of freedom at the distal end of the prosthesis. Without forearm supination/pronation and wrist extension/flexion, the participant had to approach the grasping tasks from a different angle, sometimes having to lean backwards and abduct their upper arm. The methodology of utilising a motion capture suit as a design tool to quantitatively assess the compensatory movements caused by a prosthetic device was successfully demonstrated.

Virtual environments, in conjunction with quantitative grasp quality metrics, can be used to assess the performance of the upper limb prosthesis extremity alone, uninfluenced

by user bias. A dynamic virtual environment is presented to simulate several grasping tasks with five upper limb prosthetic devices. Contact information from these grasping tasks are used to calculate the quality of the grasp and provide an overall grasping functionality score. From the simulation results, it was found that more degrees of freedom do not necessary equate to better grasping performance. The positions of force vectors during grasp formation are vital and they must be well- balanced in order to result in stable grasps. Simulated grasping and quantitative analysis in a virtual environment has been demonstrated, which can be used to better plan grasping paths and therefore improve the grasping functionality of upper limb prosthesis designs.

Prosthesis users desire their devices to have a low mass, have a low cost, and have high functionality. However, these are conflicting design objectives and decisions must be made to which design considerations to prioritise. A multi-objective model was used to balance these three objectives and select the most suitable components that make up a prosthesis. A modularity scheme was used to divide an upper limb prosthesis into three categories: socket, forearm, and terminal device. In each category, several components were considered which can either be manufactured by conventional engineering or additive manufacturing. Each component would provide a unique value determined by a several quantitative utility functions. Based on satisfaction studies in the literature, the multi-objective optimisation model found that a Split Hook terminal device with an additively manufactured socket and forearm was the optimal design as it provided a low mass and excellent grasping functionality. This model has been demonstrated to work with different user requirements to intelligently select the most appropriate upper limb components within the modularity scheme.

Overall, methods were developed which covered aspects of prosthesis design from clinical testing of prosthetic devices, functionality assessments of Computer Aided Design models, and intelligent selection of prosthesis components for individual requirements. It is hoped that these design tools may enable better communication between engineers and clinicians to ensure that users receive devices that are to their satisfaction.

Acknowledgements

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I am deeply grateful to my mother and father who have supported me undertaking my studies at the University of Nottingham. They have always told me that I could be whoever I wanted to be, as long as I was the best at it. So hopefully by selecting a niche field of research I have reached their expectations.

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Publications

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Abbreviations

3DPackRat	3D Packing Research Analysis Tool
ABA	Able-Body Adapter
ADAMS	Automated Dynamic Analysis of Mechanical Systems
ADLs	Activities of Daily Living
AM	Additive Manufacturing
ASAP	Automated Scheduling, Optimisation and Planning
BBT	Box and Block Test
CAD	Computer Aided Design
CE	Conventional Engineering
CfAM	Centre for Additive Manufacturing
DCP	Distribution of Contact Points
DoAs	Degrees of Actuation
DoFs	Degrees of Freedom
EMG	Electromyography
FFF	Fused Filament Fabrication
GII	Grasp Isotropy Index
JTHFT	Jebson Taylor Hand Function Test
MPL	Modular Prosthetic Limb
NHPT	Nine Hole Peg Test
RAPID	Rapid and Accurate Polygon Interference Detection
SHAP	Southampton Hand Assessment Procedure
SLS	Selective Laser Sintering
SMW	Smallest Maximum Wrench

Chapter 1
Introduction

1.1 Background

When a person loses an upper limb, there are significant lifestyle changes they must make to overcome the severe reduction in productivity in their work, social and daily activities[1–3]. The quality of life of people with upper limb loss is reduced as they lose the ability to reach, grasp, touch and gesture. The human hand is an intrinsic part of the body with a wide range of functionality, capable of gross and fine motor skills that are essential for independence. Prosthetic devices can be used to lessen the impact of their limb loss and restore some of their lost functionality.

A prosthetic device can be defined as an artificial extension that replaces or augments a missing or impaired part of the body[4]. There are four types of upper limb prosthetic devices which are cosmetic, functional, body powered, and externally powered[2]. These are either categorised as a passive or active prosthetic device, depending on how they are powered, as shown in *Figure 1* below.

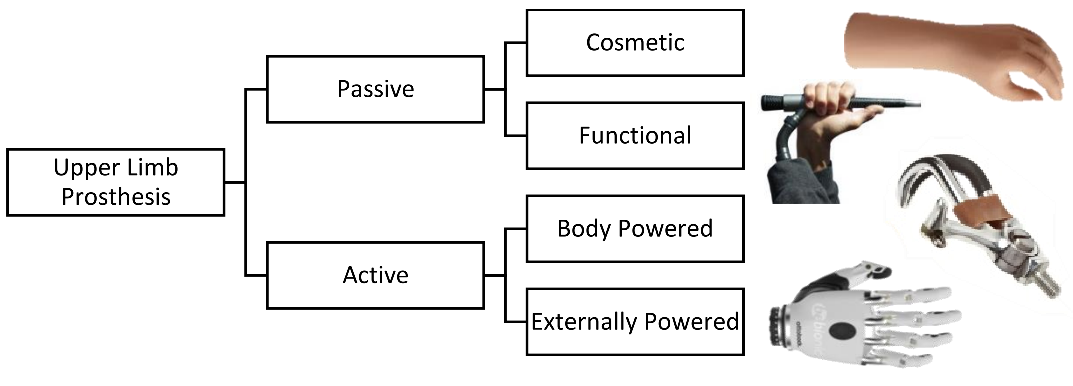

Figure 1: Upper Limb Prosthesis Categories[5–8]

Passive prosthetic devices are defined by their limited Degrees of Freedom (DoFs) and lack of Degrees of Actuation (DoAs)[4]. A DoF is defined as the number of independent parameters that define the state of the mechanical system, whereas a DoA is defined as the number of DoF that can be actively controlled. Cosmetic prostheses primarily aim to replicate the aesthetics of the missing limb[9]. Special plastics and pigments are used to match the skin tone of the user to make it look as natural as possible. Functional prostheses are designed to facilitate specific activities and maximise function during those specific tasks[4]. These are commonly designed for sporting activities that require a precise body motion which cannot be achieved with a traditional prosthesis. For example, *Figure 1* shows a golf functional prosthesis that allows users to deliver a golf swing with the correct driving performance[6]. Both types of passive prosthetic devices (cosmetic and functional) perform very well for their intended function but provide limited usefulness in other activities, especially those seen in daily life, due to their lack of grasping. Activities of Daily Life (ADLs)[10] are basic tasks that involve caring for one's self and body; examples include managing essential physical needs, grooming/personal hygiene, dressing, and eating.

Active prosthetic devices are defined by having at least one DoA that provides a prehensile grasp. Their capability to interact with a large range of objects makes them suitable for performing ADLs. Body powered prostheses are actuated by moving another part of the user's body to open and close a terminal device[11]. Typically, a terminal device consists of a claw with a single joint that is controlled by biscapular abduction[4]. A harness is strapped around the user's healthy shoulder and is connected to the claw with a cable. When the user moves their forearm forward, this causes the cable to be pulled which causes the claw to open. Body powered prostheses are the most used upper limb prosthesis today[12] as they are relatively lightweight, straightforward to operate, and provide a form of force feedback. These prostheses do however suffer certain disadvantages, such as low mechanical efficiencies which result in low grasping strength. They require the user to have the strength and range of motion to operate the harness cable and are limited to a single DoA which restricts what activities can be performed[13].

Externally powered prostheses use several motors that are powered by an external battery to form a prehensile grasp[14]. Typically, externally powered prostheses have an anthropomorphic form with each digit powered by a motor, thus allowing them to have multiple DoFs and to form a variety of prehensile grips. There are two main externally powered control systems: electric and myoelectric devices. Electric prostheses are controlled through external buttons, usually located on the device itself. Therefore, this requires the use of another limb to operate it which can be tedious and cumbersome. In contrast, myoelectric prostheses are controlled through electromyographic (EMG) signals detected by sensors located on the residual limb[15]. Even with the loss of a limb, the remaining muscles can generate an electric potential when activated neurologically. Users can learn to control the remaining muscles in their residual limb and trigger specific EMG signals that are mapped to a prehensile grasp. The consequence of having motors and sensors is that externally powered prostheses are the heaviest devices available[16]. This excessive weight, combined with the tight fit required for sensor contact, causes discomfort to the user which ultimately leads to abandonment. Despite being capable of multiple prehensile grasps, there is an issue with grasp stability, precision, and handling of small objects. With the high costs associated with upper limb prostheses, especially myoelectric devices, users expect a high degree of functionality which is not currently being fulfilled. Weighing the pros and cons, users often disagree that any prosthetic hand is inherently better than no hand[17].

Despite the large range of upper limb prostheses available there are high device rejection rates[9,16,18]. Rejection rates are calculated by dividing the number of people who no longer use their device by the number of people who have tried it at some point in their lives. Studies have shown that body-powered prostheses have the highest rate of rejection at 65%, followed by externally powered prostheses at 51% and passive prostheses at 47%[9]. For active devices, functionality and comfort have been ranked as the most important factors in upper limb prosthesis acceptance. Whilst active prostheses can perform ADLs to a certain extent due to their increased DoF, their grasping performance is too limited and unreliable. Prosthesis users have to spend extra effort to ensure that their device is positioned perfectly in order to ensure a secure grasp. It is commonly observed that prosthesis users make compensatory movements with their back and shoulders in order to perform particular tasks[10]. The weight of externally powered upper limb prostheses tends to be heavier than a human limb due to the use of motors, sensors and batteries which can also hinder natural movement. An upper limb prosthesis that is uncomfortable, painful to wear for extended periods of time, and a hindrance will likely be abandoned. In a study consisting of 252 upper limb prosthesis users[16], it was found that 98% felt they were just as functional without the device, and 95% were more comfortable without it.

The challenges of upper limb prosthesis design faced 60 years ago are still the same today[2]. The poor grasping performance and therefore poor execution of ADLs is an ongoing issue which has not been effectively addressed in modern prostheses. Whilst efforts have been made to improve the mechanical design of prostheses, the standard method of evaluation does not provide a quantitative measure to improve further iterations. The standard method of performance evaluation is through the use of aftercare surveys[19], where the demands, expectations, and satisfaction of prosthesis users are obtained. It is assumed that if a client is satisfied with their device then it has high functionality. However, these surveys are qualitative in nature and the results can be subjective. A more quantitative measure for upper limb prosthesis performance is required to identify and understand which design features are beneficial to the user. Whilst there are a few quantitative evaluation methods available that use time based activities, they have their limitations and oversimplify the performance of an upper limb prosthesis[20,21]. They are highly dependent on user skill and do not provide an informative measure for a designer to improve on. This can make it difficult to design an upper limb prosthesis which meets the user's actual requirements.

The design of upper limb prosthetic devices is greatly limited by its current design process. Despite the large range of devices available, there are high rates of rejections across all types. A common theme across modern devices is over-engineering which does not address the true needs of their users. The design for functionality is restrained by the methods of evaluation currently employed which do not accurately represent the ability of the device. Typical design cycles have long feedback loops, where physical models must be built in order to evaluate their performance. However, this results in fewer design iterations and product refinement. Better design tools are needed to enable active communication between engineers, clinicians, and users to produce highly functional devices.

This thesis addresses current weaknesses in the design cycle of upper limb prosthetic devices. Smaller design feedback loops can be achieved using a virtual environment, where the performance can be numerically assessed early in development. Furthermore, models have been created from motion capture data to predict compensatory movements in order to inform design. Trade-offs between design and cost need to be balanced to produce a prosthesis that is financially accessible. An optimisation architecture is introduced for an Additive Manufacturing based workflow that combines production time, cost, and functionality to provide bespoke designs. Such design tools could result in lower rejection rates for upper limb prostheses, therefore improving the quality of life of prosthesis users.

1.2 Thesis Layout

A description of the content of each chapter of this thesis is listed below.

Chapter 2 – Literature Review I: Design of Upper Limb Prosthetic Devices

In this chapter, a brief history of upper limb prosthesis design is given with a focus on technological events which has advanced the field. Key design features of upper limb prosthesis are discussed and examined in both commercial and academic devices. The operation and mechanics of these designs are described to highlight the capabilities of the current state of the art and note key areas for improvement. An overview of the current evaluation measures used in upper limb prosthesis performance assessment are given. The methods have been categorised into three subsections: time-based tasks, computational models, and motion capture analysis.

Chapter 3 – Research Methodology

In this chapter, the significance and novelty of this research is highlighted. The aims and objectives of this research are stated, and the methodology used to achieve them is discussed in detail.

Chapter 4 – Characterising Compensatory Movements using Motion Capture

In this chapter, work carried out using motion capture systems to measure compensatory movements of the upper body is described. This work was done to gain a better understanding of the combined motion of the upper body in order to improve future prosthesis design to prevent overuse injuries.

Chapter 5 – Using Virtual Environments for the Evaluation of Upper Limb Prostheses

This chapter describes a virtual environment that is used to assess the grasping performance of five prosthetic devices. Simulations of the Southampton Hand Assessment Procedure are partially carried out and the quality of the grasp is assessed by several numeric grasp quality metrics. This provides an in-depth quantitative measure of grasping which can be optimised to inform design specifications.

Chapter 6 – Multi-Objective Optimisation of Mass, Cost, and Functionality

In this chapter, a multi-objective optimisation model is described that manages the design of upper limb prostheses using both conventional and additive manufacturing. This scheme allows conflicting design objectives to be jointly optimised and can be used as an intelligent selector for prosthesis components, resulting in a complete device that matches the user’s requirements.

Chapter 7 – Discussion

This chapter describes the main outcomes of using motion capture analysis and virtual environments for the evaluation of upper limb prostheses and multi-objective optimisation to improve design workflow. Their usefulness as design tools is discussed in the context of the state-of-the-art to highlight their significance, limitations, and novelty.

Chapter 8 – Conclusions and Future Work

In this chapter, the main conclusions of this research are highlighted and recommendations for future work are given.

Chapter 2
Literature Review
Design and Evaluation of Upper Limb Prosthetic Devices

2.1 Introduction

The human hand is capable of sophisticated movement due to its 22 Degrees of Freedom (DoF), 35 muscle actuators, and over 17,000 mechanoreceptor sensors[22]. The hand's prehensile abilities allow it to perform tasks from fine digit manipulation to the handling of heavy objects. In addition, the structure of the upper limb, as shown in Figure 2, provides abduction/adduction and flexion/extension of the arm, flexion/extension and pronation/supination of the forearm, and wrist flexion/extension which provides flexible positioning for the hand. With upper limb loss, either congenital or from amputation, there are detrimental impacts on the quality of life which an upper limb prosthesis may be able to lessen. Prosthetic devices are tools that allow their users to gain some of the lost functionality of their missing limb. However, even state of the art designs are not a true limb replacement. There are strict weight, power, and size constraints which limit the design of prosthetic devices[23], along with ever growing demands from users to increase their functionality[2].

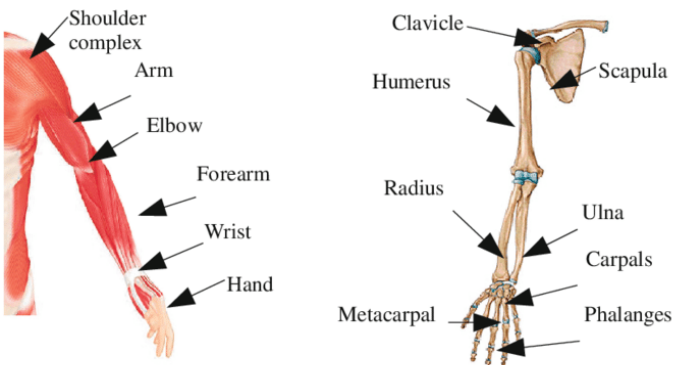

Figure 2. Upper limb anatomy.
Left: Body Segments. Right: Main Bones[24].

Measuring the functionality of an upper limb prosthesis can be challenging. Functionality as a concept is difficult to measure, and it has been typically assumed to correlate with user satisfaction in upper limb prosthesis users. The current standard of prosthesis evaluation is through patient and prosthetist interactions, where after-care surveys are usually employed to assess the satisfaction of the prosthesis user. If they are satisfied with their device, it is assumed that the prosthesis has a high functionality[19]. In reality, functionality is multi-faceted and can be further broken down into specific types, such as grasping functionality and motion strategies.

This chapter describes the design of upper limb prosthetic devices and various quantitative methods available to measure their functionality.

2.2 Design of Upper Limb Prosthetic Devices

In this section, a brief history of upper limb prostheses development is examined to show the progress that has been made in prosthetic design. Key design aspects that are commonly linked with user satisfaction are explored such as grasping functionality, weight, and aesthetics. Upper limb prosthetic devices from the market and research are identified and examined. The operations and mechanics of these devices are discussed to highlight the capabilities of the current state of the art and identify in what areas these prostheses are lacking.

2.2.1 History of Upper Limb Prosthesis Design

Throughout history, the majority of upper limb amputations occur in working age males, often from occupational or combat trauma[25]. The first documented prosthesis was in 218 BC[23], where Roman emperor general Marcus Sergius lost his right hand during the second Punic War. A passive prosthesis made of iron was constructed after his amputation that allowed him to wield a shield and return to the battlefield. One of the earliest prosthetic devices that featured articulation was seen in the early 16th century, wielded by the German knight Götz von Berlichingen[26]. After Götz lost his hand during the Siege of Landshut in Bavaria, an artisan built him an iron hand with independently controlled digits that were capable of flexion and extension of the metacarpophalangeal, proximal interphalangeal, and distal interphalangeal joints. It was driven by racks and recalled by springs which allowed Götz to make a range of prehensile grasps, thus enabling him to hold reins, grip weapons and return to battle. In the mid-16th century Ambroise Paré[27], a French military surgeon, designed a hand with digits that could be operated independently with levers and gears. The fingers were brought into flexion by four springs fixed in the palm and kept in position by sprockets which provided a very firm and reliable grasp on objects. These prosthetic devices were highly complicated designs at the time and only the very wealthy could afford them. However, these prosthetic devices had to be operated by the gross movements of a contralateral healthy hand, which limited it to unilateral activities.

It wasn’t until 1818 that Peter Baliff[28], a German dentist, pioneered the first transradial body-powered upper limb prosthesis. By using leather straps between a terminal device and a shoulder harness, Baliff’s device could be used to control the strap tension to open and close a spring-loaded hook. This enabled prosthesis users to perform bilateral tasks and gave a greater degree of independence. However, this was still limited by the condition of the residual limb as operating a body-powered upper limb prosthesis would put pressure onto the sensitive skin[28]. With the introduction of anaesthetic in 1846 and aseptic surgery in 1870, surgeons were able to prioritise cleaner amputations over surgery time, resulting in viable residual limbs that could support a prosthesis.

A significant proportion of the workforce was lost during the First World War, which prompted countries to equip amputees with affordable prosthetic devices to ensure they could still generate output. In Great Britain, amputees were given sockets and a universal terminal device that could hold a wide range of work tools[28], as shown in *Figure 3*. Industrial utensils were easily fixed to the ends of sockets where they operated as an extension of the arm. In the United States, the split hook was developed and patented in 1912 by D.W. Dorrance[29]. The split-hook was a body-powered device that was controlled by changing the tension in a cable between the terminal device and a shoulder harness, using the same mechanics of Baliff's device. It featured two prong hooks and a range of prehension surfaces that allowed it to produce tip, lateral, and cylindrical prehensile grasps[23]. Elastic bands held the two prong hooks together, where the relative movement of the user's shoulder and/or upper arm could overcome the force of the elastic bands and open the hooks. Whilst the design was simple, it was highly versatile and allowed users to complete tasks more efficiently[25]. In 1984, the bulky leather straps were replaced by the Bowden cable system with thinner and stronger cables. Even with a greater range of materials and manufacturing techniques, all modern body-powered upper limb prostheses have been adapted from the Bowden and split hook design. It is a durable, portable, and affordable design that enables users to have a sense of tactile feedback from sensing the cable's tension and to perform a wide range of motion quickly[11]. Unfortunately, prolonged use of a shoulder harness can be uncomfortable, execution of complicated motor tasks is limited, and the device has a poor aesthetic. Despite its shortcomings, the body-powered split hook is the most widely used upper limb prosthesis in the world today[12].

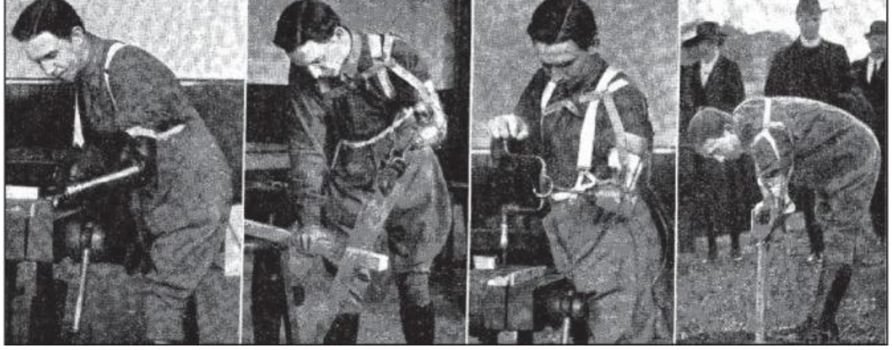

Figure 3. Interchangeability of the universal terminal device allowed amputees to quickly re-join the workforce[25]

In order to develop an upper limb prosthesis that did not rely on a harness, a breakthrough in the control mechanism was required. It was in 1948 that Reinhold Reiter, a physicist working for the Bavarian Red Cross, created the first myoelectric prosthesis[22]. He used a vacuum tube amplifier to boost the detection of surface electromyography (EMG) signals that were then used to control an electromagnet that would open and close a terminal device[30]. However, it was not portable as it required very large equipment to operate. The aim of his system was to allow a user to work more effectively at a workstation, rather than to aid everyday life. In 1960, the first portable myoelectric prosthesis was designed by Alexander Kobrinski[25], a Russian scientist. The use of transistors reduced the size of the control system to the point that the batteries and electronics could be worn on a belt, connected to the terminal device by wires. However, this device was still heavy with slow movements, weak pinch force, and wire connections that were easily damaged[31]. Interest in the development of powered prosthetics was expanded in the 1960s by what was considered one of the biggest man-made medical disasters; the severe birth malformations caused by the drug Thalidomide[30]. Pneumatic systems saw an increase in use due to the ease of control and low weight actuators, which were ideal for children with congenital limb loss. In the United Kingdom, the Hendon Arm[32] and the Edinburgh Arm[33] were one of the first few pneumatic multifunctional limb prostheses controlled by pneumatics, as seen in *Figure 4*. The Hendon Arm featured a position servomechanism control principle that allowed multiple grasping functions depending on how the device was positioned in relation with the user[32]. The Edinburgh Arm was controlled from the relative protraction-retraction and elevation-depression movements of the two shoulders[33]. However, the use of pneumatic systems never gained traction in the industry due to the difficulties of using compressed gas, which was expensive and difficult to distribute and maintain[30].

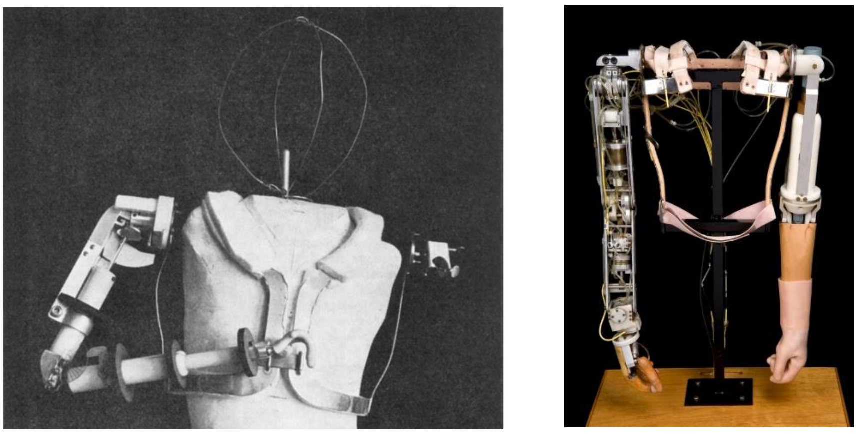

Figure 4. Early pneumatic multifunctional limb prostheses.
Left: Hendon Arm[32]. Right: Edinburgh Hand[34].

The first generation of portable batteries, electronic circuits using transistors, and small electric motors in the early 1970s gave rise to the Viennatone Myoelectric Hand. This was the first commercial prosthesis available in the United States as a result of the cooperation of Ottobock and Viennatone[35]. The myoelectric signals detected on the surface of skin were rectified and smoothed to produce a voltage that was roughly proportionate to the myoelectric power, allowing the user to control the amount of force exerted by the terminal device. The Viennatone Myoelectric Hand used a three-jaw chuck prehension mechanism that was controlled by a geared high-speed DC motor and was usually paired with a cosmetic silicone glove. Ottobock, as of 2020, offers the Sensor Hand which still has the same appearance as its predecessor with the same mechanics but with much improved batteries, motors, and materials. Advances in electrical components have enabled myoelectric prostheses to become commonplace in the market, where there are now dozens of designs available. Whilst upper limb prosthesis design has come a long way since the first commercial myoelectric device in the 1970s, some of the same design problems are still faced today. The following section discusses contemporary design considerations of upper limb prostheses in detail and describes the barriers and trade-offs in design optimisation.

2.2.2 Modern Upper Limb Prosthesis Design Considerations

The aim of an upper limb prosthesis is to improve the user's quality of life, where the main contributing attribute is the device's grasping functionality. The functionality of a prosthesis can be assessed by how closely it can reproduce the movements of the human hand. The movements of the human hand are categorised into two main groups: prehensile movements – "movements in which an object is seized and help partly or wholly within the compass of the hand"[36]; and non-prehensile movements – "movements in which no grasping or seizing is involved but by which objects can be manipulated by pushing of lifting motions of the hand as a whole of or the digits individually."[36]

Passive prosthetics with no Degrees of Freedom (DoF), which are primarily used for aesthetic reasons, can provide non-prehensile movements. An example of this would be using a passive prosthesis to hold and stabilise the position of a sheet of paper, whilst using the opposite intact hand to write. During unilateral limb loss, the remaining opposite hand becomes the dominant hand[37]. In these cases it is therefore not essential to provide a device that is capable of intricate movements when a prosthesis user would prefer to use their healthy hand instead[38]. However, passive prosthetics are not able to provide any grasping functionality for bilateral prehensile tasks, as a prehensile grasp requires at least two forces being applied in opposition to each other against an object.

The human hand's prehensile capabilities can be classified by discrete grasp taxonomies, which were first proposed by Schelsinger in 1919[39]. These grasps were associated with the shape of the objects: cylindrical, tip, hook, palmer, spherical, and lateral. These classifications lacked practical utility due to the reliance on the shape of the object held and did not consider the function of the grasp[11]. Napier concluded that all prehensile grasps could be categorised as either a power grasp or a precision grasp[36]. The power grasp provides the ability to handle heavier weights by utilising a wider contact surface formed by the hollow space between the palm and the digits in flexion. The palm and the proximal phalanges enhance the stability of the grasp. The proximal grasp provides the ability to accurately grasp small objects with low forces. The contact points are in the tips of the digits when the thumb is in opposition with the index and/or middle digits. In 1986, Cutkosky[40] extended Napier's grasping classification to show the relationship between the task force requirements and the object's attributes as shown in *Figure 5*. Cutkosky's discrete grasp taxonomy has served as the basis for grasping for many commercial prostheses.

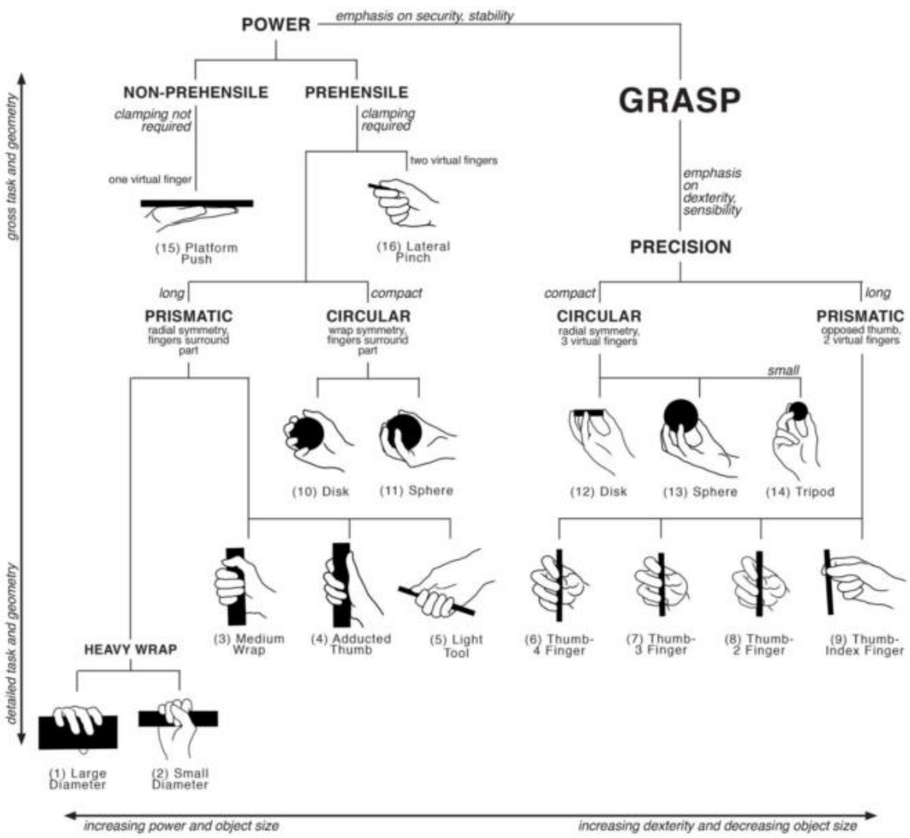

Figure 5. Cutkosky's Grasp Taxonomy[40]

The split hook design, as shown in *Figure 6,* consists of two titanium or stainless-steel rods shaped to form a hook[29]. The inside surfaces of these hooks are coated with a layer of Nitrile or Neoprene grip material to increase friction and prevent objects being damaged during grasping.[11] In a voluntary-opening style split hook, the hooks are connected to each other by a hinge at the base and are held together by a set of elastic bands. A control cable is connected between one of these hooks and a body harness, which allows the user to manipulate the tension on the elastic bands to open and close the hooks. Whilst the split hook only has a single DoF, it is capable of three prehensile grasps due to the varied grip surfaces along the two prong hooks. The hooks themselves also offer non-prehensile movements making it a very versatile device.

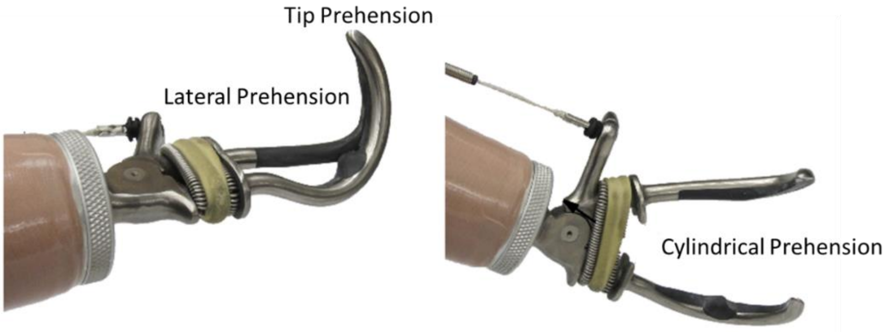

Figure 6. Hosmer Dorrance 5XTi Split Hook featuring three prehension grasp surfaces[11]

Despite their versatility, a split hook design has some drawbacks. With these devices it can be difficult to grasp large diameter objects due to the elastic bands which hold the hooks together, where a significantly high amount of tension is required. In a voluntary-opening design, the maximum strength of the grasp is determined by the elastic bands and there is no ability to increase the grasping forces. Whilst voluntary-closing designs exist, such as the Sierra Hook[41], these require the tension to be maintained throughout grasping, which can easily tire out the user. To overcome the negatives of single DoF body-powered devices, an electric motor with a battery has been historically used, as seen in the Viennatone Hand. Ottobock has greatly improved upon their original design with the Sensor Hand which gives superior grasping forces and speeds. That said, the single DoF limits the performance of split hook devices to three prehensile grasps. Prosthesis users may be forced to use a suboptimal grasp where the device is unable to perform a specific desired grasp, which can result in poor posture due to positioning difficulties. In order to truly perform all the prehensile grasps classified by Cutkosky, a minimum of three degrees of freedom is needed assuming that the digits are rigid and hard with non-rolling and non-sliding contacts[42].

2.2.3 Multi-Articulating Commercial Prosthetic Devices

Multi-articulating myoelectric prostheses can provide a wide range of prehensile grasps which body-powered devices are unable to do. Several small electric motors can be accommodated in the palmar and forearm areas of a prosthesis, allowing it to achieve multiple degrees of actuation. In addition, these devices can adopt a more anthropomorphic appearance. Anthropomorphism is the practice of applying human attributes to an object[43]; in the field of upper limb prostheses it is the design of devices which mimic the human hand's shape and kinematics[44]. A description of the mechanics of the following commercial multi-articulating upper limb prosthetics are described below: BeBionic V2, iLimb Ultra, Michelangelo and Hero Arm.

The BeBionic V2 by Ottobock (Berlin, Germany), previously RSL Steeper, has an anthropomorphic design featuring powered flexion and extension of the individual finger and thumb digits, as shown in *Figure* 7. The BeBionic has a passive circumduction joint that allows the thumb to be positioned into multiple states manually, which effectively changes the grasp. The BeBionic is myoelectric controlled and can achieve fourteen prehensile grasps. The finger digits are composed of two bodies: a proximal and a distal body, analogous to the human metacarpal phalange and proximal interphalange. However, in contrast to human anatomy, there is a fixed angled feature at the end of the distal body to represent the distal interphalange. The BeBionic is an under actuated design where all the DoFs in a finger digit are controlled by a single linear drive. There is a fixed relationship between the motions of the proximal and distal joint due to its four-bar linkage system as illustrated in *Figure* 7. The linear actuator drivers of each digit have internal positioning feedback which enables a greater degree of precision.

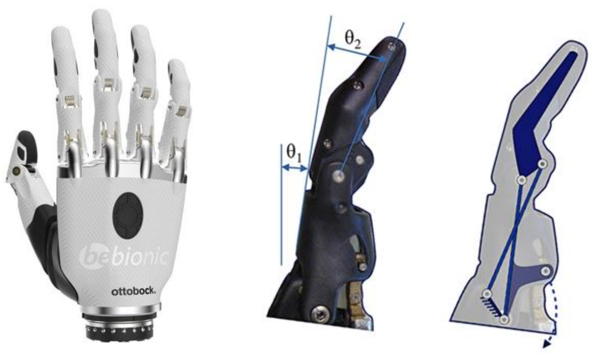

Figure 7: Ottobock BeBionic V2 Design [45]

The iLimb Ultra by Ossur[46](Reykjavík, Iceland), previously TouchBionics, shares many similarities with the BeBionic: both have an anthropomorphic design with the same finger design, as shown in *Figure 8*. The iLimb is myoelectric controlled with powered flexion and extension of the individual finger and thumb digits, capable of eighteen grasps. It uses a tendon transmission system, in which a fibrous cable connects the base and distal end of the digits. Small rollers guide the tendon cable and help to control the moment arm across the proximal interphalange joint[45]. This allows the iLimb to achieve adaptive grasping, where the distal phalanx can continue to move even after the proximal phalanx path has been blocked. Adaptive digits like these can interact with grasped objects with a greater surface area and can therefore better distribute the grasping force[47]. The finger and thumb digits are actuated by small DC motors in the proximal phalanx with high gear reductions. The motor controls the position of the proximal phalanx, and the transmission system controls position of the distal phalanx.

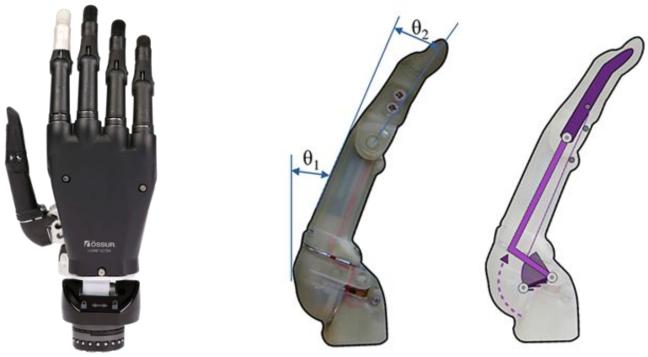

Figure 8: Ossur's iLimb Finger Design[45]

The Ottobock Michelangelo[48] has a unique design utilising a single brushless motor in the palm of the hand to actuate flexion and extension of the finger and thumb digits, and a small second motor to control thumb circumduction. The central drive system actuates all five digits simultaneously through several linkage mechanisms. The finger digits also adduct during extension and abduct during flexion, which can be used to hold thin flat objects regardless of the thumb digit position. Whilst the Michelangelo has an anthropomorphic design as shown in *Figure 9*, the finger digits are composed of a single curved body that is only actuated at a single point. The simple cam design, rigid body digits, and central drive system allows the Michelangelo to achieve the highest grasping force of all commercial upper limb prosthesis[45]. However, this also limits the Michelangelo to two prehensile grasps.

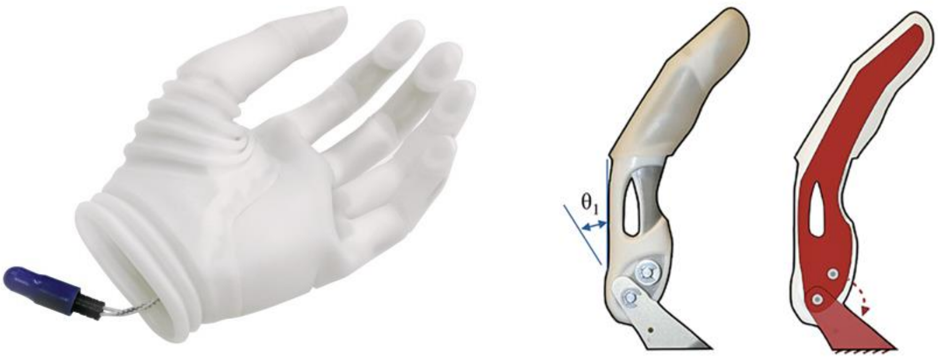

Figure 9: Ottobock Michelangelo Design[48]

OpenBionics's (Bristol, UK) Hero Arm[49] is the newest prosthetic to reach the market - launched in 2018 - and is the first device to be built primarily by Additive Manufacturing (AM). It uses a combination of Fused Filament Fabrication (FFF) and Selective Laser Sintering (SLS) to make custom-sized devices at a lower cost compared to conventional engineering methods. The Hero Arm has an anthropomorphic design, as shown in *Figure 10*, with powered flexion. The Hero Arm is controlled by a tendon system, where a cable connects the tips of the fingers to servo motors located in the forearm unit. When these cables are pulled, the digits flex and form a grasp. Springs in-between the interphalange joints that run along the top of the digits are used to extend the fingers. This system is under actuated which allows the Hero Arm to form adaptive grips. The Hero Arm makes use of single channel myoelectric sensors which reduces cost but relies on actions of the user's healthy opposite arm to change grasping modes. A button on the back of the Hero Arm is used to cycle through six grasps. The Hero Arm, compared to the other commercial models, has a lower grasping force due to the low mechanical efficiency associated with tendon cable systems. In addition, the springs that are used to extend the fingers have an antagonistic interaction with the servo motors during flexion.

Figure 10. Open Bionics Hero Arm Design[49]

For multi-articulating myoelectric prosthetics to operate, they require a large number of components, which adds to weight and cost of a device. The ideal weight of a prosthesis would be equal to or less than that of the original human limb: the average adult male hand is 400 grams[50] (distal phalanges to the wrist and not including the forearm muscles). This is particularly important for upper limb prosthetic devices as they are suspended from the soft tissue of the residual limb instead of the skeleton: the device is perceived as an external load where the effects of weight are compounded[51]. Therefore, the weight of the prosthesis has a high impact on the comfort of the device and user fatigue.

In order to be effective, prosthetics need to be worn for periods of at least 8 hours a day[23]. A survey of 54 upper limb prosthesis users by Pylatiuk et al. found that 79% considered their devices too heavy[52]. In a design priority study by Biddiss et al., 242 participants (including healthcare providers, community support groups, and one prosthesis manufacturer) rated the importance of device weight on a scale of 0-100 (least to most important): the average rating was 70. If a prosthesis is uncomfortable to wear and use, it will very likely lead to abandonment[16]. Despite weight being identified as a key factor of device abandonment, there is no set specification for the maximum weight of the device within the prosthetics community[51]. For optimal device performance and retention, the benefit from additional grasps must be compared to the detrimental increase in weight associated with the electrical components required. The most used myoelectric prosthetic devices all weigh above the ideal 400 grams: the BeBionic V2, iLimb Pulse, and Michelangelo Hand weigh 527, 539, and 746 grams respectively[45]. Only the most recent upper limb prosthesis manufacturing company, Open Bionics, has released a product that is 345 grams[53]. The cost of these commercial myoelectric prostheses are: £30 000 (Bebionic V2[54]), £40 000 (iLimb Access[55], previously known as iLimb Pulse), £47 000 (Michelangelo Hand[56]), and £10 000 (Hero Arm[57]). The majority of commercial myoelectric prostheses have a high weight and cost due to their complex electromechanical designs in order to provide multiple functions. However, it is important to assess the benefit of such functions, and its potential drawbacks, to determine ideal levels of functionality.

Devices capable of a large range of prehensile grasps place correspondingly large cognitive burdens on the user during operation. Most myoelectric prostheses have a binary state, where a myoelectric signal will trigger digits to either flex or extend. This requires the user to actively identify and select the most appropriate grasp for their task. The Bebionic V2 requires the user to generate a specific myoelectric signal that cycles through a list of grasps, which takes significant rehabilitation and training to use reliably. The Bebionic V2 also comes in a variant with several buttons on the back of the palm to navigate through a list of grasps to make this easier. The Hero Arm has a similar approach, where a single button on the back palm needs to be physically pressed to cycle through a list of grasps. The iLimb Pulse offers the user the possibility to select grasps via a smart phone application. Since all of the above procedures for changing grasping modes require a healthy hand, most prosthesis users avoid using their device for uni-lateral tasks[58]. An observational study of usage of multi-articulating prosthetics in a home setting found that 95% of tasks were completed/could be completed with only four grasps: the power (35%), precision (30%), lateral (20%), and extension (10%) grasps[59]. In addition, the reliability of grasping with a prosthesis is sufficiently low that prosthesis users feel more confident using their healthy hand[18]. Therefore, in order to achieve true multiple prehensile grasps in everyday living, the prosthesis must possess more than a single degree of control[60]. Salisbury theorised that a minimum of three finger digits and nine degrees of actuation are needed to achieve dextrous manipulation[61]. This level of control allows grasped objects to be held in place and fully restrained. However, the operation and control of prosthesis hands with nine or more degrees of actuation can be very difficult with myoelectric control. The lack of an intuitive and reliable human machine interface that can translate the prosthesis user's grasping intentions to real movements prevents widespread acceptance of multi-articulating prosthetics[62]. Therefore, prosthesis designers need to consider whether there is a good benefit from adding extra functions when it comes at the cost of weight and price[15].

2.2.4 Open-Source Prosthetic Devices

Alternate manufacturing methods can be used to reduce weight and cost of upper limb prostheses, particularly Additive Manufacturing (AM). AM processes join materials selectively, usually layer-by-layer, to fabricate objects from 3D model data[63]. There are zero to minimal material losses associated with AM, in contrast to conventional manufacturing methods such as injection moulding and machining, which significantly reduces costs. Due to the selective process of AM, geometrically complex parts can be produced utilising weight optimisation methods (e.g. internal lattice structures). One of the first AM-produced prostheses was the Robohand, which was designed as part of a community-driven project to develop low-cost open-source devices. This movement is now being driven by e-NABLE, with the goal of creating a global network of volunteers and recipients in order to increase the accessibility of cheap hand prostheses[64]. FFF is the most common AM technique used to manufacture open-source designed prostheses, and selectively deposits filament material through a heated nozzle. It has become a highly accessible technology due to the rise of low cost plug-and-play desktop printers[65]. The majority of open-sourced hand prosthesis designs utilise a reverse tenodesis transmission system[66], where the digits of the hand prosthesis close with flexion of the wrist, and open with extension of the wrist[64]. An example of such a design is shown in *Figure 11*, where all five digits are connected to a wrist tensioning system and move in unison. This body powered style of prosthesis is very simple to manufacture, costing under £50, and has been mainly aimed at children as they will require several models until they reach adulthood. However, such designs are limited to transcarpal amputees, who lack fingers but retain the wrist movement necessary for device operation. Additionally, finger articulation is very primitive as all the digits extend and flex at the same time, which severely limits the activities it can be used for. There is insufficient evidence to support that reverse tenodesis hand prosthesis designs truly aid a range of activities for daily living[65]; there may not even be an improvement in functionality between no prosthesis and reverse tenodesis hand prosthesis usage[3].

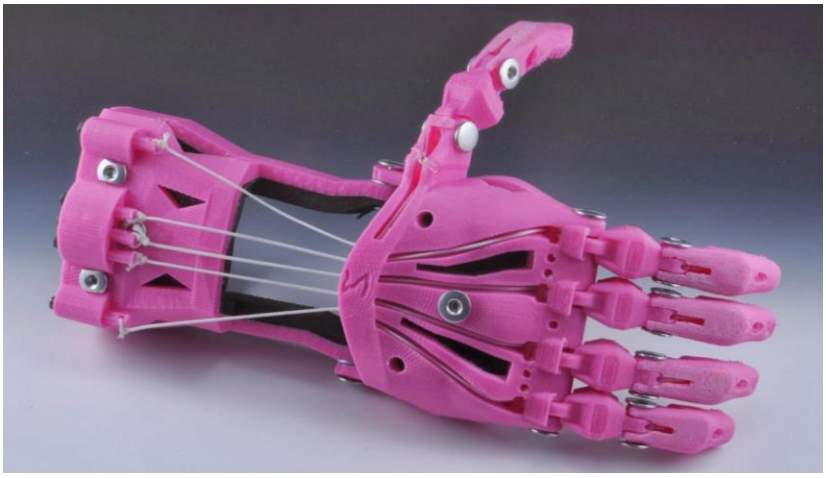

Figure 11. Cyborg Beast Hand design from the e-NABLE movement[67]

Open-source hand prosthesis designs are not just limited to body-powered types. Exii (Tokyo, Japan), no longer trading, developed and publicly shared the design of an externally-powered device, the Hackberry[68] as shown in *Figure 12*. The Hackberry can be configured in either myoelectric or electrical modes[69]. It was designed to be manufactured by FFF with a material cost of £150. It is controlled by three servos which control the index finger, thumb abduction, and the remaining fingers which are coupled together. The index finger consists of several linkages which give it an adaptive grip, whereas the other fingers do not. This makes the Hackberry better at grasping small objects between the index finger and thumb than it is at grasping large objects which requires all the fingers. It has a passive wrist that is capable of extension and flexion, as well as rotation. Whilst there are a small number of cases of people using the Hackberry, there have been no clinical trials to validate the design. Open Bionics is the first company to make commercial AM models; the design of the Hero Arm has been discussed above. The use of AM processes (specifically, FFF and SLS) to reduce material costs and weight enables them to offer the cheapest multi-articulating model currently available.

Figure 12. Exii's Opensource Hackberry Design[68]

2.2.5 Academic Upper Limb Prosthesis Design

In academic literature, a variety of novel designs have been developed to improve functionality per unit weight by focusing on one specific aspect: either weight reduction or enhanced grasping and control systems. One method of reducing weight is to reconsider the actuation system. The majority of multi-articulating prosthetics use DC motors; however, they have a low power to weight ratio of 10 W/kg[70], with low torque and high rotation speeds. They therefore require a gear train to provide an appropriate grasping speed and force. However, this adds significant weight especially with multiple fingers. Smit et al. at the Delft University of Technology have developed a novel body-powered hydraulic design[71], as shown in *Figure 13.* It has a very high power to weight ratio of 2000 W/kg[70], which overcomes the low mechanical efficiencies of cabling from body powered devices[72], and weighs only 217 grams. The Delft Cylinder Hand has seven DoFs which are actuated by miniature hydraulic slave cylinders which are connected to a master cylinder. When the body-powered shoulder strap is pulled it causes the master cylinder to extend and actuate the slave cylinders, causing the fingers to flex. The slave cylinders are all connected to one another which maintains a constant fluid pressure throughout, therefore enabling adaptive grasping without the need for heavy linkage systems.

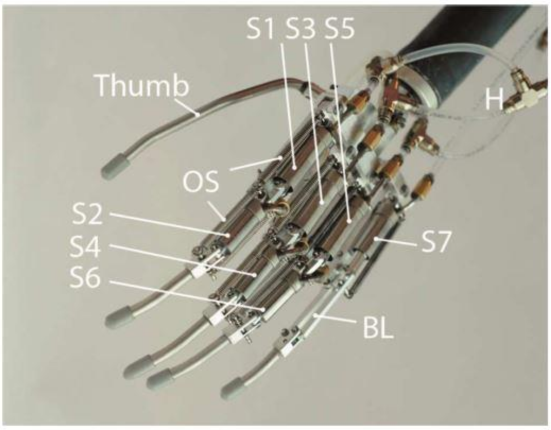

Figure 13. Delft University of Technology's Novel Hydraulic Design: Delft Cylinder Hand[71]

The opposite approach for increased functionality to weight ratios – i.e. increasing the functionality of a device – has also been explored in academia. Novel designs for improved grasping performance have been achieved by discarding the conventional prosthesis design. While the majority of externally powered upper limb prostheses are anthropomorphic in design, to appeal to their users with a familiar shape and size, there is a bottleneck of control which means that designs comprising four digits and a thumb may not be the most optimal for reliable grasping. Amend and Lipson, from Columbia University, have excluded anthropomorphism and weight as design criteria to develop a novel design that provides dexterous manipulation with minimal actuation[73]. Their design, the Jam Hand, comprises a two-prong claw with coated tips, where the coating consists of an elastic membrane which contains pockets of granular material, as shown in *Figure 14*. It uses two servo motors for motion control: one for closing the finger prong and another for the rotation of the thumb prong. The air pressure within the pockets of granular material is controlled by a three-position valve so that the interstitial space can either be evacuated, positively pressurised, or equalised with the atmosphere. These pockets of granular material can passively conform to the shape of the target object and then be vacuum hardened to create a rigid grasp, a process which is known as jamming[74]. The Jam Hand can achieve multiple precision and power grasps due to the flexibility of the granular material pockets. Performance of some fine manipulation tasks has also been demonstrated such as picking up a pen and writing. However, it has not yet been evaluated under any formal clinical tests to assess its function as an upper limb prosthesis.

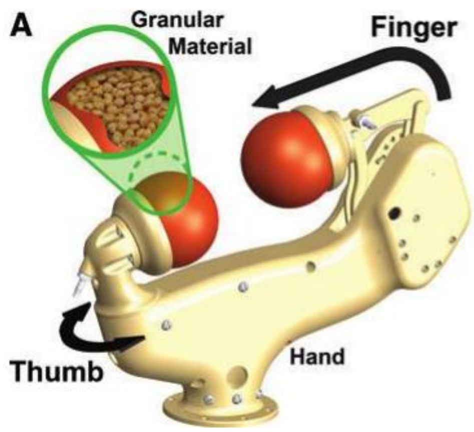

Figure 14. Columbia University's Minimal Actuation Hand: Jam Hand[73]

One of the major bottlenecks of multi-articulating prosthetics is the myoelectric control system, where there is only a single degree of control at a given instance. Without being able to manipulate individual fingers, the maximum grasping potential given the number of digits cannot be realised. A neutrally-controlled device called the Modular Prosthetic Limb (MPL)[75] has been developed by researchers in the Revolutionising Prosthetics Program (Defense Advanced Research Projects Agency, John Hopkin's University Applied Physics Laboratory) to overcome the issue of control. Intercranial electrodes are embedded in the prosthesis user which can identify high gamma activity made during reaching and grasping movements[76]. These electroencephalographic signals are then mapped to the MPL's seventeen actuators with a total of twenty-six DoFs[77], as shown in *Figure 15*. This novel brain-machine interface allows simultaneous control of multiple DoFs, allowing a greater level of control. The actuation system is similar to the BeBionic and iLimb, consisting of DC motors and gear chains embedded within the forearm and palm units[78]. However, to accommodate all the electronics required for actuation and control, the trans-radial model of the MPL weights 2.09 kg. It also involves a highly invasive surgery and considerable rehabilitation time. Whilst the MPL solves the issue of control for multi-articulating prostheses, the high cost and surgical requirements will be a barrier to most prosthesis users.

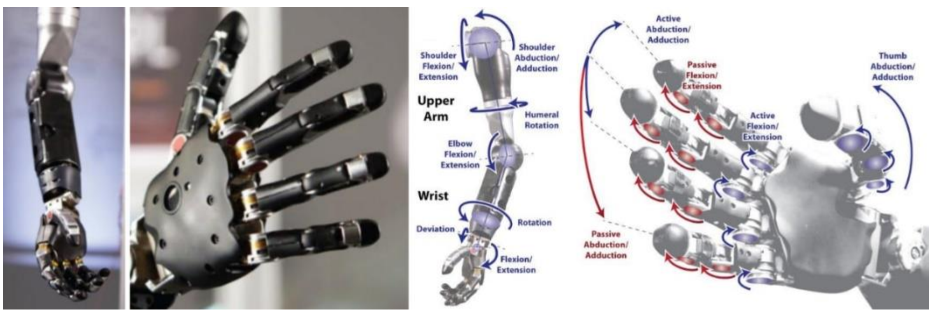

Figure 15. Johns Hopkins University's Modular Prosthetic Limb[79]

2.2.6 Summary of Upper Limb Prosthetic Devices

In summary, commercial multi-articulating prosthetic hands are capable of many prehensile grasps, which are mostly unused in tasks of everyday living[59]. To form these grasps, additional components are required which adds weight and increases cost. This results in a device which is heavier than it needs to be, which increases the probability of device rejection. Body powered hooks on the other hand, whilst more limited in number of prehensile grasps, are the most used upper limb prosthesis as they benefit from low cost, light weight, and high reliability[80]. Comfort, which is determined by the weight, is commonly reported as the top reason for device rejection[27-28,50-51] regardless of the functions available.

For an upper limb prosthesis to be accepted by its user and used in the long term, it needs to have a high functionality to weight ratio. This can be achieved in two ways. Firstly, by reducing the weight of the prosthesis through advanced manufacturing methods, using either new materials and/or processes. Whilst there are no weight restrictions on an upper limb prosthesis when it enters the market, numerous researchers have recommended that these devices should have a maximum weight of 500 grams[60,81–83]. Conventional methods of design have not achieved this. AM processes enable design with great geometric freedom for weight reduction and have been under-utilised in prosthetic manufacture to date. The only commercial multi-articulating upper limb prosthesis under the weight of 500 grams is the Hero Arm by Open Bionics. Open Bionics have used AM effectively to provide a lightweight device of 346 grams, whilst still providing a range of prehensile grasps. The second approach to improve prosthesis acceptance is to increase the functionality of the prosthesis. The aim of multi-articulating prostheses is to increase the functionality of the user by enabling them to grasp objects more easily; however, design factors for producing prostheses where the functionality translates into utility in daily life is limited. Utilisation of device functionality is restricted by the limitations of current control systems due to the lack of individual channels causing a large cognitive burden. Existing myoelectric control systems fail to make full use of the DoF of multi-articulating prostheses, while neural control systems are nascent and require highly invasive surgeries with long rehabilitation times.

A common theme across prosthesis designs, in both commercial and academic devices, is a lack of consistent methods of device evaluation. The level of functionality that an upper limb prosthesis provides is not well understood and it makes comparisons difficult. This is one of the reasons why there is a lack of functionality-to-weight optimised designs. It has therefore not been possible to weigh the pros and cons of design features effectively to create a better prosthesis. Various methods of measuring functionality of an upper limb prosthesis are discussed in *Section 2.3*.

2.3 Evaluation of Upper Limb Prosthetic Devices

Several methods of upper limb prosthesis evaluation are used to ensure that an appropriate device has been prescribed and is capable of meeting the needs of the user. The current standard for evaluating an upper limb prosthesis is with after-care surveys and direct feedback between the users and their prosthetists. These qualitative methods are very effective for procuring the satisfaction of the user and what they expect from their device. Feedback gained from these personal interactions is usually used by the prosthetist to recommend a more appropriate device. However, the results of after-care surveys are very subjective and are often about the user's individual needs rather than a holistic measure of the device performance. Therefore, the standard of using after-care surveys is not suitable for measuring the functionality of the prosthesis but rather how the user perceives their device. To measure the functionality of an upper limb prosthesis, academics have developed several quantitative methodologies. In this section, three distinct types of quantitative evaluation methods are discussed: time-based testing, virtual testing, and motion capture.

2.3.1 Time-based Testing

Time-based testing was originally developed by clinicians to assess the biological hand functionality of patients with impaired upper limb ability. The time taken for patients to complete tasks was monitored; these metrics were used to track the progress of their rehabilitative treatments/rehabilitation. Due to the similarities between grasping with a healthy limb and with an upper limb prosthesis, there have been multiple studies that use time-based testing to assess the functionality of prosthetic devices[15,78,84]. Time-based testing requires a prosthesis user to complete an activity which is either timed, or within a certain time frame. Time-based testing is the most common type of quantitative evaluation method due to the ease of implementation as set-up is simple: it only requires the user, the prosthesis, and an activity. Examples of time-based testing are the Box and Blocks Test (BBT) and the Nine-Hole Peg Test (NHPT), as shown in *Figure 16*. The BBT is a test for manual dexterity[85]; it consists of a box with a centre partition with small wooden blocks on one side and the other side empty. The prosthesis user will be asked to pick up the wooden blocks with their device and transfer it to the empty side one by one. The assessment score of the BBT is represented as the number of blocks transferred in 60 seconds, with more blocks equating to better hand function. The NHPT is a test for finger dexterity[86] which consists of nine pegs and a peg board. The prosthesis user will have to pick up the pegs from the shallow dish one at a time and slot them into the holes on the peg board in any order. Then afterwards remove the pegs one at a time and place the pegs back into the shallow dish. In total there are eighteen peg movements to be performed with a time limit of 90 seconds. The results of this test are measured in seconds per peg movement, with a lower time equating to better hand function. Both these tests are popular among academics as it takes a short time to complete and allows prosthesis users to perform several iterations of the test without any significant physical burden on the user[84]. However, these tests are highly repetitive and only require a single prehensile grasp. Therefore, the results of these tests are unable to differentiate between single DoF devices and multi-articulating designs.

Figure 16. Hand Assessment Tests
Left: Box and Blocks Test[87]. Right: Nine-Hole Peg Test[88].

There are alternate methods which use Activities of Daily Living (ADLs)[19] to better understand the benefit of different prehensile grasps. ADLs are basic activities which have been identified as essential for independent living[12,89,90]. Examples of these are the Jebson Taylor Hand Function Test (JTT)[91] and the Southampton Hand Assessment Procedure (SHAP)[92–94], shown in *Figure 17*. The JTT assesses a person's overall hand function using seven tasks: writing, turning over cards, picking up small objects (e.g. coins), stacking checkers, simulate feeding, picking up large objects, and picking up large heavy objects (e.g. tin cans). The prosthesis user will be asked to perform these tasks with their remaining healthy hand and again with their prosthesis, to use the healthy limb as a benchmark value. The total time for each task presents the final score for the JTT, where faster times indicate better hand function. The objects used in the JTT were selected based on their utility and usefulness in daily living, with no specific prehensile grasp. The SHAP addresses this weakness by using tasks that are specific to a type of prehensile grasp[95]. The SHAP focuses on the evaluation of the unilateral performance of the prosthesis through twenty-six timed tasks. The first segment consists of six abstract objects with lightweight and heavy variants. The prosthesis user is asked to start the timer, then to pick up an object and place it down in an allocated spot. The second segment consists of fourteen ADLs: picking up coins, fastening buttons, cutting food, turning pages, opening a jar lid, pouring from a jug, pouring from a carton, lifting a heavy object, lifting a light object, rotating a key, opening and closing a zip, rotating a screwdriver, and turning a door handle. Since each task is linked to a prehensile grasp, it is possible to identify key weaknesses and strengths in specific grasping performances and to identify tasks that will present difficulties to users. Both the JTT and the SHAP use a wide selection of objects and tasks which allows them to assess a range of prehensile grasps. These testing conditions better simulate typical real-world activities through the inclusion of ADLs; this higher ecological validity makes these tests more suitable for evaluating device functionality in everyday usage than other repetitive time-based tests like the BBT and NHFT.

Figure 17. Activities of daily life focused assessment tests
Left: Jebsen Taylor Hand Function Test[96]. Right: Southampton Hand Assessment Procedure[97].

Whilst time-based testing offers a quantitative measure to assess hand functionality, there are a few fundamental flaws which limit their utility for design evaluation. Their greatest weakness is that the results of all time-based tests are not intuitive measures of design quality and cannot be interrogated further to reveal design details of the actual prosthesis. For example, the results from two prostheses with different JTT scores will not be able to identify which design feature provided improved functionality. These oversimplified methods of assessing the functionality of prostheses lacks depth and is unable to directly feedback into a design process due to poor construct validity.

Additionally, the tasks in time-based testing have been selected so they can be performed unilaterally and assess solely the prosthesis device with no influence from a healthy limb. However, it does not consider how a prosthesis user moves the rest of their body and therefore does not detect excessive compensatory movements[10]. Using an upper limb prosthesis effectively requires the coordination of reaching and grasping movements/motions with the remaining articulations in the arm[98]. The musculoskeletal system of the upper body contains redundant degrees of freedom so, even with the loss of a hand, a prosthesis user has several motor strategies to complete a goal orientated task. Upper limb prostheses typically do not replace the forearm pronation/supination and wrist flexion/extension, which tends to result in prosthesis users making excessive motions at the shoulder and trunk[10]. However, these compensatory movements can result in overuse injuries and place the body in unnatural positions likely leading to device rejection [99–101]. There is little value in assigning a high time-based score to a prosthetic device if it requires unhealthy motor strategies to achieve this.

Another limitation of time-based testing is the discrepancy of scores of the same prosthetic device but different users. There are severe learning effects from time-based testing: on average it takes prosthesis users seven attempts at the SHAP to reach their performance plateau[102]. As a result, the SHAP ends up evaluating the skill of the user to manipulate their prosthesis rather than the functionality of the prosthesis, which is true for all task-based tests[102]. Time-based testing has strong internal validity in tracking the performance progress of a single user but has poor test-retest reliability and population validity which makes it unsuitable for evaluating the functionality of the prosthetic device itself.

2.3.2 Motion Capture Analysis

As discussed in *Section 2.3.1*, compensatory movements are an important aspect of upper limb prosthesis usage. The user's body adapts to the limited number of DoFs of a prosthesis by making compensatory movements[98]. However, these compensatory movements can lead to risk of joint degeneration, muscle overload, and repetitive strain injury[99–101]. Therefore, it is important to obtain quantitative data on prosthesis users to understand the effects from the combined motion of the shoulder, elbow, and wrist. Quantitative motion capture analysis can be used to measure these compensatory movements.

Previous motion capture studies have characterised the motor strategy of prosthesis users but they have mainly been used as a clinical tool to prescribe a training routine to reduce further compensatory movements[72,76,95-96]. Hebert and Lewickes[99] proposed a novel methodology to use motion capture to assess the quality of movement of a prosthesis user whilst they performed a time-based task (i.e. Box and Block Test). They used a videogrammetric technique as recommended by the International Society of Biomechanics for the study of upper limb gait using a marker set and joint coordinate system[105]. Eight motion cameras were used to track six markers placed on the participant:

sternum, seventh cervical spinous process (C7), acromion (bilaterally), lateral elbow hinge and wrist. The triangle formed by the sternum, C7, and acromion was used to show the orientation and position of the shoulder girdle segment. The markers on the wrist, lateral elbow hinge, and acromion were used to calculate the elbow flexion. The tilt of sternum and C7 markers in the vertical plane were used to track the trunk's anterior and posterior tilt. To ensure that the markers would be tracked during the time-based task, the BBT was slightly modified to start with fewer cubes and organised in a grid rather than randomly arranged as shown in *Figure 18*. Hebert and Lewickes used this methodology to compare the compensatory movements of a single participant using a body-powered voluntary open Hosmer Hook and a myoelectrical two prong Motion Control Inc. terminal device after receiving Targeted Muscle Reinnervation treatment. Whilst the participant had slower performance with the myoelectric prosthesis, there was significantly less compensatory adjustments required. The range of movement of the trunk required to operate the body-powered prosthesis was almost twice as much as the myoelectric prosthesis. Time-based evaluation methods alone do not take into account the co-ordination of reaching and grasping which has a significant impact on how well a task is performed[98]. However, this case study uses a highly repetitive time-based task which does not showcase a large range of movements.

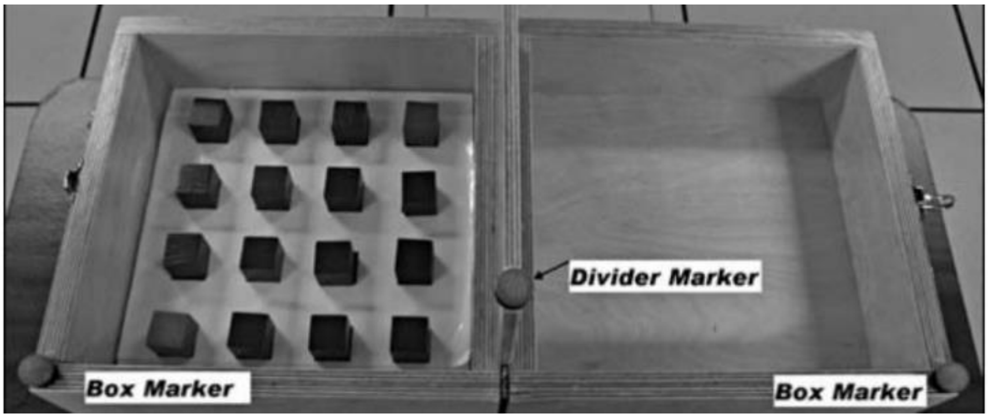

Figure 18. Placement of blocks for Herbert's and Lewicke's Modified Box and Block Test[99]

Major et al.[103] have also performed a similar videogrammetric motion capture study but used a different set of time-based tasks (i.e. Southampton Hand Assessment Procedure). Upper body kinematics were collected on six able-bodied participants and compared to seven myoelectric prosthesis users. A range of upper limb prostheses were used which include single DoF grippers and multi-articulating prosthetic devices. Participants were seated at a table and requested to perform the following ADLs from the SHAP: food cutting, page turning, carton pouring, and lifting & transferring a weighted object. Able-bodied participants were requested to use their non-dominant hand to match the prosthesis users. Shoulder flexion/extension, shoulder ab/adduction, elbow flexion/extension, trunk lateral flexion, trunk transverse rotation, and trunk forward flexion were calculated from five trials of the selected ADLs. Across all tasks, prosthesis users saw an increase in trunk motion in all three planes and higher shoulder abduction compared to controls. There was also high kinematic variability for prosthesis users for all DoFs among all tasks, due to the differences in prosthesis mechanics. These compensatory movements are a result of the absence of active distal DoFs in the prosthetic device. Major et al. concluded that the prosthesis users may benefit from dedicated training to foster adaptable but reliable motor strategies. However, this motion capture data should instead be used to closely examine the mechanics of the prosthesis and how it could be modified to reduce compensatory movements.

Not only have compensatory movements been observed with prosthesis users, the way in which users interact with objects have also changed. Researchers at the University of Alberta have developed their own set of time-based tasks paired with synchronised motion capture cameras and eye tracking to investigate visuomotor adaptations among upper limb prosthesis users[106–110]. To enable easier motion capture segmentation, two new time-based tasks were made with discrete motions: the Pasta Box task mimics moving objects from a counter to a cupboard, between cupboards of different heights, and across the body's midline; and the Cup Transfer was designed to involve risk by using compliant cups with contents that could be spilled, and therefore requiring careful grasping and placement[110]. A twelve-motion capture system was used to track participants performing the Pasta Box and Cup Transfer task, with reflective markers on the main body of a prosthetic device and on the user's trunk, pelvis, upper arms, and head. A Dikablis headset was used to track pupil movements of the user in infrared and a high-definition scene camera for a first-person view to identify the target of fixation[107]. These two systems were used in tandem to track the participant's gaze and segment the movements, as visualised in *Figure 19*. The time-based tasks were segmented into reaching, grasping, transportation, and releasing phases to measure how long participants took for each type of motion. In a case study of eight prosthesis users (7 body-powered devices and 1 myoelectric device), they found that prosthesis users had 30% more eye fixation on their prosthesis during reaching than able-bodied controls[109]. As prosthesis users reached out with their device to position their grasp, they had to use additional focus to not unintentionally collide with the target object. Whereas able-bodied controls would fixate on the target object instead and, upon grasping, would fixate on the drop-off location during transportation. Prosthesis users instead fixated on the object during the majority of the transportation phase, and only changed their fixation once the drop-off location was reached. This gave prosthesis users less time to prepare for releasing their object at the drop-off location and resulted in longer times taken for the release phase. These results show an aspect of prosthesis usage which cannot be captured in traditional evaluation methods. However, this experimental setup has very high technological and administrative barriers and will not be practical for all prosthesis designers to adopt.

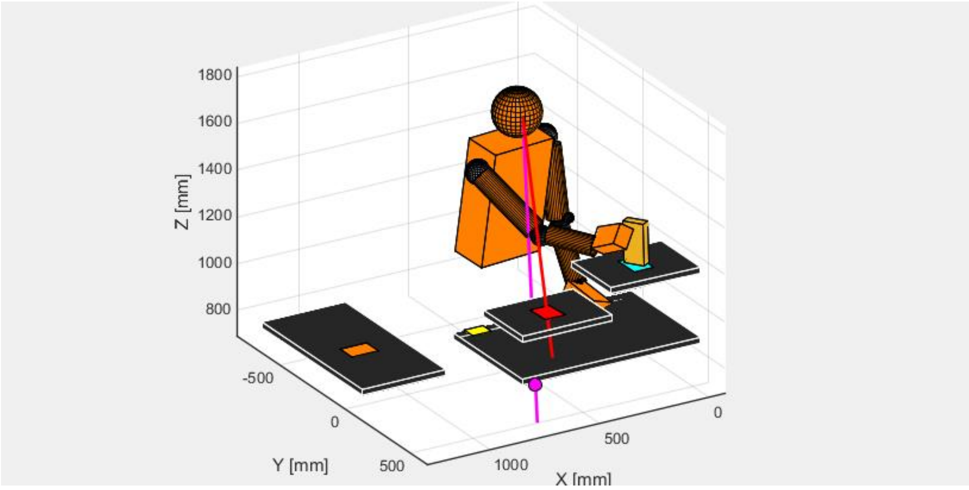

Figure 19. Visualisation of the Pasta Box Task using Motion Capture and Eye Tracking[111]

2.3.3 Virtual Evaluation

After-care surveys, direct feedback, and time-based tasks all require a prosthesis user to be actively wearing their prosthetic device and engaging with their environment. It has already cost them money and time to be fitted with an upper limb prosthesis which may not be suitable for them. Therefore, a method of benchmarking the performance of upper limb prosthesis before it reaches the end user is needed. Robotic simulators are an indispensable tool for understanding, planning, designing, and programming of industrial robotic systems[112]. These simulators enable robotic operators to program and test their applications without the need for real hardware. By considering the contact forces, torques, and contact areas, grasps can be analysed and evaluated numerically[113]. This allows the robotic system to be assessed and improved upon before manufacturing a prototype. Examples of such robotic simulators are OpenGrasp[114], Syngrasp[115], and GraspIt![116].

OpenGrasp is a modular architecture based toolkit for the simulation of grasping dexterous manipulation[112]. It was designed for human-centric autonomous robot applications (i.e. service robots), and features a humanoid robot acting in a kitchen as shown in *Figure 20*. OpenGrasp allows users to specify linear characteristics between contact force and compression ratio of deformable surfaces, therefore allowing pressure reading emulation in a manner similar to real tactile matrix sensors[114]. OpenGrasp aims to find poses of the robotic gripper relative to the target object and vectors of the joint angles, which together represent force closure: a state where the target object cannot be displaced when external forces or torques are applied[114]. The position of the robotic grippers is determined through the use of one of two alternate grasp planners: Bereson et al.'s[117] algorithm uses the surface normal of the target object as the approach direction, and a user-defined number of roll angles around the approach direction to generate a grasp; whereas Przybylski et al.'s[118,119] algorithm analyses the symmetry of the target object contained in their medial axis. Both grasp planners test candidate positions and compute each one for force closure. This allows users to identify successful grasps which will result in force closure, and therefore have greater confidence during hardware operation. However, OpenGrasp is not capable of assessing the quality of the actual grasp as it is limited to a binary decision: either successful or unsuccessful force closure.

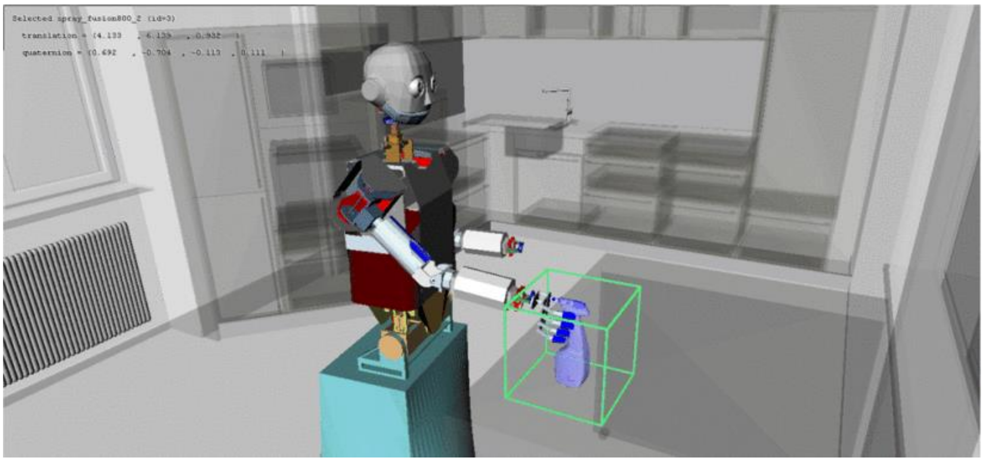

Figure 20. OpenGrasp's Kitchen Grasping Scenario[112]

Syngrasp is a MATLAB toolbox for grasp analysis of fully or under-actuated robotic hands which can assess the performance of a grasp with a large range of numeric quality metrics[115]. Grasps can either be positioned by defining the contact points on the hand with the respective contact normal angles or using Syngrasp's own grasp planner: this grasp planner aligns the palm of the robot hand with the centre of mass of the target object and generates a set of pre-grasp positions with a user defined number of orientations along this axis. From the pre-grasp position, the hand is then closed until all the fingers are in contact with the target object or have reached their joint limits, as shown in *Figure 21*. The grasp quality is computed and sorted according to the metric selected by the user; these grasp quality metrics are based on the concept of the Grasp Matrix and the Hand Jacobian[120], matrices which define the velocity kinematics and the force transmission properties of the contact points respectively. In addition, the structural stiffness of finger digits and the compliance of joints can be modelled. This kinematic coupling enables accurate simulation of underactuated robotic hands, which are common in upper limb prosthesis design. However, Syngrasp has its limitations as it only considers the contact forces and the shape of a target object that is fixed in space. It is not a dynamic simulation and therefore does not consider inertia.

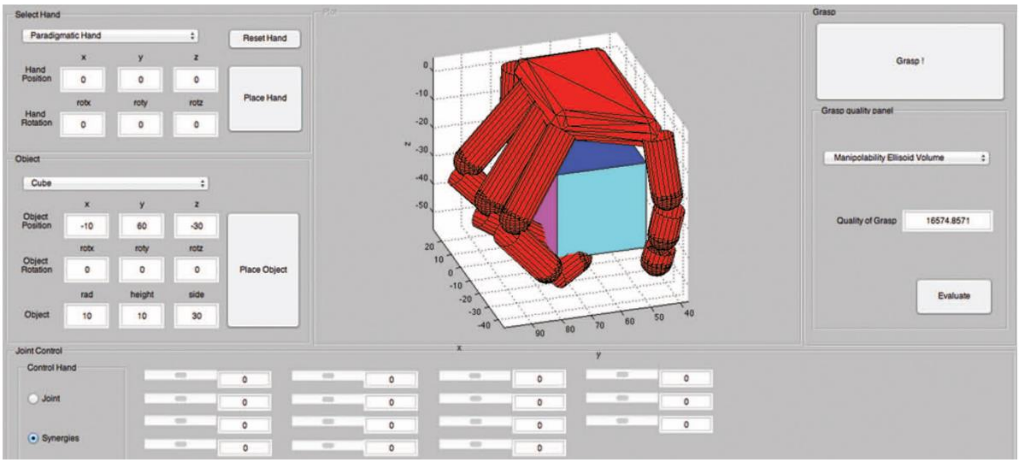

Figure 21. The Synsgrasp Graphics User Interface

GraspIt! addresses the limitations of both OpenGrasp and Syngrasp by offering a dynamic robotic simulator with in-depth grasp analysis tools. It uses rapid collision detection and contact determination, which allows users to interactively manipulate a robotic gripper to grasp objects in real time[116]. It can be used as a test bed for new grasp evaluation, grasp synthesis, and grasp planning algorithms. GraspIt! quantifies the performance of a grasp using a concept called a Grasp Wrench Space, proposed by Ferrari and Canny[121]. This criterion favours grasps which can resist arbitrary forces (wrenches) with the least effort[122]. It also considers the type of task that is being performed by the robot, where just achieving force closure is not enough to be considered a high-quality grasp. The Grasp Wrench Space could simply be a wrench in an upward direction with no torque to suspend an object against gravity, or it could be a series of wrench spaces for a complex manipulation task[116]. It also considers the ratio of task wrench forces with the magnitudes of the contact forces to measure the efficiency of a grasp. By using the Grasp Wrench Space, several relevant grasp quality metrics from the GraspIt! library can be used to numerically assess the quality of the grasp. Visualisation methods as shown in *Figure 22* also allow the user to identify contact points and check for weak points of the grasp.

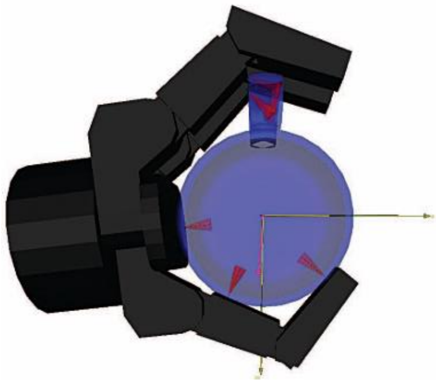

Figure 22. Example of a Force-closure Grasp in GraspIt!

Due to the similarities between grasping of industrial robotic grippers and upper limb prostheses, these simulators could also be used to assess the performance of prosthetic devices. These simulators are focused on the quality of the grasp which allows the reaching and grasping motions of using an upper limb prosthesis to be decoupled, therefore allowing the functionality to be measured without user bias. Kretchetov et al.[123] used GraspIt! to compact the impact of grasp quality of two kinematic schemes of a twenty-two DoF anthropomorphic manipulator (RUBionic), as shown in *Figure 23*. In humans, the plane of the thumb is opposed to the little finger which allows the little finger to be bent along the same plane. In primates the thumb is opposed to the middle finger instead. The RUBionic model was simulated to grasp cubes, spheres, and cylinders of various sizes with a thumb longitudinal axis of 45° (human scheme) and 22.5° (primate scheme). They found that the primate scheme for grasping small objects and long cylindrical objects had better grasp quality metrics than the human scheme. When performing fine manipulations with the RUbionic, the main digits used are the thumb, index, and middle finger. With the primate scheme, this allows the plane of motion for these digits to become parallel during grasping and provides a more reliable grasp. Whereas with the human scheme it is easier to grasp larger objects as it able to better distribute contact points over a wider area. Therefore, it could be advantageous to design a RUBionic variant that can change the orientation of the longitudinal axis of the thumb to take advantage of both kinematic schemes.

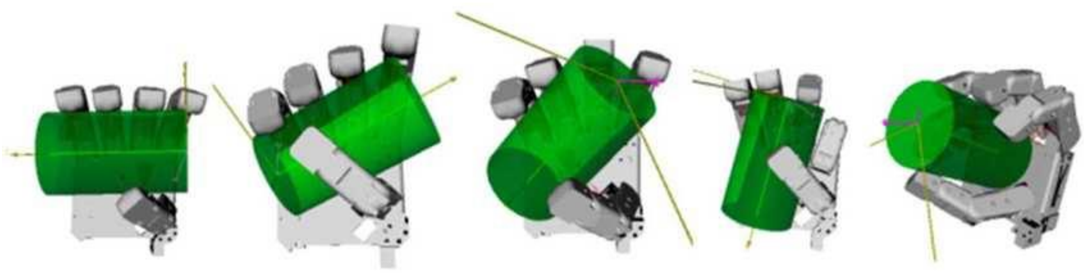

Figure 23. Configuration of grasping for a cylindrical object with the RUbionic model[123]

Each grasp quality metric has been designed to assess a specific aspect of a grasp and not all may be suitable for the evaluation of upper limb prostheses with an anthropomorphic design. There are over twenty-four grasp quality metrics used in GraspIt!, OpenGrasp, and SynGrasp which are either associated with contact points or grasper configuration[124]. Within these simulators there is no clear criterion to select a grasp quality metric for anthropomorphic grasping. To address this issue, Rubert et al.[125] has used OpenRave, the framework that OpenGrasp is based on, to characterise the ten most commonly used grasp quality metrics in robotic grasping. Their aim was to identify the most suitable set of metrics that could be used to evaluate a design, where the metrics are mutually independent and therefore do not measure the same property multiple times. To overcome the weakness of the specificity of grasp quality metrics, variability studies were performed to select suitable metrics that could be combined in parallel to produce a single unique evaluation index[126]. Expanding on these results, Leon et al.[127] characterises these grasp quality metrics for anthropomorphic robotic hand prehension, taking in to account the typical size of objects that a person would interact with. Sensitivity analyses were also performed to determine the robustness of the grasp quality metrics from translating a grasp from simulation to the real environment. Five appropriate metrics were then selected to study the effect of finger abduction using the Michelangelo hand[128], as shown in *Figure 24*. Using OpenRave, the Michelangelo hand was simulated to grasp a fixed cylinder in 12 different grasping postures. From minimum to maximum abduction angles there was a 25% increase in grasping performance. This demonstrates how information on key design features for performance can be obtained from simulation.

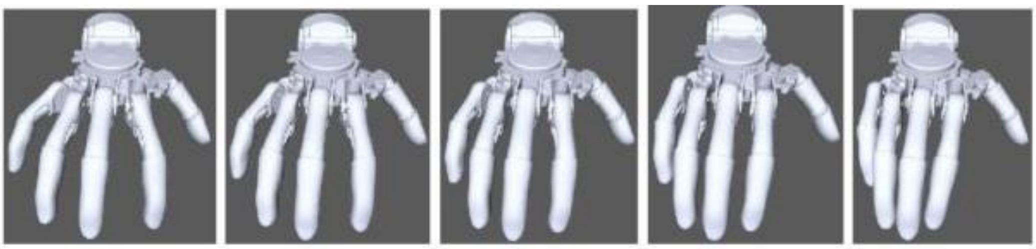

Figure 24. Michelangelo Hand with Varying Finger Abduction Values[128]

Whilst virtual environments do offer a test bed for measuring the grasping of an upper limb prosthesis, there are two main limitations. Firstly, the robotic simulators described in these studies all used static grasping. Only the kinematic movements of the fingers were modelled, and the model collisions were purely based on geometric calculations. The target object was fixed in space and therefore had no weight. In addition, the inertia of the fingers upon collision into the target object was not modelled. These assumptions reduce the computational power required but reduce the realism of the simulation. Secondly, robotic simulators at their core are designed for identifying the best performing grasp according to a quality metric. However, it does not consider how the robotic manipulator needs to be positioned to achieve that grasp. There are no positioning issues with a multi-axis robotic arm but with an upper-limb prosthesis user there are more physical limitations. Prosthesis users need to orientate their device accordingly to make the best performing grasp, but in doing so they may need to contort their body into uncomfortable and strenuous positions. As with time-based tasks, virtual environments are unable to track these compensatory movements. There is little value in a prosthesis design that can achieve high quality grasps if it requires unhealthy motor strategies to achieve this.

2.3.4 Summary of Evaluation of Upper Limb Prosthetic Devices

Multiple methods of upper limb prosthesis performance evaluation have been highlighted in this chapter, with various limitations to their use. The current standard is to use clinical observations, after-care surveys, and direct feedback to gain insight on how the prosthesis user feels about their device. Whilst it is important to ensure that an appropriate device has been prescribed, this qualitative evaluation method is not useful for improving future prosthesis design. To better evaluate the performance of an upper limb prosthesis, academics have used quantitative outcome measures in the form of time-based tasks. These activities are adopted from clinical tests that measure the functionality of hands of able-bodied people. Due to the similarities in grasping activities, they have been applied to assess upper limb prostheses. However, these time-based tasks are heavily influenced by the skill of the prosthesis user and do not accurately measure the performance of the device itself.

Therefore, to address the issue of user bias, virtual environments have been used to simulate grasping scenarios. These virtual environments were initially designed for grasp planning for robotic manipulators, but due to similarities in grasping they have also been used to study prosthetic devices. Relevant grasp quality metrics have been identified for anthropomorphic grasping and have been demonstrated to optimise design parameters. However, these robotic simulators are all based on static grasping where only the kinematic movements of fingers are modelled. It assumes that the target object is fixed in space and the model collisions are purely based on geometric calculations. These robotic simulators also assume that the user will be able to position their device in the correct posture to achieve the calculated grasp quality.

Prosthetic devices have limited DoF which can force their users to make compensatory movements to position themselves correctly and cannot be fully assessed by simulation alone, despite the advantages of simulating prosthetics in a virtual environment. Motion capture can be used to measure these compensatory movements to further inform design. Whilst there are a wide range of studies available that use motion capture, the results from these are not used to improve the design process. Instead, they are used as a clinical tool to recommend training regimes to reduce compensatory movements.

2.4 Motivation

Despite the advances made to upper limb prosthetics in recent years, there are high device rejection rates among upper limb prosthesis users[9,16,18]. There are various reasons for rejection but the most common feedback is that the devices cause discomfort, are too heavy, and have low functionality[52,15]. An upper limb prosthesis that is uncomfortable, painful to wear for extended periods of time, and a hindrance is likely to be abandoned[16]. Prosthetic devices are undoubtedly more technologically advanced than 60 years ago, but still face the same challenges[2]. Advances in prosthesis design and treatment are expanding faster than the ability to measure their effectiveness[99]. Therefore, without effective evaluation methods with the objective of improving prosthesis design, rejection rates will continue to rise. Due to the sensitive weight requirement of upper limb prosthesis, every feature and function must be carefully selected to provide an optimal device. An extensive review paper focusing on the reasons for device rejection has been discussed by Elaine Biddiss and Tom Chau[18], who state that:

> *"Perhaps the greatest pitfall awaiting engineers and clinicians involved in the development of prostheses is an eagerness to tackle the looming design challenges with a battalion of technology before the needs and desires of the end user are clearly defined and translated into specific engineering requirements."*

There is a growing demand for a golden standard in upper limb prosthesis design evaluation. The current standard of using after-care surveys and prosthesis user/prosthetist interactions provides qualitative data which can be difficult to integrate into the design process. It is assumed that if a client is satisfied with their device, then it has high functional capabilities. Therefore, more quantitative measures of upper limb prosthesis functionality are required to identify and understand which design features are beneficial to the user. Existing quantitative methods are time-based tasks such as the Southampton Hand Assessment Procedure (SHAP)[95] and the Jebson Taylor Hand Function Test (JTHF)[91]. However, time-based tasks are highly dependent on the skill of the user and do not evaluate the functionality of the device itself[129]. To truly evaluate the functionality of an upper limb prosthesis, it needs to be decoupled from its user. Whist virtual environments have been demonstrated to effectively assess the grasping of anthropomorphic manipulators; they have not seen widespread use in the prosthesis community. There is a need for more tools for upper limb prosthesis engineers and clinicians to evaluate their designs.

2.5 Gap in the Knowledge

Upper limb prostheses have high rejection rates, which has been attributed to conflicting design objectives. Prosthesis users ideally want their devices to achieve the same functionality as a biological hand. However, the desired objectives of high functionality, ease of use, comfort, low weight, and reduced cost are closely intertwined with one another[18]. Modern upper limb prosthetic devices are unable to provide enough functionality to justify the increase in weight and cost. The level of functionality that an upper limb prosthesis provides is not well understood and it makes comparisons difficult. This is one of the reasons why there is a lack of designs with optimised functionality to weight. It has not been possible to weigh the pros and cons of design features effectively and create a better prosthesis.

Contemporary clinical assessment of upper limb prosthesis performance relies on time-based assessments and user feedback, which are assumed to correlate to device functionality despite extraneous variables. To overcome this limitation, grasping simulations in virtual environments have been used in the literature to attempt to obtain quantitative measures for device performance. Various studies have been performed to calculate the ideal grasping positions of anthropomorphic manipulators, and even optimise a design parameter of an upper limb prosthesis. However, dynamic simulations have not been demonstrated, which would enable more realistic behaviour in virtual environments. Nor have they been used to compare the performance of multiple upper limb prosthetic devices.

Another method presented in the literature to assess the performance of upper limb prostheses is videogrammetric motion capture analysis. Motion capture studies can track the motion strategy of an upper limb prosthesis and identify the resulting compensatory movements. However, the results of these studies are used solely to improve rehabilitation procedures and recommended training regimes to reduce compensatory movements. Motion capture has not been used to study the design of an upper limb prosthesis with the objective of reducing compensatory movements. The use of virtual environments and motion capture to assess the mechanical design of an upper limb prosthesis is not well established. Evaluation of upper limb prostheses with respect to design considerations would enable informed design choices which could reduce rejection rates and improve user satisfaction.

Overall, there is no single solution to evaluate a prosthesis. A combination of existing methods is required to truly evaluate the functionality of a prosthesis:

- Aftercare surveys to assess the user’s satisfaction,
- Time-based tasks to assess the user’s skill,
- Simulated grasping to solely assess the upper limb prosthesis extremity,
- Motion capture to assess the motor strategy of the user.

Even with effective measures of upper limb prosthesis evaluation, it is still difficult to optimise their design due to the conflicting design objectives. There is no existing framework that considers functionality, weight, and cost. Most existing upper limb prosthetics are modular in nature when split into their main components: socket, forearm, and terminal device. There is an opportunity to optimise different combinations of components to create an upper limb prosthesis that balances the key design objectives.

This thesis addresses weaknesses in the design process of upper limb prosthetic devices by introducing three design tools that uses motion capture suit analysis, dynamic simulated grasping, and an intelligent selection method with the goal of reducing future upper limb prosthesis rejection rates.

Chapter 3
Research Methodology

3.1 Aims and Objectives

The overall aim was to improve future upper limb prostheses through the development of quantitative evaluation tools. These design tools seek to quantify the functionality of a prosthesis through the use of virtual environments, motion capture, and multi-optimisation. By developing these tools there is greater opportunity to optimise prostheses earlier in the design cycle which can result in improved functionality. With higher functionality it is anticipated that it will improve user satisfaction and therefore reduce device rejection rates.

This aim is to be achieved by completing the following objectives:

1. Develop a dynamic virtual environment to simulate grasping of upper limb prostheses and evaluate those grasps using numerical grasp quality metrics to drive design specifications.

2. Utilise an intelligent selection method to produce optimised upper limb prosthesis designs which combine both additive manufacturing and conventional engineering techniques.

3. Assess the motor strategy of an upper limb prosthesis user using motion capture analysis to gain an understanding of how prosthesis design affects compensatory movements.

3.2 Novelty and Rationale of Research

Time-based tasks are the most used quantitative method for measuring upper limb prosthesis functionality, despite their limitations as described in *Section 2.2.1*. In order to use these methods, the prosthetic device needs to have been designed, manufactured, passed mechanical testing, and gained ethical approval. These processes are expensive and time-consuming, but the only way to gain feedback regarding the functional design of an upper limb prosthesis. Another disadvantage of time-based tasks is that they suffer from user bias, where these tests are not able to assess the performance of the prosthetic device, but rather the skill of the user. Virtual environments in conjunction with quantitative grasp quality metrics can be used to assess the performance of solely the upper limb prosthesis extremity and can be performed before manufacturing. This would greatly reduce the size of the design feedback loop, and therefore allows engineers and designers to optimise their designs more quickly. While the use of virtual environments in grasping evaluation has been previously carried out in the literature, it has never been used for the purposes of design optimisation. In addition, existing simulation studies investigate static grasps where the target object is assumed to be fixed in space and inertia is not considered. To address this weakness, a dynamic virtual environment is presented in this work to simulate the grasping of abstract shapes from the Southampton Hand Assessment Procedure (SHAP). To achieve this, a kinematic modelling software called Automated Dynamics Analysis of Mechanical Systems (ADAMS) by MSC Software Corporation (California, USA) was used. MSC ADAMS is a multibody dynamics simulation software that is used to improve and optimise designs through the analysis of moving parts. This software was chosen as it eliminates the need to write complex equations of motion and allows upper limb prosthesis models to be quickly generated. MSC ADAMS considers inertia of all moving parts and can perform dynamic grasping simulations unlike the robotic simulators discussed in *Section 2.3.3*. Contact information can then be exported from MSC ADAMS into Mathworks MATLAB (Massachusetts, USA) to evaluate with a selection of numerical grasp quality metrics. Prosthetic devices with different design features can be compared with one another, allowing beneficial elements to be identified. This will provide a Virtual Prosthesis Grasp Analysis Tool (VPGAT) for engineers to evaluate their designs for grasping functionality without the need for manufacturing.

While Additive Manufacturing (AM) offers many opportunities to improve upper limb prosthesis design, there are significant barriers to its adoption. The current commercial use of AM is the manufacture of whole prosthesis models, such as Open

Bionic's Hero Arm which includes the socket, forearm, and terminal device. Where prostheses companies use conventional manufacturing techniques, the manufacturing facilities are well-established and it would be cost-prohibitive to pursue an alternative manufacturing method, despite the potential benefits to the end user. A novel model is proposed to produce hybrid designs which combine parts made via AM and conventional engineering, to leverage the advantages of both techniques. A modularity scheme is used to segment prosthesis design, where each component is assigned a utility value based on its cost, mass, and functionality. This scheme allows conflicting design objectives to be jointly optimised and can be used as an intelligent selector for prosthesis components which will result in a complete device that matches the user's requirements. This would effectively reproduce similar results to what an experienced prosthetist would recommend, but with a wider range of components and the possibility of adding new devices within its database.

One of those limitations is the over-simplification of prosthesis performance, where a rapid task completion is assumed to correlate with high functionality. To improve existing time-based tasks as an evaluation method, they can be combined with motion capture (MOCAP) analysis to provide a more in-depth understanding of how the prosthesis user moves their body during grasping. Videogrammetric MOCAP studies have been used in literature to study compensatory movements and prepare suitable training regimes for user skill. However, MOCAP has not been used to study how the design itself influences the motor strategy. This work outlines how a MOCAP suit can be used to study the compensatory motions caused from an in-house prosthesis design. A Rokoko Smart Suit Pro motion capture suit[130] (Copenhagen, Denmark) was used instead of a videogrammetric technique due to its lower technological and cost barrier, which makes it more likely to be adopted as a design tool.

Together, these three design tools can be used by engineers and clinicians to better evaluate upper limb prostheses, and therefore result in superior products and improved user satisfaction.

Chapter 4
Using Virtual Environments for the Evaluation of Upper Limb Prosthesis Grasping Functionality

4.1 Introduction

Whilst there have been significant advances in the development of upper-limb prostheses in the last 50 years[131–136], there remain high rejection and abandonment rates.[9,16,18] This is mainly attributed to a lack of functionality, where users feel the device is more of a burden than an aid. Therefore, there is a demand for upper limb prostheses with greater functionality. Despite this, evaluation methods have been unable to keep up with advances in technology. Typically, prostheses are evaluated through after care satisfaction surveys and it is assumed that if a client is satisfied with their device then it has high functionality[19]. However, the results of these qualitative surveys can be ambiguous and are prone to subjective interpretation. Contemporary quantitative methods to measure functionality are through the use of time-based tasks; several of the most commonly used procedures are outlined in *Chapter 2.3.1*. However, the results of these time-based tasks are dependent on the skill of the prosthesis user and are unable to assess the functionality of the device without bias[102]. The co-ordination of reaching and grasping makes a significant impact on how well a task is performed[98]. As a result, time-based tasks evaluate the operating skill of the prosthesis user instead of the device itself.

Currently, there is no method available to evaluate the performance of an upper limb prosthesis without manufacturing and physical testing. Without a quick process to test prosthetic devices, it can be difficult to ensure that design objectives are being met until very late in the design cycle. A solution to this problem is the use of virtual environments to simulate only the grasping action of an upper limb prosthesis. Essential hand function and the speed of performing a task can be determined by the ability of a prosthesis to form a natural and stable grip around a desired object[124]. By taking into account the contact forces, torques and contact areas, the grasp can be analysed and evaluated numerically[113]. This provides a form of evaluation which is not reliant on physical testing, which is heavily dependent on user skill, and can be performed without the need for manufacturing. Therefore, allowing designs to be compared and checked against a criteria before it is even worn by a prosthesis user. By focusing on the functionality of the hand extremity of an upper limb prosthesis, the reaching and grasping motions can be decoupled. Therefore, the prosthesis design can be evaluated through several quantitative grasp quality metrics without being impacted by other articulations or the user[125,127,128]. Virtual environments for assessing grasp quality already exist as robotic simulators, such as GraspIt!, Syngrasp, and Open Grasp, which are discussed in *Section 2.3.3*. However, these simulators assume that the target object is fixed in space and therefore inertia is not considered. They rely on geometric modelling and perfect grasping positioning, which is not a realistic representation of actual grasping of upper limb prosthesis users.

In summary, improved methods of upper limb prosthesis evaluation are required to better inform future design. The aim of this chapter is to show the development and evaluation of a novel dynamic virtual environment to simulate five upper limb prosthesis designs. The necessary virtual environment settings, contact properties, and modelling procedures are described in order to simulate grasping tasks. These grasping tasks can then be assessed using a range of numeric grasp quality metrics, which allows the design of the prosthetic to be evaluated without any user bias. Virtual environments as a design tool allows engineers to assess their design before manufacturing; this can enable more iterative designs which allows beneficial design features to be identified early, resulting in better prosthetic devices.

4.2 Methodology

In this chapter, five different upper-limb prostheses were simulated and numerically evaluated using four grasp quality metrics. The simulation task selected for evaluation of the Virtual Prosthesis Grasp Analysis Tool (VPGAT) consists of a simplified SHAP that only considers the grasping and lifting of six abstract shapes, as shown in *Figure 25* and *Table 1.* Since the objective of this research is purely to study the design of the prosthetic hand, the activities of daily living are not considered as they require use of articulations along the arm.

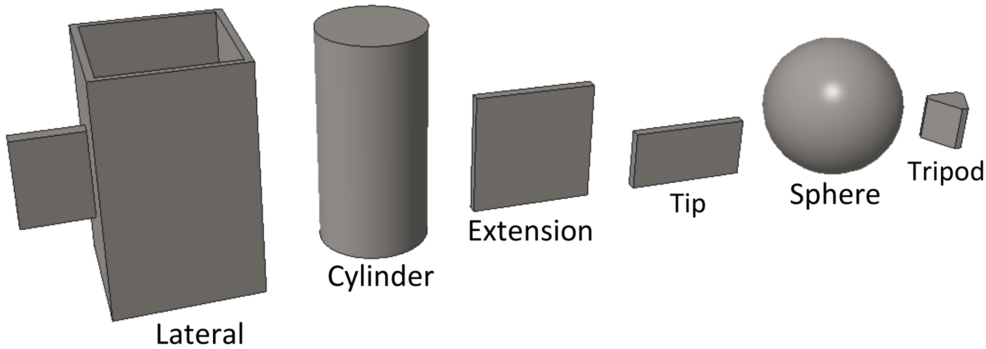

Figure 25. Southampton Hand Assessment Procedure abstract shape models[95]

SHAP Shape	Dimensions (mm)	Weight (g)
Lateral	Width – 64 Height – 107 Thickness – 3.3 Handle width – 30 Handle height – 40 Handle thickness – 6.3	228.96
Cylinder	Diameter – 50 Height – 103	536.2
Extension	Width – 60 Height – 60 Thickness – 5	142.54
Tip	Width – 60 Height – 30 Thickness – 5	71.62
Sphere	Diameter – 72	532.34
Tripod	Long side – 20 Short side – 4 Height – 24	21.08

Table 1. Heavy abstract shape properties from the Southampton Hand Assessment Procedure

4.2.1 Upper Limb Prosthesis Designs

The five prostheses tested in this study are shown in *Figure 26*: Split Hook, Clamp, eNable Raptor-Hand, CfAM-2 Hand (an in-house design) and Handii Hackberry[68]. These devices were selected due to their availability, range of design complexity and Degrees of Freedom (DoFs). A summary of the design features of the prostheses is shown in *Table 2*.

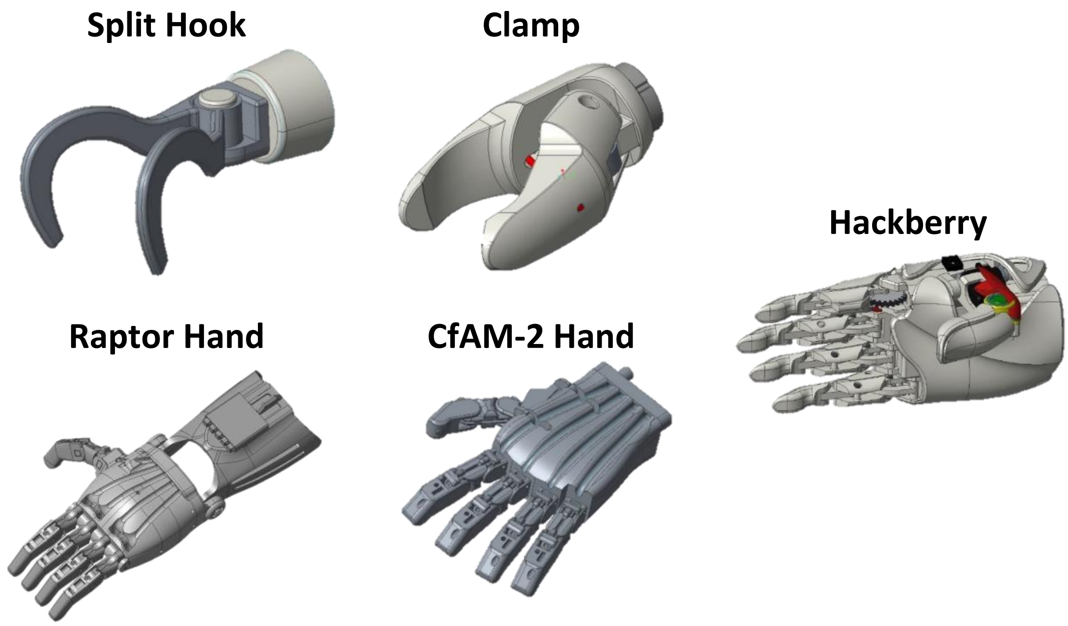

Figure 26. Upper-limb Prostheses Computer Aided Design Models

Prosthesis Design	Degrees of Freedom	Degrees of Motion	Grasping Features
Split Hook	1	1	• Opposing flat surfaces
Clamp	2	2	• Large grasping area • Opposing flat surfaces • 2 grasping topologies • Rotatable wrist
eNable Raptor-Hand	10	1	• Multi-digits • 5 coupled digits
CfAM-2 Hand	10	3	• Multi-digits • Thumb abduction • 3 prehensile grasps • 3 coupled digits
Handii Hackberry	8	3	• Multi-digits • Geared index flexion • 2 prehensile grasps • 3 off-set coupled digits

Table 2. Summary of Prosthesis Design Features

The Split Hook and Clamp design are body-powered, single DoF designs which are controlled by biscapular abduction[4]. A harness is typically strapped around the user's opposite shoulder and is connected to the claw with a cable. When the user moves their shoulder, the cable is either pulled or released, which causes the claw to open and close. Split Hooks are made from bent aluminium rods which allow them to be used as a hanging point when used as a passive device. The Clamp design has a large grasping surface area and has been designed with two grasping locations: the rounded central area for larger objects and the flat distal edges for smaller objects. It also features a rotatable wrist unit which can be manually positioned to 45° intervals.

The Raptor-Hand utilises a reverse tenodesis system[66], where all five digits are connected to a wrist tensioning system which all move in unison. The digits close with flexion of the wrist, and open with extension of the wrist[64]. It is an under-actuated prosthesis, where it has ten DoFs but only a single Degree of Actuation (DoA).

The CfAM-2 Hand[137] uses tendon cables which are attached to the tip of the finger digits and a servo motor. It features ten DoFs and three DoAs: flexion and extension of the fingers and thumb digits, and thumb abduction. The flexion and extension of the finger digits are all coupled together as they are controlled by the same servo motor. The thumb digit can be rotated to oppose the index and middle finger digits to create a tripod or cylinder prehensile grasp. The thumb digit can be held parallel with the side of the palm to produce a lateral prehensile grasp.

The Hackberry[68] has eight DoFs and three DoAs, as controlled by three servo motors. The first servo motor is used to control the extension and flexion of the index finger digit, in addition it has a tab on its intermediate phalange that allows it to curl if enough pressure is applied to it. The second servo motor is used to control the remaining finger digits. The middle, ring and little finger digits are coupled together through a gear train with a slight angle offset. The third servo motor is used as an aesthetic feature to position in the thumb in a more natural position whilst not in use, but for all functional grasps the thumb opposes the index finger digit. The Hackberry is capable of two prehensile grasps: Tip/Extension and Cylinder.

4.2.2 Software and Contact Properties

To create a virtual environment to perform dynamic simulations of upper limb prosthesis grasping, a kinetic modelling software was used: Automated Dynamics Analysis of Mechanical Systems (ADAMS) by MSC Software Corporation (California, USA). MSC ADAMS is a multibody dynamics simulation software that is used to improve and optimise designs through the analysis of moving parts. Specifically, ADAMS/View has been used in this chapter to provide a graphical user interface that allows systems to be modelled and simulated. Objects in contact generate forces which are discontinuous and non-linear, and therefore require iterative calculations. ADAMS calculates these forces using an in-built geometry engine called Rapid and Accurate Polygon Interference Detection (RAPID)[138,139]. When 3D objects are imported into ADAMS, a triangular mesh is tessellated on to the surface using tight fitting oriented bounding boxes[140]. During simulation, RAPID tests for overlaps between these bounding boxes to determine contact.

Contact forces consist of two components: the normal force and the frictional force. The normal force is the contact force that is perpendicular to the contact surface. An impact function was used to calculate the normal force which is based on four user-defined variables: stiffness, force exponent, damping, and penetration depth. This function makes any applied forces behave like a compression only spring-damper system. Contact stiffness, damping values, and penetration depths of 20 N/mm, 0.2 Ns/mm, and 0.01 mm were used respectively as recommended by the ADAMS help documentation[138,139]. The force exponent is a measure of non-linearity of the impact function's spring force and is a dependent on the material's hardness. Contact forces were expected to be low so deformations were not modelled during the simulation. Therefore, a force exponent value of 2.2 was used which is recommended for hard materials.

The frictional force is tangential to the contact surface and was approximated by the Coulomb friction model[138,139]. The frictional force can be divided into static and dynamic regimes depending on the relative velocity between the two objects. Static friction opposes any applied force as long as the object remains stationary, and once the relative velocity is non-zero dynamic friction will oppose the applied force. These frictional forces are calculated by using the coefficient of static and dynamic friction, as shown in *Equation (1)*. The values of coefficient of friction are dependent on the contact materials. Nearly all upper limb prostheses are fitted with rubber pads along the grasping outer surfaces to increase friction; to represent this coefficient of static and dynamic friction values of 1 and 0.95 were used respectively[138].

$$
\begin{gathered}
F_s \leq \mu_s F_n \\
F_d = \mu_d F_n
\end{gathered}
\tag{1}
$$

In reality, the transition between static friction and dynamic friction is a discontinuity. The coefficient of friction will instantly change from zero to μ_s at a contact velocity of zero which is known as stiction. However, ADAMS is unable to model stiction and models it instead as a continuity as shown in *Figure 27*. Values of 10mm/s and 2000mm/s were used for static and dynamic transition velocities respectively, as recommended by the ADAMS help documentation[138,139]. A summary of all the contact properties used in this simulation study is shown in *Table* 3.

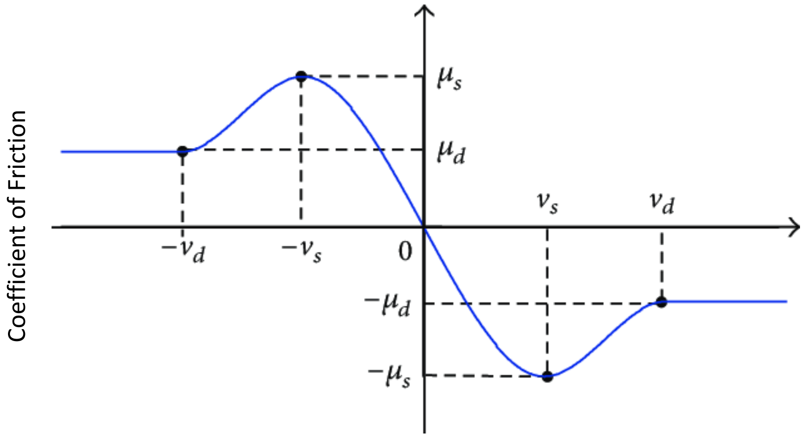

Figure 27. Coefficient of Friction vs Slip Velocity[141]

Contact Properties	Values
Static Friction Coefficient	1.0
Dynamic Friction Coefficient	0.95
Static Transition Velocity	10 mm/s
Dynamic Transition Velocity	2000 mm/s
Contact Stiffness	20 N/mm
Force Exponent	2.2
Damping	0.0001Ns/mm
Penetration Depth	0.01mm

Table 3. ADAMS Contact Properties

4.2.3 Virtual Evaluation Procedure

For the purpose of this work, only the bodies of the prostheses were modelled and not the actuation system that controls them. The force exerted by a prosthesis can vary greatly even with the same design, therefore for this study each prosthesis was modelled with the same 350 Nmm of rotational grasping torque. The simulation consisted of the prosthesis grasping an abstract shape on a flat platform and lifting it off the surface.

A Computer Aided Design (CAD) model of the prosthesis, in a Parasolid file format, was imported into ADAMS, along with the relevant test objects. The test objects were positioned to maximise the contact area with the fingers once the grasp was complete. The degrees of freedom of the fingers and thumb were defined as revolute joints which restricts their rotation to a single axis. A rotational torque was applied to each joint.

The contact properties were then defined between the test object and the hand model. Grasping forces used in this simulation are low and by approximating a hertzian contact[142], a maximum indent of 0.95 microns would be observed which is not significant. Therefore, a linear dynamic model was used which assumes all bodies are rigid with no deformations possible.

With all the model properties defined, the simulation was run over 10 seconds with 10,000 discrete steps to incrementally arrive at a steady state solution. After the prosthesis had completed its grasp around the test object, it was raised above the platform. This was to ensure that the object was fully supported by the grasp of the prosthetic hand under the effects of gravity.

The contact forces, contact torques, contact vectors, contact points, test object centre of mass (CoM), and angular positions of the finger digits at the end of the simulation were exported into a post processor to calculate the four grasp quality metrics. The complete virtual evaluation procedure is shown in *Figure 28*.

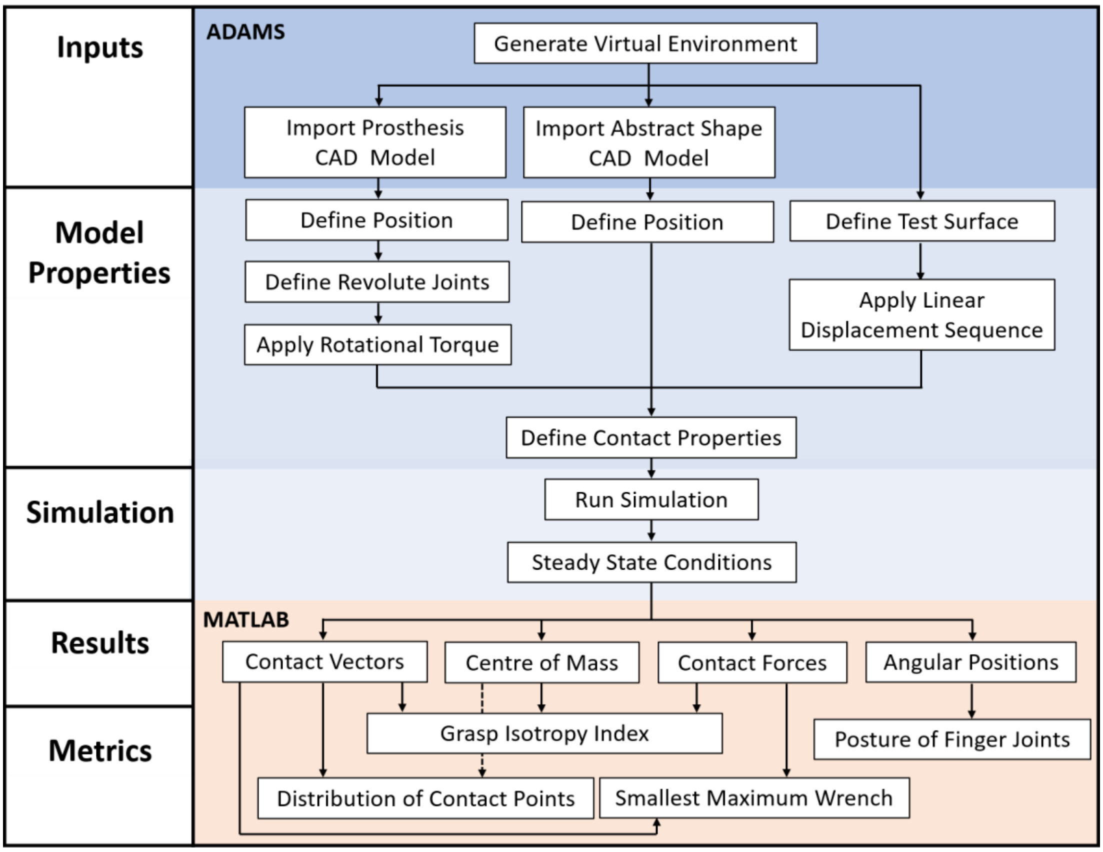

Figure 28. Virtual modelling methodology flowchart

4.2.4 Grasp Quality Metrics

Five appropriate anthropomorphic grasp quality metrics were recommended by Leon et al.[127,128] for the study of human and robotic grasps. Of those five, the following four grasp quality metrics have been considered for this study: Grasp Isotropy Index (GII), Distribution of Contact Points (DCP), Smallest Maximum Wrench (SMW) and Posture of Finger Joints (PFJ)[128]. The fifth metric, Manipulability, was omitted from this study as the resulting value would always be zero since the articulations along the arm are not being considered. These metrics were all normalised to allow comparison, where the best theoretical value is one and the worst value is zero. A value of one represents perfect balance and alignment of force vectors on a grasped object that will result in a stable grasp, which can only be displaced with a significant external force vector. In comparison, a value of zero represents force vectors which induce spin on a grasped object and are unable to reach a steady state.

To calculate the grasp quality metrics, the model shown in *Figure 29* was used. It uses three sets of frame references which are:

1. **{N}** is fixed to a stationary point which does not change throughout the simulation.
2. **{B}** is fixed to the object's CoM with its origin defined relative to **{N}** by the vector ***p***.
3. **{C}** is a frame for each contact point(i). It is defined with a new set of axes $\{\hat{n}_i \ \hat{t}_i \ \hat{o}_i\}$.

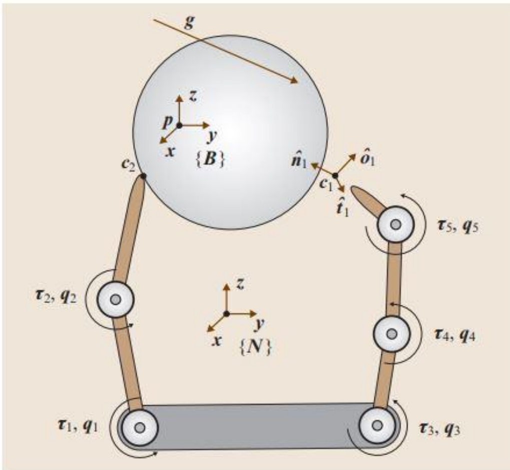

Figure 29. Diagram of a 2D Grasp of a Rigid Object with a Multi-digit Grasper[120]

The Grasp Matrix is considered one of the most important metrics[143] in grasp analysis. It defines the velocity kinematics and force transmissions properties of the contacts. Each contact between the hand's digits and the grasped object generates a *Partial Grasp Matrix*. The combination of all the Partial Grasp Matric for every contact makes up the Complete Grasp Matrix. For each contact point, the *Partial Grasp Matrix* can be calculated using *Equation (2)*.

$$\tilde{G}_i^T = \bar{R}_i^T P_i^T \qquad (2)$$

The first term is defined by:

$$\bar{R}_i = \begin{pmatrix} R_i & 0 \\ 0 & R_i \end{pmatrix} \quad (3)$$

$$R_i = [\hat{n}_i \;\; \hat{t}_i \;\; \hat{o}_i] \quad (4)$$

R_i represents the orientations of the i^{th} contact frame $\{\boldsymbol{C}\}_{\boldsymbol{i}}$ with respect to the inertial frame. The unit vectors $\hat{n}_i$, $\hat{t}_i$, $\hat{o}_i$ are expressed in $\{\boldsymbol{N}\}$. Vector $\hat{n}_i$ is the normal to the contact tangent plane and directed towards the object, $\hat{t}_i$ and $\hat{o}_i$ are vectors that are orthogonal and lie in the contact tangent plane.

The second term is defined by:

$$P_i = \begin{pmatrix} I_{3\times3} & 0 \\ S(c_i - p) & I_{3\times3} \end{pmatrix} \quad (5)$$

$I_{3\times3}$ is the identity matrix. c_i is the point of contact and is expressed in $\{\boldsymbol{N}\}$.
$S(c_i - p)$ is the cross product matrix where $S(r)$ is defined as:

$$S(r) = \begin{pmatrix} 0 & -r_z & r_y \\ r_z & 0 & -r_x \\ -r_y & r_x & 0 \end{pmatrix} \quad (6)$$

Once a *Partial Grasp Matrix* has been calculated for each contact point, they can be combined to make the *Complete Grasp Matrix* using *Equation (7)*.

$$\tilde{G}^T = \begin{pmatrix} \tilde{G}_1^T \\ \vdots \\ \tilde{G}_{n_c}^T \end{pmatrix} \quad (7)$$

Grasp Isotropy Index (GII): GII measures how similar the magnitudes of the internal forces and torques are to each other. An object is more likely to fall out of a grasp when there are large differences. It is the ratio between the smallest singular value of the Grasp Matrix (G) and the largest value, as shown in *Equation (8)*. The Grasp Matrix represents the mapping between the contact forces to the net force within the wrench space.

$$GII = \frac{\sigma_{min}(G)}{\sigma_{max}(G)} \quad (8)$$

When this metric has a value of one it is at a desirable isotropic configuration. At a value of zero it is in a singular configuration, where at least one of the DoFs cannot be controlled. This metric should be maximized. As it is a ratio it is already normalised between zero and one.

Distribution of Contact Points (DCP): The DCP describes how close the contact points are to the centre of gravity. Objects are more stable when grasped near their centre of gravity as the effects of inertia are minimised. It measures the distance between the CoM of the grasped object (*p*) and the centroid of the contact points (p_c) as shown in *Equation (9)*.

$$DCP = distance(p, c_i) \tag{9}$$

The smaller this metric is, the closer the contact points are to the CoM which results in a more stable grip; therefore, this value should be minimised. Optimal performance was set to correspond to a value of 1 for each metric. To achieve this in the case of the DCP metric, the maximum and minimum normalisation values were chosen to correspond to the shortest and longest length between the test object's CoM and outer shell, respectively.

Smallest Maximum Wrench (SMW): SMW is the largest force or torque that a grasp is capable of resisting**.** This gives an indication of how inclined an object is to be destabilised by an external force. Only the directions of the forces are used, and their magnitudes have a maximum value of 1. The origin of reference that was used was the CoM of the grasped object (*p*).

$$\mathrm{SMW} = \min_{\mathrm{w} \in \mathrm{CW}} ||\mathrm{w}|| \tag{10}$$

where w is the set of all possible wrenches. This value should be maximised so that the grasp is able to resist larger forces. The maximum value is $\sqrt{2}$ and the minimum value is zero.

Posture of Finger Joints (PFJ): The PFJ metric measures the relative position of the finger joints to their maximum limits. It provides a measure of dimension suitability, so an appropriate hand design is grasping appropriately sized objects.

$$PFJ = \frac{1}{n_q} \sum_{i=1}^{n_q} \left(\frac{y_i - a_i}{R_i} \right)^2 \tag{11}$$

$$R_i = \begin{cases} a_i - y_{im} & if\ y_i < a_i \\ y_{iM} - a_i & if\ y_i > a_i \end{cases} \tag{12}$$

Where n_q is the number of joints, y_i is the joint angle, R_i is the joint angle range between the middle position a_i and either the maximum y_{iM} or minimum y_{in} limit. This metric should be maximised where a value of zero is achieved in a completely open or closed grasp, and a value of one is achieved by the centre position.

Individual scoring of these grasp quality metrics can provide insight into the functionality of particular design features. Each of these metrics individually can be used to drive design and maximise grasping performance in a single aspect. These metrics can also be considered together by using a weighted sum, as shown in *Equation (13)*, to provide an overall measure of grasping functionality. In this thesis, the weightings of these metrics are considered in parallel with equal weights. Whilst four grasp quality metrics have been used in this study, any number of metrics can be implemented with different weightings depending on the requirements of the prosthesis.

$$Grasping\ Functionality = \begin{pmatrix} w_1 & w_2 & w_3 & w_4 \end{pmatrix} \begin{pmatrix} GII \\ DCP \\ SMW \\ PFJ \end{pmatrix} \tag{13}$$

4.3 Results

4.3.1 Individual Grasping Performance by Prosthetic Design

The grasp quality values for each prosthetic device and SHAP abstract shapes are presented as bar charts in this section, where a value of one represents the maximum possible grasping performance and a value of zero represents the lowest grasping performance.

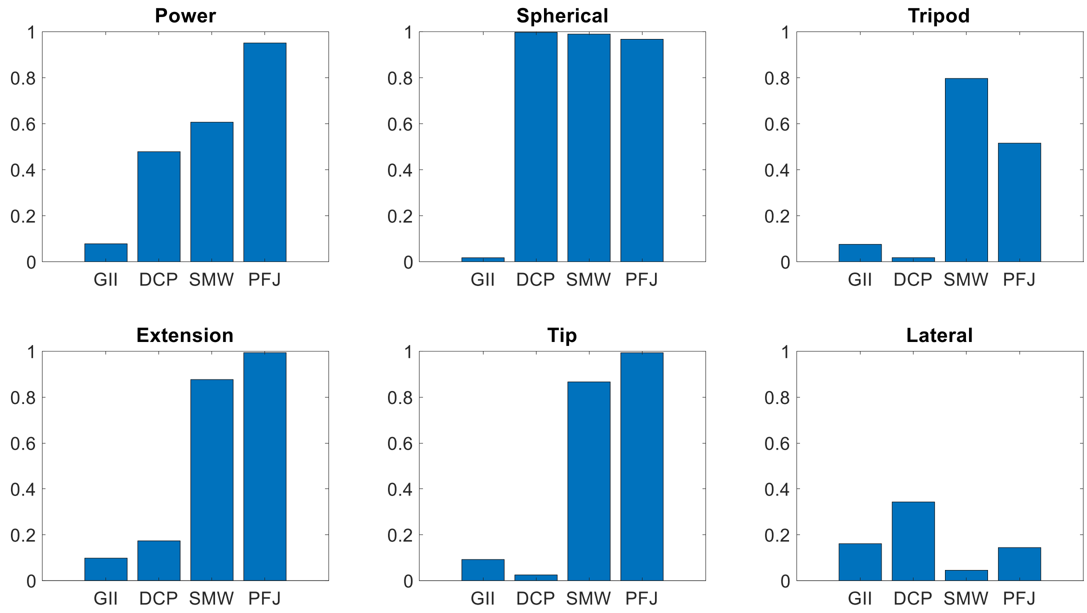

Figure 30. Split Hook Grasp Quality Metrics

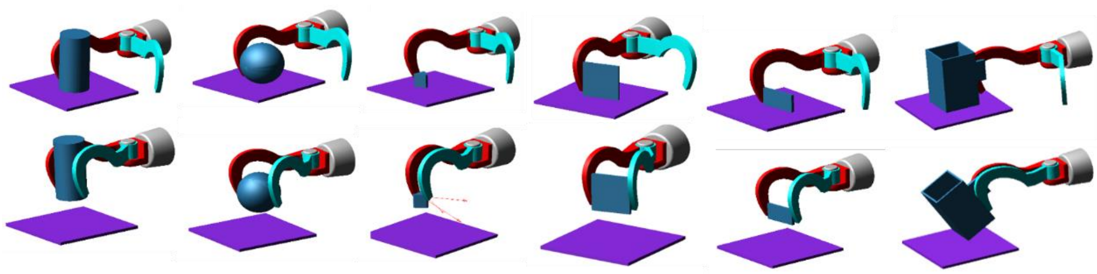

Figure 31. Initial and End Positions of the Split Hook Simulation

The grasp quality metrics for the Split Hook and a visualisation of the simulation are shown in *Figure 30* and *Figure 31* above. It is clear that the performance of the Split Hook varied greatly depending on the abstract shape being grasped. Grasping of the Spherical shape appeared to be particularly good with near-maximum DCP, SWM, and PFJ. However, this had the lowest GII metric of all of the shapes tested with the prosthetic, indicating that the forces in this grasp were anisotropic. Due to the flat grasping surfaces of the two-prong design with a single DoF, the force vectors generated when grasping anything with a curvature will always be perpendicular to one another. If not for high coefficients of friction used in this simulation model and the high weight of the abstract shapes, the grasp would have failed. Overall, the Split Hook had low GIIs due to the inability to position digits around any of the abstract shapes. High values were obtained for SMW and PFJ metrics, though performance was generally limited by poor GII and DCP metrics, particularly for the Extension and Tip shapes.

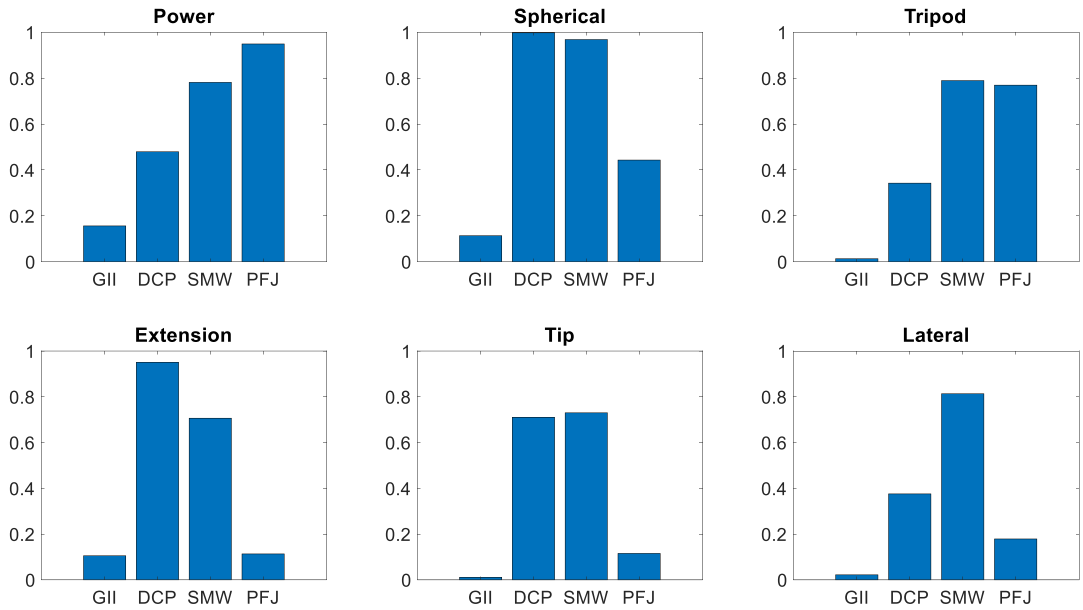

Figure 32. Clamp Grasp Quality Metrics

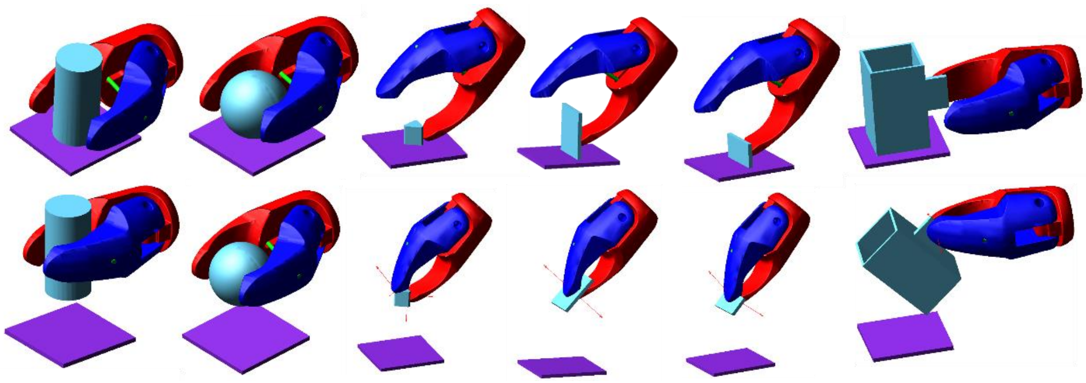
Figure 33. Initial and End Positions of the Clamp Design Simulation

The grasp quality metrics for the Clamp design and a visualisation of the simulation are shown in *Figure 32* and *Figure 33* above. The performance profile is very similar to that of the Split Hook, which is expected as they share many design features. Both have relatively flat grasping surfaces and a single DoF. GII was generally better with the clamp than with the Split Hook. This is particularly the case with the Power and Spherical shapes, where the larger rounded central grasping area could better accommodate their diameters. Grasps for the power and spherical shapes had good overall metrics. DCP and SMW were typically good across the board, with lower values obtained for GII and PFJ.

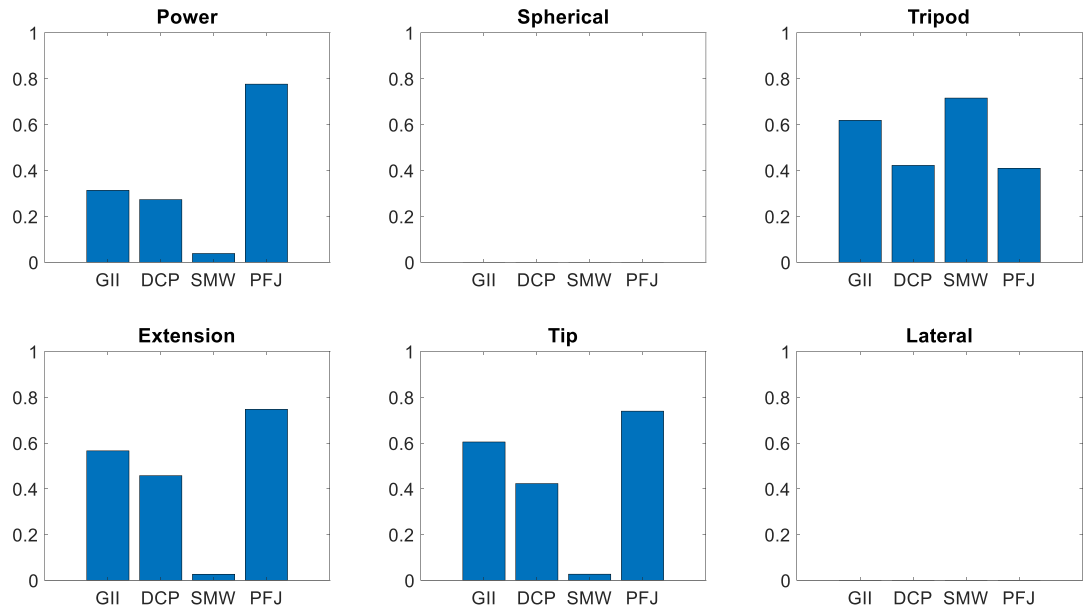

Figure 34. Raptor Hand Grasp Quality Metrics

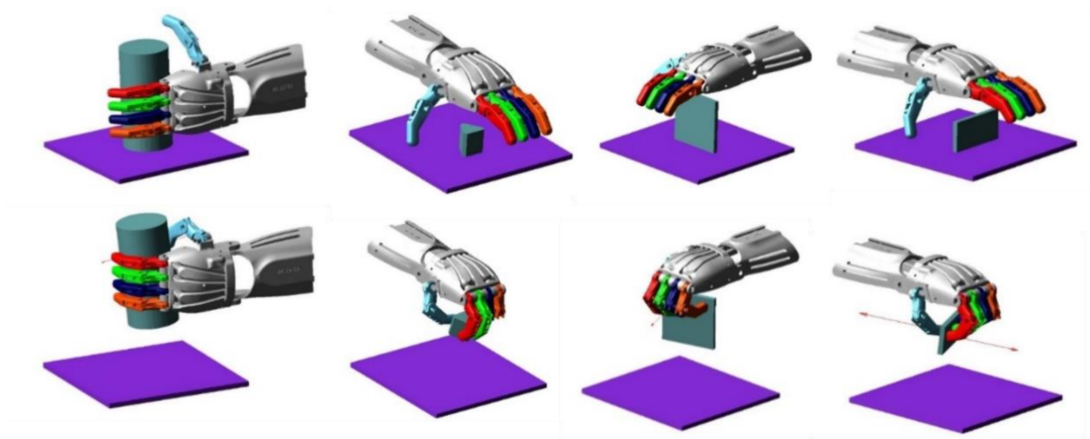

Figure 35.Initial and End Positions of the Raptor Hand Simulation

The grasp quality metrics for the Raptor Hand and a visualisation of the simulation are shown in *Figure 34* and *Figure 35* above. Data could not be obtained for the Spherical and Lateral shapes due to failure to form a grasp. The coupling of the finger digits of the Raptor Hand makes it difficult to grasp large objects and make precision grasps. Similar profiles are seen for the Power, Extension, and Tip shapes with reasonably good GII and DCP metrics, low SMW, and high PFJ. In contrast, the Tripod shape had much more even metrics across the board. It also had a notably high SMW when compared to the other shapes, and lower PFJ. The contact points were very favourable for the Tripod shape as there was a single digit on each of the three sideward faces as seen in *Figure 35*. This generated force vectors that were oriented towards the CoM, therefore improving the GII and SMW metrics.

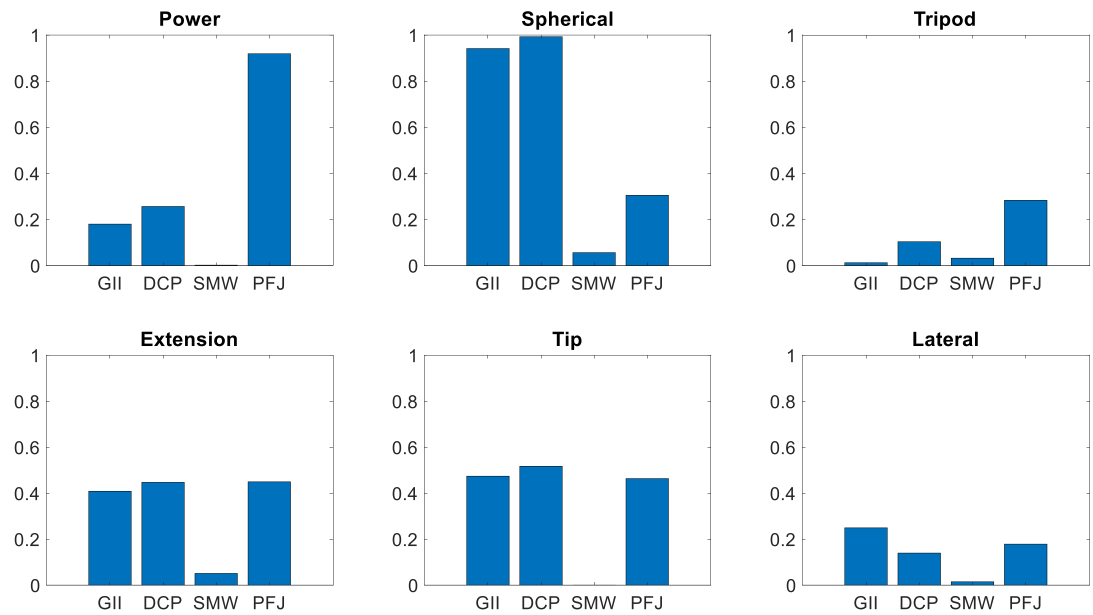

Figure 36. CfAM-2 Hand Grasp Quality Metrics

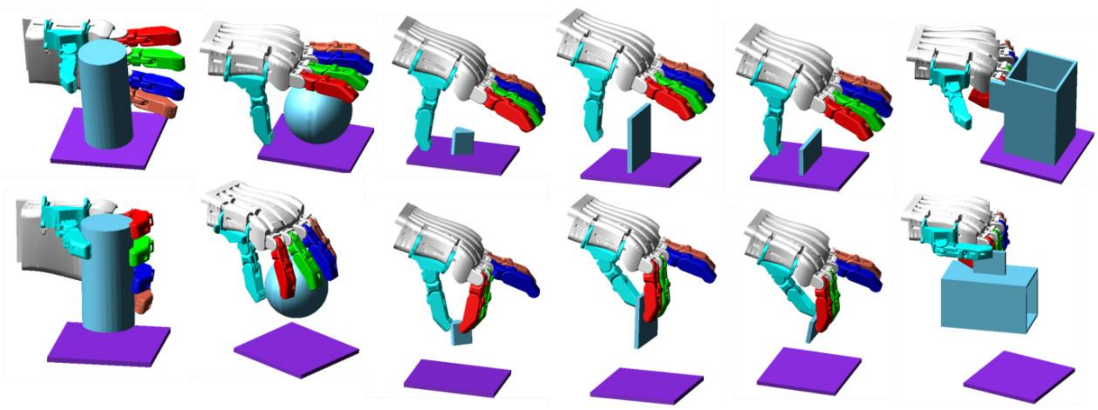

Figure 37. Initial and End Positions of the CfAM-2 Hand Simulation

The grasp quality metrics obtained for the CfAM-2 Hand and a visualisation of the simulation are shown in *Figure 36* and *Figure 37*. The CfAM-2 Hand typically had reasonably good values for all metrics except for SMW, which was low for all shapes. Grasping of the Spherical shape resulted in the highest overall metrics, with high GII and DCP. Lateral grasp had the lowest overall metrics, with a very low value for PFJ. As seen in *Figure 37*, the Lateral shape is being held by the thumb digit asserting a force against the knuckle of the index finger. Both joints are at their extreme operating limits which results in a very low PFJ value. The grasp for the Tripod shape also had very poor metrics which is due to how the forces vectors of the three digits were balancing against each other. Unlike the Raptor Hand, each digit has its own rotational torque and whilst grasping the Tripod shape some torque was induced. This caused the Tripod shape to twist and resulted in an unstable grasp, as seen in its low metric values.

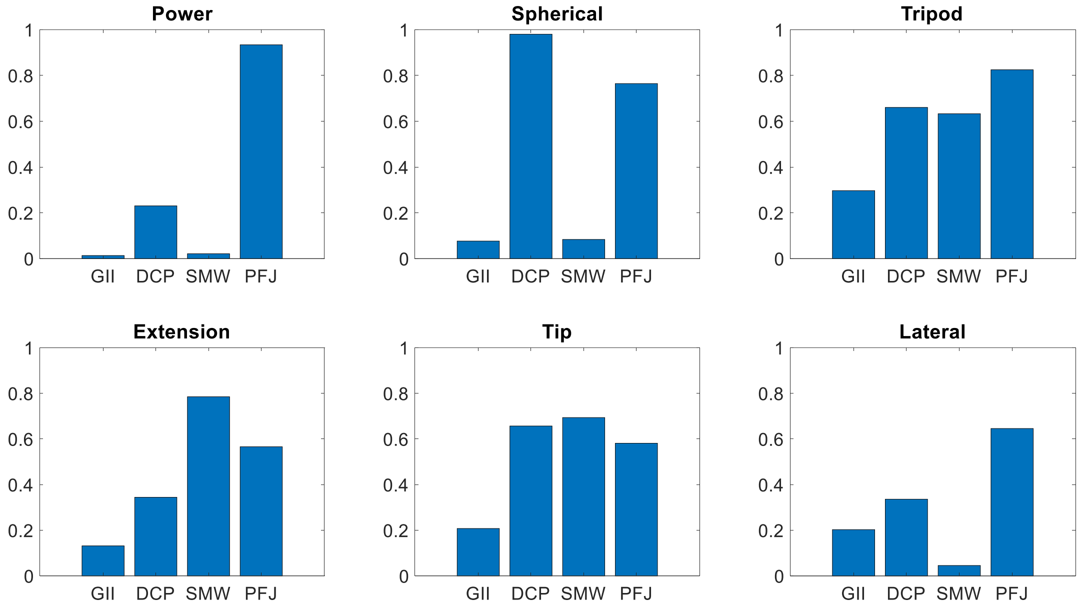

Figure 38. Hackberry Grasp Quality Metrics

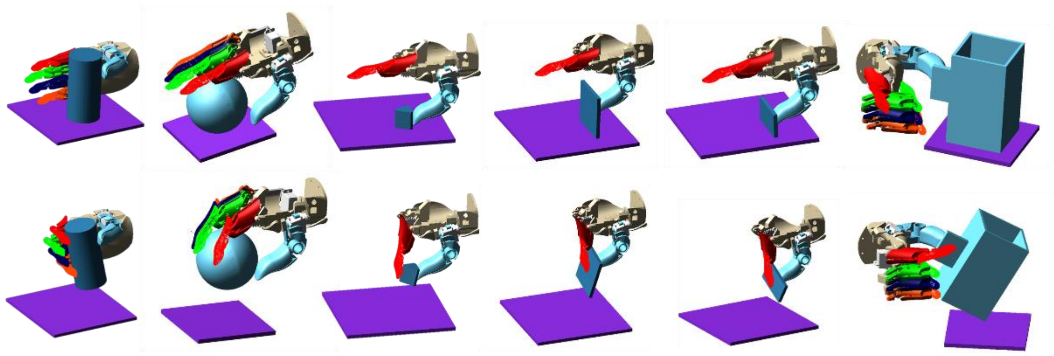
Figure 39.Initial and End Positions of the Hackberry Simulation

The Hackberry grasp quality metrics and a visualisation of the simulation are shown in *Figure 38* and *Figure 39* above. The tripod shape had the greatest overall performance, with high PFJ and low GII. The Hackberry uses two opposing digits for precision grasps and was used for the Tripod, Extension, Tip, and Lateral shapes. This allows a very controlled grasp to be achieved with minimal unwanted force vectors from other digits. This helps to improve the GII and SMW metrics, with an exception for the Lateral shape which had a very low SMW value. The grasps for the Power and Lateral shapes appeared to be the weakest, with high PFJ but performing worst for all other metrics.

The DCP metrics for the Spherical shape for all prosthetic devices were all very close to 1, with the exception of the Raptor Hand where grasping was not possible. DCP is normalised between the shortest and longest length between the CoM and the outer shell. Since a sphere has a constant radius, the maximum and minimum normalisation values are the same. Therefore, regardless of where the contact points are on the spherical shape, the DCP metric would always be 1. This was not observed with the Power shape despite having a grasping surface with a constant radius. Due to the curvature of the spherical shape, there is only a single contact point per digit. Whereas with the power shape, there can still be multiple contact points along its length which changes the average position of the contact points.

Metrics from the Extension and Tip shapes from all of the prosthetic devices almost always had the same profile. This was expected as both these shapes are geometrically similar which results in the same grasping positions. The Extension shape is double in height which changes the CoM and weight when compared to the tip shape. However, because both these shapes are always grasped close to CoM there are little differences reflected in the metrics.

Across all prosthetic devices, the Lateral shape had the lowest grasp quality metrics with especially low SMW values. This was to be expected due to the position of the grasping surface at the front of the object which is significantly offset from its CoM. The Lateral shape is also the heaviest object in this simulation study. These two factors combined cause a large torque to be generated when the Lateral shape is lifted off the surface. Any grasp that solely relies on frictional forces to keep an object suspended will result in low SMW values.

4.3.2 Average Grasping Performance by Prosthesis Design

An overall grasping functionality for the five prosthesis designs can be calculated using Equation *(13)* outlined earlier in *Section 4.2.4.* The grasping functionality scores are shown in *Table 4* below, which allows the prosthesis design as a whole to be evaluated and compared to one another. To assess the individual grasp quality metrics, the results from the VPGAT were averaged across the SHAP abstract shapes and are shown in *Figure 40*.

Prosthesis Design	**Grasping Functionality**
Split Hook	0.471
Clamp	0.485
Raptor	0.298
CfAM-2	0.309
Hackberry	0.447

Table 4. Prosthesis Grasping Functionality Scores

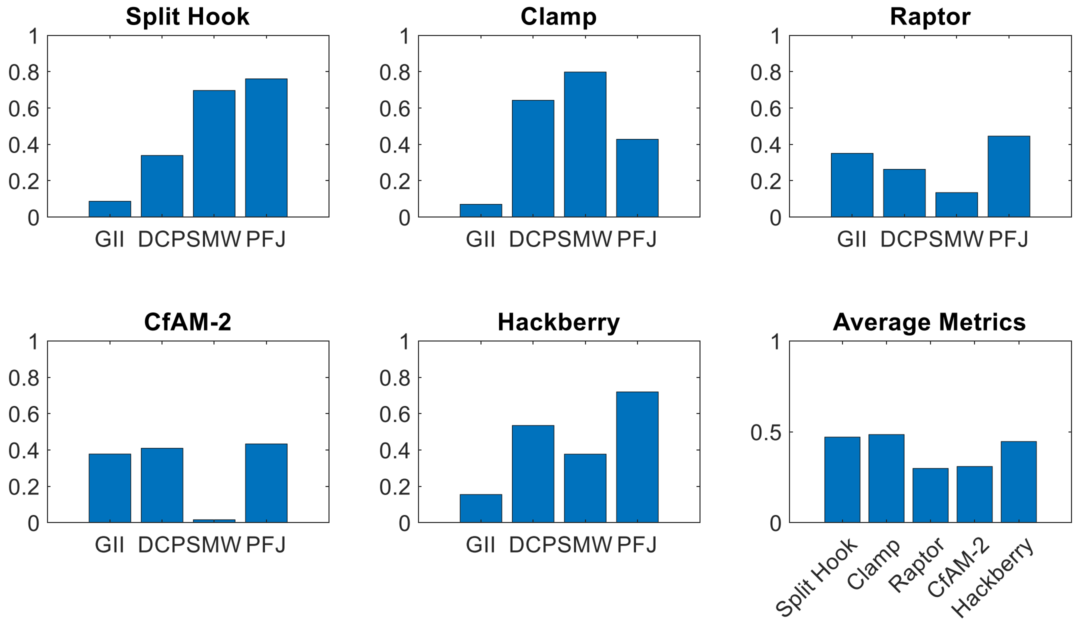

Figure 40. Average Grasp Quality Metrics for Grasping and Lifting SHAP Abstract Shapes

The Split Hook had a grasping functionality value of 0.471. It performed well on the SWM metric as it has a large surface area which increased the number of force vectors generated. The Split Hook has a small operating range which benefited the PSJ metric. However, it performed low on the DCP and GII metrics. The single DoF design prevented digits from being positioned favourably to ensure that the force vectors were equal and in opposite directions. Objects that do not have flat areas to grip create force vectors that are not parallel, therefore the grasp is more likely to be unstable during the initial grasping motion. The high aspect ratio of the hooks also created unwanted force vectors far away from the object's CoM.

The Clamp design had the best grasping functionality value of 0.485. It performed particularly well on the DCP and SMW metrics due to its large grip surface and directly opposing digits. Multiple force vectors were generated when the Clamp design grasped an object. This provided a high level of stability, as it is difficult for an external force to generate enough torque to move the object. As with the Split Hook, the Clamp design performed poorly on the GII metric due to its single DoF. Both the Clamp prosthesis and Split Hook are body powered devices, where the PFJ metric is particularly important. In this case it represents how much a patient will need to move their body to operate their prosthesis, where a larger value is more favourable.

The Raptor Hand had a grasping functionality value of 0.298. This design has ten DoFs and a single DoA, where the multiple digits were capable of curling around an object during grasping. This conforming grasp allows the force vectors to be somewhat aimed towards a single point which increases stability, as seen in the relatively high GII value. However, without active control of the additional DoFs, the Raptor hand is unable to keep the object in a stable condition after the initial grasping. The thumb digit is offset by 90° from the finger digits which creates a torque around the grasped object, thus resulting in a poor SMW value. Due to the limited volume that the Raptor hand is capable of grasping, it was unable to grasp and lift the Lateral and Sphere objects.

The CfAM-2 Hand, despite having the same DoFs as the Raptor hand and more DoAs, delivered a very similar grasping functionality value of 0.309. Each finger could conform around the grasped object which allowed for a high GII value, as compared to the other prosthesis devices, however the finger digits are controlled by a single motor, and therefore the movements are coupled. This prevents a stable grasp from being achieved after the initial contact between the finger digits and the object, as the CfAM-2 Hand was not capable of an adaptable grip. This resulted in a very low SMW value as destabilising torques were generated and additional contact points could not be added to counteract this. Whilst the CfAM-2 Hand had a higher Objective Function value than the Raptor Hand, this was due to the Raptor Hand not being capable of grasping the larger test objects. Comparing the objects which the Raptor Hand could grasp, it actually outperforms the CfAM-2 Hand. This indicates that the additional DoAs are not beneficial in a prosthesis when the finger digits can be back driven, as the opposing fingers will create force vectors which may cause joints to buckle.

The Hackberry, which is the most complicated design of the five prostheses, only achieved a grasping functionality value of 0.447. The middle, ring, and little finger digits are coupled together with a slight offset. This prevents the Hackberry from wrapping all its digits around an object. Whilst the index finger has a unique design which allows it to curl around objects if enough pressure is applied on the intermediate phalange, it is not enough to improve its GII value. It has a much better SMW than the Raptor hand and CfAM-2 Hand due to its fixed motion path. The Hackberry is a geared system where the finger digits, except the index, always move in the same path and therefore is less susceptible to being back driven whilst it is grasping an object. The movements of the actual digits are very sensitive to rotations of the servo motor that drive the gear chain, which improves its PFJ value.

The Split Hook and Clamp design outperformed the other designs, for the tests selected for this demonstration, despite being the least complex with only a single DoF. The positions of the digits in the multi-digit designs were poorly located where the resulting force vectors would cause a destabilising torque. The abduction angles between the finger digits were constant and most likely designed in this manner to appear more anthropomorphic to create an aesthetically pleasing design. Whilst it is useful to study the grasp metrics as a tool to understanding how and why a particular prosthetic has better overall functionality, it is also worth considering that for some users or activities, excellence in a particular function may be more important. Hence, in the next section, the virtual evaluation metrics are used to compare and explain how the various prosthetics performed in the different tests.

4.3.3 Grasping Performance by Abstract Shape

Each of the prostheses showed different grasping performances for the six SHAP abstract shapes, as different prehensile grips were being evaluated. The grasping functionality is represented as a radar chart as seen in *Figure 41,* where the centre of the radar charts is a value of zero and the edge a maximum value of one. Larger areas on these radar charts represent greater prosthesis performance.

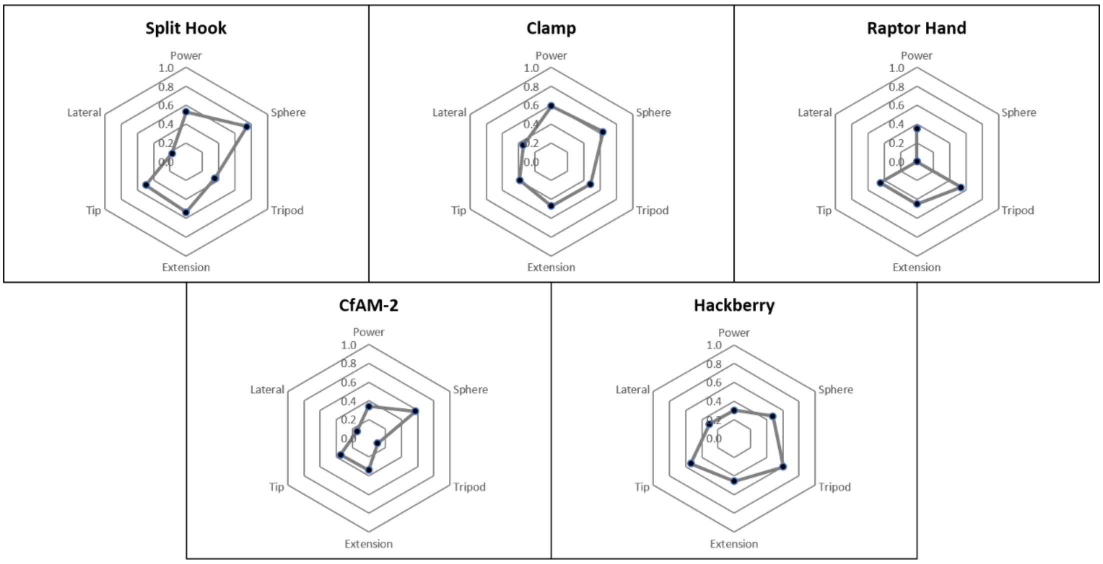

Figure 41. Prosthesis Grasping Functionality for Specific SHAP Abstract Shapes

The Split Hook and the Clamp design both feature two grasping regions, one at the tip of the claw and one along the centre. This allows both these devices to effectively form two different prehensile grasps, one for small objects and one for large objects respectively. Due to this feature, both the Split Hook and the Clamp design were able to grasp most of the SHAP objects fairly well, the exception being the Lateral shape. The CoM of the Lateral shape is significantly offset from its intended grasping location, which causes it to pivot when lifted from the table surface. This behaviour was observed with all the prosthesis devices in this study. The curvature of the central grasping region increases the grasping surface area which creates contact vectors which are directed towards a grasped object's CoM. The Clamp design performed slightly better than the Split Hook as it was much larger and created more contact points. However, with the relatively large size of the Clamp design, there was some difficulty in manipulating the smaller objects such as the Tip and Tripod shapes.

The Raptor Hand had the worst overall performance, being unable to grasp the Sphere and Lateral Shape. Due to the coupling of the finger digits at all joints, the grasping volume shrinks very rapidly which limits the size of objects it can grasp. In addition, the Raptor Hand is not capable of forming a precise opposable grasp. Both limitations prevent the Raptor Hand from grasping large objects and specific locations. The CfAM-2 Hand performed just as poorly as the Raptor Hand despite being able to grasp all the SHAP shapes. The movements of the fingers are coupled to a single motor which prevents a grasp with full contact from each finger digit. The mechanics of the digits of the Raptor Hand and CfAM-2 Hand create force vectors that push the CoM of grasped objects away resulting in limited grasping functionality.

The use of gears in the Hackberry allows it to achieve very precise grasps. The index finger and the thumb can form a stable opposable grasp that allows it to grasp smaller objects such as the Tip and Tripod shapes. However, the finger digits of the Hackberry have been designed to look anthropomorphic and have very curved surfaces, which reduces the number of contact points. As previously mentioned, the other three finger digits are coupled together with a slight offset. This prevents a grasp with full contact from each digit as seen in the poorer values for the Power and Sphere shapes.

Poor execution of mechanical design to enable prehensile grasp will result in poorer performance. CfAM-2 was designed with three prehensile grasps as its main feature but it did not consider the objects that a prosthesis user would be grasping. The grasping surfaces themselves consist of flat edges which typically result in unfavourable force vectors. For future iterations of the CfAM-2 Hand, a slightly concave surface on the digits would be recommended in order to better direct the resulting force vectors during grasping.

Overall these results suggest that the ideal prosthesis would have a large contact surface area to provide as many contact points as possible which do not generate any torques around the CoM of the grasped object; this would improve the SMW and DCP metrics. A conforming grasp with multiple finger digits will ensure that during the initial grasping the object will not be displaced and improve the GII metric. Balancing the force vectors which are created from multi-digit prosthesis is desirable to ensure that the object stays stabilised and does not compromise the SWM metric. It is important that each finger digit can apply a contact force that shifts the grasped object into a desirable location.

4.4 Summary

Using virtual environments as a test bed for assessing the performance of an upper limb prosthesis has been demonstrated in this chapter. A subset of the Southampton Hand Assessment Procedure has been replicated in a dynamic model which simulates the grasping and lifting of five upper limb prostheses. The grasping functionality performance values for these prostheses were: Clamp design (0.485), Split Hook (0.471), Hackberry (0.447), CfAM-2 Hand (0.309) and Raptor Hand (0.298). The simulation results show that single degree of freedom designs, such as the Split Hook and Clamp design, perform better than anthropomorphic designs. Each finger digit during grasping transfers a force vector onto the test object, which needs to be balanced with an opposing digit to achieve a stable grasp. Multi-articulating designs tend to have small abduction angles which results in the majority of the finger digits creating several force vectors in the same direction and therefore unable to maintain a stable grasp. Whereas prosthetic designs with fewer DoFs have an easier time achieving this balance as there are less force vectors to consider.

Simulated grasping tests in virtual environments can potentially change the typical design cycle, which allows upper limb prostheses to be evaluated and optimised before manufacturing. As a design tool, it allows engineers to modify and tweak their designs according to a quantitative measure. Whilst four grasp quality metrics have been presented in this chapter, the results from this virtual environment can be interrogated with any metric. By adopting a virtual evaluation workflow in addition to existing methods, weaknesses and strengths can be identified at an early stage and re-iterated, which can reduce time to market and ultimately saves costs. By improving the prosthesis evaluation process, better devices can be designed, and rejection and abandonment rates can be reduced.

Chapter 5
Multi Objective Optimisation of Mass, Cost, and Functionality

5.1 Introduction

The original content for this chapter was intended to be a multi-objective optimisation framework which could produce a bill of materials for prosthesis manufacturers based on mass, cost, functionality, and makespan. This was a joint research project between the Centre of Additive Manufacturing (CfAM), the Automated Scheduling and Planning (ASAP), and the Mixed Reality Lab (MRL) at the University of Nottingham. However, due to the global pandemic of the coronavirus disease in 2019, funding was cut short before completion of the project. Whilst I had developed the design of the framework, the programming and execution of the framework, which was to be led by the ASAP and MRL research groups, was not able to be carried out. Due to the lack of results with this framework, a simplified model is presented in this chapter and details of the original model is described in Section 7.3.3.

There are conflicting design objectives which prosthesis users typically desire from their devices. Users ideally want their prosthesis to achieve the same function as a biological hand. However, the desired objectives of increased functionality, ease of use, comfort, low weight and reduced cost are closely intertwined with one another[18]. Unfortunately, the current level of technology is not sufficient to meet all these demands concurrently. Improving one characteristic results in trade-offs in other characteristics. For example, while the level of functional grasping a prosthesis provides varies between models, generally the more expensive products tend to provide more functions. However, the more functions a prosthesis is capable of, the more parts are necessary, which adds more weight. Subsequently, a high-weight prosthesis could have high functionality but with high cost and poor comfort. Prosthesis users require an optimal balance which best meets their specific requirements and needs, which can be a long and difficult journey to achieve.

Due to the complex interactions between prosthesis manufacturers, prosthetists, healthcare insurance, and prosthesis users, it is very difficult to find the optimal upper limb prosthesis for a given user. There exists no tool that can help make decisions regarding the selection of upper limb prosthesis components, outside of relying on qualitative recommendations from medical professionals. To provide a quantitative tool, this chapter proposes a model that balances the top three most important criteria for upper limb prostheses: cost, weight, and functionality. Multi-objective optimisation may be carried out to assess outcomes by considering these inter-related objectives in tandem based on the prosthesis user's requirements. Using a modularity scheme, discrete utility values can be assigned to components which allows the total utility of a prosthetic device for a given user to be expressed quantitatively.

5.2 Methodology

The model proposed in this chapter presents a method by which prostheses can be optimised by using modular components, which can be selected to give the optimal combinations. A typical prosthesis consists of three main components: socket, forearm, and terminal device[144]. The socket connects the user to their device and is the component with the greatest influence on device rejection[90-145]. The socket needs to be customised to the user's residual limb in order to provide a comfortable attachment, which is then held in place either through a vacuum seal or external straps[80]. The socket is generally made by creating a plaster mould of the patient's residual limb followed by vacuum casting[146-147]. The forearm is usually a hollow shell made from silicone and plastic which houses the socket and the attachment to the terminal device. Depending on the type of terminal device, the forearm also holds any electronics or cable systems required. The terminal device provides the grasping functionality of the prosthesis, which can range from a simple two-claw split hook to a powered five-digit prosthetic hand.

It is common practice in clinical settings to select the component size, typically available as small/medium/large that is closest fitting to the user when assembling an upper limb prosthesis. These components will then be modified to fit the patient's requirements if such modifications are within budget. Whilst a bespoke design has been shown to reduce rejection rates[148], the high cost of such a design prevents this from being the standard option. Nonetheless, there is an opportunity to optimise the composition of prosthesis design by considering components built using a combination of both conventional engineering (CE) and Additive Manufacturing (AM).

5.2.1 Design Space Reduction

Multi-objective optimisation problems can become increasingly difficult to solve as more variables are considered. A modularity scheme was, hence, adopted to reduce the design space and to limit the number of possible design outcomes, as shown in *Figure 42*. With two socket designs, two forearm designs, and four terminal device designs, there are a total of sixteen possible combinations. All these components are assumed to be compatible with each other. This restriction of the design space enables simple graphical representation of the general concept of multi-objective optimisation of upper limb prosthetics. However, it should also be recognised that the general concepts can easily be expanded to more variables and different objective functions if required.

The AM components make better use of material to reduce weight due to AM's inherent design freedom[63]. In this scheme, the AM socket is modelled after a 3D scan of the prosthesis user's residual limb to provide a well-fitting attachment[149]. The AM forearm features a pattern of holes along the side to reduce mass. The CE components have stricter geometry restraints that result in simpler designs. The standard method of socket production using CE is based on castings of the users' residual limb which results in higher overhead costs compared to a 3D scan. The CE forearm is a simple extrude with little to no attempt to reduce weight. The properties of all components in the modularity scheme are shown in *Table 5*.

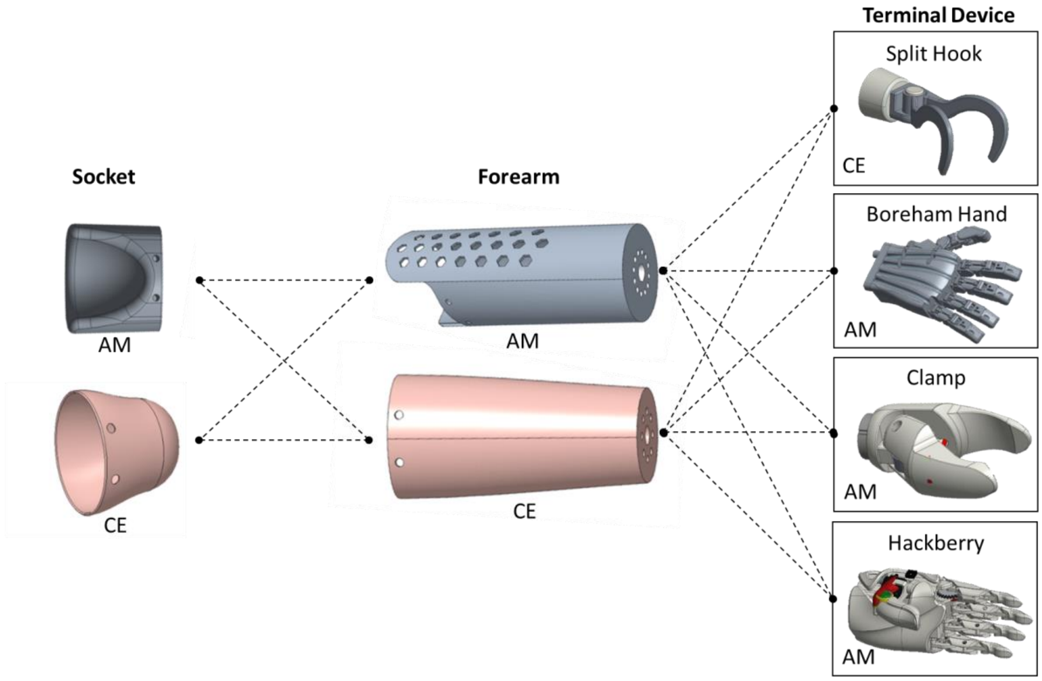

Figure 42. Prosthesis Modularity Scheme

	Mass (g)	Cost (£)	Grasping Functionality
Hosmer Split Hook #7[150]	298	490.92	0.471
Clamp	500	319.45	0.485
CfAM-2 Hand	492	244.22	0.309
Hackberry[68]	418	326.07	0.447
AM Forearm	195	178.00	-
CE Forearm[151,152]	275	256.59	-
AM Socket	120	55.29	-
CE Socket[153]	245	458.83	-

Table 5. Prosthesis Component Properties

In this model, all of the AM components are manufactured using a technique called Selective Laser Sintering (SLS). The masses of the AM components were estimated using their Computer Aided Design (CAD) model volumes and the density of sintered Nylon-12. The cost to manufacture them was based on a third-party manufacturer, 3DPrintUK[154]

(London, UK). The CE socket and forearm properties are based on a range of commercial prosthesis catalogues[151,155], and the design seen in *Figure 42* is a representative example.

The CfAM-2 Hand and Hackberry are myoelectric devices which require additional electronics which been included in the mass and cost in *Table 5;* further details can be found in *Appendix B*. The Split Hook and Clamp designs are both body-powered prosthetic devices which require a shoulder harness. The cost of a Figure of 8 Prosthesis Harness has been included in the cost in *Table 5,* but the mass has been omitted. Only the mass of components which are connected to the prosthesis user's residual limb are considered since this has the greatest influence on the comfort of the device. The CE Split Hook properties are based on the commercially available Hosmer Split Hook #7[150]. The grasping functionality values for the four terminal devices are based on the results of simulated testing in a virtual environment, as reported in *Table 4* found in *Section 4.3.2*.

5.2.2 Model Assumptions

Several assumptions have been made in order to reduce the complexity of the model. The model considers both body-powered and myoelectric prosthetic devices, which require their own additional components to control. The mass and cost of these components have been considered. However, the mechanics of the actuation systems on the socket and forearm designs have not been considered. The modularity scheme assumes that all of the components (socket, forearm, and terminal device) are compatible with one another, with no reduction in ability regardless of combination.

Prosthesis users will see their prosthetists on a regular basis to ensure that their device is performing as intended and modifications/repairs may be made affecting mass, functionality, and cost. Therefore, to simplify the after-care of upper limb prosthesis users, the model does not consider the life cycle of components and assumes that the mechanical properties of AM and CE components are identical.

The quality of the socket greatly impacts the user's comfort through design features such as better distribution of contact points on the residual limb and venting holes to reduce heat and sweating[17,146,156,157]. However, it is difficult to quantify the overall value of generic sockets made from AM and CE. Therefore, in this model, the forearm and socket are assumed to not provide any additional functionality beyond providing an attachment point for the terminal device. In addition, the quality of a socket so that it can be used by a prosthesis user are different between AM and CE sockets due to how the shape of the residual limb is captured. The residual limb shape can appear differently when it is placed into a cast where physical pressure can be applied, which typically results a better fitted socket. Whereas there is no physical pressure on the residual limb during a 3D scan and it may require multiple iterations to obtain a good socket fit. The quality of the socket is highly dependent on the skill and experience of the socket designer using either CE or AM techniques. Therefore, in this model, it is assumed that only a single iteration of sockets is manufactured for determining the costs.

5.2.3 Utility Functions

The loss of a hand causes a disutility where people with limb loss have lost functionality and productivity in their work, social, and daily lives. Applying the Hotelling model of utility[158], it can be said that a person with limb loss would have the minimum utility value of zero. On the other hand, a person with healthy limbs would have the maximum utility value of one. Different combinations of components in the modularity scheme will provide different levels of utility, and it is possible to maximise this utility based on the following three objectives: minimise cost, minimise mass, and maximise grasping functionality. The contribution to the utility of a prosthesis can be calculated using *Equations (14) - (16)* detailed below.

The cost contribution utility equation, as shown in *Equation (14)*, calculates the summation of cost for each component of the prosthesis, manufactured either through AM or CE. This is normalised between one and zero; one represents a complete device consisting of the cheapest components and zero represents a complete device consisting of the most expensive components.

$$Cost(P) = \sum_{p \in P} c(p)$$

$$\begin{aligned} &where \ldots \\ &\quad c(p)\ gives\ the\ mass\ of\ subcomponent \\ &\quad P = \{socket, forearm, terminal\ device\} \\ &\quad socket \in \{S_{AM}, S_{CE}\} \\ &\quad forearm \in \{FA_{AM}, FA_{CE}\} \end{aligned} \tag{14}$$

An analogous equation can be used to calculate the mass contribution utility, as shown in *Equation (15)*. This calculates the summation of mass for each component of the prosthesis, manufactured either through AM or CE. This is normalised between one and zero; one represents a complete device consisting of the lightest components and zero represents a complete device consisting of the heaviest components.

$$Mass(P) = \sum\nolimits_{p \in P} m(p)$$

$$\begin{aligned} &where \ldots \\ &\qquad m(p) \textit{ gives the mass of subcomponent} \\ &\qquad P = \{socket, forearm, terminal\ device\} \\ &\qquad socket \in \{S_{AM}, S_{CE}\} \\ &\qquad forearm \in \{FA_{AM}, FA_{CE}\} \end{aligned} \qquad (15)$$

Equation *(16)* gives an estimation of the grasping functionality contribution of the prosthesis, which is determined by the selection of terminal device. Grasping Functionality is normalised between one and zero, one represents the poorest performing terminal device and zero represents the best performing terminal device.

$$Grasping\ Functionality(TD) = \sum\nolimits_{td \in TD} f(td)$$

$$\begin{aligned} &where \ldots \\ &\qquad f(td) \textit{ gives the functionality of a terminal device} \\ &\qquad TD = \{terminal\ device\} \\ &\qquad terminal\ device \in \{SH, BH, C, HB\} \end{aligned} \qquad (16)$$

The three utility contribution equations need to be normalised between one and zero, to allow them to be compared to one another. The minimum and maximum values used for normalising the three utility equations are shown in *Table 6.* The normalisation limits are self-contained as they are only based on the properties of the components in the modularity scheme. While this range could be further expanded to encompass a greater variety of designs in future, the scarcity of specific information and breakdown costs of prosthetic devices makes it difficult to diversify this model without requesting data from the manufacturers. This model is also limited in that additional costs, such as service costs, are not considered; such limitations are discussed in more detail in *Section 7.3.2*.

	Minimum (Lowest Utility)	Maximum (Highest Utility)
Mass	1245 g	616 g
Cost	£1020.10	£353.44
Grasping Functionality	0.309	0.485

Table 6. Utility Function Normalisation Ranges

The total utility of the prosthesis is calculated from the product of the three utility contribution functions, *Equations (14) - (16)*, against a weighted vector as shown in *Equation (17)*. A higher value in the weighted vector indicates a higher impact on the total utility of a fully assembled prosthesis. Therefore, it is possible to identify the ideal combination of prosthesis components given the requirements.

$$Objective\ Function = \begin{pmatrix} C_w & M_w & GF_w \end{pmatrix} \begin{pmatrix} Cost(P) \\ Mass(P) \\ Grasping\ Functionality(TD) \end{pmatrix} \quad (17)$$

5.3 Results

In this section, several weighted vectors have been used to rank all possible design combinations. Each of the weighted vectors represents a specific optimisation criterion.

5.3.1 Equal Priority Objective Function

An equal priority objective function was used with the following equal vector weights: $C_w = 0.\dot{3}$, $M_w = 0.\dot{3}$, $GF_w = 0.\dot{3}$. The results of this weighted vector are shown in *Table 7*.

Rank	Socket	Forearm	Terminal Device	Total Cost (£)	Total Mass (g)	Grasping Functionality	Objective Function
1	AM	AM	Split Hook	741.21	613	0.471	0.8609
2	AM	AM	Clamp	569.74	815	0.485	0.8002
3	AM	AM	Hackberry	576.36	733	0.447	0.7924
4	AM	CE	Split Hook	821.21	693	0.471	0.7588
5	AM	CE	Clamp	649.74	895	0.485	0.6982
6	AM	CE	Hackberry	656.36	813	0.447	0.6904
7	CE	AM	Split Hook	1144.75	738	0.471	0.5743
8	CE	AM	Clamp	973.28	940	0.485	0.5136
9	AM	AM	CfAM-2	494.51	807	0.309	0.5078
10	CE	AM	Hackberry	979.90	858	0.447	0.5058
11	CE	CE	Split Hook	1224.75	818	0.471	0.4723
12	CE	CE	Clamp	1053.28	1020	0.485	0.4116
13	AM	CE	CfAM-2	574.51	887	0.309	0.4057
14	CE	CE	Hackberry	1059.90	938	0.447	0.4038
15	CE	AM	CfAM-2	898.05	932	0.309	0.2212
16	CE	CE	CfAM-2	978.05	1012	0.309	0.1192

Table 7. Prosthesis component combinations with an equal priority objective function

The top three performing prosthetic device combinations all used AM sockets and forearms, which provide the lower cost and weight option compared to the CE components. The best scoring design (0.8609) was with the Split Hook, which has the lowest possible design mass (613 g), second highest grasping functionality (0.471), and a relatively low cost (£741.21). The second highest performing design (0.8002) was with the Clamp terminal device, which has one of the lowest possible costs (£569.74) and the highest grasping functionality (0.485), at the detriment of total mass (815 g).

The lowest performing design (0.1192) consists of a CE socket, CE forearm, and the CfAM-2 Hand. Interestingly, it is neither the most expensive (£978.05) nor the heaviest (1012 g) device combination. The weakest aspect of this design combination is its low grasping functionality (0.309). The second lowest performing design (0.2212) also features a CfAM-2 Hand, even when combined with an AM forearm. The decrease in cost and mass by using the AM forearm was not enough to offset the poor grasping functionality of the CfAM-2 Hand.

5.3.2 Cost Priority Objective Function

A cost priority objective function was used with the following vector weights: $C_w = 0.8$, $M_w = 0.1$, $GF_w = 0.1$. The results of this weighted vector are shown in *Table 8.*

Rank	Socket	Forearm	Terminal Device	Total Cost (£)	Total Mass (g)	Grasping Functionality	Objective Function
1	AM	AM	Clamp	569.74	815	0.485	0.8680
2	AM	AM	Hackberry	576.36	733	0.447	0.8593
3	AM	AM	CfAM-2	494.51	807	0.309	0.8523
4	AM	CE	Clamp	649.74	895	0.485	0.7607
5	AM	CE	Hackberry	656.36	813	0.447	0.7520
6	AM	CE	CfAM-2	574.51	887	0.309	0.7450
7	AM	AM	Split Hook	741.21	613	0.471	0.7218
8	AM	CE	Split Hook	821.21	693	0.471	0.6145
9	CE	AM	Clamp	973.28	940	0.485	0.3951
10	CE	AM	Hackberry	979.90	858	0.447	0.3865
11	CE	AM	CfAM-2	898.05	932	0.309	0.3795
12	CE	CE	Clamp	1053.28	1020	0.485	0.2879
13	CE	CE	Hackberry	1059.90	938	0.447	0.2792
14	CE	CE	CfAM-2	978.05	1012	0.309	0.2722
15	CE	AM	Split Hook	1144.75	738	0.471	0.2490
16	CE	CE	Split Hook	1224.75	818	0.471	0.1417

Table 8. Prosthesis component combinations with a cost priority objective function

Much like the previous scenario, the top three performing prosthetic device combinations all used AM sockets and forearms. The highest scoring design combination (0.8680) was with the Clamp terminal device which has second lowest possible cost (£569.74), average mass (815 g), and highest grasping functionality. Despite cost being highly favoured in the weighted vector, the cheapest design has not scored first. Instead, it has scored third (0.8523) with the CfAM-2 Hand. The grasping functionality of the CfAM-2 Hand (0.309) is sufficiently low that, even though it is cheaper and lighter than the Clamp design, it has still ranked lower.

The least optimal design combination with the cost priority objective function is a CE socket, CE forearm, and the Split Hook. This was to be expected as all of these components are the most expensive in their categories.

5.3.3 Mass Priority Objective Function

A mass priority objective function was used with the following vector weights: $C_w = 0.1$, $M_w = 0.8$, $GF_w = 0.1$. The results of this weighted vector are shown in *Table 9.*

Rank	Socket	Forearm	Terminal Device	Total Cost (£)	Total Mass (g)	Grasping Functionality	Objective Function
1	AM	AM	Split Hook	741.21	613	0.471	0.9583
2	AM	CE	Split Hook	821.21	693	0.471	0.7901
3	AM	AM	Hackberry	576.36	733	0.447	0.7313
4	CE	AM	Split Hook	1144.75	738	0.471	0.6573
5	AM	AM	Clamp	569.74	815	0.485	0.5926
6	AM	CE	Hackberry	656.36	813	0.447	0.5631
7	AM	AM	CfAM-2	494.51	807	0.309	0.5187
8	CE	CE	Split Hook	1224.75	818	0.471	0.4891
9	CE	AM	Hackberry	979.90	858	0.447	0.4304
10	AM	CE	Clamp	649.74	895	0.485	0.4244
11	AM	CE	CfAM-2	574.51	887	0.309	0.3505
12	CE	AM	Clamp	973.28	940	0.485	0.2917
13	CE	CE	Hackberry	1059.90	938	0.447	0.2622
14	CE	AM	CfAM-2	898.05	932	0.309	0.2177
15	CE	CE	Clamp	1053.28	1020	0.485	0.1235
16	CE	CE	CfAM-2	978.05	1012	0.309	0.0495

Table 9. Prosthesis component combinations with a mass priority objective function

The top two designs both have the Split Hook and AM socket, where the best design has the AM forearm and the second-best design has the CE forearm. The top design (0.9583) has the lowest mass combination possible (613 g), with the second highest grasping functionality (0.471) and a below average cost (£741.21). The low weight of the Split Hook still made it rank second even when combined with a CE forearm. Though there is a significant drop in the objective function score (0.7901).

The two lowest performing combinations both have CE sockets and CE forearms, where the least optimal combination has the CfAM-2 Hand and the second worst has the Clamp terminal device. The poorest design (0.0495) has a high cost (£978.05), high weight (1012 g), and the lowest grasping functionality (0.309, CfAM-2 Hand). The second worst design (0.1235) with the Clamp terminal device has an objective score almost 2.5 times higher than that with the CfAM-2 Hand. The Clamp terminal device increases the cost and the weight albeit compared to the CfAM-2 Hand but provides a significant increase in grasping functionality (0.485).

5.3.4 Grasping Functionality Priority Objective Function

A grasping functionality priority objective function was used with the following vector weights: $C_w = 0.1$, $M_w = 0.1$, $GF_w = 0.8$. The results of this weighted vector are shown in *Table 10.*

Rank	Socket	Forearm	Terminal Device	Total Cost (£)	Total Mass (g)	Grasping Functionality	Objective Function
1	AM	AM	Clamp	569.74	815	0.485	0.9401
2	AM	CE	Clamp	649.74	895	0.485	0.9095
3	AM	AM	Split Hook	741.21	613	0.471	0.9026
4	AM	CE	Split Hook	821.21	693	0.471	0.8720
5	CE	AM	Clamp	973.28	940	0.485	0.8541
6	CE	CE	Clamp	1053.28	1020	0.485	0.8235
7	CE	AM	Split Hook	1144.75	738	0.471	0.8166
8	AM	AM	Hackberry	576.36	733	0.447	0.7866
9	CE	CE	Split Hook	1224.75	818	0.471	0.7860
10	AM	CE	Hackberry	656.36	813	0.447	0.7560
11	CE	AM	Hackberry	979.90	858	0.447	0.7006
12	CE	CE	Hackberry	1059.90	938	0.447	0.6700
13	AM	AM	CfAM-2	494.51	807	0.309	0.1523
14	AM	CE	CfAM-2	574.51	887	0.309	0.1217
15	CE	AM	CfAM-2	898.05	932	0.309	0.0664
16	CE	CE	CfAM-2	978.05	1012	0.309	0.0357

Table 10. Prosthesis component combinations with a grasping functionality priority objective function

Due to the limited options for grasping functionality, it is no surprise that the top four performing designs consisted of either the Split Hook (0.471) or Clamp (0.485) terminal device. The best design combination (0.9401) has an AM socket, AM forearm, and Clamp terminal device, which provides a low cost, average weight, and the highest grasping functionality.

The worst four designs are all the possible permutations with the CfAM-2 Hand as the terminal device which has the lowest grasping functionality (0.309). As expected, the worst design (0.0357) consisted of a CE socket, CE forearm, and the CfAM-2 Hand, which results in an above average cost (£978.05), high mass (1012 g), and the lowest grasping functionality.

5.3.5 Literature-based Objective Function

The previous four scenarios have used vector weights based on maximising a design objective. A literature-based objective function can be made by using actual results from satisfaction surveys of upper limb prosthesis users. An example of how existing literature can be used to generate the objective function is given below, based on the satisfaction studies compiled by Biddiss et al.[9]. In this study, prosthesis users were asked to rank a selection of design aspects in order of most to least important. Priority scores were generated from these surveys, where design aspects that were most frequently considered the first priority were given the highest priority scores. The relevant design aspects were grouped together and converted into a ratio as seen in *Table 11*, further details of the design aspects and their priority scores can be found in *Appendix C*. Therefore, a literature-based objective function can be made with the following vector weights as informed by user feedback: $C_w = 0.29$, $M_w = 0.56$, $GF_w = 0.15$. This was used to generate results shown in *Table 12.*

Design Aspect	Total Priority Score	Proportion
Weight	174	0.57
Cost	53	0.17
Grasping Functionality	80	0.26

Table 11. Weight, Cost, and Grasping Functionality design priority scores based on upper limb prosthesis users[9]

Rank	**Socket**	**Forearm**	**Terminal Device**	**Total Cost (£)**	**Total Mass (g)**	**Grasping Functionality**	**Objective Function**
1	AM	AM	Split Hook	741.21	613	0.471	0.9219
2	AM	CE	Split Hook	821.21	693	0.471	0.7912
3	AM	AM	Hackberry	576.36	733	0.447	0.7567
4	AM	AM	Clamp	569.74	815	0.485	0.6996
5	CE	AM	Split Hook	1144.75	738	0.471	0.6529
6	AM	CE	Hackberry	656.36	813	0.447	0.6261
7	AM	CE	Clamp	649.74	895	0.485	0.5689
8	CE	CE	Split Hook	1224.75	818	0.471	0.5222
9	CE	AM	Hackberry	979.90	858	0.447	0.4877
10	AM	AM	CfAM-2	494.51	807	0.309	0.4683
11	CE	AM	Clamp	973.28	940	0.485	0.4306
12	CE	CE	Hackberry	1059.90	938	0.447	0.3571
13	AM	CE	CfAM-2	574.51	887	0.309	0.3376
14	CE	CE	Clamp	1053.28	1020	0.485	0.2999
15	CE	AM	CfAM-2	898.05	932	0.309	0.1993
16	CE	CE	CfAM-2	978.05	1012	0.309	0.0686

Table 12. Prosthesis component combinations with a literature-based priority objective function

The optimal combinations were near-identical to rankings obtained using the mass priority objective function, despite the weighting for mass dropping to 0.57 from 0.8 between the literature based objective function and the mass priority function. The top design (0.9219) has the lowest mass combination possible (613 g), with the second highest grasping functionality (0.471), and a below average cost (£741.21). The worst design combination (0.0686) with the literature based objective function was a CE socket, CE forearm, and the CfAM-2 Hand. This has a high cost (£978.05), high weight (1012 g), and the lowest grasping functionality (0.309), which once again matches the results from the mass priority objective function.

5.3.6 Overall Trends

Across all scenarios with varying weighted vectors for the objective function, there were several similarities seen. Firstly, the optimal designs had AM sockets and AM forearms as these were cheaper and lighter than their CE equivalents. The AM components were assumed to have no impact on functionality, so these AM parts would always provide greater utility. Using an AM socket over a CE socket always provided a greater improvement in the objective function, than using an AM forearm over a CE forearm: the AM forearm is 88% lighter and 68% cheaper than its CE counterpart, whereas the AM socket is 36% lighter but 157% cheaper than the CE socket. There is significant cost saving with the AM socket that, even in scenarios where mass has a higher value in the weighted vector, the AM socket still provided greater utility.

The Split Hook consistently scored highly in all scenarios, with the exception of the cost priority objective function. The Split Hook may be the most expensive terminal device available, but it has the lowest mass and provides the second highest grasping functionality; indeed, it's grasping functionality score is only 3% lower than the highest performing terminal device, the Clamp. At the other end of the spectrum, the CfAM-2 Hand typically scored very low in all scenarios due to its detrimentally low grasping functionality. Only in a single instance did the CfAM-2 Hand make it to the top 3 designs: under the cost priority objective function when combined with AM components.

Overall, the weighted vectors in the objective functions has worked as intended. However, more component options are required to provide a better spread of data points, particularly for grasping functionality as there are three terminal devices with very similar values and the CfAM-2 Hand with significantly lower values, which somewhat skews the resulting plots.

5.4 Summary

There are high rates of device rejection among upper limb prosthesis users. This is due to high expectations of prosthesis users where they desire high functionality, ease of use, comfort, low weight and reduced costs. These are conflicting design objectives and need to be addressed as a whole rather than individually. The model described in this chapter can be used as an intelligent selector for a range of prosthesis components to select the most optimal parts. This is achieved using a modularity scheme that gives each component a discrete value, where different combinations of these components will result in different levels of utility. Utility contribution functions were used to define the value of cost, mass, and grasping functionality. The product of these contribution functions and a weighted vector gave the total utility of the prosthesis, known as the objective function.

Several scenarios were given with different values for the weighted vector: equal priority, cost priority, mass priority, grasping functionality priority, and a literature-based objective function. An example was given on how existing research studies of prosthesis satisfaction could be used to generate the weighted vector. Using the literature-based objective function, the Split Hook combined with AM sockets and forearm were the optimal designs. The AM components were cheaper and lighter than their CE counterparts. The Split Hook whilst more expensive than an AM terminal device, offers great grasping functionality, and low mass.

In summary, this design tool allows the identification of optimal components within its modularity scheme for a given objective function, which replicates what an experienced prosthetist may do in practise when prescribing devices to their patients.

Chapter 6
Characterising Compensatory Movements using Motion Capture

6.1 Introduction

There has been an identified need to improve and standardise upper limb prosthesis evaluation methodologies[2]. Whilst there are a number of quantitative evaluation methods available[20,84], as discussed in Chapter 3.2, they have severe limitations and over-simplify the performance of a hand prosthesis. These evaluation methods are based on time-based tasks which were originally designed for upper limb impairments and not specifically for prosthesis users.

Using an upper limb prosthesis effectively requires coordinating all the remaining articulations in the arm. The musculoskeletal system of the upper body contains redundant degrees of freedom, so even with the loss of a hand, a prosthesis user has several motor strategies to complete a goal orientated task. Upper limb prostheses typically do not replace the forearm pronation/supination and wrist flexion/extension, which tends to result in prosthesis users making excessive motions at the shoulder and trunk[10]. However, these compensatory movements can result in overuse injuries and place the body in unnatural positions, likely leading to device rejection[101]. Understanding the combined motion of the shoulder, elbow, wrist, and hand is thus required to produce a functional prosthesis.

Previous videogrammetric motion capture (MOCAP) studies have characterised the motor strategy of prosthesis users, but they have mainly been used as a clinical tool to prescribe a training routine to reduce future compensatory movements[101,103]. The results of these studies are not typically fed back into the design process. This is potentially a missed opportunity, and the results of these studies could instead be used to assess upper limb prosthetic devices and used to reduce compensatory movements of future design iterations.

The goal of this chapter is to outline how a MOCAP suit could be used to complement existing quantitative evaluation methods to improve future upper limb prostheses. The motor strategy with and without a prosthetic device is compared by obtaining MOCAP data. Multiple inertial based sensors track several key positions on the upper body, and the deviation between the two motor strategies can be used as a measure of compensatory movements. These compensatory movements can then be attributed to design features of the prosthetic device and used to inform future design.

6.2 Methodology

In this chapter, the motion strategy of a participant using their healthy limb and an upper limb prosthesis to perform a series of tasks was measured using a MOCAP suit. These tasks were selected from the Southampton Hand Assessment Procedure (SHAP) and are explained in greater detail in *Section 6.2.3* below. The difference in translational and rotational displacement of the MOCAP suit sensors between the healthy limb and the upper limb prosthesis was used to determine the compensatory movements caused by the device.

6.2.1 Participant

A single able-bodied participant was used as a sample of convenience, due to lack of access to upper limb prosthesis users. The participant was right-handed and had no previous experience in operating an upper limb prosthesis. In order to allow the participant to operate the upper limb prosthesis, an in-house designed able-bodied adapter[159] was used as shown in *Figure 43*. The able-bodied adapter allowed the participant to operate the grasp of a prosthesis by flexing and extending the wrist. It was secured to the participant with two touch fasteners on the distal and proximal ends of the forearm. This restricted the rotational DoFs in the forearm and prevents the wrist from impacting the positioning of the prosthesis. Several studies of upper limb prosthesis evaluation have used able-bodied adapters to measure their functionality through time-based tasks[3-84]. The learning capabilities of motor tasks by amputees have been shown to be similar to those of able-bodied subjects in a study by Dromerick et al[160]. The prosthesis and able-bodied adaptor was attached to the participant's left arm in order to mimic prosthesis users, where the side of upper limb loss becomes their non-dominant side[84].

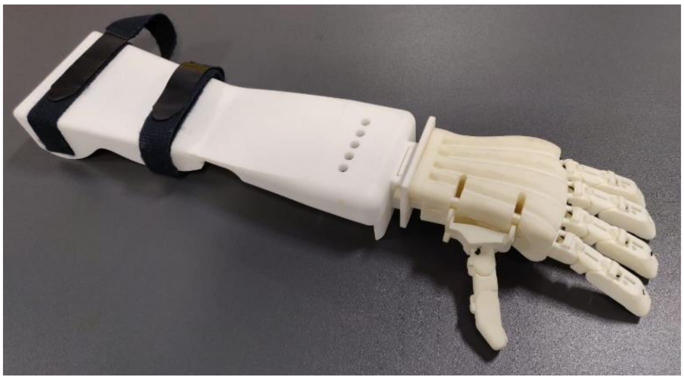

Figure 43. In-house design able-bodied adapter[159]

Ethical approval for this study was obtained from University of Nottingham Faculty Research ethical committee. Before any data collection began, the test protocol and the handling of their digital data were explained to the participant. Informed consent was acquired from the participant.

6.2.2 Prosthetic Device

The CfAM-2 Hand[137], the second in-house upper-limb prosthesis design developed at the Centre for Additive Manufacturing[161] (Nottingham, UK), was used for this study. It uses a cable pulley system where the tips of the fingers are attached to the body powered attachment on the ABA. In this configuration it features ten DoFs and two DoAs: flexion and extension of all the digits (fingers and thumb are coupled together), and thumb abduction. These DoAs provide the three prehensile grasps: Tripod, Cylinder and Lateral, as shown in *Figure 44*. An elastic material that runs along the back of the digits is used to extend the fingers, while operation of the ABA would cause flexion. The thumb digit can be rotated to oppose the index and middle finger digits to create a Tripod or Power prehensile grasp. The thumb digit can be held parallel with the side of the palm to produce a Lateral prehensile grasp. This device was selected as it features a unique thumb abduction design which is not currently employed on commercial prostheses. Therefore, the results of the MOCAP analysis could be used to comment on the motor strategy influenced by this design feature.

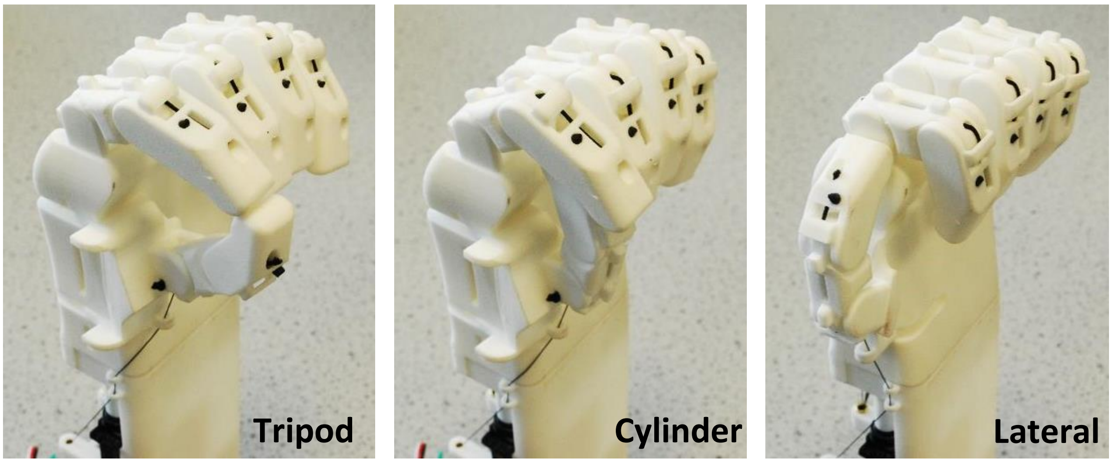

Figure 44. CfAM-2 Hand Prehensile Grasps

6.2.3 Test Protocol

The series of time-based tasks the participant was asked to perform were selected from the SHAP[92–94]. This quantitative evaluation method was used as it categorised different objects into prehensile grasps, therefore allowing the prehensile grasps of the CfAM-2 Hand to be assessed. The SHAP also features a wide range of objects which will provide a more diverse motion strategy compared to other repetitive time-based tasks such as the Box and Block Test. Due to the low energy transfer efficiencies of the cables in the CfAM-2 Hand and the able-bodied adapter, only the light variants of the abstract shapes could be lifted consistently. The six lightweight abstract shapes are shown in *Figure 45*, and their properties and intended CfAM-2 Hand prehensile grasps are shown in *Table 13*. Following SHAP guidelines[162], the test was completed in a sitting position with the seat adjusted such that the participant's elbow was at a 90° angle with the table. The SHAP foam board was placed in front of the participant, approximately 8cm from the edge of the table. The participant was given a maximum of five minutes to practise and familiarise themselves with using the prosthesis before any data collection took place. As instructed in the SHAP, the participant was asked to start an over-sized button timer with the prosthesis, then to pick up the abstract shape, place it down in an allocated spot and stop the timer.

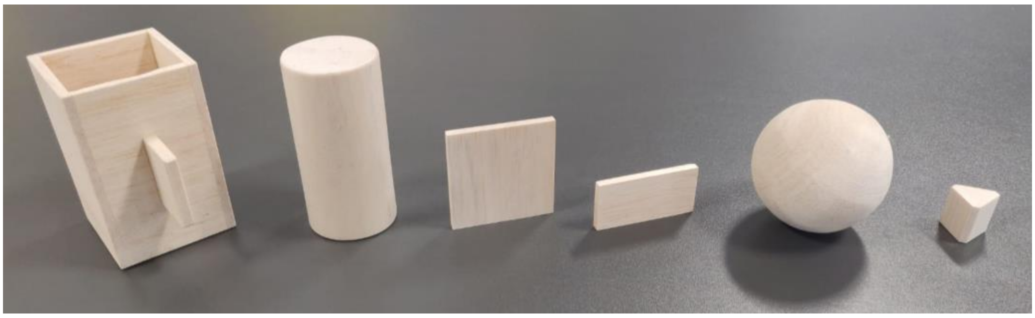

Figure 45. Southampton Hand Assessment Procedure abstract shape[95]

SHAP Shape	Dimensions (mm)	Weight (g)	CfAM-2 Hand Prehensile Grasp
Lateral	Width – 64 Height – 107 Thickness – 3.3 Handle width – 30 Handle height – 40 Handle thickness – 6.3	17.38	Lateral
Power	Diameter – 50 Height – 103	22.87	Power
Extension	Width – 60 Height – 60 Thickness – 5	1.79	Tripod
Tip	Width – 60 Height – 30 Thickness – 5	0.76	Tripod
Spherical	Diameter – 72	25.96	Power
Tripod	Long side – 20 Short side – 4 Height – 24	0.99	Tripod

Table 13. Light abstract shape properties from the Southampton Hand Assessment Procedure

6.2.4 Data Collection

An inertial based MOCAP system, a Rokoko Smart Suit Pro motion capture suit[130] (Copenhagen, Denmark), was used to measure the motion strategy of the participant. Whilst marker-based MOCAP systems are considered the gold standard when used in lower limb kinematic studies[163], in a study by Karatsidis et al.[164], it was found that the results for upper limb studies were essentially the same as an inertial based MOCAP system. MOCAP suits offer a relatively low-cost method to study human movements as they are self-contained and require less technological expertise to set up, typically costing in the range of £300 to £7 000. Whereas, videogrammetric MOCAP techniques cost between £1 400 and £10 000 requiring a blank screen backdrop, placed motion markers, commercial grade video cameras, and a dedicated space for video capture.

The Rokoko Smart Suit Pro uses a total of nineteen inertial measurement unit sensors to track the articulations of the wearer. Each sensor is fitted with an accelerometer and gyroscope which allows it to provide full spatial and rotational tracking. In this study, only the articulations of the left upper limb are relevant for performing the test protocol. Therefore, only the forearm, upper arm, shoulder, trunk (both sides) sensors were considered, as shown *Figure 46*.

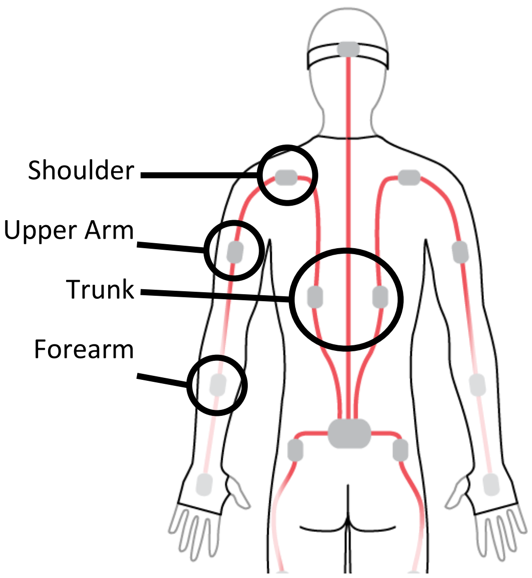

Figure 46. Motion Capture Suit Sensors of Interest

The participant was fitted with the MOCAP suit and was calibrated using Rokoko's recommended guidelines[130] at the start of the testing session. The participant completed three repeats of the test protocol without the prosthesis to set the baseline of their control motor strategy, and completed a further three repeats with the CfAM-2 Hand. The translational and rotational displacements of the five MOCAP suit sensors, which are measured from their calibration position, were exported into Mathworks MATLAB (Massachusetts, USA) for data processing. As there were 2 sensors for tracking the trunk, the results from both sensors were averaged before data smoothing and peak identification.

6.2.5 Data Processing

MOCAP analysis yields a significant quantity of data points and requires condensing so that it can be presented in a meaningful way. To achieve this, the total linear and rotational displacements for each sensor are calculated using *Equation (18)* and *Equation (19)* respectively.

$$Total\ Translational\ Displacement = \sqrt{|x_l|^2 + |y_l|^2 + |z_l|^2}$$
$$where\ ...$$
$$x_l\ is\ the\ linear\ displacement\ in\ the\ X\ plane \qquad (18)$$
$$y_l\ is\ the\ linear\ displacement\ in\ the\ Y\ plane$$
$$z_l\ is\ the\ linear\ displacement\ in\ the\ Z\ plane$$

$$Total\ Rotational\ Displacement = \mathrm{x_r} + \mathrm{y}_r + \mathrm{z}_r$$
$$where\ ...$$
$$x_r is\ the\ rotational\ displacement\ in\ the\ X\ plane \qquad (19)$$
$$y_r\ is\ the\ rotational\ displacement\ in\ the\ Y\ plane$$
$$z_r\ is\ the\ rotational\ displacement\ in\ the\ Z\ plane$$

As seen in *Figure 47* below, the total displacements can be plotted over the duration of the grasping task to show the motor strategy of the participant. Three distinct peaks are expected from repeating the grasping task three times. However, the MOCAP data can contain a lot of noise due to micro movements from the participant.

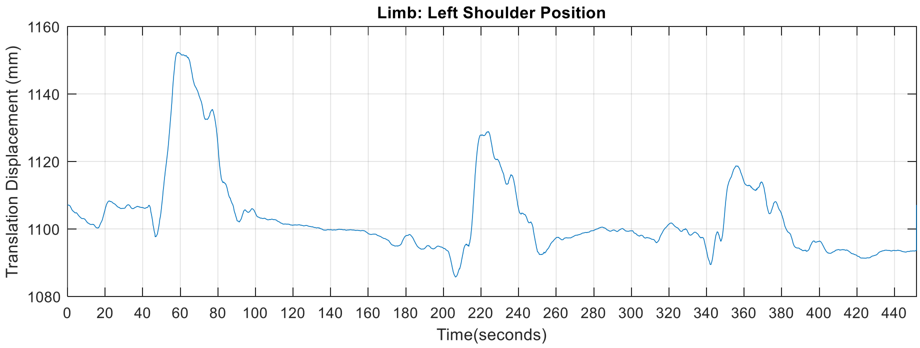

Figure 47. Total Translational Displacement of the Left Arm Sensor

In order to smooth this data for analysis, a rolling average of a hundred frames was used, as shown in *Figure 48*. This allows a peak identification function to be used, based on its intrinsic height and its location to other peaks. The prominence of these peaks equates to the maximum sensor displacement. However, the smoothing function is not capable of removing all noise as there are still multiple small peaks seen in seen in *Figure 48*. Only the largest three prominences are considered from each data set; the average is calculated to give the total displacement. The standard deviation between the three repeats of the grasping task was used as the error. The peak onset may differ between repeats due to the participant not returning to exactly their previous starting position. This does not lead to frame reading errors as the peak identification function only starts at the onset of a peak rather than from a fixed point.

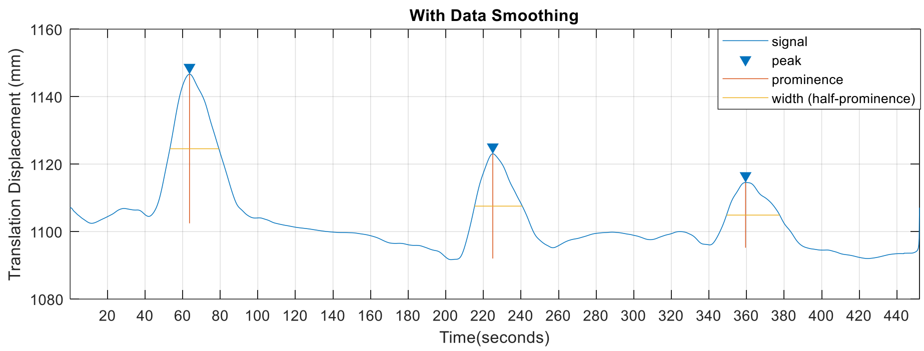

Figure 48. Total Translational Displacement of the Left Arm Sensor Peak Identification

The MOCAP data of the limb was used as the control, so that the motor strategy deviation resulting from the prosthetic device could be measured. To calculate the deviation between the control and prosthesis motor strategy, Equation *(20)* was used. This represents the difference in magnitude to complete the same grasping task and can be used as a measure of compensatory movements caused by the prosthesis.

$$Deviation(\%) = \frac{Control\ Total\ Displacement}{Prosthesis\ Total\ Displacement} \times 100 \qquad (20)$$

6.3 Results

6.3.1 Motion Capture Data

Due to the large amount of data yielded from MOCAP analysis, a subset of these data plots is shown below, *Figure 49* to *Figure 51*. These plots have been selected to show the common patterns that emerged during the testing procedure. The top half of each plot shows the control motor strategy and the bottom half shows the motor strategy influenced by the prosthetic device. The complete MOCAP data plots for all sensor positions and all grasping tasks can be found in *Appendix A*.

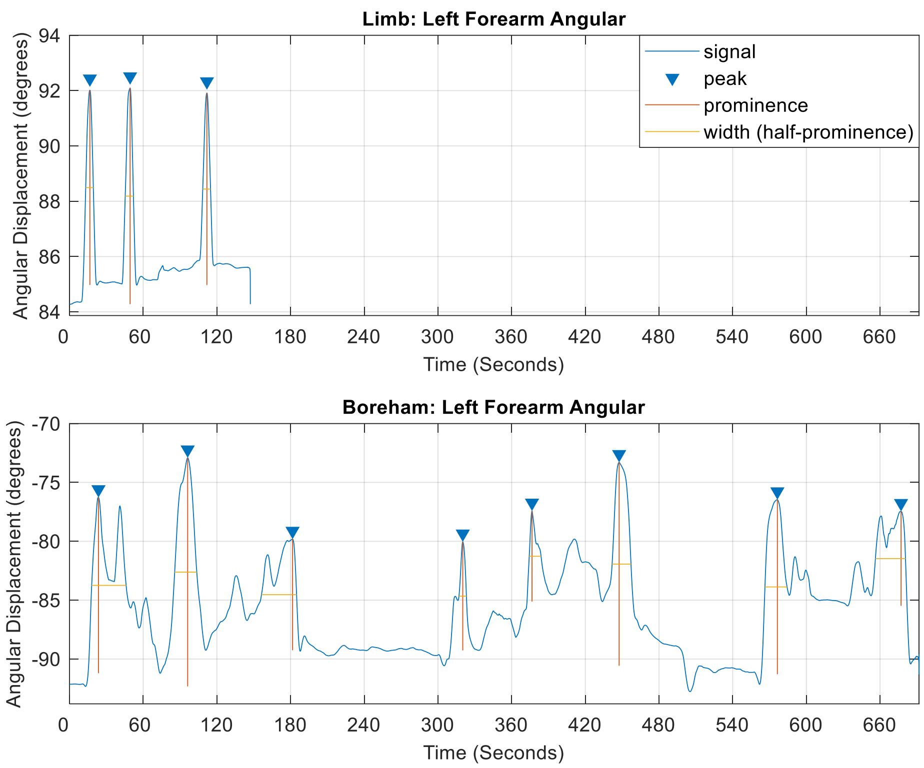

Figure 49. Forearm Angular Displacement when Grasping the Lateral Shape: Top) Control Motor Strategy, Bottom) CfAM-2 Hand Motor Strategy

It is immediately apparent from the forearm data for grasping the Lateral shape, *Figure 49*, that it took longer to complete the task using the CfAM-2 Hand than the control: <150 s with the biological hand compared to >660 s with the CfAM-2 Hand. The signal peaks were correspondingly larger in the CfAM-2 hand as well. The maximum angular displacement was larger, with approximately 20° with the CfAM-2 Hand and 6° with the biological hand. Angular displacement was highly controlled and repeatable with the biological hand, with three clear and distinct peaks. In contrast, with the CfAM-2 Hand, multiple peaks were observed which suggests several attempts were made to manoeuvre the device into better grasping positions. In the third repeat with the CfAM-2 Hand, there are only two large peaks, compared to the earlier two repeats which had three peaks. This suggests that the participant was learning how to better use the CfAM-2 Hand to grasp the Lateral shape.

The upper arm data for grasping the Tripod shape, as seen in *Figure 50,* was relatively similar between the control and the CfAM-2 Hand. Both showed three evenly spaced signal peaks with repeatable angular displacements peaks. The control had much narrower half-prominence widths and was more consistent compared to the CfAM-2 Hand. The angular displacement using the CfAM-2 Hand was roughly twice that of the control, 17° compared to 30°. The CfAM-2 Hand led to peaks with a plateau with a small decrease before returning to the baseline value. The first peak had the largest half-prominence width, whilst the latter two were roughly equal. This suggests that the participant had some difficultly grasping the Tripod shape on the first attempt, but was better able to perform the task for the two remaining repeats.

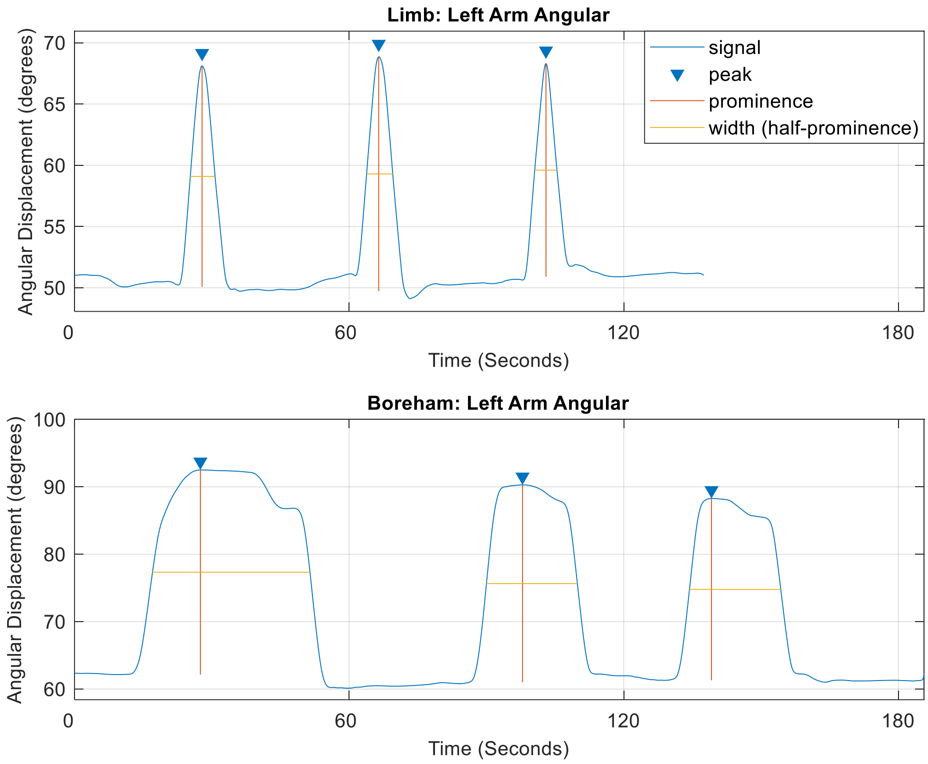

Figure 50. Upper Arm Angular Displacement when Grasping the Tripod Shape: Top) Control Motor Strategy, Bottom) CfAM-2 Hand Motor Strategy

As seen in *Figure 51,* three evenly spaced peaks were observed in the trunk data for grasping the Power shape. The control peaks were narrow and repeatable, whereas the peaks for the CfAM-2 Hand were broader and irregularly shaped, indicating that the movements were much less precise. In the second trial of the CfAM-2 Hand, there are two peaks of similar magnitude, suggesting that the Power shape slipped from the grasp and a second attempt was needed to complete the trial. The maximum angular displacement was very similar between the control and CfAM-2 Hand, with values of approximately 4° and 5° respectively. Whilst it took the CfAM-2 Hand longer to grasp the Power shape, significant compensatory movements of the trunk were not required.

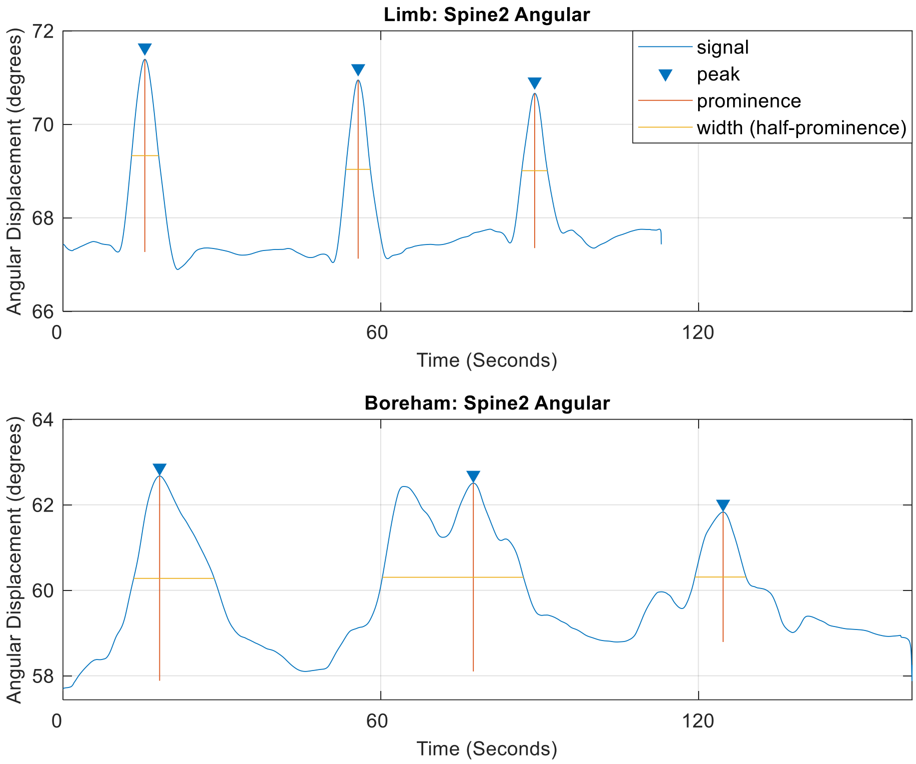

Figure 51. Trunk Angular Displacement when Grasping the Power Shape: Top) Control Motor Strategy, Bottom) CfAM-2 Hand Motor Strategy

In all of the MOCAP datasets, the control motor strategy had three distinct peaks with consistent prominences and half-prominence widths. The CfAM-2 Hand had broader peaks and larger prominences which indicate the grasping tasks took longer to perform and with greater movement. The broader peaks can be explained by the participant's motor strategy, which consisted of positioning the CfAM-2 Hand as close to the SHAP shape as possible, before taking their time to operate the grasping mechanism whilst maintaining that position. The CfAM-2 Hand motor strategy was also more inconsistent where multiple peaks within a single repeat were sometimes observed, which occurred either where multiple attempts were required to grasp the SHAP shape or several distinct movements were made to better position the CfAM-2 Hand for grasping.

6.3.2 Motor Strategy Deviation

The translational and rotational displacement deviations of the four MOCAP sensors are shown in *Figure 52 & Figure 53.* The deviations from using the prosthetic device are presented as percentages to show the altered motor strategy using *Equation (20),* outlined in *Section* 6.2.5.

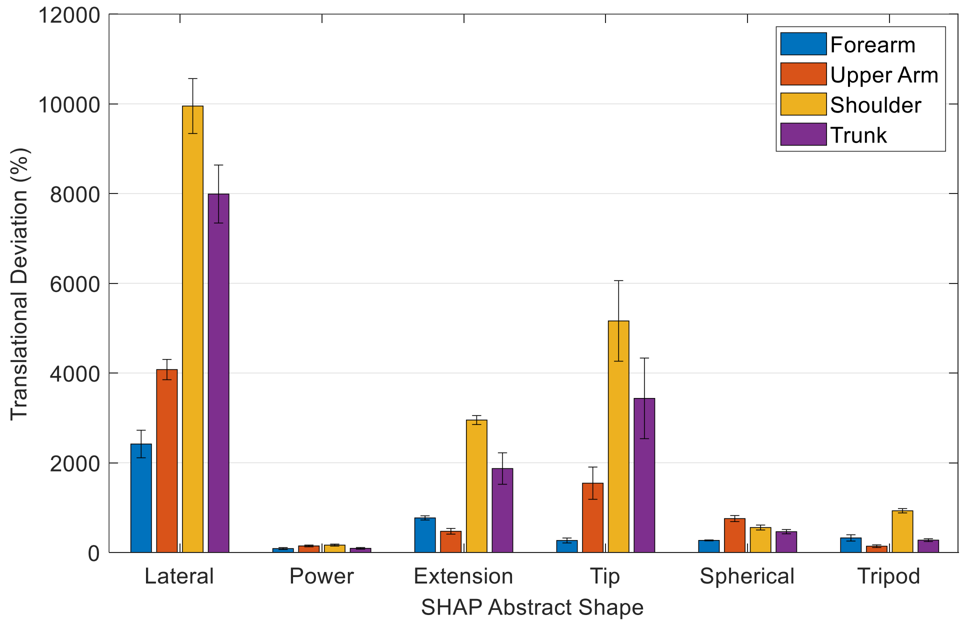

Figure 52. Motion Capture Sensor Translational Displacement

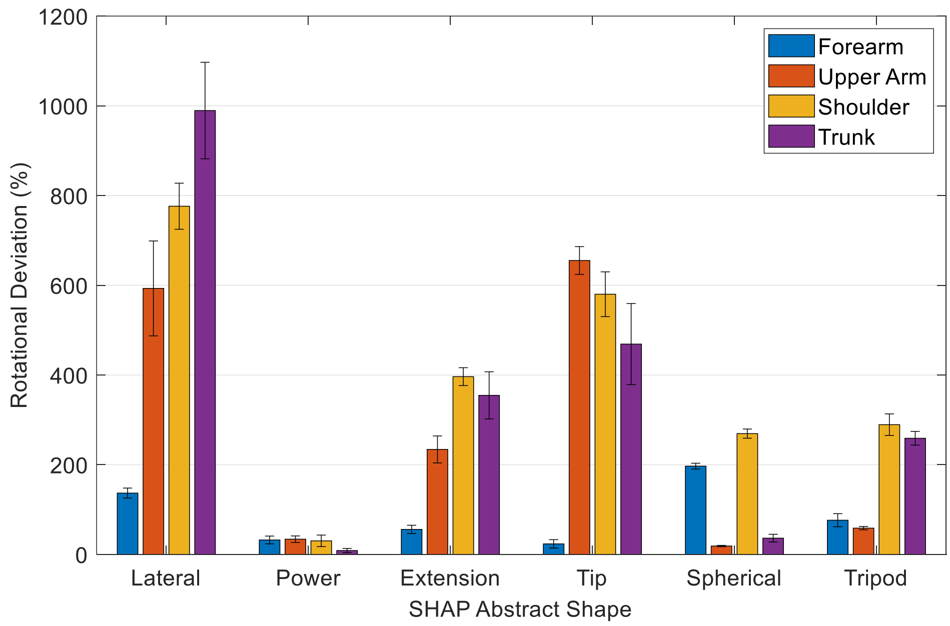

Figure 53. Motion Capture Sensor Displacement Rotational Displacement

Across all grasping tasks and sensor positions, there were significant translational and rotational deviations between the prosthesis motor strategy and the control motor strategy. This was observed to different extents for different grasping tasks and sensor positions. Translational deviation was typically a magnitude greater than the corresponding rotational deviation, since rotations at joints have a larger impact on positioning at a terminal point compared to solely translational movements. Generally, tasks with higher translational deviations also resulted in higher rotational deviations. The average deviation across the sensor locations for each of the grasping tasks is shown in *Figure 54* to show general trends.

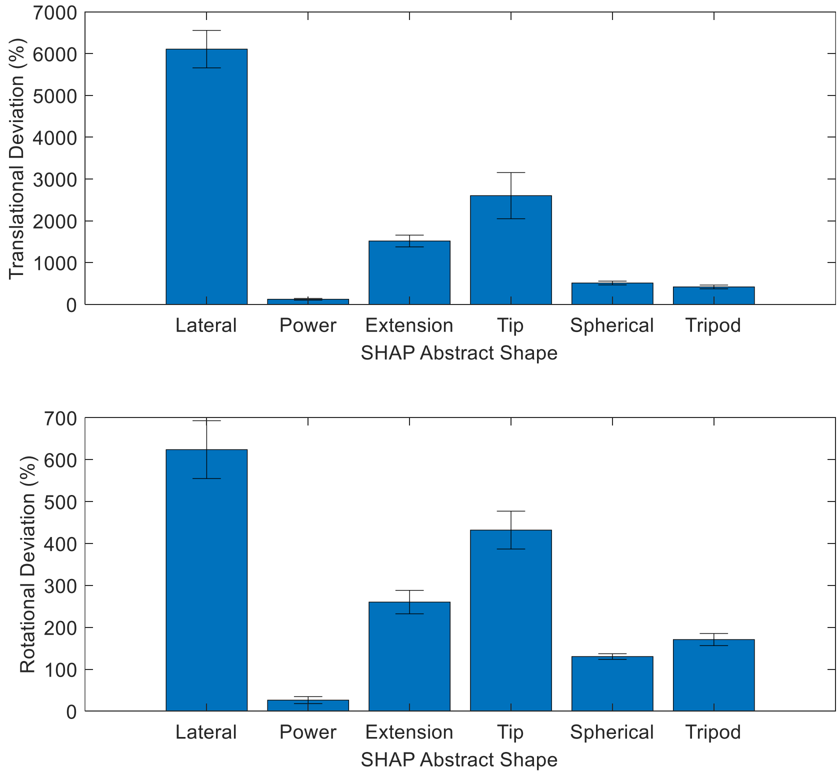

Figure 54. Average Deviation by SHAP Abstract Shape
Top) Translational Displacement, Bottom) Rotational Displacement

The task that resulted in the most motor strategy deviation, for both translational and rotational deviation, was the Lateral shape. The Lateral shape was designed to be grasped by its vertical tab, which required the participant to rotate their hand or prosthesis. Despite the CfAM-2 Hand being designed with a dedicated Lateral prehensile grasp, it was unable to provide the forearm pronation/supination required for this manoeuvre. The other DoFs in the upper body had to compensate, as illustrated in *Figure 55* in frames taken using the MOCAP suit's proprietary operating software. With the prosthesis motor strategy, it can be seen that the shoulder and upper arm abducts significantly higher than the control. With the Lateral shape directly in front of the participant, the shoulders and trunk had to effectively move backwards to create enough space to position the CfAM-2 Hand.

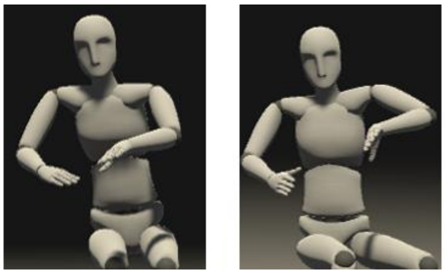

Figure 55. Motion Capture Snapshot Grasping the Lateral Shape: Left) Control Motor Strategy, Right) Prosthesis Motor Strategy

In contrast, the Power grasping task resulted in the least motor strategy deviation as it required the least positioning: the shape was relatively large which led to a predictable grasping motion and it was therefore easy to align the palm of the CfAM-2 Hand with the side of the shape. In addition, the Power prehensile grasp has a large grasping volume which allows more room for error during positioning.

The Extension and Tip grasping tasks both used the same Tripod prehensile grasp, but resulted in drastically different deviations. As seen in *Figure* 54, there was approximately a third more translational and rotational deviation grasping the Tip shape as compared to Extension shape. The Tip shape has half the height of the Extension shape which provides a smaller grasping area. This therefore required much more careful positioning and planning of the CfAM-2 Hand to achieve a successful grasp. Due to the lack of wrist flexion/extension, the participant had to raise their arms above the Tip shape moving their trunk, shoulder and upper arm. However, for the Extension grasping task, the participant approached the shape at an angle and used the fingers of the prosthesis to push it down into the thumb. Therefore, it resulted in fewer compensatory movements despite using the same prehensile grasp.

The Spherical grasping task had the second lowest translational and rotational deviation. However, the rotational deviations for the forearm and shoulder sensors were unexpectedly high, as seen in *Figure 53*. In order for the participant to grasp and move the Spherical shape, they had to rotate the CfAM-2 Hand to rest the shape on the palm of the device. The CfAM-2 Hand was not capable of forming a stable grasp on the Spherical shape with the digits, and instead the participant had to resort to a scoop-like strategy.

Lastly, the Tripod grasping task had relatively low translational and rotational deviation compared to the Extension and Tip grasping tasks, despite using the same prehensile grasp. The Tripod shape has three large faces which align well to the three digits of the Tripod prehensile grasp which aided positioning. The three digits oppose each other to form a relatively stable grasp. However, in order to get into this position, the participant had to make significant compensatory movements of the shoulder and trunk, as seen in *Figure 53*.

Overall, significant compensatory movements were observed for the Lateral, Extension, and Tip grasping tasks. These tasks required the use of prehensile grasps with small grasping volumes for relatively larger shapes. Therefore, this required careful positioning to successful grasp these shapes, thus resulting in higher translational and rotational deviation across all sensors. Deviation appeared to be minimalised where the ratio of the grasping volume and size of the shape was closer to one. The Power and Tripod grasping tasks used prehensile grasps specifically designed for their corresponding shape, which resulted in lower motor strategy deviation. When the same grasp was used for a different shape, positioning was more challenging and more compensatory movements were necessary.

6.3.3 Prosthesis Design Recommendations

The further away the MOCAP sensor was from the CfAM-2 Hand, the higher the motor strategy deviation was. The shoulder sensor had the most translation displacement deviation and the trunk sensor had the most rotational displacement deviation. This was expected based on previous compensatory studies[10,101,103] as the shoulder's and trunk's wide range of motion compensates for the loss of forearm pronation/supination and wrist flexion/extension. Due to the flexibility of the trunk, the participant used it to position their upper body to a favourable grasping location. Therefore, to reduce compensatory movements of the shoulder and trunk, future iterations of the CfAM-2 Hand should consider a rotatable wrist which replicates the forearm's pronation/supination. This is a common design feature in upper limb prostheses where the wrist is either rotated manually with the opposite limb or electrically controlled. This will allow the prosthesis user to position the CfAM-2 Hand without needing to lean back and raise their shoulder and upper arm to perform the Lateral prehensile grasp.

A less common feature which would also reduce compensatory movements is to replicate the extension and flexion of the wrist. Without this capability, the participant had to raise the forearm, upper arm, and shoulder higher than the control motor strategy to position the CfAM-2 hand directly above the grasping shapes. This was most significant for the Tip and Extension grasping tasks using the Tripod prehensile grasp. Positioning the CfAM-2 Hand was difficult for both of these tasks as the thumb digit had to be correctly aligned behind the shape first before the grasp could be performed. Due to the small grasping volume, there was little room for error. If the participant was able to flex the CfAM-2 Hand at the wrist, it would have been easier to align the shape along the length of the thumb digit rather than just the tip.

The Power prehensile grasp of the CfAM-2 Hand performed well for its intended function but struggled when used to grasp the Spherical shape. This is due to the positioning of the thumb relative to the other finger digits. During grasping, the participant found it difficult to reliably hold the Spherical shape in the palm of the CfAM-2 Hand. They found that the little and ring finger digits, when not carefully controlled, would push the Spherical shape out of the palm. Therefore, the participant had to resort to a scoop-like strategy to perform the task, which was not intended by design. Better control of the individual finger digits, or the ability to choose which digits the thumb could oppose to, would circumvent this issue.

In summary, while it was possible for the participant to complete the SHAP tasks using the CfAM-2 Hand, the performance could be improved with several design changes identified by the MOCAP analysis. This would not have been possible if the time-based tasks procedure had been used on its own.

6.4 Summary

Motion Capture analysis has been used to investigate the motor strategy of an able-bodied participant performing the abstract shapes subset of the Southampton Hand Assessment Procedure. A MOCAP suit was used to track the positions of their forearm, upper arm, shoulder, and trunk. A method of processing the high volume of data from MOCAP analysis was used, which allowed the deviation to be measured between a control and prosthesis motor strategy, as a measure of compensatory movements. This methodology allows upper limb prosthesis prototypes to be evaluated outside the clinical setting; direct testing with prosthesis users can require lengthy and expensive medical/ethical approval processes. Therefore, upper limb prosthesis prototypes can be screened for any major design flaws that lead to undesirable compensatory movement. This workflow allows more preliminary testing and can lead to a more efficient design cycle, which saves time for engineers, clinicians, and prosthesis users.

Using this method, it was found that the CfAM-2 hand motor strategies were slower than the control strategies and resulted in significant compensatory movements, where the trunk and shoulder made the greatest contributions to compensatory movements. Without forearm pronation/supination and wrist flexion/extension, the participant found it difficult to orientate the prosthetic device. This resulted in a disjointed movement, decoupling the reaching and grasping steps. Using these results, recommendations for future design iterations of the CfAM-2 Hand have been made to include wrist rotation and flexion/extension to allow better positioning. Upon redesign, the effect of these design features should see a reduction in compensatory movements.

Even when time-based testing is augmented with MOCAP analysis, this method still suffers from user bias. The compensatory movements of a prosthesis user who is skilled with their device and has gone through a training regime will be significantly lower than that of a new prosthesis user. Whilst physical evaluation is important to ensure that prosthesis users are operating their devices as intended, other forms of evaluation -as proposed in Chapter 4- are needed to solely assess the functionality of a prosthetic device so that it can be further improved without user bias.

Chapter 7
Discussion

7.1 Introduction

In this thesis, three quantitative tools have been proposed to aid in the design of upper limb prosthetic devices with the aim of reducing rejection rates.

Firstly, motion capture analysis was used to study the motor strategy of a participant, comparing performance with their biological hand against use of a prosthetic device. The positional deviation between the two motor strategies whilst performing a series of grasping tasks was used as a measure of compensatory movements. Secondly, simulated grasping within virtual environments was used to assess the grasping functionality of five prosthetic devices. A series of grasping tasks were replicated in a dynamic model and the steady state conditions of the final grasp was used to calculate four numeric grasp quality metrics. Thirdly, a multi-objective optimisation model was used to intelligently select the most optimal prosthetic device components based on user requirements. A modularity scheme was used to segment the prosthesis design to limit the design space and allow discrete utility values to be assigned to each component based on their mass, cost, and grasping functionality. An objective function could then be used to rank the prosthesis design combinations based on the overall utility.

In this chapter, the capabilities and limitations of the three design tools are discussed together with the extent that they fulfilled the objectives laid out in this thesis. Upon addressing the limitations of these design tools, a workflow which combines them all is proposed to produce high utility upper limb prostheses.

7.2 Simulated Grasping within Virtual Environments as a Design Tool

The second objective of this thesis was to develop a dynamic virtual environment to simulate grasping of upper limb prostheses and evaluate those grasps using numerical grasp quality metrics to drive design specifications.

To achieve this objective, a virtual environment was used to assess the grasping capabilities of five prosthetic devices with a range of degrees of freedom (DoFs). Conventional time-based tasks are not suitable for assessing the functionality of a prosthesis, as they primarily assess the user's skill at operating that prosthesis. Simulated grasping in a virtual environment removes the user bias and allows the use of quantitative metrics to assess the quality of grasping. A series of grasping and lifting tasks, based on a subset of the Southampton Hand Assessment Procedure[162], were performed in a dynamic model. Under steady state conditions the contact forces, contact torques, contact vectors, contact points, and centre of masses were used to calculate four commonly used grasp quality metrics. These grasp quality metrics were considered in parallel to provide an overall grasping functionality performance. The results of these simulations indicated that multiple DoFs designs have lower grasping performance than those with a single DoF, due to the difficulty of balancing multiple force vectors.

7.2.1 Feasibility

This investigation has shown that virtual evaluation provides an opportunity to gain useful design feedback without a requirement to produce physical functional prototypes with lengthy physical trials. Some manufacturers[15] quote the number of prehensile grasps and DoFs as a measure of prosthesis performance. However, the results from the simulations disagree with this as the single DoF prosthesis outperformed the anthropomorphic devices with multiple DoFs. A range of prehensile grasps can enhance grasping performance by planning finger and thumb digit motion paths that result in balanced force vectors for their given application. Mechanical design ill-considered for grasping was seen to result in poor performance, particularly in the case of the CfAM-2 hand. This prosthetic device was designed to achieve three prehensile grasps but used unfavourable grasping surfaces. These consisted of flat edges which led to unfavourable force vectors. The use of slightly concave surfaces would help to improve grasp stability by directing force vectors towards a single point. This flaw in grasping has been identified from the simulation and can now be used to influence future iterations of the CfAM-2 Hand.

Four grasp quality metrics were demonstrated in this study to showcase how the results from a prosthesis simulation could be used; however, it could be adapted to be used with other metrics if deemed appropriate. Whilst an equal weighting for each metric was used to calculate the grasping functionality, it can be varied with different user requirements to identify suitable devices. For example, open voluntary body powered hooks require their users to move their shoulder to operate them. Typically, the grasping force is generated from an elastic band that holds the hooks together, and more effort from the user is required to grasp larger objects. Therefore, a device with a higher PFJ value would be easier to operate and would be more beneficial to the user. In addition, the Objective Function can take grasping requirements into account. For example, a job may require the manipulation of power tools, so a stable Power grasp is desirable. This can be achieved with a high weighting for GII and SMW metrics.

Simulated grasping within virtual environments can enable a shorter design feedback loop which can allow engineers and clinicians to tailor their devices to better grasp specific objects and more quickly optimise their designs. Virtual environments would be used in addition to current standards of time-based testing, as physical testing is still important as it considers both reaching and grasping. However, before manufacturing a prototype, the grasping function of a prosthesis can be optimised using any numerical metric therefore enabling its use as a versatile design tool.

Overall, the objective of evaluating grasps via virtual environments was successfully achieved. A methodology was developed that allows a prosthetic device to simulate a grasping task and evaluate those grasps using numerical grasp quality metrics. This allows the grasping functionality of prosthetic devices to be assessed without the need for manufacturing and removes user bias.

7.2.2 Limitations

Although it was intended for virtual environments to decouple the grasping and reaching regimes to remove user bias, the initial position of the prosthesis in reality is determined by the reaching regime. The simulations carried out assume that the prosthetic devices were in an ideal position for grasping. However, controlling the position of a prosthesis accurately in practice can be difficult. Prosthesis users have to rely on visual feedback to position their devices and predict the motion path of the digits to grasp an object. To overcome this limitation, the use of MOCAP as discussed in *Section 7.4* could be used to determine the initial position for grasping, based on: the angle of approach, distance between the prosthetic device and the object, and the timing of the actuation systems.

In the simulations carried out, all of the bodies were assumed to be rigid bodies due to the low forces involved. When two rigid bodies make contact, there is a certain amount of friction which can be supported within the tangential plane of contact[165]. One of the reasons why grasping with biological hands is so successful is due to soft deformable finger tips[166,167]. This causes contact to occur over an area rather than at a single point which better distributes the contact forces[168,169]. In addition, as the contact forces increase there will be more deformation which will increase the contact area leading to greater stability. Deformable fingertips are also able to resist some moments in the contact tangential plane which would improve the SMW metric. Therefore, to replicate this, a common feature in upper limb prosthetic devices it to use rubberised digit tips, in addition to increasing the friction. Whilst the simulations did use frictional values of rubber, deformable parts were not modelled due to the high computational power required. Future iterations of using virtual environments to simulate grasping tasks should consider thin layers of deformable bodies at the end of digits to better emulate real prosthetic devices.

Due to how different this approach is to traditional methods of upper limb prosthesis evaluation, comparisons to existing studies are difficult. There is a need to perform physical validation studies to confirm that the grasping functionality values generated from the virtual environment correlate with better performance when used in practice. Whilst the quantitative grasp quality metrics selected for this analysis have been used in robotics with great success, the control systems between a robotic arm and a prosthetic device are drastically different. One method which could be used to validate the virtual environment results is to perform grasping and lifting tasks with the SHAP abstract shapes, and measure how much force can be lifted by the prosthetic device before grasp failure. This testing apparatus would include a series of wires and pulleys attached to a digital newton meter mounted underneath the SHAP abstract shapes. In theory, the prosthetic devices with a higher grasping functionality score, as determined through the virtual environment, should correlate with higher force values. The overall grasping functionality scores can be further interrogated looking at the individual metrics. In particular, the SMW metric (which is a measure of how inclined an object is to be destabilised by an external force) should have the greatest correlation to the physical tests proposed. Virtual environments as a test bed for upper limb prosthesis evaluation has the potential to greatly improve design workflow but does require physical validation studies in order to be adopted by the clinical community.

7.3 Multi-objective Optimisation as a Design Tool

The third and last objective of this thesis was to utilise an intelligent selection method to produce optimised upper limb prosthesis designs which combine both additive manufacturing and conventional engineering techniques.

To achieve this objective, a modularity scheme and a series of utility functions were used to quantitatively evaluate the value of an upper limb prosthesis. The modularity scheme splits prosthetic devices into three categories: socket, forearm, and terminal device. There were several component options in each category, manufactured either by conventional engineering (CE) or additive manufacturing (AM), and each component provided a different value of utility which was determined by its cost, mass, and grasping functionality. An overall objective function was used to control the proportion of these three utility contribution functions to give results that favoured a particular outcome. Several scenarios were run with different objective functions to demonstrate how this model could be used and providing a ranking of the best combinations of components accordingly. Using a literature-based objective function, it was found that a Split Hook terminal device with an additively manufactured (AM) socket and forearm was the optimal design which provided a low mass and high grasping functionality. The results of this model can be used to intelligently select the most appropriate components to address the needs of a given prosthesis user.

7.3.1 Feasibility

Although this model met the objective of selection of optimal designs combining AM and CE techniques, it should not replace the advice and recommendations of experienced medical professionals. It is useful as a comparative tool as it assigns quantitative values to prosthetic device components, however, it does not indicate the future development of prosthesis design. This model is only capable of ranking prosthesis component combinations within its modularity scheme. However, a digital tool to intelligently select prosthesis components based on user requirements does not exist and could be useful for non-medically trained staff.

The modularity scheme provides choices between AM and CE components. However, the model does not take full advantage of AM as a technology. Using AM, it is possible to manufacture a product as soon as it is required which could save on storage costs, as compared to having a warehouse stocked with CE products. AM allows for greater levels of personalisation such as sizing, shape, and aesthetics, which was not utilised in this model. An alternate model which better makes use of these advantages of AM is proposed in *Section 7.3.3*.

7.3.2 Limitations

The main limitation of the multi-objective optimisation model proposed in this thesis is the lack of trade-offs between the AM and CE components. The AM socket and forearm was cheaper to manufacture and lighter in mass compared to their CE counterparts. Therefore, in all scenarios the model would always pick AM parts since there is zero benefit to using a CE part. This was a result of assuming that the functionality of CE and AM were equal, which is not necessarily true. AM sockets are unable to provide the same level of suspension security as a moulded socket. Olsen et al.[157] performed a study with four participants with trans-radial limb absence and fitted them with several sockets made by Fused Filament Fabrication (an AM technique). Overall, the comfort of the AM sockets were well received but the suspension was found to be unsatisfactory. The majority of the participants preferred to use their original CE sockets over the AM sockets. Future iterations of the multi-objective optimisation model should look to giving functionality values to the prosthesis sockets based on satisfaction studies, which will ensure there is a trade-off between AM and CE components.

The prosthesis costs used for this model may not be accurate representations due to the inconsistent sources. For example, the CE socket cost is based on the labour and material cost to make a traditional laminated socket using a casting method[153], whereas the CE forearm cost is taken from the product catalogue of a major prosthesis company, Steeper Group[156] (Leeds, UK). All of the AM components are made through a third party manufacturer, 3DPrintUK[154], where their prices will fluctuate with demand. Therefore, a full cost analysis is required to find the manufacturing costs of prosthesis components so they do not include profit margins, overheads, and third parties as this will vary greatly with time and distributors.

In the model, it was assumed that the mechanical properties of AM and CE components were the same with similar life cycles. However, it is common for prosthesis users to return to their prosthetists for routine maintenance and repairs. On average, Chan et al.[170] found that the total cumulative 5 year prosthesis cost per upper limb amputee was found to be 53% spent in the first year. AM components do not have the same level of durability as some CE components, especially those which have been moulded, and will likely incur higher repair costs over its lifetime. Life cycle studies of AM upper limb prosthesis should be taken into consideration for future iterations of the multi-objective optimisation model.

7.3.3 Alternate Model

The model proposed in Chapter 6 is capable of ranking the best combinations of prosthetic device components for a given objective function. However, it does not consider workflow for manufacture and does not maximise the productivity of AM. The model could be expanded to consider the requirements of several prosthesis users and produce a bill of materials which match their requirements. This will output the required components and package them according to their manufacturing technique, which enables it to be more easily integrated into existing manufacturers.

This alternate model builds on an existing system called the 3D Packing Research Analysis Tool[171] (3DPackRAT), developed by the Automated Scheduling, Optimisation and Planning (ASAP) research group at the University of Nottingham. 3DPackRAT is a framework that can optimise the packing efficiency of the build volume of a SLS machine and optimise the scheduling of build volumes. The final Z-height of the build volume determines the cost of operating the machine. By increasing the packing efficiency, more models can be fitted into the build volume, therefore reducing operating costs. Unfused powder between finished models in the build volume can be reused to an extent. The powder has been subject to a heat cycle which causes some thermal degradation, and there will be a point where a powder can no longer be recycled. It is considered best practice to maximise the packing efficiency. However, the procedure of packing the build volume for SLS is usually done manually[17] and is highly dependent on the experience of the user. 3DPackRAT can pack the models within the build volume in different orientations and positions to find the optimal packing efficiency.

Using AM, it is possible to manufacture a product as soon as it is required which could save on storage costs. However, building a single component with SLS is not cost-effective, as the packing efficiency would be very low. To overcome this problem, 3DPackRat is also capable of scheduling build volumes to further improve the packing efficiency whilst meeting the required due dates. Consider a simple example, where Product A is due on Day 1 and Product B is due on Day 2. Instead of manufacturing both products on the day they are due, they are both manufactured on Day 1. As illustrated in *Figure 1* below, by manufacturing Product B early, the packing efficiency of the build volume is much higher, and therefore less powder is subject to a thermal cycle. This could also be applied for choosing to deliver a product late to maximise the packing efficiency, in cases where there is more value to be gained by saving costs then meeting the due date.

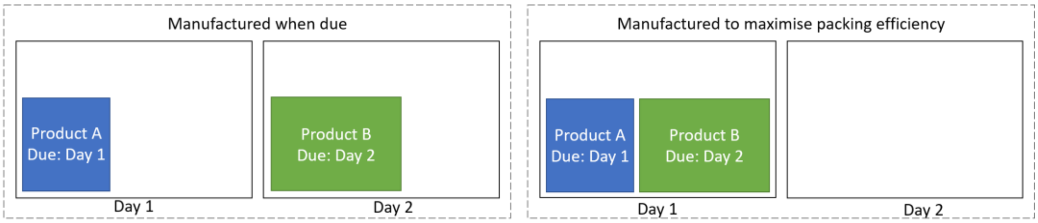

Figure 56. Comparison of build volume utilisation between two manufacturing strategies

There can be greater benefits from performing joint optimisation instead of performing packing efficiency and scheduling optimisation in sequence. These aspects are not independent and have underlying trade-offs between them. 3DPackRAT is capable of balancing these two objectives to provide an improved solution. This model aims to expand 3DPackRAT to include design optimisation to create a single tool that takes advantage of the benefits of AM: geometric freedom, on-demand manufacturing, and no additional tooling costs for bespoke designs.

The costing for AM parts is now more complex: rather than relying on quotes from third party manufacturers, the cost can be determined by the utilisation of the build volume. The new cost contribution utility equation, as shown in *Equation (21)*, calculates the true cost of AM components and CE components. The first half calculates the cost of the build volume that is utilised by AM components. In SLS, all parts have a bounding volume: a six-sided geometric volume that envelops the more complex components. These bounding volumes are required to make sure there are no intersections with other components whilst packing, which therefore prevents any unintentional fusing. The costs of the AM components are dependent on the packing efficiency in their build volumes. The second half of the cost equation calculates the cost of parts which have been manufactured by CE, which all have fixed costs.

$$Cost(P) = \sum\nolimits_{p \in P} c(b_t) \times \frac{v(p)}{v_{utilised}(b_t)} + \sum\nolimits_{p_{CE} \in P} c(p_{CE}) \tag{21}$$

where ...

$c(b_t)$ *gives the cost of the build volume*
$v(p)$ is the bounding volume of the part p.
$v_{utilised}(b_t)$ gives the utilised (bounding) volume of the build volume at time t
$P = \{socket, forearm, palm, digits\}$
$socket \in \{S_{AM}, S_{CE}\}$
$forearm \in \{FA_{AM}, FA_{CE}\}$
$palm \in \{P_{AM}, P_{CE}\}$
$digits \in \{D_{AM}, D_{CE}\}$

AM as a manufacturing process has not been well adopted by existing prosthetic device companies. There are no commercial designs which utilise both CE and AM components; instead, there are either designs made entirely by CE or AM. This alternate model leverages the advantages of SLS printing by utilising multiple build volumes and provides the opportunity to optimise scheduling to reduce overall costs. Not only can it be used to select the most optimal prosthesis components according to an objective function, it can also be used by manufacturers to produce a bill of materials, including the optimised build volume. This thereby provides an easy integration option for existing prosthesis manufacturers to expand in AM whilst keeping their CE products.

7.4 Motion Capture Analysis as a Design Tool

The first objective outlined in this thesis was to assess the motor strategy of an upper limb prosthesis user using motion capture analysis to gain an understanding of how prosthesis design affects compensatory movements.

To achieve this objective, the motor strategy of a participant was analysed using a motion capture suit. The participant performed a series of time-based tasks, based on the Southampton Hand Assessment Procedure, with and without a prosthetic device (CfAM-2 Hand). Conventional methods of assessing functionality rely on the time taken to complete these time-based tasks. However, these do not take into account compensatory movements which can result in overuse injuries and place the body in unnatural positions, leading to higher device rejection rates. A motion capture suit was used to track positional data along the forearm, upper arm, shoulder, and trunk to monitor the participant's motor strategy. The percentage difference between the participant's motor strategy with and without the CfAM-2 Hand was used as a measure of compensatory movements. In all tasks, the motor strategy with a prosthesis was slower and showed significant deviation from the control. The trunk and shoulder made the greatest contributions to compensatory movements. The participant was limited by the CfAM-2 Hand's lack of forearm pronation/supination and wrist flexion/extension, which made it difficult to orientate. These results highlighted the limitations of the CfAM-2 Hand design and can be used to inform future design.

7.4.1 Feasibility

MOCAP analysis of time-based tasks has been demonstrated to produce a quantitative value of compensatory movements and therefore provide a better understanding of how prostheses are operated. Time-based tasks such as the Box and Block[85], Nine Pegs[86] and SHAP[95] used in isolation are unable to provide information on how a device could be improved as it is not clear from those test results what beneficial design changes can be made. By using MOCAP, a particular compensatory motion can be identified and future design iterations can be optimised to minimise that motion. This quantitative measure of performance can be used in a feedback design loop to design an upper-limb prosthesis that aims to reduce compensatory movements, instead of solely relying on the users to take additional training regimes. It is important to minimise compensatory movements as they can result in overuse injuries and lead to device rejection[101].

The methodology of using the positional deviation between a biological hand and a prosthetic device as a measure of compensatory movement is novel. Such a method allows a more complete view of motor strategy to be obtained, as currently there are no evaluation methods which use compensatory movements as a measure of performance. The use of a motion capture suit and an able-bodied adapter allows the same procedure to be easily reproduced by other researchers, allowing it to be incorporated into design workflows. As the method proposed only requires able-bodied users, this allows prototypes to be quickly assessed in a workshop setting, without the need for medical grade sockets or prosthesis users. Therefore, this provides the opportunity to perform initial testing for compensatory movements which could identify any major flaws; thus, allowing a better product to reach physical assessments with upper limb prosthesis users. In addition, the MOCAP data could be used to anticipate suitable training regimes as the clinicians can be informed of what compensatory movements to expect. This may also guide selection of prosthesis devices which will not exacerbate existing conditions.

Overall, the objective of assessing the motor strategy of operating a prosthetic device was achieved. However, the data captured in this thesis is only able to comment on a single design and therefore only a rudimentary understanding of how prosthesis design affects compensatory movements was achieved. Without testing a wider range of prosthetic devices, the effects of design features on compensatory movements cannot be extrapolated to other devices. Nonetheless, it was shown that the MOCAP data can be used to gain a better understanding of how a prosthesis user operates their device.

7.4.2 Limitations

The results of the MOCAP suit presented in this thesis are limited as it only had a single able-bodied participant to operate a single design. Firstly, an able-bodied adapter (ABA) was used to allow a participant to operate a prosthetic device on their non-dominant arm. Whilst studies have indicated that the learning curve of operating a prosthesis with an able-bodied adapter were similar to an actual prosthesis user[160], the results with an able-bodied adapter can be exaggerated due to its size. Significant trunk movement was observed in the participant who moved backwards in order to accommodate the length of the ABA. The bulkiness of the ABA obscured the vision of the participant making it difficult to position accurately, especially with the smaller SHAP abstract shapes (i.e. Tripod, Tip, Extension). The prosthetic device protruded from the end of the participant arm which made it unwieldy to move, and due to the combined weight, it also caused significant fatigue. These impediments are also expected in upper limb prosthesis users albeit to a lesser extent. Therefore, further studies are required to investigate the effects of operating a prosthetic device with an ABA compared to a prosthesis user.

Secondly, future studies are required with a larger sample size to gain statistically significant data in order to comment more on the actual design. There was also no opportunity to compare how different prosthesis design features affected the motor strategy of the user, which would be one of the strengths of using MOCAP as an evaluation tool. The test protocol with the MOCAP suit only consisted of the abstract shape subset of the SHAP, where only simple reach and grasp tasks were assessed. The Activities of Daily Living (ADLs) in the latter half of the SHAP require more complex movements and are much more relevant to prosthesis users. It would be beneficial to capture the motor strategy of multiple participants performing the complete SHAP to build a library of compensatory movements. Not only can these results be used to inform future design iterations by engineers and designers, it can also be used by clinicians and prosthetists to prepare training regimes tailored to specific prosthetic devices.

The final limitation of the MOCAP analysis method used is the resolution of the inertial based system. The Rokoko Smart Suit Pro[173] has a comparatively poor resolution of 1° as compared to other leading motion capture products, such as the Perception Neuron 3 developed by Neuron Mocap[174] (Florida, USA), which has a resolution of 0.02°. With the poorer resolution of the Rokoko Smart Suit Pro, there is a risk of error propagation due to how the spatial position of the inertial sensors are calculated: the positions of the sensors are calculated relative to the previous sensor leading back to the base unit on the lower back of the MOCAP suit. Therefore, there are larger error margins the further the sensors are from the base unit. To overcome this limitation, either a higher resolution motion suit or a marker-based videogrammetric method could be used. However, both these solutions are significantly more expensive than the method used. Since the aim was to identify trends in the difference between able-bodied and prosthesis motion strategy, the resolution of the Rokoko suit is acceptable for this task, although higher resolution equipment could be used in future work.

7.5 Future Design Workflow

By addressing the limitations of the three design tools, as discussed above, they can potentially be used together to produce high utility upper limb prostheses. In order to achieve this, an updated multi-objective optimisation model is required which takes further variables into account. Instead of just considering grasping functionality, the model should incorporate functionality values of the socket and forearm components. Designing AM sockets relies on the use of 3D scanners instead of conventional casting techniques, which can result in a socket which feels less secure. Satisfaction studies should be performed to examine the impact of AM components and provide comparisons to their CE counterparts. Mechanical testing will also be required to compare the durability of prosthesis components and should be paired with life cycle studies to investigate the true costs of owning a prosthesis over a length of time. The prosthesis component costs in *Table 5* for the model described in *Chapter 5* uses an inconsistent range of sources. A cost analysis is required to find the true manufacturing costs of prosthesis components which do not include profit margins, overheads, and third parties as this will vary greatly between distributors. With these changes, the multi-objective optimisation model can be used to compile a list of ideal prosthesis components for a given set of customers with different requirements.

The grasping functionality values from the simulated grasping tasks within a virtual environment can be improved by using the results of MOCAP analysis. The initial grasping position will be more realistic if it was based on actual positioning data based on prosthesis users, rather than being manually positioned in the virtual environment. A library of motor strategies should be built using multiple prosthesis designs with multiple participants. This library can then be interrogated to match the design capabilities of prosthetic devices to compensatory movements. For new prosthetic devices, the similarities in mechanics can be cross referenced with those in the library to approximate the initial grasping position. Taking into account the range of motion of the shoulder and elbow joints, there is a finite number of combinations the upper arm and forearm can take as the prosthesis position is fixed. Therefore, it is possible to predict the possible compensatory movements caused by a prosthetic device.

This workflow is summarised in *Figure 57*. To allow the workflow described above to work as effectively as possible, close collaboration with prosthesis manufacturers, prosthesis users, clinicians, prosthetists, and designers are required. This workflow provides optimised prosthetic devices which are catered for their respective users. This personalisation can come in the form of size, shape, function, and aesthetic. Achieving this requires the collection of data which are the quantified results of satisfaction studies, cost analysis and mechanical testing. These are all needed in order to drive the multi-objective optimisation model, to achieve high utility upper limb prostheses and reduce rejection rates.

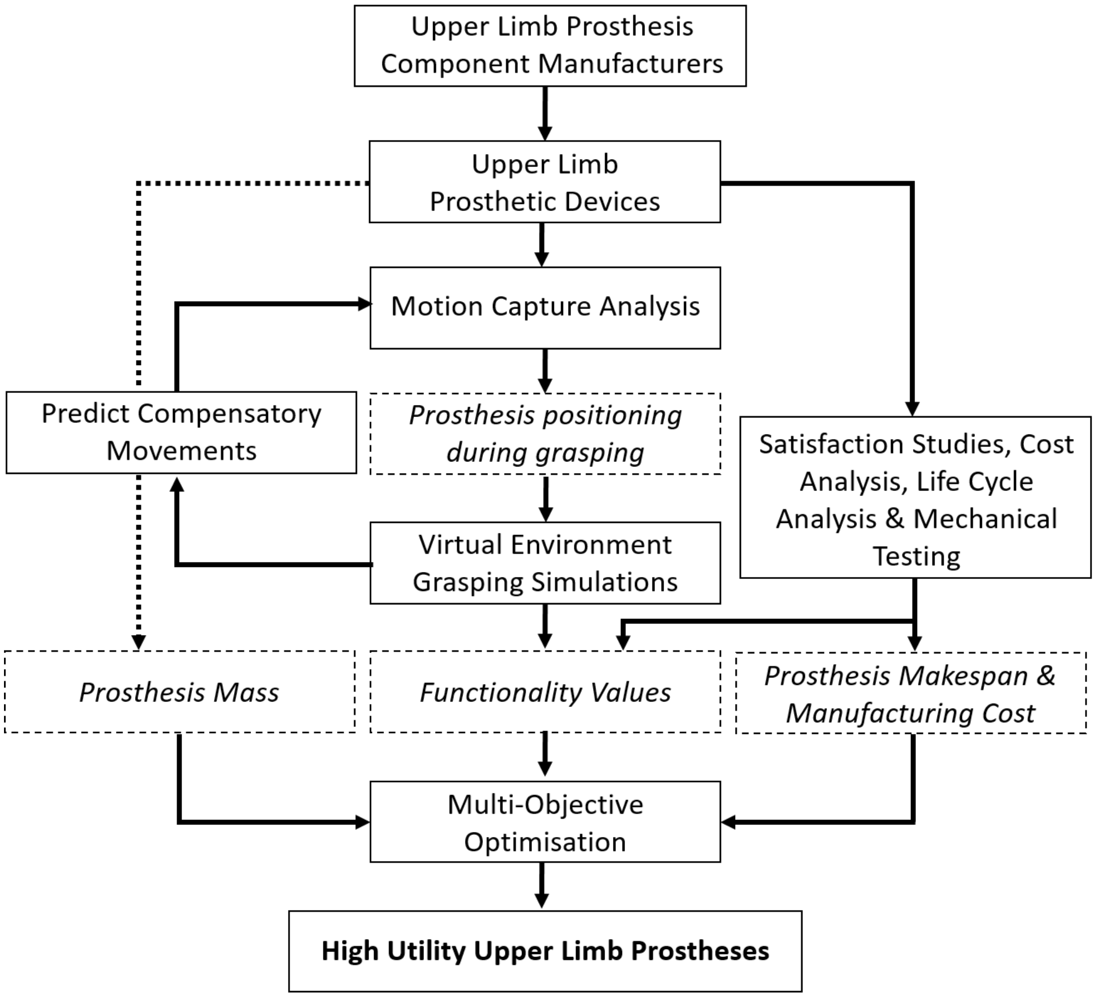

Figure 57. Proposed flow chart combining motion capture analysis, virtual environments, and multi-objective optimisation to produce high utility upper limb prostheses

Chapter 8
Conclusions and Future Work

8.1 Conclusions

Upper limb prostheses rejection rates are high for various reasons which include discomfort, high mass, and low functionality. A prosthetic device that is uncomfortable, painful to wear for extended periods of time, and a hindrance will likely be abandoned. This has been attributed to the limitations of the current design process. Despite the variety of prosthetic devices available, there are high rejection rates. A common theme across modern devices is over-engineering, where the prosthetic devices do not meet user requirements well. The design for functionality is restrained by current evaluation methods which do not accurately represent the ability of the device. In this thesis, three quantitative design tools were proposed to improve upper limb prosthesis functional design.

The first tool was used to assess the motor strategy of an upper limb prosthesis user using motion capture (MOCAP) analysis to understand how prosthesis design affects compensatory movements. A MOCAP suit was used to track positions of their forearm, upper arm, shoulder, and trunk during grasping tasks with their biological hand and a prosthetic device via an abled-body adapter. Compared to the control strategy, the prosthetic motor strategies were slower and resulted in significant compensatory movements, especially with the trunk and shoulder. The conclusions that can be made are:

- Reducing compensatory movements with the design of the prosthetic device is not currently addressed in the literature and is potentially a missed opportunity.
- A method of processing high-volume MOCAP data was proposed and the deviation between motor strategies was used as a measure of compensatory movements.
- Prosthetic devices should consider a form of wrist control in order to replicate forearm pronation/supination and wrist flexion/extension for positioning.
- Studies of motor strategy using MOCAP analysis can be used to identify prosthesis design features that correlate with compensatory movements.

The second tool was used to simulate grasping of upper limb prostheses and evaluate those grasps using numerical grasp quality metrics. Five prosthetic devices with different degrees of freedom (DoFs) were assessed within a dynamic virtual environment which removed any user bias. A series of grasping and lifting tasks, based on a subset of the Southampton Hand Assessment Procedure[162], were carried out. At the end of the simulation, under steady state conditions, the contact forces, contact torques, contact vectors, contact points, and centre of masses were used to calculate four commonly used grasp quality metrics. These grasp quality metrics were considered in parallel to provide an overall grasping functionality performance. The conclusions and outcomes of this tool are:

- A novel method of using a virtual environment to replicate time-based tasks for the purposes of prosthetic device evaluation.
- Numeric quality metrics, which are directly related to grasping performance variables, were obtained.
- Designs with multiple DoFs designs were observed to have lower grasping performance than those with a single DoF, due to the difficulty of balancing multiple force vectors.
- This tool can allow designers to assess their designs without solely relying on physical testing procedures.

The third design tool was used to intelligently select optimal prosthetic device components for a given set of requirements. Prosthesis users desire their devices to have low mass, low cost, and have high functionality. However, these are conflicting design objectives which need to be balanced in order to meet the needs of a prosthesis user. Several scenarios were run with different objective functions to demonstrate how this tool could be used and providing a ranking of the best combinations of components accordingly. The conclusions that can be made from this tool are:

- A novel modularity scheme was used which allowed components to be assigned unique values of utility based on cost, mass, and grasping functionality.
- It was possible to rank combinations of components based on user requirements.
- Using a literature-based objective function, it was found that a Split Hook terminal device with an additively manufactured socket and forearm was the optimal design as it provided a low mass and high grasping functionality.

8.2 Implications for the Field of Upper Limb Prosthetic Devices

The three quantitative design tools proposed in this thesis can be used together in a single workflow to address the high rejection rates of upper limb prosthetic devices and integrate into existing prosthesis manufacturers. MOCAP analysis can be used to better inform the positioning required for simulated grasping in a virtual environment. Upon building a library of motor strategies of multiple prosthesis users performing tasks, it could be used to match the design capabilities of prosthetic devices to particular compensatory movements. For new prosthetic devices, the similarities in mechanics can be cross referenced with those in the library to approximate the initial grasping position. This allow the virtual environment to be used a tool to predict the compensatory movements by considering the position of the prosthetic device and the range of motion of the shoulder and elbow joints. The results of the simulated grasping tasks combined with further user satisfaction studies, cost analysis studies, mechanical testing, and life cycle studies of upper limb prosthetic devices manufactured by both AM and CE can be used in the multi-objective optimisation model. By developing these tools there is greater opportunity to optimise prostheses using quantitative results which can result in greater utility for the prosthesis user, and therefore improve user satisfaction and reduce device rejection rates.

8.3 Recommendation for Future Work

8.3.1 Virtual Environments for Simulated Grasping

The simulated grasping tasks performed within the virtual environments were manually positioned to maximise the contact areas of the digits during grasping. However, this is not a realistic representation of grasping by prosthesis users. Therefore, to overcome this limitation, virtual environment testing should be paired with MOCAP analysis to determine the initial position of the simulation. However, this will require a certain number of prosthetic devices to be studied with MOCAP analysis before the virtual environment can be consistently used to study grasping functionality. Once the motor strategy library has been compiled, new prosthesis designs can then be evaluated within the virtual environment by cross referencing similar upper limb prosthesis designs.

The virtual environment can be expanded to not only assess the grasping functionality of prosthetic devices but could also be used to predict compensatory movements. The similarities in mechanics of new prosthetic devices can be compared to devices within the motor strategy library to approximate the initial grasping position. In this position, there is a finite number of configurations that the forearm and upper arm can have due to the limited range of motions of the elbow and shoulder joints. Once this process has been refined, it can be used to predict compensatory movements from the design stage and not rely on physical testing.

8.3.2 Multi-Objective Optimisation Model

The model presented in this thesis was simplified to only consider the prosthetic device costs upon manufacturing with no further modifications or repairs. However, a significant portion of costs occur after the initial prosthetic device fitting. To improve the multi-objective optimisation model, life cycle and cost analysis studies of conventionality engineered (CE) and additively manufactured (AM) components should be undertaken. The results of this studies will provide accurate costing statistics and collaborations with prosthesis component manufacturers would be highly beneficial.

The functionality of AM sockets and forearms were assumed to be the same to simplify the model. This resulted in the model always recommending AM sockets and forearms, which may not best meet user needs. Satisfaction studies should be taken to investigate the differences between the AM and CE components, and used to provide a quantitative functionality value. This will ensure that there are trade-offs between using AM and CE components.

8.3.3 Motion Capture Analysis

The main limitation of the motion capture (MOCAP) analysis carried out in this thesis was the lack of participants to create a statistically sound set of results. With only one participant operating a single prosthetic device, anomalies with their motor strategy could not be identified. Using the same methodology outlined in *Section 6.2*, future work should look to sample a greater number of participants with a wider selection of prosthetic devices. The devices in *Figure 26* found in *Section 4.2.1* would be suitable as they provide a range of degrees of freedom. In addition, the grasping tasks should be expanded to include the Activities of Daily Living from the Southampton Hand Assessment Procedure to provide a better picture of the motor strategy of prosthesis users in a home setting. By testing several designs with several participants, a library of motor strategies can be built with two key applications. The first application would be for engineers to compare the mechanics of their designs to existing designs in the motor strategy library. This will allow engineers to visualise the expected compensatory movements and make the appropriate changes if required. The second application would be for clinicians and prosthetists to better plan and structure training regimes for prosthesis users. If they can visualise the expected motor strategy and take the necessary precautions before their patients even operate the prosthetic device, it could speed up the rehabilitation process.

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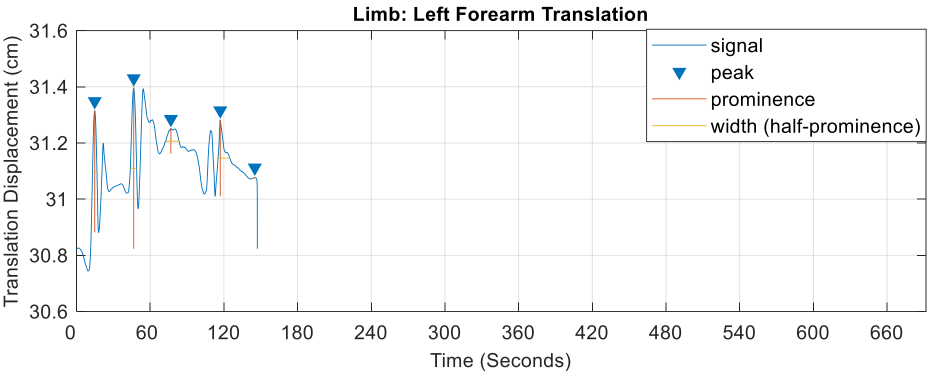
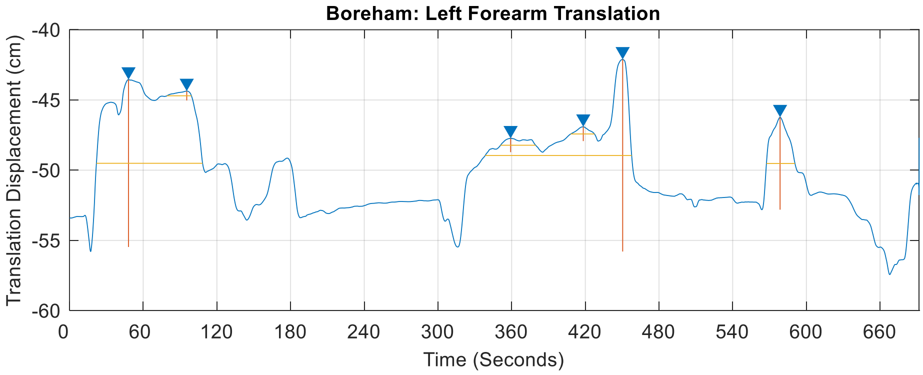
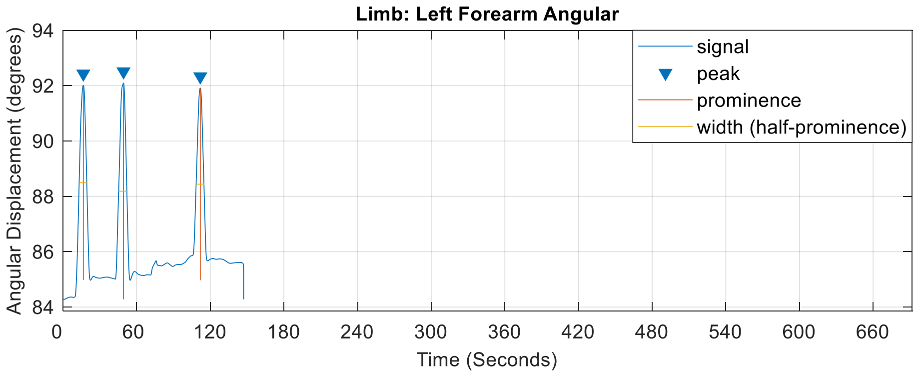
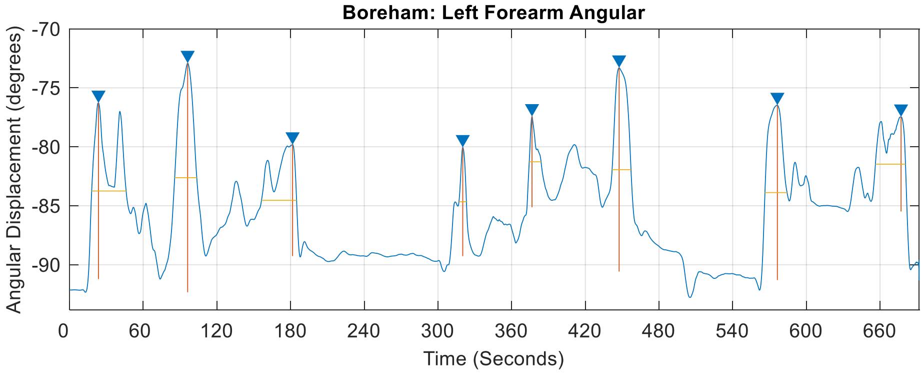
Appendix A

Lateral MOCAP Forearm

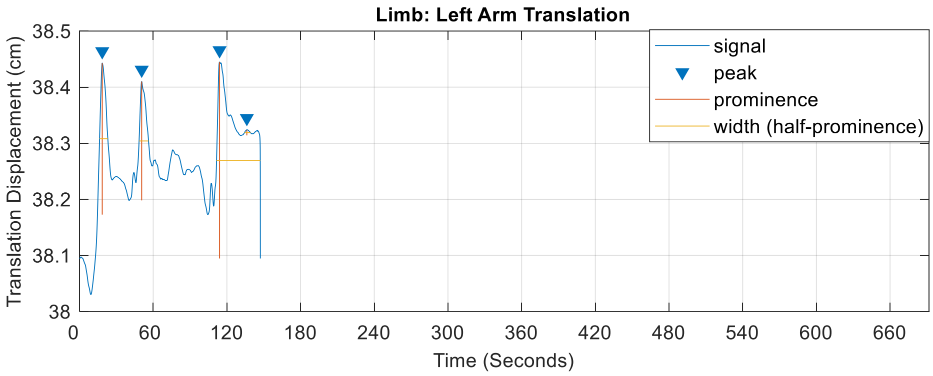


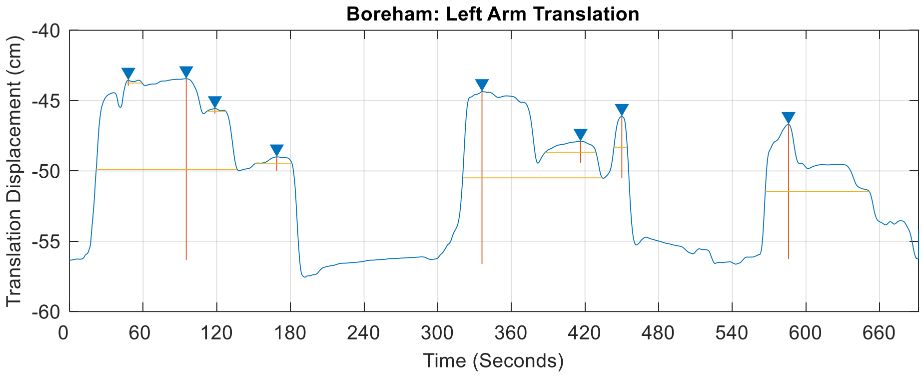


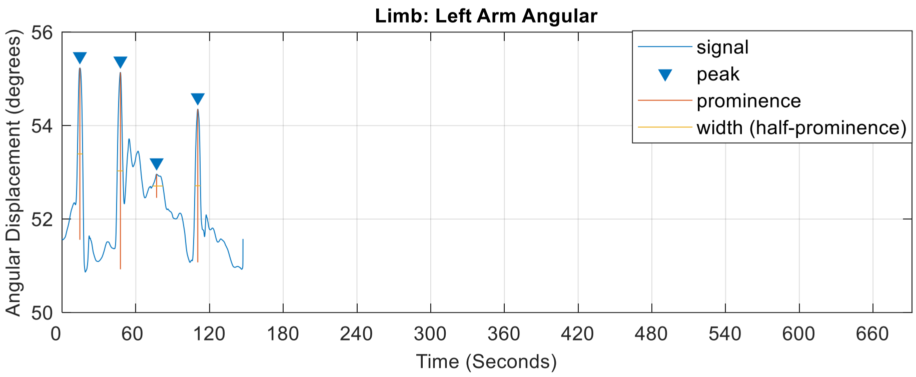


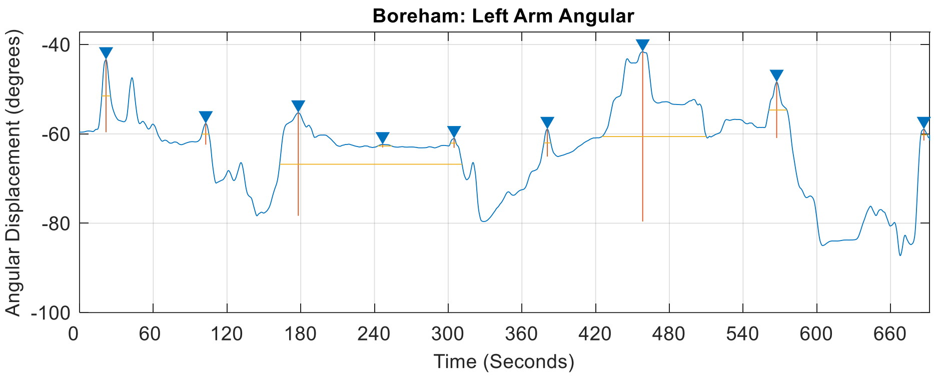


Lateral MOCAP Upper Arm









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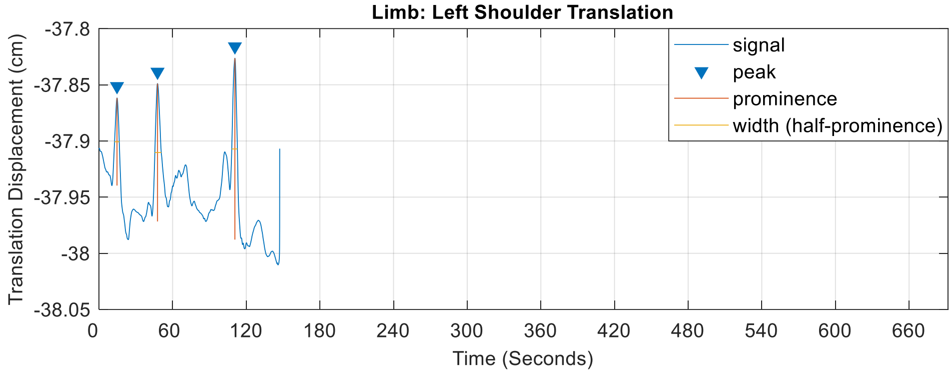
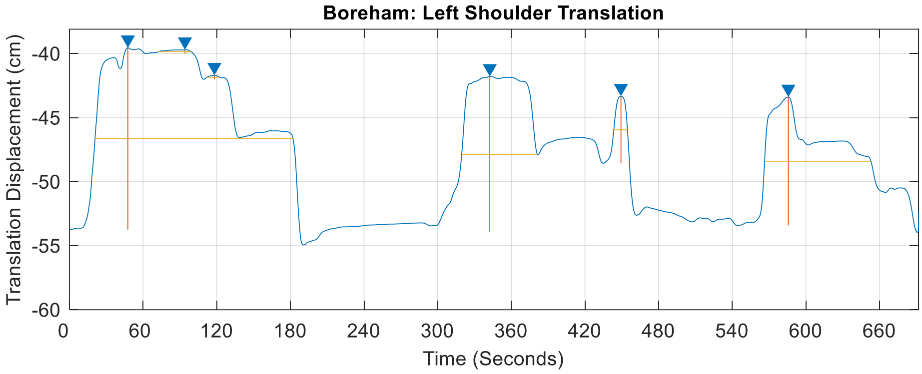
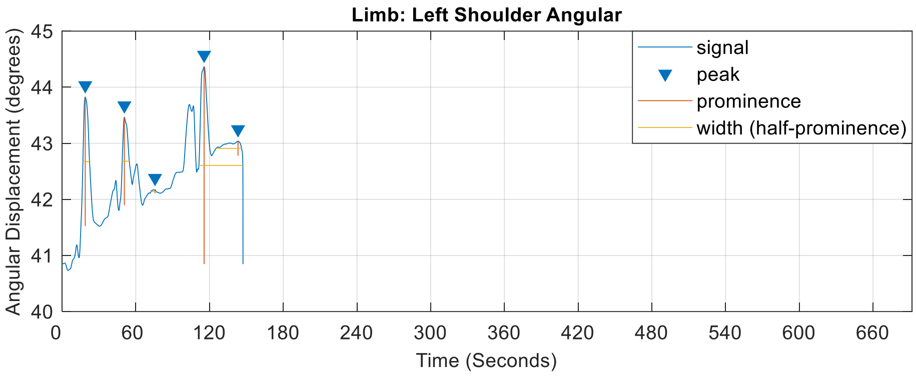
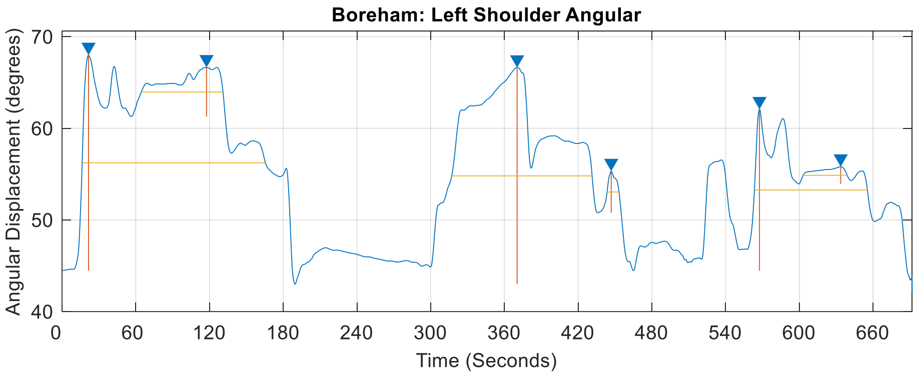
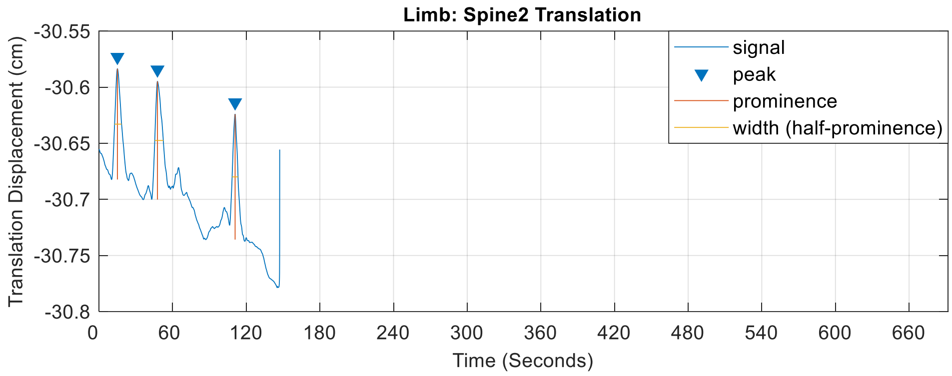


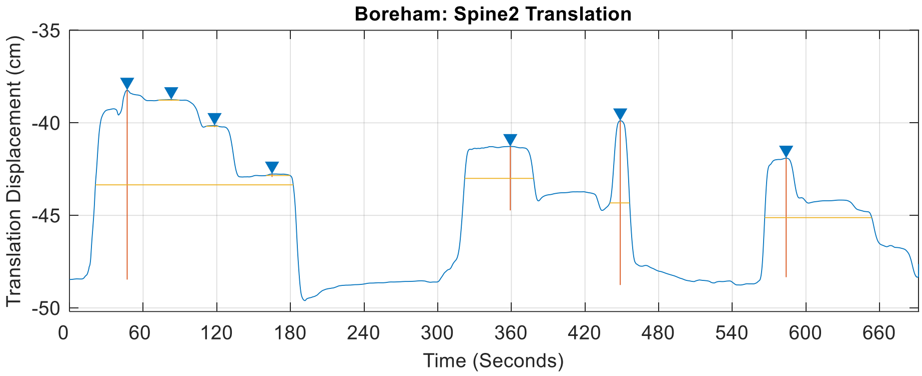


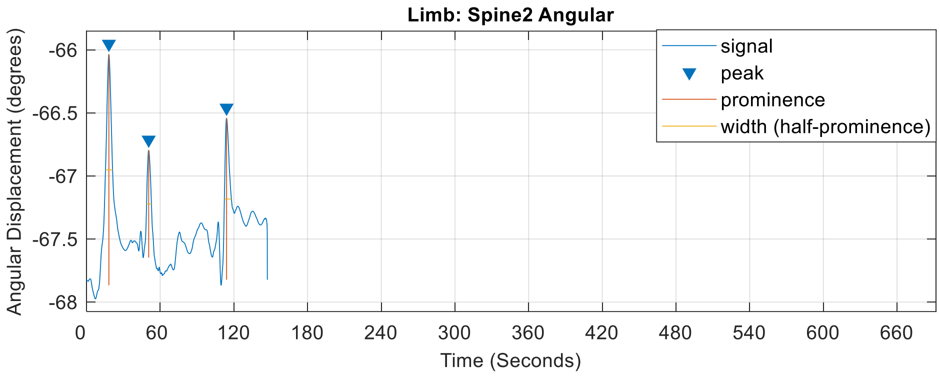


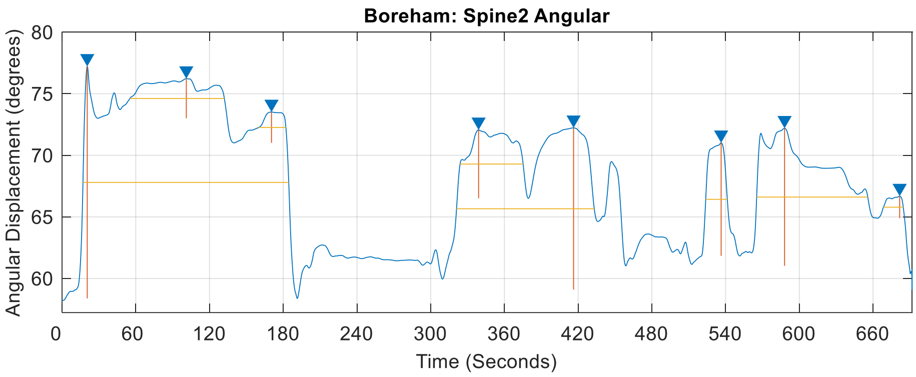


Lateral MOCAP Trunk

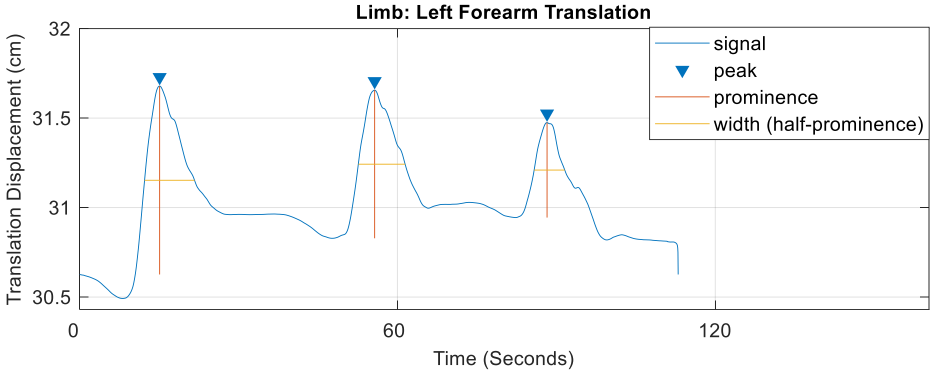


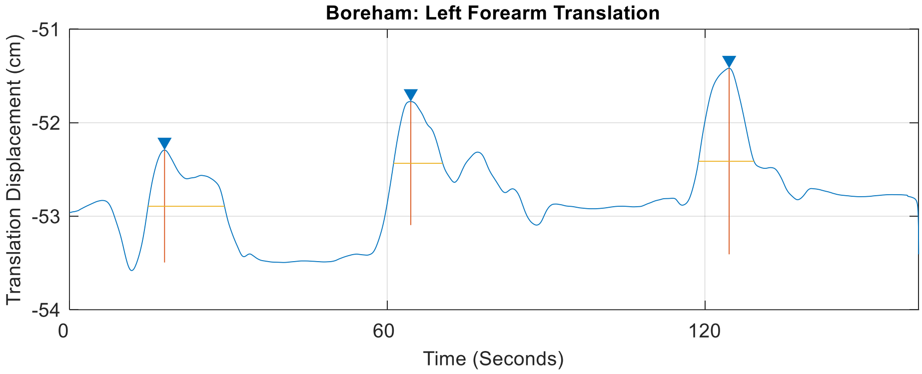


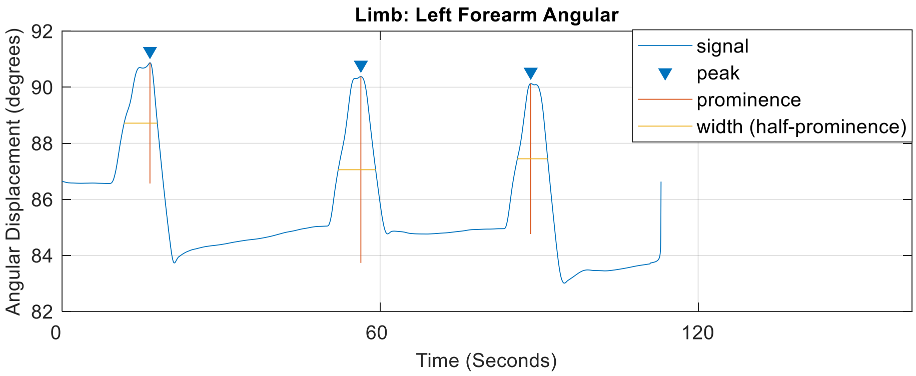


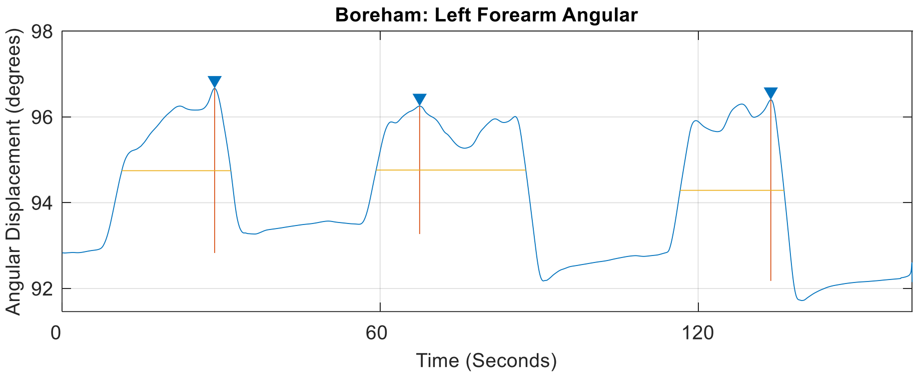


Power MOCAP Forearm









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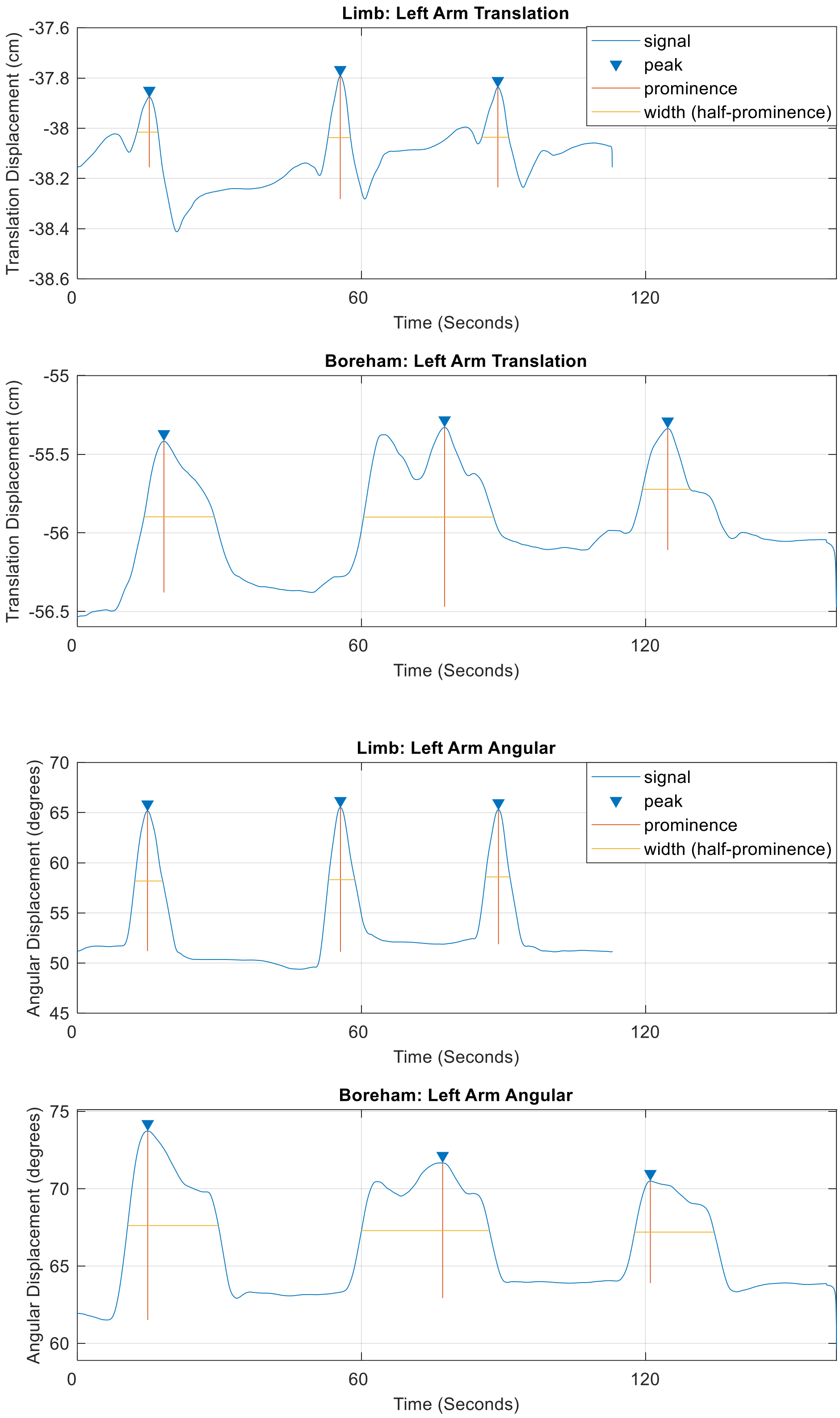

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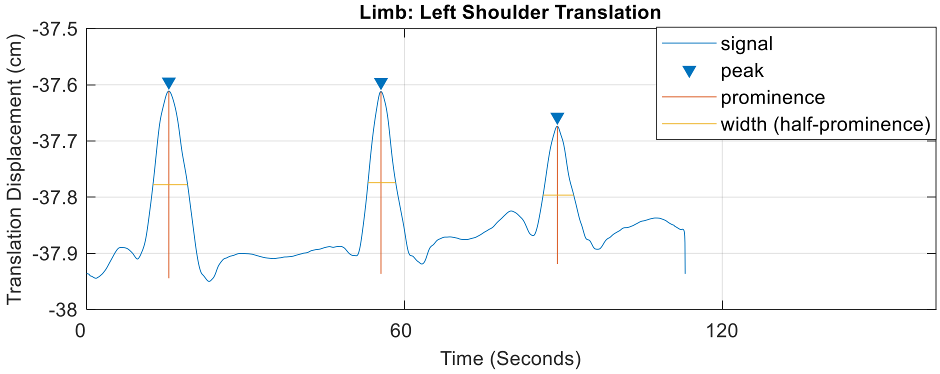

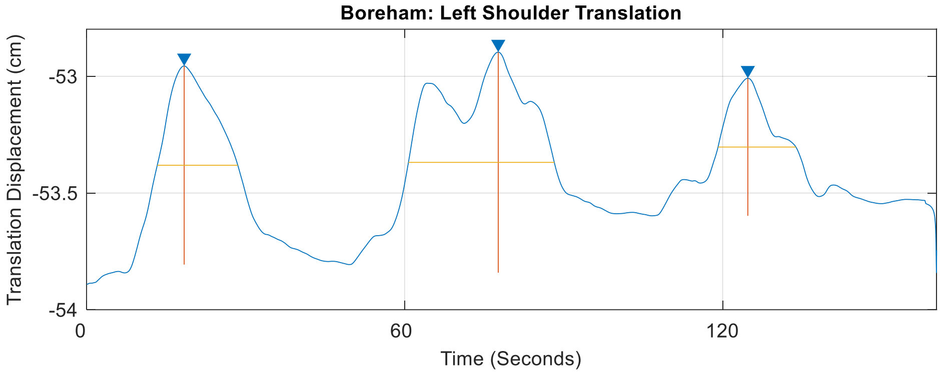

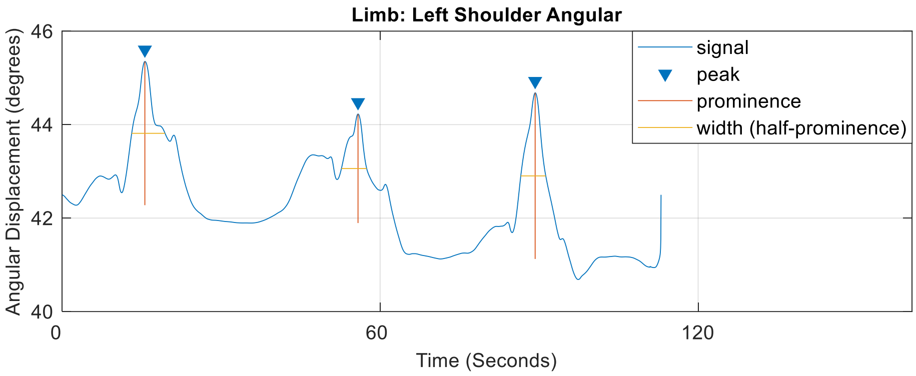

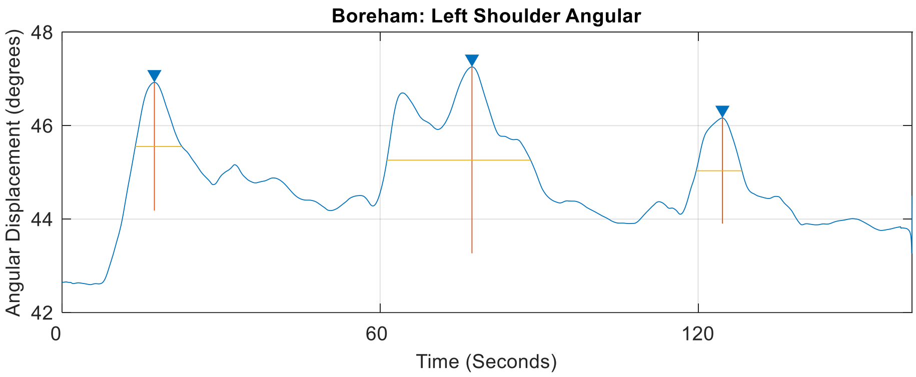

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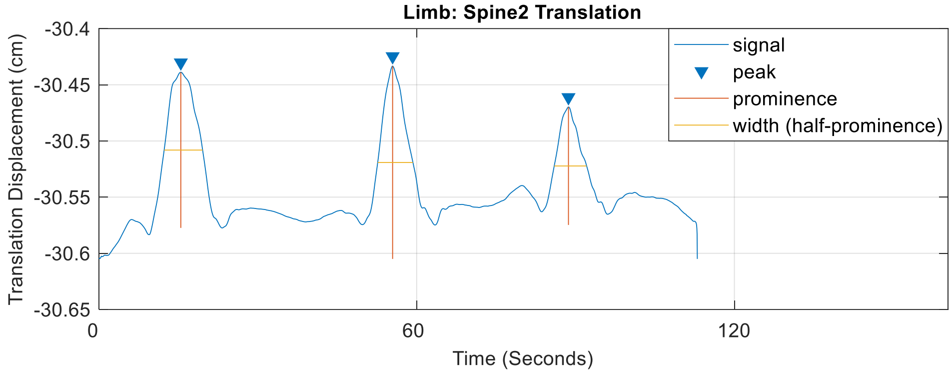
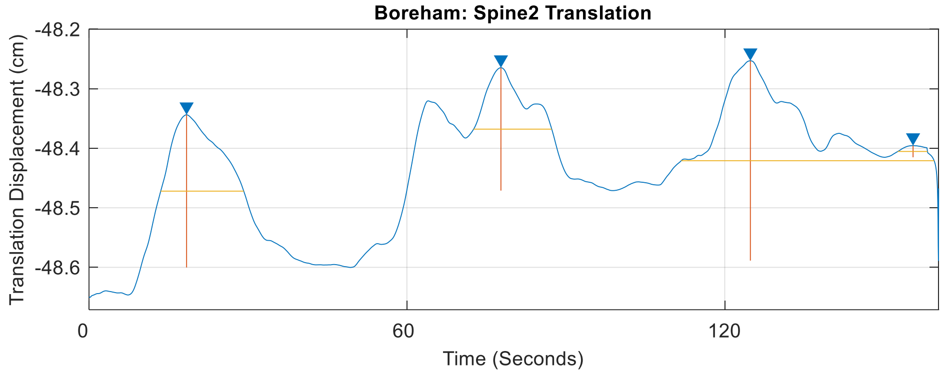
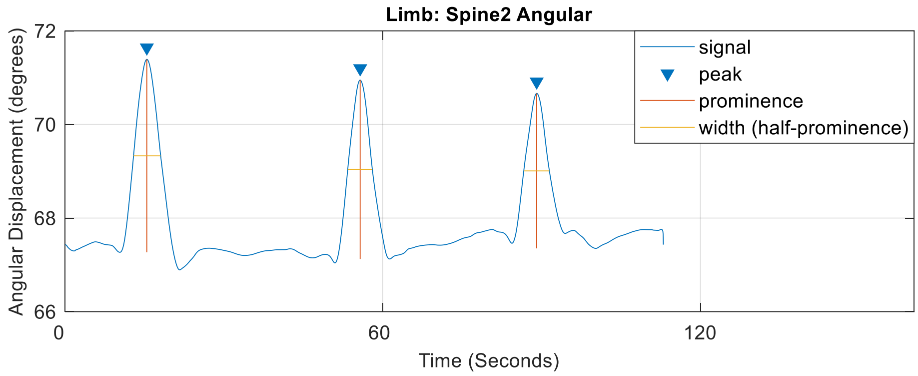
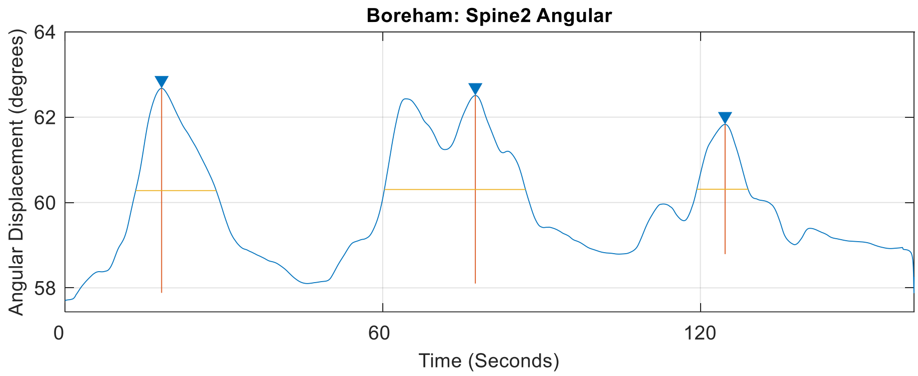
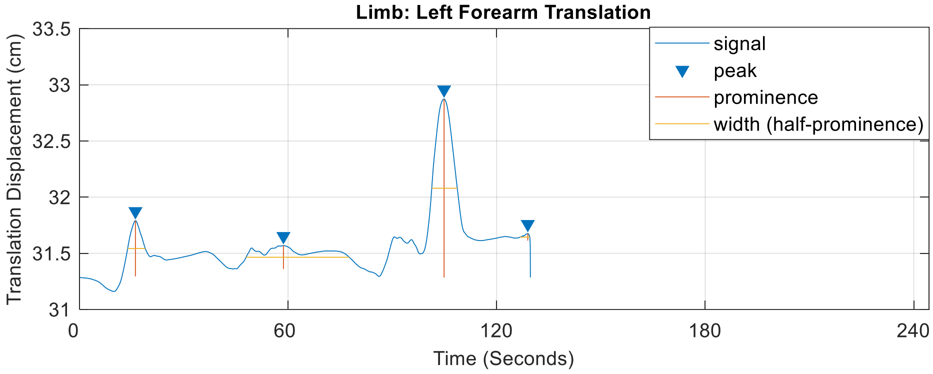


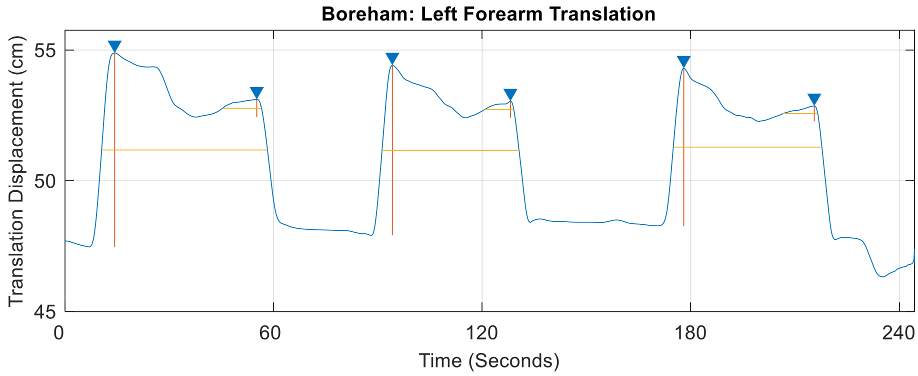


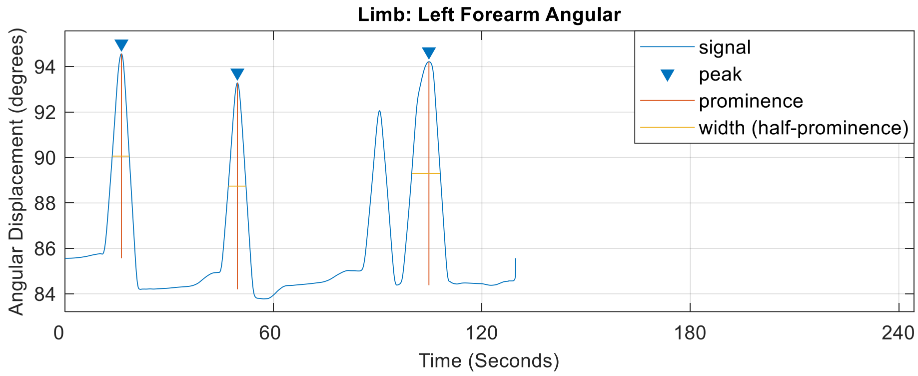


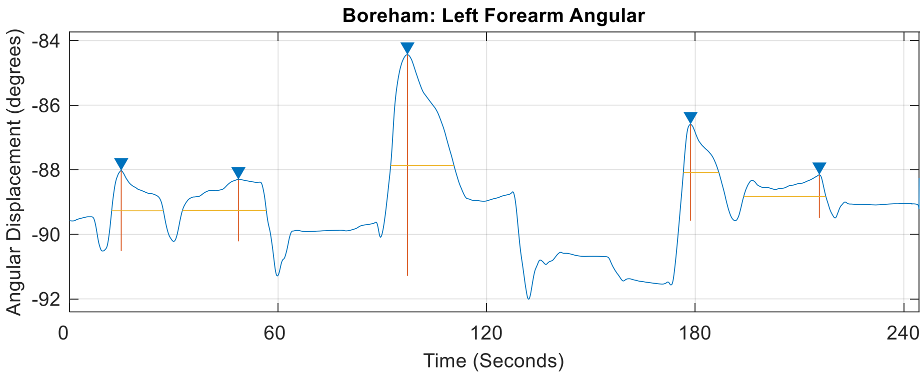


Extension MOCAP Forearm









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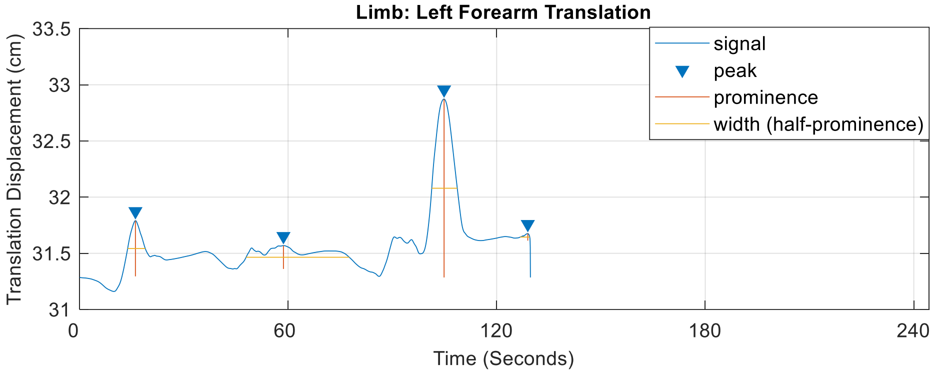

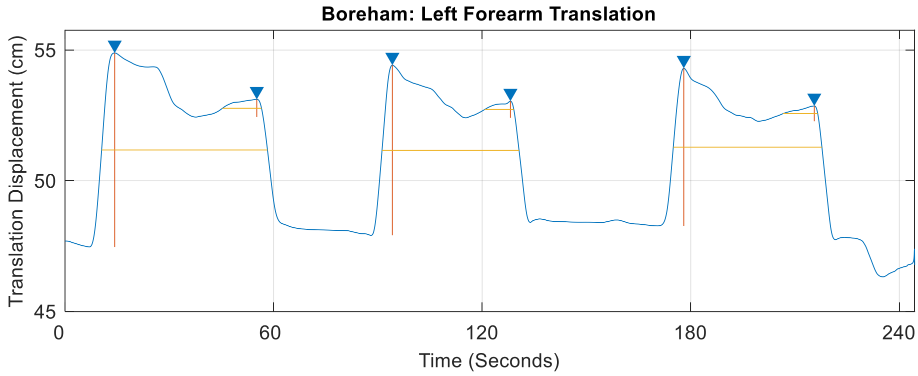

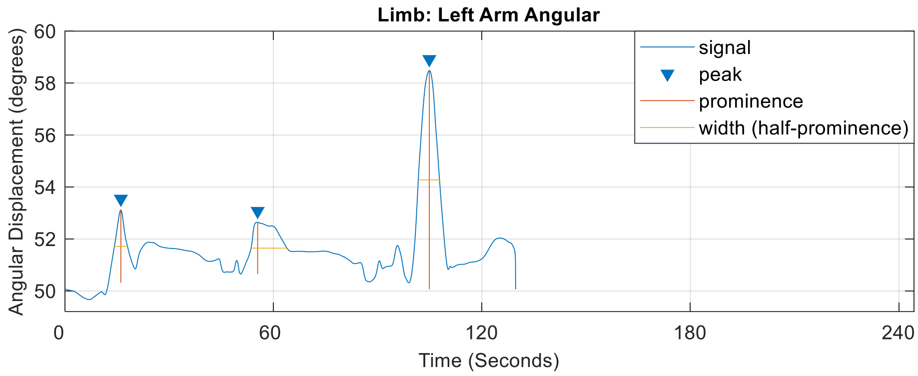

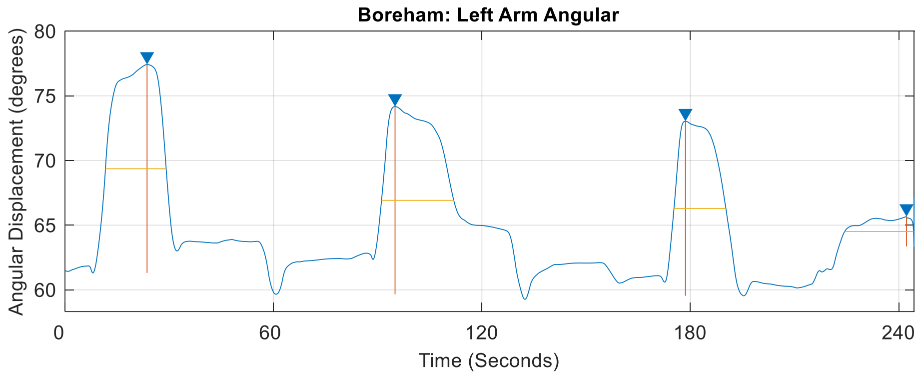

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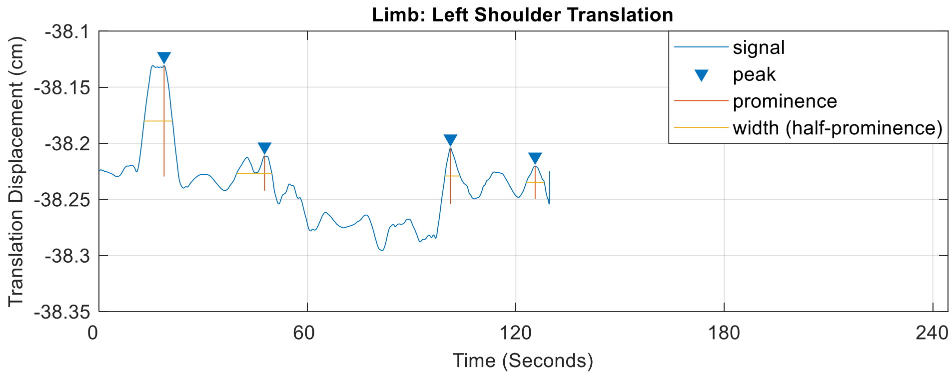

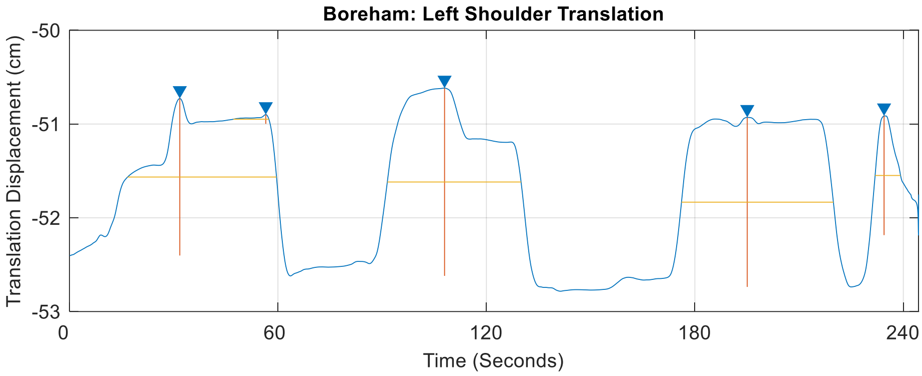

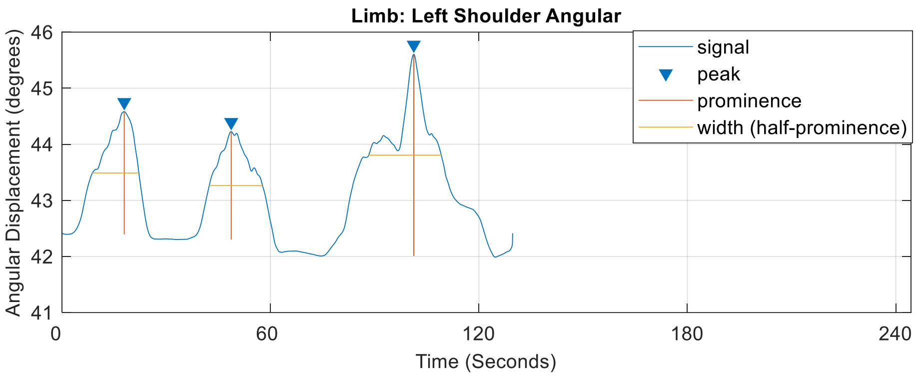

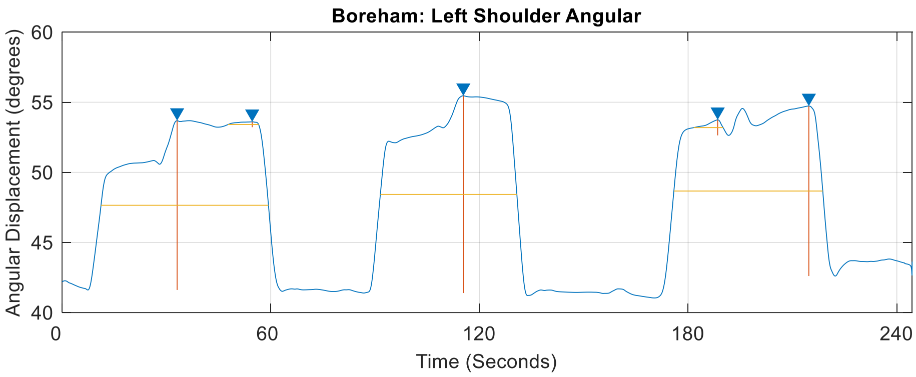

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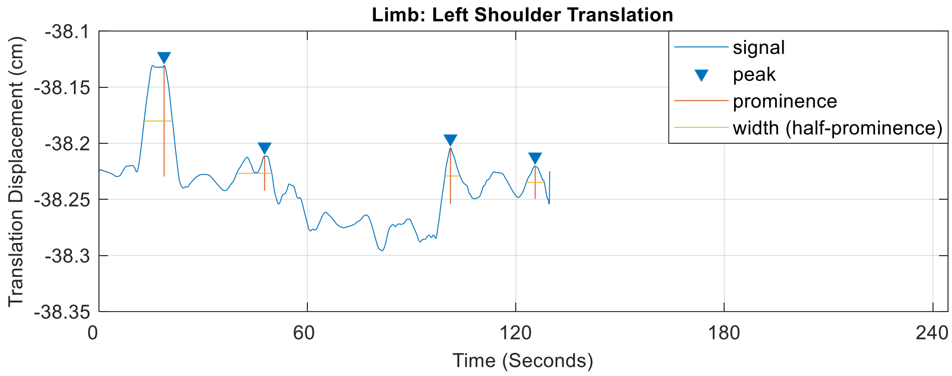
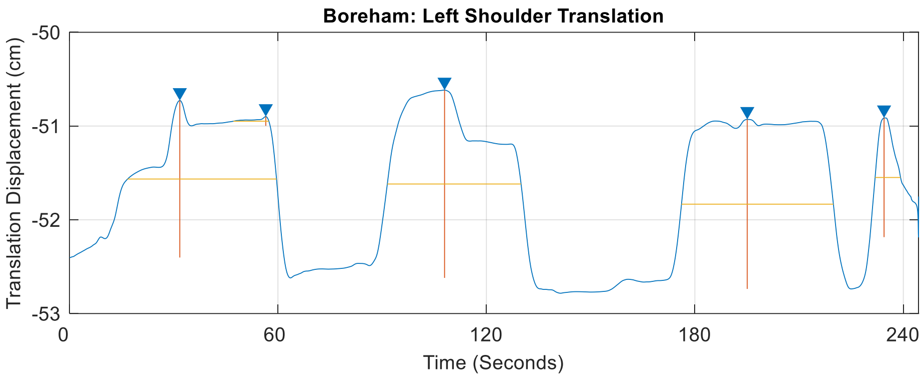
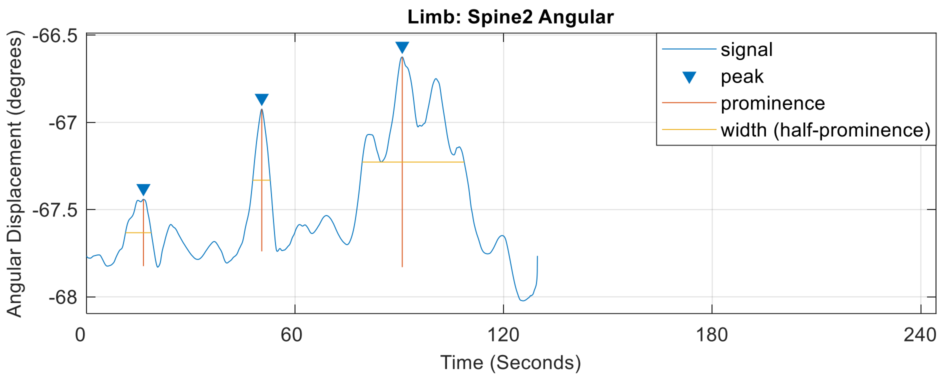
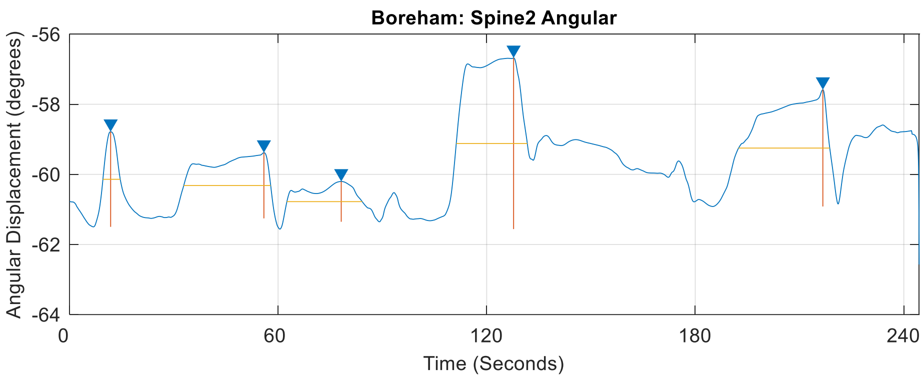
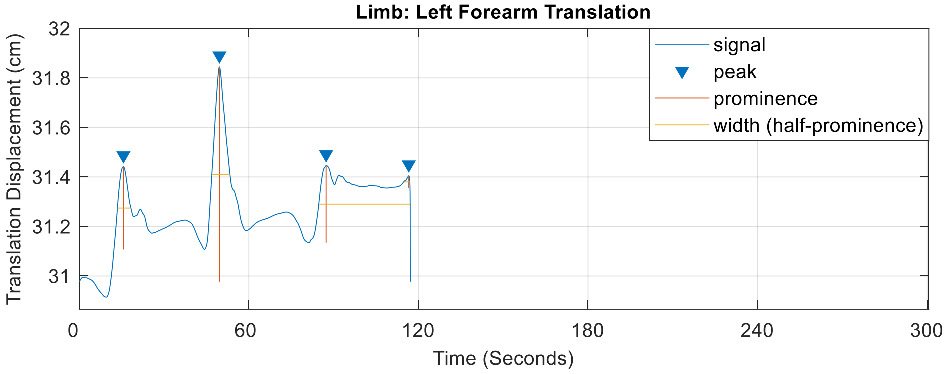


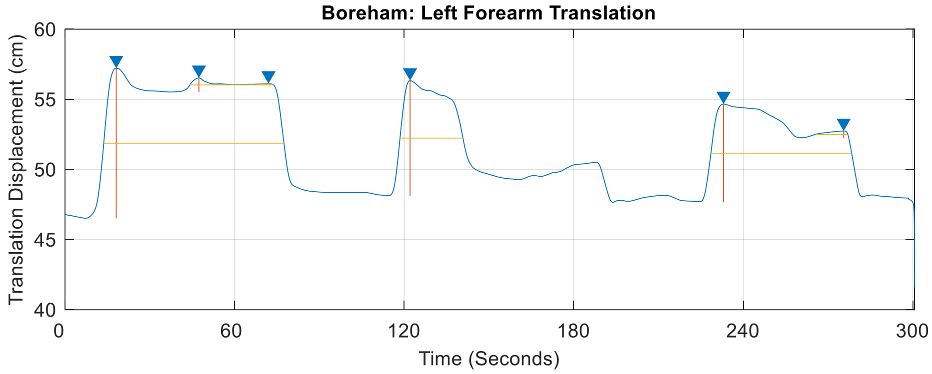


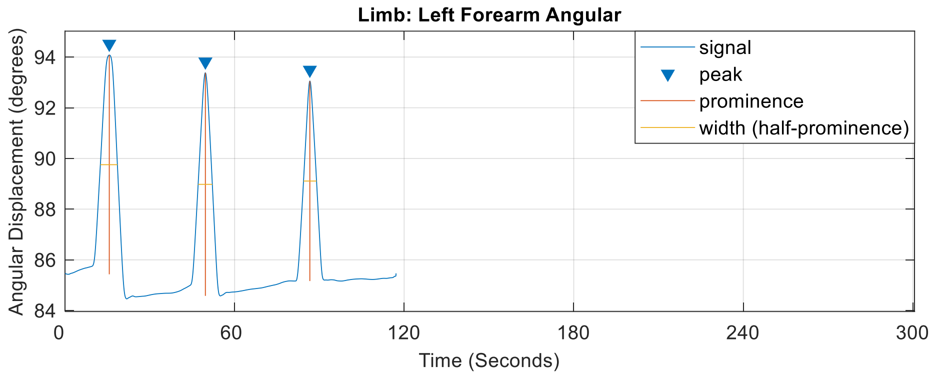


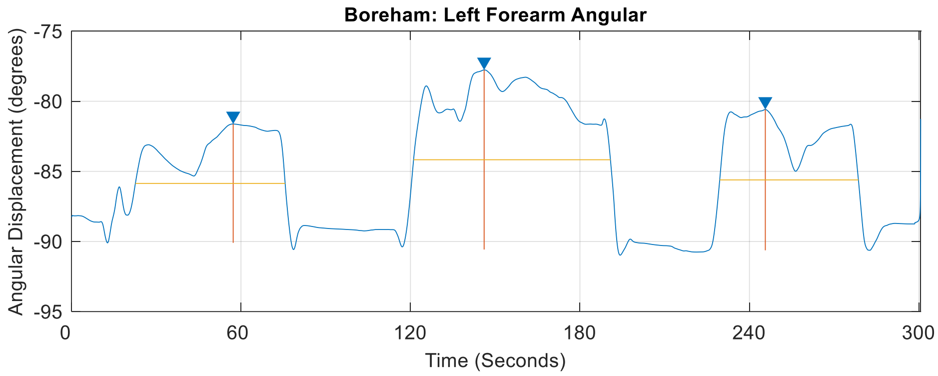


Tip MOCAP Forearm









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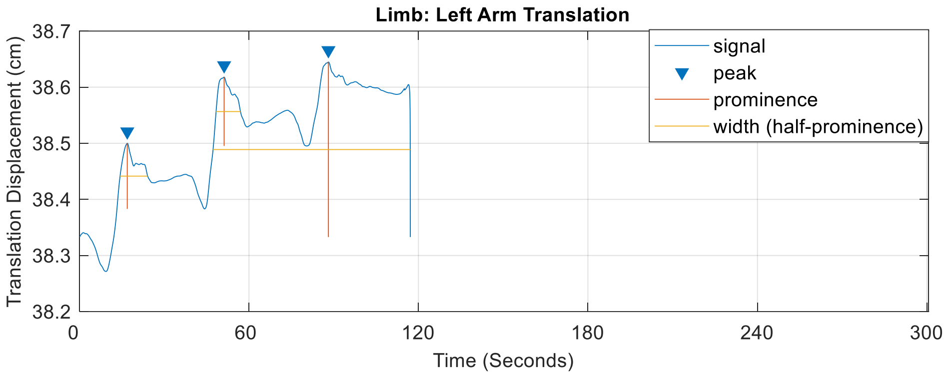

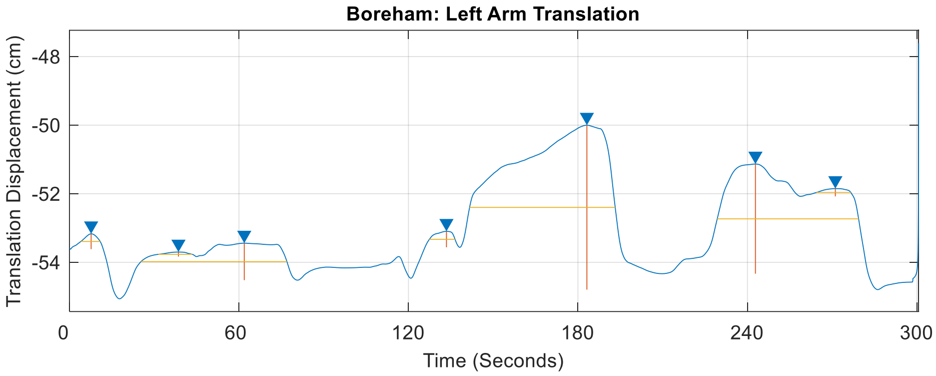

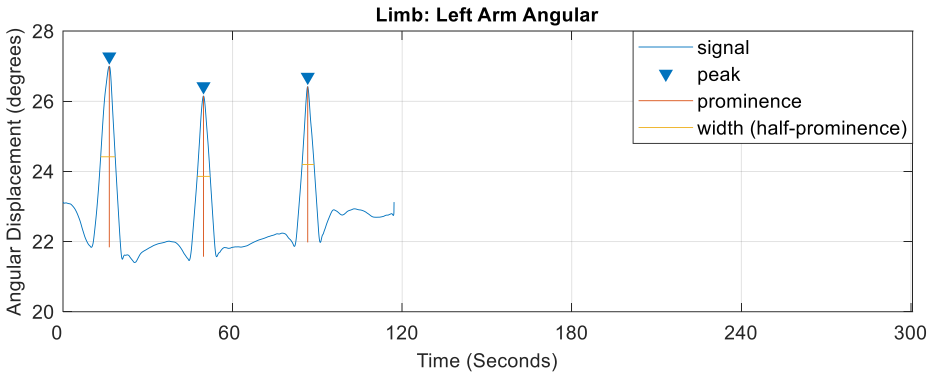

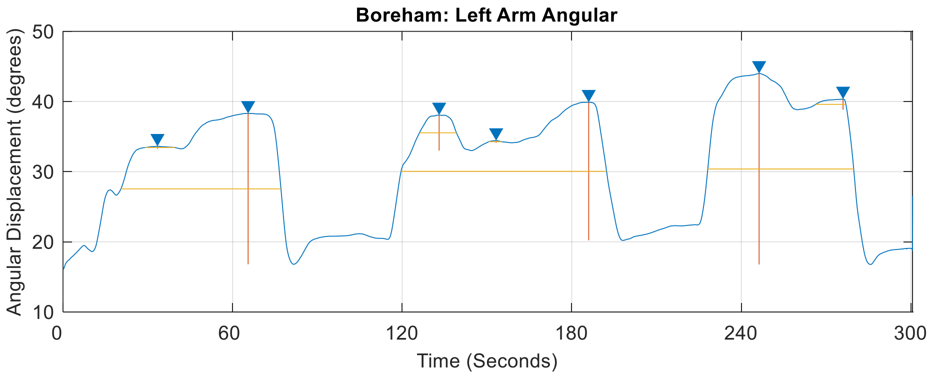

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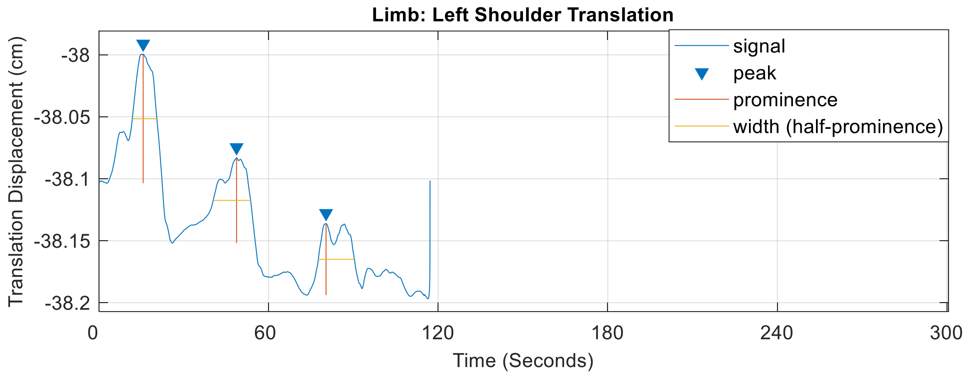

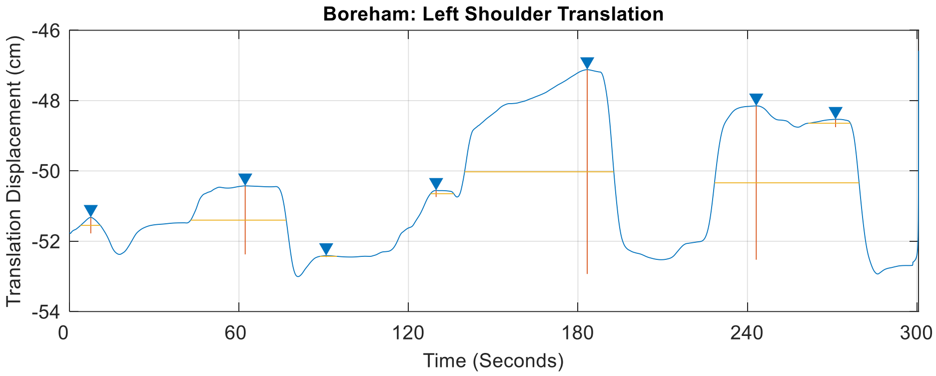

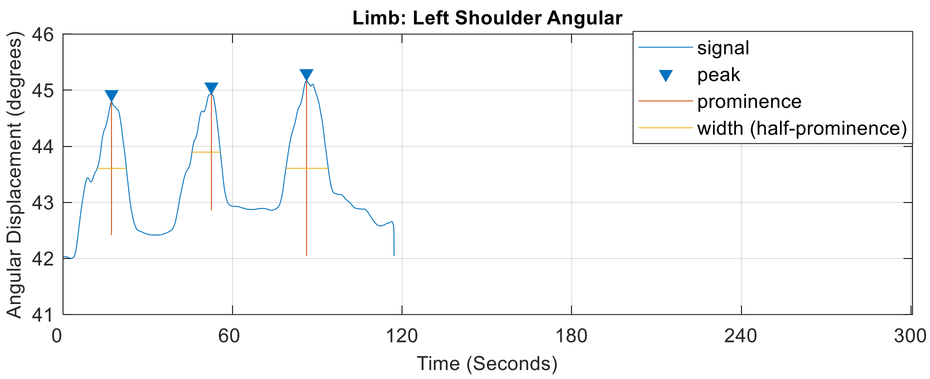

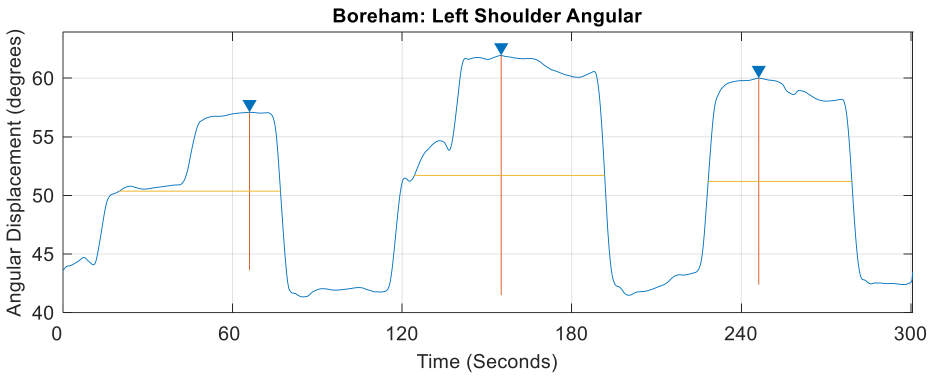

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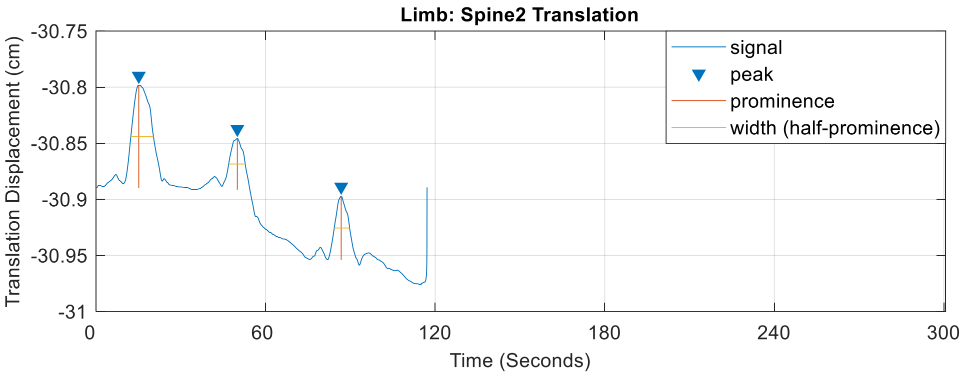
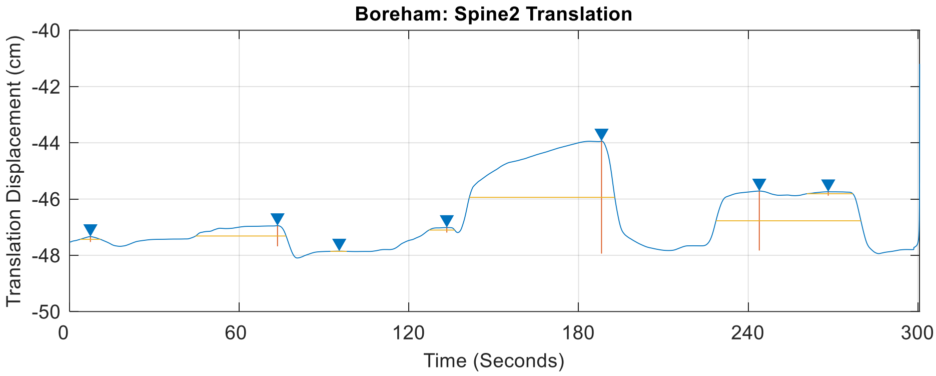
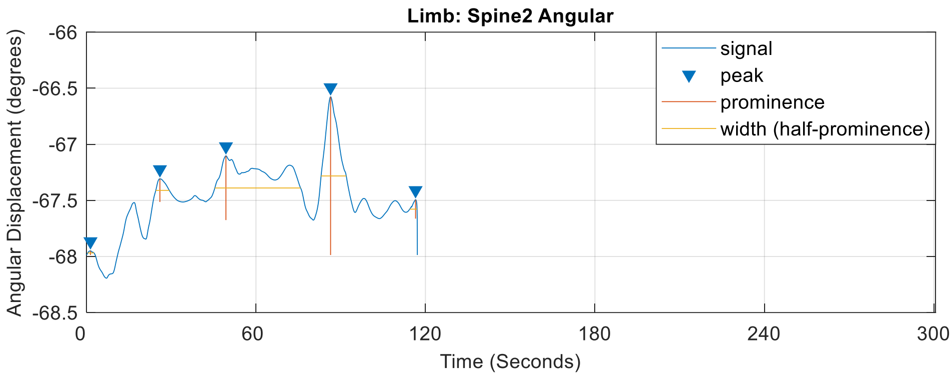
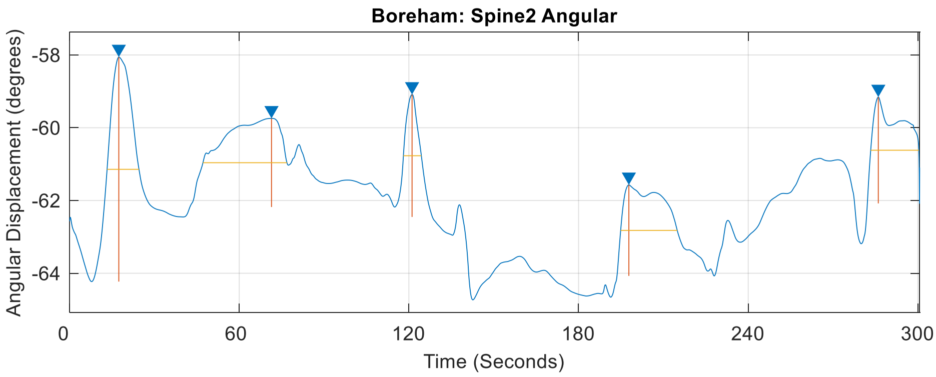
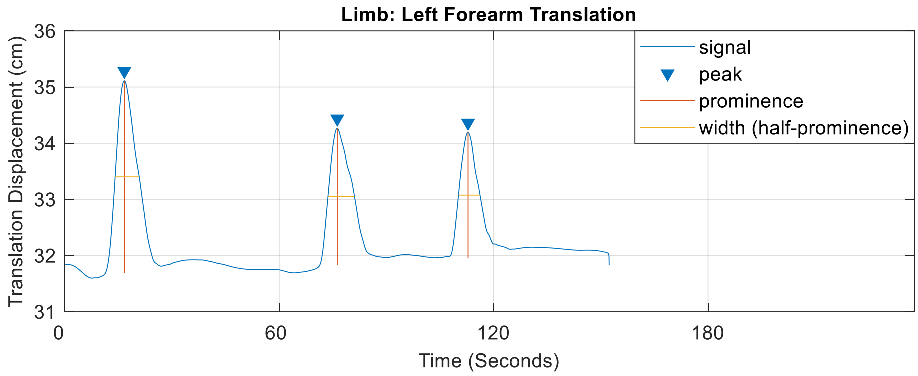


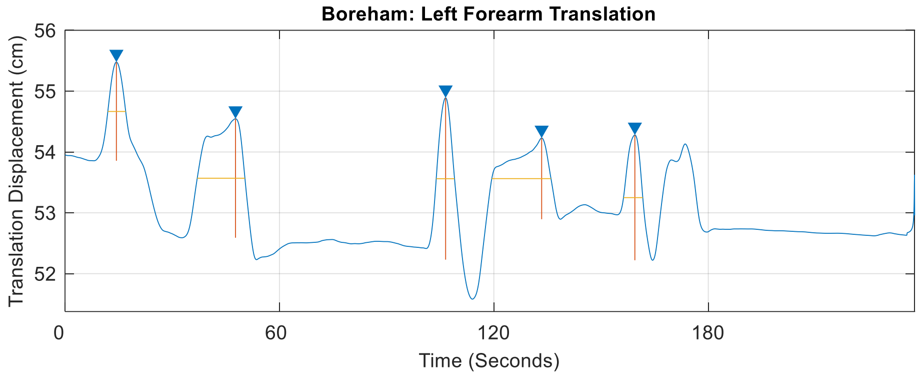


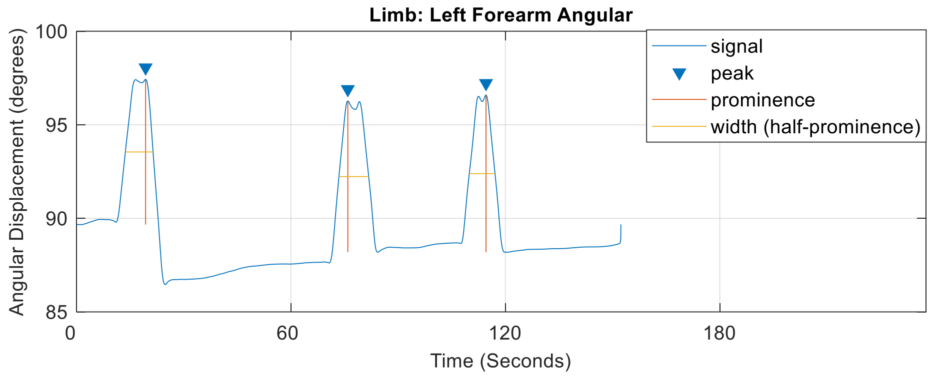


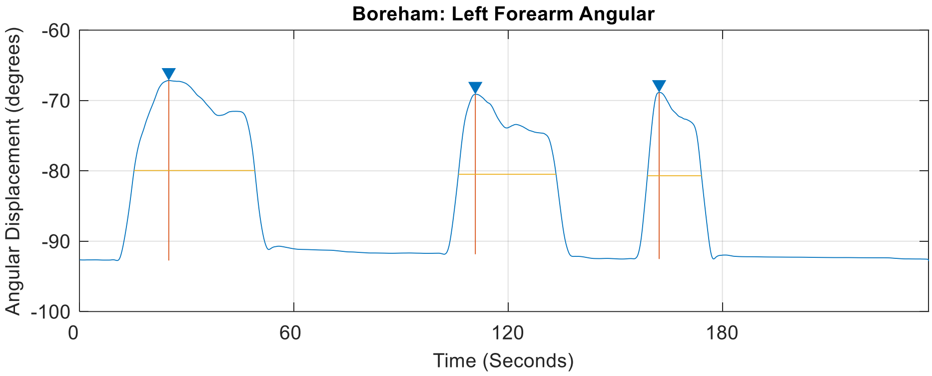


Sphere MOCAP Forearm

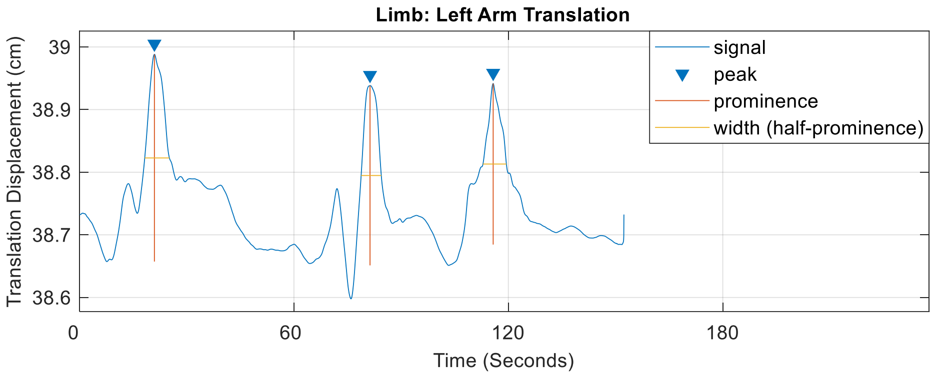


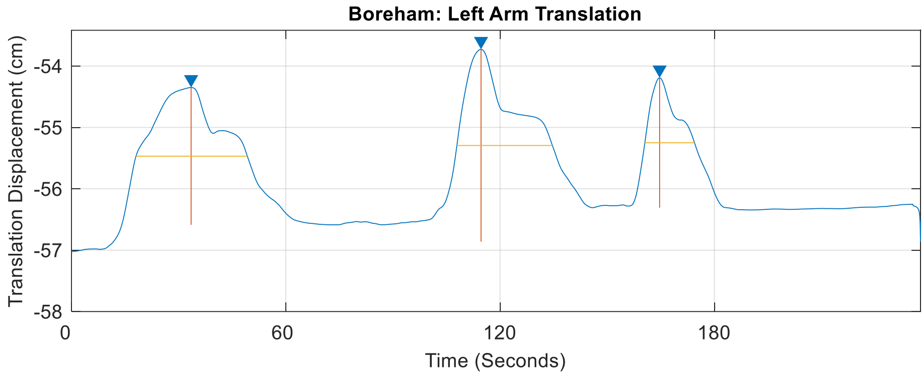


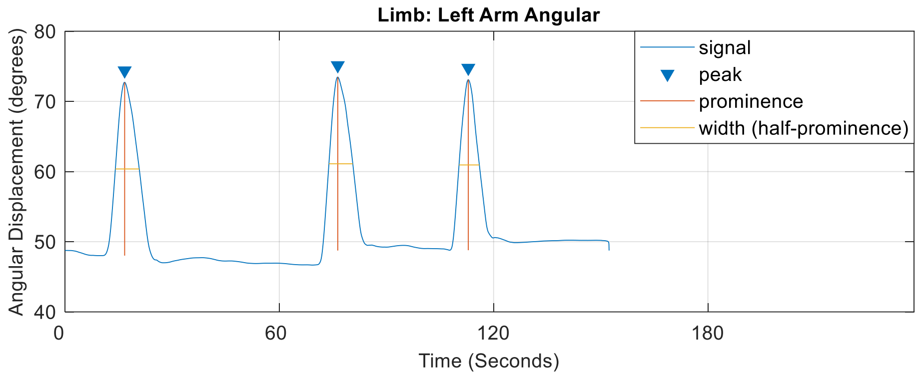


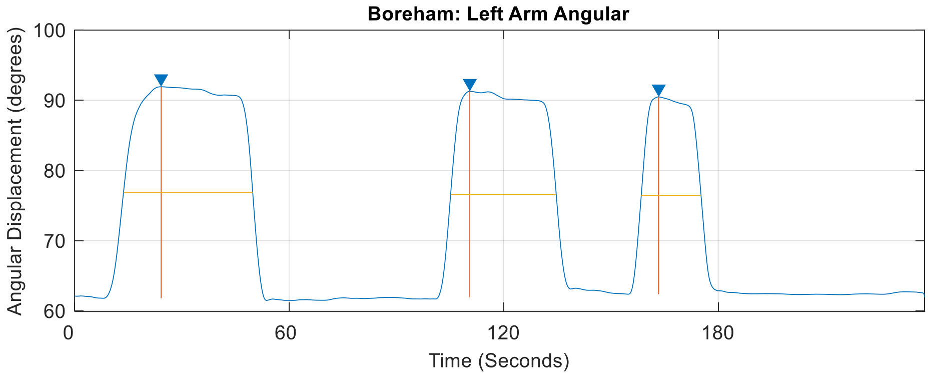


Sphere MOCAP Upper Arm









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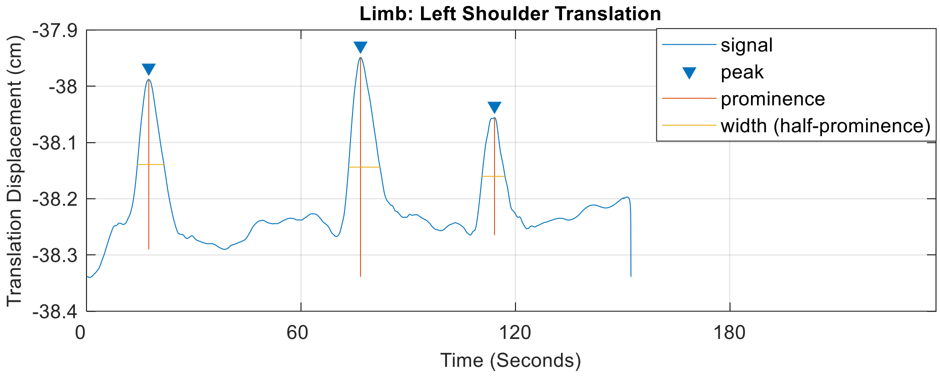
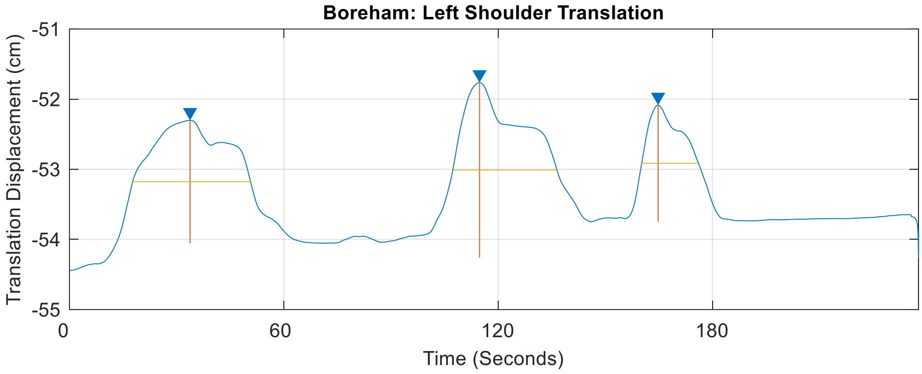
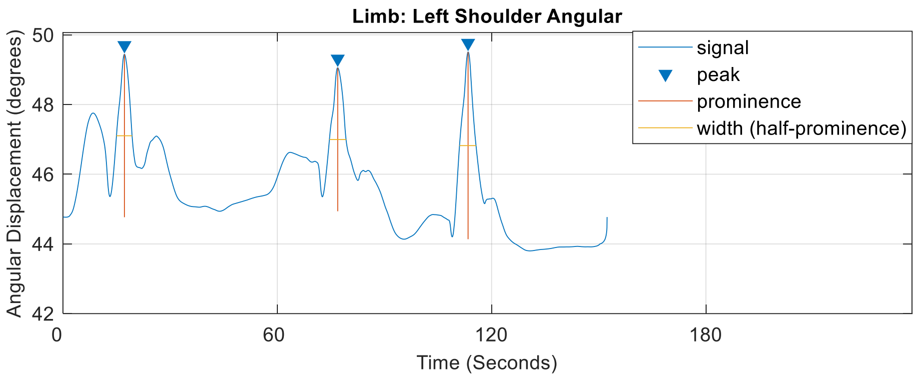
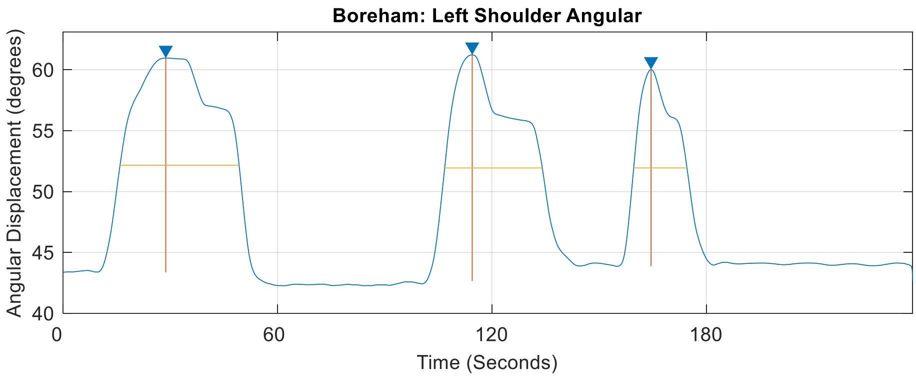
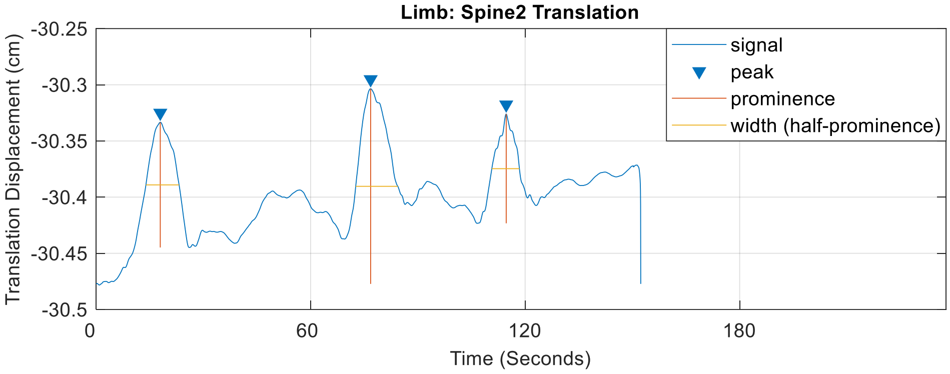


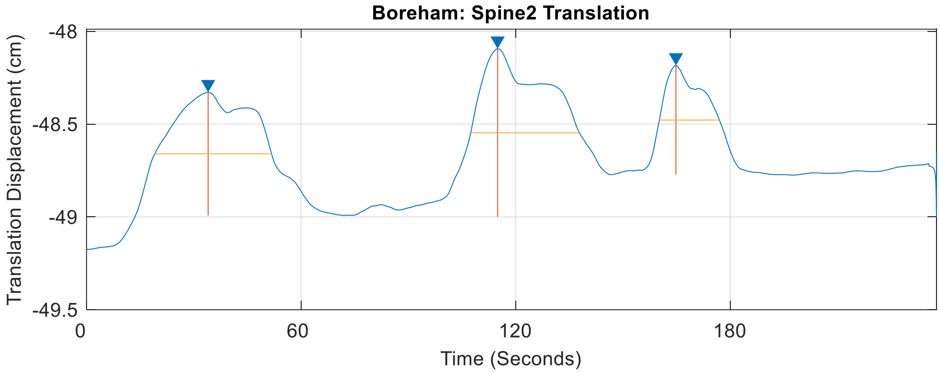


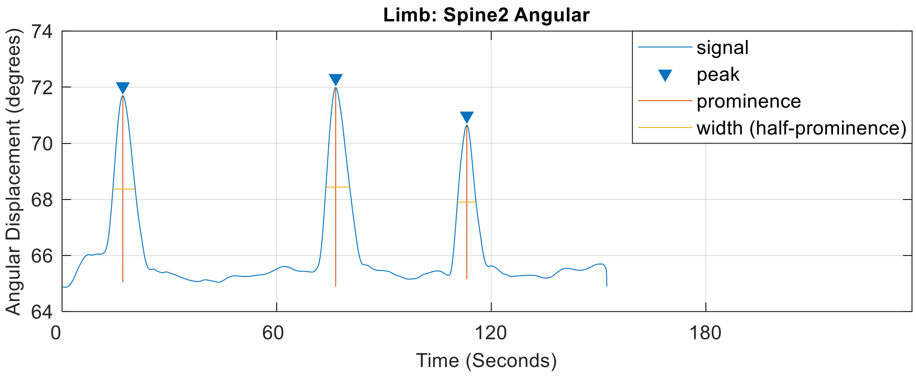


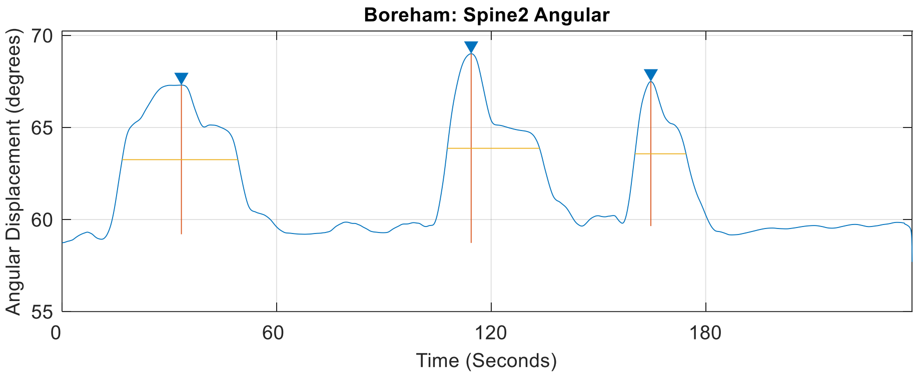


Sphere MOCAP Trunk

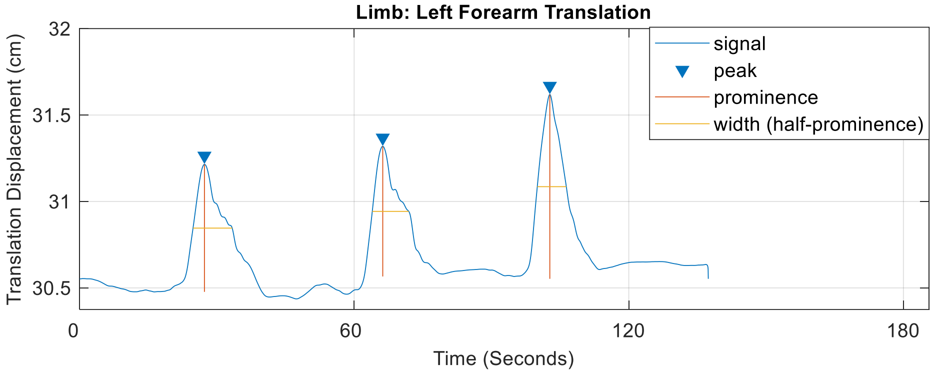


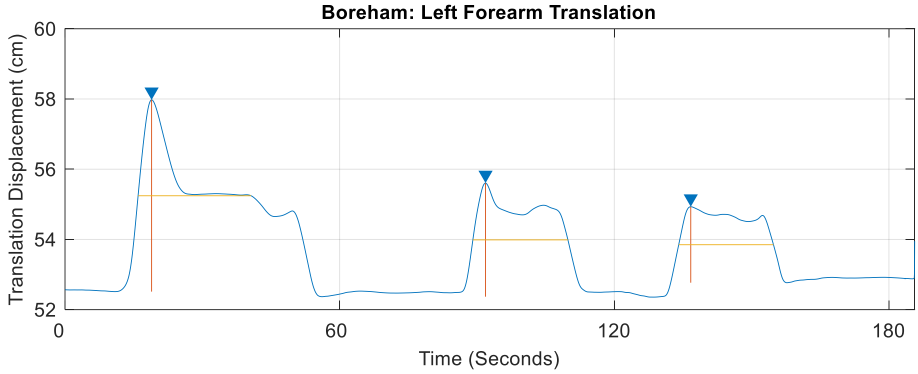


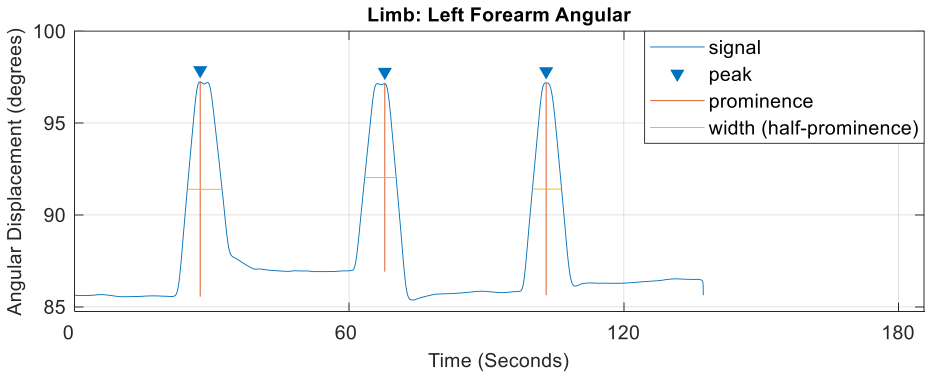


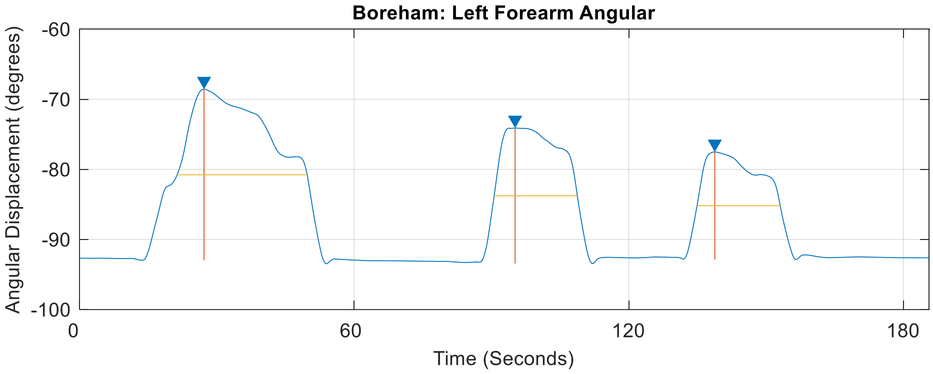


Tripod MOCAP Forearm

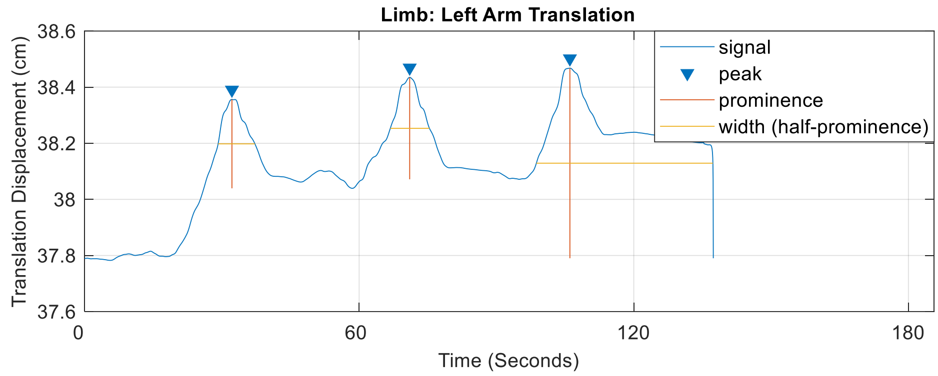


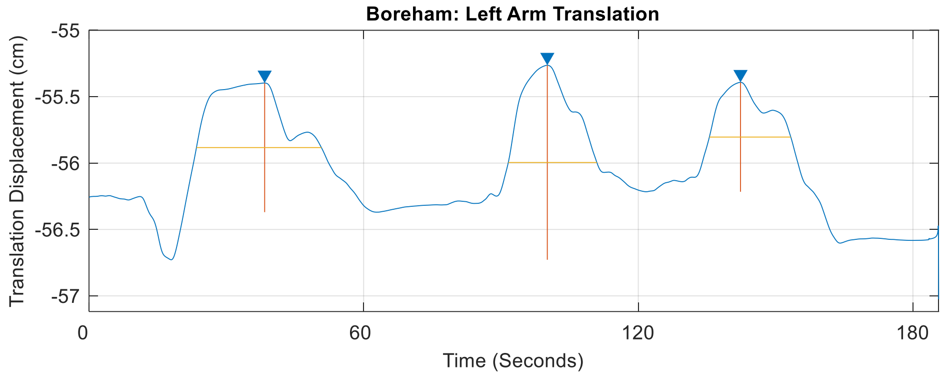


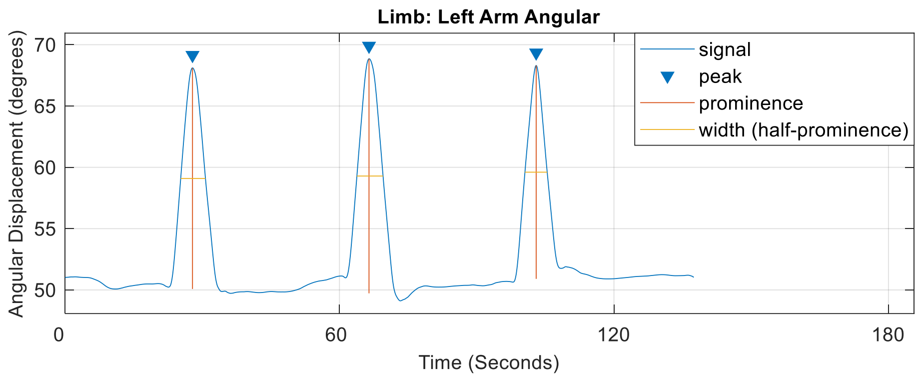


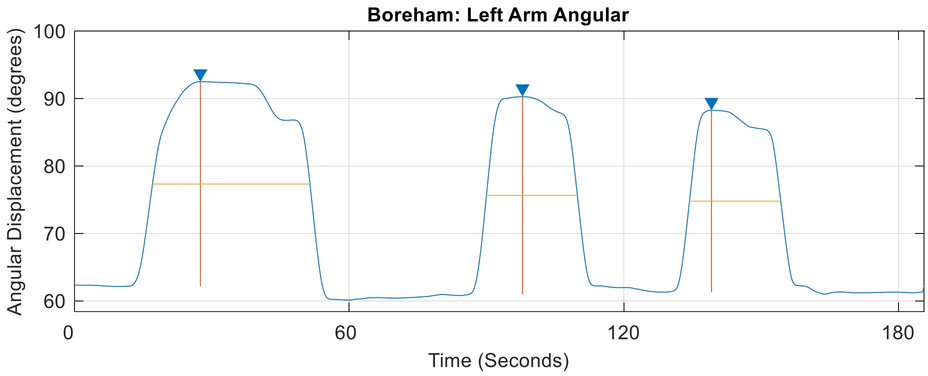


Tripod MOCAP Upper Arm









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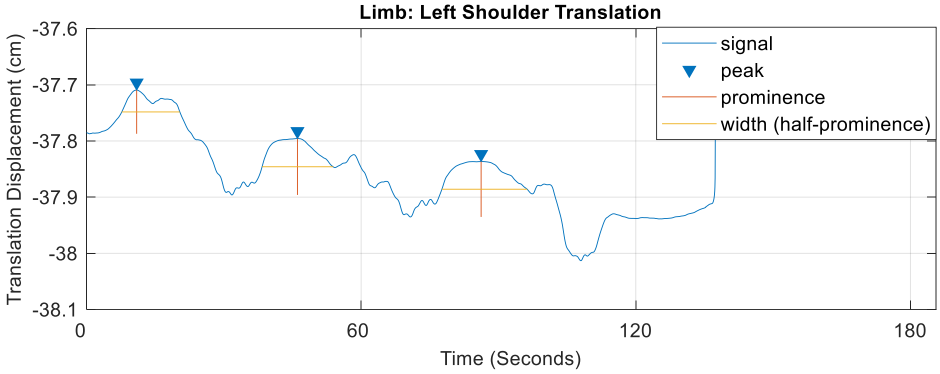
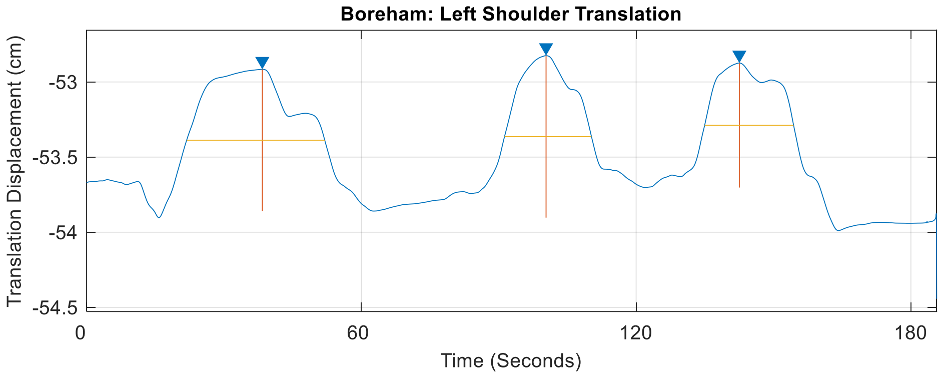
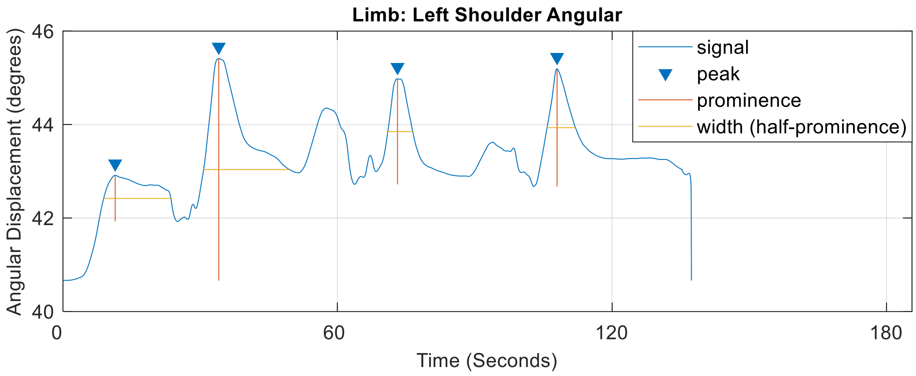
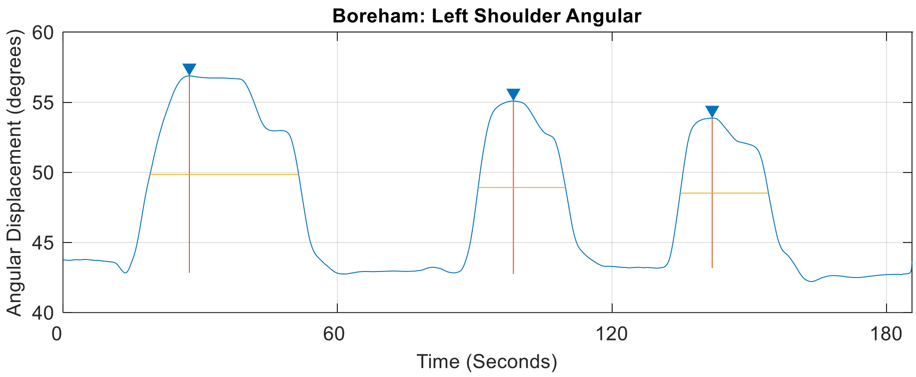
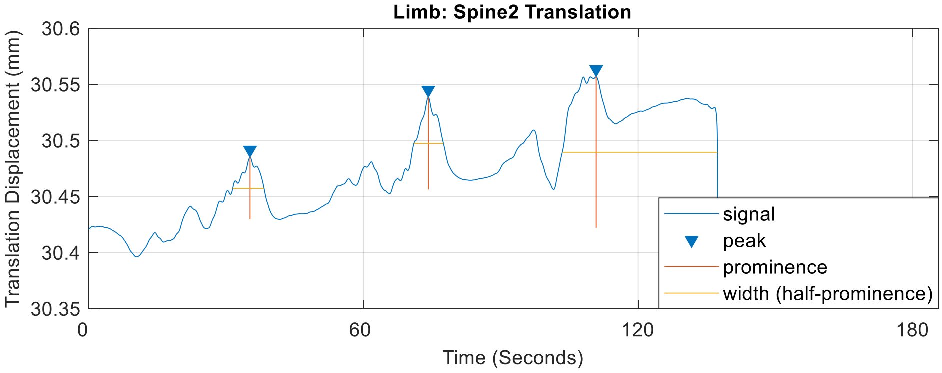


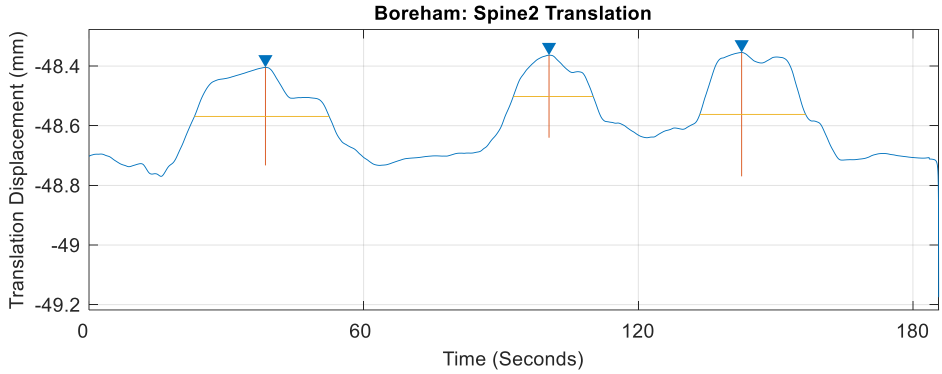


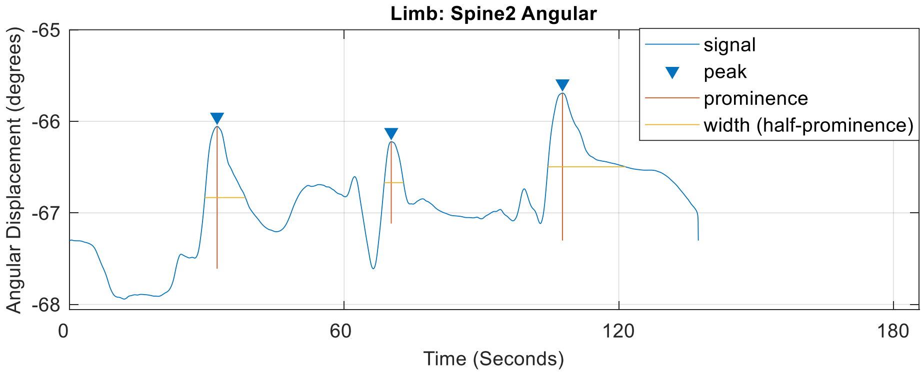


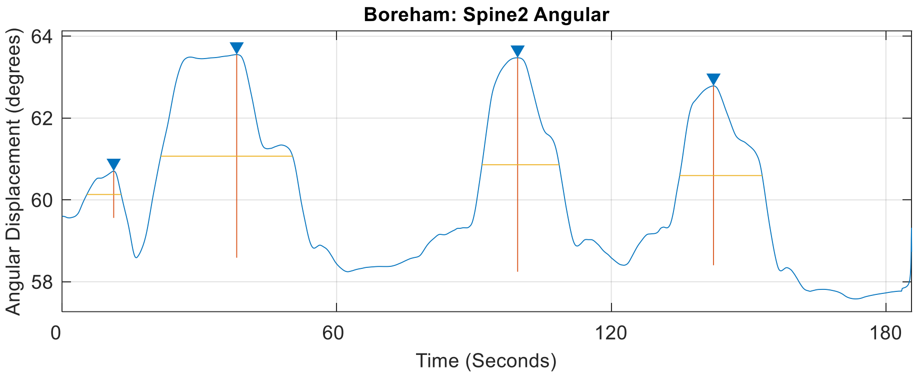


Tripod MOCAP Trunk









Appendix B

Bill of Materials for Terminal Devices

Item	Weight	Cost
Terminal Devices w/o add-ons		
Hosmer Split Hook #7[150]	298	381.6
Clamp	500	210.13
CfAM-2 Hand	330	120.15
Hackberry	256	202

Electronics		
Arduino Uno	-	18.65
GWS S03N Standard Servo	-	6.47
Goteck Micro Servo	-	5.38
Myoselectric Muscle Sensor V3	-	49.44
Breadboard	-	1.7
9V Battery	-	4.99
DFRobot 7.4V Lipo 2200mAh Battery	-	18.612
Accessories (Resistors, Wires)	-	7.37
Electronics Total	162	124.072

Body-Powered Accessories		
Harness Figure 8 Single Control	-	109.32

Appendix C

***Consumer design priorities for upper limb prosthetics from Biddiss et al.*[9]**

Adult consumer design priorities.

Electric hand ($n = 48$)	P_d	Body-powered hook ($n = 37$)	P_d
1. Weight	45	1. Comfort of harness/straps	29
2. Glove durability	23	2. Weight	23
3. Cost	20	3. Cost	20
4. Sensory feedback	16	4. Wrist movement/control	20
Fine motor skills/dexterity	16	5. Grip strength	18
5. Heat	15	Fit	18
6. Frequency of unplanned	14	6. Reliability	16
movements	12	7. Heat	12
7. Life-like	10	8. Sensory feedback	9
8. Comfort of harness	9	9. Ability to maneuver in awkward	8
9. Reliability	9	positions	8
Size	9	10. Donning/doffing	
Independently moving fingers	7	Physical effort needed to use	
10. Fit	7		
Wrist movement/control			

Pediatric consumer design priorities.

Electric hand ($n = 25$)	P_d	Body-powered hook ($n = 9$)	P_d
1. Weight	70	1. Weight	36
2. Heat	28	2. Overall appearance	29
3. Glove durability	23	Overall comfort	29
4. Sensory feedback	21	Overall function	29
5. Noise	14	3. Size	18
6. Cost	13	4. Reliability	13
7. Life-like	12	5. Life-like	11
8. Wrist movement/control	9	Fit	11
9. Size	8	Usefulness	11
Ease of cleaning	8	Harness comfort	11
10. Donning/doffing	7	Ease of control	11
		6. Heat	9
		Grip strength	9

Mass related design priorities
Cost related design priorities
Grasping functionality related design priorities