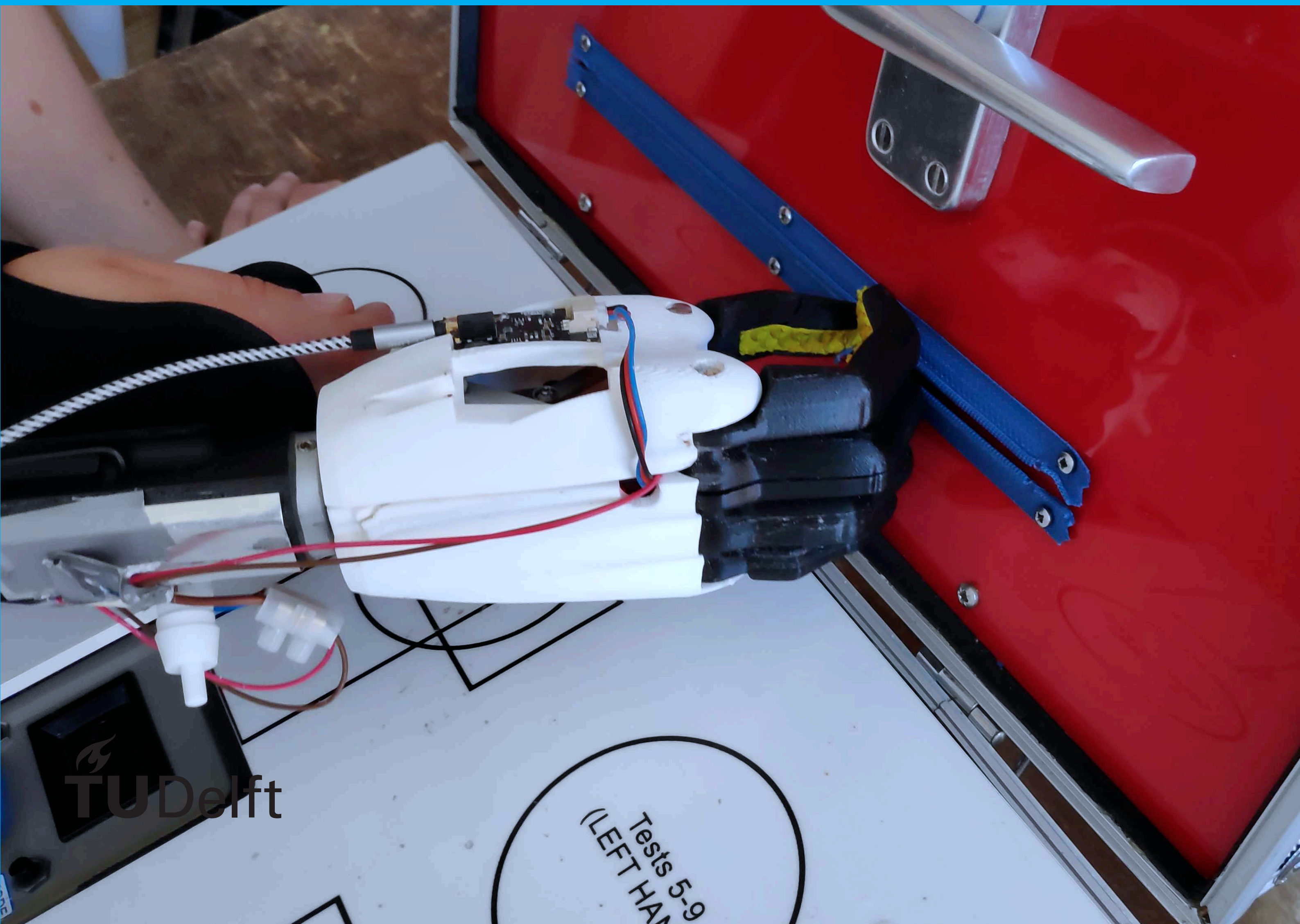
The MyoGrab Hand

Development of a functional low cost myoelectric upper limb 3D printed prosthesis

J.P. de Wit



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Development of a functional low cost myoelectric upper limb 3D printed prosthesis

by

J.P. de Wit

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Abstract

Background: Last decades myoelectric prostheses have become more usable, functional and reliable. Yet, the cost of myoelectric upper limb prostheses are still a hindering factor of widespread usage of the prostheses. In developing countries, where the demand for prostheses is relatively higher, commercial myoelectric prostheses are out of reach for the majority of amputees because of limited financial resources. Low cost myoelectric prostheses with the aim of use in developing countries are not commercially available at this moment.

Objectives: The goal of this study is to develop a functional low cost myoelectric upper limb prosthesis. The developed prosthesis will be tested with standardized and validated test methods, to evaluate if the prosthesis is usable and functional. The Box and Blocks Test (BBT) and the Southampton Hand Assessment Procedure (SHAP) will be used for this purpose.

Results: A low cost myoelectric prosthesis prototype was developed in this study; the MyoGrab Hand. The material cost of the prototype was 99.33 euro. The weight of the prosthesis was 352 g. The maximum pinch force of the hand prosthesis was 54.8 N. The MyoGrab Hand was tested by 20 able-bodied participants with BBT and the SHAP. In a first evaluation with ten participants, the prototype proved its functionality with an average score of 17.0 (± 2.2) with the BBT and an Index of Functionality (IoF) of average 41 (± 8.3) with the SHAP. After optimizing the Arduino code structure, the MyoGrab Hand was evaluated again. Ten able-bodied participants carried out the BBT. The average score on the BBT with the optimized MyoGrab Hand was 18.4 (± 2.4). The result of the BBT in the second evaluation was not significantly better than in the first evaluation (t = -1.3, df = 18, p = 0.21). In a durability experiment, critical failure of the prototype occurred after 16539 cycles of opening and closing of the hand.

Conclusion: The goal of this study, the development of a functional low cost myoelectric prosthesis, has been achieved. The MyoGrab Hand was evaluated with standardized and validated test methods and proved itself functional. This is one of the first studies that was focused on the functionality, usability and durability of a low cost myoelectric prosthesis. The functionality of the MyoGrab Hand does not equal the functionality of commercial myoelectric prostheses, commercial myoelectric prostheses are more functional. The MyoGrab Hand was able to perform most of the activities of daily life (ADL) tasks in the SHAP, however, carrying out these ADL tasks took longer, as compared to commercial prostheses. This study is a first step in the direction of making a functional myoelectric prosthesis available to larger part of the amputee population with limited financial resources. Future research should focus on a lower weight battery solution, waterproof design and testing in a non-clinical setting.

Preface

During a study trip with the study association Labyrint in my second year of the Psychology bachelor at the University of Leiden, we visited a clinic at the university in Valencia. Both physically and mentally impaired patients were helped with assistive technology to improve their functioning in daily life. Shown examples during the visit were a self balancing spoon for Parkinson's patients with a tremor, the so called tremor spoon and a computer controlled with eye tracking. I was inspired and enthusiastic about how the clinic improved the quality of life for patients by applying sometimes simple, sometimes advanced solutions for activities in daily life they would struggle with in daily life. After the visit I started thinking, if working in this field would be something for me. Looking back, it was the first step of a long path. For I was studying Psychology, and I did not yet exactly know what I found so interesting in this field, I started to explore the opportunities. I ended up following the first semester of the first year of Mechanical Engineering at the TU Delft for my elective courses. The elective courses were a success. I enjoyed mechanical engineering. So I started looking for my next goal, a master program. I thought back to a certain lecture during my minor. The lecturer, Gerwin Smit, mentioned the prosthesis he was working on at the time. This was something that made me think back to the clinic in Valencia, improving the quality of life of patients with the help of technology. I contacted Gerwin Smit with the questions which master program would connect to working on projects like the prosthesis project he was working on. Biomedical Engineering was the answer. I got in touch with Dick Plettenburg (coordinator of the Biomedical Engineering master program at the time). I was allowed to start the masters program with my background. The only condition was: a pre-masters program. A tough year with the essential Mechanical Engineering bachelor courses followed. I succeeded and was allowed to start the Biomedical Engineering masters program. A year later I contacted Gerwin Smit for the same reason that inspired me to start with the program in the first place: a prosthesis. I started with this graduation project, developing a myoelectric prosthesis. At this moment I am at the end of the end of the path of education, that started five years ago at the clinic in Valencia.

I want to thank Gerwin Smit, who inspired me to start with this master program, and guided me through the process of the literature study and the gradation project. I want to thank Dick Plettenburg for his help when I started with the pre-masters program and the master, and for taking place in the graduation committee. I want to thank Bob van Vliet for taking place in the graduation committee as well. I am grateful for the support of Jan van Frankenhuyzen and Jos van Driel for their support during the process of manufacturing and testing.

J.P. de Wit
Delft, June 2020

Contents

List of Abbreviations

ADL	Activities of Daily Living
BBT	Box and Blocks Test
DoF	Degrees of Freedom
EEG	Electroencephalography
EMG	Electromyography
FLC	Fuzzy Logic Control
FSM	Finite State Machine
IDE	Integrated Development Environment
IoF	Index of Functionality
LiPo	Lithium Polymer
MCU	Microcontroller Unit
NiMH	Nickel Metal Hydride
PLA	Polylactic acid
PR	Pattern Recognition
SHAP	Southampton Hand Assessment Procedure

1 Introduction

1.1. Background

This study describes the development, testing and evaluation of a functional low cost myoelectric hand prosthesis.

The first document describing about the idea of powering a prosthesis externally was a German book Ersatzglieder und Arbeitshilfen (Substitute Limbs and Work Aids), published in 1919 and written by G. Schlesinger [96]. In 1948, Reinhold Reiter developed the first myoelectric prosthesis [86]. The initially developed myoelectric prostheses lacked functional applications because of large size, heavy weight, slow speed and limited pinch force [116]. Over the years myoelectric prostheses have gradually become more functional. By 1980, the myoelectric prosthesis was a relevant clinical alternative in rehabilitation [97, 116]. Biddiss and Chau (2007) reviewed upper limb prosthesis use of 25 years [13]. During these 25 years years myoelectric prostheses became more cosmetically appealing and more functional, but abandonment rates of prostheses were still high (23%).

Recently developed myoelectric prostheses have been found to be improved on multiple fronts. Prosthetic devices have an increased number of joints and actuators [11]. A number of different grips is available and the appearance is realistic [11, 68]. The state-of-the-art Michelangelo hand proved to have increased manual dexterity, functionality in Activities of Daily Life (ADL), more natural posture and increased user satisfaction [33, 68, 82]. One characteristic has not improved over the years: the costs.

Calado et al. (2019) recently reviewed commercial myoelectric upper limb prostheses. The prosthesis with the lowest cost included in this review was 6600 USD [20]. The high costs of myoelectric prostheses are blocking the accessibility of commercial myoelectric prostheses [20, 87, 109]. These prosthetic devices are especially too expensive for developing countries [14, 70]. The incidence of traumatic amputation is higher in developing countries compared to modern countries [101]. In developing countries only a small part of the population is (partially) covered with health insurance [57]. The quality of healthcare is low in these countries and the out-of-pocket expenses are high for individuals requiring healthcare [14, 57]. But also in developed countries, like the United States (US), the coverage of healthcare insurance is limited [87]. Consequently, the current commercial myoelectric prosthesis market only serves a small portion of the worldwide amputee population, in countries with more financial resources.

Recent advances in 3D printing technology [69] and the advent of low cost electronics have the potential to change the myoelectric prosthesis costs [59, 92]. A recent trend of development of low cost myoelectric prostheses has been observed [36]. Although the research focus has intensified on this subject, no functional upper limb prosthesis has made it to the commercial market [20]. One of the factors contributing to the limited availability is the lack of mechanical and functional testing of the developed myoelectric prostheses [23, 59], which was also concluded in the conducted low cost myoelectric prosthesis review [36].

Not only the purchase of a prosthesis is costly, maintenance has to be taken into account as well [31]. Maintenance accounts for a major cost compared to the initial purchase of the prosthesis [15]. A solution would be a low cost prosthesis that is easy to assemble and has low maintenance [31]. This study will try to fill this gap by developing a myoelectric prosthesis that is low cost and functional. The prosthesis should be practical, usable and reliable in functioning.

As recommended in a recently conducted literature review and endorsed by other publications, the prosthesis should be 3D printed and use open source hand models and microcontrollers [36, 69]. 3D printed manufacturing improves the accessibility of prostheses and is low cost [20, 69]. The use of widely available and easy to manufacture solutions is preferred, to support local manufacturing. The electromyography (EMG) analysis algorithm was recommended to be reliable and simple [36], to ensure reliable functioning. Pattern recognition (PR) algorithms have been found to lack sufficient reliability outside laboratory conditions [53, 59]. The recommendations stated in the literature review will be adopted in this study [36].

1.2. Problem definition

Commercially available upper limb myoelectric prostheses are expensive and not within financial reach of the majority of amputees [69]. In recent years more low cost upper limb myoelectric prostheses have been developed. A number of studies describe the development of low cost 3D printed upper limb prostheses [24, 36, 42, 62]. However, the majority of those studies did not evaluate and test the developed prostheses using standardized and validated methods [36], if any testing was conducted at all. Hobbyists are developing open source prostheses as well. Those prostheses have the same issue, no testing data is available that proves their functionality [62]. Low cost prostheses are still not available on the market at the time of writing [20, 36].

1.3. Goal

This study has the goal of developing a low cost and functional upper limb myoelectric prosthesis. Known shortcomings, as described in the literature review, are the lack of test and evaluation data in prosthesis development studies regarding the actual functionality of the prosthesis for ADL tasks. This study will provide exactly those details, in an effort to develop a prosthesis that is functional and reliable and can be used for ADL.

2

Design requirements

2.1. System functions

The working principle of a myoelectric upper limb prosthesis can be divided in a number of functions. For every function a number of requirements are listed that have to be taken into account during the search for solutions. Table 2.1 visualizes the functions, parameters and requirements. The requirements will be elaborated on in the following section.

Table 2.1: Functions of design with the requirements.

Aspect	Parameter	Requirement
Prosthetic hand	Test data Manufacturing Availability	Available evidence proving functioning and durability Partly 3D printable Open source
Actuation	Output force Size Moving speed	Result in 30 N pinch force Able to fit inside the prosthesis $>$ 172 degree/s
Control system	Connectivity Size	Able to connect electrodes and actuation Able to fit inside the prosthesis
EMG measurement	Applying of electrodes	Multiple use Non-invasive Non-irritating
EMG analysis	Number of variables Computational load	Control of all DoF of prosthetic hand* Be able to run on selected microcontroller
Power source	Supply MCU	+5 V connection
Costs	Material costs	$<$ 180 euro
Weight		$<$ 400 g
Durability	Cycles till failure	$>$ 130.000 cycles

*Depends on the selected prosthetic hand.

2.1.1. Prosthetic hand

For this study, an already existing prosthetic hand design will be adopted. The adoption of an existing prosthetic hand allows us to purely focus on the myoelectric control and functionality of the prosthesis rather than the development of the prosthetic hand design. A number of previously conducted myoelectric prosthesis development studies have been adopting existing prosthetic hands in their research [2, 18, 32]. This requires the design to be open source. In the process of selecting suitable designs, it should be taken into account that the actuation mechanism and control mechanism should be able to fit in the hand.

The recommendation from the literature review [36], to use 3D printing as manufacturing method, will be adopted as a second requirement. 3D printing allows for local manufacturing in developing countries. Online sources and communities offer a great variability of 3D printable prosthetic hand designs [62].

Research documentation and functional and mechanical testing data are required to draw conclusions about the feasibility of the prosthetic hand for the application of the hand in this study. Therefore the availability test data and documentation are set as an design requirement.

2.1.2. Actuation

Actuation is required to move the prosthetic hand. The actuation output force should be able to achieve a pinch force of 30 N with the prosthetic hand. A pinch force of 30 N is necessary for functional use in ADL [100].

The space inside the prosthesis is limited. The actuation should be able to fit inside the prosthetic hand. The size of the actuation mechanism is therefore the second requirement for this function.

Although the maximum grasping speed of the human hand is high (2290 degree/s) [54], the grasping speed to grasp objects during ADL is much lower. The grasping speed of a human hand for pick and place tasks is 172 degree/s [54]. The grasping speed of at least 172 degree/s is set as the design requirement for the actuation mechanism.

The number of actuators depends on the degrees of freedom (DoF) that the prosthetic hand offers. If the selected prosthetic hand allows for control of more than one DoF and if there is enough space for more than one actuator, more actuators will be implemented.

2.1.3. Control system

The activation of the actuator and the analysis of the EMG signal requires a control system. Microcontrollers (MCU's) are generally used for this purpose [36]. The MCU is integrated in a printed circuit board (PCB). To avoid damage of the MCU, incorporation of the MCU inside the prosthetic hand is required. Consequently the size of MCU is limited to the available free space in the prosthetic hand.

The EMG analysis algorithm will run on the MCU. The computational load of the EMG analysis algorithm should fit the computational capacity of the MCU. The EMG analysis will be discussed in one of the following sections.

2.1.4. EMG measurement

Electromyography (EMG) is the technique of measuring muscle activation. The user of the prosthesis will have control over the prosthesis with the use of muscle activation which is measured with electrodes [48]. Electrodes measure the difference in electrical potential on the skin at two proximal points, as a result of muscle activation [55, 56]. The electrodes are attached to the skin, close to a muscle, or implanted [20].

For comfortable use of the prosthesis, the electrodes should be easy to apply and detach [48]. Multiple use and comfortable putting on and off of the electrodes is of importance. Since a prosthesis is an integrated accessory in daily life for the user, the electrodes should be safe to use. The electrodes should not irritate or hurt the skin. Published research focusing on the safety of electrodes is able to serve as proof of safe use.

2.1.5. EMG analysis

The EMG signal measured by the electrodes is rectified, amplified, filtered and sampled, and usable for analysis [75]. A number of algorithms is available for analysis [48, 75]. Algorithms differ in the number of degrees of freedom that can be controlled. The number of DoF of the prosthesis in this study relies on the choice of prosthetic hand. Reliability and stability are most important for functionality as concluded in the literature review [36]. With the selection of solutions, the reliability of the analysis method should be maximized.

2.1.6. Power source

The system needs power to supply the PCB and the actuation mechanism. The power source has to be electric for the control unit and may need an additional source of power to supply the actuation mechanism depending on the type of actuation. The replacement or charging of the power source has to be taken into account. Functionality and safety requires a reliable and stable functioning prosthesis, of which low maintenance is a contributing factor. The usability drops with frequent maintenance, such as having to replacing the power source.

2.1.7. Costs

The goal of this study is to develop a prosthesis that is affordable, also in lower income countries. Middle lower income countries are defined as countries with a GDP per capita of 1.026 USD and 3.995 USD by the World Bank [10]. One of the countries with worlds largest population is India (1.3 billion [9]), a lower middle income country. The average annual income per capita in India is 1835 euro (2010 USD) [8]. The maximum cost requirement for the prosthesis to be developed in this study will be set at a maximum cost of 10% of the Indian average yearly income, a cost of maximum 180 euro. The cost requirement is limiting the material costs of the prosthesis. Other cost factors, such as labor and equipment (e.g. a 3D printer), will not be taken into account.

2.1.8. Weight

Previously conducted research found heavy weight as a reason for disuse of the prosthesis [12, 13, 72]. As concluded in the literature study, most in research developed low cost myoelectric prostheses are heavier than an average human hand [36]. The weight of an average human hand is 400 g [25]. The weight of an average human hand is set as the design requirement for the weight of the prosthesis, at 400 g. As the power source will not be integrated in the hand, it can be worn elsewhere on the body. Therefore the battery weight will not be included in the design requirements for the prosthesis weight.

2.1.9. Durability

The durability of a prosthesis is an important factor, especially for use in developing countries where the availability of service centers for repair of prostheses is limited [31]. The user relies on the prosthesis, and therefore the user has to known what to expect in terms of durability. Repair of the prosthesis results in inconvenience for the user and extra costs [15]. As estimated by Luchetti et al., the yearly use of a commercial myoelectric prosthesis is around 130.000 cycles [68]. The 130.000 cycles will be adopted as the durability design requirement in this study. The durability design requirement of 130.000 cycles is comparable to design requirements set in previous research (100.000 cycles requirement for the Delft Cylinder Hand) [100].

3 Design solutions

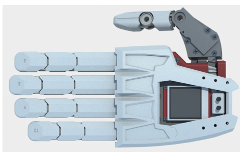
3.1. Overview

For each function of the prosthesis multiple solutions will be proposed. The proposed solutions will use existing technologies, in line with the purpose of manufacturing and use in developing countries with limited resources. The details regarding the advantages and disadvantages will be discussed for every solution. The best proposed solution will be selected and discussed at the end of each section.

3.2. Prosthetic hand

A search was conducted to find open source robotic and prosthetic hands. This resulted in four models. The hands are visualized in Table 3.1. The properties and details of each proposed robotic or prosthetic hand will be discussed in the Solution selection section.

Table 3.1: Prosthetic hand design solutions.

Function	Solutions			
Prosthetic hand	100 dollar hand 	Galileo hand 	InMoov hand 	Nazree hand 

3.2.1. 100 dollar hand

The 100 dollar hand has been developed at the Technical University of Delft. The hand was developed to function as a body powered prosthesis. The prosthesis is 3D printed. Small aluminum, laser cut bars are incorporated in the fingers for extra reinforcement. The prosthetic hand is shaped like a human hand with one DoF. The functionality and durability of the 100 dollar hand has been tested and documented in three graduation theses [17, 91, 110].

Boere [17] studied the functionality of the prosthesis by evaluating the gross manual dexterity. The Box and Blocks Test (BBT) was used in the evaluation of the body powered 100 dollar hand. The BBT is a test used to evaluate gross hand dexterity [71] and is known as a validated test to evaluate prosthesis functioning [29, 52]. An average score of 32 with the BBT was found. As concluded by Boere the 100 dollar hand is functional, the score was comparable to the performance of the commercially available TRS Hook [17].

Roovers [91] focused on the durability of the prosthesis in both normal and extreme conditions. The durability of the 100 dollar hand was tested under normal conditions, in salty, humid condition in the hand and with sand in the hand. 219.000 cycles were achieved under normal and sanded condition. 35.000 cycles were achieved in the salty, humid condition.

Donselaar [110] improved the design of the prosthesis and reported its mechanical properties. The 100 dollar hand is capable of providing a pinch force of 30 N at the fingers. Donselaar reported a mass of 220 g.

3.2.2. Galileo hand

The Galileo hand prosthesis was developed by Fajardo et al. [41, 42]. The hand is open source available and published papers are the available documentation. Fajardo et al. focused mainly on design and the reliability of EMG control in their study. No mechanical evaluation experiments or functional testing with validated methods were conducted.

3.2.3. InMoov hand

The InMoov hand was originally developed as an art project of the artist Gael Langevin [46]. The goal of the project was to build an open source 3D printable human robot. The hand has been adopted in previously published prosthesis development studies [24, 27, 51]. Condori et al. focused on the development and evaluation of a PR algorithm to control the InMoov hand with electroencephalography (EEG) signals [27]. Canizares et al. studied the feasibility of 3D printing technology for the development of low cost prostheses [24]. Hasan et al. developed a EEG controllable prosthesis with integration of the InMoov hand [51]. None of the studies reported the use of standardized and validated tests for evaluation of the developed prostheses built around the InMoov hand. Also no mechanical tests were reported.

3.2.4. Nazree hand

The Nazree hand [73] was found online at a community based open source 3D design database. The 3D printable hand was developed by a hobbyist. No documentation and test data about its functioning is available.

3.2.5. Solution selection

The documentation and testing data of the proposed prosthetic hands is very limited, except for the 100 dollar hand. The 100 dollar hand is relatively simple in design, mechanically tested [91, 110] and functionally evaluated [17]. The Galileo hand, the InMoov hand and the Nazree hand lack both mechanical testing data and functional evaluation. A difference between the 100 dollar hand and the other proposed hands is the number of DoF. The other hands offer control over more DoF.

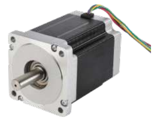
Given the available mechanical testing data available and proof of functionality, the 100 dollar hand is the most probable prosthetic hand that will result in functional, reliable and stable performance. The 100 dollar hand is therefore selected as prosthetic hand in this study.

3.3. Actuation

In previously conducted research in the field of upper limb prosthesis development the following actuation methods have been used: pneumatic [64, 77, 80, 81] and hydraulic actuators [61] and electrically driven by a motor [11, 24, 34, 42]. Electrically driven motors can be split in three categories: stepper motors, DC motors and servo motors.

One of the requirements for the actuation is the 30 N pinch force [100]. With the available documentation of the 100 dollar hand, the required input force for actuation of the hand was estimated. Donselaar found a input/output force ratio of 0.25 [110]. Thus, to achieve a 30 N pinch force of the prosthetic hand, the force at the linkage mechanism inside the hand is estimated at 120 N.

Table 3.2: Actuation design solutions.

Function	Solutions				
Actuation	Pneumatic	Hydraulic	Stepper motor	DC motor	Servo motor
					

3.3.1. Pneumatic

Pneumatic actuation in hand prostheses has been evaluated by Peerdeman et al. [77]. Peerdeman et al. concluded that pneumatic cylinders are a viable actuation method for prostheses. Although the properties of pneumatic actuation are desirable, such as the achievable output force and weight [77, 80], additional research endorsing the use of pneumatic actuation in prostheses is limited compared to electrical actuation. Pneumatic cylinders for prosthetic application are not widely commercially available.

3.3.2. Hydraulic

Miniaturised hydraulic actuation systems have been developed for prosthetic application [61]. Although miniature hydraulic actuation systems are developed in research [26, 61], their commercial availability remains limited. As concluded by Campana et al., more researched is needed for hydraulic actuation in small robotic applications, to improve the relatively heavy weight and power consumption [21].

3.3.3. Electric

Electric actuators are commonly used in prostheses [28]. An advantage of an electrical actuation mechanism is that only one power source is needed to feed both the actuation and the MCU. Three types of electric actuation for prosthetic actuation have been identified in literature: stepper motors [44, 93, 102], DC motors [11, 28, 42] and servo motors [24, 44, 50].

Stepper motors are designed for precise position control [3]. Stepper motors are relatively more expensive than servo motors [44]. Stepper motors offer high torque at low rotational speeds [102], which is a desirable property in prosthetic hand actuation.

Geared DC motors are used for prosthetic applications in combination with lead screw or worn gear [11] to minimize the backdrivability. Although the customization possibilities with geared DC motor actuation are wide-ranging, which is an advantage, the lower extent of off-the-shelf availability and with that, the relatively higher costs, pose as disadvantages.

Servo motors are designed for high torque and precise movement. Servo motors are commonly used in robotic systems for their precise movement and high torque [1]. Both characteristic are desirable for the actuation of the prototype. A large variety of off-the-shelf servo motors, that suit the purpose of prosthesis actuation, are available on the market. Servo motors are compact and able to fit inside the prosthesis. Servo motors are easy to control with Arduino software, which is recommended in the literature review [36].

3.3.4. Solution selection

Part of the goal of this study is to develop a prosthesis that is relatively easy to manufacture in developing countries with limited resources. The limited commercial availability of miniature pneumatic cylinders does not comply with this goal, and therefore pneumatic cylinders are not feasible for the actuation of the prototype. Because of the limited use of hydraulic actuation in previous prosthesis research and the limited commercial availability of miniature hydraulics, hydraulic actuation will be abandoned as solution as well.

Electrical actuation has favorable properties for actuation application in for the prototype. Electric actuation is widely commercially available, relatively easy to implement and low cost. The MCU and actuation can be powered with the same power source. Stepper motors are relatively expensive and the fundamental working principle of stepper motors do not match the actuation properties required for this study. Geared DC motors in combination with a non-backdrivability system comply with the needs of actuation in this study. However, geared DC motors with a non-backdrivability system are not as off-the-shelf available and cost effective as servo motors. Servo motors are capable of providing the required force, cost effective and widely available. The off-the-shelf availability and cost effectiveness match the goal of this study, to develop a low cost prosthesis. Therefore, servo motor actuation is selected for the prototype.

3.4. Control system

In the literature study is found that Arduino micro-controllers (MCU's) and software are generally used to control low cost myoelectric prostheses [36]. A number of MCU's are proposed. An overview of the details of the specifications of the proposed MCU's is added to Appendix A.7.

Table 3.3: Control design solutions.

Function	Solutions				
Control	Arduino Uno 	Arduino Nano 	Raspberry Pi 	Digispark Pro 	Digispark 

3.4.1. Arduino Uno and Nano

Arduino has a number of MCU's on the market, varying in size, pin connections and computational resources. For this study the Arduino Uno [5] and Arduino Nano [4] are proposed, both have been used in previously published prosthesis development studies [36]. The Arduino Nano is the smallest MCU available manufactured by Arduino. The Arduino Uno has larger dimensions, but is comparable in specifications (Appendix A.7).

3.4.2. Raspberry Pi

Raspberry Pi is capable of relatively more complex software processes compared to Arduino MCU's [89]. The Raspberry Pi was used for control in previously developed myoelectric prostheses [19, 94]. In case of a more complex EMG analysis algorithm, that Arduino is not able to compute, Raspberry Pi may offer a solution. The Raspberry Pi is relatively large compared to the Arduino MCU's.

3.4.3. Digispark

Digispark boards are similar to MCU's offered by Arduino but smaller, cheaper and they have a slightly less computational capacity [38]. Digispark MCU's are a solution if size is a limiting factor and less computational capacity is not an issue. Digispark has two MCU's on the market. The Digispark ATtiny85 is the smallest (18 x 23 mm) MCU available. The Digispark Pro is slightly larger than the Digispark ATtiny85 but smaller than the Arduino Nano. The Digispark ATtiny85 has the least computational power. Digispark boards run on Arduino software, they are compatible with the Arduino integrated development environment (IDE).

3.4.4. Solution selection

The use of Arduino compatible MCU's for prosthesis control was recommended in the literature review [36]. With the integration of commonly used control hardware and software, previously conducted, open source work can be adopted and further developed [36]. The Raspberry Pi will be dropped as a solution. The space inside the 100 dollar hand is limited. Therefore, space is an important limiting factor. The Digispark ATtiny85 is the smallest MCU and compatible with the Arduino IDE. The Digispark ATtiny85 is selected as MCU to be used in the prototype.

3.5. EMG measurement

The electrical potential resulting from muscle activation is measured with electrodes. Multiple types of electrodes are on the market. Three main types can be distinguished: invasive electrodes, gel electrodes and dry electrodes.

Table 3.4: EMG measurement design solutions.

Function	Solutions			
EMG measurement	Invasive electrodes 	Gel electrodes 	Dry electrodes 	Myo armband 

3.5.1. Invasive electrodes

Invasive electrodes are either inserted into the muscle (needle electrodes) or implanted to measure EMG [112]. These are the most accurate and reliable methods of measuring EMG, because crosstalk (noise from neighbouring muscles) of other muscle signals is minimized [48, 112]. The implantable electrodes require surgery, which makes application complicated and expensive [20]. Needle electrodes are inserted into the muscle, which is uncomfortable and can be experienced as painful [103]. The use of invasive electrodes results in the most reliable and accurate EMG measurement, but application has its safety risks and is relatively expensive.

3.5.2. Gel electrodes

Electrodes of silver/silver chloride (Ag/AgCl) are most commonly used for EMG measurement in clinical settings [7, 90]. A conductive gel required to ensure good electrical contact, the gel is applied between the skin and the electrode [39]. Gel electrodes are simple, reliable, light weight and cost effective [7]. A disadvantage of the gel electrodes comes with long term measurements. The conductive electrolyte gel used with Ag/AgCl electrodes dries over time, which deteriorates the performance with long term measurements [90]. Long term use of gel electrodes may cause skin irritation [7, 83, 90]. Gel electrodes are a reliable and cost effective solution for EMG measurements, but have drawbacks in long term measurements.

3.5.3. Dry electrodes

Dry electrodes are metal electrodes directly placed on the skin. No conductive substance is required to ensure its functioning [90]. Dry electrodes are reusable and allow for multiple EMG recordings without replacement [90]. EMG measurements with dry electrodes are prone to noise, unstable and depend on good skin contact [20]. Dry electrodes are considered as a solution for the use of EMG measurements in myoelectric prostheses [20]. Dry electrodes are easy to apply, multiple use and do not harm the skin, but have disadvantages, such as being prone to noise.

3.5.4. Myo armband

The myo armband contains multiple dry electrodes, is able to detect multiple hand movements and has been implemented in previously developed prostheses [47, 84, 94]. The costs of a myo armband are high (200 USD) [85], compared to a single dry electrode (37.50 USD) [37]. For the control of a prosthesis with multiple DoF, the myo armband is a solution. The prosthesis developed by Sanchez-Velasco et al. had eight different grasp types [94]. The myo armband was able to identify eight different hand movements to control the different grasps. The myo armband is able to detect multiple hand movements [84], but the disadvantage is the relatively high cost.

3.5.5. Solution selection

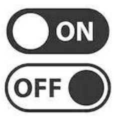
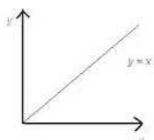
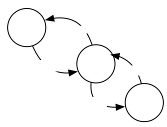
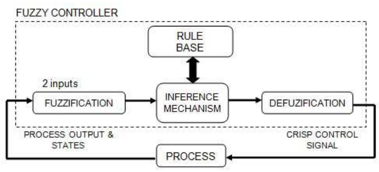
The use of invasive electrodes is complex, expensive and has safety risks. The invasive electrodes are abandoned as solution for EMG measurement, they do not meet the requirements. The gel electrodes have desirable properties, such as easy use and cost effectiveness. However, the electrode use in prosthetic application involves long term measurements, for which gel electrodes are less suitable. The skin health risks of long term measurements do not match the requirements set for EMG measurement.

The dry electrodes are non-invasive, non-irritating and applicable for multiple use. Since the 100 dollar hand has one DoF, the added functionality of the myo armband, which allows for detection of multiple grasps, is not needed. The more cost effective solution of single dry electrodes is selected as the EMG measurement method for the prototype.

3.6. EMG analysis

The EMG signal measured with the electrodes has to be processed and analyzed to detect muscle activation of the user. A number of commonly used EMG analysis algorithms has been found in literature: on/off, proportional, Finite State Machine (FSM), Fuzzy Logic Control (FLC) and PR [36, 48, 75]. Most PR algorithms lack sufficient reliability outside laboratory conditions [53, 59]. Commercial PR control systems are available (FDA approved), but those are expensive (around 15.000 USD) [20]. Therefore, PR algorithms will not be considered in this study.

Table 3.5: EMG analysis design solutions.

<table>
<tr><th>Function</th><th colspan="4">Solutions</th></tr>
<tr><td>EMG analysis</td><td>On/off
</td><td>Proportional
</td><td>FSM
</td><td>FLC
</td></tr>
</table>

3.6.1. On/off control

On/off control is the most basic control method. One DoF can be controlled. Depending on the input, the hand of the prosthesis will either open or close.

3.6.2. Proportional control

Proportional control is comparable to on/off control, but offers more control over the controllable variable, e.g. the hand closing speed. The amplitude of the EMG signal or the muscle activation time can be used for proportional control. The height of the amplitude can be coupled to the closing speed.

3.6.3. Finite state machine control

In FSM control, the postures of the hand are defined as states [48]. Transition between states is bound to fixed rules and depends on the input. FSM allows for control over multiple DoF.

3.6.4. Fuzzy logic control

In FLC systems complex input data is simplified, this process is called fuzzification [63]. A crisp input value is converted into a fuzzy value. The fuzzy value is evaluated against a set of rules, and a decision is made for each rule. After all the rules have been evaluated, the fuzzy value is converted into a crisp value again, called defuzzification. The output crisp value can be applied to the system [63]. FLC systems are able to deal with noisy input data [49]. This property makes FLC useful for EMG applications, since EMG input data can be noisy [20]. FLC allows, like FSM, for control over multiple DoF.

3.6.5. Solution selection

The 100 dollar hand has one degree of freedom. A simple, but reliable EMG analysis method is needed. FSM and FLC offer multiple degrees of freedom but, since this is not required, a more simple solution is preferred. Therefore, FSM and FLC are not selected.

On/off and proportional control offer control over one DoF. To maximize the control over the prosthesis, the proportional control is selected as control strategy for the prototype.

3.7. Power source

The power source for the actuation depends on the design choice of the actuation type. An electric motor will require a battery, but e.g. pneumatic actuation will require compressed gas.

An electric power source is required to supply the control unit. All MCU's proposed in this study require a power supply of at least 5 V. MCU's have a low energy consumption, therefore, the battery used to power just the PCB may have a low capacity.

Both Lithium polymer (LiPo) and Nickelmetal hydride (NiMH) batteries have been proposed as electric power source. LiPo batteries have a high energy density [58], thus they are relatively low weight. LiPo batteries are more sensitive and unstable [30, 58], over-discharging, overcharging, over temperature and mechanical damage may cause safety hazards, such as fire, explosion, heat and smoke [104].

NiMH batteries have a lower energy density compared to LiPo batteries [104]. NiMH batteries are more stable and are easier to handle. The safety hazards of NiMH batteries as a result of mis-handling include generation of heat, released of hydrogen gas, water vapor, and corrosive electrolyte aerosols [65].

Table 3.6: Power source design solutions.

Function	Solutions		
Power source	Compressed gas 	LiPo battery 	NiMH battery 

3.7.1. Solution selection

Since electric actuation is selected, a battery will be used to power the prototype. LiPo batteries have extensive safety hazards, mis-handling of the battery may cause fire or explosion [30, 58, 104]. NiMH batteries have less safety hazards compared to LiPo batteries [79]. This is the first prototype and the functioning of the prosthesis still has to be proved, safety has to be assured. Although LiPo batteries have favourable properties, safety is valued as valued as more important. Therefore, the NiMH battery is selected as power source for the prosthesis.

3.8. Summary

For each function of the prosthesis solutions have been proposed and selected. The selected solutions for each function is summarized in the following list.

- Prosthetic hand: 100 dollar hand
- Actuation: Servo motor
- Control system: Digispark
- EMG measurement: Dry electrodes
- EMG analysis: Proportional control
- Power source: NiMH battery

4

Prototype

4.1. Prosthetic hand

The 100 dollar hand was originally designed to function as a body powered prosthesis (Figure 4.1a). The design was revised for externally actuated use. The spring mechanism that was implemented for the body powered prosthesis was removed (Figure 4.1b). The actuation mechanism and MCU had to be integrated inside the hand. Appendix A.9 describes the details regarding the revision of the original 100 dollar hand. Printing details of the developed hand are added to Appendix A.10. To improve the grip of the prosthesis, anti-slip material has been attached to the fingers. Cricket bat grip strips have been used for this purpose (Figure 4.1c).

4.1.1. Cosmetic glove

Covering the hand prosthesis with a glove to improve cosmetic appearance is common with commercial myoelectric prostheses [11]. The focus of this study is to develop a prosthesis as simple as possible. The cosmetic glove is relatively heavy, around 90 g [100]. As mentioned by Smit et al. (2012), the cosmetic glove significantly adds to the weight of the prosthesis [99]. This is almost one fourth of the design requirement set for the weight.

The cosmetic appearance of a prosthesis is highly valued by the user [88, 95, 113]. A glove highly adds to realistic appearance of a prosthesis [22]. The glove adds more grip as well. This study will focus exclusively on the functionality, therefore the cosmetic glove will not be used. The 100 dollar hand itself has the shape of a human hand. Added cosmetic appearance can be achieved using 3D printing resin colour approximately resembling the users skin colour.

4.2. Actuation

A servo motor with an output torque of $24\ kg{\cdot}cm$ is required to provide a pinch force of 30 N (elaborated on in Appendix A.5). Initial tests have been performed with a Goteck HC1621S digital servo, with a torque of $19\ kg \cdot cm$. The tests confirmed proof of concept, but stalling the servo motor resulted in a loud noise. Efforts have been made to obtain feedback about the actual position of the servo motor (in case of stalling) to solve the noise problem, but this did not result in a satisfying outcome.

For the prototype, a continuous rotation servo motor has been selected. Continuous rotation servo motors control the rotation speed, instead of the servo position. The position can still be influenced by the controlling the time of movement and the speed. Stalling of a continuous rotation servo motor does not produce noise. For the prototype the TD-8130MG Waterproof continuous rotation digital servo has been selected with $30\ kg \cdot cm$ torque. The costs of this servo motor are 14.50 euro [106]. Specifications of the servo motor are added to Appendix A.12. The servo motor has five different rotational speed modes, which differ in output torque.

4.2.1. Transmission

The closing mechanism of the prosthetic hand depends on a small trajectory of connected links within the hand. This trajectory can be bridged by the servo. The servo arm is be connected to the linkage

(a) (b) (c)

Figure 4.1: (a) The original 100 dollar hand. (b) The revised 100 dollar hand, (c) Realization of the prototype.

mechanism (Figure 4.1b). Therefore, no additional transmission is required.

4.3. Control system

The Digispark ATtiny85 was selected for controlling the prototype. However, the MCU was not compatible with the Arduino library required for the filtering of the EMG signal. Therefore, the slightly larger board (26.7 x 18.3 mm), the Digispark Pro was used for the prototype. The Digispark Pro is compatible with the Arduino library for EMG filtering. The Digispark Pro was bought at a cost of 6.50 euro [105]. The electrical scheme of the system is attached in Appendix A.8. Appendix A.13 contains a detailed description of the connection of the Digispark Pro with the EMG sensor, the servo and the power source.

4.4. EMG measurement

The DFRobot EMG sensor by Oymotion is used in the prototype [37]. The EMG sensor costs 37.50 USD (Figure 4.2). Specifications of the EMG sensor can be found in the Appendix A.11. The sensor was found to be suitable for the purpose of prosthetic control [20].

Figure 4.2: The DFRobot EMG sensor by Oymotion.

4.4.1. Muscle activation and EMG sensor placement

The muscles that are used for control have to be close to the skin surface, to prevent crosstalk. Radial wrist deviation is a movement of the wrist that does not interfere with the use of upper and lower arm (Figure 4.3). Radial wrist deviation is generated by the extensor carpi radialis longus [98]. The extensor carpi radialis longus is known for myoelectric prosthesis control in previous research [6, 45, 53, 107]. Figure 4.4a visualizes the extensor carpi radialis.

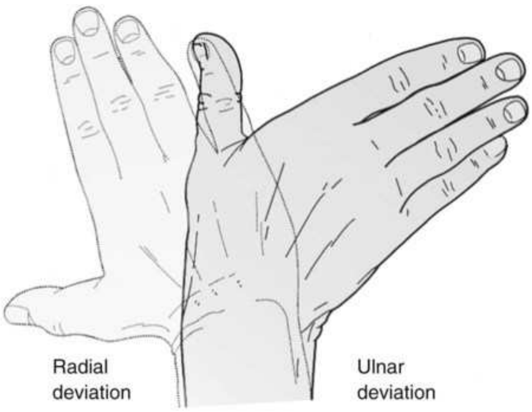
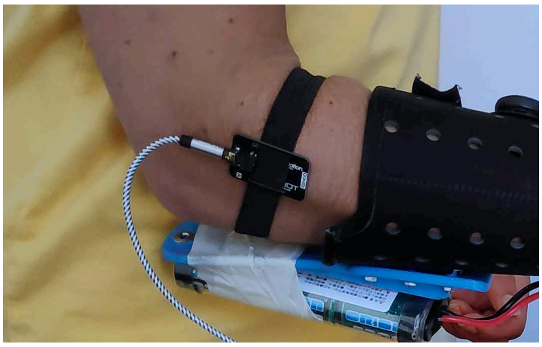


Figure 4.3: Radial wrist deviation, the movement used control of the prosthesis.

The activation of the extensor carpi radialis longus is measured with an EMG sensor at the elbow. The EMG measurement is most optimal at the belly of the muscle. The placement of the EMG sensor is shown in Figure 4.4b.

(a)



(b)

Figure 4.4: (a) Visualisation of the extensor carpi radialis longus, the muscle used for EMG control. (b) EMG sensor position at the extensor carpi radialis longus of a participant.

4.5. EMG analysis

Proportional control was initially selected for implementation in this study. However, the EMG signal measured with the Oymotion dry electrodes was not stable enough to control the servo reliably with proportional control of the amplitude of the EMG signal. The muscle activation time proved to be a more reliable variable. To ensure reliable and stable control, a threshold was used to differentiate between two different grip speeds, instead of proportional control.

4.5.1. Threshold control

The threshold was set at a muscle activation time of 150 ms. Muscle activation time of less than 150 ms triggers the fast grasp, muscle activation time of more than 150 ms triggers the slow grasp. The slow mode, with a pinch force of 35 N, where the closing of the hand takes 1.3 s and a fast mode, with a pinch force of 55 N, where closing of the hand takes 0.5 s. The slow mode can be used for precision tasks and the handling of fragile objects. The fast mode allows the user to swiftly grasp an object and can be used when a strong grasp is required. The Arduino code for the control of the prosthesis is included in Appendix A.1.

4.5.2. Signal processing

The EMG sensor detects muscle activity in a range of -1.5 mV and +1.5 mV (Appendix A.13). The measured signal is sent to the amplification circuit of the Oymotion EMG sensor. The signal is rectified and amplified a 1000 times. The output signal is analog and has a voltage range between 0 and +3.0 V. The sensor is connected to the Arduino compatible MCU, the Digispark Pro. Oymotion (developer of the dry electrode) provided a Arduino library to further process the signal from the sensor [76]. An anti-hum notch filter eliminates 50 Hz power line noise, a low pass filter is used to filter noises above 150 Hz and a high pass filter filters noises below 20 Hz. The signal is sampled at a frequency of 500 Hz. The filter settings are a common configuration and are integrated in a set of simple functions (part of the EMG Filters library of Oymotion), which was recommended by Oymotion [76].

4.6. Power source

The energy consumption of the TD-8130MG servo is 200 mA in neutral position (Appendix A.12). The maximum stalling current of the servo is 3400 mA. The opening and closing movement of the hand prosthesis takes 500 ms. The daily usage of a prosthesis hand is based on the findings of Luchetti et al. [68]. Luchetti et al. (2015) reported an estimated median usage of 130.000 cycles annually, which corresponds to approximately 360 cycles a day. The 360 cycles take six minutes, based on the 1 s duration of one cycle (0.5 s opening time and 0.5 s closing time). 340 mAh of capacity is consumed in these six minutes and 200 mAh is consumed every hour in resting state. 2340 mAh is theoretically consumed with average usage of the servo motor, wearing the prosthesis for ten hours. The Digispark Pro has a very low power consumption, compared to the servo. The Digispark Pro consumes 22 mAh in resting state. A minimum capacity of 3000 mAh will offer a safety buffer for an intense day of prosthesis usage, without switching batteries. A 7.2 V NiMH battery pack with a capacity of 3300 mAh is implemented in the prototype. This is comparable to the low cost upper limb prosthesis Federica battery [40].

5

Evaluation

5.1. Overview

The prototype will be evaluated to check whether the main goal and the design requirements are met. The prototype is named the 'MyoGrab Hand'. 'Myo' refers to its control method for the user, myoelectric, and 'Grab' to the one DoF functioning of the prosthesis that allows to grabbing objects. The functionality of the MyoGrab Hand will be evaluated using standardized and validated tests. The Box and Block Test (BBT) and the Southampton Hand Assessment Procedure (SHAP) are proposed to make conclusions about its functionality and to compare the currently developed prosthesis to commercial myoelectric prostheses [67, 71].

5.2. Design requirements evaluation

5.2.1. Costs

The material costs will be evaluated. The labor and used equipment, such as a 3D printer, will not be taken into account. The costs of the 3D printed parts are calculated in terms of the used polylactic acid (PLA) quantity.

5.2.2. Weight

The prosthesis weight will be measured to evaluate if the design requirement of a maximum weight of 400 g has been met. The battery will not be included in the measurement. A Mettler PM4800 Delta Range scale will be used for this purpose.

5.2.3. Pinch force

The pinch force of the MyoGrab Hand will be evaluated. The requirement was set at 30 N. Once the prosthesis hand is closed, the servo stops. The servo has five different levels of speed. A higher speed results in a higher pinch force. The pinch force at all levels of speed will be measured. For this purpose, a Farnell FX19 compression load cell [43] will be used. The sensor will be positioned at the tip of the index and middle finger. When the hand closes, the thumb pushes the force contact point of the sensor. A visualisation of the experiment setup is shown in Figure 5.1. The Arduino code used in this experiment is attached in Appendix A.4.

When the prosthesis hand is closed, counter forces can be applied to force the prosthesis hand to open. The counter force needed to open the fingers of the hand will be measured with a pull force sensor. The cable of the pull force sensor will be connected to the fingers, perpendicular to the joint to create a maximum momentum. The minimum amount of force to open the hand will be measured. Assuming that the static friction is higher than the dynamic friction in the prosthesis, the highest measured force will be considered to be the force required to open the prosthesis hand. Figure 5.2 shows the setup with the pull force cable attached to the top of the thumb.

Figure 5.1: Pinch force experiment setup.

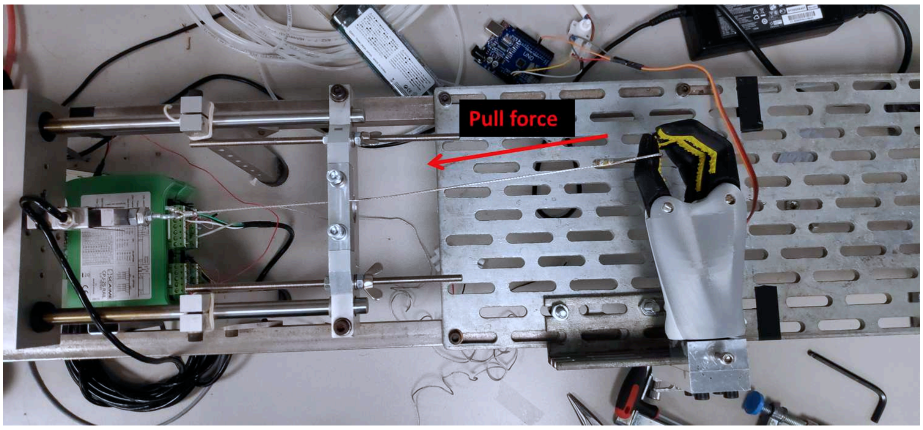


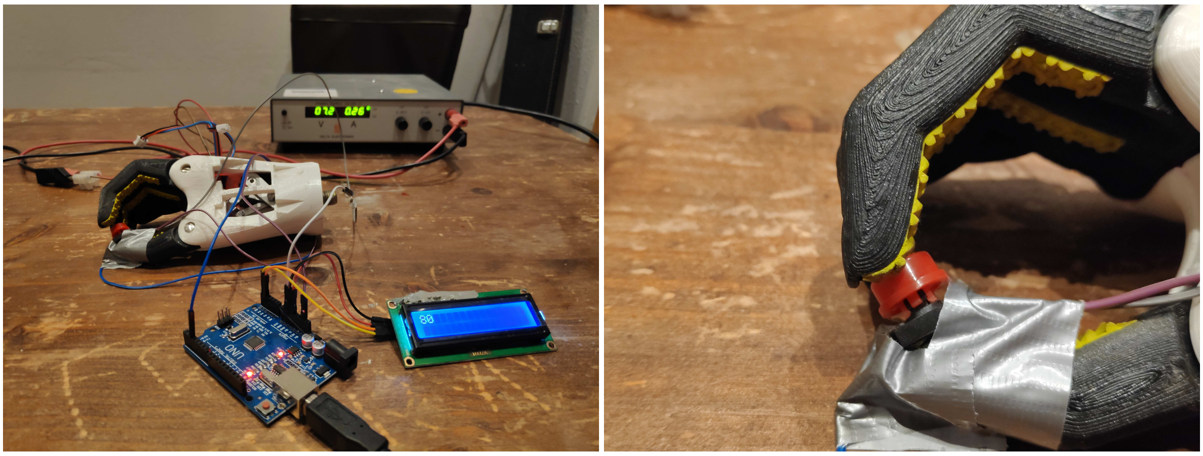
Figure 5.2: The experiment setup for measuring the pull force causing opening of the developed prosthesis hand.

5.2.4. Durability experiment

Durability of the prosthetic hand is one of the design requirements. The design requirement was set a minimum of 130.000 cycles. The durability of the developed prosthesis will be tested with an experiment. The number of cycles of closing and opening till prosthesis failure will be tested. The MyoGrab Hand will be connected to a 7.2 V power source and the prosthesis will be programmed to open and close every two seconds. An Arduino Uno MCU with a button and LCD connected will be used to count the number of cycles. Figure 5.3a shows the experiment setup. The button is attached to the thumb of the prosthesis (Figure 5.3b). When the hand closes, the button is pushed and the cycle is registered and visualized on the LCD. The Arduino code of the prosthesis (opening and closing the hand) and the code of the Arduino Uno (registering the number of cycles) are attached to Appendix A.4.

5.3. Functional evaluation

The prosthesis will be tested with the use of a simulator, which enables non-amputee participants to test the prosthesis. A simulator has been used in previously conducted research to test and evaluate upper limb prostheses [52, 111]. The commercially available TRS prosthetic simulator [108] was used in this study. The cable liner (used for body powered prostheses) was removed from the simulator. Figure 5.4 shows a participant wearing the simulator. A total of 20 participants (13 males and seven females, average age 23.6 ± 1.9, age range 20-27) were recruited to participate in the experiment. All participants had their right hand as dominant hand.



(a) (b)

Figure 5.3: (a) Overview of the durability experiment setup. (b) Close-up of the button positioned at the top of the thumb of the MyoGrab Hand.

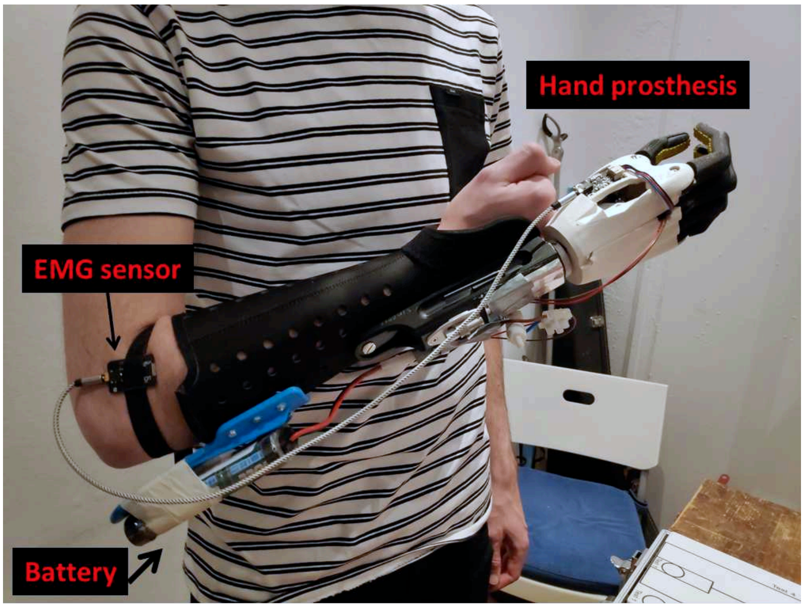


Figure 5.4: A participant wearing the simulator with the hand prosthesis attached.

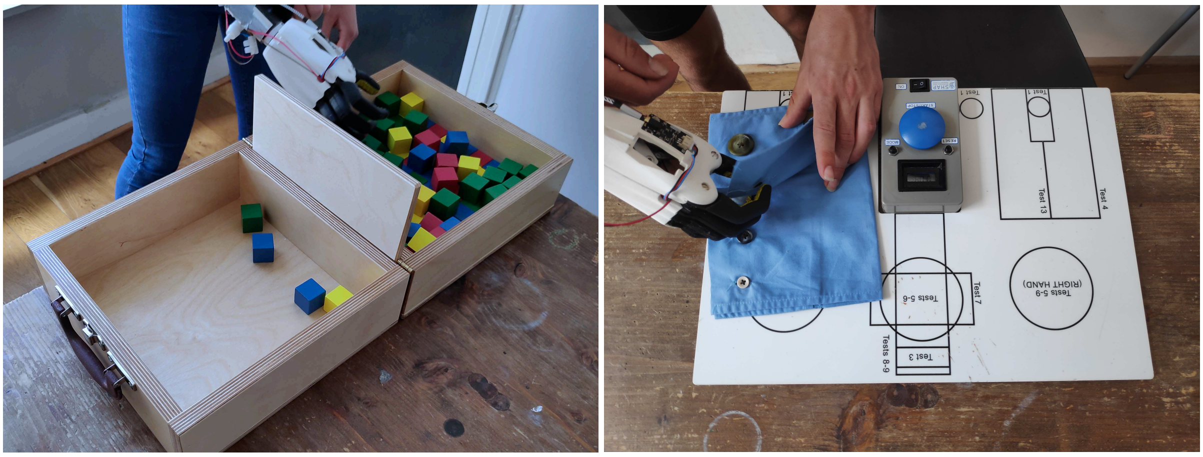
5.3.1. Box and Blocks Test

The BBT is a test to measure gross manual dexterity [71]. In the BBT, a participant has to move wooden square cubes from a box to an adjacent box. The boxes are separated by a small facade. The number of blocks transported within a minute is scored. The BBT is visualized in Figure 5.5a.

The BBT is commonly used to evaluate upper limb prosthesis functioning [29, 52]. Possible shortcomings or advantages can be identified. The developed prosthesis can be compared to market equivalents. Also may the outcome of the BBT be used to compare future research prostheses to the current prototype.

5.3.2. Southampton Hand Assessment Procedure

The SHAP is a clinically validated hand function test [67]. This test is suitable to evaluate upper limb prostheses, as shown in previous research [66]. The test consists of the grasping and movement of six abstract objects and 14 ADL tasks, e.g. opening and closing a zip and filling a glass with water (Figure 5.5b). Each task is timed by the participant. Commercial myoelectric prostheses are evaluated and compared using the SHAP [74], as well as low cost, externally powered prostheses [114, 115]. The



(a) (b)

Figure 5.5: (a) Participant carrying out the BBT. (b) Participant carrying out the button board task of the SHAP.

SHAP was used to evaluate the prototype and the results will be compared with other prostheses that have SHAP performance data available, such as the DMC Plus Otto Bock hand prosthesis [74]. Each SHAP task is classified with one of six hand grasps. The SHAP result gives a score for each hand grasp and an Index of Functionality (IoF), the IoF provides an overall assessment of hand functioning. The IoF is scored between 0 and 100. A score of 0 is very impaired hand functioning and a score between 95-100 indicates normal functioning of the hand, usually achieved by healthy non-amputee participants [66].

5.3.3. First evaluation

In the first evaluation ten participants (six males and four females, average age 23 ± 1.3, age range 20-25) performed the BBT (three trials) and the SHAP (one trial). The participants had to sign the informed consent form (added to Appendix A.15), before participation. The participants were fitted with the simulator, with attached the prosthesis, and the EMG sensor. The battery was taped to the simulator, close to the elbow. The participants practiced with the prosthesis before the experiment, to make sure the EMG sensor was positioned correctly and to get used to the feeling of the simulator. After practice, the BBT was performed three times, followed by carrying out the SHAP. According to the protocol, both tests have to be carried out from a sitting position. However, due to the lack of wrist mobility and extended length of the arm because of the simulator, the participants were allowed to carry out the tests in a for them comfortable position, standing was allowed.

5.3.4. Optimization

After the first evaluation, the Arduino code running the MyoGrab Hand was optimized, resulting in a faster response of the actuator to muscle activation. The Arduino code of the optimized prosthesis is attached to Appendix A.2.

5.3.5. Second evaluation

To evaluate the effects of the improvements of the MyoGrab hand, the prosthesis has been tested with the BBT by ten participants (seven males and three females, average age 24.1 ± 2.2, age range 20-27). The participants have their right hand as their dominant hand and did not participate in the first evaluation. The effect of the improved reaction time of the MyoGrab Hand is best measured with the BBT. Therefore, only the BBT was used in the second evaluation. Ten participants signed the informed consent for the second evaluation (Appendix A.16). The exact same experiment protocol was used for the second evaluation as for the first evaluation. An independent samples t-test is used to compare the results of the first evaluation and the second evaluation.

6
Results

6.1. Design requirements

6.1.1. Costs

The material costs of the developed prototype are summarized in Table 6.1. The total of material costs were 99.33 euro. The 3D printing costs of the MyoGrab Hand are based on the use of 250 grams of PLA. Labor and equipment costs are not included in the cost calculation.

Table 6.1: The material costs of the prototype

Part	Costs in euro
3D printed parts	13.00
TD-8130MG servo	14.50
Digispark Pro	6.50
EMG sensor by Oymotion	33.00
7.2 V NiMH 3300 mAh battery	23.90
Jumper wires	0.44
Servo arm	7.99
Total	99.33

6.1.2. Weight

The weight of the MyoGrab Hand is 352 g. A visualisation of the measurement is shown in Figure 6.1. The battery is not included in this measurement. The weight of the NiMH battery, used during testing, is 352 g.

6.1.3. Pinch force

The pinch force has been measured at five different levels of servo speed. The highest pinch force is 54.8 N. The lowest pinch force is 11.7 N. The measured forces are shown in Table 6.2. The pinch force was measured ten times for each servo speed. The stated forces are the averages over ten measurements.

The pull force resulting in opening of the hand was measured for both the thumb and the fingers separately with the servo motor in active, neutral state (no movement). A pull force of 41 N at the top of the thumb resulted in opening of the hand. For the fingers the pull force cable was attached to the top of the middle finger. A pull force of 11.5 N resulted in opening of the hand.

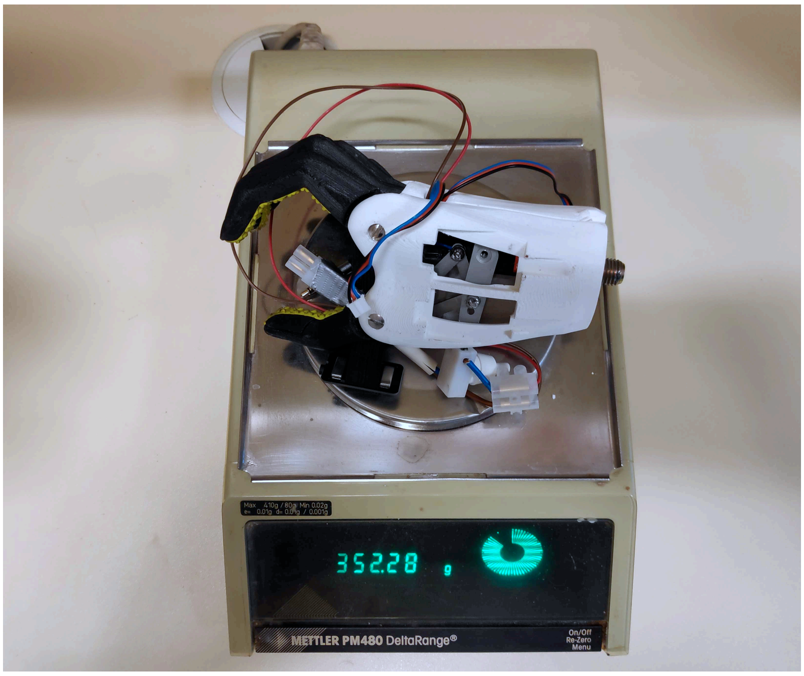


Figure 6.1: Weight of the prosthesis and EMG sensor (excluding the battery).

Table 6.2: Pinch force of the servo at the five available speeds.

Servo speed level	Pinch force [N]
1 (slowest)	11.7 (± 1.8)
2	34.8 (± 2.2)
3	48.3 (± 1.0)
4	53.3 (± 0.4)
5 (fastest)	54.8 (± 0.6)

6.2. First evaluation

6.2.1. Box and Blocks Test

The scores of the participants on the BBT are visualized in Figure 6.2 and 6.3. The average score on the third trial was 17.0 (± 2.2) blocks. The highest score was 21 blocks on the third trial. The lowest score on the third trial was 13 blocks. The scores of all participants on all trials are added to Appendix A.17.

6.2.2. Southampton Hand Assessment Procedure

The average IoF score of the SHAP is 41 (± 8.3). The highest IoF score was 54, the lowest score was an IoF of 24. Figure 6.4 shows the distribution of the SHAP IoF scores of all participants.

6.3. Second evaluation

6.3.1. Box and Blocks Test

The scores of the participants on the BBT are visualized in Figure 6.3 and Figure 6.5 The average score on the third trial was 18.4 (± 2.4) blocks. For participant number nine, the score on the first trial is used for the average calculation, because in the second and third trial problems with the EMG measurement were observed which heavily affected the control of the prosthesis. The scores of all participants on all trials are added to Appendix A.17. The highest score was 24 blocks on the third trial.

The result of the BBT in the second evaluation was not significantly better than in the first evaluation (t = -1.3, df = 18, p = 0.21).

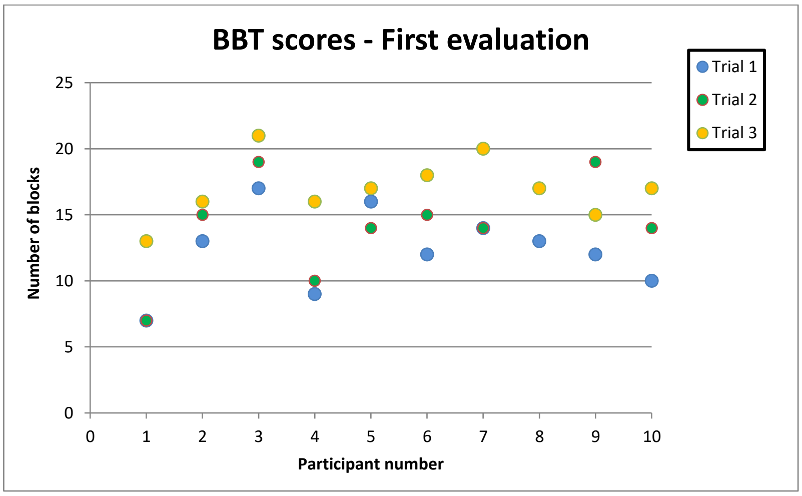
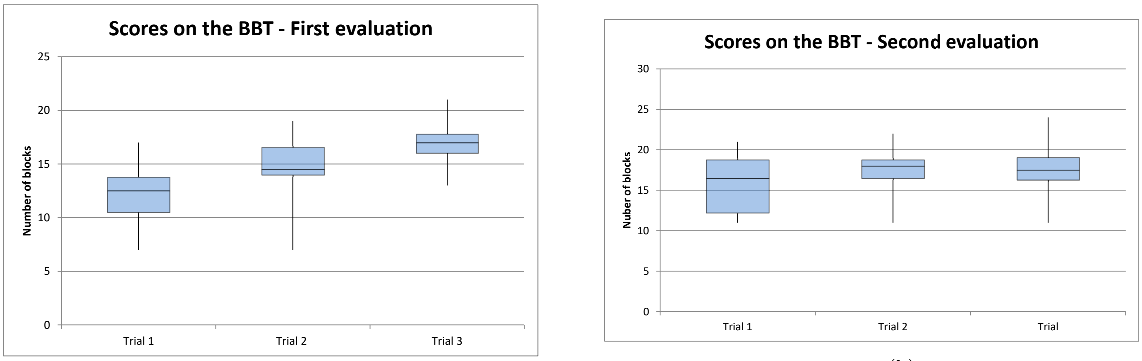


Figure 6.2: Participant scores on three trials of the BBT. Average score trial 1: 12.3 (± 2.9), average score trial 2: 14.4 (± 3.5), average score trial 3: 17.0 (± 2.2).



(a)

(b)

Figure 6.3: Result per BBT trial in a boxplot, (a) the first evaluation and (b) the second evaluation.

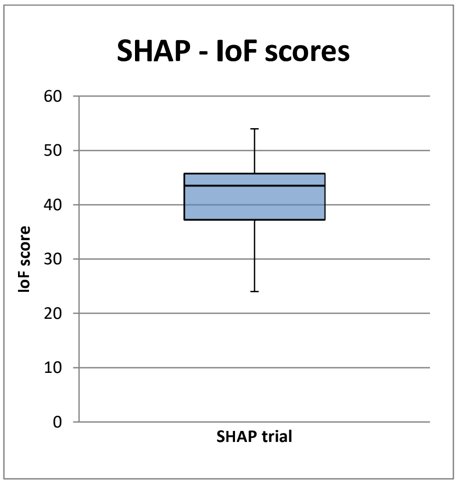


Figure 6.4: Visualisation of the distribution of the SHAP scores.

6.4. Durability experiment

The prosthesis was functioning for 16539 cycles until critical failure of the servo motor. The servo motor was the failing factor in the prosthesis. The experiment was paused a number of times to solve non-critical issues, as elaborated on in Appendix A.14.

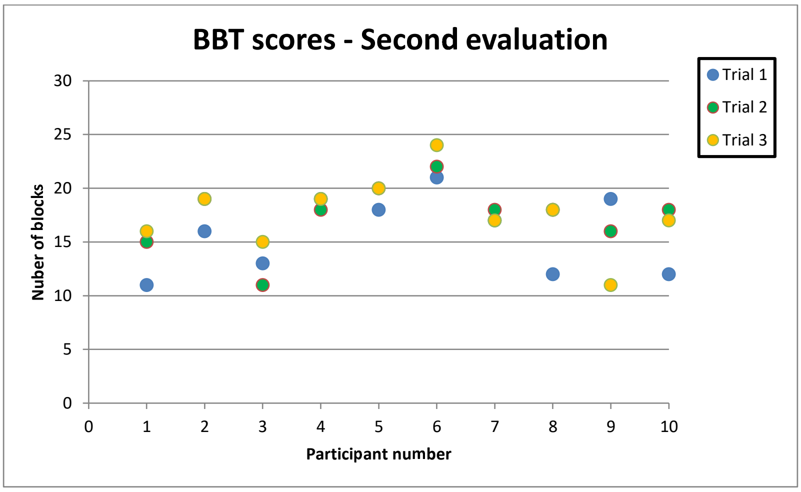


Figure 6.5: Participant scores on three trials of the BBT. Average score trial 1: 15.8 (± 3.4), average score trial 2: 17.5 (± 2.8), average score trial 3: 17.6 (± 3.2).

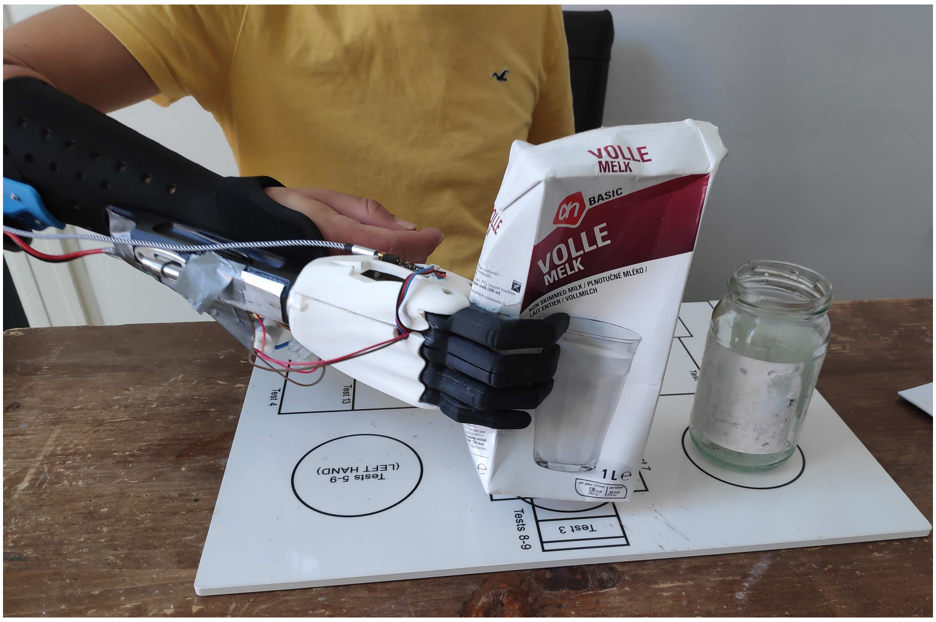


Figure 6.6: Participant completing the carton pouring task of the SHAP.

7 Discussion

7.1. Evaluation of design requirements

The goal of this study was to develop functional low cost myoelectric hand prosthesis. In the prototype a number of design requirements have been met. Several design requirements needed testing to evaluate if they have been met. The costs, weight and results of the mechanical testing will be discussed. The test results of the functionality test will be evaluated whether functionality has been proven.

7.1.1. Costs

The costs of the prosthesis were 99.33 euro. The cost of the prosthesis is well within the maximum costs of 180 euro. The described costs are exclusively the material costs. The material costs give insight in only a part of the total costs of providing a prosthesis to an amputee. Additional costs, such as equipment, labor, certification of the prosthesis, the prosthesis socket and custom fitting, have to be taken into account in order to get a clear perspective on the total costs of providing prostheses. Although commercialization will probably result in higher costs, this is a first step on the road to lower cost myoelectric prostheses.

7.1.2. Weight

The weight of the MyoGrab Hand, excluding the battery, is 352 g. The design requirement of a maximum weight of 400 g has been met. The weight includes all the components of the prosthesis except for the battery. The battery weight is 352 g. The weight of the TRS simulator is 511 g [108]. The weight of the total system (MyoGrab Hand, the battery and the simulator) is 1215 g.

7.1.3. Pinch force

The pinch force design requirement was set at a minimum of 30 N. A servo with a torque of $24\ kg \cdot cm$ was estimated to be able to achieve this (Appendix A.5). The TD-8130MG servo has a torque of $30\ kg \cdot cm$ and resulted in a maximum pinch force of 54.8 N at maximum speed of the servo. The design requirement for the pinch force has been met. With the lowest speed mode of the servo a pinch force of 11.7 N was achieved. The slow speed mode allows the user to grasp more fragile objects without the risk of crushing the object.

7.1.4. Durability

The prosthesis lasted for 16539 cycles until critical failure occurred. The design requirement of 130.000 cycles has not been met. The servo motor was the component that broke down, which caused discontinuation of the durability experiment. The EMG sensor and battery were not included in the experiment setup for testing. Therefore, no conclusions can be drawn regarding the durability of the EMG sensor and the battery.

7.1.5. Functionality

The Box and Blocks Test was carried out three times by each participant. A learning curve is observed for most participants. The average score on the BBT was 17.0 in the first evaluation and 18.4 in the second evaluation. The optimized MyoGrab Hand did result in a higher average score, however, the improvement is not significant. A larger sample size of the experiment is required to drawn reliable conclusions regarding the improvement of the optimization.

Two tasks of the SHAP could not be completed with the MyoGrab Hand. Grasping of the heavy sphere with the SHAP was impossible with the prosthesis. The combination of its heavy weight and the polished surface made the prosthesis slip while grasping it. The third ADL task of the SHAP, simulated food cutting, was not feasible as well. The knife was difficult to grasp and the user could not apply enough force to cut the food. The rest of the grasping tasks and ADL tasks were completed by most of the participants.

7.2. Comparison with commercial prostheses

7.2.1. Cost

Table 7.1 summarizes the properties of a number of prostheses that are compared. Commercial myoelectric prostheses are generally costly. The price of the MyoGrab Hand is low compared to the commercial prostheses, 109.34 USD (99.33 euro). The Hero arm is the closest in price range with a price of 6600 USD [20]. As discussed previously, the costs described of the developed prosthesis include only the costs of material, the material costs are only a part of the total cost of a commercial prosthesis. Open source availability of the prosthesis and locally manufacturing has the potential of minimizing the costs.

7.2.2. Weight

The weight of the MyoGrab Hand is lower compared to the commercially available prostheses. The prosthesis weight is comparable to the DMC Plus hand. The weight of the cosmetic glove is not included in the weight calculation of the DMC Plus hand. The developed prosthesis does not require a cosmetic glove for a hand-shaped appearance, although the appearance of the skin with a glove is more realistic.

The Hero Arm, which is 3D printed as well, has a lower weight compared to the MyoGrab Hand with a weight ranging from 280 to 346 g [20]. The publication reporting the weight of the Hero Arm is unclear what components (such as battery, actuation) are included in the weight calculation [20]. The Hero Arm has, as well as the MyoGrab Hand, no cosmetic glove. The Hero Arm aimed at a more futuristic design of the hand.

The battery was excluded in the weight calculation for the developed prosthesis and the battery of the DMC Plus hand is included in the weight calculation. The battery has the same weight as the prosthesis, the summed weight is around 700 g. Since the battery is a crucial component of the prosthesis, the used NiMH battery is not feasible for actual functional use. The MyoGrab Hand is far from comparable with commercial prostheses, when it comes to the weight with the battery included.

Table 7.1: Overview of properties and performance results on the BBT and the SHAP (IoF) of the developed prosthesis and commercial myoelectric prostheses [16, 20, 74, 109].

	MyoGrab Hand	DMC Plus	Michelangelo	i-Limb	Hero arm
Cost [USD]	109.34	?	60.000	33.000	6600
Weight [g]	352	355	498	479	280-346
Max grip force [N]	54.8	121	70	48	?
Number of grips	1	1	7	11	4-6
BBT score	17	?	29	?	?
IoF	41	74	83	52	?

Unpublished data is referred to with a '?'.

7.2.3. Grip force

The maximum achieved pinch force of the developed myoelectric hand of 54.8 N. The commercial i-Limb has a maximum achievable grip force of 48 N [74]. Compared to the i-Limb the developed prosthesis has better grip force. The design of the MyoGrab Hand is more similar to the DMC Plus hand. The DMC Plus hand outperforms the prototype, able to deliver more than twice the pinch force of the prototype. The MyoGrab Hand cannot compete with its commercial equivalent when it comes to pinch force. The limited grip force of the MyoGrab Hand may result in limitations of use of the hand for daily functioning, e.g. grasping heavy objects. The effect of the difference in grip force on ADL tasks and functionality will be further discussed in the following section.

7.2.4. Functionality

As concluded in the literature review almost none of the reviewed low cost myoelectric prostheses was tested with validated and standardized test methods. This study is one of the first studies developing a low cost myoelectric prosthesis and proving its functionality for ADL tasks. A score of 18.4 (± 2.2) on the BBT and an IoF of 41 (± 8.3) were achieved. Van der Niet et al. found an IoF of 52 for the i-Limb [74]. The highest IoF score obtained with the MyoGrab Hand was 54, which is better than the i-Limb. The developed low cost prosthesis has similar functional capabilities as a commercial prosthesis. The i-Limb is capable of multiple grasp types. The added functionality of multiple grasps can be questioned, when the grip force is similar, since the IoF score is comparable with the developed prosthesis. The Michelangelo hand has multiple grips and a higher grip force, resulting in the highest functionality score on the SHAP compared to the other prostheses.

The DMC Plus hand has an IoF of 74, which is way better than the MyoGrab Hand. The working principle of the DMC Plus hand and the developed prosthesis are comparable. A part of the difference in functionality is likely to be attributed to the difference in grip force. The grip force of the MyoGrab Hand might be a limiting factor for the functionality of the prosthesis.

The functionality tests have been carried out with able-body participants with the use of a simulator. The participants did not have prior experience using EMG or a prosthesis. For functional testing of the commercial prostheses, amputee participants were used. They have experience with both EMG and prosthesis use and might score relatively better on the functionality tests. A part of the difference in scores on the BBT and the SHAP may be explainable because of the difference in experience of participants.

Only the Michelangelo arm was evaluated with the BBT, with a score of 29. The MyoGrab Hand had an average score of 18.4 with the BBT in the second evaluation. The score on the BBT is heavily influenced by the speed of the prosthesis. It can be concluded that grasping objects with the MyoGrab Hand is less fast than with the Michelangelo hand. Although the BBT is a suitable test and evaluation method for prostheses, no BBT test data was available of the commercial prostheses selected for comparison.

7.3. Limitations and strengths

7.3.1. Limitations

The developed prosthesis has only been tested with able-body participants. The test results with able-body participants give proof of concept, but effect of difference in muscle state between able body participants and participants with an amputation has not been researched. Myoelectric prostheses rely on the activation of muscles in what is left of the arm. For participants with an amputation, arm muscle activity might be more difficult to measure due to impaired muscles as a result of dis-use. Differences muscle activity measurements may effect control [35].

To test with non-amputee participants, a simulator was used to connect the MyoGrab Hand. Previously conducted research used a simulator to test and evaluate developed prostheses, and this solution was adopted in this study. However, research endorsing the use of a simulator to simulate prosthesis use of a person with an amputation is limited. It is unclear whether the test results obtained with simulator research can be compared with test results from persons with an amputation.

The MyoGrab Hand is tested with the BBT and the SHAP. Although the SHAP was developed to represent ADL tasks, an IoF score on the SHAP indicating good functionality does not fully guarantee functionality in actual ADL. A test is a simulation of reality. To draw more valid conclusions about the functionality for ADL tasks, the prosthesis has to be tested for a longer period of time with real ADL tasks in actual daily life. Lucetti et al. tested the Michelangelo hand with the SHAP and during a three month trial [68], allowing to draw solid conclusions about functionality.

In this study the tests were conducted in a clinical environment, stable and controlled. The prosthesis has not been tested in a non-clinical setting where the environment is not as stable and external factors may impair the functioning of the prosthesis. Prostheses tend to show decreased functionality in non-clinical environments due to natural variations in EMG patterns and noise-sources [59]. Sweat or dry skin, changing arm positions, small displacements of the electrode and variations in the muscle contraction of the user affect the EMG measurement [59, 60]. EMG measurements are heavily affected by the environment and are sensitive to noise. The functionality of the prosthesis has to be put into perspective, since the functionality outside the clinical setting is unknown [59].

The goal of this study was to develop a low cost prosthesis with the focus on less developed countries. Although the MyoGrab Hand is relatively simple in design and is easy to assemble, the feasibility to manufacture the prosthesis in less developed countries has not been investigated.

7.3.2. Strengths

In several reviews, the lack of testing of (low cost) prostheses was criticized [23, 36], however, this study tackled this problem by using the BBT and SHAP to be able to evaluate functionality. This study wishes to set an example of evaluating functionality with validated and standardized testing for future research of low cost myoelectric prostheses.

Although the costs are low, the design is relatively simple and only one DoF is used, the prosthesis is functional. The majority of participants managed to complete 13 out of the 14 ADL tasks of the SHAP. A relatively high functionality is achieved, given the costs, required resources for manufacturing and number of DoF of the prosthesis.

The use of a simulator allows for testing with non-amputee participants. This study may contributed to a wider interest in the evaluation of prostheses with a simulator. If the availability of participants with an amputation is very limited, the use of a simulator and non-amputee participants, to be able to evaluate functioning of a developed prosthesis, is a solution.

7.4. Implications for future research

In the design of the MyoGrab hand the placement of the EMG amplification circuit is easily accessible but also exposed. For safe and general use of a prosthesis, waterproof functioning is required. The servo is waterproof, but the microprocessor and EMG amplification circuit are not water proof. Future research may focus on designing a solution to protect the circuits from damage.

For practical and reliable application of the MyoGrab Hand, a longer lifetime is required. More research is needed to analyse the factors causing servo motor failure in the prosthesis.

The MyoGrab Hand has been tested in a clinical setting and proved functionality. The prosthesis has to be tested in daily life to evaluate the effect of external noise on the EMG signal stability. Complications in control and safety might arise, that did not came up in the clinical testing environment.

The MyoGrab Hand has been tested with able-body participants and proved itself functional. The actual user is an amputee and the prosthesis should be tested by participants with an amputation to strengthen the proof of functionality.

The battery is a crucial component of the prosthesis currently heavily affecting the weight. In this study a NiMH battery was used to power the prosthesis because of safety precautions. The battery is however the same weight as the prosthesis, the current battery is not feasible for functional and comfortable use. Future research may be able to find a low weight, safe alternative to power the prosthesis.

8 Conclusion

The incentive for this study was the lack of proof of functionality of low cost myoelectric prostheses. In this study a low cost myoelectric prosthesis was developed and the functionality and usability for ADL tasks was tested. The developed prosthesis has a low material cost, 99,33 euro. The prosthesis was tested with standardized and validated test methods, the BBT (average result of 17.0) and the SHAP (average IoF of 41), by a group of ten participants in a first evaluation experiment. The prototype was optimized, an EMG sensor was added for more natural control and the code analyzing the EMG signal analysis was made more time-efficient. The optimized prototype was evaluated with a second experiment by a group of ten participants with the BBT (average result of 18.4). The functionality of the prosthesis was proven with these tests. This is one of the first low cost myoelectric prosthesis that has proof of functionality. The result of the BBT in the second evaluation was not significantly better than in the first evaluation (t = -1.3, df = 18, p = 0.21). In the durability experiment, critical failure of the prototype occurred after 16539 cycles of opening and closing of the hand.

The pinch force of the developed prosthesis (54.8 N) was comparable to the commercial i-Limb (48 N). On functionality the i-Limb (IoF of 52) is slightly better than the MyoGrab Hand (IoF of 41). But compared to the DMC Plus hand (IoF of 74), which is comparable in design and working principle, and the Michelangelo hand (IoF of 83) the developed prosthesis was evaluated as less functional.

Compared to commercial myoelectric prostheses, the developed prosthesis was evaluated as less functional. The functionality achieved by the MyoGrab Hand is relatively good, given the difference in costs.

The battery of the MyoGrab Hand used in this study is too heavy, future research may be able to find a lower weight solution. Making the prosthesis waterproof and testing in a non-clinical environment would be the next step in realization of commercialization of this prosthesis.

This study is a first step on the path of making a functional myoelectric prosthesis available to larger part of the amputee population with limited financial resources. The functionality and usability in a clinical setting was proved in this study. The goal of developing a low cost functional myoelectric prosthesis has been achieved.

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A

Appendix

A.1. Arduino code of initial prototype

```
/*
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   OR TORT (INCLUDING NEGLIGENCE OR OTHERWISE) ARISING IN ANY WAY OUT OF
   THE USE OF THIS SOFTWARE, EVEN IF ADVISED OF THE POSSIBILITY OF SUCH
   DAMAGE.
*/

#if defined(ARDUINO) && ARDUINO >= 100
#include "Arduino.h"
#else
#include "WProgram.h"
#endif
#include "EMGFilters.h"
#include <Servo.h>
#include "DigiKeyboard.h"

// The EMG sensor is connected to (analog) pin 12
#define emgAnInput A12

// a threshold of 1000 is used to detect muscle activation
unsigned long threshold = 1000;

EMGFilters myFilter;

// The sample frequency rate and the notch frequency to filter power line noise (50 Hz
    in the Netherlands)
SAMPLE_FREQUENCY sampleRate = SAMPLE_FREQ_500HZ;
```

```
NOTCH_FREQUENCY humFreq = NOTCH_FREQ_50HZ;

Servo motor;

void setup()
{
  // Initializing the filter function of the EMG library
  myFilter.init(sampleRate, humFreq, true, true, true);
  Serial.begin(115200);

  // The servo motor is connected to (PWM) pin 1 of the Digispark Pro
  motor.attach(1);
  // Assigning the neutral position to the servo
  motor.write(90);
}

void loop()
{
  // vector containing the time of activation, needed for proportional control
  static int Time[] = {0, 0};

  // angle contains the state of the prosthesis hand, assuming an open hand at the start
  static int angle = 0;

  // Reading, filtering and squaring the incoming EMG signal
  int data = analogRead(emgAnInput);
  int dataAfterFilter = myFilter.update(data);
  int envelope = abs(sq(dataAfterFilter));


  // If the EMG signal is lower than the threshold the value will be set to 0
  envelope = (envelope > threshold) ? envelope : 0;


  if (threshold > 0)
  {
    // The getEMGCOunt function analyses the signal for detection of muscle activation
    if (getEMGCount(envelope, Time))
    {
      // If the hand is open, the hand will be closed
      if (angle == 0)
      {
        // updating the state of the hand to closed
        angle = 1;

        // If the muscle activation time is more than 150 ms, the hand will open in slow mode
        if (Time[1] > 149)
        {
          motor.write(99);     // Closing the hand in slow mode
          delay(1300);         // Waiting for the hand to be fully closed
          motor.write(90);     // Assigning neutral position to the servo
        }

        // If the muscle activation time is less than 150 ms, the hand will open in fast mode
        if (Time[1] < 150)
        {
          motor.write(102);    // Closing the hand in fast mode
          delay(500);          // Waiting for the hand to be closed
          motor.write(90);     // Assigning neutral position to the servo
        }

      }
      // If the hand is in closed state, the hand will open
      else
      {
        // Updating status of the prosthesis to open
        angle = 0;
        motor.write(75);     // Opening the hand
```

```
        delay(500);          // Waiting for the hand to fully open
        motor.write(90);     // Assigning neutral position to the servo
      }
    }
  }
  delayMicroseconds(500);
}

// The getEMGCOunt function analyses the EMG signal
int getEMGCount(int gforce_envelope, int Time[])
{
  static long integralData = 0;
  static long integralDataEve = 0;
  static bool remainFlag = false;
  static unsigned long timeMillis = 0;
  static unsigned long timeBeginzero = 0;
  static long fistNum = 0;
  static int  TimeStandard = 400;
  static bool flag = true;

  static unsigned long startTime = 0;
  static unsigned long activationTime = 0;
  /*
    The integral is processed to continuously add the signal value
    and compare the integral value of the previous sampling to determine whether the signal is continuous
  */
  integralDataEve = integralData;
  integralData += gforce_envelope;

  // If the muscle activation starts, a timer is set to measure the activation time
  if (gforce_envelope > 1000)
  {
    startTime = millis();
    if (flag == true)
    {
      flag = false;
      activationTime = startTime;
    }
  }
  /*
    If the integral is constant, and it doesn't equal 0, then the time is recorded;
    If the value of the integral starts to change again, the remainflag is true, and the time record will be re-entered next time
  */
  if ((integralDataEve == integralData) && (integralDataEve != 0))
  {
    timeMillis = millis();
    if (remainFlag)
    {
      timeBeginzero = timeMillis;
      remainFlag = false;
      return 0;
    }
    /* If the integral value exceeds 400 ms, the integral value is clear 0,return that get EMG signal */
    if ((timeMillis - timeBeginzero) > TimeStandard)
    {
      integralDataEve = integralData = 0;
      // The activation time is saved in the vector and send to time main loop
      Time[1] = startTime - activationTime;
      flag = true;
      // return 1 gives the command to open or close the prosthesis depending on its state
      return 1;
    }
    return 0;
  }
  else {
    remainFlag = true;
    return 0;
```

```
    }
}
```

A.2. Arduino code of optimized prototype

```
/*
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   modification, are permitted provided that the following conditions
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   AND ON ANY THEORY OF LIABILITY, WHETHER IN CONTRACT, STRICT LIABILITY,
   OR TORT (INCLUDING NEGLIGENCE OR OTHERWISE) ARISING IN ANY WAY OUT OF
   THE USE OF THIS SOFTWARE, EVEN IF ADVISED OF THE POSSIBILITY OF SUCH
   DAMAGE.
*/

#if defined(ARDUINO) && ARDUINO >= 100
#include "Arduino.h"
#else
#include "WProgram.h"
#endif
#include "EMGFilters.h"
#include <Servo.h>
#include "DigiKeyboard.h"

// The EMG sensor is connected to (analog) pin 12
#define emgAnInput A12

// a threshold of 1000 is used to detect muscle activation
unsigned long threshold = 1000;

EMGFilters myFilter;

// The sample frequency rate and the notch frequency to filter power line noise (50 Hz
    in the Netherlands)
SAMPLE_FREQUENCY sampleRate = SAMPLE_FREQ_1000HZ;
NOTCH_FREQUENCY humFreq = NOTCH_FREQ_50HZ;

Servo motor;

void setup()
{
  // Initializing the filter function of the EMG library
  myFilter.init(sampleRate, humFreq, true, true, true);
  Serial.begin(115200);

  // The servo motor is connected to (PWM) pin 1 of the Digispark Pro
  motor.attach(1);
  // Assigning the neutral position to the servo
  motor.write(90);
}

void loop()
{
  // vector containing the time of activation, needed for proportional control
  static int Time[] = {0, 0};

```

```
  // angle contains the state of the prosthesis hand, assuming an open hand at the start
  static int angle = 0;

  // Reading, filtering and squaring the incoming EMG signal
  int data = analogRead(emgAnInput);
  int dataAfterFilter = myFilter.update(data);
  int envelope = abs(sq(dataAfterFilter));


  // If the EMG signal is lower than the threshold the value will be set to 0
  envelope = (envelope > threshold) ? envelope : 0;


  if (threshold > 0)
  {
    // The getEMGCOunt function analyses the signal for detection of muscle activation
    if (getEMGCount(envelope, Time))
    {
      // If the hand is open, the hand will be closed
      if (angle == 0)
      {
        // updating the state of the hand to closed
        angle = 1;

        // If the muscle activation time is more than 150 ms, the hand will open in slow mode
        if (Time[1] > 349)
        {
          motor.write(99);    // Closing the hand in slow mode
          delay(1300);        // Waiting for the hand to be fully closed
          motor.write(90);    // Assigning neutral position to the servo
        }

        // If the muscle activation time is less than 150 ms, the hand will open in fast mode
        if (Time[1] < 350)
        {
          motor.write(102);   // Closing the hand in fast mode
          delay(540);         // Waiting for the hand to be closed
          motor.write(90);    // Assigning neutral position to the servo
        }

      }
      // If the hand is in closed state, the hand will open
      else
      {
        // Updating status of the prosthesis to open
        angle = 0;
        motor.write(75);    // Opening the hand
        delay(500);         // Waiting for the hand to fully open
        motor.write(90);    // Assigning neutral position to the servo
      }
    }
  }
  delayMicroseconds(5);
}

// The getEMGCOunt function analyses the EMG signal
int getEMGCount(int gforce_envelope, int Time[])
{
  static long integralData = 0;
  static long integralDataEve = 0;
  static bool remainFlag = false;
  static unsigned long timeMillis = 0;
  static unsigned long timeBeginzero = 0;
  static long fistNum = 0;
  static int  TimeStandard = 50;
  static bool flag = true;

  static unsigned long startTime = 0;
```

```
  static unsigned long activationTime = 0;
  /*
    The integral is processed to continuously add the signal value
    and compare the integral value of the previous sampling to determine whether the
    signal is continuous
  */
  integralDataEve = integralData;
  integralData += gforce_envelope;

  // If the muscle activation starts, a timer is set to measure the activation time
  if (gforce_envelope > 1000)
  {
    startTime = millis();
    if (flag == true)
    {
      flag = false;
      activationTime = startTime;
    }
  }
  /*
    If the integral is constant, and it doesn't equal 0, then the time is recorded;
    If the value of the integral starts to change again, the remainflag is true, and
    the time record will be re-entered next time
  */
  if ((integralDataEve == integralData) && (integralDataEve != 0))
  {
    timeMillis = millis();
    if (remainFlag)
    {
      timeBeginzero = timeMillis;
      remainFlag = false;
      return 0;
    }
    /* If the integral value exceeds 400 ms, the integral value is clear 0,return that
    get EMG signal */
    if ((timeMillis - timeBeginzero) > TimeStandard)
    {
      integralDataEve = integralData = 0;
      // The activation time is saved in the vector and send to time main loop
      Time[1] = startTime - activationTime;
      flag = true;
      // return 1 gives the command to open or close the prosthesis depending on its
    state
      return 1;
    }
    return 0;
  }
  else {
    remainFlag = true;
    return 0;
   }
}
```

A.3. Arduino code of pinch force test

```
// Library for the servo
#include <Servo.h>

Servo motor;

void setup() {
  // servo is attached to pin 1 (PWM) of the Digispark Pro
  motor.attach(1);
}

void loop() {
  // The motor.write() communicates the rotation speed to the servo.
  // 83-97: No rotation
  // 98-102: counter clock wise rotation with 98 slowest and 102 fastest
  // 103-180: same speed as 102
  //
  // 78-82: clock wise rotation (82 slowest, 72 fastest)
  // 0-77: same speed as at 78

  // pinch force has been tested for 98, 99, 100, 101, 102
  motor.write(98);  // closing of the hand
  delay(3300);
  motor.write(75); // opening of the hand
  delay(500);
  motor.write(90); // no movement
  delay(6000);
}
```

A.4. Arduino code of durability test

A.4.1. Code of Arduino Uno

```
// Libraries
#include <Wire.h>
#include <LiquidCrystal_I2C.h>

// The button is connected to the second digital pin of the Arduino Uno
const int  buttonPin = 2;

// Setting the parameters for connection to the LCD
LiquidCrystal_I2C lcd(0x27, 2, 1, 0, 4, 5, 6, 7, 3, POSITIVE);

// Variables will change:
int buttonPushCounter = 0;   // Counter for the number of button presses
int buttonState = 0;         // Current state of the button
int lastButtonState = 0;     // Previous state of the button

void setup() {
  // Initialize the button pin as a input:
  pinMode(buttonPin, INPUT);
  // Initialize serial communication:
  Serial.begin(9600);

  // Initialize LCD
  lcd.begin(16,2);
  lcd.clear();               // Clear LCD screen
  lcd.print("Start");        // Display "start"

}

void loop() {
  // Read the pushbutton input pin:
  buttonState = digitalRead(buttonPin);

  // Compare the buttonState to its previous state
  if (buttonState != lastButtonState) {
    // If the state has changed, increment the counter
    if (buttonState == HIGH) {
      // If the current state is HIGH then the button went from off to on:
      buttonPushCounter++;

      // Clear LCD
      lcd.clear();

      // Display the push count
      lcd.print(buttonPushCounter);
    }

    // Delay to avoid bouncing
    delay(50);
  }
  // Save the current state as the last state, for next time through the loop
  lastButtonState = buttonState;

}
```

A.4.2. Code of the Digispark Pro

```
// Library for the servo
#include <Servo.h>

Servo motor;

void setup() {
  // servo is attached to pin 1 (PWM) of the Digispark Pro
  motor.attach(1);

  // Assigning neutral position to the servo (no movement)
  motor.write(90);
}

void loop() {
  motor.write(102);      // Closing of the hand
  delay(520);
  motor.write(90);       // Neutral position
  delay(500);
  motor.write(75);       // Opening of the hand
  delay(500);
  motor.write(90);       // Neutral position
  delay(500);
}
```

A.5. Pinch force calculation

Parameters:

ratio = 0.25
pinch force F = 30 N
estimated servo arm length l = 2 cm
input torque T

Calculation:

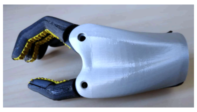
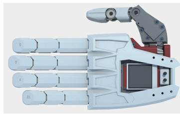
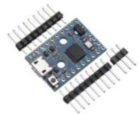
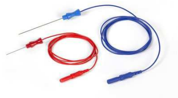
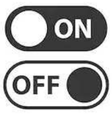
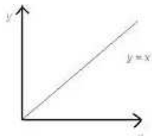
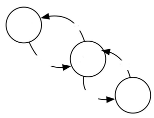
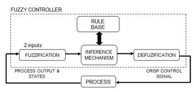
$F = ratio * T/l$

$T = (F * l)/ratio = (30 * 2)/0.25 = 240N * cm$

$T = 240N * cm \approx 24kg * cm$

A.6. Morphological scheme

Table A.1: Morphological scheme, with all proposed design solutions.

Function	Solutions				
Prosthetic hand	100 dollar hand 	Galileo hand 	InMoov hand 	Nazree hand 	
Actuation	Pneumatic 	Hydraulic 	Stepper motor 	DC motor 	Servo motor 
Control	Raspberry Pi 	Arduino Uno 	Arduino Nano 	Digispark Pro 	Digispark 
EMG measurement	Gel electrodes 	Dry electrodes 	Myo armband 	Invasive electrodes 	
EMG analysis	On/off 	Proportional 	FSM 	FLC 	
Power source	Compressed gas 	LiPo battery 	NiMH battery 		

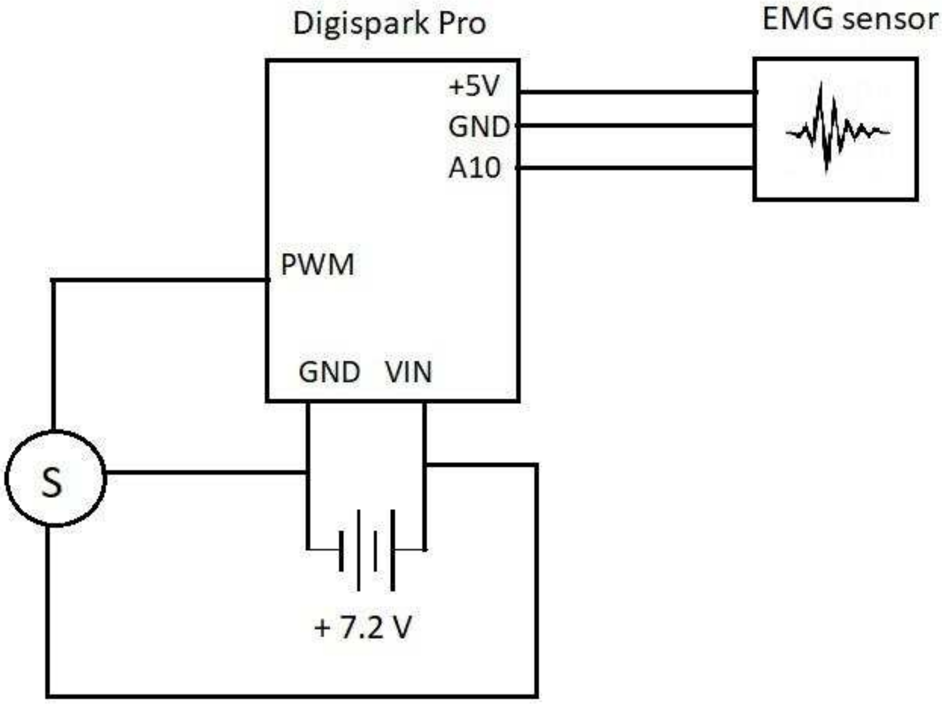
A.7. MCU specifications

Table A.2: Overview of the specifications of the proposed MCU's [4, 5, 38, 78].

	Arduino Uno	Arduino Nano	Raspberry Pi	Digispark Pro	Digispark
Dimensions [mm]	68.6 x 53.4	18 x 45	85.6 x 56.5	26.7 x 18.3	18 x 23
Flash memory [KB]	32	32	expandable	16	8
RAM	2 KB	2 KB	2-8 GB	0.5 KB	0.5 KB
Clockspeed	16 MHz	16 MHz	1.4 GHz	16 Mhz	?
Weight [g]	25	7	50	?	?
Number of analog pins	6	8	-	10	4
Number of digital pins	14	14	-	14	6
Power consumption [mAh]	?	19	540	22	?

Unpublished data is referred to with a '?'.

A.8. Electrical system scheme



A.9. 100 dollar hand CAD design refinement

The originally developed model of the 100 dollar hand was designed to function as a body powered prosthesis. In this study the selected actuation mechanism, a servo is integrated inside the hand. The mechanism inside the 100 dollar hand, to open and close the hand, had to be revised.

The original 100 dollar hand has a spring mechanism to keep the hand in open hand position when there is no force applied on the cable. When the cable is pulled, and the spring force has been overcome, the hand closes. The spring mechanism is unneeded in an externally actuated hand prosthesis, and was therefore removed. More space was available to fit in the servo and electronics. The CAD files of the original 100 dollar hand were available for SolidWorks 2019. SolidWorks 2019 was used to remodel the revisions made to the original files. Figures A.1 and A.2 show the original and revised designs. Just above the servo is a space for the Digispark Pro.

The original 100 dollar hand is printed with polylactic acid (PLA). Other printing materials have been considered, but since the documentation of the 100 dollar hand showed that the hand made form PLA, is able to withstand force and durability tests, we decided to adopt these findings and stick to printing with PLA.

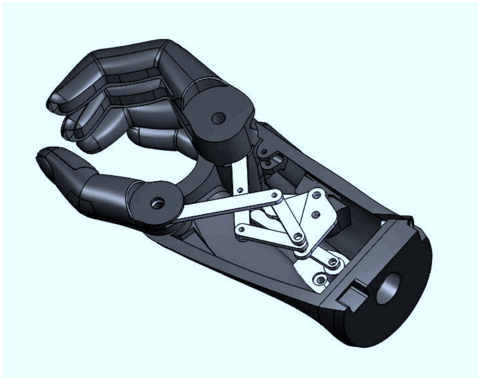


Figure A.1: Original 100 dollar hand.

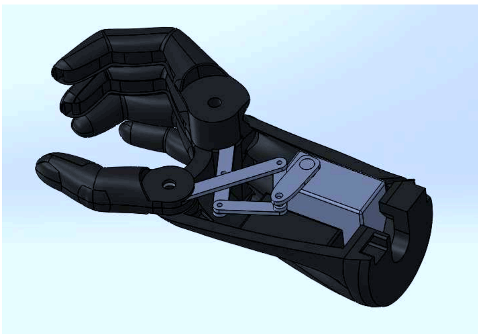


Figure A.2: Revised 100 dollar hand.

A.10. 3D printing specifications

Table A.3: 3D printing specifications and Cura settings used for printing the 100 dollar hand.

Printer type and printer settings	
Printer	Ultimaker S3
Layer height	0.2 mm
Infill density	75%
Infill pattern	Triangle
Printing material	PLA
Support material	PVA

A.11. EMG sensor specifications

Signal Transmitter Board
  Supply Voltage: +3.3 V - 5.5 V
  Operating Voltage: +3.0 V
  Detection Range: +/-1.5 mV
  Electrode Connector: PJ-342
  Module Connector: PH2.0-3P
  Output Voltage: 0 - 3.0 V
  Operating Temperature: 0-50 °C
  Size: 22 mm * 35 mm (0.87 inch * 1.38 inch)

Dry Electrode Board
  Electrode Connector: PJ-342
  Wire Length: 50 cm (19.69 inch)
  Plate Size: 22 * 35 mm (0.87 inch * 1.38 inch)
  weight: 36 g

A.12. Servo specifications

Servo specification

TD-8130MG Waterproof Digital Servo - 30kg - Continuous
Type: TD-8130MG Waterproof Digital Servo
Supply Voltage: 4.8V - 7.2V DC
Dimensions: 40 x 20 x 40.5 mm
Weight: 56 g ś 2 g
Rest current: 140 mA - 200 mA
Stall current: 2600 mA ś 10% - 3400 mA ś 10%
Torque: 29.0kg ů cm (4.8V); 32.5kg ů cm (6.6V)
No-load speed: 0.22 seconds / 60 degrees (4.8V); 0.20 sec / 60 degrees (6.6V)
Rotation range: continuous
Waterproof: yes
Gear material: metal

Control Specification:

Command signal: pulse width modification
Control type: digital controller
Pulse bandwidth range: 500 2500 usec
Neutral position: 1500 usec
Dead bandwidth: 5 usec

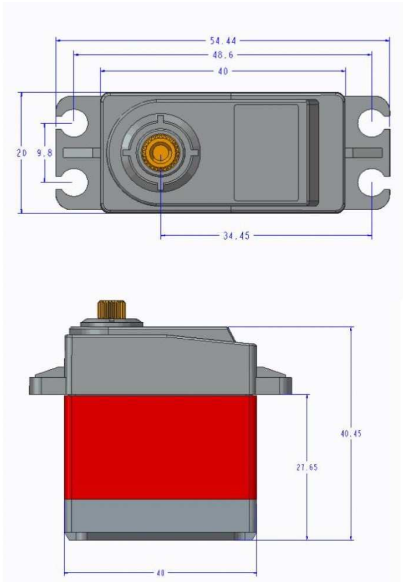


Figure A.3: Servo dimensions.

A.13. Digispark Pro connection with components

The pins of the Digispark Pro are schematically visualized in Figure A.4. The board is powered via the VIN pin, allowing voltages in a range of 6 to 20 V. The power source positive pin of the power source is connected to the VIN pin and the negative negative pin to the ground (GND) of the Digispark.

The EMG sensor has three connections: +5 V, GND and analog output. The power input pin of the sensor is connected to the 5 V output pin of the Digispark Pro. The GND pin to the GND of the Digispark and the analog output pin is connected to analog input pin 12.

The servo requires 7.2 V to produce its maximum torque. The positive pin of the servo is directly connected to the VIN pin of the Digispark Pro. Both the Digispark Pro and servo are supplied via the same VIN pin. The GND pin is also connected to the same pin of the power supply and the board. The speed of the servo is controlled via a pulse width modulation (PWM) pin. This pin is connected to pin 1 of the Digispark Pro, which supports PWM.

All connections were soldered to the board. The connections were secured and isolated with glue from a hot glue gun.

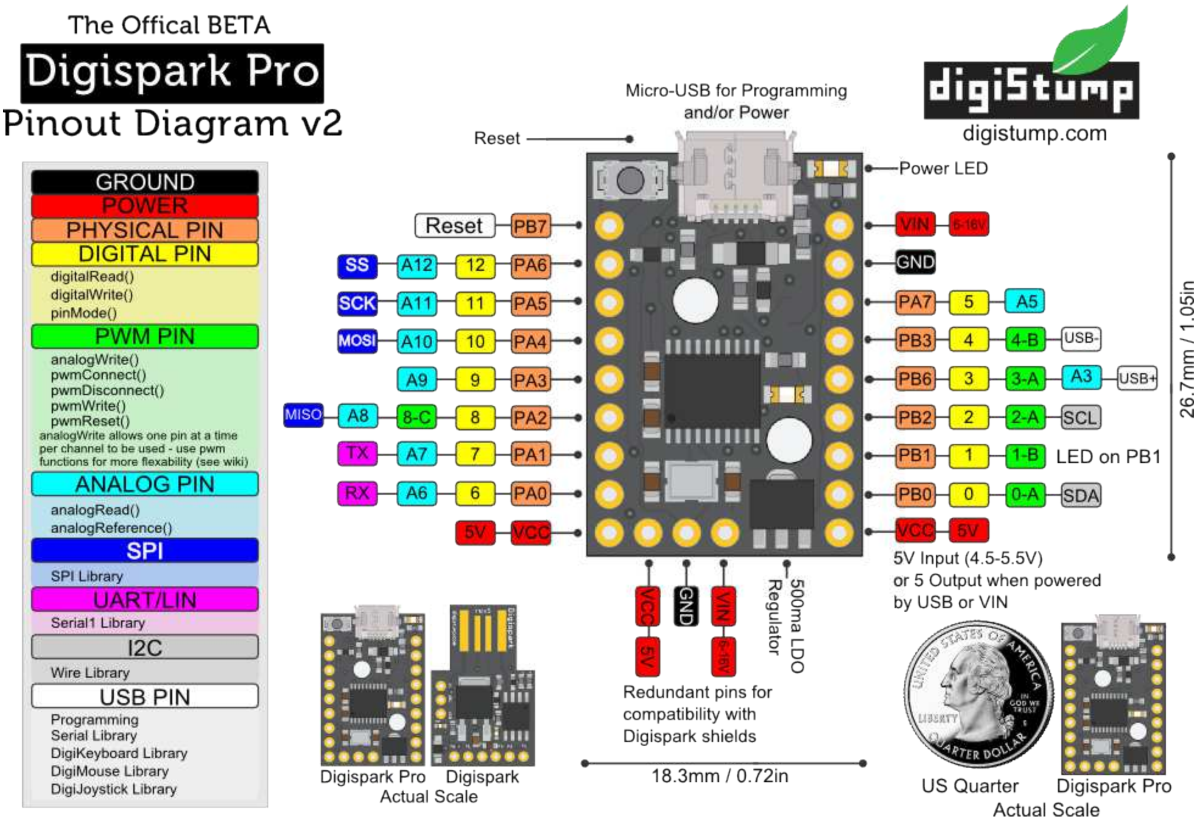


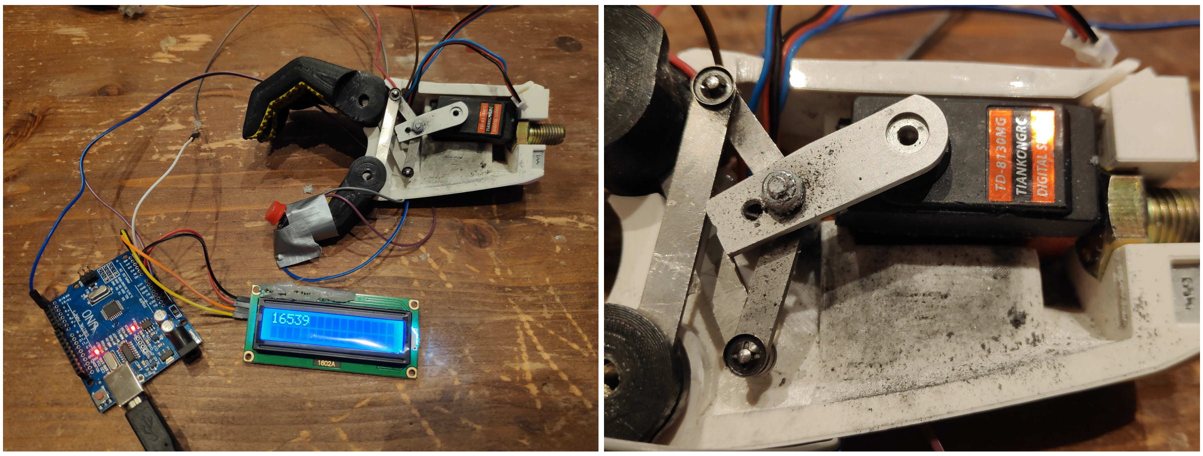
Figure A.4: Digispark Pro schematic pin visualisation.

A.14. Durability experiment

The durability experiment was paused a number of times to fix non-critical issues. The anomalies are summarized in Table A.4. After 1341 cycles, the locknut got loose. The locknut was fastened and the experiment was continued. At 4589 cycles, the locknut got loose again. Because the servo arm was not restrained anymore, the servo arm was continuously hitting the 3D printed outer shell of the prosthesis, which caused damage to the 3D printed outer shell of the prosthesis. Since this was damage caused as a result of the no longer restrained servo arm, and not as a direct result of continuous use of the prosthesis, the part was replaced and the experiment was continued. At cycle 6720, the experiment was paused to replace the locknut. At cycle 13833, a cable of the power supply connected to the Digispark Pro got loose. The cable was reconnected to the Digispark Pro and the experiment was continued. At cycle 16539, the prosthesis stopped opening the hand from closing position (Figure A.5a). After further inspection, the servo motor appeared to only move anti-clockwise. The servo motor was the limiting factor for the failure of the prosthesis. Figure A.5b shows the inside of the MyoGrab Hand after the experiment.

Table A.4: The anomalies, which caused the experiment to pause/stop, during the durability experiment.

Cycle	Anomaly
1341	Lucknut got loose.
4589	Locknut got loose.
6720	Locknut replacement.
13833	Loose power cable.
16539	Servo motor failure.



(a) (b)

Figure A.5: (a) Photo of the LCD counter and the inside of the prosthesis after critical failure. (b) Close-up of the inside of the MyoGrab Hand after the durability experiment.

A.15. Informed consent of the first experiment

Informed consent form

Myoelectric prosthesis evaluation

In this experiment the functioning of the myoelectric hand prosthesis, developed in this study, will be evaluated. Two standardized and validated tests are used for this purpose, the Blocks and Box Test and the Southampton Hand Assessment Procedure (SHAP) are tests used to evaluate hand functioning, in this case functioning of the hand prosthesis.

The Block and Box Test is a test in which small blocks have to be moved from one side of the box to the other side, within a time span of one minute. The SHAP contains of a number of tasks of grasping and moving differently shaped objects and 14 activities of daily living. The experiment takes approximately 30 minutes.

If you wish to withdrawal from the experiment you can notify the experiment instructor anytime, and you can withdrawal from this study without any further consequences.

The scores of the two tests will be anonymously used to draw conclusions about the use and functioning of the prosthesis. Personal information such as age and gender will be collected. No data that is able to identify you as a person is collected. The collected data will be only be used for this research.

Contact details of the researcher:
Jesse de Wit
j.p.dewit@student.tudelft.nl

Consent Form for testing the myoelectric prosthesis

Please tick the appropriate boxes	**Yes**	**No**
Taking part in the study		
I have read and understood the study information dated 27-04-2020, or it has been read to me. I have been able to ask questions about the study and my questions have been answered to my satisfaction.	○	○
I consent voluntarily to be a participant in this study and understand that I can refuse to answer questions and I can withdraw from the study at any time, without having to give a reason.	○	○
I understand that taking part in the study involves carrying-out several tasks while using the hand prosthesis. The performance scores of the tests will be collected.	○	○
Use of the information in the study		
I understand that information I provide will be used for evaluation of the myoelectric prosthesis and will be shared with the researchers.	○	○

Signatures

______________________ ______________________ ________

Name of participant Signature Date

I have accurately read out the information sheet to the potential participant and, to the best of my ability, ensured that the participant understands to what they are freely consenting.

Jesse de Wit

_________________________ ___________________ ________

Researcher name Signature Date

Study contact details for further information:

Jesse de Wit

j.p.dewit@student.tudelft.nl

A.16. Informed consent of the second experiment

Informed consent form

Myoelectric prosthesis evaluation

In this experiment the functioning of the myoelectric hand prosthesis, developed in this study, will be evaluated. A standardized and validated test is used for this purpose. The Blocks and Box Test is used to evaluate hand functioning, in this case functioning of the hand prosthesis.

The Block and Box Test is a test in which small blocks have to be moved from one side of the box to the other side, within a time span of one minute.
If you wish to withdrawal from the experiment you can notify the experiment instructor anytime, and you can withdrawal from this study without any further consequences.

The scores of the test will be anonymously used to draw conclusions about the use and functioning of the prosthesis. Personal information, such as age and gender, will be collected. No data that is able to identify you as a person is collected. The collected data will be only be used for this research.

Contact details of the researcher:
Jesse de Wit
j.p.dewit@student.tudelft.nl

Contact details of the TU Delft supervisor:
Gerwin Smit
G.Smit@tudelft.nl

Consent Form for testing the myoelectric prosthesis

Please tick the appropriate boxes	**Yes**	**No**
Taking part in the study		
I have read and understood the study information (dated 22-06-2020), or it has been read to me. I have been able to ask questions about the study and my questions have been answered to my satisfaction.	○	○
I consent voluntarily to be a participant in this study and understand that I can refuse to answer questions and I can withdraw from the study at any time, without having to give a reason.	○	○
I understand that taking part in the study involves carrying-out several tasks while using the hand prosthesis. The performance scores of the tests will be collected.	○	○
Use of the information in the study		
I understand that information I provide will be used for evaluation of the myoelectric prosthesis and will be shared with the researchers.	○	○

Signatures

______________________ ______________________ ________

Name of participant Signature Date

I have accurately read out the information sheet to the potential participant and, to the best of my ability, ensured that the participant understands to what they are freely consenting.

Jesse de Wit

_________________________ ___________________ ________

Researcher name Signature Date

Study contact details for further information:

Jesse de Wit
j.p.dewit@student.tudelft.nl

A.17. BBT results

Table A.5: Box and Blocks Test results of all participants on all three trials in the first evaluation.

Participant	Trial 1	Trial 2	Trial 3
1	7	7	13
2	13	15	16
3	17	19	21
4	9	10	16
5	16	14	17
6	12	15	18
7	14	14	20
8	13	17	17
9	12	19	15
10	10	14	17

Table A.6: Box and Blocks Test results of all participants on all three trials in the second evaluation.

Participant	Trial 1	Trial 2	Trial 3
1	11	15	16
2	16	19	19
3	13	11	15
4	19	18	19
5	18	20	20
6	21	22	24
7	17	18	17
8	12	18	18
9	19	16	11
10	12	18	17

A.18. SHAP results

SHAP
Southampton Hand
Assessment Procedure

Participant ID ....

Assessor ....

Your SHAP Times

Abstract Objects

Light Sphere:	14.07	Heavy Sphere:	100.00
Light Tripod:	8.68	Heavy Tripod:	5.16
Light Power:	5.18	Heavy Power:	5.44
Light Lateral:	6.66	Heavy Lateral:	6.30
Light Tip:	4.13	Heavy Tip:	4.65
Light Extension:	4.47	Heavy Extension:	4.40

Activities of Daily Living (ADLs)

Pick Up Coins:	37.53	Lifting a Heavy Object:	6.69
Button Board:	42.97	Lifting a Light Object:	6.35
Simulated Food Cutting:	100.00	Lifting a Tray:	6.19
Page Turning:	3.81	Rotate Key:	12.16
Jar Lid:	22.50	Open/Close Zip:	48.22
Glass Jug Pouring:	7.00	Rotate A Screw:	67.28
Carton Pouring:	22.09	Door Handle:	4.16

Your SHAP Scores

Functionality Profile

Spherical:	16	Tripod:	25
Power:	39	Lateral:	49
Tip:	21	Extension:	81

Index of Function Score

Index of Function: 46

SHAP
Southampton Hand
Assessment Procedure

Participant ID ....

Assessor ....

Your SHAP Times

Abstract Objects

Light Sphere:	4.97	Heavy Sphere:	100.00
Light Tripod:	7.88	Heavy Tripod:	4.62
Light Power:	4.95	Heavy Power:	6.47
Light Lateral:	5.47	Heavy Lateral:	5.94
Light Tip:	8.19	Heavy Tip:	4.62
Light Extension:	5.28	Heavy Extension:	6.22

Activities of Daily Living (ADLs)

Pick Up Coins:	152.50	Lifting a Heavy Object:	8.88
Button Board:	28.68	Lifting a Light Object:	80.94
Simulated Food Cutting:	100.00	Lifting a Tray:	8.15
Page Turning:	11.00	Rotate Key:	27.03
Jar Lid:	15.09	Open/Close Zip:	40.06
Glass Jug Pouring:	25.10	Rotate A Screw:	43.03
Carton Pouring:	14.81	Door Handle:	4.97

Your SHAP Scores

Functionality Profile

Spherical:	38	Tripod:	31
Power:	28	Lateral:	36
Tip:	17	Extension:	51

Index of Function Score

Index of Function: 38

SHAP
Southampton Hand
Assessment Procedure

Participant ID ....

Assessor ....

Your SHAP Times

Abstract Objects

Light Sphere:	8.37	Heavy Sphere:	100.00
Light Tripod:	5.47	Heavy Tripod:	4.43
Light Power:	5.75	Heavy Power:	8.94
Light Lateral:	6.50	Heavy Lateral:	7.72
Light Tip:	6.97	Heavy Tip:	4.22
Light Extension:	4.19	Heavy Extension:	3.44

Activities of Daily Living (ADLs)

Pick Up Coins:	41.25	Lifting a Heavy Object:	25.19
Button Board:	13.31	Lifting a Light Object:	20.88
Simulated Food Cutting:	100.00	Lifting a Tray:	7.12
Page Turning:	5.34	Rotate Key:	11.00
Jar Lid:	7.63	Open/Close Zip:	21.72
Glass Jug Pouring:	86.94	Rotate A Screw:	29.00
Carton Pouring:	13.12	Door Handle:	3.63

Your SHAP Scores

Functionality Profile

Spherical:	43	Tripod:	37
Power:	17	Lateral:	25
Tip:	24	Extension:	78

Index of Function Score

Index of Function: 37

SHAP
Southampton Hand
Assessment Procedure

Participant ID ....

Assessor ....

Your SHAP Times

Abstract Objects

Light Sphere:	13.64	Heavy Sphere:	100.00
Light Tripod:	12.34	Heavy Tripod:	9.00
Light Power:	6.46	Heavy Power:	5.78
Light Lateral:	6.82	Heavy Lateral:	9.79
Light Tip:	6.84	Heavy Tip:	5.78
Light Extension:	16.10	Heavy Extension:	5.50

Activities of Daily Living (ADLs)

Pick Up Coins:	68.03	Lifting a Heavy Object:	9.72
Button Board:	100.00	Lifting a Light Object:	100.00
Simulated Food Cutting:	100.00	Lifting a Tray:	6.85
Page Turning:	6.91	Rotate Key:	14.59
Jar Lid:	9.75	Open/Close Zip:	100.00
Glass Jug Pouring:	23.28	Rotate A Screw:	129.93
Carton Pouring:	25.91	Door Handle:	6.81

Your SHAP Scores

Functionality Profile

Spherical:	24	Tripod:	9
Power:	27	Lateral:	34
Tip:	13	Extension:	42

Index of Function Score

Index of Function: 31

SHAP
Southampton Hand
Assessment Procedure

Participant ID ....

Assessor ....

Your SHAP Times

Abstract Objects

Light Sphere:	8.28	Heavy Sphere:	100.00
Light Tripod:	5.66	Heavy Tripod:	5.94
Light Power:	3.72	Heavy Power:	5.53
Light Lateral:	5.00	Heavy Lateral:	4.60
Light Tip:	5.72	Heavy Tip:	4.28
Light Extension:	5.47	Heavy Extension:	3.65

Activities of Daily Living (ADLs)

Pick Up Coins:	35.37	Lifting a Heavy Object:	6.94
Button Board:	40.31	Lifting a Light Object:	100.00
Simulated Food Cutting:	100.00	Lifting a Tray:	8.84
Page Turning:	5.75	Rotate Key:	3.88
Jar Lid:	5.53	Open/Close Zip:	8.16
Glass Jug Pouring:	14.94	Rotate A Screw:	16.19
Carton Pouring:	18.41	Door Handle:	4.78

Your SHAP Scores

Functionality Profile

Spherical:	42	Tripod:	29
Power:	38	Lateral:	73
Tip:	34	Extension:	72

Index of Function Score

Index of Function: 54

SHAP
Southampton Hand
Assessment Procedure

Participant ID ....

Assessor ....

Your SHAP Times

Abstract Objects

Light Sphere:	29.94	Heavy Sphere:	100.00
Light Tripod:	6.34	Heavy Tripod:	5.81
Light Power:	5.25	Heavy Power:	6.91
Light Lateral:	22.47	Heavy Lateral:	24.01
Light Tip:	8.72	Heavy Tip:	9.25
Light Extension:	6.81	Heavy Extension:	5.78

Activities of Daily Living (ADLs)

Pick Up Coins:	133.63	Lifting a Heavy Object:	29.47
Button Board:	48.78	Lifting a Light Object:	36.34
Simulated Food Cutting:	100.00	Lifting a Tray:	8.88
Page Turning:	13.03	Rotate Key:	14.47
Jar Lid:	22.94	Open/Close Zip:	31.68
Glass Jug Pouring:	26.90	Rotate A Screw:	33.31
Carton Pouring:	40.34	Door Handle:	7.81

Your SHAP Scores

Functionality Profile

Spherical:	5	Tripod:	25
Power:	16	Lateral:	14
Tip:	10	Extension:	42

Index of Function Score

Index of Function: 24

SHAP
Southampton Hand
Assessment Procedure

Participant ID ....

Assessor ....

Your SHAP Times

Abstract Objects

Light Sphere:	4.19	Heavy Sphere:	100.00
Light Tripod:	7.56	Heavy Tripod:	5.09
Light Power:	4.93	Heavy Power:	4.68
Light Lateral:	4.47	Heavy Lateral:	5.10
Light Tip:	4.37	Heavy Tip:	4.25
Light Extension:	4.68	Heavy Extension:	4.97

Activities of Daily Living (ADLs)

Pick Up Coins:	65.31	Lifting a Heavy Object:	11.44
Button Board:	36.90	Lifting a Light Object:	5.81
Simulated Food Cutting:	100.00	Lifting a Tray:	7.03
Page Turning:	7.09	Rotate Key:	17.44
Jar Lid:	7.31	Open/Close Zip:	7.94
Glass Jug Pouring:	65.66	Rotate A Screw:	17.69
Carton Pouring:	26.01	Door Handle:	6.16

Your SHAP Scores

Functionality Profile

Spherical:	42	Tripod:	29
Power:	41	Lateral:	31
Tip:	27	Extension:	69

Index of Function Score

Index of Function: 48

SHAP
Southampton Hand
Assessment Procedure

Participant ID ....

Assessor ....

Your SHAP Times

Abstract Objects

Light Sphere:	4.69	Heavy Sphere:	100.00
Light Tripod:	12.94	Heavy Tripod:	4.84
Light Power:	4.90	Heavy Power:	4.22
Light Lateral:	4.57	Heavy Lateral:	6.15
Light Tip:	4.57	Heavy Tip:	3.82
Light Extension:	5.22	Heavy Extension:	6.60

Activities of Daily Living (ADLs)

Pick Up Coins:	157.81	Lifting a Heavy Object:	6.47
Button Board:	31.59	Lifting a Light Object:	6.81
Simulated Food Cutting:	100.00	Lifting a Tray:	9.35
Page Turning:	6.41	Rotate Key:	9.19
Jar Lid:	14.40	Open/Close Zip:	23.47
Glass Jug Pouring:	29.63	Rotate A Screw:	40.06
Carton Pouring:	21.32	Door Handle:	3.85

Your SHAP Scores

Functionality Profile

Spherical:	37	Tripod:	20
Power:	40	Lateral:	38
Tip:	26	Extension:	66

Index of Function Score

Index of Function: 44

SHAP
Southampton Hand
Assessment Procedure

Participant ID ....

Assessor ....

Your SHAP Times

Abstract Objects

Light Sphere:	18.84	Heavy Sphere:	100.00
Light Tripod:	5.02	Heavy Tripod:	6.25
Light Power:	5.38	Heavy Power:	7.38
Light Lateral:	11.88	Heavy Lateral:	5.06
Light Tip:	7.37	Heavy Tip:	6.81
Light Extension:	5.75	Heavy Extension:	4.12

Activities of Daily Living (ADLs)

Pick Up Coins:	23.88	Lifting a Heavy Object:	9.60
Button Board:	30.00	Lifting a Light Object:	7.97
Simulated Food Cutting:	100.00	Lifting a Tray:	9.90
Page Turning:	12.37	Rotate Key:	4.68
Jar Lid:	5.00	Open/Close Zip:	100.00
Glass Jug Pouring:	22.00	Rotate A Screw:	24.07
Carton Pouring:	17.00	Door Handle:	5.09

Your SHAP Scores

Functionality Profile

Spherical:	31	Tripod:	32
Power:	38	Lateral:	41
Tip:	40	Extension:	46

Index of Function Score

Index of Function: 45

SHAP
Southampton Hand
Assessment Procedure

Participant ID ....

Assessor ....

Your SHAP Times

Abstract Objects

Light Sphere:	6.88	Heavy Sphere:	100.00
Light Tripod:	4.69	Heavy Tripod:	7.07
Light Power:	6.16	Heavy Power:	5.06
Light Lateral:	5.94	Heavy Lateral:	5.38
Light Tip:	6.40	Heavy Tip:	4.47
Light Extension:	4.06	Heavy Extension:	4.44

Activities of Daily Living (ADLs)

Pick Up Coins:	57.66	Lifting a Heavy Object:	13.12
Button Board:	42.78	Lifting a Light Object:	10.50
Simulated Food Cutting:	100.00	Lifting a Tray:	8.82
Page Turning:	9.15	Rotate Key:	11.50
Jar Lid:	5.13	Open/Close Zip:	19.00
Glass Jug Pouring:	15.35	Rotate A Screw:	22.19
Carton Pouring:	20.28	Door Handle:	4.81

Your SHAP Scores

Functionality Profile

Spherical:	44	Tripod:	27
Power:	33	Lateral:	49
Tip:	20	Extension:	62

Index of Function Score

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