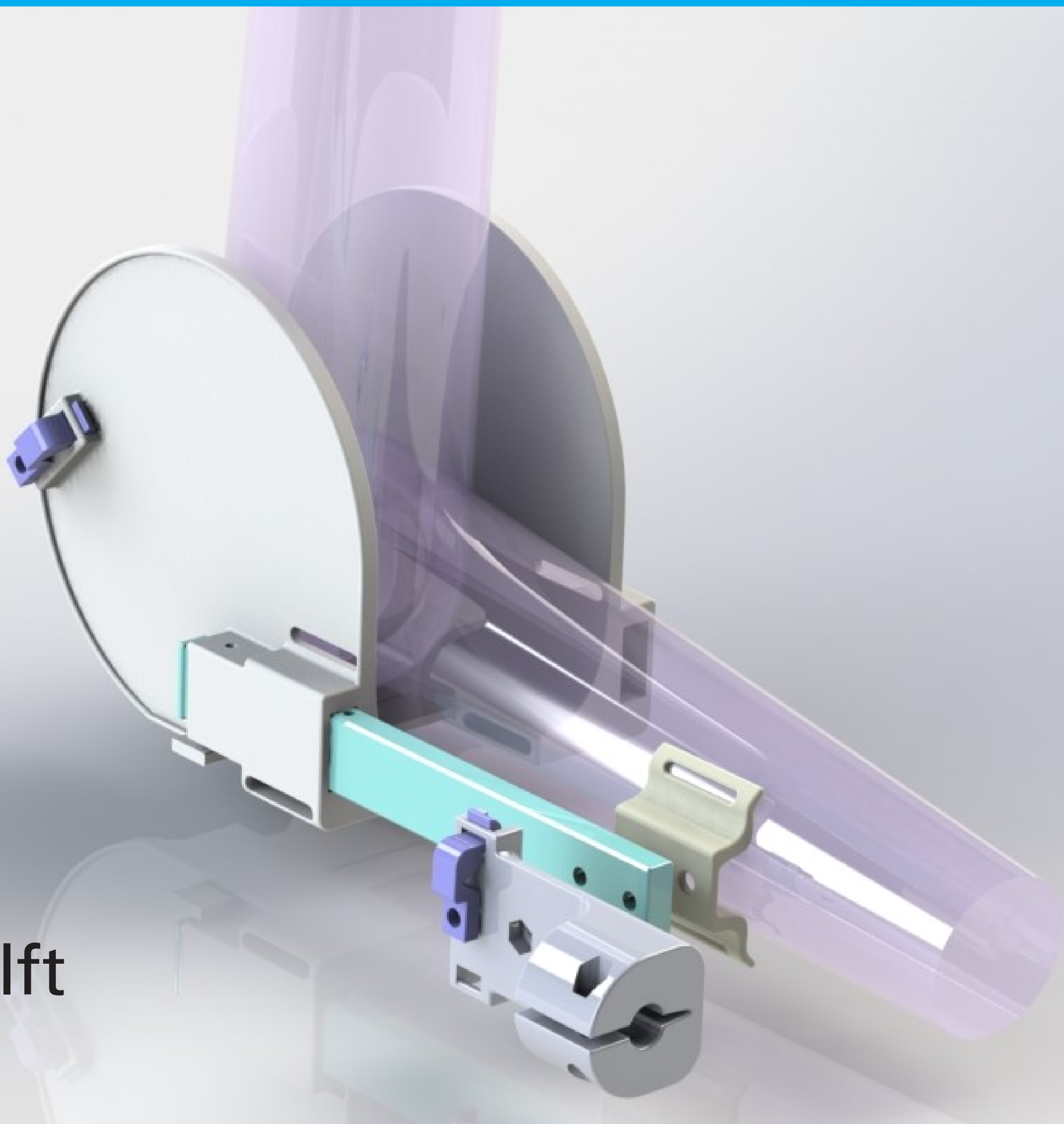
The Delft Pro Simulator

An accessible and 3D-printable prosthesis simulator to simulate transradial limb absence

M.A. Sinke



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by

M.A. Sinke

to obtain the degree of Master of Science
at the Delft University of Technology,
to be defended publicly on May 25, 2021 at 02:00 PM.

Student number:	4622596	
Project duration:	September 21, 2020 – May 25, 2021	
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This thesis is confidential and cannot be made public until May 25, 2023.

TU Delft

Preface

From a young age I have liked both sports and biology. This inspired me to study human movement science. But during my minor I realized that even though the human body has amazing qualities, it needs technology in order to perform at the highest level. In sports this could be the difference between coming in first or second. For some people it might give them the opportunity to live without the aid of others. My interest in developing technology to increase the quality of life of people grew during my study. Because of my background in human movement science I had to do a pre-master program to be accepted into the master mechanical engineering. Fortunately, I was able to gain the knowledge needed to start the master. Picking the courses was hard, considering that everything was really interesting, but luckily I followed a course in which Gerwin Smit gave lectures about prostheses. These spiked my interest for prostheses, and were the start of this thesis.

I want to thank the following people for making me able to complete this thesis. *Gerwin Smit* for every meeting and every critical note. I may not always have accepted the feedback directly, but thinking about it gave me a lot of new ideas and improved the old ones. *Aimee Sakes and Paul Breedveld* for being a member of the graduation committee and providing me with useful feedback during the green light meeting. *Jan van Frankenhuyzen* for patiently explaining every aspect of my design to me and then helping me to produce it. *Eva and Sanne* for always providing support, spare laptops, and much needed coffee breaks. *Tom, Jurjen, Israa, Menno* and all others that attended the group meetings which always gave me a lot of feedback to proceed my work (or find people at the faculty). *Jeroen* for staying up late and helping me by proofreading. And of course *my parents* for their support during my entire education (and for always asking ”are you not finished yet?”)

M.A. Sinke
Delft, May 2021

Abstract

Introduction: Prosthesis simulators are essential tools for research and rehabilitation. They can be used to increase the amount of participants in research studies or to assist in rehabilitation. The main goal of this master thesis is to design a functioning prosthesis simulator, which can be used with commonly used terminal devices. A second goal is that the design needs to be accessible for multiple research groups and rehabilitation centers. The last goal is that the design needs to solve the problems that currently exist with prosthesis simulators: sub-optimal position of terminal device and lack of restriction of degrees of freedom of intact arm.
Methods: The design requirements were based on the goals and a function analysis which included the state of the art of prosthesis simulators. Based on these design requirements the final design of The Delft Pro Simulator was developed. The design method that was used was a combination of the basic design cycle and the Fishtrap model.
Results: The Delft Pro Simulator is a novel 3D-printable design that is accessible for multiple users. The design consists of only 14 parts and weighs 356 grams. The design is minimal-assembly and the material cost is low. The simulator is able to constrain pro- and supination of the intact arm. It can be worn on the left or the right arm. The socket can be used with all terminal devices with the American standard bolt (1/2”-20) and the European standard bolt (M12x1.5). A user tested the functional performance of the design with two standardized tests, the Box and Blocks Test and the Nine Hole Peg Test.
Discussion and conclusion: This report presents a novel prosthesis simulator design. The goals have all been achieved. The design can be used with all commonly used terminal devices. Use with a myoelectric device has not been tested, but should be possible with minor adaptations. Most parts of the simulator can be 3D-printed except for some parts which are very accessible. The design can be used with two possible orientations of the terminal device with respect to the intact hand (dorsal and palmar). The simulator is able to accurately simulate transradial limb absence by constraining pro- and supination of the intact arm, while leaving flexion and extension free.

Keywords
Prosthesis simulator - Design - Upper limb - Transradial amputation - 3D-printer

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Glossary and list of abbreviations

Axial Relating to an axis. In this case the midline of the forearm.
BBT Box and Blocks Test, a standardized test to test hand dexterity.
Body-powered prosthesis A prosthesis that is operated with a cable that is connected to the prosthesis and the body. It is usually operated with a shoulder harness around the contralateral shoulder.
Contralateral On the opposite side of the body.
Distal Away from the center of the body.
Extension Stretching the elbow.
FDM Fused Deposition Modeling, a 3D-printing technique that works by melting filament in the printer head and depositing it layer by layer on the print bed, creating the product.
Flexion Bending of the elbow.
Lateral Away from the midline of the body.
Myoelectric prosthesis A prosthesis that is operated by electric motors. These are activated by electrodes that measure the electrical activity of the muscles.
NHPT Nine Hole Peg Test, a standardized test to test hand dexterity.
PLA PolyLactic Acid, a plastic material that can be used for 3D-printing.
Pronation Rotation of the forearm, moving the palm of the hand downwards and the thumb moves inward.
Proximal Towards the center of the body.
PVA PolyVinyl Alcohol, a water soluble material that can be used as support material.
Radial Part of the hand or arm at the side of the thumb.
Supination Rotation of the forearm, moving the palm of the hand upwards and the thumb moves outward.
TD Terminal device, a synonym for prosthesis.
Ulnar Part of the hand or arm at the side of the pink.

1

Introduction

1.1. Background information

1.1.1. Prosthesis simulator

Researchers working on upper limb prostheses encounter limitations caused by the small population of people with upper limb absence [1–3]. The sub-population of novice prosthesis users is even smaller. Researching training programs requires novice prosthesis users to participate. It is unethical to deny recent amputees occupational therapy to be able to participate or be the control group in a research study [1]. Especially because being able to start training early after an amputation improves prosthesis acceptance [4]. The rejection rates of prostheses by people with transradial limb absence are high, 20 to 45% of this group rejects the use of a prosthesis [5]. In literature, two ways to decrease the rejection rate were identified, by either improving the prostheses [6–8] or by improving the training programs for people with upper limb absence [1–3, 9]. A sufficient number of participants is needed for both research types, to be able to perform a reliable statistical analysis. The small population of people with upper limb absence makes this hard to achieve. A solution that could increase the population of participants is an upper limb prosthesis simulator. This device enables people with intact limbs to be fitted with a hand prosthesis. Besides increasing the population of participants, this comes with some other advantages. An able-bodied participant is usually an inexperienced prosthesis user, which is beneficial for researching training and makes the research sample more homogeneous. The increased population also allows researchers to recruit participants with specific characteristics, e.g., in terms of age or dominant hand.

A prosthesis simulator is not only useful for improving research, but it can be used in rehabilitation as well. Intermanual transfer is the principle that skills learned in one hand can be transferred to the other without additional training [2, 10]. People with upper limb absence can practice with their sound hand to improve prosthesis handling. This form of therapy seems very promising, De Boer et al. presented evidence that an experienced prosthesis user is able to use a prosthesis simulator without additional training [10]. This shows that the skills can transfer from a prosthesis to a prosthesis simulator. One of the advantages of intermanual transfer is the possibility to start training early, which reduces the prosthesis rejection rate [2]. Prosthesis use rates of 93% have been reported for patients who were fitted with an prosthesis within 30 days after amputation. For patients who were fitted later a use rate of 42% was reported [4]. Wound healing issues can make it impossible to be fitted within this 30 day period. For these patients practicing with a prosthesis simulator can be a good opportunity to start training right after the amputation.

A recent literature review reports only one commercially available prosthesis simulator, the Fillauer TRS body-powered simulator [12, 14]. This simulator was used in 2 of the 31 reviewed research studies [11, 15]. The other studies used custom-made prosthesis simulators. **Figure 1.1** shows three examples of state of the art prosthesis simulators. In one of the studies, the design process of the prosthesis simulator was reported [6]. The other studies did not report the details of the design process or building specifications of the simulator. This made it hard to reproduce studies or compare studies with each other. The literature review also focused on the differences between a simulator and a prosthesis. The key differences that were identified were the position of the terminal device and the

(a) A prosthesis simulator with the terminal device in an axial position from the intact hand. This simulator was made by adding a frame to a wrist brace [10].

(b) A body-powered prosthesis simulator with the terminal device in a palmar position from the intact hand [11]. This is the commercially available TRS simulator [12].

(c) A myoelectric prosthesis simulator with the terminal device in a dorsal position from the intact hand. This simulator was custom made for the participant [13].

Figure 1.1: Three state of the art prosthesis simulator designs, with the terminal device in three different orientations with respect to the intact hand. Pictures obtained from De Boer et al. [10], Cuellar et al. [11] and Kyberd [13].

degrees of freedom of the arm and (prosthetic) wrist. The terminal device should ideally be placed at the position of the intact hand, which is not possible for simulators. The other difference is that people with upper limb absence usually experience a restriction in degrees of freedom of the forearm, either due to the amputation or due to the prosthetic socket. In most of the simulator designs, the degrees of freedom of the forearm were not restricted.

1.1.2. 3D-printing

The process of creating a product by adding layers of material on top of each other is called additive manufacturing. A popular term for additive manufacturing is 3D-printing [16]. Multiple 3D-printing techniques make it possible to use many different materials; including but not limited to plastics, metals and even hybrid materials that combine plastics with wood fibers. Currently, the most commonly used form of 3D-printing is Fused Deposition Modeling (FDM), which is also called Fused Filament Fabrication (FFF) [17]. The printer melts material in the printer head and deposits it on the print bed, creating the product layer by layer. 3D-printing has some advantages and disadvantages when comparing it to other (subtractive) manufacturing methods. Four advantages are commonly reported. The first is that the design process can be very quick, because the product can be made directly from the 3D-drawings. This gives the opportunity to make prototypes in an early stage of the design process and implement improvements immediately. Secondly, there is no need for assembly, because products can be made out of one part. No specialized tools are needed for assembly or fabrication (except for a printer). Thirdly, complex geometries can be made in the same manner as easier shapes, this gives a large design freedom. The last is that products can be customized easily, for example scaling a product does not require other tools or machines [16, 18]. There are four commonly reported disadvantages as well. Firstly, the mechanical properties of the product are hard to predict and calculate. The exact fabrication method, printing orientation, infill percentage, and other printer settings have an influence on the strength of the printed product. Secondly, the material can shrink after printing. This also depends on the parameters of the printer and (inconsistent) post-processing, which can influence the accuracy of the printed product. The third disadvantage is that the product has a size constraint, as the print bed is not infinitely big. Especially if a certain printing orientation is needed, the size of the printer can limit the design. Finally, 3D-printing is not suitable for mass production [19, 20].

Ottobock calls 3D-printing a “revolutionising treatment for patients” [21]. They mention the possibility of quickly being able to make an individual fitting for both prostheses and orthoses. In February 2021, searching for ‘“3D-printing” AND “hand prosthesis”’ at `scholar.google.com` gives about 25 results when only selecting articles from 2021. Schools, institutions and health care facilities increasingly have access to a 3D-printer. Companies are even offering lesson packages so that students can print their own prostheses [22]. It can be hard to access a prosthesis workshop in third world countries which makes it impossible to get the knowledge and/or tools to produce a complex mechanism like a prosthesis. Making complex geometries using a 3D-printer means that the production difficulty will be kept low. Furthermore, assembly can be reduced to a minimum to keep accessibility high. It can be concluded that 3D-printing makes prostheses more accessible for different target groups. The part that is currently missing is an accessible 3D-printed prosthesis simulator. The advantages that 3D-printing offers when making a prosthesis also apply when making a prosthesis simulator. A 3D-printed prosthesis simulator can be used in schools to test their printed prostheses. And it can be used in rehabilitation or in research to test terminal devices or training programs for novice prosthesis users. The simulator can be easily customized to one's needs, and no specialized tools are needed. The 3D-printer itself is the most expensive part required for production of the simulator. This should not form a problem when the same machine can be used to print a prosthesis and the simulator.

1.2. Problem statement

Research often uses prosthesis simulators for various reasons. The problem is that design choices for prosthesis simulators are often not reported in the articles, making reproduction of current simulators impossible. Another problem is that differences between prostheses and prosthesis simulators may influence the simulation and thereby study results. The key differences are identified to be terminal device position and degrees of freedom of the intact arm.

1.3. Goal

The goal of this master thesis is to design a functioning prosthesis simulator, which can be used with commonly used terminal devices. The design should be available and easily producible for multiple research groups. The design needs to solve the problems that currently exist with prosthesis simulators: sub-optimal position of terminal device and lack of restriction of degrees of freedom of intact arm.

1.4. Layout of the report

The layout of the report is as follows: **Chapter 1** gives an introduction in prosthesis simulators and 3D-printing of prosthesis simulators. **Chapter 2** introduces the design requirements and the design approach. **Chapter 3** presents the design and the prototype of the first, structural, concept. **Chapter 4** presents the design and the prototype of the more detailed formal concept. In **Chapter 5** the features to make the design usable with a terminal device are added. This chapter ends with the final design. In **Chapter 6** the final design is evaluated by a user. **Chapter 7** is a report of the material and strength tests that were performed. The design requirements are also evaluated in this chapter. In **Chapter 8** the results are discussed and suggestions for future research are made. And, finally, **Chapter 9** concludes the report. The appendices contain more background information, the technical drawings and the Matlab scripts that were used for the calculations.

2

Design requirements

2.1. Introduction

This chapter introduces the design requirements and the design approach for the design of the Delft Pro Simulator. The design requirements are closely linked to the function of the simulator. The design requirements follow from the function analysis. The detailed function analysis is available in **Appendix A**. The function analysis is structured by a process tree, which covers the entire lifecycle of a prosthesis simulator. The function analysis also describes the state of the art of prosthesis simulators [23]. The design requirements are presented in the following two sections. They are split in two categories: the design requirements and the design wishes. The design requirements must be optimized in order to achieve the best possible simulation. The design wishes can make the design better or easier to use, but these are not necessary for the function of the simulator. The last section of this chapter describes the design approach.

2.2. Design requirements

- **Constrain degrees of freedom of the intact arm:** Transradial limb absence causes people to lose degrees of freedom of their affected arm [5]. In order to make an accurate simulation, the arm of an intact user should be constrained in the same way. Movement of the wrist should be constrained, as well as pro- and supination of the forearm. Flexion and extension of the elbow should be possible.
- **Adaptable socket:** The simulator needs to be adaptable so that it can fit most adults. It needs to adjust for forearm lengths (elbow crease to wrist) of 20 to 25 cm [24]. It also needs to adjust for arm circumferences of 20 to 25 cm and wrist circumferences of 12 to 17 cm.
- **Usable with one hand:** To enable research participants or novice users to use the simulator at home, it needs to be possible to put the simulator on with one hand.
- **3D-printable:** The majority of parts should be 3D-printable in order to increase accessibility of the design. This means that the size of the parts cannot exceed 20x20x18 cm, which is the building volume of the smallest Ultimaker and Prusa printers (**Appendix B**).
- **Amount of parts ≤ 15:** Assembly should be kept to a minimum to make the building process easy. The maximum amount of parts is therefore 15. This value does not include the parts needed to mount a terminal device to the simulator, because some terminal devices require different parts (e.g., a shoulder harness or a battery).
- **Usable with a body-powered terminal device:** The simulator design will be tested with a body-powered terminal device. In order to use such a device, a shoulder harness is needed. A Bowden cable needs to be fixated at two points to guide the cable from the prosthetic hand towards the opposite shoulder of the user.
- **Resist common loads:** The simulator needs to be strong enough to resist common loads for body-powered terminal devices (visible in **Figure 2.1**). A body-powered hand has a pinch force of 15–60 N [8]. It needs to be possible to lift and pull an object with a force of ≥ 60 N. Furthermore,

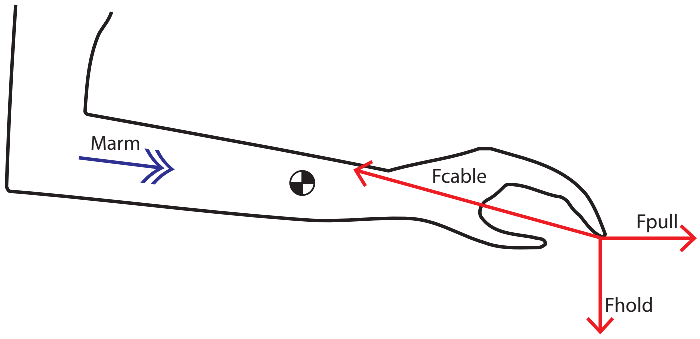


Figure 2.1: Overview of the possible forces and torque that the prosthesis simulator needs to resist. The force as caused by a cable is absent in myoelectric prostheses.

the simulator may not deform due to the force exerted by the cable. The highest cable operation force of a body-powered hand is 131 N [25]. The last common load that needs to be resisted is the torque of the forearm. This torque is between 0.5 and 4.6 Nm for male participants and between 0.4 and 2.3 Nm for female participants [26]. The simulator needs to resist 4.6 Nm to be useful for all participants.

- **Center of mass displaces ≤ |35 mm|:** The kinematics of the arm change with displacement of the center of mass. The average forearm lengths of men and women are 35.5 cm and 34.2 cm [24]. A displacement of center of mass by 10% of the forearm length corresponds with an approximate displacement of 35 mm.
- **Weight ≤ 500 grams:** A high weight makes the prosthesis simulator less comfortable to use. Values from 274 to 1008.5 grams have been reported for previous designs. The latter was including the terminal device [27]. The simulator should be as light as possible, with a maximum of 500 grams. This is excluding the terminal device.
- **Skin-safe material:** The material must be skin-safe so that users can wear it. This also means that it needs to withstand moisture and saline. After wearing it it needs to be able to be cleaned with disinfecting soap or an alcohol solution.
- **Lifespan ≥ 100 hours:** The goal for the simulator is to last longer than one study. When using it for three studies, with approximately 30 participants each and one hour of use per person, it needs to last at least 100 hours without any malfunctioning. Improving durability of the simulator is not only beneficial for the environment, but also for the people using it. A longer lasting simulator is cheaper per use, and more people are able to use the simulator.
- **Material cost ≤ €100.** To increase accessibility the design needs to be as cheap as possible, with a maximum of €100. The cost can make a design more or less accessible for different target groups. To use the design in third world countries or at high schools, the cost should be as low as possible. The material cost of this design should be limited to €100,-. Labour costs are excluded because the design needs to be minimal-assembly.

2.3. Design wishes

- **Fit all terminal devices:** Both the European standard bolt (M12x1.5) and the American standard bolt (1/2”-20) should fit in the socket.
- **Usable with a myoelectric terminal device:** In order to use it with a myoelectric terminal device there needs to be space for electrodes on the arm. There also needs to be space to put a battery.
- **Adapts in ≤ 1 min:** The device should be easy to use. It needs to change length and width in less than one minute so that it can be used by multiple users in a short time.
- **Put on and take off ≤ 1 min:** The simulator needs to be easy to wear. Putting the simulator on and securing it should take less than one minute. Taking the simulator off is expected to last the same amount of time.

- **No specialized tools:** No (specialized) tools should be needed to adjust the simulator or to put it on. This makes it more accessible and allows users to adjust the simulator at any time.
- **Recyclable:** More than 50% of the total amount of parts should be recyclable, by either reusing it or composting it. This reduces the ecological footprint of the simulator.

2.4. Design approach

The approach of this thesis was a combination of the basic design cycle and the Fishtrap model. The Fishtrap model was used to generate and evaluate solutions in the basic design cycle. The basic design cycle starts with a definition of the problem. This is followed by a function analysis, from which the design requirements can be obtained. With the requirements in mind solutions are generated, which are then evaluated. Generating and evaluating solutions is usually an iterative process. When the final design satisfies all design requirements, the basic design cycle ends. In this report, the solution finding and evaluating phase was structured by the Fishtrap model. The name of this model comes from “Catching the best solution” [28]. This model consists of three phases and three concepts are formed: a structural concept, a formal concept and a material concept [23]. The material concept is also the final design. The idea is that in every phase the possible solutions diverge to many options, which are then grouped in categories and converge back to one or more possible concept solutions. In this way, multiple solutions can be explored simultaneously. The design cycle with the Fishtrap model results in one final concept, which satisfies all requirements that have been determined in the beginning of the design process. The combination of 3D-printing and the Fishtrap model allows for quick iteration steps and evaluations of prototypes before going to the next phase.

Figure 2.2 shows the Fishtrap model chart of the design of the prosthesis simulator. The first phase, that resulted in the structural concept, was the basic structure of the simulator. The structural concept solved a spatial problem, where did different parts need to go and how could they be used together. For the simulator this was the way to restrict pro- and supination and how could the simulator be attached to the arm. The second phase, that resulted in the formal concept, determined the shape of the parts. In this phase the structural concept of the previous phase was worked out to a functioning prototype. The goal was to find the best solution for every design requirement and incorporate it in the design. The last phase, that resulted in the material concept, determined the final touches of the design. These were the production method and the materials used for the parts. For the design of the simulator, the final parts to be able to use the design with a body-powered terminal device were added in this phase as well. The last thing that was added in this phase was the part to satisfy the design wish to make the device usable with all bolt sizes for terminal devices.

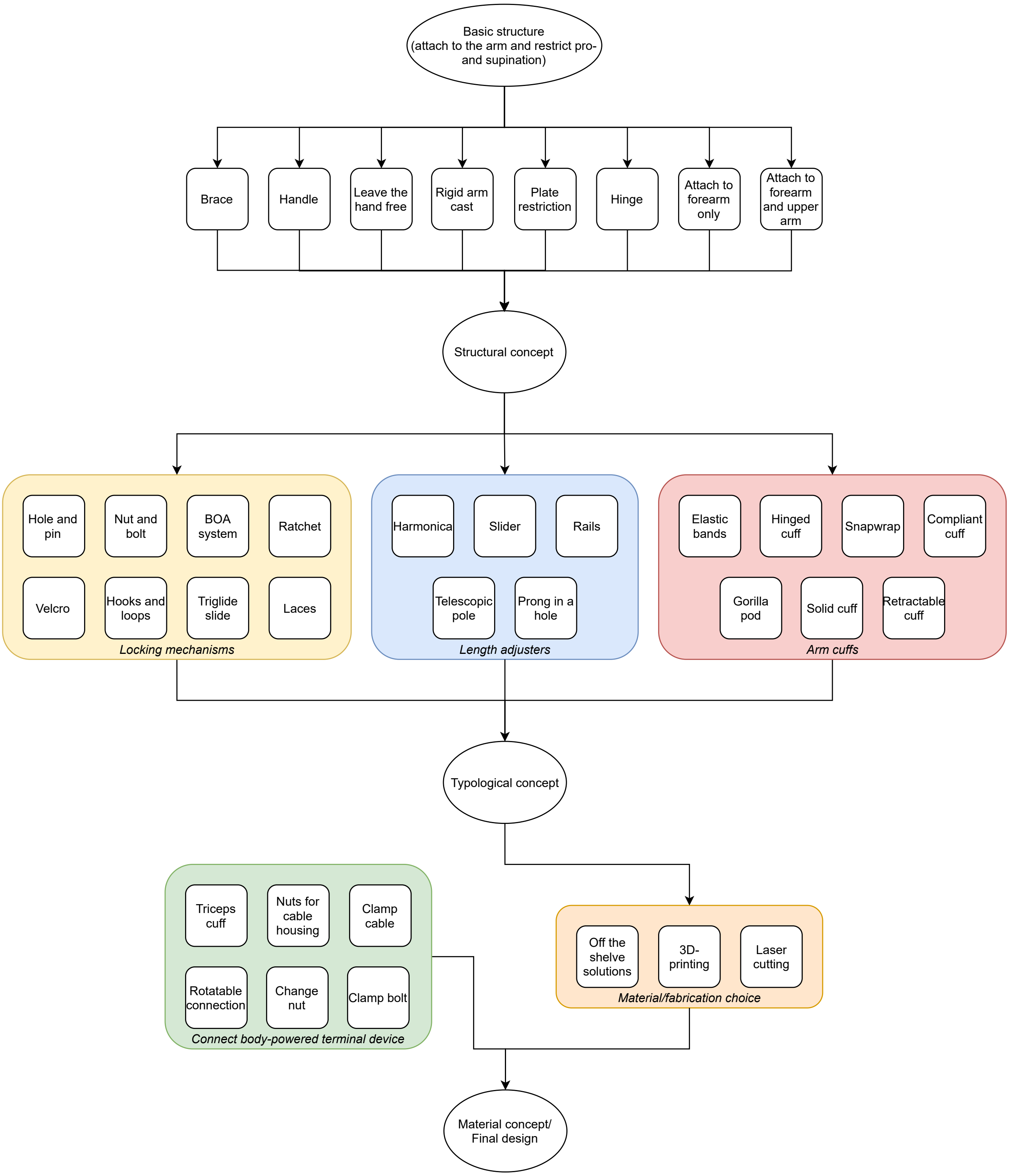


Figure 2.2: Overview of the different design solutions. They have been organized with the Fishtrap model. The basic structure of the design formed the connection to the arm and the restriction of degrees of freedom of the intact arm. This resulted in the structural concept. To create the typological concept, the details were filled in. It was the selection of the best cuffs, the best locking mechanisms and the best length adjuster. The material concept, which was also the final design, determined the material and fabrication method. For the final design the parts to connect a body-powered terminal device were added as well.

3

Structural concept

3.1. Generate solutions

In literature, four different solutions were found for constraining movement at wrist level (**Figure 3.1a–d**). These were a handle, a brace, a moulded socket around the arm or leaving the hand out of the simulator. A handle usually came with two attachment points to the arm. These were Velcro bands attached to a solid ring or frame [29]. A design with a handle did not constrain pro- and supination or flexion and extension of the elbow. A wrist brace could be a specialized solution [12] or an off-the-shelf brace that was adapted for its use [6]. The specialized solution incorporated the parts needed for the simulator on the brace. The adapted brace consisted of a brace with a frame added to it, to support simulator functions. A wrist brace enclosed the entire forearm. The wrist movement was constrained, but pro- and supination and flexion and extension were not constrained. Two designs in literature were custom made for the user [13, 30]. They were moulded around the entire forearm and restricted both the elbow as well as the hand and wrist. The hand and wrist were restricted with a handle in both cases. Pro- and supination were constrained because the socket was attached around the elbow as well. The downside was that this also restricted flexion and extension. The last proposed solution was to leave the hand out of the simulator [8]. The simulator was attached to just the wrist and the forearm with adaptable cuffs. Leaving the hand free meant that movement of the hand or wrist did not have an influence on the prosthesis orientation. It did not restrict pro- and supination, but if the arm could rotate freely in the socket, the orientation of the prosthetic hand with regard to the axis of the arm remained the same. This design did not influence flexion and extension of the elbow.

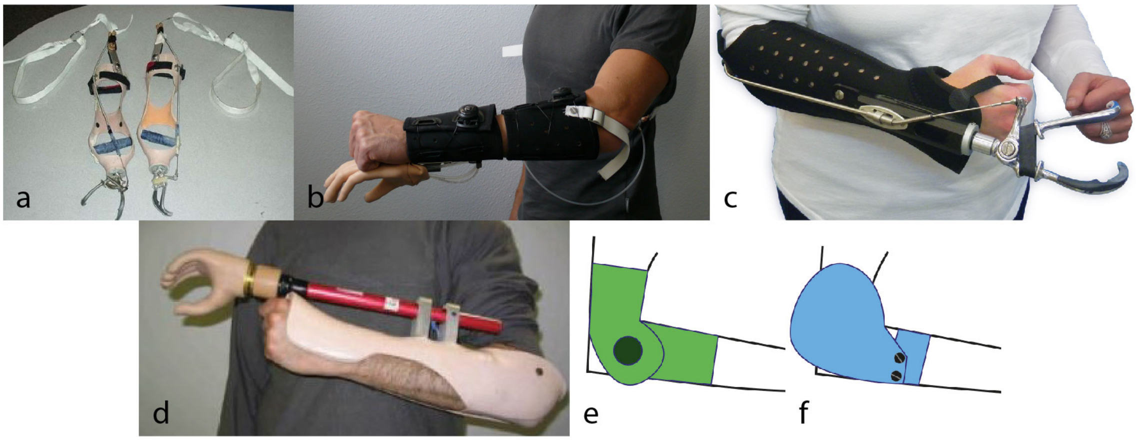


Figure 3.1: An overview of found solutions for restriction of wrist movement and the restriction of pro- and supination of the forearm. (a) Handle constraint [31]; (b) Leaving the hand free [8]; (c) Using a wrist brace [12]; (d) A moulded cast custom made for the user, this also restricts pro- and supination [13]; (e) A hinge could constrain pro- and supination while leaving flexion and extension free; (f) Two plates and a cuff that constrain pro- and supination.

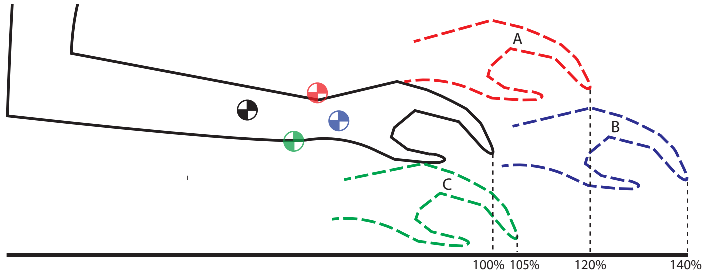


Figure 3.2: Three positions for the terminal device as found in literature. The estimated positions of the center of mass of the device are indicated. As well as the estimated arm overlength caused by the simulator.

Pro- and supination of the user were restricted in only a small amount of the custom-made designs that were found in prosthesis simulator literature. These designs also restricted the flexion and extension of the elbow. The other aforementioned designs restricted only the movement of the wrist of the user. A solution could be to add a hinge to the design (**Figure 3.1e**). This should allow flexion and extension, while constraining pro- and supination of the arm. A compliant non-assembly 3D-printable hinge had been used in a knee orthosis before [32]. This knee orthosis did need a (non-3D-printable) damper and spring to function properly. Another solution could be an off-the-shelf prosthesis hinge, which is generally made out of metal. More possible solutions were found in health care. Splints and orthoses are commonly used to allow broken bones or strained muscles to heal. Two splint designs that restricted pro- and supination were the Muenster splint and the sugartong splint. These did not leave flexion and extension completely free. The sugartong can be adapted by leaving the back of the elbow free and putting the restricting plates in the direction of the upper arm (**Figure 3.1f**). This forms another solution that leaves flexion and extension free and restricts pro- and supination.

The position of the terminal device with respect to the intact hand was the last topological feature to be determined in the structural concept. Current prosthesis simulators placed the terminal device at three spots respective to the intact hand (**Figure 3.2**). These were the axial, palmar and dorsal position. These are not the only possible positions, the prosthetic hand can be placed anywhere around the intact hand. The terminal device needs to be as close to the intact hand as possible, to improve comfort and to improve the simulation. The comfort is higher when the weight of the terminal device is closer to the elbow. The simulation is better if the device could be used in the same way someone with upper limb absence uses a prosthesis. This leaves two possible positions for the terminal device, palmar and dorsal, with respect to the intact hand. The proposed solution is to make the prosthesis simulator symmetrical. This allows the user to choose the orientation of the terminal device. It can be used at the dorsal or the palmar side and/or on both arms of the user.

3.2. Evaluate solutions

In the Fishtrap chart (**Figure 2.2**) the solutions that were found in the previous section were categorized. This evaluation converges the amount of solutions back to a structural concept. The categories are the amount of attachment points, the ways to constrain the wrist and the way to constrain pro- and supination. A lower amount of attachment points is preferred to increase comfort of the participant. For comfort it is also preferred to have as little weight as possible on the distal end of the arm. The proposed solution is therefore to attach the simulator to the arm at two points and to leave the hand (and wrist) free. The best way to restrict pro- and supination of the forearm was the last decision that needed to be made. To make the evaluation more objective, Harris profiles were made. The grading criteria for the elbow cuff are to restrict pro- and supination, leave flexion and extension free, to be 3D-printable and to be adaptable for multiple users. **Table Ia** shows a specification of these grading criteria. **Table Ib** shows the Harris profiles for restricting pro- and supination. The constraint with the two plates is the best one of the three. It leaves flexion and extension free, it is easy to 3D-print and adaptable for multiple users. A constraint based on a moulded cast can be a good solution for a simulator that is custom made for a single user. A constraint based on a hinge is hard to 3D-print. The proposed solution of leaving the hand and wrist free and attaching the simulator only to the forearm can be accepted for the plate constrain. The plates make it possible to leave the hand free, because the forearm rotates in the cuff, while the orientation of the prosthetic hand is kept the same by the plates pressing on the upper arm.

Table I: Grading criteria and Harris profiles for constraining the degrees of freedom of the intact arm.

(a) Specification of the grading criteria.

	- -	-	+	++
Pro- and supination	Full range of motion	120 degrees	90 degrees	Fully restricted
Flexion and extension	Fully restricted	90 degrees	120 degrees	Full range of motion
3D-printable	Nothing can be printed	< 50% can be printed	> 50% can be printed	Everything can be printed
Adaptable	Not adjustable	Adjusts in one dimension	Adjusts in two dimensions	Will fit everyone

(b) The Harris profiles.

	Hinge constrain	Plate constrain	Cast constrain
Pro- and supination	++	++	++
Flexion and extension	+	++	-
3D-printable	-	++	++
Adaptable	+	+	- -

The orientation of the prosthesis with respect to the intact hand was evaluated as well. A recent study compared two terminal device positions with respect to the intact hand [33]. The palmar and the axial direction were compared. This study found that the palmar position caused the users to compensate less for wearing the simulator. Other studies also reported the importance of novice users to be able to see the prosthetic hand to be able to use it [13, 34]. Putting the terminal device in another position than axial from the intact hand reduces the change in kinematics of the user [2, 3]. For some reasons, like cable orientation, it can be better to place the terminal device on another side of the intact hand. This was why the solution to make the simulator design symmetrical was accepted. In this way the simulator can be used in multiple different orientations. These should limit the overlength of the arm as much as possible. The preferred position of the prosthetic hand compared to the intact hand was the palmar side.

3.3. Prototype

The proposed solution for constraining pro- and supination was to design a cuff with plates attached to it. The goal of the prototype was to find out whether this design can constrain pro- and supination of a user. The plates need to constrain a moment of 4.6 Nm caused by pro- and supination of the forearm. This was a distributed load over the whole plate. The wall thickness needs to be high enough to restrict this moment. A calculation was made to determine whether the walls would be strong enough at a height of 3 mm. An estimation of the wall strength can be made by calculating the maximum deflection caused by the load. **Equation 3.1** was used for this. In this formula W is the distributed load on the plate; w is the width of the plate; h is the thickness of the plate; L is the length of the plate; E is the flexural modulus and M_{arm} is the moment by the arm on the plate.

$$|v_{\text{max}}| = \frac{WL^4}{8EI} \tag{3.1}$$

with:

$$I = \frac{wh^3}{12} \tag{3.2}$$

and:

$$W = \frac{2M_{\text{arm}}}{L} \tag{3.3}$$

The values to be filled in are; E = 2346.5 MPa [35], h = 3 mm, w = 50 mm, L = 75 mm and M_{arm} = 4.6 Nm. This resulted in a maximum deflection of v_{max} = 1.8 mm. This should be sufficient for the simulator, as the skin of the arm can be deformed more than this. The aforementioned measurements were

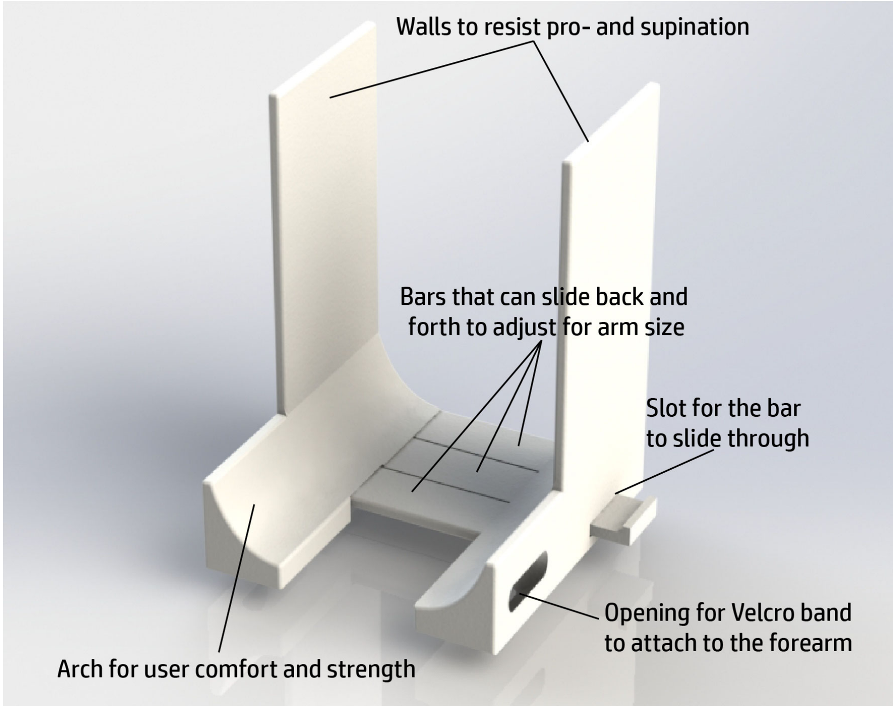


Figure 3.3: A render of the drawings of the first prototype. The function of the different parts are indicated in the figure. The three bars create a retractable cuff that could be used to adjust the prototype to different arm sizes.

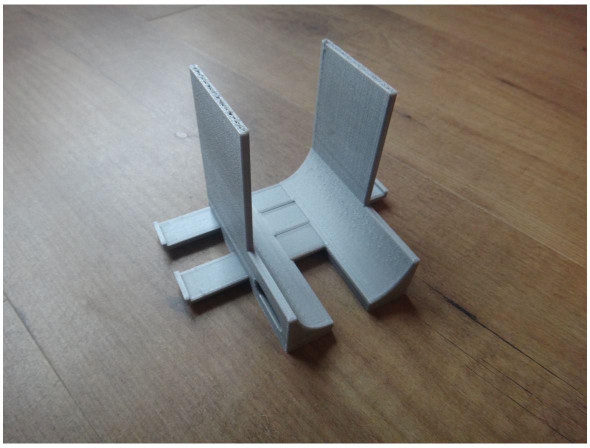
used to create the design in **Figure 3.3**. In order to make the design wearable and adjustable, a sliding mechanism was added to the walls. This mechanism consists of three parallel bars. These bars can slide in slots on the opposite wall. The connection between the walls and the slots is an arch. This is comfortable for a user to wear and it prevents creating a vulnerable spot in the design. The walls are square, which makes them only usable to constrain pro- and supination with the elbow in a 90 degree angle. This decision was made because it was still useful for testing and it saved material. The prototype was printed on the Ultimaker 3 3D-printer, using grey PolyLactic Acid (PLA) and PolyVinyl Alcohol (PVA) as support material. The printing settings are reported in **Appendix B**. The goal of the prototype was to prove that the walls can resist pro- and supination of a participant wearing it. The first examination of the prototype shows that the printer ran out of material before finishing the print (**Figure 3.4a**). The walls are approximately 3 cm lower than intended (5 cm instead of 8 cm). **Figure 3.4b** shows a picture of a user "wearing" the prototype. The size of the sliding mechanism has a good range to adjust to different arm sizes (4 to 8 cm wide). However, the walls are too low to resist any movement of the elbow. The strength of the walls was tested with a bench-top experiment, as it was not possible to test it with a user. The prototype was clamped and then a displacement was prescribed. The corresponding force was measured. **Figure 3.5a** shows the set-up of the experiment. It was not possible to test the design with a prescribed distributed load. When loading the wall at just one point, the deformation is different than in **Equation 3.1**. The deformation for a point load is:

$$|v_{max}| = \frac{PL^3}{3EI} \tag{3.4}$$

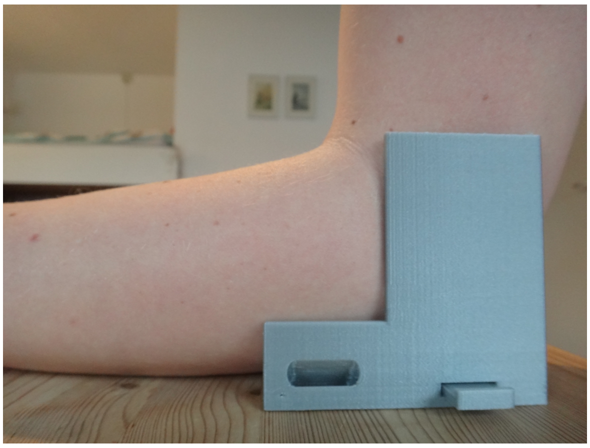
The relation between the point load (P) and the distributed load (W) is therefore:

$$P = \frac{3WL}{8} \tag{3.5}$$

Filling in **Equation 3.5** with the values W = 122 N/m and L = 75 mm results in a point load of P = 3.45 N. The graph in **Figure 3.5b** shows the result of the experiment. The measured force was 3.45 N at a displacement of 3.1 mm. This means that the walls are bending more than calculated. This can partly be prescribed to tolerance of the sliding mechanism (as the wall was not completely isolated). The figure shows a zigzag pattern, this was due to the force sensor slipping. This briefly caused the force to be smaller, while the displacement kept increasing (or decreasing). There was also a deformation of the wall visible, so the profile of the wall needs to be changed in order to improve the second prototype. This solution was added to the design in **Chapter 4**.

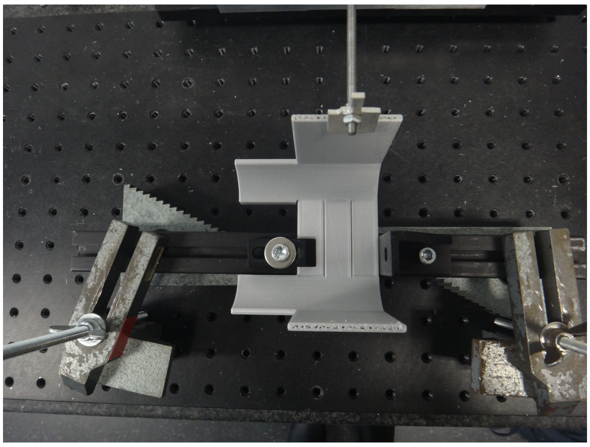


(a) Here it is clearly visible that the walls are not finished. The sliding mechanism works well.

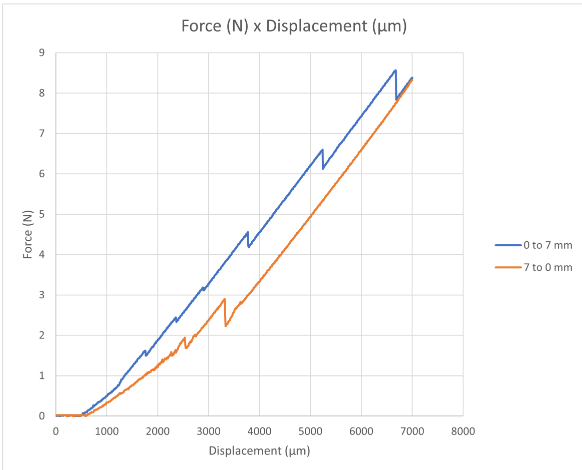


(b) The prototype can be worn by a user. The walls are too low to be able to keep the simulator in the same orientation as the upper arm.

Figure 3.4: The first prototype, which is printed in grey PLA on the Ultimaker 3. The walls are not finished due to the printer running out of material before finishing the print.



(a) Clamping the prototype for the force-displacement test.



(b) The results of the force-displacement test. Displacement was the independent variable, the force was measured.

Figure 3.5: Set-up and results of testing the wall strength of the first prototype.

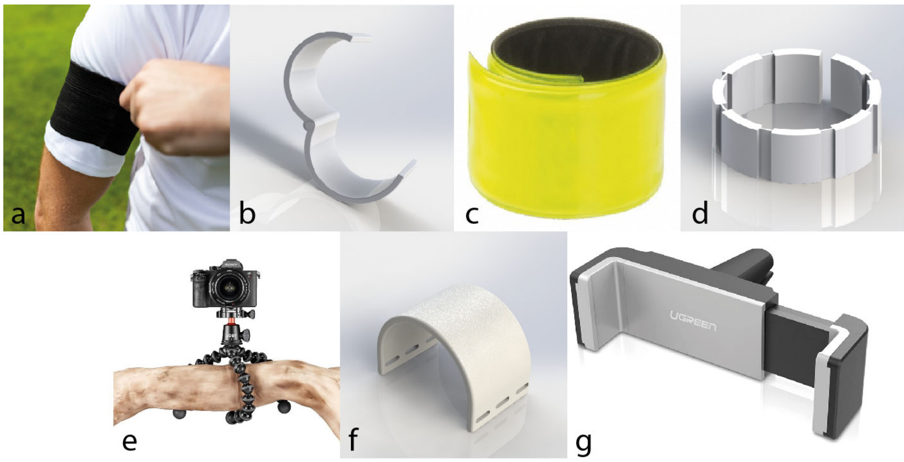
4

Formal concept

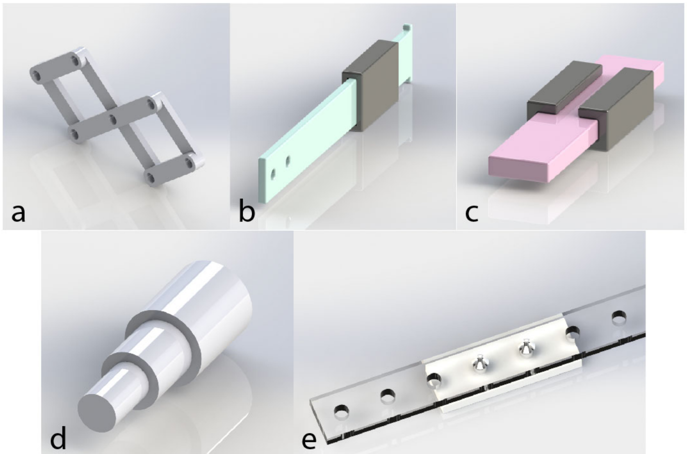
4.1. Generate solutions

In **Chapter 3** the decision was made to attach the simulator to the forearm only. It needs to be attached to the forearm at two points, with a length adjuster between them. The two cuffs need to be able to adapt for different arm sizes. Furthermore, it is important that the simulator cannot fall off during movement, so it needs to be locked as well. The solutions to adapt for arm circumference and arm length were grouped in three categories; the cuff, the length adjuster and locking mechanisms. A cuff or a length adjuster might already have a locking mechanism, but this is not the main function of that part. The distinction of the three functions allows to easily combine different concepts. In literature, two different cuffs were found. The first was a solid cuff, which was strapped to the participants wrist with Velcro [31]. The second was a compliant cuff, this was made of a flexible material that can form around the participants arm [11, 12]. A brainstorm session resulted in more solutions. In **Figure 4.1a** the results of this brainstorm are visualized in photos and drawings. A solid cuff could adapt for different arm sizes in multiple ways. A hinge could be added to the cuff, so that it could fit and be tightened around different arm sizes. Another option was to design a retractable cuff that clamps the arm. An example of a mechanism like this is a phone holder for in the car. The ends of the cuff could retract to adjust to multiple sizes. Another way to make a cuff adaptable for multiple users was by putting multiple (ball) joints in a row. A design that makes use of this solution is the GorillaPod, which is a flexible tripod (designed to resemble gorilla arms). The tripod could be wrapped around an object, which is similar to fixating something around the arm. 3D-printing created the design freedom to replace joints and hinges with flexible structures. By using different infills and structures, a compliant cuff could be formed that could adapt to fold around the arm of the user. Another compliant cuff option could be an elastic band or a fabric strap. The last option that was found during the brainstorm was the snapwrap. This is a bracelet that consists of bistable metal bands. The bands could be formed around the arm.

A length adjuster mechanism was needed for the simulator. The part needs to be strong as it forms the connection between the two arm cuffs and the connection to the terminal device. In literature, one simulator was found that was adjustable for different arm lengths [10]. The simulator was adjusted with a rails. The proximal cuff could be shoved over the rails to adjust. This rails and other solutions that were designed are drawn in **Figure 4.1b**. Most of the length adjusting systems consisted of a form-enclosed part that could slide back and forth. This could be done in two ways; a slider could go in a form enclosed slot and a part could slide on a part that forms a rails. The difference was whether the adjuster was completely enclosed by the slot. Using a rails allowed the design to have parts that stick out, which could then travel through the slot. A way to increase the length difference that the length adjuster could achieve could be by designing a telescopic pole. This is especially useful if the design needs to fit in a small 3D-printer. Another option was to use a system that can compress and decompress, like a harmonica. The harmonica can be designed with multiple hinges and rigid rods, or it could be designed with compliant hinges that can be 3D-printed. The last solution that was found is a prong-in-a-hole system like a baseball cap. This did not allow the user to slide it back and forth, but they could separate the parts and reattach them at the desired length.



(a) An overview of solutions for the attachment of the simulator to the arm. (a) An elastic band that stretches over the arm [36]; (b) A solid cuff with a hinge to adjust for arm size; (c) A snapwrap that forms around the arm [37]; (d) A compliant cuff, the anatomy of the parts allow the design to wrap around the arm; (e) A GorillaPod, many small hinges allow it to wrap around something [38]; (f) A solid cuff; (g) A retractable cuff (in this case a car phone holder) that can adjust for different sizes [39].



(b) An overview of solutions for the length adjuster system. (a) A harmonica, in this case drawn with hinges. A compliant harmonica is also possible. (b) A slider with a form enclosed slot. (c) A rails with a slider. (d) A telescopic pole. (e) A prong-in-a-hole system, the two parts can be separated and reattached to adjust length.

(c) An overview of solutions for locking mechanisms. (a) Pin and holes, the pin can be put in the holes to lock and adjust; (b) A bolt and wing nut [40]; (c) A BOA system, the dial can be used to tighten the laces [41]; (d) A ratchet [42]; (e) Velcro on a glove; (f) Hooks and loops; (g) Triglide slide; (h) Laces on a shoe.

Figure 4.1: Overview of proposed solutions for making the design adjustable for multiple users.

The same locking mechanisms could be used for the arm cuffs and the length adjuster mechanisms. **Figure 4.1c** shows an overview of the solutions that were found. In literature, two locks that were used to attach the simulator to the arm were reported. These were Velcro straps and the BOA system [6, 10]. Velcro sticks to itself and some other fabrics, making it easy to incorporate in a socket. The BOA system is a patented system that uses a dial to fasten laces [41]. The mechanism is a ratchet, which makes it useful to tighten the socket. One lock for a length adjuster was found in literature. This was a bolt and a nut [10]. The parts featured a knob that could be turned, so that no additional tools were needed to adjust the simulator size. When a knob is not available, a wing nut or a regular nut could also be used to lock the mechanism. A source of inspiration for more solutions was found in clothing. Laces, hooks and loops and a triglide slide are common ways of adjusting and fastening pieces that fit multiple people. Laces could be put through holes and tied to secure the simulator. A design with hooks and loops would need multiple loops (or hooks) so that the design is adjustable for multiple participants. The triglide slide, which is commonly used to adjust a backpack strap, could be used in combination with a piece of fabric only. The prong-in-a-hole system that was proposed as a length adjuster mechanism could also be made flexible and used as a lock for an arm cuff. Other solutions could be a ratchet, a mechanism that allows movement in only one direction. The cuff could be released by pulling the lock up and using the other hand to loosen the strap. A non-compliant lock could consist of one (or multiple) holes and a pin. This needs two parts, of which the holes could be aligned. The pin could be used to secure the mechanism by putting the pin through the two holes.

4.2. Evaluate solutions

The proposed solutions need to be evaluated to find the best solution for making the simulator adjustable for multiple users. Two important design requirements for the cuffs and the locking mechanisms were that the mechanism could be operated with one hand and that the mechanism was 3D-printable. Designs that could not be operated with one hand and that were not 3D-printable should be discarded. Designs that were not operable with one hand (but are 3D-printable) should be discarded as well. For adjusting the length of the simulator, being operable with one hand was regarded less important. The length of the simulator could be adjusted before putting the simulator on. However, being 3D-printable was still an important design requirement for this part. To quickly discard the unsuitable options, a C-box grid was created (**Figure 4.2**). The designs in the upper right corner are the ones that are operable with one hand and 3D-printable, these are the interesting solutions that can be worked out further to fit in the final design. The designs in the upper left corner are operable with one hand, but not (fully) 3D-printable. These may form the best solution when the others are not satisfying the design requirements. In order to evaluate the remaining solutions with the other design criteria that are important for adjusting the simulator Harris profiles were made. The full overview of Harris profiles can be found in **Appendix C**. **Table II** shows the grading criteria and the Harris profiles of the three best solutions for each category (the cuff, the length adjuster and the locking mechanisms). The grading criteria were not ranked in order of importance in the table. The best designs were found by making a summation of the scores.

The three best scoring solutions for the arm cuff were a retractable cuff, a solid cuff and a compliant cuff (**Table IIb**). Two cuffs were needed for the design; one at the elbow and one at the wrist. Both cuffs need to fulfil another function and it is not necessary to use the same cuff at both places. The elbow cuff needs to make the design adjustable for multiple users, while the function of the plates restricting pro- and supination remains intact. The wrist cuff needs to stabilize the simulator, so that the design is more comfortable for the user to wear. The three proposed cuffs each have advantages and disadvantages. The solid cuff is less precise and adjustable than the other two, but it is a good solution because it is very durable. The compliant cuff is more precise, but it requires a more flexible material to 3D-print it. This can make the design more expensive and therefore less accessible. The retractable cuff is also precise, but its disadvantages are the strength and that it consists of two parts. A retractable design is very useful for the elbow cuff, because it will make sure that the two plates to restrict pro- and supination will remain in the same orientation. The solid cuff is useful at the wrist. The arm can rotate in the cuff, while the elbow cuff will keep the prosthesis simulator in the desired orientation. The precision at this part is not needed, as long as the cuff is fixated to the arm. Both cuff types do not have a lock from its own. The three best solutions for locking the device are holes and pin, Velcro and the triglide slide (**Table IId**). These three solutions are all not fully 3D-printable. The triglide

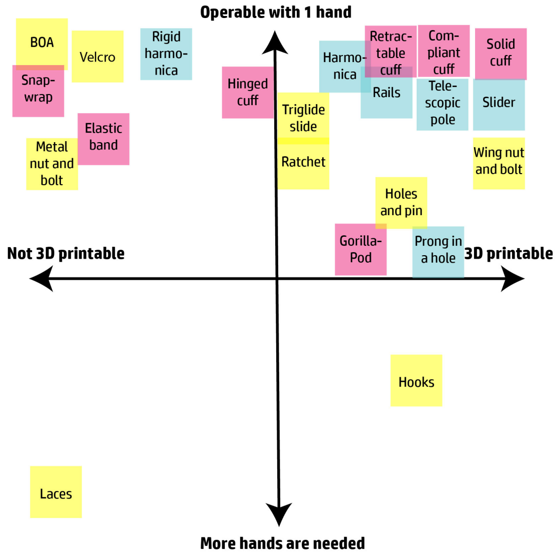


Figure 4.2: A C-box overview of the design solutions for making the design adjustable for different users. The arm cuffs were placed on the grid in red frames. The length adjusters were placed on the grid in blue frames. The locking mechanisms were placed on the grid in yellow frames.

slide needs a strap to attach to something and the Velcro itself is also not 3D-printable. The holes and pin can be made fully 3D-printable, however, it is more durable to use a metal pin. When ignoring the requirement that the design must be 3D-printable, the Velcro is the best solution. This solution can be paired with both cuffs, at the wrist and at the elbow.

The three best scoring solutions for the length adjuster were the slider, the rails and the telescopic pole (**Table IIc**). These three can all be combined with the proposed solutions for the cuffs. The advantage of a telescopic pole is that it does not take up more space than necessary. The downside is that it is made out of many parts. The design requirement for the length adjuster was that it needs to adapt from 20 to 25 cm, so it needs to adjust for at least 5 cm. The building volume of the smallest printer is 18 cm (**Appendix B**). The building volume of the printer is big enough to print the length adjuster at once. This makes the telescopic pole less feasible for the design. The rails and the slider are then the two options that are left. The difference between a slider and a rails is that the slider is stronger than the rails. This is an important feature, as the forces on the length adjuster can be high as that also is the part that connects the terminal device with the simulator. The assumption is that by securing the cuffs around the forearm and by internal friction of the slider, the length adjust does not need a lock. If an extra lock is needed, the best solutions are the holes and pin, Velcro and the triglide slide (**Table IId**). The disadvantage of holes and pin is the amount of parts, but when pairing the holes and pin with a slider system, it only needs one more part (the pin) to work. It can also be hard to operate with one hand, but adjusting the length of the simulator can be done without wearing it, so this is not a problem. The Velcro and triglide slide are less strong than the holes and pin. They are also harder to combine with a slider mechanism. This is why the holes and pin are chosen as lock for the length adjuster.

4.3. Prototype

The first prototype showed that the wall needed to be improved. Furthermore, a smaller tolerance in the sliding system could also prohibit the displacement of the walls. Instead of adding more thickness to the walls, the decision was made to create a ledge around the edges of the walls in order to change the profile and improve the strength. This increases the thickness of the walls (h), which increases the inertia (I) and reduces $|v_{max}|$. The benefit is that this causes a minimal increase of weight, contrary to increasing the overall wall thickness. The ledge increased the wall thickness with 3 mm to 6 mm. The estimated $|v_{max}|$ was then 0.15 mm. In contrast with the first prototype, this prototype should be able to resist pro- and supination of the forearm with the arm in all possible orientations. This is why the plates have a circular shape. **Figure 4.3** shows a Solidworks render of the prototype, the functions of

Table II: The specification of grading criteria and Harris profiles for making the design adaptable for multiple users.

(a) Grading criteria for making the design adaptable for multiple users. These grading criteria can be used for designing the cuff, the length adjuster and the locking mechanisms.

	- -	-	+	++
Use with one hand	Assistance is needed	Two loose parts (but doable)	One loose part	No loose parts
3D-printable	Nothing can be printed	<50% can be printed	>50% can be printed	Everything can be printed
Amount of parts	4 or more parts	3 parts	2 parts	1 part
Easy accesible	Specialized part	3D-print special materials	Common part	Easy to print (with PLA)
Precise	Not adjustable	> 5 mm	5 mm > step > 1 mm	< 1 mm
Strength	One time use	Fragile	Durable	Extremely hard to break

(b) The Harris profiles for adaptable arm cuffs.

	Retractable cuff	Solid cuff	Compliant cuff
Use with one hand	++	++	++
3D-printable	++	++	++
Amount of parts	+	++	++
Easy accessible	++	++	-
Precise	++	- -	++
Strength	+	++	+

(c) The Harris profiles for length adjusters.

	Rails	Slider	Telescopic pole
Use with one hand	++	++	++
3D-printable	++	++	++
Amount of parts	+	+	-
Easy accessible	++	++	++
Precise	++	++	++
Strength	++	+	+

(d) The Harris profiles for locking mechanisms for both arm cuffs and length adjusters.

	Holes and pin	Velcro	Triglide slide
Use with one hand	+	++	++
3D-printable	+	- -	+
Amount of parts	+	++	+
Easy accessible	+	+	+
Precise	+	++	++
Strength	++	++	+

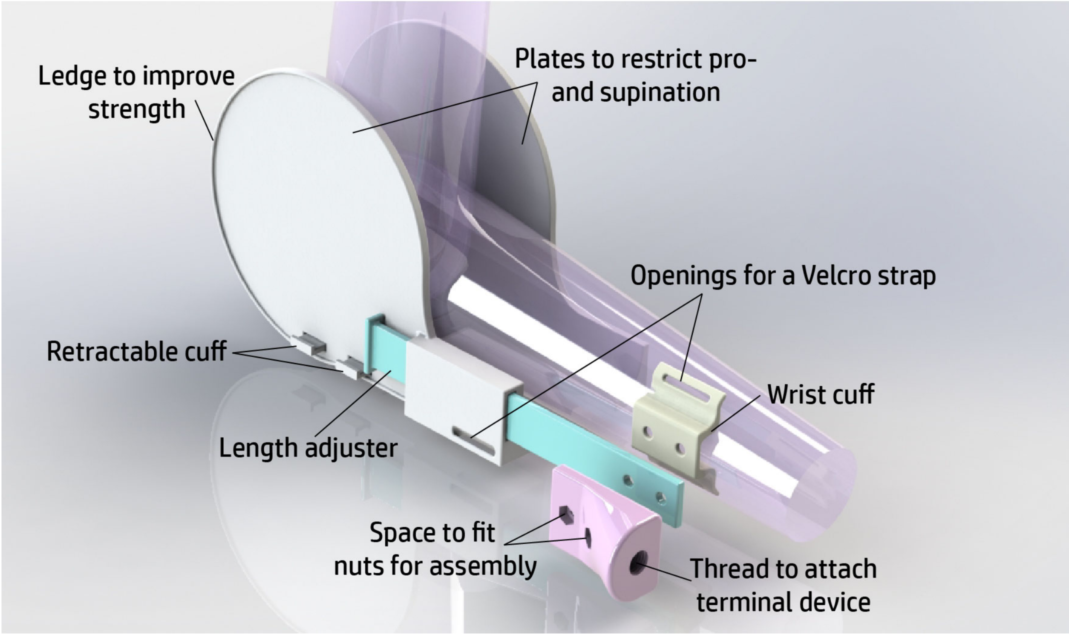


Figure 4.3: Render of the Solidworks drawings of the second prototype with a model of an arm visualized in transparent pink. The function of the parts is indicated in the figure.

the different parts are indicated in the figure. The solution evaluation showed that the cuff at the elbow should be a retractable cuff with a Velcro strap. The first prototype showed that the tolerance of the sliding system was too big. This is why the tolerance was adjusted from 0.5 mm to 0.25 mm for this prototype. For the cuff at the wrist, the decision was made to design a solid cuff. The shape of the wrist is not round, but it is more like an oval. A solid cuff that follows the shape of the flat side of the wrist was designed for this. For people with smaller wrists, the cuff might stick out a bit to the side. For people with bigger wrists, the cuff might be placed on the wrist instead of around the wrist. It should fit most participants with a wrist height ranging from 4 to 9 cm. To attach it to the user, holes were made to put a Velcro strap through. The wrist cuff also has two holes that can be used to put bolts through to connect the wrist cuff with the length adjuster and the part that connects to the terminal device. A form enclosed slider was found to be the best solution for the length adjuster. For this purpose the elbow cuff will be equipped with a slot in which a bar can slide back and forth to adjust the length. In this design it was made without a lock. The hypothesis is that the friction of the slot with the adjuster and the fastening of the socket around the muscle belly of the forearm will be enough to keep the simulator in place. To be able to test this prototype with a prosthesis, a European standard thread with size M12x1.5 was 3D-printed into the part that forms the connection between the simulator socket and the terminal device. This part was designed to be lightweight, by removing bulky material from the sides. The strength of the part is maintained by reducing the amount of sharp edges and keeping the wall thickness around the printed thread high (5 mm). The part also incorporates two hexagonal holes, which make it possible to secure the bolts for assembly without using a wrench. The size of the bolts is M5. The wrist part, the length adjuster and the socket-terminal device connection can all be screwed together with two bolts and nuts. A ledge at the end of the length adjuster prevents the simulator from falling apart after assembly.

The prototype was made using the same 3D-printer properties as the first prototype (**Appendix B**). The prototype needs two bolts and two nuts, with size M5 and length 3 cm. A total of four parts need to be printed. It also needs two pieces of Velcro to be worn by a user. A visual inspection of the printed parts show that everything is printed as intended. Assembly of the device was very quick, only 2 bolts needed to be fastened. The weight of this prototype (including the bolts and nuts) is 300 grams. **Figure 4.4** shows a picture of a user wearing the prototype. Something that is immediately evident is that the Velcro strap of the elbow piece is located too far from the elbow crease. This makes it hard to secure the simulator in the right place. The second observation is that the prosthesis bolt (M12x1.5) is a bit loose when not entirely screwed in. This is confirmed by **Figure 4.4**, where the slightly bigger

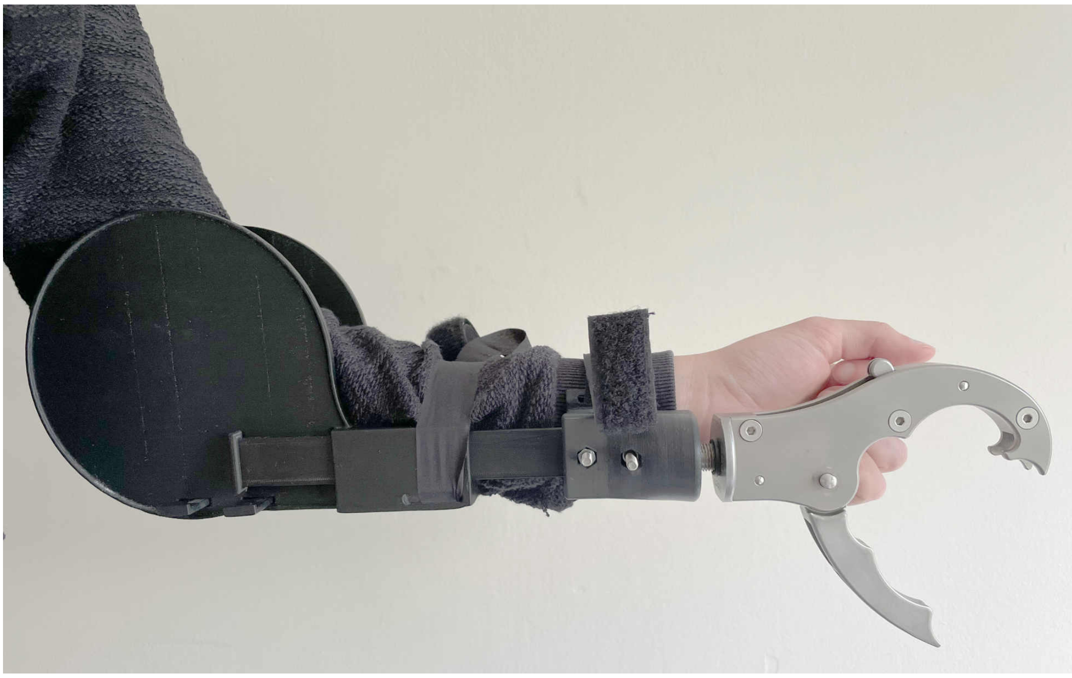


Figure 4.4: The second prototype, which is printed with black PLA on the Ultimaker S5. It can be worn by a user. The thread as printed in the distal end securely fits a M12x1.5 bolt, which means that a prosthesis can be attached to the simulator. The walls are high enough and strong enough to resist rotation of the forearm.

American standard bolt (1/2”-20) could fit in the first millimeter of the socket. One of the goals of this prototype was to conduct an experiment to determine whether the length adjusting mechanisms needs a hole and pin lock to work properly. To test this, the simulator was clamped at the elbow cuff. The length adjuster was retracted to a starting position that was marked with tape. A bucket was attached to the length adjuster with a piece of non-stretching dyneema rope. **Figure 4.5** shows the set-up of the experiment. The mass of the empty bucket is 376 grams. In steps of 10 ml or 10 grams water will be added to the bucket, until the mechanism starts to move. The friction force is equal to the gravitational force, so this can be calculated with $F = m * g$. The mechanism started to move with a total mass of 596 grams. The friction force is therefore $0.596 * 9.81 = 5.85N$. This is lower than the force needed to operate a body-powered prosthesis [43]. The cuffs will also prevent the length adjuster from retracting when operating a prosthesis, but this causes strain on the forearm of the user. This experiment concludes that a lock is needed for the length adjusting mechanism.

Figure 4.5: Experimental setup to determine the friction force between the length adjuster and the slot of the second prototype. In steps of 10 ml, water is added to the bucket until the system starts to slide. (a) The marked starting position; (b) The weight of the bucket when it started to move; (c) The end position.

5 Final design

5.1. Generate solutions

To make the final design usable with a body-powered terminal device, two features need to be added. A design wish was to make the design usable with the two bolt sizes commonly used for prostheses. These were the European standard (M12x1.5) and the American standard (1/2”-20). The other feature that needs to be added is the connection of the Bowden cable to the socket. Because different terminal devices have different bolt sizes, tapping the thread in the socket or using a state of the art prosthetic wrist is not a suitable solution. **Figure 5.1a** shows a solution that allows the use of a 3D-printed thread. This thread is created on both sides of the part. This allows the user to rotate the part 180 degrees to use the thread with the desired size. A second solution is to create a non-threaded hole in the socket that is big enough to fit either bolt size. At the other end of the hole, the user can fasten the prosthesis with the nut that is the right size for the specific prosthesis **Figure 5.1b** . The last proposed solution, that does not require spare parts or changing of parts is a clamp **Figure 5.1c**. By clamping the bolt, the prosthesis will be attached to the socket as well. This solution can also be used as a friction wrist. This allows the user to rotate the terminal device in the desired position.

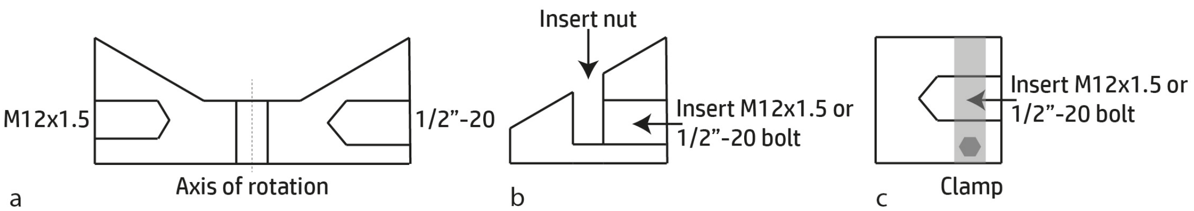


Figure 5.1: Overview of solutions to attach prostheses with two different bolt sizes to the prosthesis simulator socket. (a) A design that can rotate 180 degrees, with two different threads on the ends; (b) A design that replaces the nut; (c) Clamping the bolt so that both sizes fit.

A body-powered prosthesis needs a Bowden cable to operate the device. The housing of a Bowden cable needs to be fixated at both end points to make optimal use of the benefits of such cable [44]. Furthermore, the endpoints of the housing should be directed in the same direction as the force that acts on the cable to minimize dissipation of energy. The housing of the cable usually is a tight spring. Current prosthesis sockets use a custom sized nut to screw on the cable. This nut is equipped with a hook. A slot on the socket then allows the user to attach the cable to the socket. The slots can be put on different places of the simulator socket to give the user freedom in where to attach the cable to the socket. Such a nut is a specialized part, which is not commonly found worldwide. To make the part more accessible a nut with the same properties can be 3D-printed. This nut can be paired with a slot on the simulator to fixate the cable housing in the right place. A second solution that is commonly used is a specialized triceps cuff to which the proximal end of the cable is attached [7]. This gives the endpoint of the cable housing the right direction towards the opposite shoulder, reducing the dissipation of energy. A similar design was patented in 2016 [45]. The cable housing is not fixed to a rigid part of the socket, but to a piece of non-stretch fabric. This offers a big range of motion for the user, while always maintaining the right orientation for the cable. The last proposed solution is to clamp the cable housing. This can be done by creating a form enclosing piece for the cable housing and fastening it with a bolt and nut. This solution also works on cables that have a housing with a smooth surface.

5.2. Evaluate solutions

Three possible solutions were given for making the connection between the terminal device and the simulator. These were a rotatable part, replacing the nut which is located behind a universal hole or a clamp to clamp the bolt instead of screwing it in. The connection to the terminal device is located at the wrist, which is on the distal end of the forearm. It is important that this part is as light as possible, because the terminal device is usually heavy and additional weight at the distal end needs to be reduced for user comfort. This is why the rotatable connection part is not a suitable solution for the connection of the terminal device. This part is estimated to be twice the size of the other solutions, making it twice as heavy as well. The solution that uses a universal hole with a nut that can be replaced is a good solution, but the downside is that it requires the user to carry around a spare nut. Especially at big research facilities or schools, these spare parts create a great risk of losing them. The proposed solution is therefore to clamp the bolt. An advantage of clamping the bolt is that it can be used as a friction wrist, which gives the user the possibility to determine the orientation of the prosthesis. The clamp can be secured with a bolt and a nut. It is possible to use a butterfly nut, which allows the user to adjust the tension on the clamp without any tools.

The three solutions to fixate the Bowden cable are all able to secure the housing. The main differences are the amount of parts and the possibility to 3D-print the parts. The solution that uses the least amount of parts and may be fully 3D-printed is the nut to screw on the cable. An attempt was made to 3D-print the nut for the cable to test the solution. The outer diameter of the cable was measured to be 5 mm and the thread size was measured to be slightly less than 1 mm. Two test blocks were printed both with a female thread with outer diameter 5 mm and one with a pitch of 0.9 mm and one with a pitch of 1.0 mm. Both 3D-printed nuts did not fit on the cable, meaning that another solution was necessary. The two remaining solutions are to clamp the cable or to use a triceps pad or piece of fabric to fixate the cable. At the distal end, it is necessary to clamp the cable, because the other solutions are only usable at the proximal end. This clamp can be fixated with a bolt and a nut, the clamp itself can be 3D-printed. To decrease the amount of parts that cannot be 3D-printed, the decision was made to also clamp the bolt at the proximal end.

5.3. Final design

Figure 5.2 shows the Solidworks render of the final design. The designs of the wrist cuff and the plates to restrict pro- and supination are the same as in the previous prototype. The other parts have been changed or are new. The previous prototype showed that it was not comfortable for the user to wear the design without locking the length adjuster. The best locks to be combined with a slider were evaluated in **Chapter 4**. The proposed solution was a pin and holes lock. In the final design, the holes are 3.5 mm in diameter. In this way they fit a pin with a 3 mm diameter. A M3 bolt can be used as a pin. To prevent breaking of the holes, the holes are placed at a distance of approximately 3 mm from each other. The centers of the holes are separated 6 mm. There are 10 holes, so the length can be adjusted over 6 cm (with a precision of 6 mm). The length adjuster needs to be strong enough to resist deflection and torsion, caused by using the simulator. The thickness of the length adjuster needs to be high enough to add the 3.5 mm holes. The minimum thickness of the length adjuster is 9 mm, which makes the walls around the holes approximately 3 mm thick. **Appendix E** contains the Matlab code for calculating deflection and torsion of the length adjuster. The results show that the length adjuster is strong enough at a thickness of 9 mm. The slot for the length adjuster at the elbow cuff is moved more towards the elbow. The Velcro band of the elbow cuff is also moved to a location at the elbow crease instead of halfway through the forearm. This improves user comfort and the additional benefit is that the design requires less material. The proposed solution for connecting the terminal device to the socket was to clamp the bolt. **Figure 5.3a** shows the shape of the clamp. The dotted lines indicate where the clamping bolt is located. The shape of the clamp allows stresses to be distributed over the entire part. The wall thickness around the bolt is 14 mm. This also reduced the accumulation of stresses at one critical point. The last parts that were added in this design were the parts to fixate the housing of the Bowden cable. The housing of the Bowden cable needs to be fixated to the socket on both ends of

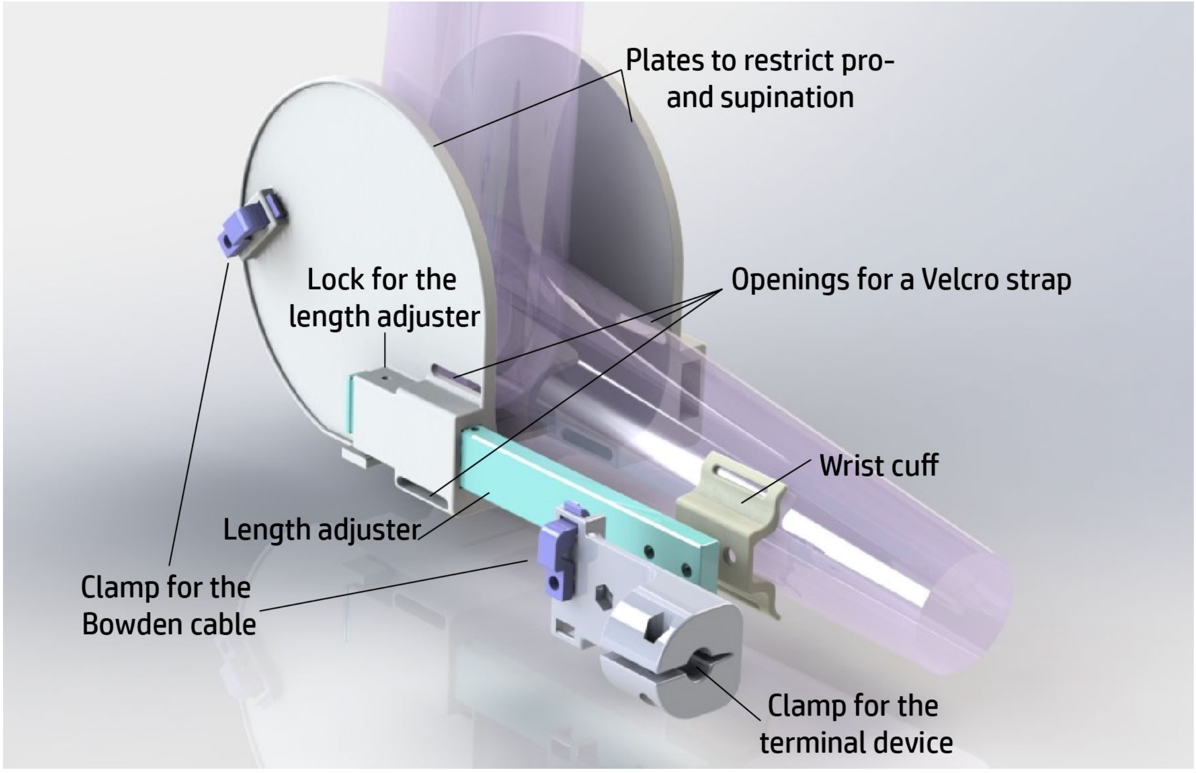
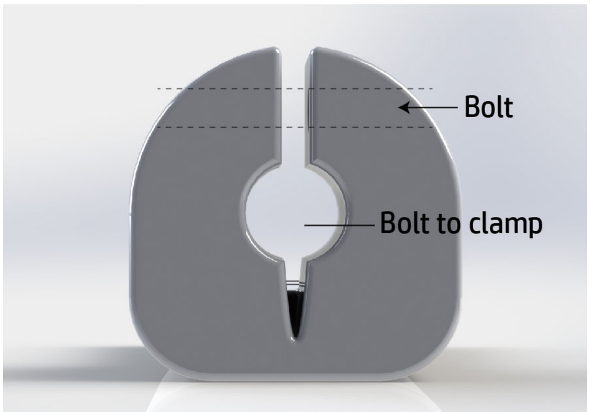
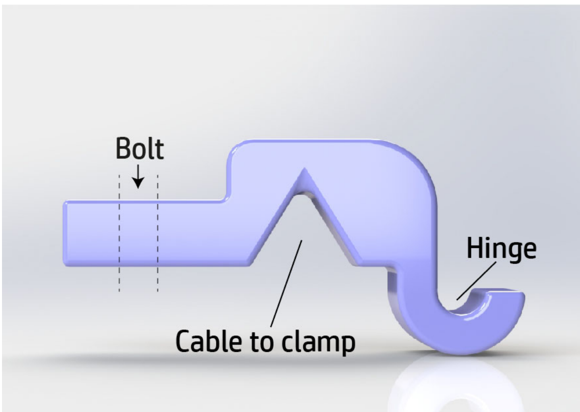


Figure 5.2: A Solidworks render of the final design. The functions of the parts are indicated in the figure.



(a) A clamping mechanism to fit two common prosthesis bolt sizes (M12x1.5 and 1/2-20 inch). The dotted lines indicate where the clamping bolt will be. This design distributes the stresses evenly, so that the part does not break when tightening the bolt.



(b) Side view of the clamping mechanism to clamp the cable housing. The triangular shape makes sure that the cable is properly clamped. A hinge on the right side makes assembly easier. The place for the bolt is indicated with dotted lines.

Figure 5.3: The profiles of two parts needed to make the final design usable with a (body-powered) terminal device.

Table III: Part list of the final design. This includes the weight of each part. The table is split in simulator parts and parts needed to fit the TRS Grip 2S.

Simulator parts	**Quantity**	**Weight (g)**
Wrist cuff	1	11
Prosthesis connection piece	1	51
Length adjuster	1	39
Elbow part left and right	2	215
Bolts M5	3	21
Nuts M5	3	6
Bolt M3	1	4
Velcro band	2	9
Total	**14**	**356**

Prosthesis specific parts	**Quantity**	**Weight (g)**
TRS Grip 2S	1	353
Cable and harness	1	100
Bolts M5	1	7
Nuts M5	1	2
Clamps	2	6
Total	**6**	**468**

Figure 5.4: A user is wearing the final design. A TRS Grip 2S terminal device with a figure-of-9-harness is fitted to the simulator.

the cable in order to operate the prosthesis without any problems [44]. The proposed solution was to clamp the housing of the cable. **Figure 5.3b** shows the design of this clamp. To reduce the amount of parts, the clamp hinges around a rod and is secured with one bolt and nut, instead of with a bolt and a nut on two sides. At the wrist unit, this can be combined with the already existing bolt. But at the elbow unit, a separate attachment is needed. This attachment point is placed on the plates. The angle of the clamp with regard to the upper arm is 45 degrees when the elbow is flexed at 90 degrees. This is the most optimal direction for the cable towards the contralateral shoulder when the clamp is rigid and the design needs to be symmetrical. The profile of the clamp shows that the clamp has a triangular hole. This was done instead of a circular hole to create more contact points between the cable and the clamp. The shape of the triangle was optimized so that cable housing with a 5 mm diameter could be clamped properly.

Technical drawings of the final design are printed in **Appendix D**. The design was printed on two different printers. The elbow cuffs were printed on the Ultimaker 3. The length adjuster and the terminal device-simulator connection part were both printed on the Ultimaker S5. The material that was used for the prints is PLA, with PVA as support material. Inspection of the print showed that the bottom of the elbow cuff was warped. This happens when the material shrinks during printing [46]. The warp also caused the walls to be not entirely straight. This deviation was minimal and the design could still be worn by a user (**Figure 5.4**). The prosthesis simulator socket weighs a total of 356 grams. This is excluding the parts needed for the body-powered prosthesis (harness, cable and TRS Grip). The total weight of the simulator, including the terminal device, is 824 grams. **Table III** shows the part list of the final design, with the weight per part.

5.4. Materials

One of the design requirements was that the design needs to be 3D-printable. One of the most commonly used materials to 3D-print prostheses and orthoses is PolyLactic Acid (PLA) [47]. This can be used in combination with the water-soluble PolyVinyl Alcohol (PVA) as support material. PLA is safe for direct contact with the skin. A second material that is often used for prostheses is Acrylonitrile Butadiene Styrene (ABS). Both materials can be used on FDM printers, which are the most common ones to use [47]. Some prostheses are printed with flexible materials, like polyurethane. This material is more expensive than PLA and ABS. In order to keep the prosthesis simulator accessible for all groups PLA was chosen for the prosthesis simulator. PLA or PVA can be used as support material. Using PVA as support material makes it easier to remove the material afterwards. The design also allows PLA as support material by following the 3D-printing guidelines for prostheses from Cuellar et al. [48]. Another reason why using PLA is preferred is that it can be made of corn and is biodegradable, which reduces the impact on the environment [49]. Most parts of the design were 3D-printed with PLA; the elbow cuff, the wrist cuff, the length adjuster, the clamps for the Bowden cable and the connection for the terminal device. The decision was made to use off-the-shelf solutions for the other parts. The bolts and nuts technically can be 3D-printed, but the decision was made to use metal ones. These are more durable and stronger. They are also very common all over the world, so it does not reduce accessibility of the design. The other part that was not 3D-printable is the Velcro. This is also a common and non-expensive part. When necessary it can be replaced by a fabric strap, which can be tied around the arm as well. **Table IV** shows the list of materials and the cost of the parts. It also shows the supplier that they were obtained from in this thesis.

Table IV: Cost of materials for the prosthesis simulator socket. The parts to include a terminal device are not included, as they can be very different for different terminal devices.

Part	**Quantity**	**Cost**		**Supplier**
PLA Filament	322 grams	€ 6.43	€19.95/kg	**Meer3D**
PVA Support material	114 grams	€ 9.11	€39,95/500g	**Meer3D**
M5 Nuts	3	€ 0.03	€1/100pc	**Toolstation**
M5x30 mm Bolts	3	€ 0.16	€0.51/10 pc	**Toolstation**
M3x20 mm Bolts	1	€0.06	€1.35/25pc	**Toolstation**
Velcro band	1 m	€0.36	€0.72/2m	**Action**
Total:		**€16.15**		

6

User evaluation

6.1. Participant information

Due to COVID-19 it was not possible to do user testing with different participants. The ethics committee was not able to approve the research due to the strict measures at the time. Instead of testing the simulator with participants, it was tested by the author. The participant was 24 years old, female and right handed. She had no prior experience with using a body-powered (or myoelectric) prosthesis simulator. **Table V** contains the anthropometric measurements of the participant before the evaluation. These measurements could be used to determine the adjustability of the prosthesis simulator. Before the start of the trials, the requirements of resisting pro- and supination of the arm and leaving flexion and extension free were tested. **Figure 6.1** shows the participant wearing the prosthesis simulator. The participant was able to fully flex and extend the elbow while wearing the simulator (**Figure 6.1a and b**). The participant was also able to fully pro- and supinate the forearm. **Figure 6.1c–e** show that the orientation of the terminal device was not changed by pro- and supinating the arm.

Table V: All data and characteristics of the participant who tested the simulator, measured prior to the experiment.

(a) Anthropometric measurements of the participant.

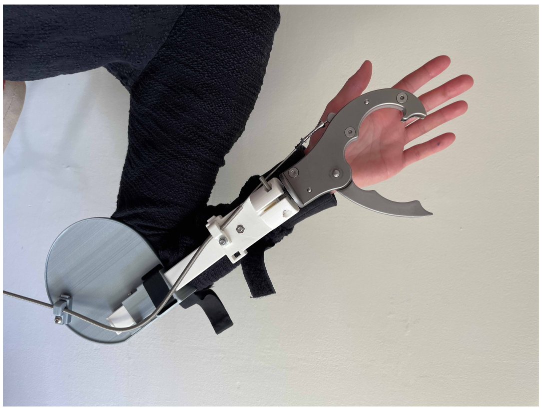
Anthropometric measurements	Value
Upper arm circumference (at the elbow)	23 cm
Upper arm length	25 cm
Forearm circumference (at the elbow)	24 cm
Forearm length (elbow crease to wrist)	20 cm
Wrist circumference	15 cm
Hand length	17 cm

(b) Characteristics of the participant.

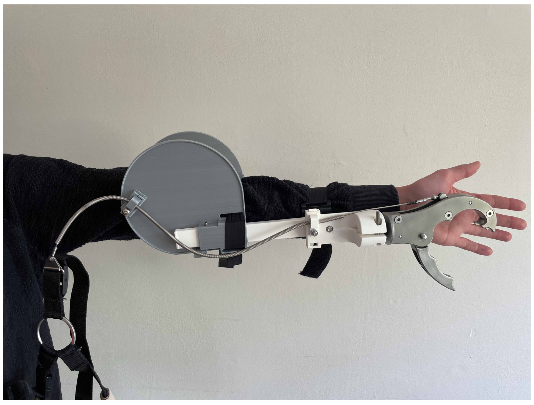
Data	Value
Dominant hand	Right hand
Age	24 years
Sex	Female
Skill level	No prior prosthesis use

6.2. Adapt for user

The elbow cuff can be adjusted from 5 to 10 cm wide. Arm circumferences of approximately 17 to 30 cm fit. The wrist cuff can be used for wrist heights that range from (at least) 4 to 9 cm. Wrist circumference is less relevant, as the size of the Velcro band is the limiting factor. The length adjuster adjusts the simulator length from 18.6 to 24.6 cm, measured from the terminal device clamp to the locking pin (which is approximately at the elbow crease). The sizes obtained from the user show that the simulator could be adjusted properly to the user (**Table V**). To measure how long it takes to adjust the simulator for the user an experiment was done. The simulator was put in one of two starting positions. The participant was required to fit the prosthesis to her left (non-dominant) arm as fast and accurate as possible. One of the starting positions was the widest one that was possible; with all straps loose, the length adjuster fully extended and the elbow piece fully extended as well. The other was the smallest one that was possible; with the straps fully tightened, the length adjuster completely shoved in and the elbow piece at the smallest size. In order to assess how easy the simulator is to adjust, the simulator was put in both positions three times. The first position was chosen randomly, the other five trials



(a) The maximum amount of flexion that was possible while wearing the simulator. The intact arm was not restricted by the simulator.



(b) The maximum amount of extension that was possible while wearing the simulator. The intact arm was not restricted by the simulator.

(c) The elbow was flexed to 90 degrees and pro- and supination was neutral.

(d) The elbow was flexed to 90 degrees and the forearm was fully pronated.

(e) The elbow was flexed to 90 degrees and the forearm was fully supinated.

Figure 6.1: Range of motion of the intact arm while wearing the prosthesis simulator. (a) and (b) show maximum flexion and extension; (c)–(e) show a neutral position and a fully pronated and supinated position. The user was able to achieve full range of motion for all five movements. The orientation of the prosthesis does not change by pro- or supinating the arm.

Table VI: Time recorded that was needed to put the simulator on and adjust it starting from the biggest and the smallest position.

Starting position		**Biggest**			**Smallest**	
Time recorded	**Total**	**Adjust length**	**Put on**	**Total**	**Adjust length**	**Put on**
Trial 1	**01:12**	00:22	00:50	**00:49**	00:29	00:20
Trial 2	**00:41**	00:19	00:21	**00:37**	00:15	00:22
Trial 3	**00:31**	00:09	00:22	**00:28**	00:10	00:17

alternated between the positions. The participant timed her own trials. Standing up was allowed, but the user remained seated during the entire task. **Table VI** shows the scores of the six trials. There is a distinction in the time it took to adjust (and secure the length) and the time needed to put the simulator on and secure the straps. The first position was the biggest position. During the first trial, the pieces of Velcro were completely loose and out of the holes of the simulator. This was not repeated in the other trials. Three observations were made during the experiment. In the smallest position the cable gets "stuck" under the elbow cuff adjuster. Pushing the elbow cuff out far enough to be able to adjust the length apparently made it approximately the right size for the participant. The second observation was that it is easiest to put the simulator on while it is laying on a flat surface like a table. The last observation was that taking the simulator off (not measured) was very easy. When the Velcro was loosened a little, the arm could slide out of the cuffs. This was measured once and took only 6 seconds.

6.3. Functional performance

In order to test the functional performance of the Delft Pro Simulator, two standardized test were performed. These tests were the Box and Blocks Test (BBT) and the Nine Hole Peg Test (NHPT). The experiment protocol was based on the protocol used in Haverkate et al. [7]. The simulator was worn on the non-dominant arm of the participant, which is the left arm. The harness was adjusted so that the prosthesis was slightly engaged when the arm was in a neutral position with the elbow extended on the side of the body. During the tests the participant was seated, the chair height made it possible to hover the simulator over the table with the elbow at an angle of 90 degrees. The BBT requires the participant to move as many blocks from one box to the other in 60 seconds. The boxes are separated by a divider (**Figure 6.2**). The score is the number of displaced blocks during the 60 seconds. The NHPT requires the participant to move the pegs from the reservoir into the holes. When all the pegs are in the holes, they need to be returned to the reservoir. The test is stopped after 90 seconds to prevent fatigue. The score is the time per peg displacement. The BBT and NHPT were administered alternately. The participant had a break after every six trials. During this break the prosthesis simulator was completely removed from the arm. The six trials were repeated three times, so that in the end, there were 18 trials in total. **Table VII** shows the scores of all trials. The participant reported that the compensating movements were causing the upper arm to get tired during the trials. The ab- and adduction of the upper arm that can be seen in **Figure 6.2** compensated for the lack of pro- and supination. The results of the experiment were plotted over the results of the experiment by Haverkate et al. (**Figure 6.3**) [7]. The figure shows that most trials would be outliers in the study of Haverkate et al., while the protocol was the same. A learning curve is visible from trial 1 to 7 for both tests. In trial 8 and 9 the results seem to be reaching a plateau. The NHPT shows two outliers from the general trend at trial 4 and 5. The participant reported that the pegs were lying in the wrong orientation during those trials and it was hard to put them in the right orientation.

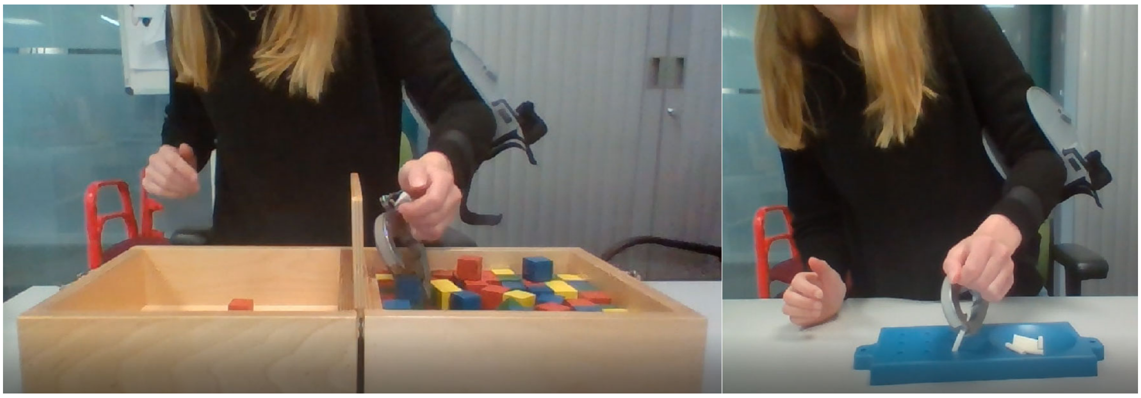


Figure 6.2: Performing the Box and Blocks Test (left) and the Nine Hole Peg Test (right) with the Delft Pro Simulator with the body-powered TRS Grip 2S attached.

Table VII: Results of the Box and Blocks Test and the Nine Hole Peg Test.

Trial number	BBT (block movements)	NHPT (peg movements)	NHPT (s/peg displacement)
1	6	4	22,5
2	8	3	30
3	9	4	22,5
	break		
4	13	2	45
5	11	2	45
6	15	7	12,9
	break		
7	17	8	11,3
8	11	4	22,5
9	15	4	22,5

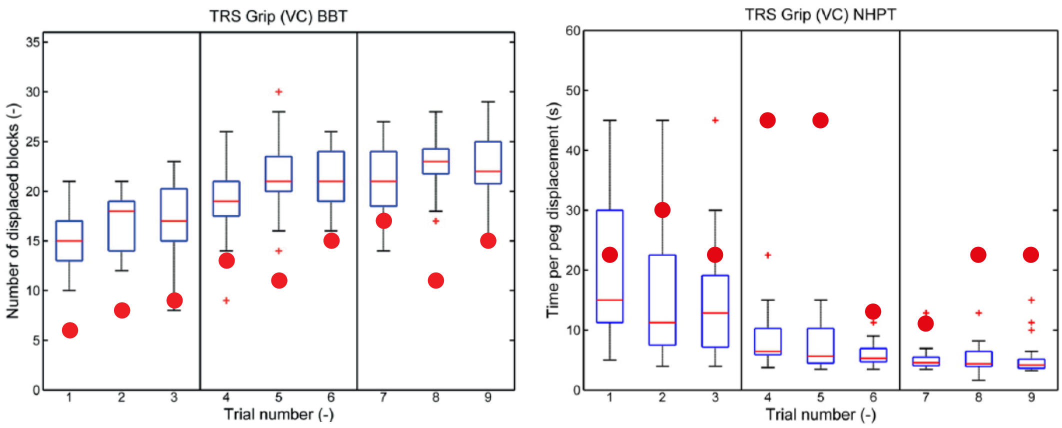


Figure 6.3: The results of the 9 trials of the Box and Blocks Test and the Nine Hole Peg Test plotted as red dots over the results of Haverkate et al. [7]. The Box and Blocks Test shows a learning curve, where the Nine Hole Peg Test results appear to be more random.

7

Final design evaluation

7.1. Calculation: Displacement of center of mass

One of the requirements was that the center of mass of the forearm should not displace more than |35 mm|. **Equation 7.1** shows the formula for calculating the center of mass of a system. In this equation M is the total mass of the system, x is the location of center of mass of a part and m is the mass of a part. This results in x_{com}, which is the center of mass of the entire system. The Matlab script in **Appendix E** was used to perform the calculations.

$$x_{com} = \frac{\sum_{i=1}^{N} x_i m_i}{M} \tag{7.1}$$

The parameters that were used to determine the center of mass without the simulator were obtained from the article by De Leva [24]. The assumption was made that the center of mass of the arm in the lateral direction is in the middle of the arm. The origin of the coordinate system was placed at the elbow. The absolute displacement of center of mass without a terminal device was calculated first. The center of mass displaces 21.56 mm for male users and 26.94 mm for female users. Then the center of mass displacement with the TRS Grip 2S was calculated. This displacement is 19.65 mm for male users and 23.99 mm for female users. The center of mass displaces more for the average female than for the average male, this is due to differences in body composition. The center of mass moves towards the elbow in the axial direction and towards the side of the terminal device in the lateral direction. **Appendix E** provides a table with all calculated values, including the original position of center of mass and the relative displacement in the axial and lateral direction.

7.2. Experiment: Simulator strength

The goal of the experiment was to test the strength of the simulator. The simulator should be usable with commonly used body-powered terminal devices. To achieve this, it needs to withstand a lifting and a pulling force of at least 60 N. Furthermore, the cable force should not deform the simulator. The band around the forearm allows the user to rotate the arm inside the prosthesis simulator, which is why an isolated strength test of the walls was not necessary anymore. In order to test the pull and lifting strength, a user was asked to pick up a bucket filled with water. This was done in two conditions. One was with the elbow flexed in a 90 degree angle and one was with the arm fully extended which is similar to pulling. The experiment started with lifting an empty bucket (376 grams). The weight was increased by adding water in steps of 500 grams. Lifting with an extended and a flexed arm was done alternately. The maximum weight that was lifted with an extended arm was 8 kg of water. Adding the weight of the bucket gives a total weight of 8.38 kg. This gave a maximum pulling force of $8.38 * 9.81 = 82.16$ N. The maximum weight that was achieved with the elbow flexed was 5.5 kg of water. Adding the weight of the bucket gives a total weight of 5.88 kg. The maximum lifting force was therefore $5.88 * 9.81 = 57.64$ N. The photos in **Figure 7.1** show the final trial for both conditions. There are no visible deformations for either condition. The right photo in **Figure 7.1** shows that the user is unable to fully close the terminal device while holding the weight.

Figure 7.1: The final trial for the experiment to determine the maximum weight that could be carried with the simulator as a pulling force (left) and a lifting force (right). The simulator shows no deformations for both conditions.

7.3. Experiment: Material testing

The goal of the experiment was to determine whether the material is influenced by moisture and saline and cleaning with a 70% alcohol cleaning solution. In order to test the breakdown of PLA, a part of the second prototype was weighed. The original weight was 28 grams. It was then put in a salty bath, which consisted of water with 0.9 gram of salt per liter. This corresponds to the amount of salt in sweat [50]. After 7 days (168 hours), the part was taken out of the bath, rinsed with fresh water and dried. After this it was weighed again. **Figure 7.2a and b** show a side by side comparison of the piece before and after submerging it in the saline solution. The experiment shows that after 168 hours of contact with moisture and saline, the product was not damaged at all. The material had the same weight and the appearance had not changed. The same experiment was done to assess the influence of cleaning it with alcohol. For this experiment, a piece of PLA was put in a 70% alcohol cleaning solution. It was left submerged in this solution for 7 days (168 hours). Afterwards the piece was rinsed and allowed to dry for at least an hour. **Figure 7.2b and c** show the side by side comparison of this experiment. The piece of material had the same weight as before, which shows that the material was not affected. However, the material did change colour, the colour went from black to dark grey. The complete experiment shows that the material can withstand moisture for 14 days or 336 hours without the material breaking down.

7.4. Evaluation of design requirements

Table VIII shows the design requirements and whether they were met by this design. When applicable, the measured or calculated value of the requirement was also reported. One of the design requirements was not tested. The simulator should be used at least 100 hours without breaking. The simulator was only used by one participant in this thesis. The total wearing time was approximately 3 hours in total. There were no deformations visible when performing the strength tests with the simulator, indicating that the lifespan should be long during normal use. The moisture and saline experiment showed that the lifespan is not influenced by moisture and saline. However, the total wear time was too short to draw an informed conclusion about the lifespan of the simulator. Two design wishes could not be tested. There was no myoelectric prosthesis available to test the simulator with. The prosthesis only covers

(a) Before submerging the piece in the saline solution. The weight is 28 grams, which will be used to determine whether there was any breakdown of material during the experiment.

(b) After submerging the piece in a saline solution (0.9% salt). The weight of the piece remained the same, which shows that no material was lost. The piece is not discoloured and visual inspection shows no signs of other damage.

(c) After submerging the piece in 70% alcohol for 168 hours. The weight of the piece remained the same, which shows that no material was lost. The piece discoloured during this experiment. The black PLA turned dark grey. Visual inspection shows no further signs of damage.

Figure 7.2: Side by side comparison of the piece that was used to determine the effect of soaking it in a saline solution for 168 hours and afterwards soaking it in an alcohol solution for 168 hours. The piece shows no signs of damage after the saline bath and is discoloured after the alcohol bath, with no other damage visible.

about 90 cm^2 of the skin of the forearm and 234 cm^2 of the upper arm. This gives a lot of space to place the electrodes on the skin. Furthermore, the slot for the length adjuster on the opposite side of where the prosthesis hand is placed can be used to attach a battery pack to. This places the weight in the proximity of the body. The expectation is that with some minor adaptations, use with a myoelectric terminal device should be possible. Another wish was that more than 50 % of the parts are recyclable. 3D-printed material can be molten and then made into new filament. The bolts and nuts are made of steel, which means they are very durable. If one of the bolts or nuts breaks, it can be heated and then formed into something else. Only the Velcro is hard to recycle, but small pieces may be reused for other purposes. Discarding the material can be done by composting the PLA. PLA was made from corn, which makes it a biodegradable material. Under the right circumstances it can be decomposed into water and carbon dioxide in 47 to 90 days [49].

A few requirements were met partially. The length adjuster is able to adjust the length from 18.6 to 24.6 cm. The design requirement was that it needed to adjust length from 20 to 25 cm. Even though the desired range was possible (it was adjustable over 6 cm instead of 5 cm), the design did not reach the desired 25 cm. This might place the terminal device in a sub-optimal position for individuals with longer than average forearms [24]. The difference between 24.6 and 25 cm is only 4 mm, which is smaller than the precision of the locking mechanism of the length adjuster. This is why the final design was accepted. Another requirement that was met partially is that the design is not fully 3D-printable. For some aspects of the simulator, the decision was made to use an off-the-shelf solution instead of a 3D-printed solution. It is possible to 3D-print a thread (or a complete bolt and nut). But using a metal bolt and nut is more durable. Especially because the resulting product is dependent on the quality of the printer. The Ultimaker S5 can achieve a higher resolution than some other printers. This makes the quality of threads on other printers questionable and may cause the design to fail. The other not 3D-printable part is the Velcro band, which is used to secure the simulator to the participants arm. The decision to use Velcro instead of a compliant solution was made to increase accessibility as well. For a compliant solution, a more flexible (and expensive) 3D-printable material is needed. Furthermore, deformation of the material over and over again makes it less durable than using rigid parts and connecting it with Velcro. The last requirement that was met partially was that the material of the simulator discoloured after 168 hours in a 70% alcohol solution. This does not form a problem for use, as usually the parts are not submerged for such a long time, but only wiped with the cleaning solution.

Table VIII: The design requirements that need to be met by the prosthesis simulator. Some requirements are indicated as "wish", this means that they can make the design better but are not required for the design to work properly.

Description	**Requirement**	**Requirement fulfilled (measured value)**
Connect to socket	Wish: Fits both M12x1.5 and 1/2"-20 bolts	Yes
Use with myoelectric terminal device	Wish: Space for battery (size: 10x30x55mm)	Not tested
	Wish: Space for electrodes on arm	Not tested
Use with body-powered terminal device	Guides shoulder harness cable towards opposite shoulder	Yes
	Bowden cable fixated at two points	Yes
Position terminal device	Center of mass displaces ≤ |35 mm|	Yes (26.94 mm for female users and 21.56 mm for male users)
Weight	≤ 500 grams (excluding terminal device)	Yes (356 grams)
Adaptable socket to fit multiple users	Forearm length of 20 to 25 cm	Yes partially (18.6 to 24.6 cm)
	Circumference 20 to 25 cm (arm)	Yes (15 to 30 cm)
	Circumference 12 to 17 cm (wrist)	Yes (all sizes)
	Wish: Change length in ≤ 1 min	Yes (slowest trial: 29.3 s)
	Wish: No tools are needed to adjust	Yes
Easy to wear	Possible to put on with one hand	Yes
	Wish: Put on and secure in ≤ 1 min	Yes (slowest trial: 50.4 s)
Degrees of freedom intact arm	Pro- and supination constrained	Yes
	Flexion and extension elbow free	Yes
Accessibility	3D-printable	Yes mostly (all parts except the bolts, nuts and Velcro)
	≤ 15 parts (excluding terminal device)	Yes (14 parts)
	Material cost ≤ €100	Yes (€16.15)
Material	Skin-safe	Yes
	Withstand moisture and saline	Yes
	Withstand disinfectant	Yes partially (material discoloured)
Lifespan	≥ 100 hours of use	Not tested
Discard	Wish: ≥ 50% of parts are recyclable	Yes (all parts except the Velcro)
Strength	Constrain forearm torque ≥ 4.6 Nm	Yes
	Withstand lift/pull force ≥ 60 N	Yes partially (lift 57.64 N and pull 82.16 N)
	Withstand cable force	Yes

8 Discussion

8.1. Final design

Table VIII gives an overview of the design requirements and whether they were met by the design. The evaluation in **Chapter 7** showed that the design was able to fulfil most of the design requirements and wishes. It was not possible to test whether the lifespan of the simulator is longer than 100 hours of use. The material testing showed that the material could withstand this duration of use, but unexpected factors that could influence the lifespan like dropping the simulator were not tested. Some design requirements were fulfilled partially. These did not influence the usability of the design for the purpose of this study. One of the design requirements that was met partially was the strength of the design. The requirement that the design should withstand a lift force greater than 60 N was measured with the TRS Grip 2S and a user wearing the design. A bench-top experiment would have been a good addition to this experiment, as the design showed no signs of deformation at a lift force of 57.64 N. The design may have been able to withstand a higher force, but this was not measured. The user and/or terminal device was the limiting factor. The pull force that is possible with the simulator is higher than the requirement at a maximum of 82.16 N. Another important observation of this experiment is that the simulator does not fall off quickly, even though the cuffs are only secured around the forearm. The length of the socket was 0.4 cm shorter than required. This was accepted, because the difference is approximately the same size as the precision of the lock for the length adjuster. The precision is 0.6 cm, which means that the simulator can be 0.3 cm too short or too long in the least optimal scenario.

The user testing experiments showed that the simulator is able to resist pro- and supination while leaving flexion and extension free. The final design was user tested with a TRS Grip 2S with a figure-of-9 shoulder harness. The Box and Blocks Test and Nine Hole Peg Test were performed. The user reported fatigue of the upper arm during the tests, because she was compensating for the lack of pro- and supination of the forearm. The mass of the simulator socket can also cause discomfort. The mass of the socket was 3 grams more than the terminal device that was used during the trials (respectively 356 and 353 grams). The expectation is therefore that the mass of the socket caused less fatigue of the arm than the compensating movements or the mass of the terminal device. The performance of the seventh trial was the highest for both tests. The eight and the ninth trial were lower. This is prescribed to fatigue of the arm. The Box and Blocks Test shows a learning curve, while the Nine Hole Peg Test looks more like random trials. Possible explanations are that the Nine Hole Peg Test is harder to perform than the Box and Blocks Test. The pegs need to be rotated from a horizontal to a vertical orientation to be put in the holes. The dexterity needed for this test is higher than the dexterity needed for the Box and Blocks Test. The required rotation of the pegs can also add more compensating movements, causing the user to get tired more easily.

8.2. Comparison with the state of the art

The orientation of the prosthesis is not influenced by pro- and supination of the intact arm. A recent literature study found only two prosthesis simulators that were also able to resist pro- and supination. These designs also influence flexion and extension, while the Delft Pro Simulator does not influence these movements. The highest measured score of the BBT was 17 blocks per minute. The highest measured score of the NHPT was 11.3 seconds per peg displacement. When comparing the results of this study with the results of Haverkate et al. [7], the score of the Delft Pro Simulator is consequently lower with the same terminal device (**Figure 6.3**). The participant reported that her upper arm got tired from compensating for the lack of pro- and supination. In personal correspondence with the authors of Haverkate et al. it was confirmed that the pro- and supination of the forearm was not restricted by their prosthesis simulator. It is possible that the participants of that study were able to rotate the terminal device by pro- and supinating their intact arm to perform the tasks more easily. In that case, the Delft Pro Simulator simulates the transradial limb absence better, causing a lower performance. Another difference between the state of the art simulators and the Delft Pro Simulator is the number of possible ways to use the simulator. The connection between the terminal device and the socket makes it possible to use all commonly used terminal devices, with the American or the European standard bolt. This connection offers the possibility to be used as a friction wrist as well, giving the user the opportunity to choose the optimal orientation for the terminal device. There are two connection points for the cable for a body-powered prosthesis. This means that the cable can originate from the terminal device from three (palmar, dorsal and ulnar/radial) positions without any issues. The design is completely symmetric, which means that it can be worn on both the left and the right arm without any adaptations. Using a myoelectric terminal device should also be possible with a minimal amount of adaptations. In literature, no prosthesis simulators were found that could fit both a myoelectric and a body-powered prosthesis on the same simulator socket. No designs were found in the literature that allow the user to choose the terminal device position.

8.3. Strengths and limitations

A way to validate the simulation is to reproduce a study that was performed by people with upper limb absence who were using a prosthesis with a prosthesis simulator. A successful simulation should give the same results. However, there were a few reasons why this was not possible in this study. First, the number of studies that did prosthesis user testing with participants with upper limb loss is very small. In addition to this, it was not possible to do a user test with participants (either anatomically intact or with upper limb loss) due to the COVID-19 restrictions. The third reason is that the TU Delft owns a limited amount of (commercially available) prostheses, which are all body-powered. This caused a restriction in possible studies to reproduce, as there are no myoelectric prostheses available. Finally, the amount of standardized objective tests for terminal devices is limited as well [51]. Other limitations in validating the simulation are that statistical power could not be achieved with one participant. A report of user satisfaction and comfort would certainly be biased, because the author was the only user. The experiments that were conducted to test the time it takes to adjust the simulator to the user are also biased. Especially the first trial is expected to be slower for other users than the author. The design was made by the user, which means that the user knows exactly how the design works.

The cable attachment for a body-powered prosthesis is one of the critical aspects of this design. In order to use the Bowden cable properly, it is important that the forces that the cable needs to exert on the system are in the same direction as the orientation of the fixated point of the housing. At the distal end (the wrist) it is possible to attach the cable at two different positions. These two positions make the cable suitable for attaching a terminal device with the cable connection at the dorsal or the palmar side of the intact hand. Depending on the intended arm to use the prosthesis simulator on, the radial (for left-side simulator use) or ulnar side (for right-side simulator use) is also in a good position. The other side is not guaranteed to have the cable in the most optimal direction. The two orientations provided in this design were sufficient for the Otto Bock hand and the TRS Grip 2S hook. It is possible that for different prosthetic devices another orientation is needed, which the current attachment points cannot provide. The proximal end (the elbow) of the cable attachment proved to be beyond the scope of this thesis. After a few unsuccessful attempts to 3D-print a custom-sized nut to screw the cable in, the decision was made to clamp the cable. This clamping had to be in a neutral position, so that the prosthesis simulator could still be used on the other arm, or with the prosthesis placed in a different orientation. This causes the end of the housing of the Bowden cable to be oriented in a sub-optimal position. Normally, the orientation of the Bowden cable on the proximal end is not fixated in one position, however, the parts needed to attach it to the socket in such a way are not 3D-printable or highly specialized. The last critical aspect is the strength of the 3D-printed connection between the elbow and the wrist pieces of the simulator. Calculations were made to determine the strength of the piece, and the decision was made to make it 9 mm thick. This makes it strong enough to resist most forces and torques. However, all calculations are based on material properties from the Ultimaker website, which are based on a infill of 90%. The simulator parts are all printed with an infill of only 40%. This was done to save weight and material. The assumption was that the strength of the parts is determined by the dimensions of the parts. Torsion and deflection of the part feature thickness to the third power (h^3). There is an optimum between thickness of the part and the infill percentage. This requires an elaborate material study, because no information on material properties relating to infill percentage was found in literature. No deformations of the part were observed while doing the user testing. But the user testing was not done with male participants, who generally can achieve higher cable forces than female participants [52].

A strength of the design is the possibility to adjust it to the wishes of the user. The optimal position of the prosthesis with respect to the intact hand was not determined, but the design lets the user wear the prosthetic hand in two different orientations (palmar or dorsal). It is also possible to wear the simulator on the left or the right arm. In all orientations the displacement of the center of mass of the arm is kept to a minimum. This will make sure that the kinematics of the arm are changed less than in other designs. Another strength is that it is possible to produce the Delft Pro Simulator without a prosthetists workshop. There are no specialized tools or materials necessary. This simulator only needs a 3D-printer and some very common parts to work. Many research groups, schools and others already use a 3D-printer to print a terminal device. This makes the 3D-printer an accessible way of creating a simulator. The non-3D-printable materials are bolts, nuts and Velcro. These are very common and available all over the world. The design is very accessible. The material cost was only €16.15, which was way below the requirement of the maximum material cost of €100,-.

8.4. Recommendations for future research

A combination of easy accessible manufacturing techniques was mentioned in this thesis. The choice was made to 3D-print the entire design, which constrains the design possibilities. In a future design other accessible production methods, like laser cutting, can be explored to improve the design. This can form a solution for possible critical parts of the design, like the length adjuster or the way to fixate the housing of the Bowden cable for a body-powered prosthesis. Using another material can be used to find a more compliant solution for attaching the cable at the proximal end of the socket. The decision was made to create two different attachment points for the Bowden cable on the connection between terminal device and the socket. This makes the cable orientation optimal for three positions (palmar, dorsal and ulnar/radial). A next version of the simulator could add another attachment point that makes the cable orientation optimal for all possible cable orientations. The last recommendation for future research is to perform user testing trials. More data is needed to evaluate the design. The simulator, or the next version of the simulator, should also be compared with prosthesis users for verification of the simulation.

9 Conclusion

The first goal of this thesis was to design a functioning prosthesis simulator that can be used with common terminal devices. This thesis presents a functioning prosthesis simulator. All terminal devices with a M12x1.5 or 1/2”-20 connection bolt are compatible. The design has proven to be functional with a body-powered device. Using it with a myoelectric device was not tested, but this should be possible with minor adaptations. The second goal was to make the design available and easy to produce for different research groups. This goal was achieved by designing only 3D-printed parts and furthermore only using common materials that are available worldwide, like bolts, nuts and Velcro. The third and last goal was that the device should solve the problems that have been identified with other prosthesis simulators. These were sub-optimal position of the terminal device and degrees of freedom of the intact arm. Even though the terminal device position was not researched directly, this design provides the possibility to choose between two orientations without adapting the simulator. The terminal device can be placed at the palmar side or the dorsal side of the intact hand. The simulator can be worn on the left or the right arm. The design restricts the degrees of freedom of the intact arm in the desired directions. Pro- and supination of the forearm are restricted by the design, while flexion and extension of the elbow are left free. Future research should focus on validating the simulation by testing it with more users and by comparing the results to prosthesis users.

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A

Function analysis

A.1. Introduction

Figure A.1 shows the process tree of a prosthesis simulator. This gives an overview of everything the prosthesis simulator will encounter during its lifecycle. This chapter will explain the different parts of the process tree and why certain functions are important. It will start with a short overview of the state of the art and then follow the process tree until the disposal of the simulator.

A.2. Process tree

1. Originate

In a recent literature review, the Fillauer TRS bodypowered simulator was reported to be the only commercially available prosthesis simulator [12, 14]. This simulator was used in two of the 31 research studies in the review [11, 15]. Only one study focused on the design of a prosthesis simulator [6]. The sockets of the prosthesis simulators were made to fit multiple participants. Only in two studies the design was custom made for a single user [13, 30]. Velcro or BOA cable closure technology were the most popular options to tighten the simulator around different arm sizes. The sockets could not be adjusted for arm length differences and most sockets cover only the hand, wrist and forearm. The optimal position for a prosthetic device is usually at the same position as where the intact hand would have been. However, for people with intact limbs, their own hand would be in the way of the prosthesis. In literature three positions for the terminal device were found; dorsal, axial or palmar from the intact hand. In **Figure 3.2** the options are visualized. Most studies put the prosthesis in the axial position, only two were dorsal [13, 15] and three were palmar [7, 8, 11]. The type of terminal device influenced the position. For body-powered devices it is important that the position of the cable is in a straight line to the prosthesis, so that the cable cannot get stuck [11]. One simulator was made with traditional molding techniques, using polypropylene [53, 54]. Some simulators were made by altering a wrist brace by adding a frame of a rigid material on it. These were made of carbon fiber, lightweight aluminum or fiberglass [27, 55, 56]. There was one study that used a 3D-printer to build a frame out of ABSplus that was mounted on a wrist brace [6]. The Fillauer TRS simulator was made from EXOS® laminates that could be formed around the arm when heated [12].

Comfort is an important factor for prosthesis acceptance [57]. Literature about prosthesis simulators has no mention of comfort. There have been multiple mentions of an unexpected difference in performance of participants. Less strong participants performed worse than the other participants. They described the prosthesis simulator as heavy. The actual weight of the prosthesis simulator in these studies was not reported, which makes it difficult to reproduce the results. It also makes it impossible to find out whether the prosthesis was indeed too heavy or other factors are at play. For two simulators the weight has been reported. These are respectively 1008.5 grams, including a myoelectric terminal device [27, 34] and 511 grams, without a terminal device [12]. A simulator consisting of two cuffs was also found, these cuffs weigh 187 grams each, so 274 grams in total, without terminal device [8]. An amputation at transradial level causes people with upper limb absence to lose degrees

1. Originate

- 1.1 Study the current situation
  - 1.1.1 Existing products
- 1.2 Develop product
  - 1.2.1 Design product
  - 1.2.2 Build prototype
  - 1.2.3 Test prototype
- 1.3 Prepare for production
  - 1.3.1 Open source
  - 1.3.2 Upload files

2. Distribute

- 2.1 Produce
  - 2.1.1 Download files
  - 2.1.2 3D print parts
  - 2.1.3 Laser cut parts
  - 2.1.4 Postprocess fabricated parts
  - 2.1.5 Assembly of all parts

3a. Use (with body-powered terminal device)

- 3a.1 Attach terminal device
  - 3a.1.1 Attach shoulder harness
- 3a.2 Adapt for user
  - 3a.2.1 Adjust simulator length
  - 3a.2.2 Adjust shoulder harness
  - 3a.2.3 Change cuff size
- 3a.3 Put on
  - 3a.2.1 Fasten cuffs
  - 3a.2.2 Put on shoulder harness
- 3a.4 Manipulate object
  - 3a.4.1 Pick up object
  - 3a.4.2 Hold object
  - 3a.4.3 Release object
- 3a.5 Take off
  - 3a.5.1 Remove shoulder harness
  - 3a.5.2 Release cuffs

3b. Use (with myoelectric terminal device)

- 3b.1 Attach terminal device
  - 3b.1.2 Attach batteries
  - 3b.1.3 Attach electrodes
- 3b.2 Adapt for user
  - 3b.2.1 Adjust simulator length
  - 3b.2.2 Change cuff size
- 3b.3 Put on
  - 3b.2.1 Put electrodes on muscle
  - 3b.2.2 Fasten cuffs
- 3b.4 Manipulate object
  - 3b.4.1 Pick up object
  - 3b.4.2 Hold object
  - 3b.4.3 Release object
- 3b.5 Take off
  - 3b.5.1 Release cuffs
  - 3b.5.2 Remove electrodes

4. Maintenance

- 4.1 Clean
  - 4.1.1 Desinfect parts
- 4.2 Repair
  - 4.2.1 Inspect for defects
  - 4.2.2 Remove broken parts
  - 4.2.3 Replace broken parts

5. Discard

- 5.1 Reuse parts
- 5.2 Recycle materials

Figure A.1: The process tree of a prosthesis simulator. This shows everything a prosthesis simulator encounters during its lifecycle. A process tree can be used to establish the correct design requirements.

of freedom of their arm. At wrist level no movement is possible. Dependent on the exact level of amputation, pro and supination are possible. However, this is usually constrained by the amputation or the prosthesis socket. In two of the current designs the degrees of freedom of the forearm of the user are constrained [13, 30]. The other designs are only constrained at the wrist. This makes it possible for able-bodied prosthesis users to compensate.

2. Distribute

Current simulators need to be made in a specialized workshop. The prosthesis simulator that is designed in this thesis can be produced at any place in the world. There are no special tools needed for the assembly. There is also no specific knowledge needed to create the device. All parts are easy to produce (e.g., by 3D-printing) or very common (e.g., a piece of cloth or a shoelace). The building instructions are easy to share and spread over the world, so that production can happen at the location where the device will be used. The product can be distributed by sharing the CAD files. These can be sent to the 3D-printer and printed at the location where the simulator is going to be used.

3. Use

The simulator can be used for different purposes, like rehabilitation training or research. An important factor for this is that it can be used with every terminal device. There are two types of terminal devices. These are a body-powered and a myoelectric terminal device. A bodypowered terminal device is controlled with body movement. This is usually done with a shoulder harness or a cuff around the upper arm. The mechanism is operated by putting tension on a cable in between the mechanism and the harness or cuff. Properties of three commercially available bodypowered terminal devices are reported in **Table IX**. A myoelectric terminal device is controlled by detecting EMG activity of muscles of the forearm. A motor then operates the device. The properties of two commonly used myoelectric terminal devices are reported in **Table X**. The prosthesis simulator should be able to be used with both a myoelectric and body-powered terminal device. Because these both have a different process the process tree is split in two parts.

3a. Use (with bodypowered terminal device)

First, the terminal device needs to be screwed in the simulator socket. To operate the terminal device, a cable attached to a shoulder harness is needed. The cable needs to be attached to certain parts of the simulator to make sure that it runs smoothly and will not get stuck somewhere. Secondly, the device needs to be fitted to the user. The socket needs to be adjusted for their specific arm size. The length of the simulator needs to be altered, as well as the cuff size. Then the shoulder harness can be adjusted so that there is a slight tension in the cable when the user has the simulator in a relaxed position. The third step is putting on the device. The user should be able to do so themselves. Then the simulator is ready to be used in interaction with the environment. It can be used to pick up objects or hold objects and release them. When finished, the user can take off the device by releasing the cuffs and taking off the shoulder harness. The prosthetic hand can be unscrewed from the socket if use with another terminal device is desired.

Table IX: Body-powered device specifications.

Specification	Ottobock Hand	Hosmer 5Xa Hook	TRS Grip 2S
Opening width	NA	NA	75 mm
Gripping force	9 N	9-12 N	N/a
Weight	395 g	87 g	318 g
Mechanism	Voluntary opening	Voluntary opening	Voluntary closing
Length	155 mm	124 mm	141 mm
Cable position	Dorsal side	Radial/ulnar side	Dorsal side

A.2.1. 3b. Use (with myoelectric terminal device)

First, the terminal device needs to be screwed in the simulator socket. Then, the electrodes and battery pack should be installed on the socket. Secondly, the simulator needs to be adjusted to the users arm length and arm circumference. When this was done, the placement of the electrodes on the arm of the user need to be determined. Thirdly, the user can put on the simulator. The electrodes need to be put on the muscles. Then, the simulator can interact with the environment. Finally, the cuffs can be released and the electrodes can be removed from the arm muscles. The prosthetic hand can be unscrewed from the socket if use with another terminal device is desired.

Table X: Properties of commonly used myoelectric terminal devices [58, 59].

Specification	MyoHand VariPlus Speed	Myo Select RSL Steeper
Opening width	100 mm	100 mm
Proportional speed	15-300 mm/s	125 mm/s
Gripping force	100 N	80 N
Weight	460 g	470 g
Power supply	Energypack 757B21 (7.2 V)	S-Cell 7.2 V DC
Battery weight	51 g	44 g
Battery size	54 mm long	6 x 30 x 47 mm (x2)
Length		140 mm
Electrode position	Center	Center

4. Maintenance

To make sure that the prosthesis simulator stays clean, which is especially important when the simulator is used by multiple users, it needs to be cleaned after every use. This makes sure that it is safe for every participant to wear. After every use and especially when removing the terminal device the simulator needs to be inspected for any defects. The defect parts need to be removed and repaired or replaced before using the device again.

5. Discard

When the device is broken, it might be possible to reuse certain parts for making the replacement. These should be taken out and saved for this purpose. If the device is broken and cannot be fixed anymore, it should be recycled. Plastic and metal parts can be melted and reused. Biodegradable materials can be composted.

B

3D-printer properties

B.1. Details of Fused Deposition Modeling (FDM) 3D-printers

Table XI: Details of three commonly used 3D-printers.

	Prusa i3 MK3S+	**Ultimaker 3 (Extended)**	**Ultimaker S5**
Build volume	200 x 200 x 180 mm	197 x 215 x 200 mm (197 x 215 x 300 mm)	330 x 240 x 300 mm
Filament diameter	1.75 mm	2.85 mm	2.85 mm
Dual extrusion	Possible with adaptation	Yes	Yes
Layer resolution	50 micron	20-200 micron	20-150 micron
XYZ resolution	-	12.5, 12.5, 2.5 micron	6.9, 6.9, 2.5 micron
Compatible materials	ABS, PLA, HIPS, PVA, Nylon, Wood	PLA, PVA, ABS, CPE, Nylon	PLA, Tough PLA, ABS, Nylon, CPE, CPE+, PC, TPU 95A, PP, PVA, Breakaway
Price (incl. taxes)	€999	€3629 (€4469)	€6648.95

B.2. Printer properties used for printing the prototypes

Table XII: Printer properties for printing the prototypes.

	Ultimaker S5	**Ultimaker 3**
Build volume	330 x 240 x 300 mm	197 x 215 x 200 mm
Layer thickness	0.2 mm	0.2 mm
Wall thickness	1 mm	1 mm
Infill percentage	40%	40%
Material	PLA	PLA
Support material	PVA	PVA

C

Harris profiles and specification of grading criteria

C.1. Structural concept

Table XIII: Grading criteria and Harris profiles for constraining the degrees of freedom of the intact arm.

(a) Specification of the grading criteria.

	- -	-	+	++
Pro- and supination	Full range of motion	120 degrees	90 degrees	Fully restricted
Flexion and extension	Fully restricted	90 degrees	120 degrees	Full range of motion
3D-printable	Nothing can be printed	< 50% can be printed	> 50% can be printed	Everything can be printed
Adaptable	Not adjustable	Adjusts in one dimension	Adjusts in two dimensions	Will fit everyone

(b) The Harris profiles.

	Hinge constrain	Plate constrain	Cast constrain
Pro- and supination	++	++	++
Flexion and extension	+	++	-
3D-printable	-	++	++
Adaptable	+	+	- -

C.2. Formal concept

Table XIV: Grading criteria for making the design adaptable for multiple users. These grading criteria can be used for designing the cuff, the length adjuster and the locking mechanisms.

	- -	-	+	++
Use with one hand	Assistance is needed	Two loose parts (but doable)	One loose part	No loose parts
3D-printable	Nothing can be printed	<50% can be printed	>50% can be printed	Everything can be printed
Amount of parts	4 or more parts	3 parts	2 parts	1 part
Easy accesible	Specialized part	3D-print special materials	Common part	Easy to print (with PLA)
Precise	Not adjustable	> 5 mm	5 mm > step > 1 mm	< 1 mm
Strength	One time use	Fragile	Durable	Extremely hard to break

Table XV: Harris profiles for cuff design, these are the solutions for making the design adaptable for different arm circumferences.

	Retractable cuff	Solid cuff	Compliant cuff	Hinged cuff	Elastic band	GorillaPod	Snapwrap
Use with one hand	++	++	++	++	++	+	++
3D-printable	++	++	++	-	- -	++	- -
Amount of parts	+	++	++	+	++	- -	+
Easy accessible	++	++	-	+	+	++	- -
Precise	++	- -	++	++	++	+	++
Strength	+	++	+	+	+	- -	-

Table XVI: Harris profiles for length adjusters, to determine the best way to adjust for arm length.

	Slider	Rails	Telescopic pole
Use with one hand	++	++	++
3D-printable	++	++	++
Amount of parts	+	+	-
Easy accessible	++	++	++
Precise	++	++	++
Strength	++	+	+

	Compliant harmonica	Prong in a hole	Rigid harmonica
Use with one hand	++	+	++
3D-printable	++	++	-
Amount of parts	++	+	- -
Easy accessible	-	-	+
Precise	++	+	++
Strength	-	-	+

Table XVII: Harris profiles for locking mechanisms that can both be used for the arm cuffs and the length adjuster.

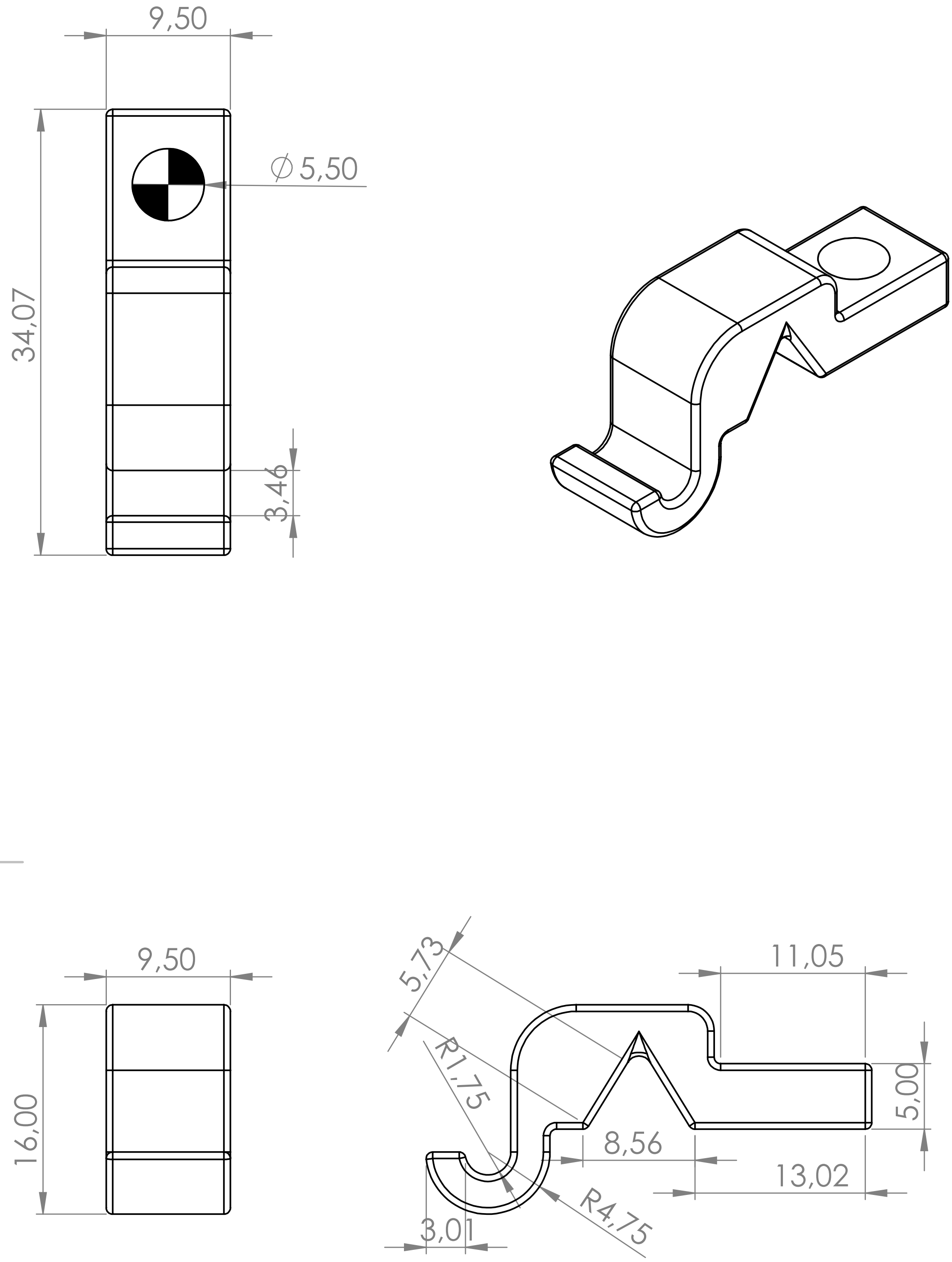
	Holes and pin	Velcro	Triglide slide
Use with one hand	+	++	++
3D-printable	+	- -	+
Amount of parts	+	++	+
Easy accessible	+	+	+
Precise	+	++	++
Strength	++	++	+

	Nut & bolt (printed)	Nut & bolt (metal)	Ratchet
Use with one hand	+	+	++
3D-printable	++	- -	+
Amount of parts	+	+	-
Easy accessible	++	+	+
Precise	-	+	+
Strength	-	++	- -

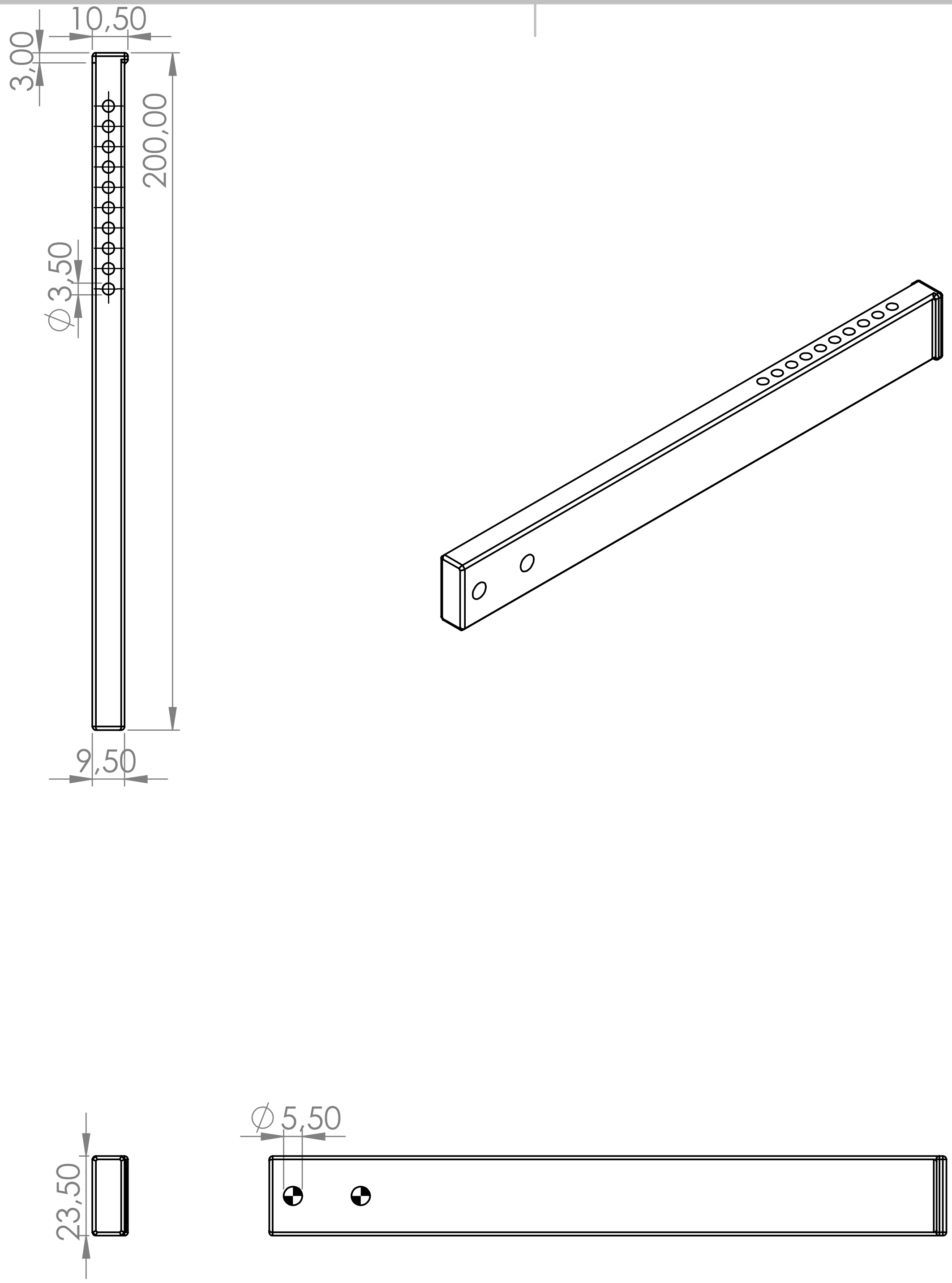
	BOA system	Laces	Hooks
Use with one hand	++	- -	-
3D-printable	- -	- -	+
Amount of parts	-	+	+
Easy accessible	- -	+	+
Precise	++	+	-
Strength	++	++	-

D

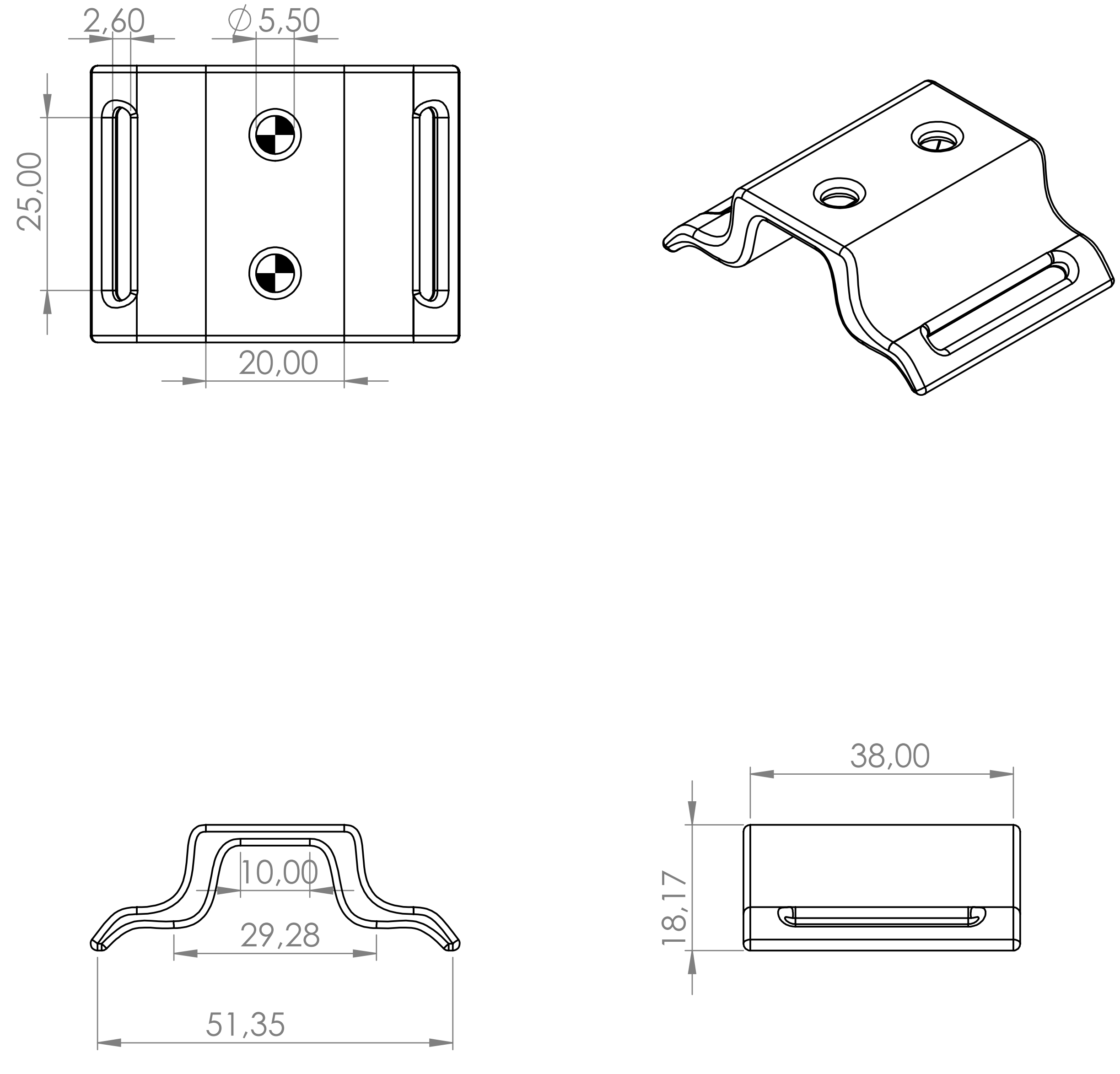
Technical drawings



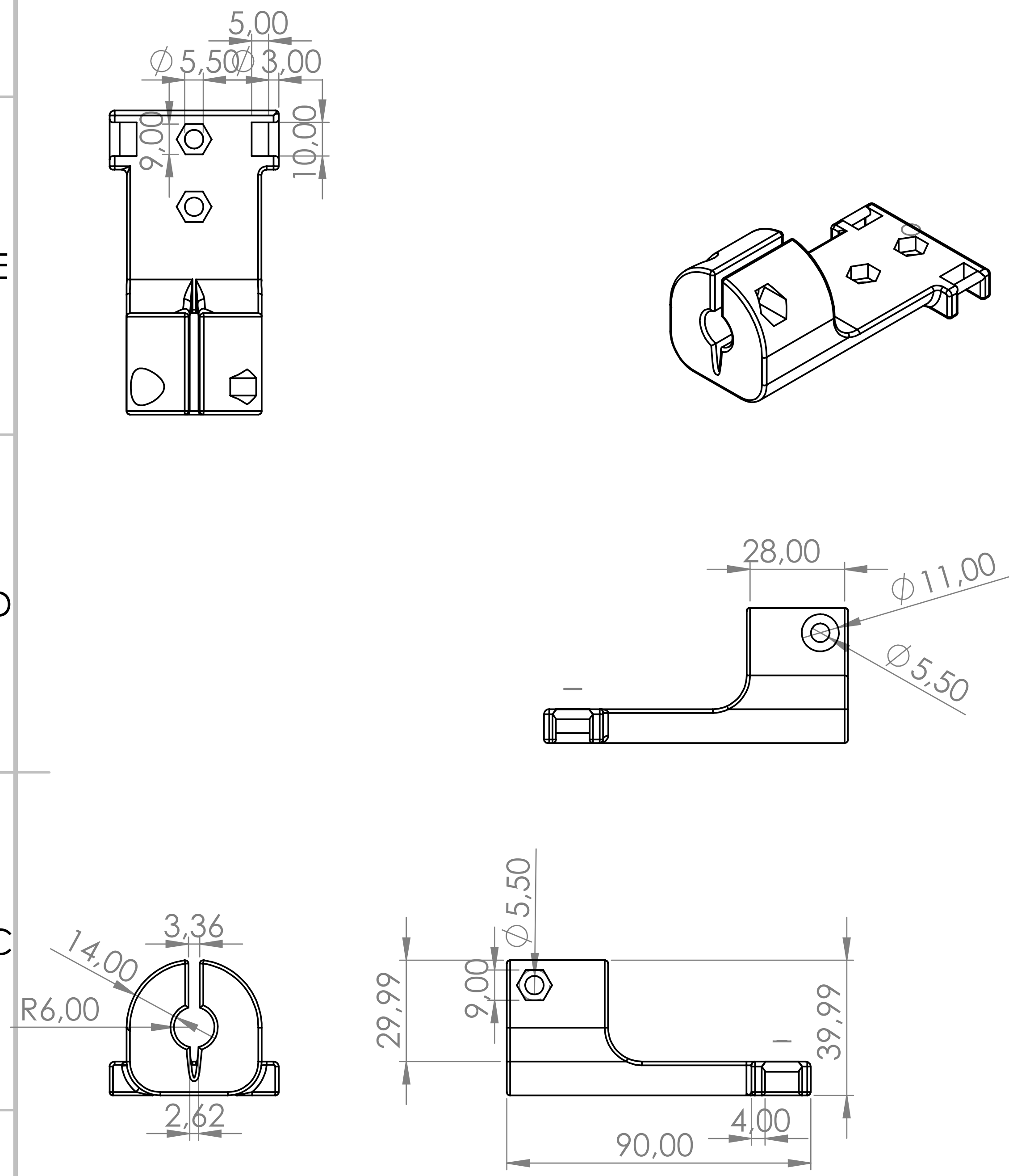
DRAWN BY: Maaike Sinke 6-4-2021	TITLE: Cable clamp	
MATERIAL: PLA	DWG NO. 1	A4
UNIT OF MEASURE: MM	WEIGHT: 3 grams	SCALE:2:1	SHEET 1 OF 1



DRAWN BY: Maaike Sinke 6-4-2021	TITLE: Length adjuster
MATERIAL: PLA	DWG NO. 2 — A4
UNIT OF MEASURE: MM	WEIGHT: 39 grams
SCALE:1:2	SHEET 1 OF 1



DRAWN BY: Maaike Sinke 6-4-2021	TITLE: Wrist cuff	
MATERIAL: PLA	DWG NO. 3	A4
UNIT OF MEASURE: MM	WEIGHT: 11 grams	SCALE:1:1 SHEET 1 OF 1



DRAWN BY:

Maaike Sinke
6-4-2021

TITLE:

Terminal device connection

MATERIAL:

PLA

DWG NO.

4

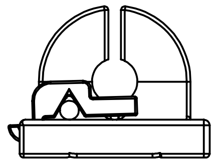
A4

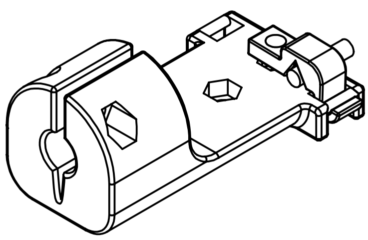
UNIT OF MEASURE: MM

WEIGHT: 51 grams

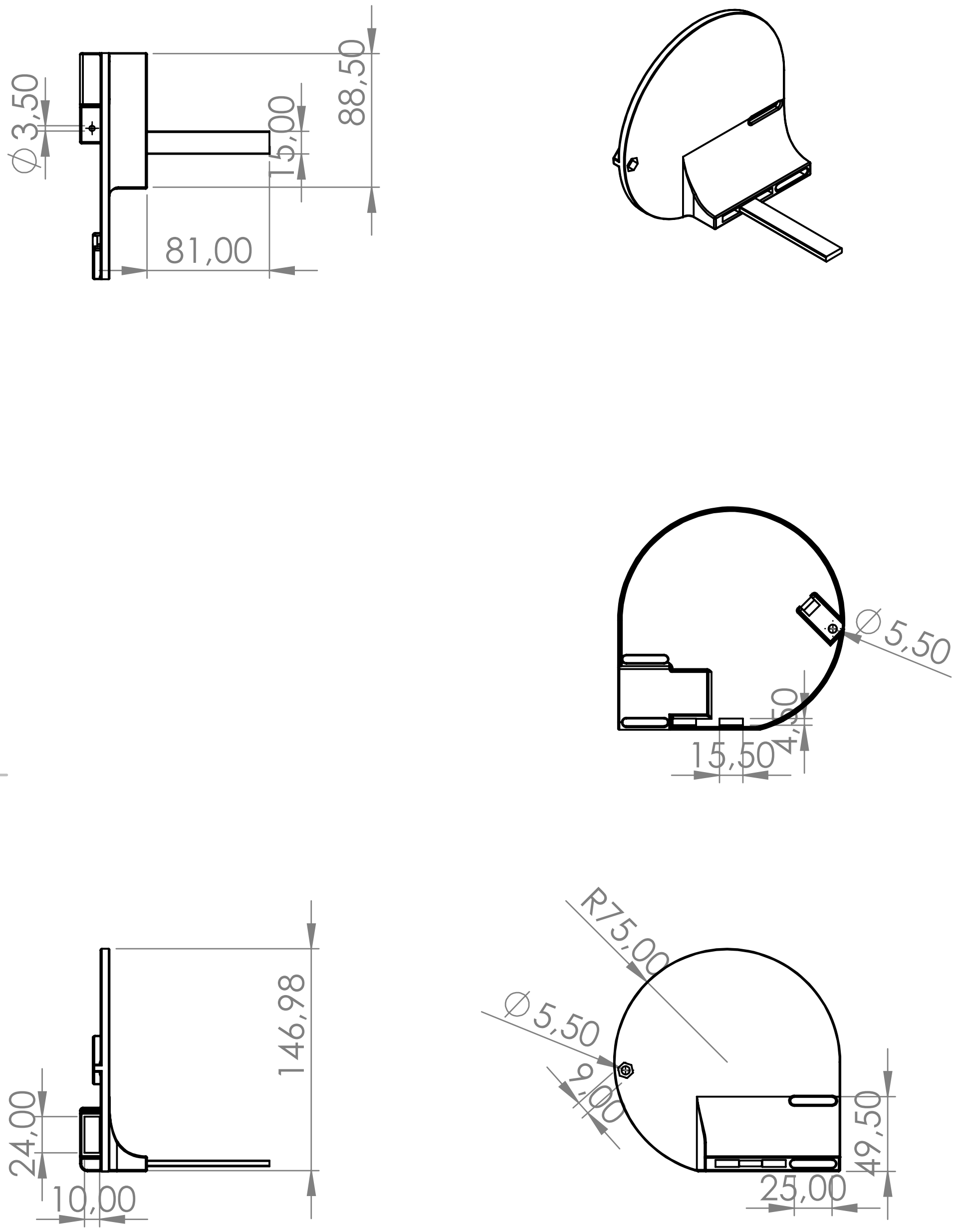
SCALE:1:2

SHEET 1 OF 2

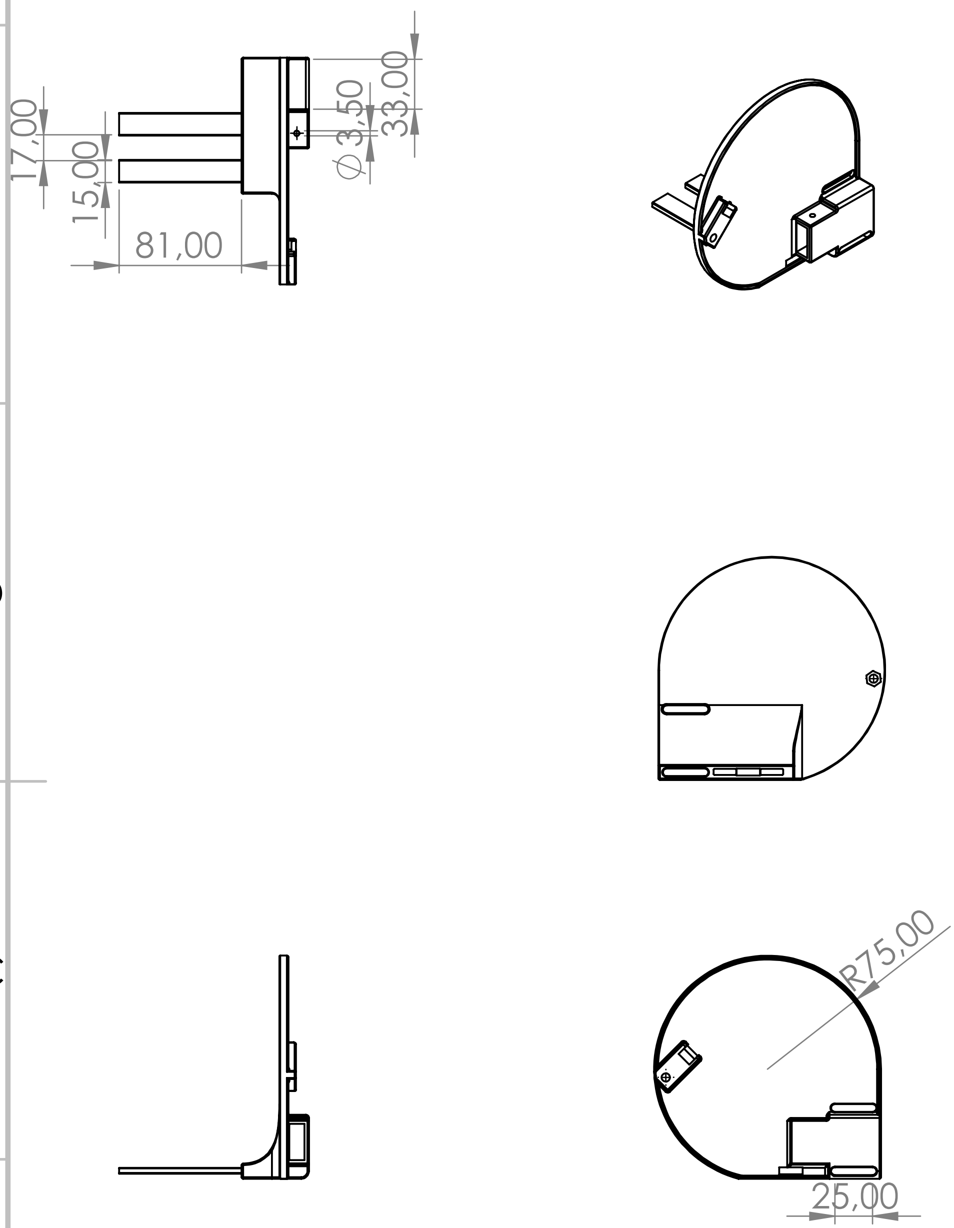




DRAWN BY: Maaike Sinke 6-4-2021	TITLE: Terminal device connection		
MATERIAL: PLA	DWG NO. 4		A4
UNIT OF MEASURE: MM	WEIGHT: 51 grams	SCALE:1:2	SHEET 2 OF 2



DRAWN BY: Maaike Sinke 6-4-2021	TITLE: Elbow cuff (left)
MATERIAL: PLA	DWG NO. 5 / A4
UNIT OF MEASURE: MM / WEIGHT: 101 grams	SCALE:1:5 / SHEET 1 OF 1



DRAWN BY: Maaike Sinke 6-4-2021	TITLE: Elbow cuff (right)
MATERIAL: PLA	DWG NO. 6 — A4
UNIT OF MEASURE: MM	WEIGHT: 114 grams
SCALE:1:5	SHEET 1 OF 1

E
Matlab

E.1. Deformation of walls

```
%% 1.1 Wall deformation theory %%
Marm = 4.6; % moment in Nm
L = 0.075; % wall length in m
w = 0.050; % wall width in m
h = 0.003; % wall thickness in m
E = 2.3465*10^9; % Pa (PLA 90% infill) Ultimaker website

W = (2*Marm)/L; % distributed load
I = (w*h^3)/12; % moment of inertia of wall
vmax = abs((W*L^4)/(8*E*I)) % absolute maximum deflection with distributed load

P = abs((vmax*(3*E*I))/L^3); % estimated point load

%% 1.2 Wall deformation experiment %%
v = 0.003; % prescribed displacement in mm
F = 3.45; % measured force in N

P_calculated = abs((v*(3*E*I))/L^3); % calculated force (N) corresponding to prescribed ...
    displacement

E_calculated = abs((F*L^3)/(3*v*I)); % Pa Calculated with prescribed displacement and ...
    measured force (40% infill)

%% 1.3 Wall deformation prototype 2 %%
Marm = 4.6; % moment in Nm
L = 0.075; % wall length in m
w = 0.075; % wall width in m
h = 0.006; % wall thickness in m
E = 2.3465*10^9; % Pa (PLA 90% infill) Ultimaker website

W = (2*Marm)/L; % distributed load
I = (w*h^3)/12; % moment of inertia of wall
vmax = abs((W*L^4)/(8*E*I)) % absolute maximum deflection with distributed load
```

E.2. Torsion and deflection of length adjuster

```
%% 2.1 Torsion of length adjuster %%
G = 2.4*10^9; % Pa shear modulus
L = 0.2; % part length in m
a = 0.0235; % part width in m
b = 0.0095; % part height in m

ratio = a/b; % ratio width/height
```

```
c1 = 0.258;
c2 = 0.249;

T = 0:0.01:100;
theta = (L*T)/(c2*a*b^3*G);

T1=4.6;
theta1 = (L*T1)/(c2*a*b^3*G);

figure;
plot(T,theta)
xlabel('Torsion (Nm)')
ylabel('Angle of deflection (rad)')
title('Torsion vs deflection (theoretically)')
hold on
plot(T1,theta1,'o')

%% 2.2 Deflection of length adjuster %%
L = 0.2; % m
h = 0.0095; % 9.5 mm
b = 0.0235; % 23.5 mm
W = 0:0.01:100; %force on beam N
I = 1/12*b*h^3;
E = 2.3465*10^9; % Pa (PLA 90% infill) Ultimaker website
vmax = abs((W*L^4)/(8*E*I));

figure;
plot(W,vmax)
xlabel('Force (N)')
ylabel('Deflection (m)')
title('Force vs deflection of the length adjuster (theoretically)')
```

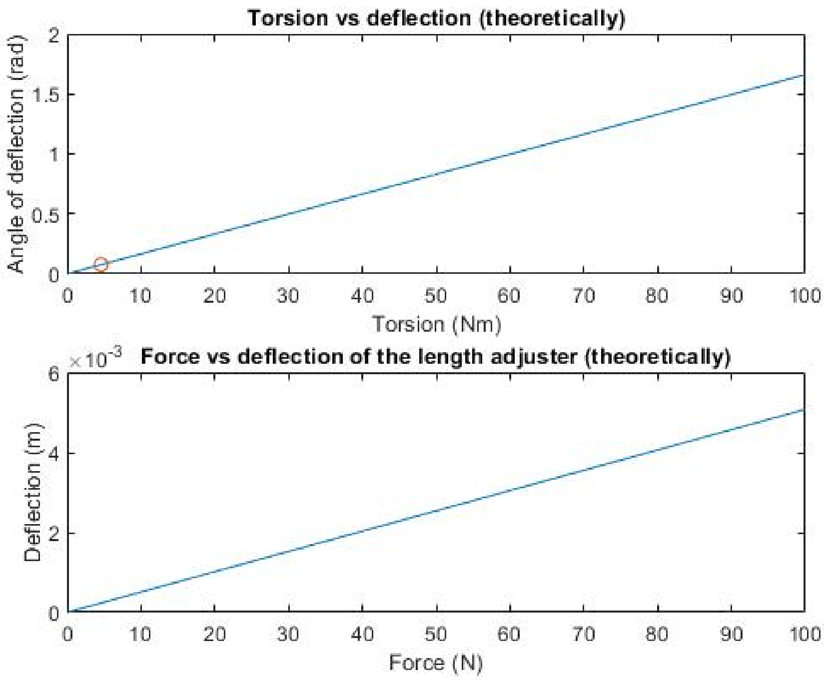


Figure E.1: The calculated torsion-deflection and force-deflection graphs of the length adjuster. The red dot in the upper figure indicates the maximum value of the torsion caused by a user on the simulator.

E.3. Calculation center of mass

```
%% Center of mass displacement axial direction %%
% female
FA_l = 264.3; % mm
FA_m = 1.38; % percentage of 61.9 kg
FA_com = 45.59; % percentage of FA_l
Com_pos = FA_com/100*FA_l; % longitudinal com position
FA_Mass = FA_m/100*61.9;

H_l = 78.0;
H_m = 0.56;
H_com = 74.74;
HCom_pos = H_com/100*H_l; % longitudinal com position
H_Mass = H_m/100*61.9;

% Calculate COM of entire arm
HCom_pos1 = FA_l+HCom_pos;
COMbined = (HCom_pos1*H_Mass + Com_pos*FA_Mass)/(H_Mass+FA_Mass);

% Prosthesis simulator no TD
Sim_mass = 0.356; %kg
Sim_com = 65; %mm

COMbined_sim = (HCom_pos1*H_Mass + Com_pos*FA_Mass + Sim_com*Sim_mass ...
        )/(H_Mass+FA_Mass+Sim_mass);
Test = COMbined_sim-COMbined

% Prosthesis simulator w/ TD
SimTD_mass = 0.824; %kg
SimTD_com = 165; %mm

COMbined_TDsim = (HCom_pos1*H_Mass + Com_pos*FA_Mass + SimTD_com*SimTD_mass ...
        )/(H_Mass+FA_Mass+SimTD_mass);
TestTD = COMbined_TDsim-COMbined

% male
mFA_l = 268.9; % mm
mFA_m = 1.62; % percentage of 73.0 kg
mFA_com = 45.74; % percentage of FA_l
mCom_pos = mFA_com/100*mFA_l; % longitudinal com position
mFA_Mass = mFA_m/100*73;

mH_l = 86.2;
mH_m = 0.61;
mH_com = 79;
mHCom_pos = mH_com/100*mH_l; % longitudinal com position
mH_Mass = mH_m/100*73;

% Calculate COM of entire arm
mHCom_pos1 = mFA_l+mHCom_pos;
mCOMbined = (mHCom_pos1*mH_Mass + mCom_pos*mFA_Mass)/(mH_Mass+mFA_Mass);

% Prosthesis simulator no TD
Sim_mass = 0.356; %kg
Sim_com = 65; %mm
mCOMbined_sim = (mHCom_pos1*mH_Mass + mCom_pos*mFA_Mass +  Sim_com*Sim_mass ...
        )/(mH_Mass+mFA_Mass+Sim_mass);
mTest = mCOMbined_sim-mCOMbined

% Prosthesis simulator w/ TD
SimTD_mass = 0.824; %kg
SimTD_com = 165; %mm
mCOMbined_TDsim = (mHCom_pos1*mH_Mass + mCom_pos*mFA_Mass + SimTD_com*SimTD_mass ...
        )/(mH_Mass+mFA_Mass+SimTD_mass);
mTestTD = mCOMbined_TDsim-mCOMbined

%% Center of mass displacement lateral direction %%
% Assumption that COM is in the center of the arm
```

```
L_sim = 70; % mm in the maximum size
M_sim = 0.132; % kg only the parts that are not symmetric
Com_arm = 0;
M_arm = (H_Mass+FA_Mass);
mM_arm = (mH_Mass+mFA_Mass);

displaced = (L_sim*M_sim+Com_arm*M_arm)/(M_sim+M_arm)
mdisplaced = (L_sim*M_sim+Com_arm*M_arm)/(M_sim+mM_arm)

% with TD
L_sim = 70; % mm in the maximum size
M_sim = 0.6; % kg only the parts that are not symmetric
Com_arm = 0;
M_arm = (H_Mass+FA_Mass);
mM_arm = (mH_Mass+mFA_Mass);

displacedTD = (L_sim*M_sim+Com_arm*M_arm)/(M_sim+M_arm)
mdisplacedTD = (L_sim*M_sim+Com_arm*M_arm)/(M_sim+mM_arm)


female = sqrt((displaced^2)+(Test^2))
male = sqrt((mdisplaced^2)+(mTest^2))
female_withTD = sqrt((displacedTD^2)+(TestTD^2))
male_withTD = sqrt((mdisplacedTD^2)+(mTestTD^2))
```

Table XVIII: The displacement of center of mass caused by the simulator with and without terminal device (TD). Displacement (1) is the difference between the situation without simulator and the situation with simulator without TD. Displacement (2) is the difference between the situation without simulator and the situation with simulator with TD. All values are in mm and the origin is at the elbow.

	Male	**Female**
Axial direction		
Without simulator	181.53	178.83
With simulator (no TD)	160.62	152.80
Displacement (1)	-20.91	-26.03
With simulator (with TD)	175.98	173.20
Displacement (2)	-5.56	-5.63
Lateral direction		
Without simulator	0	0
With simulator (no TD)	5.25	6.93
With simulator (with TD)	18.85	23.32
Both directions combined		
Total displacement (no TD)	21.56	26.94
Total displacement (with TD)	19.65	23.99