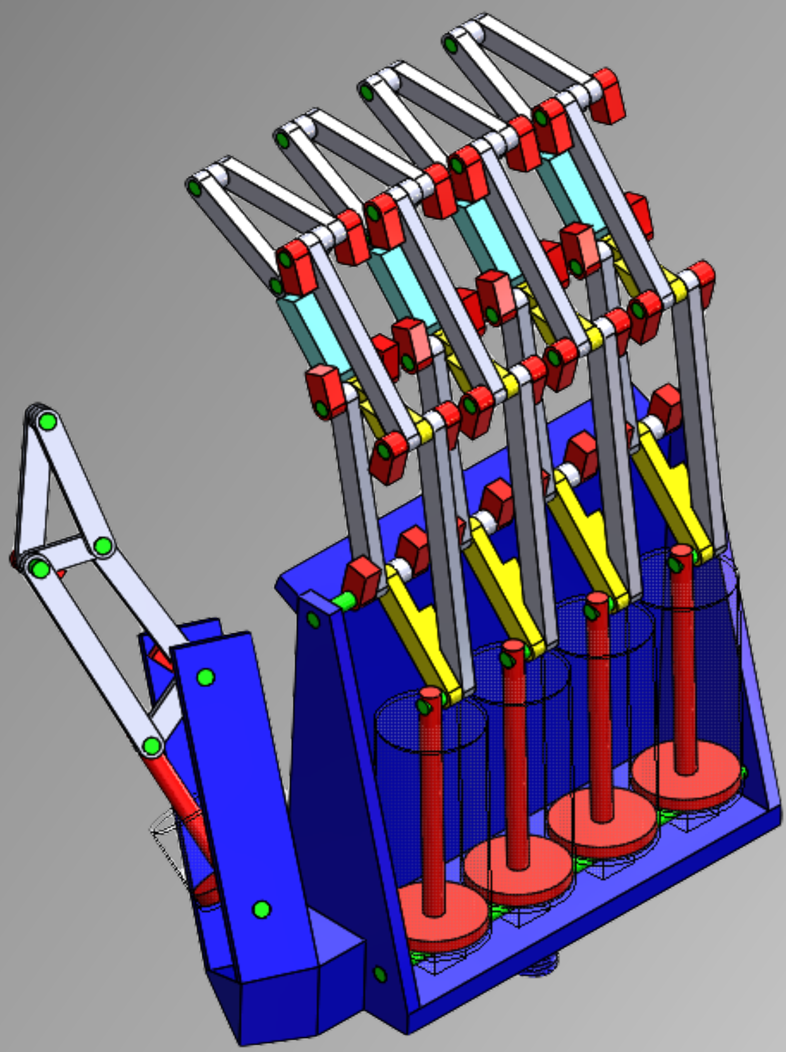
Design of a Pneumatically Powered Hand Prosthesis for Toddlers

Introducing an underactuated 3-DOF linkage-based finger transmission mechanism

J.C. Vervaet

Master of Science Thesis



Delft University of Technology

Department of Biomechanical Engineering

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Introducing an underactuated 3-DOF linkage-based finger transmission mechanism

Master of Science Thesis

For the degree of Master of Science in Biomedical Engineering at Delft University of Technology

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Contents

Preface

After I graduated high school I wanted to study medicine. However, I was not able to pursue this career due to the draw. It forced me to choose something else. Due to my interests in mechanics and technology I decided to try Mechanical Engineering. However, during that year I learned that I still had a large interest in the human anatomy and physiology. I found out that a study exists that combines both natural sciences and medicine. For my bachelor's degree I studied Medical Natural Sciences, only to come back to the Mechanical Engineering department a few years later. During these lasts years of studying Biomedical Engineering, I learned a lot about engineering and also what things interests me. This thesis represents my final work of finishing this master's degree. I would like to thank Dick Plettenburg for his supervision and also Jan van Frankenhuyzen who helped me realize the prototype. In the years to come, I hope to bridge the gap between the medical and technical worlds, thereby providing useful and innovative insights or concepts, that could ultimately help improve medical care.

Introduction

Background

Prevalence, causes and solutions

In 1996 the National Health Interview Survey reported that in the United States the prevalence of absence or loss of an upper extremity was approximately 102,000. Those who lost an arm or had an amputated hand conclude 41,000. The remaining 61,000 were described by those who lost one or more fingers (LaPlante et al., 1996). Ziegler-Graham et al. (2008) estimated that in the United States the number of persons living with the loss of a limb is expected to more than double from 1.6 million in 2005 to 3.6 million in 2050. This estimated trend shows that there is an increasing need for upper extremity prostheses, in order to alleviate the debilitation and distress. Many different designs have been proposed to restore the function and appearance of the missing limb.

Types of prostheses

Prosthesis for upper extremity amputees can be categorized into cosmetic, body-powered and externally powered prostheses. A cosmetic prosthesis is aesthetically pleasing and cheap, though limited in functionality due to its single configuration. A body-powered prosthesis is controlled by a harness, which is commonly attached to the shoulder. Through movements of the shoulder, i.e. flexion-extension and/or abduction-adduction, the prosthetic hand will open and close accordingly. Such a device provides the user with force and displacement feedback. Externally powered prostheses generally consist of a power source, controller and actuator. These prostheses can be divided into three subcategories by actuator type and energy source, namely electric, hydraulic and pneumatic. Depending on the design and control of such a prosthesis, feedback can be provided to the user.

Abandonment

Creating an upper extremity prosthesis that functions well has been a challenge, since the simplest movements require complex controls (Freund, 1983; Schieber, 1995). The human hand consists of 27 bones, of which most are controlled through the activation of many muscles and subsequently tendons. There is an alternating synergy between different muscles, where some are activated and others inhibited at a certain time (Freund, 1983; Schieber, 1995). In addition, the hand contains many different sensors that provide one with feedback. The control and feedback of a human hand make for a complex overall system, which is hard to replicate into an anthropomorphic upper extremity prosthesis. Such a prosthesis has many requirements, which currently can only be partly met. Biddiss et al. (2007) analyzed and compared surveys of upper limb prosthesis acceptance and abandonment. They reported mean rejection rates of 45% and 35% for body-powered and electric prostheses respectively in pediatric populations. Significantly lower rates of rejection for both body-powered 26% and electric 23% devices were observed in adult populations. 22% of cessations attributable to prosthetic problems or discomfort. In general upper limb prostheses are abandoned due to heavy weight, low functionality and limited degrees of freedom (DOF) (Silcox et al., 1993; Atkins et al., 1996; Schulz et al., 2001). Despite a comparable mass between a prosthetic and human hand, prostheses are experienced as too heavy, since the generated torque from the mass of a transradial prosthesis is applied at a relative short part of the forearm (Belter et al., 2013). The low functionality and limited degrees of freedom are explained by the inadequacy of finger control and different grips. Plettenburg (1998) suggests that if any of the the following three requirements: cosmetics, comfort and control, is not fulfilled the prosthesis will be abandoned. Schultz et al. (2007) reported that experts ranked comfort first, followed by function and cosmetics. Each of these three requirements will result in a list of design criteria. However, some criteria will be compromised and others met depending on the type of prosthesis.

Pneumatically powered prosthesis

Pneumatic actuators have a high power density, which allows for a small and lightweight actuator to be included in the design of a hand prosthesis. In addition, there is no power consumption required, assuming that there is no leakage, when a grasp is maintained, i.e. during isometric contraction.

Problem Statement

Upper extremity prostheses have high rejection rates up to 45% (Biddiss et al., 2007). Generally, the abandonment of such prostheses are caused by discomfort, heavy weight, low functionality and limited degrees of freedom(Silcox et al., 1993; Atkins et al., 1996; Schulz et al., 2001; Biddiss et al., 2007). These shortcomings form a problem, which leads to the following problem statement:

Upper extremity prosthesis have high abandonment rates due to discomfort and a lack of functionality.

Objective

The stated problem will be addressed by the main objective of the thesis:

To design and create a working pneumatically powered hand prosthesis for toddlers, which is highly functional and lightweight.

Rejection rates are highest in the pediatric populations (Biddiss et al., 2007). In addition, the design of a prosthesis for a toddler leads to a result, which is more likely to be expanded to an adult version than vice versa. Fulfilling the main objective will be attempted by performing a literature study, which is necessary to gather information regarding the requirements and working principles of pneumatically powered upper extremity prostheses. This information is then used to create a design, which will be translated into a working prototype. Consequently, this prototype is tested to show if the list of requirements are met and if the prototype works correctly.

Outline

This thesis is divided into three chapters. Chapter 1 provides a scientific paper, which concludes the most important findings of this thesis in a short organised manner. Chapter 2 includes the literature study that was performed prior to the start of this thesis. In this study an in depth comparison of different pneumatic actuation methods for upper extremity prosthesis can be found. After the validation of the choice of researching pneumatics, the different types of pneumatic prosthesis are presented and analyzed. Lastly, chapter 3 includes different appendices, where the different stages of the design are presented and elaborated in more detail. Appendix A, elaborates the concept synthesis, where the process and reasoning behind certain choices are explained. Appendix B, presents the manufacturing and assembling of the prototype. Appendix C, shows and discusses the performance of the prototype. Appendix D, explains the future recommendations of the design. Lastly, appendix E, shows the images of the technical drawings.

Chapter 1

Scientific Paper

Design of a pneumatically powered hand prosthesis for toddlers

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Abstract

A large proportion of upper extremity prostheses are abandoned by their users. Commonly, the main reasons for abandonment are related to comfort, control and cosmetics. Prostheses are often experienced as too heavy and have limited functionality. A pneumatic power source is able to provide a relatively high force using a small and lightweight actuator. However, state-of-the-art pneumatically powered upper extremity prosthesis offer either high grasp forces or the ability to adapt to the size and shape of an object. This article presents the design of a pneumatically powered hand prosthesis for toddlers, which focuses on being lightweight and highly functional. An underactuated 3-DOF linkage-based finger transmission mechanism was created and served as a proof-of-principle. The resulting prototype was 3D printed and tested following an elaborate list of criteria. The results demonstrate that the mechanism is capable of transmitting an actuator force of 100.5 N to a fingertip force of 32.3 N using 5 bar of pressure. The proposed mechanism also offers the functionality of adaptive grasping. The conceptual design and prototype show promising capabilities for the development of a lightweight and highly functional prosthesis for toddlers.

Keywords: prosthesis, hand, pneumatic, finger, transmission, linkage, adaptive

1-1 Introduction

Creating an upper extremity prosthesis that functions well has been a challenge, since the simplest movements require complex controls (Freund, 1983; Schieber, 1995). The human hand consists of 27 bones, of which most are controlled through the activation of many muscles and subsequently tendons. There is an alternating synergy between different muscles, where some are activated and others inhibited at a certain time (Freund, 1983; Schieber, 1995). In addition, the hand contains many different sensors that provide one with feedback. The control and feedback of a human hand make for a complex overall system, which is hard to replicate into an anthropomorphic upper extremity prosthesis. Biddiss et al. (2007) showed that upper extremity prostheses have high rejection rates up to 45%. Generally, the abandonment of such prostheses are caused by discomfort, heavy weight, low functionality and limited degrees of freedom (Silcox et al., 1993; Atkins et al., 1996; Schulz et al., 2001; Biddiss et al., 2007). Despite a comparable mass between a prosthetic and human hand, prostheses are experienced as too heavy, since the generated torque from the mass of a transradial prosthesis is applied at a relative short part of the forearm, which ultimately leads to discomfort (Belter et al., 2013). The low functionality and limited degrees of freedom are explained by the inadequacy of finger control and different grips. Unfortunately, none of the articles provided quantitative design criteria. Plettenburg (1998) suggests that if any of the the following three requirements: cosmetics, comfort and control, is not fulfilled the prosthesis will be abandoned.

In order to reduce abandonment rates of upper extremity prostheses, solutions are required that counteract the main reasons thereof. Consequently, comfort will be addressed in terms of minimizing the weight of the prosthesis and control will be addressed by mimicking anthropomorphic functionality.

The weight of a prosthesis largely depends on the weight of the actuator. Peerdeman et al. (2012) showed that a pneumatic cylinder actuator offers equal forces and higher closing speeds compared to an average DC motor used in state-of-the-art hand prosthesis, whilst having a mass of over 10 times less. This high energy-to-mass ratio allows for a small and lightweight actuator to be included in the design of a hand prosthesis. In addition, there is no power consumption required, assuming that there is no leakage, when a grasp is maintained, i.e. during isometric contraction.

The functionality of a prosthesis is mostly determined by its ability to grasp objects of different shapes, sizes and weights. Obtaining this ability can be achieved when the prosthetic hand adapts to the shape and size of the object, whilst also providing a sufficiently high force to uphold the weight of the object. Vervaet (2019) showed that pneumatically powered upper extremity prostheses often fulfill one of two criteria having either an adaptive grip or a relatively high grip force (Chapter 2, Literature Study). In this study two types of pneumatically powered prosthesis were distinguished namely, pneumatic cylinders and fluidic actuators. Pneumatic cylinders use compressed gas to move the piston, thereby allowing displacement of the valve stem. Fluidic actuators are actuators that manipulate a material by expanding it using pressurized gas and thereby exerting force and/or movement in a desired direction. The former showed prostheses with relatively large grip forces and the latter showed prostheses that were capable of adaptive grip. Despite the adaptive grip capabilities of the transmission mechanisms with fluidic actuators, the average force output was 10 times lower than the average values found with pneumatic cylinder actuators. Reducing the high rejection rates of prostheses requires a combination of the instantaneous force delivered by pneumatic actuators with a transmission mechanism that allows for adaptive grip.

In order to achieve this, a transmission mechanism is required that is capable of adaptive grip, whilst also transmitting and sustaining high forces. In general, two types of transmission mechanism are distinguished, being linkage-based and tendon-pulley mechanisms (Carrozza et al., 2006). Tendon-pulley mechanisms can lead to friction and elasticity (Ceccarelli, 2004). In addition, Carbone et al. (2015) reported that the tendon-pulley design led to very high friction losses and very low mechanical efficiency. They also reported that the use of tendons limits the maximum amount of input force, which leads to lower achievable grasping forces. Contrarily, Smit et al. (2013) showed that a linkage mechanism is most suitable for prosthetic hands, due to its higher energy efficiency compared to tendon-pulley mechanisms. Carbone et al. (2015) showed that linkage-based mechanisms result in very limited friction losses and good mechanical efficiency. They reported that linkage-based mechanisms allow for higher input torques and higher achievable grasping forces.

This study aims to design and create a working pneumatically powered hand prosthesis for toddlers, which is highly functional and lightweight. The main focus of this thesis concerns the creation of an underactuated linkage mechanism, which is effective in transmitting high forces, whilst allowing adaptive grip. A design for toddlers is preferred, due to the relatively high rejection rates in the pediatric populations (Biddiss et al., 2007). In addition, the design of a prosthesis for a toddler leads to a result, which is more likely to be expanded to an adult version than vice versa. This paper presents the design criteria and conceptual design of a pneumatically powered hand prosthesis. Accordingly, due to time limitations of the thesis project, only the finger transmission mechanism is created and tested.

1-2 Methods and Materials

1-2-1 Design criteria

A number of design criteria were created to address the requirements obtained from the main reasons of abandonment and the literature study. Figure 1-1 shows data from Molenbroek (1993), which equal the criterion for the hand size of the prosthesis. In addition, the

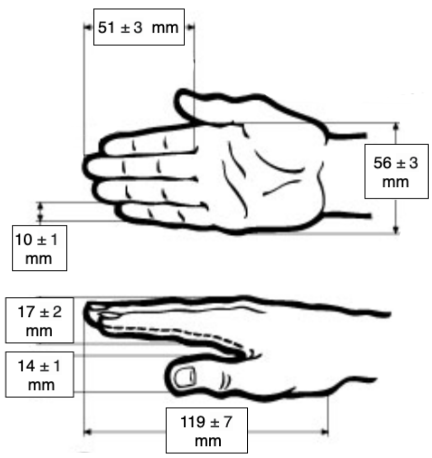

Figure 1-1: Hand dimensions of an average 4 year old child. (Molenbroek, 1993)

comfort of the prosthesis is mainly addressed by minimising the weight thereof. A maximum weight of 106 grams was set as the criterion. Despite potential proportionality differences between adults and children, linear scaling was applied. The weight criterion was determined by taking the average weight of a 4 year old child, being 17.7 kg Scholtens et al. (2007), and subsequently applying the determined hand mass percentage of 0.6 % from Tözeren (1999).

Furthermore, a force criterion of 46 N is suggested, which is derived from a grasp force study from Ploegmakers et al. (2013). However, the amount of force that the prosthetic hand should deliver is difficult to determine, due to a number of factors. Prosthetic hands are often made of non-yielding material, which results in a relatively small contact area between the hand and the object. Kargov et al. (2004) showed that the number of contact points and amount of contact area decreased in the following order: human hand, adaptive prosthetic hand, non-adaptive prosthetic hand. It can be assumed that the human hand is most capable of firmly and comfortably grasping objects of different shapes, sizes and masses. In addition, the grasp force exerted by the prosthesis is also a function of the transmission mechanism and the friction between the contact points of the prosthesis and object. The forces that a prosthesis should deliver are therefore not representative of the performance of a grasp. Nonetheless, an indication of the force that the prosthesis should deliver is suggested to develop a fitting design.

Besides that, the prosthetic hand must have adaptive grip. The performance of a prosthesis highly depends on the number of contact points between the prosthesis and the object. An adaptive grip can increase the number of contact points, thereby increasing the hand's ability to perform varying grasping tasks. This mechanism should allow for multiple Degrees of Freedom (DOF) to be operated using a single actuators. Furthermore, the adaptive grip criterion can be extended and explained by means of two stages. The first stage is comprised of the pre-shaping phase, which is explained as the movement of the fingers as a whole, where the trajectory is set. The second stage is comprised of the adaptive phase, which is only active when an object makes contact with a phalanx or when a motion boundary is reached.

Lastly, the movement capabilities of the prosthetic hand should equal those of a human hand. Therefore, the prosthesis should have 15 degrees of freedom and must fulfill the following range of motion in the joints Kapandji (1971), Stillfried et al. (2014):

- MCP $< 90^\circ$
- PIP $> 90^\circ$
- DIP $< 90^\circ$

1-2-2 Conceptual design

A full CAD representation of the prosthetic hand design is shown in figure 1-2. Essentially, the hand consists of a frame, finger transmission mechanism and pneumatic cylinder actuator. The finger transmission mechanism is shown in more detail in fig 1-3.

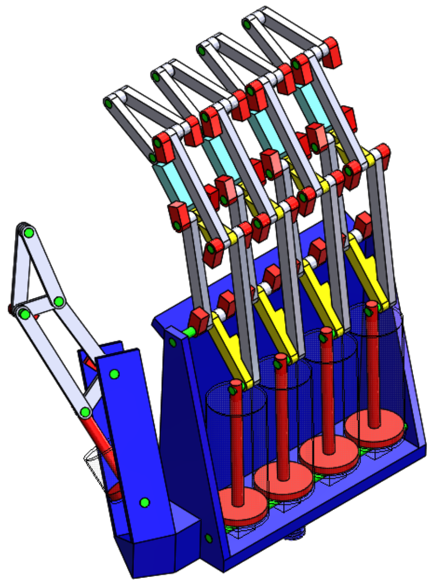

Figure 1-2: The CAD of the prosthetic hand. This figure shows the frame, pneumatic cylinder actuators and the finger transmission mechanism.

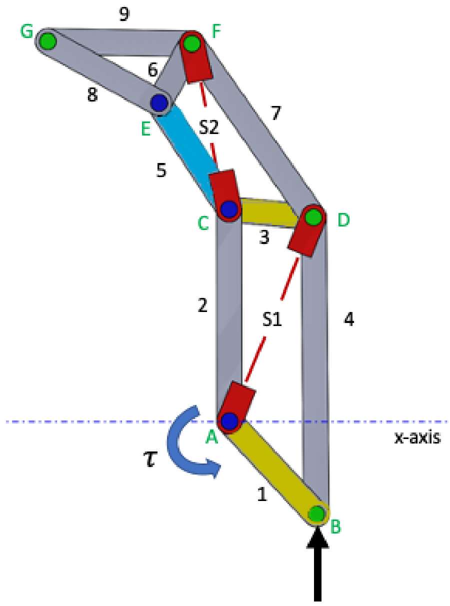

Figure 1-3: The CAD model of the finger transmission mechanism for a toddler sized prosthesis. This schematic indicates the different links (with numbers), springs (S1 and S2) and pins (with letters). The black arrow indicates the force provided by the actuator, which subsequently causes a torque in A.

The mechanism is underactuated, providing 3 DOF with a single actuator. The three joints that each provide a single DOF are the joints A, C and E (in blue). Respectively, these joints are the equivalents of the MCP, PIP and DIP of a human hand. Joint A, which intersects the x-axis, is fixed in the frame. In addition, the mechanism consists of nine links and two tension springs. Link 1,2,3 and 4 together with spring 1 (S1) make up the proximal phalanx. Link 3,5,6 and 7 together with spring 2 (S2) make up the middle phalanx. Lastly, 6, 8 and 9 make up the distal phalanx. The pneumatic cylinder actuator is connected at B and drives link 1 in a counterclockwise rotation around A when actuated.

To support the upcoming explanation of the working principle of the finger transmission mechanism, the grasping motion is demonstrated in figure 1-4. The trajectory of the operation of the finger mechanism can be split up into two motion stages, namely the pre-shaping phase and adaptive phase. The first phase, pre-shaping, is comprised of the whole finger moving as a rigid body around joint A. The actuator force in B causes a torque around joint A, whilst both springs prevent the phalanges to change shape, i.e. keeping the same configuration. The second phase, adaptive phase, is initiated when link 2 reaches its final position or when one of the following links is obstructed by an object during the first phase: 2,5 or 8. Depending on the location of contact a following motion is initiated. When link 2 is obstructed, the actuation force in B will no longer cause rotation of the finger in

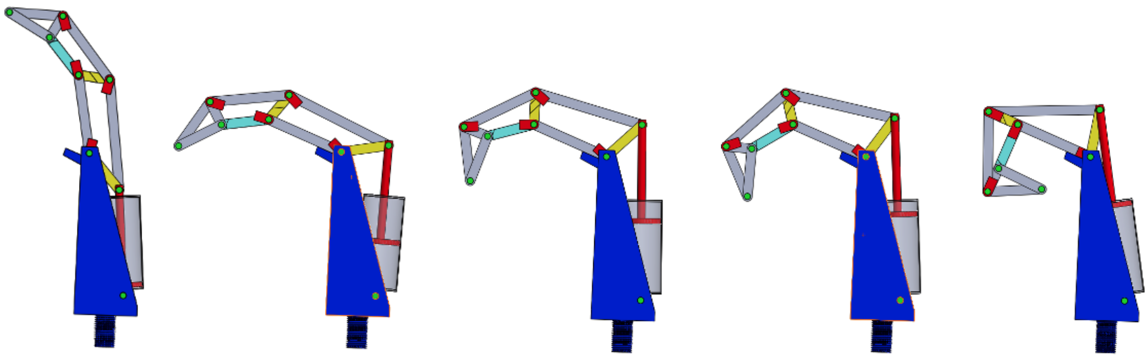

Figure 1-4: The motion of the finger transmission mechanism at different actuator stroke lengths without the presence of an object. The mechanism starts in a neutral position. Then, upon activation of the actuator the mechanism moves as a whole until link 2 reaches its end position, i.e. pre-shaping phase. Subsequently, the adaptive phase starts and causes the middle and distal phalanges to gradually close until their final positions are reached.

joint A as a whole, but will instead only cause link 4 to move relatively to link 2. This motion will occur simultaneously with the elongation of spring 1. When the middle phalanx is free to move, the former movement of link 4, will push the whole middle phalanx around joint C, since link 5 is not blocked and spring 2 is not elongated. Alternatively, link 5 could be hindered, which triggers a similar motion as described for the proximal phalanx. This motion includes the elongation of spring 2, whilst link 7 moves relatively to link 5. As a result, the distal phalanx will then rotate around joint E. Lastly, the order can be reversed, where an object blocks link 5 first and subsequently followed by link 2. The whole motion is complete when either an object is fully grasped, or when all maximum angles of the joints A, C and E are reached.

Mechanical stops were applied at joints A, C and E to ensure that the finger mechanism ends at the right angles, hence providing the desired range of motion. Rotation in joint A stops, when link 2 is approximately 9 degrees from the x-axis due to an extension of the frame. This construction is shown in figure 1-5. Furthermore, rotation in joints C and E stop when their connecting links become perpendicular. This also prevents a configuration where the springs pull the system further into a flexed state, which is then irreversible, meaning that the fully extended configuration can no longer be reached. This mechanism is depicted in figure 1-6.

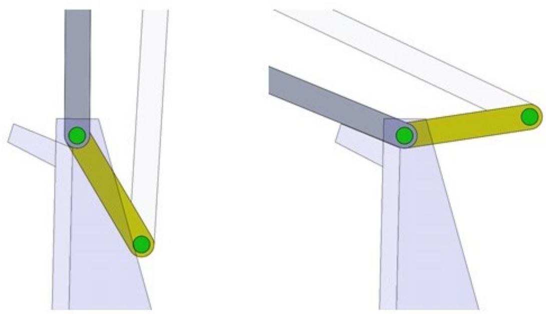

Figure 1-5: The mechanical stop at the MCP joint, that forces the mechanism into its grasping stage.

To ensure that the finger mechanism is at rest when there is no actuation force, as depicted in 1-3, two mechanical stops are applied at joints B and D. These stops are necessary, since the springs naturally want to be in their shortest state, thereby creating the smallest possible angle at both joints B (between links 1 and 4) and D (between links 3 and 7). The resulting mechanical stops are shown in figure 1-7.

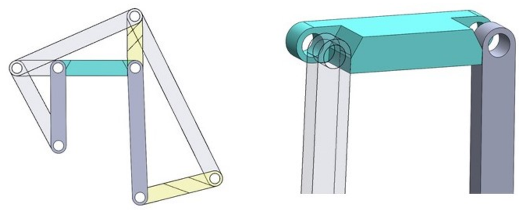

Figure 1-6: The mechanical stops embedded in link 5, which cause the PIP and DIP joints to stop rotating at an angle of 90 degrees relative to their adjacent link.

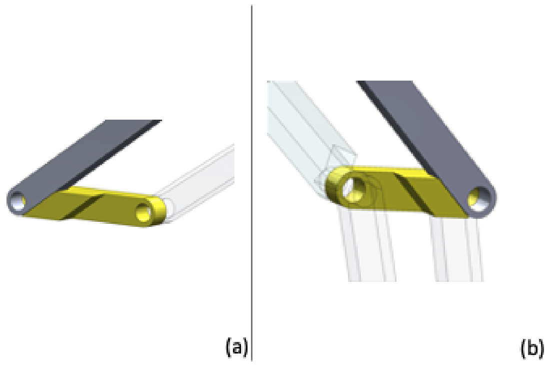

Figure 1-7: The mechanical stops, that create a stable rest position. A, shows the rear view of link 1 (in yellow). B, shows the front view of link 3 (in yellow).

1-2-3 Prototype design

Prior to creating the presented conceptual prosthesis as a whole, the finger transmission mechanism was built as an initial prototype. Building a prototype will show how well it is capable of being manufactured and assembled. More importantly, the prototype of the finger mechanism will serve as a proof-of-principle, where its working principle is tested along with its capabilities of fulfilling the criteria created for the full hand prosthesis.

A frame had to be designed to combine the finger transmission mechanism and pneumatic cylinder actuator into a working prototype. This frame mainly provides two fixed pivot points, namely one at the bottom of the cylinder and one at the MCP joint. Additionally, a connector piece was made to connect the pneumatic cylinder actuator to the finger transmission mechanism. The full design of this prototype in CAD is given in figure 1-8.

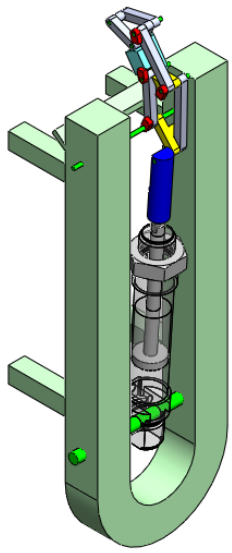

Figure 1-8: The CAD of the proof-of-principle prototype. The full design includes the frame (green), cylinder (grey), connector (blue) and finger transmission mechanism.

1-2-4 Materials

The prototype design was translated into a working prototype of the finger transmission mechanism. Instead of using AL-7075 T6 as intended for the whole hand, this prototype mainly consists of Polylactic acid (PLA), since it will serve as a proof-of-principle. The nine links and frame were made using an Ultimaker S3, which is a 3D printer that employs a technique known as fused deposition modeling (FDM). The printer was set to use an infill density of 70 percent and triangle infill pattern. Moreover, a layer height of 0.1 mm and line width of 0.4 mm were used. Stainless steel pins of 2 mm diameter were used as pivots in the joints and were capped with star-locks. Furthermore, a FESTO DSNU-16-25-P-A was used as the pneumatic actuator. This actuator is double acting, has a bore of 16mm and a stroke of 25mm. In addition, two sets of springs were used, which are shown in the following two tables 1-1 and 1-2. Where, c is the spring stiffness, l the spring length without tension, s the maximum stretch, Fn the maximum force and Dm the diameter center to center of the spring. Lastly, the connector piece was made of stainless steel and was manufactured in the TU Delft workshop.

Table 1-1: Spring properties of set 1

Properties	Spring 1	Spring 2
	T330	T540
c (N/mm)	0.15	0.78
l (mm)	30.40	17.50
s (mm)	35.1	16.9
Fn (N)	9.5	15.4
Dm (mm)	5.45	6.8

Table 1-2: Spring properties of set 2

Properties	Spring 1	Spring 2
	T290	TR390
c (N/mm)	0.9	1.44
l (mm)	26.40	15.90
s (mm)	14.7	9.04
Fn (N)	15.7	15.2
Dm (mm)	2.95	3.87

1-2-5 Prototype evaluation

The size and weight criteria are irrelevant to measure, due to the purpose of this prototype, being a proof-of-principle. Two criteria, degrees of freedom and range of motion, can be determined using prior knowledge of the design and visual inspection. Essentially, there are only two criteria that can be tested using this prototype, which are the force and adaptive grip.

Despite the force criterion being measured using a cylindrical grip on a dynamometer, the force for this prototype is measured at the fingertip using a pinching motion. The cylindrical grip could only be properly tested when a full hand would be assembled. The fingertip force will be measured using a load cell. The placement of this load cell relative to the fingertip was achieved by an additional frame. This frame fits over the legs of the main frame. The resulting set-up is shown in figure 1-9. This set-up will be used to test the fingertip force with an increasing supply pressure using two sets of springs. Supply pressure increments of 0.5 bar were used. The finger should move in its fully extended state, pre-shaping phase, for it to firmly reach the load cell. The load cell was operated using LabView. The force was measured over time and exported to a text file. Subsequently, the values from the text file were copied to MATLAB. The data was processed in MATLAB, consequently plotting the results in one figure.

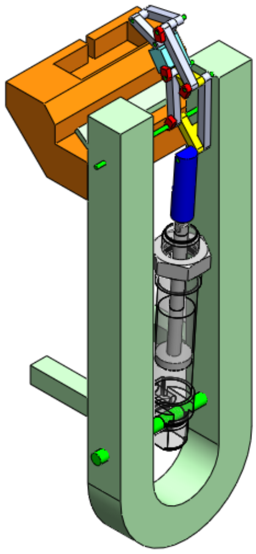

Figure 1-9: The test set-up of the prototype with the additional frame for the load cell.

The construction of the test frame and mechanical stops of the prototype will show if the mechanism will adapt to these obstructions when the cylinder is pressurized. In addition, the middle phalanx will be hindered in another test, to show if the order of hindrance causes a different or similar adaptive effect. These motions will be captured using the rear main camera of a OnePlus 6T. This camera uses the Sony IMX 519 sensor and has 16 megapixels with a pixel size of 1,22 micrometer. The included aperture is f/1.7.

1-3 Results

1-3-1 Prototype

The resulting physical prototype is shown in figure 1-10. Besides that, figure 1-11 shows an enlarged image of the finger transmission mechanism part of the prototype. The link sizes of the proximal, middle and distal phalanges are 25 mm, 15 mm and 15 mm respectively. The outer dimensions of the finger transmission mechanism, excluding springs, are (length x width x depth) 55 x 23 x 15 mm. The range of motion of the mechanisms MCP, PIP and DIP are 76, 58 and 56 degrees respectively. In addition, the MCP, PIP and DIP each provide a degree of freedom, resulting in 3 degrees of freedom per finger.

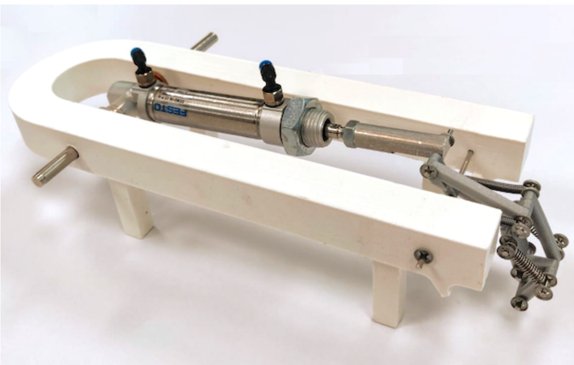

Figure 1-10: The physical prototype of the full test design. This image contains the frame, pneumatic cylinder, connector and finger transmission mechanism.

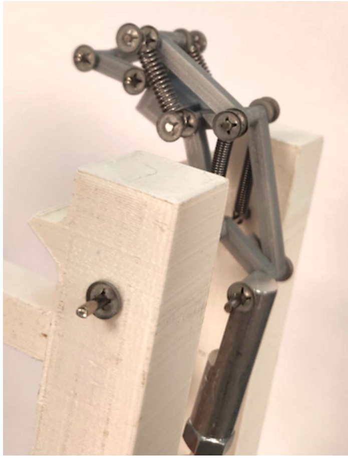

Figure 1-11: An enlarged image of the finger transmission mechanism part of the prototype. The bottom part is the connector, which is not part of this mechanism.

1-3-2 Performance

The performance of the prototype was measured with the fingertip force and the capability of adaptive grip. The results of the former are presented in figure 1-12, where the force is plotted as a function of the pressure. The test

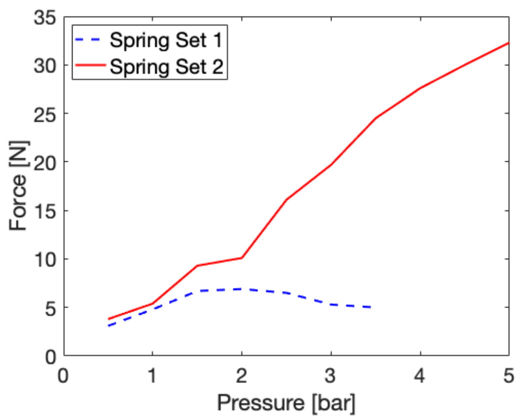

Figure 1-12: The test results of the fingertip force measurements. The force is plotted as a function of the pressure. The results using spring set 1 are shown with the blue dashed line and the results from spring set 2 are shown with the red line.

using spring set 1 was terminated after 3.5 bar. This was due the elongation of spring 1, consequently displacing the fingertip to the edge of the sensor, instead of remaining centered. Additionally, the test using spring set 2 was terminated, since the high forces broke the connection of link 2 in joint C at a pressure of 5.5 bar. The maximum force at the fingertip that was obtained using 5 bar was 32.3 N, where the actuator force was 100.5 N.

The other metric that shows the performance of the prototype is the adaptive grip. The capabilities of adaptive grip without and with obstruction are shown in figures 1-13 and 1-14 respectively.

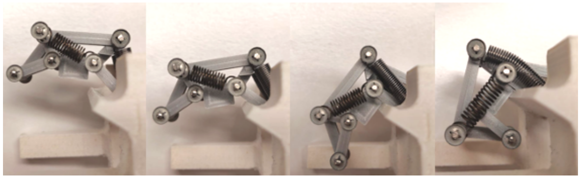

Figure 1-13: The resulting motion of the adaptive grip mechanism (with spring set 1).

Figure 1-14: The resulting motion of the adaptive grip mechanism when the proximal phalanx is obstructed (with spring set 1).

1-4 Discussion

1-4-1 Design criteria

Size

The preliminary design was supposedly made of AL-7075 T6, which is able to stay within the size as is quantified in figure 1-1. Contrarily, the prototype was made of PLA. Therefore, it is slightly longer and has a width of 23mm, including the springs of set 2. The distal phalanx was slightly up-scaled to allow for the designed range of motion to occur, thereby making it 3 mm longer than originally intended. The relatively large width of the prototype is due to the required thickness of the links needed to sustain similar forces as intended for the conceptual design. The prototype links were therefore twice as thick. Combining that with the post processing of the holes in the links, which created some rough edges, the assembled links were not as closely connected as originally intended, hence the relatively large width of the finger. Lastly, the contacting link (5) of the middle phalanx was created with extra large mechanical stops, since the level of detail required for the theoretical design requires higher quality manufacturing and assembly.

Weight

The weight of the hand should also remain within its limit of 106 grams. However, weighing the prototype was irrelevant, since it serves as a proof-of-principle. Alternatively, the weight of the theoretical concept can be estimated. Plettenburg (2005) designed the WILMER 21-t025, which is a piston in cylinder actuator. This design is characterised by its low mass and thin cylinder wall, being only 1 mm larger in diameter than the piston bore. The weight of the Wilmer design actuator used for this prosthesis should result in approximately 6.8 grams. Furthermore, the finger mechanism was calculated using its CAD in SolidWorks, resulting in approximately 3.5 grams. A similar approach for the hand frame was taken and was approximated at 42 grams. Adding these values together would result in a hand with a total weight of 93.5 grams. Despite being below the threshold of 106 grams, the presented weight of 93.5 grams could be drastically reduced by optimizing the contact area of the hand frame, whilst using as little material as possible.

Force

The fingertip force of 32.3 N was measured with an actuator force of 100.5 N, whilst using spring set 2. A higher actuator force resulted in the failure of the ending of link 2 at joint C (PIP). This failure is probably caused by the quality and direction of the 3D printed part. Fused deposition modelling ensures the strength of a part by printing in a single direction, therefore resulting in either yield or shear strength. The yield strength of the links was determined and was supposedly sufficient. However, a combination of both tension and shear force probably resulted in the breaking of the part. The force transmission ratio of 3.1 is mainly caused by the varying distance and angle of the line of action of the actuator force and the resulting reaction force at the fingertip. Additional force losses are caused by frictional forces and probably some tension in the springs. Furthermore, slight deviations could be possible due to the margin of error caused by the quality and calibration of the sensor and also the consistency of the contact point between the fingertip and sensor. To reduce the error of the contact point, springs with a high initial tension and high stiffness were used. Belter et al. (2013) performed a review of anthropomorphic prosthetic hands that were commercially available. The study showed that the average fingertip force of the middle finger of five different prosthetic hands was 8.5 N with a maximum of 14.5 N. The fingertip force of this prototype is much higher, whilst also being designed in a small and lightweight form factor. The fingertip force is, however, not directly related to the force design criterion, which was the grasping force that was measured with a dynanometer. Fortunately, the sum of added forces offers a promising perspective for achieving the grasping force of 46 N. Contrarily, the theoretical achievement of this criterion is not representative of the performance of the grasp. This is in part due to the force distribution of the prosthesis, which is more evenly and wider spread in human hands (Kargov et al., 2004). Contributing to this, is the non-yielding characteristic of materials chosen for prosthesis, which cause a relatively smaller contact area between the prosthesis and objects. Moreover, the friction between the prosthesis and objects can drastically change the performance of a grasp.

Adaptive grip

Adaptive grip was tested using two sets of springs. The first set of springs being lowest in stiffness, showed that is was easily and fully capable of adapting to obstructions. However, when the input force got too high, the finger would rotate around the PIP and DIP prior to completing the MCP rotation, i.e. prematurely entering the adaptive phase. Alternatively, the second set of springs showed that with a higher stiffness the finger is more likely to rotate around the MCP as a whole, but has more difficulty with the subsequent motions in the PIP and DIP. Ideally, a spring is required that has a high initial tension and low stiffness. Another observation was made, where the pin at joint C interferes with the elongated spring 1 (figure 1-13).

Degrees of freedom

The prototype showed that the finger transmission mechanism has 3-DOF. The conceptual design provides 3-DOF in each finger and 2-DOF in the thumb. However, the conceptual design shows a total of 14-DOF, where the thumb is incapable of circumduction. Thus, lacking one degree of freedom.

Range of motion

The range of motion as defined by Kapandji (1971), Stillfried et al. (2014), excluded specifics concerning the exact start and end angles of the appointed joints. The obtained range of motion from those articles were assumingly measured from a fully extended configuration to a fully flexed configuration. However, in practice the human hand does not fully extend before grasping an object, and it is therefore not required to possess the full range of motion. Besides that, the position and stroke length of the actuators forced compromises in the starting angles of the mechanism. The conceptual design contains a larger range of motion, due to the specific design of the cylinder stroke and dimensions of the links. Contrarily, the prototype used a cylinder with a slightly shorter stroke, thereby forcing the finger into a slightly more flexed starting configuration. In addition, the thicker links required an extension of link 5, which also caused a slightly more flexed starting configuration. Despite the smaller range of motion of the prosthetic finger than quantified as the design criterion, the finger transmission mechanism is ought to have sufficient range of motion to complete activities of daily living.

1-4-2 Recommendations

The conceptual and prototype design both require additional improvements and research to increase their eventual acceptance rates. The size and layout of the components of the prosthesis can be optimized. To pursue an anthropomorphic design, the lengths of the links should be adjusted according to the lengths of the subsequent fingers. Besides that, the index and middle finger, and/or the ring and pink finger could be coupled through the use of longer pins, thereby creating more space in-between the finger for different spring sizes. Furthermore, the comfort of the prosthesis largely depends on the weight of the prosthesis and the distribution of its components. The hand frame should be updated to get a lighter weight whilst keeping the maximum amount of contact area. Besides that, a socket should be designed that prioritises comfort and aims to position the pressure regulator, valve and power supply strategically. This could be done by positioning the heavier components proximal and lighter components distal. Additionally, the socket should be designed that is usable by amputees of varying residual limb lengths. In order to improve the grasp performance of the prosthesis, grip tape could be applied to the links that make contact with objects. The added friction could potentially reduce the ejection phenomenon of grasping objects. Moreover, the placement and transmission of the cylinder and its force require improvements. The current design causes uneven torque distribution throughout the stroke, due to the constantly changing line of action of the force applied at joint B.

Two important factors that influence the performance of a prosthesis are control and feedback. However, these factors were considered, but not pursued during this stage of the design. The input signal should either have varying gain or it should be able to distinguish to different input signals. These are two different ways of achieving the same principle, where the first phase uses less force to move the whole finger, and the second phase uses higher forces to firmly grasp an object. In addition, this would require springs with lower stiffness, which could result in a larger grasping force due to less opposition of the springs. Contrarily to the input signal variations, it should be considered that the supply pressure becomes sub-optimal related to the gas usage, consequently also the total number of operations achievable on a single CO_2 cartridge. Preferably, the input signal originates from myoelectric signals that are either measured at different muscles, or from one muscle but more easily distinguishable due to the controllability of their input gain, i.e. level of contraction of the residual underlying muscle.

1-5 Conclusion

This study presented the design of a pneumatically powered hand prosthesis for toddlers. The main reasons that cause abandonment of prostheses were addressed by creating and partly fulfilling design criteria that address the functionality and weight of a prosthesis. The theoretical concept showed a hand prosthesis design that is:

- sized for 4 year old children.
- lightweight, weighing less than 106 g.

The resulting prototype showed:

- a 3-DOF finger transmission mechanism.
- a fingertip force of 32.3 N at 5 bar.
- the capability of adaptive grip.

Altogether, the theoretical concept and prototype showed that the complete hand prosthesis could provide a promising solution to the high abandonment rates of upper extremity prosthesis.

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Chapter 2

Literature Study

An Overview of Pneumatically Powered Upper Extremity Prostheses from the 21ST century

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Abstract

A large proportion of upper extremity prosthesis are abandoned by their user. Commonly, the main reasons for abandonment are related to comfort, control and cosmetics. More specifically, due to heavy weight, low functionality and limited degrees of freedom. Pneumatically powered upper extremity prosthesis provide promising solutions, such as a high power to mass ratio, to reduce abandonment rates. A literature study was performed to explore the pneumatically powered upper extremity prostheses published since 1999. This study provides an overview and extensive discussion of the pneumatically powered upper extremity prostheses found in the literature study. The specifications, performance and working principles of the prostheses are elaborated and subsequently analyzed. This gives insight to the important requirements and features of pneumatically powered upper extremity prostheses. The most influential features to meet the requirements comfort and control, were determined to be adaptive grip and weight. In addition, there were no clear correlations found between the features of the prostheses. Furthermore, great variations in content between different articles was observed. Due to the different research and development stages of the prosthesis as well as different reasons for publication, some features were highlighted where others were neglected. As a result thereof, design trade-offs occurred, which due to its varying reasons caused inconsistent comparisons. This inconsistency should be addressed in the near future by creating standards for a set of requirements and the testing of a prosthesis. As a consequence, researcher should be able to evaluate and compare prosthesis more consistently, thereby eventually decreasing abandonment rates.

***K*eywords** Pneumatic · Prosthesis · Hand · Arm · Upper Extremity · Transradial · Transhumeral · Overview

2-1 Introduction

In 1996 the National Health Interview Survey reported that in the United States the prevalence of absence or loss of an upper extremity was approximately 102,000. Those who lost an arm or had an amputated hand conclude 41,000. The remaining 61,000 were described by those who lost one or more fingers (LaPlante et al., 1996). Ziegler-Graham et al. (2008) estimated that in the United States the number of persons living with the loss of a limb is expected to more than double from 1.6 million in 2005 to 3.6 million in 2050. This estimated trend shows that there is an increasing need for upper extremity prostheses, in order to alleviate the debilitation and distress. Many different designs have been proposed to restore the function and appearance of the missing limb. Prosthesis for upper extremity amputees can be categorized into cosmetics, body-powered and externally powered prosthesis. Figure 2-1 shows an overview of the general types of upper extremities prosthesis.

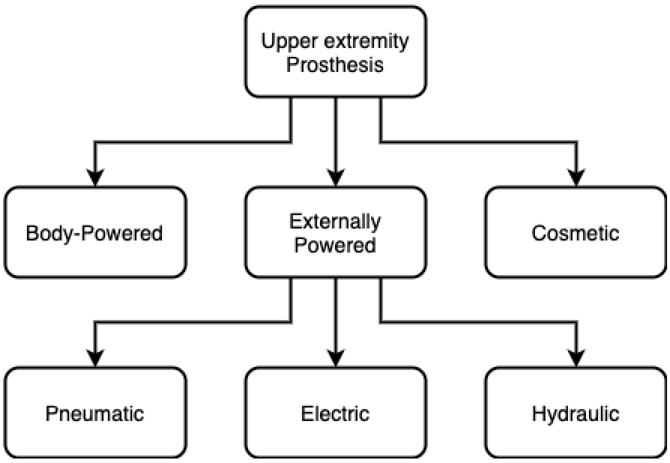

Figure 2-1: Overview of different types of upper extremity prosthesis

A cosmetic prosthesis is aesthetically pleasing and cheap, though limited in functionality due to its single configuration. A body-powered prosthesis is controlled by a harness, which is commonly attached to the shoulder. Through movements of the shoulder, i.e. flexion-extension and/or abduction-adduction, the prosthetic hand will open and close accordingly. Such a device provides the user with force and displacement feedback. Externally powered prostheses generally consist of a power source, controller and actuator (figure 2-2).

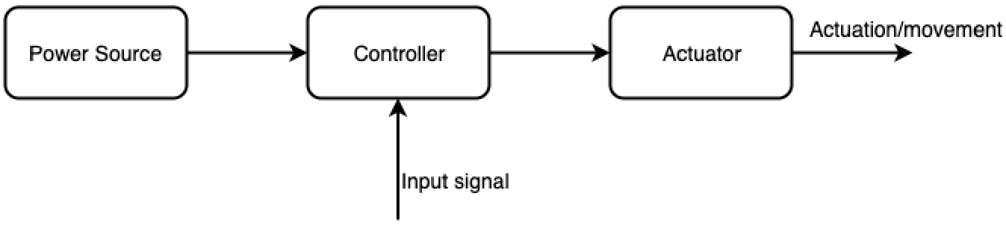

Figure 2-2: Overview of the components of an Externally Powered Prosthesis

These prostheses can be divided into three subcategories by actuator type, namely electric, hydraulic and pneumatic. The actuators draw power from their power source, which subsequently are battery packs, liquids and gasses. A controller is in place to regulate the supply from the power source to the actuator. This controller needs an input signal upon which it regulates the outgoing supply. This input signal acts as a kind of gatekeeper and is in general either mechanical or electrical. A mechanical activation requires a switch or button, where an electric activation requires an electric signal. A prosthesis that uses electromyographic (EMG) signals from the muscles of the user's residual limb to control the prosthesis is called a myoelectric prosthesis.

Creating an upper extremity prosthesis that functions well has been a challenge, since the simplest movements require complex controls (Freund, 1983; Schieber, 1995). The human hand consists of 27 bones, of which most are controlled through the activation of many muscles and tendons. There is an alternating synergy between different muscles, where some are activated and others inhibited at a certain time (Freund, 1983; Schieber, 1995). In addition, the hand contains many different sensors that provide one with feedback. The control and feedback of a human hand make for a complex overall system, which is hard to replicate into an anthropomorphic upper extremity prosthesis. Such a prosthesis has many requirements, which currently can only be partly met. Biddiss et al. (2007) analyzed and compared surveys of upper limb prosthesis acceptance and abandonment. They reported mean rejection rates of 45% and 35% for body-powered and electric prostheses respectively in pediatric populations. Significantly lower rates of rejection for both body-powered 26% and electric 23% devices were observed in adult populations. 22% of cessations attributable to prosthetic problems or discomfort. Plettenburg (1998) suggests that if any of the the following three requirements: cosmetics, comfort and control, is not fulfilled the prosthesis will be abandoned. Each of these three requirement will result in a list of design criteria. However, some criteria will be compromised and others met depending on the type of prosthesis (figure 2-1).

In general upper limb prostheses are abandoned due to heavy weight, low functionality and limited degrees of freedom (DOF) (Silcox et al., 1993; Atkins et al., 1996; Schulz et al., 2001). Despite a comparable mass between a prosthetic and human hand, prostheses are experienced as too heavy, since the generated torque from the mass of a transradial prosthesis is applied at a relative short part of the forearm (Belter et al., 2013). The low functionality and limited degrees of freedom are explained by the inadequacy of finger control and different grips. These shortcomings can result in a larger required force to grasp an object. Subsequently, amputees with a long residual limb could require a larger force to operate the prosthesis, since the actuator is located distal to the elbow (Fite et al., 2008). Besides that, a long residual limb also leaves little room for components of a prosthetic hand, such as the power source, controller and its actuators. Atkins et al. (1996) identified elements that users deemed necessary in the design of an improved upper limb prosthesis. These improvements include additional wrist movements, better control mechanisms that require less visual attention and ability to make coordinated motions of two joints. These shortcomings and improvements contribute to the three main criteria, namely cosmetics, comfort and control. Schultz et al. (2007) reported that experts ranked comfort first, followed by function and cosmetics. Despite this order, fulfilling one criteria will often compromise another. This trade-off will become apparent when the specific types of prosthesis are clarified.

Body-powered prostheses are limited to one or two control motions, which are often confined by the opening and closing of the hand. Despite the feedback that the device provides the user with, the device is in need of many improvements. Atkins et al. (1996) reported factors that needed to be included to satisfy users, which included better cables, harness comfort and gloving material. Furthermore, users complained about excessive wear temperature, abrasion of clothes, wire failure, unattractive appearance and harness discomfort and/or breakage (Biddiss et al., 2007).

Externally powered prostheses are commonly actuated by electromagnetic motors. These motors, with an appropriate form factor, have a low torque density compared to human joint actuation, therefore they require a transmission, which is commonly obtained with gears (Fite et al., 2008). The use of gear ratios reduces the power and speed of the motor due to an increased actuator weight and reduced efficiency (Fite et al., 2008; Peerdeman et al., 2012; Belter et al., 2013). Ideally, an electromagnetic motor for upper extremity prostheses, has a high torque and power density, while being fast and efficient, but also remaining compact and lightweight. Achieving these factors simultaneously is a challenge, since electromagnetic motors that provide enough speed and force are often heavy and bulky. Moreover, users of electrically powered prostheses can expect high cost and weight along with increased maintenance through glove and battery replacement (Biddiss et al., 2007). In contrast, Biddiss et al. (2007) also report that electrically powered prosthesis offer advantages in appearance, increased pinch strength, ease of operation and lack of harness.

The input signal for modern electrically powered prostheses is an electromyographic signal acquired from the muscles that lie underneath the skin of the residual limb, i.e. a myoelectric prostheses. Although the electromyographic signal is commonly used with electrically powered prostheses, it is not exclusive to it. The myoelectric prosthesis uses myoelectric sensors, which receive signals and send them to the controller. This controller processes these signals and will carry out a signal to the dedicated actuator. Depending on the design of the prosthesis, the complexity of its control increases with the amount of sensors and independent actuators. Peerdeman et al. (2011) reports that there are many classifiers that provide good accuracy for different wrist movements, but that there are far fewer results for different grasps. They also argue that results are not completely valid, since the studies used non-disabled subjects and a number of restraints in the contractions. Besides that, the classifiers for different grasps and wrist movements could not be used simultaneously and are not used in commercially available prostheses. Despite promising control capabilities, Peerdeman et al. (2011) concluded that the main reasons for abandonment of a myoelectric prosthesis are due to non-intuitive control, lack of sufficient feedback and insufficient functionality. In addition, Atkins et al. (1996) reports that myoelectric prostheses require better gloving material, better batteries and charging units, and improved reliability for the hand and its electrodes. Furthermore, they reported that future developments should include greater finger movement, less visual attention, and greater wrist movement. Peerdeman et al. (2011) combined literature from (Atkins et al., 1996; Biddiss et al., 2007) with their own workshop to determine requirements that could reduce the abandonment rate. In their workshop participants determined the requirements needed for multiple activities of daily living. The basic structure of this list consists of three requirements: feedback, EMG sensing and control. Peerdeman et al. (2011) found that only part of these requirements are met by modern research prototypes. Research in the areas of feedback, EMG sensing and control were found to be mainly technology driven. Instead, more attention should be paid to the integration and validation of myoelectric prosthesis. Due to limitations of electromagnetic motors, Peerdeman et al. (2011) highly recommends research into an alternative actuation.

The alternative actuation for externally powered prosthesis is pneumatic. These type of actuators can be categorized into pneumatic cylinders or fluidic actuators. Pneumatic cylinders use compressed gas to move the piston, thereby allowing displacement of the valve stem. Fluidic actuators are actuators that manipulate a material by expanding it using pressurized gas and thereby exerting force and/or movement in a desired direction. This type of actuation can be controlled in various ways. Three different types of fluidic actuators are distinguished: bellows, McKibben Pneumatic Artificial Muscle (PAM) and Pneumatic Networks (PneuNets). The most basic actuator from these three are bellows. Bellows can be described as a flexible material that expands when pressurized with air. Alternatively, a PAM works similar to a biological muscle, where it shortens with contraction and elongates when relaxed, thereby providing linear actuation. Such an actuator is composed of a tube, which is covered by a braided mesh. The tube will expand when pressure is applied. This expansion causes the braided mesh to shorten in longitudinal direction, thereby generating a force. The last option, PneuNets, consist of embedded channels inside an elastomer. Upon pressurization these actuators expand more in certain regions (flexible) than others (stiff), thereby creating motion. This actuation method can vary greatly due to its large variety in available parameters. Variations in the geometry of the channels (shape and thickness) and the material choice (possibly combining multiple) alters the properties of the actuator, which allows for different controlled deformations of the actuator and its surrounding structures. Pneumatic actuators have a high power density, which allows for a small and lightweight actuator to be included in the design of a hand prosthesis. In general, the pressurized gas used in a pneumatic drive source is compressible, which improves the safety of such devices. In addition, there is no power consumption required, assuming that there is no leakage, when a grasp is maintained, i.e. during isometric contraction. Contrarily, pneumatic actuators suffer from the nonlinearity of the compressible gas, which makes it difficult to control. Compact power sources supply the user with a limited amount of grasping cycles, where larger sources lack an appropriate form factor.

Recent advancements in technology seem to favor the direction of externally powered prostheses. This concluding section of the introduction justifies why this study researched the different pneumatically powered upper extremity prostheses. The main advantages and disadvantages of pneumatic systems compared to electrical systems will be summarized to clarify this choice. Pneumatic actuators are an alternative to electromagnetic motors, since they can exert equal force with over 10 times less mass (Peerdeman et al., 2012). Moreover, pneumatic systems are often safer than electric systems since they use gas, where electric systems are more susceptible to environmental conditions. The storage of the pneumatic power source is lighter and smaller than the batteries of an electric power source. However, the capacity of pneumatic power source is smaller than that of an electric power source (Peerdeman et al., 2012). So, in general a pneumatically powered prosthesis is more capable of supplying sufficient power with a lower mass and an appropriate form factor than electrically powered prosthesis. They are safer to use and have similar opening and closing speeds compared to electrically powered prostheses. Peerdeman et al. (2012) claims that a pneumatic system can outperform electric systems when the design is improved. This study explores the pneumatic possibilities and innovations used in upper extremity prosthesis. A literature study will be performed to provide an overview of pneumatically powered upper extremity prostheses.

2-2 Method

A literature search was performed using Scopus and Espacenet. The following search terms were used:

(pneumatic OR gas-actuated)* AND *(arm OR hand OR transhumeral OR transradial* OR *(upper AND extermity))* AND *prosthe**

Prior to this search some inclusion and exclusion criteria were determined. The search was limited to articles published from 1999 until present and also to English language. The criteria concerning publication years was implemented due to the prior research performed by this study's supervisor about the design and development of a pneumatically powered hand prosthesis for children (Plettenburg, 2002). The search resulted in 152 articles in Scopus and 9 patents in Espacenet. Further reduction of these numbers are explained by the discarded articles, which title and abstracts did not relate to the purpose of this overview. Furthermore, other articles were discarded due to a lack of detailed information. Contrarily, articles from other sources were added. Eventually, a total of 11 articles and 2 patents were used in this overview. The process of the literature study using Scopus is depicted in figure 2-3.

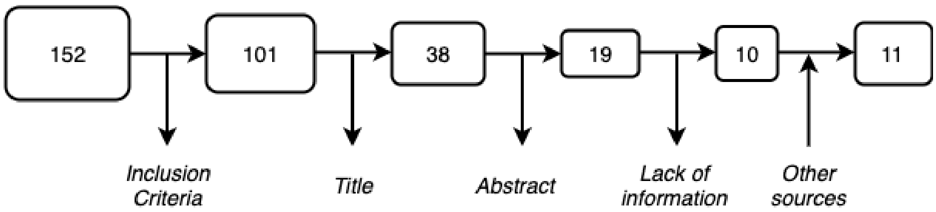

Figure 2-3: Article selection process using Scopus

2-3 Results

The results are split up into three main subsection. The first subsection contains articles that provided a prosthetic device with pneumatic actuation. The second subsection contains an article that used pneumatics alternatively. The third subsection contains article references, which were not included due to a lack of information or misleading abstract.

2-3-1 Pneumatically powered upper extremity prostheses

This subsection shows and describes twelve prosthetic devices. First, the specifications of the different prostheses are presented. Followed by, the explanation of the working principles of the prostheses. Figure 2-12 and 2-13 provide images of the prostheses and can be found in the appendix.

Specifications

Table 2-1 shows the specifications of the devices provided by the article or deducted from it. In order to compare and analyze the results as objectively as possible the prosthetic devices are further categorized. A differentiation has been made between the following parts that can be moved upon actuation: hand, wrist and forearm. The articles that are categorized by *'Hand'* contain prosthetic devices in which solely finger movements are possible. The articles that are categorized by *'Hand + Forearm'* contain prosthetic devices in which also solely finger movements are possible. The articles that are categorized by *'Hand + Wrist'* contain prosthetic devices that allow for both finger and wrist movements. The articles that are categorized by *'Hand + Forearm + Wrist'* contain prosthetic devices that are capable of both finger and wrist movements. The latter category contains one prosthetic device from (Fite et al., 2008) which is capable of elbow movement.

The specifications of the table consist of actuation method, number of actuators, number of joints, degrees of freedom, adaptive grip, weight, size, grip force, speed and pressure. In order to verify that the table is interpreted properly, three of these specifications will be shortly elaborated. The number of degrees of freedom (DOF) can be described by the number of independent movements a system can make. That is if the state of multiple joints can be determined with the state of a single actuator, than this is described as a single DOF. Furthermore, adaptive grip is when the state of the joints depends on the contact state of the other joints. This means that different parts of the finger that are actuated by the same actuator, i.e. underactuation, will adapt to the size and shape of the object. Finally, the speed of the prostheses in milliseconds represents the time it takes for full flexion and extension to complete.

In addition to the specifications from table 2-1, two other specifications will be presented, namely operation cycles and movement of joints. The number of operation cycles has been mentioned by only three articles. Fite et al. (2008) provided that 200ml of their power source hydrogen peroxide can provide 55KJ of work. Moreover, Nemoto et al. (2018) stated that their device could be used 150 times with 38g liquid CO2. Additionally, Devi et al. (2018) used a 12V battery to power a mini compressor to generate pressurized air. Lastly, Kim et al. (2018) were able to operate their hand 1250 times with 80ml of 70% hydrogen peroxide. The use of these different power sources will be elaborated in the upcoming section that covers working principles. Furthermore, the range of motion of the joints from the different prostheses is shown in table 2-2.

Working Principle

The working principles of the different prostheses will be explained. This includes elaborations of the power source, control system, actuation method and finger kinematics. The same categories and order will be applied as presented in table 2-1.

Hand + Forearm + Wrist
Fite et al. (2008) designed a prosthesis capable of elbow, wrist and hand movements. The prosthesis was powered by cold gas nitrogen, but was designed to work with a monopropellant hydrogen peroxide. This substance releases energy through exothermic chemical decomposition. When the valves were opened, a catalyst was utilized to convert the liquid hydrogen peroxide to pressurized gas. This gas was used to power nine actuators through nine servovalves. The elbow consisted of one cylinder and one servovalve to control its flexion. Movements in the hand and wrist were realized through eight servovalves and eight cylinders, both of which used five for the hand and three for the wrist. A linear proportional and integral (PI) controller was used to operate the valves using feedback from force and angle sensors. The force generated by the actuator was transferred into motion by pulling a series of tendons that span multiple joints. The joints were fitted with wound torsional springs, which allowed for passive extension of the joints due to its compliant joint forces.

The prosthesis from Polhemus et al. (2013) used a hybrid control system, which translated shoulder movements into input signals. A single gas reservoir was used to power the prosthesis, but its location was not provided. In addition, a microprocessor was used to distinguish five different movements and subsequently control six solenoid valves. These valves operated eight actuators, consisting of six PAMs and two pneumatic rotary actuators. The former being used to control flexion in the fingers and wrist. The latter was used to control wrist rotation, i.e. suppination and pronation. The solenoid valves were strategically coupled to create five predetermined movements. Furthermore, the thumb was controlled by an internal mechanism and could be manually placed in two different positions. The neutral positions of the joints were reached by a return force from torsion springs, that were placed in each joint. Unfortunately, the article provided no further technical details concerning the finger kinematics.

Takeda et al. (2009) developed a prosthetic device that used eighteen PAMs to operate the hand and two motors to operate the wrist. Their PAM design is similar to the McKibben-type actuator, but was capable to operate with low pressure and low volume of air. Besides that, its size was greatly reduced, allowing it to fit inside the finger joints. All the interphalangeal joints (IP, PIP, DIP) contained a single PAM for flexion and a rubber gum for extension. The metacarpophalangeal joints (MCP) contained two PAMs to operate both flexion and extension. The carpometacarpal (CMC) joint was also controlled by two PAMs to allow abduction and adduction of the thumb. Moreover, an electro-pneumatic regulator was used to control the air pressure, which originated from an external source. Two distinct wrist movements, suppination/pronation and radial/ulnar deviation, could be achieved with two motors, that were controlled with fuzzy control.

Hand + Wrist

Schulz et al. (2001) created a transradial prosthesis that allowed controlled wrist and hand movements. The bellows actuator is shown in figure 2-4. This actuator design was fitted inside all the joints in the prosthesis. However, the PIP and DIP joints are coupled. The fingers contained actuators, flex sensors and touch sensors. The metacarpus housed a microcontroller, microvalves and micropump.

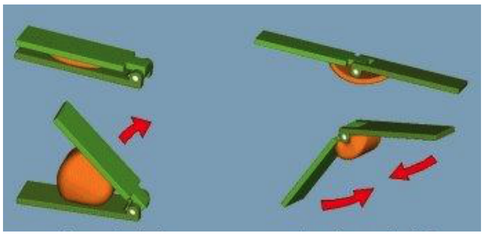

Figure 2-4: This figure shows the bellows actuator design that is incorporated in every joint in the prosthesis. The images on the left show the expansion principle of the actuator. The images on the right show the contraction principle of the actuator. This image was obtained from Schulz et al. (2001).

Hand + Forearm

Nemoto et al. (2018) designed an underactuated prosthetic hand that used PneuNets to create movements. The hand contained five of these actuators to control motion in all of the joints. The working principle of the actuator is depicted in figure 2-5. The natural motion of the actuator was constrained by an exoskeleton. The combination of the actuator with the exoskeleton created compliant finger movements. The design described in this article was controlled using a simple mechanical switch. Once switched, the solenoid valve opened, which allowed gas to flow from a CO2 cylinder through a pressure regulator into the actuators. This pressurized the actuators, subsequently causing flexion of the fingers. The material properties of the actuator, caused the finger to extent passively once the pressure was released. The CO2 cylinder, pressure regulator and solenoid valve can be worn separately from the prosthesis in a so called drive unit. In addition, the prosthesis contained a couple of passive joints. Two of these are placed in the carpometacarpal (CMC) joint, allowing the thumb to be placed in three different starting positions. The other two passive joints could be found in the wrist. These joints were locked in different configurations of wrist flexion/extension and suppination/pronation.

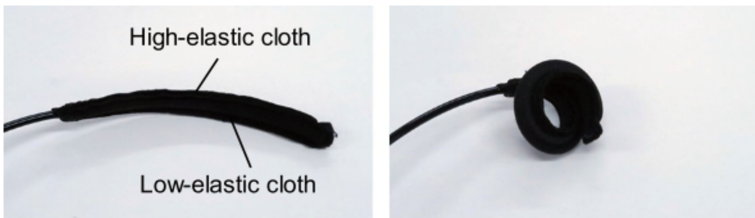

Figure 2-5: This figure shows the actuator without its exoskeleton, such that the natural motion of the actuator is unconstrained. The left image shows the neutral state of the actuator. The right image shows the pressurized state of the actuator. These images were obtained from Nemoto et al. (2018).

Whitesides et al. (2017) invented the PneuNets and created an embodiment in which the PneuNets were used as finger actuators for a prosthetic hand. A schematic illustrations of the actuation is depicted in figure 2-6. The finger actuator included fluidically interconnected chambers and a inextensible bottom layer. The pressurization of the elastic chambers in combination with the inelastic bottom, prevented radial expansion, but allowed lateral expansion, thereby creating a bending motion. This pressure was supplied by a compressor, located in the hand. A microprocessor, powered by a battery, controlled the air compressor and solenoid valves. A total of five finger actuators were used, which each connected to its own solenoid valve. Moreover, myoelectric sensors were used to recognize different muscle flexing patterns, resulting in different combinations of solenoid valve openings, thus different grasp configurations.

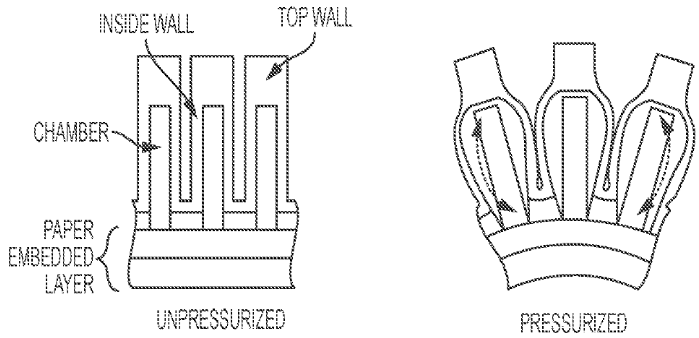

Figure 2-6: This figure shows the unpressurized state of the actuator on the left and the pressurized actuator on the right. These images were obtained from Whitesides et al. (2017).

Devi et al. (2018) investigated a PneuNet type actuator for prosthetic applications. The actuator is an Assymetric Bellow Flexible Pneumatic Actuator (ABFPA), which is made of embedded structures within an elastomer and varies in thickness in different areas. The research led to the actuator shown in figure 2-7, which could obtain the largest bending angle. A prosthesis was created that contained ten actuators to control the bending in ten different joints. Each actuator was coupled to its own solenoid valve, subsequently controlled by an Arduino microcontroller. The input signal for this controller was generated by the press of a button in a mobile app. Upon receiving this signal, the controller generated a signal for the specified solenoid valves to open and for a mini-compressor to generate a certain air pressure. The DC motors of the mini-compressor were powered by a 12V battery, which was also placed within the forearm of the device. This process caused the desired actuators to bend and subsequently create motion in the desired fingers. When the actuators were relieved of pressure the fingers passively returned to its neutral (extended) state, due to the material properties. This version of the prosthesis contained no sensors to provide feedback, but the developers claimed that with a tactile sensor a suitable feedback control system could be developed.

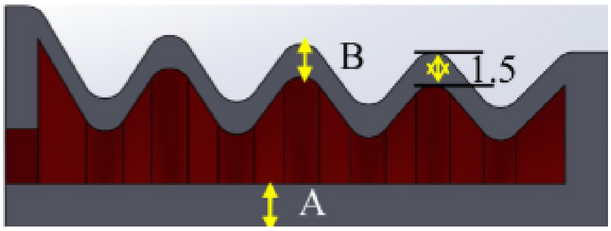

Figure 2-7: This figure shows the cross section of the ABFPA with A and B varying in thickness. This image was obtained from Devi et al. (2018).

Hand

Fras et al. (2018) proposed an underactuated prosthetic hand with adaptive grip. This hand used six actuators with six solenoid valves, to control five fingers with a total of twenty-one joints. Each finger was equipped with an actuator, except the thumb, which contained two. These actuators were used to cause flexion of the fingers, with the exception of the additional thumb actuator, which allowed the thumb to change between opposition and apposition positions. The actuator is a fiber-reinforced conical tube consisting of a helical thread surrounded by two silicone layers, as shown in figure 2-8. An external power source was used to apply pressure to the silicone structure, which caused it to expand. However, the helical thread constrained radial expansion, thereby only allowing longitudinal expansion. The actuators were surrounded by an exoskeleton that converted this longitudinal expansion into a bending motion. The actuators were controlled independently by Pulse-Width-Modulation (PWM) signals that operated the solenoid valves, allowing for varying pressure levels. These signals were generated by a Raspberry PI control unit that ran a Python code. Despite the adaptive grip, different grasps were pre-loaded in the control unit. This resulted in predetermined actuation of certain actuators, which caused flexion in certain fingers. The material properties of the actuators caused the fingers to move back to the neutral position when the pressure was released.

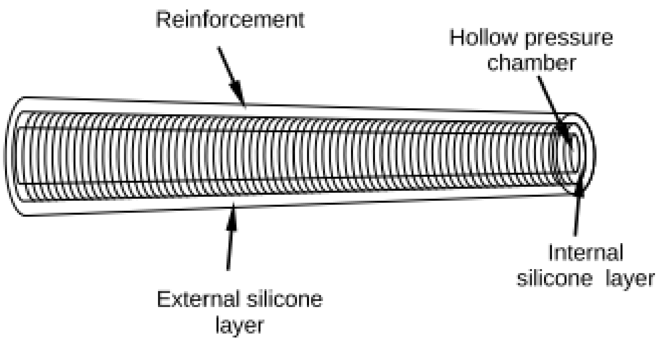

Figure 2-8: This schematic shows the actuator without its surrounding exoskeleton. This fiber-reinforced conical tube consists of a helical thread, which is surrounded by two silicone layers. This image was obtained from Fras et al. (2018).

Kim et al. (2018) provided a innovative actuator design that is targeted for prosthetic applications. The actuator design is shown schematically in figure 2-9. The actuator was referred to as a pneumatic dual-mode actuation mechanism, which used two single-acting cylinder actuators combined with a rack-and-pinion clutch. The MODE1 actuators had a smaller diameter with a large stroke, where the MODE2 actuators had the opposite with a large diameter and small stroke. The MODE1 actuator used a nylon monofilament instead of a rod to transfer the motion of the piston. This nylon monofilament was attached to the pinion clutch together with a Kevlar tendon. This Kevlar tendon transferred the movements from the clutch into bending of the finger. Due to a lack of space inside the actuator, the return spring was placed inside the finger. The MODE2 actuator used a conventional pneumatic cylinder design with a return spring. The rod of this actuator was connected to a rack clutch, which then drove the pinion clutch. The finger bent due to the actuation of MODE1 until it reached an object, since this then activated the MODE2 actuator. This switch signals was provided by a tension sensor in the third link of a finger. The MODE1 actuator was used to provide high movement speeds, whereas the MODE2 actuator was used to realize a high grasp force. Ultimately, reducing the amount of gas that needed to be consumed. The researcher tested this actuation unit in a five fingered prosthetic hand. The force generated by the Kevlar tendon was applied to each finger through a four-bar linkage structure. Due to space restrictions, the actuation unit was applied to the index finger, middle finger and thumb. The little and ring finger were operated by one conventional pneumatic actuator, where a differential mechanism was used to ensure adaptive grasping. The abduction and adduction of the thumb were performed by manually changing the position and subsequently locking or unlocking a passive rack-clutch mechanism. Furthermore, Matlab Simulink was used to control the device. This was done through ten solenoid valves, of which two where placed in the hand and the remaining eight in the forearm. Upon activation of the prosthesis, hydrogen peroxide was decomposed into oxygen gas and water. Subsequently, a pressure regulator was used to provide the right amount of pressure for the actuators. Depending on the state of the different solenoid valves, thereafter a type of grasp was performed due to pressurization of specific actuators and their modes.

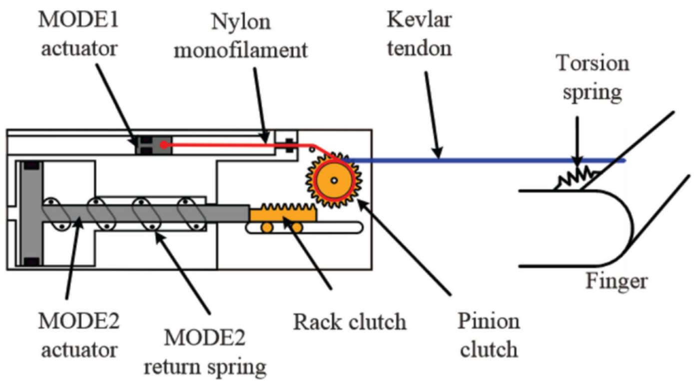

Figure 2-9: This figure shows a schematic of the pneumatic dual-mode actuation mechanism. This mechanism consists of a MODE1 actuator, MODE2 actuator, MODE2 return spring, nylon monofilament, kevlar tendon, torsion spring, rack clutch, pinion clutch and finger. This image was obtained from Kim et al. (2018).

Low et al. (2015) created a test setup with a prosthetic hand with three fingers. Each finger was made of a single actuator and has three joints, shown in figure 2-10 below. The actuators have the same working principle as PneuNets using embedded chambers and a restraining bottom layer. When pressure was applied into the finger, the channels inside the elastomer expanded in the most compliant areas. The alteration of thickness in different regions, caused the actuator to create flexion in the three joints. The amount of flexion that was achieved, depended on the pressure inside the actuator. In this test setup an external power source and external controller were used.

Figure 2-10: This figure shows the actuator design used to operate as a whole finger with 3 joints. The three stripped bottom parts represent, from left to right, the DIP, PIP and MCP joints. Furthermore, a restraining layer (black) is attached to the bottom of the actuator. The power source is connected to the adapter at the end of the actuator. This image was obtained from Low et al. (2015).

Vorob'ev et al. (2017) designed a prosthetic hand that was controlled with foot and toe movements. Four actuators with four valves allowed independent finger movements, with the exception of one pair of coupled fingers. The actuators were bellows cylinders, which applied a linear force in a similar way as a pneumatic cylinder. Through rotation levers the finger flexed when pressure was applied to the actuators. On the contrary, the extension of the fingers occurred through the return springs in the bellows actuators. Furthermore, an insole consisted of six tactile sensors, which measured the force provided by the toes and various parts of the foot. These signals were processed by a microcontroller that generated two signals. One was a signal for the coupled solenoid valve to open and the other signal was to task the air compressor to supply a certain pressure. This pressure level depended on the measured force in the dedicated tactile sensor, thus allowing variations in grip force.

Nishikawa et al. (2016) created a prosthetic hand that had an actuator in each joint. It used fourteen actuators to control fourteen joints with seven solenoid valves. The actuators were categorized as bellows, since they use a flexible material that expanded when pressurized with air. The design of this specific actuator is shown in figure 2-11. When the flexible material, being rubber, of this actuator expanded, it moved the attached links in a certain direction. This direction was controlled by the constraints of the actuator, which caused a rotation around a hinge, thereby bending the joint. This actuation caused flexion of the fingers, where alternatively extension of the fingers occurred due to the restoring force of the rubber when the pressure was released. A microcomputer was used to control the solenoid valves with a PWM signal. This allowed varying pressure levels to be used in the bellows, thereby altering the bending angle of the joints. Each solenoid valve was strategically coupled to certain actuators that allowed different grasps. These grasps were pre-loaded onto the microcomputer, but could be manually switched into three different modes. In order for the microcomputer to generate a PWM signal it required an input signal, which in this case was an EMG envelope. This version of the prosthesis used the EMG signal as a trigger signal, which was not proportional to generated force by the actuators. Although, the prosthesis was equipped with an angle and pressure sensor, these were not yet used in a control loop, but served the purpose of analytic data. This version of the prosthetic hand was equipped with an external power source and external microcomputer, since it was only used in a test setup.

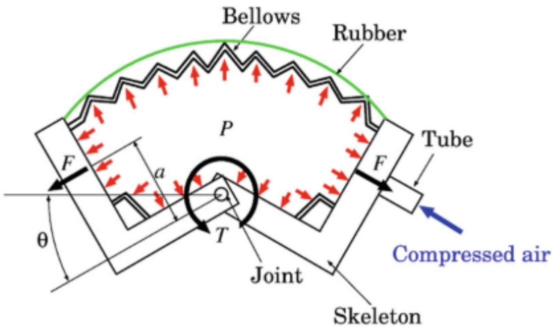

Figure 2-11: This figure shows the bellows actuator used in each of the fourteen joints. It consists of a bellows, rubber, tube, joint and skeleton. This image was obtained from Nishikawa et al. (2016).

2-3-2 Upper extremity prosthesis with pneumatic parts

Meza (2019) made a modular arm prosthesis, with a module that comprised of a pneumatic system. This module was responsible for the attachment of the prosthesis to the residual limb of the user. The module consisted of a pneumatic circuit, which expanded upon pressurization, thereby imprisoning the stump.

2-3-3 Excluded articles

Several articles were excluded due to a misleading abstract or due to a lack of details concerning the working principle and/or the specifications of the prosthesis. The following articles were discarded due to that reason: (Abboudi et al., 1999), (Omar et al., 2001), (Caldwell et al., 2002), (Ide et al., 2016) and (Tsujiuchi et al., 2006). In addition, Withrow et al. (2008) presented a design, which was very similar to that shown in the article from Fite et al. (2008). The article was discarded since there was no reasoning behind the differences in the design.

Table 2-1: This table shows the specifications of the prostheses from different developers. The specifications consist of the actuation method, number of actuators, number of joints, DOF, adaptive grip, weight, size, grip force, speed and pressure. The sizes of the prosthesis show the dimensions in the following order: length x width x height. *The weight presented is the weight of a single finger. ** Weight reduction of approximately 500g possible, when control unit is worn separately. ***Present the weight of the prosthesis without a power supply

Developer	Actuation Method	n x Actuators	n x Joints	DOF	Adaptive Grip
Hand + Forearm + Wrist					
(Fite et al., 2008)	Cylinder	9	21	21	Yes
(Polhemus et al., 2013)	PAM	8	16	7	No
(Takeda et al., 2009)	PAM (+ motors)	18 (+ 2)	17	17	No
Hand + Wrist					
(Schulz et al., 2001)	Bellows	18	15	18	No
Hand + Forearm					
(Nemoto et al., 2018)	PneuNet	5	30	17	Yes
(Whitesides et al., 2017)	PneuNet	5	15	15	Yes
(Devi et al., 2018)	PneuNet	10	10	10	No
Hand					
(Fras et al., 2018)	PneuNet	6	14	14	Yes
Kim et al. (2018)	Cylinder	4	11	6	No
(Low et al., 2015)	PneuNet	3	9	9	Yes
(Vorob'ev et al., 2017)	Bellows Cylinder	4	4	4	No
(Nishikawa et al., 2016)	Bellows	14	14	14	No

Developer	Weight (g)	Size (mm)	Grip Force (N)	Speed (ms)	Pressure (kPa)
Hand + Forearm + Wrist					
(Fite et al., 2008)	2000	-	-	-	2100
(Polhemus et al., 2013)	-	95 x 66	-	-	-
(Takeda et al., 2009)	-	-	-	-	-
Hand + Wrist					
(Schulz et al., 2001)	20*	-	3	100	0.525
Hand + Forearm					
(Nemoto et al., 2018)	755 **	-	1.5	1000	300
(Whitesides et al., 2017)	-	-	-	-	-
(Devi et al., 2018)	950**	-	0.46	100	0 - 500
Hand					
(Fras et al., 2018)	-	105 x 100 (child)	0.35	-	0 - 1000
Kim et al. (2018)	420***	198 x 79 x 31	29.1	600	500
(Low et al., 2015)	125***	146 (finger)	-	-	75 - 100
(Vorob'ev et al., 2017)	-	-	-	-	800
(Nishikawa et al., 2016)	240***	107 x 128 x 211	-	-	0 - 100

Table 2-2: This table shows the range of motion of different joints. MCP is short for metacarpophalangeal, PIP for proximal interphalangeal and DIP for distal interphalangeal. The table shows the results in degrees. If a - is shown, the prosthesis was capable of movement in that joint, but did not give a specific value. In addition, when a red cross is shown, the prosthesis was not capable of movement in that joint. *Instead of an PIP or DIP, this joint is an IP joint.

Developer	MCP	PIP	DIP	Thumb Flexion/ Extension	Thumb Abduction/ Adduction	Wrist Flexion/ Extension	Wrist Suppination/ Pronation	Wrist deviation Radial/ Ulnar
Hand + Forearm + Wrist								
(Fite et al., 2008)	-	-	-	-	-	-	-	-
(Polhemus et al., 2013)	-	-	-	-	90	±45	±45	x
(Takeda et al., 2009)	75	75	75	75	-	x	±30	±10
Hand + Wrist								
(Schulz et al., 2001)	100	100	100	100	-	-	±30	x
Hand + Forearm								
(Nemoto et al., 2018)	-	-	-	-	75	-	-	x
(Whitesides et al., 2017)	-	-	-	-	x	x	x	x
(Devi et al., 2018)	87	87*	x	87	x	x	x	x
Hand								
(Fras et al., 2018)	-	-	-	-	-	x	x	x
Kim et al. (2018)	-	-	x	-	-	x	x	x
(Low et al., 2015) (hand with 3 fingers)	55.7	141	126	x	x	x	x	x
(Vorob'ev et al., 2017)	-	x	x	x	x	x	x	x
(Nishikawa et al., 2016)	90	90	90	90	x	x	x	x

2-4 Discussion

In this section the specifications and working principles of the prostheses will be discussed. This includes a comparison between the different prostheses, and a comparison between the features of the prostheses and requirements found in literature. Besides that, a few other topics related to the design and performance of the prosthesis will be discussed.

Actuation method

A certain type of actuation method did not result in the same usage. Each of the methods were used in unique ways to generate movement in the prosthetic device. The cylinder and PAM have a similar effect, where a linear force is produced. The way that this force is transferred into motion, however, differs. The prosthetic devices that were actuated by pneumatic and bellows cylinders displaced a link by pulling a tendon, which due to joint coupling could move multiple links at once. PAMs were used in the exact same way, but were alternatively also used within joints. A reduction in size allowed these actuators to be placed inside the joints. Similarly, the bellow type actuators have been used to accommodate one DOF in each joint. In addition, PneuNets operated as a whole finger or were also incorporated in a joint. In contrast to PAMs and cylinders, PneuNets did not use tendons to move finger links. Variations in size have been used to either function as a single finger actuators or to function as an actuator in a single joint. Despite an identical actuation method, the applied method could slightly differ such that the actuator is used in its own unique manner.

Number of actuators, number of joints, DOF and adaptive grip

The relation between the number of actuators, number of joints, DOF and adaptive grip will be discussed. The number of actuators is related to the desired number of DOF. Fewer actuators are required when a linkage system allows multiple joints to be moved with a single actuator, i.e. adaptive grip. Alternatively, each DOF in a joint is obtained by a single actuator, with the exception of passive joints. A single joint can provide more than one DOF, but might not have actuators to accommodate these movements. These passive joints can be manually moved and locked in a certain direction, e.g. in the thumb and wrist. The use of passive joints could create DOF, that are not operated by the actuators. Since one joint can provide multiple DOF, it is possible that the amount of DOF surpasses the number of joints. Contrarily, the number of joints can also exceed the number of DOF, where a less intuitive joint coupling design is used or specifically when double joints are used as done in the prosthesis from Nemoto et al. (2018). They believed that the use of double joints would allow the finger to follow the movement of the actuator more closely. Furthermore, adaptive grip was seen in four different articles, where the means of achieving this differed. The prosthesis from Fite et al. (2008) contained joints that lacked shafts and bearings, but consisted of pairs of oppositely wound torsional springs. Together, these factors made the joints to be fully compliant, thereby enabling adaptive grip. Alternatively, the prostheses from Low et al. (2015), Fras et al. (2018), Nemoto et al. (2018) were capable of adaptive grip due the nature of their actuator, namely PneuNets. All these actuators bend in a certain direction due to set constraints, such as varying actuator properties or an exoskeleton. However, if a natural constraint, like an object, appeared than the constraint part will stop moving whilst if possible the remainder of the actuators keeps bending. This enables prosthesis with adaptive grip to contain multiple configurations with fewer actuators, thereby also simplifying control.

Weight

The weight of the prosthesis greatly impacts the comfort of the prosthesis. This is partly, due to the torque that is applied at a certain part of the limb, thereby impacting the way the weight is perceived. This does not happen with a normal human hand, where the torque is applied to the skeleton. A normal human hand weighs 400g on average(Chandler et al., 1975). The weight of prosthetic hands, both commercial and from research, lies between 350 and 2200g (Belter et al., 2013; Nemoto et al., 2018). The weights of the prosthetic devices from this research ranges from 125 to 2000g. However, note that these weights can differ drastically mostly due to the included or excluded prosthetic limbs. In addition, the weights of the prosthesis are often not consistent, since for example the control unit and/or power supply are external and not added to the total weight of the prosthesis as seen in Polhemus et al. (2013), Low et al. (2015), Nishikawa et al. (2016), Kim et al. (2018). In contrast to the prostheses with incomplete weights, the weights of the prostheses from Fite et al. (2008), Devi et al. (2018), Nemoto et al. (2018) are complete. Despite being a complete prosthesis, that does not mean that the different parts are all mounted on the prosthesis. The prosthesis from Nemoto et al. (2018), had a drive unit that could be worn separate from the prosthesis. On the other hand, the prosthesis from Fite et al. (2008) is hard to compare since this it the most extensive prosthesis, due to its additional elbow control. The prostheses from Nemoto et al. (2018) and Devi et al. (2018) consisted of similar parts, hand and forearm, and used the same type of actuation method. Both prostheses used PneuNets, though the prosthesis from Nemoto et al. (2018) used each actuator as finger actuator, where the prosthesis fromDevi et al. (2018) used an actuator in each joint. The prosthesis from Nemoto et al. (2018) weighs 755g, and is therefore lighter than the prosthesis from Devi et al. (2018), which weighs 950g. In addition, the prosthesis from Nemoto et al. (2018) used fewer actuators, with more DOF and adaptive grip while also exerting a larger grip force of approximately 3 times. Despite these advantages, the prosthesis was 10 times slower than the prosthesis from Devi et al. (2018). Both developers claimed that the weight can be reduced by 500g when the drive unit, which contained a compressor and battery, was mounted separately. The comparison of these two prostheses shows that trade-offs occur between different features of their designs.

Size
The dimensions of an adult human hand range between a length of 180 and 198mm and a width of 75 to 90mm (Belter et al., 2013). The prosthetic hand from Kim et al. (2018) is the only one that stays within this range. The other hands are a lot smaller, where only Fras et al. (2018) mentioned that their prosthesis was specifically developed for children. Furthermore, Low et al. (2015) showed the length of the finger actuator, thus excluding the palm of the hand. In addition, Nishikawa et al. (2016) showed a large deviation in height, which is explained by the dimensions of their L-shape socket that was used to test the device with. Moreover, the configuration of the hand could impact the measured size. This difference could be noticeable when the thumb is positioned differently. Furthermore, the smaller prosthesis that consisted solely of a hand part, were all unable to house a power source.

Grip Force
The grip force presented in this study showed the grip force measured at a fingertip. The required grip force for a prosthesis is difficult to determine, since there are multiple factors involved. The performance of a grip involves the force delivered by the actuator, grasp type, joint coupling, contact points and shape and size of the object. An adaptive grip will allow more contact points between the object and hand, thereby creating more friction between them, thus resulting in a better grip. A larger grip force might be required when adaptive grip is not implemented in the design of the prosthesis. Some articles only gave an actuator

force, but this was discarded, since this does not directly result in a grip force. The actuator needs to overcome the static friction and return force of the hand to generate a movement and eventually also a force at the fingertip. Moreover, some tasks demand more speed where other demand more grip force. The largest grip force of 29.1N was measured by Kim et al. (2018), with the dual-mode pneumatic cylinder design. The second largest force of 3N was measured in Schulz et al. (2001) using bellows actuators. A comparably small force of 0.46N was measured in Devi et al. (2018), which used PneuNets in each joint. None of these prostheses had adaptive grip. Contrarily, Fras et al. (2018), Nemoto et al. (2018) designed prosthetic hands that had adaptive grip. Despite a relative low force of 1.5 and 0.35N respectively, the grip might perform better. Some articles used a different metric to describe grip force, which was the capability of a hand to carry a certain amount of weight. The grip forces that resulted from those articles were discarded, since the values are not accurate, due to the multiple factors involved in grip force.

Speed

The data presented in table 2-1 concerning speed, consisted of a full cycle of complete flexion and extension, i.e. opening and closing the hand. The four articles that presented this data ranged between 100 and 1000ms. The speed in the smaller actuators seen in Schulz et al. (2001), Nemoto et al. (2018) were capable of performing a full cycle in 100ms. The prosthetic device from Nemoto et al. (2018) was approximately 10 times slower, due to its much larger size. The former having actuators incorporated in their joints and the latter having a finger actuator as a whole. All of these prostheses relied on their material properties to passively extend after voluntary closing. Lastly, the dual-mode pneumatic cylinder from Kim et al. (2018) was capable of completing a full cycle in approximately 600ms. This actuator was the only one that mentioned both flexion and extension speeds, where the flexion by actuator had a duration of 200ms to complete and the extension by spring 400ms. Peerdeman et al. (2012) showed that the pneumatic cylinder actuator needs approximately 0.3 seconds to overcome the spring force of the prosthetic hand. Thereby, proving that this adds significant time to the full cycle of opening and closing a hand. A similar scenario could occur with the use of Bellows or PneuNets actuators, where pressure builds up and eventually overcomes the material stiffness. It should be noted that the articles did not mention if the recorded speed includes potential pressure build up time. Furthermore, some articles showed a graph of the pressure level in the actuator and the time. However, reaching the pressure level does not mean that a certain bending angle ore motion is achieved.

Pressure

The operating pressure differs drastically between the prostheses, where it ranges between 0.525 kPa to 2100 kPa. The articles that used bellow actuators to operate their prosthesis, used the least operating pressure ranging from 0-100 kPa. However, these actuators were used to operate a single joint, therefore consisting of a smaller actuator that requires less pressure. The highest operating pressure is seen in Fite et al. (2008), which a pressure of 2100 kPa to drive their pneumatic cylinders. This pressure, more than twice as high as the second highest operating pressure, can be partly explained by the capability of elbow movement. Comparatively, the prosthesis from Kim et al. (2018) also used pneumatic cylinders to actuate their prosthesis, but this prosthesis consisted merely of a hand. Therefore, a much lower operating pressure of 500 kPa was used. Lastly, the middle class, consisting of the prostheses that used PneuNets as their actuation method. These actuators showed variations in operating pressure between 0-1000 kPa. Similarly to the bellow actuators, some PneuNets were used to control a single joint, where others were used to operate a whole finger. Despite these different purposes, the operating pressures for the different actuator types overlapped.

Operation cycles
The three articles that gave information about the durability or number of operation cycles, each did it in a different way. Fite et al. (2008) presented the work that their power source could provide, but did not present the work required for a single operation cycle. Besides that, Devi et al. (2018) provided that a 12V battery was used to power a mini compressor and solenoid valves. However, the power consumption of the mini compressor was not given, therefore not resulting in any valuable durability information. In contrast to the former two developers, Nemoto et al. (2018) stated that their prosthetic device was capable of 150 operation cycles. Moreover, Kim et al. (2018) provided that their prosthesis could be fully operated for 1250 times. Other developers were unable to provide information concerning the number of operation cycles, since their device was unfinished and possibly only used within a test setup. Noticeably, the number of operation cycles could differ, since in practice a desired grasp might not require full flexion and extension. In addition, this gas consumption can also change depending on the shape and weight of the object. Furthermore, the number of operation cycles is less important when the power source can be changed or charged rapidly.

Movement of joints
First of all there is a difference between types of joints, where some are mechanical and others are the actuator themselves. The variation in joint angles can be caused either due to mechanical constraints or due to the limitations of an actuator. Despite these variations, the capability of movements in different joints is a design choice from the developer. Furthermore, movement in joints could also be passive. A large variation in movable joints was noted, but there were no articles that provided any reasoning concerning those choices. An important parameter for the number of capable grasps is the possibility of the thumb to perform abduction and adduction.

Types of grasps
The types of grasps a prosthetic device can perform depends on the working principle thereof. The prostheses with adaptive grip were capable of performing multiple different grasps, which depend on the object and its placing. Alternatively, prostheses without adaptive grip were capable of a number of grasp that were predetermined. The capability of positioning the thumb in a configuration, which is either due to abduction or adduction, can greatly increase the number of possible grasps. In this study, there was no article that provided a finite number of possible grasps. The researcher often showed a few examples of different grasps to prove that their prosthesis worked accordingly. Thus, the amount of different grasps could be much larger than the few examples given.

Working Principle
The prostheses discussed in this study all use unique combinations of a power source, controller, actuator and finger kinematics. Multiple options for power sources were presented, which include a monopropellant, air compressor or external gas supply. The number of valves and their design can have significant impact on the working principle of the prosthesis. Since most articles did not mention any specifics concerning the valve design, the independent use of certain actuators could not be deducted. Furthermore, with the exception of one prosthetic from Nemoto et al. (2018), using a single mechanical switch to actuate the device, all others had a computerized approach using microcontrollers and microprocessors. The

control systems used in most studies were feed-forward. However, the two studies that used pneumatic cylinders to actuate their prosthesis used a control loop with feedback. Despite the feed-forward control systems used in most studies, different sensors were implemented providing analytic data. Furthermore, various input signals were used to control the prostheses. Ranging from translations of body movements, such as shoulder, foot and toe, to translations of myoelectric signals into input signals. Despite these efforts to differentiate multiple input signals, most prosthetic hands have more outputs, thereby creating a deficiency in the number of inputs. Alternatively, this problem was partly solved by creating a prosthesis capable of adaptive grip. Therefore, finger kinematics are important since this influences many features of a prosthesis such as the required number of actuators, obtained number of DOF and complexity of control.

Generally, pneumatic systems are more difficult to control due to their nonlinear behaviour. Note that the PneuNets require an increasing amount of pressure to change the state of the finger, that is if there is no contact with an object. The relation between pressure and valve opening is nonlinear, thereby creating an input signal that will not result in the desired pressure increase or decrease. Furthermore, the compressibility of gas creates another nonlinearity. Such hysteresis and nonlinearities were only briefly mentioned in two articles, being Devi et al. (2018), Kim et al. (2018).

Another important factor to ensure good control of the prosthesis is responsiveness. Responsiveness is influenced by the speed of the signalling, actuator and finger kinematics. It can be described as the time it takes from command to action. A high responsiveness will result in a quick response and therefore also a more intuitive control. There were, however, no reports of responsiveness in any of the studied articles.

Testing
Testing the performance of the various prostheses differed rigorously. Some articles did not test their prosthesis and others tested only certain aspects. The variation in tested aspects and lack of description of the test setup, often implied that the results were less valid. In addition, the testing was often performed in a simple test setup to show that the prosthesis had certain mechanical capabilities, where it is equally, if not more, important to show its use by an actual amputee. In some instances the testing showed only a part of the capabilities of a prosthesis, thereby not crossing the limits or showing the boundaries thereof. Furthermore, some testing was performed to explicitly show that an improved or innovative aspect of the prosthesis actually worked. Lastly, the evaluation of the testing by discussing the article's method and results was also limited.

Alternative metrics
A few aspects of the discussed prostheses have not yet been mentioned. These few aspects consists of the following: socket, housing of components, cost and noise. The design choices for a socket were not elaborated and also not evaluated by users. However, as mentioned earlier, comfort is an import requirement for a prosthesis, which can be greatly improved by a well designed socket. In addition, the housing of the components was often external, since the articles only showed the performance of a specific aspect of the prosthesis. In contrast, a few prosthetic devices included all their components within the prosthesis, but did not specify the reasoning behind the placing of their components. Furthermore, the length of the residual limb could impact both the socket design as well as the left over space for the components. Moreover, an important aspect to implement the device into practice is the cost of the device.

Only Devi et al. (2018) provided a method that reduced manufacturing cost by simplifying the process and using cheap materials. Solely this developer indicated the cost of a prosthesis resulting in approximately 800 USD.

Requirements

The different developers did not design their prosthesis according to an extensive list of design requirements. They often focused on only a few. In general, the articles listed the following shortcomings: bulky, heavy, rigid fingers, lack of compliance, complex and limited control and functionality. The requirements found in literature could be categorized within and as comfort, control and cosmetics. Most articles focused on showcasing a new control method, thereby neglecting the other two requirements, namely comfort and cosmetics. Despite heavy weight being an important drawback, this requirement was only addressed by using pneumatics as power source due to its power to mass ratio. However, this hardly increases comfort, since no attention was paid to a socket. Neglecting these requirements can be explained by the motivation behind the prosthesis being for the purpose of showcasing a specific feature instead of showcasing a prosthesis for commercial purposes. The increased focus on one aspect resulted in less attention for other requirements, thereby compromising them. For the same reason, many of the specifications were not given in the articles. In general, requirements often include improvements of certain aspects, but are not specific in their demands. It is therefore difficult to determine whether the prosthesis meets these requirements. Nonetheless, feedback was deemed important by users, but was rarely included in the control systems of the prostheses found in this study. Although requirements are important, prosthesis can also range in control, comfort and cosmetics to address a different type of lifestyle.

2-5 Conclusion

This study provided an overview and analysis of upper extremity prostheses that were pneumatically powered. The specifications and working principles of the different prostheses were elaborated and compared. The number of prostheses that were provided was limited, partly due to the relative unexplored possibilities of pneumatics. Additionally, studies often included the research of a specific part of a pneumatically powered device, which would be possibly used for prostheses. The articles included in this overview consisted of prosthesis that were partly and fully complete. Some articles were published to prove a concept, meaning that they showcased a specific feature of the prosthesis. Others published their article to show the development that their prosthesis had made. This means that the versions presented in this article can be an iteration of the final product. Furthermore, the focus of the articles on a specific feature, caused them to neglect other features and requirements since the prosthesis were not yet meant for commercial use. Therefore, the comparison of the specifications of these prostheses is slightly inconsistent. Moreover, there were no clear correlations found between the different specifications. Additional inconsistencies were present, due to the variation in the extent to which an amputee had a remaining limb and also due to the variation between actuation methods. On the contrary, the working principles were presented, providing information about the different features of the prostheses. The discussion of both specifications and working principle gave insight to important requirements and features of a modern pneumatically powered prosthesis for upper extremities.

Generally, the three requirements comfort, control and cosmetics should be met. However, the prosthetic devices used in this overview commonly compromised one feature for another. The most influential requirement for improving control amongst the presented articles is adaptive grip. Adaptive grip can be achieved using each of the actuation method, when an innovative coupling mechanism is used or when PneuNets are used as finger actuators. This feature allows a single actuator to operate multiple DOF, whilst reducing control complexity. Moreover, it makes for a larger variation of grasps with better grip due to a larger contact area. The number of grasps is, however, dependent on the movements of joints, especially in the thumb. With respect to comfort, weight is considered an important specification. The weight of the prosthesis is applied at the skin of the residual limb instead of the skeleton. A well designed socket can greatly improve comfort, but more importantly still is the weight and its distribution. The weight of the prosthesis should be lower than the average weight of a human hand, where heavy components should either be worn externally or placed as close to the residual limb as possible. Contrarily, an important feature that greatly improves control, but is missing in all prosthesis but one, the prosthesis from Fite et al. (2008), is feedback.

In general, a trade-off occurs between requirements due to the focus on specific features for either research purposes or for personal preferences. The requirements of a prosthesis will differ depending on the needs of its user. Despite the personalized requirements, standards should be introduced. These standards should apply for a specific set of requirements, and also for the testing thereof. This should allow researcher to evaluate and compare their work in a standardized manner, thereby decreasing the chance of abandonment of their prosthesis when eventually incorporated in practice.

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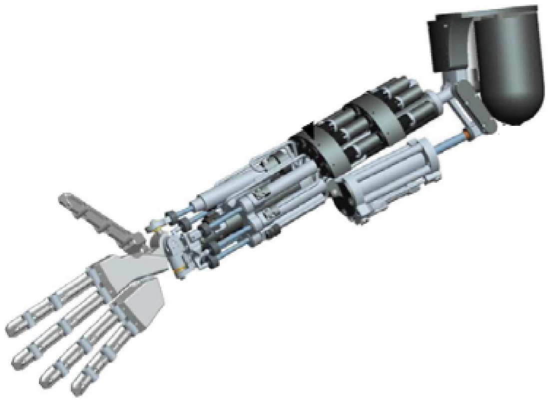
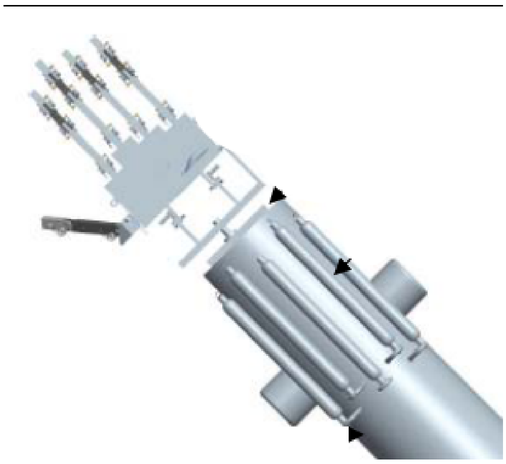
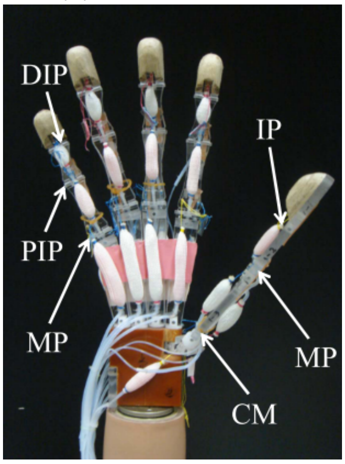
Appendix



(a) (Fite et al., 2008)

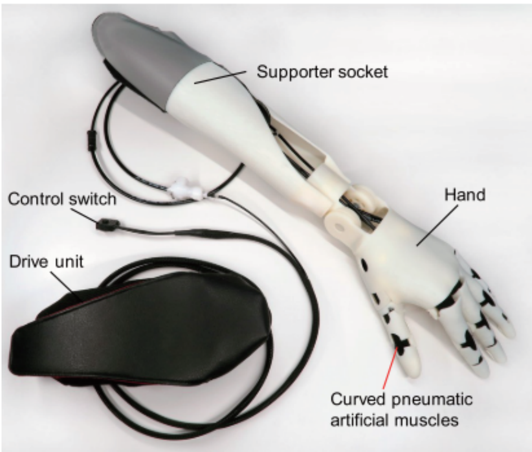


(b) (Polhemus et al., 2013)

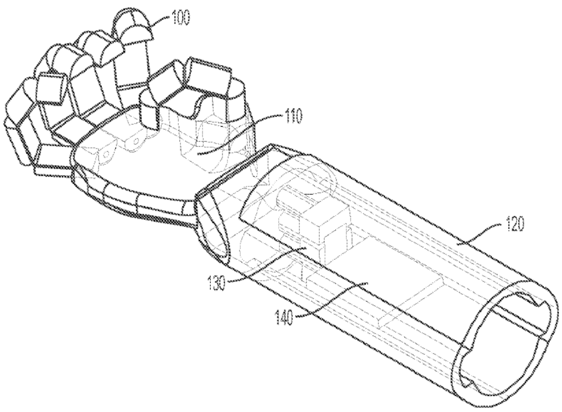


(c) (Takeda et al., 2009)

(d) (Schulz et al., 2001)

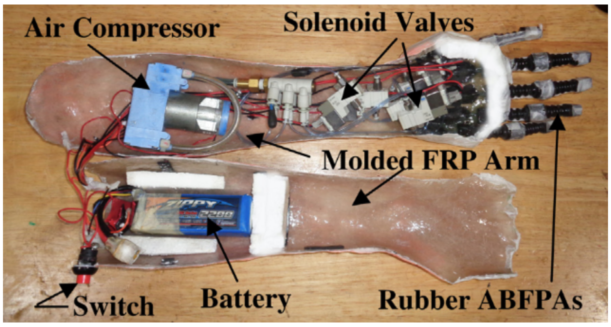


(e) (Nemoto et al., 2018)

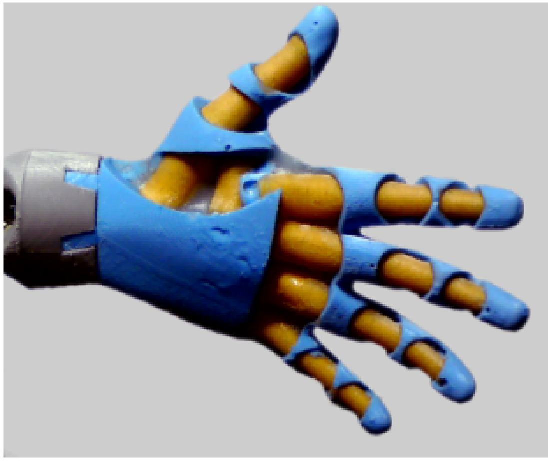


(f) (Whitesides et al., 2017)

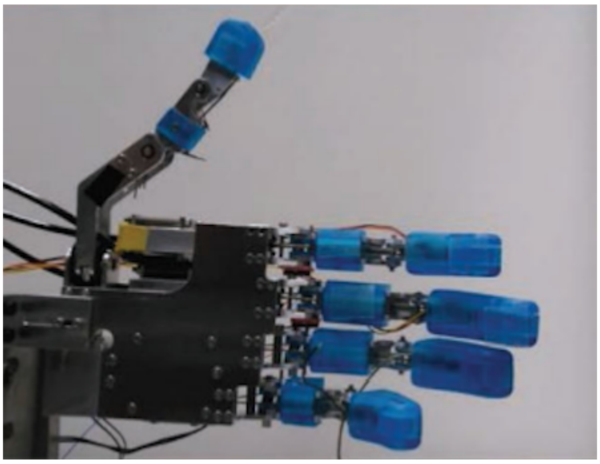
Figure 2-12: This figure shows an overview of the design of the prosthetic devices. The order is identical to table 2-1.



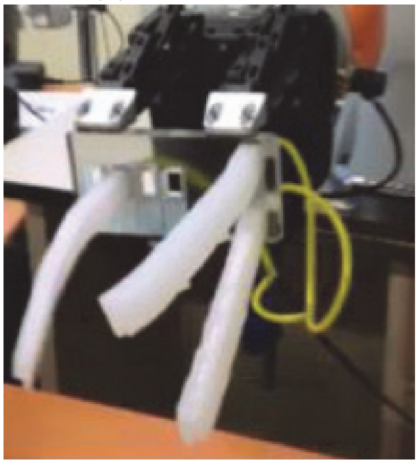
(a) (Devi et al., 2018)



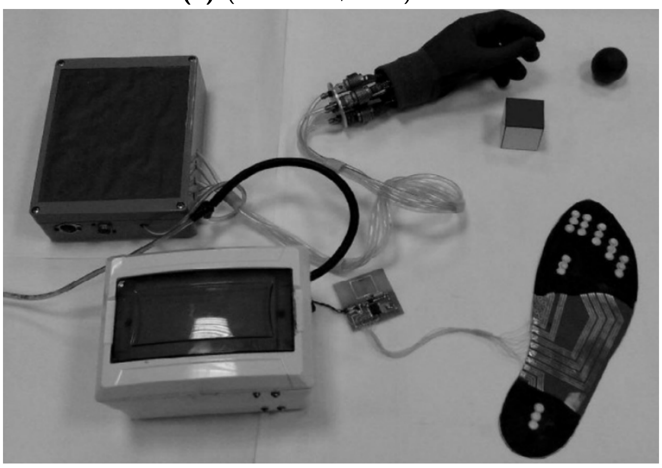
(b) (Fras et al., 2018)



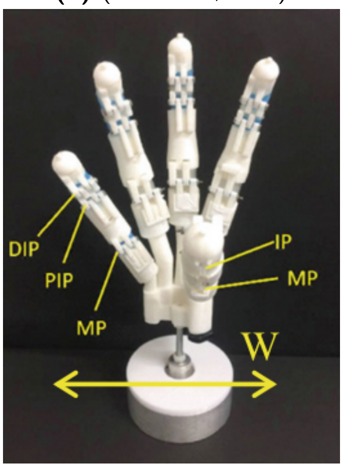
(c) (Kim et al., 2018)



(d) (Low et al., 2015)



(e) (Vorob'ev et al., 2017)



(f) (Nishikawa et al., 2016)

Figure 2-13: This figure shows an overview of the design of the prosthetic devices. The order is identical to table 2-1 and continues where figure 2-12 stopped.

Appendix A

Concept Synthesis

A-1 Design Criteria

Prior to designing a concept, the different requirements for a pneumatically powered upper extremity prosthesis are identified and specified. This section describes the purpose/meaning of the individual requirements and also quantifies them. A summary of all the requirements is given in A-1.

Table A-1: Summary of the design criteria.

Criterium	Description and/or Quantification
Size	See figure A-2
Weight	Maximum of 106 g
Force	Minimum of 46.1N
Adaptive grip	Yes
Degrees of Freedom	15
Range of Motion	MCP $< 90°$, PIP $> 90°$, DIP $< 90°$

A-1-1 Size

The size of the prosthesis must equal the size of the average 4 year old child. The required data was obtained from Molenbroek (1993). Figure A-1 shows the dimensions of the average 4 year old child. In addition, table A-2 summarizes these measurements including their standard deviations using the right terminology.

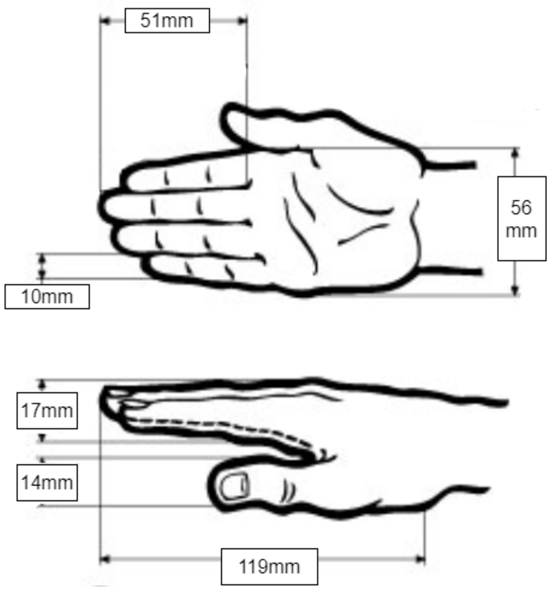


Figure A-1: Hand dimensions of an average 4 year old child. (Molenbroek, 1993)

Table A-2: Values of the hand dimensions of an average 4 year old child, including standard deviations.

Measures	**mean**	**sd**
Middle finger length (mm)	51	3
Hand width without thumb (mm)	56	3
Pink breadth (mm)	10	1
Hand thickness (mm)	17	2
Thumb breadth (mm)	14	1
Hand length (mm)	119	7

A-1-2 Weight

The weight of the prosthesis should be as low as possible, but no higher than 106g. This value is obtained by taking the mean weight of 4 year old girls, who are on average 0.4kg lighter than 4 year old boys (Scholtens et al., 2007). The mean weight of 4 year old girls is 17.3kg (-+2.4kg) (Scholtens et al., 2007). The accompanying mean length of these children was similar to the mean length of children reported by Molenbroek (1993). Despite, different sources the mean weight is applicable for both groups. An adult human hand of males makes up 0.6 percent of the body weight (Tözeren, 1999). For females this percentage is 0.5 percent (Tözeren, 1999). Linear scaling was applied despite the potential differences between adults and children. The weight of the hand of a boy should average at 106.2g. The weight of the hand of a girl should average at 86.5g.

A-1-3 Force

The amount of force that the hand should deliver upon a certain grasp is difficult to determine. Metrics like grasp force and finger tip force of a prosthesis are difficult to compare to that measured in a human hand. This is mainly due to the different force distribution seen in a human hand and prosthesis. Furthermore, prosthesis are often made of non-yielding material, which result a relative small contact area between the hand and the object. Kargov et al. (2004) showed that the number of contact points and amount of contact area decreased in the following order: human hand, adaptive prosthetic hand, non-adaptive prosthetic hand. In addition, Kargov et al. (2004) showed that, with similar joint torques, the adaptive prosthetic hand exerted low contact forces during grasping, due to the distribution of forces amongst many and wide contact areas. Contrarily, the non-adaptive prostheses exerted high grip forces, which were concentrated on less and smaller contact areas. The human hand used the most contact points with the widest contact area, whilst exerting the lowest average force. The average contact forces were found to be 6.5 times larger in non-adaptive prosthesis when compared with human hands. It can be assumed that the human hand is most capable of firmly and comfortably grasping objects of different shapes, sizes and masses. The forces that a prosthesis should deliver are therefore not representative of the performance of a grasp. In addition, the grasp force exerted by the prosthesis is also a function of the transmission mechanism and the friction between the contact points of the prosthesis and object. Nonetheless, an indication of the force that the prosthesis should deliver must be known to develop a fitting design, that is accompanied by a transmission mechanism and actuator force. In literature two different force measurements of hand prosthetics are used alternately, namely finger tip force and (cylindrical) grasp force. However, the force exerted by 4 year old children were only reliably presented in articles with a large number of participants, who used a cylindrical grasp. Table A-3 presents the results of a grasp force study from Ploegmakers et al. (2013). These values led to the following minimal force requirement of 46.1N, which is the force exerted by the non-dominant hand of a 4 year old girl.

Age (yr)			**Boys**		
	n	Dominant(N)	Non-dominant(N)	Height(cm)	Weight(kg)
4	124	55.9	52.0	111	19
			Girls		
	n	Dominant(N)	Non-dominant(N)	Height(cm)	Weight(kg)
4	109	50.0	46.1	111	19

Table A-3: This table presents the grip forces exerted by 4 year old boys and girls. The forces of both dominant and non-dominant hands are depicted. Besides that, the number of participants, height and weight are shown.

A-1-4 Adaptive grip

As mentioned in the subsection above the performance of a prosthesis highly depends on the number of contact points between the prosthesis and the object. An adaptive grip can increase the number of contact points, thereby increasing the hand's ability to perform varying grasping tasks. This mechanism allows for multiple DOF to be operated using a single actuators. Therefore, The transmission mechanism of the prosthesis must use the principle of adaptive grip.

A-1-5 Degrees of Freedom

The amount of DOF should equal the number of DOF that a normal human hand possesses. The prosthetic hand should have 15 active DOF. Each finger, except the thumb, should have a MCP, PIP and DIP joint. The thumb should have a MCP and IP joint.

A-1-6 Range of motion

The range of motion of the prosthesis should model the range of motion of a human hand as closely as possible. A similar range of motion accommodates a wide variety of grasps and could improve the adaptation of certain grasps. Independent measurements were performed by Kapandji (1971), Stillfried et al. (2014), who both concluded the same results concerning the range of motion of MCP, DIP and PIP flexion. These results are the following:

- MCP $< 90°$
- PIP $> 90°$

- DIP $< 90°$

Contrarily, the following observations were purposefully discarded, since their contributions to the functioning is expected to be minimal, but the effort required to fulfill them as additional requirements would be time consuming. Kapandji (1971), Stillfried et al. (2014) found that the range of motion of the MCP joint progressively increased with from the index finger towards the little finger. Furthermore, results were presented that showed the side to side motion of the fingers in the MCP joint. Lastly, no results were presented, which concluded the joint angles when the hand was relaxed. To conclude, only the values from the bullet points above are used as design criteria.

A-2 Wishes

In addition to the 'must do' requirements of the prosthesis, this chapter describes the 'wish' requirements.

Safety

The device must not cause harm to the user or others.

Pressure

Plettenburg (2002) determined the pressure level for minimal gas consumption, which is 1.2 MPa. The same value is used as the pressure requirement for this prosthesis, since the size of the pneumatic cylinder is comparable.

Operation cycles

The number of operation cycles that the device could perform should be as high as possible. The first step in this process is choosing the right pressure for minimal gas consumption as described above. The remainder consists of choosing a vessel that is able to supply as many operation as possible whilst having a small embodiment.

Socket

The prosthesis should include a socket, which is enables users with different residual limb lengths to use the prosthesis. Moreover, the socket should be made as comfortable and light as possible. Lastly, the socket should use up as little space as possible in order to leave space for components of the hand.

Placing of components

In order to minimise the torque applied at the residual limb, heavier components should be placed proximal and lighter components distal.

Complexity

Should be easy to do maintenance and also easy to use.

Environment

The device should be able to sustain environmental factors such as temperature, sand and possibly water.

Aesthetic

The device should be aesthetically pleasing.

Durability

The device should be able to withstand outside forces.

Reliability

The device should be reliable, and therefore not malfunction.

A-3 Outside research scope

Control and feedback

Due to the adaptive grip mechanism, the prosthesis will require only a single input signal. The obtainment of this signal does not lie within the scope of this research. In addition, the device is not designed with the providence of feedback. However, in the recommendation section some remarks are made considering the control and feedback of the prosthesis.

A-4 Concept creation

This section describes the choices and complications that came with the creation of a concept using the design criteria. Firstly, additional literature will be discussed, which describes different types of pneumatic actuators and transmission mechanisms. Secondly, the concept creation will be discussed and the working principle will be explained. Finally, the quantified design criteria will be checked, to see if the concept theoretically meets these criteria.

A-4-1 Pneumatic actuator

The literature study did not suggest a particular pneumatic actuator. However, as stated in the design criteria, both adaptive grip and a relatively high force must coexist in the desired prosthesis. The prostheses from literature that used adaptive grip where commonly actuated by PneuNets. A careful inspection of the presented grasp examples seen in prosthesis actuated by PneuNets, showed that the objects grasped often rested on the area between the thumb and index finger, since the hand was used in an upright position. The total load of the object was therefore not completely carried by a certain grasp type. It is therefore invalid to conclude that the force or grasp were sufficient to hold certain objects. Despite tremendous adaptive grip capabilities, the average force output was much lower than recommended in literature. Contrarily, high force outputs were measured in the prosthesis that used pneumatic cylinders and PAMs. Although both actuators can have high resulting force outputs, PAMs are not suitable for the desired prosthesis, because the force output depends on the stroke length. A combination of the instantaneous force delivered by pneumatic actuators with a transmission mechanism that allows for adaptive grip make for a promising prosthesis.

A-4-2 Transmission mechanisms

A suitable transmission mechanism should be designed to accommodate the design criteria, adaptive grip. In order to achieve adaptive grip, the prosthesis should be actuated using fewer actuators than degrees of freedom, i.e. underactuation. A small additional literature research was performed to study the transmission mechanisms employed in underactuated robotic and prosthetic hands.

In general, two types of transmission mechanism are distinguished, being linkage-based and tendon-pulley mechanisms (Carrozza et al., 2006). Linkage mechanisms are used in many different prosthetic and robotic hand designs (Mu et al., 2007; Wu et al., 2009; Yang et al., 2009; Zhao et al., 2010; Sheng et al., 2014; Omarkulov et al., 2015; Lee et al., 2016; Herath et al., 2017; X. Liu et al., 2017; Yoon et al., 2017; Zhang et al., 2018). Additionally, the same applies for tendon-pulley mechanisms (Carrozza et al., 2004; Carrozza et al., 2006; Inouye et al., 2012; Ozawa et al., 2014; Sun et al., 2014; Gao et al., 2015; Chen et al., 2016; Deng et al., 2016; Wang et al., 2017).

Alternatively, articles opted to differentiate the two different mechanisms (Birglen et al., 2006; Y. Liu et al., 2009; Smit et al., 2013; Carbone et al., 2015; Andrés et al., 2019). Tendon-pulley mechanisms can lead to friction and elasticity (Ceccarelli, 2004). Smit et al. (2013) showed that a linkage mechanism is most suitable for prosthetic hands, due to its higher energy efficiency compared to tendon-pulley mechanisms. In addition, Carbone et al. (2015) reported that the tendon-pulley design led to very high friction losses and very low mechanical efficiency. Furthermore, they reported that the use of tendons limits the maximum amount of input force. Subsequently, this leads to lower achievable grasping forces.

Contrarily, Carbone et al. (2015) showed that linkage-based mechanisms result in very limited friction losses and good mechanical efficiency. Furthermore, they reported that linkage-based mechanisms allow for higher input torques and higher achievable grasping forces.

Despite the efforts and successes of some of the above mentioned researcher to develop a transmission mechanism for a hand prosthesis, none of them fit the criteria set for this thesis.

The adaptive grip criteria can be extended and explained by means of two stages. The first stage is comprised of the pre-shaping phase, which is explained as the movement of the fingers as a whole, where the trajectory is set. The second stage is comprised of the adaptive phase, which is only active when an object makes contact with a phalanx or when a motion boundary is reached. A few articles did not present a prosthetic or robotic hand capable of adaptive grip (Mu et al., 2007; Omarkulov et al., 2015; Herath et al., 2017; X. Liu et al., 2017). Moreover, there were articles that reported only partly adaptive capabilities in their hands, more specifically between the middle and distal phalanges (Wu et al., 2009; Yang et al., 2009). Furthermore, Sheng et al. (2014) created a compliant finger mechanism, which had adaptive grip capabilities, but was only capable of moving the middle and distal phalanges with a fixed proximal phalanx. Besides the adaptive grip criteria, some designs were discarded due to other reasons. Generally, the presented finger mechanisms were sized for adults, which is difficult to replicate in a toddler sized finger mechanism. In addition, most articles used cross four-bar linkages, which create oddly shaped and angled planes at which the phalanges make contact with objects (Y. Liu et al., 2009; Wu et al., 2009; Yang et al., 2009; Zhao et al., 2010; Sheng et al., 2014; Omarkulov et al., 2015; Zhang et al., 2018). This creates the need for a cover to be used around these links, which requires additional material and space to be used, causing them to be bulkier. Lastly, Zhang et al. (2018) presented a finger mechanism for a prosthetic hand, but used compression springs which absorbs a part of the input force, thereby lowering the fingertip and grip force of the prosthesis. In conclusion, a linkage-based mechanism is desired, which fits in a toddler sized prosthesis and is operable by a pneumatic drive source. Furthermore, cross four-bar linkage should be avoided as well as compression springs that decrease the force output of the prosthesis.

A-4-3 Size

The dimensions of the hand are set in the design criteria section. However, the sectioning of the total finger length into three phalanges was not specified. The total finger length was determined to be 51mm. Buryanov et al. (2010) reported the proportions of hand segments using x-ray images. These proportions were applied to the total length of 51mm and resulted in the following lengths of the phalanges:

- Proximal phalanx - 25mm
- Middle phalanx - 15mm
- Distal phalanx - 12mm

A-4-4 Working principle

The mechanism is depicted in figure A-2. A pneumatic cylinder actuator is connected at B and drives the point up when actuated. The green pin that intersects with the (blue) x-axis is fixed. The mechanism is made up of 9 links and two tension springs. Link 1,2,3 and 4 make up the proximal phalanx. Link 3,5,6 and 7 make up the middle phalanx. Lastly, 6, 8 and 9 make up the distal phalanx. Furthermore, the joint between link 1 and 2 will be referred to as the MCP joint (A). Subsequently, the PIP (C) and DIP (E) are between the links 2 and 5, and 5 and 8 respectively. The MCP, PIP and DIP are all limited in various ways to ensure that their final angles match the desired range of motion. The MCP stops at approximately 9 degrees from the x-axis, where the PIP and DIP stop when their connecting links become perpendicular. The mechanics of these stops are shown later in this section. When the finger mechanism moves freely, as there is no object intersecting its path, the path is as depicted in figure A-3. Alternatively, there are different motions, which are caused by varying contact points between an object and the finger mechanism. These motions are shown in figure A-4 and A-5.

The paths can be split up into two motion stages, namely the pre-shaping and adaptive phases. The first phase is comprised of the whole finger moving as a rigid body around the MPC joint. The actuator force in B causes a torque around the MCP joint, whilst both springs prevent the phalanges to change shape, i.e. keeping the same configuration. The second phase is initiated when the MCP joint reaches its maximum angle or when the finger is in contact with an object. Depending on the location of contact a following motion is initiated. One option is that the object makes initial contact with the proximal phalanx (figure A-4), the other option is that the object makes initial contact with the middle phalanx (figure A-5).

When link 2, the proximal phalanx, is obstructed by an object, the actuation force in B will no longer cause motion in the MCP joint, but will cause link 4 to move relatively to link 2. This motion will occur simultaneously with the elongation of spring 1, whilst ultimately moving the middle phalanx and the distal phalanx. When the middle phalanx is free to move, the former movement of link 4, will push the whole middle phalanx around its DIP joint, since link 5 is not blocked and the spring is not elongated. Alternatively, link 5 could be hindered by an object or due to its maximal range of motion, which will trigger a similar motion as described before. This motion includes the elongation of spring 2, whilst link 7 is moved relatively to link 5. As a result, the distal phalanx will then rotate around its DIP joint. Lastly, the order can be reversed, where an object blocks link 5 first. The whole motion is complete when either an object is fully grasped, or when all maximum angles of the MCP, PIP and DIP are reached.

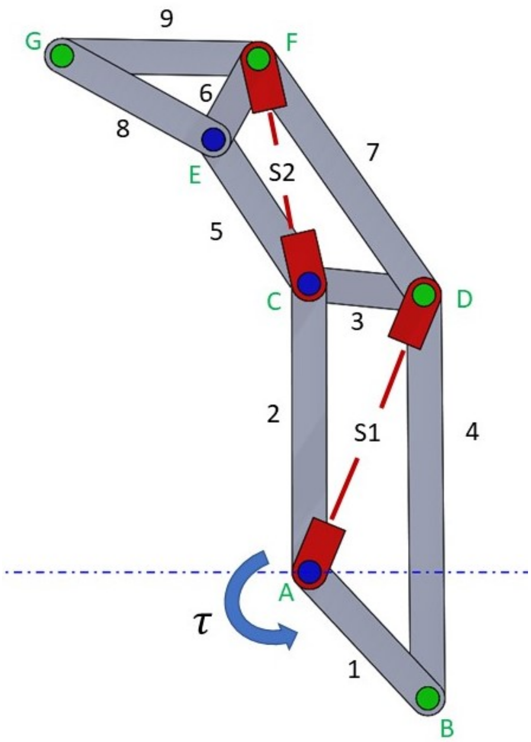


Figure A-2: The transmission mechanism for a finger of a toddler sized prosthesis. This schematic indicates the different links (numbers), springs (S1 and S2) and pins (letters).

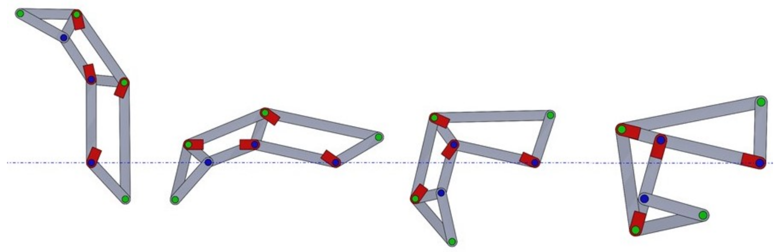


Figure A-3: Multiple configurations of the transmission mechanism for a finger of a toddler sized prosthesis when moving freely.

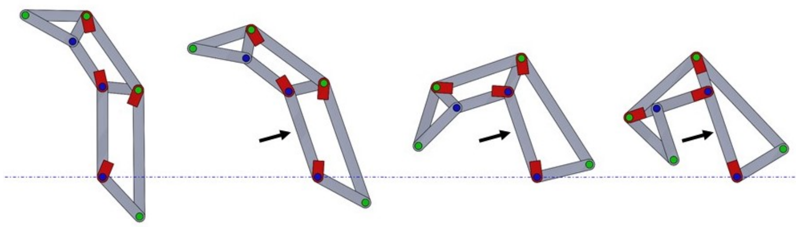


Figure A-4: Multiple configurations of the transmission mechanism for a finger of a toddler sized prosthesis when encountering an object at the proximal phalanx.

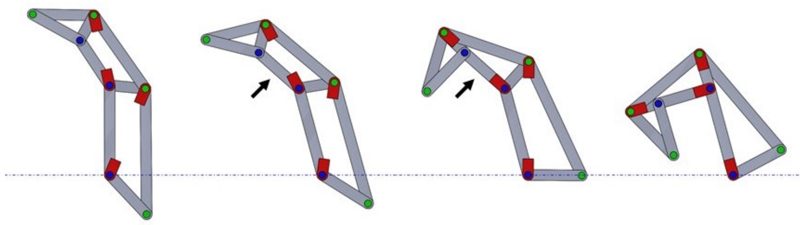


Figure A-5: Multiple configurations of the transmission mechanism for a finger of a toddler sized prosthesis when encountering an object at the middle phalanx.

Numerous mechanical stops are carefully designed to enable the finger mechanism to work as described above. To ensure that the finger mechanism is at rest when there is no actuation force, as depicted in A-2, two stops are applied between links 1 and 4 (B), and links 3 and 7 (D). These stops are necessary, since the springs naturally want to be in their rest/shortest state, thereby creating a small as an angle possible at both B and D. Similarly, when the finger is put in a certain position within the gravitational field, both C and D will be free to move as long as it does not elongate the springs. Further demonstration of this problem and its solution is shown in figure A-6.

In addition, mechanical stops were applied at joints A, C and E to ensure that the finger mechanism started and ended at the right angles, hence providing the desired range of motion. The stops at C and E were achieved using an adjustment to a single link, which working principle is shown in figure A-7. These stops also prevent a configuration where the springs pull the system further into a flexed state, which is then irreversible, meaning that the fully extended configuration can no longer be reached.

Besides that, the stop at A, the MCP joint, is caused by the body that houses the finger mechanisms and pneumatic cylinders. A small extension in this frame restricts the movement of the MCP joint, which causes link 2 to stop moving as if it where touching an object, thereby shifting to the grasping stage.

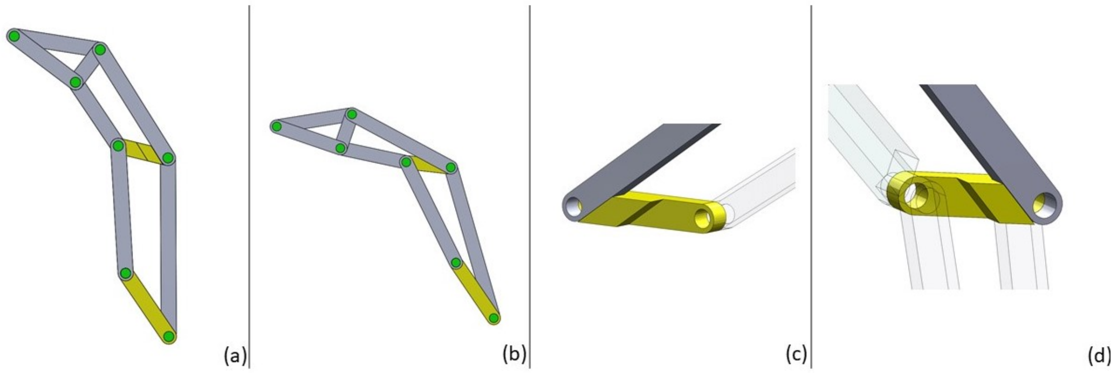


Figure A-6: This figure depicts the problem and solution for the neutral position. A shows the neutral position. B shows the effect of the springs, when there are no mechanical stops. C and D show the exact placement and principle of the mechanical stops, which concern links 1 and 3.

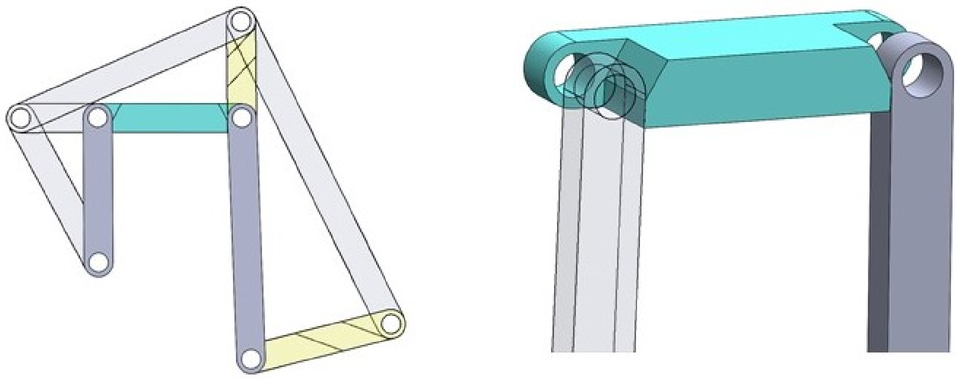


Figure A-7: The mechanical stop at the green link to prevent smaller angles than 90 degrees between the links 2 and 5 and links 5 and 8.

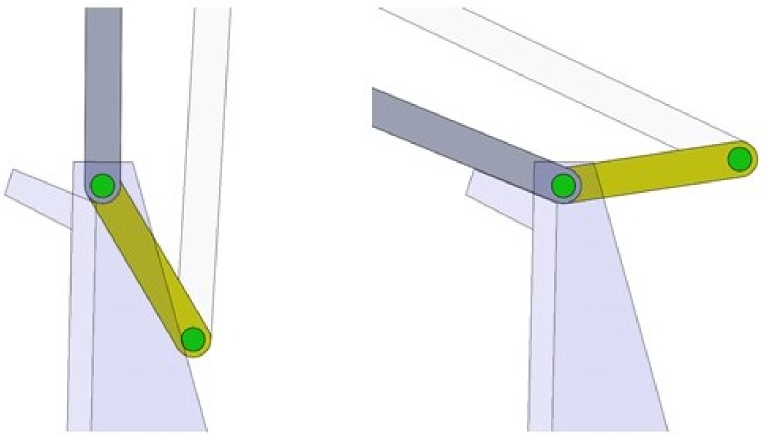


Figure A-8: The mechanical stop caused by the frame of the prosthesis. This stop ensures that the rotation of link 2 ends.

A-4-5 Pneumatic drive source

Now that the finger mechanism is established, the remaining space that is available for the pneumatic cylinder can be determined. Since point A lies underneath the MCP joint from where the original finger length was measured, the available length for the cylinder in the palm of the hand is reduced. This reduction leaves a length 56 mm to be used by the cylinder. Fortunately, a stroke of 28 mm is required to complete a full grasp. Furthermore, connection point A is located to the right of the centre of the mechanism. This means that the available space for the cylinder is slightly reduced. Despite this reduction, the available width per cylinder is still smaller than the hand thickness, thereby dominating the available space. The average width available for each cylinder is estimated using A-3, which resulted in 13 mm. This assumption is based on a 2 mm thick frame that would be needed on both sides of the hand. Plettenburg (2005) designed the WILMER 21-t025, which is a piston in cylinder actuator. This design is characterised by its low mass and thin cylinder wall, being only 1 mm larger in diameter than the bore. Utilizing such a design would result in a bore of 12 mm. As mentioned before, the optimal pressure for minimal gas consumption was determined by Plettenburg (2002) at 1.2 MPa. These values are used in the following formula to obtain the force:

$$F = P \cdot A$$

where, F is the force in N, P the pressure in Pa and A the area in m^2. The force was determined to be 136 N.

A-4-6 Static model

A static model was used to determine the required yield strength for the links and the initial tension required in the springs. This type of model uses different configurations of the system, where it assumes equilibrium in all parts. Therefore, accelerations and moments of inertia are not taken into account. For this transmission mechanism the stroke length, joint angles and object placement were varied to obtain the resulting forces in different configurations. The highest tensile force that could occur in an equilibrium configuration is 186 N. The required material could be determined using this tensile force and the cross sectional area of the link, such as in the following formula:

$$\sigma = \frac{F}{A}$$

In addition, to the required yield strength, the material must have a low density. The material found to be most suitable is an aluminium alloy, AL 7075 T6, which has the following material properties:

$$\text{Density} = 2,77 * 10^3 - 2,83 * 10^3 \ kg/m^3$$
$$\text{Yield strength} = 435 - 520 \text{ MPa}$$

This material allows for very small links and pins to be used, whilst maintaining a low weight and sustain high forces. The initial tension in the spring forces was also determined, but due to the use of a static model instead of a dynamic model, and the uncertainties of friction forces as well as the large spread of the stiffness of springs, this value was only used as an initial guess. Similarly, the fingertip force was estimated, but will be much lower due to the friction as well. The fingertip force using the static model was approximately 40N.

A-4-7 Pneumatically powered hand prosthesis

Figure A-9 shows the design of the pneumatically powered hand prosthesis. Apart from the isometric view, figure A-10 shows the rear, top and side views. The finger transmission mechanism is combined with the pneumatic actuator to form a whole finger. Aligning these fingers within a frame creates part of a hand prosthesis. In order to form a full functioning hand, a thumb and palm of the hand have to be implemented as well. In this configuration the hand has 14 DOF, where the thumb lacks 1 DOF in its CMC joint. A pressure regulator, valve and power supply need to be placed inside the socket of the prosthesis.

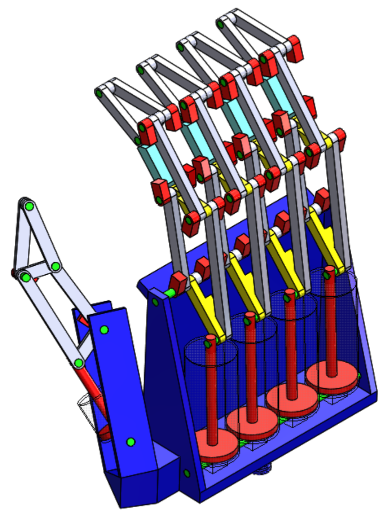


Figure A-9: Isometric view of the concept of the prosthetic hand

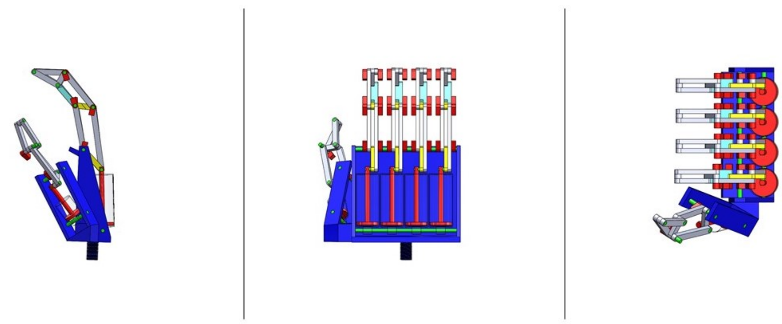


Figure A-10: This figure shows the prosthetic hand in a few configuration, which are from left to right: side, rear and top view.

A-4-8 Challenges of the design

In order to simultaneously reach the end configuration whilst also optimizing both cylinder position and the initial configuration, a lot of key aspects need to be considered. Firstly, the length of the proximal, middle and distal links were final, whilst the remaining links had to be adjusted to obtain proper starting and ending angles in the joints. The length of these residual links were limited to the outer dimensions of a finger of a four year old child. In addition, the required material dictated the width and thickness of the links, thereby potentially increasing or reducing its range of motion. The difficulties of the design were mostly caused by the pneumatic cylinder actuator. A transmission mechanism between the finger mechanism and cylinder was required to translate the force of the cylinder onto the finger mechanism. The torque applied to the finger mechanism was maximized by connecting the cylinder rod to the end of link 1. Instead of using a slider-crank mechanism, the cylinder also pivots underneath its body. The cylinder and finger mechanism had to be aligned along the same line, due to the restriction in dimension. This resulted in a cylinder stroke, which was mostly directed upwards, whilst the driving link of the finger mechanism, link 1, was rotating anticlockwise. The end position of the finger mechanism was dominated by this design. Pushing link 1 further anticlockwise was not possible due to the limit of the cylinder stroke. When surpassing this limit, the spring return of the cylinder would no longer pull the mechanism back into its initial configuration, but would instead push the mechanism into an irreversible flexed state. In addition, the vertical position of link 1 needed to be reached by the cylinder stroke. The challenge of this principle is that the cylinder rod was likely to intersect with the MCP joint, when it was placed right underneath the finger mechanism. The cylinder was therefore placed slightly to the right of the finger mechanism, whilst still starting in an upright position. The upright position was necessary to ensure efficient force transmission and distributed torque along the stroke path. The optimal placement of the cylinder was therefore determined by the maximum finger dimensions, cylinder width, stroke length and valve stem diameter. The lengths of links 1 and 4 were adapted to optimise these variables. Combining these requirements resulted in a small reduction of the range of motion of the MCP joint, namely it would theoretically no longer reach 90 degrees, but 82 degrees instead.

The implementation of the springs also came with its challenges and trade-offs. Initial spring lengths are related to the initial angles at which the mechanism starts. As mentioned in the working principle section, the springs should not be allowed to cause an irreversible state of the mechanism. This was prevented using the mechanical stops.

Lastly, some design considerations were made concerning the frame of the hand. In this preliminary design the palm of the hand was made rigid, such that a multitude of objects could be grasped. However, if weight reduction is required then another material and/or pattern could be used to optimise the contact area its weight.

A-4-9 Design criteria vs Concept

In this subsection the design criteria from table A-1 are compared with the theoretical capabilities of the created concept.

The preliminary design made of AL-7075 T6 is able to stay within the size as is quantified in figure A-1. The weight of the hand should also remain within its limit of 106 grams. As mentioned before, the cylinder design to be used for this prosthesis is presented in Plettenburg (2005). This piston in cylinder actuator has 1.5x larger bore than the actuator designed for this prosthesis. Nonetheless, the formula given to determine the weight using a certain stroke length with the same 1.2 MPa pressure is used to estimate the actuators weight. The weight of the Wilmer design actuator used for this prosthesis should result in approximately 6.8 grams. Furthermore, the finger mechanism was calculated using its CAD in SolidWorks, resulting in approximately 3.5 grams. A similar approach for the hand frame was taken and was approximated at 42 grams. Adding these values together would result in a hand with a total weight of 93.5 grams. This value is, however, exaggerated since all the parts together are probably lighter. The finger used to calculate this weight was the middle finger, which is also the largest finger of the five. Additionally, the actuator that is to be used in this prosthesis is smaller than the Wilmer design, but its weight is assumed to be the same as the Wilmer prosthesis. Lastly, the frame of the hand is very bulky and is not optimized for contact area using as little material as possible. Hence, the total weight of the hand could be reduced by quite a bit. Despite the possible reduction, the hand shows that it is already capable of staying within the weight limit quantified in table A-1.

The force criteria is difficult to predict. This depends on a large variety of factors, that will only become evident upon testing the mechanism. The actuator force that can be generated is sufficient to create such forces, but the large variety of factors could drastically reduce the grasp force provided by each individual finger.

In theory, the adaptive grip criteria should be met. This criteria can however only be validated through testing. Alternatively, the number of degrees of freedom is not met, since the thumb lacks one degree of freedom. The range of motion of the is only partly met, since the MCP joint falls about 8 degrees short of the criteria.

Appendix B

Manufacturing and Assembling

B-1 Design Embodiment

Prior to creating the presented prosthesis as a whole, the finger mechanism is build as an initial prototype. In practice, there are many different factors that could influence the theoretical concept. Building a prototype will show how well it is capable of being manufactured and assembled. More importantly, the prototype of the finger mechanism will serve as a proof-of-principle, where its working principle is tested along with its capabilities of fulfilling the criteria created for the full hand prosthesis.

In this section, the manufacturing and assembling of a proof-of-principle finger transmission mechanism are explained. The theoretical concept was designed using Al-7075 T6 as the main material. The embodiment of such a concept would have required much time and high cost to manufacture, whilst also allowing for less iterations to be made. At this stage more readily available off-the-shelf products are preferred along with a faster manufacturing speed. A way more time efficient and easy way to manufacture the finger concept, that allows for rapid prototyping, is a 3D printing technique known as fused deposition modeling (FDM). A 3D printed finger transmission mechanism could serve as a proof-of-concept, where the working principle and capabilities of fulfilling the criteria are tested. The material used in the available 3D printers was polylactic acid (PLA). Therefore, the creation of the prototype was reconsidered using PLA as the main material. Furthermore, a pneumatic cylinder actuator as exactly required was also not readily available and production of such a low mass construction cylinder is difficult and time consuming. Consequently, a substitute was required. However, industrial pneumatic cylinders are often much bulkier than the low mas construction WILMER 21-t025 created by Plettenburg (2005). A different frame was required to enable this prototype to move as intended. The remainder of this section will discuss and explain the choices made during this remodelling of the finger transmission mechanism using PLA and an industrial pneumatic cylinder actuator.

B-1-1 Dimensions

The use of a different material requires redimensioning of the finger transmission mechanism. In order to sustain the tensile forces generated inside the links, the links were adjusted by making them wider and thicker. The order of magnitude of this enlargement was determined using the results from Pandzic et al. (2019). Moreover, other reports of infill density and patterns were also included, but showed varying results (Khan et al., 2018; Seol et al., 2018). Pandzic et al. (2019) used a printer of the same company that will be used to print the parts for the prototype. Besides that, the study of Pandzic et al. (2019) performed an extensive research on the yield strength of 3D printed parts using different infill densities and patterns. They concluded that a concentric infill pattern would result in the highest yield strength. A higher infill density, results as expected in a higher yield strength. Due to the size of the parts required for this prototype an infill of 70 percent was chosen, since this already is a demanding task. This infill pattern and density should result in a part with yield strength of 23.9 MPa. Therefore, the width and thickness of the links were adjusted to 4mm both, to ensure a large enough cross sectional area. By doing so, the range of motion of the joints was reduced. In order to alleviate this reduction, some links were elongated. Lastly, the contacting link (5) of the middle phalanx was created with extra large mechanical stops, since the level of detail required for the theoretical design requires higher quality manufacturing and assembly. A detailed description of the 3D printers properties and settings are given in section B-2 below.

B-1-2 Cylinder choice

There are a few aspects that determine the positioning of the cylinder relative to the finger transmission mechanism. Firstly, the dimensions of the cylinder determines the distance between the connection points of the cylinder to the finger mechanism and to the frame. Secondly, the valve stem diameter determines how closely the cylinder can be put directly underneath the finger mechanism. Lastly, the stroke length determines in what configuration the finger transmission mechanism is able to start and end. A cylinder that could provide the actuator strength, as conceptually designed, should be chosen. In addition, its stroke length should be sufficiently long to show the working principle of the finger transmission mechanism. The double acting piston in cylinder Festo DSNU-16-25-P-A was chosen for the prototype. Figure B-1 shows an illustration of the product and table B-1 shows the relevant specifications thereof. Falling slightly short of the desired stroke length the mechanisms start position was therefore slightly flexed already. The newly redefined finger transmission mechanism had to be combined with the Festo cylinder to create a working prototype.

Figure B-1: The product illustration of the DSNU-16-25-P-A from www.festo.com

Table B-1: Technical data from the data sheet of the DSNU-16-25-P-A Festo cylinder actuator.

Feature	Value
Stroke	25 mm
Piston diameter	16 mm
Piston rod thread	M6
Piston-rod end	Male thread
Mode of operation	Double-acting
Theoretical force at 6 bar, return stroke	103,7 N
Theoretical force at 6 bar, advance stroke	120,6 N

B-1-3 Frame

A frame had to be designed to combine the finger transmission mechanism and cylinder into a working prototype. This frame should mainly provide two fixed pivot points, namely one at the bottom of the cylinder and one at the MCP joint. Furthermore, it is required that this frame is sturdy, such that it can withstand extensive testing and is easy to work with. The resulting frame is shown in figure B-2.

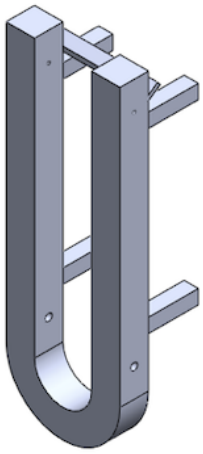


Figure B-2: This figure shows the CAD of the test frame that is used to fix the pivots of the MCP joint and cylinder bottom.

B-1-4 Springs

The calculations of the static model led to an initial estimate of the required initial tension in the springs. However, these calculations give no information concerning the spring stiffness. It is assumed that the spring stiffness should be as low as possible, whilst the initial tension should be the opposite. Such a combination is rare and difficult to obtain, since these springs are used in such a small environment. Fortunately, off-the-shelf springs were largely available. Thus, springs with the right stiffness and initial tension will be empirically found, whilst using the static model values as initial estimates.

B-1-5 Additional parts

Other parts such as pins and star-locks were used to complete the assembly. Stainless steel pins of 2 mm in diameter were used through the axis of the joints and pivots. Multiple star-lock closing mechanisms were used to tighten the parts together along the stainless steel pins. Furthermore, a connector piece was designed that allowed a connection between the cylinder valve stem and finger transmission mechanism. This part includes a female m6 thread, for the cylinder-rod thread. Moreover, it has a flat side, thereby creating extra space, which allows for the cylinder to be placed more underneath the finger transmission mechanism. This connector piece is made of stainless steel and its design shown in figure B-3.

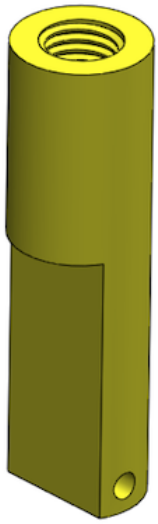


Figure B-3: The CAD of the connector piece is made of the pneumatic cylinder actuator and the finger transmission mechanism.

B-1-6 Design in full

The complete CAD version and physical prototype as well as the finger transmission mechanism prototype are shown in figure B-4, B-5 and B-6.

B-2 Printer properties and settings

The CAD assemblies were converted into a stl-files, which were interpreted by a software program Cura. Cura is the printer software that comes with the Ultimaker S3 printer. Cura allows the Ultimaker S3 to comprehend the stl-files created in SolidWorks. In addition, it allows the user to alter the settings of the printer and lets users choose how the parts should be aligned. The relevant settings that were used for this prototype are shown in table B-2.

Table B-2: Summary of the design criteria. These settings apply for the printing of the links. The frame was printed using a line height of 2mm and infill of 40 percent.

Category	Setting	Value
Quality	Layer Height	0.1 mm
	Line Width	0.4 mm
Infill	Infill Density	70%
	Infill Pattern	Triangles
Material	Main Extruder	PLA
	Print Core	AA 0.4
Support	Support Extruder	PVA
	Print Core	BB 0.4

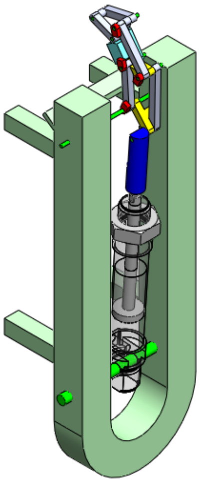


Figure B-4: The CAD model of the fully assembled prototype.

Figure B-5: An image of the fully assembled prototype.

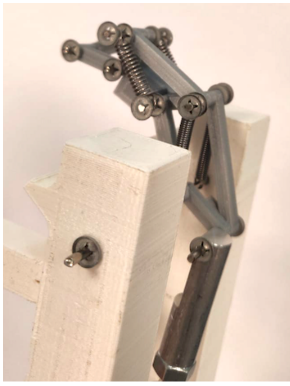


Figure B-6: An image of the fully assembled finger transmission mechanism.

B-3 Technical Drawings

The CAD model was converted into technical drawings using SolidWorks. The technical drawings of the parts of this prototype are included in Appendix E.

Appendix C

Prototype Performance

C-1 Method

The prototype, being a proof-of-principle finger mechanism, is tested to see if it meets or is able to meet the design criteria of the full prosthetic hand. The size and weight criteria are irrelevant to measure, due to the purpose of this prototype, being a proof-of-principle. Two criteria, degrees of freedom and range of motion, can be determined using prior knowledge of the design and visual inspection. Essentially, there are only two criteria that can be tested using this prototype, which are the force and adaptive grip.

Despite the force criteria being measured using a cylindrical grip on a dynamometer, the force for this prototype is measured at the fingertip using a pinching motion. The cylindrical grip could only be properly tested when a full hand would be assembled. The fingertip force will be measured using a load cell, figure C-2. The placement of this load cell relative to the fingertip was achieved by an additional frame, figure C-1. This frame fits over the legs of the main frame.

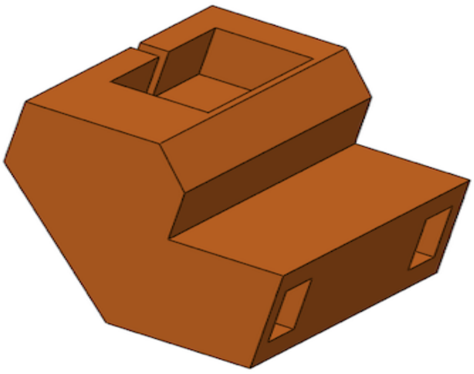


Figure C-1: The CAD image of the frame designed for positioning and holding the load cell.

Figure C-2: An image of the load cell.

The resulting set-up is shown in figure C-3. Additionally, the physical set-up including more detail is shown in figure C-4. This set-up will be used to test the fingertip force with an increasing supply pressure using two sets of springs. The specifications of these springs are given in tables C-1 and C-2.

Table C-1: Spring properties of set 1

Properties	Spring 1	Spring 2
	T330	T540
c (N/mm)	0.15	0.78
l (mm)	30.40	17.50
s (mm)	35.1	16.9
Fn (N)	9.5	15.4
Dm (mm)	5.45	6.8

Table C-2: Spring properties of set 2

Properties	Spring 1	Spring 2
	T290	TR390
c (N/mm)	0.9	1.44
l (mm)	26.40	15.90
s (mm)	14.7	9.04
Fn (N)	15.7	15.2
Dm (mm)	2.95	3.87

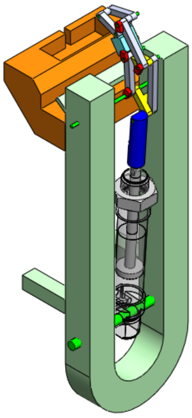


Figure C-3: The CAD image of the implementation of the frame designed for positioning and holding the load cell.

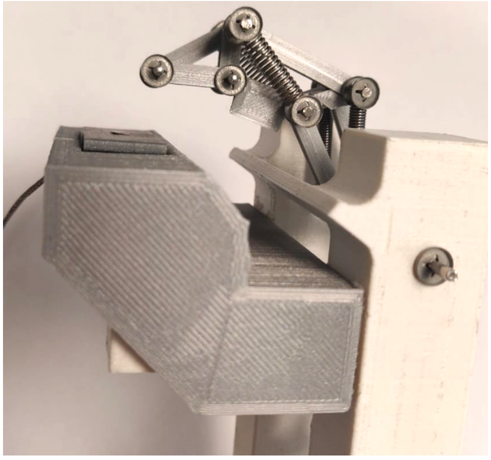


Figure C-4: The test setup showing the additional frame, load cell and finger transmission mechanism.

The fingertip force was measured using supply pressure increments of 0.5 bar. The finger should move in its fully extended state for it to firmly reach the load cell. The load cell was connected to a PLC, which was operated by LabView. The force was measured over time and exported to a text file. Subsequently, the values from the text file were copied to MATLAB. The data was processed in MATLAB, consequently plotting the results in one figure.

The construction of the test frame and mechanical stops of the prototype will show if the mechanism will adapt to these obstructions when the cylinder is pressurized. In addition, the middle phalanx will be hindered in another test, to show if the order of hindrance causes a different or similar adaptive effect. These motions will be captured using the rear main camera of a OnePlus 6T. This camera uses the Sony IMX 519 sensor and has 16 megapixels with a pixel size of 1,22 micrometer. The included aperture is f/1.7.

C-2 Results

The performance of the prototype was measured with the fingertip force and the capability of adaptive grip. The results of the former are presented in figure C-5, where the force is plotted as a function of the pressure. The test using spring set 1 was terminated after 3.5 bar. This was due the elongation of spring 1, consequently displacing the fingertip to the edge of the sensor, instead of remaining centered. Additionally, the test using spring set 2 was no longer able, since the high forces broke the connection of link 2 in joint C. The maximum force at the fingertip that was obtained using 5 bar was 32.3 N, where the actuator force was 100.5 N.

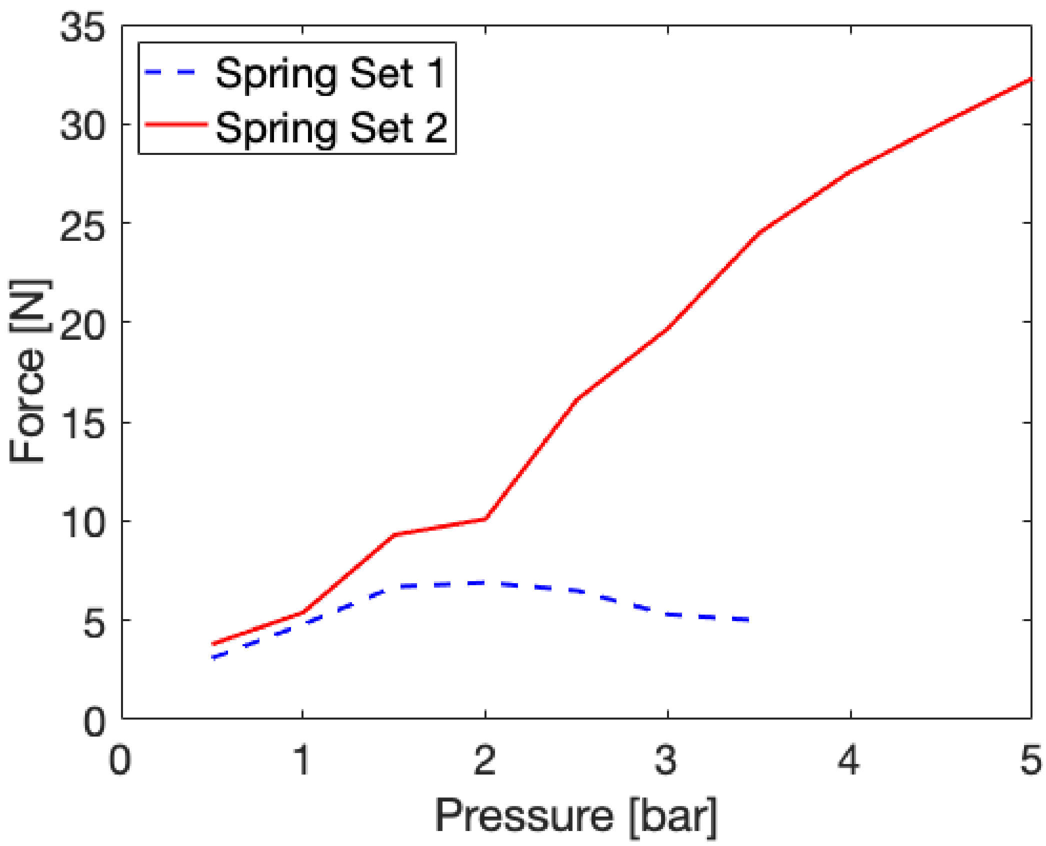


Figure C-5: The test results of the fingertip force measurements combined in a single figure.

The other metric that shows the performance of the prototype is the adaptive grip. The capabilities of adaptive grip without and with obstruction are shown in figures C-6 and C-7 respectively.

C-3 Discussion

The fingertip force of 32.3 N was measured with an actuator force of 100.5 N, whilst using spring set 2. A higher actuator force resulted in the failure of the ending of link 2 at joint C (PIP). This failure is probably caused by the quality and direction of the 3D printed part.

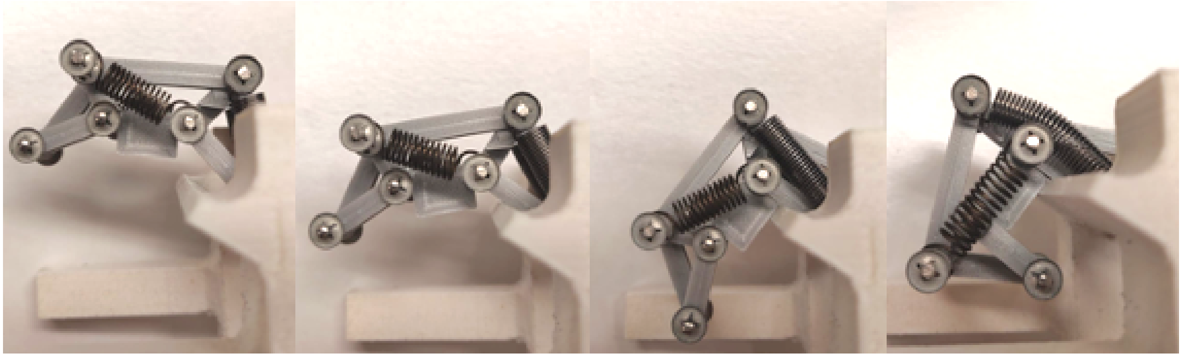


Figure C-6: The resulting motion of the adaptive grip mechanism (with spring set 1).

Figure C-7: The resulting motion of the adaptive grip mechanism when the proximal phalanx is obstructed (with spring set 1).

Fused deposition modelling ensures the strength of a part by printing in a single direction, therefore resulting in either yield or shear strength. The yield strength of the links was determined and was supposedly sufficient. However, a combination of both tension and shear force probably resulted in the breaking of the part. The force transmission ratio of 3.1 is mainly caused by the varying distance and angle of the line of action of the actuator force and the resulting reaction force at the fingertip. Additional force losses are caused by frictional forces and probably some tension in the springs. Furthermore, slight deviations could be possible due to the margin of error caused by the quality and calibration of the sensor and also the consistency of the contact point between the fingertip and sensor. To reduce the error of the contact point, springs with a high initial tension and high stiffness were used. Belter et al. (2013) performed a review of anthropomorphic prosthetic hands that were commercially available. The study showed that the average fingertip force of the middle finger of five different prosthetic hands was 8.5 N with a maximum of 14.5 N. The fingertip force of this prototype is much higher, whilst also being designed in a small and lightweight form factor. The fingertip force is, however, not directly related to the force design criteria, which was the grasping force that was measured with a dynanometer. Fortunately, the sum of added forces offers a promising perspective for achieving the grasping force of 46 N. Contrarily, the achievement of this criteria is not representative of the performance of the grasp. This is in part due to the force distribution of the prosthesis, which is more evenly and wider spread in human hands (Kargov et al., 2004). Contributing to this, is the non-yielding characteristic of materials chosen for prosthesis, which cause a relatively smaller contact area between the prosthesis and objects. Moreover, the friction between the prosthesis and objects can drastically change the performance of a grasp.

Adaptive grip was tested using two sets of springs. The first set of springs being lowest in stiffness, showed that is was easily and fully capable of adapting to obstructions. However, when the input force got too high, the finger would rotate around the PIP and DIP prior to completing the MCP rotation. Alternatively, the second set of springs showed that with a higher stiffness the finger is more likely to rotate around the MCP as a whole, but has more difficulty with the subsequent motions in the PIP and DIP. Ideally, a spring is required that has a high initial tension and low stiffness. There should be noted that the springs have a high spread in their stiffness. This spread is accompanied by different forces and lengths that the springs are able to achieve. Additionally, the manufacturers of the springs are cautious with presenting the maximum stretch length. Another observation was made, where the pin at joint C interferes with the elongated spring 1 C-6.

Appendix D

Recommendations

D-1 Expansion

The presented prototype shows promising results for a prosthetic hand for 4 year old children. Nonetheless, there are numerous improvements that could be made by either extensive research or small adjustments. Meeting all the criteria is key for creating a good prosthesis. Therefore, an extra degree of freedom in the thumb should be created. This could either be a passive locking mechanism or an active DOF operated by an actuator. In addition, the design can be expanded by adding the capability of wrist movements. Moreover, to pursue an anthropomorphic design, the lengths of the links should be adjusted according to the lengths of the subsequent fingers. Alternatively, this could be achieved by altering the fixed point of the MCP joint for the different fingers. Furthermore, the static model showed that the forces in the lower links of the finger transmission mechanism experience considerable higher levels of stress. The links towards the middle and top of the mechanism could therefore be made thinner, which could allow for a larger range of motion to be achieved, whilst also becoming slightly lighter. Also, grip tape could be applied to the links that make contact with objects. The added friction could increase the performance of certain grips and could potentially reduce the ejection phenomenon of grasping objects. Besides that, a socket should be designed that prioritises comfort and aims to position the pressure regulator, valve and power supply strategically. This could be done by positioning the heavier components proximal and lighter components distal. Additionally, the socket should be designed that is usable by amputees of varying residual limb lengths. Besides that, the index and middle finger, and/or the ring and pink finger could be coupled through the use of longer pins, thereby creating more space in-between the finger for different spring sizes. Lastly, the constantly changing line of action of the force applied at link 1, causes inequality in the applied torque throughout the motion of the finger. An alternative transmission mechanism should be designed that opts for an equal distribution of the torque applied to the finger through the actuation of link 1.

D-2 Research

The implementation of techniques to operate the hand could be investigated. These research topics include the various ways of triggering an input signal and feedback. The input signal should either have varying gain or it should be able to distinguish to different input signals. The former would allow for an increasing grip strength to be applied due to an increase in pressure, regulated through a pressure regulator. Such a system could allow for more fragile objects to be handled. Alternatively, the latter could use one signal for the pre-shaping phase and another signal for the adaptive grasping phase, where additional force it required, thus allowing a higher pressure in the cylinder. Effectively, these are two different ways of achieving the same principle, where the first phase uses less force and the second phase uses higher forces to firmly grasp an object. In addition, this would require spring with lower stiffness, which could result in a larger grasping stage due to less opposition of the springs. Preferably, the input signal originate from myoelectric signals that are either measured at different muscles or are easily distinguishable due to the controllability of their input gain, i.e. level of contraction of the residual underlying muscle. Another option would be the use of a reed switch, which detects the position of the piston due to a built-in magnet. Consequently, this detection could be used to increase the pressure in the cylinder through the pressure regulator, thereby creating a larger grasp force from the desired piston position. Contrarily to the input signal variations, it should be considered that the supply pressure becomes sub-optimal related to the gas usage, thus total number of operations on a single CO_2 cartridge. Alternatively, a PAM could be investigated and designed, which naturally increases its force when the stroke is continued.

D-3 Adaptation

The theoretical concept of the hand prosthetic is designed for four year old children. Ploegmakers et al. (2013) showed the grip force of children. The difference between a 4 and 5 year old child is already quite large, see table D-1. The exponential growth that children go through would mean that a new prosthesis is required after each year. An adaptive model should be created, which allows scaling of the links, springs and pneumatic cylinder actuators for a prosthesis that can be used by people with varying ages. It might even be possible to 3D print the whole hand for adults, since the parts can be made much larger.

Age (yr)			**Boys**		
	n	Dominant(N)	Non-dominant(N)	Height(cm)	Weight(kg)
4	124	55.9	52.0	111	19
5	102	73.5	66.7	117	22
			Girls		
	n	Dominant(N)	Non-dominant(N)	Height(cm)	Weight(kg)
4	109	50.0	46.1	111	19
5	105	65.7	58.8	118	22

Table D-1: This table presents the grip forces exerted by 4 and 5 year old boys and girls. The forces of both dominant and non-dominant hands are depicted.

D-4 Evaluation

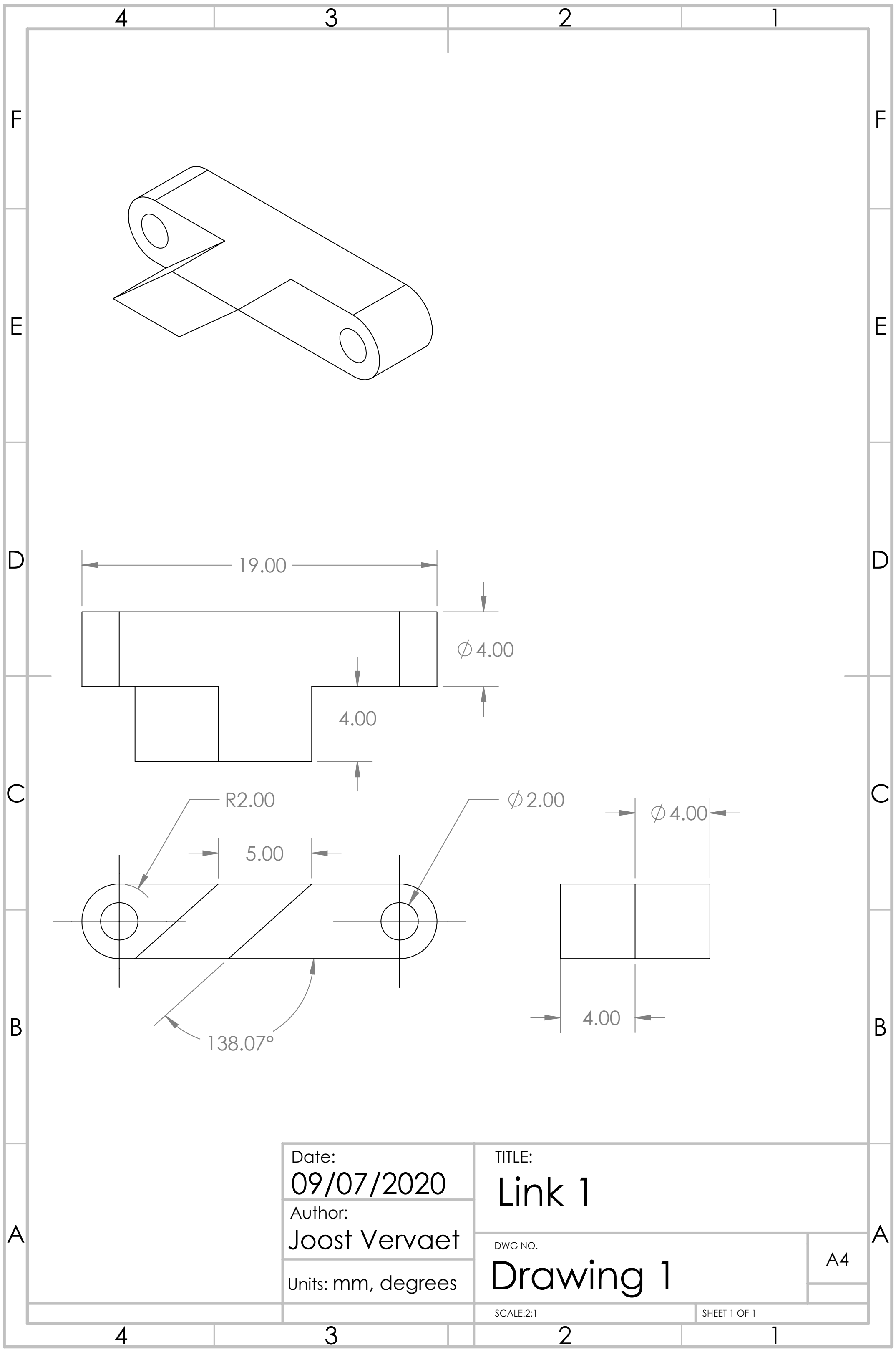
Once a prosthetic hand is fully manufactured and assembled, it should be tested by amputees. The tests should include the capability of performing several activities of daily living. Furthermore, a number of grasps on objects of varying shapes and sizes should be tested. Lastly, the distribution of the forces amongst the different contact points between the finger and the object should be measured and evaluated.

Appendix E

Technical Drawings

Note

The remaining pages are filled with technical drawings of the prototype. These technical drawings are provided to enable exact reproduction of the prototype using any manufacturing method. In addition, the technical drawings might give a better understanding of the design of parts that were used for the prototype.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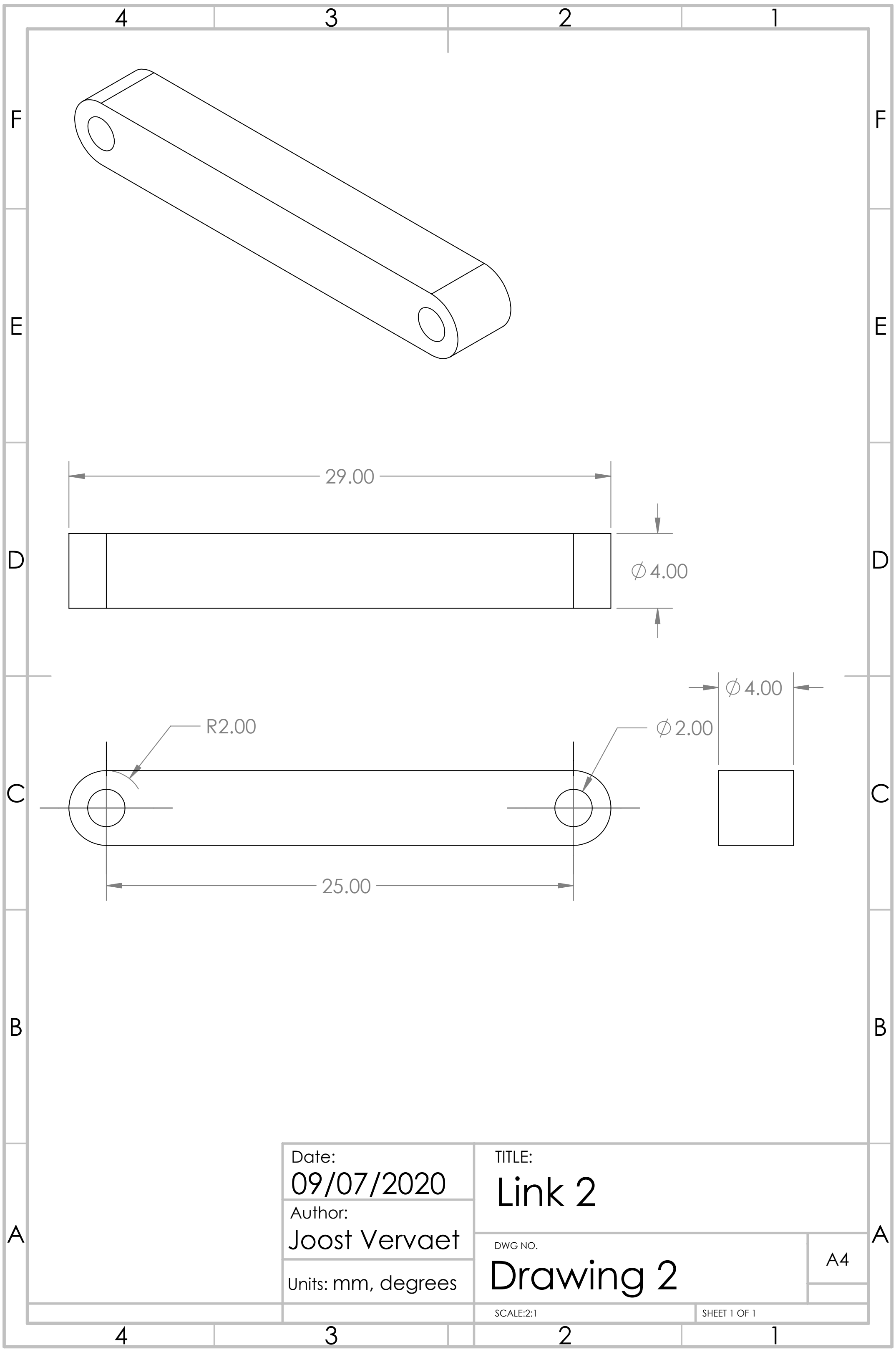
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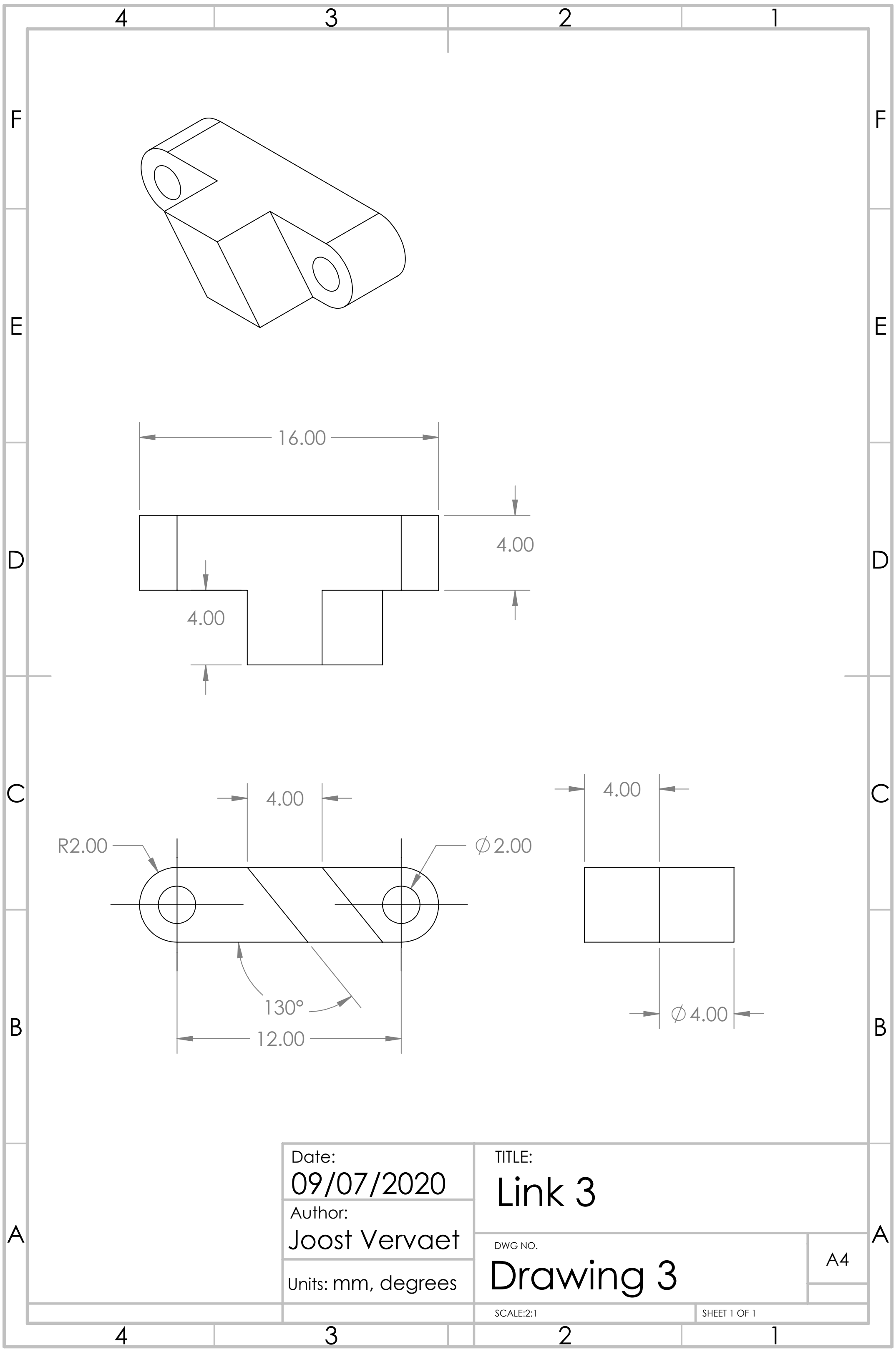
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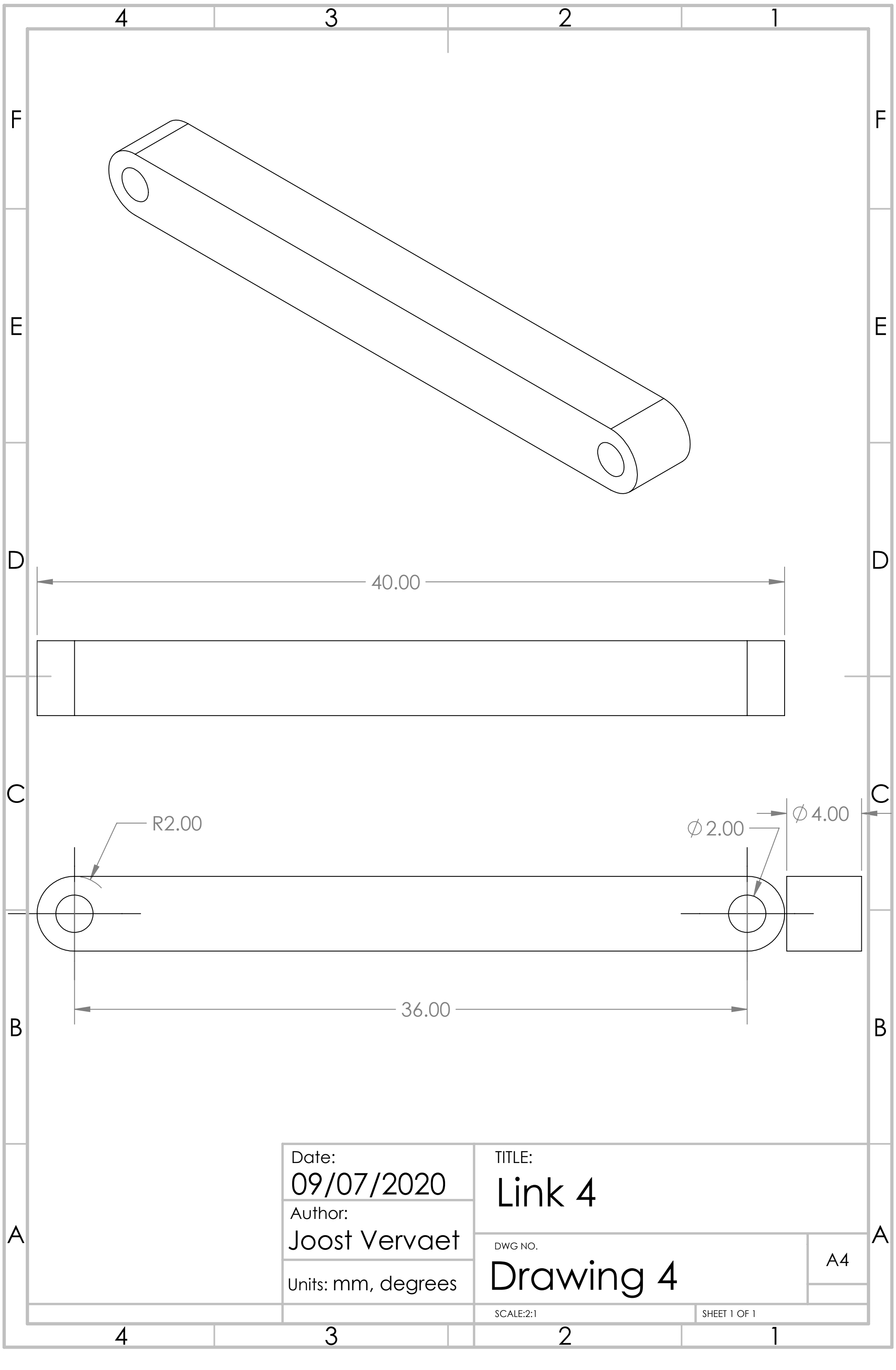
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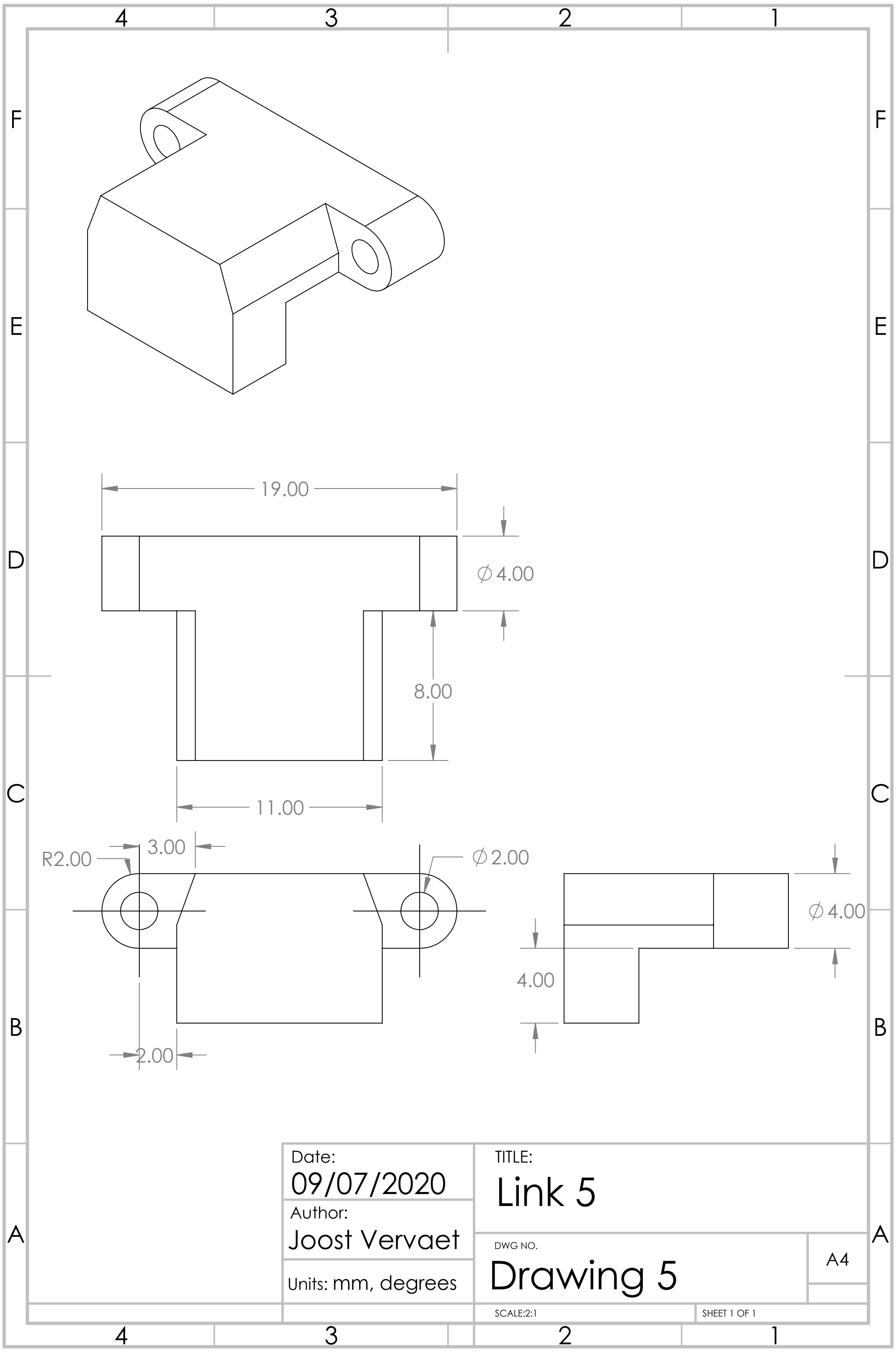
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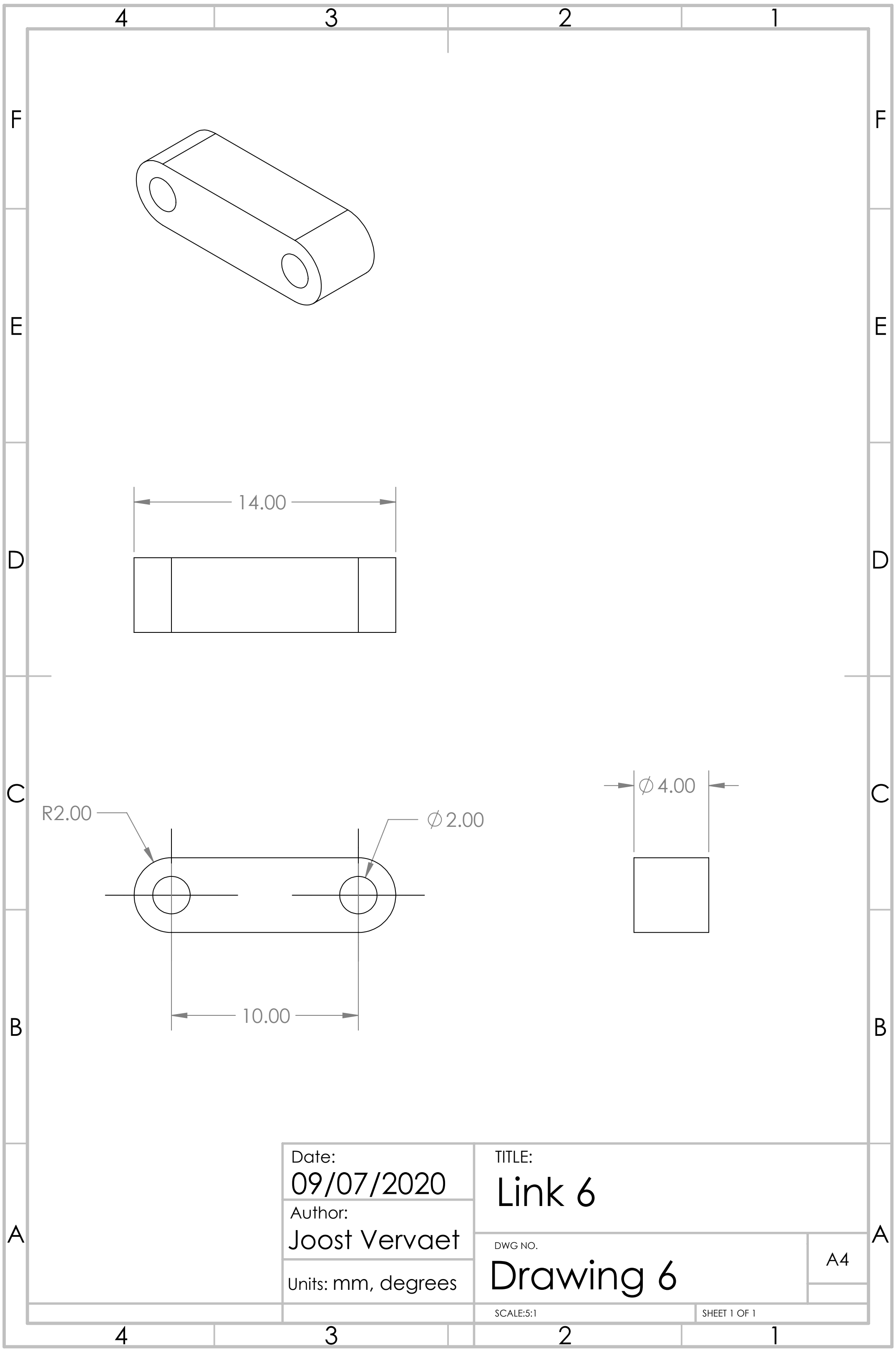
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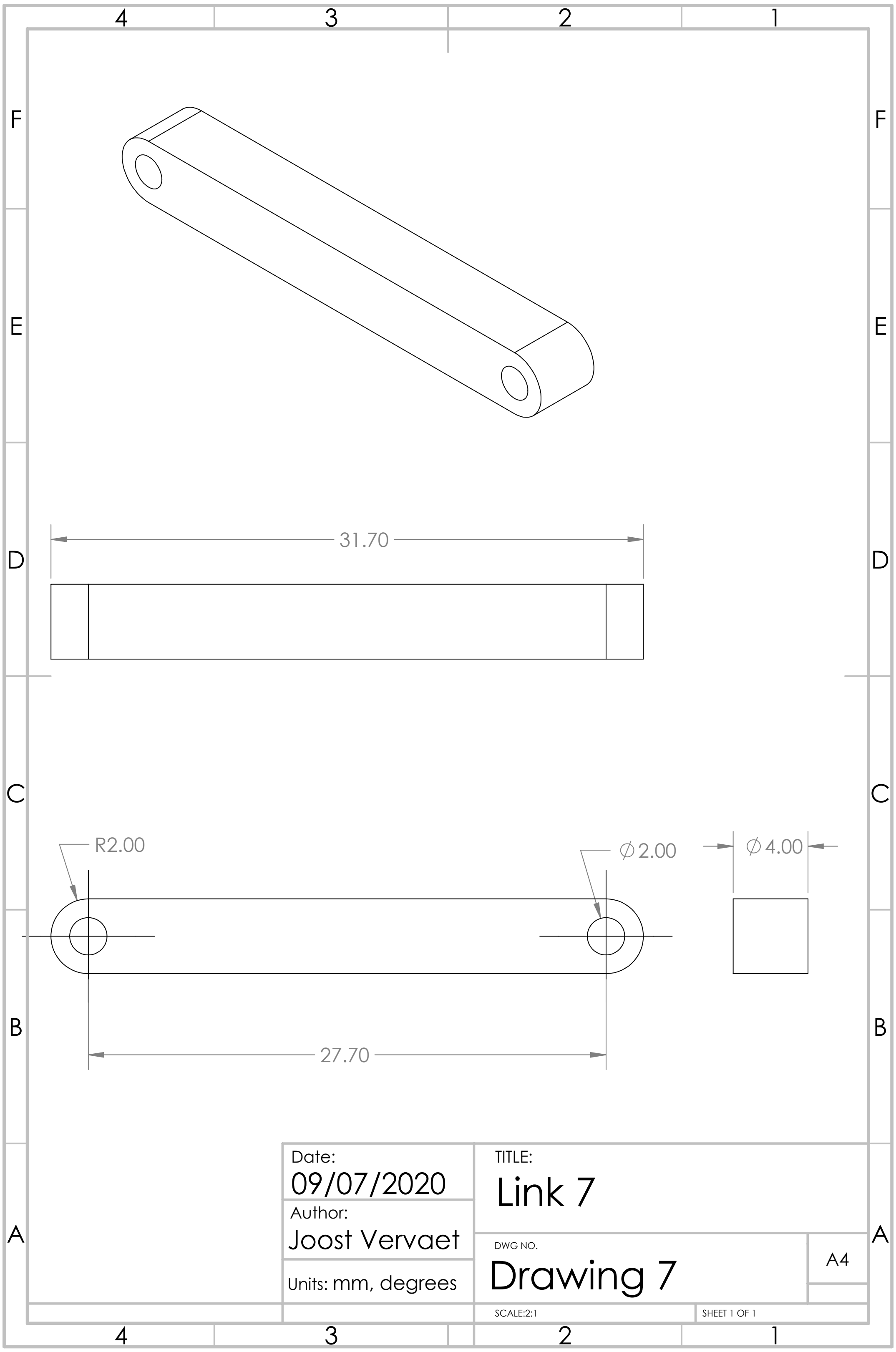

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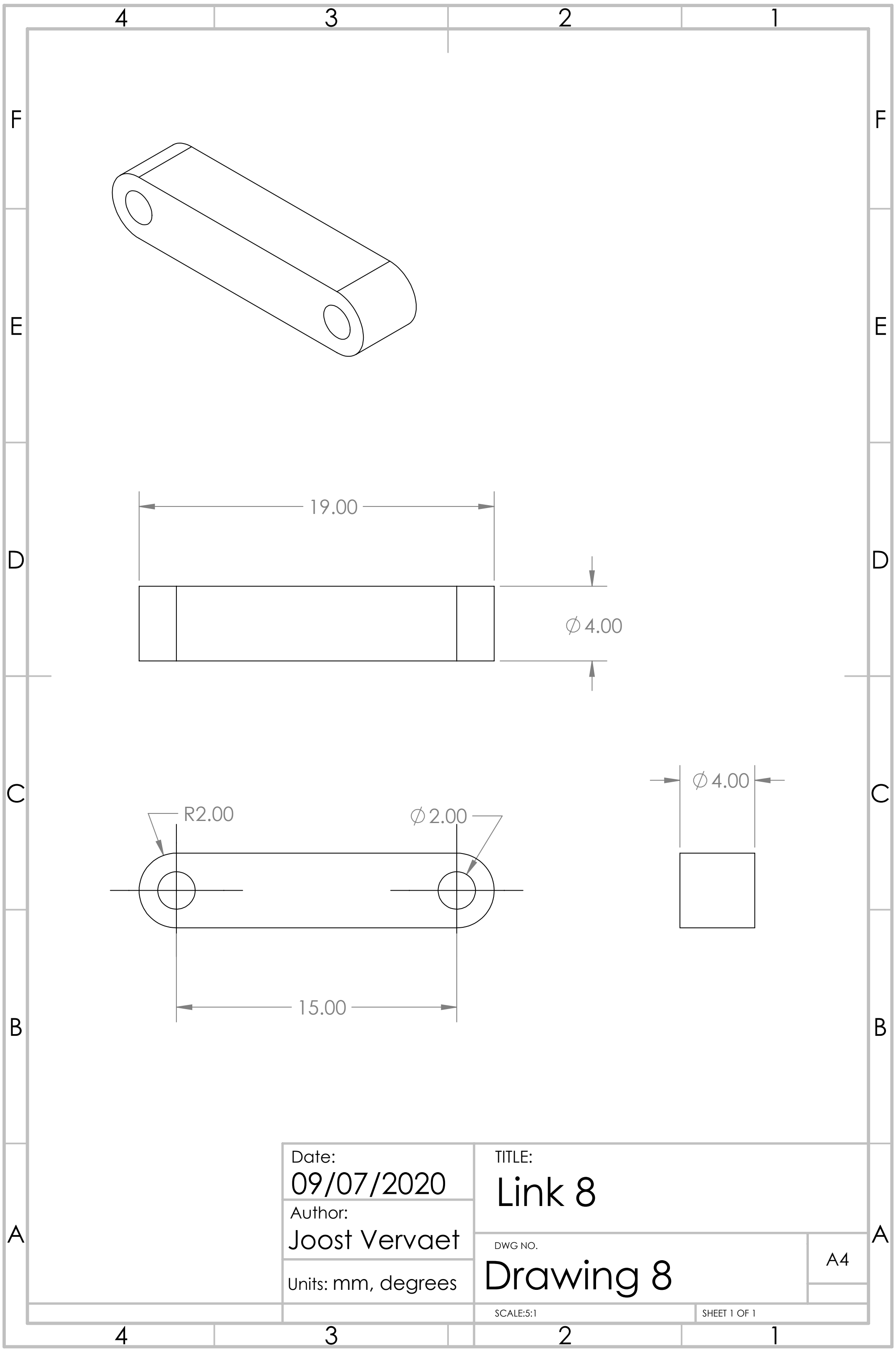
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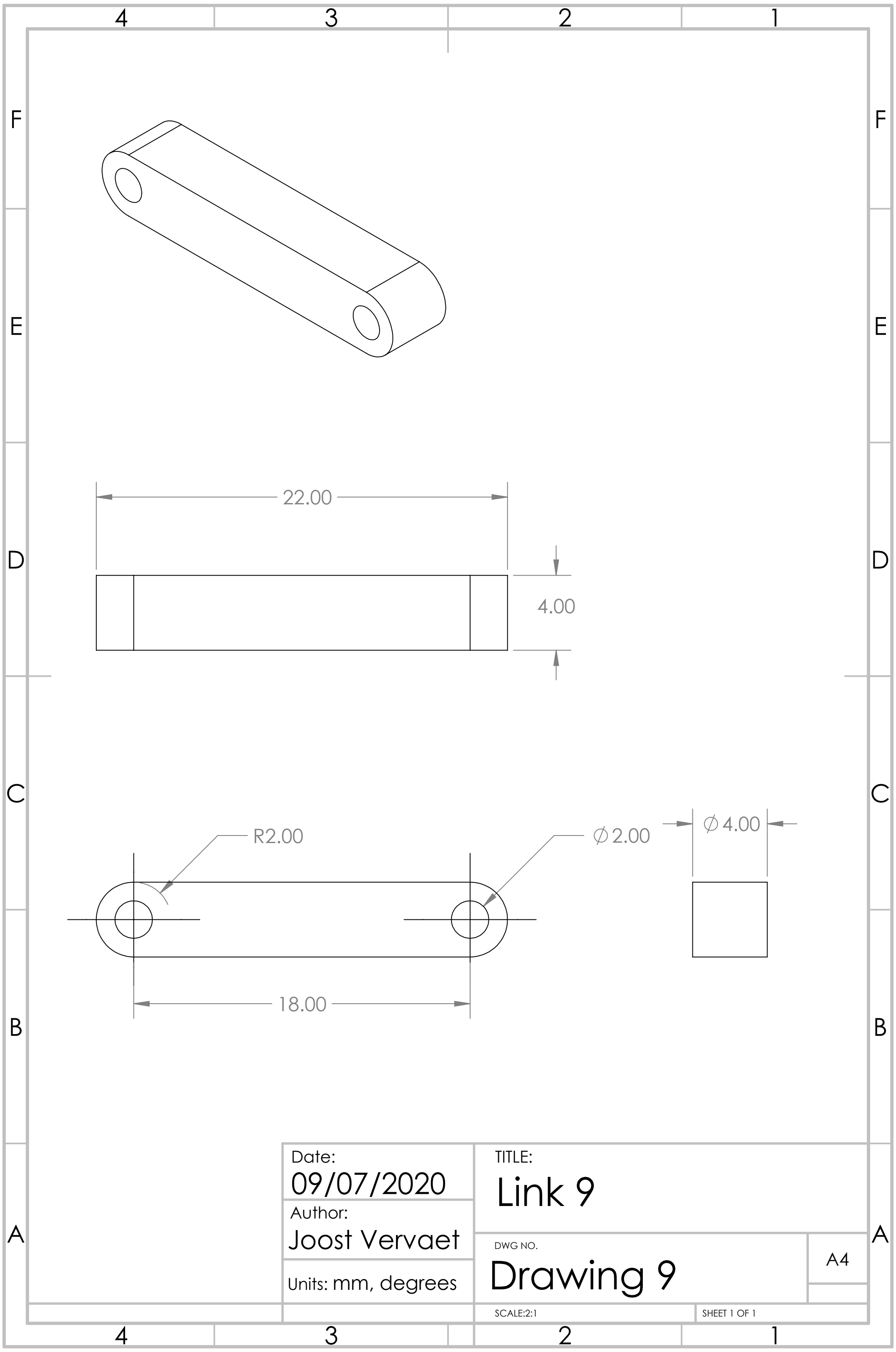
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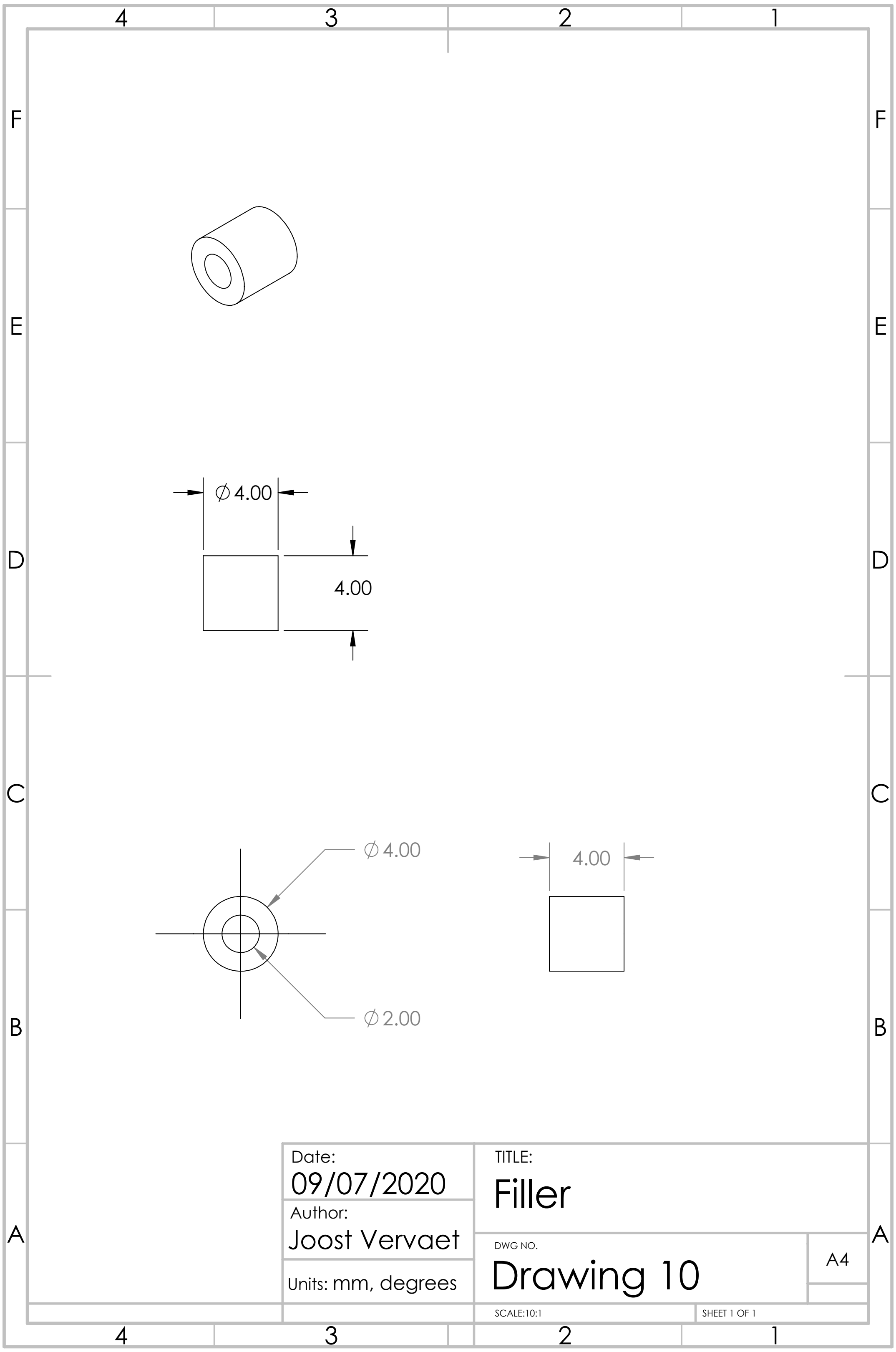

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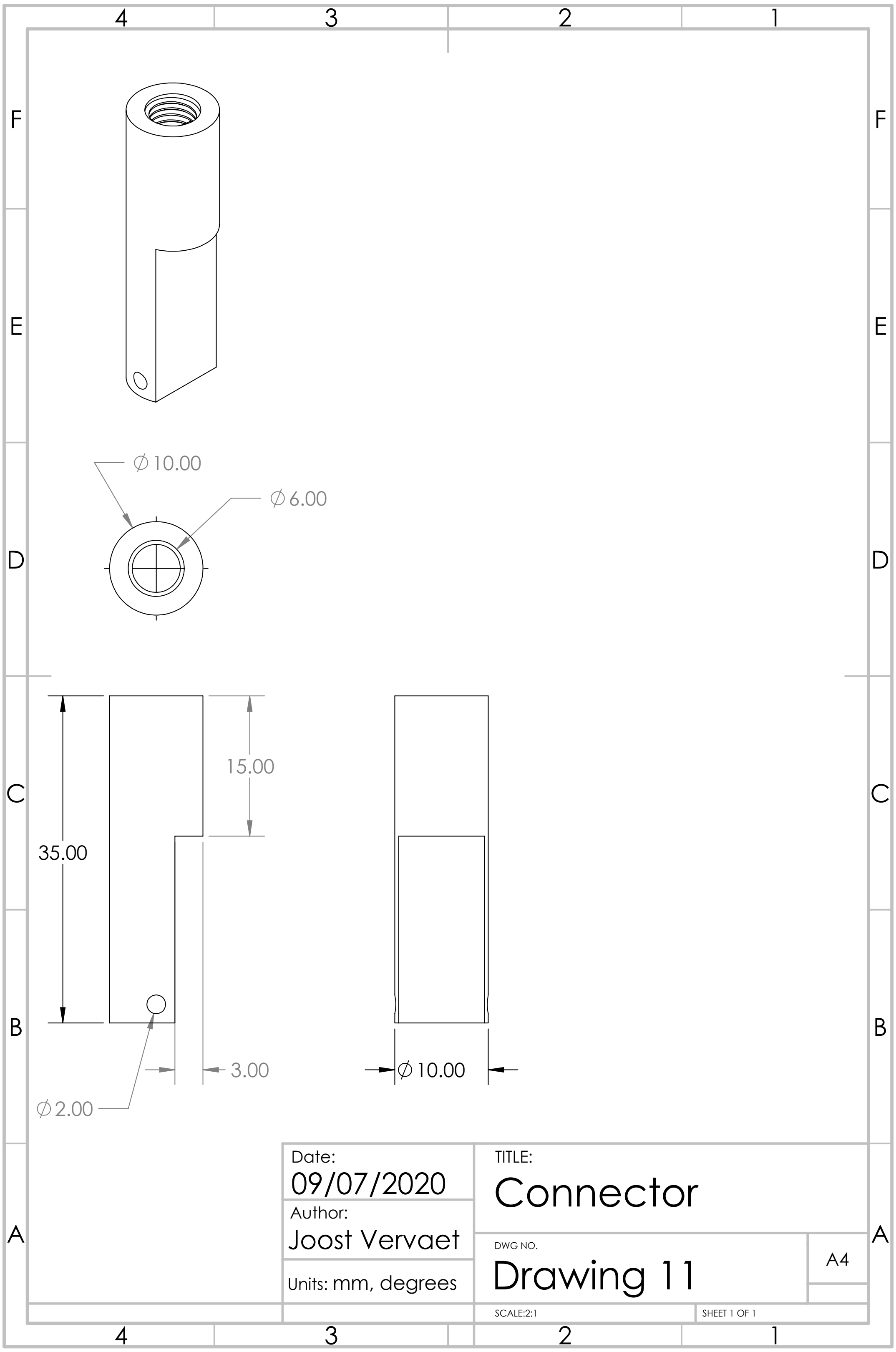

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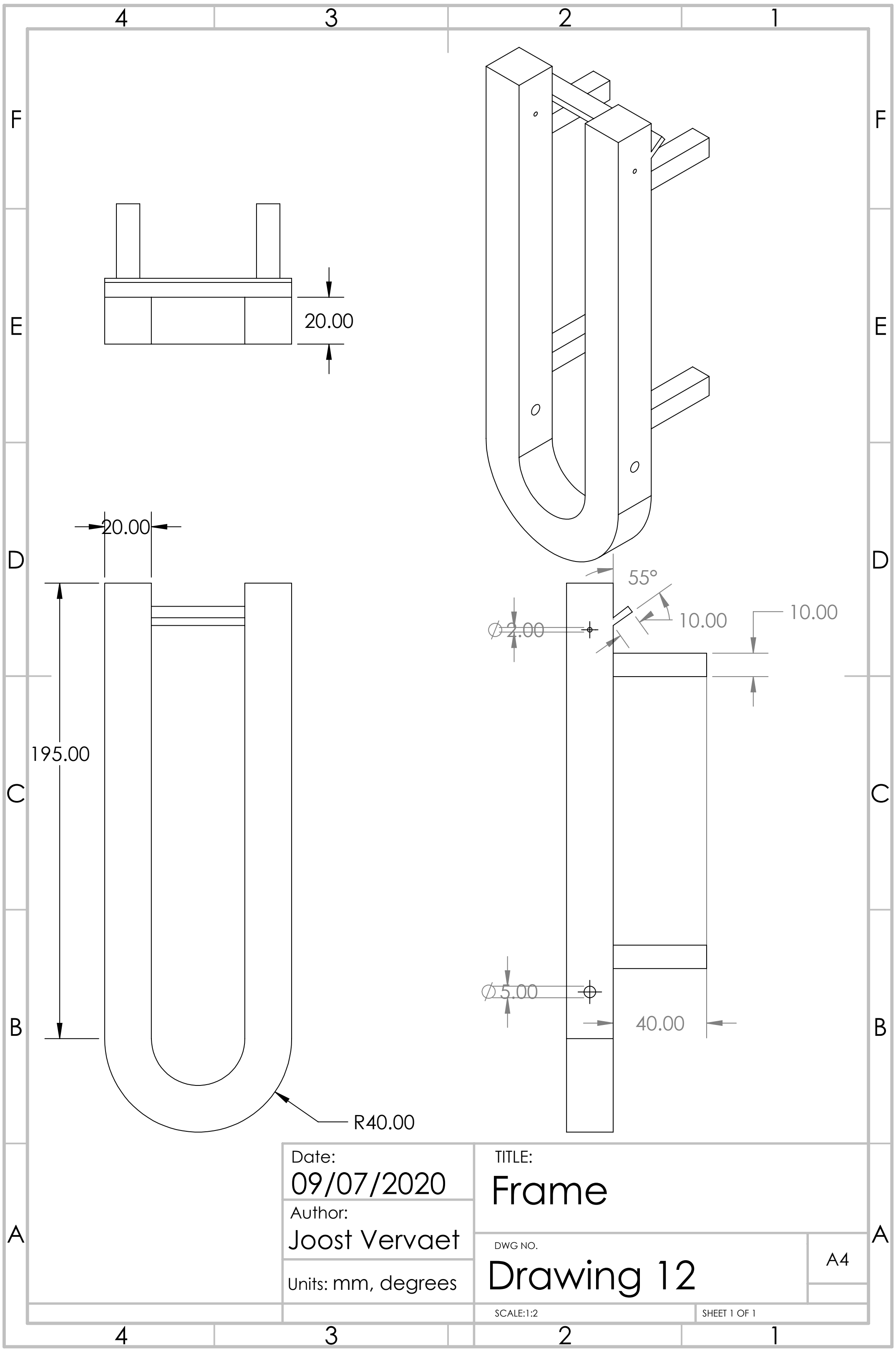
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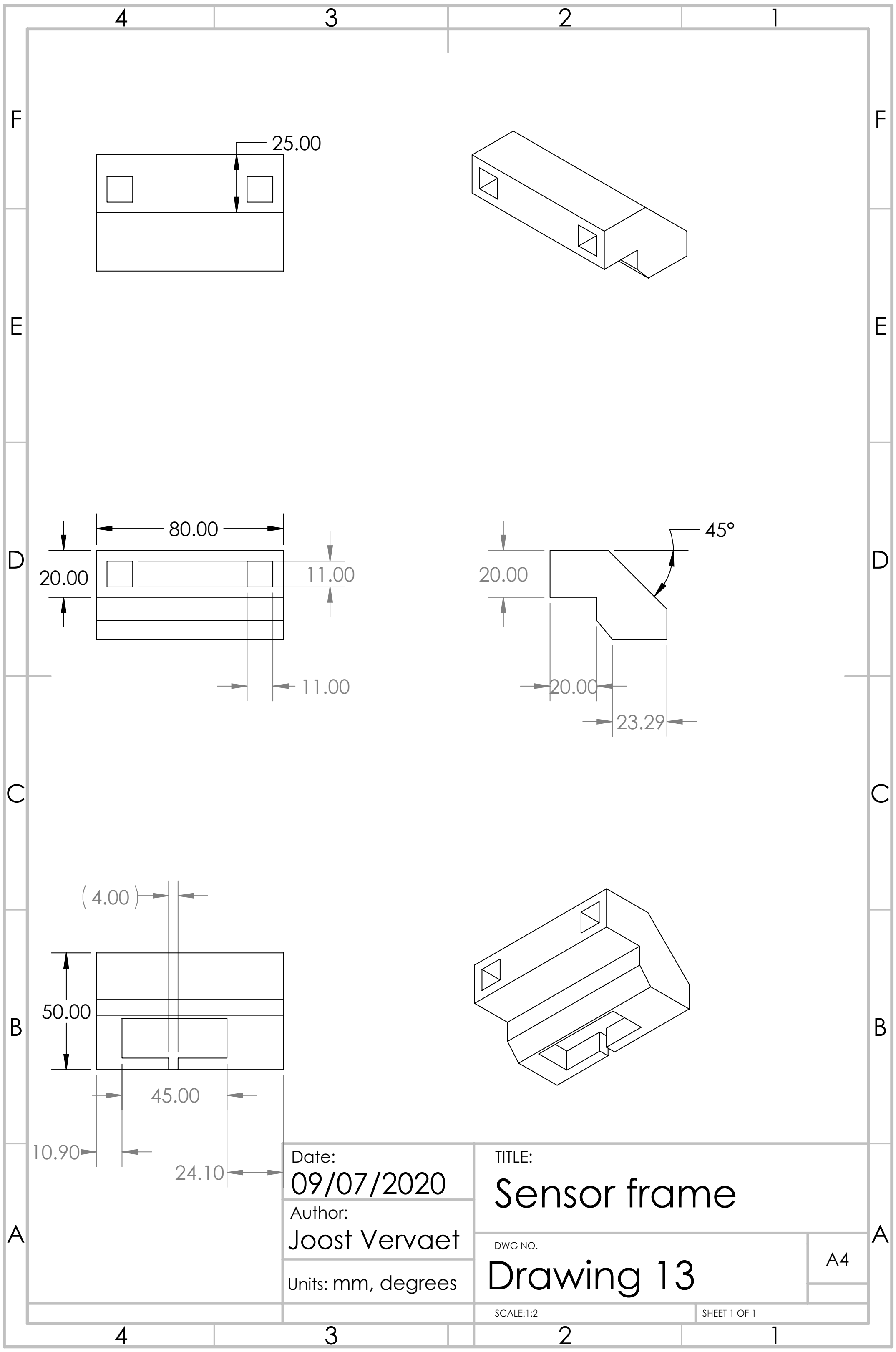
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