The Power of Pneumatics

Design of a Novel Elbow Prosthesis

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Design of a Novel Elbow Prosthesis

by

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An electronic version of this thesis is available at http://repository.tudelft.nl/.
An abbreviated version for journal publication of this thesis is available at Appendix B.

Abstract

Introduction & Goal Almost 57 percent of above elbow amputees stop using their prosthesis. One of the main factors of non-wear is the high weight. Pneumatic actuation can be more lightweight than electronic actuation, and might therefore help with reducing the mass of a prosthesis and increase its comfort. To illustrate this potential, the goal of this thesis was: ”To show the potential of pneumatic prostheses by designing a lightweight pneumatic elbow prosthesis, with a functionality equivalent to existing prostheses”

Design Methodology The design process was divided into five phases: Analysis, Conceptualization, Embodiment, Manufacturing and Assembly, and Testing. The analysis brought to light the functions, requirements, wishes, and design values. The two design phases, conceptualization and embodiment, translated these into a potentially viable prosthesis. Afterwards, a prototype was manufactured, assembled, and tested to see if the functions, wishes, and requirements were met.

Results The prototype weighs almost 1300 [g]. A payload of 4.0 [kg] can be lifted to 87 [°]. In theory, 2.5 [kg] can be lifted throughout the entire range of motion, up to 140 [°]. The maximum pronation/-supination torque is over 2.8 [Nm] throughout the entire range of motion of $\pm$ 90 [°]. The locking mechanism is theoretically capable of passively holding over 6 [kg] in any position. In theory, an average of 125 cycles of use can be achieved per 25 [g] CO_2 cartridge.

Discussion The prototype does not have an integrated fuel source. The weight of the prototype is $\approx$ 1300 [g]. With minor changes the weight can be reduced to 900 [g] Switching to electronic control significantly decreases the prosthesis mass. Removing pronation/supination and the active locking mechanism reduces the weight to less than 600 [g]. The frame of the prototype was under-constrained, leading to an increased friction for higher angles of flexion.

Conclusion The combination of the high functionality and low mass of the prototype shows a potential for pneumatic actuation for prostheses.

Keywords ***Pneumatic, Elbow, Prosthesis, Lightweight***

Preface

Before you lies the thesis ”The Power of Pneumatics: Design of a Novel Elbow Prosthesis”. The inspiration for this design comes from dr.ir. Dick Plettenburg, who has been one of the sole advocates for pneumatic actuation within upper-extremity prostheses. As a student of (bio-)mechanical engineering and as a person who believes that technology should be used to better the lives of everyone, the idea of designing a prosthesis in a way that has not been done before spoke to me. Whilst researching the topic of elbow prostheses, my passion for this subject grew by the day. The realization of the level of impact a prosthesis has on the life of the user and the shear amount of users worldwide, especially in developing countries, instilled a drive in me to continue pursuing the improvement of prostheses for the rest of my life.

I could not have achieved the design as it it proposed in this thesis without the following people. First of all, my supervisor Dick Plettenberg, who provided me with advice and subtle guidance, allowing me to design the prosthesis as I had envisioned it, without being forced in any direction. It was the trust of professor Plettenberg in my capabilities that drove me to honor his original idea of creating a pneumatic elbow prosthesis. Furthermore, Jan van Frankenhuyzen and Damian de Nijs for helping me to transfer the design from the realm of theory into a tangible prototype. Finally, my family for supporting me unconditionally during the writing of this thesis. Thank you all,

J. Kieft
Delft, October 2020

Contents

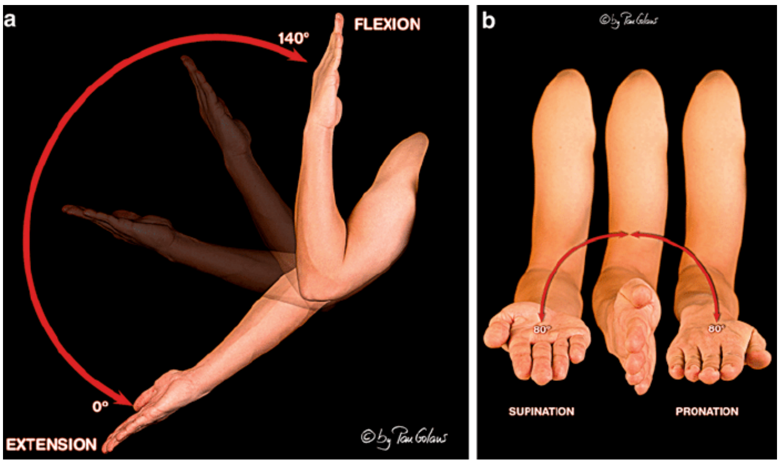
1

Introduction

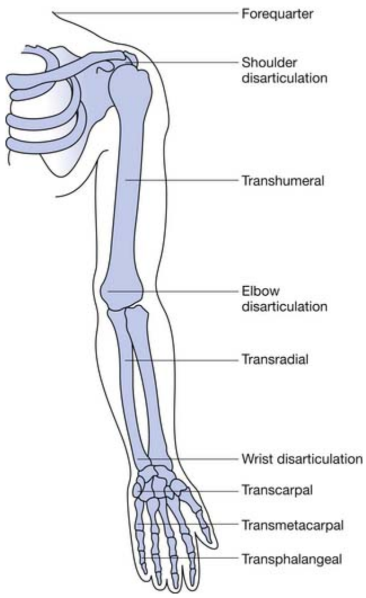
There may be over one million people with an above elbow amputation worldwide [1] [2]. It is estimated that 80 percent of these amputees live in developing countries [2]. Despite progress achieved over the last decades, or even centuries, rejection rates are still high for elbow prostheses: 57 percent of above elbow amputees stop using their prosthesis compared to 6 percent of trans-radial amputees [3]. A survey in 2007 [4] asked a group of 266 amputees, 21 percent of which had a transhumeral amputation, for factors of non-wear. The responses can be clustered into the three basic requirements for a prosthesis [5]: control, comfort and cosmetics. An overwhelming 98 percent of prosthesis non-users reported a lack of functionality (control) as a reason. 95 percent reported comfort as a reason, with 88 percent of non-users saying the prosthesis is too heavy. The appearance of the prosthesis played a role in not wearing the prosthesis for 70 percent of the non-users. Progress in the field of elbow prostheses is required to reduce the number of prosthesis rejections.

1.1. Elbow Anatomy

The unique build of the anatomical elbow allows for two degrees of freedom: flexion/extension (FE) and pronation/supination (PS), see Figure 1.1a.



(a) The two ranges of motion of the elbow: Flexion/Extension and Pronation/Supination. Image taken from [6]



(b) The different classifications of upper extremity amputations. Image taken from [7]

Figure 1.1: The anatomy of the upper extremity, with a focus on the elbow range of motion (RoM)

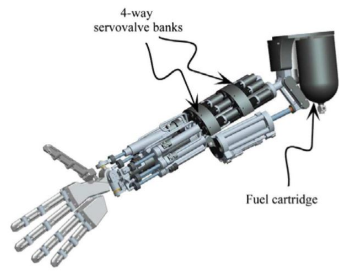
A distinction should be made between different amputees, see Figure 1.1b. Elbow disarticulations are amputations through the elbow joint, transhumeral amputees have the amputation at some location through their humerus. The importance of this classification is the relevance for design. Because of the presence of an anatomical structure, there is less space for a mechanical structure in the case of elbow disarticulations. For transhumeral prostheses, more space is available for design aspects such as actuation units or power supply, depending on the location of amputation.

1.2. Elbow Prostheses

A review of the current state-of-the-art of elbow prostheses was performed in the form of a literature review, shown in appendix C. This review discusses the advantages and disadvantages of the three main types of actuation for elbow prostheses: Body-powered, Electric & Pneumatic. It's shown that each type has their own drawback(s) and area(s) of expertise. Body-powered prostheses eliminate the need for an external power-source, are lightweight and (can) provide proprioceptive feedback, but the harnass is cited as being uncomfortable and not visually appealing. Electric prostheses are strong and often look beautiful, but are known to be heavy, expensive and lack proprioceptive feedback. One of the conclusions of this literature review is that pneumatic actuation is underrepresented in the field of elbow prostheses. The benefits that pneumatic actuation offers align with the reasons for non-wear given by users, such as being lightweight. The prototypes that do exist show that the often cited downsides of pneumatic actuation, such as a lack of control and safety, are overemphasized [8]. However, the last relevant publication regarding such a design dates back to 2008. For a complete list of all sources, a broader explanation of elbow prostheses, a summary different types of (1) actuation, (2) control, and (3) topology, and an overview of the most prominent current prostheses, please see the literature review (Appendix C).

(a) An example of a body-powered prosthesis. Image taken from [9]

(b) An example of an electric prosthesis. Image taken from [10]



(c) An example of a pneumatic prosthesis. Image taken from [8]

Figure 1.2: Examples of the three main types of actuation for elbow prosthesis: Body-Powered, Electric & Pneumatic

1.3. Goal

This thesis will try to show the potential of pneumatic actuation by designing and testing a prototype. It should be noted that the goal is not to create the perfect prosthesis, as this is impossible due to the wide range of different requirements and wishes by the user. Patients have already shown the ability to switch between different types of prostheses depending on the required activity. A well-functioning pneumatic prosthesis could be another tool in the arsenal of a patient, to be used when the situation requires it. They could bridge the gap between electric and body-powered prostheses, functioning as a general use prosthesis during the day. The expected low mass and lack of a harnass lead to good comfort for long-time use. Furthermore, the lack of said harnass also means that the device can be designed to be more anthropomorphic and therefore be more visually appealing. To summarize, the goal of this paper is:

To show the potential of pneumatic prostheses by designing a lightweight pneumatic elbow prosthesis, with a functionality equivalent to existing prostheses.

2
Design Process

The aim of the thesis is to have a working prototype of an elbow prostheses. The design process will be subdivided into five phases, based on the Delft Design Guide [11]. For an optimal final design, there are feedback moments within each phase and between the different phases. By assigning specified inputs and outputs to each phase, the design process becomes more manageable and easier to follow. The phases of the design are: Analysis, Conceptualisation, Embodiment, Manufacturing & Assembly, and Testing & Evaluation. The first three phases are mainly focused on design, whereas the fourth and fifth phase are more concerned with prototyping. The desired output are shown in Table 2.1. The phases design process is visualised in Figure 2.1 and with more detail in Figure 2.2.

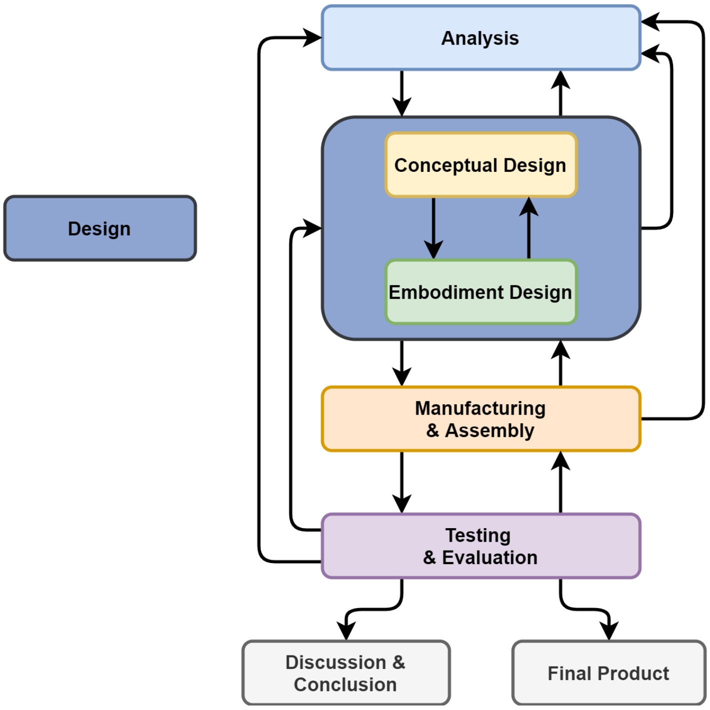


Figure 2.1: The design process visualised (summarized)

2.1. Phases

Analysis The analysis phase is aimed to achieve a better understanding of the design values, goals and wishes. The analysis will be done from three different perspectives: Other prostheses (competitors), users, and literature. The goal of these analyses is generating a function tree. Evaluating the functions is vital for proper time management. This will be done using the MoSCoW method. The output of this phase will be the design values, requirements, wishes, and functions.

Conceptual Design The second phase in the design methodology concerns conceptualization, or synthesis. Functions will be clustered to form modules. Possible concepts to the modules will be synthesized and evaluated to achieve the output of a concept choice per module.

Embodiment Design During the embodiment design, the chosen concepts will be translated into a final design. The method of achieving this desired output is to create a parametric design and to determine the parameters based on models (such as MATLAB) or available components. This combined effort leads to a feasible CAD design, which will be optimized to reduce costs and time during manufacturing.

Manufacturing & Assembly When the embodiment design has been completed, manufacturing and assembly can begin. The manufacturing of the custom parts will be done at the workshop at 3mE. For this, technical drawings are required.

Testing & Evaluation If the manufacturing and assembly have been successful, the prototype can be tested to see if the requirements and wishes are met.

Table 2.1: An overview of the outputs of each phase

Phase	Output(s)
Analysis	Design Values
	Design Requirements
	Design Wishes
	Design Functions
Conceptual Design	Prototype Modules
	Concept Choice per Module
Embodiment Design	Finished Design
	CAD Model
Manufacturing & Assembly	Technical Drawings
	Prototype
Testing	Evaluation of the Requirements
	Concrete Future Recommendations

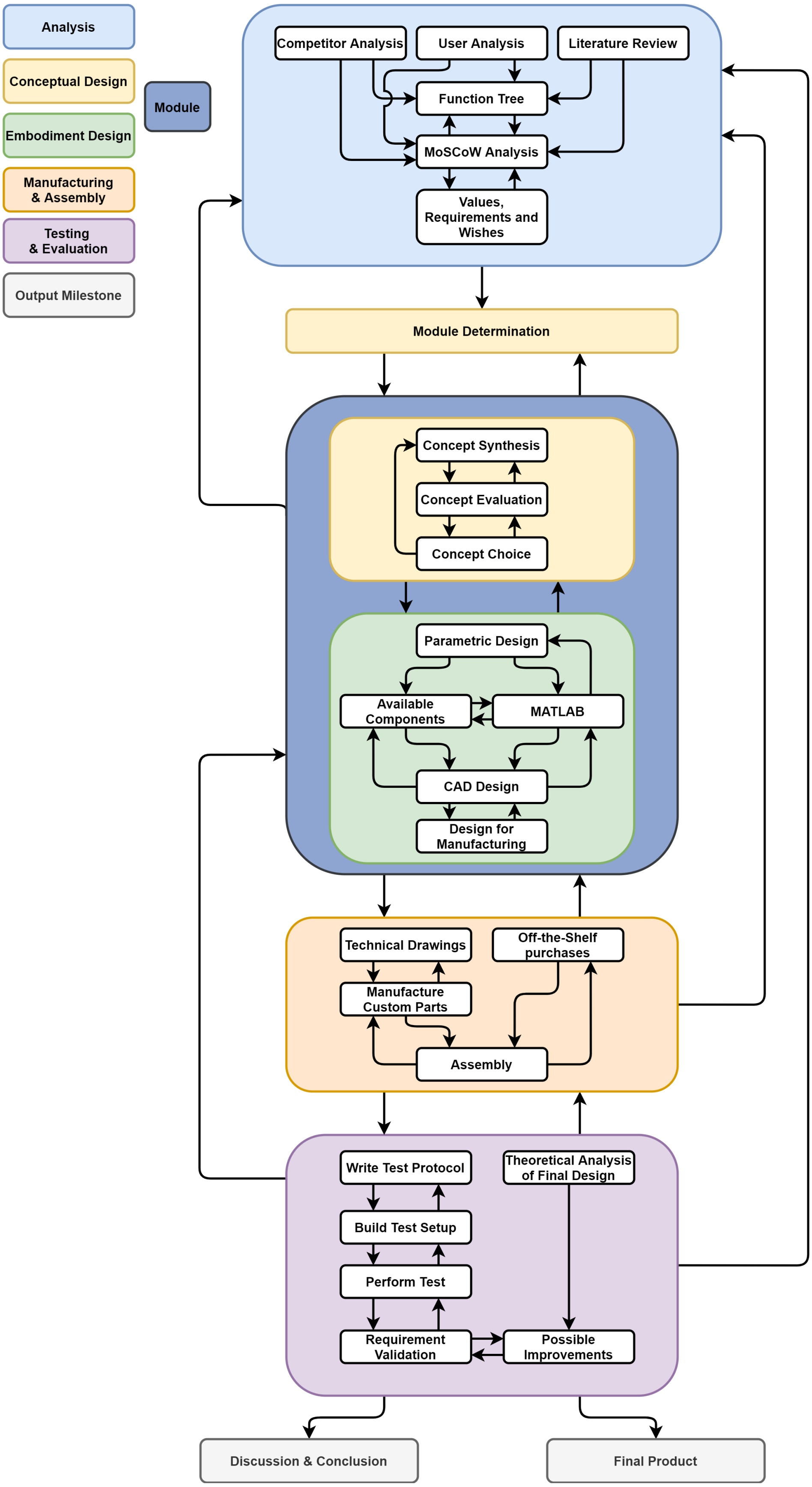


Figure 2.2: The design process visualised (detailed)

3
Analysis

To achieve an understanding of the design values, requirements, and wishes, the analysis phase is required. The methods used to achieve a broad view of this field are doing a user analysis, competitor analysis, literature review, generating a function tree, and doing a MoSCoW analysis of the discovered functions.

3.1. User Analysis

A user analysis will help with determining the requirements and wishes of the user, which may differ from the technical requirements and wishes. Since prostheses have a high rejection rate [3], user analyses are vital to achieve daily use of the prosthesis. However, within the time-frame of the graduation project, a user analysis is not feasible. Therefore we must rely on user analyses done in the past. A reliable analysis about device abandonment factors for upper-limb prostheses was published in 2007 by Biddiss et al. [4]. From this user review, multiple factors proved to play a role in device rejection, the most important ones are discussed below.

Control & Functionality On the top of the list of factors for device abandonment is functionality. Multiple factors from the Biddiss analysis fall in this category. Respondents said they were just as functional or more without the device and that they had more sensory feedback without the prosthesis. Furthermore they called the prosthesis inconvenient, and too difficult or tiring to work with.

Comfort Comfort is an important factor for the use of prostheses. Even if the functionality was similar to a human arm, an uncomfortable arm would still disappear in the closet for some users. In the review by Biddiss et al, a vast majority the respondents said they were more comfortable without the prosthesis. An important factor for comfort is the fitting, i.e. the connection to the user arm. This importance has been known for some time and is one of the reasons for e.g. the development of the WILMER approach back in 1998 [12]. The most important factor for non-wear due to discomfort is weight [4].

Cosmetics The users are human and therefore have their personal opinion regarding the appearance of a prosthesis. Appearance can be more important than functionality or comfort in some countries because of social, economic, cultural, psychological or religious reasons [13].

Costs The importance of the price of a prosthesis is a debatable subject. According to the Biddiss analysis, costs did not play a major role in prosthesis rejection. Even though Biddiss et al. avoided bias as much as possible, by including all demographics as much as possible and carefully thinking about how to acquire data, this data set comes from the developed world. In low-income countries, where an estimated eighty percent of the amputee population lives [2], it is important that any public service is cost-effective [14].

3.2. Literature Review & Competitor Analysis

A literature review concerning actuation for elbow prostheses has been performed prior to starting the thesis. It is included in the appendix: Appendix C. The most significant elbow prostheses, as discussed in the literature review, were studied to find additional functions. These prostheses were also used to quantify the requirements, see section 3.5.2. The data from this competitor analysis is shown in Appendix D.. Most elbow prostheses achieve pronation and supination using a terminal device. To further accentuate the potential for pneumatic actuation, the designed prosthesis will also implement active pronation and supination.

3.3. Function Tree

Generating a function tree is a method to systematically analyse the functions to be achieved in the design. This process might not help with finding obvious functions, such as ”generate movement”, but for less-distinct functions, this method can be an important step, i.e. ”allow for maintenance”. The function tree should be structured in such a way that functions that might be hidden during this stage of the design can be added later in the design process, enabling a feedback moment of the design. The functions were generated by brainstorming and looking at other prostheses and deducing the functions. The function tree, after passing the MoSCow analysis, is shown in Figure 3.1. The functions are explained in detail in Appendix E.

3.4. MoSCoW Analysis

A MoSCoW analysis helps with prioritizing functions by categorizing them in in one of four categories; (1) Must have, (2) Should have, (3) Could have and, (4) Won’t have [15]. Anything in the “must-have” category is considered essential for the project to be viable. ”Should-haves” are important to the project, but they are not vital. If left out, the product still functions. However, if they are included, they add significant value. ”Could-haves” can be regarded as ”nice-to-haves”, since they would add value to the project, but not to such a significant degree as the ”should-haves”. If there is a limited amount of time available, the ”could-haves” are the first to be de-prioritized. One of the powers of the MoSCoW analysis lays in the ”won’t-have” category. These are functions that will not be included (this version), no matter the time-frame. This helps with expectation management for both the designer and client.

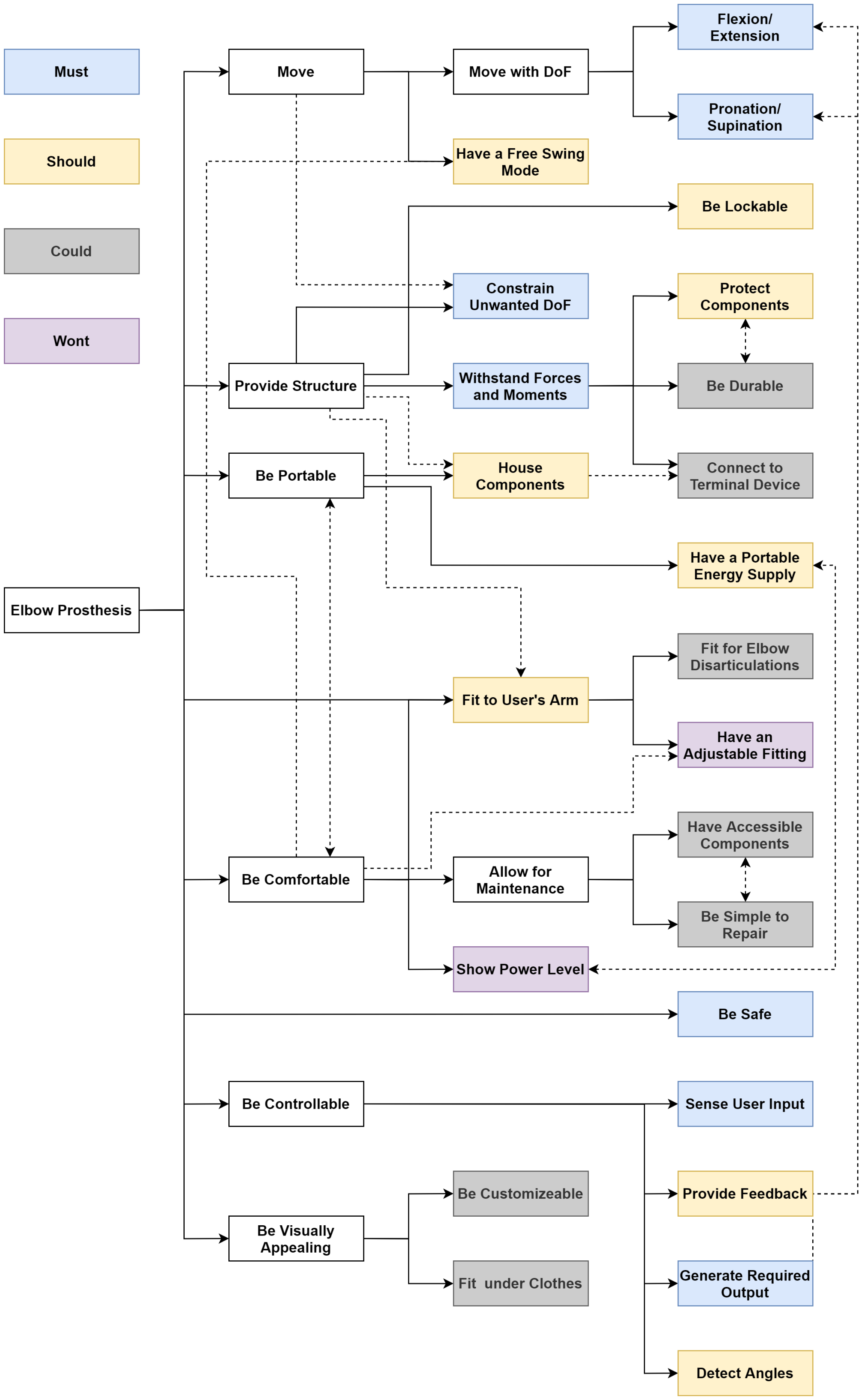


Figure 3.1: The final function tree after passing the MoSCoW analysis

3.5. Design Values, Requirements and Wishes

Design values guide the designer when making choices, such as ”modularity” or ”durability”. A requirement is a goal which must be fulfilled in order for the product to function properly, such as a minimum required range of motion. A wish is a design parameter that could be optimized when the opportunity presents itself, such as appearance.

3.5.1. Design Values

The design values for this thesis are:

1. Durability
2. Safety
3. Simplicity
4. Easy to manufacture & assemble

Durability In the field of externally powered elbow prostheses, pneumatic actuation is likely to be more durable than electric actuation. One of the reasons for this is the vulnerability of electric motors to sand and other micro-particles. By having durability as a design value, the benefit of durability can further be exemplified. Furthermore, as discussed earlier, more and more prostheses are required in developing countries where access to healthcare is limited. Having a more durable prostheses means that less repairs and replacements are needed.

Safety Safety should always be a number one priority. Since it is impossible to know all the dangers beforehand, safety can not truly be quantified. By having safety as a design value, it is ensured to be taken into account during every decision of the design.

Simplicity If the user has a better understanding of the device, they might feel more comfortable with using it. This straightforward approach plays into one of the reasons why amputees end up abandoning a device: disappointment. As stated by Plettenburg in 2002, ”In the domain of information and education a large discrepancy is observed between the expectations of the novice user of a prosthesis and the reality” [5]. A simple design might not lower the expectations of the user, but it might help with explaining certain design choices and thereby have a higher level of engagement with the user.

A simple design could lead to a more modular design. Modularity has two main advantages: (1) Separation of modules allows for several designers working independently on different parts of a device and (2) Parts can be interchanged between similar devices. Aside from these benefits, modularity can be especially valuable for prostheses because of two main cases: children and patients in developing countries. The grow-rate of a child means that an updated prosthesis is required every so often. A modular prosthesis could replace certain components whilst keeping other. This helps with reducing the cost of prosthesis replacement. For patients in developing countries, where access to healthcare is often limited, modularity of components can extend the durability of the full prosthesis. In case of a failure, instead of having replace the entire prosthesis, only the failed parts can be replaced, saving costs and time.

Easy to manufacture & assemble The simplicity of the design greatly influences this value. By reducing the amount of components and their complexity, the price of manufacturing and assembly decreases. This is vital for market viability in the developed world, where body-powered and electric prostheses have had decades to optimize their marketability. Furthermore, devices that are easier to manufacture and assemble could have a higher chance of being used in developing countries.

The easy manufacturing and assembly reduces the time spent in this phase of the design process, see Section 2.1. Without this, designing, building and testing a prototype within the time-limit of a thesis is not feasible.

3.5.2. Requirements

The different analyses led to the following requirements, summarized in Table 3.2. The requirements are based on an average human man, weighing 80 kilograms and having a length of 180 centimeters. Existing prostheses will be used to help hone the requirements. Regrettably, specific information was hard to find, especially since most elbow prostheses generate pronation/supination in the terminal device. A complete list of the information found for each of the thirteen most significant elbows is found in appendix D. By looking at the activities of daily living (ADL) the requirements for the ranges of motion, torques and angular velocities can be determined. The reasoning for the requirements can be found in Appendix F.

3.5.3. Wishes

The analyses led to several wishes, shown in Table 3.1.

Table 3.1: An overview of the wishes for the prosthesis

Category	**Details**
Cosmetics	Fit under clothes
	Have customize-able covers
Control	Lockable in all-positions
	Actuated locking mechanism

Table 3.2: An overview of the requirements for the prosthesis

Requirement	Value		Comment	Reasoning (Appendix E)
				Requirements based on Cosmesis
Length Forearm	< 87	mm	Goal	$L_{Forearm} = 1/3 * 0.145 * H_{User}$
	< 174	mm	Maximum	$L_{Forearm} = 2/3 * 0.145 * H_{User}$
Length Upper Arm	< 50	mm	Goal	To fit for users with elbow dis-articulations
	< 113	mm	Goal	$L_{UpperArm} = 1/3 * 0.189 * H_{User}$
	< 170	mm	Maximum	$L_{UpperArm} = 1/2 * 0.189 * H_{User}$
				Requirements based on Comfort
Weight	< 800	gr	Goal	$M_{Prosthesis} = 0.5 * (2/3 * 0.016 + 1/3 * 0.028) * M_{User}$
	< 1000	gr	Maximum	Based on other prostheses (Appendix D)
Sound Level	< 20	dB	Goal	The sound level produced by rustling leaves [16]
	< 50	dB	Maximum	The sound level produced by a conversation at home [16]
Costs	< €1000		Maximum	Thesis Budget
Cycles of Use until Power Depletion	> 300		Goal	If the elbow is used once every two minutes
	> 100		Minimum	Replace the fuel source twice per day
				Requirements based on Control
Flexion/ Extension	0-150	°	Goal	Based on most ADL [17]
	5-135	°	Minimum	Based on other prostheses (Appendix D)
Pronation/ Supination	±90	°	Goal	Based on most ADL [18]
	±50	°	Mnimum	Based on minimum ADL [19]
FE Lifting Payload	> 4	kg	Goal	According to Magermans et al. [18]
	> 1.5	kg	Minimum	According to Murray et al. [19]
FE Holding Payload	> 23	kg	Goal	The Boston Elbow & The Utah Arm (Appendix D)
	> 6	kg	Minimum	> 1.5 * Payload Goal
PS Torque	> 4.2	Nm	Goal	Based on other prostheses (Appendix D)
	> 1.5	Nm	Minimum	Based on other prostheses (Appendix D)
FE Stroke Time	< 0.7	s	Goal	Based Buckley et al. data [20]
	< 1.5	s	Maximum	Based on other prostheses (Appendix D)
PS Stroke Time	< 0.5	s	Goal	Based Buckley et al. data [20]
	< 1.0	s	Maximum	Based on other prostheses (Appendix D)

4

Conceptual Design

4.1. Module Determination

The main body of the thesis will follow the design methodology as discussed in Section 2.1. For the reader who prefers reading the complete design process per module, please see the appendix. The modules, corresponding appendix and functions are shown in Table 4.1.

Table 4.1: An overview of the modules and corresponding functions

Priority	**Actuation** Appendix H	**Locking Mechanism** Appendix I	**Frame** Appendix J	**Control** Appendix K
Must Have	Flexion Extension	-	Constrain Unwanted DoF	Generate Required Output
	Pronation Supination	-	Withstand Forces and Moments	Sense User Input
Should Have	-	Be Lockable	Protect Components	Detect Angles
	-	Have a Free-Swing Mode	House Components	Provide Feedback
	-	-	Fit to User Arm	Have a Portable Energy Supply
Could Have	-	-	Connect to Terminal Device	-
Won't Have	-	-	Have an Adjustable Fitting	Show Power Level

4.2. Actuation

One of the most fundamental decisions in the design process is the choice of type of actuation. One of the consequences of this choice is the available configuration of movement generation, especially since electric actuators mainly cause rotation and pneumatic actuators mainly cause a linear motion. The choice for pneumatic actuation is an integral part of the goal of the thesis. The reasoning for this choice is further elaborated upon in appendix C and appendix G.

4.2.1. Pneumatic Actuator

For the purposes of compactness and weight limitations, this thesis will focus on simple, linear pneumatic cylinders. These can be categorized into single-acting or double-acting cylinders. Single-acting cylinders use one gas port to allow compressed gas to enter the cylinder to move the piston to the desired position, by either pushing or pulling the piston. A spring returns the piston to a "home" position when the pressure is removed. A "push" type pushes the piston outwards when the cylinder is pressurized. An absence of pressure means the piston is in its retracted position. The "pull" type is its counterpart, where the piston is extended in the absence of pressure and will retract when the cylinder is pressurized. Double-acting cylinders have an gas port at each end and move the piston forward and back by alternating the port that receives the pressurized gas. It is possible to use a spring for a double-acting cylinder to help with either pushing or pulling.

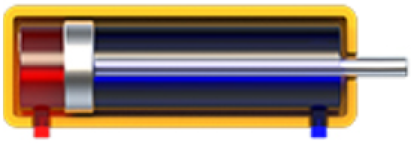


Figure 4.1: A double acting pneumatic cylinder. Image taken from [21]

Medium When using pneumatic actuation, the medium for energy storage and actuation has to be chosen. Instead of storing energy, a compressor could be used, but the low power density of such an approach would nullify the advantages of pneumatic actuation [8]. Two gasses were considered to serve as this medium: carbon dioxide (CO_2) and hydrogen-peroxide (H_2O_2). Despite the great promise shown by the use of monopropellants in the prototype by the Vanderbilt university, carbon-dioxide will be used as the medium for two reasons: (1) this prototype will mainly focus on the mechanical viability of a pneumatic prosthesis, thus energy storage plays a significantly smaller role compared to commercial products. (2) the use of monopropellants requires specialised knowledge of chemistry and pneumatic control, for both of which there is no time during this design cycle.

General Pneumatic Actuator The conceptual design has led to the design for a general pneumatic actuator as shown in Figure 4.2.

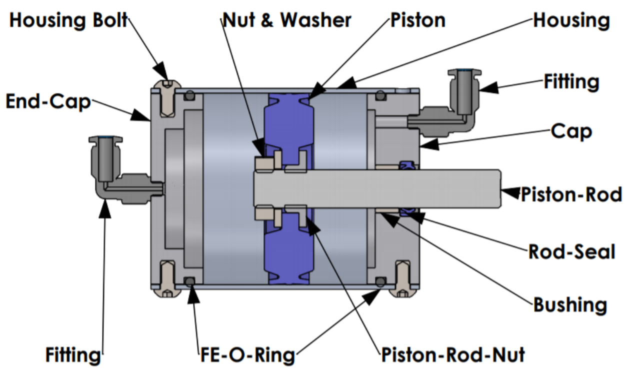


Figure 4.2: A schematic cross-section of the general actuator, with the parts labeled

4.2.2. Topology

Linear actuators will be used to generate rotary movement. An important decision that should be made regarding flexion/extension and pronation/supination is whether to place these motions in parallel or series.

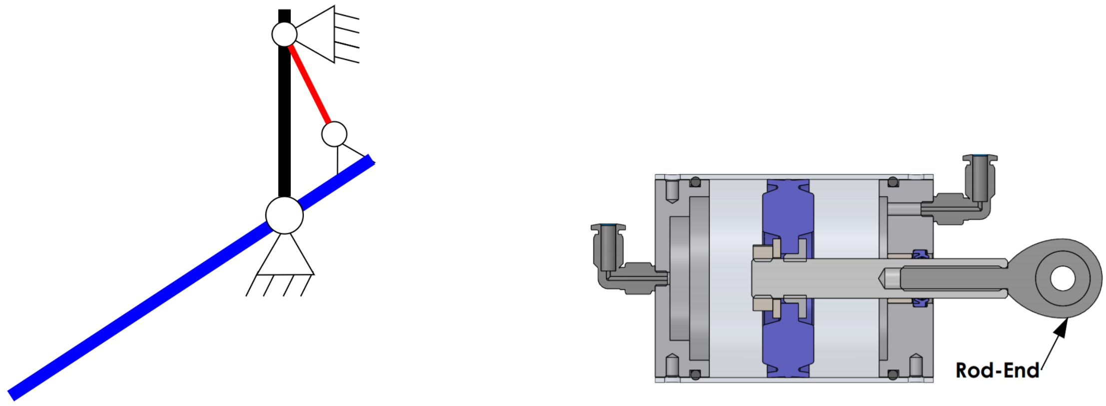
The actuators are placed in series for the following reasons (mainly based on the design values of simplicity and being easy to manufacture & assemble): (1) a series topography enables modularity, which in turn enables the prioritization of motion. A focus can be placed on generating an adequate flexion/extension before moving on to pronation/supination, (2) the ”dead-weight” factor plays less of a role due to the lightness of pneumatic actuators, and (3) the complexity of the kinematic model might cause insecurities in the control scheme.

4.2.3. From Linear Motion to Rotation

The chosen type of actuator is linear, the required movements are circular. A method of translating the generated motion is required. There is an abundance of options to use, but one of the design values is simplicity, therefore unnecessarily complex methods will not be discussed. The following methods will be seen as viable options: (1) a lever, (2) a slider-crank mechanism, and (3) a lead screw. A decision will be made separately for each of the two degrees of freedom.

4.2.4. Flexion/Extension Final Concept

The chosen configuration to achieve flexion and extension is a lever attached to the backside of the elbow prosthesis. This is illustrated in Figure 4.3.



(a) The chosen configuration for linear to rotational motion of the flexion/extension actuator

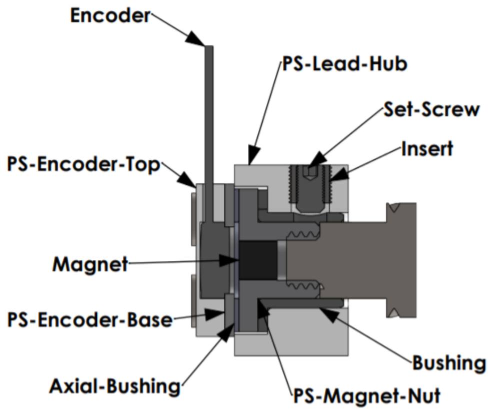
(b) A schematic cross-section of the FE-Actuator concept, with the end-effector parts labeled

Figure 4.3: The chosen configuration for flexion/extension of the elbow prosthesis

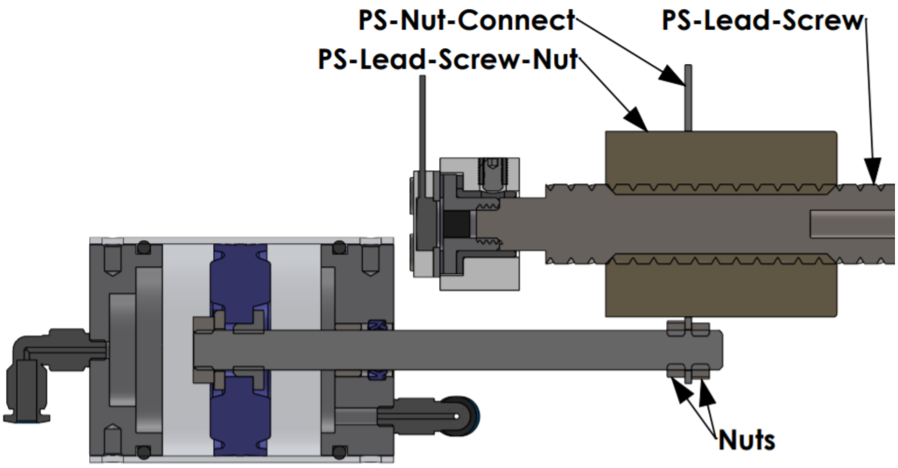
4.2.5. Pronation/Supination Final Concept

The required axis of rotation is aligned with the forearm. This combined with the lower torque requirements gives the screw-principle a good chance of succeeding for this movement. This would result in a compact and innovative design. Close attention should be paid to the configuration for proper constraints on the system, especially in the collar of the lead-screw.

(a) The chosen configuration for linear to rotational motion of the pronation/supination actuator. Image taken from [22]



(b) A detail-section of the PS-Actuator concept, focused in the collar



(c) A schematic cross-section of the PS-Actuator concept, with the end-effector parts labeled

Figure 4.4: The chosen configuration for pronation/supination of the elbow prosthesis

4.3. Locking Mechanism

The elbow prosthesis being ”lockable” will greatly improve its comfort since only movement will require an input. This improvement will be significantly better if the locking is actuated. This means that the user does not have to use their sound arm to activate or deactivate the locking mechanism. Furthermore, the usability and ”duration per charge” will be improved by being able to hold a position without constantly requiring the prosthesis to be powered, thus having the disengagement of the lock be active. The locking mechanism should be as lightweight as possible. Having a heavy and bulky locking mechanism would decrease the usability, comfort and even cosmetics of the prosthesis.

4.3.1. Be Lockable

To reduce the required force, a force amplifier can be used. An example of a force amplified brake can be seen in the book ”Werktuigkundige Systemen” by Jan C. Cool [23]. This method uses the geometry to amplify the force, illustrated in Figure 4.5. Equation (4.1) shows the torque/input ratio if no ”blocking” of the brake disc would occur. However, the system is designed in such a way that the torque generated friction force (W) works in the same directions as the torque generated by the input force (F). This will increase the normal force (N) and in turn increase the friction force. This positive feedback loop leads a high locking torque despite having a relatively low input force. The concept chosen to achieve continuous active locking is a single push-type pneumatic cylinder combined with two locking levers and a brake. This is illustrated in Figure 4.6a.

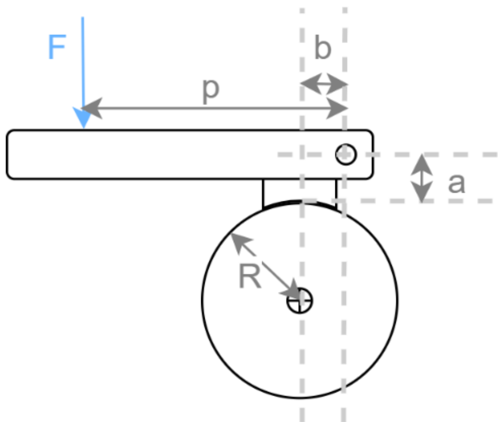


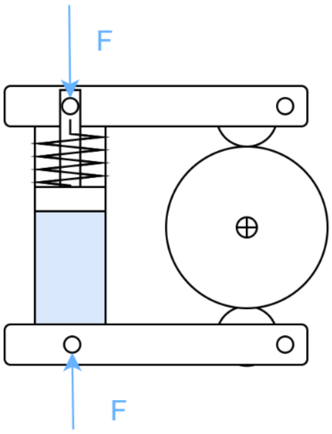
Figure 4.5: The specific geometry to achieve a force-amplified locking mechanism

$$\frac{M_w}{F_u} = \frac{pRf}{b - af} \tag{4.1}$$

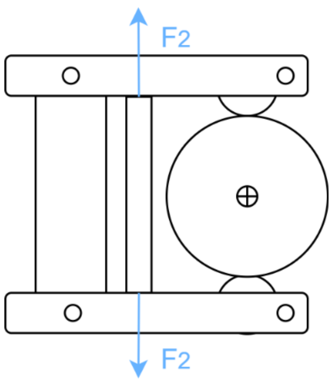
4.1: Where M_w is the friction moment on the disk [Nm], F_u is the input force [N], f is the coefficient of friction between the lever and the disk [-], R is the radius of the disk [m], a is the vertical distance between the connection point and the pivot point [m], b is the horizontal distance between the connection point and the pivot point [m], and p is the horizontal distance between the acting line of the input force F_u and the pivot point [m]

4.3.2. Free-swing Mode

Active engagement ensures that the system is always in a ”free-swing mode”, without requiring an input. Active disengagement means that without conscious thought, the prosthesis will hold its position. To achieve both, two modes of engagement will be implemented. Without any input, the locking mechanism will be engaged. Active disengagement is to be achieved by directing air low to the actuator. To achieve free-swing mode, a second mode of disengagement will be implemented. The chosen concept is using a bolt for active dis-engagement. By tightening this bolt, a pushing force, similar to that of the actuator, is generated. To activate the free-swing mode, the bolt simply has to be tightened, to deactivate this mode, loosen the bolt.



(a) The proposed concept to achieve active locking of FE: a single-acting push pneumatic cylinder



(b) The proposed secondary method of actuation to achieve active disengagement

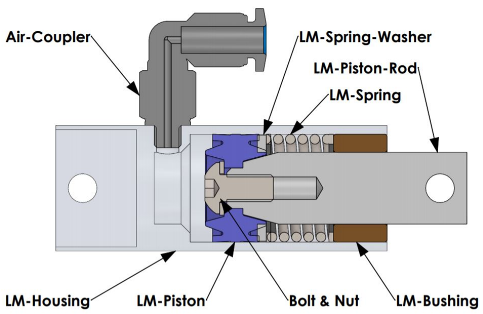
Figure 4.6: The chosen concept for the locking mechanism

4.3.3. PS-Locking

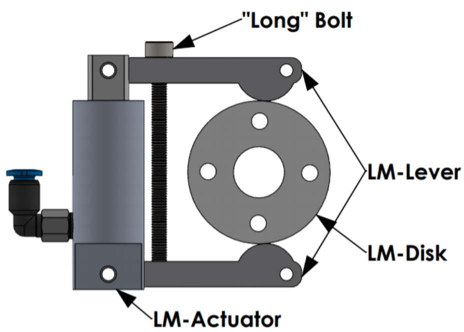
The concept discussed above was meant to be used for locking of both the flexion/extension and pronation/supination of the elbow. However, the concept is ambitious and comes with insecurities considering functionality. The decision was therefore made to only implement this for the locking of the FE motion, since this is the motion with the biggest load and the one where free-swing is more important. In order to achieve locking of the pronation/supination, a set-screw will be used in the PS-Collar will be used (see Figure 4.4b)

4.3.4. Locking Mechanism Final Concept

The actuator used for the locking mechanism is a single-acting, pneumatic ”push” cylinder. The dimensions of the actuator are determined by the available springs, pistons and required input for for locking engagement. The force generated by the cylinder, when activated, should be higher than the force generated by the spring. Furthermore, it is desired that the actuator is as short as possible to reduce the bulkiness of the mechanism. The function of the frame is to transfer the forces from the actuator to the brake disc. Figure 4.7 shows the final concepts for the locking mechanism.



(a) A cross section of the LM actuator



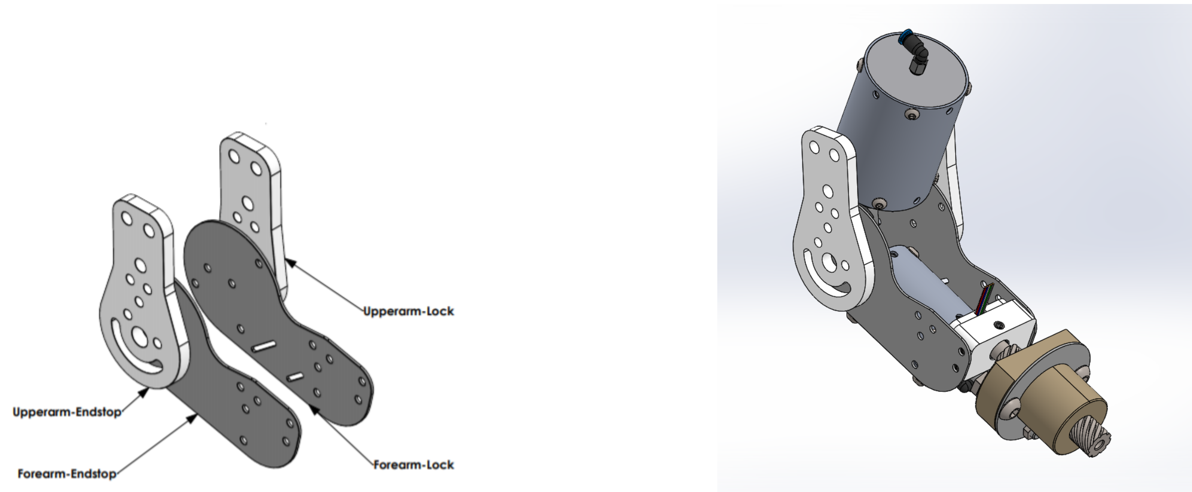
(b) The frame of the locking mechanism

Figure 4.7: The chosen concept for the locking mechanism

4.4. Frame

4.4.1. Withstand Forces and Moments

The frame consists of four load bearing structures: Two for the upper arm and two for the forearm. This design choice causes the frame to be strong, lightweight and simple to manufacture, only requiring a laser cutter.



(a) The chosen concept for the load bearing structures of the frame

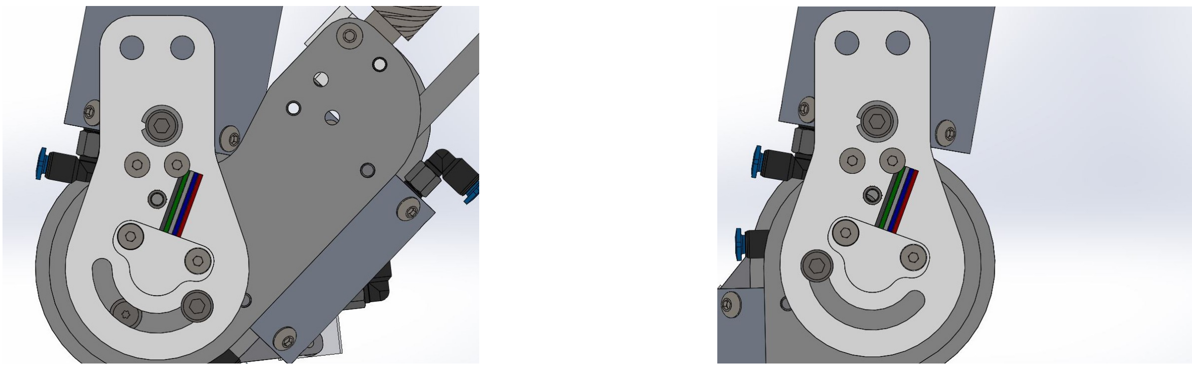
(b) The chosen concept for the load bearing structures of the frame, including the actuators

Figure 4.8: The chosen concept for the load bearing structures of the frame

4.4.2. Constrain Unwanted DoF

Bushings will be used to reduce the friction at the rotary constraints.

Endstops Despite being lightweight, the desired velocity of the prosthesis can lead to a high amount of energy in the system. If an electronic control circuit is used, the control can be programmed in such a way that impact is prevented or at least mitigated. However, this prototype will use slower, manual, control (see appendix K). To prevent the actuator from being damaged, endstops will be used, see Figure 4.9



(a) The endstop limiting the movement of the forearm at 140 [°] flexion

(b) The endstop limiting the movement of the forearm at 0 [°] flexion

Figure 4.9: The chosen concept for the FE-endstop

4.4.3. Protect Components

The main function of the covers is to protect the internal components. Because the prosthesis will not be a static device, a method to resolve the cover around the elbow joint is needed. Two concepts will be discuss, a sleeve around the elbow and a modern day version of a medieval couter (also spelled cowter), shown in Figure 4.10.

(a) An elbow sleeve. Image taken from [24]

(b) A medieval elbow couter. Image taken from [25]

Figure 4.10: A visualisation of the two proposed covers

This design choice is a choice between functionality and cosmetics. The cowter protects the elbow joint at the cost of being heavier, bulkier and more complex. This choice will be highly dependent on the user, if the prosthesis is not exposed to high external forces, the downsides of the cowter are not worth the trade-off. However, in cases where the user is expected to encounter high external forces and protection is desired over appearance, the cowter should be chosen. If an easy transition between the two types of covers is available, both types can be designed and the user can choose on a daily, or even hourly, basis which cover to use. Because of the limited time available, only one type of cover will be designed: The ”sturdy” cowters. This is to be able to test the strength of the covers. Furthermore, to understand what would make a prosthesis visually appealing, a more in-depth user analysis is required. Symmetry is desired, so that the covers can be implemented on both right and left arms. The amount of cowters was limited to two, to limit bulkiness in the design.

4.4.4. House Components

Due to a limit in available time, the decision was made to forgo this function in parts. The pneumatic and electronic circuits to control the prosthesis will be placed outside on a control board.

Actuator Implementation The chosen method of connecting the actuators to the frame is shown in Figure 4.11. The threaded holes in the actuator caps will be used for this connection, as this will prevent the need for additional parts.

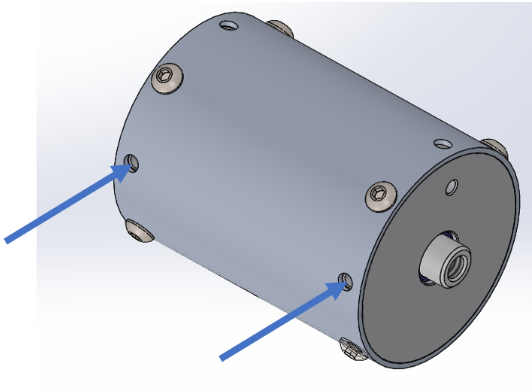
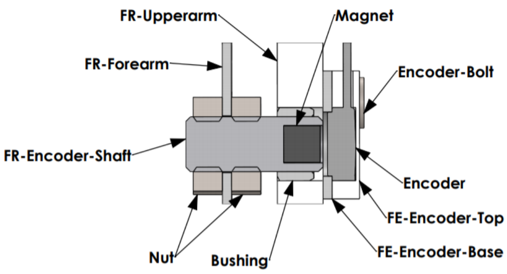
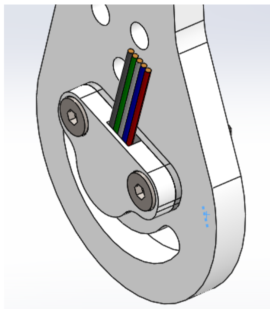


Figure 4.11: A concept of attaching the actuator to the frame using the caps of an actuator

FE Encoder Implementation As discussed in section 4.5.3, a rotary encoder is required to measure the flexion/extension angle. The chosen implementation can be seen in Figure 4.12



(a) A cross section of the FE Encoder Implementation



(b) A isotropic view of the FE Encoder Implementation

Figure 4.12: The chosen concept for the locking mechanism

4.4.5. Fit to User's Arm

Due to a limit in available time, the decision was made to forgo this function.

4.4.6. Connect to Terminal Device

Due to a limit in available time, the decision was made to forgo this function.

4.4.7. Final Concept

Prototyping Similar to the final conceptual design of the actuators, 3D-printed prototypes were used to determine the final concept, see Figure 4.13.

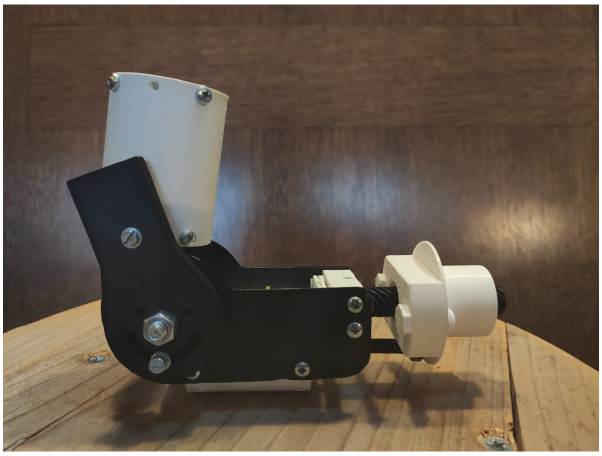


Figure 4.13: A 3D printed prototype, used for conceptual and embodiment design of the frame

Holes labelled Figure 4.14 shows the function of each of the holes of the skeleton.

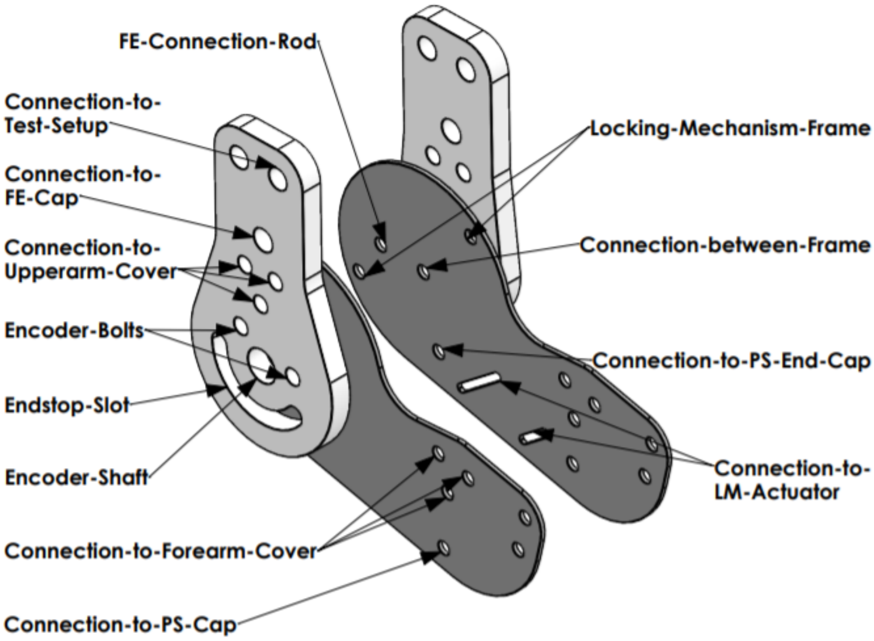


Figure 4.14: The skeleton of the frame, with the function of each hole labelled

4.5. Control

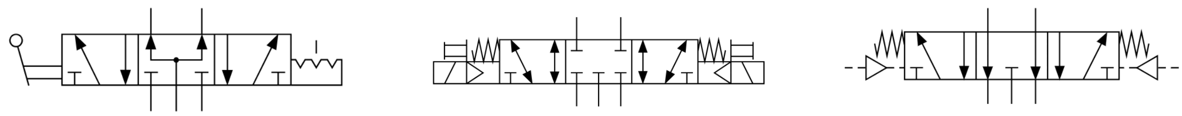
4.5.1. Sense User Input

Control of the prosthesis can be done (1) mechanically, by i.e. opening valves using a harnass, (2) manually, with the hands of the operator, or (3) electronically, using e.g. an Arduino. The goal of the designed prosthesis is to show that pneumatic actuation is a viable option. Manual control performed by the operator eliminates any insecurities about electronic signalling and data processing, meaning that only the hardware of the prosthesis is tested. As a first step in the design process, testing one aspect of the device at a time is advisable to determine future improvements. Therefore, manual valves will be the method for this prosthesis.

4.5.2. Generate Required Output

Valves Solenoid valves are classified as X/Y, where X is the amount of connections and Y is the amount of states. For the single-acting LM actuator only two states are desired: actuated and non-actuated. This cylinder requires three connections: (1) input, (2) output, and (3) exhaust. The required valve is therefore a 2/3 valve. These valves can be either normally-open, normally-closed or bi-stable. A bi-stable valve will stay in its position until further instruction. Normally-open and normally-closed valves have a home position to which they return if there are no instructions. The majority of the time, the locking mechanism should be engaged and the actuator should therefore be non-actuated. This means that the valve should be 2/3 normally-closed.

For the double-acting cylinders, three states are desired: No motion, flexion, and extension. For these two cylinders, a 5/3 valve will be used. These valves can be mid-position-pressurized, mid-position-exhausted or mid-position-closed, as shown in Figure 4.15. A mid-position-pressurized valve supplies both chambers of the actuator with pressurized gas whereas mid-position-closed valve has residual pressure in on chamber of the cylinder. This is useful for holding a position, but this prevents the free-swing mode. The holding of a position is achieved using the locking mechanism, thus the 5/3 mid-position-exhausted type is chosen to be the valve for the double-acting cylinders.



(a) 5/3 mid-position-pressurized (b) 5/3 mid-position-closed (c) 5/3 mid-position-exhausted

Figure 4.15: The different versions of a 5/3 valve

Pressure Regulation To control the force exerted by the other cylinders, the pressure should be controlled. This can be achieved by using pressure regulators or using ”bang-bang control”. Bang-bang control of pressure is achieved in valves with only a discrete open or closed state. By regulating the time the valve is opened, the pressure is regulated. Pressure regulators are a less crude method, but cause the prosthesis to become heavier and bulkier, with the lightest versions found on `https://www.festo.com/` weighing over 150 grams. Therefore, bang-bang control will be used to regulate the pressure.

Circuit The final step in the conceptual design of the pneumatic circuit is to design the circuit itself. This circuit is shown in Figure 4.16

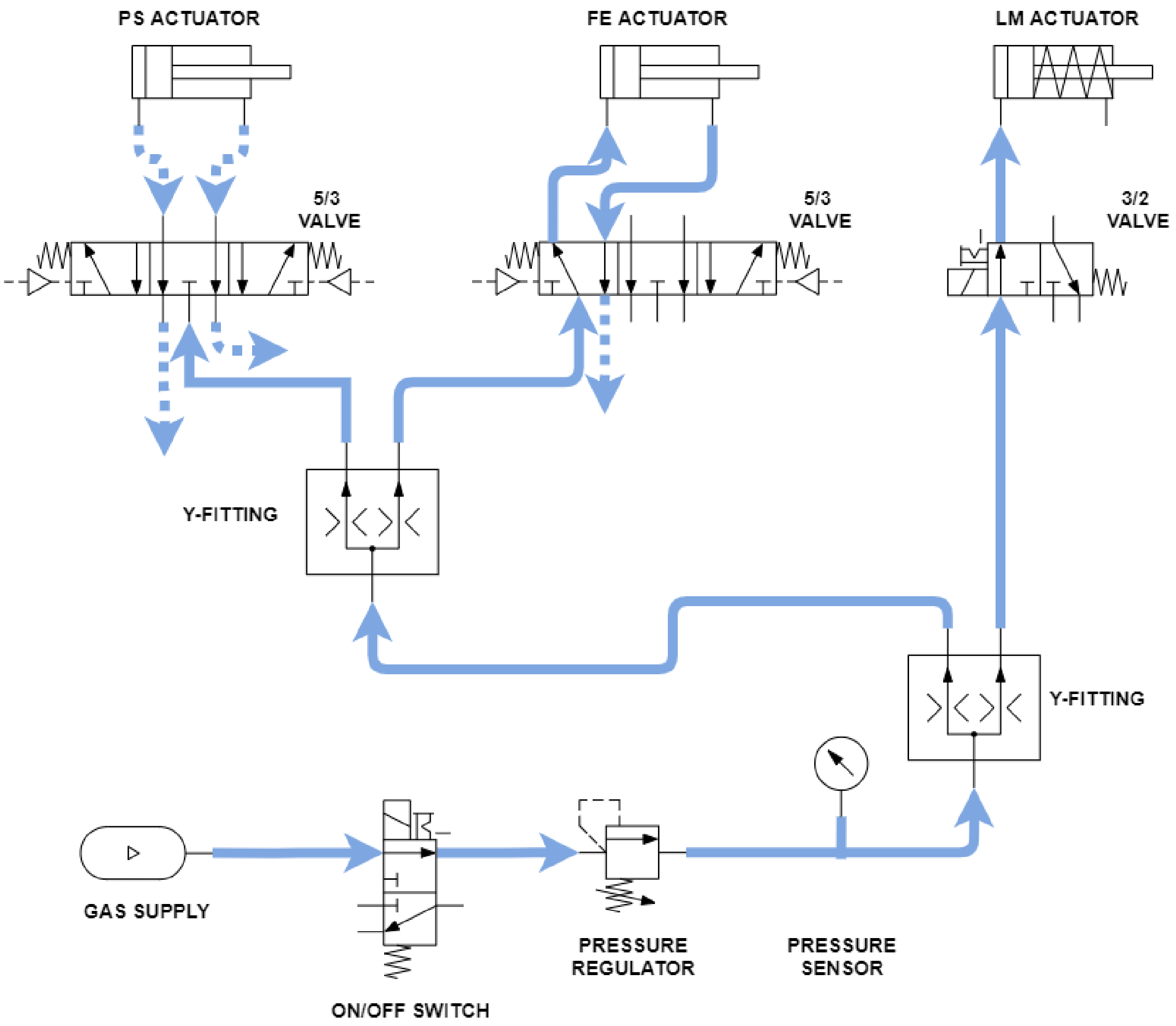


Figure 4.16: The pneumatic circuit with three valves

4.5.3. Detect Angles

Both the flexion/extension and pronation/supination angle provide valuable information required for safe control of the prosthesis. The angles can be detected visually in case of a manual control scheme. However, if the control will be done using electronics, i.e. solenoid valves, encoders should be used to determine the angle. The encoder can either be linear, placed on the actuator to determine the piston position, since each position has an unique corresponding angle. A different option is using a rotational encoder, to be placed directly on the axis-of-rotation. These are much smaller and lighter than linear encoders, but their implementation can be more difficult. With weight optimization being one of the design wishes, the rotational encoders are preferred.

4.5.4. Have a Portable Energy Supply

Due to a limit in available time, the decision was made to forgo this function.

4.5.5. Provide Feedback

Due to a limit in available time, the decision was made to forgo this function.

5
Embodiment

In the embodiment phase, the details of the design are finalized. A complete list of all required parts and renders of the custom parts can be found in the appendix (Appendix L and Appendix M respectively).

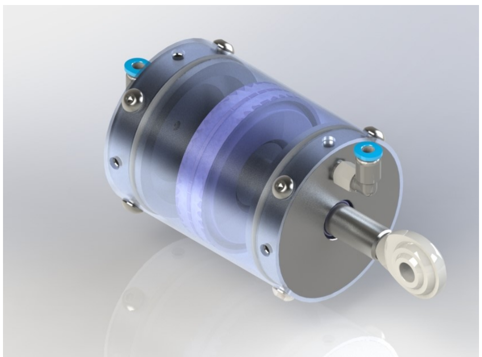
5.1. Actuation

Working Pressure The working pressure can not be increased indefinitely, as there are three main limiting factors: (1) cylinder thickness, (2) allowed pressure for off-the-shelf components, and (3) gas consumption. It was advised to use a working pressure of 12 [bar] for optimal gas efficiency [26]. This is in-line with the maximum pressure of the off-the-shelf components and not limited by the cylinder thickness. Therefore, 12 [bar] will be used as the working pressure for the system.

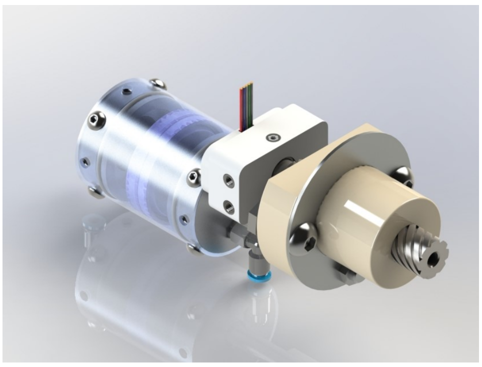
Actuator Parameters Table 5.1 shows the final parameters determined in the embodied design and Figure 5.1 shows renders of these actuators.

Table 5.1: An overview of the embodiment design of the FE-, and PS-actuator

Actuator	Flexion/Extension	Pronation/Supination
Piston Diameter	⌀ 40 [mm]	⌀ 32 [mm]
Actuator Force	1507 [N]	965 [N]
Stroke Length	26 [mm]	16 [mm]
Output	±4 [kg]	3.0 [Nm]
End-effector	Rod-end	Lead-screw



(a) A render of the embodied FE actuator



(b) A render of the embodied PS actuator

Figure 5.1: Renders of the embodied actuators

5.1.1. Flexion/Extension

For the embodiment, MATLAB was used to determine parameters, such as the stroke length. The final parameters were determined in an iterative process, using the tools and plots shown in Figure 5.2. The used script can be seen at Appendix H

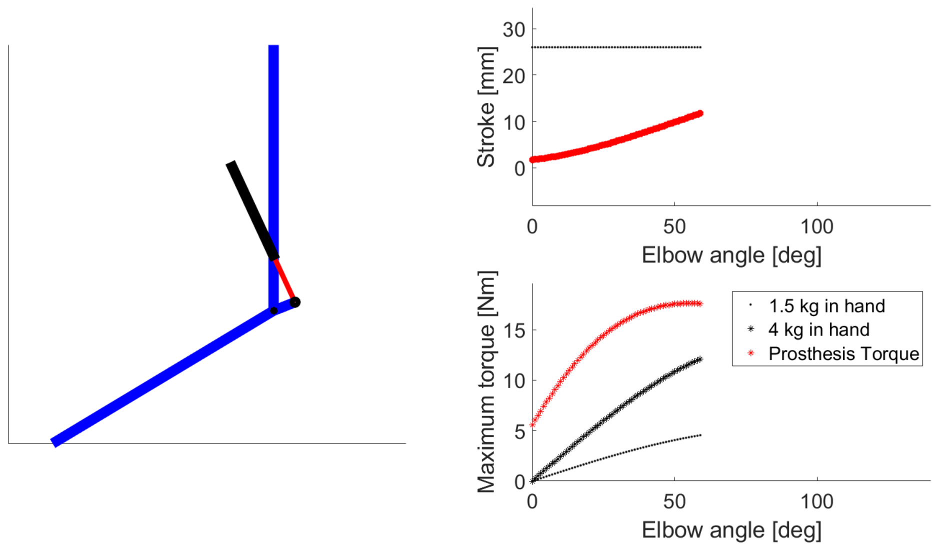


Figure 5.2: The plots used to determine the parameters of the flexion/extension actuator an configuration

Pistons can be bought off-the-shelf at `https://eriks.nl/nl/`. The most relevant diameters are ⌀ 32 [mm], ⌀ 40 [mm], and ⌀ 50 [mm]. The corresponding maximum torques are dependent on the topology of the prototype. An example is shown in Figure 5.3, which shows that a 32 [mm] actuator would be too weak and a 50 [mm] actuator would be overpowered, and thus cause the prosthesis to become unnecessarily heavy and bulky. The ⌀ 40 [mm] piston will be used, as the peak torque requirement and delivered are of similar values.

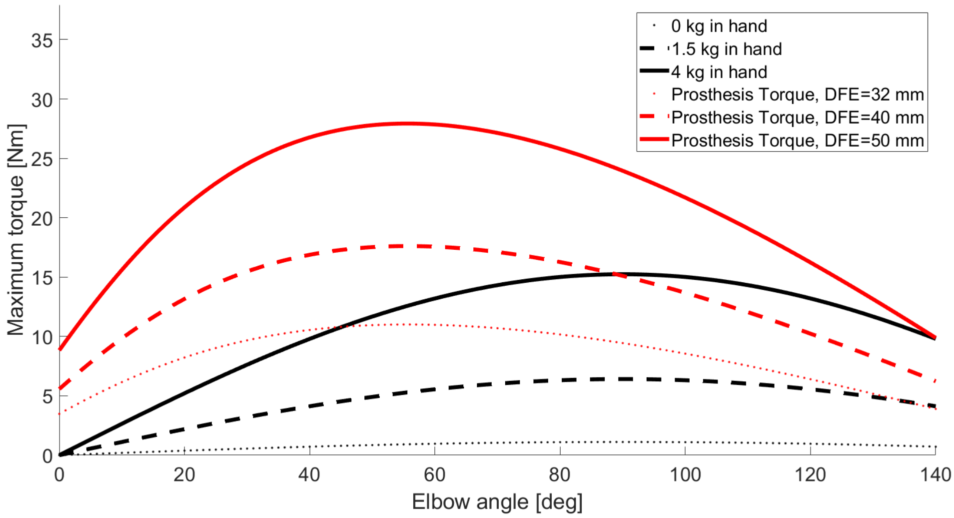
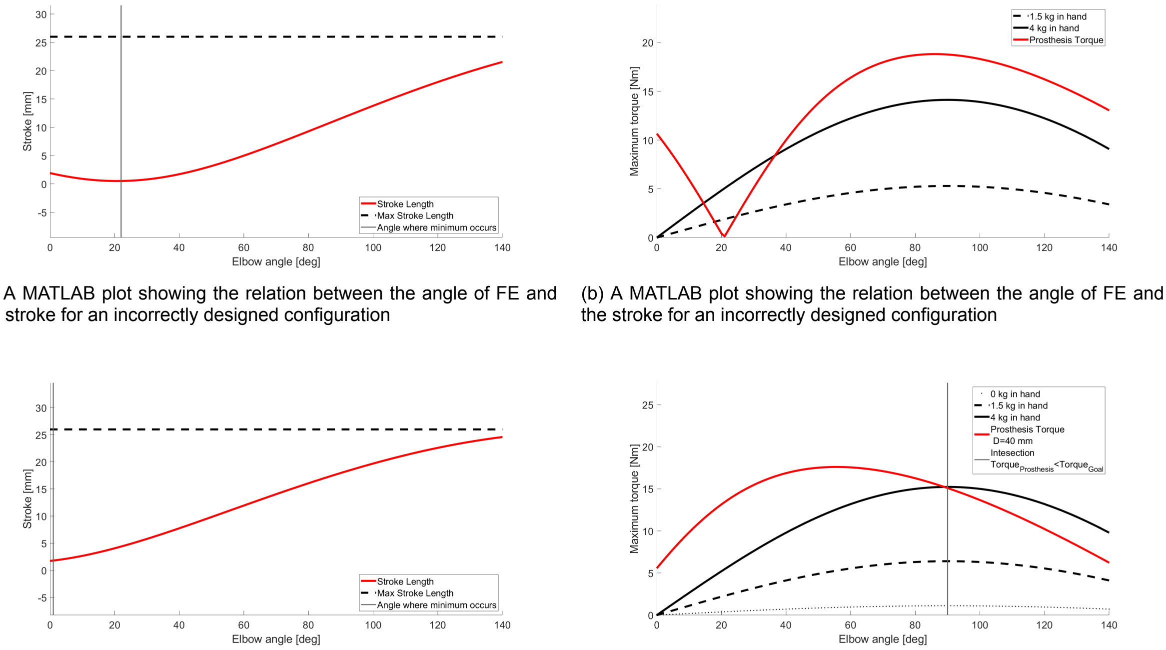


Figure 5.3: The torques achieved with different piston diameters and the required torques shown dependent on the angle of flexion

The goal is overlap the maximum of the prosthesis torque and desired torque as much as possible to achieve the most efficient actuation. A perfect overlap is not possible due to geometrical limitations. This is illustrated in Figure 5.4, where a close overlap of maximum torques comes at the cost of significantly reduced torques in other regions of the RoM (Figure 5.4b). Furthermore, the stroke length has a minimum. This is a singularity within the system, where an extension of the actuator can go either way and therefore the system is not fully deterministic, see Figure 5.4a. The optimal configuration has been determined iteratively, and is shown in Figure 5.4c and Figure 5.4d. The torque is translated back into payload and shown in Figure 5.5. It should be noted that in this configuration, the desired payload of 4 [kg] can not be lifted over the entire RoM. This is the trade-off between functionality and comfort/cosmetics.



(a) A MATLAB plot showing the relation between the angle of FE and the stroke for an incorrectly designed configuration

(b) A MATLAB plot showing the relation between the angle of FE and the stroke for an incorrectly designed configuration

(c) A MATLAB plot showing the relation between the angle of FE and the stroke for a correctly designed configuration

(d) A MATLAB plot showing the relation between the angle of FE and the torques for a correctly designed configuration

Figure 5.4: The plots used to determine the parameters of the flexion/extension actuator an configuration

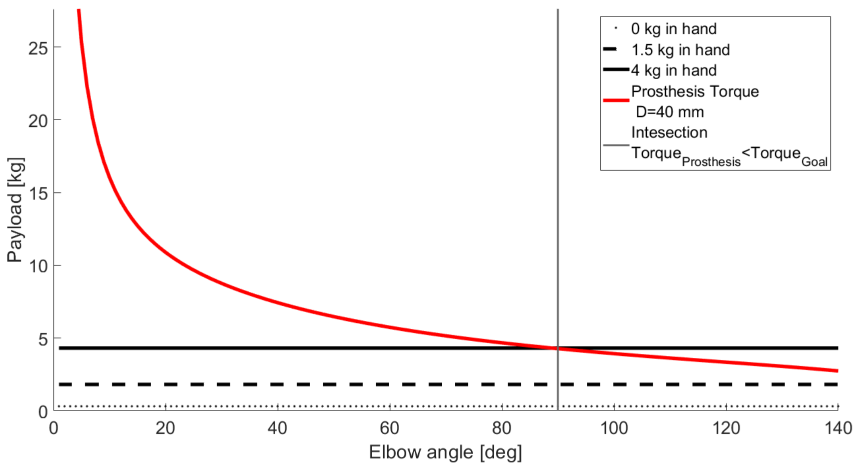


Figure 5.5: A MATLAB plot showing the relation between the angle of FE payload

5.1.2. Pronation/Supination

The selection lead-screw and actuator size is a combined effort. The lead-screw will be bought off-the-shelf https://www.igus.nl/. The input is a force and the output is a torque, for a lead-screw this means the torque is generated is the back-drive torque. Therefore, the PS torque can be calculated using Equation (5.1) and Equation (5.2), and the required lead-screw length can be calulated using Equation (5.3). For the calculations, the following parameters are used $\mu = 0.12$, $L_{Buffer} = 4[mm]$, and $L_{Collar} = 10[mm]$.

Appendix H shows sixteen options. 3.0 [Nm] was decided upon as the designed output torque, as it can be achieved with a relatively light and small combination of lead-screw and actuator: **12x25 [mm], ⌀ 32 [mm]**.

$$T_{PS} = \frac{F * p * \eta}{2\pi} \tag{5.1}$$

5.1: The equation used to determine the PS torque. T_{PS} is the torque of pronation/supination [Nm], F is the actuator force [N], p is the lead-screw lead [m], and η is the efficiency [-]

$$\eta = \frac{\tan\lambda}{\tan(\lambda + \phi)} \tag{5.2}$$

5.2: The equation used to determine the efficiency of a lead-screw. η is the efficiency [-], λ is the lead angle [°] ,and ϕ is the friction angle , equal to $\arctan(\mu)$[°]

$$L_{Lead-screw} = \frac{p}{2} + 2 * L_{Buffer} + L_{Collar} + L_{\text{Nut}} \tag{5.3}$$

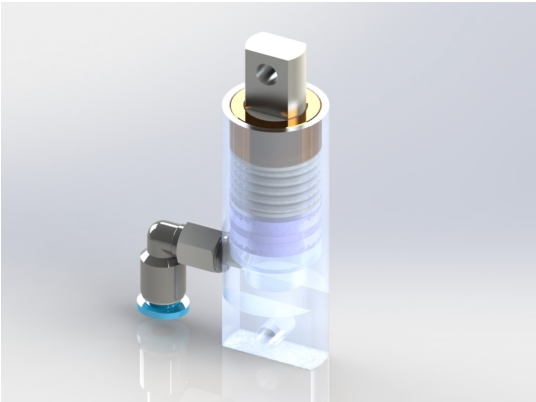
5.3: The equation used to determine the length of the lead-screw. $L_{\text{Lead-screw}}$ is the length of the lead-screw [mm], p is the lead-screw lead [mm], L_{Buffer} is the minimum distance between the end of the lead-screw and the nut [mm] , L_{Collar} is the length of the lead-screw used for the suspension within the collar [mm], and L_{Nut} is the length of the nut [mm]

Table 5.2: The parameters for two of the combinations of lead-screw and piston diameter

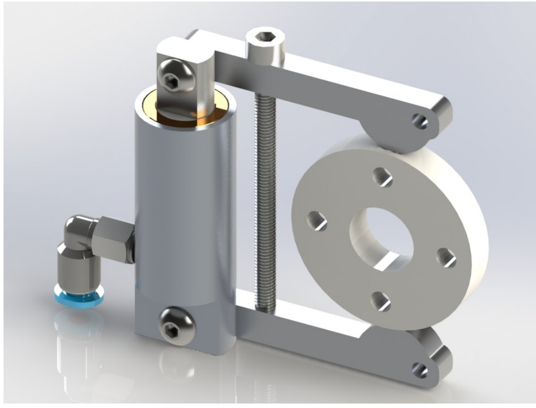
					Piston Diameter		
					⌀ 25 [mm]	**⌀ 32 [mm]**	**⌀ 40 [mm]**
Diameter x Pitch	λ	η	**Length**	**Mass**	**Torque**		
12x25 [mm]	33.6 [°]	77.9 [%]	65.5 [mm]	97.5 [g]	1.8 [Nm]	3.0 [Nm]	Lead-screw nut too weak
16x35 [mm]	34.8 [°]	78.2 [%]	70.5 [mm]	146.7 [g]	2.6 [Nm]	4.2 [Nm]	6.6 [Nm]

5.1.3. Locking Mechanism

The selected concept consists out of two sub-modules: The actuator and the frame. The goal of the actuator is not to generate movement, but to cancel the spring force. The actuator of the locking mechanism is single-acting pneumatic push cylinder. The primary function of the frame is to transfer the input spring force from the actuator to the brake disc. A secondary function is to transfer the forces of the free-swing mode disengagement method.



(a) A render of the embodied LM actuator



(b) A render of the embodied LM Frame

Figure 5.6: Renders of the embodied locking mechanism

LM Actuator The dimensions of the actuator are determined based the available springs, pistons and required input for for locking engagement. The force generated by the cylinder, when activated, should be higher than the force generated by the spring. Furthermore, it is desired that the actuator is as short as possible to reduce the bulkiness of the mechanism. The working pressure of the LM actuator will be the source pressure: 12 [bar], to eliminate the need for pressure regulation.

Pistons can be bought off-the-shelf at `https://eriks.nl/nl/`. The smallest available diameters are 8, 10, 12, and 16 [mm]. In an ideal world, the topology of the system can amplify an input (spring) force of 1 [N] to generate a locking moment of ∞. However, since there are losses and inefficiencies, a minimum spring force of 50 [N] is desired. The spring is selected from the catalogue of `https://www.tevema.com/nl/`. The selection of piston and spring lead to the combination shown in Table 5.3

Table 5.3: An overview of the embodiment design of the LM-actuator

Piston Diameter	⌀ 12 [mm]
Actuator Force	135 [N]
Spring Force	87 [N]
Stroke Length	2 [mm]
End-effector	Levers

LM Frame The topology of the lever is constrained by Equation (4.1), the size of the actuator, available space on the frame of the prosthesis, and the radius of the disc. The material used for the locking lever will be aluminum, chosen for its strength, low density, and availability to laser-cut. The embodiment design led to the parameters shown in Table 5.4.

Table 5.4: The embodied parameters of Equation (4.1)

Parameter in eq. (4.1)	p	R	a	b	b/a
Value	35 [mm]	14 [mm]	6 [mm]	4 [mm]	0.67

These parameters are used to determine the ratio of normal force/input force: 17.5 [-], thus the normal force on the disc is ≈1500 [N]. The locking torque without positive feedback is calculated around at 21 [Nm]. If the coefficient of frictionis higher than b/a (= 0.67) in theory, $M/f \rightarrow \infty$. The limiting factor is the generated normal force (N) on the brake disc. The maximum torque generated by the flexion/extesion actuator is 18 [Nm]. To achieve a locking moment of locking payload of $1.5 * P_{Lift}$, the locking mechanism should hold a torque of 27 [Nm]. To determine the required width of the brake disc, a force analysis was performed, see Equation (5.4). A material with a high coefficient of friction and one that can be easily manufactured in the desired shape by 3D-printing is: TPU (Thermoplastic polyurethane)

$$L = \frac{(\frac{T}{R*f}) * E^*}{\sigma_e^2 * \pi * R^*} \tag{5.4}$$

5.4: The equation used to determine the maximum force based on contact stress of two parallel cylinders.L is the width of the connection [m], T is the required holding moment [Nm], R is the radius of the disc [m], and f is the coefficient of friction [-], E^* is the equivalent module of elasticity [Pa] σ_e is the material compressive strength, and R^* is the equivalent radius [m]. Based on the equation from [27]

Material	f	σ_e	E^*	R^*	L
TPU[1]	1	66 [MPa]	0.32 [GPa][2]	10 [mm][3]	4.5 [mm]

1. Material properties based on ”TPU (Ether, aliphatic, Shore D60)”
2. Based on contact with aluminum. E_{TPU} = 0.25 [GPa] & ν_{TPU} = 0.46
3. The lever radius is set at 35 [mm]

Holding Payload The lifting payload and holding payload are shown in Figure 5.7. This figure shows that the goal of locking the arm whilst holding 23 [kg] is extremely high compared to the regular payload distribution. The maximum payload that can be help in any position is 7.6 [kg]

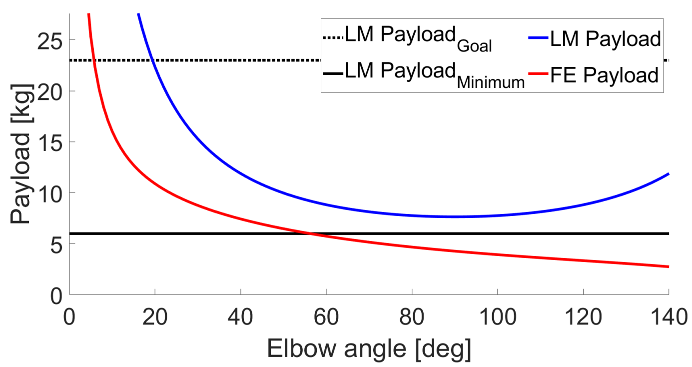


Figure 5.7: A MATLAB plot showing the relation between the angle, the LM payload and the FE payload

5.2. Frame

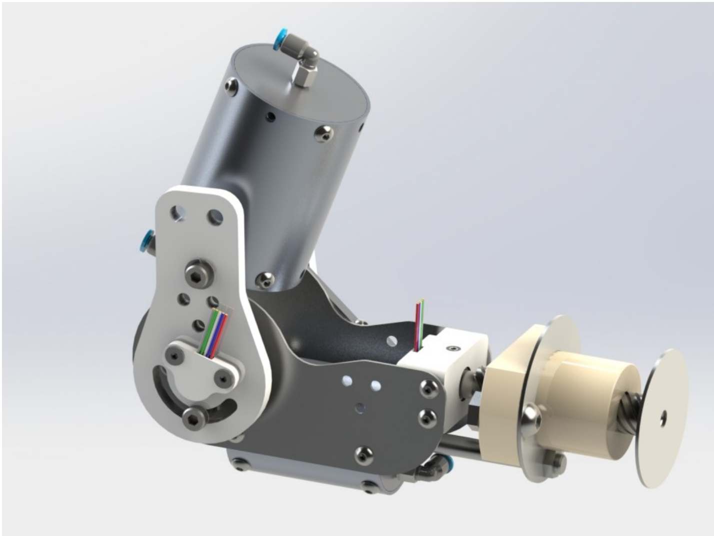


Figure 5.8: The prosthesis without covers

5.2.1. Connections

"Regular Connections" All bolts used in the system are M3, which is the smallest bolt to easily work with. Equation (5.5) will be used to determine the minimal required bushing length.

$$L = S * \frac{F}{P * n * d} \tag{5.5}$$

5.5: The equation to calculate the minimal bushing length (L) [mm], S is a safety factor [-], F is the radial force on the holes [N], P is the permissible load [MPa], n is the amount of holes the force is divided over, and d is the hole diameter [mm]

Encoder Shaft The encoder shaft has to house the encoder magnet, which is 4 [mm] in diameter. A diameter of 6 [mm] is chosen for this shaft. Manufacturing is easier if the part can be made with the same diameter all over, therefore, the threaded connection will be M6. Furthermore, to prevent interference with the magnet, the encoder shaft is made from aluminum.

5.2.2. Sheet Material

The material selection for the sheets is based on the forces on the holes, whilst being as lightweight as possible. The stress on the material is approximated using Equation (5.6). The stresses found are compared to the yield stresses of three selected materials: Stainless steel (304), Aluminum (6061) and POM.

$$\sigma_{min} = S * \frac{F}{n * d * t} \tag{5.6}$$

5.6: The equation to calculate the stress (σ_{min}) [MPa], S is a safety factor [-], F is the radial force on the holes [N], n is the amount of holes the force is divided over, d is the hole diameter [mm], and t is the sheet thickness [mm]

$$m = A * t * \rho \tag{5.7}$$

5.7: The equation to calculate the mass on the sheets (m) [g], A is the surface area of the sheet [mm^2], t is the sheet thickness [mm], and rho is the material density [g/mm^3]

Forearm The highest force on these sheets occurs at the connection-rod to the FE actuator. Everything connected to the forearm sheets can be a rigid connection, no bushing is required.

Upperarm The highest force on the upperarm sheets occurs on the connections to the FE cap and the connection between the two sets of sheets, both 1500 [N]. Because of the implemented bushings, the minimal thickness is equal to the bushing length: 4 [mm].

Table 5.5: The material and thickness selection for the frame sheets

	Material	Thickness	Area	Mass
Forearm	Steel	1 [mm]	≈3200 [mm^2]	≈26 [g]
Upperarm	PMMA	4 [mm]	≈1800 [mm^2]	≈8.6 [g]

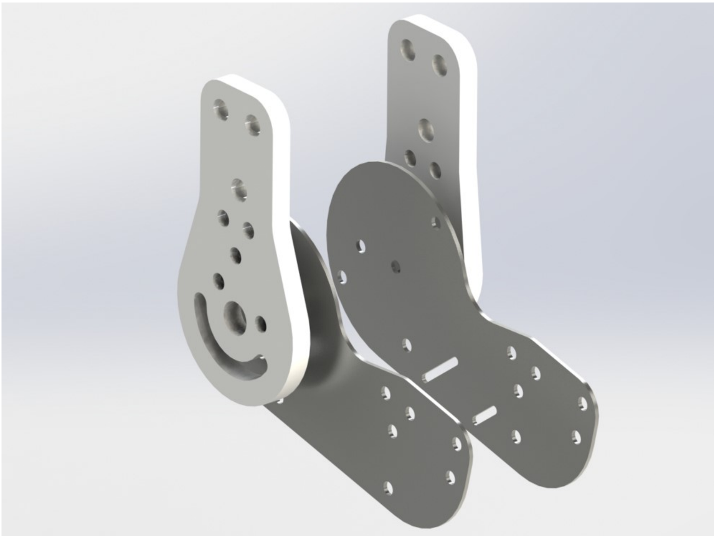


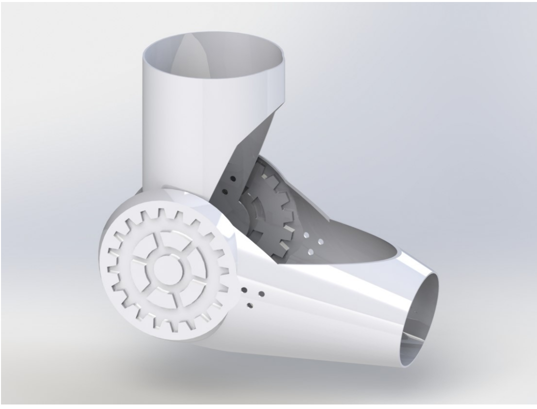
Figure 5.9: A render of the four sheets, which will function as the main load bearing structures of the prototype

5.2.3. Covers

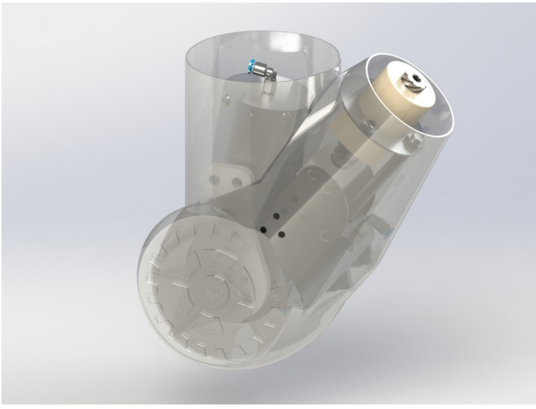
To facilitate easy manufacturing of the covers, they will be 3D-printed. The chosen material is ABS (Acrylonitrile Butadiene Styrene), as this material can be post-processed using acetone vapors. These vapors will settle on the outer layer of the covers and "melt" the ABS. If handled correctly, this can be used to create an appearance close to that achievable by ejection molding, see Figure 5.10.

Figure 5.10: The effects of an acetone treatment on ABS. On the left: Untreated, on the right: Treated with acetone

Figure 5.11a shows a render of the covers. It can be noted that the inside of the elbow is open. This is needed to provide enough space for the actuator at 140 [°] flexion, shown in Figure 5.11b. This could have been solved by either redesigning the topology so that the space is not needed, or by designing the covers in such a way that the they are "closed" on the top side, by following the required contour. The first solution should be avoided because it will change the FE angle-torque relation. The second possible solution was rejected as it limits the accessibility of the components whilst not improving the appearance greatly. Since accessibility is a design value, and especially important for a prototype, the covers will be kept "open". An elastic fabric can be used to cover the holes and keep out dust and other debris.



(a) A render of the designed covers



(b) A render illustrating why the inside of the elbow covers are "open"

Figure 5.11: Renders of the designed covers

The covers have been designed in such a way that they can be easily interchanged. Furthermore, an image can be implemented within the covers to allow for personalising the prosthesis. For the prototype, a gear has been engraved to symbolize mechanical engineering.

The covers have become bulkier that expected, causing the entire prototype to become significantly bulkier. The reason for this is twofold. First of all, the positioning of the locking mechanism leads to a larger required forearm cover, which is further increased because of the desire for a symmetrical cover. A second reason for the increased bulkiness is safety: a moving assembly always has the potential to cause harm. In the case of the covers, if a finger gets stuck whilst the prosthesis is moving, the high forces can cause serious injuries. The final covers are shown in Figure 5.12.

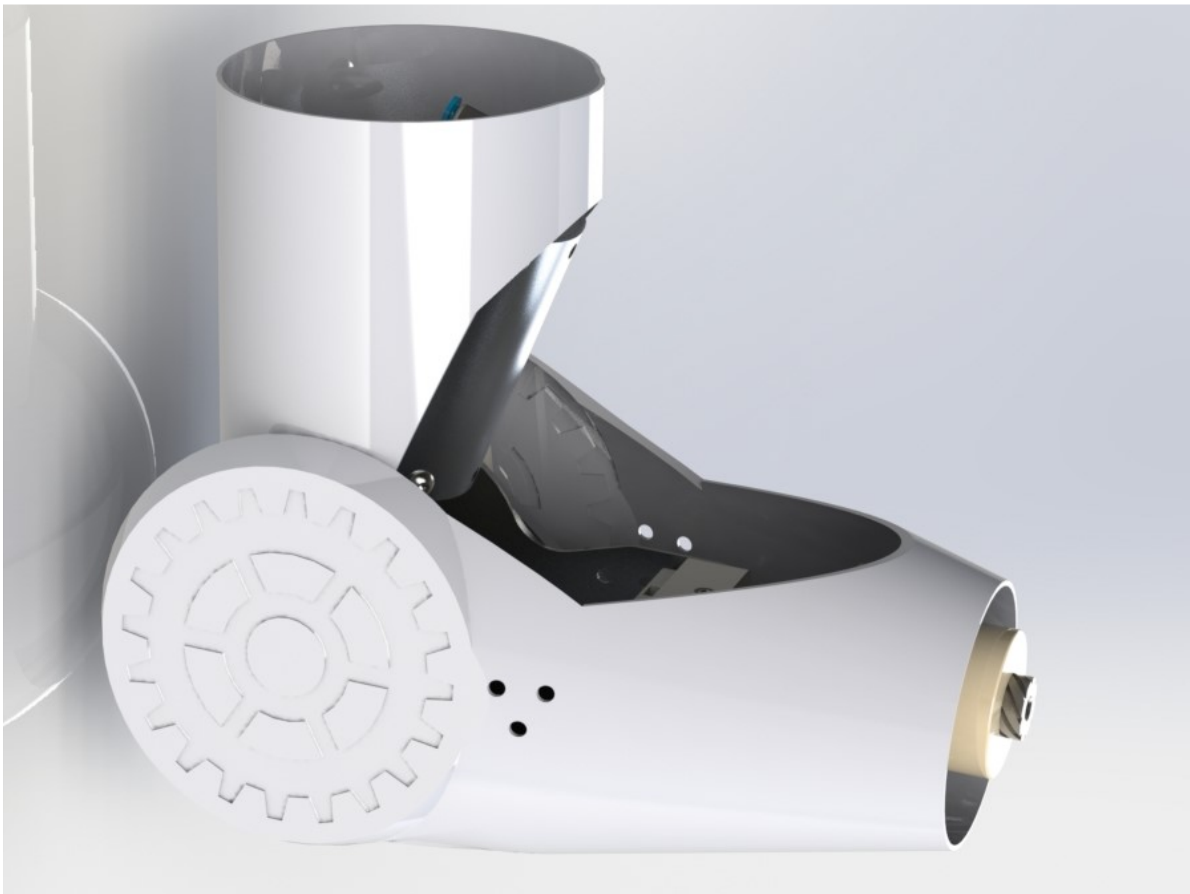


Figure 5.12: A render of the designed covers, implemented on the prototype

5.3. Control

5.3.1. Circuit Embodiment Design

Valve Selection The product finder from https://www.festo.com/ was used to find the valves. The 3/2 valve for the prototype differs slightly from the concept, instead on a mono-stable normally-closed, valve a bi-stable valve will be used. This is done to still be able to test the free-swing mode even if the secondary actuation method of the locking mechanism does not work. The selected valves are shown in Table 5.6.

Table 5.6: An overview of the available valves

Type	Control	Weight	Flow Rate	Switching Frequency	Code
5/3	Selector, Sideways	235 [g]	530 [L/min]	0.5 [Hz]	VHEF-ES-P53E-M-G18
3/2	Finger Lever	156 [g]	750 [L/min]	0.5 [Hz]	VHEF-LT-M32-M-G18

Accessories Several accessories can be used for a valve. First of all, fittings transfer the gas from the valve to the tube or the other way around. Secondly, silencers can be used to dampen the noise from the exhaust outlets on the valve. The FESTO product finder was used to find the required products. The fittings requirements and choices are shown in Section 5.3.1 and the silences is shown in Section 5.3.1.

Table 5.7: The required fittings

Input	Output	Design	Used for	Amount	Weight p.u.	Code
G-1/8	⌀ 4 mm	Straight	Valves	8	9.1 [g]	QS-G1/8-4
M3	⌀ 4 mm	L-Shape	Actuators	5	3 [g]	QSML-M3-4
⌀ 4 mm	2x ⌀ 4 mm	Y-Shape	Circuit	2	3.5 [g]	QSMY-4

Table 5.8: The two available compatible exhausts to the chosen valves

Silencer	Material	Noise level[1]	Q	Weight	Max. Pressure
U-1/8	POM	<74 [dB]	2000 [L/min]	2.3 [g]	10 [bar]

1. Measured at 6 [bar] with respect to atmosphere at a distance of 1 [m]

5.3.2. Detect Angles

RLS sells extremely lightweight, small and robust rotatory encoders, perfectly suitable for the required implementation, available on https://www.rls.si/eng/. The chosen encoder and corresponding magnet are shown in Table 5.9. However, at this point in the design process, it was decided that the control will be done manually and therefore the electronic detection of the angles is no longer required. To reduce the costs of the prototype, it was decided to forgo the encoders. Future implementation would require no design changes.

Table 5.9: The chosen encoder and magnet

Component	Weight	Size	Resolution
Encoder: RM08	2.0 [g]	⌀ 8 [mm]	±0.3
Magnet	0.4 [g]	⌀ 4 [mm] x 4 [mm]	-

5.3.3. Gas Consumption

The gas consumption depends on the intensity of the activity and the stroke length of the actuators. The consumption will be estimated using the ideal gas law (Equation (5.8)). The amount of gas used for a minimal intensity (minimal force, full RoM) cycle is 0.12 [g]. For a maximum performance cycle this is 0.95 [g]. If one out of twenty movements requires full power and with 25 [%] losses, the expected average gas consumption is 0.20 [g] for a full RoM movement. Thus 20 [g] CO_2 is required to achieve the minimum of 100 cycles. The goal of 300 cycles can be reached with a gas supply of 60 [g]. In reality, the gas consumption depends heavily on the type of use. Some of the most easily accessible gas cartridges are those used for cycling. The largest of these cartridges contains 25 [g] CO_2 and has a threaded nozzle for easy implementation. Per 25 [g] cartridge, an average of 125 use cycles can be achieved.

$$m = M * \frac{p * V}{R * T} \tag{5.8}$$

5.8: The molar form of the ideal gas law. m is the gas used [g], M is the molar mass [g/mole], p is the pressure [Pa], V is the volume [m^3], R is the universal gas constant: 8.314 [J/K*mole] and T is the temperature [K]

Table 5.10: An estimation of the gas used for flexion/extension and pronation/supination. Used gas: CO_2, Molar mass 44 [g/mole], Pressure 1.2 [MPa], Temperature 293 [K], Universal Gas Constant 8.314 [J/K*mole]

Motion	Flexion/Extension	Pronation/Supination
Stroke Volume	31 [ml]	10 [ml]
Minimal Pressure to Achieve Movement		
Required Pressure	0.15 [MPa]	0.15 [MPa]
Gas Used	0.09 [g]	0.03 [g]
Maximum Pressure for Maximum Torque Output		
Required Pressure	1.2 [MPa]	1.2 [MPa]
Gas Used	0.72 [g]	0.23 [g]

5.3.4. Stroke Time

The stroke time of the actuators can be estimated using the flow rates of the valves and supply. The flow rate of valves is extremely high compared to the relatively small volume of the actuators. The flow rate of the gas supply system to be used in the test set-up is unknown. However, if CO_2 cartridges are used, an estimation can be made. When opened and left by itself, a 12 gram CO_2 cartridge will empty in about 3 seconds. The mass flow rate is thus ≈ 4 [g/s]. Aside from the flow rates, there are delays in the system such as activation times of the valves. These are unknown parameters in the embodiment design and should be tested in the next phase. However, for the moment, the total delay is estimated at 0.4 [s]. The expected stroke times are shown in Table 5.11.

Table 5.11: An estimation of the stroke times for flexion/extension and pronation/supination. Used gas: CO_2, Molar mass 44 [g/mole], Pressure 1.2 [MPa], Temperature 293 [K], Universal Gas Constant 8.314 [J/K*mole]

Motion	Volume	Gas Used	Stroke Time	"Delayed" Stroke Time
Flexion/Extension	31 [ml]	0.67 [g]	0.17 [s]	0.57 [s]
Pronation/Supination	10 [ml]	0.21 [g]	0.05 [s]	0.45 [s]

5.3.5. Final Circuit

The final circuit is shown in Figure 5.13.

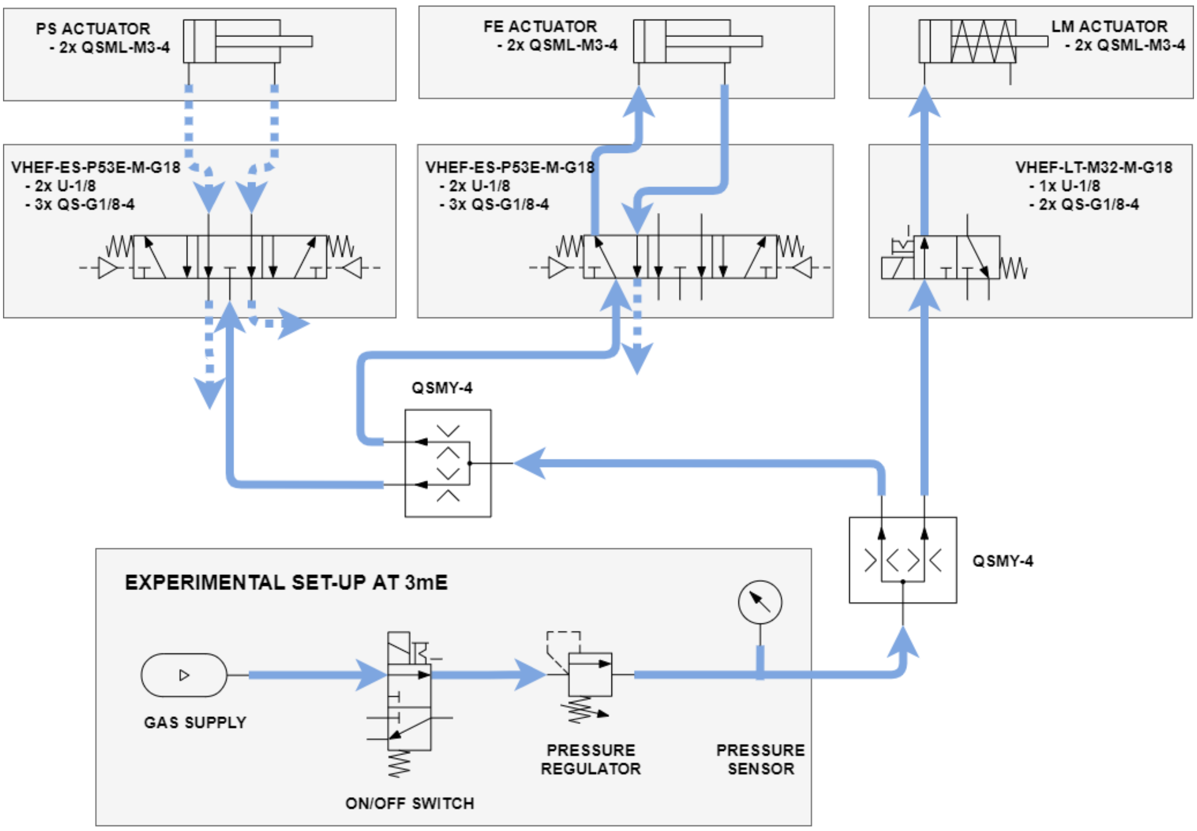


Figure 5.13: The final designed pneumatic circuit

5.4. Embodiment Design Evaluation

5.4.1. "Mistakes" in the Design

When evaluating the design, several errors came to light. However, the design was already in production at this point, thus these could no longer be changed. Possible solutions to these "mistakes" are shown in Section 8.2.4 and details are shown in Appendix R.

PS Lead Screw After revising the lead-screw selection, some mistakes things were noted. The equation to calculate did not include the friction. With friction included the output PS torque would only be 1.8 [Nm].

LM Brake Disc Whilst evaluating the brake disc, it was discovered that an assumption made in Equation (5.4) was wrong. In the equation, 'R^*' is used to determine the area the force is acting upon. This would only be true if the indentation in the material is unlimited. This is not the case, as the stroke of the LM-actuator is limited. Where a high p/b ratio is very useful for generating a higher force on the brake disc, it limits the indentation and therefore the surface area the force is acting upon. With the parameters shown in Table 5.4 and an actuator stroke of 2 [mm], the indentation depth can be calculated to be 0.22 [mm]. Equation (5.9) can be used to determine the maximum normal force on the brake disc, which is 249 [N]. Any higher forces will not act on the brake disc but act on the cylinder. The maximum torque the brake disc can hold is a mere 2.5 [Nm], higher torques have a chance to slip. The maximum payload that can be help in any position is 0.7 [kg], significantly lower than the desired value or even the lifting torque.

$$F = \frac{\pi}{4} * E^* * L * d \tag{5.9}$$

5.9: The equation used to determine the maximum force based on contact stress of two parallel cylinders. F is the normal force on the disc [N], E^* is the equivalent module of elasticity [Pa], L is the width of the connection [m], and d is the indentation depth [m]. Based on [27]

To increase the holding payload, a different material with a higher compressive modulus can be chosen to function as a brake disc, the width of the disc can be increased or the stroke of the LM actuator can be increased. When this likely failure was noticed, the LM actuator and disc width could no longer be changed. It was noticed that the spring washer has the same "footprint" on the piston as the spring. Therefore, it will not distribute the forces anymore than the spring would. Removing the 1 [mm] thick spring washer increased the LM actuator stroke 1 [mm], which is an increase of 50 [%]. The updated indentation depth is 0.34 [mm] and the corresponding maximum force and holding payload are 384 [N] and 1.1 [kg] respectively. As the spring is 1 [mm] less compressed, the output force is lowered to 80 [N]. The force amplification and high p/b ratio mean that this is no problem, as the limiting factor is still the material failure of the brake disc.

5.4.2. Final Designed Requirement Values

Table 5.12: An evaluation of the requirements for the theoretical prosthesis

	Designed Value	Comment
Weight	≈1250 [g]	Can be reduced by over 400 [g] with minor changes, mainly changing the valves
Flexion/Extension RoM	0-140 [°]	-
Pronation/Supination RoM	±90 [°]	-
FE Lifting Payload (P)	P>4 [kg]	For RoM 0-90 °, see Figure 5.5
	4 [kg] > P > 1.5 [kg]	For RoM 90-140 °, see Figure 5.5
FE Holding Payload	1.1 [kg]	-
PS Torque	1.8 [Nm]	Can be increased to 3.0 [Nm] by changing the lead-screw
FE Stroke time	0.57 [s]	A lot of speculation: to be tested
PS Stroke time	0.45 [s]	A lot of speculation: to be tested
Sound Level	<74 [dB]	-
Cycles of Use per Charge	125	Average use, per 25 [g] cartridge
Cost	662.50 €	The cost of the machining and metal parts is not known/included

6

Manufacturing and Assembly

"Manufacturing and assembly" is the penultimate phase of the design process. During this phase, the theoretical design and CAD-model are translated into a functioning real-world prototype.

6.1. Manufacturing

For the custom parts to be produced, technical drawings had to be created. The technical drawing for the part PS-1-1-05 is shown in Figure 6.1. All technical drawings can be found in Appendix N

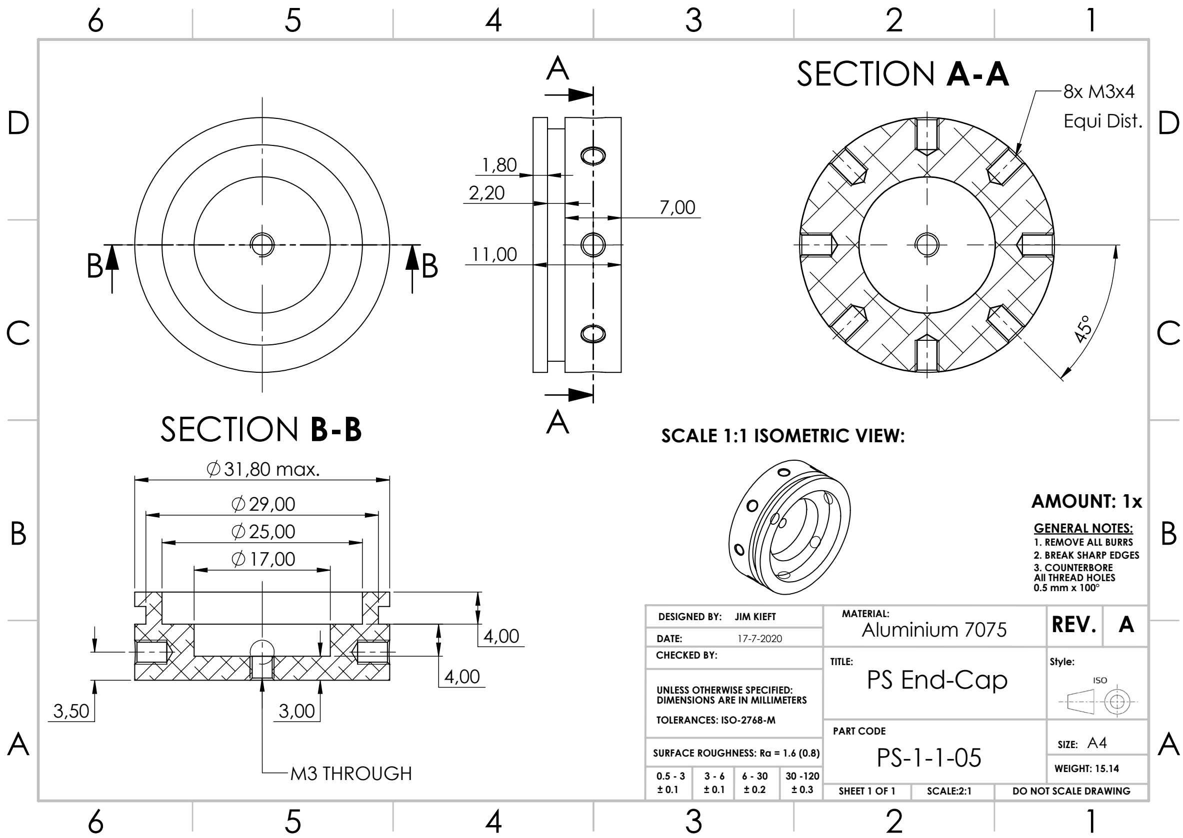


Figure 6.1: The technical drawing of part: PS-1-1-05: PS End-Cap

6.1.1. Problems during Manufacturing

COVID-19 Delays The COVID-19 Virus led to several problems during the manufacturing of custom parts and ordering of off-the-shelf products. The most influential effect was caused by the delay of the laser-cut custom parts. Despite being designed in such a way that the production time would be less than an hour, the delivery time of these parts was more than two months. As the frame consists mostly of laser-cut parts, the final test were delayed by months. This meant that there was no time for iteration of the prototype.

Locking Mechanism Frame The levers used for the locking mechanism frame could not be laser-cut. However, because of the aforementioned delay, this was not communicated until October 15th, almost three months after the .DXF files had been send to the manufacturer. The consequence is that the locking mechanism cannot be tested within the time-frame of the thesis.

Actuator Housing For both the FE-, and PS-Actuator the housing was not manufactured according to the tolerances. A possible reason given by the manufacturer is the thin-walled cylinders being deformed during the processing into an oval shape. This causes the assembled actuator to slightly leak as air can bypass the piston. This will lead to a reduction of efficiency.

The manufacturer proposed several solutions to fix this for a next iteration, as it is not a design flaw but rather a manufacturing challenge. However, there was no time to re-manufacture the housings.

Lead-Screw Nut The wrong lead-screw nut was delivered and had to be manufactured in order for the designed configuration to work.

(a) The desired lead-screw-nut

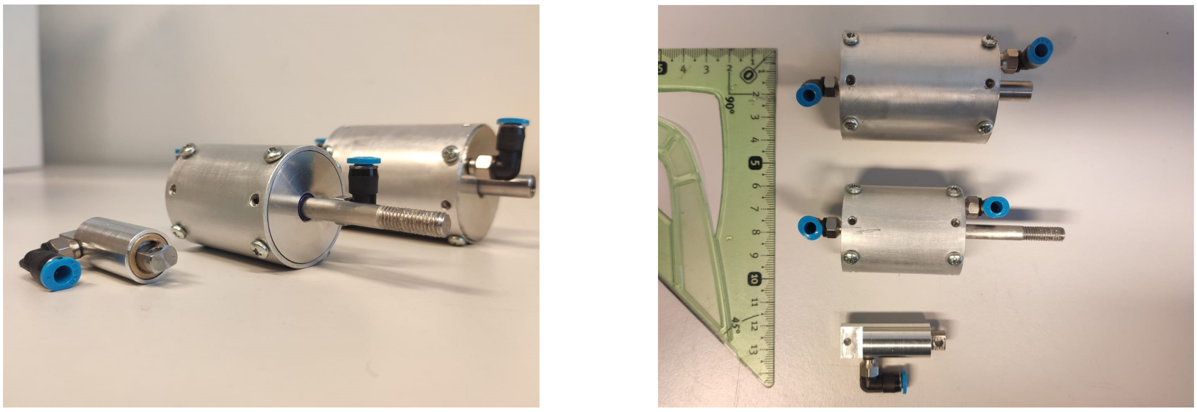
(b) The delivered lead-screw-nut

Figure 6.2: The reason for customizing the lead-screw-nut: A wrong delivery

6.2. Assembly

Appendix O shows a guide for the the assembly of the prosthesis.

6.2.1. Assembled Components



(a) The assembled three actuators. Left to right: LM-Actuator, PS-Actuator, and FE-Actuator

(b) The assembled three actuators with a ruler for scale. Top to bottom: FE-Actuator, PS-Actuator, and LM-Actuator

Figure 6.3: The assembled FE-, PS-, and LM-actuators

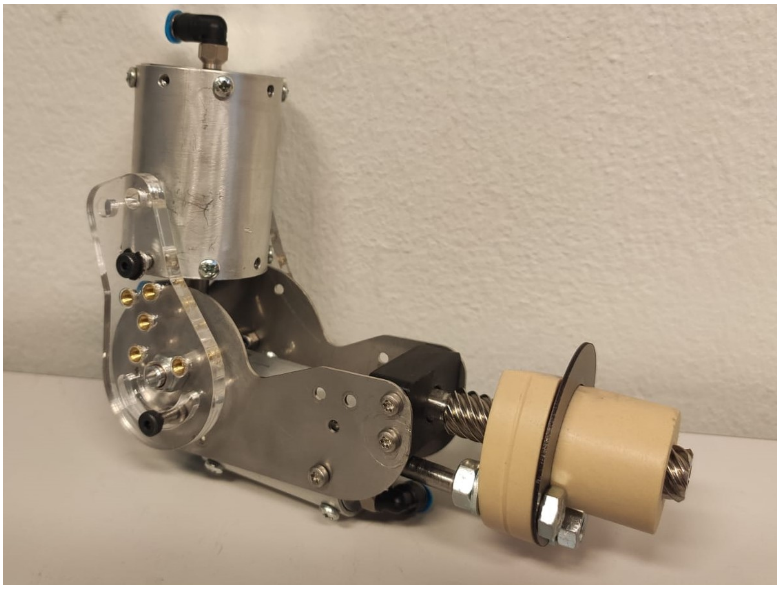
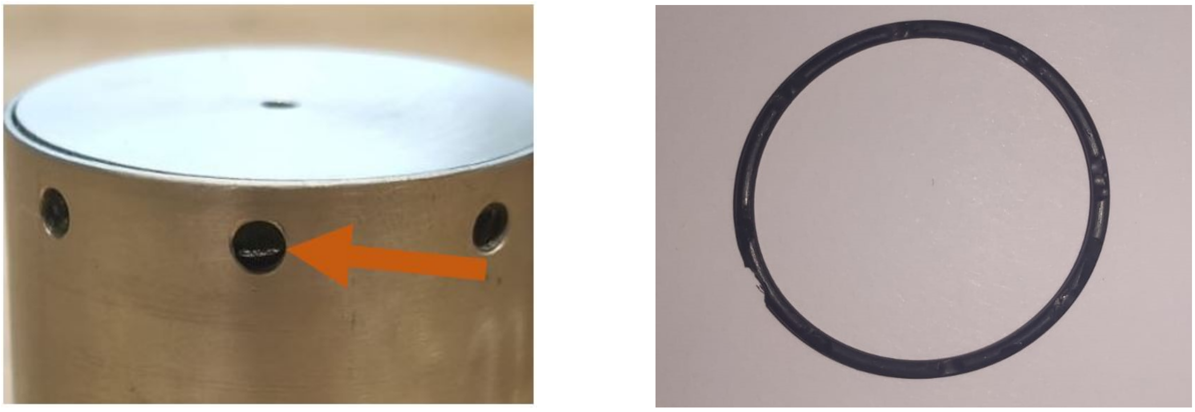


Figure 6.4: The prototype of the pneumatic elbow prosthesis

6.2.2. Discoveries during Assembly

O-Ring The method of assembling the o-ring within the cylinder proved to be problematic. The o-rings were chosen to be rather large, to prevent leakage. The outer diameter of the o-ring is 1 [mm] larger than the inner diameter of the cylinder housing. This is done intentionally to avoid leaks. By using grease during assembly, this would not have been a problem, were it not for one design flaw: The holes in the housing. When the cap with the o-ring is being pressed into the housing, these holes give the o-ring an opportunity to expand, and ”bulge-out”, see fig. 6.5a. Pressing the cap further into the housing causes the o-ring to be damaged at the location of the holes, see fig. 6.5b. A damaged o-ring is more likely to leak gasses and reduce the efficiency of the cylinder.

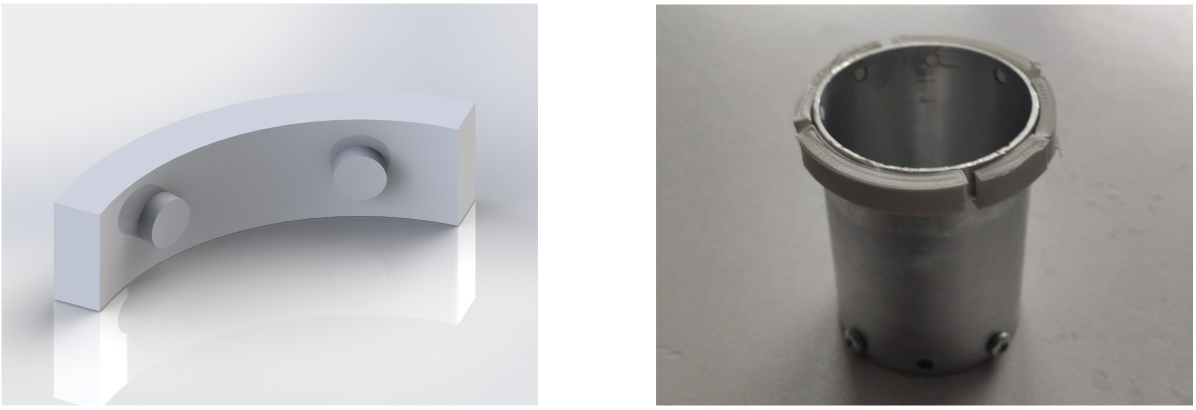


(a) The ”bulged-out” o-ring

(b) The damage caused to the o-ring

Figure 6.5: The damage caused to the o-ring by the holes in the cylinder housing

To solve this, a custom part was designed and 3D-printed to fill these holes during the pressing of the caps, but easy to remove after the o-ring has passed the holes without taking damage. This part is shown in Figure 6.6. It functioned as designed and protected the o-ring during the assembly.



(a) A render of custom part used to protect the o-ring during assembly

(b) The o-ring protectors implemented

Figure 6.6: The method used to prevent the o-ring from being damaged during assembly

PS-Actuator During the process of discovering why the o-rings failed, the inside of the PS-Housing got slightly scratched. It is expected that this will cause a reduction in efficiency, as the gas can bypass the piston by flowing through the scratch.

7 Testing

The goal of the testing phase is to measure if the prototype achieves the designed values/requirements. There are two types of tests: validation and final. Validation tests will happen throughout the assembly, to see if it has been assembled correctly. These are shown in Table 7.1. The final tests will determine whether or not the designed values/requirements are met, shown in Table 7.2. Furthermore, before starting the assembly and at some times during the assembly, the mass of a part, component or sub-assembly has to be measured. The designed test protocol and set-up is shown in Appendix P.

Table 7.1: An overview of the validation tests

Test #	Output	Comment
1	FE & PS Output Forces & Stroke Time	**Do Before**: Attaching the end-effector to the piston-rod
2	LM Force	**Do Before**: Implementing the LM-Actuator on the frame

Table 7.2: The final test that measure the result(s)

Test #	Output	Comment
1	PS Output Torque	**Do Before**: Attaching the PS-Actuator to the frame
2	FE Lifting Payload	-
3	LM Holding Payload	-
4	Sound Level	-
5	Uses until Charge	Will be skipped because of leaking actuators

Data Acquisition The stroke times were recorded using the 60 [fps] camera of the Xiaomi 9T. The FE- and PS-Actuator forces were measured with a load cell (model: Futek LCM300 loadcell 2KN (500lb)) For the pronation/supination torque, a load cell was used to determine the radial force at a distance of 34 [mm] from the lead screw axis (model: FUTEK Miniature S-Beam Jr. Load Cell). The force data was fed to a data acquisition (DAQ) device (model: NI USB-6008, 12-bit, 10 kS/s) and into the computer. The angles for the varying FE payloads were captured by taking a photograph at the maximum angle using the camera of the Xiaomi 9T. The sound levels were measured at a distance of 10 [cm] from the valves, using a decibel meter (model: Center 322 Sound Level Meter).

Data Processing The measurements involving the forces and PS-torque were averaged per pressure and poly-fitted using a first order polynomial. For the FE payloads, the measurement were averaged per payload and poly-fitted using a third order polynomial.

7.1. Validation

7.1.1. Components Mass

The SolidWorks model gives an indication of the weight of the part or component. The mass (and prize) of each of the individual parts can be found in Appendix L. A summarized version is given in Table 7.3. This table can be used as a starting point for eventual weight reduction for a next iteration of the design.

Table 7.3: The expected and measured mass for the sub-assemblies

Sub-Assembly	CAD "Mass"	Measured Mass
FE-Actuator	137 [g]	154 [g]
PS-Actuator	206 [g]	220 [g]
Frame Sheets	72 [g]	75 [g]
Prototype[1]	478 [g]	491 [g]
LM-Actuator	21 [g]	22 [g]
Covers	67 [g]	70 [g]
Control Circuit	718 [g]	705 [g]
Full Prototype [2]	1263 [g]	1278 [g]

1. Without valves, cover, and locking mechanism, see Figure 7.1
2. With all components

Figure 7.1: The weight of the prototype

Additional Notes on Prototype Weight Some of the components that were required for the prototype, such as the valves, have significantly lighter available versions. The main example are the valves of the control circuit. If electronic control is used instead of manual control, The prototype can be 460 [g] lighter, see Appendix R.

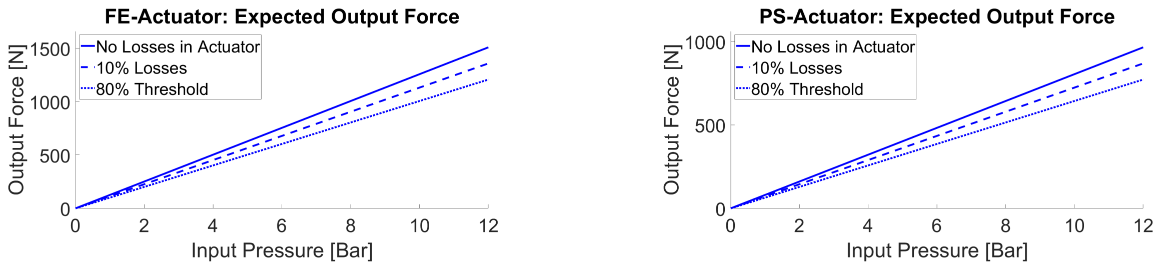
If a more bare-bones flexion/extension prosthesis is desired, the removal of the PS-Actuator and corresponding valve would reduce the total weight by over 250 [g]. The elimination of the actuated locking mechanism would lead to a reduction of at least 85 [g]. This means that a bare-bones prototype, with an integrated fuel source and control circuit, could weigh less than 600 [g].

7.1.2. FE/PS-Actuators: Force Output & Stroke Time

This test has to be performed before the end-effector is placed on the piston-rod

Expected Results: Forces It is expected that the force will increase linearly with the pressure. The efficiency of the pneumatic cylinder will be estimated based on the slope of the force-pressure relation. If the efficiency is low, a low output torque/payload can be attributed to the cylinder and not the configuration. This is useful for further design iterations, as gives an indication of which component or sub-assembly improve or change in order to increase the output torque/payload.

A new piston will have to wear in before working optimally. Furthermore, as mentioned, the housings were not perfectly smooth on the inside, further reducing the efficiency. To account for this, 10 [%] losses are expected to occur. The expected outcome forces are plotted in Figure 7.2. A threshold value of 80 [%] efficiency is set. If the efficiency is below this threshold, something has gone wrong during the assembly and the actuator should be re-assembled.



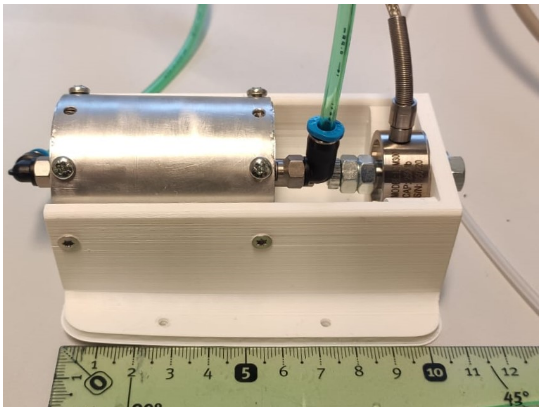
(a) The expected results of the FE output force

(b) The expected results of the PS output force

Figure 7.2: The expected results of the general actuator test

Expected Results: Stroke Times CO_2 cartridges will not be used in the prototype, thus the stroke time of the prosthesis cannot be determined. As mentioned in Section 5.3.4, the flow rate from the cartridge is most likely not the limiting factor. The ”delays” of the system could be estimated if the flow rate from the gas supply is known, by subtracting the theoretical stroke time from the measured stroke time. However, the prototype differs from an eventual prototype, especially in the control department. To have the prototype be operated manually, there are different valves and longer connection cables, thus the measured delays would not be fully representative.

Results The setup for tests to determine the actuator forces are shown in Figure 7.3. The same setup is used to determine the stroke times.


(a) The setup of the FE-Actuator tests

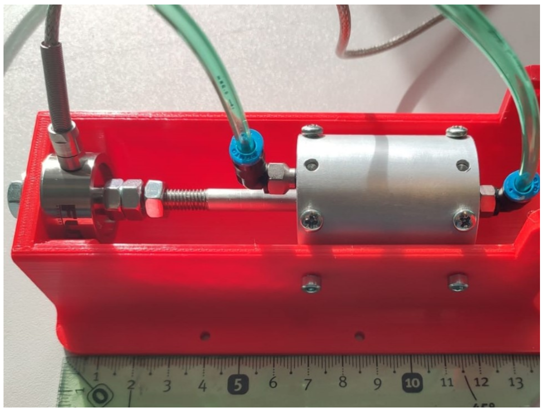
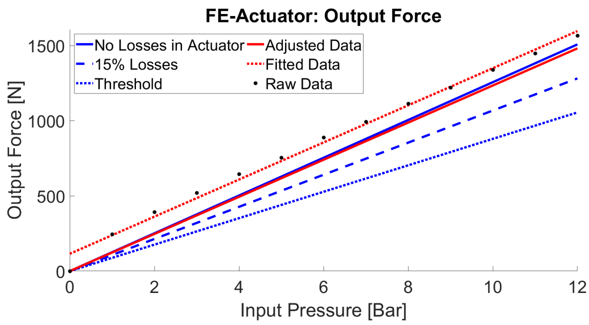

(b) The setup of the PS-Actuator tests

Figure 7.3: The setup of the FE and PS Actuator tests. The colors do not mean anything, the white 3D-print filament ran out during the test preparation

The measurements concerning the FE-Actuator force are shown in Figure 7.4a. It can be seen that the measured force is higher than the ideal force. Most likely, the used manometer is not perfectly accurate and has an offset. There were two manometers available, one ranging from 0 - 16 [bar] and one ranging from 0 - 40 [bar]. The smaller one was used, as it was estimated to be more accurate. However, the data shows this might not be the case. The difference between the manometers is ≈1.3 [bar], see Figure 7.4b. For the fitted data to go through the origin, an x-offset of 1.0 [bar] is required. Neither manometer is perfectly accurate. The smaller manometer will be used for the rest of the tests, accounting for a 1.0 [bar] offset.



(a) The results of the FE output force


(b) The difference between the two manometers

Figure 7.4: The consequences of using a manometer with an offset

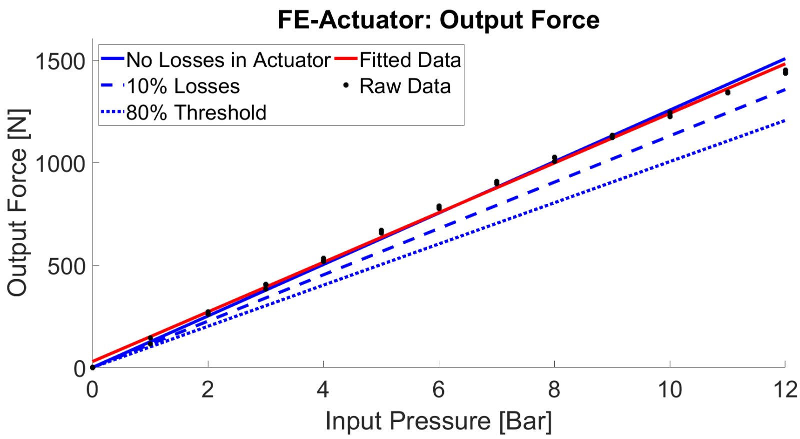


Figure 7.5: The results of the FE output force test

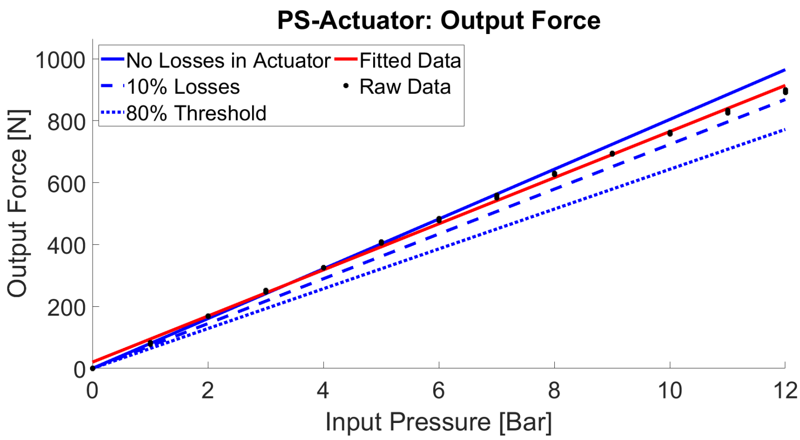


Figure 7.6: The results of the PS output force test

For both tests, all three data sets are very similar, illustrated by the low scatter of the ”Raw Data”. It can be seen that the efficiency of the PS-Actuator is lower than the FE-Actuator, as was expected due to the scratch in the housing. Nevertheless, the efficiency is high enough to continue with, and there were no options to manufacture a new PS-Housing within the time frame. The efficiencies are shown in Table 7.4

Table 7.4: An estimate of the efficiency of the FE- and PS-Actuator

Motion	Efficiency
Flexion/Extension	97 [%]
Pronation/Supination	94 [%]

The stroke times are shown in Table 7.5. They were measured for an unloaded configuration.

Table 7.5: The stroke times for the FE- and PS-Actuators for an unloaded configuration

Motion	Expected Stroke Time	Stroke Time	Expected Stroke Time (with delays)	Time for full movement
Flexion/Extension	0.17 [s]	0.15 [s]	0.57 [s]	0.36 [s]
Pronation/Supination	0.05 [s]	0.08 [s]	0.45 [s]	0.30 [s]

7.1.3. LM Actuator: Force Output

Expected Results As with the FE-, and PS-Actuators, the force is expected to increase linearly with with the pressure. However, for the LM-Actuator, the spring causes a starting force of of - 80 [N]. This force is negative as it pushes the piston "inwards". The expected outcome forces are plotted in Figure 7.7. This graph shows that when the cylinder is not pressurized, the force on the brake-disc will be at its maximum. Pressurizing the actuator will decrease this force and therefore decrease the holding torque. An output force of 0 [N] would mean that the locking mechanism is completely disengaged and the friction for FE-movement is therefore minimized.

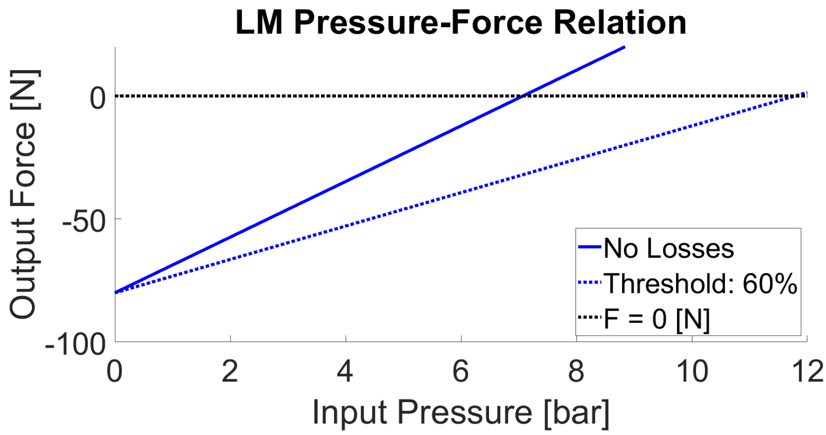


Figure 7.7: The expected results of the LM-Actuator Test. A postive force is directed outwards.

Results The test setup is shown in Figure 7.8

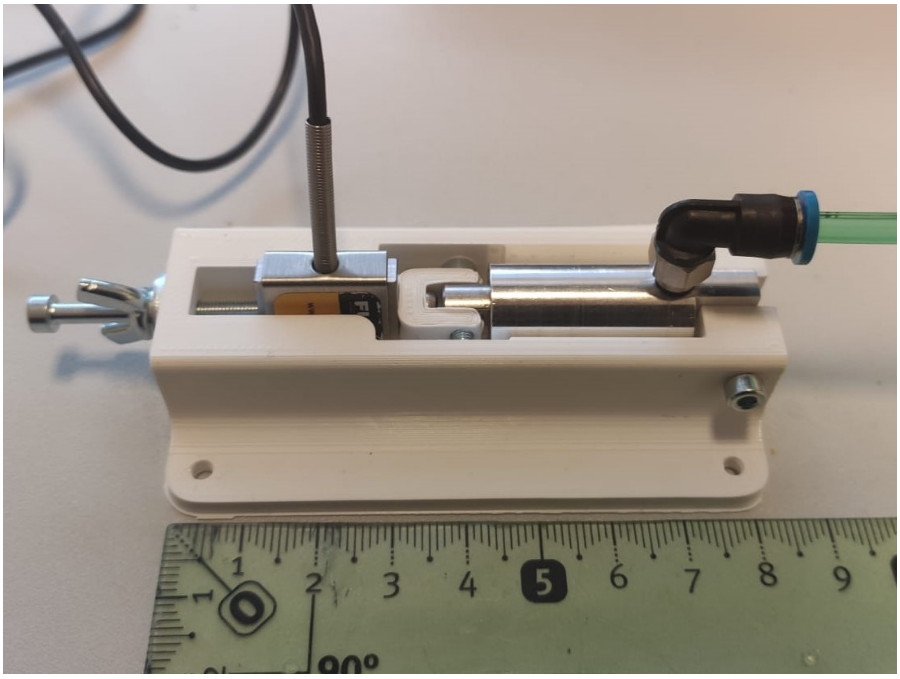


Figure 7.8: The setup of the LM-Actuator test

The results of the LM-Force test are shown in Figure 7.9. It can be seen that the actuator loses about 85 [%] of its force. This is no problem, as the spring force is still nullified at 12 [bar]. The LM actuator has passed the validation test.

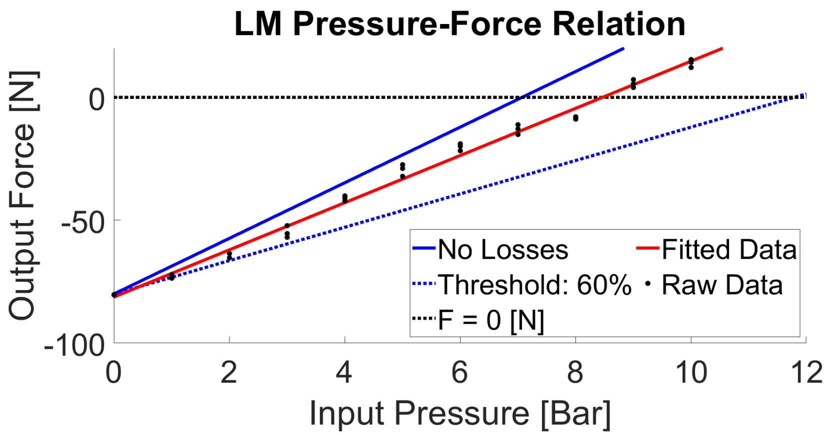


Figure 7.9: The results of the LM-Actuator test

7.2. Final Tests

No torque sensors were available. For the PS torque, the force was measured at a distance of 34 [mm]. The FE lifting torque and LM holding torque were not measured, their functionality was based on the payload.

7.2.1. PS Actuator: Torque Output

It is expected that, just like the force, the torque will increase linearly with the pressure.

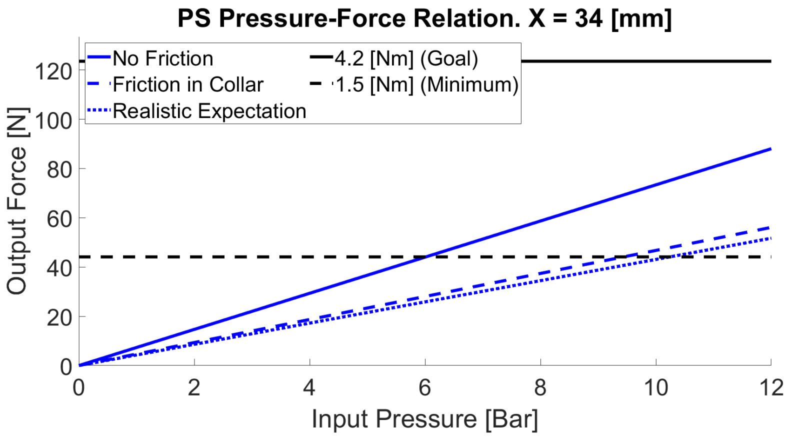


Figure 7.10: The expected results of the PS torque test

The test setup is shown in Figure 7.11.

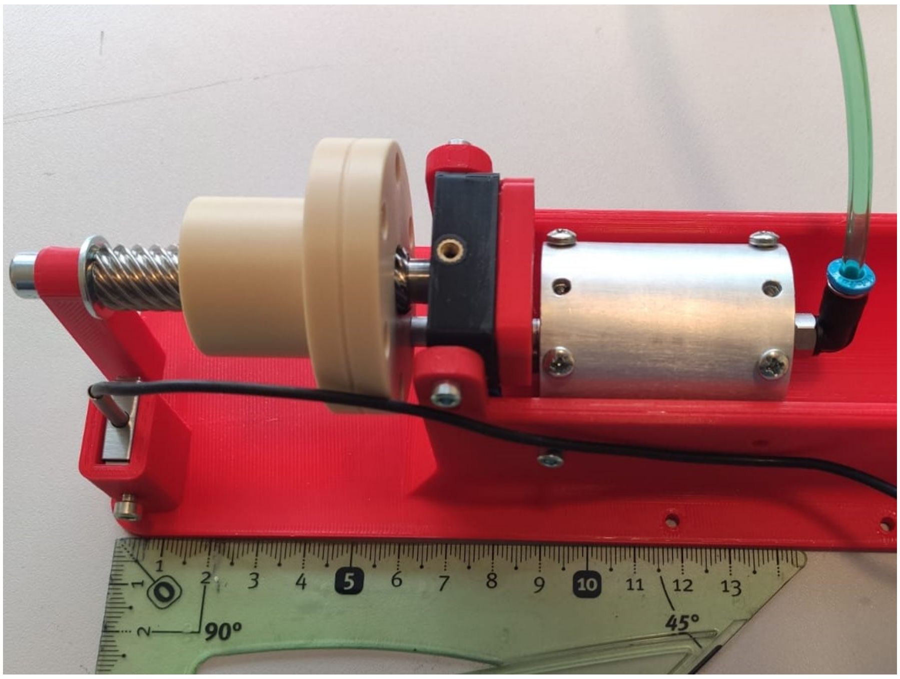
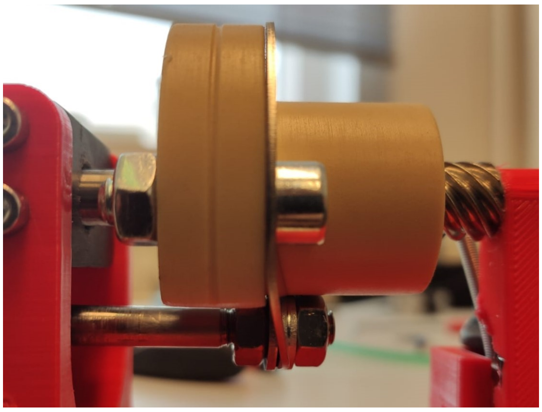
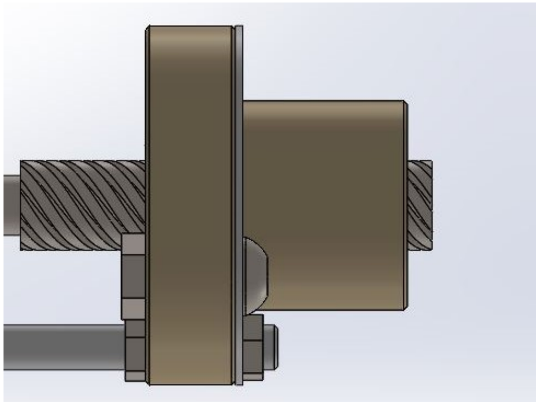


Figure 7.11: The setup of the PS torque test

The torque test brought to light a design flaw: The configuration and thickness of the PS-Nut-Connect. The designed configuration minimized the size of the nut and therefore its mass. The thickness was set at 1 [mm] in order to achieve easy manufacturing of the part, as it can be cut from the same steel sheet as other parts. This thickness was not enough to withstand the actuator force, as shown in Figure 7.12a. This could be fixed by (1) increasing the thickness or (2) changing the configuration. Because of the earlier issues with delays of laser-cut parts, the latter option was chosen. To change the configuration, a different than designed lead-screw-nut is required. Luckily, the lead-screw-nut which was wrongly delivered provided the opportunity to change the configuration. The new configuration is shown in Figure 7.12b. With this new configuration, the test could be performed.



(a) The failure of the PS-Nut-Connect



(b) The new configuration of the force transfer between the piston-rod and the lead-screw-nut

Figure 7.12: The consequence and solution to a too weak connection between the actuator and lead screw

Figure 7.13 shows the results of the torque test. The friction caused by the collar was severely overestimated. The output torque at 12 [bar] is ≈ 2.8 [Nm].

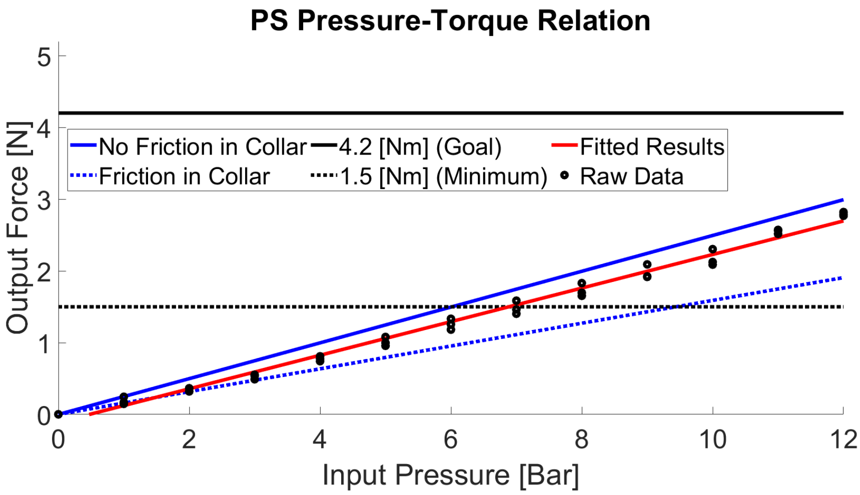


Figure 7.13: The results of the PS torque test

7.2.2. FE Lifting Payload

This test is aimed to determine the maximum angle a payload can reach at a pressure of 12 [bar].

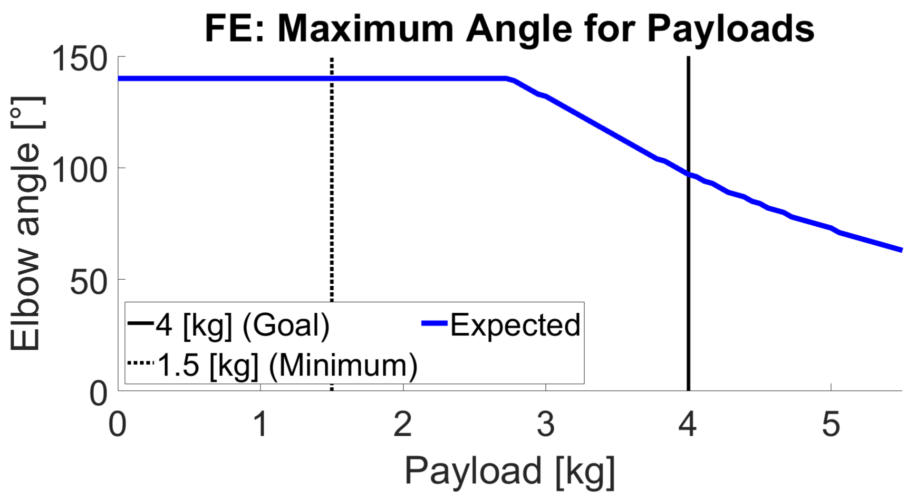


Figure 7.14: The expected results of the FE output force

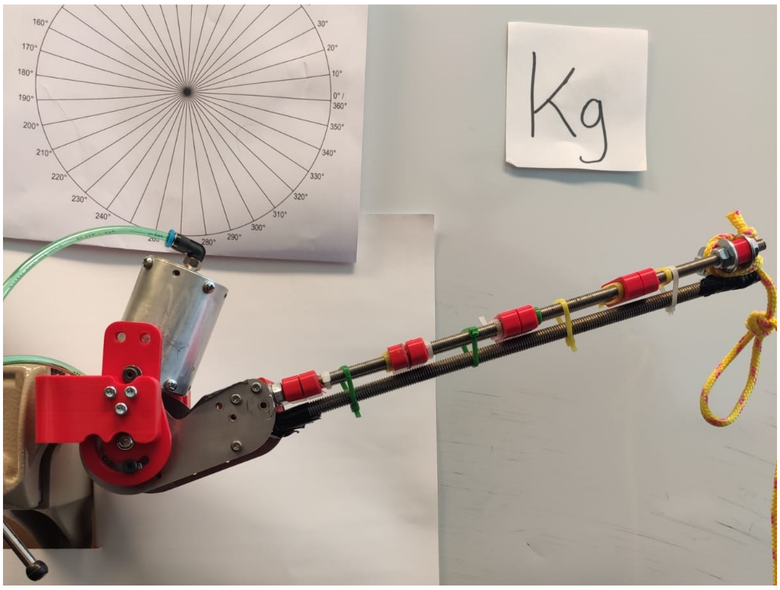
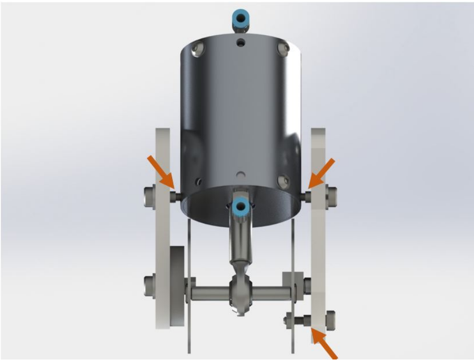
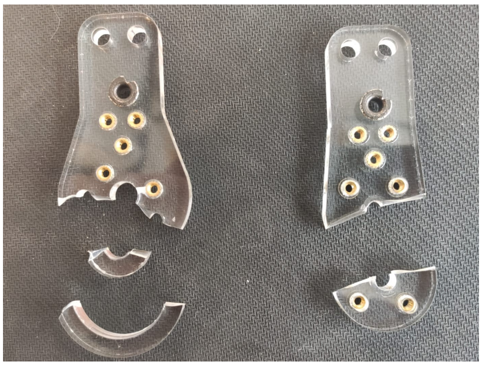


Figure 7.15: The setup of the FE payload test

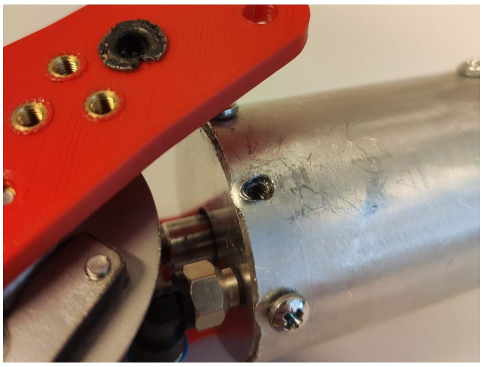
During the preparation of the test, it was discovered that the frame is not fully constrained at several locations, see Figure 7.16a. Because of this bending can occur and the PMMA sheets broke, see Figure 7.16b. 3d-printed back-up sheets were used. For higher FE angles, the frame starts to bend inwards, greatly increasing the friction. Furthermore, during the assembly and testing phases, the FE-actuator was constantly dis-assembled and re-assembled. This put an excessive strain on threaded holed in the aluminum caps. Eventually, the thread started to fail, as can be seen in Figure 7.16c.



(a) The locations where the frame is under-constrained and can therefore move



(b) The broken PMMA sheets



(c) The failure of the threads in the the FE-caps

Figure 7.16: Two failures discovered during the testing of the maximum liftable payload

The results of the FE Payload test are shown in Figure 7.17.

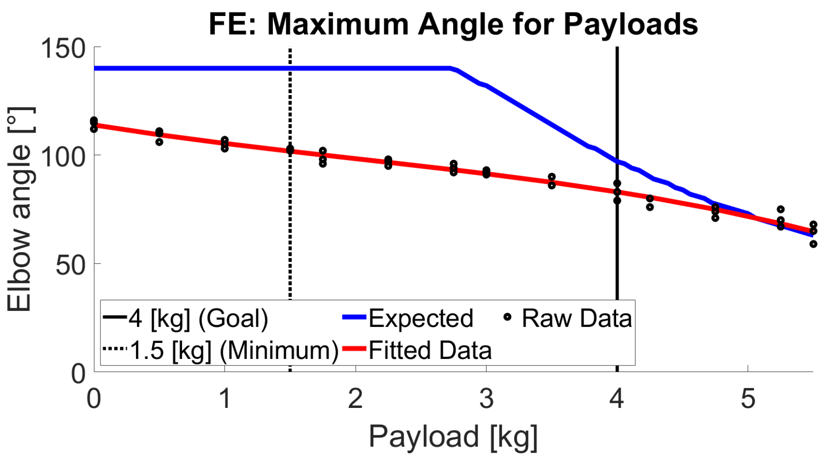


Figure 7.17: The results of the FE Payload Test

7.2.3. LM Holding Payload

This test is aimed to determine the maximum holding payload of the locking mechanism. It will be tested at the 90 [°], as this is the angle which causes the highest holding torque to be required. The under-constrained frame leads to an under-constrained locking mechanism. This prevents the levers from acting a force upon the brake disc, and the maximum payload the locking mechanism can hold is therefore 0 [kg].

7.2.4. Sound Level

It is expected that the gas exhaust of the actuators will be the loudest process of the prototype. The silencers should ensure that the sound levels will stay below 74 [dB]. The expected and measured results are shown in Table 7.6.

Table 7.6: The expected and measured sound levels produced during for flexion/extension and pronation/supination

Motion	Expected Sound Level	Measured Sound Level
Flexion/Extension	<74 [dB]	52 [dB]
Pronation/Supination	<74 [dB]	51 [dB]

7.2.5. Uses until Charge

Whilst all the tests thus-far have been using an ”infinite” gas-supply, to test the functionality of the prototype, the uses until the cartridge is depleted should be tested. However, as the actuators leak, something that can be easily avoided in a next iteration, the results of this test will not represent the prosthesis potential. It was therefore decided to forgo this test until it can be performed with fully functional, non-leaking actuators.

8 Discussion

8.1. Design Process

8.1.1. Analysis

The analysis phase provided an overview of reasons for non-wear, produced a function tree, with the functions ranked according to the MoSCoW method, and provided a list of design values, requirements and wishes. These outcomes were extremely useful for the design of the prototype. Furthermore, they provide a solid foundation for a completely different design, even if the chosen method of actuation were to be either electronic or body-powered.

8.1.2. Conceptualization

The conceptualization phase shows the use of working with the chosen modules and the power of 3D-printed prototyping. The limited available time for the thesis prevented the concepts from being extremely out-of-the box.

8.1.3. Embodiment

During the embodiment phase, the large variety of tools and skills required to design a prosthesis is highlighted. MATLAB is used for topology determination and optimization of the FE implementation, an analysis of available off-the-shelf components was required for the PS-lead-screw selection, SolidWorks was required to have a 3D-model, and understanding and creating technical drawings is a necessity to bring the theoretical design into reality.

8.1.4. Manufacturing & Assembly

Because of the COVID-19 virus, there were delays in the manufacturing and ordering of the required parts. Some off-the-shelf parts had to be interchanged for a similar counterpart. The manufacturing shows that it is possible to design a pneumatic prosthesis consisting of relatively simple parts and off-the-shelf product. The assembly guide will be useful for anyone willing to recreate the prototype.

8.1.5. Testing

The testing phase brought to light some issues that needed solving for the prototype to function properly. The results will be discussed in detail in the next section.

8.2. Prototype

8.2.1. Functions & Modules

Table 8.1 shows an overview of the modules, the corresponding function, their ranking in the MoSCoW analysis and whether or not they have been implemented in the prototype. It can be noted that for the modules: Frame and Control, a majority of the functions was not achieved. This was expected, as the focus of the prototype was on showing the potential of the method of actuation. It was therefore not required to house components, such as a gas cartridge.

Table 8.1: An overview of the modules, the corresponding function, their ranking in the MoSCoW analysis and whether or not they have been implemented: Green = implemented, yellow: partially implemented, red: not implemented

Priority	Actuation	Locking Mechanism	Frame	Control
Must Have	Flexion Extension	-	Constrain Unwanted DoF	Generate Required Output
	Pronation Supination	-	Withstand Forces and Moments	Sense Operator Input
Should Have	-	Be Lockable [1]	Protect Components	Detect Angles
	-	Have a Free-Swing Mode	House Components	Provide Feedback
	-	-	Fit to User Arm	Have a Portable Energy Supply
Could Have	-	-	Connect to Terminal Device	-
	-	-	Be Customize-able	-
Won't Have	-	-	Have an Adjustable Fitting	Show Power Level

General Module Evaluation It can be seen that modules with more functions are more likely to not achieve all functions. There are two possible reasons for this: (1) the larger modules are more likely to have "could-haves" and "won't-haves", which will almost certainly not be implemented in the first design cycle, and (2) the larger modules lack a certain overview and are therefore more difficult to manage. The first reason is inherent to the chosen design process. The second reason can be mitigated by limiting the amount of functions in a certain module by creating more modules or to use sub-modules to create a more coherent overview. For example, splitting the module "Frame" into multiple sub-modules: "Structure", "Covers", and "Fitting". This distribution of functions is shown in Appendix Q.

Frame The designed upper arm sheets do not fully constrain the undesired degrees of freedom. This leads to additional forces on the sheets, causing failure, and to additional friction in the system, especially regarding flexion/extension. For a future iteration, the constraints should be redesigned.

8.2.2. Requirements

Table 8.2 compares the results of the prototype to the requirements found in the analysis phase. If a requirement has not achieved its goal, possible reasons and solutions are discussed.

Table 8.2: The results of the prototype compared to the requirements. Green: The goal is achieved, Yellow: The minimum/maximum value is not surpassed, and Red: The minimum/maximum value is surpassed

Requirement	Design	Prototype	Goal	Extreme
Requirements based on Cosmesis				
L Forearm	172 [mm]	172 [mm]	< 87 [mm]	< 174 [mm]
L Upper Arm	144 [mm]	144 [mm]	< 113 [mm]	< 170 [mm]
Requirements based on Comfort				
Weight Prototype	1260 [g]	1278 [g]	< 800 [g]	< 1000 [g]
Implementing Electronic Valves	860 [g]	Not Build		
”Bare-Bones” Version	< 600 [g]	Not Build		
Sound Level	< 74 [dB]	52 [dB]	< 20 [dB]	< 50 [dB]
Cycles of Use until Charge	125	Not Tested	> 300	> 100
Cost	662.5 €	662.5 €	-	< 1000 €
Requirements based on Control				
Flexion/Extension RoM	0-140 [°]	0-140 [°]	0-150 [°]	5-135 [°]
Pronation/Supination RoM	±90 [°]	±90 [°]	±90 [°]	±50 [°]
FE Lifting Payload[1]	2.5 [kg]	0-4 [kg]	> 4 [kg]	> 1.5 [kg]
FE Holding Payload	1.1 [kg]	0 [kg]	> 23 [kg]	> 6 [kg]
Redesign the Frame and Disc	> 6 [kg]	Not Tested		
PS Torque	1.8 [Nm]	2.8 [Nm]	> 4.2 [Nm]	> 1.5 [Nm]
FE Stroke time	0.57 [s]	0.36 [s]	< 0.7 [s]	< 1.5 [s]
PS Stroke time	0.45 [s]	0.30 [s]	< 0.5 [s]	< 1.0 [s]

1. Depends highly on the desired lifting angle

Length The limited stroke of a pneumatic cylinder restricts the possible range of motion, thus in order to achieve the required ranges of motion, the cylinder length had to increase.

Weight The weight of the prototype exceeds the maximum by almost 280 [g]. Switching to electronic control significantly decreases the prosthesis mass. Despite not being lighter than the goal, the prosthesis would still be lighter than comparable prostheses, whilst achieving similar or better functionality. If the additional functionality such as the active pronation/supination is removed, the ”bare-bones” version can weigh less than 600 [g].

Sound Level The sound level was measured at a distance of 10 [cm] without barrier between the source (valves) and the sensor, and even then only slightly exceeded the maximum allowed sound level. If the valves are implemented within the frame in a next iteration, the covers will help with reducing the sound level.

Cycles of Use until Charge The use cycles are based an expected average use and a 25 [g] CO_2 cartridge as gas supply. The amount of cycles can be increased by (1) using a bigger cartridge, (2) decreasing $M * p * V$, and (3) increasing $R * T$. Implementing a bigger canister means a more bulky and heavier design, decreasing both the cosmetics and comfort of the prosthesis. $M * p * V$ depends on the gas used (M), and the payload, Using a lower pressure reduces the amount of gas used, but also reduces the output torque. An increase of $R * T$ is only achieve able by increasing the temperature, since R is a constant. Actively increasing the temperature of CO_2 will require an energy source and thus require additional components/added mass.

Flexion/Extension RoM The limited stroke size of a pneumatic actuator restricts the RoM. Achieving the full range of motion would add additional mass to the system and cause the FE-torque-angle curves to differ more, leading to less efficient actuation. A range of motion of 0-140 [°] should be enough for almost all ADL. Because of the additional friction due to the unconstrained frame, the 140 [°] angle is not achieved by actuation. It can be achieved by manually moving the forearm.

FE Lifting Payload In theory, 2.5 [kg] can be lifted throughout the entire range of motion, and 4.0 [kg] can be lifted to almost 100 [°] flexion. The main limitation in this design is the topology, thus increasing the actuator force would not necessarily help with achieving the goal. The topology of FE-implementation could be re-evaluated to overlap the angle-torque relation of the achieved- and desired payload.

In reality, the under-constrained frame caused an additional friction. The desired FE-payload of 4 [kg] can be lifted to a maximum of 87 [°]. Even with no payload in hand, the maximum achieved angle is 115 [°]. However, as the actuator force suffices to the theory, it is expected that redesigning the frame constraints will significantly increase the payload capacities for the angles above 90 [°].

FE Holding Payload The goal of a non-actuated holding payload of 23 [kg] (achieved the Boston Elbow and Utah Arm [28]) is incredibly high, but could be achieved with the use of the designed force-amplification method. However, in the prototype the levers cannot exert a force on the brake disc. Therefore, the maximum holding payload of the prototype is 0 [kg].

PS Torque The desired PS-torque of 4.2 [Nm] is not achieved. This was a calculated decision to help reduce the mass of the prototype. Contrary to the FE lifting payload, increasing the actuator force directly influences the output torque.

8.2.3. Wishes

Table 8.3 shows whether or not a wish has been fulfilled. If a wish has not been fulfilled, possible reasons and solutions are discussed.

Table 8.3: An overview of the wishes for the prosthesis. Green: The wish is achieved, Yellow: The wish is partially achieved, and Red: The wish is not achieved

Category	Details
Cosmetics	Fit under clothes
	Have customize-able covers
Control	Lockable in all-positions
	Actuated locking mechanism

Locking of Pronation/Supination To reduce the mass and bulkiness of the prosthesis, no actuated system to achieve locking of pronation/supination has been implemented. The friction within the system should provide enough resistance to movement for most cases, and if heavier loads will be lifted, the set-screw can be used to hold the position.

8.2.4. Possible Improvements to the Prototype

During the embodiment design, several choices have been made that could be done differently in future iterations. The most influential of these ”possible improvements” and the consequences are discussed below. A more in-depth explanation of each of these improvements is given in Appendix R.

Actuator Caps The threads in the aluminum caps failed after extensive use. For a future prototype, stronger threads should be used. This can be achieved by changing the material to steel or by using threaded inserts. Furthermore, the material used for the caps can be changed from aluminum to a plastic, such as POM. The density would be decreased from 2.9 [kg/m^3] to 1.4 [kg/m^3], a reduction of 52 [%]. If all caps were to be changed to POM, the total reduction in weight of the prosthesis would be about 45 [g]. It is vital to use inserts if this modification is chosen.

PS Lead-Screw A 16 [mm] diameter lead-screw is available in aluminum, reducing the complete mass (including the nut) by more than 50 %. This new lead-screw would increase the torque by 1.2 [Nm] and decrease the mass of the actuator by about 20 [g]. Furthermore, it is stronger and could be implemented with a ⌀ 40 [mm] actuator to further increase the output torque.

LM Brake-Disc The holding payload depends on the friction generated on the brake disc, which depends on the normal force on the disc, the coefficient of friction, and the disc radius. The disc radius cannot be made much larger without much design changes, and the coefficient of friction of TPU on aluminum is one of the highest possible. Thus the normal force on the disc should be increased. The generated force is self-amplifying, thus the actuator does not need to be stronger. The problem lays with the maximum force until the LM-piston-rod is fully retracted. This force depends on the compressive modulus, disc width, and indentation. Off these parameters, only the compressive modulus can be changed without altering the design.

Valves Manual valves are significantly heavier than solenoid valves, thus in the future, solenoids valves are highly recommended. To control solenoid valves, a circuit board, battery and other electronic components should be implemented, leading to additional mass. A simple Arduino Uno and 9V battery could suffice, which would weigh 25 [g] and 45 [g] respectively. If the assumption is made that all other components would weigh a total of 20 [g], the total reduction in mass would still be almost 400 [g]!

8.3. Design Goal

The design goal was:

To show the potential of pneumatic prostheses by designing a lightweight pneumatic elbow prosthesis, with a functionality equivalent to existing prostheses

The prototype is heavier than desired. However, concrete, relatively minor, changes are shown to reduce the mass to be lighter than most commercial- and academic prostheses, whilst achieving a similar or better functionality. A prototype achieving the shown functionality, whilst being designed, manufactured and tested in less than six months, indicates that the potential for pneumatic actuation in prostheses merits further investigation. It can therefore be stated that the design goal has been achieved.

8.4. Further Research

8.4.1. Socket

As it was not part of the main research goal, no attention was paid to the socket. However, for the prototype to become a viable product, a socket is a necessity. Comfort should be a main goal in this research. To achieve a higher level of comfort, it would be preferable if the socket was adaptive and lightweight.

8.4.2. Covers

Despite being able to interchange the covers, a lot of progress can be made considering the appearance of the prosthesis. First of all, an in-depth user analysis could be done to understand what would make an elbow prosthesis visually appealing to the user, i.e. should it resemble a human arm as much as possible or is a high-tech robotic appearance preferred? With this knowledge in mind, covers can be designed that are suited to the user's wishes. Furthermore, the possibility of using carbon-fibre as a material for the covers can be investigated, as these might help with the reduction of weight.

8.4.3. Force-Amplified Locking Mechanism

The potential of a force-amplified mechanism is high due to its high possible locking forces and low mass. It should therefore be tested in the future and researched to see if such a method can be applied for more uses in either this or other prostheses.

8.4.4. Control

The control of the prototype has been achieved using manually. If the prototype would become a viable product, the patient should be able to control the movements. Possible types of sensing the user input, e.g. EMG or body-powered, should be evaluated and implemented. Furthermore, the possible reduction of weight by switching to electronic valves should not be taken lightly. The implementation of electronic valves would require a method of translating the user input into electronic signals and a control scheme to properly translate the user input into the required output.

8.4.5. Integration

To achieve a viable full prosthesis, a terminal device such as a hand or hook should implemented. If a pneumatically actuated terminal device is desired, research could focus on an implementation of the Delft Cylinder Hand [29]. Furthermore, the fuel supply and control circuit should be integrated within the prosthesis.

8.4.6. Gas Type

Monopropellants such as H_2O_2 have a much higher energy density than CO_2 [8], thus to achieve more cycles of use without adding additional mass to the prosthesis, monopropellants as a fuel source for prostheses should be researched.

9
Conclusion

This thesis presents the design of a pneumatically powered elbow prosthesis. Aside from the prototype, the thesis proposes a prioritized function tree, design values, requirements, wishes, and modules which can be used to guide the design of any elbow prosthesis. These can be of value for any design, especially those regarding elbow prostheses.

The prototype has an equivalent functionality to commercial and academically proposed prostheses:

- The flexion/extension payload depends on the desired lifting angle.
  - 4.0 [kg] can be lifted to 87 [°].
  - In theory, 2.5 [kg] can be lifted throughout the entire range of motion, up to 140 [°].
- The pronation/supination torque is over 2.8 [Nm] throughout the entire range of motion of $\pm$ 90 [°]
- The locking mechanism is theoretically capable of passively holding over 6 [kg] in any position
- In theory, an average of 125 cycles of use can be achieved per 25 [g] CO_2 cartridge.
- The covers can be interchanged easily, to allow for customization of the device.
- The prototype consists of simple components, allowing for easy maintenance and repair
- The weight of the prototype is $\approx$ 1300 [g]. With minor changes the weight can be reduced to 900 [g].
- Removing pronation/supination and the active locking mechanism reduces the weight to less than 600 [g].

The combination of high functionality and the low mass of the prototype shows a potential for pneumatic actuation for prostheses.

Bibliography

[1] K. Ziegler-Graham, E. J. MacKenzie, P. L. Ephraim, T. G. Travison, and R. Brookmeyer, “Estimating the prevalence of limb loss in the united states: 2005 to 2050,” *Archives of physical medicine and rehabilitation*, vol. 89, no. 3, pp. 422–429, 2008.

[2] M. LeBlanc, “Give hope - give a hand - the ln-4 prosthetic hand.”

[3] T. W. Wright, A. D. Hagen, and M. B. Wood, “Prosthetic usage in major upper extremity amputations,” *Journal of Hand Surgery*, vol. 20, no. 4, pp. 619–622, 1995.

[4] E. Biddiss and T. Chau, “Upper-limb prosthetics: critical factors in device abandonment,” *American journal of physical medicine & rehabilitation*, vol. 86, no. 12, pp. 977–987, 2007.

[5] D. H. Plettenburg, *Upper extremity prosthetics: current status & evaluation*. Delft University of Technology, 2002.

[6] F. Malagelada, M. Dalmau-Pastor, J. Vega, and P. Golano, “Elbow anatomy,” *Sports injuries: prevention, diagnosis, treatment and rehabilitation*, vol. 2, pp. 527–53, 2014.

[7] T. Sheehan, “Rehabilitation and prosthetic restoration in upper limb amputation,” *Cifu D. authors. Braddom’s Physical medicine & rehabilitation 5th Ed*, pp. 182–1835, 2015.

[8] K. B. Fite, T. J. Withrow, X. Shen, K. W. Wait, J. E. Mitchell, and M. Goldfarb, “A gas-actuated anthropomorphic prosthesis for transhumeral amputees,” *IEEE Transactions on Robotics*, vol. 24, no. 1, pp. 159–169, 2008.

[9] Ottobock, “Body-powered upper limb prostheses.”

[10] Ottobock, “Above-elbow prosthesis with dynamicarm.”

[11] A. Van Boeijen, J. Daalhuizen, R. van der Schoor, and J. Zijlstra, *Delft design guide: Design strategies and methods*. 2014.

[12] D. H. Plettenburg, “Basic requirements for upper extremity prostheses: the wilmer approach,” in *Proceedings of the 20th Annual International Conference of the IEEE Engineering in Medicine and Biology Society. Vol. 20 Biomedical Engineering Towards the Year 2000 and Beyond (Cat. No. 98CH36286)*, vol. 5, pp. 2276–2281, IEEE, 1998.

[13] D. Cummings, “Prosthetics in the developing world: a review of the literature,” *Prosthetics and orthotics international*, vol. 20, no. 1, pp. 51–60, 1996.

[14] C. S. Harkins, A. McGarry, and A. Buis, “Provision of prosthetic and orthotic services in low-income countries: A review of the literature,” *Prosthetics and Orthotics International*, vol. 37, no. 5, pp. 353–361, 2013.

[15] P. Plan, “Moscow prioritization.”

[16] T. U. D. of Civil/Environmental Engineering, “Noise sources and their effects.”

[17] A. Oosterwijk, M. Nieuwenhuis, C. van der Schans, and L. Mouton, “Shoulder and elbow range of motion for the performance of activities of daily living: A systematic review,” *Physiotherapy theory and practice*, vol. 34, no. 7, pp. 505–528, 2018.

[18] D. Magermans, E. Chadwick, H. Veeger, and F. Van Der Helm, “Requirements for upper extremity motions during activities of daily living,” *Clinical biomechanics*, vol. 20, no. 6, pp. 591–599, 2005.

[19] I. A. Murray and G. R. Johnson, “A study of the external forces and moments at the shoulder and elbow while performing every day tasks,” *Clinical biomechanics*, vol. 19, no. 6, pp. 586–594, 2004.

[20] M. Buckley, A. Yardley, G. Johnson, and D. Cams, “Dynamics of the upper limb during performance of the tasks of everyday living—a review of the current knowledge base,” *Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine*, vol. 210, no. 4, pp. 241–247, 1996.

[21] P. Team, “Know your pneumatics: Single or double acting? choosing the right cylinder.”

[22] RS, “Igus flanged round nut for lead screw, dia. 25mm.”

[23] J. C. Cool, *Werktuigkundige systemen; 3e dr.* Delftse Uitgevers Maatschappij, 1992.

[24] McDavid, “Mcdavid 512 elastic elbow sleever.”

[25] Wikipedia, “Briquet détail armure vienne.”

[26] D. Doedens, “Optimal co2 pressure for a pneumatic system,” *Master Thesis*, 2015.

[27] V. L. Popov, *Contact mechanics and friction*. Springer, 2010.

[28] A. Ramirez-Garcia, C. Toledo, L. Leija, and R. Munoz, “Status of elbow myoelectric prosthesis: Cinvestav-ipn prosthesis,” *Revista Mexicana de Ingenieria Biomedica*, vol. 30, no. 1, pp. 66–73, 2009.

[29] G. Smit, “Delft cylinder hand: Natural grasping.”

[30] U. Arm, “Motion e2 elbow.”

[31] L. Resnik, S. L. Klinger, and K. Etter, “The deka arm: Its features, functionality, and evolution during the veterans affairs study to optimize the deka arm,” *Prosthetics and orthotics international*, vol. 38, no. 6, pp. 492–504, 2014.

[32] D. Gow, W. Douglas, C. Geggie, E. Monteith, and D. Stewart, “The development of the edinburgh modular arm system,” *Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine*, vol. 215, no. 3, pp. 291–298, 2001.

[33] Steeper, “Espire pro elbow.”

[34] J. H. APL, “Modular prosthetic limb.”

[35] M. S. Johannes, J. D. Bigelow, J. M. Burck, S. D. Harshbarger, M. V. Kozlowski, and T. Van Doren, “An overview of the developmental process for the modular prosthetic limb,” *Johns Hopkins APL Technical Digest*, vol. 30, no. 3, pp. 207–216, 2011.

[36] I. Vujaklija, D. Farina, and O. C. Aszmann, “New developments in prosthetic arm systems,” *Orthopedic research and reviews*, vol. 8, p. 31, 2016.

[37] Ottobock, “Prosthetics 2019/2020.”

[38] T. Lenzi, J. Lipsey, and J. W. Sensinger, “The ric arm—a small anthropomorphic transhumeral prosthesis,” *IEEE/ASME Transactions on Mechatronics*, vol. 21, no. 6, pp. 2660–2671, 2016.

[39] J. M. Zuniga, A. M. Carson, J. M. Peck, T. Kalina, R. M. Srivastava, and K. Peck, “The development of a low-cost three-dimensional printed shoulder, arm, and hand prostheses for children,” *Prosthetics and orthotics international*, vol. 41, no. 2, pp. 205–209, 2017.

[40] D. A. Winter, *Biomechanics and motor control of human movement*. John Wiley & Sons, 2009.

[41] B. Morrey, L. Askew, and E. Chao, “A biomechanical study of normal functional elbow motion.,” *The Journal of bone and joint surgery. American volume*, vol. 63, no. 6, pp. 872–877, 1981.

[42] L. O’Sullivan and T. J. Gallwey, “Forearm torque strengths and discomfort profiles in pronation and supination,” *Ergonomics*, vol. 48, no. 6, pp. 703–721, 2005.

[43] C. JW'Bill'Limehouse, T. Farnsworth, and C. FAAOP, "A preliminary study of 40+ upper extremity patients using the animated control system," 2005.

[44] J. T. Belter and A. M. Dollar, "Performance characteristics of anthropomorphic prosthetic hands," in *2011 IEEE International Conference on Rehabilitation Robotics*, pp. 1–7, IEEE, 2011.

[45] □. □. пневматика, "How does a pneumatic cylinder work? pneumatic cylinder design."

[46] J. Mendoza-Vázquez, E. Tlelo-Cuautle, J. Vázquez-Gonzalez, and A. Escudero-Uribe, "Simulation of a parallel mechanical elbow with 3 dof," *Journal of applied research and technology*, vol. 7, no. 2, pp. 113–123, 2009.

[47] H. . Pneumatics, "Mounting style can dramatically improve hydraulic and pneumatic cylinder performance."

[48] S. Tutorials, "How slider crank mechanism works."

[49] C. R. Source, "Acme lead screws and associated loads."

[50] H. K. M. Solutions, "Lead screws 101."

A
Nomenclature

Abbreviation	Name
ADL	Activities of Daily Living
DoF	Degree of Freedom
FE	Flexion/Extension
PS	Pronation/Supination
RoM	Range of Motion

B

Journal Version

Pneumatic Actuation for a Lightweight Elbow Prosthesis

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***Abstract*—Almost 57 percent of above elbow amputees stop using their prosthesis. One of the main factors of non-wear is the high weight. Pneumatic actuation can be more lightweight than electronic actuation, and might therefore help with reducing the mass of a prosthesis and increase its comfort. This paper presents the mechanical design of a pneumatically actuated elbow prosthesis to show the potential of pneumatic actuation in the design of prostheses. The design focused on the two main degrees of freedom of the elbow: flexion/extension and pronation/supination. Each degree of freedom can be controlled using independent actuators. A small-scale third actuator facilitates actuated disengagement of a locking mechanism, designed to reduce gas consumption. A prototype was built to determine experimental results for the validation of the theoretical design. The functionality of the prototype is similar to, or better than, electronic or body-powered elbow prostheses, whilst being more lightweight.**

***Index Terms*—Pneumatic, Elbow, Prosthesis, Lightweight**

I. Introduction

THERE may be over one million people with an above elbow upper extremity amputations worldwide [1] [2]. It is estimated that 80 percent of these amputees live in developing countries [2]. Despite progress achieved over the last decades, or even centuries, rejection rates are still high for elbow prostheses: 57 percent of above elbow amputees stop using their prosthesis compared to 6 percent of trans-radial amputees [3]. A survey in 2007 [4] asked a group of 266 amputees, 21 percent of which had a transhumeral amputation, for factors of non-wear. The responses can be clustered into the three basic requirements for a prosthesis [5]: control, comfort and cosmetics. An overwhelming 98 percent of prosthesis rejecters reported a lack of functionality (control) as a reason. 95 percent reported discomfort as a reason, with 88 percent of rejecters saying the prosthesis is too heavy. The appearance of the prosthesis played a role in not wearing the prosthesis for 70 percent of the rejecters. Progress in the field of elbow prostheses is required to reduce the number of prosthesis rejections.

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This paper is an abbreviated version of a thesis titled: The Power of Pneumatics. Additional information, such as background information, in-depth discussions, and test protocols, can be read in this thesis. The thesis "The Power of Pneumatics" can be found at https://repository.tudelft.nl/

A. Elbow Anatomy

The unique build of the anatomical elbow allows for two degrees of freedom: flexion/extension (FE) and pronation/supination (PS). A distinction should be made between different amputees. Elbow disarticulations are amputations through the elbow joint, transhumeral amputees have the amputation at some location through their humerus. The importance of this classification is the relevance for design. Because of the presence of an anatomical structure, there is less space for a mechanical structure in the case of elbow disarticulations. For transhumeral prostheses, more space is available for design aspects such as actuation units or power supply, depending on the location of amputation.

In a recent literature review [6], it's shown that each of the main types for actuation of elbow prostheses has their own drawbacks and areas of expertise. Body-powered prostheses eliminate the need for an external power-source, are lightweight and (can) provide proprioceptive feedback, but the harnass is cited as being uncomfortable and not visually appealing. Electric prostheses are strong and often visually appealing, but are known to be heavy, expensive and lack proprioceptive feedback. One of the conclusions of this literature review is that pneumatic actuation is underrepresented in the field of elbow prostheses. The benefits that pneumatic actuation offers align with the reasons for non-wear given by users, such as being lightweight. The prototypes that do exist, show that the often cited downsides of pneumatic actuation, such as a lack of control and safety, are overemphasized [7]. However, the last publication about a pneumatic elbow prosthesis dates back to 2008 [7].

II. Methods

Three phases were identified in order to develop a novel pneumatic elbow prosthesis: Analysis, Design and Testing.

A. Analysis

The analysis phase is aimed to achieve a better understanding of the design requirements. The analysis will be done from three different perspectives: Other prostheses (competitors), users, and literature. The different analyses led to the requirements summarized in Table I.

User Analysis

A user analysis will help with determining the requirements and wishes of the user, which may differ from the technical requirements and wishes. Since prostheses have a high rejection rate [3], user analyses are vital to achieve daily use of the prosthesis. A reliable analysis about device abandonment factors for upper-limb prostheses was published in 2007 by Biddiss et al. [4]. Multiple factors proved to play a role in device rejection, the most important ones are discussed below.

a) Control & Functionality: On the top of the list of factors for device abandonment is functionality. Respondents said they were just as functional or more without the device and that they had more sensory feedback without the prosthesis.

b) Comfort: Comfort is an important factor for the use of prostheses. In the review by Biddiss et al, a vast majority the respondents said they were more comfortable without the prosthesis. The most important factor for non-wear due to discomfort is weight.

c) Cosmetics: The users are human and therefore have their personal opinion regarding the appearance of a prosthesis. According to Biddiss et al, prosthesis rejecters were significantly less satisfied. Furthermore, appearance can be more important than functionality or comfort in some countries because of social, economic, cultural, psychological or religious reasons [8].

Literature Review & Competitor Analysis

A literature review ([6]) concerning actuation for elbow prostheses has been performed prior to starting the design process. Other elbow prostheses were studied to find additional functions and used to quantify the requirements.

B. Design

Based on the requirements, a novel design was conceived. A render of this design is shown in Fig. 1. The design process from conceptualization to embodiment can be found in the supplementary information.

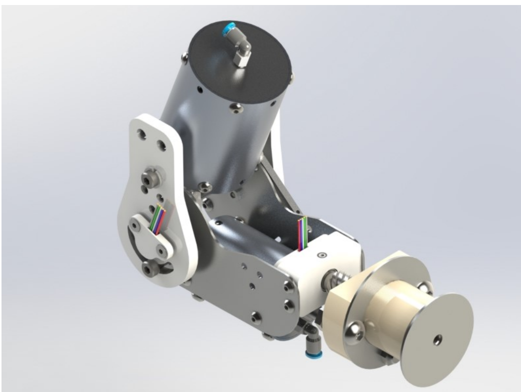

Fig. 1: The CAD-Model of the pneumatic elbow prosthesis

TABLE I: An overview of the requirements for the prosthesis, based on an average man, weighing 80 kilograms and having a length 180 centimeters

Requirement	Value		Comment
Requirements based on Cosmesis			
Length Forearm	< 87	mm	Goal
	< 174	mm	Maximum
Length Upper Arm	< 50	mm	Goal[1]
	< 113	mm	Goal
	< 170	mm	Maximum
Requirements based on Comfort			
Weight	< 800	gr	Goal
	< 1000	gr	Maximum
Sound Level	< 20	dB	Goal
	< 50	dB	Maximum
Cycles of Use until Power Depletion	> 300		Goal
	> 100		Minimum
Requirements based on Control & Functionality			
Flexion/ Extension	0-150	°	Goal
	5-135	°	Minimum
Pronation/ Supination	± 90	°	Goal
	± 50	°	Minimum
FE Lifting Payload	> 4	kg	Goal
	> 1.5	kg	Minium
FE Holding Payload	> 23	kg	Goal
	> 6	kg	Minimum
PS Torque	> 4.2	Nm	Goal
	> 1.5	Nm	Minimum
FE Stroke Time	< 0.7	s	Goal
	< 1.5	s	Maximum
PS Stroke Time	< 0.5	s	Goal
	< 1.0	s	Maximum

1. To fit with patients with an elbow disarticulation

a) Prototype Gas Supply: The prototype does not entail a build-in gas supply. The gas consumption depends on the intensity of the activity and the stroke length of the actuators. The consumption will be estimated using the ideal gas law. The used medium is CO_2. The amount of gas used for a minimal intensity (minimal force, full RoM) cycle is 0.12 [g]. For a maximum performance cycle this is 0.95 [g]. If one out of twenty movements requires full power and with 25 [%] losses, the expected average gas consumption is 0.20 [g] for a full RoM movement.

b) FE Holding Payload: Due to an force-amplified mechanism based on chapter 5.5 of "Werktuigkundige Systemen" [9], the locking moment can achieve holding moments $\rightarrow \infty$. The limiting factor is a combination of the compressive modulus and yield strength of the brake disc.

C. Manufacturing & Assembly

A prototype of the newly designed prosthesis was manufactured, see Fig. 2. Manual control is desired for testing different aspects of the prototype independently. Manually controlled valves are significantly larger (and heavier) than electronic valves. These valves are too big to be implemented on the frame of the prototype and were placed on a separate control board.

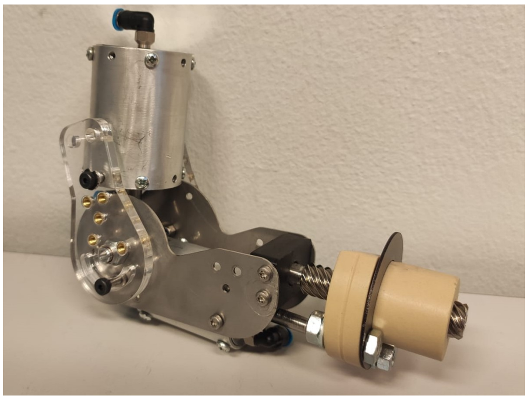

Fig. 2: The prototype of the pneumatic elbow prosthesis

D. Testing

The goal of the testing is to measure if the prototype achieves the designed values/requirements. The following parameters will be tested/measured:

- Mass
- Stroke Times
- Pronation/Supination Torque Output
- Flexion/Extension Lifting Payload
- Sound Level

a) Data Acquisition: The stroke times were recorded using the 60 [fps] camera of the Xiaomi 9T. For the pronation/supination torque, a load cell was used to determine the radial force at a distance of 34 [mm] from the lead screw axis (model: FUTEK Miniature S-Beam Jr. Load Cell). The force data was fed to a data acquisition (DAQ) device (model: NI USB-6008, 12-bit, 10 kS/s) and into the computer. The angles for the varying FE payloads were captured by taking a photograph at the maximum angle using the camera of the Xiaomi 9T. The sound levels were measured at a distance of 10 [cm] from the valves, using a decibel meter (model: Center 322 Sound Level Meter).

b) Data Processing: The measurements involving the PS torque were averaged per pressure and poly-fitted using a first order polynomial. For the FE payloads, the measurement were averaged per payload and poly-fitted using a third order polynomial.

III. Results

TABLE II: The measured mass for the sub-assemblies

Sub-Assembly	Mass
FE-Actuator	154 [g]
PS-Actuator	220 [g]
Frame Sheets	75 [g]
LM-Actuator	22 [g]
Covers	70 [g]
Control Circuit	705 [g]
Full Prototype [2]	1278 [g]

1. Without valves, cover, and locking mechanism
2. With all components

a) Additional Notes on Prototype Weight: Some of the components that were required for the prototype, such as the valves, have significantly lighter available versions. The main example are the valves/control circuit. If electronic control is used instead of manual control, The prototype can be 460 [g] lighter. Furthermore, the designed prosthesis has more functionality than most conventional prosthesis, such as active pronation/supination in the elbow. The removal of the additional functionality could reduce the weight to less than 600 [g].

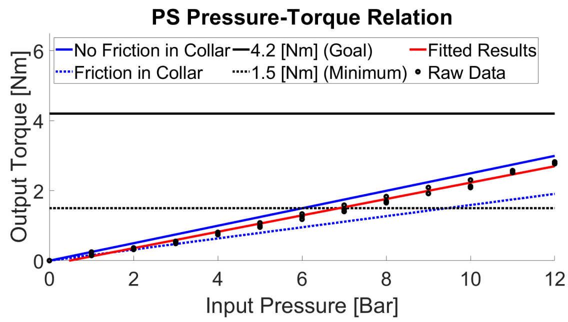

Fig. 3: The results of the PS torque test

Whilst testing, it was discovered that the prototype frame bends inwards for flexion angles above 90 [°], greatly increasing the friction.

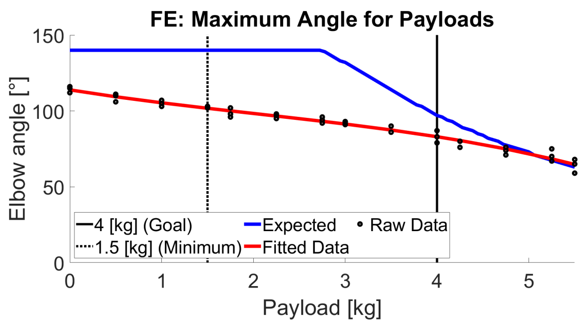

Fig. 4: The largest achieved flexion angle, pressurized at 12 [bar]

Table III compares the results of the prototype to the requirements found in the analysis phase. If a requirement has not achieved its goal, possible reasons and solutions are discussed.

TABLE III: The results of the prototype compared to the requirements. Green: The goal is achieved, Yellow: The minimum/maximum value is not surpassed, and Red: The minimum/maximum value is surpassed

Requirement	Prototype
Requirements based on Cosmetics	
L Forearm	172 [mm]
L Upper Arm	144 [mm]
Requirements based on Comfort	
Weight	600 - 1278 [g]
Sound Level	52 [dB]
Cycles of Use until Power Depletion[1]	125 [2]
Requirements based on Control	
Flexion/Extension RoM[1]	0-140 [°]
Pronation/Supination RoM	$\pm$ 90 [°]
FE Lifting Payload[1]	0-4 [kg]
FE Holding Payload	> 6 [kg] [2]
PS Torque	2.8 [Nm]
FE Stroke time	0.36 [s]
PS Stroke time	0.30 [s]

1. Highly correlated

2. Theoretical

IV. Discussion

a) Length: The limited stroke of a pneumatic cylinder restricts the possible range of motion, thus in order to achieve the required ranges of motion, the cylinder stroke length had to increase.

b) Weight: The weight of the prototype exceeds the maximum by almost 280 [g]. Switching to electronic control decreases the prosthesis mass by 460 [g]. Despite not being lighter than the goal, the prosthesis would still be lighter than comparable prostheses, whilst achieving equivalent, or better, functionality. If additional functionality such as the active pronation/supination is removed, the "bare-bones" version can weigh less than 600 [g].

c) Sound Level: The sound level was measured at a distance of 10 [cm] without a barrier between the source (valves) and the sensor, and even then only slightly exceeded the maximum allowed sound level. If the valves are implemented within the frame in a next iteration, the covers will function as barrier, reducing the sound level.

d) Cycles of Use until Power Depletion: The cycles of use until power depletion have not been tested as no power supply has been implemented. If a 25 [g] CO_2 cartridge is implemented as the gas supply, an average of 125 cycles of use can be achieved until depletion. In reality, the gas consumption depends heavily on the type of use.

e) Flexion/Extension RoM: The limited stroke size of a pneumatic actuator restricts the RoM. Achieving the full range of motion would add additional mass to the system and cause the FE-torque-angle curves to differ more, leading to less efficient actuation. A range of motion of 0-140 [°] should be enough for almost all activities of daily life. The prototype cannot actively achieve 140 [°], because of the increased friction at higher angles of flexion. It can be achieved by manually moving the forearm.

f) FE Lifting Payload: In theory, 2.5 [kg] can be lifted throughout the entire range of motion, and 4.0 [kg] can be lifted to almost 100 [°] flexion. For the prototype, the desired FE-payload of 4 [kg] can be lifted to a maximum of 87 [°]. Even with no payload in hand, the maximum achieved angle is 115 [°]. If moved manually, 140 [°] flexion can be achieved.

g) FE Holding Payload: The shown holding payload of > 6 [kg] is theoretical. In a next iteration of the prototype, the force-amplified locking mechanism should be tested.

h) PS Torque: The desired PS-torque of 4.2 [Nm] is not achieved. This was a calculated decision to help reduce the mass of the prototype. The goal of 1.5 [Nm] has been easily surpassed.

V. Conclusion

This paper presents the design of a pneumatically powered elbow prosthesis. The prototype has an equivalent functionality to commercial and academically proposed prostheses:

- The flexion/extension payload depends on the desired lifting angle.
  - 4.0 [kg] can be lifted to 87 [°].
  - In theory, 2.5 [kg] can be lifted throughout the entire range of motion, up to 140 [°].
- The pronation/supination torque is over 2.8 [Nm] throughout the entire range of motion of $\pm$ 90 [°]
- The locking mechanism is theoretically capable of passively holding over 6 [kg] in any position
- In theory, an average of 125 cycles of use can be achieved per 25 [g] CO_2 cartridge.
- The weight of the prototype is $\approx$ 1250 [g]. With minor changes the weight can be reduced to 900 [g].
- Removing pronation/supination and the active locking mechanism reduces the weight to less than 600 [g].

The combination of high functionality and the low mass of the prototype shows a potential for pneumatic actuation for prostheses.

Declaration of conflicting interests

The author declared no potential conflicts of interest with respect to the research, authorship, and/or publication of this article.

References

[1] K. Ziegler-Graham, E. J. MacKenzie, P. L. Ephraim, T. G. Travison, and R. Brookmeyer, "Estimating the prevalence of limb loss in the united states: 2005 to 2050," *Archives of physical medicine and rehabilitation*, vol. 89, no. 3, pp. 422–429, 2008.

[2] M. LeBlanc. (). "Give hope - give a hand - the ln-4 prosthetic hand," [Online]. Available: https://web.stanford.edu/class/engr110/2011/LeBlanc-03a.pdf. (accessed:05.03.2020).

[3] T. W. Wright, A. D. Hagen, and M. B. Wood, "Prosthetic usage in major upper extremity amputations," *Journal of Hand Surgery*, vol. 20, no. 4, pp. 619–622, 1995.

[4] E. Biddiss and T. Chau, "Upper-limb prosthetics: Critical factors in device abandonment," *American journal of physical medicine & rehabilitation*, vol. 86, no. 12, pp. 977–987, 2007.

[5] D. H. Plettenburg, *Upper extremity prosthetics: current status & evaluation*. Delft University of Technology, 2002.

[6] J. Kieft, "A state-of-the-art review of actuation for powered elbow prostheses," *TU Delft Literature Review*, 2020.

[7] K. B. Fite, T. J. Withrow, X. Shen, K. W. Wait, J. E. Mitchell, and M. Goldfarb, "A gas-actuated anthropomorphic prosthesis for transhumeral amputees," *IEEE Transactions on Robotics*, vol. 24, no. 1, pp. 159–169, 2008.

[8] D. Cummings, "Prosthetics in the developing world: A review of the literature," *Prosthetics and orthotics international*, vol. 20, no. 1, pp. 51–60, 1996.

[9] J. C. Cool, *Werktuigkundige systemen; 3e dr.* Delftse Uitgevers Maatschappij, 1992.

[10] D. Doedens, "Optimal co2 pressure for a pneumatic system," *Master Thesis*, 2015.

[11] C. S. Harkins, A. McGarry, and A. Buis, "Provision of prosthetic and orthotic services in low-income countries: A review of the literature," *Prosthetics and Orthotics International*, vol. 37, no. 5, pp. 353–361, 2013.

[12] F. Malagelada, M. Dalmau-Pastor, J. Vega, and P. Golano, "Elbow anatomy," *Sports injuries: prevention, diagnosis, treatment and rehabilitation*, vol. 2, pp. 527–53, 2014.

[13] A. Ramirez-Garcia, C. Toledo, L. Leija, and R. Munoz, "Status of elbow myoelectric prosthesis: Cinvestav-ipn prosthesis," *Revista Mexicana de Ingenieria Biomedica*, vol. 30, no. 1, pp. 66–73, 2009.

[14] D. H. Plettenburg, "Basic requirements for upper extremity prostheses: The wilmer approach," in *Proceedings of the 20th Annual International Conference of the IEEE Engineering in Medicine and Biology Society. Vol. 20 Biomedical Engineering Towards the Year 2000 and Beyond (Cat. No. 98CH36286)*, IEEE, vol. 5, 1998, pp. 2276–2281.

[15] G. Smit. (). "Delft cylinder hand: Natural grasping," [Online]. Available: https://www.bitegroup.nl/category/prosthetic-devices/delft-cylinder-hand/. (accessed:07.08.2020).

[16] Ottobock. (). "Above-elbow prosthesis with dynamicarm," [Online]. Available: https://www.ottobockus.com/prosthetics/upper-limb-prosthetics/solution-overview/above-elbow-prosthesis-featuring-dynamicarm/. (accessed:06.08.2020).

[17] ——, (). "Body-powered upper limb prostheses," [Online]. Available: https://www.ottobock-export.com/en/prosthetics/upper-limb/solution-overview/arm-prostheses-body-powered/. (accessed:06.08.2020).

C

Literature Review

A State-of-the-Art Review of Actuation for Powered Elbow Prostheses

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***Abstract*—**
***Background*: Despite progress achieved over the last decades, rejection rates are still high for elbow prostheses. The reasons for non-wear can be clustered into three categories: Control, Comfort, Cosmetics. Further development of elbow prostheses is needed to decrease the levels of device abandonment.**

***Objective*: This review has a primary and secondary objective: (1) To give an overview of the current state-of-the-art of powered elbow prostheses, with a focus on actuation and (2) prove the hypothesis "Pneumatic actuation is underrepresented in the field of upper extremity prostheses"**

***Methods*: Five databases were searched using a Boolean combination of relevant keywords: IEEE Xplore, Scopus, Web of Science, PubMed & Espacenet.**

***Results*: 64 articles were reviewed in total. The advantages and disadvantages of the three main types of actuation; electric, body-powered, and pneumatic, have been summarized in a single table. A second table lists the most prominent elbow prostheses designed or updated in the past twenty years. A trend analysis shows the percentage of type of actuation used in design, from the 1950's until the current day.**

***Conclusion*: The low expected mass for a pneumatic prosthesis might be a solution for one of the biggest reasons of prosthesis rejection: Low comfort. Furthermore, raised disadvantages of pneumatic prosthesis, such as control and safety, are overemphasized. With only one out of twelve most prominent prostheses using pneumatic actuation, it can be said that pneumatic actuation for upper extremity prostheses is underrepresented.**

***Index Terms*—Transhumeral, Elbow, Prostheses, Actuation, Electric, Body-powered, Pneumatic, Review,**

I. Background

It was estimated that there will be 2.2 million amputees in the USA alone by 2020 [1]. Thirty percent of all amputations are upper extremity amputations, 28 percent of which are above elbow [2]. This leads to a required $\approx$ 180 thousand elbow prostheses in the USA. It is estimated that 80 percent of the amputees live in developing countries [2]. There is a huge global market for elbow prostheses. Despite progress achieved over the last decades, or even centuries, rejection rates are still high for elbow prostheses: 57 percent of above elbow amputees stop using their prosthesis compared to 6 percent of trans-radial amputees [3]. A survey in 2007 [4] asked a group of 266 amputees, 21 percent of which had a transhumeral amputation, for factors of nonwear. The responses can be clustered into three categories as proposed by Plettenburg in 2002 [5]: Control/Functionality, Comfort and Cosmesis. An overwhelming 98 percent of proshesis rejecters reported a lack of functionality as a reason. 95 percent reported comfort as a reason, with 88 percent of rejecters saying the prosthesis is too heavy. The appearance of the prosthesis played a role in not wearing the prosthesis for 70 percent of the rejecters. Further progress in these, and more, fields is required to improve prosthesis design and decrease device abandonment. For innovative design, two aspects should be considered: knowing the advantages and disadvantages of certain design choices and knowing what others have done.

The main topic of this review will be methods of actuation. Because of their importance to understanding prosthesis design, topology and control will be briefly discussed. The primary goal of this literature study is twofold (1) to create a clear summary of advantages and disadvantages for the actuation methods for powered elbow prostheses and (2) to generate an overview of the existing prostheses. These subgoals can be combined as:

To give an overview of the current state-of-the-art of powered elbow prostheses, with a focus on actuation

Whilst researching the different types of actuation, an pneumatic actuation seemed to be underrepresented. Because of the relevance to the state-of-the-art of such an underrepresentation, a secondary goal of this review is to prove the hypothesis:

Pneumatic actuation is underrepresented in the field of upper extremity prostheses

II. METHOD

A. Primary Goal: State-of-the-Art Review

The following search electronic engines are used to scope as much in the field of elbow prostheses as possible: IEEE Xplore, Scopus, Web of Science and PubMed. The searches were performed during the month of February 2020. The following string of key words and Boolean operators was used to search in the title, abstract and keywords, this is visually represented in Figure 1. The results have been limited to the English language. After eliminating the duplicate entries, the unique entries remained will be screened using the inclusion criteria to filter the relevant papers. Further publications are found using the snowball method.

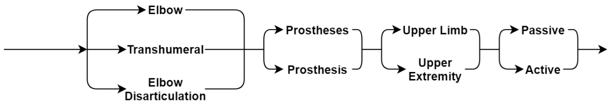


Fig. 1: A visual representation of the search query for the primary goal. Parallel elements mean a Boolean "OR" and elements in series means "AND"

1) Inclusion Criteria: An entry is regarded to be relevant when its topic considers the design, development, use or review of (1) an arm prostheses including the elbow, (2) solely an elbow prostheses, (3) a certain vital aspect for prostheses (e.g. a review about types of actuation). Articles considering elbow arthroplasty (inner prostheses), surgical articles and EMG-only articles are excluded. With a previous study by Carey et. al [6], 260 articles were excluded because of their focus on development of specific control algorithms or EMG processing but did not test these with actual prosthetic devices or prosthesis users. In some cases, several articles are published about the same design, without any changes to said design. Of these articles, the last published article is reviewed in this study.

B. Secondary Goal: Research into Pneumatic Popularity

The analysis will be done in twofold, quantitative and qualitative. For the quantitative analysis, two graphs will be generated relating different types of actuation of the years, one concerning articles and one concerning patents. Articles and patents are two distinctly different publications. Articles relate to academic research whereas patents relate to commercial viability. The previouly mentioned search engines are used to find articles. For patents, the used search engine is Espacenet. The string used is visualised in Figure 2. This search query will try to cover all publications from the 1960's until the present day. Per year, the percentage of articles/patents resulting from the search has been determined. These will be averaged and curve fitted (using MATLAB "polyfit") within their category: articles or patents. A similar quantitative trend analysis is done for attention paid to different upper extremity prostheses: hand, elbow, and shoulder. This analysis does not fit within the narrative scope of this review and will therefore be discussed in the appendix.

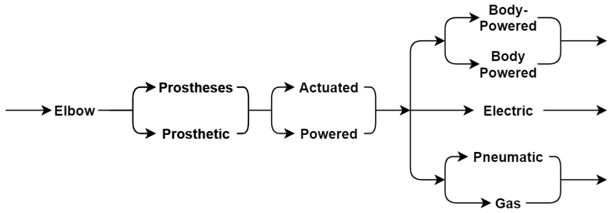


Fig. 2: A visual representation of the search query used for the secondary goal. Parallel elements mean a Boolean "OR" and elements in series means "AND"

For the qualitative analysis, reviews concerning actuation for upper extremity or elbow prostheses are read. The attention paid to pneumatics, or lack thereof, will be discussed. These reviews will be picked from the publications found using the search query for the primary goal.

III. RESULTS

The results of the first, qualitative, analysis are visualised in Figure 3. The results of the first query will be discussed in three categories: Anatomy and Topology, Control, and Actuation. The results of the trend analysis are shown in Section VIII.

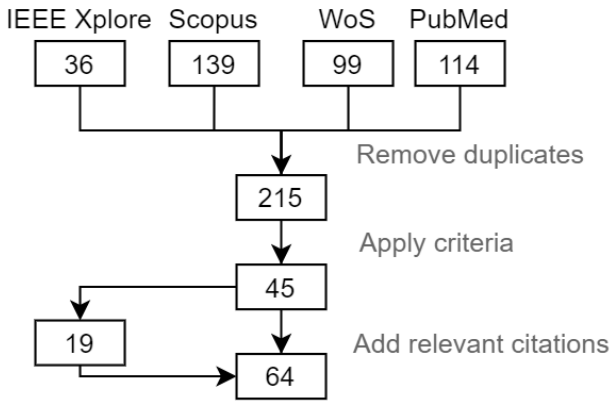


Fig. 3: A visual representation of the search methodology and the results quantified

IV. Anatomy and Topology

A distinction should be made between different above elbow amputees. Patients with a functioning shoulder, but missing an elbow can be classified in one of two classifications: Elbow disarticulations are amputations through the elbow joint, transhumeral amputees have the amputation at some location through their humerus. The importance of this classification is the relevance for design. Because of the presence of an anatomical structure, there is less space for a mechanical structure in the case of elbow disarticulations. For transhumeral prostheses, more space is available for design aspects such as actuation units or power supply, depending on the location of amputation.

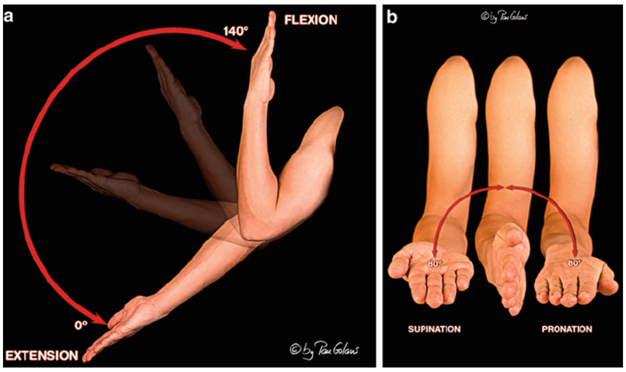


Fig. 4: The degrees of freedom of an elbow joint. Taken from [7]

Most elbow prostheses have been designed with a single degree of freedom: flexion/extension (EFE). However, the unique build of the anatomical elbow actually allows for two degrees of freedom: flexion/extension (FE) and pronation/supination (PS) Figure 4 [8]. Instead of a combined workspace for the elbow, the pronation/supination is often generated by a wrist movement, independent of the elbow [9]. PS is even regarded as the most important wrist movement, surpassing wrist flexion/extension (WFE) and radial/ulnar deviation (RUD) in terms of desirability by the users [10]. For the purposes of this literature study, humeral rotation will be regarded as a motion finding is origin in the shoulder joint, thus it will not be discussed in-depth. Two layers of a parallel and series topography will be discussed next; (1) EFE and PS are done in parallel or series, (2) within either EFE or PS, actuators can work in parallel.

A. Parallel and Series Topography

The musculo-skeletal system of the human body more often than not works with antagonistic muscle pairs, where as one muscle contracts, the other relaxes. The biceps/triceps combination is such a muscle pair. The contraction of the biceps will cause the elbow to flex and an triceps contraction will cause extension. These muscles work parallel to each other. A method to provide the prosthesis with both EFE and PS at the elbow levels, is to use parallel actuation. If a series actuation is used, a prosthesis is more likely to be less efficient. E.g. if PS would follow EFE in a myoelectric elbow prosthesis, the motors to achieve powered PS would be dead weight for the EFE motors, causing a higher power consumption and lower torques/velocities [9]. When using a pair of linear actuators, like the muscles of the human bodies, the antagonist can help the agonist by pushing, unlike the human body, where only pulling is possible. This will improve the torque density and power/weight ratio of the prosthesis. Rotational actuators, like electric motors, are often designed on-axis. As this is a single motion, this is neither series nor parallel. The advantage of serial systems are that, more often than not, they are simpler and easier to make modular.

V. Control

Over the decades of research intro prostheses, several control methods have developed. The choice of control method and actuation method go hand in hand. E.g., body-powered prostheses have a better synergy with straps as a control method, because neither rely on an external power source. However, it is important to have a clear distinction between control and actuation.

The world outside a laboratory is full of uncertainties. Prostheses, when used in the real world, are subject to external disturbances. The human body deals with these disturbances by having a great number of feedback loops. A well-functioning anthropomorphic prosthesis should apply feedback to the operator for optimal control [5]. While there are research groups considering feedback, including thermal and tactile, within their (electric) prosthesis [6], body-powered prostheses are the only prostheses with proprioceptive feedback at the moment. Without proprioceptive feedback, vision closes the control loop, which is mentally more draining. Proprioceptive feedback will work optimally if there are direct relations between the input and output parameters of the prosthesis, such as position or velocity. Furthermore, it should be simple enough for the user to understand and feel natural. There have been some types of control which are now hardly used for the control of prostheses and will therefore not be discussed/explained in this actuation-focused review; (1) Foot Control [11], (2) Muscle Bulging [5], (3) Myo-acoustics [5].

A. The most common types of control

a) Body-powered harness: In 2002, straps were considered the most commonly used method to control an active prosthetic device [5]. Dozens of different configurations of cables were used to achieve movement of the prostheses. Most strap control methods worked by changing the distance from a fixed point and guiding the displacement using a Bowden cable. Sometimes, the same cable could generate different movements, depending on whether or not the elbow was locked. For elbow control with a body-powered prosthesis, a shoulder harness is often used as a control method [5]. One of the problems for control using straps, and therefore the human body, is the limited number of independent movements of the human body.

b) EMG: The skin surface electromyogram (EMG) signals of user's residual muscles are used as the input signal to control the prosthesis. Myoelectric control were studied in laboratories and brought to the market during the late 50's. The simple and robust control scheme scales linearly with electrical activity of the residual muscles in the stump. This concept of direct proportional control is still in use to date [12]. An unamplified signal is difficult to read out from the skin surface, therefore a technique called Targeted Muscle Reinnervation (TMR) is used to ensure a usable signal. In short, with TMR additional nerves are placed in the stump and a layer of subdermal fat is removed to reduce the amount of surface area required to read a signal. TMR also creates more independent control sites by isolating muscle segments using fat, thus more freedom of control of the prosthesis is possible.

c) Direct muscle attachment (Cineplasty): At the dawn of the twentieth century Giuliano Vanghetti was the first to aim at exploiting natural movements of the remnant muscles to activate the mechanical prosthesis. By doing so, he was the first to directly connect muscles and tendons to a mechanical prosthesis [13]. This work eventually led to the modern day neuroprosthetic control methods. The main working principle has not changed drastically over the years. An antagonistic muscle pair is used to generate either a signal to move, or the movement itself. An example is a biceps/triceps combination, where a biceps contraction corresponds to the opening of the prosthetic hand and the triceps contraction corresponds to the closing of the prosthesis. The human operator is "in-the-loop" and should have a right amount of control of the prosthesis. Due to the invasive nature of this control method and the surgery required, in combination with a lack of proper hygiene for a long time, this control method had fallen into oblivion. A final nail in the coffin of cineplasty is its relatively unattractive appearance [5].

d) Cortical Control: Perhaps the most modern type of control is cortical control. The goal of cortical control is to have the human operate the prosthesis with their thought and create a truly integrated system [14]. Johannes et al. describe their goal of closed-loop cortical control as neural decoding and sensory encoding. This means that the signals coming from the brain should be translated into motion (neural decoding) and the prosthesis should be able to generate a "feeling" by stimulating nerves that lead to the brain (sensory encoding). Currently, the only facility using this type of control is the Applied Physics Laboratory (APL) of the John Hopkins University. Neural decoding is achieved by wireless myobands which read out the nerve activity, enhanced by TMR, and translate this into use able signals for the prosthesis [14]. Sensory encoding is to be achieved using vibrotactile sensors to activate brain implants in order to generate haptic feedback and thermal [15]. As one might expect. the algorithms used to translate electrical signals into neural signals can be quite complex. Research like this does have a price-tag attached to it as over 120 million dollar has been spent into developping this type of control. In 2015, APL had ten Modular Prosthetic Limbs, costing around $500,000 each [16].

VI. Actuation

In general, the actuation for prostheses can be divided into passive and externally powered prostheses [17]. Passive can be subdivided into subcategories such as static and adjustable [18]. Static prostheses are mostly designed for cosmetic reasons. A cosmetic prosthesis fulfils these needs without having to bother itself with the issues of moving components and can therefore be much more lightweight. Adjustable prostheses provide the operator with the option to move the device. They are often combined with a locking mechanism, where the device can be locked in a certain position. This is useful for e.g. changing the position of the elbow from a sitting position at a desk (elbow flexed at 90 degrees) to a standing position (elbow joint in its anatomical neutral position). A downside of passive prostheses is that they cannot move on their own and therefore need outside assistance, mostly from the sound arm. Powered prostheses can be further subcategorized into body-powered, electric, pneumatic and non-conventional actuation. Because of the immense differences between the classes, patients often have more than one type of prostheses. For instance, a passive prostheses when doing desk work, but changing into a more rugged body-powered prostheses when working in the garden. A simplified overview of the classification method for prostheses has been visualised in Figure 5.

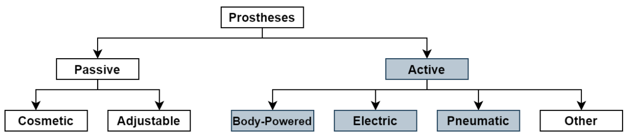


Fig. 5: A simplified overview of the classification methods for prostheses. The colored boxes are the focus of this article

The development of an elbow prosthetic device faces the same problems as most other, especially upper-limb, prostheses. These problems range from control, comfort, costs, power to weight ratio, to some lesser-known issues such as noise, manufacturing and heat. The chosen type of actuation plays a vital role in these parameters [19]. Because each type of actuation has its own benefits, hybrid designs have been developed, e.g. an electrical hand prosthesis coupled with a body-powered elbow [17].

One of the first mentions of a body-powered arm is the Ballif arm in 1812 [20], see Figure 6a. The first pneumatic hand was developed at the beginning of the 20th century, soon followed by the first electric-powered hand. At the end of the Second World War, early concepts of myoelectric prostheses were introduced [21]. At the moment, there are some more obscure methods such as piezoelectric materials and new innovations such as the Peano-HASEL artificial muscle [22]. Comparing different types of actuation to find which is superior, has been proven to be difficult, as stated by Carey et al. [6]. According to some articles, electric prostheses have become the norm, with body-powered prostheses following close on second place [23] [6]. According to other sources, passive and body-powered prostheses hold the dominant position within the field of elbow prostheses [12]. Biddiss et al. [24] mention that individuals with a higher level amputation, such as an above-elbow amputation, are more likely to use, and keep using, a body-powered terminal device.

(a) One of the oldest elbow prostheses: The Ballif Arm as visualised in the book by Pierre Ballif in 1818 [25]

(b) One of the most modern elbow prostheses on the market today: The Ottobock Dynamic Arm [26]

Fig. 6: A visualization of the trends regarding electric, body-powered and pneumatic actuation for elbow prostheses

A. *The most common types of actuation*

a) Body-Powered: A body-powered prosthesis works by harnessing the power of the human body. Since it does not need an external power source, body-powered prostheses are the oldest type of powered prostheses and are often called ”classic”. Body-powered prostheses are environmentally robust, relatively low cost, and offer some degree of proprioception and force feedback to the user [21].

b) Myoelectric: A myoelectric prosthesis uses electric motors for actuation. The most common control method for electric actuation is EMG. Unlike body-powered prostheses, myoelectric prostheses do not require donning or use of a shoulder harness, and also typically allow operation throughout a larger range of motion. [23]. Furthermore they exhibit a low torque density (relatively to human joint actuation), which results in an actuator with a power density that is approximately three to five times less than human skeletal muscle [27].

c) Pneumatic: Pneumatic actuation, or gas actuation, often uses a compressed gas to power the prosthesis. The gas is stored under high pressure in a container, causing a phase-transition into a liquid. Therefore, pneumatic actuation is also referred to as a liquid fuel cell. In the past this was done with carbon dioxide, but the more recent versions by Fite et al. [27] use the monopropellant ”hydrogen peroxide”, which can have an energy density about 5 to 12 times as high as carbon dioxide. Pneumatic actuation does not require heavy electric motors and can therefore be made more lightweight. A common argument for not using pneumatic actuation is noise. This has been brought up by Fite et al, but disputed almost immediately. As an example they use their own prosthesis, which produces noise of 50 dB or less from 1 metre away. According to Fite et al. this can be considered as ambient noise in most situations. This claim is supported when looking at ambient sound levels, where 60 dB is a normal conversation [28]. A noteworthy observation should be included; despite the promising developments of the Vanderbilt arm, no updates have been published since 2008.

TABLE I: An overview of the pros and cons for the two major types of actuation used at the moment: Body-powered and Electric [1], and the type of actuation used more prominently in the 1960's and 70's: Pneumatics[2]. The table is organised to follow the main design principles when designing for a prosthesis: Comfort, Control, Cosmesis [5]. Aspects not fitting within these categories or special features are shown at the end. There are two disputed entries (coloured yellow), these will be discussed in section IX

Metric	**Myoelecric**	**Body-Powered**	**Pneumatic**
Comfort	Medium	Low [6][24]	Unknown[2]
Range of Motion	Large [23] [29]	Limited due to harness	Limited due to short stroke [14]
Weight	Heavy [24]	Light [4][30]	Light [31]
Compliance	Low, because of rigid motors [32]	-	High, because of soft actuators [32][33]
Costs	High [6][8][29][24]	Low [6][30]	Low [33]
Sound	Somewhat loud [24], but negligible compared to ambient sound [27]	Silent Actuation [30]	Loud, but negligible compared to ambient sound [27]
Suspension[4]	Comfortable: No Harness [23]	Uncomfortable: Body Harnass [34][10]	Comfortable: No Harness[3]
Maintenance	Difficult and Expensive [6]	Easy and Cheap [6]	Low/None [33]
Training Time	Long [6]	Short [6]	Unknown[2]
Control	Good [8][29][27]	OK [6] [30]	Poor [12], disputed by [27]
Bandwidth[4]	Good [27]	-	Appropriate for human movement [27]
Feedback[4]	Thermal and tactile [6]	Proprioceptive [6][30]	None
Power/Weight	Reasonable [27]	-	High [35]
Durability	Low [8][29][24][35]	High [6] [30]	Medium - High
Efficiency	Without gears: High With gears: Low [8][36]	-	Without leaks: High With leaks: Varies
Grasp	Superior pinch force [29]	Weak(er) grasp [6]	Unknown[2]
Usability	Unsuitable for heavy work [8]	Suitable for heavy work [6]	Unknown[2]
Cosmesis	Good [6][8][29]	Poor [6]	Good[3]
Shape	Not Antropomorphic [27]	Cables [34][10]	Antropomorphic [27]

	Pros	**Cons**	**Pros**	**Cons**	**Pros**	**Cons**
Other Aspects & Features	Positive Effects on Phantom Pain [6]	-	Frequency of adjustment [6]	-	Power Regeneration [36]	Packaging [14][27]
	-	-	-	-	-	Unsafe: High Temperature and Pressure [14], disputed by [27]

[1]The paper by Carey et al. [6], titled "Differences in myoelectric and body-powered upper-limb prostheses: Systematic literature review" provided a majority of the information used considering body-powered and electric prostheses.

[2]Relatively little information is known about pneumatic actuation for prostheses, compared to electric and body-powered actuation

[3]No sources, but based on the similarities with electric actuation

[4]Not fully related to type of actuation. E.g. Multiple possibilities for a body Suspension exist independent of the actuation and feedback also depends on type of control

TABLE II: An overview of the state-of-the-art of powered elbow prostheses

Elbow Prosthesis[1]	**Creator(s)**	**Topology**	**Actuation**	**Control**	**Commercially Available?**
CINESTAV-IPN [9] [37][38]	The National Polytechnic Institute, Mexico	Parallel	Electric	EMG	No
Motion E2 Elbow [39]	Fillauer	Series	Body-powered	Body-powered	Yes
The Boston Elbow	Liberating Technologies Inc.	Series	Electric	EMG	Yes
The DEKA Arm [2] [40]	DEKA Intergrated Solutions Corp.	Series	Electric	Multiple[3]	No
The Edinburgh Arm [17] [5]	Princess Margeret Rose Orthopaedic Hospital	Series	Electric	Body-powered pressure pads	No
Espire Pro [41]	RSL Steeper Inc.	Series	Electic	EMG	Yes
The Modular Prosthetic Limb [14][42]	John Hopkins Applied Physics Laboratory	Series	Electric	Closed-loop Cortical Control	No
The Ottobock Dynamic Arm+ [26]	Ottobock	Series	Electric	EMG	Yes
The RIC Arm [43]	The Rehabilitation Institute of Chicago	Series	Electric	EMG	No
The UTAH Arm [44]	Motion Control Inc.	Series	Electric	EMG	Yes
The Vanderbilt Arm [27]	The Vanderbilt University	Series[5]	Pneumatic	Unknown	No
The Creighton Arm [45]	Creighton University	Series	Body-powered	Body-powered	No

[1]When no name is stated, the name of the university or first author is used

[2]The DEKA arm is sometimes referred to as ”the Luke arm”, named after Star Wars protagonist Luke Skywalker

[3]The DEKA arm works with several input methods such as, foot controls, EMG, Pneumatic bladders and ”other commercially available input devices”

[4]The Edinburgh arm was designed in 1999, but deemed relevant enough

[5]The Vanderbilt arm has some parallel elements on the forearm level

VII. Overview

The goal of Table II is to show an overview of the commercially available and most prominent elbow prostheses in the current state-of-the-art. Elbow prostheses predating the 21^{th} century, with no update in the past twenty years, will not be listed. For a full scope of the field of elbow prostheses, these older prostheses are listed below, including the date of the latest publication.

- The IBM Arm [46] (1954)
- The INAIL Elbow [8] (1974)
- The VA Elbow [47] (1975)
- Princess Margeret Rose Elbow [48] (1988)
- The NY Electric Elbow [49] (1989)
- The VASI Arm [50] (1991)
- The MELA Arm [51] (1996)
- The Proto-1 (2006) & Proto-2 (2007), which led into the Modular Prosthetic Limb (see Table II) [42]

Aside from the prostheses listed in Table II, there are several small research groups doing similar research into upper limb prostheses for above the elbow amputees, either at the moment or in the past twenty years. These groups all use myoelectric prostheses:

- The University of Moratuwa [52]
- The Vanderbilt University [23]. This is a different model than the one listed in Table II.
- The Gen 2 [53]
- The Saga Univesity [54]

VIII. Decline of Pneumatics?

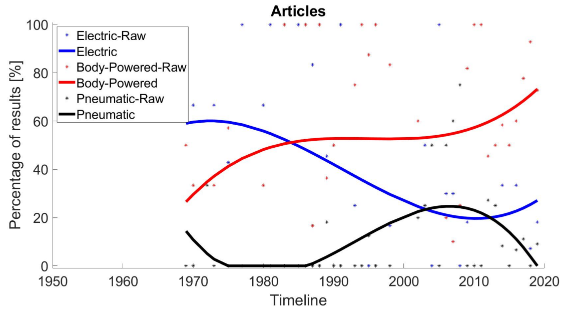
As described in Section II-B, to research a perceived decrease of attention towards pneumatic actuation within upper extremity prostheses a trend analysis has been done, both in a quantitative and in a qualitative manner.

A. Trend Analysis: Quantitative

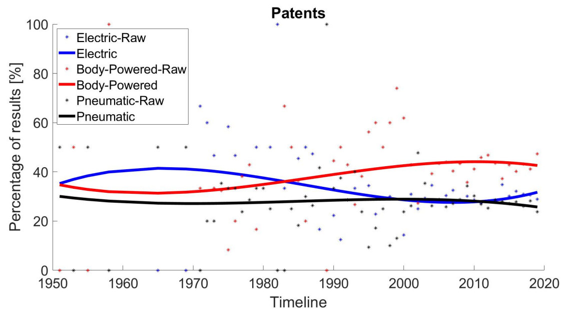
The results are shown in Table III and Figure 7.

TABLE III: An overview of the total amount of results per main type of actuation for powered elbow prostheses for different search engines: Electric, Body-powered & Pneumatic

Search Engine	Type	Electric	Body-Powered	Pneumatics
IEEE Xplore	Articles	15	17	9
Scopus	Articles	35	57	12
Web of Science	Articles	5	34	6
PubMed	Articles	17	36	4
Espacenet	Patents	1671	2299	1527



(a) A trend visualisation based on articles found through IEEE Xplore, Scopus, Web of Science and PubMed



(b) A trend visualization based on patents found through Espacenet

Fig. 7: A visualization of the trends regarding electric, body-powered and pneumatic actuation for elbow prostheses. The fits were generated using a MATLAB polyfit with three polynomial coefficients

B. Trend Analysis: Qualitative

There have been several state-of-the-art reviews about elbow/upper-limb prostheses and their actuation in the past 20 years. The tone and attention of these publications is an indication of the mindset in the field of elbow prostheses.

- 2002 Upper Extremity Prostheses: Current Status & Evaluation [5]
- 2003 Study of the Different Types of Actuators and Mechanisms for Upper Limb Prostheses [19]
- 2008 Evolution of Elbow Prosthesis Transmission [34]
- 2009 Upper limb prostheses for amputations above elbow: A review [55]
- 2015 A review of current upper-limb prostheses for resource constrained settings [56]

The book by Plettenburg [5] describes the multiple facets of upper extremity prostheses and it is one of the only reviews to pay proper attention to a body-powered prosthesis, as opposed to passive or externally powered prostheses. Furthermore, pneumatics is given as a possible method of actuation for hand prostheses. However, in its concluding remarks it is noted that “Two options are available to provide power to a prosthesis: electrical power or body power”. This statement seems insinuate that there are only these two methods to provide power for an active prostheses. In the evaluation section of his book, Plettenburg states that more research into alternative power supplies is required and that the use of electric prostheses might have been too unquestioned. This is further driven home by stating that, in general, pneumatic actuators weight less than their electric counterpart. This is especially important since, as mentioned in the introduction, 88 percent of the prosthesis rejecters mentioned weight as a major factor in abandonment. Even 65 percent of the frequent wearers listed weight as a reasons for possible rejecting a prosthesis [4].

In 2003 [19], where the entire premise is to review different types of actuators, the word “pneumatic” is mentioned once, in a section about hydraulics, whereas the word “electric” is mentioned sixteen times, with a detailed review of different types of electric motors. The article reviews some of the, at the time, newest ideas in actuation, such as memory-shape [57], polymers [58], piezoelectric [59] and ultrasonic [60]. The article devotes a section to the hydraulic McKibben artificial muscle, but references to a Japanese article [61]. The original McKibben muscle was a pneumatic artificial muscle (PAM), developed in the 1950’s. This review will not give an in-depth analysis about the McKibben muscle for two reasons; (1) these reviews have been done on multiple occasions, e.g. [62], (2) PAM’s are a complex subject and having them as a side-project within this review would not do them justice.

The lack of mentions of pneumatic actuation continues with Casolo et al. [34], Toledo et al. [55], and Phillips et al. [56]. The paper by Casolo et al. is an analysis of the evolution of the transmission of an elbow prosthesis. The paper starts off with listing some the negative aspects of active prostheses, including the problem of weight and the heavy batteries. However, the rest of the article is solely focused on electric prostheses. The article by Toledo et al. is a review focused on upper limb prostheses for amputations above the elbow. Not a single mention is made about a pneumatic alternative to the electric versions. There is no proper explanation given on why the entire article focuses solely on electric prostheses. It is noteworthy that nothing is mentioned about pneumatics or gas-actuation because articles specifically about a gas-actuated transhumeral prosthesis have been published, specifically the Vanderbilt arm [27]. The publication by Phillips et al. reviews the existing upper-limb prostheses, with a focus on resource constrained settings. The paper even states “A device that incorporates electronics will be less likely to successfully function in these conditions”. Even with this clear statement and downside of electronics, the word “electric” (or variants such as “electronic” and “myo-electric”) is mentioned 41 times. “Pneumatic” or “air” is not mentioned a single time. Again, a few articles about pneumatic and hydraulic upper limb prostheses have been published in between this review and the previous one; in 2012 [36] and in 2014 [63].

IX. Discussion

A. Actuation: Advantages and Disadvantages

From Table I it can be seen that the main three types of actuation have their advantages and disadvantages, with none being perfect. It is important to avoid taking all statements at face value. The reliability and experience of the author(s) should be checked, especially in cases of disagreement between authors. There are two main cases to be checked, both coloured yellow in the table: pneumatic control and pneumatic safety.

1) Pneumatic Control: Fite et al. of the Vanderbilt University [27] demonstrate that control of a pneumatic prostheses is achievable using a prototype. The article by Vujaklija et al. [12] is written eight years later than the article by Fite et al. Nevertheless, the authors write that ”[considering pneumatic actuation] the control was inefficient and not robust enough”. The source used is the Heidelberg Pneumatic Arm Prosthesis [64], which dates back to 1965. Surprisingly, there are no mentions of the Vanderbilt prototype in the article by Vujaklija et al. It should be noted that the Vanderbilt prototype uses a custom four-way servo valve for control, which indicates a required specialized knowledge of both control and pneumatic actuation. Compared to electric control, the control of a pneumatic prosthesis might be a bit more challenging due to this require expertise and physical components such as valves.

2) Pneumatic Safety: The second case concerns the safety of pneumatic actuation. The designers of the Modular Prosthetic Limb (MPL) have given an overview of their developmental process [14]. In this process, a monopropellant actuation system (as used by the Vanderbilt prototype), was rejected based primarily on safety reasons, since the exothermic reaction would release high-temperature and high-pressure steam. In the article by Fite et al., which predates the MPL development article by about three years, the issues regarding safety are also brought to attention. However, in this case vital information is given: The catalyst pack temperature might be high (237 degrees Celsius for 70 percent peroxide). but the temperature of the gas is cooled as it performs work. This leads to a heavily reduced temperature of about 85 degrees Celsius, which is sufficiently low that it can be used in a design, as long as proper isolation precautions are taken [27].

B. Representation Pneumatic Actuation

The results in Table III already indicate a problem for a trend analysis regarding articles; the total amount of results is rather low, thus a large deviation of percentage can happen on a yearly basis. Since the total amount of patents is an order of magnitude higher, the results might be more reliable, albeit less specific. The results of the analyses are somewhat contradicting: The quantitative trend analysis of patents applied for each year show no decrease in popularity of pneumatic actuation, staying at around $\approx$ 28 percent. The qualitative side shows that electric prostheses are mentioned a significantly more in reviews, with pneumatic being missing from some reviews altogether. This lack of pneumatics is further supported when looking at the two cases as discussed above. Both times, information given by Fite et al. [27] is not taken into account in the articles published years later.

From Table II it can be seen that most elbow prostheses that are being designed at the moment or in the near past, rely on electricity as a power supply, despite the flaws shown in Table I. Only one out of the twelve most prominent elbow prostheses ($\approx$ 8 percent) uses pneumatic actuation and two out of twelve are body-powered ($\approx$ 17 percent). There is not a single commercial prosthetic elbow using pneumatic actuation.

A lack of representation does not necessarily mean that something is underrepresented, it might simply be unsuitable for the cause. However, by looking at the benefits that pneumatic actuation has, especially being lightweight, which is a major factor is prosthesis rejection, it could be said that pneumatic actuation does have a potential and should therefore be considered as an option by designers. This is further illustrated by creation of the Vanderbilt prototype. Furthermore, there are benefits that have not been fully explored. A relevant example of this is its robustness, which might be less important for western countries, but might play a decisive role if the prosthesis is designed for a developing country, where the climate is harsh and replacements are not readily available. This is especially relevant considering 80 percent of amputees live in a developing country. Combining the lack of representation and the potential for success leads to conclusion that pneumatic actuation is indeed underrepresented in the design for upper extremity prostheses.

C. Further Research

The combined results are interesting and raises a question for further research: "Why are there patents being applied for pneumatic prosthetic actuation, when no commercial prosthesis based on pneumatics exist?". A plethora of reasons can be named, ranging from commercial devices copying each other, costs, or that we can expect a number of commercial pneumatic prostheses in the coming years. However, it is too early to speculate and further research is required to answer the question reliably.

As the main focus of this review is the type of actuation, not all aspects were represented in an in-depth manner. For a complete state-of-the-art analysis further literature reviews should be done into subjects including, but not limited to, control and topology.

D. Evaluation

As the amount of articles written about pneumatic actuation for prostheses is low and most of these are older than 20 years, insecurities in the findings exist and assumptions had to be made. It is unclear which prostheses are from different companies, e.g. the Motion E2 Elbow [39] and Hosmer Elbow [65] both being part of the Fillauer company, and also being part of the Utah Arm family. The limited amount of articles on pneumatic actuation for (elbow) prostheses also influenced the trend analysis to such a degree that no certain conclusion can be made based on said data.

X. Conclusion

Conflicting information considering the different types of actuation for powered elbow prostheses causes insecurities, which is why reviews like these are necessary. The main goal stated in Section I is "To give an overview of the current state-of-the-art of powered elbow prostheses, with a focus on actuation" The results are summarized in tables Table I and Table II. They show the advantages and disadvantages of the main types of actuation, and provide an overview of twelve of the most prominent prosthesis.

- The low expected mass for a pneumatic prosthesis might be a solution for one of the biggest reasons of prosthesis rejection: Low comfort
- Raised disadvantages of pneumatic prosthesis, such as control and safety, are overemphasized
- Pneumatic actuation for upper extremity prostheses is underrepresented

Acknowledgements

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Appendix

This query has been visualised in Figure 8. It is understood by the author that this will in no way scope the entire range of publications about the subject, since many synonyms are available, (e.g. limb substitution). Furthermore, it is also expected by the author that a lot of non-relevant papers will be found using this method. However, it is believed by the author that all terms will be equally (dis)advantaged by limiting the search terms in this way. The final results should be discussed in terms of percentages and not in absolute numbers. The analysis does not fully fit within the narrative of this article and the results are therefore discussed in the appendix.

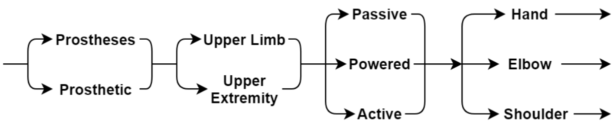
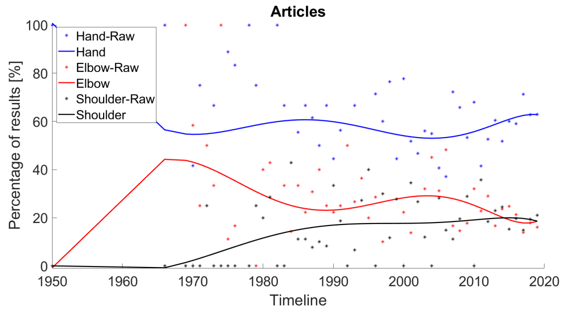


Fig. 8: A visual representation of the search query used for the quantitative research for the anatomical areas. Parallel elements mean a Boolean "OR" and elements in series means "AND".

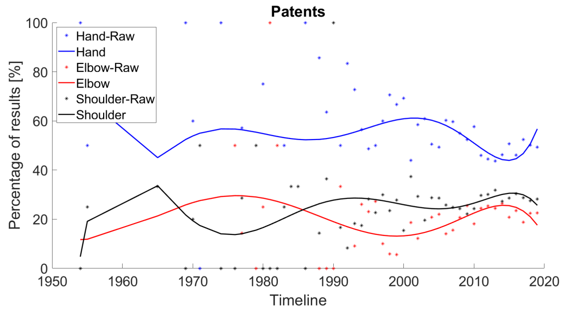
The results of the secondary trens analysis, as proposed in "Methods" are shown in Table IV and Figure 9. It can be seen that the amount of results is significantly higher than the results of the previous analysis (Table III). The results are therefore a bit more reliable. This claim is further supported by the similarities between Figure 9a and Figure 9b, which was not the case for the actuation trend analysis. However, relatively little results predate the 70's, which is why there is such a steep curve. From the 70's on wards, the results are more numerous and the graph(s) even out. As was expected, a large portion of both the articles and the patents focus on hand prostheses. The decrease in attention for elbow prostheses from the 1970's until 2000 does come as a surprise. It was expected that the development of the elbow would have a similar shape in curve as the shoulder, where slowly more and more attention is paid to the higher amputations and disarticulations. The results have been processed in a box plot, see Figure 10. The median of the results are compared to the amputee population as estimated by Maurice LeBlanc [2] Table V. This table shows that the applications for patents follows the amputee population more closely than the published articles. Furthermore, both graphs show a steady increase in attention for shoulder prostheses even though it has already over double the attention (in percentage) as the percentage of shoulder amputees. A surprising result is that the amount of attention paid to hand prostheses is lower that the amount of hand amputees. It was expected by the author that the representation of hand prostheses would be more than their respective amputee population. Tens of potential reasons can be mentioned for these discrepancies, but for a reliable discussion and conclusion, a separate study is needed.

TABLE IV: An overview of the total amount of results per main anatomical goal for prostheses for different search engines: Hand, Elbow & Shoulder

Search Engine	Type	Hand	Elbow	Shoulder
IEEE Xplore	Articles	101	33	23
Scopus	Articles	84	58	45
Web of Science	Articles	235	84	71
PubMed	Articles	393	149	133
Espacenet	Patents	3070	1303	1676



(a) A trend visualisation based on articles found through IEEE Xplore, Scopus, Web of Science and PubMed



(b) A trend visualization based on patents found through Espacenet

Fig. 9: A visualization of the trends regarding developments for hand, elbow or shoulder prostheses. The fits were generated using a MATLAB polyfit with seven coefficients

References

[1] K. Ziegler-Graham, E. J. MacKenzie, P. L. Ephraim, T. G. Travison, and R. Brookmeyer, "Estimating the prevalence of limb loss in the united states: 2005 to 2050," *Archives of physical medicine and rehabilitation*, vol. 89, no. 3, pp. 422–429, 2008.

[2] M. LeBlanc. (). Give hope - give a hand - the ln-4 prosthetic hand, [Online]. Available: https://web.stanford.edu/class/engr110/2011/LeBlanc-03a.pdf. (accessed:05.03.2020).

[3] T. W. Wright, A. D. Hagen, and M. B. Wood, "Prosthetic usage in major upper extremity amputations," *Journal of Hand Surgery*, vol. 20, no. 4, pp. 619–622, 1995.

[4] E. Biddiss and T. Chau, "Upper-limb prosthetics: Critical factors in device abandonment," *American journal*

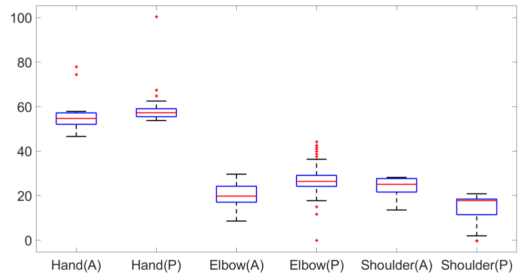


Fig. 10: Boxplots of the trend analysis regarding hand, elbow and shoulder prostheses. (A) means the results are based on articles, (P) corresponds to patents

TABLE V: A comparison between the results in the field of articles, patents and the amputee population

Source	**Hand**	**Elbow**	**Shoulder**
Amputee Population [2]	64%[1]	28%	8%
Articles	54.7%	19.8%	20.1%
Patents	57.2%	26.4%	17.8 %

[1]Including below elbow prostheses and amputees

of physical medicine & rehabilitation, vol. 86, no. 12, pp. 977–987, 2007.

[5] D. H. Plettenburg, *Upper extremity prosthetics: current status & evaluation*. Delft University of Technology, 2002.

[6] S. L. Carey, D. J. Lura, and M. J. Highsmith, "Differences in myoelectric and body-powered upper-limb prostheses: Systematic literature review.," *Journal of Rehabilitation Research & Development*, vol. 52, no. 3, 2015.

[7] F. Malagelada, M. Dalmau-Pastor, J. Vega, and P. Golano, "Elbow anatomy," *Sports injuries: prevention, diagnosis, treatment and rehabilitation*, vol. 2, pp. 527–53, 2014.

[8] F. Casolo, S. Cinquemania, and H. Giberti, "Optimization of myoelectric elbow prosthesis transmission," *International Journal of Applied Engineering Research*, vol. 12, no. 21, pp. 10 909–10 920, 2017.

[9] A. Ramrez-Garcia, C. Toledo, L. Leija, and R. Munoz, "Status of elbow myoelectric prosthesis: Cinvestav-ipn prosthesis," *Revista Mexicana de Ingeniera Biomedica*, vol. 30, no. 1, pp. 66–73, 2009.

[10] D. J. Atkins, D. C. Heard, and W. H. Donovan, "Epidemiologic overview of individuals with upper-limb loss and their reported research priorities," *JPO: Journal of Prosthetics and Orthotics*, vol. 8, no. 1, pp. 2–11, 1996.

[11] M. S. Ghiyasvand, "Graded levels of foot control for below elbow prosthesis," in *First International IEEE EMBS Conference on Neural Engineering, 2003. Conference Proceedings.*, IEEE, 2003, p. 271.

[12] I. Vujaklija, D. Farina, and O. C. Aszmann, "New developments in prosthetic arm systems," *Orthopedic research and reviews*, vol. 8, p. 31, 2016.

[13] P. Tropea, A. Mazzoni, S. Micera, and M. Corbo, "Giuliano vanghetti and the innovation of "cineplastic operations"," *Neurology*, vol. 89, no. 15, pp. 1627–1632, 2017.

[14] M. S. Johannes, J. D. Bigelow, J. M. Burck, S. D. Harshbarger, M. V. Kozlowski, and T. Van Doren, "An overview of the developmental process for the modular prosthetic limb," *Johns Hopkins APL Technical Digest*, vol. 30, no. 3, pp. 207–216, 2011.

[15] R. S. Armiger, F. V. Tenore, K. D. Katyal, M. S. Johannes, A. Makhlin, M. L. Natter, J. E. Colgate, S. J. Bensmaia, and R. J. Vogelstein, "Enabling closed-loop control of the modular prosthetic limb through haptic feedback," *Johns Hopkins APL Tech Dig*, vol. 31, no. 4, pp. 345–53, 2013.

[16] T. N. Y. Times. (). Robotica episode 3: Prosthetic limbs, controlled by thought, [Online]. Available: https://www.nytimes.com/2015/05/21/technology/a-bionic-approach-to-prosthetics-controlled-by-thought.html. (accessed:12.03.2020).

[17] D. Gow, W. Douglas, C. Geggie, E. Monteith, and D. Stewart, "The development of the edinburgh modular arm system," *Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine*, vol. 215, no. 3, pp. 291–298, 2001.

[18] B. Maat, G. Smit, D. Plettenburg, and P. Breedveld, "Passive prosthetic hands and tools: A literature review," *Prosthetics and orthotics international*, vol. 42, no. 1, pp. 66–74, 2018.

[19] V. O. D. Cura, F. L. Cunha, M. L. Aguiar, and A. Cliquet Jr, "Study of the different types of actuators and mechanisms for upper limb prostheses," *Artificial organs*, vol. 27, no. 6, pp. 507–516, 2003.

[20] A. L. Muilenburg and M. A. LeBlanc, "Body-powered upper-limb components," in *Comprehensive management of the upper-limb amputee*, Springer, 1989, pp. 28–38.

[21] D. S. Childress, "Historical aspects of powered limb prostheses," *Clin Prosthet Orthot*, vol. 9, no. 1, pp. 2–13, 1985.

[22] N. Kellaris, V. G. Venkata, G. M. Smith, S. K. Mitchell, and C. Keplinger, "Peano-hasel actuators: Muscle-mimetic, electrohydraulic transducers that linearly contract on activation," *Science Robotics*, vol. 3, no. 14, 2018.

[23] D. A. Bennett, J. E. Mitchell, D. Truex, and M. Goldfarb, "Design of a myoelectric transhumeral prosthesis," *IEEE/ASME Transactions on Mechatronics*, vol. 21, no. 4, pp. 1868–1879, 2016.

[24] E. Biddiss, D. Beaton, and T. Chau, "Consumer design priorities for upper limb prosthetics," *Disability*

and Rehabilitation: Assistive Technology, vol. 2, no. 6, pp. 346–357, 2007.

[25] P. Ballif, *Description d'une main et d'une jambe artificielles*. nicht ermittelbar, 1818.

[26] O. Bock. (). The dynamic arm+, [Online]. Available: https://www.ottobock.com/en/. (accessed:24.02.2020).

[27] K. B. Fite, T. J. Withrow, X. Shen, K. W. Wait, J. E. Mitchell, and M. Goldfarb, "A gas-actuated anthropomorphic prosthesis for transhumeral amputees," *IEEE Transactions on Robotics*, vol. 24, no. 1, pp. 159–169, 2008.

[28] G. o. B. C. Healthwise Staff. (). Harmful noise levels, [Online]. Available: https://www.healthlinkbc.ca/health-topics/tf4173. (accessed:02.04.2020).

[29] S. Millstein, H. Heger, and G. Hunter, "Prosthetic use in adult upper limb amputees: A comparison of the body powered and electrically powered prostheses," *Prosthetics and orthotics international*, vol. 10, no. 1, pp. 27–34, 1986.

[30] N. A. Hashim, N. A. Abd Razak, N. A. Abu Osman, and H. Gholizadeh, "Improvement on upper limb body-powered prostheses (1921–2016): A systematic review," *Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine*, vol. 232, no. 1, pp. 3–11, 2018.

[31] D. Plattenburg, "Pneumatic actuators: A comparison of energy-to-mass ratio's," in *9th International Conference on Rehabilitation Robotics, 2005. ICORR 2005.*, IEEE, 2005, pp. 545–549.

[32] M. Sekine, K. Kita, and W. Yu, "Designing and testing lightweight shoulder prostheses with hybrid actuators for movements involved in typical activities of daily living and impact absorption," *Medical devices (Auckland, NZ)*, vol. 8, p. 279, 2015.

[33] M. Irshaidat, M. Soufian, A. Al-Ibadi, and S. Nefti-Meziani, "A novel elbow pneumatic muscle actuator for exoskeleton arm in post-stroke rehabilitation," in *2019 2nd IEEE International Conference on Soft Robotics (RoboSoft)*, IEEE, 2019, pp. 630–635.

[34] F. Casolo, S. Cinquemani, and M. Cocetta, "Evolution of elbow prosthesis transmission," in *2008 5th International Symposium on Mechatronics and Its Applications*, IEEE, 2008, pp. 1–6.

[35] D. Plettenburg, "Electric versus pneumatic power in hand prostheses for children," *Journal of medical engineering & technology*, vol. 13, no. 1-2, pp. 124–128, 1989.

[36] M. Foglia and M. Valori, "A high performance wire device for an elbow prosthesis," in *2012 4th IEEE RAS & EMBS International Conference on Biomedical Robotics and Biomechatronics (BioRob)*, IEEE, 2012, pp. 494–499.

[37] A. Escudero, J. Alvarez, and L. Leija, "Development of a parallel myoelectric prosthesis for above elbow replacement," in *Proceedings of the Second Joint 24th Annual Conference and the Annual Fall Meeting of the Biomedical Engineering Society][Engineering in Medicine and Biology*, IEEE, vol. 3, 2002, pp. 2404–2405.

[38] A. Ramirez-Garcia, L. Leija, and R. Munoz, "Active upper limb prosthesis based on natural movement trajectories," *Prosthetics and orthotics international*, vol. 34, no. 1, pp. 58–72, 2010.

[39] Fillauer. (). Nexo, [Online]. Available: http://fillauer.com/Upper-Extremity-Prosthetics/nexo.php. (accessed:24.02.2020).

[40] L. Resnik, S. L. Klinger, and K. Etter, "The deka arm: Its features, functionality, and evolution during the veterans affairs study to optimize the deka arm," *Prosthetics and orthotics international*, vol. 38, no. 6, pp. 492–504, 2014.

[41] R. S. Inc. (). The espire series, [Online]. Available: https://www.steepergroup.com/prosthetics/upper-limb/wrists/espire-classic,-classic-plus-basic/. (accessed:24.02.2020).

[42] J. H. A. P. Laboratory. (). Revolutionizing prosthetics program overview, [Online]. Available: https://www.jhuapl.edu/Prosthetics/Program. (accessed:27.02.2020).

[43] T. Lenzi, J. Lipsey, and J. W. Sensinger, "The ric arm—a small anthropomorphic transhumeral prosthesis," *IEEE/ASME Transactions on Mechatronics*, vol. 21, no. 6, pp. 2660–2671, 2016.

[44] S. C. Jacobson, D. F. Knutti, R. T. Johnson, and H. H. Sears, "Development of the utah artificial arm," *IEEE Transactions on Biomedical Engineering*, no. 4, pp. 249–269, 1982.

[45] J. M. Zuniga, A. M. Carson, J. M. Peck, T. Kalina, R. M. Srivastava, and K. Peck, "The development of a low-cost three-dimensional printed shoulder, arm, and hand prostheses for children," *Prosthetics and orthotics international*, vol. 41, no. 2, pp. 205–209, 2017.

[46] S. W. Alderson, "The electric arm," *Human Limbs and Their Substitutes*, pp. 359–408, 1954.

[47] E. Lewis, C. Sheredos, T. Sowell, and V. Houston, "Clinical application study of externally powered upper-limb prosthetics systems: The va elbow, the va hand, and the va/nu myoelectric hand systems.," *Bulletin of prosthetics research*, no. 10-24, pp. 51–136, 1975.

[48] D. W. Lamb, T. D. Dick, and W. B. Douglas, "A new prosthesis for the upper limb," *The Journal of bone and joint surgery. British volume*, vol. 70, no. 1, pp. 140–144, 1988.

[49] W. Prout, "The new york electric elbow, the new york prehension actuator, and the nu-va synergetic prehensor," in *Comprehensive Management of the Upper-Limb Amputee*, Springer, 1989, pp. 221–226.

[50] W. Sauter, "The use of electric elbows in the rehabilitation of children with upper limb deficiencies," *Prosthetics and orthotics international*, vol. 15, no. 2, pp. 93–95, 1991.

[51] M. E. Cupo and S. J. Sheredos, "Clinical evaluation of the modular electromechanical lock actuator (mela)

for above-elbow prostheses: A final report," *Journal of rehabilitation research and development*, vol. 33, no. 1, p. 56, 1996.

[52] D. Bandara, R. Gopura, K. Hemapala, and K. Kiguchi, "Development of a multi-dof transhumeral robotic arm prosthesis," *Medical engineering & physics*, vol. 48, pp. 131–141, 2017.

[53] G. Li, Y. Liu, and Z. Li, "The prosthetic arm: A dramatic improvement for the limb amputation from the humerus," in *2019 IEEE 4th International Conference on Advanced Robotics and Mechatronics (ICARM)*, IEEE, 2019, pp. 475–480.

[54] S. K. Kundu, K. Kiguchi, and E. Horikawa, "Design and control strategy for a 5 dof above-elbow prosthetic arm," *Int. J. Assist. Robot. Mechatron*, vol. 9, pp. 61–75, 2008.

[55] C. Toledo, L. Leija, R. Munoz, A. Vera, and A. Ramirez, "Upper limb prostheses for amputations above elbow: A review," in *2009 Pan American Health Care Exchanges*, IEEE, 2009, pp. 104–108.

[56] B. Phillips, G. Zingalis, S. Ritter, and K. Mehta, "A review of current upper-limb prostheses for resource constrained settings," in *2015 IEEE global humanitarian technology conference (GHTC)*, IEEE, 2015, pp. 52–58.

[57] K. DeLaurentis, C. Mavroidis, and C. Pfeiffer, "Development of a shape memory alloy actuated robotic hand," in *Proc. 7th International Conference on New Actuators*, 2000, pp. 281–284.

[58] D. Brock, "Review of artificial muscle based on contractile polymers," MASSACHUSETTS INST OF TECH CAMBRIDGE ARTIFICIAL INTELLIGENCE LAB, Tech. Rep., 1991.

[59] L. Rosenberg, *Piezoelectric motivator for prosthetic devices*, US Patent 4,895,574, 1990.

[60] J. L. Pons, H. Rodriguez, I. Luyckx, D. Reynaerts, R. Ceres, and H. V. Brussel, "High torque ultrasonic motors for hand prosthetics: Current status and trends," *Technology and Health Care*, vol. 10, no. 2, pp. 121–133, 2002.

[61] Y. Lee, "A micro rubber artificial muscle driven by a micro compressor for artificial limbs," in *International Conference on New Actuators*, vol. 3, 2000, pp. 272–275.

[62] B. Tondu, "Modelling of the mckibben artificial muscle: A review," *Journal of Intelligent Material Systems and Structures*, vol. 23, no. 3, pp. 225–253, 2012.

[63] V. Bevilacqua, M. Dotoli, M. M. Foglia, F. Acciani, G. Tattoli, and M. Valori, "Artificial neural networks for feedback control of a human elbow hydraulic prosthesis," *Neurocomputing*, vol. 137, pp. 3–11, 2014.

[64] E. Marquardt, "The heidelberg pneumatic arm prosthesis," *The Journal of bone and joint surgery. British volume*, vol. 47, no. 3, pp. 425–434, 1965.

[65] Cascade. (). Hosmer, [Online]. Available: https://www.cascade-usa.com/hosmer-e-200-elbow-9640.html. (accessed:10.03.2020).

[66] N. M. Bajaj, A. J. Spiers, and A. M. Dollar, "State of the art in prosthetic wrists: Commercial and research devices," in *2015 IEEE International Conference on Rehabilitation Robotics (ICORR)*, IEEE, 2015, pp. 331–338.

[67] J. Mendoza-Vázquez, E. Tlelo-Cuautle, J. Vázquez-Gonzalez, and A. Escudero-Uribe, "Simulation of a parallel mechanical elbow with 3 dof," *Journal of applied research and technology*, vol. 7, no. 2, pp. 113–123, 2009.

[68] M. Sekine, N. Tsuchiya, K. Kita, and W. Yu, "Designing and testing a hybrid lightweight shoulder prosthesis," in *2014 36th Annual International Conference of the IEEE Engineering in Medicine and Biology Society*, IEEE, 2014, pp. 2500–2503.

[69] B. N. Perry, C. W. Moran, R. S. Armiger, P. F. Pasquina, J. W. Vandersea, and J. W. Tsao, "Initial clinical evaluation of the modular prosthetic limb," *Frontiers in neurology*, vol. 9, p. 153, 2018.

D
Competitor Analysis

Table D.1: An overview of the most prominent "competitors": existing prostheses. Some prostheses cannot achieve pronation/supination. In these cases, a grey "x" is used to show that there is no data to be represented. If information should be presented, such as FE RoM, but is missing from the source a yellow "?" is used to illustrate missing information

Elbow Prosthesis	Source	Weight [grams]	RoM FE [°]	RoM PS [°]	FE Payload	FE Holding Payload	PS Torque [Nm]	FE Angular Velocity [°/s]	PS Angular Velocity [°/s]	Costs [€]	Time until Recharge [hour]
CINESTAV-IPN	[28]	1050	10-115	±90	1 [kg]	x	x	45	180	?	?
Motion E2 Elbow	[30]	439	?	x	?	x	x	?	x	?	?
The Boston Elbow	[28]	1100	0-145	±90	2,3 [kg]	22.65 [kg]	x	145	x	?	8
The DEKA Arm	[31]	1200	?	x	?	x	x	?	x	?	?
The Edinburgh Arm	[32]	1766	?	x	4 [kg]	x	x	?	x	?	?
Espire Pro	[33]	1085	-5 - 135	x	4 [kg]	11	x	?	x	?	?
The Modular Prosthetic Limb	[34]	4800 [1]	?	?	16 [kg]	x	6,8 [kg]	120	120	?	?
	[35]	x	?	x	60 [Nm]	x	x	?	x	?	?
The Ottobock Dynamic Arm+	[36]	700	?	x	18 [Nm]	x	x	?	x	?	?
	[37]	1000	15 –145	x	5	x	x	?	x	?	?
The RIC Arm	[38]	982	?	?	12 [Nm]	x	2,2	80	500	?	05:23
The UTAH Arm	[28]	913	0-135	±90	1	15,9-22,7 [kg]	x	112	x	?	?
The Vanderbilt Arm	[8]	1550	0 - 160	150	30 [Nm]	x	4,2	80	x	?	?
The Creighton Arm	[39]	382	0-90	0-90	?	x	x	?	x	200 USD	?

[1] This is the weight for a full-arm prosthesis

E

Function Explanation

Must Have

These functions are essential to the prosthesis, when missing even one of these in the final design, it cannot be considered a viable product. The following functions are fundamental in the design of an elbow prosthesis.

Flexion/Extension The main elbow movement is flexion/extension. It is used for a wide range of ADL.

Pronation/Supination Pronation/supination is an important degree of freedom for a great variety of ADL since it helps with the positioning of the hand and must be included in a prosthesis to have full functionality.

Constrain Unwanted DoF An important aspect of the structure is to constrain the unwanted degrees of freedom.

Withstand Forces and Moments The structure of the prosthesis should be build in such a way that it can handle the forces and moments generated by its own actuators, but also external forces and moments. Without this function, the structure and therefore the prosthesis will fail.

Be Safe Safety is a number one priority. Therefore it is vital that the amputee is shielded from all kinds of danger including, but not limited to, electrical and thermal.

Sense User Input A prosthesis is useless if the operator cannot operate it. The prosthesis is therefore required to be able to sense the user input. Multiple methods of sensing user input have been devised, ever since the first non-static prosthesis. These methods range from cables to EMG and cortical control.

Generate Requested Output The input given by the user must somehow be translated into the requested output for the prosthesis to work.

Should Have

These functions are highly desired, but missing them does not endanger the core goal of the prosthesis. Priority in design will go to the ”must haves”.

Have a Free Swing Multiple other elbow prostheses have a mode typically referred to as ”free-swing”, which is used during walking. This mode enables the elbow to swing alongside the body, just as a normal elbow would do. The result is a more natural feel and look. This increases the comfort of the prosthesis

Be Lockable An elbow is not constantly moving and there are a multitude of ADL in which the elbow should be kept in a constant position. This could be achieved by constantly powering the joint torque to hold the elbow in this position, increasing power consumption. It can also be achieved by having a locking mechanism independent of the power supply. Locking can be either continuous or discrete, with a set amount of degrees between the different lockable position.

Protect Components A prosthesis is bound to have vulnerable components. To avoid constant failure, it is important to shelter these components from external factors such as forces, moments and micro-particles such as sand.

House Components In order for the prosthesis to be considered viable, it has to be portable. An optimal design would house all required components within the confines of the prosthesis. However, there are examples of prostheses where i.e. a battery pack is located on the belt. These designs are still viable elbow prosthesis, thus housing all components within the prosthesis is not essential.

Have a Portable Energy Supply A prosthesis would not live up to its full use if it were not portable. Therefore, a portable energy supply is needed. For body-powered prostheses, this comes in the form of human muscle, for electric prostheses batteries are used, and pneumatic prostheses use storage tanks or gas cartridges.

Fit to User Arm The user should be able to wear the prosthesis, thus it should fit to the user arm for the prosthesis to be regarded viable. However, the prosthesis designed in this thesis will not yet be used by a patient, it is therefore not yet a must.

Provide Feedback The world outside a laboratory is full of uncertainties. Prostheses, when used in the real world, are subject to external disturbances. The human body deals with these disturbances by having a great number of feedback loops. A well functioning anthropomorphic prosthesis should apply feedback to the operator for optimal control [5]. Without feedback, the prosthesis will still work, but the user is required to close the control loop visually, which is mentally more straining and slower.

Detect Angles To properly control the device, the orientation of the varying degrees of freedom is desired to be known.

Could Have

These are the functions that will be implemented if they do not endanger the schedule of the higher priority functions.

Connect to Terminal Device An elbow without a terminal device, e.g. a hand, might seem useless. There are companies selling a terminal device, which could be connected to a prosthesis. Since the functionality of an elbow does not depend on the presence of a hand, the prosthesis can be considered a viable product, without the existence of a terminal device.

Fit for Elbow Disarticulations As discussed in Section 1.1, there are different types of patients requiring elbow prostheses. However, the additional design limitations to design for patients with an elbow disarticulation restrict a major part of the design options. Furthermore, non of the existing prostheses have this function, thereby missing it does not make the designed prosthesis ”less functional”.

Be Customizeable Some amputees want the prosthesis to closely resemble the missing limb and thus use cosmetic gloves. Other amputees want to show-off the prosthesis and want a brightly colored one or one resembling the arm of superhero. There is no ”one-size fits all” in the field of cosmetics. Having a customizeable prosthesis enables the user to adjust the appearance of the prosthesis to their own wishes.

Fit under Clothes By being able to cover the prosthesis underneath e.g. a sweater, might help with user acceptance.

Wont Have

These are functions that might be implemented in a later version, but no further attention will be paid to them during this design cycle.

Have an Adjustable Fitting Changes in lifestyle of the user, such as losing or gaining weight, or doing different activities with the prosthesis, have an effect on the fitting. If the prosthesis were to have an adjustable fitting, this would help greatly with user comfort. However, the human-machine interaction is quite an extensive subject, too large to be covered as a side-project within this design process. Furthermore, a different graduation project that is being worked on at the moment concerns 3D-printed fittings. Depending on the results of this project, this might be implemented in this design. For these reasons, no further attention will be paid to the option of an adjustable fitting.

Show Power Level This is feature that would help the user with determining when to recharge the device and therefore help with use in daily life. However, since this will be the first version of the device and it will not be used in daily life, this feature is unneeded at the moment.

F

Reasonsing Requirements

Length Looking at other elbow prostheses, an average space 1/3 of the upper-arm and 2/3 of the forearm are used for the generation of flexion/extension and pronation/supination. The reason why the entire forearm is not used is the need for a connection to, control, and actuation of a terminal device. Furthermore, the available space at the side of the upper arm highly depends on the patient. For patients with an elbow dis-articulation, the entire upper-arm is present and therefore no components can be placed there. Please note that this is a rough estimate, since the majority of prostheses do not clearly list the lengths required. However, by combining this estimate with data provided by Winter [40], a guideline for the length limits can be generated. According to Winter, the average forearm length for a man is 14.5 percent of his height. For the upper arm, this is 18.9 percent. For our user, this would correspond to $L_{Forearm} = 2/3 * 0.145 * H_{User} = 174mm$ and $L_{UpperArm} = 1/3 * 0.189 * H_{User} = 113mm$.

Weight With an estimation of the length and with the Winter data, the weight of the replaced body parts can be estimated. A forearm and an upper arm account for 1.6 percent and 2.8 percent of the total body mass respectively. The weight of the replaced body parts therefore corresponds to $M_{Anatomical} = (2/3 * 0.016 + 1/3 * 0.028) * M_{User} = 1600grams$. However, this does not mean that the prosthesis can be as heavy as 1.6 kilograms. Because of a variety of reasons, including but not limited to the connection to the human body, a prosthesis can feel heavier than it actually is. To counter this, the prosthesis should weigh at most half the replaced body part in order to feel natural, this would be 800 grams. For patients with an elbow disarticulation, the allowed weight is even lower at $M_{Anatomical} = 0.5 * (2/3 * 0.016) * M_{User} = 420grams$.

Comparing different prostheses on basis of weight is quite difficult. Some prostheses can replace an entire arm, whilst others focus on the elbow joint. The weight given, if any, is not traceable to a specific degree of freedom in some cases. Nevertheless, it seems that the elbow modules of most prostheses weighs around 1000 grams, including prostheses where pronation/supination is not included in the elbow but in a separate wrist module. There are two outliers: The Motion E2 Elbow [30] and arm developed by the university of Creighton [39], weighing 439 grams and 382 grams respectively. In order to achieve these low weights, they have sacrificed functionality. The Motion E2 Elbow can only perform FE and is meant for ”light-duty” applications. The Creighton university arm has a limited range of motion for both FE (0°to 90°) and PS (90°) and the corresponding torques are not stated.

Ranges of Motion An article by Oosterwijk et al. [17] investigates the range of motion for both the elbow and shoulder. They conclude that a range of 0°to 150°for elbow flexion is required for ADL. This is more than what the generally accepted study by Morrey et al.[41] states: 30°to 130°. For pronation and supination, Magermans et al.[18] propose a required range of motion of ±90°, whereas Murrey et al. [19] consider ±50°to suffice for most ADL.

For flexion/extension, the existing prostheses have an average range of motion of ±5°to 135°. The prosthesis designed in this thesis should have a similar RoM in order to have an ”equivalent or higher functionality”, which is part of the goal of this thesis. The range of motion for pronation/supination can greatly differ, some prostheses have a RoM of ±90°whilst others do not enable this motion at all and require a terminal device to achieve this motion. Any PS motion at all would be better than some of the other prostheses, thus the lower limit is based on the ADL.

Payloads & Torques Similar to the RoM, the torques required of an elbow prosthesis can be determined by looking at the ADL and other prostheses. For FE, the term "torque" is not suitable as a requirement for a pneumatic prosthesis. The reason for this is the dependency on the angle of the elbow. When lifting a certain load, the required torque can be estimated using Equation (F.1). This equation leads to a graph as shown in Figure F.1. An electric motor can deliver a constant torque, but this is not the optimal solution to lifting a load with a changing angle. Most of the time the electric motor is much stronger than required. Depending on the design requirements, the electric motor might be too weak to lift the load at the peak required torque (elbow at 90°). This illustrates a further possible benefit of pneumatic actuation, since this type of actuator can provide a constant force and will therefore have a graph similar in shape as Equation (F.1). An optimal design configuration follows the required curve as closely as possible, more on this in Chapter 4.

$$T = m * g * L * sin(\theta) \tag{F.1}$$

F.1: Where T is the required torque [Nm], m is the lifted mass in hand [kg], g is the gravity constant [kg/s^2], L is the distance between the elbow joint and the hand [m] and θ is the flexion angle of the elbow [°]. Friction and mass of the prosthesis are neglected

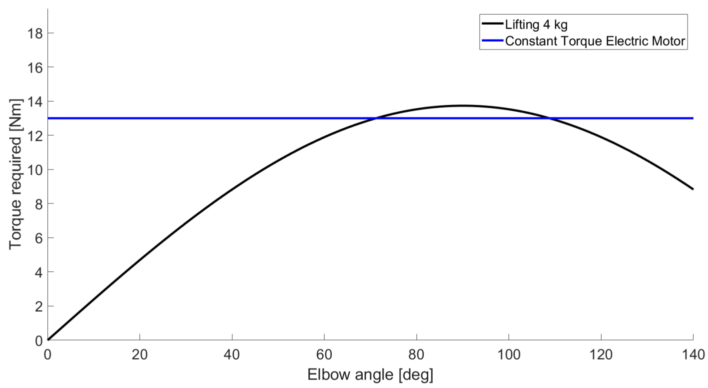


Figure F.1: The torque required to lift a certain load (m = 4 kg) and a possible torque provided by an electric motor (T = 13 Nm)

The requirement will therefore be a certain payload in hand. Magermans et al. [18] mention a payload of 4 kg is required to perform all ADL, whereas 1.5 kg suffices according to Murray et al. [19]. Most existing prostheses show a similar range in payload capacity, with the exception of the Modular Prosthetic Limb, which can supposedly lift 35 lbf, or almost 16 kg. However, the entire arm prosthesis weighs about 4.8 kg.

Aside from a lifting payload, some prostheses have an additional "holding" payload. This is the payload which the prosthesis can hold in a certain position without moving, by e.g. locking elbow joint. Not all prostheses have, or mention, this functionality. There are two prosthesis that show a remarkable "holding" payload: The Boston Elbow and The Utah Arm, both of which can support almost 23 kg in a certain position. As this "holding" mode is considered a "should-have" (see Figure 3.1), there is no minimum value. However, if implemented, a holding force of twice the payload will be used as a minimum, to prevent failure of the device.

The requirement for pronation/supination should be in terms of a torque instead of a payload, since it is mostly independent of the angle. A study by O'Sullivan et al. [42] shows that humans can achieve up to 15 [Nm] PS. This is severely above the capabilities is most prostheses. Commercial prostheses, such as the AxonRotation by Ottobock, achieve 1.5 [Nm]. The strongest PS torque found in prostheses was 4.2 [Nm] [8].

Velocities & Accelerations The final biological metrics are the required angular velocities and accelerations, these are summarized in Table F.1. The human body can achieve significantly higher velocities than prostheses, with the fastest prosthesis, The Boston Elbow, achieving 145 [°/s]. For most prostheses, the angular accelerations are not discussed and can therefore not be compared to the human body. Without accelerations, discussing the velocities is quite difficult. A better requirement would be a combination of these two: the time for a full RoM stroke. By using the data from Table F.1, an estimate of full stroke time for a human can be determined, this will serve as the goal of the prosthesis. The existing prostheses will serve as an upper limit to determine to full stroke time.

Table F.1: The angular velocities and accelerations of the elbow

DoF	Parameter	Buckley et al. data [20]
Flexion/ Extension	Angular Velocity	250 deg/s
	Angular Acceleration	2000 deg/s^2
Pronation/ Supination	Angular Velocity	400 deg/s
	Angular Acceleration	8000 deg/s^2

Table F.2: The time for full stroke of the elbow

DoF	Human	Average Other Prostheses
Flexion/Extension	0.7 s	1.5 s
Pronation/Supination	0.5 s	1.0 s

Sound Level The sound level of a prosthesis plays a role in the comfort and acceptance of said prosthesis. A sound level of 20 [dB] is about the sound level rustling leaves produce [16]. Conversation at home produce around 50 [dB], any louder than this will disrupt daily life and should therefore be avoided.

Cycles of Use per Charge The fuel source for the prosthesis should last for one day. However, it is difficult to find exactly how many cycles an elbow performs a day. A study by J.W. Limehouse et al. [43] seems to provide an answer to this, but it nowhere to be found online. If an elbow is used an average of once every three minutes, and a regular waking day is 15 [hours], the elbow is used around 300 times per day. If we compare it to an estimated 1200 grasping movements per day [44], this usage seems reasonable. Please note that this is heavy speculation. If a waking day is divided into three sections: Morning, afternoon, and evening, each section can have its own cartridge. The minimal use per cartridge is 100 cycles.

Cost The upper limit for the cost is the budget of the thesis. An important note is that, since custom parts will be produced at 3mE for the prototype, these will not affect the cost.

G

Method of Actuation

In the literature review, three main types of actuation were selected and evaluated: Electric, Body-powered and pneumatic. The advantages and disadvantages were discussed in detail in the review, a short summary is given below. For the sources and further explanations, see Appendix C.

Electric Electric prosthesis stand out due to the following advantages; (1) a large range of motion, (2) synergy with EMG-control, (3) appearance. The downsides of an electric prosthesis include: (1) weight, (2) costs, (3) durability, (4) difficult maintenance, (5) long training time, and (6) lack of proprioceptive feedback.

Body-powered The main advantages of having a body-powered prosthesis are: (1) proprioceptive feedback, (2) robustness, (3) low cost, (4) silent actuation, (5) independent of an external power supply, (6) light weight and (7) easy maintenance. Body-powered prostheses do have some drawbacks: (1) low comfort, (2) limited range of motion, (3) limited control options, and (4) appearance.

Pneumatic The advantages include: (1) light weight, (2) compliance, (3) durability, (4) costs, (5) simple maintenance, and (6) anthropomorphic actuators. Further advantages can be inferred by looking at similarities between electric and pneumatic actuation (lack of harness): (7) comfort, and (8) appearance. The disadvantages include (1) limited range of motion, (2) lack of any feedback, (3) uncertainties in design, (4) packaging of required gas, and (5) concerns about safety due to high-pressure gasses.

Actuation Method Choice

To make a decision regarding actuation choice, the results of the user analysis will be used. The selection between these vastly different methods is highly dependent on the goal of the prosthesis. For developing countries where access to electricity might be limited, a body-powered prosthesis might be preferred. The decision will be based on the main factors in prostheses design and use: Control, Comfort, and Cosmesis.

Electric actuation will not be chosen for the method of actuation because of discomfort, mainly due to the high weight. Body-powered actuation will not be chosen as the method of actuation for this prostheses: (1) the harness is not comfortable, (2) despite the major benefit of proprioceptive, the options for control are limited, and (3) unattractive appearance because of the harness. This leaves us with pneumatic actuation as the actuation method. Not a lot is known about the control of pneumatic prosthesis, but a prototype by Fite et al. [8] shows control is possible, albeit with a custom valve set-up. Choosing pneumatic actuation has a final advantage: designs for electric and body-powered prosthesis already exist, limiting the options for an innovative design.

H
Actuation

Conceptual Design

Cylinder Type

There is a plethora of available pneumatic actuators. For the purposes of compactness and weight limitations, this thesis will focus on simple, linear pneumatic cylinders. These can be categorized into single-acting or double-acting cylinders. Single-acting cylinders use one air port to allow compressed air to enter the cylinder to move the piston to the desired position, by either pushing or pulling the piston. Often a spring to return the piston to a “home” position when the air pressure is removed. A ”push” type pushes the piston outwards when the cylinder is pressurized. An absence of pressure means the piston is in its retracted position. The ”pull” type is its counterpart, where the piston is extended in the absence of pressure and will retract when the cylinder is pressurized. Double-acting cylinders have an air port at each end and move the piston forward and back by alternating the port that receives the high pressure air. The ifferent pistons are visualized in Figure H.1.

(a) A ”push type” single acting pneumatic cylinder. Image taken from [21]

(b) A ”pull type” single acting pneumatic cylinder. Image taken from [21]

(c) A double acting pneumatic cylinder. Image taken from [21]

Figure H.1: Examples of the types of pneumatic actuation

Amount of Cylinders The force exerted by a pneumatic cylinder can be calculated using Equation (H.1). This shows that the force scales with he square of the cylinder radius. This is a useful realisation for design, since it can help with determining whether to use a single or multiple cylinders to achieve the desired torque. Two cylinders with radius ”r” could be replaced with a single cylinder with radius ”$r\sqrt{2}$” and still generate the same force. Whilst multiple cylinders might provide more design opportunities, a single large actuator would be more lightweight. Furthermore, having multiple cylinders would mean having a multitude of cables, tubes, valves, and more secondary actuator components, which would further increase the cost, weight, complexity, and size of the prosthesis. A single actuator for each degree of freedom will be used.

$$F = p * \pi * \frac{d^2}{4} \tag{H.1}$$

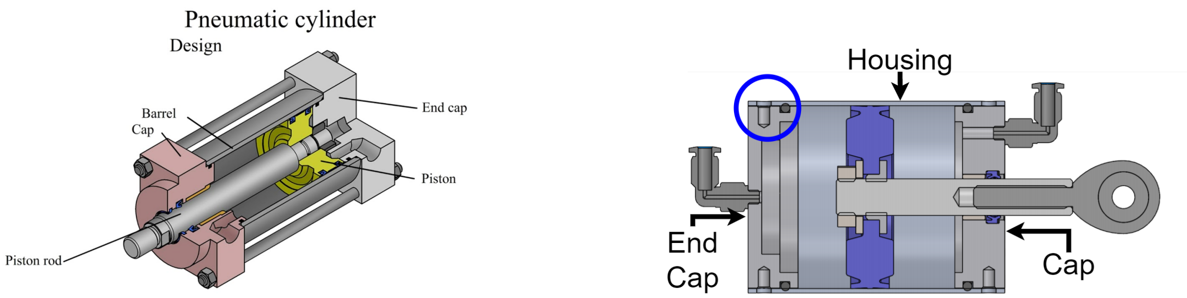
H.1: The force exerted by the pneumatic cylinder on the piston, without losses. Where F is the cylinder force [N], p is the cylinder pressure [mPa], and d is the cylinder diameter [mm]

Springs A spring in the system might increase the time-until-charge of the prosthesis by reducing the required force of the actuator. Springs can be implemented within the actuator or within the frame. Implementing a single spring offsets the system. Such an offset might prove beneficial for a single movement, such as helping with the flexion. However. the prosthesis will be use for a variety of ADL, an offset might help with one motion but limit another. Furthermore, with an offset the free-swing mode of the system is limited. Multiple springs can be used to design a prosthesis stable in all positions. In such a design, the actuators only have to provide enough force to lift the payload, the weight of the prosthesis will not influence the required force. However, such a design adds additional complexity and weight to the prosthesis. For these reasons, the design will continue without using a springs.

Type of Prosthesis Choice Based on all the information stated above, for both flexion/extension and pronation/supination one actuator without springs will be used: a double-acting cylinder.

Pneumatic Actuator Design

A generic pneumatic actuator consists out of five main components: a barrel, two end caps, a piston and a piston rod, as shown in Figure H.2a. The barrel and the end caps provide the sealed structure and support for the piston to move within. The piston rod transmits the movement of the piston to the end-effector of choice. In most designs, the barrel is sealed by using tie rods to pull the end caps to the barrel . However, this external structure causes the actuator to become larger and heavier than needed. A different solution is internal caps and radial bolts, as shown in Figure H.2b.



(a) The main components for a pneumatic cylinder. Image taken from [45]

(b) A method of attaching the caps to the housing in limited space environments

Figure H.2: The three selects concepts to translate the rectilinear motion of the pneumatic actuator

Prototyping 3D-printed prototypes were used to determined the final design of the pneumatic cylinder(s), see Figure H.3.

Figure H.3: Some of the 3D-printed prototypes that were used to determined the final design of the pneumatic cylinder(s)

Final Concept The housing of the actuator fulfills several functions: Guide the piston, create a sealed chamber, and withstand the external forces and moments. The cap and end-cap: create a sealed chamber with chamber with a connection to the push-fitting, and provide a connection to the frame. The o-rings are used to prevent leakage from the chambers. Furthermore, the cap houses the bushing and rod-seal. The bushing guides the piston rod and the rod-seal prevents leaks.

The piston rod transfers the movement of the piston to the end-effector. It is connected to the piston using a nut and washer on one side and the piston-rod-nut. The piston rod nut and nut & washer clamp the piston to the piston rod. The leaking of air between the two chambers should be avoided as much as possible. To prevent leakage, the piston rod nut is designed to fit within the piston. For this reason, the piston rod nut is a custom part an not a second ”nut & washer”.

- Custom: Housing
- Custom: Piston-Rod
- Custom: Piston-Rod-Nut
- Custom: Cap
- Custom: End-Cap
- Piston
- O-ring (2x)
- Rod-Seal
- Bushing
- Fitting (2x)
- Nut & Washer
- Housing Bolt (8x)

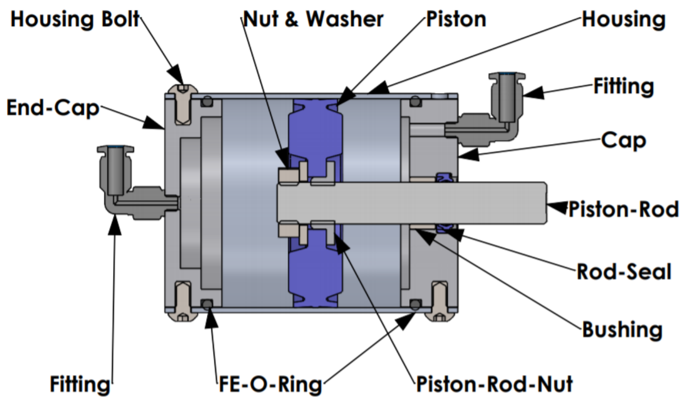


Figure H.4: A schematic cross-section of the general actuator, with the parts labeled

Topology

Since the choice of actuation is pneumatic, linear actuators will be used to generate movement.

Series or Parallel Actuation

An important decision that should be made regarding flexion/extension and pronation/supination is whether to place these motions in parallel or series. This design choice is for the combination of these two different degrees of freedom, i.e. if the series configuration is chosen, a parallel configuration can still be used within a single degree of freedom.

Series Most elbow prostheses have placed the two degrees of freedom in series. This enables a more modular and simple design. The disadvantage can be ”dead-weight” of the actuators for the second degree of freedom, i.e the actuators used to generate PS come after the generation of FE and are therefore needlessly carried around, increasing the inertia and power consumption of the system.

Parallel Only a select few prostheses use parallel actuation to generate both flexion/extension and pronation/supination simultaneously. The most prominent group is the Department of Electric Engineering, Bioelectronics of the Center of Research and Advanced Studies of the IPN in Mexico, [46], Figure H.5. Parallel mechanisms can produce a greater torque since the actuators can help each other. This eliminates the ”dead-weight”. The trade-off is a more complex kinematic model as the motion of every actuator relates to all other actuators.

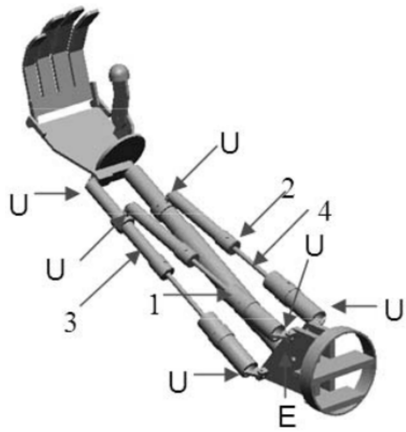
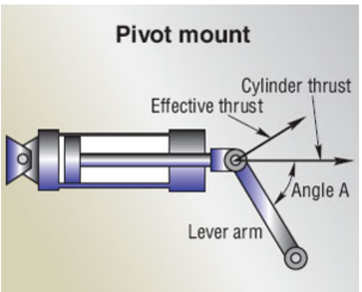


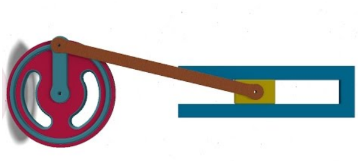
Figure H.5: An example of an elbow prosthesis where flexion/extension and pronation/supination are done in parallel [46]. 1, 2, and 3 are linear actuators and 4 is the supporting mechanical structure Image taken from [46]

FE and PS will be placed in series for the following reasons, mainly based on the design values of simplicity and being easy to manufacture and assemble. (1) A series topography enables modularity, which in turn enables the prioritization of motion. A focus can be placed on generating an adequate flexion/extension before moving on to pronation/supination, (2) the ”dead-weight” factor plays less of a role due to the lightness of pneumatic actuators, and (3) the complexity of the kinematic model might cause insecurities in the control scheme.

From Rectilinear Motion to Rotation



(a) An example of a lever. Image taken from [47]



(b) An example of a crank-slider mechanism. Image taken from [48]

(c) An example of a lead-screw. Image taken from [22]

Figure H.6: The three selects concepts to translate the rectilinear motion of the pneumatic actuator

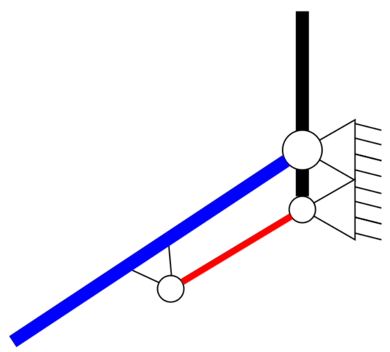
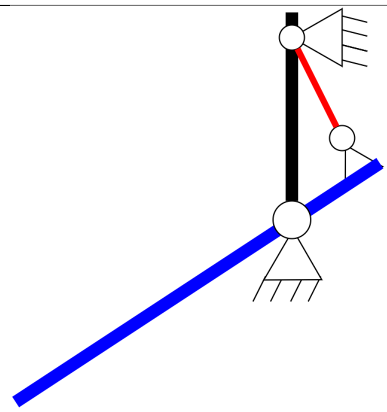
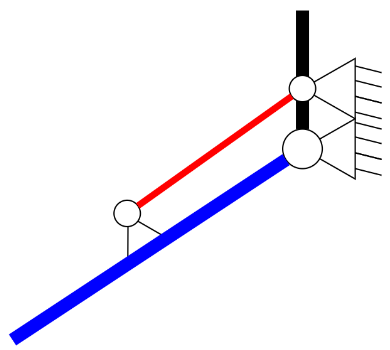
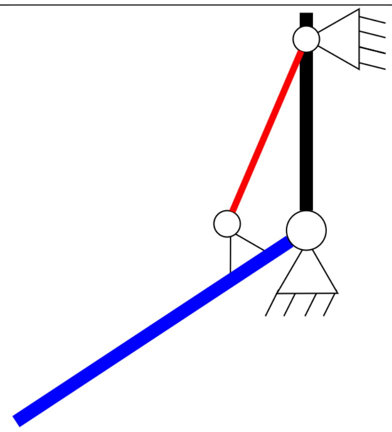
Table H.1: An overview of the three selects concepts to translate the rectilinear motion of the pneumatic actuator

	Lever	**Crank-Slider**	**Lead Screw**
Axis of Rotation	Perpendicular to linear motion	Perpendicular to linear motion	Aligned with linear motion
Range of Motion	Depends on arm length and stroke length	Depends on arm length and stroke length	Depends on screw lead and stroke length
Torque	Depends on arm length and piston force	Depends on arm length and piston force	Depends on screw lead and piston force
Holding Torque	Nothing extra	Nothing extra	High, because of friction on screw lead
Maximum Torque	High	High	Low-Medium
Moving Cylinder?	Yes, rotation	No	No
Size	Medium	Medium	Medium
Simplicity of Configuration	High	Medium	Medium

Flexion-Extension Configuration

A screw principle would not be applicable because of the working axis combined with the high torque requirements. A large cylinder would have have to be placed perpendicular to the forearm, resulting in an non-anthropomorphic design, affecting both comfort and cosmetics. Keeping the design value of simplicity in mind, the preference goes to a simple lever. The trade-off is a rotating cylinder, thus additional attention should be paid to the stress relief of the gas tubes and cables. The choice of a double-acting cylinder eliminates the need of having two cylinders for one powered movement. The four configuration concepts are shown in Appendix H.

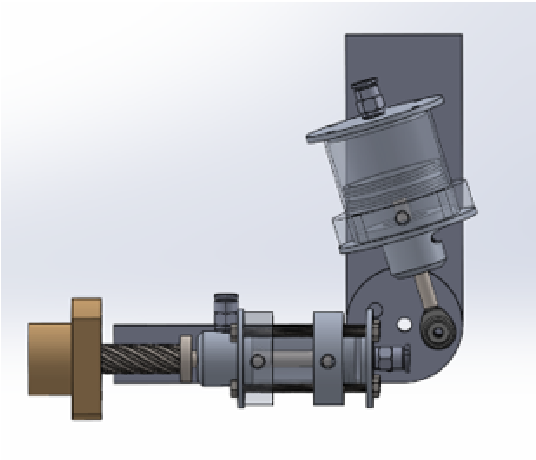
Table H.2: The four different proposed configurations to achieve flexion/extension of the elbow

Base	**Forearm**	**Upper Arm**
	Configuration 1	**Configuration 3**
		
	Configuration 2	**Configuration 4**
		

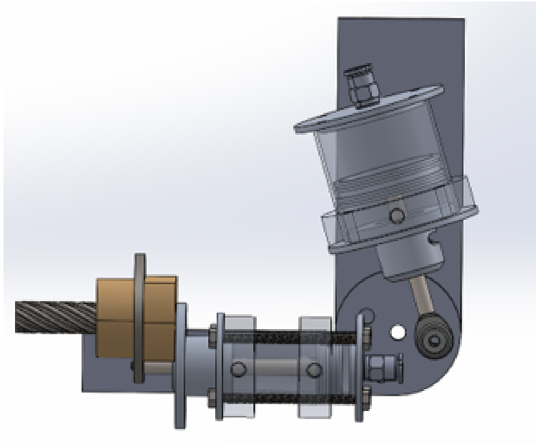
Using the upper arm as a base, the prosthesis will not be usable for patients with a elbow disarticulation. Therefore, the preference goes to using the forearm as a base. During the further conceptualization, using the forearm as a base was no longer an option. With required cylinder size due to the maximum pressure and with the decided upon configuration of pronation/supination (Appendix H) there was too little space available for two cylinders in the forearm. The resulting full concept was bulky, visually non-appealing and limiting the RoM of the flexion extension. The same reasoning eliminates configuration 4. The chosen configuration is configuration 3.

Pronation-Supination Configuration

For pronation/supination, the desired axis of rotation is aligned with the forearm. A lever or crank-slider method to achieve rotation would require the cylinder to be placed perpendicular to this axis. The only viable option for this is to place it in the upper arm, but this should be avoided to have the prosthesis be usable for patients with an elbow disarticulation. The remaining concept is using a lead-screw. The relatively low torque requirements, stationary cylinder and additional holding torque provide further benefits to this concept. Two configurations are discussed: Connect the lead-screw to the piston rod or connect the lead-screw nut to the piston rod.



(a) A configuration where the PS actuator drives the lead-screw



(b) A configuration where the PS actuator drives the lead-screw nut

Figure H.7: The two discussed configurations to achieve pronation/supination

Rotation The piston can freely rotate within the cylinder. This is no problem for the FE configuration since the end-effector restricts this rotation. However, for a lead-screw, this poses a challenge as it depends on rotation for movement. If the rotation of the piston is not restricted, the piston could rotate in stead of the desired output rotation of the forearm. For the first configuration the lead-screw and piston-rod are in series, thus a secondary piston rod is required to restrict the rotation of the piston, see Figure H.8. This leads to additional mass and components. Furthermore, this is only viable with a custom-built piston. For the second configuration, the lead-screw and piston-rod are parallel. If the lead-screw is constrained in all DoF except the desired rotation around its axis, this configuration prevents rotation of the piston.

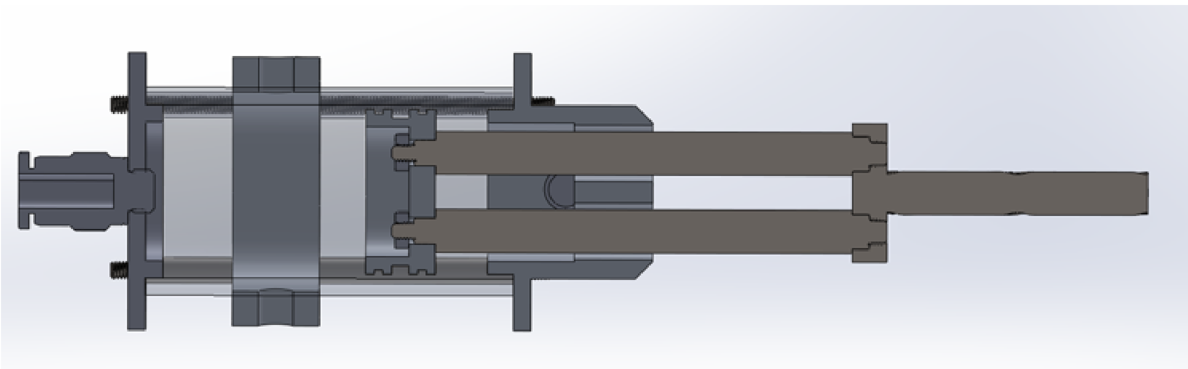
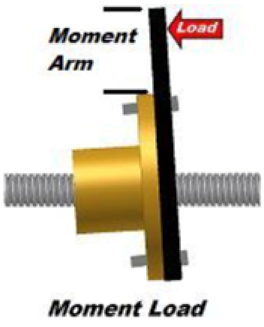
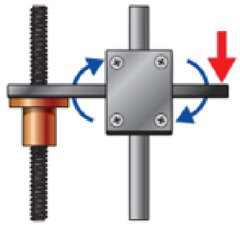


Figure H.8: A configuration where two piston-rods restrict the rotation of the piston

Force Distribution The first configuration has the lead-screw and actuator in series, which is the most suitable configuration considering force distribution. For the parallel configuration, the off-axial force on the nut can cause additional friction, as shown in Figure H.9a. However, the piston-rod serves as a guiding rod, partially negating this issue, see Figure H.9b.



(a) The problems of a non-aligned force on a lead-screw nut. Image taken from [49]



(b) The problems of a non-aligned force on a lead-screw nut solves by using a guide. Image taken from [50]

Figure H.9: Two configurations to achieve pronation/supination

Size Size plays a further role in the selection between the two configurations. The when driving the lead-screw, the prosthesis becomes quite long as twice the length of the lead-screw plus the length of the actuator is used. when driving the lead-screw nut, only the actuator size and length of the lead-screw will be used.

Encoder Implementation The implementation of an encoder is desired, see Appendix K. A rotational encoder is desired for its small size. However, such an encoder should be implemented on-axis. This is not possible for the first configuration. This, combined with the parameters discussed above, leads to a choice for the second configuration: Drive the lead-screw nut.

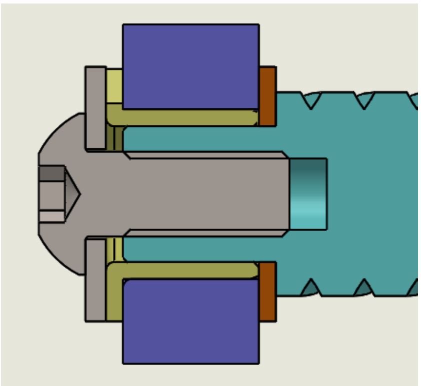
Lead-screw Collar

The collar provides the lead-screw with the constraints in DoF, whilst limiting friction as much as possible. The lead-screw should only be allowed to rotate around its own axis of rotation. Radial translation and rotation other axis other than principle axis can be constrained with a bushing in the collar. The lead-screw should have a flush section to achieve this without damaging the bushing. Axial translation can be constrained by clamping the collar between the lead-screw and and a bolt/washer. To achieve this, the diameter of the flush section should be smaller than the lead-screw diameter, this is illustrated in Figure H.10a.

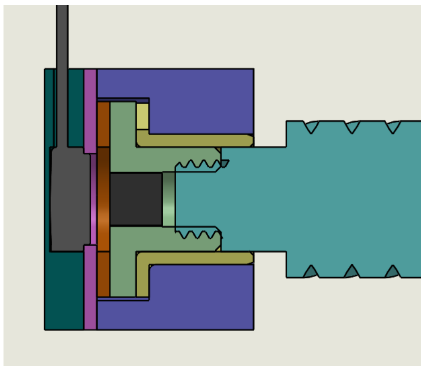
Encoder Implementation This concept does have a major drawback: No encoder is implemented. This can be solved by either implementing a linear encoder on the actuator or by revising the concept so that an encoder can be implemented. The first was not preferable, thus the latter option is chosen. This is achieved by removing the bolt and washer, and inserting new custom parts: The magnet-nut, encoder-base and encoder-holder. The magnet-nut holds the magnet used for the encoder and functions to constrain the axial movement of the lead-screw, thanks to being clamped between the shoulder of the bushing and the axial-bushing, which is now placed between the magnet-nut and encoder-base. The encoder-base and -holder function to properly constrain the encoder.

Locking of Pronation/Supination After a revision, the locking of PS will not be done using the same mechanism as the locking for FE. For more details on this decision, see Appendix I. The simplest method to achieve locking is using a set screw in the collar.

Frame Connection The lead-screw is only properly constrained if the collar is fully constrained. This is achieved by connecting the collar to the frame using four bolts, as shown in Figure H.11.



(a) A concept for the lead-screw collar



(b) A concept for the lead-screw collar with the implementation on an encoder

Figure H.10: Two concepts for the collar to suspend the lead-screw. Cyan: Lead-screw, Purple: Collar, Yellow: Bushing, Orange: Axial-Bushing, Grey: Nut & Washer, Light green: Magnet-Nut, Pink: Encoder-base, Teal: Encoder-Holder, Dark grey: Encoder, and Black: Magnet

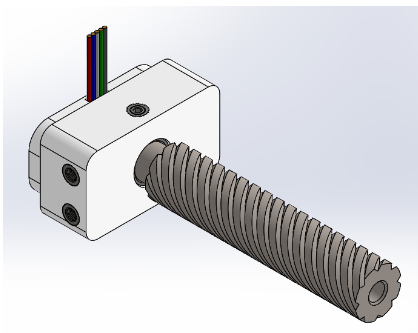


Figure H.11: The method to connect the collar to the frame

Embodiment

General Actuator

Working Pressure Pneumatic actuators have the option to be small and lightweight, whilst still providing a high force. If the working pressure is doubled, the output force is doubled. The working pressure can not be increased indefinitely, as there are three main limiting factors: (1) allowed pressure for off-the-shelf components, (2) gas consumption, and (3) cylinder thickness, Most of the pneumatic components will be bought at either https://eriks.nl/nl/ or from https://www.festo.com/cms/nl_nl/index.htm. The components in the size/weight categories suitable for the prosthesis are mostly limited at 10 - 12 bar. The pressure of the systems directly influences that gas consumption. Professor Plettenburg has studied this and advised to use a working pressure of 12 [bar] for optimal gas efficiency. Since this is in-line with the maximum pressure of the off-the-shelf components 12 [bar] will be used as the working pressure for the system. The cylinder thickness will be designed to be strong enough to handle this pressure.

Caps Conceptual design led to the actuator caps fulfilling two functions: To seal the housing and to allow for attachment to the frame. For the attachment to the frame, two M3 bolts will be used at a 4 [mm] depth, creating a 24 [mm^2] surface area for the force to work on the caps. For simplicity sake, it is desired to have the actuators be as similar as possible. Thus the heaviest load scenario will be used to determine the allowed materials. This is the 1500 [N] generated by the FE-actuator. The yield strength of the chosen material for the caps should be at least 62.5 [MPa] to prevent deformation and leaks. Steel, aluminum and POM are considered as candidates, shown in Table H.3. Depending on the type of POM used, it might be a suitable material for the caps. However, the safety margin would be too low, and the time pressure of the thesis assignment led to the decision to avoid this risk. Aluminum is chosen as the material since the yield is sufficiently high and the density is about three times lower than steel.

Housing For a cylinder, an internal pressure causes two distinct types of stress in the cylinder wall: hoop stress and longitudinal stress. If $\frac{D}{t} > 20$, the cylinder can be regarded as being "thin-walled" and the following equations can be used to estimate the stresses, see Appendix H and Appendix H. It can be seen that for a cylinder with a constant thickness, the hoop stress will be twice as high. Furthermore, the chosen concept for the actuators use caps which will be significantly thicker than the wall thickness. Thus, only the hoop stress will be taken into consideration. To estimate the required thickness, the mean diameter (D_m) is assumed to be the inner diameter. Table H.3 shows that all three required thicknesses are < 1 [mm], which will cause problems with manufacturing. A minimum thickness of 1 [mm] is set and used to calculate the mass of the housing. POM would be the lightest option, however the aluminum housing can be bought at the required sizes at https://salomons-metalen.nl/. This saves time and costs during the manufacturing. Furthermore, pneumatic cylinders using POM as a housing have not been seen before and the <10 [g] gain is not worth risking this novel method. Finally, materials with a higher yield strength can support the load bearing functions of the frame. For all these reasons, aluminum is chosen as the housing material.

$$\sigma_\theta = \frac{p * D_m}{2 * t} \tag{H.2}$$

H.2: The equation to estimate the hoop stress for a thin-walled cylinder. σ_θ is the hoop stress [MPa], p is the internal pressure [MPa], D_m is the mean diameter (which is the inner diameter + t) [m], and [t] is the cylinder thickness

$$\sigma_z = \frac{p * D_m}{4 * t} \tag{H.3}$$

H.3: The equation to estimate the longitudinal stress stress for a thin-walled cylinder. σ_z is the hoop stress [MPa], p is the internal pressure [MPa], D_m is the mean diameter (which is the inner diameter + t, and [t] is the cylinder thickness

Table H.3: The thee material candidates for the actuator. FE-actuator parameters used: F = 1500 [N], d = 40 [mm], $L_{Housing}$ = 56.5 [mm], $L_{Piston\text{-}rod}$ = 52 [mm]

	Steel 304L[1]	**Aluminum 6060** [1]	**POM**
Yield Strength [MPA]	180	160	22-78
Density [kg/m^3]	8.1.	2.7	1.4
		Housing	
t [mm]	0.13	0.15	0.4
m [g] (for t = 1 [mm])	58.9	19.6	10.2

[1]Chosen from the available materials at https://salomons-metalen.nl/

Piston Rod The end-effector for the FE-actuator is a rod-end, connected to the piston-rod using an M5 thread. It is desired to keep the diameter of the piston-rod as small as possible. This can best be achieved using a steel rod. This was also advised to be best for production by Jan van Frankenhuyzen.

Pronation/Supination

The nut can be implemented directly on the lead-screw or use ball-bearings to achieve even lower friction and higher efficiencies. By implementing a ball-screw, the efficiency could be increased from around 75% to over 90%. However, this increases the mass of the system and decreases the resistance of the prosthesis to micro-particles such as sand. Furthermore, a friction within the transmission can be helpful, up to a certain degree, by dampening the effects of external forces.

Table H.4: The most relevant information of the selection of lead-screws to be used for PS torque generation (https://www.igus.nl/)

Lead Diameter	**⌀ 10 [mm]**	**⌀ 12 [mm]**	**⌀ 14 [mm]**	**⌀ 16 [mm]**
Allowed Force [N]	780	1405	1440	1669
Material Lead-Screw	Steel	Steel	Steel	Steel
Mass Steel Lead-screw [g/mm]	0.62	0.89	1.22	1.59
Mass Nut [g]	23.7	39.2	37.2	34.6
Diameter Nut [mm]	42	48	48	48
Length Nut [mm]	25	35	35	35
Possible Thread Pitch [mm]	12	5	25	35
	25	25	30	-
	50	-	40.6	-

Table H.5: The table used to determine which combination of piston diameter and lead-screw to use for the generation of torque for pronation/supination. The "mass" is the mass of the lead-screw for the required length + the mass of the nut.

Piston Diameter	**⌀ 25 [mm]**	**⌀ 32 [mm]**	**⌀ 40 [mm]**
Actuator Force [N]	589	965	1508
Lead Diameter: 10 [mm]	Pass	Fail	Fail
Lead Diameter: 12 [mm]	Pass	Pass	Fail
Lead Diameter: 14 [mm]	Pass	Pass	Fail
Lead Diameter: 16 [mm]	Pass	Pass	Pass

Original Data

Diameter x Pitch	λ	η	**Length**	**Mass**	**Torque**		
[mm]	[°]	[%]	[mm]	[g]	[Nm]		
10x12	20.9	72.6	49.0	54.1	0.8	-	-
10x25	38.5	78.6	55.5	58.1	1.8	-	-
10x50	57.9	75.2	68.0	65.9	3.5	-	-
12x5	7.6	51.7	55.5	88.6	0.2	0.4	-
12x25	33.6	77.9	65.5	97.5	1.8	3.0	-
14x25	29.3	76.9	65.5	116.1	1.8	3.0	-
14x30	34.3	78.1	68.0	120.2	2.2	3.6	-
14x40.6	42.7	78.7	73.3	126.6	3.0	3.8	-
16x35	34.8	78.2	70.5	146.7	2.6	4.2	6.6

Flexion/Extension MATLAB Code

```
clear all; clc; close all
% Jim Kieft
% Flexion/Extension
%% Input Parameters
% Human Topology
L_User = 1.80;                    % Length user [m]
L_Upperarm = 186*L_User;          % Upper arm length [mm]
L_Forearm = 146*L_User;           % Forarm length [mm]
L_Hand = 106*L_User;              % Hand length [mm]
L_toHand = L_Forearm+54*L_User;   % Distance from elbow joint to mass inhand [mm]
L_CoG_Below_Elbow = 0.5*(L_Forearm+L_Hand); % Distance from elbow joint to CoG Lower ...
    arm [mm]
RoM = [0;140];                    % Range of Motion [deg]

% Goals
m1 = 1.5;      % Desired mass in hand [kg]
m3 = 3;    % Minimum mass in hand [kg]
m4 = 4;    % Desired mass in hand [kg]
m_prosthesis = 0.5; % Expected mass of proshesis
% Natural parameters
g = 9.81;    % Gravity constant [kg/ms^2]

% Cylinder parameters
Pressure = 1.2;       % Pressure in cylinder [Mpa] (12 bar)
Radius_C16 = 16;          % Inner radius cylinder [mm]
Radius_C20 = 20;          % Inner radius cylinder [mm]
Radius_C25 = 25;          % Inner radius cylinder [mm]
Radius_P = 4;          % Radius piston [mm]

L_Act = 56.5;          % Cylinder length [mm]
L_Stroke = 26;         % Stroke [mm]
L_Rod_offset = 13;     % Distance between rod end and Cap when maximally retracted [mm]
L_Act_End_Cap = 52;    % Distance between connection to the frame and End-Cap[mm]
L_Act_Cap = 4.5;       % Distance between connection to the frame and Cap[mm]

% Attachment points
X_Fore = -12;  % Connection point forearm [mm]
Y_Fore =   2;  % Connection point forearm [mm]
X_Upper = -1;  % X_location attachement upperarm [mm]
Y_Upper = 31;  % Y_location attachement upperarm [mm]

X_E = [0;0];              % Location elbow [mm]
X_S = [0;L_Upperarm]; % Location shoulder [mm]
X_2 = [X_Upper;Y_Upper];      % Location Connection 2 [mm]
%% Constants & Calculations before starting
% Force
Area_P = pi*Radius_P^2;        % Area piston [mm^2]

Area_C16 = pi*Radius_C16^2;        % Area cylinder [mm^2]
Area_E16 = Area_C16-Area_P;        % Effective pressure area [mm^2]
F_max_16 = Pressure*Area_E16;      % Max piston force [N]

Area_C20 = pi*Radius_C20^2;        % Area cylinder [mm^2]
Area_E20 = Area_C20-Area_P;        % Effective pressure area [mm^2]
F_max_20 = Pressure*Area_E20;      % Max piston force [N]

Area_C25 = pi*Radius_C25^2;        % Area cylinder [mm^2]
Area_E25 = Area_C25-Area_P;        % Effective pressure area [mm^2]
F_max_25 = Pressure*Area_E25;      % Max piston force [N]

%%
A = sqrt(X_Fore^2+Y_Fore^2); % Distance attachement lower arm to origin [mm]
B = sqrt(X_Upper^2+Y_Upper^2); % Distance attachement upper arm to origin [mm]
%% Angles
theta = linspace(90-RoM(1),90-RoM(2),max(RoM)-min(RoM)+1)';
FE = -(theta-90);
Gamma = theta+90;

```

```
Alpha_Upper = atand(X_Upper/Y_Upper);    % Attachment connector angle C2 [deg]
Alpha_Fore = atand(Y_Fore/X_Fore);       % Attachment forearm angle C2 [deg]
%% Topology
L_X = -cosd(theta).*L_Forearm;
L_Y = -sind(theta).*L_Forearm;

Y2_1 = sind(theta).*X_Fore;
X2_1 = cosd(theta).*X_Fore;
C2_1 = [-X2_1,-Y2_1]; % Attachment point actuator 2

Y2_2 = Y2_1+cosd(theta).*Y_Fore;
X2_2 = X2_1-sind(theta).*Y_Fore;
C2_2 = [-X2_2,-Y2_2]; % Attachment point actuator 2 + displacement

S = (Y_Upper+Y2_2)./(X_Upper+X2_2);
Psi2 = atand(S);

E_A = [X_Upper-L_Act_End_Cap*cosd(Psi2),Y_Upper-L_Act_End_Cap*sind(Psi2)]; % End ...
    position Actuator 2 A
E_B = [X_Upper+L_Act_Cap*cosd(Psi2),Y_Upper+L_Act_Cap*sind(Psi2)]; % End position ...
    Actuator 2 B

LA2 = sqrt(B^2+X_Fore^2-2*B*X_Fore*cosd(Gamma+Alpha_Upper-Alpha_Fore));    % Length ...
    Actuator 2

%% Torques
T_mp = m_prosthesis*g*L_CoG_Below_Elbow*sind(FE)/1000;

T_m1 = m1*g*L_toHand*sind(FE)/1000;
T_m2 = m4*g*L_toHand*sind(FE)/1000;

Tm1 = T_mp+T_m1;
Tm2 = T_mp+T_m2;


%Phi = FE-abs(Alpha_Fore)-abs(Alpha_Upper_;
Phi = FE-Alpha_Fore-Alpha_Upper;

C = sqrt(A^2 + B^2 - 2*A*B*cosd(Phi));
Phi2 = acosd((B^2-A^2-C.^2)./(-2*A*C));
Phi3 = acosd((A^2-B^2-C.^2)./(-2*B*C));

T_16 = F_max_16*B*sind(Phi3)/1000;  % Max torque
T_20 = F_max_20*B*sind(Phi3)/1000;  % Max torque
T_25 = F_max_25*B*sind(Phi3)/1000;  % Max torque


%% Finding the intersection between T and Tm2
for i = 1:length(T_20)
    if T_20(i) ≤ Tm2(i)
        Intersection = i;
        break
    end
end

%% Stroke Length
Max_Stroke = L_Stroke*ones(size(theta));
Stroke = C-L_Rod_offset-L_Act_Cap;
if min(Stroke) ≤ 0
        disp('Stroke < 0')
end

for i = 2:size(Stroke)
    if Stroke(i) ≤ Stroke(i-1)
        disp('minimum')
    end
end
%% Plot
Vsh=zeros(length(LA2),3);% initalise video

fig1 = figure(1);
```

```
for i = 1:length(LA2)

    % Subplot 1: Movement
    p1 = subplot(1,2,1);
    axis equal
    hold on
    set(gca,'XTick',[])
    set(gca,'YTick',[])
    %axis([-350 100 -300 300])
    axis([-140 70 -70 140])

    plot([X_E(1),X_S(1)],[X_E(2),X_S(2)],'b','LineWidth',10)   % Plot upper arm
    plot([L_X(i),X_E(1)],[L_Y(i),X_E(2)],'b','LineWidth',10)   % Plot forearm
    plot([X_E(1),X_2(1)],[X_E(2),X_2(2)],'b','LineWidth',4)   % Plot extension piece ...
        upper arm
    plot([C2_2(i,1),0],[C2_2(i,2),0],'b','LineWidth',10)   % Plot extension piece forearm

    plot([C2_1(i,1),C2_2(i,1)],[C2_1(i,2),C2_2(i,2)],'k','LineWidth',2)   % Plot linear ...
        actuator configuration 2
    plot([E_A(i,1),X_2(1)],[E_A(i,2),X_2(2)],'k','LineWidth',10)   % Plot linear ...
        actuator end 2A
    plot([E_B(i,1),X_2(1)],[E_B(i,2),X_2(2)],'k','LineWidth',10)   % Plot linear ...
        actuator end 2B
    plot([C2_2(i,1),E_B(i,1)],[C2_2(i,2),E_B(i,2)],'r','LineWidth',5)   % Plot linear ...
        actuator configuration 2

    plot(X_E(1),X_E(2),'k*','LineWidth',5)    % Plot elbow location
    plot(X_S(1),X_S(2),'k*','LineWidth',4)    % Plot shoulder location
    plot(X_2(1),X_2(2),'ko','LineWidth',2)    % Plot normal attachment
    plot(C2_2(i,1),C2_2(i,2),'ko','LineWidth',5)    % Plot attachment location forearm

    % Subplot 2: Angle vs. Actuator length
    p2 = subplot(2,2,2);
    axis([min(RoM) max(RoM) min(Stroke)-10 max(Stroke)+10 ])
    xlabel('Elbow angle [deg]')
    ylabel('Stroke [mm]')
    set(gca,'FontSize',20)
    hold on
    plot(FE(i),Stroke(i),'r*','LineWidth',4)
    plot(FE(i),Max_Stroke(i),'k.','LineWidth',1)
    %plot([0,180],[L_Stroke,L_Stroke],'k:','LineWidth',2)   % Plot upper arm
    %plot([0,180],[0,0],'k:','LineWidth',2)   % Plot upper arm

    % Subplot 3: Angle vs. Max Torque
    p3 = subplot(2,2,4);
    axis([min(RoM) max(RoM) 0 max(max(T_20))+2 ])
    xlabel('Elbow angle [deg]')
    ylabel('Maximum torque [Nm]')
    set(gca,'FontSize',20)
    hold on
    plot(FE(i),Tm1(i),'k.')
    plot(FE(i),Tm2(i),'k*')
    plot(FE(i),T_20(i),'r*')
    legend('1.5 kg in hand','4 kg in hand','Prosthesis Torque')
    %pause(0.2)
    drawnow

    % Clear subplot 1 for clarity
    if i < length(LA2)
        cla(p1)
    end
end
%% Stroke graph: Angle vs. stroke
figure
axis([min(RoM) max(RoM) min(Stroke)-10 max(Stroke)+10 ])
xlabel('Elbow angle [deg]')
ylabel('Stroke [mm]')
set(gca,'FontSize',20)
hold on
plot(FE,Stroke,'r','LineWidth',4)
plot(FE,Max_Stroke,'k--','LineWidth',4)
```

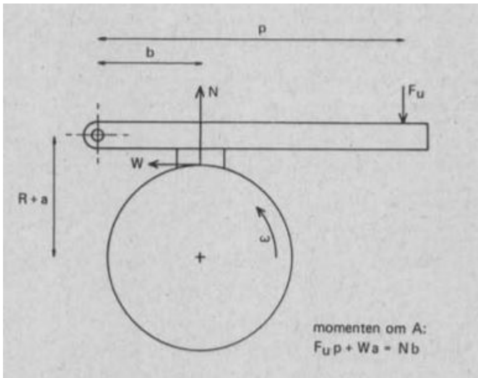
```
plot( [find(Stroke==min(Stroke)) find(Stroke==min(Stroke))],[min(Stroke)-10 ...
    max(Stroke)+10],'Color',uint8([100 100 100]),'LineWidth',2)
legend('Stroke Length','Max Stroke Length','Angle where minimum ...
    occurs','Location','southeast')

%% Torque graph: Angle vs. Max Torque
figure
axis([min(RoM) max(RoM) 0 max(T_20)+10 ])
xlabel('Elbow angle [deg]')
ylabel('Maximum torque [Nm]')
set(gca,'FontSize',20)
hold on
plot(FE,T_mp,'k.','LineWidth',4)
plot(FE,Tm1,'k--','LineWidth',4)
plot(FE,Tm2,'k','LineWidth',4)
plot(FE,T_20,'r','LineWidth',4)
plot([Intersection Intersection],[0 max(T_20)+15],'Color',uint8([100 100 ...
    100]),'LineWidth',2)
legend('0 kg in hand','1.5 kg in hand','4 kg in hand',sprintf('Prosthesis Torque\n D=40 ...
    mm'), sprintf('Intesection\nTorque_{Prosthesis}<Torque_{Goal}'))
```

I

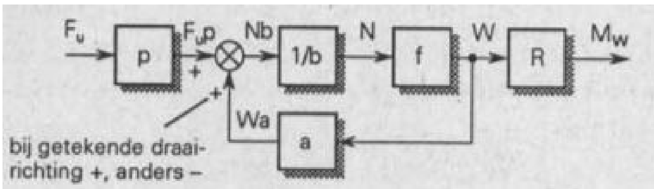
Locking Mechanism

Conceptual Design

The goal is thus to have a small active locking mechanism which can hold a considerable torque in all positions without requiring actuation to lock. Professor Plettenburg advised to check the chapter ”Mechanische Versterkers” of the book ”Werktuigkundige Systemen” by Jan C. Cool [23]. In this book, a ”meekoppelinig” is discussed. The main principle of this mechanism is redirecting a portion of the generated force to the input, thus increasing said input force. The book gives an example, see Figure I.1, where the generated friction force W helps with generating a moment around pivot point A, thereby increasing normal force N and itself (W) as a consequence. This example can be expressed mathematically as Equation (4.1). If $\frac{b}{a} < f$ then $\frac{M_W}{F_u} \rightarrow \infty$, thus a small input force can generate a high locking moment if the mechanism is properly designed.



(a) ”Met de aangegeven draairichting zorgt de optredende wrijvingskracht voor ondersteuning van het aandrijfmoment Fu’p. Bij wijziging van de draairichting verandert de meekoppeling in een tegenkoppeling, zie [Figure I.1b]”. Image and caption taken from [23]



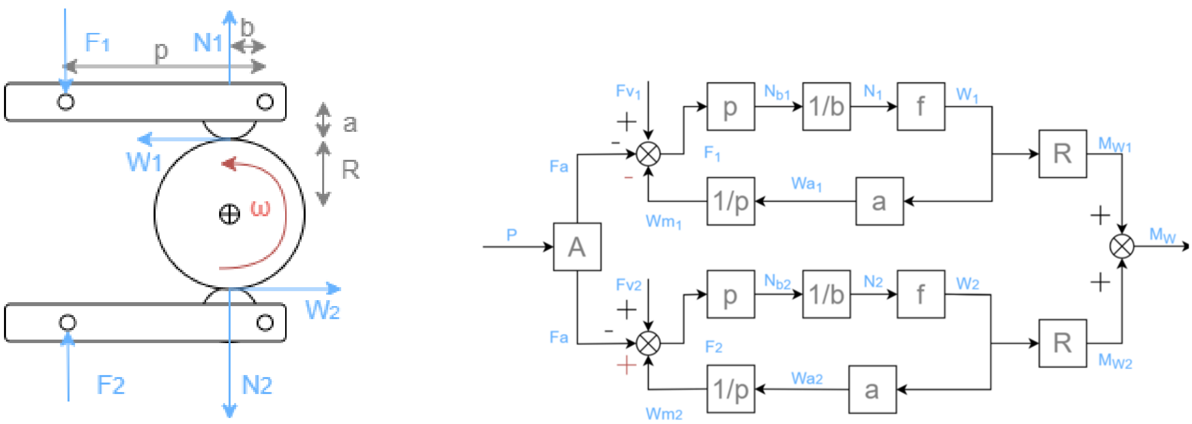
(b) ”Het blokschema van de opstelling van [Figure I.1a]. Het + teken (mee koppeling) geldt voor de aangegeven draairichting. Indien H = -a·f/b < -1, dus als b/a < f treedt blokkeren van de remschijf op.” Image and caption taken from [23]

Figure I.1: A ”Meekoppeling” mechanism proposed by J. Cool [23]

$$\frac{M_w}{F_u} = \frac{pRf}{b - af} \tag{I.1}$$

I.1: Where M_w is the friction moment on the disk [Nm], F_u is the input force [N], f is the coefficient of friction between the lever and the disk [-], R is the radius of the disk [m], a is the vertical distance between the connection point and the pivot point [m], b is the horizontal distance between the connection point and the pivot point, p is the horizontal distance between the acting line of the input force F_u and the pivot point [m]

Topology This mechanism will only work for rotation in one direction, counterclockwise in the example. Since the elbow prosthesis requires locking in both directions, the example provided by J. Cool is mirrored to generate a concept, see Figure I.2. The geometric requirement of $\frac{b}{a} < f$ still holds up to achieve $\frac{M_W}{F_u} \to \infty$.



(a) A schematic diagram

(b) An updated version of the block diagram

Figure I.2: An updated version of the "meekoppeling" with two levers to restrict both clockwise and counter-clockwise motion

Actuation A method to generate the input force is required. The other actuators use pneumatic actuation, thus to keep the prosthesis simple, the same method of actuation will be used. The locking is engaged when the forces push towards each other, and disengaged when there is no force or if the forces are pushing away from each other. As one of the wishes is to have passive engagement/active disengagement of the locking mechanism, the best choice for actuation is to use a single-acting push pneumatic cylinder. This is illustrated in Figure I.3.

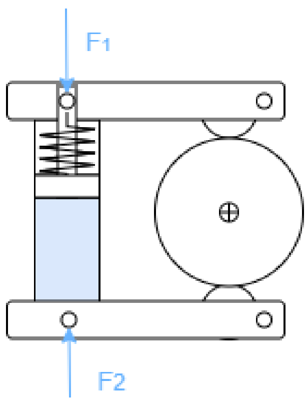


Figure I.3: The proposed method of actuation, a single-acting push pneumatic cylinder, implemented in the concept shown in Figure I.2

LM Actuator

The goal of the actuator is not to generate movement, but to cancel the spring force. The actuator of the locking mechanism is single-acting pneumatic push cylinder. The dimensions of the actuator are determined by the available springs, pistons and required input for for locking engagement. The force generated by the cylinder, when activated, should be higher than the force generated by the spring. Furthermore, it is desired that the actuator is as short as possible to reduce the bulkiness of the mechanism.

To eliminate the need for a (second) pressure regulator, the working pressure of the LM actuator will be the same as the pressure used for both the FE actuator and PS actuator: 12 bar. The designed cylinder consists out of the following parts (also illustrated in Figure 4.7a:

- Custom: LM-Housing
- Custom: LM-Piston-Rod
- Custom: LM-Spring-Washer
- LM-Piston
- LM-Spring
- LM-Bushing
- Air-Coupler
- Bolt & Washer

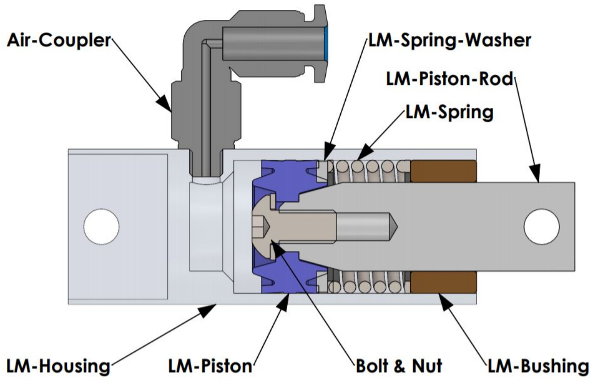


Figure I.4: A cross section of the LM actuator

It can be seen than, in contrast to i.e. the PS- and FE actuators, there are no ”caps”. The reason for this is that the housing can be designed in such as way that it will function as a cap on one side. On the other side of the piston, there is no need for a sealed chamber, since the actuator is single-acting.

LM-Housing The housing of the LM actuator is a complex part which fulfills several functions: (1) Guide the piston, (2) Provide a connection to a lever, (3) House the bushing, (4) Create an ”airtight” chamber with connection to the air coupler, and (5) Withstand the external forces and moments.

LM-Piston-Rod The piston rod transfers the movement of the piston to a lever. It is connected to the piston using a bolt and washer on one side and the shape of the piston rod on the other. The bushing guides the motion of the piston rod, whilst simultaneously keeping the friction forces low.

LM-Piston The piston moves when force generated the air pressure in the chamber overcomes the spring force.

LM-Spring The spring serves as the F_u of Equation (4.1). It pushes the piston rod ”in” by pushing at its connected piston

LM-Spring-Washer A washer is needed to prevent the spring from damaging the piston.

LM Frame

The primary function of the frame is to transfer the input force (F_u) from the actuator to the braking disk, following the topology design limitations shown in Equation (4.1). A secondary function is to transfer the forces of the free-swing mode disengagement method. The frame of the locking mechanism consists out the following components (illustrated in Figure I.5:

- Custom: LM-Lever
- Custom: LM-Disk
- ”Long” Bolt
- Bolts

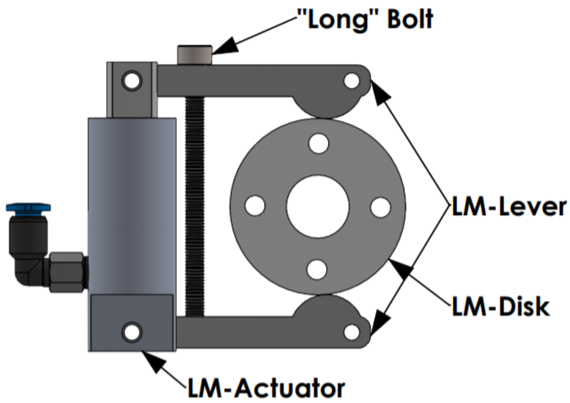


Figure I.5: The frame of the locking mechanism

LM-Lever The topology of the lever is constrained by Equation (4.1), the size of the actuator, available space on the frame of the prosthesis, and the radius of the disk. The further away the attachment points of the forces, the higher the locking moment. The tedious iterative process to achieve the final dimension will not be explained. The material used for the locking lever will be aluminum, chosen for its strength, low density, and availability to laser-cut.

LM-Disk The disk should fulfill two main functions: (1) provide an attachment for the lever to work upon, thereby generating the desired locking moment, and (2) connect to the upper arm frame. The hole in the middle is required for a connection in the frame, see Appendix J. The four smaller holes will serve to connect the disk to the upper arm using four bolts.

Embodiment

The selected concept consists out of two main modules: The actuator and the frame.

LM-Housing Aluminum is chosen as the material for the housing since it is lightweight, strong and can achieve a low surface roughness. The same air fitting as those used for the FE- and PS actuator is chosen.

LM-Piston-Rod The radius of the piston rod should be at least 7 [mm], to provide enough strength to the connection to the lever. The piston rod will have a required maximum surface roughness. Jan van Frankenhuyzen, who will help with manufacturing the custom parts, advised to use a 8 [mm] rod, since these can be bought with a predefined surface roughness and this will save time during manufacturing. Steel is chosen as the material for the piston rod to achieve the required strengths and surface roughness.

LM-Piston Pistons can be bought off-the-shelf at Eriks.nl. The smallest available diameters are 8, 10, 12, and 16 [mm].

LM-Spring In an ideal world, the topology of the system can amplify an input force of 1 [N] to generate a locking moment of ∞. However, since there are losses and inefficiencies, a minimum spring force of 50 [N] is desired. The springs are selected from the catalogue of Tevema.nl.

Piston & Spring Selection The 8 [mm] diameter piston can not be used because this is the desired diameter of the piston rod, leaving no space for the spring/bushing. Using the search engine on the site, the spring possibilities can be whittled down based on a minimum required force, a selected range of diameters, and by selecting springs with a high stiffness, which means that they can be rather small whilst generating a high force. Because of the minimum required piston rod diameter, all remaining spring have at least a minimum required housing diameter of [mm], thereby eliminating the 10 [mm] diameter piston. Furthermore, the remaining spring reach the required 50 [N], thus the 12 [mm] diameter piston is the smallest available option. The inner diameter of this chosen piston is 4.50 [mm]. The piston rod diameter at this end will thus be 4.50 [mm], the largest hole for a bolt is M3.

The inner diameter of the LM Housing will be 12 [mm]. The spring should be selected to fit within this housing. Two springs fit this criterion, with a minimum housing diameter of 12.01 [mm]: D12200 & D22200. These two springs are shown in Appendix I, where it can be seen that all parameters are identical, except for the stiffness and related force. A 12 [mm] piston can produce a force of 135 [N], both springs are weaker and can thus be overcome by actuation. The strongest of the two springs should be chosen to realize the highest possible locking force.

Spring	Price	L	L_{min}	C	F_{max}
D12200	2.75 €	20.00 [mm]	7.82 [mm]	7.11 [N/mm]	86.59 [N]
D22200	3.98 €	20.00 [mm]	7.82 [mm]	6.10 [N/mm]	74.37 [N]

LM-Bushing A bushing with an outer diameter of 12 [mm] and inner diameter of 8 [mm] is required. The only available option is a sintered bronze bushing. The forces on the bushing are low, thus the shortest available length is chosen: 6 [mm].

J Frame

Conceptual Design

Withstand Forces and Moments

It is vital for the prosthesis to withstand forces and moments exerted on it. The structure can be provided with an endoskeleton, like the bones of a human body, or in the form of an exoskeleton, like the shell of a beetle. A major advantage of using an exoskeleton is that a single module fulfills two vital functions: To provide strength and to protect internal components. Furthermore, such an exoskeleton would have a relatively high area moment of inertia, providing additional resistance against bending. However, since such an exoskeleton has has to protect the internal components as well, the size is larger than would be required of an endoskeleton which would only have to provide strength. Furthermore, the cosmetics of the prosthesis play a role in the selection of type of skeleton. An exoskeleton made of metal can be designed to be simple in order to be manufactured, however this will lead to an non-anthropomorphic design. It is possible to design sheet metal into an anthropomorphic shape, leading to a cosmetics. The downside of this is the complexity of the design. For this prototype, simplicity, production time and low costs are important, thus the ”Endoskeleton + Covers” concept is chosen.

For symmetry and simplicity, the decided upon concept is using four laser-cut sheets, see Figure J.1. To see the function of the holes, see Figure 4.14.

- Forearm-Lock
- Forearm-Encoder
- upper arm-Lock
- upper arm-Encoder

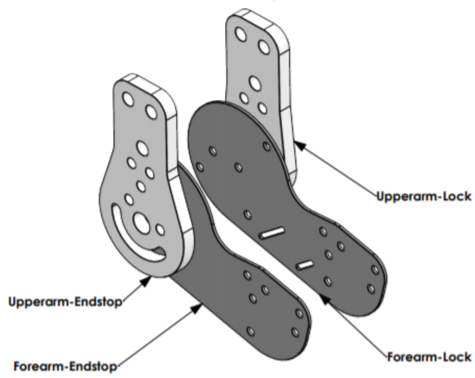


Figure J.1: The chosen concept for the load bearing structures of the frame, with the actuators hidden for better visibility

Constrain Unwanted DoF

An important aspect of the structure is to constrain the unwanted degrees of freedom, whilst producing as little friction as possible for the desired DoF. To prevent friction in rotary constraints, a bushing or bearing can be used. A bearing can have a coefficient of friction as low as 0.001, compared to an average 0.15 of a PTFE lined bushing. The downside of of bearings is the associated mass, bulkiness and vulnerability. The reduction of friction does not weight up to these negative aspects, which are aligned with the values and wishes of the design. SKF has a large variety of different bushings, shown in Table J.1. A light, flanged bushing is desired, the remaining options are a PTFE composite or PTFE Polyamide bushing. Of the two, the PTFE composite bushing has a higher permissible load and will therefore be used in the design.

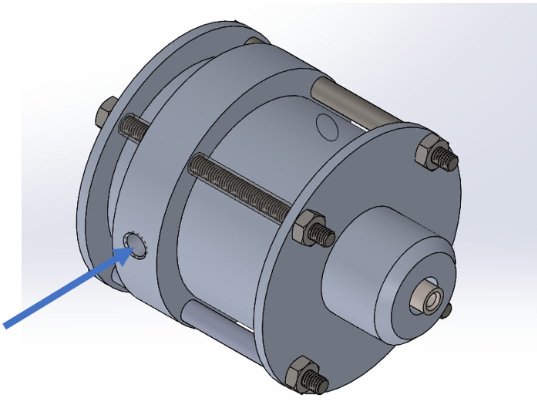
Table J.1: An overview of the different bushings. (g) means the bushing is greased

Bushing	**Permissible Load**		μ	**Mass**	**Maintenance Free Operation?**	**Available Flanged?**
	Static	Dynamic				
Solid Bronze	45 [MPa]	25 [MPa]	0.08-0.15(g)	Heavy	No	Yes
Sintered Bronze	20 [MPa]	10 [MPa]	0.05-0.10(g)	Heavy	Good	Yes
Wrapped Bronze	120 [MPa]	40 [MPa]	0.08-0.15(g)	Heavy	Suitable	Yes
PTFE Composite	250 [MPa]	80 [MPa]	0.03-0.25	Light	Excellent	Yes
POM Composite	250 [MPa]	120 [MPa]	0.02-0.20	Light	Good	No
PTFE Polyamide	80 [MPa]	40 [MPa]	0.06-0.15	Light	Excellent	Yes
Filament Wound	200 [MPa]	140 [MPa]	0.03-0.08	Light	Excellent	No

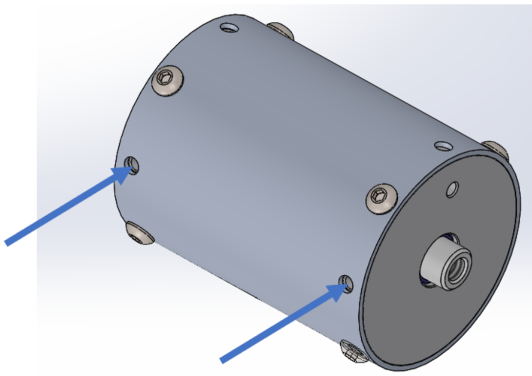
House Components

Due to a limit in available time, the decision was made to forgo this function in parts. The pneumatic and electronic circuits to control the prosthesis will be placed outside on a control board.

FE/PS-Actuator Implementation To house the actuators, there should be a connection between the actuators and the frame. The first concept uses the tie-rods from the actuator to constrain a custom part, which has threaded holes to allow for a connection to the frame, see Figure J.2a. Using spacers, the distance of the custom part to the cap(s) can be set. A downside of this concept is the additional mass of the custom part. Furthermore, the actuators were changed to not use tie-rods, thus this concept is not feasible. A second concept uses the threaded holes of the caps to connect to the frame, see Figure J.2b. A huge advantage of this concept is its relatively low bulkiness and mass. A downside of this concept is that the distance between the ends of the actuator and the connection can not be changed. This is especially relevant for the embodiment of the FE-actuator, since the actuator rotates around this connection. For the PS-actuator, the second concept is more beneficial since it allows for multiple connections to the frame. Playing around with the MATLAB script for the FE embodiment, see Appendix H, showed that the closer the axis of rotation is to the end of the actuator, the better the toque/angle graph. Thus, the second concept is chosen.



(a) A concept of attaching the actuator to the frame using the tie-rods of an actuator

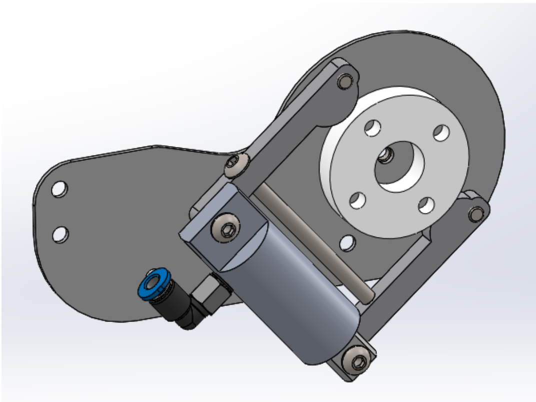


(b) A concept of attaching the actuator to the frame using the caps of an actuator

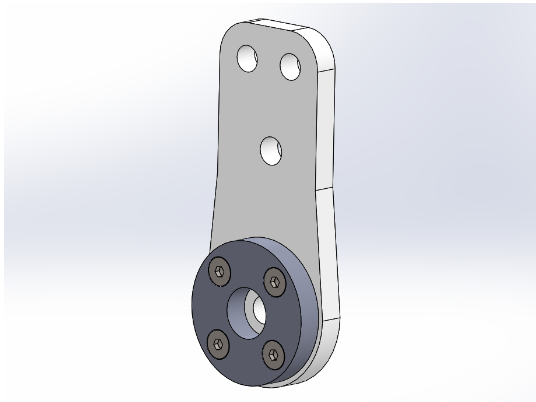
Figure J.2: The concepts for attaching the actuator to the frame

For the connection, a shoulder bolt will be used. Such a bolt has an integrated design of a threaded part and a shaft, eliminating the need for a custom part or more than one part. To reduce friction, a flanged bushing will be implemented.

LM Implementation To implement the locking mechanism, both the actuator and the LM-frame have to be connected to the general frame. For the locking to work properly, the brake disc should be attached to either the upper arm or forearm frame, and the levers should be attached to the other. Furthermore, the actuator should be attached to the same base sheet as the levers. The FE-actuator will have a larger diameter than the PS actuator. The distance between the forearm sheets will be smaller than the distance between the upper arm sheets. To keep the bulkiness of the prosthesis minimal, the LM-actuator should thus be attached to the forearm sheets. However, as the LM-actuator will be rather small, the options for attaching it to the frame are limited. There are no ”cap” and ”end-cap”, thus the same method used for the FE/PS-actuator implementation will not be applicable. Furthermore, if the actuator cannot transfer an axial force to the frame, this force will be transferred to the levers, increasing the locking torque. An attachment to the levers would suffice to constrain the actuator in almost all degrees of freedom. One DoF should still be constrained: Rotation around the axis of rotation of the brake disc. To constrain this DoF, two tie-wraps will be used. The brake disc will be connected to the upper arm sheet using four bolts to prevent slipping.



(a) The chosen concept of the implementation of the locking mechanism



(b) The chosen concept of the implementation of the locking mechanism brake disc

Figure J.3: The chosen concept for the locking mechanism

Embodiment

Connections

"Regular Connections" All bolts used in the system are M3, which is the smallest bolt to easily work with. The smallest shoulder bolts available on https://www.jeveka.com/en are M3, with a ⌀ 4 [mm] x 7 [mm] shaft. Steel backed, PTFE lined bushing will be used, which have an permissible dynamic load of 80 [MPa]. Equation (J.1) will be used to determine the minimal required bushing length. For the FE-connection, the load is 1500 [N], two connections are responsible for handling this load, both of which have a diameter of 4 [mm]. The safety factor used is 1.5. The minimal thickness to suffice is 3.5 [mm]. Bushing lengths are sold in integers of millimeters, to prevent additional manufacturing by trimming the bushings, a bushing length of 4 [mm] will be used. This process is similar for the connections between the sheets. The side where the locking mechanism will be implemented will use the same shoulder bolt and therefore use the same bushing.

$$L = S * \frac{F}{P * n * d} \tag{J.1}$$

J.1: The equation to calculate the minimal bushing length (L) [mm], S is a safety factor [-], F is the radial force on the holes [N], P is the permissible load [MPa], n is the amount of holes the force is divided over, and d is the hole diameter [mm]

Encoder Shaft The encoder shaft has to house the encoder magnet, which is 4 [mm] in diameter. A diameter of 6 [mm] is chosen for this shaft. Manufacturing is easier if the part can be made with the same diameter all over, therefore, the threaded connection will be M6. Furthermore, to prevent interference with the magnet, the encoder shaft is made from aluminum.

Sheet Material

The material selection for the sheets is based on the forces on the holes, whilst being as lightweight as possible. The stress on the material is approximated using Equation (J.2). The stresses found are compared to the yield stresses of three selected materials: Stainless steel (304), Aluminum (6061) and PMMA.

$$\sigma_{min} = S * \frac{F}{n * d * t} \tag{J.2}$$

J.2: The equation to calculate the stress (σ_{min}) [MPa], S is a safety factor [-], F is the radial force on the holes [N], n is the amount of holes the force is divided over, d is the hole diameter [mm], and t is the sheet thicknes [mm]

$$m = A * t * \rho \tag{J.3}$$

J.3: The equation to calculate the mass on the sheets (σ) [g], A is the surface area of the sheet [mm^2], t is the sheet thickness [mm], and rho is the material density [g/mm^3]

Table J.2: The thee material for the skeleton sheets

	Stainless Steel 304	**Aluminum 6061**	**PMMA**
Yield Strength [MPa]	≈505	≈260	≈65
Density [g/mm^3]	8.1E-3	2.7E-3	1.2E-3

Forearm The highest force on these sheets occurs at the connection-rod to the FE actuator. Everything connected to the forearm sheets can be a rigid connection, no bushing is required. Based on the overview shown in Table J.3, the lightest permissible material/thickness combination is a 2 [mm] aluminum sheet. However, at the time this was not possible to be laser-cut at 3mE. The second best option is a tie between 1 [mm] steel and 3 [mm] aluminum. Not only is the 1 [mm] steel less bulky, the same sheet can be used for multiple other parts: LM Spring Washer, FE Encoder Base, PS Encoder Base, and PS Nut Connect. The sheet chosen for the forearm frame will therefore be 1 [mm] stainless steel.

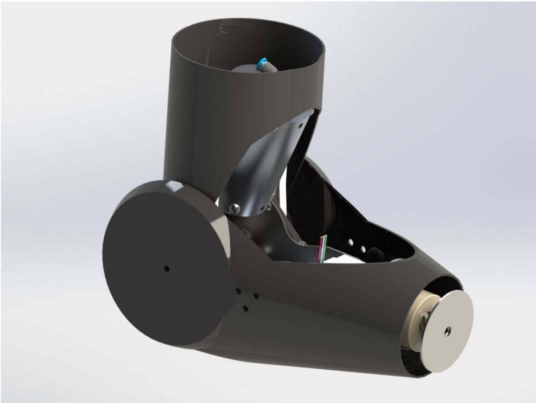
Table J.3: The material selection for the forearm sheets. Based on Equation (J.2) and Equation (J.3) : S = 1.5, F = 1500 [N], n = 2, d = 3.2 [mm], and A ≈3200 [mm^2]. If a material/thickness combination cannot handle the load without yielding, the cell is colored red

Thickness	**1 [mm]**	**2 [mm]**	**3 [mm]**	**4 [mm]**	**5 [mm]**
σ_{min}	351[MPa]	175 [MPa]	117 [MPa]	88 [MPa]	70 [MPa]
Mass					
Stainless Steel 304	25.9 [g]	51.8 [g]	77.7 [g]	103 [g]	129[g]
Aluminum 6061	8.6 [g]	17.3 [g]	25.9 [g]	34.6 [g]	43.2 [g]
PMMA	3.8 [g]	7.7 [g]	11.5 [g]	15.3 [g]	19.2 [g]

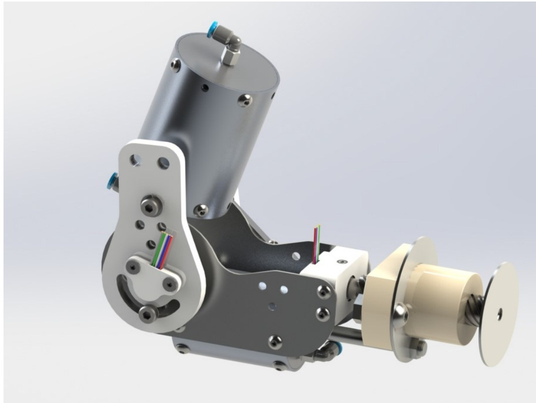
Upper arm The highest force on the upper arm sheets occurs on the connections to the FE cap and the connection between the two sets of sheets, both 1500 [N]. Because of the bushing length of 4 [mm], the thickness should be at least 4 [mm]. The outer diameter of the bushings is ⌀ 5.5 [mm]. This increase in outer diameter leads to an minimal required yield stress of 51 [MPa]. The lightest option for the upper arm sheets is a 4 [mm] PMMA sheet

Table J.4: The material selection for the forearm sheets. Based on Equation (J.2) and Equation (J.3) : S = 1.5, F = 1500 [N], n = 2, d = 5.5 [mm], and A ≈1800 [mm^2]. If a material/thickness combination cannot handle the load without yielding, the cell is colored red

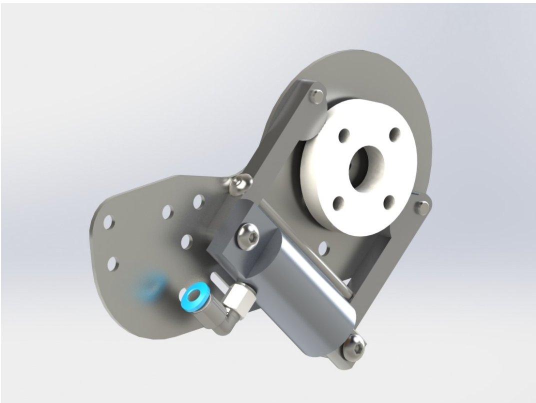
Thickness	**4 [mm]**
σ_{min}	51 [MPa]
Stainless Steel 304	58.3 [g]
Aluminum 6061	19.4 [g]
PMMA	8.6 [g]



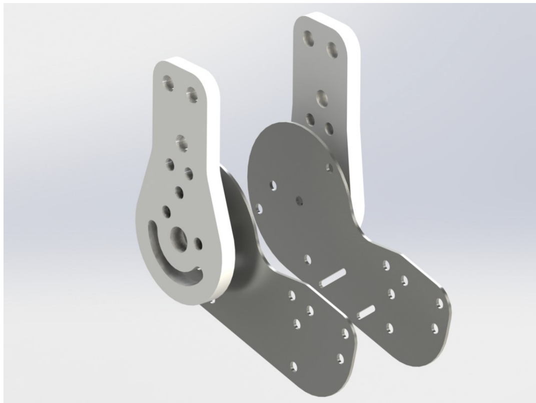
(a) A render of the embodied design of the implementation of the locking mechanism



(b) The chosen concept of the implementation of the locking mechanism brake disc



(c) A render of the embodied design of the implementation of the locking mechanism



(d) The chosen concept of the implementation of the locking mechanism brake disc

Figure J.4: The chosen concept for the locking mechanism

K

Control

Pneumatics

Conceptual Design

Pneumatic cylinders are the chosen method for actuation, with two double-acting cylinders to achieve flexion/extension and one single-acting push cylinder used to disengage the locking mechanism.

Gas Type

When using pneumatic actuation, the medium for energy storage and actuation has to be chosen. Instead of storing energy, a compressor could be used, but the low power density of such an approach would nullify the advantages of pneumatic actuation [8]. The required gas will therefore be stored in a pressurized cylinder. Two gasses will be considered to serve as this medium: carbon dioxide (CO_2) and hydrogen-peroxide (H_2O_2). Carbon-dioxide is considered as it is the gas used most for this purpose in history. Hydrogen-peroxide is considered because it has been implemented and proven useful in the pneumatic elbow prosthesis of the Vanderbilt University [8]. This second gas is considered a "monopropellant": "A monopropellant is a substance that reacts or decomposes exothermically when in contact with a catalyst. Since these reactions break molecular bonds, they can provide a significantly greater energy density than a phase change which does not alter the molecular structure of the substance." [8]. A 70% concentration of H_2O_2 can have an energy density which is almost four times higher than that of CO_2: 210 [kJ/kg] and 56 [kJ/kg] respectively. Higher concentrations of hydrogen-peroxide can achieve significantly higher energy densities, at the cost of reduced safety, e.g. higher exhaust temperatures.

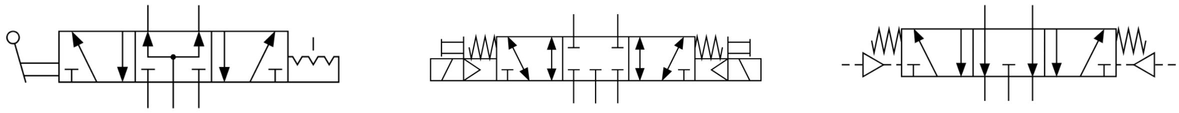
Despite the great promise shown by the use of monopropellants in the prototype by the Vanderbilt university, carbon-dioxide will be used as the medium for two reasons: (1) this prototype will mainly focus on the mechanical viability of a pneumatic prosthesis, thus energy storage plays a significantly smaller role compared to commercial products. (2) The use of monopropellants requires specialised knowledge of chemistry and pneumatic control, for both of which there is no time during this design cycle.

Gas Regulation

Valves Solenoid valves are classified as X/Y, where X is the amount of connections and Y is the amount of states. The single-acting LM actuator only two states are desired: actuated and non-actuated. This cylinder requires three connections: (1) input, (2) output, and (3) exhaust. The required valve is therefore a 2/3 valve. These valves can be either normally-open, normally-closed or bi-stable. A bi-stable valve will stay in its position until further instruction. Normally-open and normally-closed valves have a home position to which they return if there are no instructions. The majority of the time, the locking mechanism should be engaged and the actuator should therefore non-actuated. This means that the valve should be 2/3 normally-closed.

For the double-acting cylinders, three states are desired: No motion, flexion, and extension. Five connectors are needed for this cylinder (1) input, (2) FE output, (3) FE exhaust, (4) PS output, and

(5) PS exhaust. For these two cylinders, a 5/3 valve will be used. These valves can be mid-position-pressurized, mid-position-exhausted or mid-position-closed, as shown in Figure K.1. A mid-position-pressurized valve supplies both chambers of the actuator with pressurized gas whereas mid-position-closed valve has residual pressure in on chamber of the cylinder. This is useful for holding a position, but this prevents the free-wing mode. The holding of a position is achieved using the locking mechanism, thus the 5/3 mid-position-exhausted type is chosen to be the valve for the double-acting cylinders.



(a) 5/3 mid-position-pressurized (b) 5/3 mid-position-closed (c) 5/3 mid-position-exhausted

Figure K.1: The different versions of a 5/3 valve

Control of the prosthesis can be done (1) mechanically, by i.e. opening valves using a harnass, (2) manually, with the hands of the operator. or (3) electronic/solenoid valves, using an Arduino. The goal of the designed prosthesis is to show that pneumatic actuation is a viable option. Manual control performed by the operator eliminates any insecurities about electronic signalling and data processing, meaning that only the hardware of the prosthesis is tested. As a first step in the design process, testing one aspect of the device at a time is advisable to determine future improvements. Therefore, manual valves will be the method for this prosthesis. These manual valves are significantly heavier than solenoid valves, thus in the future, solenoids valves are highly recommended.

Pressure Regulation The LM cylinder is designed to disengage the lock when the pressure is 12 bar. However, to control the force exerted by the other cylinders, the pressure should be controlled. This can be achieved by using pressure regulators or using ”bang-bang control”. Bang-bang control of pressure is achieved in valves with only a discrete open or closed state. By regulating the time the valve is opened, the pressure is regulated. Pressure regulators are a less crude method, but cause the prosthesis to become heavier and bulkier, with the lightest versions found on https://www.festo.com/ weighing over 150 grams. Therefore, bang-bang control will be used to regulate the pressure.

Circuits

The final step in the conceptualization of the pneumatic circuit, is to design the circuit itself. The LM actuator only has to be actuated whenever the FE actuator is actuated in either flexion or extension. Therefore, with proper design, only two valves are needed, one to control PS and one to control both FE and LM. This is illustrated in Figure K.3. A different option is to use a valve to control the actuator of the locking mechanism, as shown in Figure K.2. The circuit with three valves might weigh more, due to the added mass of the valve. However, there are two reasons why this circuit will be chosen to work with. First of all, the complexity of the 2-valve circuit is significantly higher than the 3-valve circuit. The additional mass of the components required to achieve a 2-valve circuit, partly, negates the mass of the third valve. Furthermore, the 2-valve circuit has a delay in supplying the LM actuator with gas because of the significantly longer path the gas has to take. Flexion/extension can not occur whilst the locking mechanism is not disengaged. By having independent control of the locking mechanism, the response time of the prosthesis will be significantly reduced.

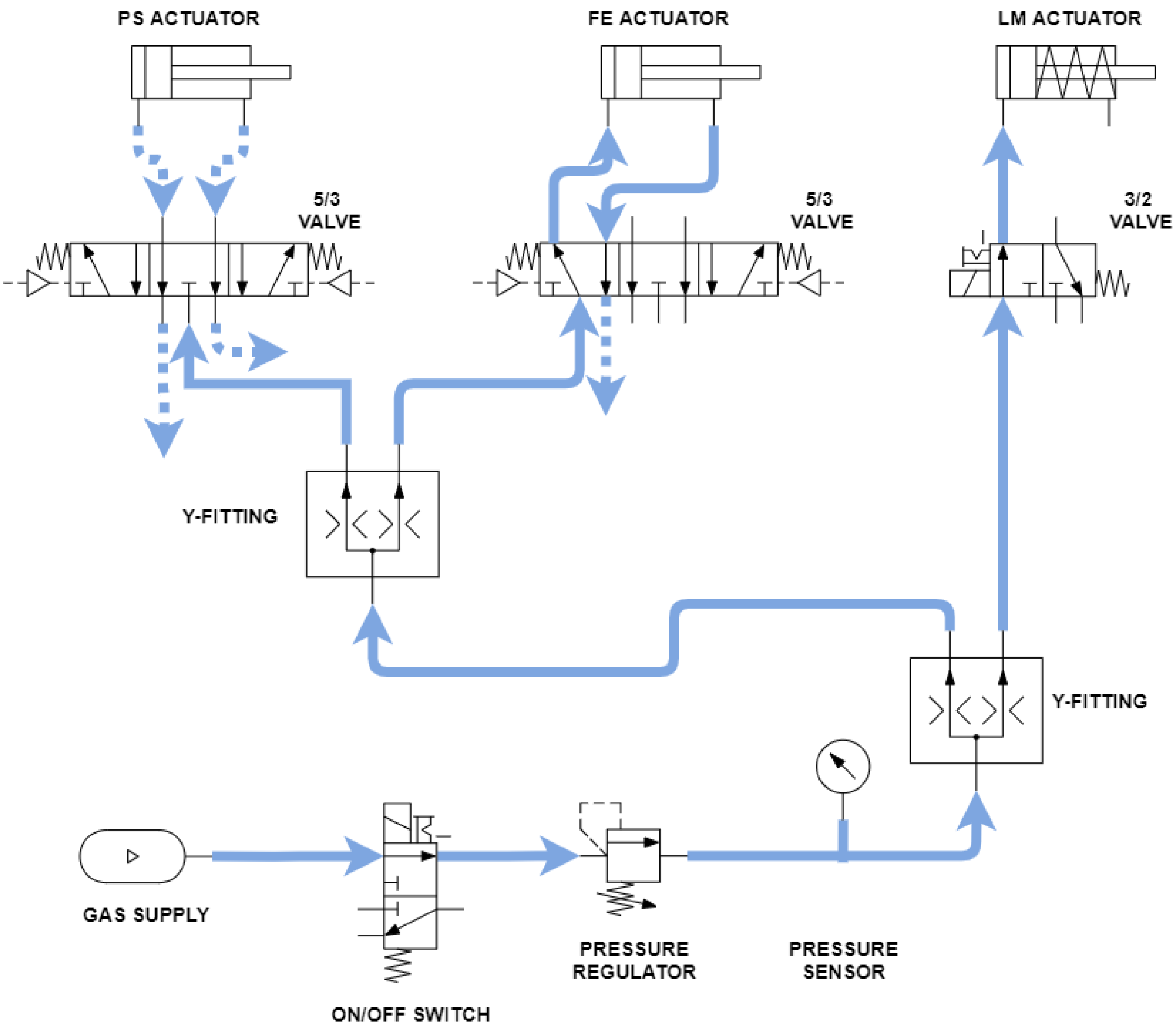


Figure K.2: The pneumatic circuit with three valves

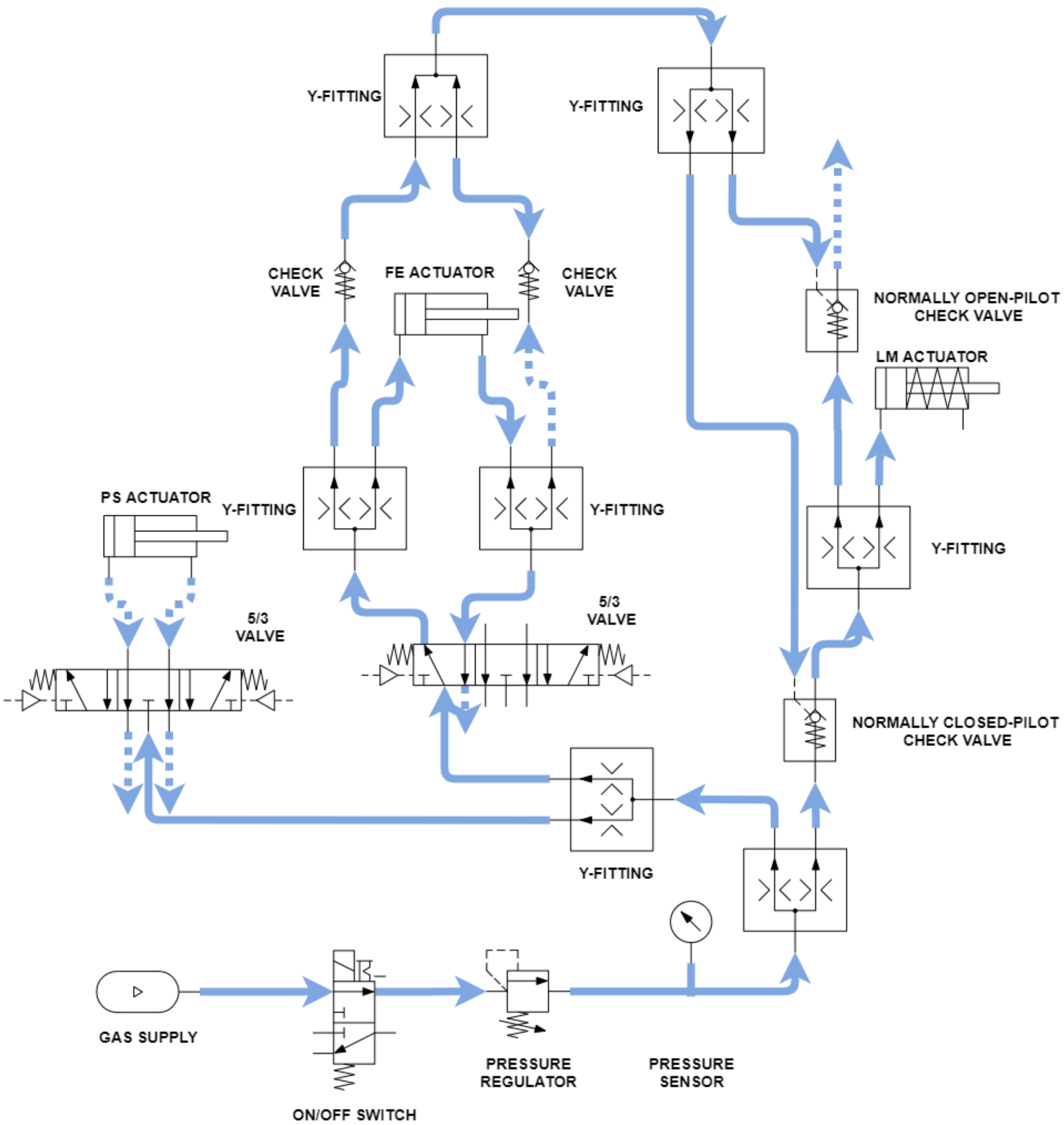


Figure K.3: The pneumatic circuit with two valves

Embodiment

Valves

Flow Rate Since the details of the actuators are known, the flow rate required of the valves can be determined using Equation (K.1) and Equation (K.2). The results are shown in Table K.1.

$$Q = 60 * \frac{V}{t} \tag{K.1}$$

K.1: The equation to calculate the required flow rate of the valves. Q is the flow rate [L/min], V is the displaced volume [L] , t is the stroke time [s]

$$V = \frac{\pi * d^2 * L}{4} \tag{K.2}$$

K.2: The equation to calculate the displaced volume. V is the stroke volume [L], d is the inner diameter [dm], L is the stroke length [dm]

Table K.1: An overview of the calculations for the required flow rates

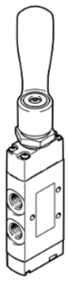
				Goal		Requirement	
Motion	**d** [mm]	**L** [mm]	**V** [mL]	**t** [s]	**Q** [L/min]	**t** [s]	**Q** [L/min]
FE	40	25	31	0.7	0.85	2.7	1.3
PS	32	12.5	10	0.5	0.38	1.2	0.6

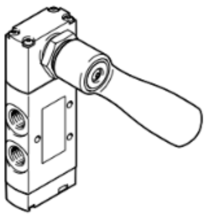
Valve Selection The product finder from `https://www.festo.com/` was used to find the valves which led to the ”VHEF” series. The remaining choices concern the type of lever to use for actuation, the size of the connectors, whether they have an internal or external air pilot supply, and the maximum pressure. The size of the connections are the smallest possible to fit the desired small tubes. An external supply for the pilot air is useful when operating with a vacuum, which is not the case. An internal supply is simpler and will therefore be chosen. The maximum allowable pressure for all of these valves ranges from 8-10 bar. After a discussion with professor Plettenburg and Jan van Frankenhuyzen, it was decided that the 10 bar versions could still be used with a 12 bar pressure. The 3/2 valve will differ slightly from the concept, instead on a mono-stable normally-closed, valve a bi-stable valve will be used. This is done to still be able to test the free-swing mode even if the secondary actuation method of the locking mechanism does not work.

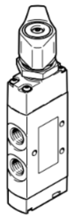
The available options are shown in Table K.2 and visualised in Figure K.4. To show the improvements a solenoid valve could offer in the future concerning weight, an example is shown in the table as well. It can be noted that the flow-rate for both PS and FE is easily reached with these valves. For both type of valves, the lightest option is chosen. This results in VHEF-LT-M32-M-G18 and VHEF-ES-P53E-M-G18.

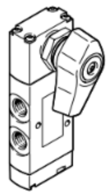
Table K.2: An overview of the available valves found on https://www.festo.com/cms/nl_nl/index.htm

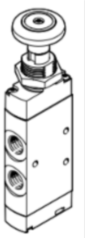
Type	Control	Weight [g]	Flow Rate [L/min]	Code
5/3	Hand Lever, Sideways	265	530	VHEF-HS-P53E-M-G18
5/3	Hand Lever	265	530	VHEF-H-P53E-M-G18
5/3	Selector, Sideways	235	530	**VHEF-ES-P53E-M-G18**
5/3	Selector	235	530	VHEF-E-P53E-M-G18
3/2	Bushbutton	168	750	VHEF-PTC-B32-G18
3/2	Toggle Lever	174	750	VHEF-VT-B32-G18
3/2	Finger Lever	156	750	**VHEF-LT-M32-M-G18**
3/2	Hand Lever, Sideways	236	750	VHEF-HST-B32-G18
3/2	Hand Lever	236	750	VHEF-HT-B32-G18
3/2	Selector, Sideways	206	750	VHEF-EST-B32-G18
3/2	Selector	206	750	VHEF-E-B32-G18

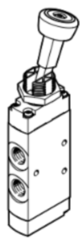

(a) Manual valve: Hand Lever

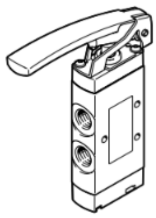

(b) Manual valve: Hand Lever Sideways


(c) Manual valve: Selector Valve


(d) Manual valve: Selector Valve Sideways


(e) Manual valve: Pushbutton


(f) Manual valve: Toggle Lever


(g) Manual valve: Finger Lever

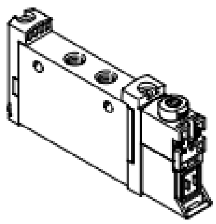

(h) Electric valve: VUVG

Figure K.4: The different options of valves

L

Part List

Part Code Legend

General		Specific	
OS	Off-the-Shelf	**FE**	Flexion/Extension
		PS	Pronation/Supination
		LM	Locking Mechanism
		FR	Frame
OS-1 OS-2 OS-3 OS-4	Pneumatic Components Mechanical Components Electronic Components Bolts, Nuts, Washers, etc.	XX-1	Custom Part
OS-X-1 OS-X-2 OS-X-3 OS-X-4 OS-X-5	General Use Used in FE Used in PS Used in LM Used in FR	XX-1-X XX-1-X	Custom Metal Part Custom Non-Metal Part
OS-4-1 OS-4-2 OS-4-3 OS-4-4	Bolts Nuts Washers Inserts		
OS-4-x-MxL	Thread Size & Length		

Table L.1: An overview of all the parts used to build the prototype. Some parts were custom made or could be taken from the workshop at 3mE free of charge, these cells are colored grey

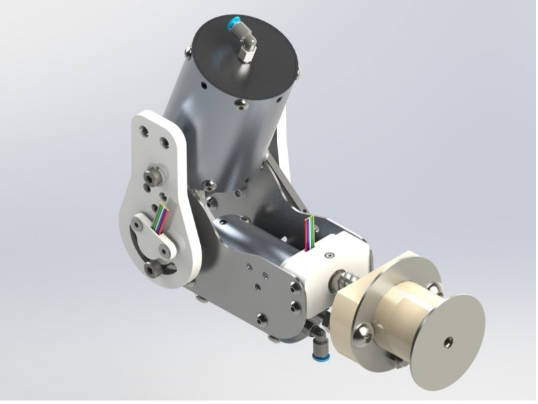
Name	Code	Amount	Expected Total Mass	Measured Total Mass	Cost
Full Prosthesis	**A**	**1**	**1250**		**€ 654,29**
Flexion/Extension	**FE**	**1**	**157,6**	**154**	**€ 44,69**
FE Housing	FE-1-1-01	1	19,3	19	€ -
FE Piston-Rod	FE-1-1-02	1	17,7	16	€ -
FE Piston-Rod-Nut	FE-1-1-03	1	1,1	0	€ -
FE Cap	FE-1-1-04	1	30,5	31	€ -
FE End-Cap	FE-1-1-05	1	21,8	22	€ -
Push-in Fitting	OS-1-1-01	2	6,0	6	€ 9,12
FE Piston	OS-1-2-01	1	35,1	35	€ 12,02
FE Rod-Seal	OS-1-2-02	1	0,1	0	€ 6,77
FE O-Ring	OS-1-2-03	2	1,0	0	€ 1,24
FE Rod-Bushing	OS-2-2-01	1	1,0	0	€ 0,65
FE Rod-End	OS-2-2-02	1	13,0	14	€ 5,80
M3 Round Head	OS-4-1-M3x6-1	8	4,0	0	€ 0,88
M8 Nut	OS-4-2-8	1	5,0	4	€ -
M8 Washer	OS-4-2-8	1	2,0	0	€ -
Locking Mechanims	**LM**	**1**	**16,8**	**22**	**€ 11,18**
LM Housing	LM-1-1-01	1	4,1	7	€ -
LM Piston-Rod	LM-1-1-02	1	8,1	8	€ -
LM Spring-Washer	LM-1-1-03	1	0,5	0	€ -
LM Piston	OS-1-4-01	1	1,2	0	€ 8,80
LM Rod-Bushing	OS-2-4-01	1	3,0	0	€ 2,38
Push-in Fitting	OS-1-1-01	1	3,0	3	€ 4,56
LM Spring	OS-2-4-03	1	1,5	0	€ 2,75
M3 Round Head	OS-4-1-M3x6-1	1	0,5	0	€ 0,11
M3 Washer	OS-4-3-M3	1	0,1	0	€ -

Name	Code	Amount	Expected Total Mass	Measured Total Mass	Cost
Pronation/Supination	**PS**	**1**	**243,5**	**220**	**€ 107,84**
PS Housing	PS-1-1-01	1	12,5	19	€ -
PS Piston-Rod	PS-1-1-02	1	17,7	17	€ -
PS Piston-Rod-Nut	PS-1-1-03	1	0,7	0	€ -
PS Cap	PS-1-1-04	1	18,9	20	€ -
PS End-Cap	PS-1-1-05	1	15,1	15	€ -
PS Nut-Connect	PS-1-1-06	1	8,7	9	€ -
PS Magnet-Nut	PS-1-1-07	1	1,6	0	€ -
PS Encoder-Base	PS-1-1-08	1	3,4	4	€ -
PS Lead-screw	PS-1-1-09	1	58,3	38	€ 34,73
PS Lead-Hub	PS-1-2-01	1	6,5	5	€ -
PS Encoder-Top	PS-1-2-02	1	1,2	0	€ -
Push-in Fitting	OS-1-1-01	2	6,0	6	€ 9,12
PS Piston	OS-1-3-01	1	20,0	19	€ 10,84
PS Rod-Seal	OS-1-3-02	1	0,1	0	€ 6,04
PS O-Ring	OS-1-3-03	2	0,8	0	€ 1,04
PS Rod-Bushing	OS-2-3-01	1	1,0	0	€ 0,58
PS Lead-Bushing	OS-2-3-02	1	1,0	0	€ 1,15
PS Lead-Axial-Bushing	OS-2-3-03	1	1,0	0	€ 2,60
PS Lead-Nut	OS-2-3-04	1	39,2	45	€ 38,05
M3x6 Round Head	OS-4-1-M3x6-1	12	6,0	7	€ 1,32
M3x8 Countersunk	OS-4-1-M3x8-2	12	6,0	4	€ 1,32
M6x16 Round Head	OS-4-1-M6x16-1	2	1,2	3	€ -
M6 Nut	OS-4-2-M6	5	12,5	12.5	€ -
M6 Washer	OS-4-3-M6	1	1,0	0	€ -
M3 Insert	OS-4-4-M3x4.5	9	2,7	0	€ 0,81
M3 Set Screw	OS-4-5-M3x5	1	0,3	0	€ 0,24
Electronics	**EL**	x	**4,8**	**x**	**€ 171,00**
RM08 Encoder	EL-3-1-01	2	4,0	x	€ 166,00
RM08 Magnet	EL-3-1-02	2	0,8	x	€ 5,00

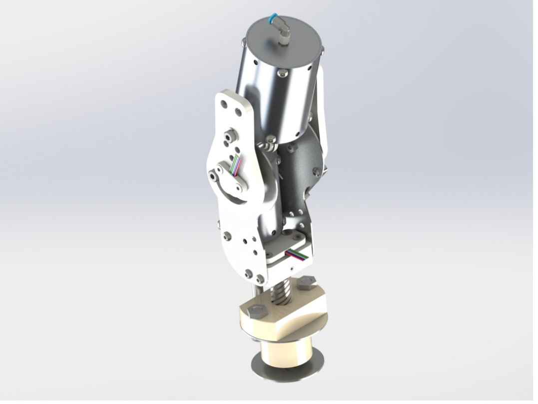
Name	Code	Amount	Expected Total Mass	Measured Total Mass	Cost
Frame	**FR**	**1**	**112,6**		**€ 16,64**
FR Forearm-Endstop	FR-1-1-01	1	25,5		€ -
FR Forearm-Lock	FR-1-1-02	1	25,2		€ -
FR Connection-Rod	FR-1-1-03	1	4,6	5	€ -
FR FE-Lever	FR-1-1-04	2	3,6		€ -
FR Encoder-Shaft	FR-1-1-05	1	1,0		€ -
FR Encoder-Holder-Base	FR-1-1-06	1	1,4		€ -
FR Endplate	FR-1-1-07	1	9,7		€ -
FR Upperarm-Encoder	FR-1-2-01	1	11,5		€ -
FR Forearm-Lock	FR-1-2-02	1	9,5		€ -
FR Brake-Disc	FR-1-2-03	1	3,4		€ -
FR Encoder-Holder-Top	FR-1-2-04	1	0,5	0	€ -
FR Upperarm-Cover	FR-1-2-05	1	19		€ -
FR Forearm-Cover	FR-1-2-06	1	43		€ -
FR Cover-Cowter	FR-1-2-07	2	16		€ -
FR Shaft Bushing	OS-2-5-01	1	1,0	0	€ 0,58
FR FE-Housing Bushing	OS-2-5-02	3	3,0	0	€ 2,40
M3x8 Countersunk	OS-4-1-M3x8-2	6	3,0	2	€ 0,66
M3x8 Shoulderbolt	OS-4-1-M3x8-3	4	7,2	8	€ 11,68
M3x8 Low Head Bolt	OS-4-1-M3x8-4	2	1,3	0	€ 0,96
M3 Insert	OS-4-4-M3x4.5	4	1,2	0	€ 0,36
M3 Nut	OS-4-2-M3	2	5,0	4	€ -
M6 Nut	OS-4-2-M6	2	5,0	5	€ -
Pneumatics	**PN**	**1**	**718,3**	**705**	**€ 311,15**
Push-in Fitting G-1/8	PN-1-02	8	72,8	73	€ 14,80
Push-in Fitting Y	PN-1-03	2	5,0	7	€ 8,10
FE-PS Valve	PN-2-01	2	470,0	449	€ 190,48
LM Valve	PN-2-02	1	159,0	115	€ 73,67
Silencer	PN-3-01	5	11,5	15	€ 24,10

M
Renders

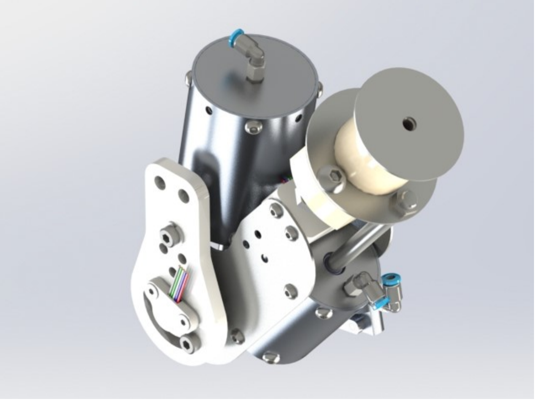
Assembly



(a) Isometric view of prosthesis: 90 degrees flexion

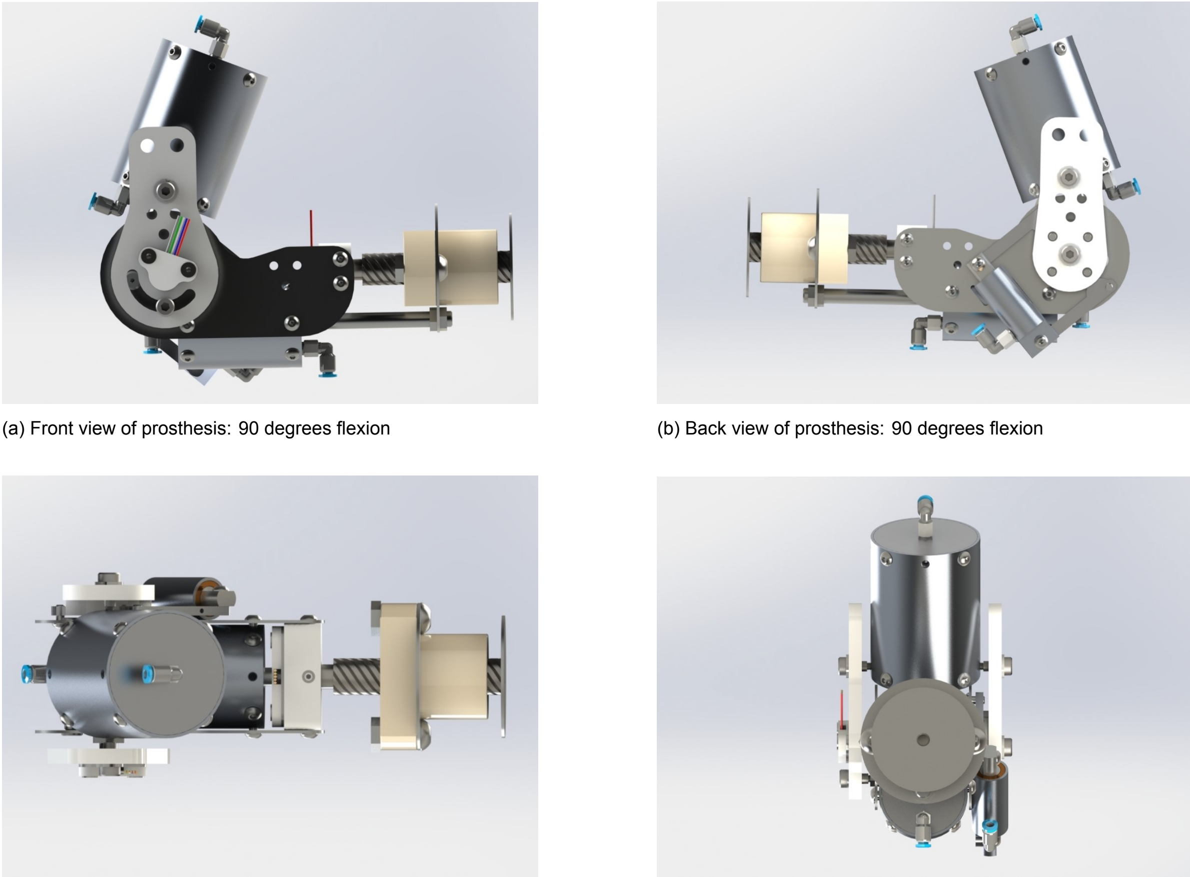


(b) Isometric view of prosthesis: 5 degrees flexion



(c) Isometric view of prosthesis: 140 degrees flexion

Figure M.1: Renders of the assembly: Isometric view



(a) Front view of prosthesis: 90 degrees flexion

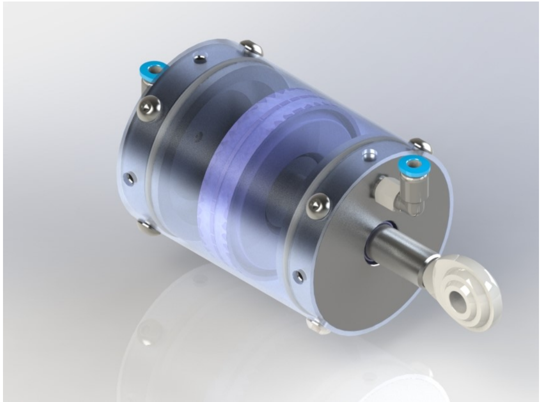
(b) Back view of prosthesis: 90 degrees flexion

(c) Top view of prosthesis: 90 degrees flexion

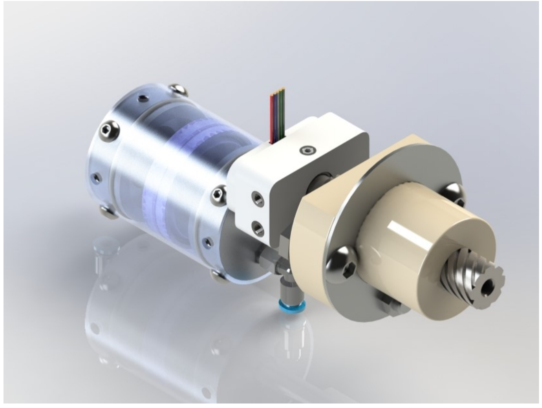
(d) Right view of prosthesis: 90 degrees flexion

Figure M.2: Renders of the assembly: Different views

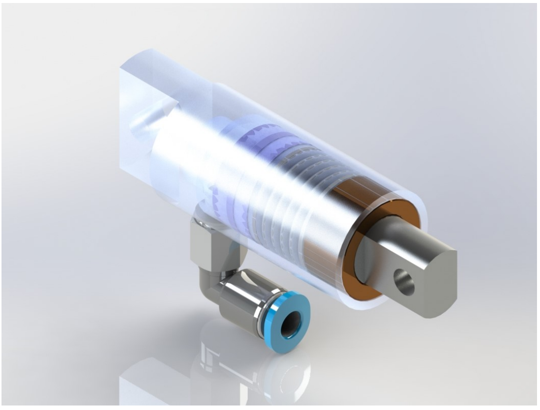
Actuators



(a) FE-0-0-00: FE Actuator



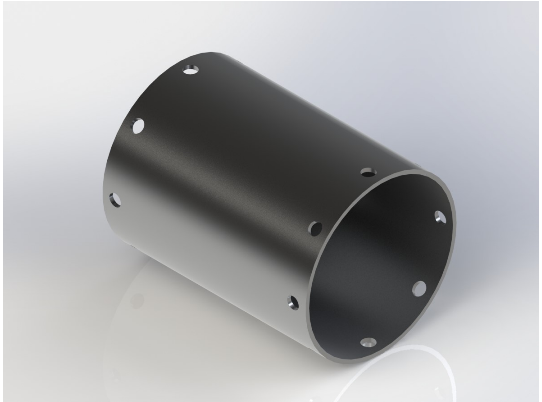
(b) PS-0-0-00: PS Actuator



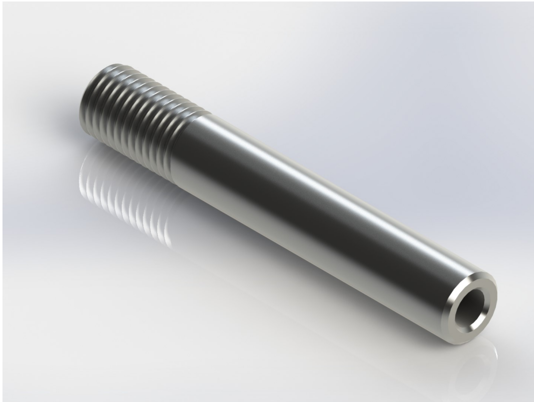
(c) LM-0-0-00: LM Actuator

Figure M.3: Renders of the actuators

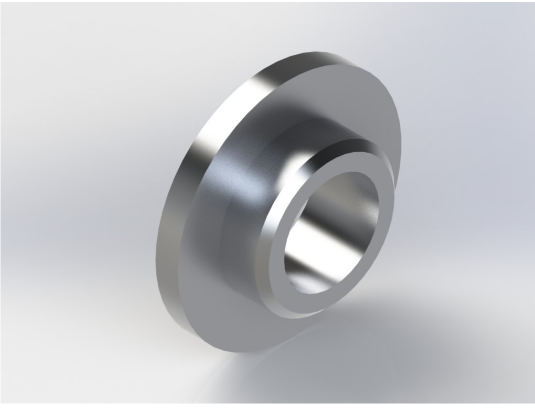
Flexion/Extension Actuator



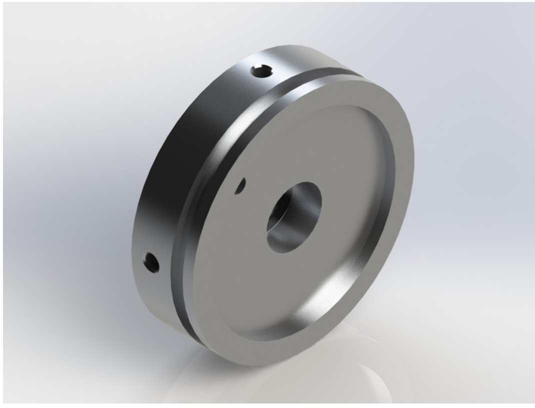
(a) FE-1-1-01: FE Housing



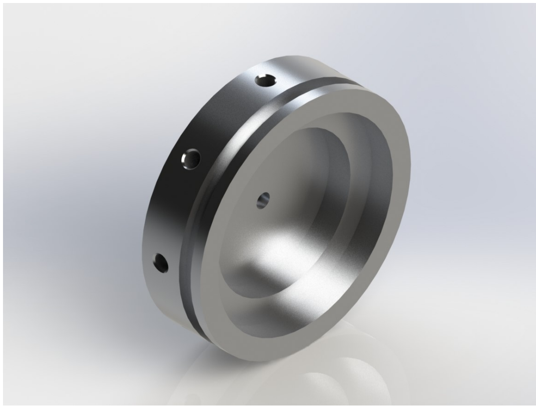
(b) FE-1-1-02: FE Piston-Rod



(c) FE-1-1-03: FE Piston-Rod-Nut



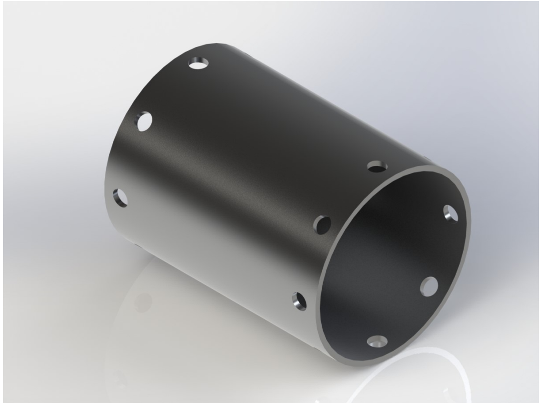
(d) FE-1-1-04: FE Cap



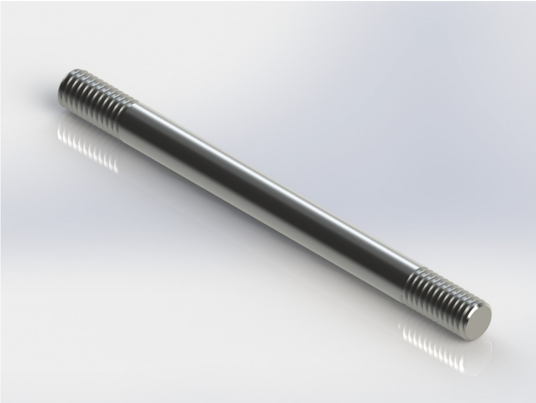
(e) FE-1-1-05: FE End-Cap

Figure M.4: Renders of the custom parts required for the FE actuator assembly

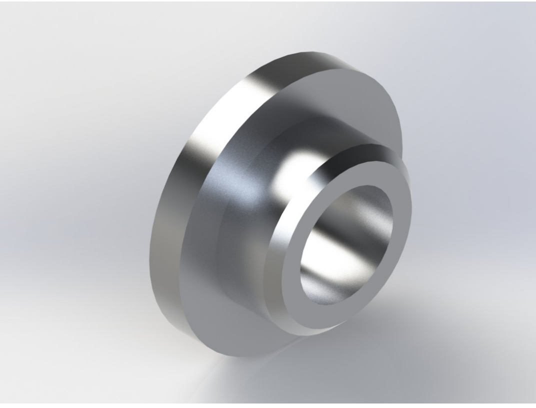
Pronation/Supination Actuator: Metal Custom Parts



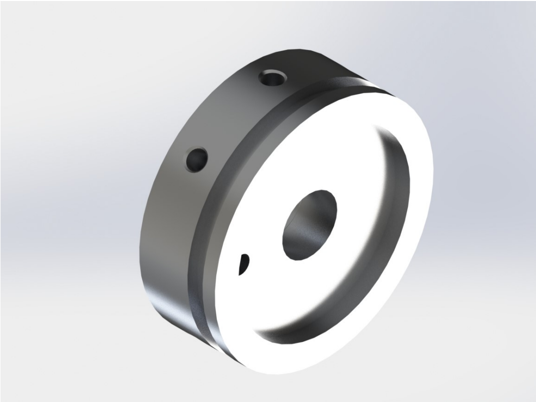
(a) PS-1-1-01: PS Housing



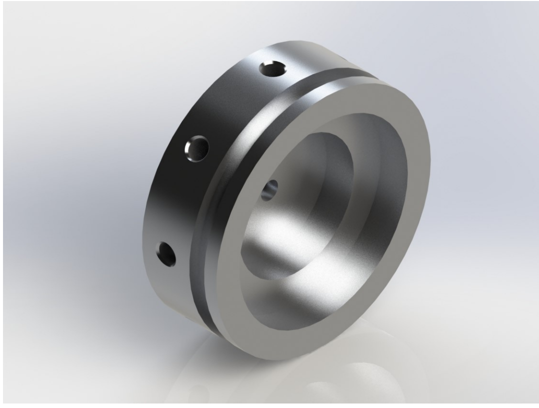
(b) PS-1-1-02: PS Piston-Rod



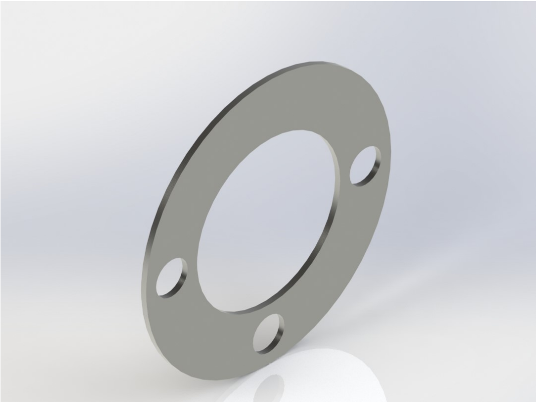
(c) PS-1-1-03: PS Piston-Rod-Nut



(d) PS-1-1-04: PS Cap

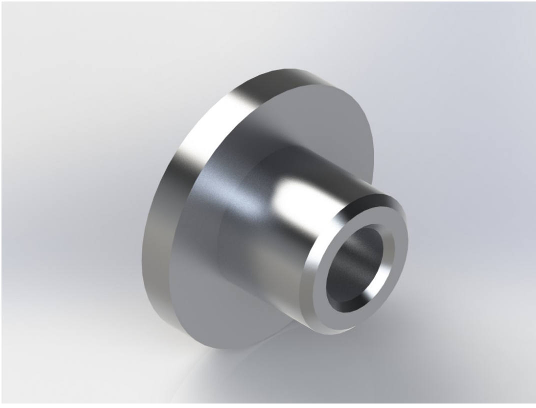


(e) PS-1-1-05: PS End-Cap

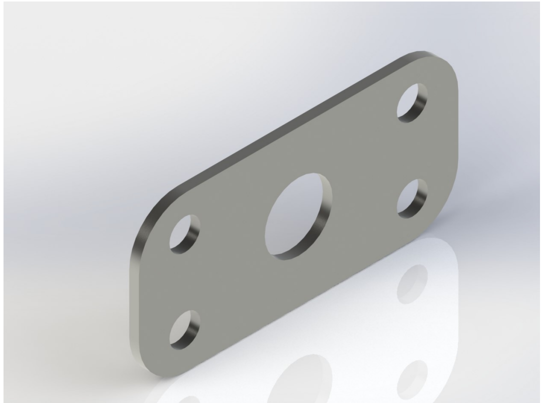


(f) PS-1-1-06: PS Nut-Connect

Figure M.5: Renders of the metal custom parts required for the PS actuator assembly



(a) PS-1-1-07: Magnet-Nut

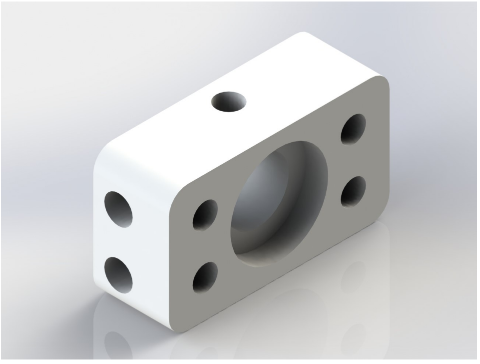


(b) PS-1-1-08: PS Encoder-Base

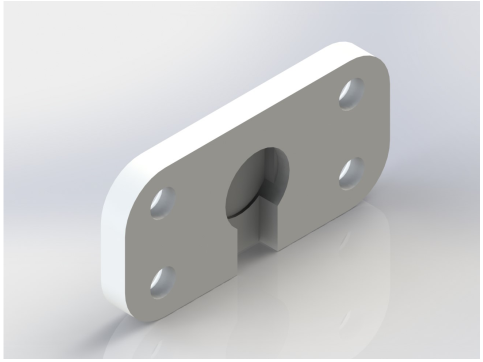
(c) PS-1-1-09: PS Lead-Screw

Figure M.6: Renders of the metal custom parts required for the PS actuator assembly: Continued

Pronation/Supination Actuator: Non-Metal Custom Parts



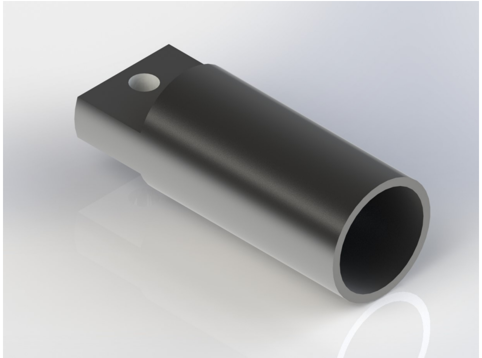
(a) PS-1-2-01: PS Lead-Hub



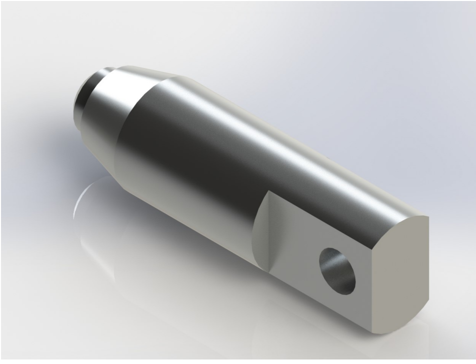
(b) PS-1-2-02: PS Encoder-Top

Figure M.7: Renders of the non-metal custom parts required for the PS actuator assembly

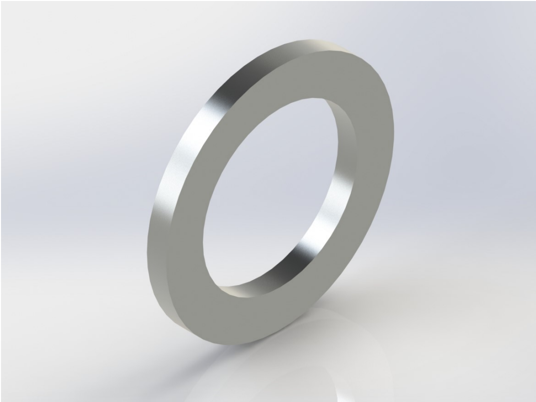
Locking Mechanism Actuator



(a) LM-1-1-01: LM Housing



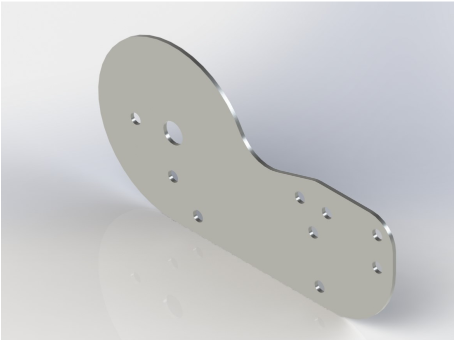
(b) LM-1-1-02: LM Piston-Rod



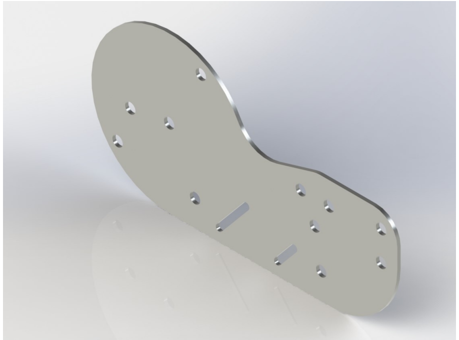
(c) LM-1-1-03: LM Spring-Washer

Figure M.8: Renders of the custom parts required for the LM actuator assembly

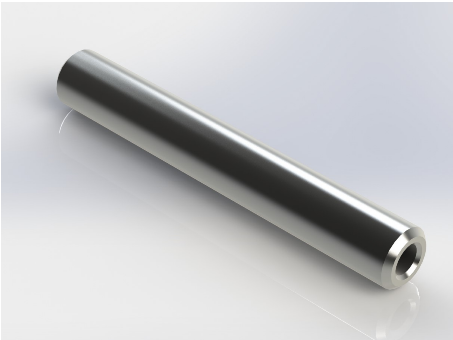
Frame: Metal Parts



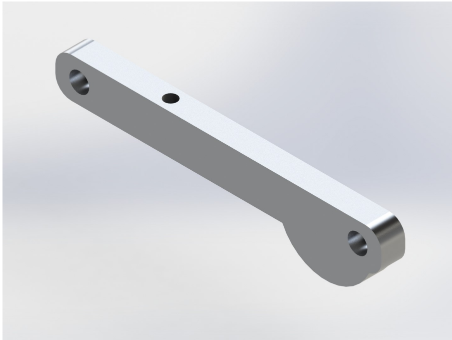
(a) FR-1-1-01: FR Forearm-Endstop



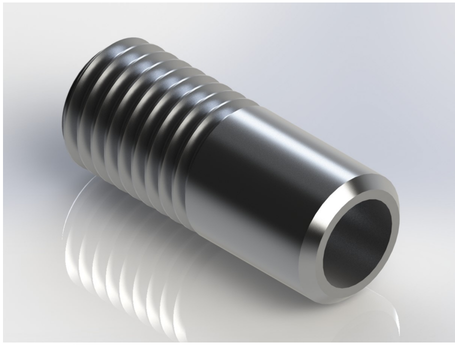
(b) FR-1-1-02: FR Forearm-Lock



(c) FR-1-1-03: FR FE-Connection-Rod

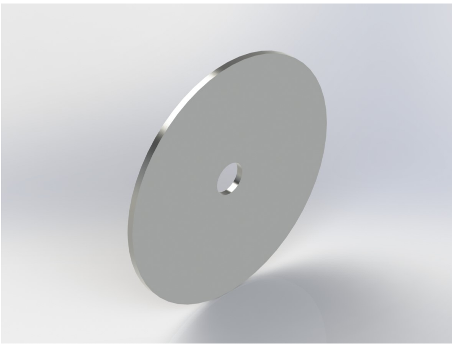


(d) FR-1-1-04: FR FE-Lever



(e) FR-1-1-05: FR Encoder-Shaft

(f) FR-1-1-06: FR Encoder-Holder-Base



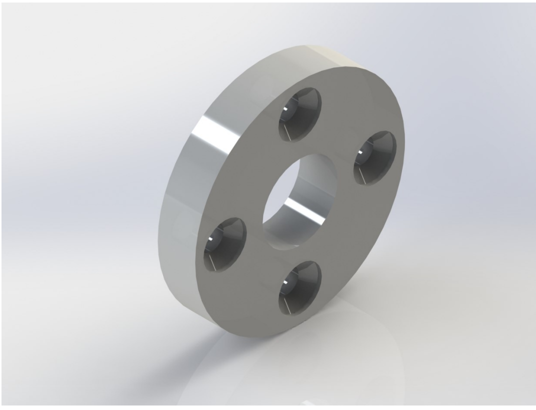
(g) FR-1-1-07: FR Endplate

Figure M.9: Renders of the metal custom parts required for the Frame

Frame: Non-metal Parts

(a) FR-1-2-01: FR Upperarm-Encoder

(b) FR-1-2-02: FR Upperarm-Lock



(c) FR-1-2-03: FR FE-Lock

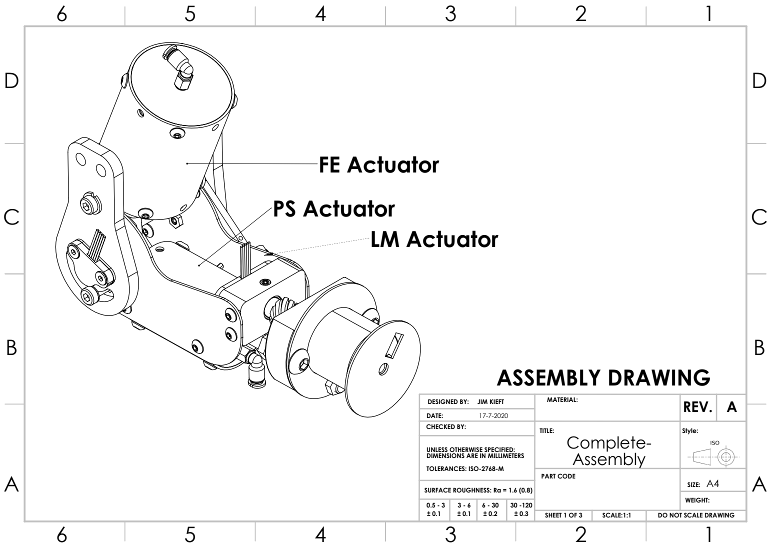
(d) FR-1-2-04: FR Encoder-Holder-Top

Figure M.10: The non-metal custom parts required for the Frame

N

Technical Drawings

Complete Assembly



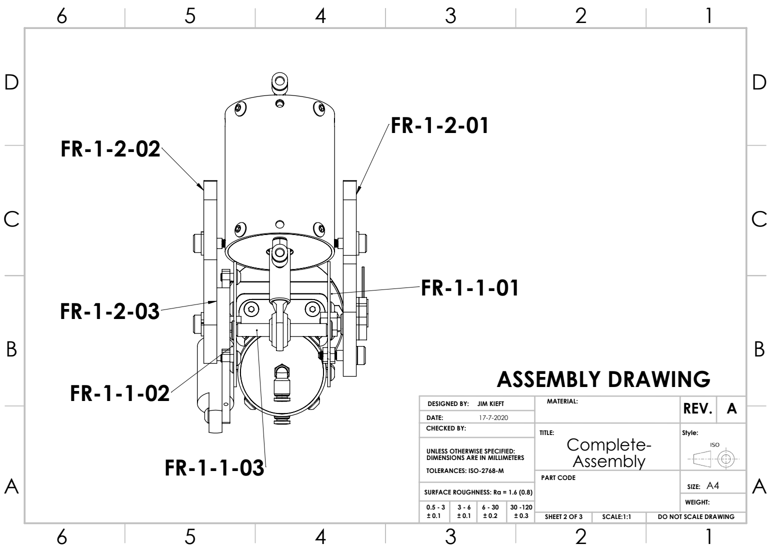


Figure N.1: The technical drawing of the complete assembly

FE-0-0-00: FE Actuator

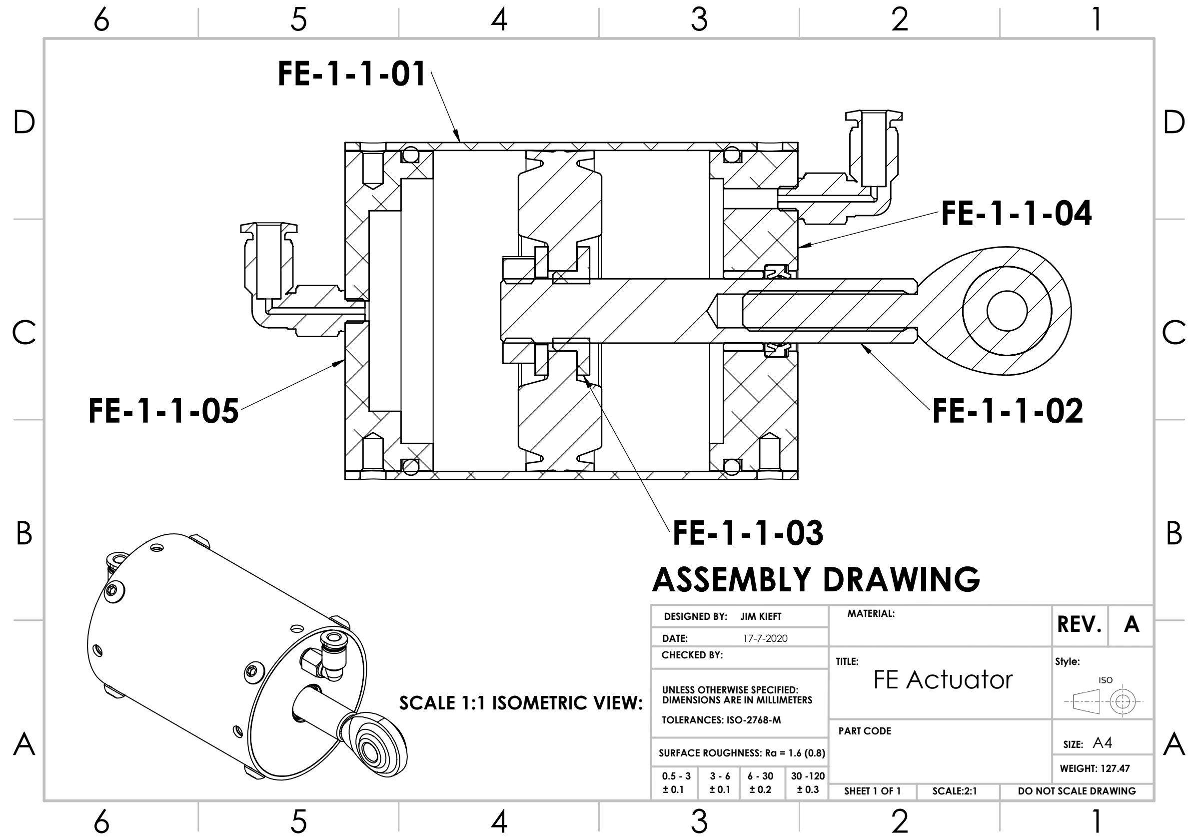


Figure N.2: The technical drawing of assembly: FE-0-0-00: Actuator

PS-0-0-00: PS Actuator

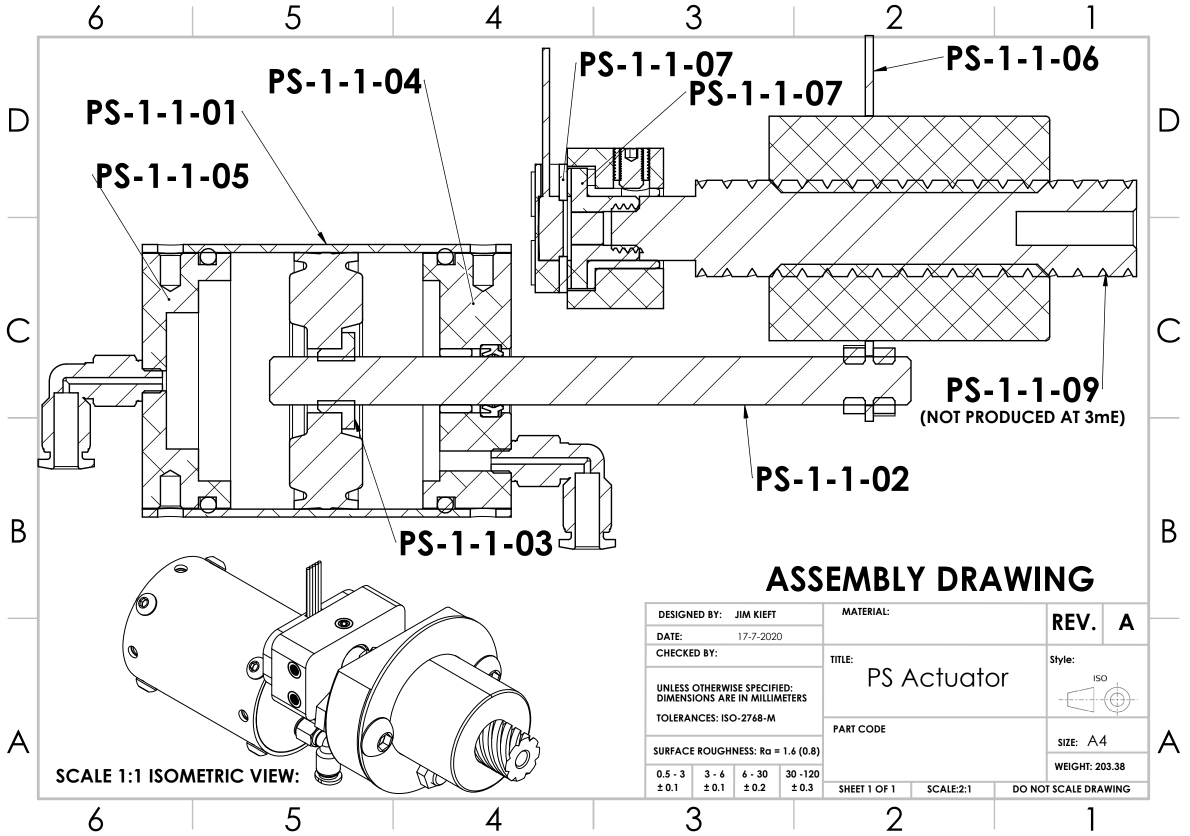


Figure N.3: The technical drawing of assembly: PS-0-0-00: PS Actuator

LM-0-0-00: LM Actuator

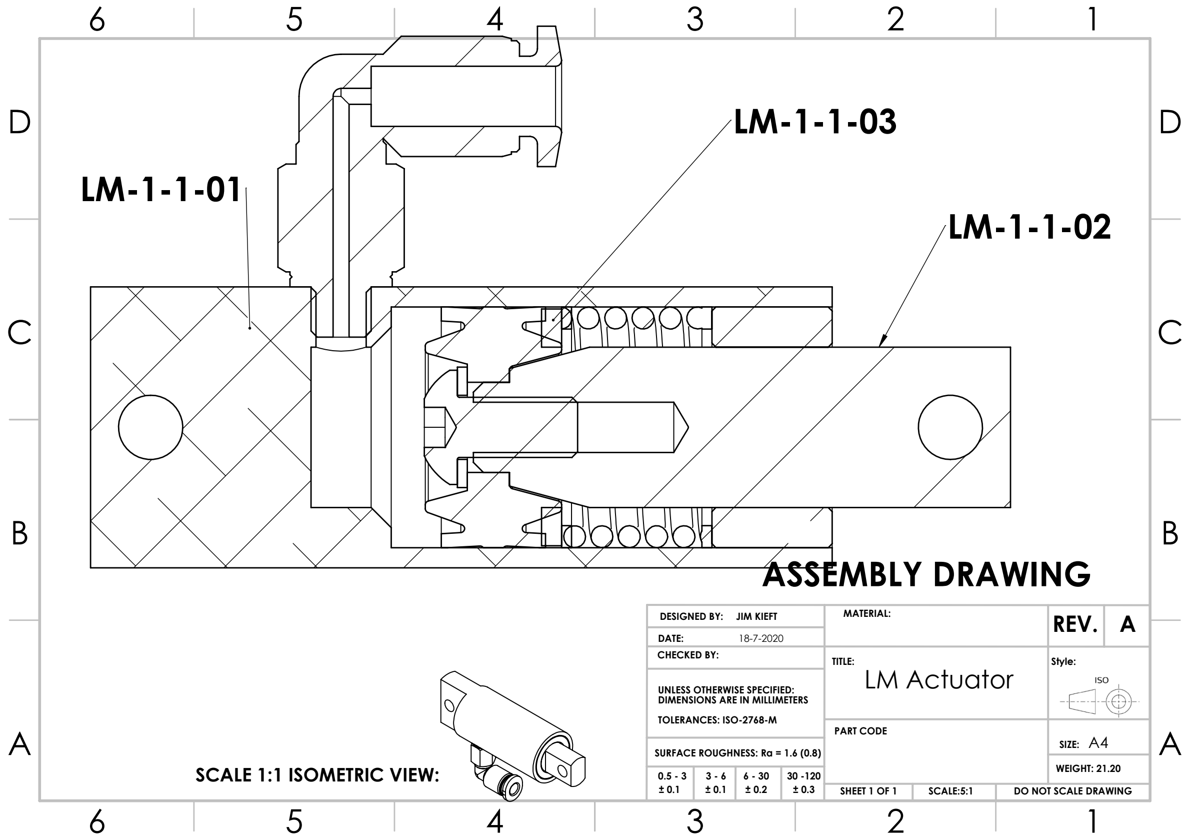


Figure N.4: The technical drawing of assembly: LM-0-0-00: LM Actuator

FE-1-1-01: FE Housing

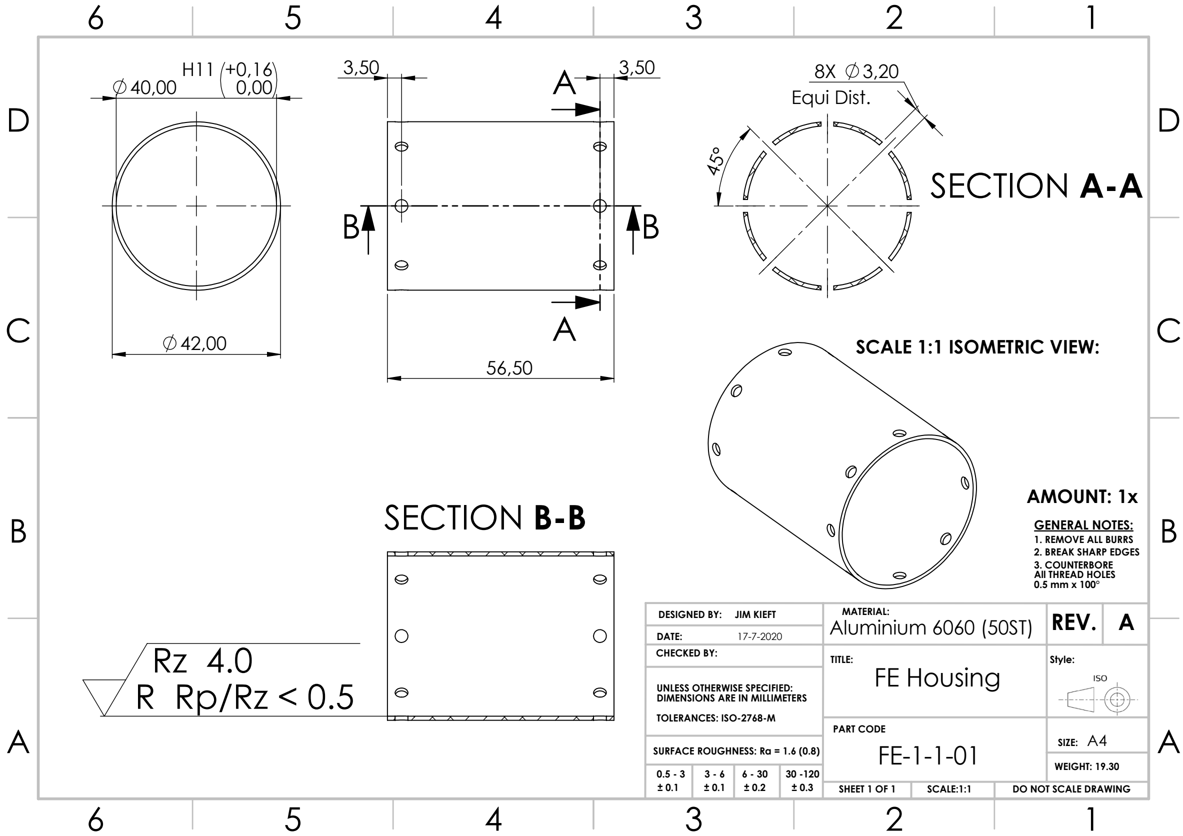


Figure N.5: The technical drawing of part: FE-1-1-01: FE Housing

FE-1-1-02: FE Piston-Rod

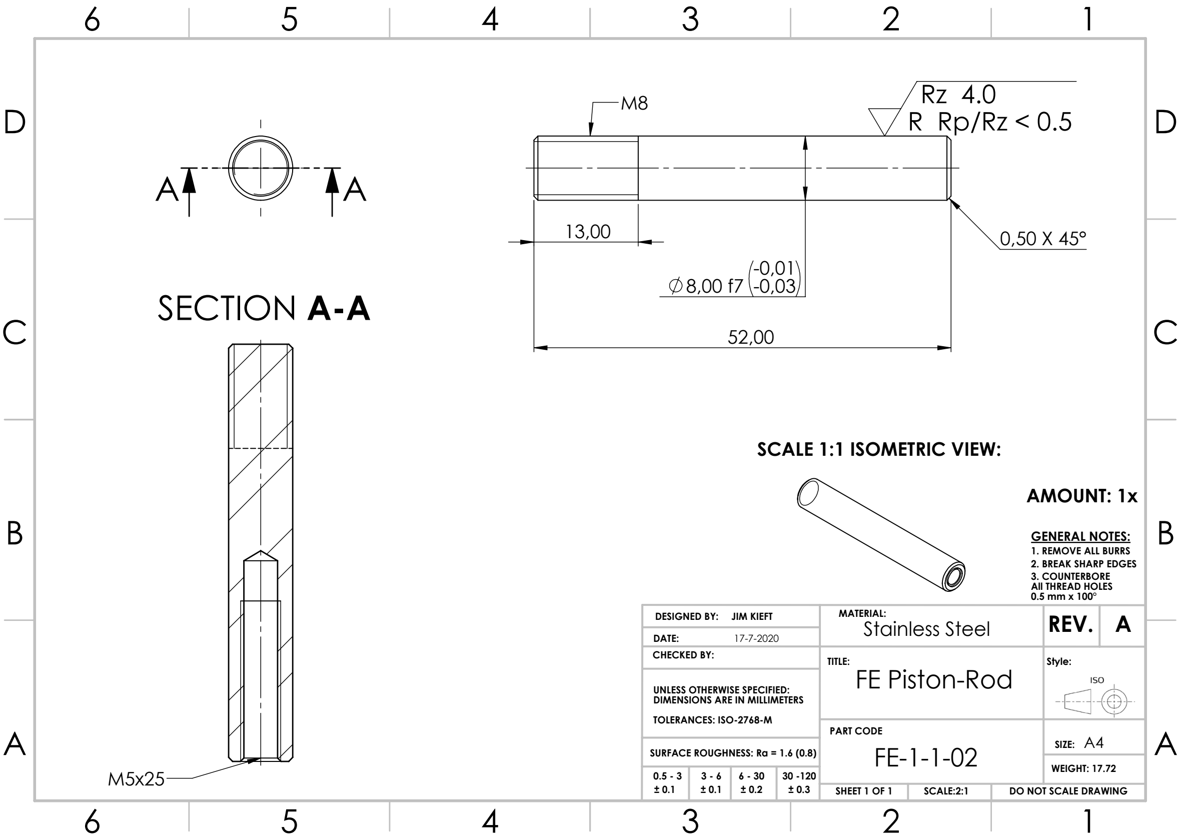


Figure N.6: The technical drawing of part: FE-1-1-02: FE Piston-Rod

FE-1-1-03: FE Piston-Rod-Nut

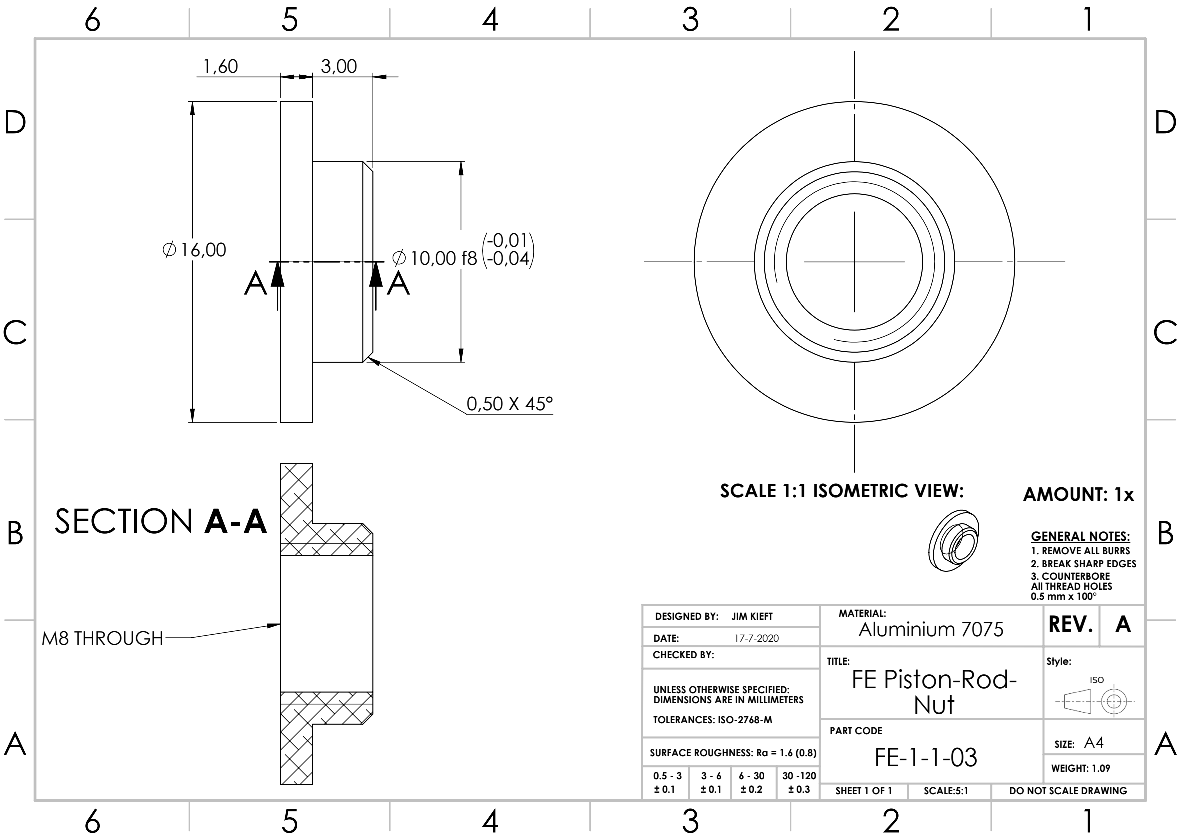
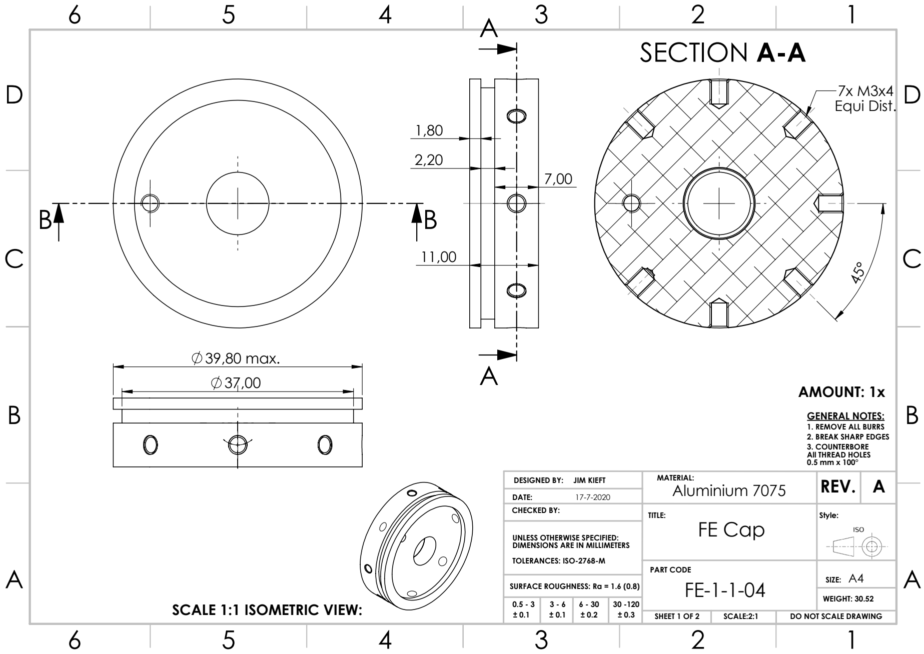


Figure N.7: The technical drawing of part: FE-1-1-03: FE Piston-Rod-Nut

FE-1-1-04: FE Cap



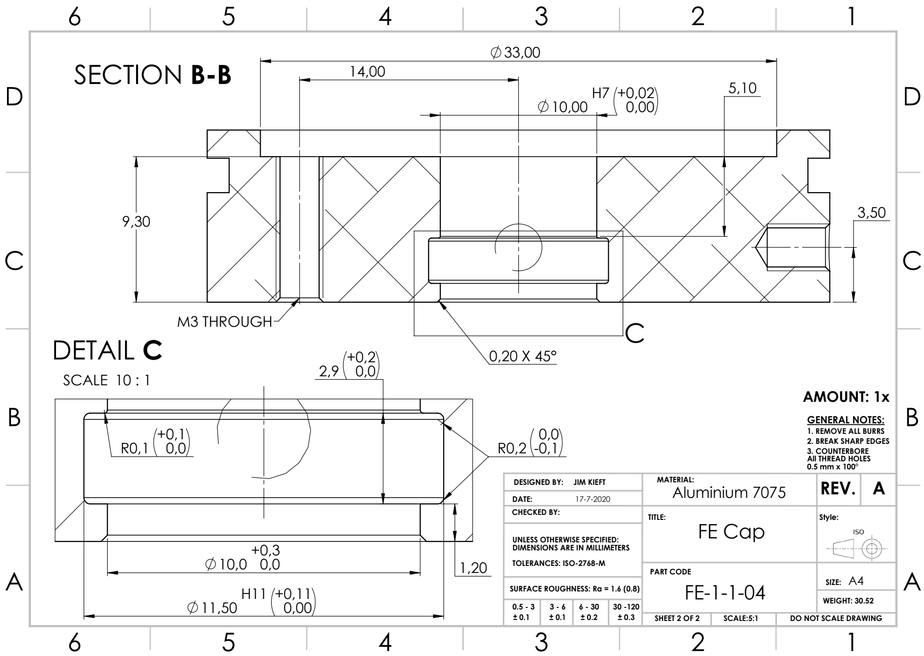


Figure N.8: The technical drawing of part: FE-1-1-04: FE Cap

FE-1-1-05: FE End-Cap

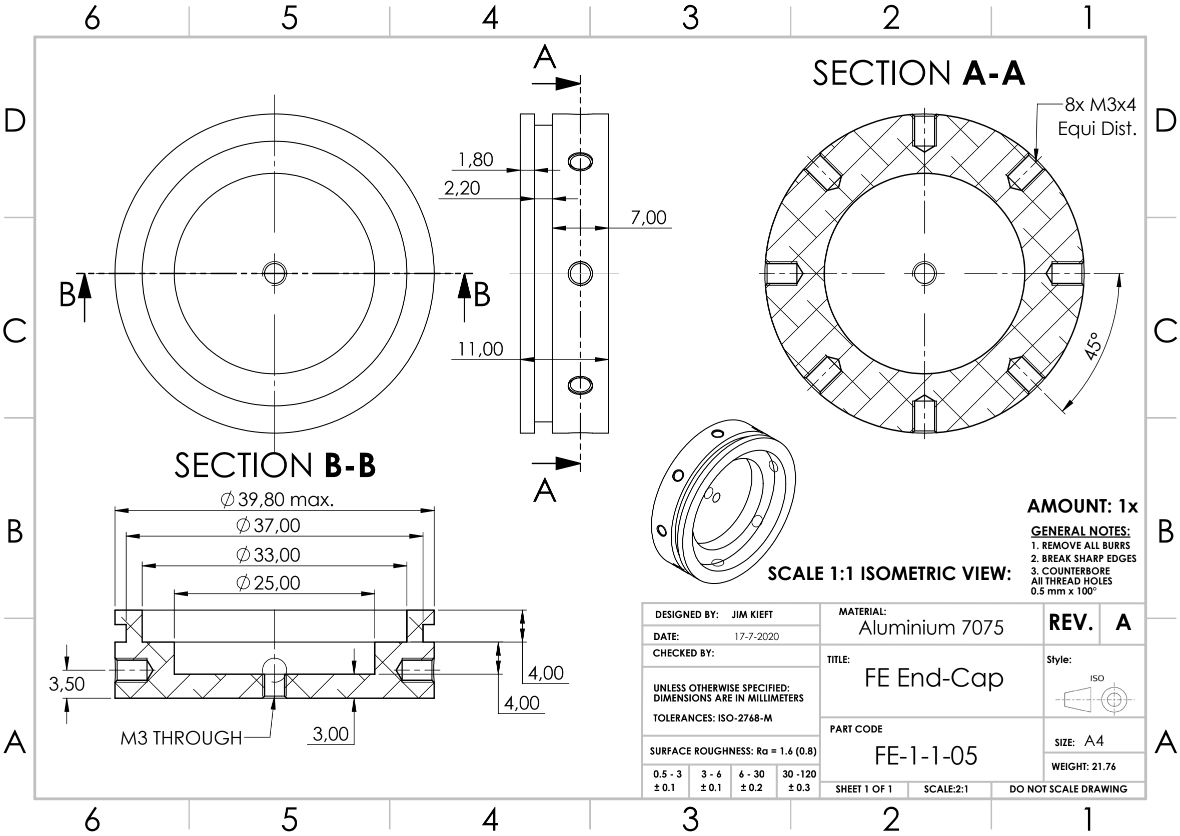


Figure N.9: The technical drawing of part: FE-1-1-05: FE End-Cap

PS-1-1-01: PS Housing

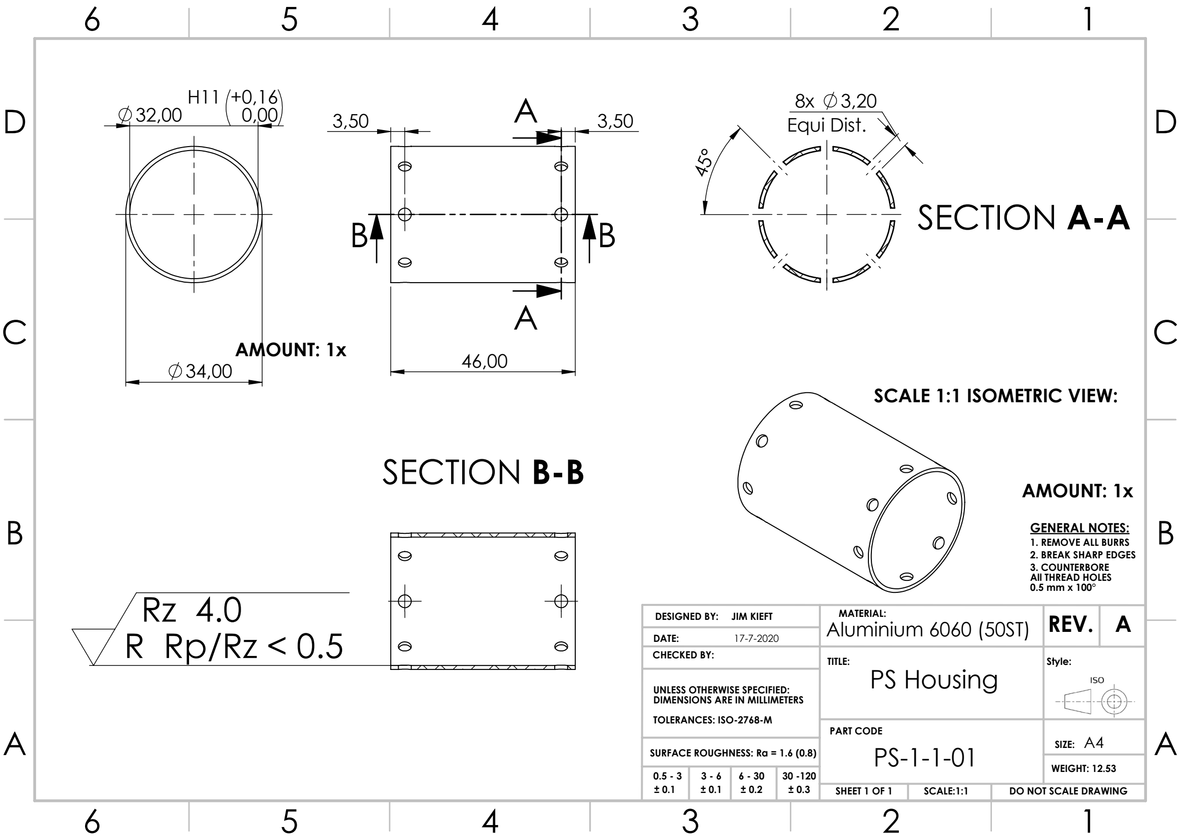


Figure N.10: The technical drawing of part: PS-1-1-01: PS Housing

PS-1-1-02: PS Piston-Rod

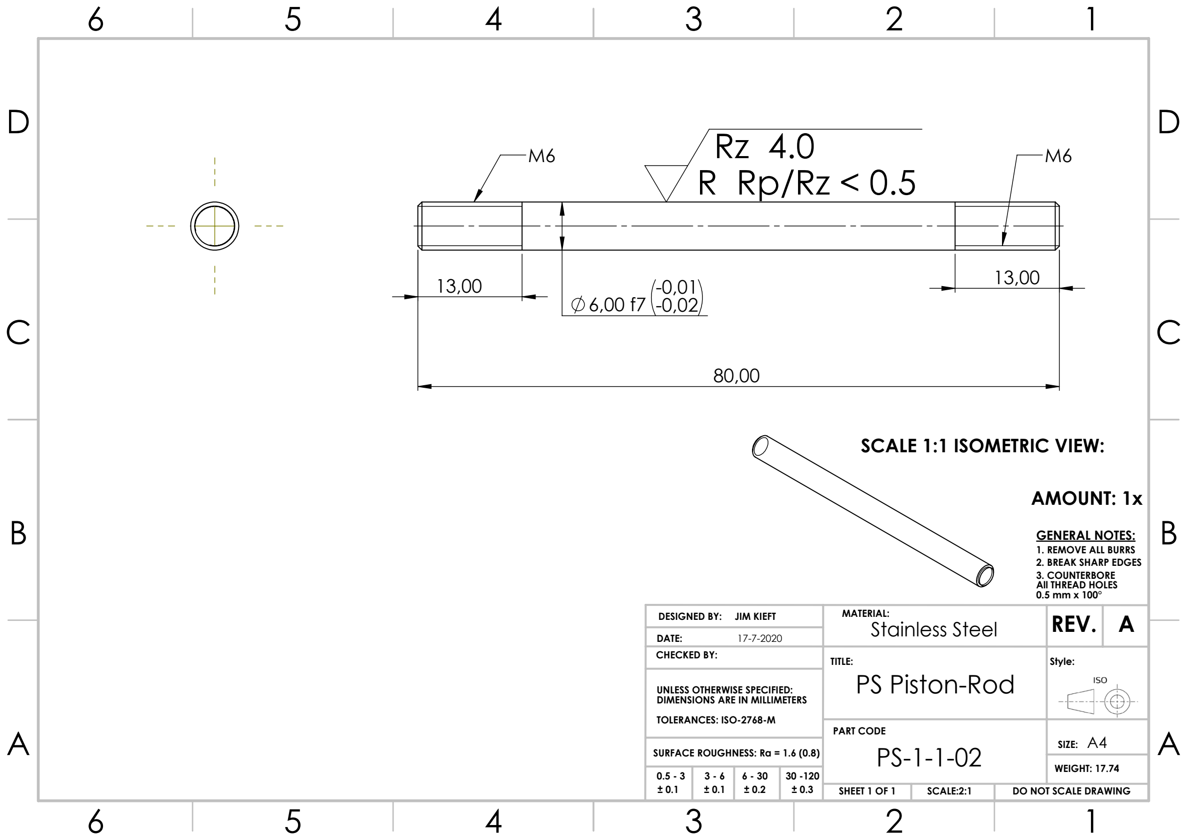


Figure N.11: The technical drawing of part: PS-1-1-02: PS Piston-Rod

PS-1-1-03: PS Piston-Rod-Nut

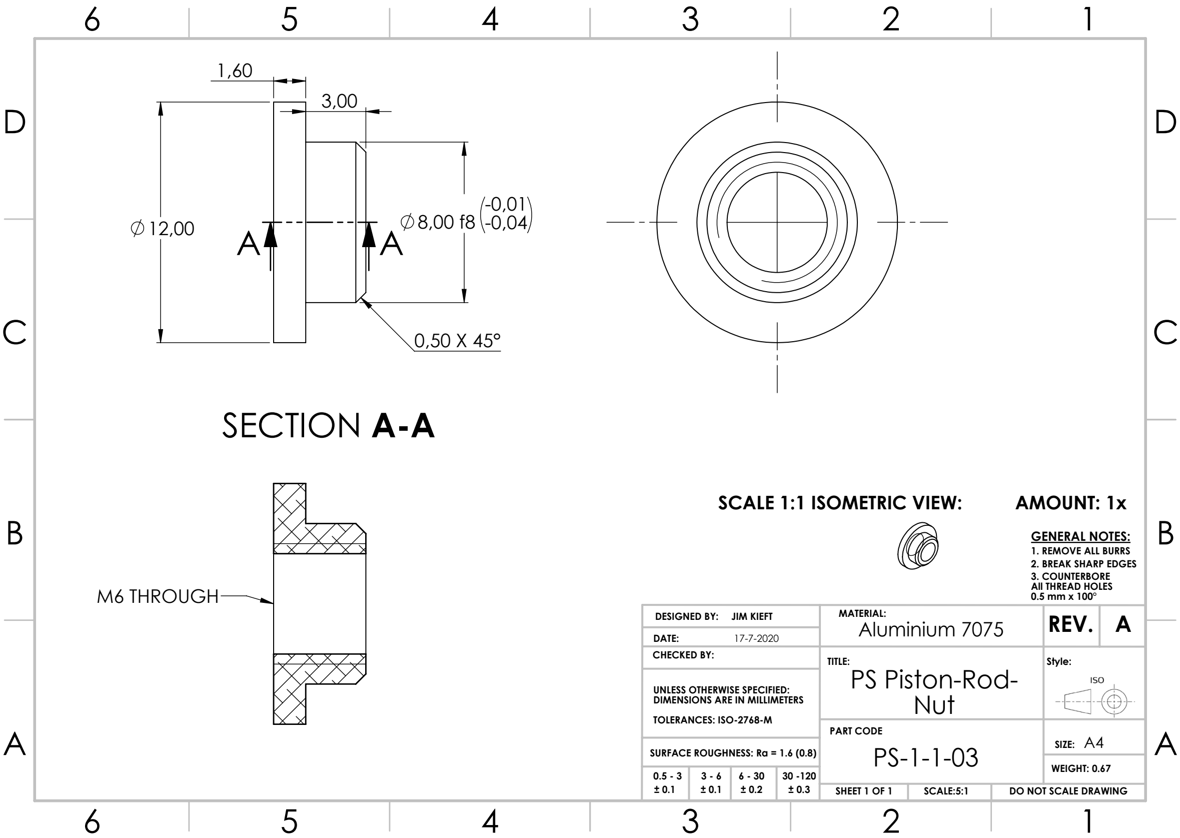


Figure N.12: The technical drawing of part: PS-1-1-03: PS Piston-Rod-Nut

PS-1-1-04: PS Cap

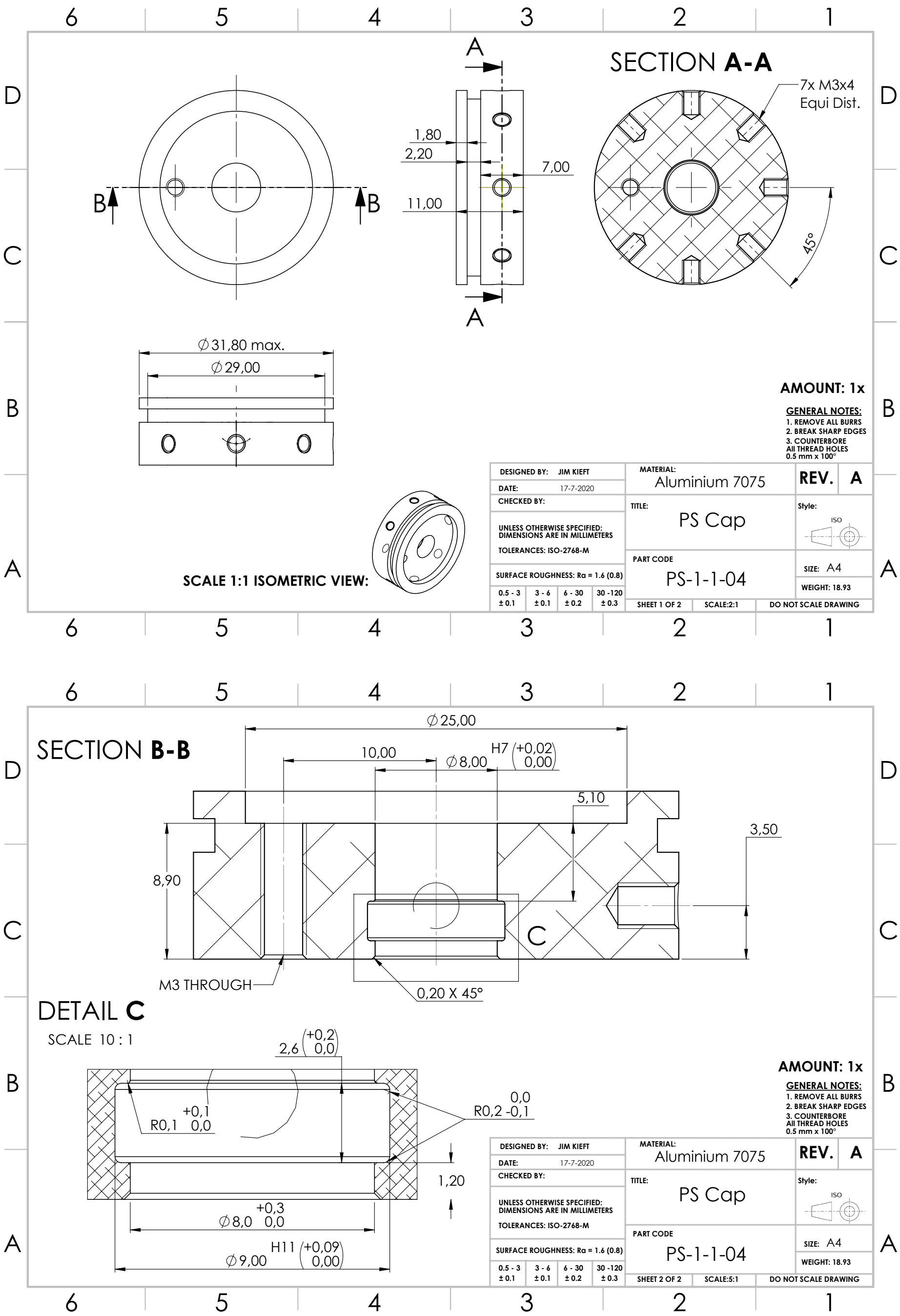


Figure N.13: The technical drawing of part: PS-1-1-04: PS Cap

PS-1-1-05: PS End-Cap

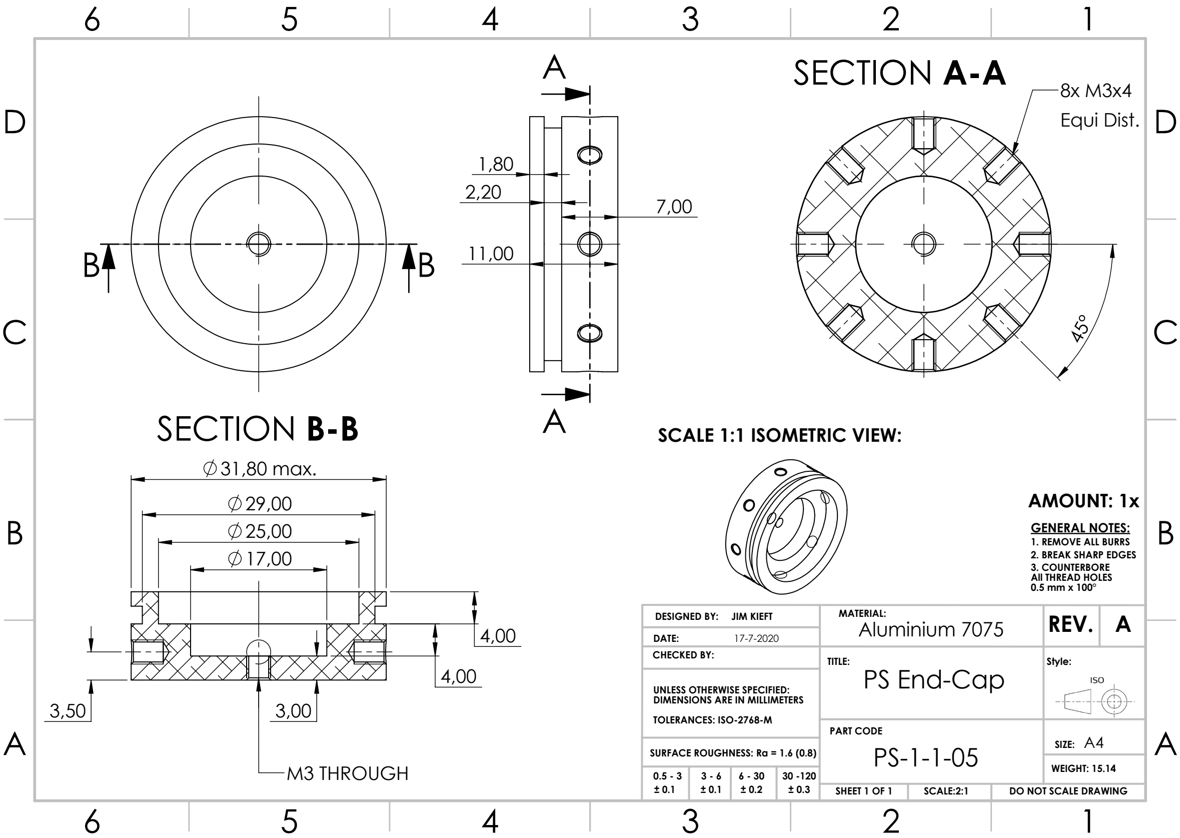


Figure N.14: The technical drawing of part: PS-1-1-05: PS End-Cap

PS-1-1-06: PS Nut-Connect

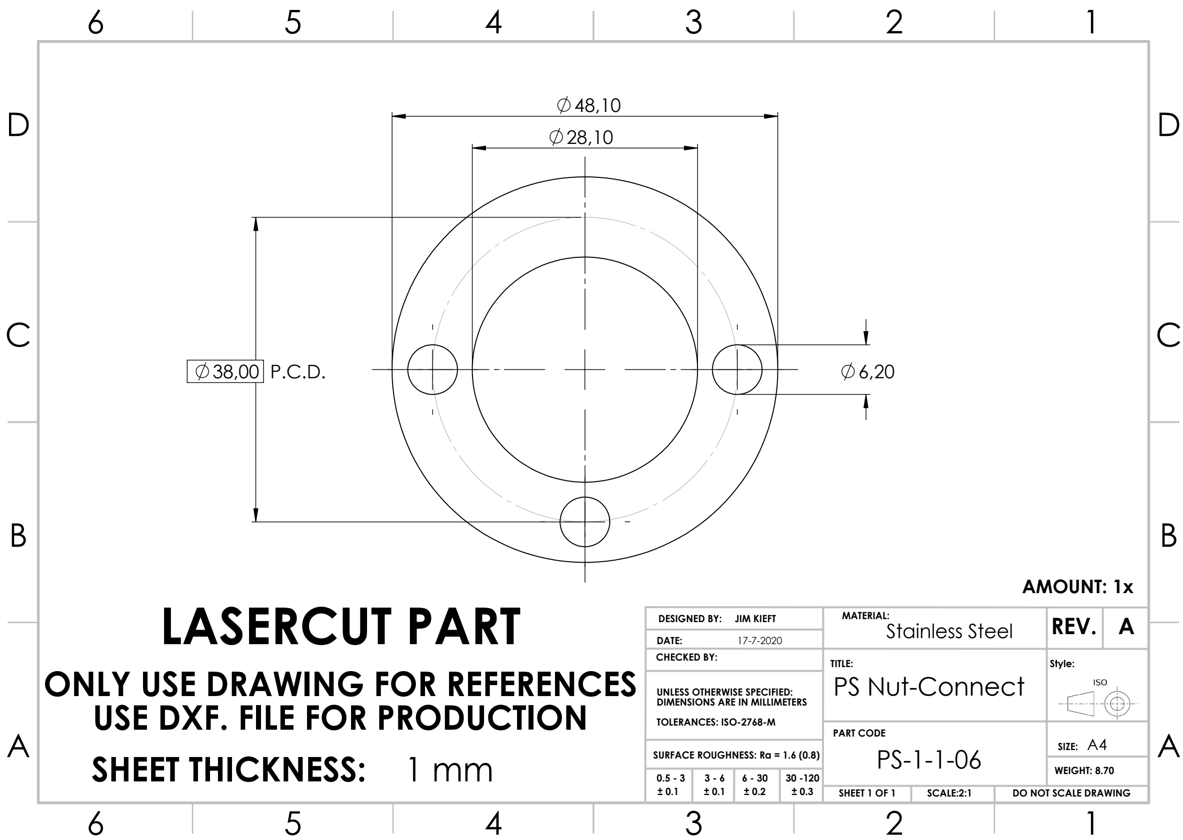


Figure N.15: The technical drawing of part: PS-1-1-06: PS Nut-Connect

PS-1-1-07: PS Magnet-Nut

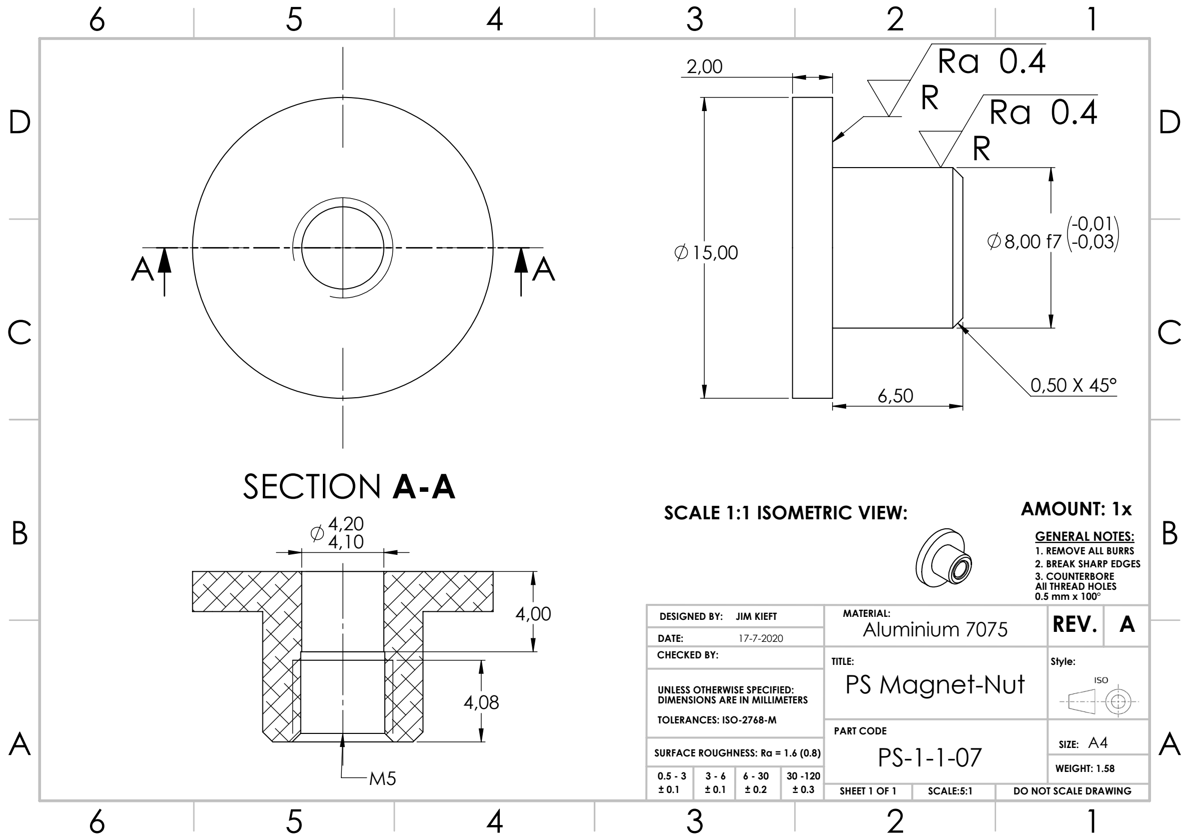


Figure N.16: The technical drawing of part: PS-1-1-07: PS Magnet-Nut

PS-1-1-08: PS Encoder-Base

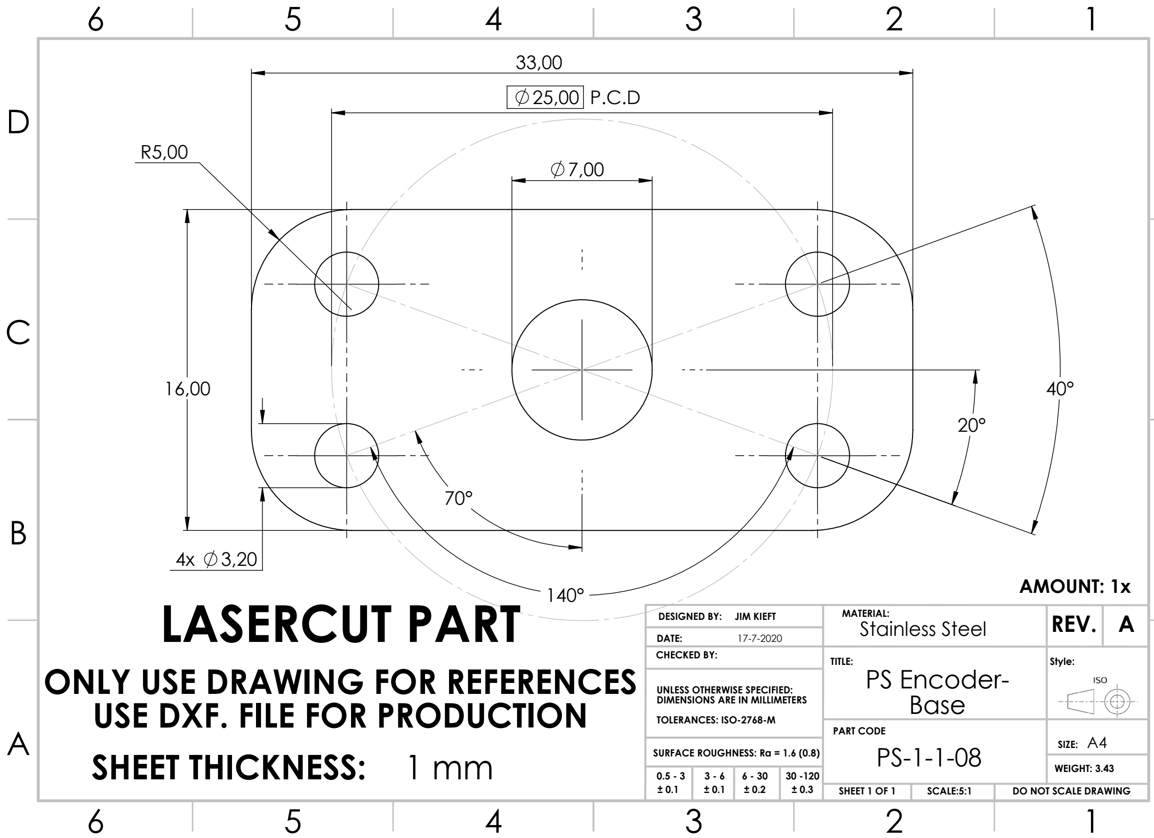


Figure N.17: The technical drawing of part: PS-1-1-08: PS Encoder-Base

PS-1-1-09: PS Lead-Screw

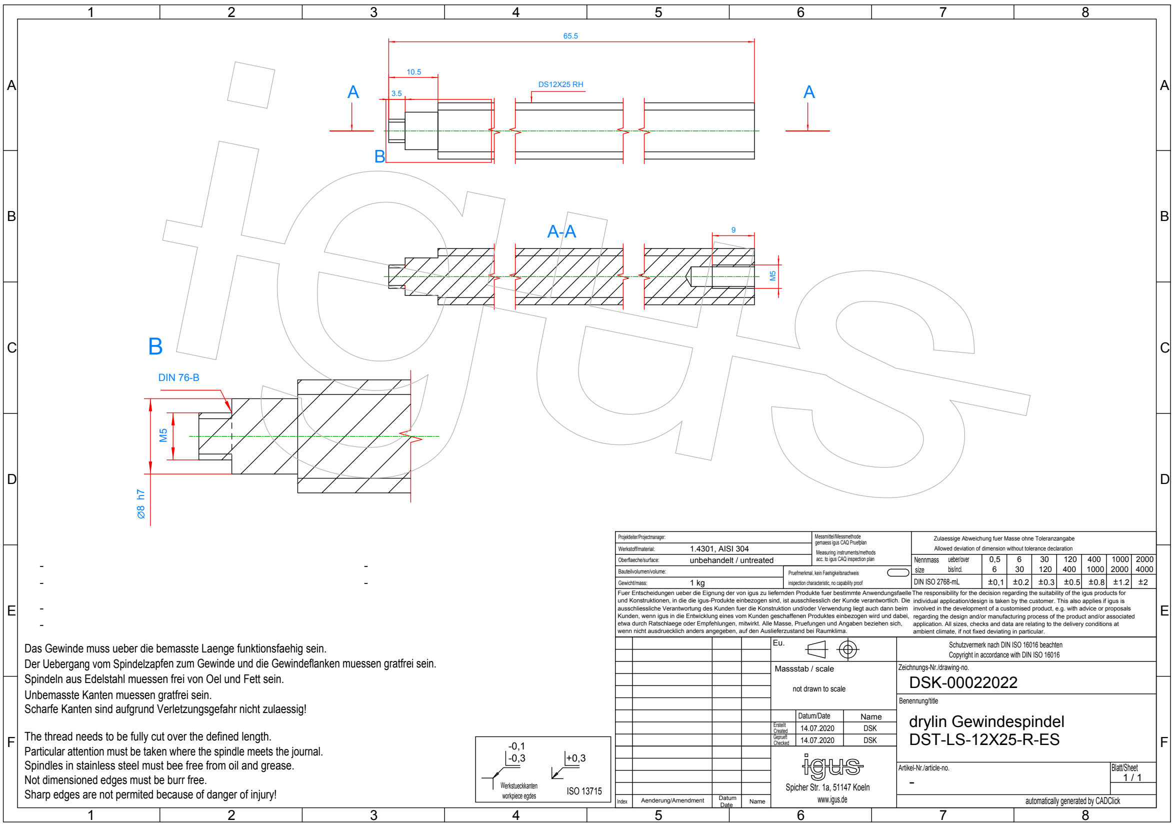


Figure N.18: The technical drawing of part: PS-1-1-09: PS Lead-Screw

LM-1-1-01: LM Housing

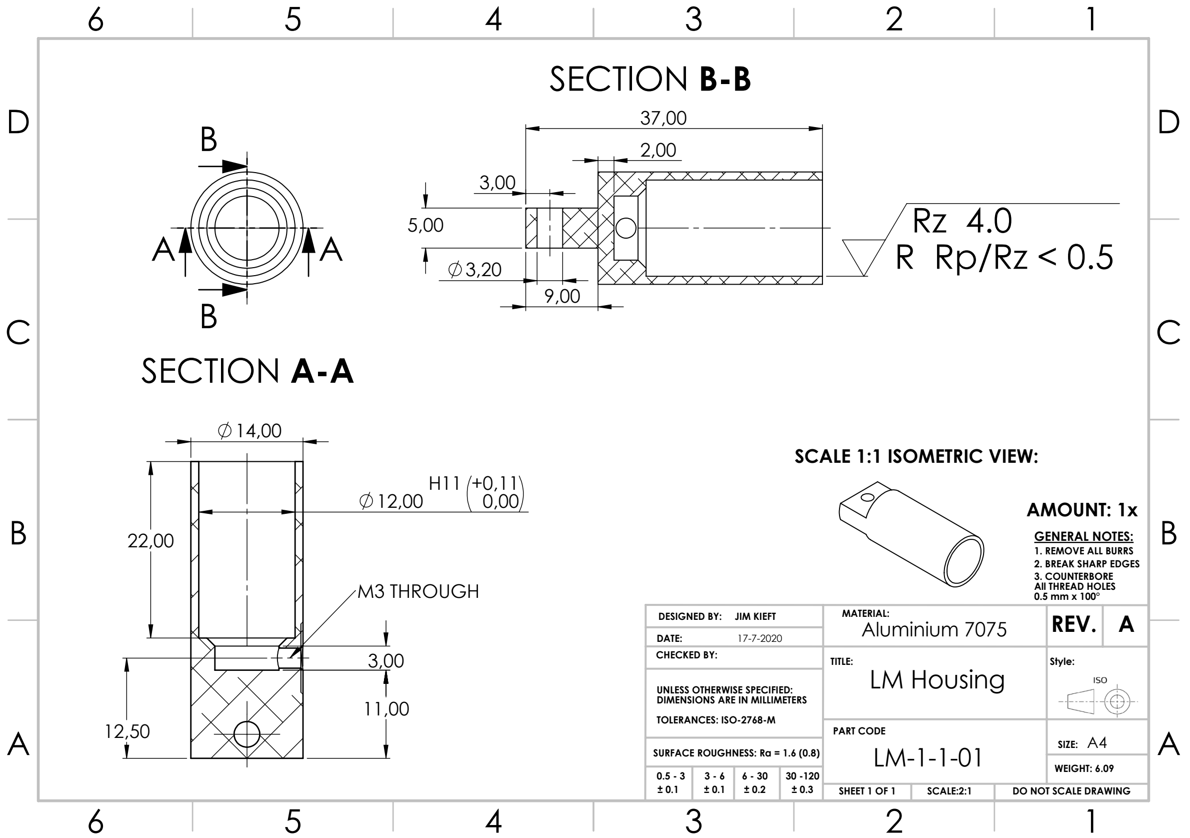


Figure N.19: The technical drawing of part: LM-1-1-01: LM Housing

LM-1-1-02: LM Piston-Rod

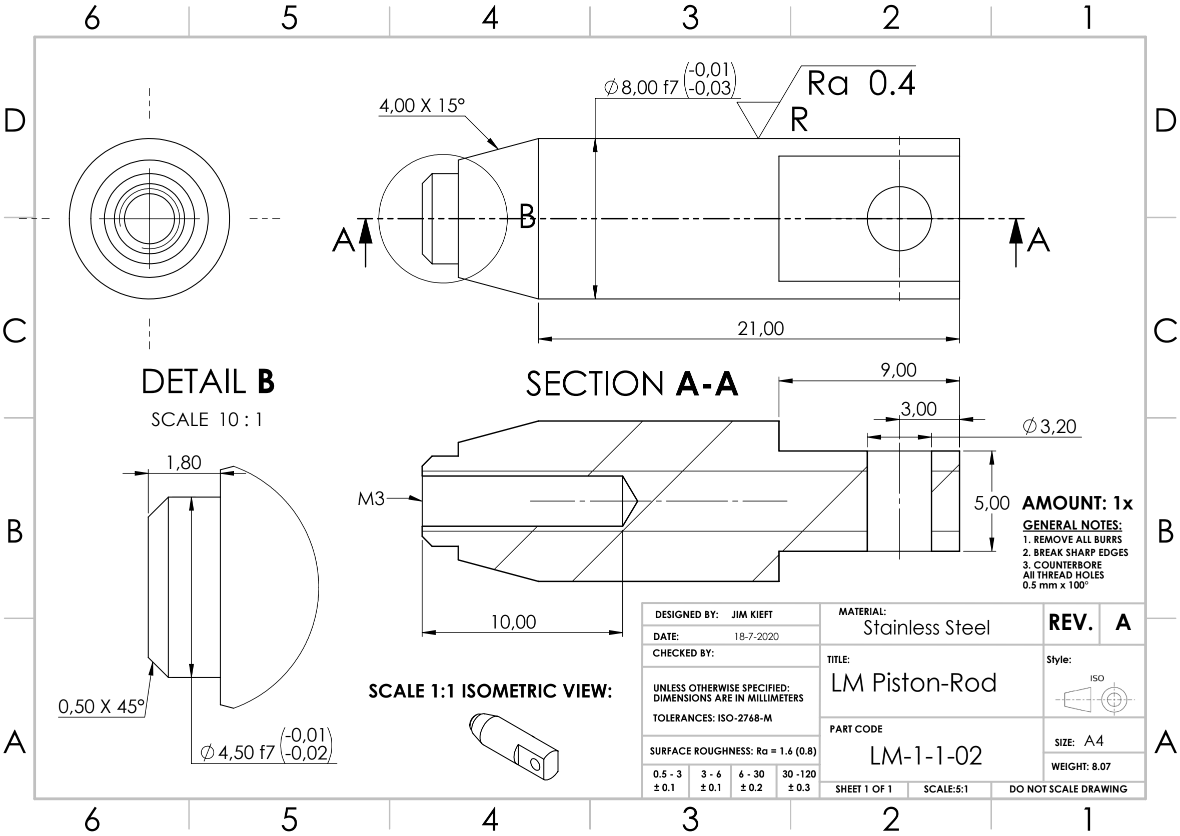


Figure N.20: The technical drawing of part: LM-1-1-02: LM Piston-Rod

LM-1-1-03: LM Spring-Washer

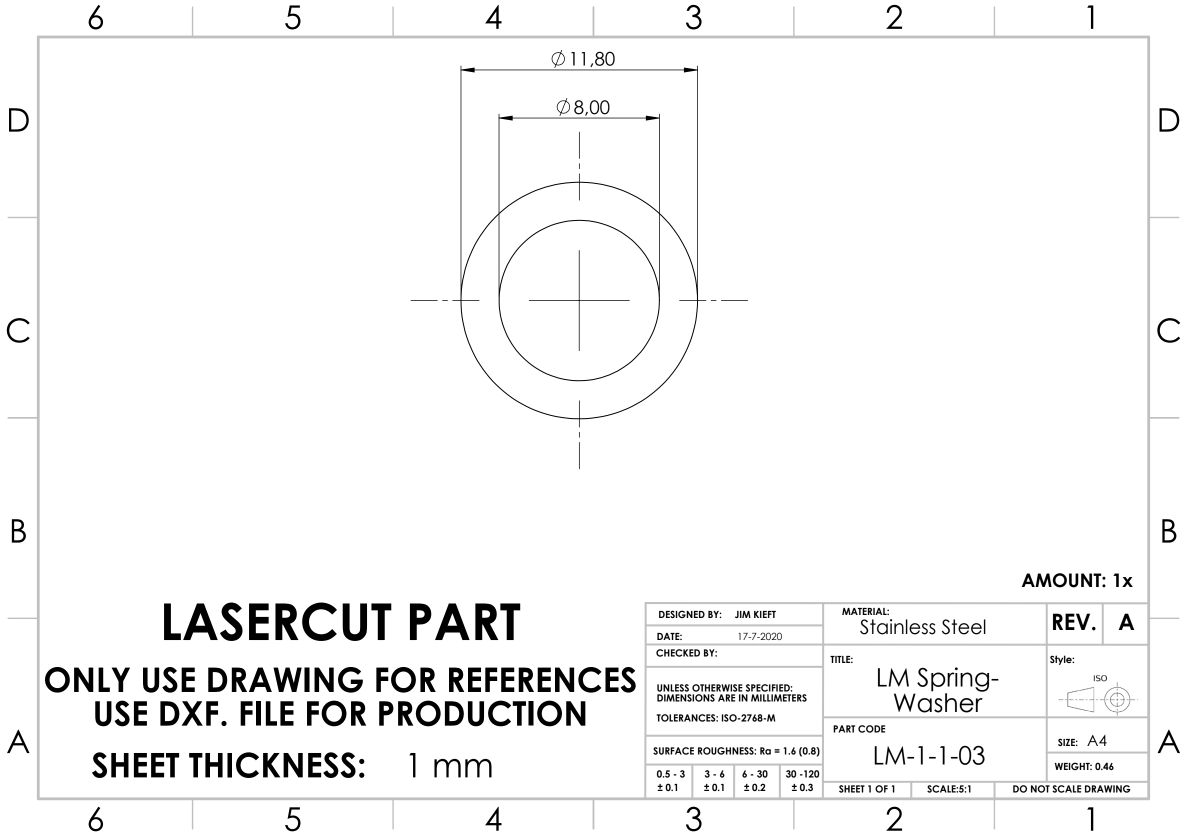


Figure N.21: The technical drawing of part: LM-1-1-03: LM Spring-Washer

FR-1-1-01: FR Forearm-Endstop

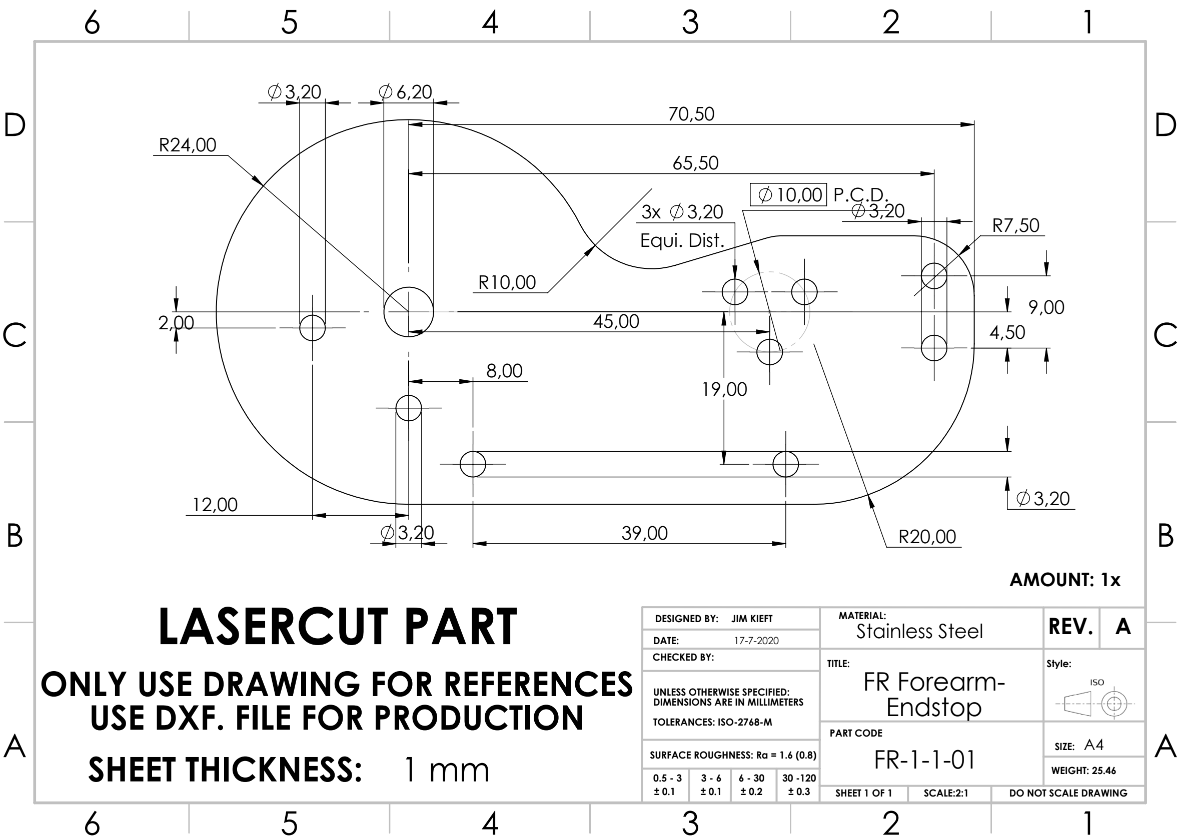


Figure N.22: The technical drawing of part: FR-1-1-01: FR Forearm-Endstop

FR-1-1-02: FR Forearm-Lock

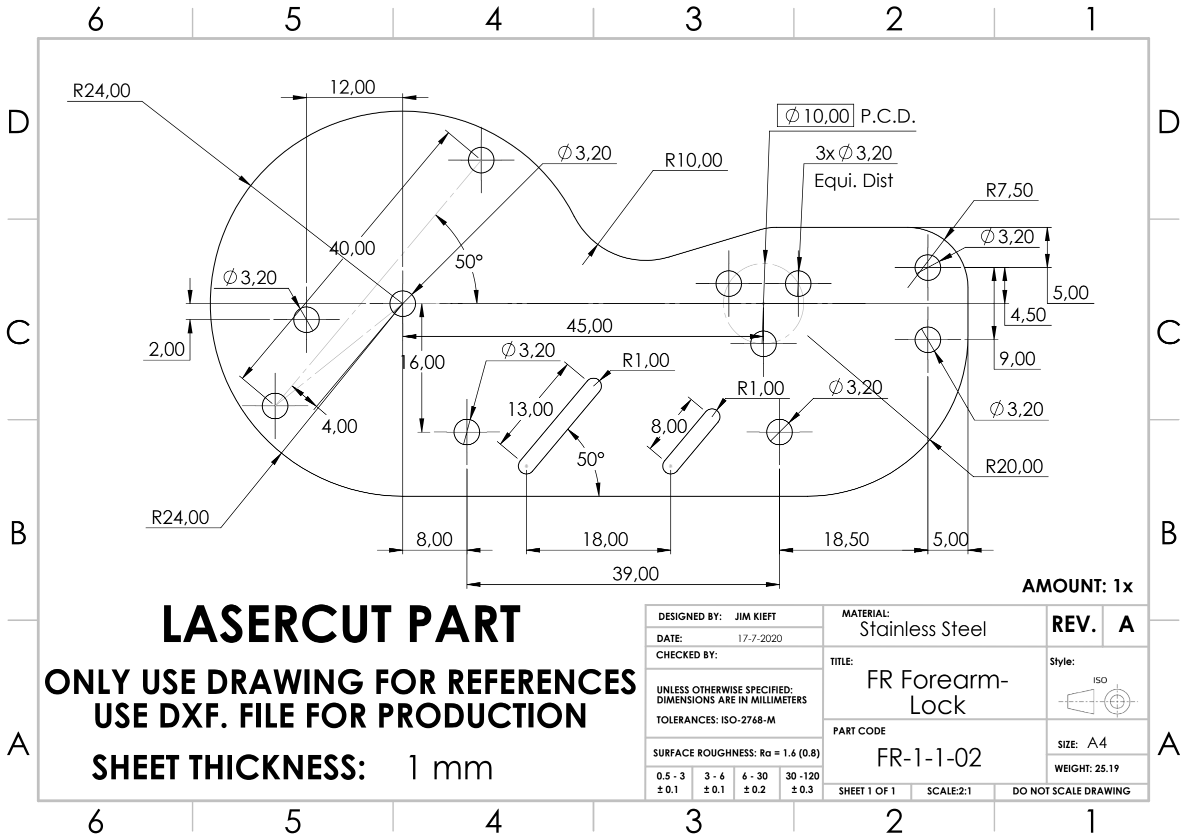


Figure N.23: The technical drawing of part: FR-1-1-02: FR Forearm-Lock

FR-1-1-03: FR FE-Connection-Rod

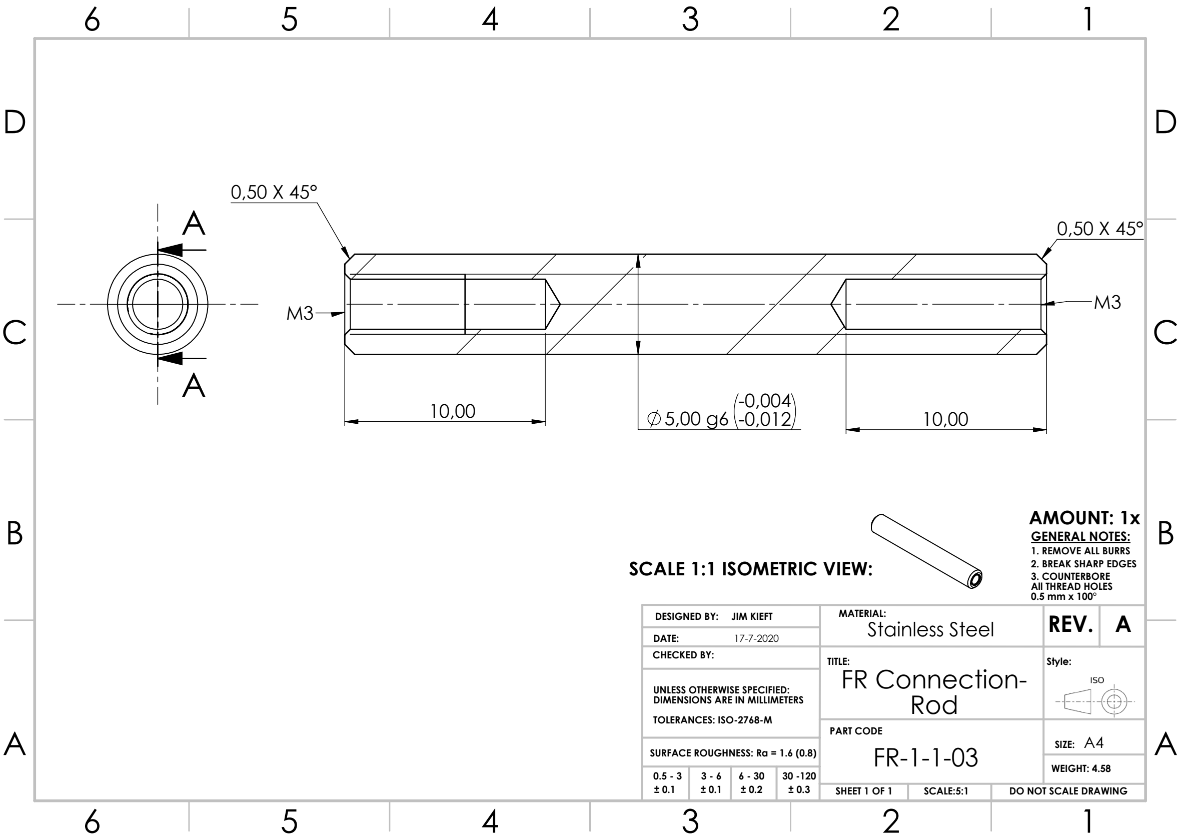


Figure N.24: The technical drawing of part: FR-1-1-03: FR FE-Connection-Rod

FR-1-1-04: FR FE-Lever

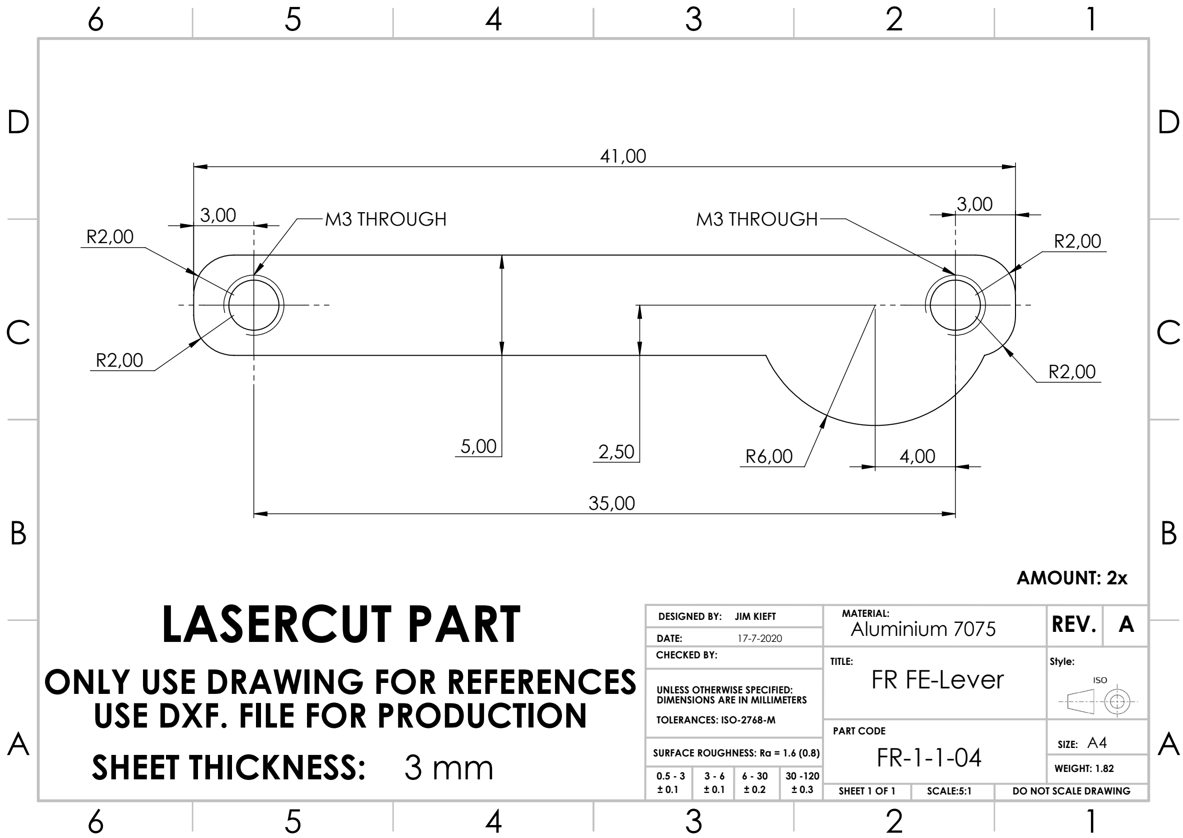


Figure N.25: The technical drawing of part: FR-1-1-04: FR FE-Lever

FR-1-1-05: FR Encoder-Shaft

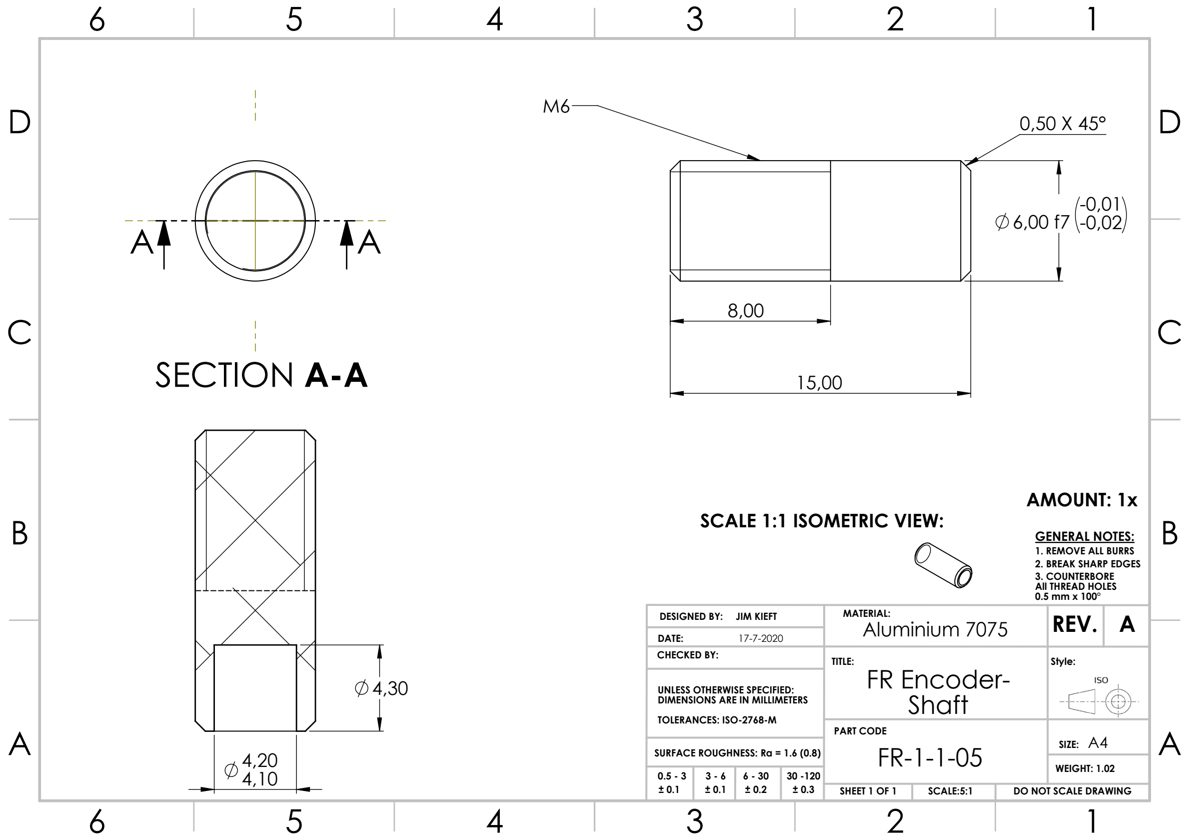


Figure N.26: The technical drawing of part: FR-1-1-05: FR Encoder-Shaft

FR-1-1-06: FR Encoder-Holder-Base

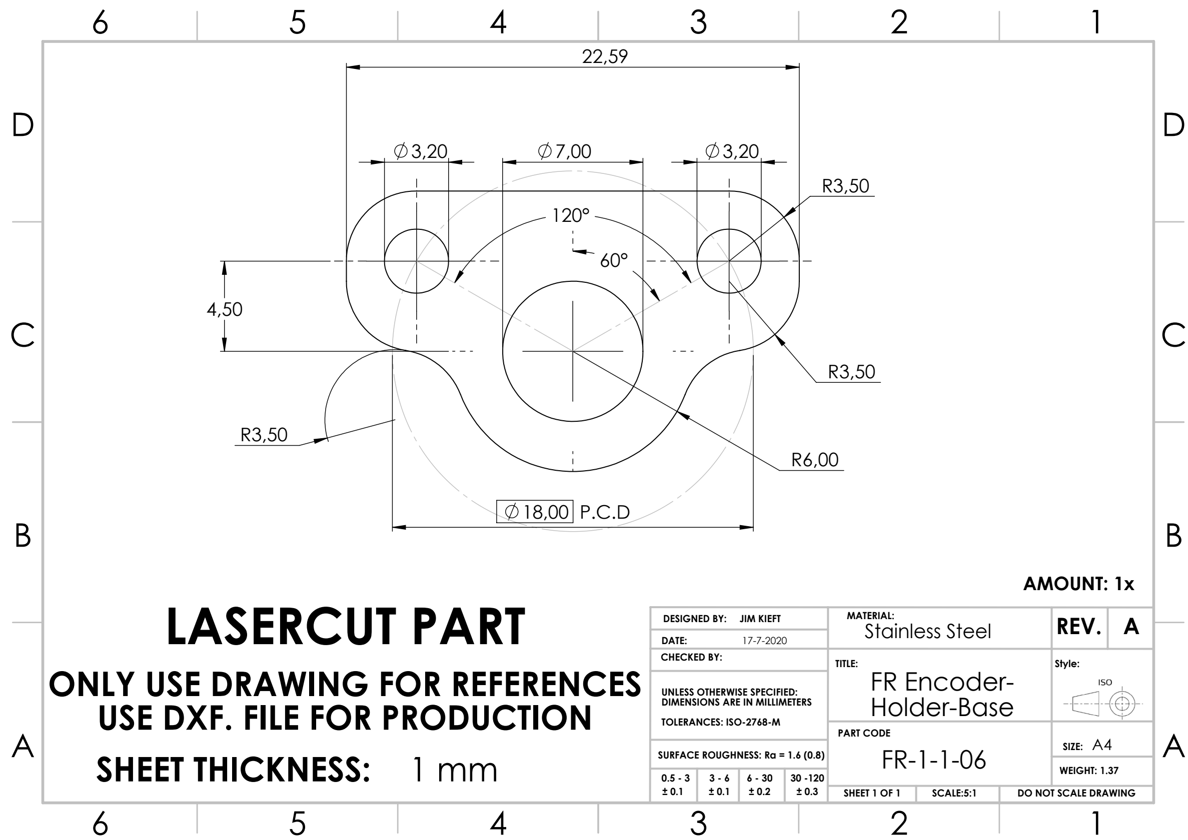


Figure N.27: The technical drawing of part: FR-1-1-06: FR Encoder-Holder-Base

FR-1-1-07: FR Endplate

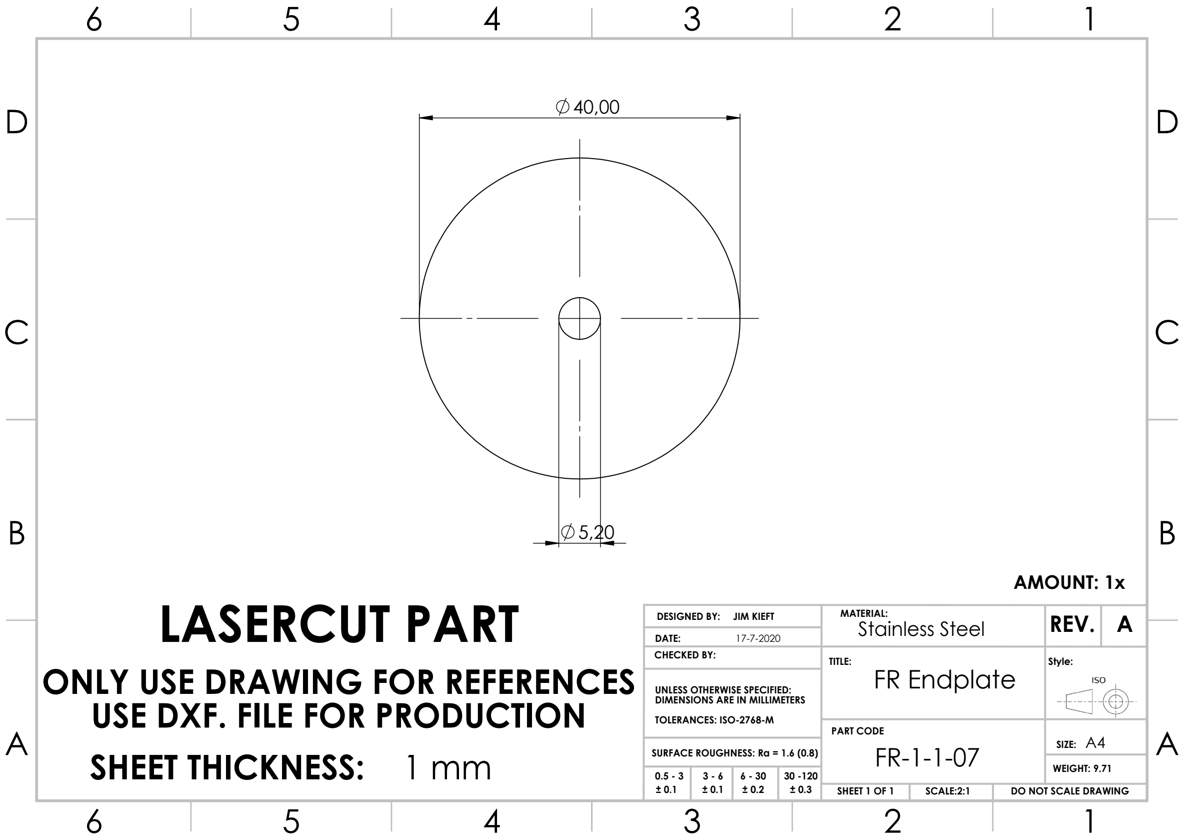


Figure N.28: The technical drawing of part: FR-1-1-07: FR Endplate

FR-1-2-01: FR Upperarm-Encoder

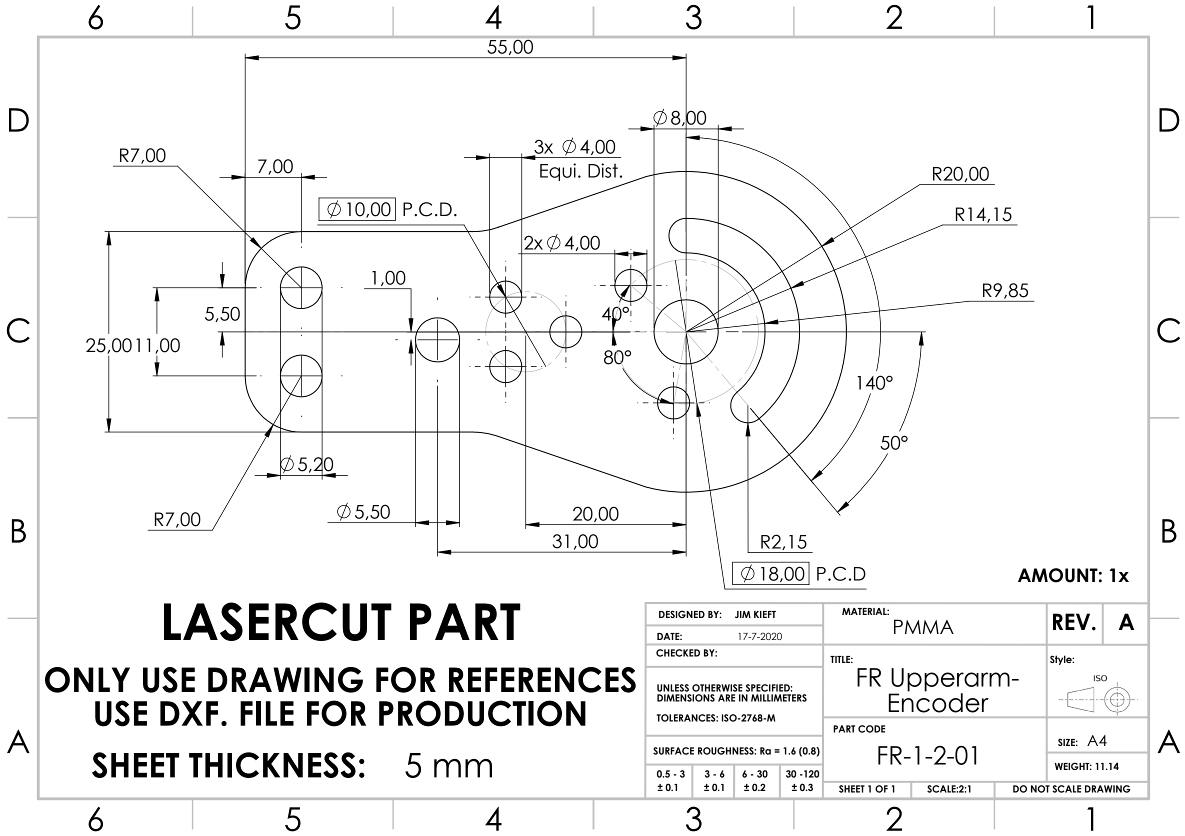


Figure N.29: The technical drawing of part: FR-1-2-01: FR Upperarm-Encoder

FR-1-2-02: FR Upperarm-Lock

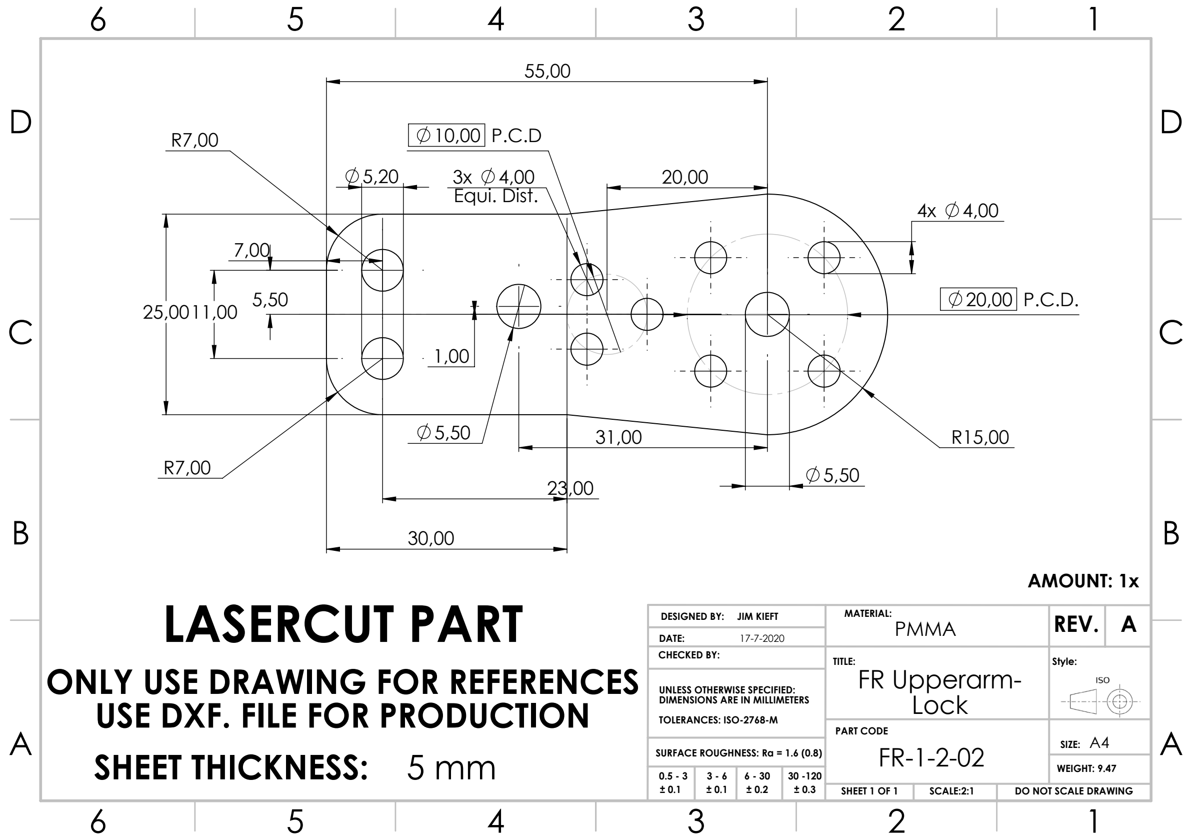


Figure N.30: The technical drawing of part: FR-1-2-02: FR Upperarm-Lock

FR-1-2-03: FR FE-Lock

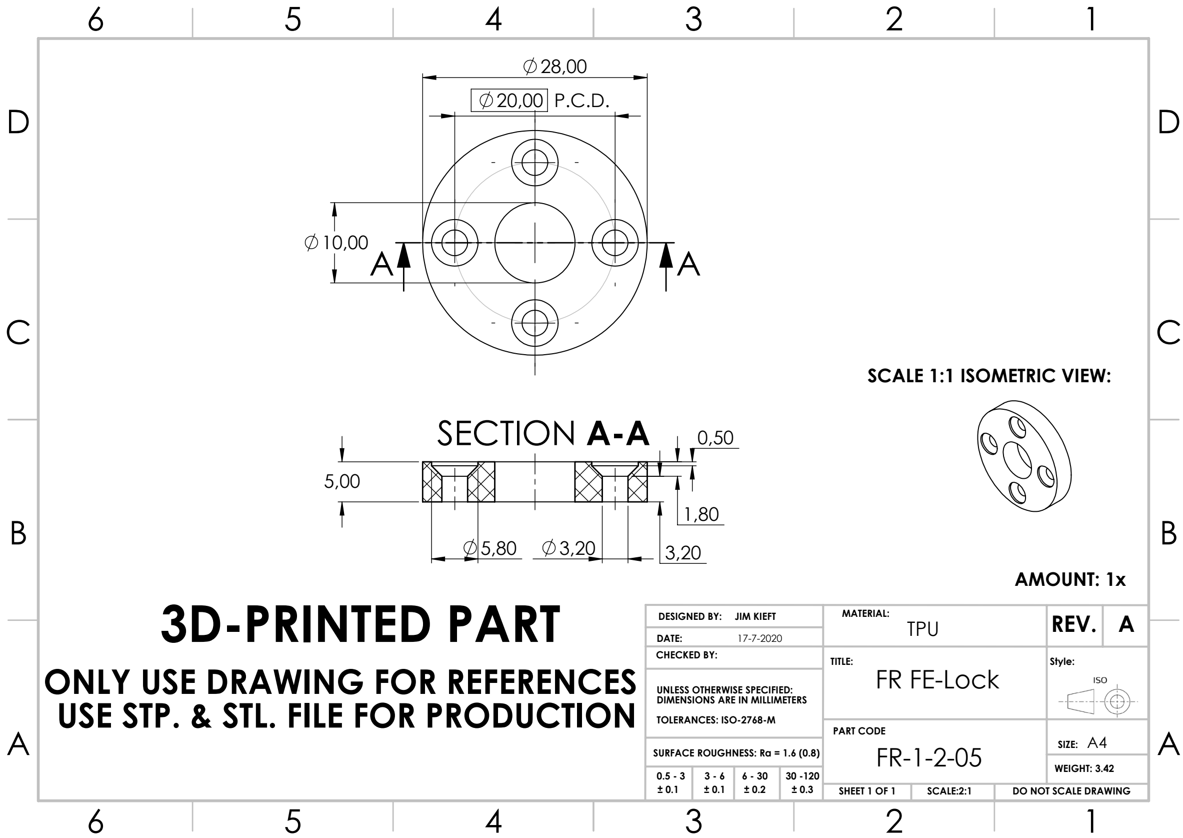


Figure N.31: The technical drawing of part: FR-1-2-03: FR FE-Lock

O

Assembly

General Actuator

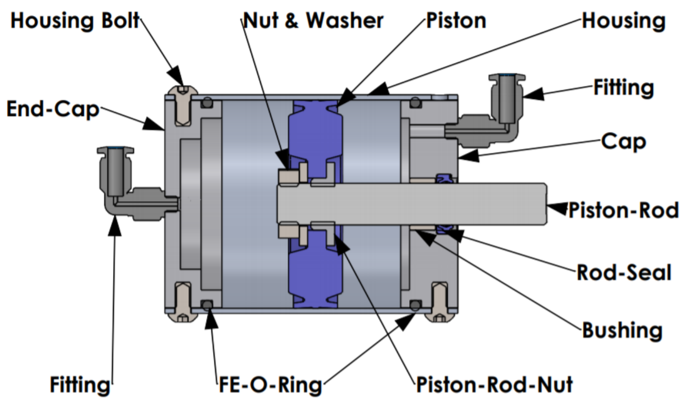
The following guide is for the assembly of both the FE-, and PS-Actuator. Please ensure you use correct parts, e.g. use the PS-O-Ring for the PS assembly.

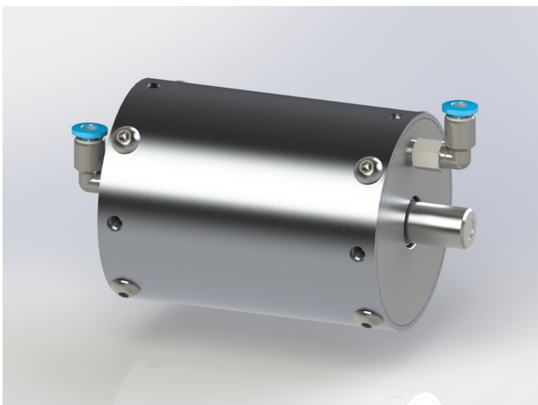
Parts/Sub-Assembly required:

- Housing
- Piston-Rod
- Piston-Rod-Nut
- Cap
- End-Cap
- Piston
- O-ring (2x)
- Rod-Seal
- Bushing
- Push-in Fitting (2x)
- M8 Nut & Washer
- Housing Bolt (8x)

Tools/Supplies Required

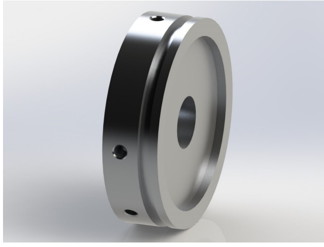
- Hand Press
- M3 Wrench (5.5 [mm])
- M6 Wrench (10 [mm])
- M8 Wrench (13 [mm])
- Pick
- Glue
- Lint-free Cloth



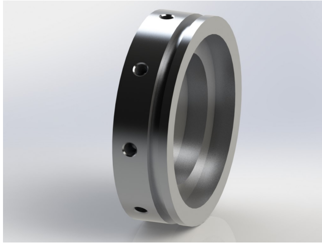


The actuator without end-effector

Step 1: Prepare the Caps



Cap



End-Cap

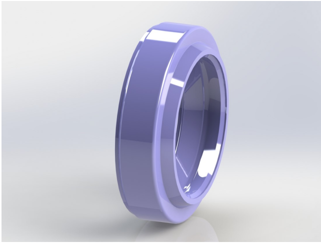
Parts/Sub-Assembly required:

- Cap
- End-Cap
- Bushing
- Rod-Seal
- O-Ring (2x)
- Push-in Fitting (2x)

Tools/Supplies Required

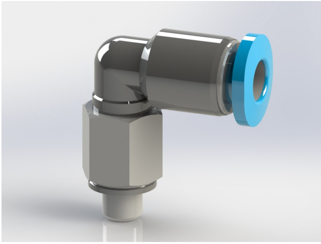
- Hand Press
- M3 Wrench (5.5 [mm])
- Lint-free Cloth

Rod-Bushing)



Rod-Seal

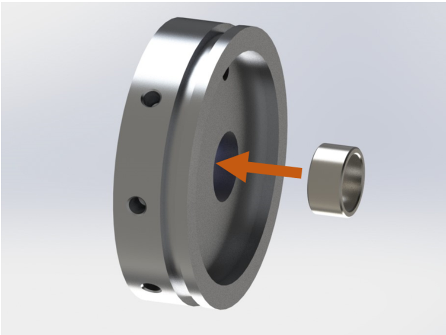
O-Ring (2x)



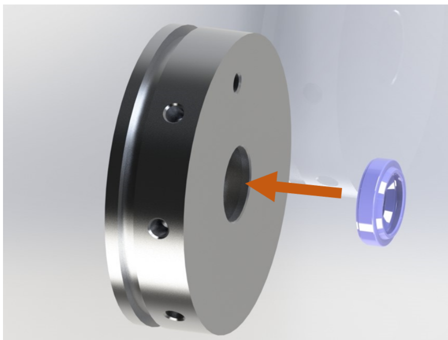
Push-in Fitting (2x)

Cap Sub-Assembly

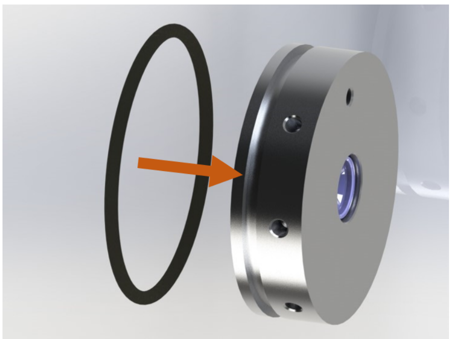
1. Use the hand press to press the bushing into the cap
   - Use the lint-free cloth to prevent damaging the bushing



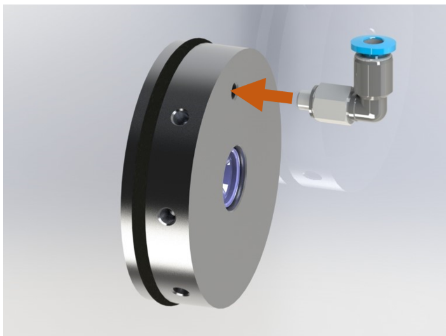
2. Use the pick to place the rod-seal into the cap



3. Place the o-ring in the groove at both caps



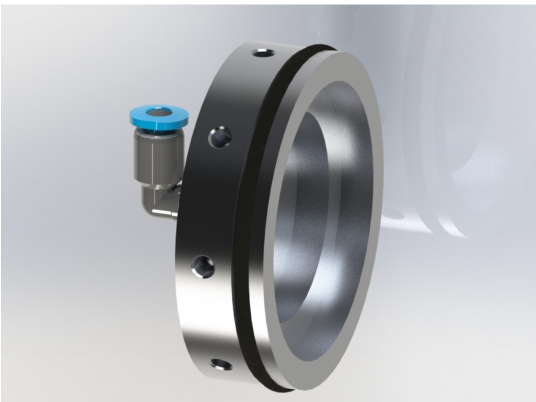
4. Screw the push-in fitting to the cap
    - Use the M3 wrench to tighten at ≈0.45 [Nm]



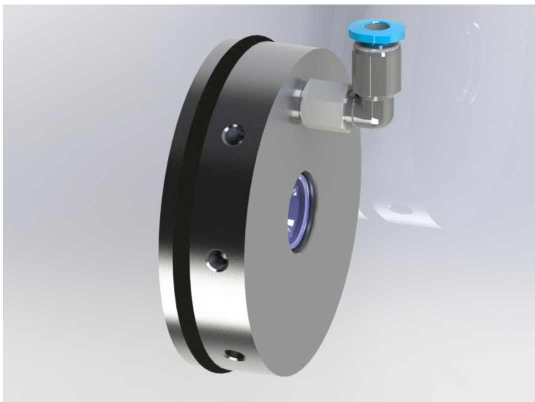
End-Cap Sub-Assembly

1. Repeat steps 3 & 4 of the Cap Sub-assembly

Output Step 1



End-Cap Sub-Assembly



Cap Sub-Assembly

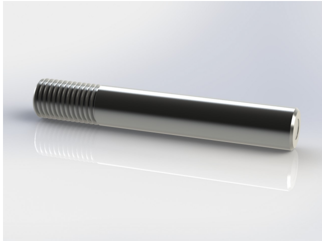
Step 2: Prepare the Piston-Sub-Assembly

Parts/Sub-Assembly required:

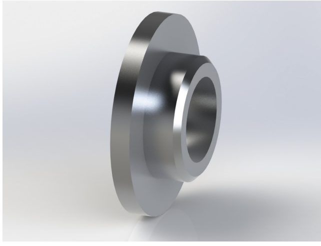
- Piston-Rod
- Piston-Rod-Nut
- Piston
- Nut & Washer

Tools/Supplies Required

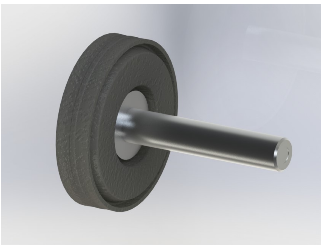
- M6 Wrench (10 [mm])
- M8 Wrench (13 [mm])
- Glue
- Lint-free Cloth



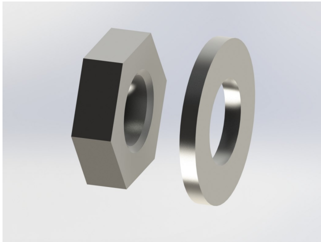
Piston-Rod



Piston-Rod-Nut

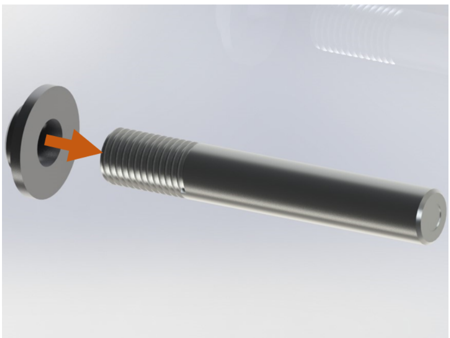


Piston

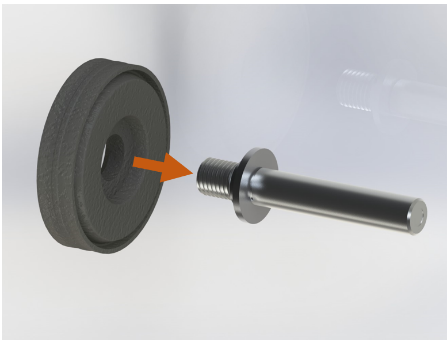


Nut & Washer

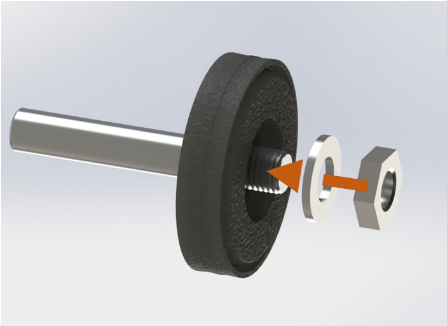
1. Screw the piston piston-rod-nut on the piston-rod, with the smaller diameter facing outwards if needed use a clamp to hold the piston-rod. Use the lint-free cloth to prevent damaging the piston-rod.
   - For the FE sub-assembly: Use a clamp to tighten at ≈15[Nm]
   - For the FE sub-assembly: Use a clamp to tighten at ≈5 [Nm]



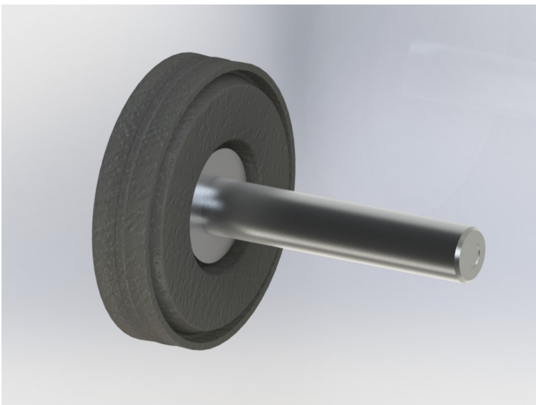
2. Place the piston on the piston-rod-nut, rotate to disperse the glue evenly
   - Put a droplet of glue on the piston-rod-nut if dis-assembly is no longer desired



3. Apply the washer and nut, if needed use a clamp to hold the piston-rod. Use the lint-free cloth to prevent damaging the piston-rod.
    - For the FE sub-assembly: Use the M8 wrench to tighten at ≈23 [Nm]
    - For the FE sub-assembly: Use the M6 wrench to tighten at ≈9 [Nm]



Output Step 2



Piston Sub-Assembly

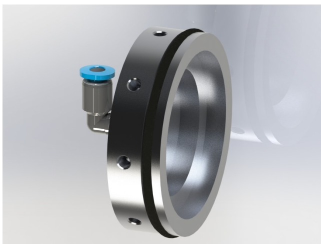
Step 3: Assemble the Actuator

Parts/Sub-Assembly required:

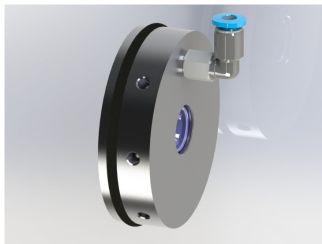
- Cap-Sub-Assembly
- End-Cap-Sub-Assembly
- Piston-Sub-Assembly
- Housing
- Housing Bolt M3x5 (8x)

Tools/Supplies Required

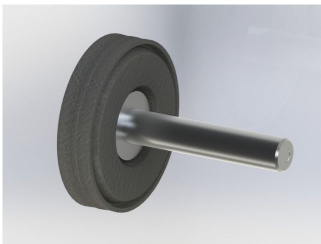
- Hand Press
- Screwdriver
- Lint-free cloth
- O-ring Protector (4x)
- Grease
- Rocol kilopoise 0001



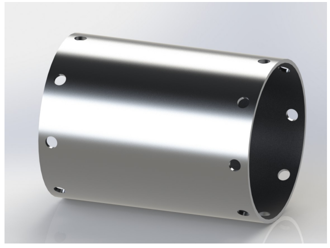
End-Cap Sub-Assembly



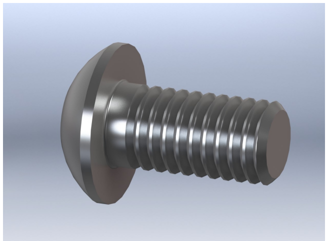
Cap Sub-Assembly



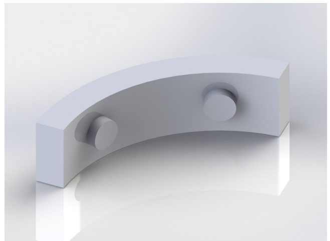
Piston Sub-Assembly



Housing

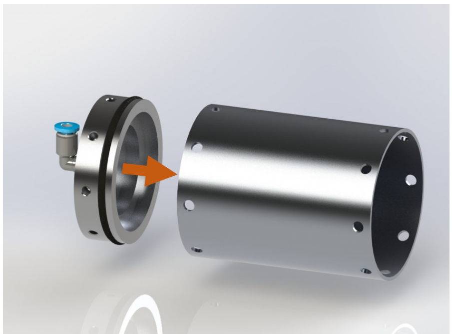


Housing Bolt (8x)

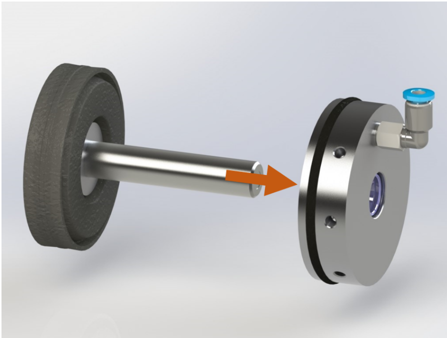


O-ring Protector

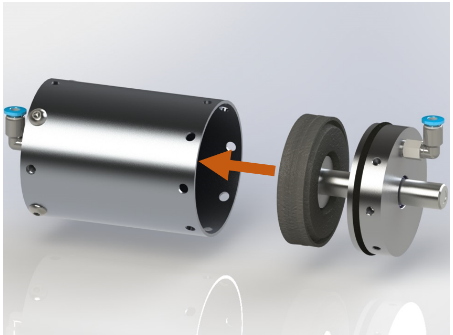
1. Press the End-Cap-Sub-Assembly into the housing. Use the lint-free cloth to prevent damaging the parts
   - Make sure to align the holes in the housing with the threaded holes in the end-cap
   - Use grease for an easier implementation of the cap
   - Use the o-ring protectors to prevent the o-ring from "bulging" out of the holes in the housing.
2. Screw in four of the bolts, leave one hole open between the bolts
   - Use the screwdriver to tighten at ≈0.45 [Nm]



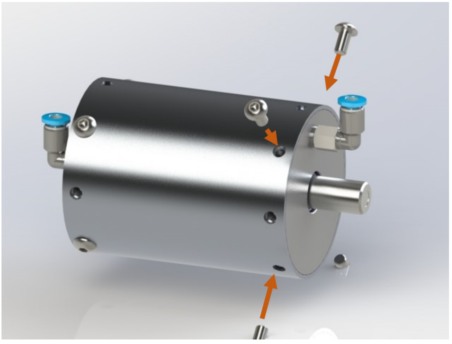
3. Push the piston-rod through the bushing hole in the cap.
    - Use Rocol kilopoise 0001 as a lubricant



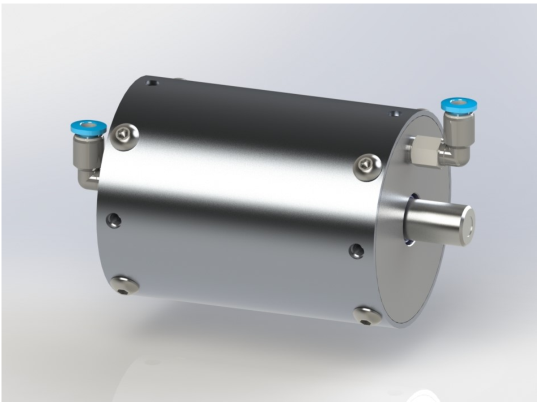
4. Press the End-Cap/Piston-Sub-Assembly into the housing. Use the lint-free cloth to prevent damaging
    - Make sure to align the holes in the housing with the threaded holes in the end-cap
    - Use grease for an easier implementation of the cap
    - Use the o-ring protectors to prevent the o-ring from "bulging" out of the holes in the housing.



5. Screw in four of the bolts, in-line with the bolts of step 3.2
    - Use the screwdriver to tighten at ≈0.45 [Nm]



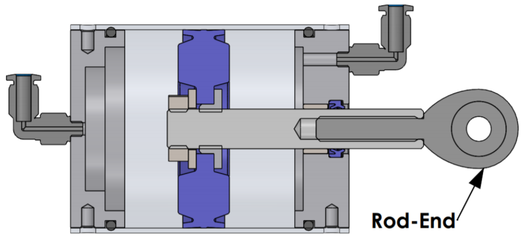
Output Step 3



The actuator without end-effector

FE Actuator

Perform Validation Test 1 before starting this assembly

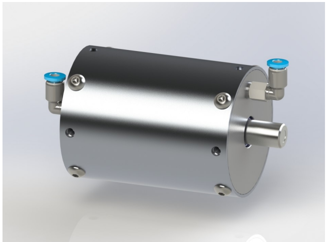


Parts/Sub-Assembly required:

- FE General Actuator
- Rod-End

Tools/Supplies Required

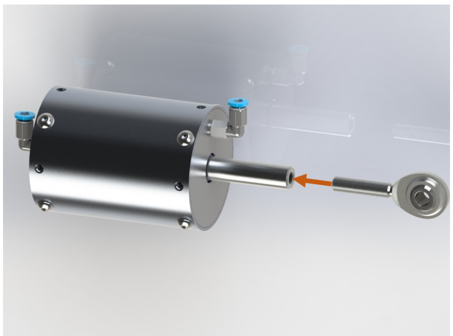
- Clamp
- Lint-free cloth
- Screwdriver



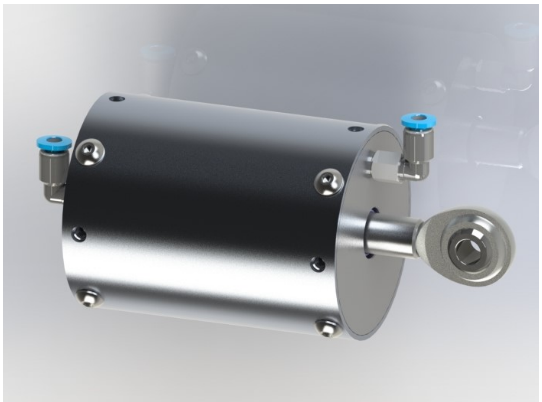
The FE-actuator without end-effector

Rod-End

1. Screw the rod-end into the piston-rod. Clamp the piston-rod with the lint-free cloth to prevent damaging
   - Use the screwdriver to tighten at ≈5 [Nm]



Output FE Actuator



The FE-actuator

PS Actuator

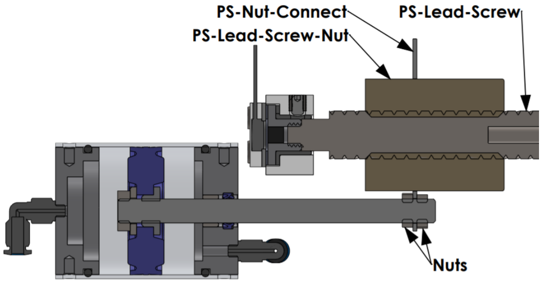
Perform Validation Test 1 before starting this assembly

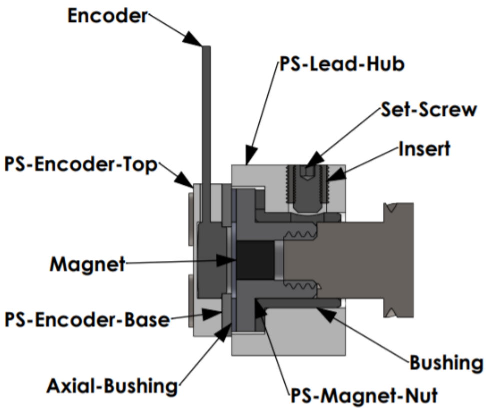
Parts/Sub-Assembly required:

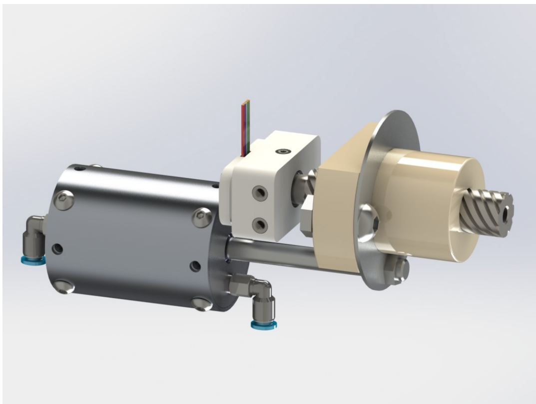
- PS General Actuator
- PS-Nut-Connect
- PS-Magnet-Nut
- PS-Encoder-Base
- PS Lead-Screw
- PS-Collar
- PS-Encoder-Top
- PS Lead-Screw Nut
- Encoder
- Magnet
- Flanged Bushing
- Axial-Bushing
- Insert (9x)
- M6 Nut (4x)
- Set-screw
- Lead-nut Bolts (2x)
- Countersunk Bolts (4x)

Tools/Supplies Required

- Soldering Iron (with a sharp tip)
- Hand Press
- Clamp
- Lint-free Cloth







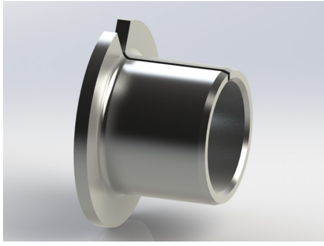
Step 1: Prepare the Collar

Parts/Sub-Assembly required:

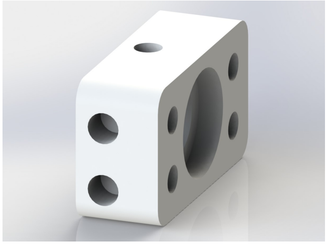
- Flanged Bushing
- PS-Collar
- Insert (9x)
- Set-screw

Tools/Supplies Required

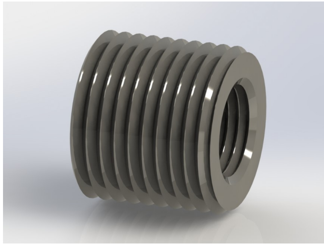
- Soldering Iron (with a sharp tip)
- Hand Press
  - ⌀ 15 [mm] cylinder



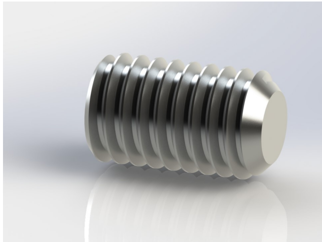
Flanged Bushing



PS-Collar

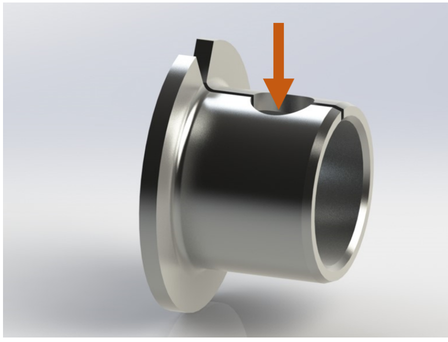


Insert (9x)

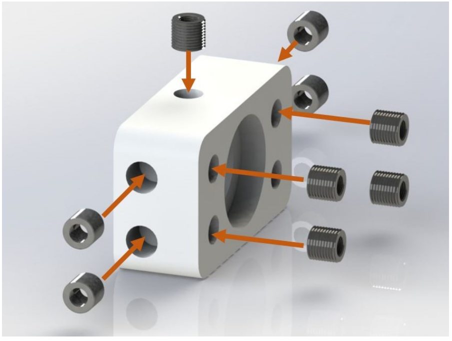


Set-screw

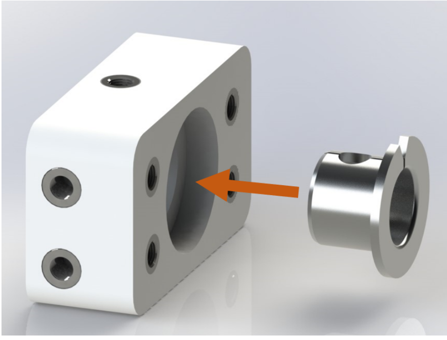
1. Drill a 4 [mm] hole in the flanged bushing, x [mm] from the non-flanged side.



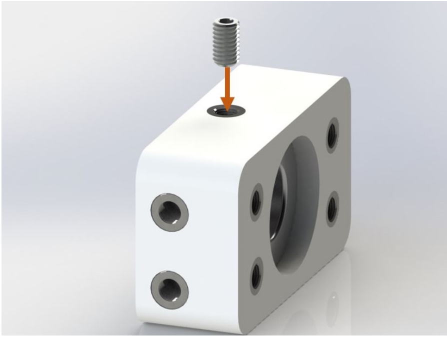
2. Use the soldering iron to heat the inserts and push them in the desired holes



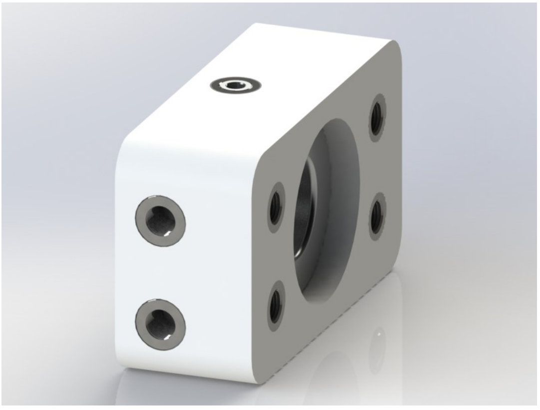
3. Press the flanged bushing into the collar. Use the ⌀ 15 [mm] cylinder to push it in fully and use the lint-free cloth to prevent damaging the bushing.



3. Screw the set-screw into the top threaded hole



Output Step 1



PS-Collar Sub-Assembly

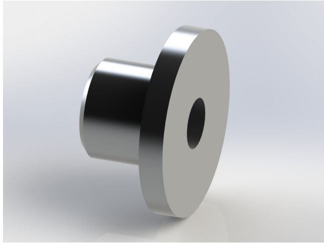
Step 2: Prepare the Lead-Screw-Sub-Assembly

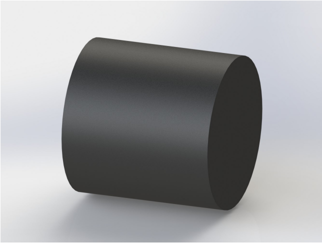
Parts/Sub-Assembly required:

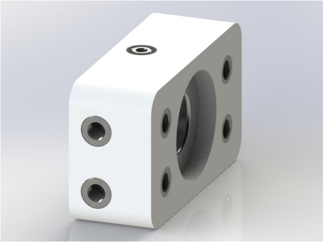
- PS-Magnet-Nut
- Magnet (not used in prototype)
- PS-Collar Sub-Assembly
- PS-Lead-Screw
- Axial-Bushing
- PS-Encoder-Base
- Encoder (not used in prototype)
- PS-Encoder-Top
- Countersunk Bolts M3x8 (4x)

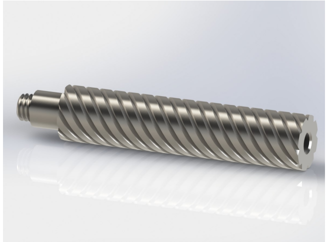
Tools/Supplies Required

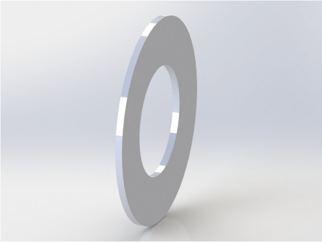
- Hand Press
- Lint-free Cloth

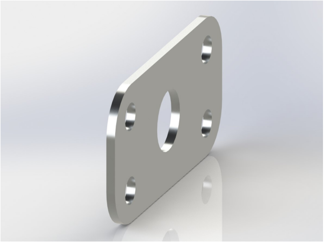

PS-Magnet-Nut

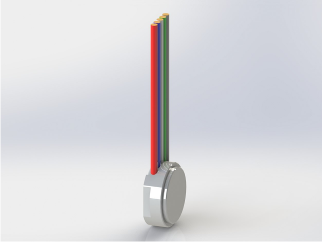

Magnet


PS-Collar Sub-Assembly

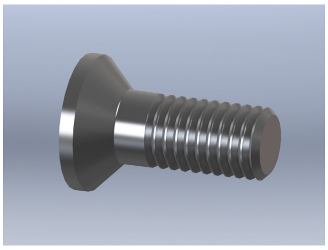

PS-Lead-Screw


Axial-Busing

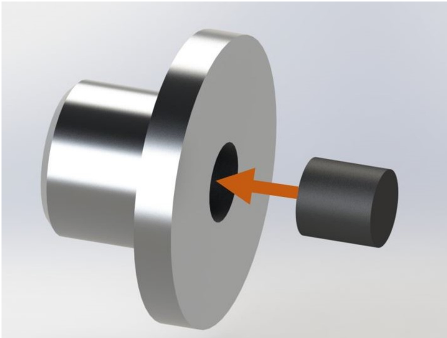

PS-Encoder-Base


Encoder

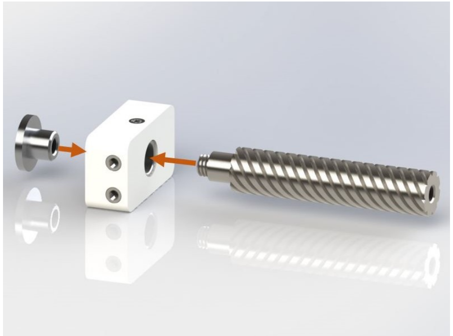

PS-Encoder-Top


Countersunk Bolts M3x8 (4x)

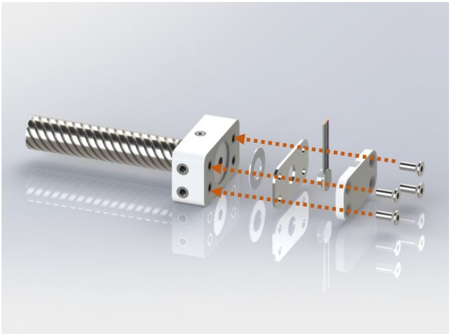
1. Press the magnet into the magnet-nut. Use the lint-free cloth to prevent damage.



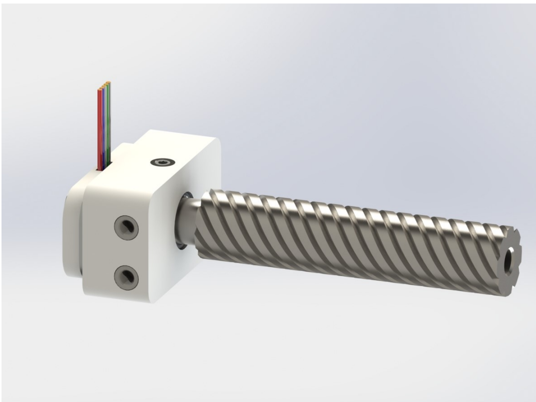
2. Put the lead-screw through the collar and screw into the magnet-nut
   - **Do not clamp the lead-screw on the lead**
   - If needed, it is allowed to carefully clamp the lead-screw at the flush part. Use the lint-free cloth to prevent damage
   - Clamp the magnet-nut on the outer edges of the ⌀ 15 [mm]. Use the lint-free cloth to prevent damage
   - Tighten to ≈1.5 [Nm]



3. Place the axial bushing, encoder base, encoder, and encoder top. Attach to the frame using the countersunk bolts.
   - Tighten to ≈0.45 [Nm]



Output Step 2



Lead-Screw Sub-Assembly

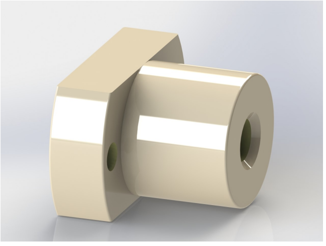
Step 3: Prepare the Lead-Screw-Nut-Sub-Assembly

Parts/Sub-Assembly required:

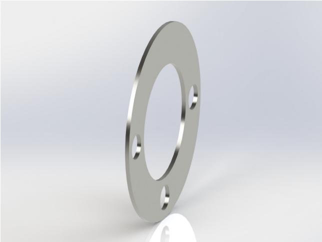
- Lead-screw-nut
- PS-Nut-Connect
- M6 Nut (2x)
- Lead-screw-nut Bolts M6x16 (2x)

Tools/Supplies Required

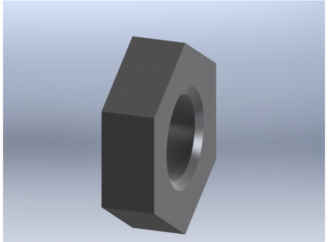
- M6 Wrench (10 [mm])
- Screwdriver



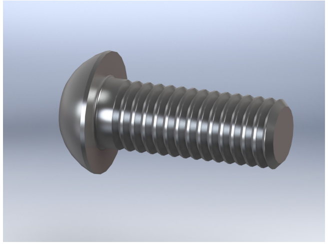
Lead-screw-nut



PS-Nut-Connect

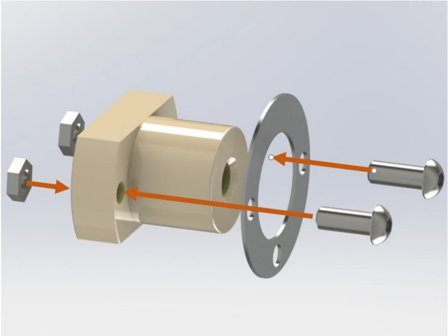


M6 Nut (2x)

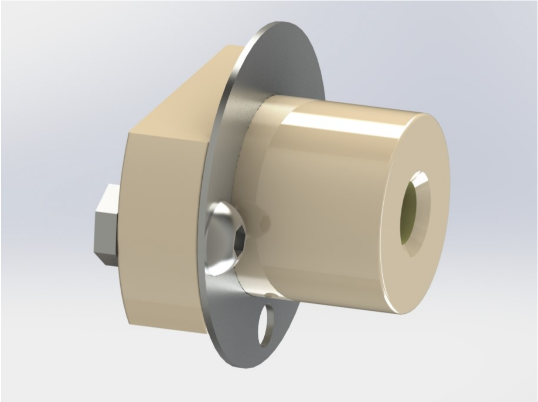


Lead-screw-nut Bolts M6x16 (2x)

1. Place the nut-connect over the lead-screw-nut and attach together using the lead-nut bolts and M6 nuts
   - Tighten to ≈6 [Nm]



Output Step 3



Lead-Screw-Nut Sub-Assembly

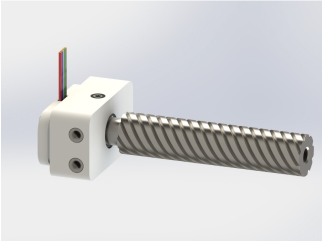
Step 4: Create the PS-Actuator

Parts/Sub-Assembly required:

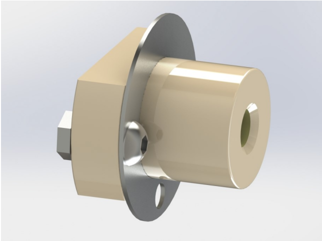
- Lead-Screw Sub-Assembly
- Lead-Screw-Nut Sub-Assembly
- PS General Actuator
- M6 nut (2x)

Tools/Supplies Required

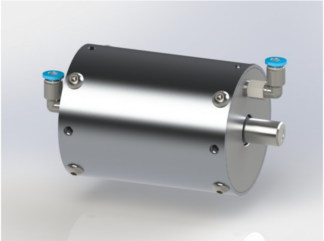
- Lint-free Cloth
- M6 Wrench (10 [mm])



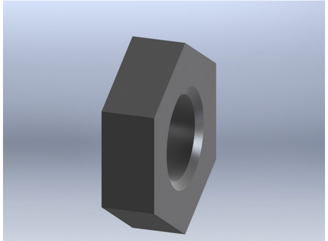
Lead-Screw Sub-Assembly



Lead-Screw-Nut Sub-Assembly)

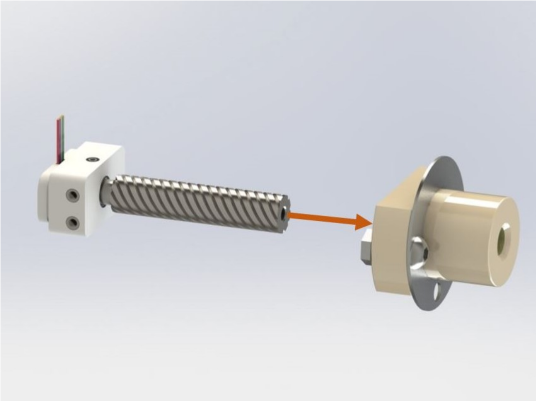


PS General Actuator

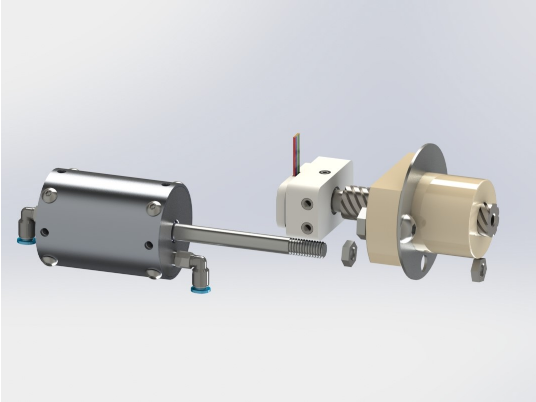


M6 nut (2x)

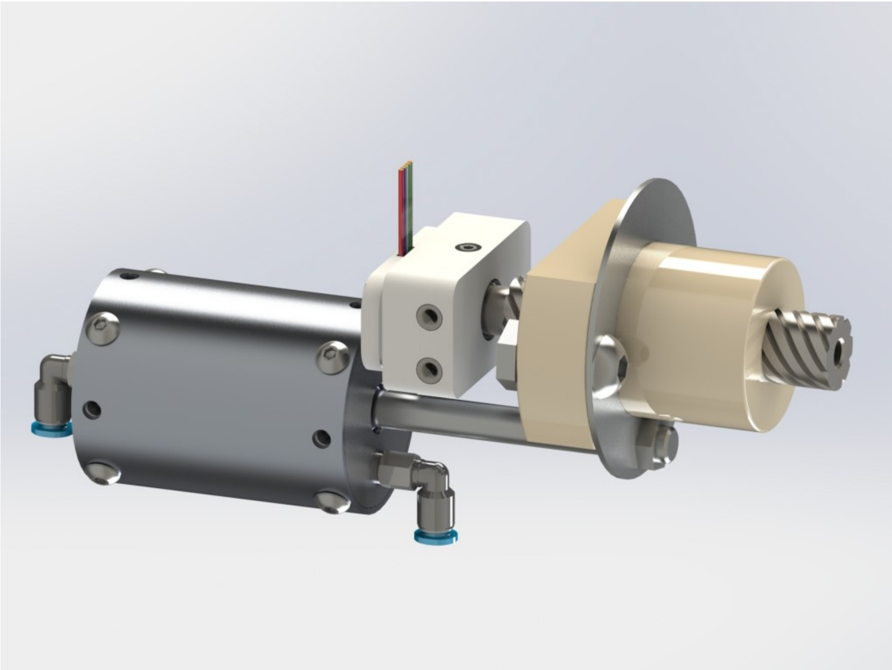
1. Slide the lead-screw-nut over the lead screw



1. Attach the nut-connect to the piston-rod using a M6 nut on either side

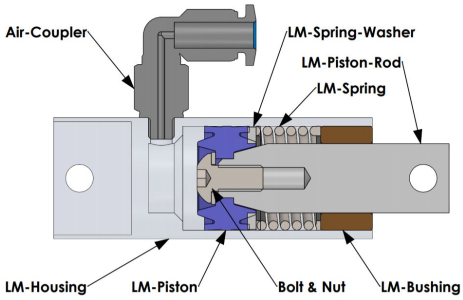


Output Step 4



Countersunk Bolts M3x8 (4x)

LM Actuator

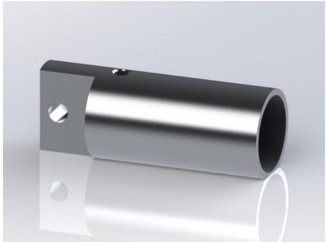


Parts/Sub-Assembly required:

- LM-Piston-Rod
- LM Piston
- M3x5 bolt & Washer
- LM-Housing
- LM-Spring-Washer
- Spring
- Bushing
- Push-in Fitting

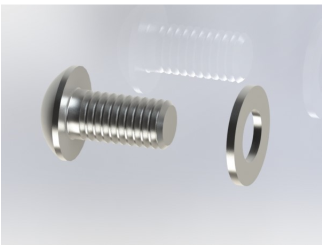
Tools/Supplies Required

- Hand Press
- Clamp
- Lint-free Cloth

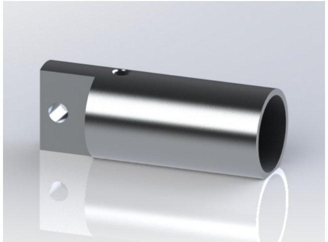


LM-Piston-Rod

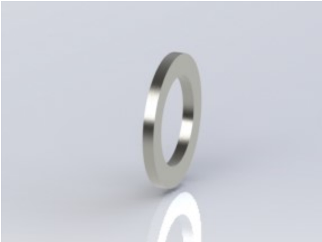
LM-Piston



M3x5 Bolt & Washer

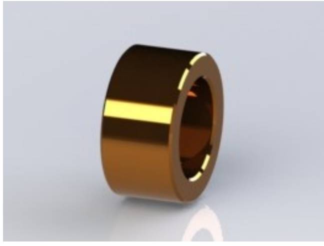


LM-Housing

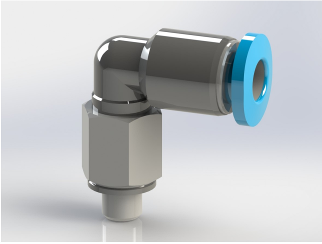


LM-Spring-Washer

Spring

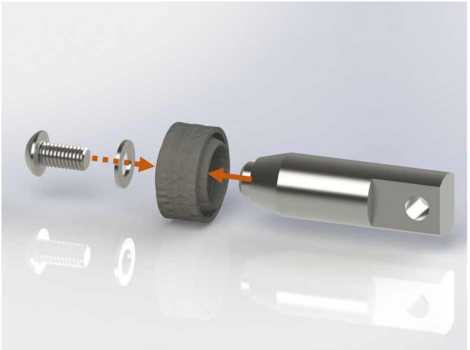


Rod-Bushing



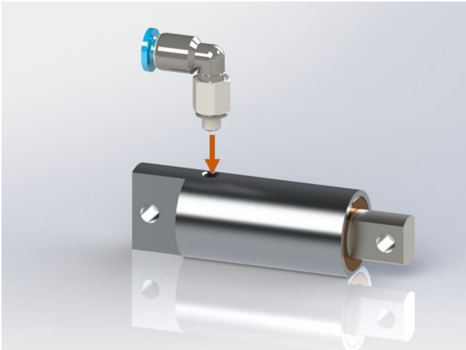
Push=in Fitting

1. Attach the piston-rod tot the piston using the M3x5 bolt.
    - Use the M3 washer
    - Tighten to ≈0.45 [Nm]
    - Use the clamp and lint-free cloth to hold the piston-rod in position if necessary

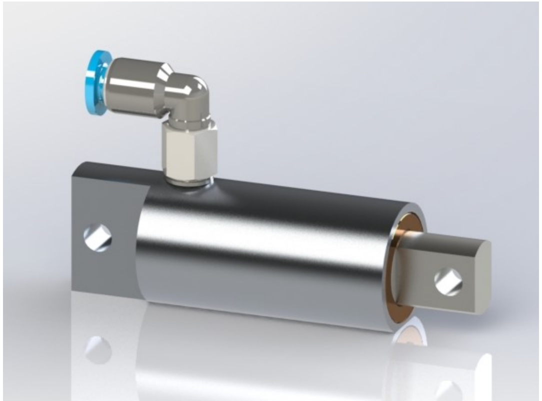


2. Place the spring-washer, spring and bushing over the piston rod. Press into the housing. Use the lint-free cloth to avoid damaging the bushing

3. Screw the push-in fitting to the cap
    - Use the M3 wrench to tighten at ≈0.45 [Nm]



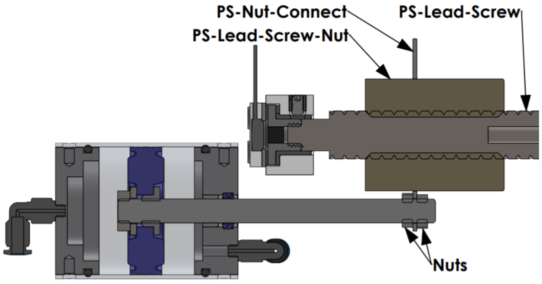
Output: LM Actuator



The LM Actuator

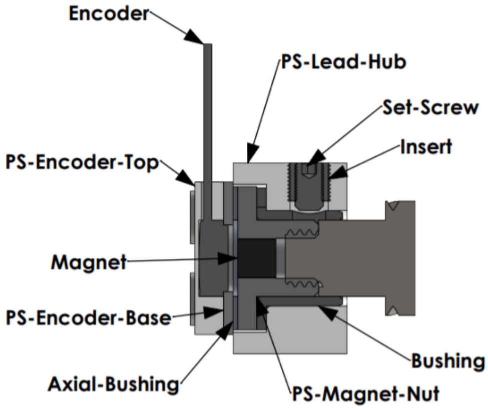
Frame

Perform Validation Test 1 before starting this assembly



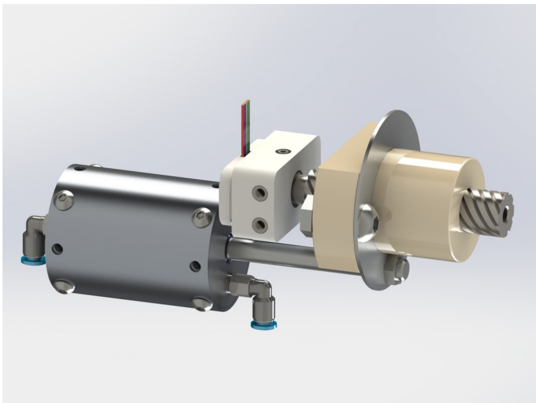
Parts/Sub-Assembly required:

- FR-Upperarm-Lock
- FR-Upperarm-Encoder
- FR-Lowerarm-Lock
- FR-Lowerarm-Encoder
- FR Connection Rod
- FR Encoder Shaft
- Bushing
- Flanged Bushing (3x)
- Inserts (12x)
- Shoulder Bolt (3x)
- M3x4 Bolt (8x)
- M6 Nut (2x)
- M3 Nut (8x)



Tools/Supplies Required

- Soldering Iron (with a sharp tip)
- Hand Press
- Clamp
- Lint-free Cloth



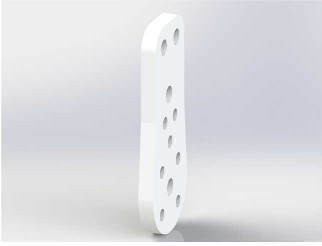
Step 1: Prepare the Upper Arm Sheets

Parts/Sub-Assembly required:

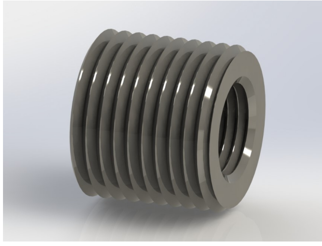
- FR-Upperarm-Lock
- Inserts (12x)
- Flanged Bushing (3x)
- FR-Brake Disc
- M3x10 Countersunk Bolts (4x)
- FR-Upperarm-Encoder
- Bushing

Tools/Supplies Required

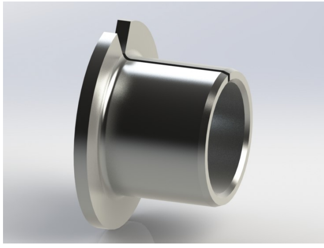
- Soldering Iron (with a sharp tip)
- Hand Press
- Lint-free cloth
- Screwdriver



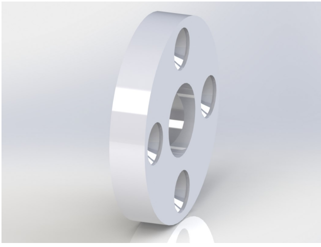
FR-Upperarm-Lock



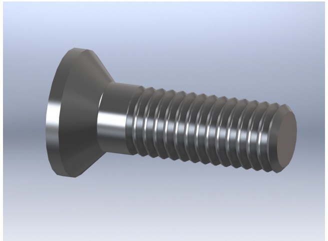
Insert (9x)



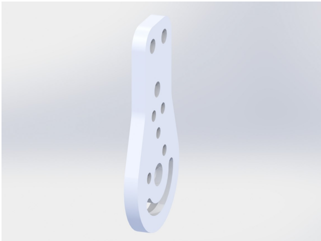
Flanged Bushing (3x)



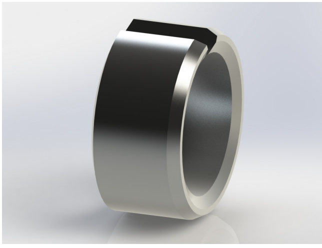
FR-Brake-Disc



M3x10 Countersunk Bolts (4x)



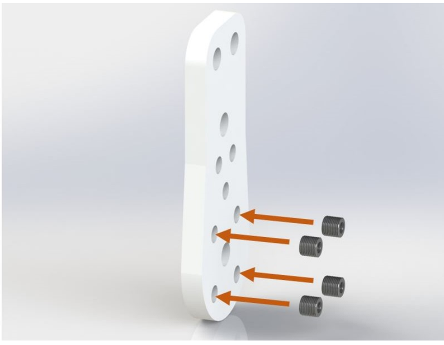
FR-Upperarm-Encoder

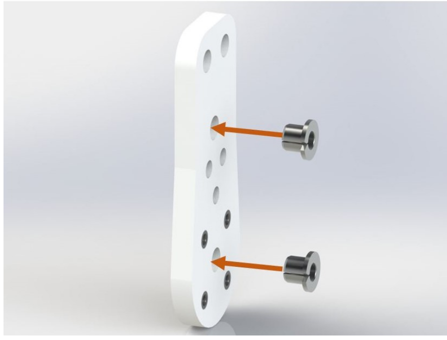


Bushing

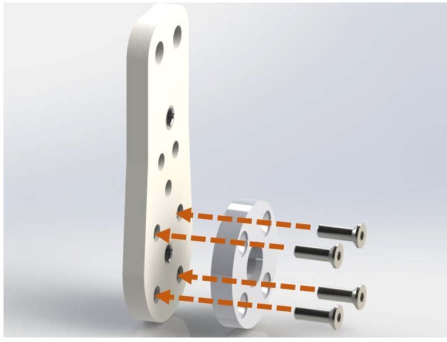
Upperarm-Lock Sub-Assembly

1. Use the soldering iron to heat the inserts and push them in the desired holes

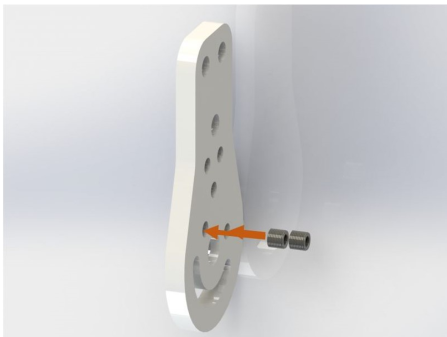




2. Press the flanged bushings in the desired holes. Use a lint-free cloth to avoid damaging the bushings.

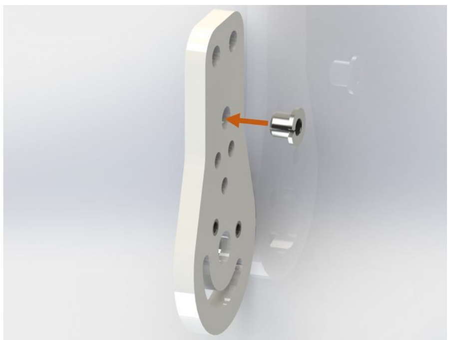


3. Attach the brake disc to the upperarm-lock, using the four M3 countersunk bolts.
   - Tighten at 0.45 [Nm]



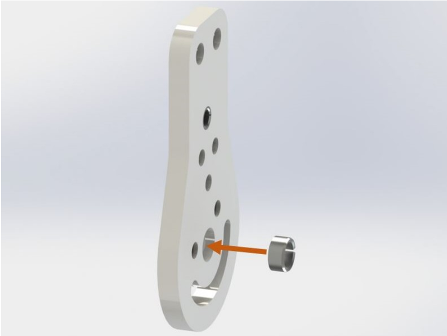
Upperarm-Encoder Sub-Assembly

1. Use the soldering iron to heat the inserts and push them in the desired holes

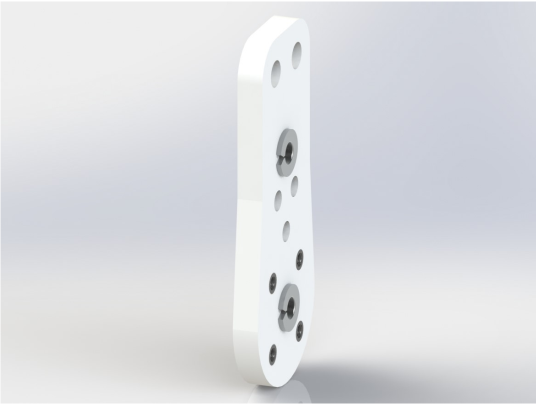


2. Press the flanged bushing in the desired hole. Use a lint-free cloth to avoid damaging the bushings.

3. Press the bushing in the desired hole. Use a lint-free cloth to avoid damaging the bushings.
    - Press the bushing in from the other side, compared to the flanged bushings. This is to ensure no obstruction of the encoder.



Output Step 1



Upperarm-Lock Sub-Assembly

Upperarm-Encoder Sub-Assembly

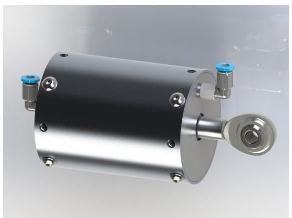
Step 2: Attach the FE Actuator

Parts/Sub-Assembly required:

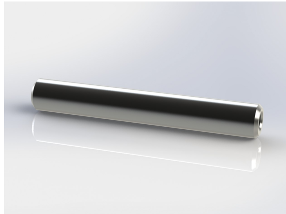
- FE-Actuator
- FR-Connection-Rod
- FR-Lowerarm-Encoder
- FR-Lowerarm-Lock
- M3x8 Low-Head bolts (2x)
- FR-Encoder-Shaft
- Magnet
- M6 Nut (2x)
- LM-Lever (2x)
- Upperarm-Lock Sub-Assembly
- Upperarm-Encoder Sub-Assembly
- M3 Shoulder-Bolt(4x)
- M3 Nut

Tools/Supplies Required

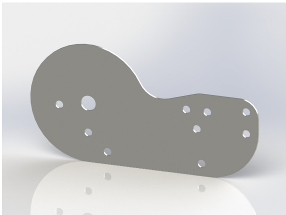
- Soldering Iron (with a sharp tip)
- Hand Press
  - ⌀ 15 [mm] cylinder
- M6 Wrench (10 [mm])



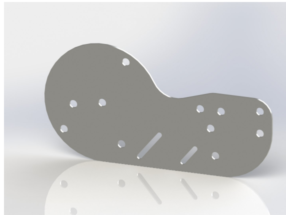
FE-Actuator



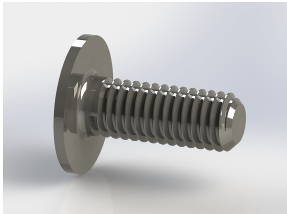
FR-Connection-Rod



FR-Lowerarm-Encoder

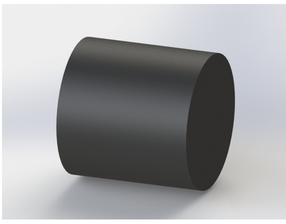


FR-Lowerarm-Lock

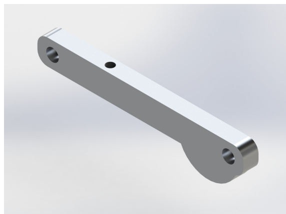


M3x8 Low-Head bolts (2x)

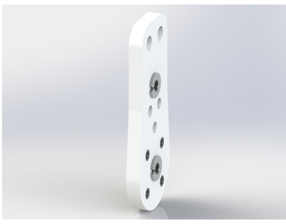
FR-Encoder-Shaft



Magnet

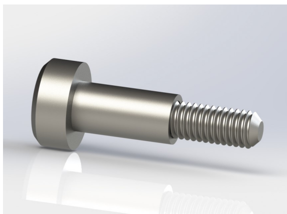


Lever

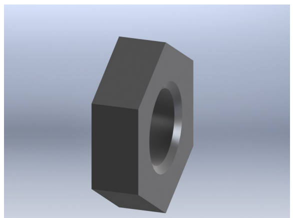


Upperarm-Lock Sub-Assembly

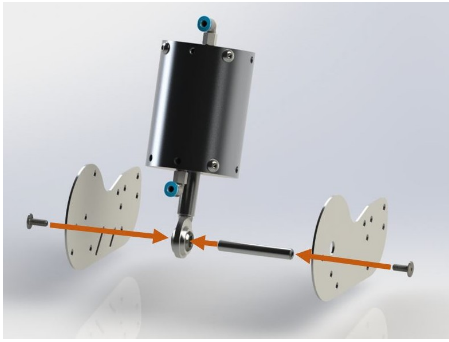
Upperarm-Encoder Sub-Assembly



M3 Shoulder-Bolt (4x)

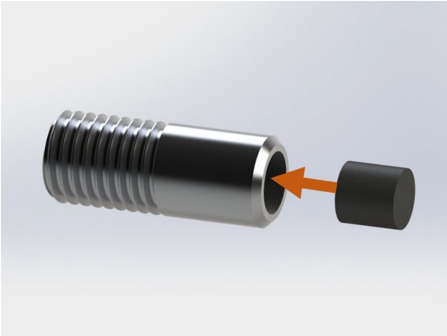


M3 Nut (4x)



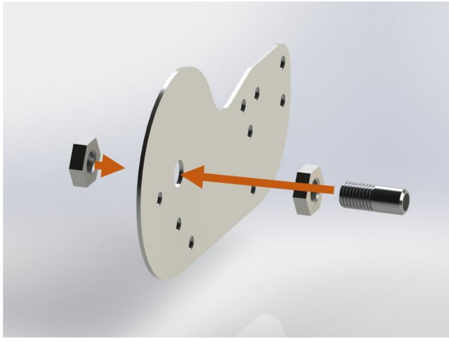
FE-Actuator, Connection 1

1. Slide the FR Connection-Rod through the rod-end and attach to the FR-Lowerarm-Lock and FR-Lowerarm-Encoder using the M3x8 Low-Head Bolts
    - Tighten at ≈0.45 [Nm]

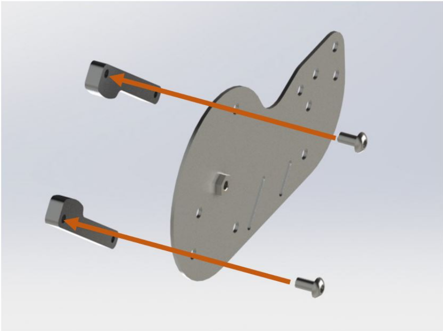


Implement the Encoder-Shaft

1. Press the magnet into the magnet-nut. Use the lint-free cloth to prevent damage.



2. Attach the encoder-shaft to the forearm-encoder, use an M6 nut on either side of the metal sheet
    - Tighten at ≈3.5 [Nm]

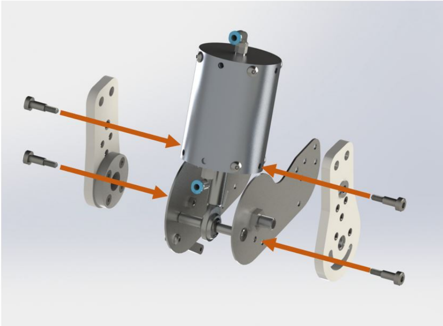


FE-Actuator, Connection 2

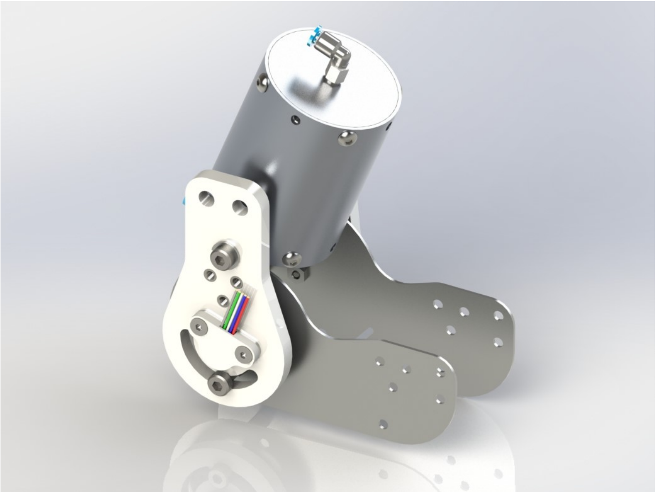
1. Attach the levers for the locking mechanism to the forearm frame sheet
    - Tighten at ≈0.45 [Nm]

FE-Actuator, Connection 2

1. Attach the upper arm sheets to the frame and FE-Actuator using the M3 Shoulder Bolts.
    - Tighten at ≈0.45 [Nm]



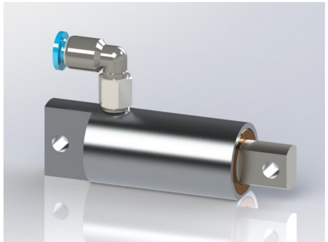
Output: Prototype 0.1



The Prototype 0.1

Step 3: Attach the LM Actuator

Prototype 0.1



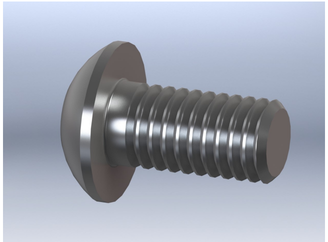
The LM Actuator

Parts/Sub-Assembly required:

- Prototype 0.1
- LM-Actuator
- M3x8 bolt (4x)
- Tie-wrap (2x)

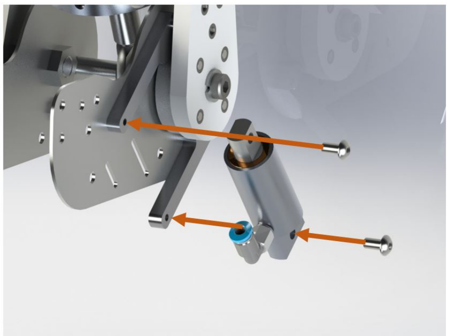
Tools/Supplies Required

- Screwdriver

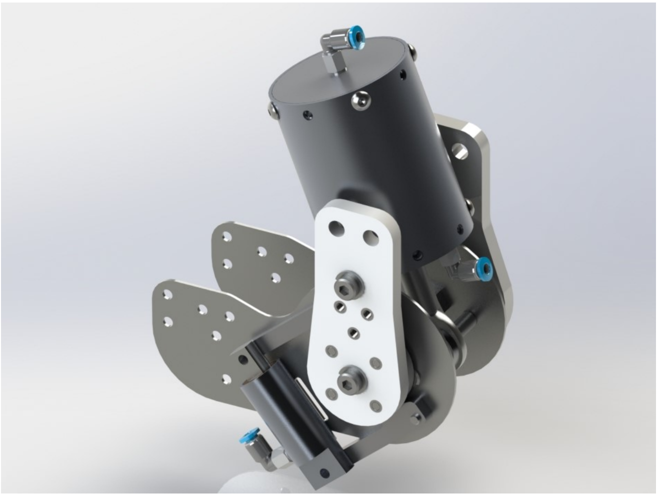


M3x8 bolt

1. Connect the LM-Actuator to the levers. To achieve this, the LM-Actuator must be pressurized > 12 [bar].
    - Tighten at ≈0.45 [Nm]



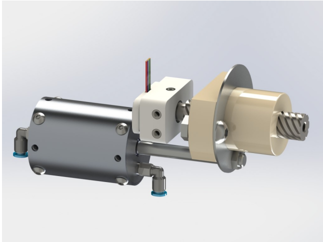
Output: Prototype 0.2



The Prototype 0.2

Step 4: Attach the PS Actuator

Prototype 0.2



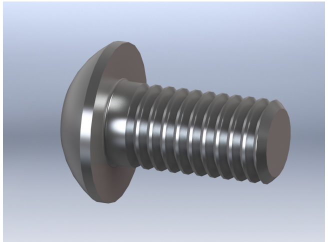
The PS-Acuator

Parts/Sub-Assembly required:

- Prototype 0.2
- PS-Actuator
- M3x8 bolt (8x)

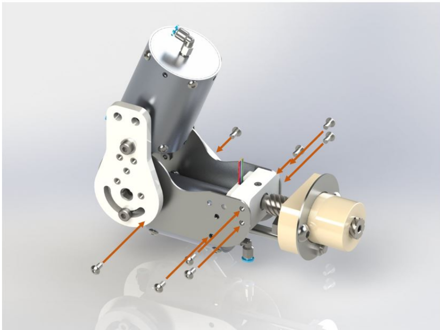
Tools/Supplies Required

- Screwdriver

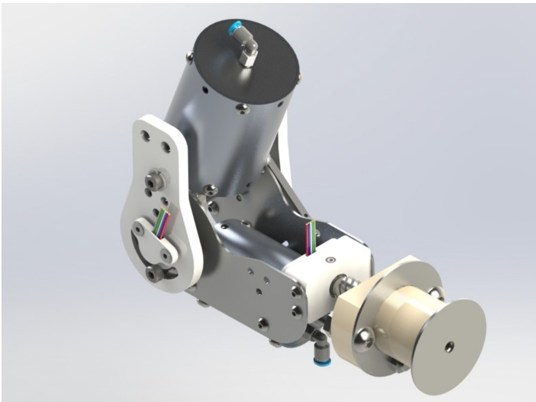


M3x8 Bolt

1. Connect the PS-Actuator to the frame.
   - Tighten at ≈0.45 [Nm]

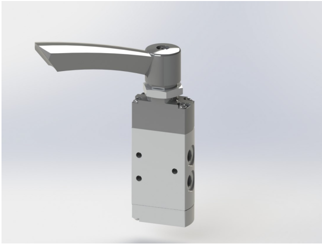


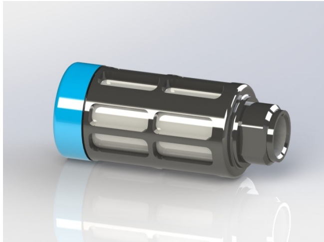
Output:Mechanical Prototype

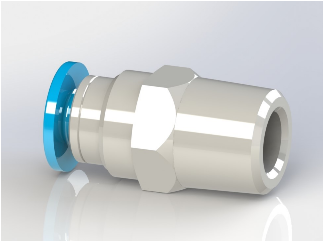


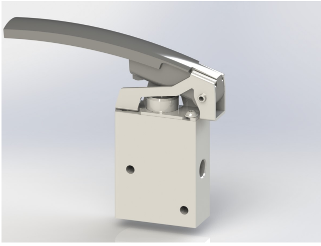
The Mechanical Prototype

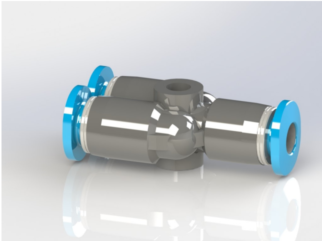
Control Circuit

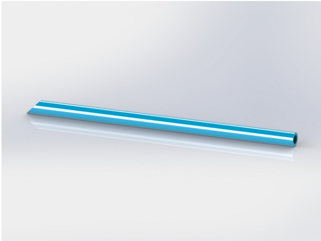

PS/FE-Valve (2x)


Exhaust Silencer


Push-In Fitting G1/8-4


LM Valve


Push-In Fitting Y (2x)


⌀ 4 [mm] Tube

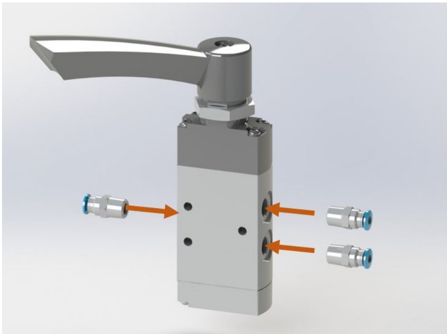
Parts/Sub-Assembly required:

- PS/FE-Valve (2x)
- Exhaust Silencer (5x)
- Push-In Fitting G1/8-4 (8x)
- LM Valve
- Push-In Fitting Y (2x)
- ⌀ 4 [mm] Tube

Tools/Supplies Required

-

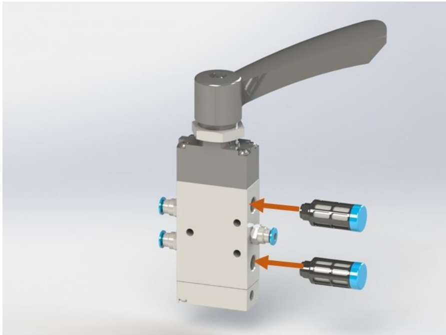
Step 1: Prepare the Valves

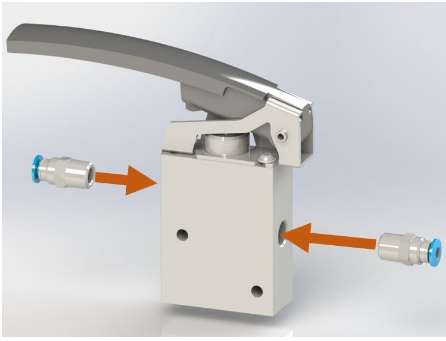


FE/PS-Valves

1. Screw in the Push-In Fitting G1/8-4 in the correct locations

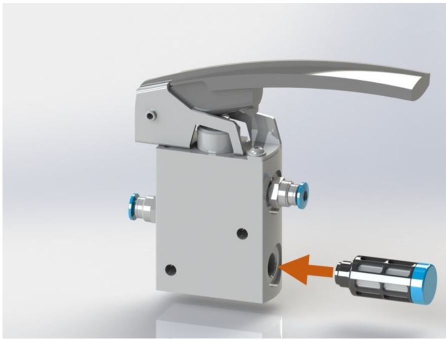
1. Screw in the Exhaust Silencer in the correct locations





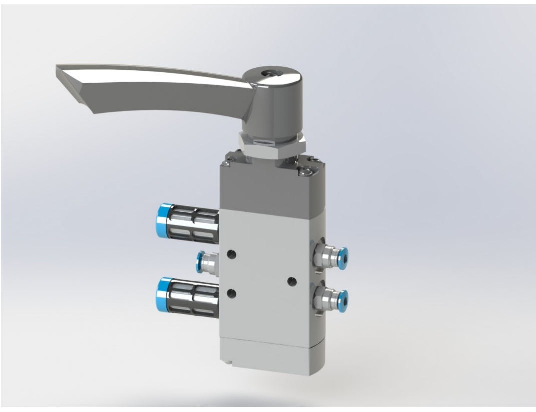
LM-Valve

1. Screw in the Push-In Fitting G1/8-4 in the correct locations

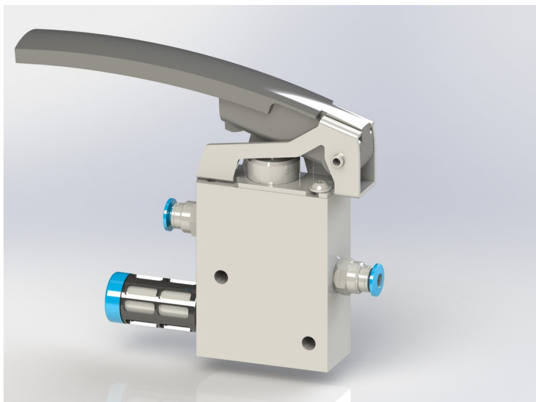


1. Screw in the Exhaust Silencer in the correct locations

Output



FE-PS-Valve Sub-Assembly

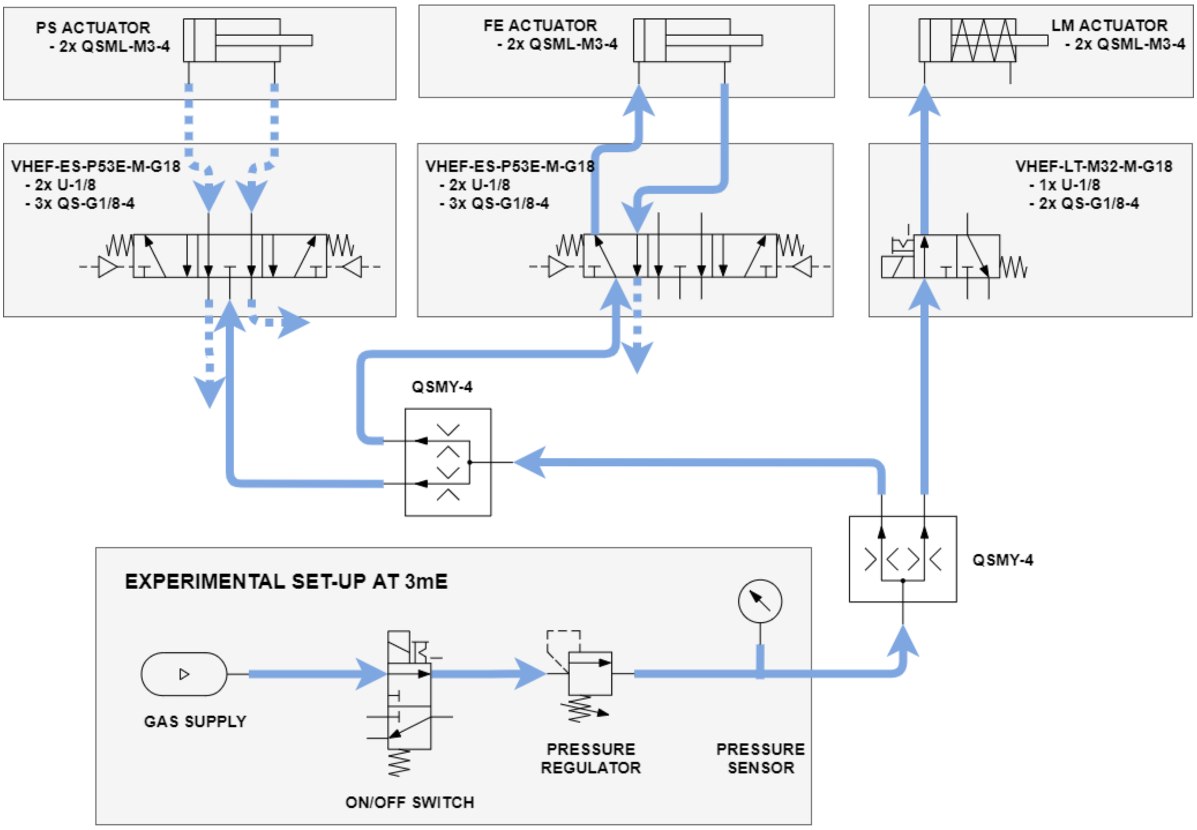


LM-Valve Sub-Assembly

Step 2: Attach the Tubes

FE/PS-Valves

1. Attach the tubes following the diagram shown below



The final designed pneumatic circuit

P

Test Protocol

Validation

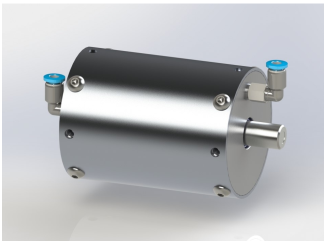
FE/PS-Actuators: Force Output & Stroke Time

Important Notes

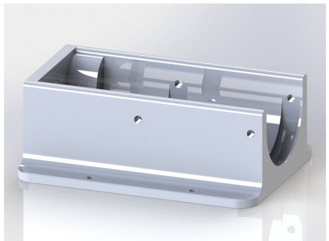
- **This test has to be performed before the end-effector is placed on the piston-rod**
- **Before starting the test, make sure the gas supply and control circuit are functioning and ready to use**

Parts/Sub-Assembly Required:

- FE Actuator, without end-effector
- PS Actuator, without end-effector
- FE Testing Frame
- PS Testing Frame
- M3x10 Bolt (8x)



Actuator without End-Effector



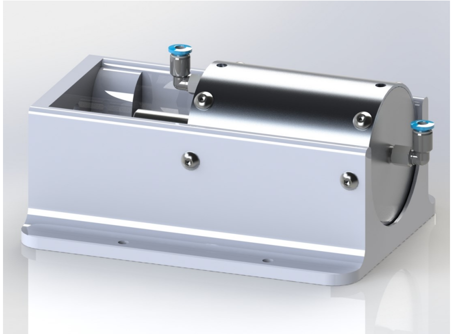
Testing Frame

Tools/Supplies/Sensors Required:

- Force Sensor: Futek LCM300 load-cell 2KN (500lb)
- Computer with LabVIEW
- Data Collection Unit
- Camera

Futek LCM300 loadcell 2KN

Prepare the Actuator and Test-Setup

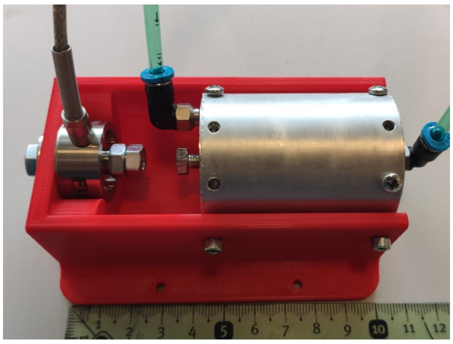


1. Prepare the actuator
   - **FE**: Screw the M5 bolt in the piston-rod
   - **PS**: Screw the M6 Nut on the piston rod
2. Place the actuator in the testing frame
   - Attach the actuator to the testing frame using the M3x10 bolts
3. Attach the push-in fittings to the valve

Setup Velocity Test

1. Set-up a camera system. Ensure that the actuator and valve are within the video frame.
   - Use a camera of which the fps (frames per second) is known

Perform Velocity Test

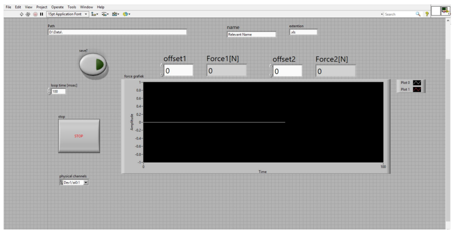


1. Increase the pressure to 1 [bar]
2. Start the recording
3. Use the valve to direct the flow to the actuator, the piston-rod will fully extend
4. Stop the recording
5. Repeat three times

Process Data Velocity Test

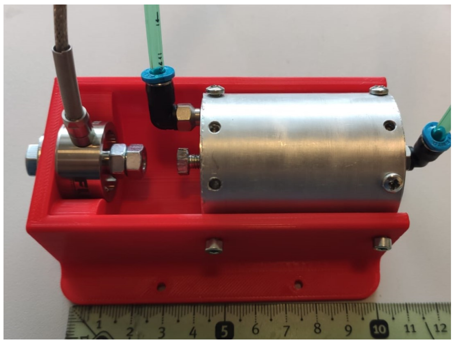
1. Read out the memory card and save the recordings
2. Count the frames required for a full extension, divide by the camera fps to find the stroke time

Setup Force Test



1. Connect the data collection unit to the computer
2. Connect the sensor to data collection unit
3. Start LabVIEW
4. Adjust the force offset
5. Screw the M6 Nut on the sensor threaded part,
6. Place the sensor in the testing frame and fasten using the M6 nut

Perform Force Test



1. Slowly increase the pressure until the end-effector gently touches the sensor
2. Increase the pressure incrementally with 1 [bar] per step.
3. Use the valve to direct the flow to the actuator
4. Read out the force for each step

   If the force exceeds 1700 [N], do not continue increasing the pressure!

5. Repeat the experiment 3 times

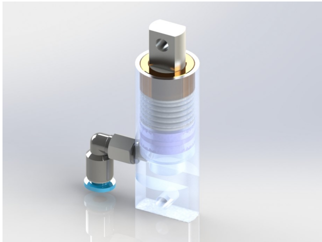
LM Actuator: Force Output

Important Notes

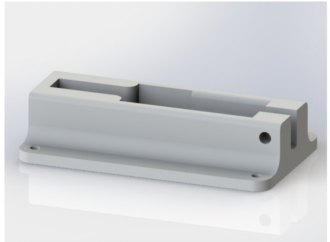
- **This test has to be performed before implementing the LM-Actuator on the frame**
- **The cable of the sensor is fragile, pay attention when working with the sensor**

Parts/Sub-Assembly Required:

- LM Actuator
- LM Test Frame
- M3x10 Bolt
- M3x12 Bolt
- M3 Nut



LM Actuator



LM Test Frame

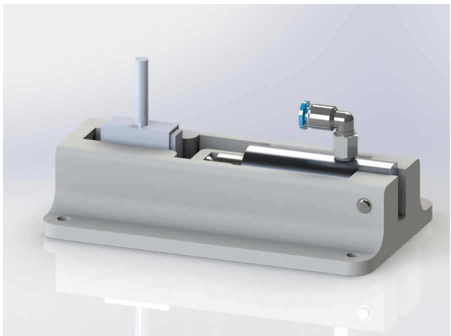
Tools/Supplies/Sensors Required:

- Mini Loadcell / krachtopnemer S beam 111N
- Computer with LabVIEW
- Data Collection Unit

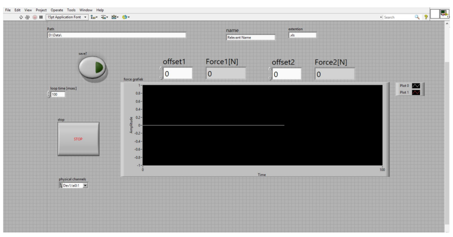
Mini Loadcell / krachtopnemer S beam 111N

Assemble the Test-Setup

1. Place the LM actuator in the LM test frame
2. Use the M3x12 bolt and the M3 nut to attach the LM housing to the LM test frame
3. Attach an M3x12 bolt to the sensor
4. Use the M3x12 bolt and the M3 nut to attach the LM housing to the LM test frame

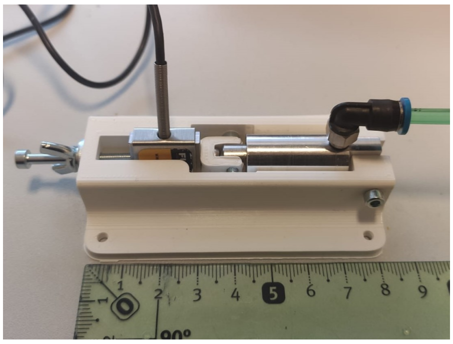


5. Connect the data collection unit to the computer
6. Connect the sensor to data collection unit
7. Start LabVIEW
8. Adjust the force offset



Perform Test

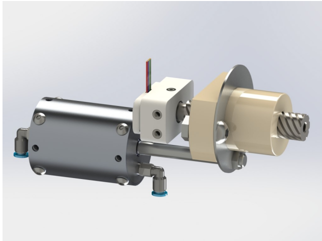
1. Increase the pressure incrementally with 1 [bar] per step.
2. Use the valve to direct the flow to the actuator
3. Read out the force for each step
4. Repeat the experiment 3 times



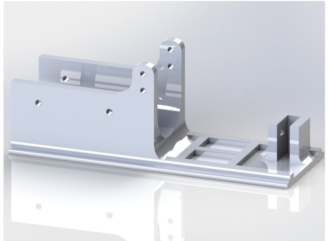
Final Test

PS Actuator: Torque Output

This test has to be performed before attaching the PS-Actuator to the frame



PS Actuator



LM Test Frame

Parts/Sub-Assembly Required:

- PS Actuator
- PS Testing Frame
- M3x8 Bolt

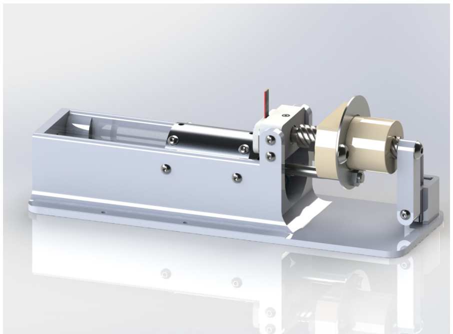
Tools/Supplies/Sensors Required:

- Mini Loadcell / krachtopnemer S beam 111N

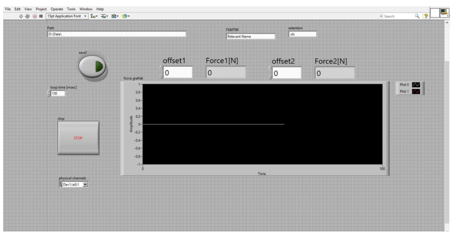
Mini Loadcell / krachtopnemer S beam 111N

Assemble the Test-Setup

1. Place the actuator in the testing frame
2. Attach the actuator to the testing frame using the M3x10 bolts
3. Attach the push-in fittings to the valve
4. Place the sensor in the testing frame and fasten using the M3x8 bolt



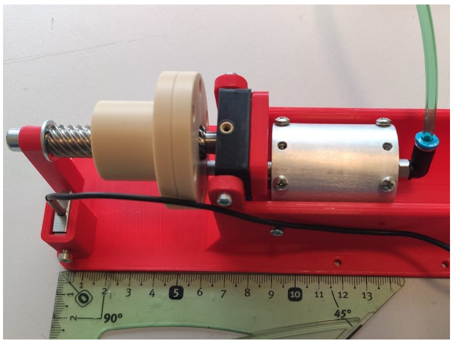
5. Connect the data collection unit to the computer
6. Connect the sensor to data collection unit
7. Start LabVIEW
8. Adjust the force offset



PS Torque Test

Important Notes

- **Ensure that the pressure is set at 0 [bar]**



1. Increase the pressure incrementally with 1 [bar] per step.
2. Use the valve to direct the flow to the actuator
3. Read out the force for each step
4. Repeat the experiment 3 times

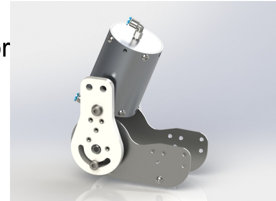
Lifting & Holding Payload

Parts/Sub-Assembly Required:

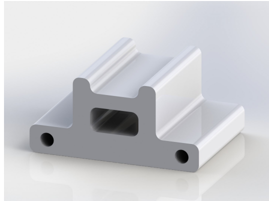
- Prototype without the PS-actuator implemented
- Stability Block
- Weight Connector
- Prototype Holder

Tools/Supplies/Sensors Required:

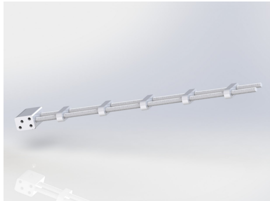
- Weight Holder (Hook)
- Weights
  - 0.5 [kg] (2x)
  - 1.25 [kg] (1x)
  - 2.5 [kg] (1x)
  - 5 [kg] (1x)



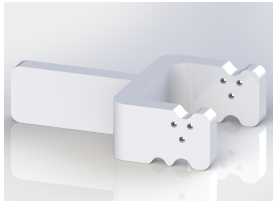
Prototype without PS Actuator



Stability Block



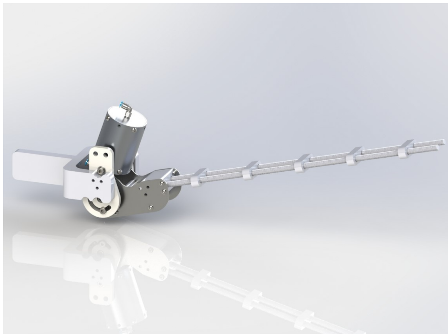
Weight Holder



rototype Holder

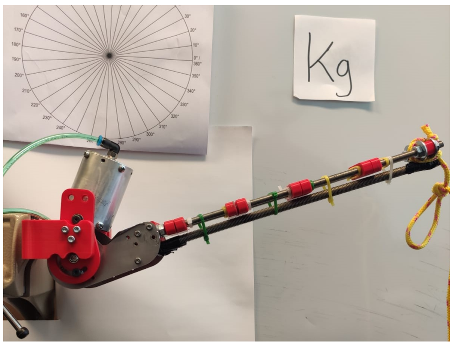
Assemble the Test-Setup

1. Attach the stability block to the forearm sheets
   - As the PS-Actuator is not implemented, the stability of the frame is compromised. This is why a ”stability block” is implemented.
2. Connect the weight connector to the frame
3. Connect the prototype holder to the upper arm sheets
4. Place the prototype holder in a vice
5. Setup the camera system



Perform the Payload Test

1. Increase the pressure to 12 [bar]
2. Increase the weight
3. Use the valve to direct the flow to the actuator
4. Take a picture
5. Repeat steps 2-4 until all desired weights are tested
6. Repeat the test three times



Process the Payload Test Data

1. Use an image processing tool, i.e. InkScape, to determine the angle of flexion

Q

Module Evaluation

Table Q.1: An overview of the most prominent "competitors": existing prostheses. Some prostheses cannot achieve pronation/supination. In these cases, a grey "x" is used to show that there is no data to be represented. If information should be presented, such as FE RoM, but is missing from the source a yellow "?" is used to illustrate missing information

Module	**Actuation**	**Locking Mechanism**	**Frame**			**Control**
Sub-Module	-	-	**Structure**	**Covers**	**Fitting**	-
Must Have	Flexion Extension	-	Constrain Unwanted DoF	-	-	Generate Required Output
	Pronation Supination	-	Withstand Forces and Moments	-	-	Sense User Input
Should Have	-	Be Lockable [1]	-	Protect Components	-	Detect Angles
	-	Have a Free-Swing Mode	House Components	-	-	Provide Feedback
	-	-	-	-	Fit to User Arm	Have a Portable Energy Supply
Could Have	-	-	Connect to Terminal Device	-	-	-
Wont Have	-	-	-	-	Have an Adjustable Fitting	Show Power Level

[1] Is a summary of several sub-functions, see Table 3.1

R

Embodiment Design Evaluation

Actuator Caps

The the material used for the caps can be from aluminum to a plastic, such as POM. The density would be decreased from 2.9 [kg/m^3] to 1.4 [kg/m^3], a reduction of 52 [%]. If all caps were to be changed to POM, the total reduction in weight of the prosthesis would be about 45 [g]. To prevent failure of a plastic due to threading, inserts will be used. Eight of these will weigh ≈2 [g] combined: This does not significantly reduce the weight loss of using plastic caps.

Table R.1: An overview of possible weight loss by changing the material of the caps to POM

Part	Aluminum Mass [g]	Plastic Mass [g]	Loss [g]
FE Cap	30	14.4	15.6
FE End-Cap	22	10.6	11.4
PS Cap	19	9.1	9.9
PS End-Cap	15	7.2	7.8

PS Lead

After revising the lead-screw selection, some mistakes things were noted. The equation to calculate did not include the friction caused by the thread angle (α) nor did it include the friction caused by the the collar (T_c), see Equation (R.1) and Equation (R.2). Where the consequences of the thread angle frictions are relatively small, the collar friction can reduce the output friction by a significant amount. The updated torques are shown in Appendix H. Despite the combination of **12x25 [mm], ⌀ 32 [mm]**, achieving only 1.8 [Nm], it still would have been selected, were it not for one thing. The 16 [mm] diameter lead-screw is available in aluminum, reducing the complete mass (including the nut) by more than 50 %. The additional 5 [mm] stroke-length required because of the larger pitch, leads to a longer required PS-Housing and PS-Piston-Rod, leading to an added mass of 1.4 [g] and 1.1 [g] respectively. The new lead-screw would increase the torque by 1.2 [Nm] an decrease the mass of the actuator by about 20 [g].

$$T_{PS} = \frac{F * d * \eta}{2} \left(\frac{\mu * \sec\alpha - \tan\lambda}{1 + \mu * \sec\alpha * \tan\lambda}\right) - T_c \tag{R.1}$$

R.1: An updated equation used to determine the PS torque. T_{PS} is the torque of pronation/supination [Nm], F is the actuator force [N], d is the lead-screw diameter [m], η is the efficiency [-], μ is the coefficient of friction between the lead-screw and nut [-], α is half the thread angle [°], λ is the lead angle [°], and T_c is the friction torque caused by the collar [Nm]

Table R.2: An updated version of Table H.4. The most relevant information of the selection of lead-screws to be used for PS torque generation (https://www.igus.nl/)

Lead Diameter	⌀ **10 [mm]**	⌀ **12 [mm]**	⌀ **14 [mm]**	⌀ **16 [mm]**
Allowed Force [N]	780	1405	1440	1669
Material Lead-Screw	Steel	Steel	Steel	Steel
	-	-	-	Aluminum
Mass Steel Lead-screw [g/mm]	0.62	0.89	1.22	1.59
Mass Aluminum Lead-screw [g/mm]	-	-	-	0.54
Mass Nut [g]	23.7	39.2	37.2	34.6
Diameter Nut [mm]	42	48	48	48
Length Nut [mm]	25	35	35	35
Possible Thread Pitch [mm]	12	5	25	35
	25	25	30	-
	50	-	40.6	-

$$T_c = \frac{F * d_c * \mu_c}{2} \tag{R.2}$$

R.2: The equation used to determine the friction torque in the collar. T_c is the friction torque in the collar [Nm], F is the force on the collar [N]. d_c is the collar diameter, and μ_c is the coefficient of friction between the collar and the lead-screw [-]

Table R.3: The invariant parameters used for the calculations of the lead-screw selection

Symbol	**Value**	
μ	0.12	
μ_c	0.15	
ϕ	8.5	°
α	14.5	°
L_{Buffer}	4	mm
L_{Collar}	10	mm
d_c	15	mm

...

Table R.4: An updated version of Appendix H. The table used to determine which combination of piston diameter and lead-screw to use for the generation of torque for pronation/supination. The "mass" is the mass of the lead-screw for the required length + the mass of the nut.

Piston Diameter					**⌀ 25 [mm]**	**⌀ 32 [mm]**	**⌀ 40 [mm]**
Actuator Force [N]					589	965	1508
Lead Diameter: 10 [mm]					Pass	Fail	Fail
Lead Diameter: 12 [mm]					Pass	Pass	Fail
Lead Diameter: 14 [mm]					Pass	Pass	Fail
Lead Diameter: 16 [mm]					Pass	Pass	Pass
$M_{Friction}$ [Nm]					0.66	1.08	1.70
Diameter x Pitch	λ	η	**Length**	**Mass**	**Torque**		
[mm]	[°]	[%]	[mm]	[g]	[Nm]		
10x12	20.9	72.6	49.0	54.1	0.1	-	-
10x25	38.5	78.6	55.5	58.1	1.1	-	-
10x50	57.9	75.2	68.0	65.9	2.9	-	-
12x5	7.6	51.7	55.5	88.6	<0	<0	-
12x25	33.6	77.9	65.5	97.5	1.1	1.8	-
14x25	29.3	76.9	65.5	116.1	1.0	1.7	-
14x30	34.3	78.1	68.0	120.2	1.5	2.4	-
14x40.6	42.7	78.7	73.3	126.6	2.3	3.8	-
16x35	34.8	78.2	70.5	72.7	1.8	3.0	4.7

Electronic & Pneumatic Circuits

Table R.5: An overview of weight reduction electrically controlled valves would lead to

Type	**Control**	**Weight [g]**	**Flow Rate [L/min]**	**Code**
5/3	Selector, Sideways	235	530	**VHEF-ES-P53E-M-G18**
5/3	**Solenoid**	55	210	VUVG-L10-P53E-ZT-M5-1P3
3/2	Finger Lever	156	750	**VHEF-LT-M32-M-G18**
5/3	**Solenoid**	55	210	VUVG-L10-P53E-ZT-M5-1P3

Gas Type

Table R.6: An estimation of the gas used for flexion/extension and pronation/supination

	Molar Mass [g/mole]	Pressure [MPa]	Universal Gas Constant [J/K*mole]	Temperature [K]	Volume [ml]	Gas Used [g]	Reduction Compared to CO_2 [%]
				CO_2			
FE	44	1.2	8.314	293	31	0.67	-
PS	44	1.2	8.314	293	10	0.21	-
				70% H_2O_2			
FE	37	1.2	8.314	505	31	0.33	51
PS	37	1.2	8.314	505	10	0.11	47
				80% H_2O_2			
FE	36	1.2	8.314	760	31	0.21	68
PS	36	1.2	8.314	760	10	0.07	67

List of Figures

List of Tables