

École polytechnique de Louvain

Lower-limb prosthesis - Design of the inversion-eversion degree of freedom for active ankle foot prosthesis

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Abstract

In Belgium, approximately a thousand people suffer from major lower-limb amputation every year. These people are affected by the loss of mobility, self-esteem and quality of life caused by amputation. To enable them to regain mobility and quality of life they are fitted with lower-limb prostheses. The majority of prostheses focus on the main degree-of-freedom of walking.

The inversion-eversion degree-of-freedom is the movement that enables the ankle to tilt in the frontal plane. This movement allows the foot to remain flat on sloping or unstable ground and therefore ensures stability. It is also involved in everyday movements such as sidestep cutting, which are steps taken to avoid an obstacle without rotating the body. This degree of freedom is still only very rarely included in current ankle-foot prostheses.

In the present work, the design of a module that can be fixed to ankle-foot prostheses and add the inversion-eversion degree-of-freedom is performed. The design methodology led to a weight activated passive solution. The aim is to allow the ankle to adjust to the slope of the ground when the weight of the amputee does not apply on it, therefore changing the neutral position of the mechanism. Once the weight acts on the prosthesis, a mechanical impedance ensures stability through a nylon rope via a locking system. The deformation of the nylon rope allows the inversion-eversion of the ankle.

Finite elements methods with a CAD software were used to quantify the deformation and the structural strength of the different parts of the mechanism.

Abbreviations

Meaning	Abbreviation
Dorsiflexion/ Plantarflexion	DP
Inversion/eversion	IE
Degree-of-freedom	DoF
Range of motion	ROM
Series elastic actuator	SEA
Transfemoral	TF
Transtibial	TT
Computer aided design	CAD
Finite elements methods	FEM

Figure 1: List of abbreviations used

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Chapter 1

Introduction

A few millions years ago, the evolution led hominids to bipedal locomotion. Instead of crawling, our ancestors started walking only on their feet releasing their hands for other tasks. Nowadays, this specific task of walking seems completely natural and intuitive. It is not the case for everyone. For example, a toddler has to learn how to walk and his bones and muscles must grow enough to support his weight. It generally takes him two years before being able to walk. The length of this learning process reflects the complexity of such a task and the adaptability to the environment needed to perform it smoothly.

For people suffering from lower limb amputations, their ability to walk is heavily impacted and therefore strongly affecting their quality of life (QoL). They suffer from a reduced mobility, pain, unemployment and altered self-image. The QoL must be assessed to guide the rehabilitation but it is difficult as many parameters are involved. It is illustrated in a review made by Avery et al. [1] in which they investigate the patient reported outcome measures to assess the impact of lower limb amputations caused by peripheral arterial disease. As shown in Figure 1.1, the aspects to be taken into account are varied and complex.

The two main causes of amputations are traffic accident (traumatic amputation) and peripheral arterial diseases or diabetes[2]. Andrea Icks et al. studied the rate of lower limb amputation in people with and without diabetes in Belgium between 2009 and 2013 [3]. They used data from the Belgian national health insurance funds and showed that approximately a thousand people suffer from lower limb amputation each year. Even if the trend for the amputation resulting from diabetes is decreasing, the number of amputations for other reasons remains roughly constant as shown in Figure 1.2. It is therefore necessary to develop technological solutions to enable the rehabilitation of these people.

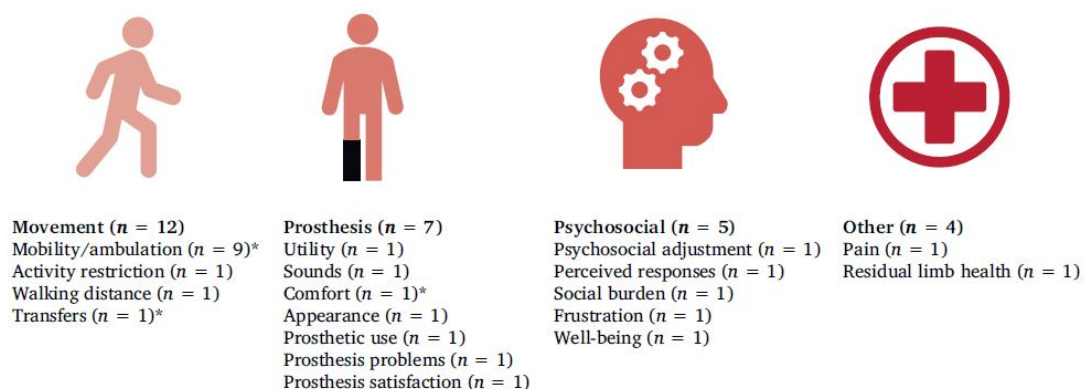


Figure 1.1: Domains of amputation specific patient reported outcome measures (PROMs) and their frequency of measurement in included studies (n=11). Domains in bold are only applicable to prosthesis users. *Domains reported in unidimensional PROMs. [1]

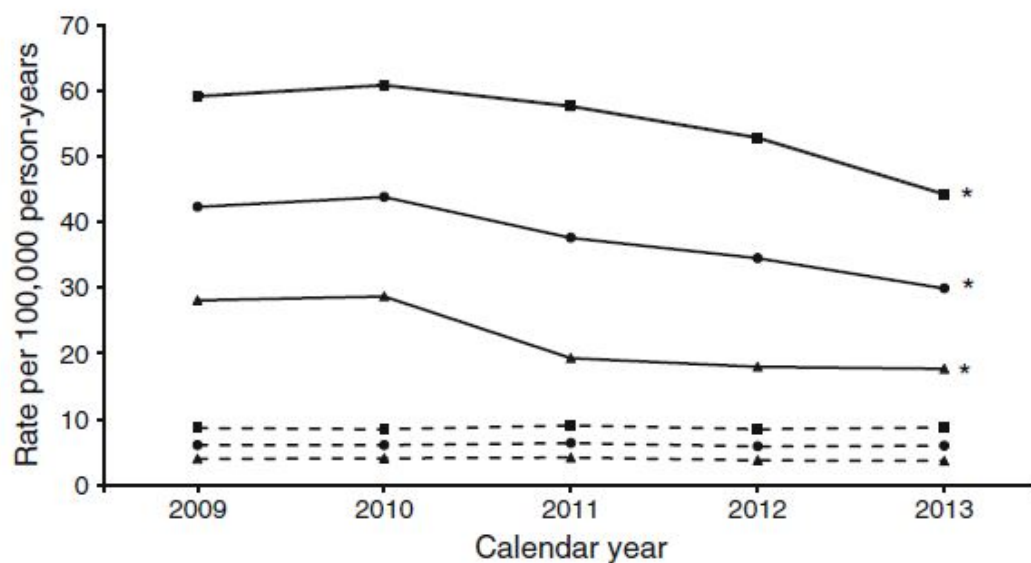


Figure 1.2: Time trend of age- and sex-standardised major amputation rate. Solid lines, people with diabetes; dashed lines, people without diabetes; circles, men and women; squares, men; triangles women. [3]

To help amputees regain their independence and improve their quality of life, lower limb prostheses were developed. Rudimentary at the beginning, they have become more and more complex thanks to scientific advances. Initially, the main concern was to enable the person to walk again thus focusing on the main degree-of-freedom (DoF) of the joints. Subsequently, stability and other issues have driven research towards the other degrees-of-freedom of the lower limb especially for the ankle.

The present thesis focuses on the design of a mechanism to add the frontal DoF to active ankle foot prosthesis. This movement is called inversion/eversion and allows the ankle to tilt laterally. This is particularly useful when walking on uneven ground. The foot adapts to the inclination of the ground, thus improving the patient's stability. Furthermore, it is implied in daily life movements such as changing his direction while walking to avoid an obstacle. For the moment, very few prototypes of ankle-foot prosthesis including this DoF exist.

The first section is a review of the current state of the art for ankle-foot prostheses. This includes a brief passage on the history of prosthesis. Then the biomechanics of a sound ankle is explained and finally the evolution of ankle-foot prosthesis towards prototypes with two degrees-of-freedom is presented.

The second section describes the methodologies used for the design process and the project management. The classical mechanical design Pahl and Beitz procedure was selected for the design and agile methods were implemented for the project management.

The third section consists in the design of the mechanism. After having established the specifications, the solution is constructed by following the chosen design procedure. Three different possible solutions are obtained and embodiment design is performed.

Numerical simulations are presented in the fourth section. The CAD software Solidworks is used to assess the feasibility of the solutions. The structural strength is also tested to verify the material selection hypothesis.

In the fifth section, a final choice is made between the three solutions and the final design is described.

Finally, the limitations of the current design are discussed and perspectives for further improvements are suggested.

Chapter 2

State of the art

This chapter will present some key elements in the history of prosthesis. Then, the evolution of ankle-foot prosthesis is overviewed. Finally, a specific focus on prototypes having two degrees-of-freedom is provided.

2.1 History of prosthesis

Two decades ago, in the Sheikh 'Abd el-Qurna necropolis in Egypt, archeologists discovered what could be the oldest prosthesis ever found. It was estimated to be 3,000 years old [4]. This wooden and leather toe prosthesis probably helped with the aesthetic aspect but also for the balance.(Figure 2.1).



Figure 2.1: Oldest prosthesis ever discovered dated to 3,000 years[4]



Figure 2.2: Artificial limb found in Capuan tomb in 1858 and dated to about 300 B.C [5]

Historically, wars have always occurred resulting in a large number of injuries and amputations among the survivors. With the perpetual evolution of techniques, the use of wood in prostheses was neglected and replaced by different materials. It is the case for an artificial limb made in copper, shown in Figure 2.2. It was found in a Capuan tomb in 1858 and dated to 300 B.C.. It was probably made for a Roman soldier who lost his leg during battles [5].

Ambroise Paré (1510-1590)[6] played a major role in this development since he was the official royal surgeon in France, improving amputation techniques by perfecting the ligation of vessels[7] and designing functional prosthetic limbs based on his knowledge about the anatomy. This earned him being called *the father of modern surgery* [8].

James Edward Hanger was a 18 years old engineering student in 1861 when the civil war began in America. Only two days after being enrolled, a cannon ball hit his leg and caused its amputation. Once he got home, he started working on a prosthesis to replace the lost limb. He ended up creating an innovative articulated knee, shown in Figure 2.3 [9].

The first world war had an important impact in the development of prosthetic devices since the demand at that time was significant. The wounded soldiers often suffered amputation as a result of infections and gangrene and the lack of drug during this pre-antibiotic era [10]. There was a real need to rehabilitate the veterans, the poor designs and materials at that time were an obstacle to the daily use of prosthetic devices. Many associations were founded to help the amputees like the American Orthotic & Prosthetic Association [11].

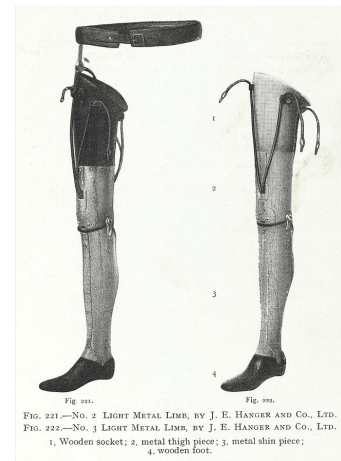


Figure 2.3: Innovative articulated knee designed by Hanger in 1861[9]

During and after the second world war, advances in medicine and technologies impacted the prosthetic technology. For example, the penicillin discovered by Alexander Fleming in 1929 started to be manufactured and used in military hospitals in 1942. The use of antibiotics reduced the risk of infection after amputation [12].

Nowadays, the major causes for lower limb amputation are diseases like diabetes or trauma from traffic accident[2]. In the case of diabetes, foot ulcerations usually result from peripheral arterial diseases that hinder the cicatrization and cause gangrene or infection that spreads in deep tissue and bone. Those complications may finally lead to amputation[13]. In Belgium, approximately a thousand people suffer from lower limb amputation each year [3]. The properties and the purpose of prosthesis have evolved with time. As it was shown, aesthetic features were enhanced with more complex mechanisms allowing movements.

Generally, modern ankle-foot prosthesis can be divided into three categories: Conventional feet (CF), Energy-storing and returning (ESR) feet and the most recent one called the bionic feet [14].

2.2 Biomechanics of the ankle

This section summarizes the biological motivation for the inversion/eversion DoF in ankle-foot prosthesis. First of all, the gait cycle is briefly introduced. Then the degrees-of-freedom of an healthy ankle are explained and quantified in term of range of motion, torque, impedance, ... Finally, some movements in which the inversion/eversion movement is involved are presented.

2.2.1 Gait cycle

The gait cycle is divided in two phases : the stance phase during which the foot is touching the ground and the swing phase when it is not touching the ground. Those phases are represented in Figure 2.4.

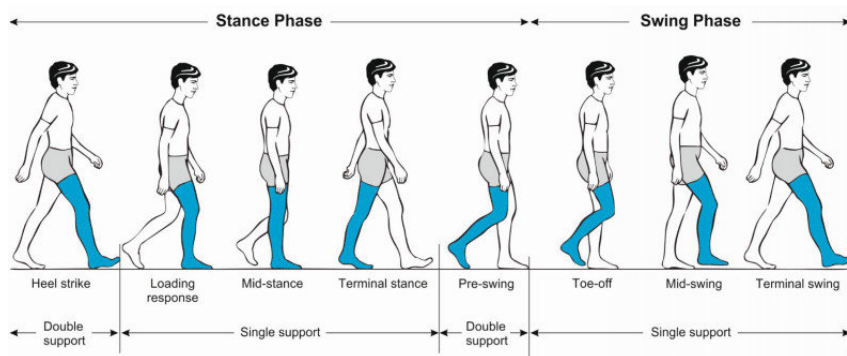


Figure 2.4: Phases of the normal gait cycle [15]

2.2.2 Degree-of-freedom

The ankle has three DoF, the main one is in the sagittal plane and is the plantarflexion and dorsiflexion (DP). Plantarflexion permits the push off after terminal stance and the dorsiflexion puts the feet back in position during the swing phase before the next gait. The second one is in the frontal plane and is called the inversion-eversion (IE) of the foot. Not yet included in active ankle-foot prostheses, this movement is the subject of this work and will be described in more details later. Finally, the last one is in the transverse plane and consists in the abduction adduction of the ankle. However, it will not be discussed in this work.

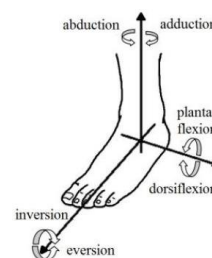


Figure 2.5: Movements of the ankle-foot [16]

The inversion-eversion being the main focus of this work, its implication in different behaviors is further explained. Figure 2.6 represents the IE in more details. This motion allows the foot to adapt to uneven ground in order to increase its stability. Furthermore, the significance of IE was discovered and studied in movements such as sidesteps and turning steps. In 2015, Ficanha *et al.* studied the impact of sidestep cutting in the frontal plane. Sidestep cutting can be described as a step taken to avoid an obstacle without rotating the body, as illustrated in Figure 2.7 [17]. Their results are presented and discussed.



Figure 2.6: Inversion and eversion [18].

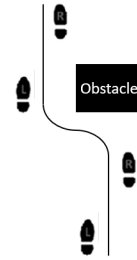


Figure 2.7: Example of sidestep cutting to avoid an obstacle [17].

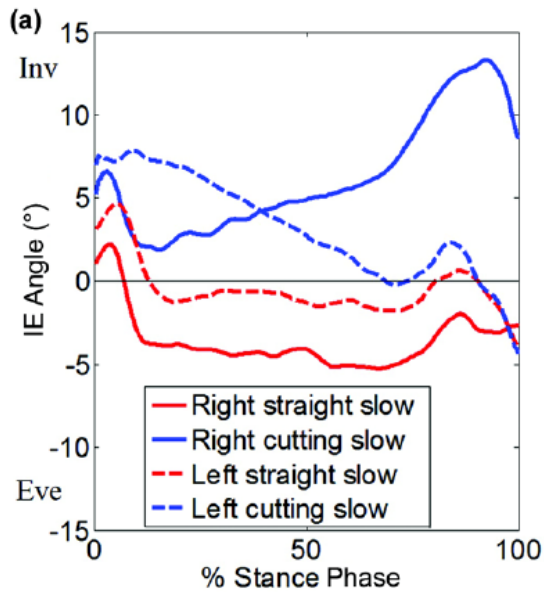


Figure 2.8: Plot of mean ankle angles in inversion-eversion (IE) at slow speed. Results obtained by Ficanha *et al.*[17] 13

In all the following graphs and figures, a negative angle describes eversion and a positive one inversion.

Ficanha *et al.* observed an increasing inversion from 2 to 13 degrees for the ankle initiating the movement when cutting right. And a decreasing inversion from 8 to -4 degrees for the contralateral step as shown in Figure 2.8, referred as *Right cutting slow* and *Left cutting slow* respectively. We also observe in Figure 2.8 angles between -5 and 5 degrees during a straight walk at slow speed (average 96 steps/min) [17].

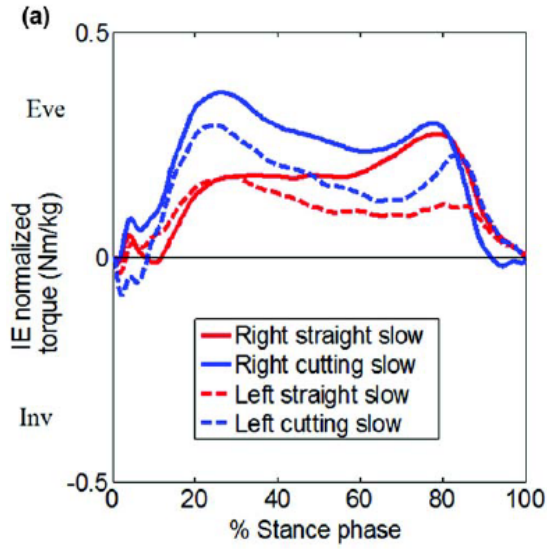


Figure 2.9: Plot of mean normalized ankle moments in inversion-eversion (IE) at slow speed. Results obtained by Ficanha *et al.*[17]

They measured higher eversion torque for sidestep cutting than straight walking with a maximum of eversion moment of 0.36 Nm/kg for the right ankle during sidestep cutting at slow speed. In both situations, sidestep cutting and straight walking, the moments are in eversion. The increase in eversion moments for sidestep cutting compared to straight walking occurred during weight acceptance and mid stance [17].

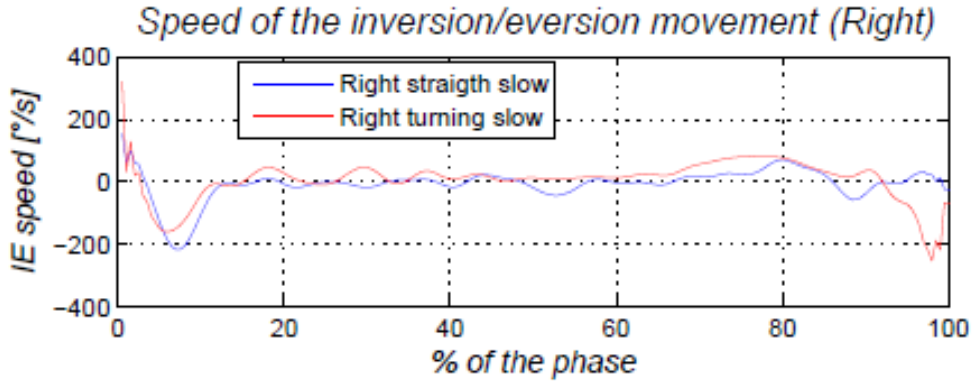


Figure 2.10: Speed of the inversion/eversion movement for the right ankle. Figure by Emile Moreau [19] based on Ficanha's data [17]

Emile Moreau's master thesis used Ficanha's data to compute the speed of the IE movement. In figure 2.10, we observe peaks up to 200 °/s for both situations during weight acceptance and again the same kind of peak at the end of the stance for the sidestep cutting [19].

Moreau also performed other experiments like lateral steps, walks on an inclined plane and unstable ground. In figure 2.11 is shown the IE angle when walking on a 15 degrees inclined plane in inversion mode (i.e. the inclined plane is angled to such a degree that the ankle is in inversion). It is observed that between the toe impact event and the heel off the ankle's angle in the frontal plane is the same as the slope of the plane.

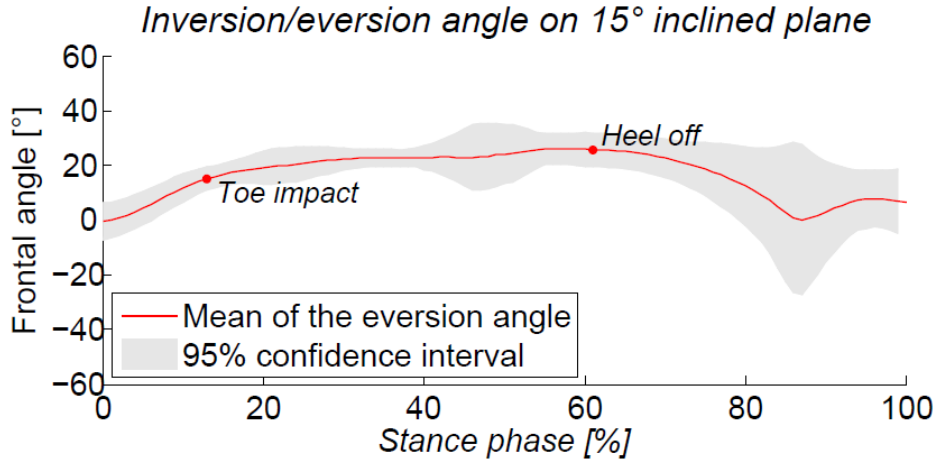


Figure 2.11: IE angle for 15 degrees inclined plane in inversion mode.[19]

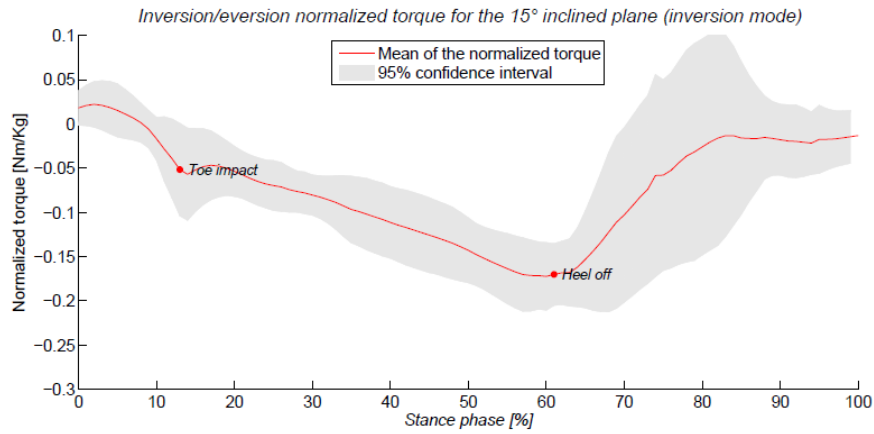


Figure 2.12: IE torque for 15 degrees inclined plane in inversion mode[19]

In Figure 2.12 is shown the normalized torque when walking on the 15 degrees inclined plane. Being in inversion mode, the torque is in eversion to counteract the gravity leaning the body. With a peak eversion torque of 0.17 Nm/kg [19].

The mechanical impedance, which correlates output force to input disturbance motion, plays an essential role in stability for the ankle. This impedance is time-varying and task dependent, by contracting our muscles we are able to modify the stiffness and damping of our joints.

In 2018, Ficanha *et al.* performed experiments to measure the time-varying impedance of the ankle during the stance phase of ground level walking. They applied random perturbation along 8 axes of rotation and measured the impedance at different moments of the stance length (SL). Their results are shown in Figures 2.13 and 2.14 [20].

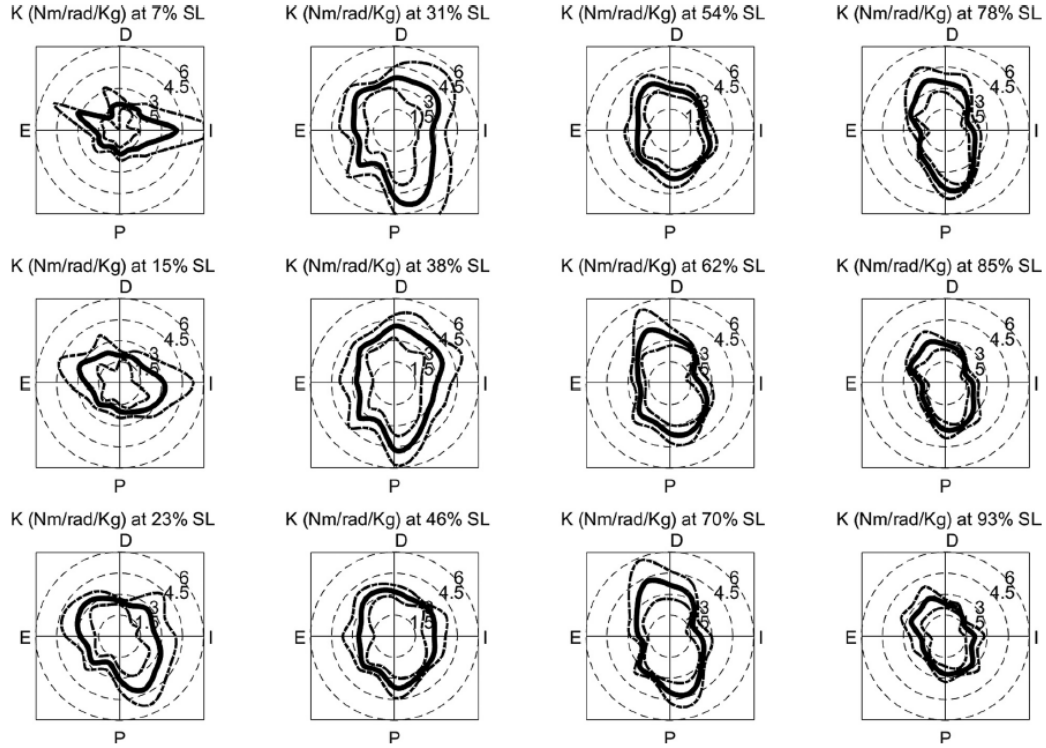


Figure 2.13: Contour plots of the normalized stiffness from 7% to 93% SL. Solid line: Stiffness. Dashed lines: $\pm$ Standard deviation. [20]

They observed that the damping and stiffness of the ankle vary continuously during the stance phase. For both components of the impedance, they observed low values at heel strike then increasing values during mid stance and decreasing values at push off. The maximum and minimum average ankle stiffness were $5.43 \left[\frac{Nm}{Rad \cdot kg} \right]$ at 31% of the stance length (plantarflexion and inversion combined) and $1.14 \left[\frac{Nm}{Rad \cdot kg} \right]$ at 7% of the stance length (dorsiflexion and eversion combined) respectively.

For the damping, the maximal and minimal average values were 0.08 and 0.016 $\left[\frac{Nm \cdot s}{Rad \cdot kg}\right]$ at 38% (plantarflexion and inversion combined) and 7% (dorsiflexion and eversion combined) of the stance length respectively [20].

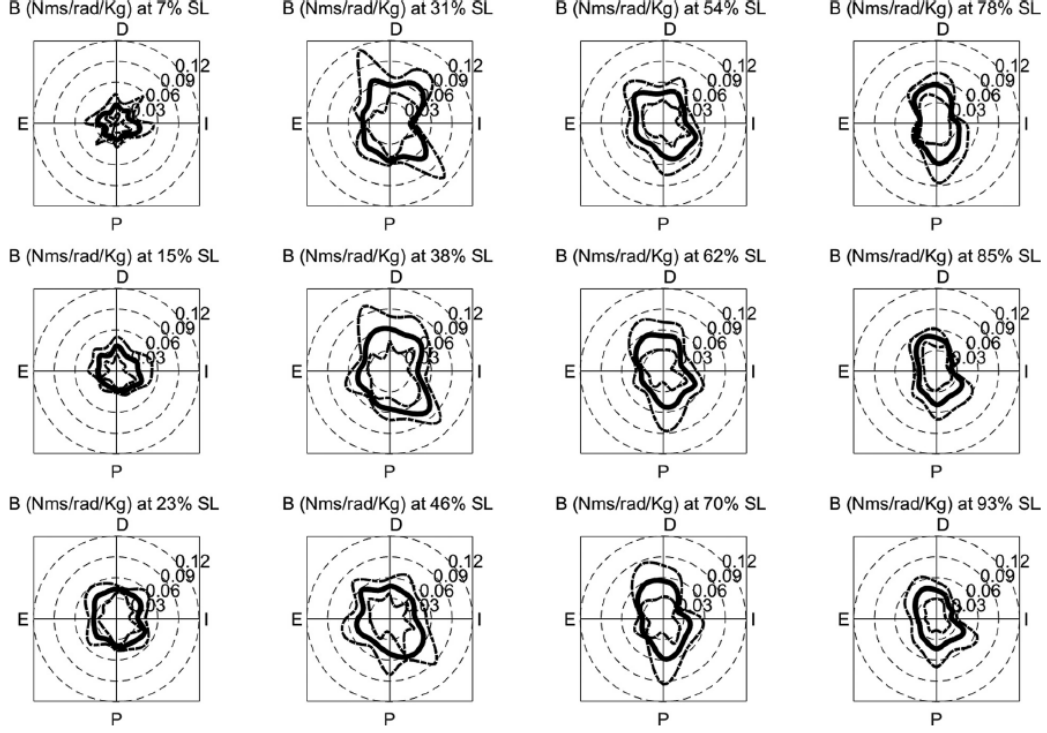


Figure 2.14: Contour plots of the normalized damping from 7% to 93% SL. Solid line: Damping. Dashed lines: $\pm$ Standard deviation.[20]

After heel strike, the stiffness and damping values are bigger in IE than DP which is probably due to stability reasons for IE while DP needs more compliance to absorb the shock. They also observed that the maximum values were measured for movements which are produced by rotations of the subtalar joint, suggesting the importance of this joint and the associated muscles in the modulation of the impedance for the ankle [20].

2.2.3 Conclusion

The degree-of-freedom in the frontal plane plays a critical role in the stability of the ankle. It is implied in many daily life movements like level ground walking, sidestep cutting or walking on uneven ground. The impedance of the ankle, which is time-varying and task dependent, is modulated by the contraction of leg muscles and ensures the right behavior of the joint. In Figure 2.15 are shown the typical values for both the sagittal and frontal DoF for level ground walking. Depending on the sources, the total range of motion for IE varies from 35° [21] to 75° [19].

Movements	Maximal ROM [°]	ROM during walking [°]	Peak Torque [Nm/kg]	Power [W/kg]
Dorsiflexion/ Plantarflexion	75 [-55 ; 20]	30	1,6	3,6
Inversion/ Eversion	35 to 75 [-12 ; 23] [-45 ; 30]	10	0,27	0,05

Figure 2.15: Comparison of the sagittal and frontal DoF of the ankle for level ground walking [17, 21, 22]

2.3 Modern prosthesis

This section provides an overview of the evolution of ankle-foot prosthesis. The objective is to understand the different implementations and their implications.

2.3.1 Conventional feet

The most basic ankle-foot prosthesis is the SACH, which stands for Solid Ankle Cushioned Heel. This passive device made of wood and cushioning material provides stability during the stance thanks to the wooden keel and absorbs shocks after heel strike thanks to the cushioned heel. It is still used (especially in developing countries) because of its robustness and its very low cost [14]. The device is shown in Figure 2.16.

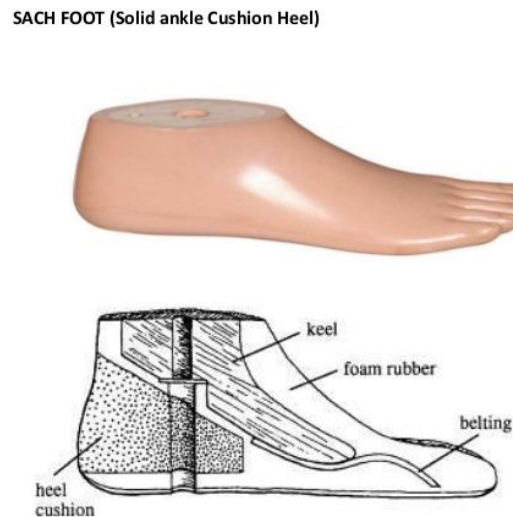


Figure 2.16: Top : physical implementation of the SACH foot, bottom : scheme with the composing elements. [23]

2.3.2 Energy storing and Returning feet

The desire of amputees to be able to achieve a more natural walking, to reduce the metabolic cost or to practice sports has led to the development of Energy Storing and Returning feet (ESR). ESR foot prosthesis are capable of storing energy by deformation during the stance and returning a part of it during the late stance to do the push off [14].

The first ESR was the Seattle Foot (see Figure 2.17 **A.**) created in 1981 and composed of a flexible keel inside a polyurethane shell that stores energy when loading during the stance phase, like an elastic spring, and returns part of it for the push off later in the gait cycle. However, a big part of this energy is dissipated in heat and sound due to friction [14].

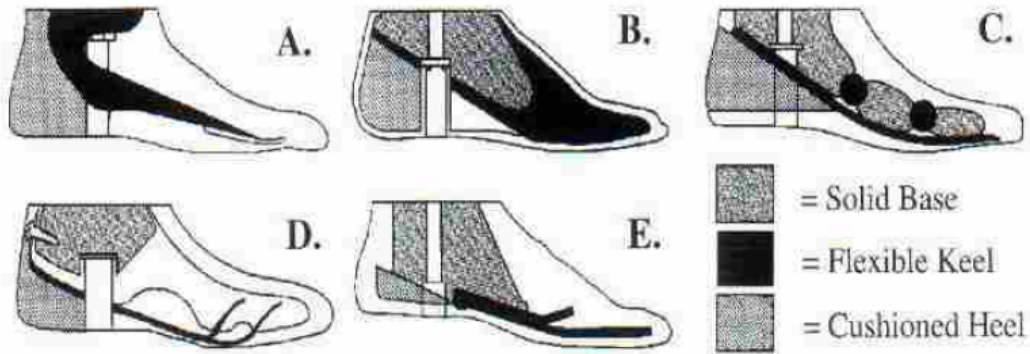


Figure 2.17: Cross sections of various energy-storing feet. **A.** Seattle foot, **B.** Dynamic foot, **C.** STEN foot, **D.** SAFE foot, **E.** Carbon Copy II [24]

The following generations of ESR were composed of an entire flexible carbon fiber shank and a heel spring, allowing the entire prosthesis to flex and store energy. For comparison, in the Seattle foot only the keel was able to flex. Some examples are shown in figure 2.18, including the pioneer of this generation the Flex-Foot (**A.**) developed in 1987 [24].



Figure 2.18: Advanced energy storing prostheses: **A.** Modular III, **B.** Reflex VSP, **C.** Advantage DP and **D.** Pathfinder [24]

The already presented mechanisms are passive, meaning they store energy by deformation and some of it can be used in the gait cycle while the rest is dissipated. Active components are required to inject more energy during the gait cycle to help amputees' ambulation. Semi-active ESR exist, they use a microcontroller and a small motor to change the configuration of the device in order to adapt to the environment, by changing the tilt of the ankle for example. They are also called stabilizing devices. The two most advanced and commercially available devices are the Meridium, Ottobock (Duderstadt,DE) and the Proprio foot, Össur (Reykjavik,IS), shown in Figures 2.19 and 2.20.



Figure 2.19: Meridium (Ottobock)



Figure 2.20: Proprio foot (Össur)

Research semi-active devices were also produced, one of them is reviewed.

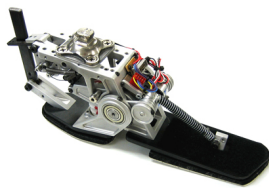


Figure 2.21: Energy recycling foot developed at the Delft University [25]

Collins and Kuo developed in 2010 an energy recycling ankle, shown in Figure 2.21. The device stores passively energy that is normally dissipated by the leg to return it for the push off. The foot is divided in two parts : the rear-foot and the fore-foot, the energy stored in the spring is released only when a sufficient load is detected on the fore-foot. The only active elements are a microcontroller and two micro-motors that are used to release the energy-storing spring. The device was compared to a conventional prosthesis (Seattle LightFoot 2) on able-bodied subjects and resulted in a greater push off and lower metabolic expenditure [25].

2.3.3 Bionic feet

The major difference between this last category and the two previous one is the embedding of an active component in the prosthesis, allowing active energy injection during the gait cycle. Indeed, a significant plantarflexion torque is required in the late stance. This requirement cannot be completely fulfilled by passive devices. Different kind of actuators can be used like pneumatic or electrical actuators. Klute was the first to try to create a powered ankle prosthesis, using McKibben actuator as pneumatic muscles [26].

The MIT developed a prototype of an active ankle-foot prosthesis in 2012 called BiOM [27]. Then the german company Ottobock bought it and made it commercially available. To the best of the author's knowledge, it remains the only commercially available active ankle-foot prosthesis. The first prototype was done in 2007, using a series elastic actuator (SEA) to inject power. The advantages of using a SEA are the possibility to do force control but also to modulate the joint stiffness [28]. Both the first prototype as well as the currently available model are shown in Figure 2.22 and 2.23.



Figure 2.22: Physical prototype of the powered ankle-foot prosthesis developed by Herr et al. [28]



Figure 2.23: Empower (Ottobock)[29]

In October 2020, the International Conference on Intelligent Robots and Systems (IROS)[30] took place online and presented some of the most advanced bionics prostheses for the lower limb. Some of them will be described.



Figure 2.24: Powered ankle prosthesis developed at the Robotics Institute University of Michigan[31]

Gregg and his team developed a prosthetic leg including a prosthetic knee and ankle, the assembled ankle actuator is shown in Figure 2.24. A single-stage stepped-planet compound planetary gear transmission coupled with a high torque-density motor is used. Hence, a low mechanical impedance is achieved for the actuator. This lead to a facilitate control of the device, low acoustic sound, high backdrivability, ... The motor is placed proximal to the body for the ankle to reduce the hip torque needed to move a mass worn distal from it. A parallelogram four-bar linkage is implemented to translate the torque from the gearbox to the ankle. Experiment were performed with a transfemoral (TF) amputee to confirm that the ankle trajectory was similar to reference trajectories. The leg prosthesis weights 6.09 kg [31].

The work of Dirk Lefeber and his team for the european project CyberlegsPlus-Plus was also presented. Their goal is to enhance/restore the mobility of transfemoral amputees [32]. The ankle is the AMP4+, shown in Figure 2.25. It is an active ankle-foot prosthesis developed at the VUB (Belgium). It uses a resettable overrunning clutch (RORC) in series with a series elastic actuator (SEA). They showed that this clutched series elastic actuator (CSEA) reduces the electrical energy consumption during the passive gait phases in which the ankle torque is generated without using the motor. They compared it with a SEA driven ankle, for a relative weight increase of 8% the electrical energy reduction for level ground walking was 22% and 4% at slow and fast speed respectively. The device weights 2.3 kg, not including batteries [33].

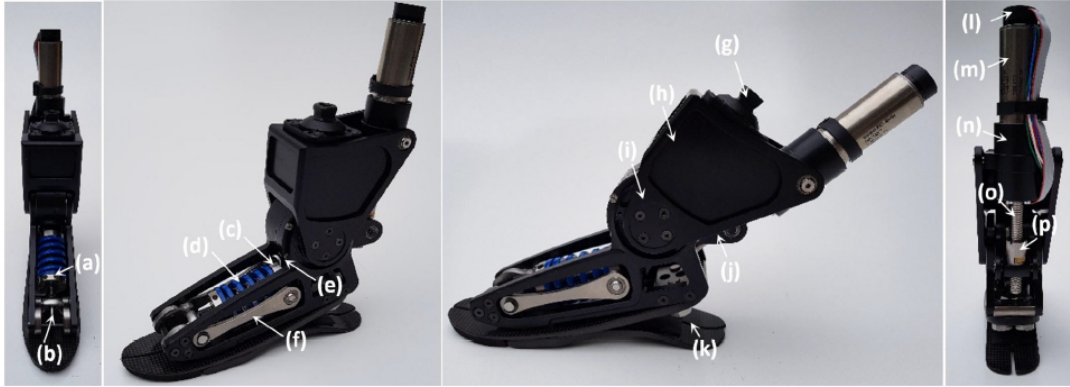


Figure 2.25: The developed AMP-Foot 4+ prototype: front view, isometric view, side view and back view [33]

Tommaso Lenzi and his team presented an active ankle prosthesis with a polycentric design, this design allows the Instantaneous Center of Rotation (ICR) to move during the gait cycle. Doing so the load on the transmission is reduced and the speed and torque required at the motor output are also affected, improving the electrical efficiency. They succeeded in matching the average weight of a female ankle/foot (1.3 kg) with a small build height (12 cm). The prosthesis uses a seven-bar mechanism with one degree of freedom actuated by a linear actuator [34]. It is shown in Figure 2.26.



Figure 2.26: The powered polycentric ankle developed by Lenzi and his team, Utah [34]

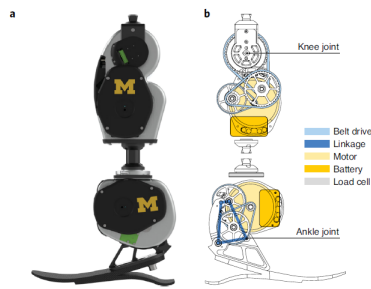


Figure 2.27: The OSL and its design components. **a** Rendering of the OSL. **b** Schematic of the OSL [35]

Elliott Rouse and his team presented an open source leg (OSL) prosthesis comprising of an knee and ankle. Their goal is to allow teams around the world to test different control strategies and to study its biomechanics. Their device is scalable, the ankle and the knee can operate independently allowing research for TF and TT amputees. In Figure 2.27, the different components are shown. They chose brushless motors similar to those in the drone industry because of their useful properties, namely their efficiency and torque density. This provides the opportunity to use belt drive transmission thanks to the low transmission ratios. With their design they achieved a build height of 21.3 cm and a weight of

1.74 kg [35].

Kevin Hollander and Thomas Sugar presented the work of the the Human Machine Integration Lab, ASU. They developed an ankle prosthesis able to walk and run. The Walk-Run ankle is composed of a 200 Watts brushless DC motor actuating a roller-screw via a belt drive transmission. A spring is also used to store and release energy during walking and running gaits. The presented design, see Figure 2.28, has a weight of 1.9 kg not including batteries and electronics. A controller detects the ambulation mode then it will compute the desired motor trajectory thanks to PD control. Different speeds were tested (walking 1.6 m/s, running 2.6 m/s, running 4 m/s) with able-bodied using a bypass system and compared with reference and experimental data. The Walk-Run was able to mimick desired torque and angle trajectories up to 2.6 m/s, at higher speed the output torque was not sufficient [36].



Figure 2.28: Walk-Run ankle developed at the Human Machine Integration Lab, ASU Polytechnic Arizona State University[36]

To conclude this section devoted to active ankle prosthesis, François Heremans' work is presented. His PhD thesis at UCLouvain lead in 2019 to ELSA (Efficient Lockable Spring Ankle), a biomimicking ankle prosthesis.

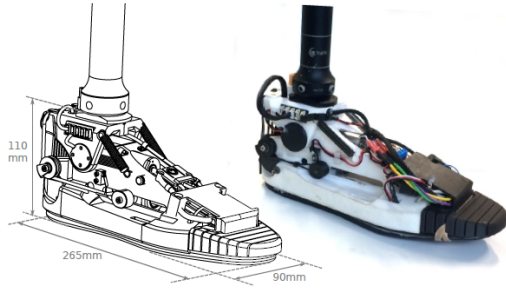


Figure 2.29: CAD model of ELSA and assembled prototype to be tested [37]

Since active devices tend to be bulkier and heavier than the passive ones which is a disincentive for their use, ELSA was designed to be lightweight, energy efficient and low cost. In order to reduce the weight, series elastic actuators (SEA) are used. They offer the opportunity to use smaller motor by decreasing the motor speed. Furthermore, reducing the motor torque using a parallel spring increases the efficiency since the actuator must then provide only a fraction of the requested joint torque [37]. The prototype is shown in Figure

2.29.

The parallel spring is a nylon rope attached on one end to the locking mechanism (ratchet and pawl) and on the other end the fixation point can be adjusted to render a desired stiffness [38]. The active part of the design is composed of a ball screw actuator with an elastic rope transmission. The system being unidirectional and only able to pull (plantarflexion), a spring is used to bring back the ankle to its neutral position during the swing phase (dorsiflexion) [37]. Both are shown in Figure 2.30 and 2.31.

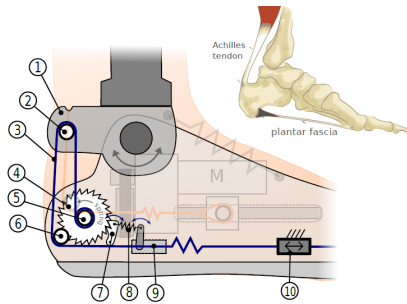


Figure 2.30: Bio-inspired lockable parallel spring [37]

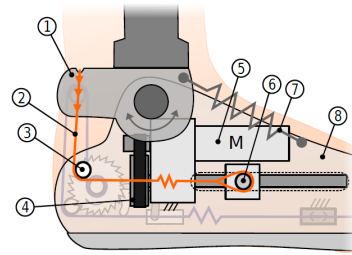


Figure 2.31: Series elastic actuator [37]

ELSA is lightweight (1.2 kg not including batteries), low-cost (~ 1000 \$) and with a small build height (11 cm).

2.4 Ankle-foot prostheses with 2 DoF

This section describes the existing research prototypes for two degrees-of-freedom ankle-foot prostheses, at least to the best of the author's knowledge. The section will be divided with respect to the way of actuation.

2.4.1 Passive

In 2016, Mai *et al.* designed MACPA which stands for multi-axis fully compliant prosthetic ankle. This prosthesis is composed of two parts, the upper one is a flexure hinge allowing IE motion and the lower one permits the DP motion with a curved-beams group. The prototype is made with polyoxymethylene (POM) and carved by a computer controlled milling machine. They optimized the shape of the foot based on the total desired stiffness of the structure (550 N/mm) with FEM. They measured a stiffness of 243 N/mm for the eversion [39]. The prosthesis is shown in Figure 2.32.

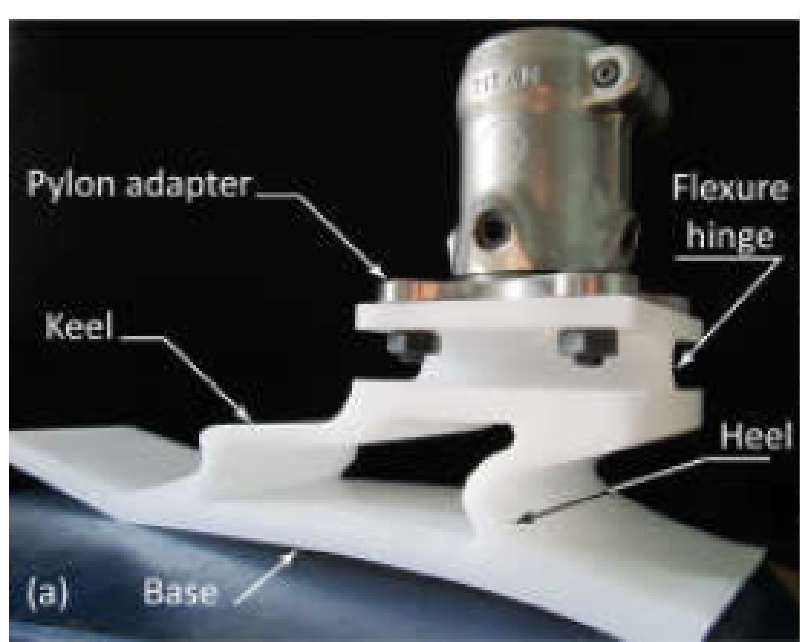


Figure 2.32: Picture of the fabricated MACPA[39]

2.4.2 Semi-active

Hugh Herr and his team published in 2020 a paper about an ankle-foot prosthesis for rock climbing augmentation. Their goal was to enhance capability of transtibial amputees for rock climbing.

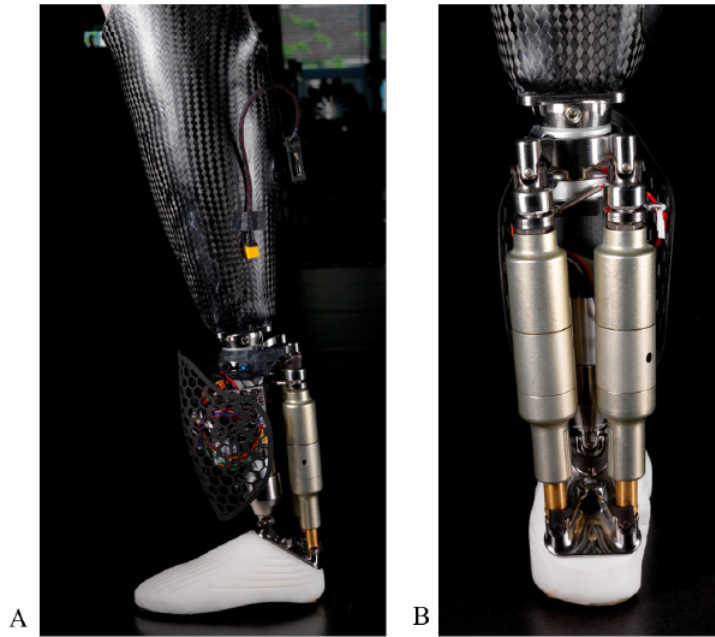


Figure 2.33: Lateral view (A) and posterior view (B) of final rock climbing ankle-foot prosthesis [40]

They designed a prosthesis controlled electromyographically. Which means that their controller is using electrical signals from muscle contractions as input and computes the corresponding motor command. This technique allows the user to control his foot orientation to climb in a more efficient way than using his passive prosthesis. The device is shown in Figure 2.33. It is a quasi-passive system because the purpose of the two linear actuators is to permit volitional position control when the foot is not touching the ground. The two actuators being non-backdrivable, it means that no output torque is needed during the stance phase. Indeed the ankle position is locked when it supports the body weight of the user. They achieved a range of motion of ± 22.34 degrees and ± 16.61 degrees for inversion/eversion and dorsiflexion/plantarflexion respectively. The total mass including batteries and electronics is 1.292 kg for a build height of 25 cm [40].

2.4.3 Active

The first attempt to make an active ankle prosthesis device with two degrees-of-freedom was done by Bellman et al, in 2008 [41].

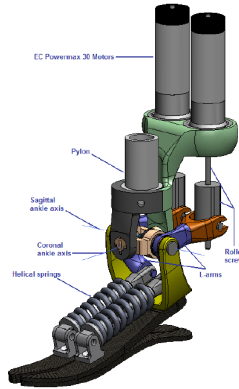


Figure 2.34: CAD design of Sparky 3 by Bellman et al. [41]

They created the SPARKy project (Spring Ankle with Regenerative Kinetics) funded by the US Army Medical Research & Material Command. Their objective was to restore ankle function for transtibial amputees especially for wounded soldiers. They made 3 different prototypes and only the last one SPARKy 3 has two DoF, to the best of the author's knowledge SPARKy 3 was never assembled in a real prototype [41]. In figure 2.34 is shown the CAD design of Sparky 3. They used two different motors that can be used in unison for powered running and jumping and separately to control the inversion/eversion DoF. They estimate based on the design a weight of 2.1 kg.

In 2016 Ficanha et al, developed a cable-driven ankle-foot prosthesis with both active plantarflexion/dorsiflexion and inversion/eversion, based on a proof-of-concept previously made [43]. They used Bowden cable to transmit the power from the motors to the ankle-foot mechanism. Also since cables are lightweight and flexible they have high actuation power over volume and over mass ratios. Their main limitation remains the loss of energy generated by friction. They use an instrumented foot to measure the ground reaction force, making impedance control possible. The passive components shown in Figure 2.35 weight 1.13 kg and the motors and transmission, which are not embedded weight 2.2 kg [42].

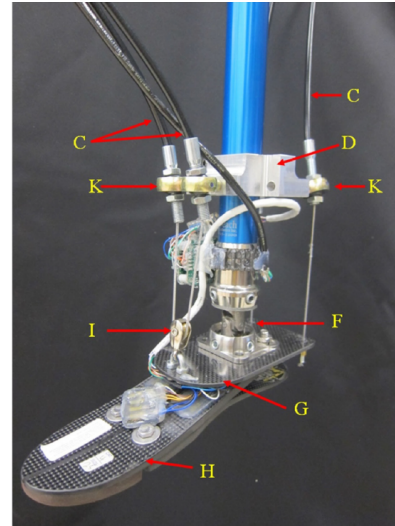


Figure 2.35: Two DOFs cable driven ankle-foot prosthesis developed by Ficanha et al. [42]

In 2018, Collins et al. designed a tethered ankle-foot prosthesis emulator composed of two independently actuated toes, its CAD design is shown in Figure 2.36. The purpose of the emulator is to test different control strategies.

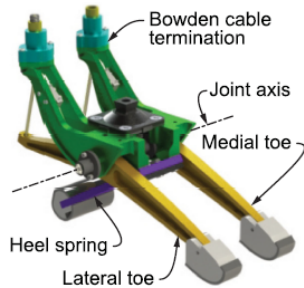


Figure 2.36: Mechanical design of the two degree of freedom ankle-foot prosthesis emulator designed by Collins et al. [44]

They designed it to be an end-effector actuated by Bowden cables, this way the prosthesis is lightweight (0.72 kg for 8 cm build height) since the motors and the controllers are placed off-board. The two degrees-of-freedom are coupled, indeed the toes shared a common axis of rotation which is similar to the one of plantarflexion in a human ankle joint. Toes have been designed with an I-beam cross section well adapted for three-points bending. The DP is achieved when both toes are rotating in the same direction and IE when they are rotating in opposite direction. The dorsiflexion is done passively thanks to rubber bands, an unactuated heel spring is also connected to the frame to absorb shocks and rubber bands are attached at the end of toes and on the heel to improve grip with the ground. The ranges of motion of the design for plantarflexion/dorsiflexion and inversion/eversion are -20 to 30 degrees and -30 to 30 degrees respectively. It was tested with a TT amputee [44].

In 2019, Tsung-Han Hsieh developed a prototype of powered ankle-foot prosthesis for his master thesis at the MIT. The physical implementation is shown in Figure 2.37. It was the first prototype of untethered, self-contained powered two degrees-of-freedom ankle-foot prosthesis for walking and stairs ascending. It is composed of two motors actuating a four-bar linkage thanks to timing belts. It forms a differential drive meaning the plantarflexion/dorsiflexion is achieved when the motors are moving in sync and inversion/eversion when they are not moving in sync. Strain gauges are used to perform force control. The prosthesis was tested with a knee immobilizer to evaluate the torque tracking performance and the robustness of the mechanical transmission. The design allows 13.5 degrees both in inversion and eversion. Furthermore it allows 9 and 21 degrees in dorsiflexion and plantarflexion respectively. It weights 2.42 kg without batteries for a build height of 21.08 cm [45].

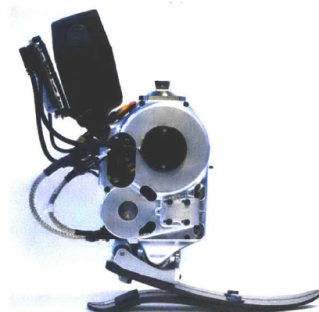


Figure 2.37: 2 DoF powered ankle-foot prosthesis developed by Hsieh [45], MIT.

2.5 Conclusion

The evolution of ankle-foot prosthesis has led to active solutions that have the advantage of providing the necessary energy for the gait cycle. However, this comes at a price in terms of weight and bulk due to the motor and other mechanical components.

We observe that two degrees-of-freedom ankle-foot prosthesis can be done in each of the three actuation ways. Nevertheless, to be able to control the mechanical impedance and to adapt it during the gait cycle or specific tasks, two motors are commonly used. Consequences of it are the bulkiness and the heaviness of such prosthesis. The lighter solutions with off-boards motors are mainly used to test control strategies and are not designed to be daily live prostheses for patients. Figures 2.38 and 2.39 summarize the different bionic feet prototypes presented and the 2 DoF prototypes respectively.

Prosthesis	Mass [kg]	Height [cm]	ROM [°]	Peak torque, Instantaneous [Nm]
Empower	2	22	45	125
Gregg	6,09 (ankle + knee)	NA	NA	NA
Lefebvre	2,03 (w/o batteries)	NA	55	NA
Lenzi	1,3	12	55	125
Rouse	1,74	21,3	30	178
Sugar	1,6 (w/o batteries)	31	NA	140
Heremans	1,28 (w/o batteries)	11	50	125

Figure 2.38: Comparison of bionic feet presented in section 2.3.3.
Abbreviations : w/o = without, NA = Not Available

Prosthesis	Actuation	Actuators	Mass [kg]	Height [cm]	DP ROM [°]	IE ROM [°]	Peak torque DP [Nm]	Peak torque IE [Nm]	Stiffness in DP [Nm/rad]	Stiffness in IE [Nm/rad]
Mai	Passive	/	NA	NA	NA	NA	NA	NA	NA	243 [N/mm]
Herr	Semi-Active	2 linear Actuators	1,292	25	±16,51	±22,34	100 (static) 0,55 (dyn)	100 (static) 0,55 (dyn)	NA	NA
Ficanha	Active	Bowden cable system	1,13 (+2,2 Motors)	NA	NA	NA	264	NA	0 to 236	1 to 33
Collins	Active	Bowden cable system	0,72 (w/o motors)	8	-20 to 30	-30 to 30	250	±30	NA	NA
Hsieh	Active	Differential drive	2,42 (w/o batteries)	21,08	-21 to 9	±13,5	162,4	59,35	NA	NA

Figure 2.39: Comparison of the 2 dof prototypes presented in section 2.4.
Abbreviations : w/o = without, NA = Not Available, DP = dorsiflexion/plantarflexion, IE = inversion/eversion, dyn = dynamic, ROM = Range Of Motion

It is therefore clear that ankle-foot prostheses with two degrees-of-freedom are still in their early stages. This work will focus on a different approach than those presented, namely to independently manage the inversion-eversion DoF assuming than that in the sagittal plane is already implemented. This way, the obtained mechanism could be fixed on top of a wide range of existing active ankle-foot prosthesis. The following chapters will be dedicated to the design of such a mechanism.

Chapter 3

Methodology

This chapter will focus on the design and project management methods. A good methodology is very important in order to find the best possible solution to a specific problem. Before going into the design, a methodology must be chosen between the different possibilities.

The goal of this chapter is not to list all the design methodologies but rather to justify the choice made for this master thesis.

3.1 Design-focused approach

Classical design methodologies are typically procedural approaches. It consists of models made of many phases and focusing on a specific audience or industry sector. They can be divided into two types:

- Descriptive models : Based on what is observed in practice. Describing the reality of the field [46, 47].
- Prescriptive models : Strategy to improve the efficiency of the design process. Guidelines to rationalize the procedure steps [46, 47].

The most classical example of descriptive model is the French's model, illustrated in Figure 3.1. For the prescriptive models one of the most famous one is the Pahl and Beitz model[48], illustrated in Figure 3.2. Since the Pahl and Beitz model is precisely made for mechanical design, its procedure will be followed in this work. It consists of a list of working steps which are considered to be strategic guidelines for design. By following them, nothing essential should be forgotten [46]. As presented in Figure 3.2, the four steps of this methods are the following:

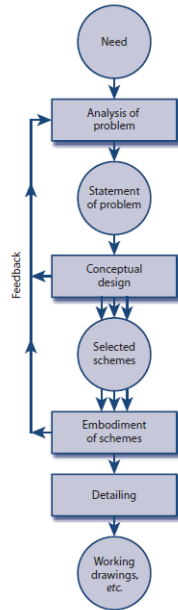


Figure 3.1: French's model of the design process [49]

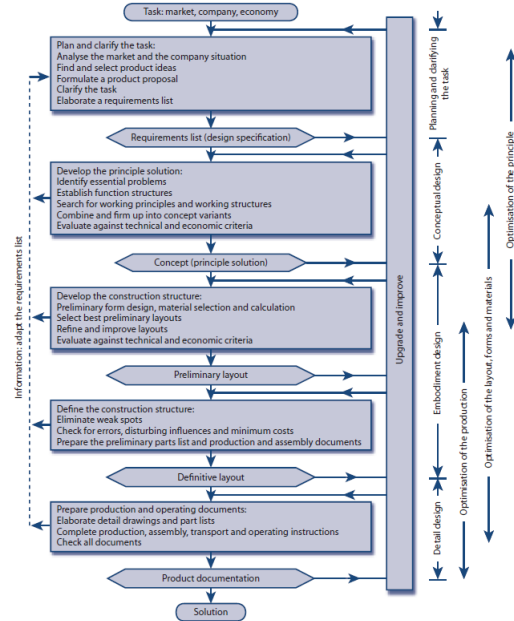


Figure 3.2: Pahl and Beitz's model of the mechanical design process [48]

1. Planning and task clarification. The goal of this step is to obtain the design specification.
2. Conceptual design. By combining concepts to answer the main problems, a first iteration of the solution is built.
3. Embodiment design. Dimensioning, material selection, preliminary design form to end up with a preliminary layout. And after final adjustments, the definitive layout.
4. Detail design. Documents related to the manufacturing (drawings, assembly, transport,...).

A common point shared by many methods is the non linearity of the process. Feedback is included making it iterative or cyclic. To implement the different steps, tools borrowed from the Cross model [50] will be used. Such as for example the morphological chart, breaking down the problem into sub-problems and their corresponding sub-solutions, comparison of combined solutions, ...

3.2 Project-focused approach

The other important methodology aspect for this work is the project management. Again, there exists plenty of methods. For this work, Agile methods more precisely SCRUM model was used. As explained by Srivastava et al [51], this model was initially developed for software development. Scrum model is both iterative and incremental because the builds are successive and complete each other. Generally speaking, the Scrum's workflow consists of collaboration with the product owner, the scrum master and the scrum team. The idea is to list the requirements in a project backlog. Then the work is divided in sprints. A sprint is the smallest block of scrum and lasts on average two to three weeks. During a sprint, daily scrum meeting are organized to check the project's progress. At the end of each sprint a review is organized with the product owner and feedback is given to the team [52]. Such typical framework is shown in Figure 3.3.

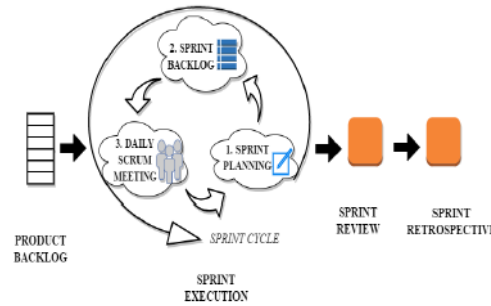


Figure 3.3: Scrum framework. Sharma et al. 2016 [52]

This global framework was adapted for this work. François Heremans, one of the supervisors, agreed to play the role of the scrum master. Weekly meetings were organised separated by sprints of one week. First, a roadmap for the second semester was made (see section 8, Figure 8.2). Then each week, a sprint backlog was set up based on the progress or difficulties encountered the previous week (see section 8, Figure 8.3). This way, complex tasks with large deadlines such as the design could be divided into simpler ones that could be completed in a week. Every three weeks or so, an update on the progress of the work was also carried out with Renaud Ronsse, the main advisor of the thesis, playing rather the role of the product owner. This way, the objectives could be adapted based on the evolution of the work. This flexibility is an advantage of agile methods when compared to more classical models such as the Waterfall models [53] which are more linear. In the context of the Covid crisis, this approach has proven to be very efficient for the motivation and the feeling of loneliness of the author.

Chapter 4

Design

4.1 Introduction

In this chapter, the design of a mechanism allowing inversion/eversion for ankle foot prosthesis will be tackled. The sagittal DoF will be considered already implemented. The objective is to provide a lightweight module that could fit on most of the existing ankle foot prosthesis. Adding the DoF in the frontal plane would permit the foot to adapt to a wider range of terrain such as uneven ground. This adaptability plays an important role in the amputee's stability. Improving stability has a positive impact on the confidence that the amputee has in the prosthesis.

The methodology presented in the previous section will be implemented. The guidelines proposed by the Pahl and Beitz model will be followed and the Cross model's toolbox will help implementing it.

4.2 List of requirements

In order to complete the information of the previously analyzed literature (see section 2), three interviews with transtibial amputees have been made. The characteristics of the participants are the following, see Figure 4.1. The goal of those interviews was to identify the main design constraints based on the patient's reality.

Transtibial amputee	Age (years)	Sex	Cause of amputation	Affected side	Type of prosthesis	Time since amputation
S1	63	M	Trauma	Left	ESR	43 years
S2	58	M	Trauma	Left	ESR	21 years
S3	38	M	Malformation	Both	ESR	38 years

Figure 4.1: Participant demographics

Based on the literature data and on the interviews one main function is identified : *The ambulation with inversion/eversion*. The main goal of the design is indeed to add the DoF in the frontal plane for the ambulation. This main function is divided into two sub functions, namely *Level ground walking* and *Sloped terrain*. These sub functions represents two situations for which data are available regarding the range of motion, the mechanical impedance,... thanks to the work of Ficanah [20] and Moreau [19]. The main constraints identified with the help of interviewed amputees are the safety, the transitions between different modes (sitting - standing - walking), the additional cost, the dimensions and the noise. The complete specifications with the criteria are shown in section 8, Figure 8.1.

4.3 Conceptual design

With the chosen design procedure, the next step further the best possible solution is a morphological chart. This step consists of dividing our main function, here *Ambulation with inversion/eversion*, into simpler sub-functions and to list all the possible ways to fulfill them. In our case, the main function was divided into four sub-functions:

1. Elements capable of generating a force.
2. Second degree-of-freedom kinematics.
3. Elements capable of modifying the impedance.
4. Limiting the range of motion.

The morphological graph obtained is shown in Figure 4.2.

For the first item we have from left to right : (1) a damping spring, (2) a motor and its controller, (3) a shrinkage in a beam, (4) a flexible material, (5) magnets, (6) hydraulic and (7) pneumatic.

For the second item, (8) two rotoid joints in series or (9) an universal joint.

For the third item, (10) changing the pneumatic pressure in a cushion, (11) blocking an elastic element, (12) wind an elastic tendon, (13) a variable impedance joint mechanism, (14) changing the lever arm of a beam and (15) hydraulic solutions.

Finally for the last item, (16) hard or soft mechanical stops can be used.

The following step in the procedure is to create different solutions combining concepts listed in the morphological chart. Five designs were developed and are presented in the next section.

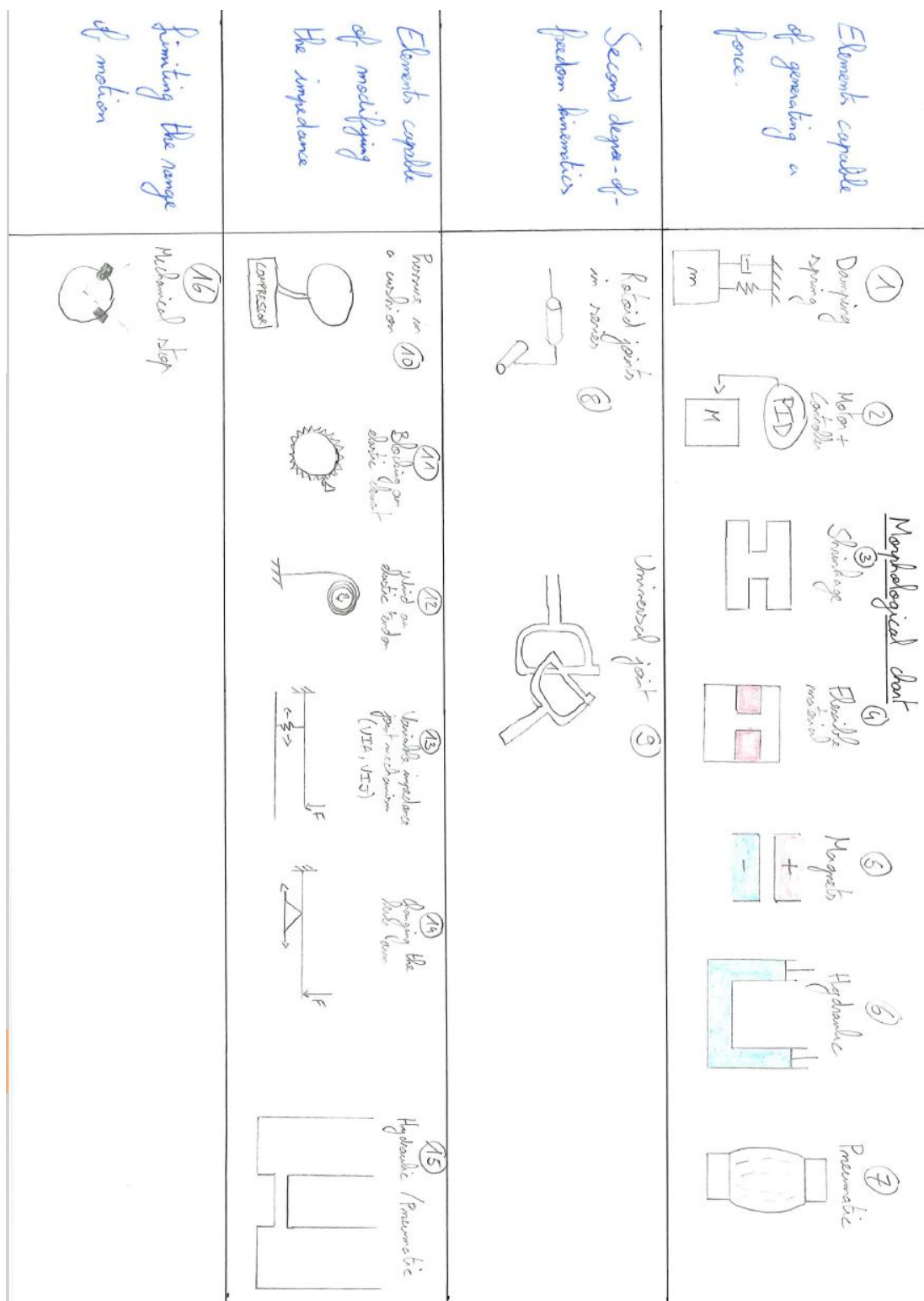


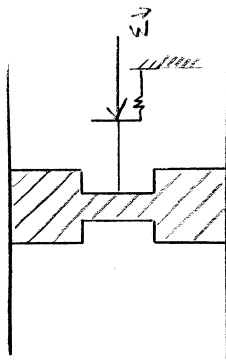
Figure 4.2: Morphological chart for the main function

4.3.1 Combined solutions

In this section, five combined solutions are presented and then evaluated according to the specifications to choose the best possible architecture.

Hydraulic solution

In the Figure 4.3, the first combined solution which is an hydraulic one is presented.

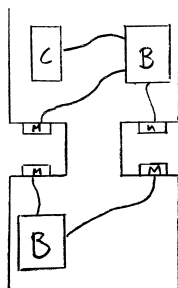


Two deformable tanks are filled with an incompressible fluid and the linking pipe between them can be closed when the weight of the person is acting on it. The inversion/eversion movement is allowed by the displacement of the fluid from one side to the other due to the compression of the tank. The impedance of the ankle is constant when the linking pipe is closed by the weight of the person acting on it. A spring is used to re open the linking pipe when the weight is not acting on it anymore, during the swing phase.

Figure 4.3: Hydraulic solution, frontal plane view

Electro-magnet solution

In the Figure 4.4, a solution with electro-magnet is presented.

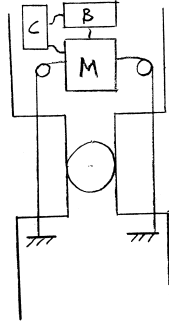


In this case, the movement is allowed thanks to the deformation of a beam which is narrow in the middle. Two batteries symbolised by the letter B are alimenting a controller (C) and four electro-magnets (M). The impedance of the ankle can be controlled thanks to the repulsive forces generated by the electro magnets during the stance phase.

Figure 4.4: Electro-magnet solution, frontal plane view

Series elastic actuator solution

An active solution using a series elastic actuator is shown in Figure 4.5

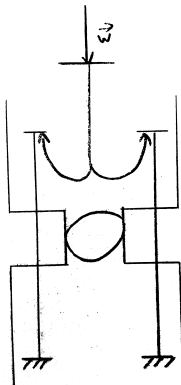


The motor is symbolised by the letter M and the batteries and the controller by the letters B and C respectively. An elastic component such as a nylon tendon is put in series with the motor forming a series elastic actuator (SEA). The controller gives the right command to the SEA to obtain the desired inversion/eversion movement. The ankle is rotating around a rotoid joint in the frontal plane.

Figure 4.5: SEA solution, frontal plane view

Passive solution

A passive solution which is weight activated is presented in Figure 4.6.

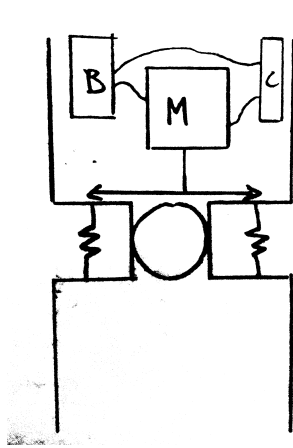


This solution is completely passive and relies on the weight of the person to change the length of an elastic component such as a nylon rope. The downward movement induced by the weight is transformed into an upward movement for the elastic element. This deformation will change the impedance of the system and the ankle joint will be more rigid during the stance phase when the weight is pushing on it. The rotation is done thanks to a rotoid joint.

Figure 4.6: Passive solution, frontal plane view

Semi-active solution

The last solution is a semi-active one, which is presented in Figure 4.7.



In this last solution, a small motor is controlled to move elastic elements such as springs. By changing the lever arm separating the elastic elements from the axis of rotation, the rendered stiffness of the system is changed. The ankle is rotating in the frontal plane thanks to a rotoid joint.

Figure 4.7: Semi-active solution, frontal plane view

4.3.2 Comparison

The combined solutions are now compared according to the specifications with objective criteria to select the best possible one. The criteria are the following:

- Robustness : the device should withstand or overcome adverse conditions or rigorous testing.
- Cost : The device should be low-cost, the additional cost should not exceed 50%.
- Weight : The device should be lightweight, no more than 200 grams.
- Bulkiness : The device should be contained within the volume of the biological limb.
- Complexity : The device should be as simple as possible.
- Autonomy : The device should last at least one full day.
- Aesthetic : The device should be unobtrusive and not easy to spot.

- **Functionality** : The device should be well suited to allow the specific movements.

The table with the comparison of the different solutions is presented in Figure 4.8.

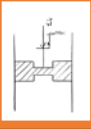
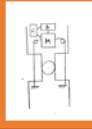
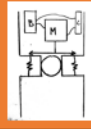
	1 	2 	3 	4 	5 
Robustness	0	0	-2	1	1
Cost	0	1	0	2	1
Weight	0	0	-1	2	1
Bulkiness	0	0	-2	1	1
Complexity	0	-1	-1	2	1
Autonomy	0	-2	-2	2	-1
Aesthetic	0	1	-1	2	2
Functionality	0	1	2	-1	1
Total	0	0	-7	11	7

Figure 4.8: Relative comparison of the different proposed solutions

Solutions are evaluated in a relative way, meaning that the first solution is taken as a reference and all the other solutions are compared to it with a scale from -2 to 2 . With a positive score meaning the solution is better suited for this criteria than the reference and vice-versa.

Passive and semi-active solutions, respectively solution 4 and 5, get the best scores. The constraints on the weight, bulkiness, complexity, robustness and autonomy being largely unfavorable to active solutions. The most promising solution seems to be a solution with a weight activated architecture taking advantage of the deformation of an elastic component to provide higher impedance during the stance phase.

4.3.3 Weight activated architecture

The choice of a weight activated architecture has been made and must now be further explored. This architecture should fulfill the design objectives, namely being able to adapt to uneven ground while providing stability thanks to correct values of mechanical impedance for the ankle joint. These two concepts can be treated separately.

Stability

In a sound ankle, the stability during the stance phase is provided by the mechanical impedance of the joint produced by the simultaneous contraction of different leg muscles. Here, it will be replaced by a nylon rope providing the desired stiffness and allowing deformations necessary for the required range of motion in inversion/eversion.

Adaptation to uneven ground

A necessary condition to achieve stability is the adaptation to uneven ground. To fulfill this objective the architecture must be able to change its neutral position depending on the slope of the terrain. This way, the patient's feet would be perfectly flat on the ground while allowing inversion/eversion motion.

First sketch

A first sketch implementing the two design objectives was made and is shown in Figure 4.9. In this first sketch, the mechanism is composed of two parts :

- The first one, the shank part, on the top can rotate around the axis of inversion/eversion but can also slide vertically when the weight of the patient is pushing on it through the pylon of the prosthesis.
- The foot part is fixed to the ankle-foot prosthesis and supports a shaft with a nylon rope wound around it.

This design allows the ankle to move freely when no weight is acting on it, and thus to adapt to the inclination of the terrain, since the shank part is in contact with the shaft via a sliding bearing. But once the weight of the patient is acting on the shank part, the latter slides down and the contact with the shaft is now done via a toothed wheel coupling the shank part and the shaft. The nylon rope wound around the shaft restricts the inversion/eversion movement and provides stability.

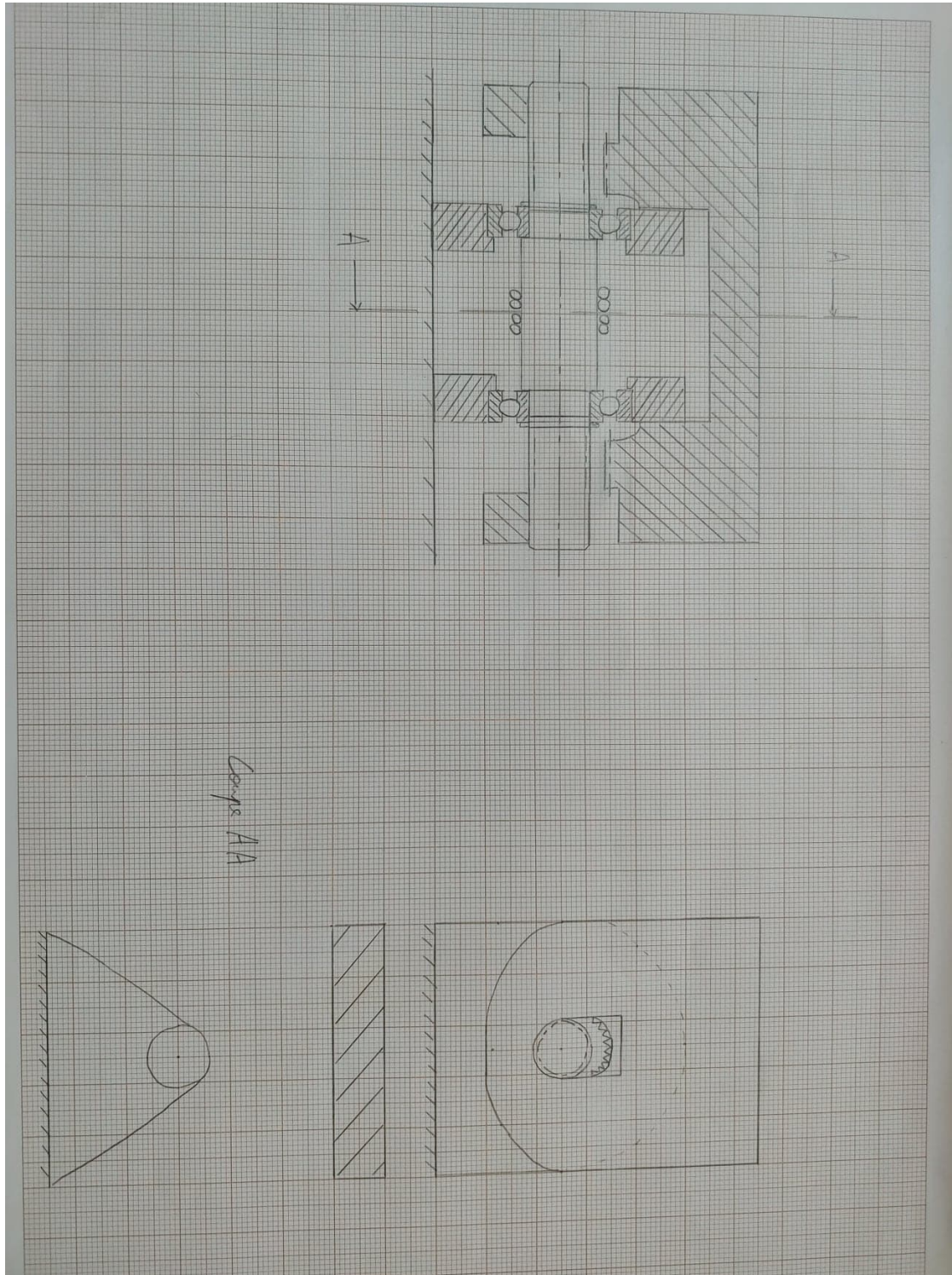


Figure 4.9: First sketch made for the weight activated architecture

This first sketch has clear vagueness like the assembly, the dimensioning of the main components and the guiding of the elements. However, it is a starting point for the architecture. A comparison of weight activated locking mechanisms coupling the shaft and the shank part must be done to select the most suitable one.

4.3.4 Locking mechanism

The listing and selection of the weight activated locking mechanisms is based on the review of locking mechanisms done by Lefeber and his team in 2015 [54]. Again a morphological chart containing all the possibilities was made and is shown in Figure 4.10.

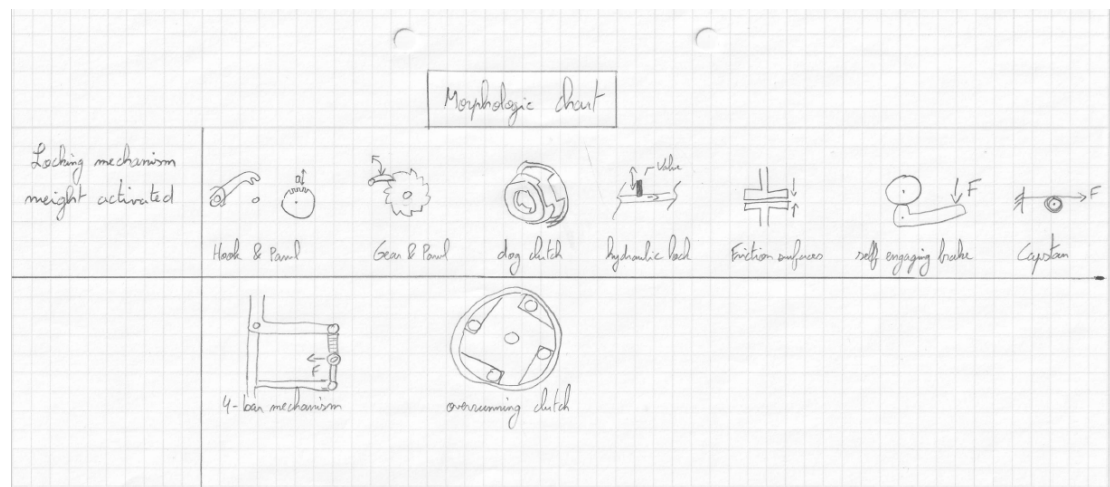


Figure 4.10: Morphological chart for the weight activated locking mechanisms

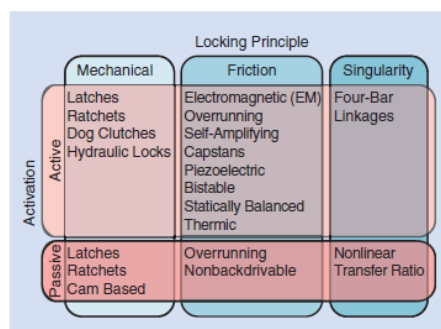


Figure 4.11: Classification of the locking devices by Lefeber *et al.* [54]

In this chart the listed solutions are the following : hook and pawl, gear and pawl, dog clutch, hydraulic lock, friction surfaces, self energizing brake, capstan, four-bar mechanism and the overrunning clutch. They can be categorized into three main locking principles : mechanical locking, friction-based locking and singularity locking. The classification is shown in Figure 4.11.

In our application, the activation of the mechanism is done by the weight of the patient. This way it is possible to use active locking devices without any motors.

The same procedure with combined solutions was followed and three locking mechanisms were chosen :

- Hook and pawl : the coupling of the shaft and the shank part is done by meshing of a hook and a pawl.
- Friction surfaces : a self energizing brake takes advantage of the weight of the patient. A brake shoe fixed to the shank part is pushed against the shaft linking the rotation of the two elements.
- A capstan : the coupling of the shaft with the shank part is achieved thanks to a rope wound around the shaft with its extremities attached to the shank part. The vertical movement induced by the patient's weight tightens the rope and couples the movements.

In order to compare properly the solutions, a first dimensioning of these is needed. The following section is devoted to it.

A first common parameter is the minimal diameter of the shaft to take up the effort. In the first instance, pure shear effort assumption is made for the shaft since it undergoes torsion due to the rope wound around it to provide stability during stance phase. According to the specifications and the data obtained by Ficanha the joint torque T_j generated in inversion/eversion is 0.27 Nm/kg . We can express the shear constraint inequality.

$$\sigma = \frac{F}{A} < S_y \quad (4.1)$$

Where σ is the shear stress and S_y is the yield strength of the material. We express the force F as $F = \frac{T_j}{r}$ and the area $A = \pi r^2$. We also take a security factor of 4 on the initial torque. Replacing F and A in the shear constraint inequality gives

$$r > \sqrt[3]{\frac{4T_j}{\pi S_e}} \quad (4.2)$$

According to the specifications the loading resistance must be calculated for a 100 kg patient which gives $T_{j,100} = 27 \text{ Nm}$. For a classical cast aluminium alloy 7075-T6 ($S_e = 505 \text{ MPa}$), this gives $r > 4.1 \text{ mm}$. Now that the minimum radius required for the shaft is known we can compute the dimensions of the other main elements.

4.4 Embodiment design

This section is devoted to the calculations related to the three possible locking solutions as well as the elastic element providing stability. The final objective is to be able to compare these solutions on the basis of objective criteria. For this, the orders of magnitude for the dimensions of the different parts are necessary. As the mechanism will be subjected to cyclical stresses caused by the gait cycle, a fatigue analysis is also performed.

4.4.1 Hook and pawl

The goal here is to calculate the thickness required for the pawl to take the effort without plastic deformation or failure. This dimensioning is similar as the one for spur gears, we will compare the results obtained with those found on catalog. On the KHK website the following gear corresponding to our specifications was found.[55]

Catalog Number	MSG1-36
Material	SCM415
Module	1
Number of Teeth	36
Pitch Diameter (mm)	36.0
Bore Diameter (mm)	12 with keyway
Bending Strength (Nm)	31.87
Surface Durability (Nm)	27.81
Precision Grade	JIS N5
Pressure Angle	20°
Tooth Finish	Ground
Heat Treatment	Overall carburizing
Tooth Hardness	55 - 60HRC
Weight	0.085kg

The characteristics of the MSG1-36 gear are shown in Figure 4.12. For the pawl we will start with the shear stress equation

$$\tau = \frac{F}{A} \quad (4.3)$$

In the appendix C10 of the Juvinall[56], the following characteristics can be found for the 7075-T6 cast aluminium alloy :

$$S_u = 570 \text{ MPa} \quad S_y = 505 \text{ MPa}$$

Let's take the same radius as the gear for the hook thus $r = 18 \text{ mm}$. With a safety factor of 2 on the loading torque, the force F acting on the pawl is:

$$F = \frac{T_j}{r} = \frac{2 * 27}{0.018} = 3000 \text{ N} \quad (4.4)$$

Figure 4.12: Characteristics of the MSG1-36 spur gear by KHK [55].

Since ambulation will produce cyclic efforts on the mechanism, the fatigue phenomenon must be taken into account. Based on the Figure 4.13, the fatigue strength for the selected material at 500 millions cycles is $S_n = 131 \text{ MPa}$. By taking $0.2 S_y = 101 \text{ MPa}$ we are sure to avoid fatigue failure. Equation 4.3 becomes:

$$101 * 10^6 \geq \frac{3000}{A} \quad (4.5)$$

$$A \geq 2.97 * 10^{-5} m^2 \quad (4.6)$$

The MGSA1-36 spur gear is 1 cm thick, by taking the same thickness for the pawl with a rectangular area we have $A = ab$ and we assume $a = 10 mm$ which gives

$$b \geq 2.97 mm \quad (4.7)$$

This value is a bit less than twice the tooth thickness of the MGSA1-36 which is $1.57 mm$ but the gear's material is different. The gear seems more interesting because of the smaller tooth thickness leading to a better resolution. Since the gear has 36 teeth, the resolution is

$$\theta = \frac{360^\circ}{36} = 10 \text{ degrees} \quad (4.8)$$

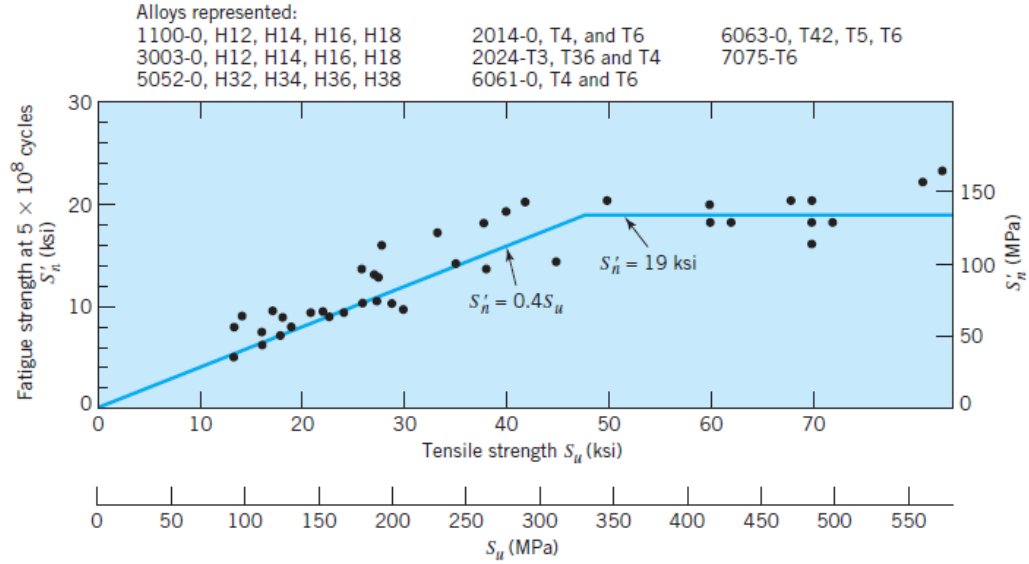


FIGURE 8.9
 Fatigue strength at 5×10^8 cycles versus tensile strength for common wrought-aluminum alloys.

Figure 4.13: Fatigue strength versus tensile strength for common wrought-aluminium alloys, Juvinall Chapter 8: Fatigue [56]

4.4.2 Clutch brake

For this solution, a friction surface on the shank part of the mechanism will be pushed against the shaft thanks to the patient's weight. This braking system similar to a clutch couples the shaft and the shank part. To avoid slippage the braking torque must be higher than the joint torque, a safety factor of 1.5 will be used.

The chapter 18 of the Juvinall[56] gives the equation for a situation similar to our case. It is illustrated in Figure 4.14.

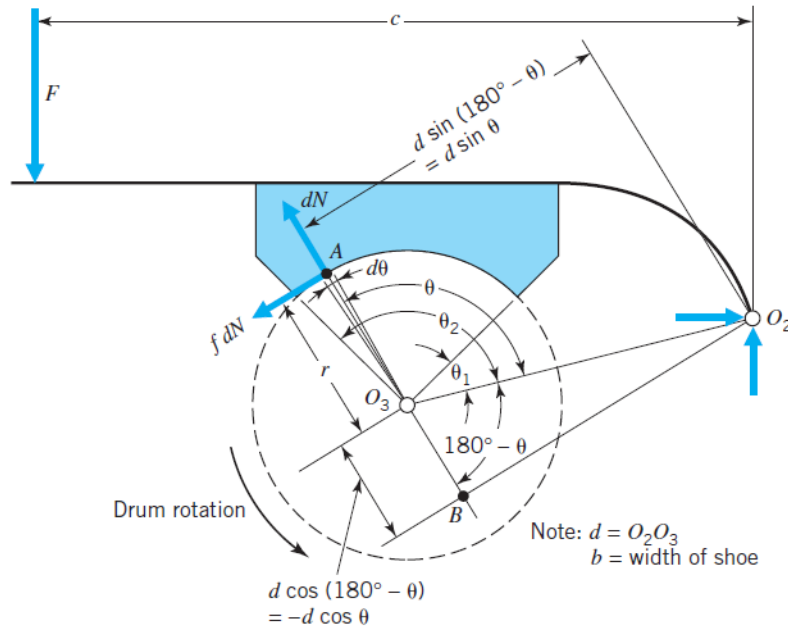


Figure 4.14: Forces acting on a brake shoe, Juvinall Figure 18.9[56]

The following equations are given for the moment of the normal forces M_n , the moment of friction forces M_f and the brake torque T .

$$M_n = -\frac{p_{max} b r d}{4(\sin\theta)_{max}} [2(\theta_2 - \theta_1) - \sin 2\theta_2 + \sin 2\theta_1] \quad (4.9)$$

$$M_f = \frac{f p_{max} b r}{(\sin\theta)_{max}} \left[r(\cos\theta_1 - \cos\theta_2) + \frac{d}{4}(\cos 2\theta_2 - \cos 2\theta_1) \right] \quad (4.10)$$

$$T = -\frac{r^2 f b p_{max}}{(\sin\theta)_{max}} (\cos\theta_1 - \cos\theta_2) \quad (4.11)$$

Our situation is slightly different since the shank part is sliding down vertically and is not rotating around a point O_2 . Also, the force is directly acting on the shoe. Two assumptions are made to change the model presented :

- the distance d tends to infinity, this way the rotating movement around the pivot O_2 becomes a vertical movement for the shoe.
- θ_1 and θ_2 can be taken from the horizontal since d tends to infinity.

Furthermore, we are in static so the maximum pressure p_{max} can be expressed as $p_{max} = \frac{F}{A} = \frac{W}{(\theta_2 - \theta_1)br}$. Where W is the weight of the patient and b is the width of shoe.

The braking torque becomes

$$T = -\frac{rfW}{(\theta_2 - \theta_1)}(\cos\theta_1 - \cos\theta_2) \quad (4.12)$$

The critical parameter in this equation is the radius of the drum, by isolating it in the equation we have

$$r = \frac{T(\theta_2 - \theta_1)}{(\cos\theta_1 - \cos\theta_2)fW} \quad (4.13)$$

If we set the following parameters:

- $\theta_1 = 55^\circ$ and $\theta_2 = 125^\circ$.
- Friction coefficient $f = 1.2$ (rubber shoe)
- 100 kg amputee's weight $W = 981N$.
- 100 kg amputee's joint torque $T_{j,100} = 27Nm$.

The braking torque must cancel the joint torque thus with the safety factor

$$T = 1.5 * T_j \quad (4.14)$$

The radius of the drum is

$$r = \frac{1.5 * T_{j,100}(\theta_2 - \theta_1)}{(\cos\theta_1 - \cos\theta_2)fW} = 36.6 \text{ mm} \quad (4.15)$$

It is important to note that this result does not depend on the patient's weight since the joint torque also depend on the patient's weight ($T_j = 0.27Nm/kg$).

For the fatigue analysis, the wear can be calculated thanks to equation 9.1a of the Juvinall [56].

$$W = \frac{K}{H}FS \quad (4.16)$$

Where W is the volume of material worn away (mm^3), F is the compressive force between the surfaces (N), S is the total rubbing distance (mm), K is the wear coefficient and H is the surface hardness (MPa).

The wear coefficient is obtained thanks to the figure 9.12 in the Juvinall (Figure 4.15), because we are in the case of abrasive wear with two bodies we take $K = 10^{-2}$. The surface hardness is estimated from value obtained by Shen *et al.*[57]. They estimated the surface hardness of nitrile rubber (NBR) as 72 Shore A which is equal to 401 Bhn and the conversion from Brinell hardness to MPa is done by multiplying by 9.81. H is equal to 3933.81 MPa. The compressive force is simply the weight of the patient (100 kg) and the total rubbing surface is estimated by making the assumption that it takes maximum 1° for the brake to stop completely when it is engaged. Thus $S = 0.036 * 1 * \frac{2\pi}{360} = 0.628 \text{ mm}$.

If we compute the wear for 10^6 cycles, we have

$$W = \frac{10^{-2}}{3933.81} \times 981 \times 0.628 \times 10^6 = 1566.045 \text{ mm}^3 = 1.56 \text{ cm}^3 \quad (4.17)$$

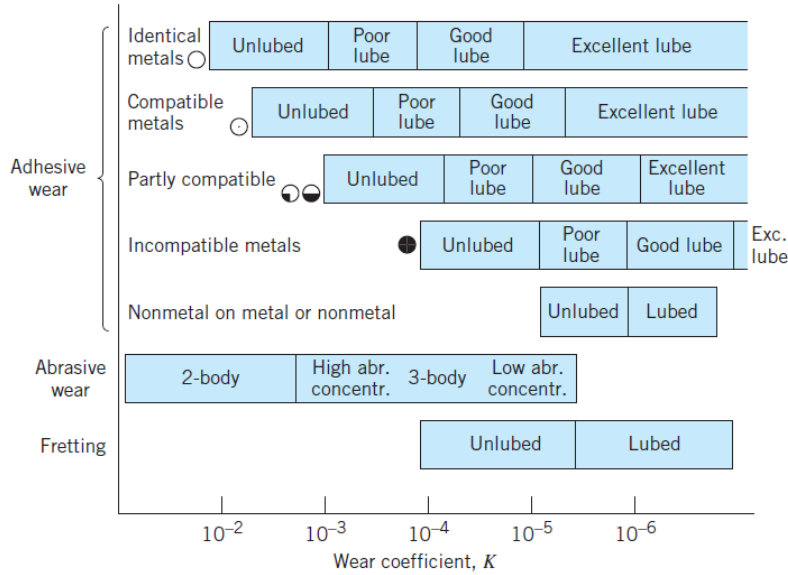


Figure 4.15: Figure 9.12 of the Juvinall [56], estimated wear coefficient for various sliding situations.

4.4.3 Capstan

In this section, the dimensions of both the elastic component used to render the impedance and the capstan locking mechanism are computed. We begin by describing the capstan effect and the elastic component.

When a rope is wound several times around a shaft, thanks to friction, the tension force in one end of the rope to sustain an heavy load acting on the other end is small. It is expressed by the following equation [58]:

$$T_{load} = T_{hold}e^{\mu\phi} \quad , \phi = 2\pi n \quad (4.18)$$

Where T_{load} is the load tension in one end of the rope, T_{hold} is the tension in the other end of the rope to sustain the load, μ is the friction coefficient between the rope's material and the shaft's material and n is the number of turns the rope makes around the shaft.

We will take advantage of the capstan effect to minimise the fastening force required for the nylon cable used to render the desired impedance through its deformation.

We have to compute T_{hold} , assuming the same radius for the shaft as the hook and pawl solution (6 mm) and the joint torque from the specifications ($T_{j,100} = 27$ Nm) we can first compute T_{load} with a safety factor on the load torque.

$$T_{load} = \frac{2 * 27}{0.006} = 9000N \quad (4.19)$$

If we make 3 turns around the shaft in each direction and the strand goes through the shaft inside a bore, knowing the friction coefficient between aramid and steel is 0.4 then

$$T_{hold} = \frac{T_{load}}{e^{\mu\phi}} = \frac{9000}{e^{0.4*2\pi*3}} = 4.78N \quad (4.20)$$

It requires only a 4.78 N tension force in the strand fixed, the strand could be for example glued inside the bore.

The range of motion given by the specifications is 23° in both directions. It means that when the patient's foot is in stance phase the patient should be able to lean his body thanks to inversion or eversion of his ankle. Since our mechanism uses a nylon rope wound around the shaft, it is its deformation that allows this movement.

We compute the deformation of the rope d_r caused by the movement:

$$d_r = \theta r = 23 * \frac{2\pi}{360} * 0.006 = 2.4 \text{ mm} \quad (4.21)$$

In the website of *Kanirope*[59], we find the following properties for a nylon rope corresponding to our needs.

- Diameter = 6 mm.
- Breaking strength = 610kg.
- Failure deformation = 16 to 27%.

We want the deformation d_r to be at most a 5% deformation of the rope, we can compute the required length L_r for our application

$$L_r = 20 * 0.0024 = 0.048 \text{ m} = 4.8 \text{ cm} \quad (4.22)$$

While nylon rope can be used to allow the movement and render a mechanical impedance, aramid rope can be used for the locking system since it has a much higher breaking strength and therefore very small deformations induced by the load. This property will be used in the locking mechanism to tighten the rope around the shaft when the patient's weight induced the displacement. The characteristics of an aramid rope corresponding to the specifications are the following [59]:

- Diameter = 3 mm.
- Breaking strength = 560 kg.
- Failure deformation = 2 to 4%.

4.4.4 Sketches

The 3 solutions presented in the previous sections are sketched with the size and dimensions computed for the main elements, see Figures 4.16, 4.17 and 4.18. Intermediate drawings were also previously made and are presented in the appendix, see Figures 8.4, 8.5 and 8.6. The guiding of the shank part is done by deformation of the material instead of a vertical sliding displacement to avoid blocking due to misalignment. The deformation needed is in a range of 2 mm to permit the locking. Such small displacement should be unnoticeable for the amputee wearing the device. Cantilever will be used because this technique is simple to implement and allows for large deformation.

The CAD software Solidworks will be used to finish the dimensioning and positioning of the different elements. Thanks to simulations with finite elements methods deformations will also be quantified. The assumptions made for material selection could also be checked.

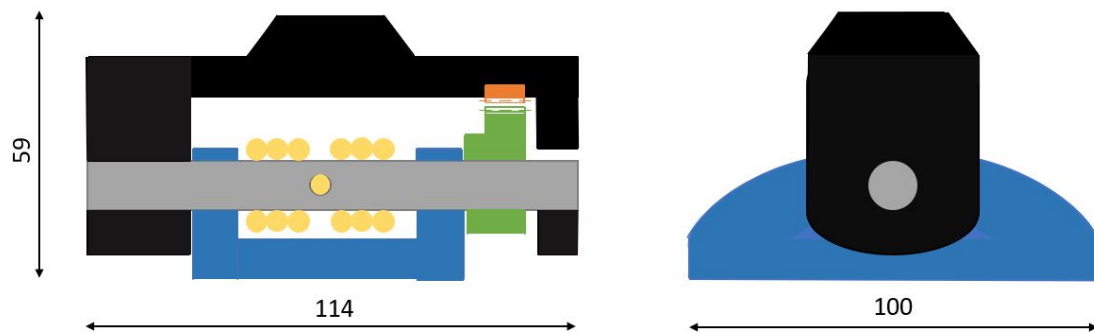


Figure 4.16: Schematic representation of the hook and pawl solution. Left: sectional view, Right: side view. (Blue) Foot part, (Grey) Shaft, (Black) Shank part, (Yellow) Nylon rope, (Green) partial gear, (Orange) Teeth for the locking.

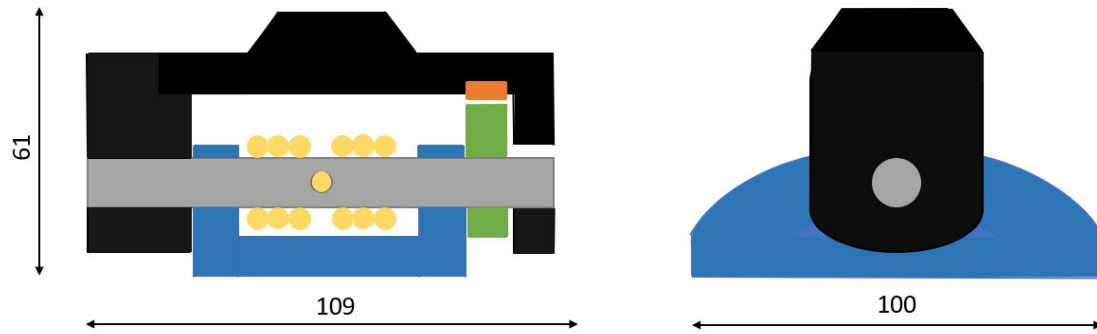


Figure 4.17: Schematic representation of the clutch brake solution. Left: sectional view, Right: side view. (Blue) Foot part, (Grey) Shaft, (Black) Shank part, (Yellow) Nylon rope, (Green) Friction gear, (Orange) Rubber brake shoe.

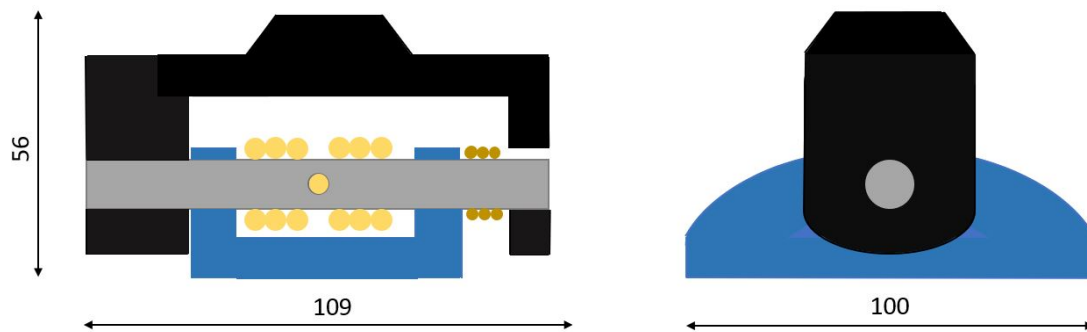


Figure 4.18: Schematic representation of the capstan solution. Left: sectional view, Right: side view. (Blue) Foot part, (Grey) Shaft, (Black) Shank part, (Yellow) Nylon rope, (Brown) Aramid rope.

Chapter 5

Numerical simulations

In this chapter, the three mechanisms are implemented in the CAD software Solidworks. Thanks to finite elements simulations the functionality, deformation and structural strength of the parts will be evaluated. Since the three solutions share the same actuation principle (cantilever) and global architecture, simulations will be performed for the hook and pawl solution only. The initial dimensions will be based on the drawings made but will be adapted according to the results of the simulations.

The three mechanisms are shown in Figures 5.1, 5.2, 5.3.

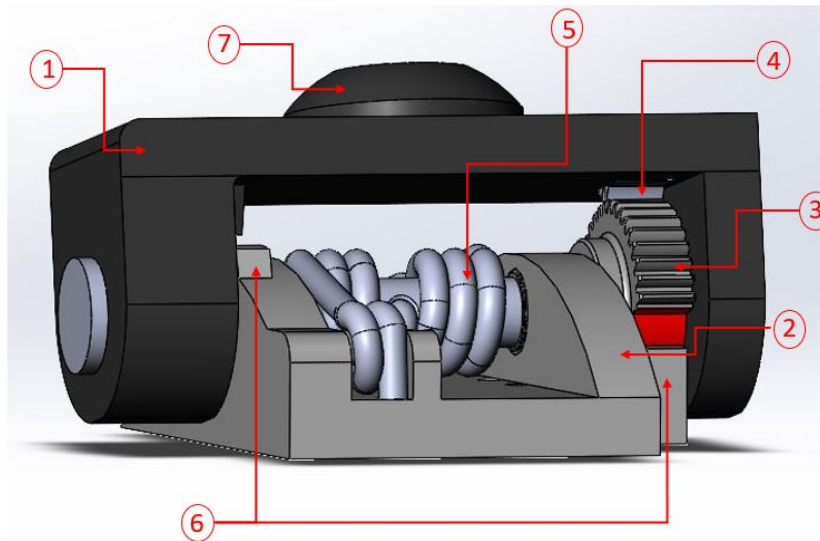


Figure 5.1: Assembly in Solidworks of the Hook and pawl solution. (1) Shank part, (2) Foot part, (3) Partial gear, (4) Locking teeth, (5) Nylon rope, (6) Mechanical stops, (7) Pyramidal adapter.

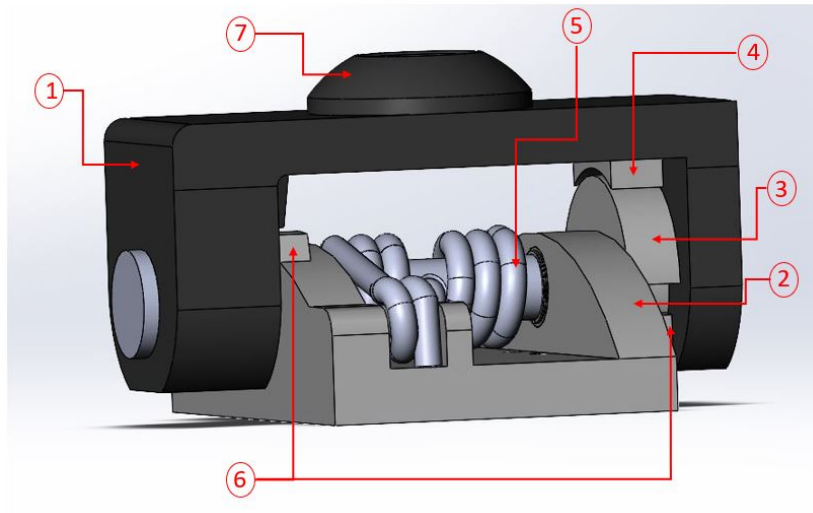


Figure 5.2: Assembly in Solidworks of the clutch brake solution. (1) Shank part, (2) Foot part, (3) Friction gear, (4) Rubber brake shoe, (5) Nylon rope, (6) Mechanical stops, (7) Pyramidal adapter.

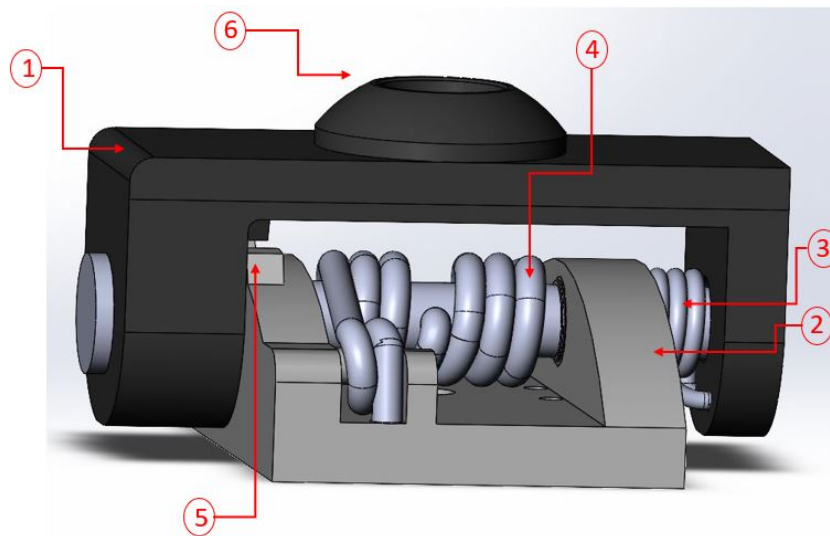


Figure 5.3: Assembly in Solidworks of the capstan solution. (1) Shank part, (2) Foot part, (3) Aramid rope, (4) Nylon rope, (5) Mechanical stop, (6) Pyramidal adapter.

5.1 Material selection

Some assumptions must be made for the material selection of the different parts. For the shank part an elastic material with a low Young's modulus is needed for the activation of the locking mechanism by deformation. In addition, we want to reduce as much as possible the manufacturing cost. For this reason, polymer such as the polyamide 6 which can be used for 3D printing seems interesting. The characteristics of the polyamide 6 provided by the Solidworks' library are the following (Figure 5.4):

Property	Value	Unit
Young's modulus	2,62	GPa
Poisson ratio	0,34	/
Shear modulus	970,4	MPa
Mass density	1120	kg/m ³
Tensile strength	90	MPa
Yield strength	103,64	MPa

Figure 5.4: PA type 6 properties, Solidworks' library

For the foot part fixed to the ankle foot prosthesis, a more robust material is needed. It is important to be aware that this part must sustain the load for the inversion/eversion but also for the plantarflexion/dorsiflexion. In the sagittal plane, most of the active ankle-foot prostheses provide a 120 Nm torque to propel the foot off the ground at the end of the stance phase [27, 34, 37]. For the frontal plane, there is a joint torque of 0.27 Nm/kg (Figure 8.1). The part must withstand both. For this reason, cast aluminium alloy 7075-T6 which has a high yield strength and Young's modulus will be selected. Its characteristics provided by the Solidworks' library are the following (Figure 5.5):

Property	Value	Unit
Young's modulus	72	GPa
Poisson ratio	0,33	/
Shear modulus	26,9	GPa
Mass density	2810	kg/m ³
Tensile strength	570	MPa
Yield strength	505	MPa

Figure 5.5: Cast aluminium alloy 7075-T6 properties, Solidworks' library

The same aluminium was previously selected for the shaft.

5.2 Finite elements methods

This section is devoted to the numerical simulations performed thanks to the CAD software Solidworks on the previously made CAD of the solutions. Two kinds of tests are performed:

- **Functionality.** First the feasibility of the actuation principle with the selected materials is checked. The amputee's weight is therefore applied to the shank part to check that the deformation achieved by the cantilever is sufficient.
- **Structural strength.** As the mechanism is subject to high stresses, it must be checked that the parts do not leave the elastic deformation zone. We will therefore apply the combination of the patient's weight and the torques in the frontal and sagittal planes to the parts and verify that the maximum stress does not exceed the yield strength of the material.

5.2.1 Functionality

Above all the functionality of the mechanism must be tested. The actuation principle relies on the deformation of the shank part to activate the locking mechanism. For the hook and pawl solution, the deformation must be slightly smaller than the tooth depth which is 2.25 mm. For the other solutions, the same range of deformation of 2 mm is required. Two extreme cases will be tested: for a patient of 50 kg and 100 kg. In the first case, the critical parameter will be the deformation that could be not sufficient due to the lighter weight. In the second case, the stresses on the parts could be too high and cause permanent deformation.

In Figure 5.7 are presented the results obtained for the simulation of the interaction of the shank part with the shaft in the case of the 100 kg patient. In the following simulations the shaft is fixed, the displacement of the shank part is restricted to the vertical plane and a bearing connection is used between the two on the left hand. It is illustrated in Figure 5.6.

The maximum Von Mises stress computed is 218.174 MPa. This stress is located on the left hand of the shaft and is caused by the shank part pushing on it because of the cantilever. Nevertheless, this stress remains twice as small as the yield strength of the aluminium selected for the shaft (505 MPa).

In Figures 5.8 and 5.9, the partial gear and the tooth are added to the assembly. The displacement of the shank part caused by a 50 kg load is shown. The maximum displacement is 1.854 mm and is located at the right extremity of the shaft. This displacement is sufficient for the meshing of the gear and the guiding is done by

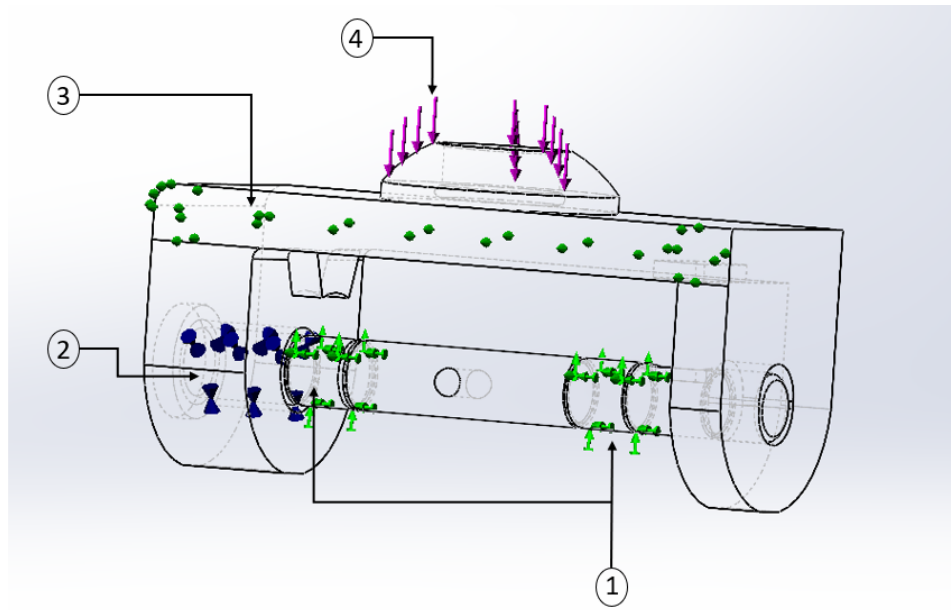


Figure 5.6: Parameters used for the simulation of the interaction of the shank part with the shaft. (1) fixed geometry, (2) bearing connection, (3) plane support, (4) load.

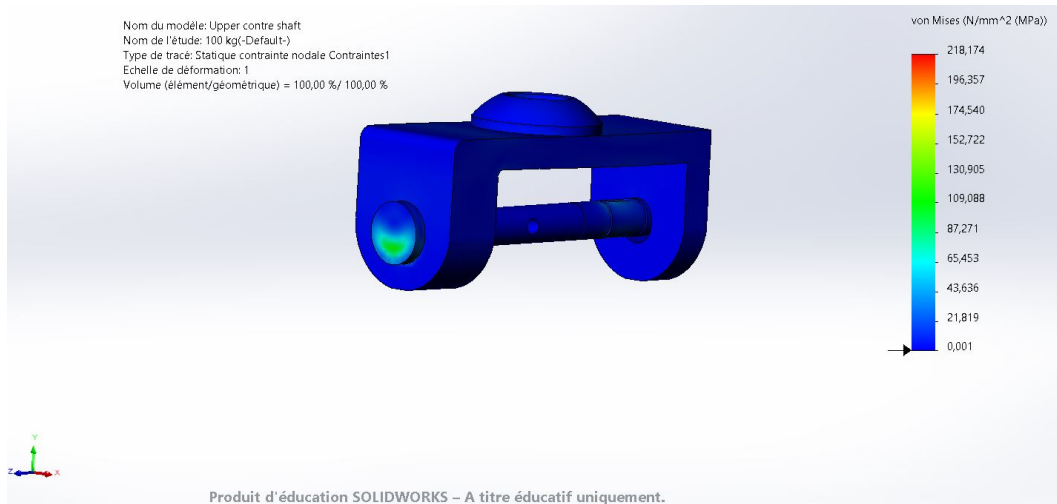


Figure 5.7: Von Mises stress on the shank part and the shaft caused by the 981N amputee's weight.

the elongated hole at the right end of the shank part.

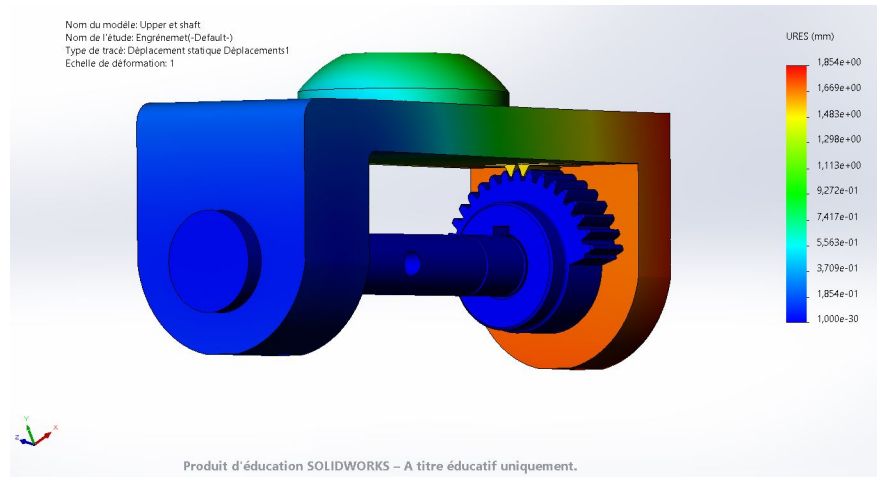


Figure 5.8: Displacements of the shank part with a 50 kg load applied on it causing the meshing of the gear.

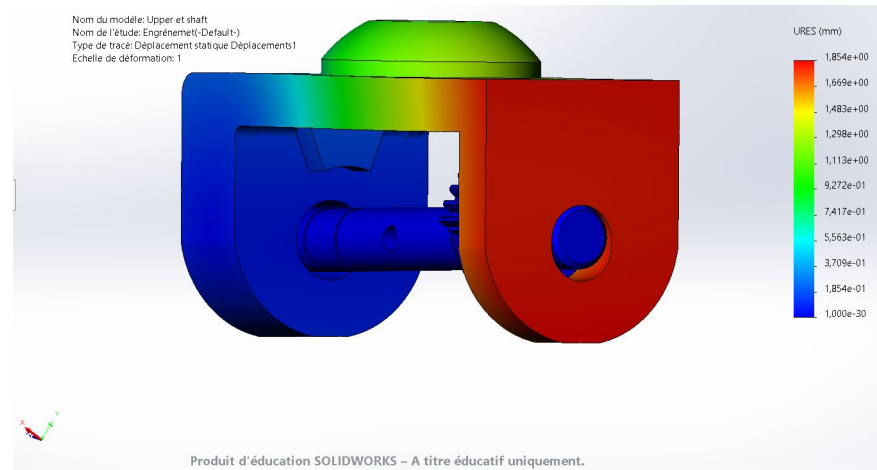


Figure 5.9: Displacements of the shank part with a 50 kg load applied on it guided by the elongated hole.

Discussion

The material selected for the shank part (i.e. polyamide 6) seems appropriate as it allows the desired deformation. As predicted, the displacement of 1.829 mm needed for the gearing is a bit smaller than 2 mm. Nevertheless, the yield strength and the fact that the movement is limited by the contact between the elongated hole and the shaft are sufficient to avoid plastic deformation. A cantilever is therefore appropriate to take advantage of the deformation to activate the mechanism.

5.2.2 Structural strength

Now that the feasibility of actuation has been demonstrated, the structural strength of the parts must be tested. Each of them must be able to withstand the forces and torques acting on them, namely the amputee's weight and the joint torques in the frontal and the sagittal planes. Due to technical limitations of Solidworks, only the shank part, the shaft and the foot part will be tested. The behaviour of the nylon rope will not be able to be tested.

Shank part

The first simulation performed on the shank part is the torsion caused by the joint torque in the frontal plane (0.27 Nm/kg). The hypothesis made, illustrated in Figure 5.10, are the following:

- The teeth's location is fixed to simulate the activation of the locking mechanism.
- Two bearing supports are used for the extremities of the part.
- The 27 Nm joint torque is applied at the pyramidal connection level.

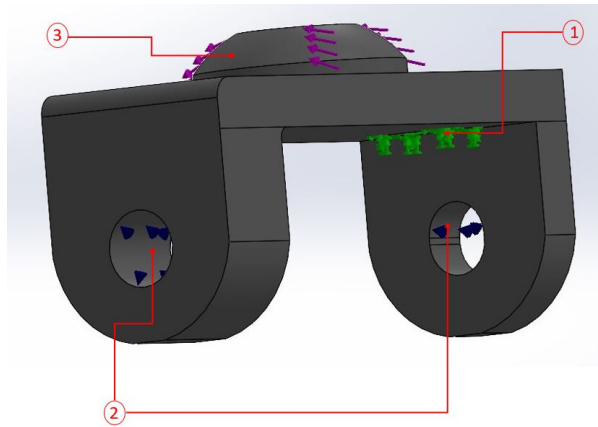


Figure 5.10: Parameters used for the simulation of the torsion of the shank part in the frontal plane. (1) Fixed geometry, (2) Bearing support, (3) Joint torque in the frontal plane.

The results of the simulation are shown in Figure 5.11. The maximum Von Mises stress computed is 43.081 MPa and this stress concentration is located at the corners of the tooth's location. In the rest of the part, the stress remains significantly below the yield strength (103.649 MPa).

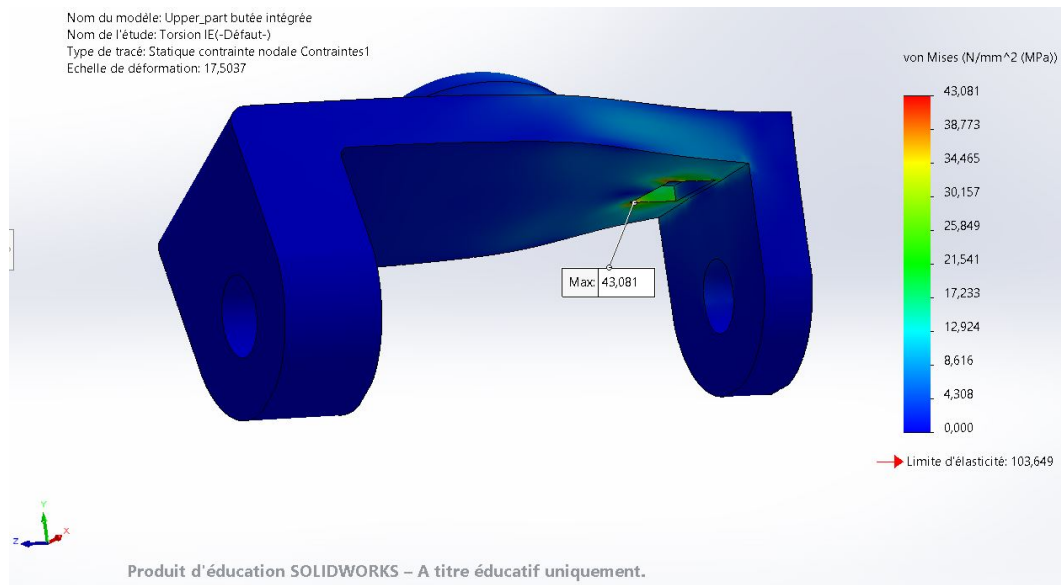


Figure 5.11: Von Mises stress on the shank part caused by the 27 Nm joint torque in the frontal plane.

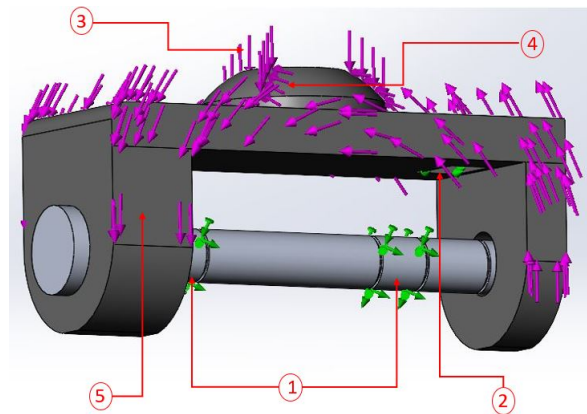


Figure 5.12: Parameters used for the simulation of the combination of efforts applied to the shank part. (1) Fixed geometry, (2) Plane support, (3) Amputee's weight, (4) Joint torque in the frontal plane, (5) Joint torque in the sagittal plane.

The next simulation looks at the effect of all the loads applied at the same time. The hypothesis are quite the same as in the previous case but the joint torque in the sagittal plane and the amputee's weight are added. Furthermore, the shaft is included because the torque in the sagittal plane reinforces the deformation and the locking effect. To simulate the locking displacement a plane support is added at the tooth's location (Figure 5.12).

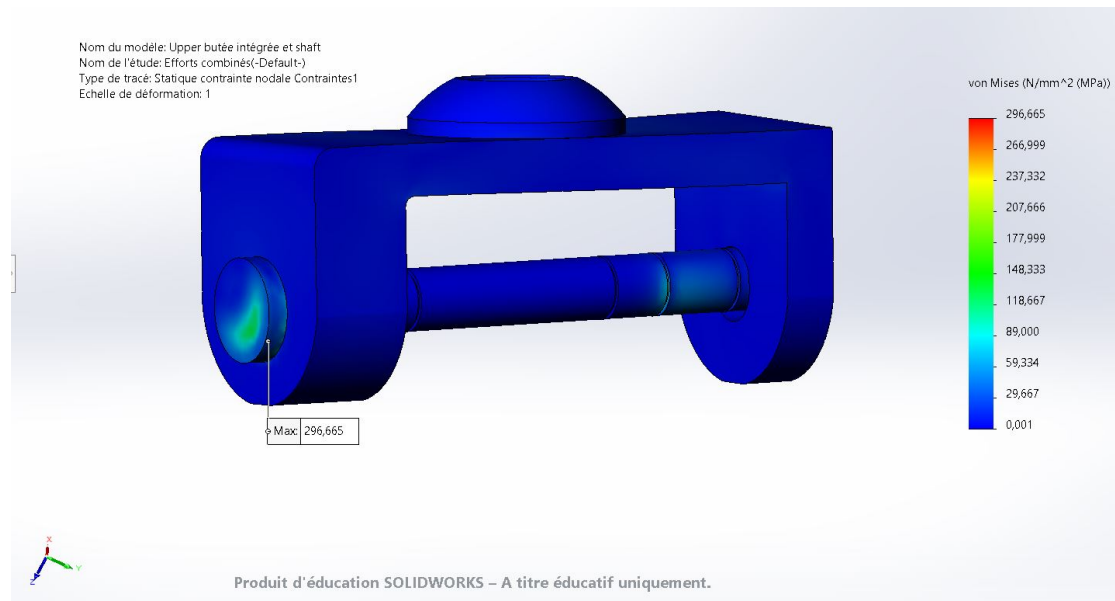


Figure 5.13: Von Mises stress on the shank part caused by the combination of the 27 Nm torque in the frontal plane, the 120 Nm torque in the sagittal plane and the 981 N amputee's weight.

The results presented in the Figure 5.13 show again a stress concentration at the left end of the shaft. The maximum value is greater than the previous simulation because of the torque in the sagittal plane applied on it and increasing the leverage effect. It reaches 296.665 MPa.

The last simulation for the shank part is about the mechanical stop. To do so, the part is held stationary while the joint torque is applied to the mechanical stop. It is illustrated in Figure 5.14.

The results shown in Figure 5.15 proves that the mechanical stop included into the shank part is able to withstand the joint torque in inversion/eversion. Indeed, the maximum Von Mises stress computed is a bit less than 40 MPa which gives a safety factor greater than 2.5.

Discussion

The simulations demonstrate that the shank part withstands the efforts. The actual geometry of the design causes stress concentration at the tooth's location and the end of the shaft. Nevertheless, it remains under the yield strength of the material and should not provoke plastic deformation.

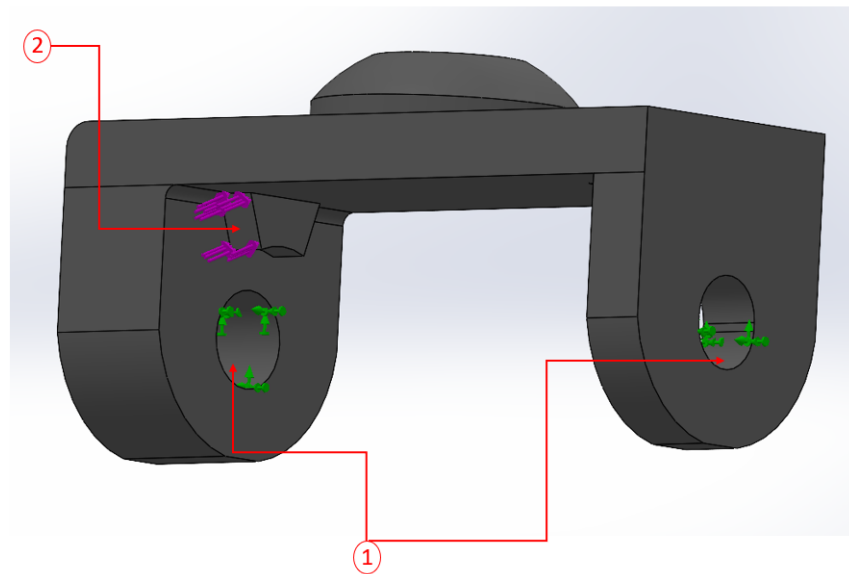


Figure 5.14: Parameters used for the simulation of the torque applied to the mechanical stop. (1) Fixed geometry, (2) Joint torque in the frontal plane.

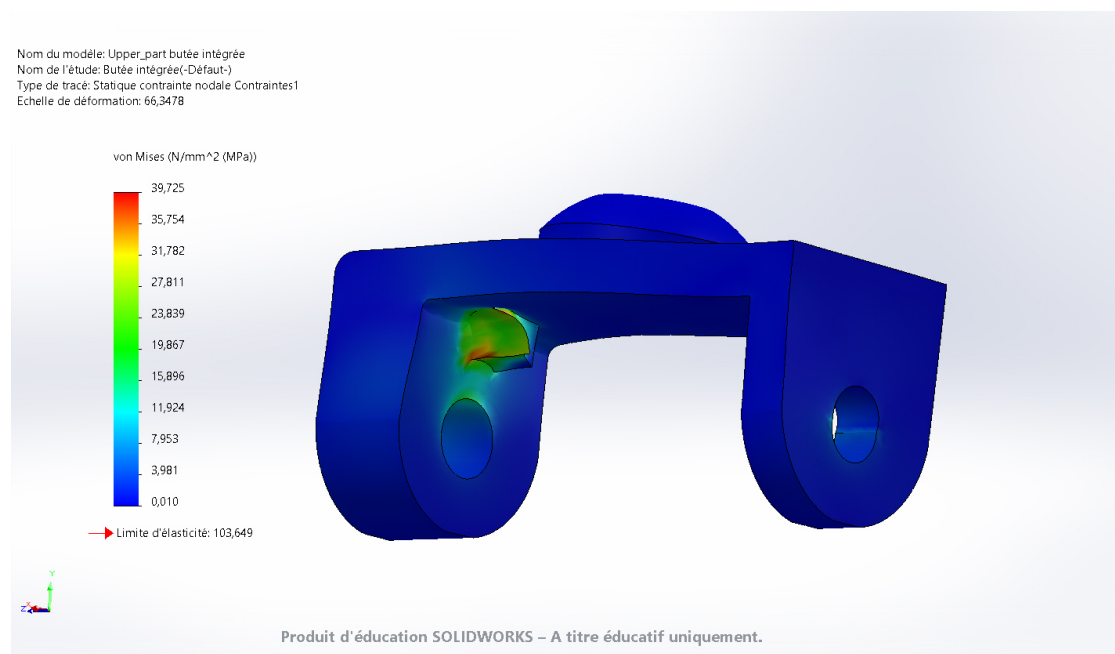


Figure 5.15: Von Mises stress on the mechanical stop of the shank part caused by the 27 Nm joint torque in the frontal plane.

Foot part

The same kind of tests will be performed on the foot part. For the frontal plane, the joint torque does not directly apply on the part but it creates a tension in the nylon rope that pulls on the anchor pins used for the support.

We start the simulations with the impact of this tension on the anchor pins used to wound the nylon rope before fixing it into the foot part. Knowing the diameter of the shaft and the joint torque in the frontal plane, we can compute the tension force in the strand.

$$F = \frac{T_{j,100}}{r} = \frac{27}{0.006} = 4500 \text{ N} \quad (5.1)$$

As shown in Figure 5.16, this force is applied horizontally towards the shaft to simulate the tension in the nylon rope. The foot part is held fixed for the simulation.

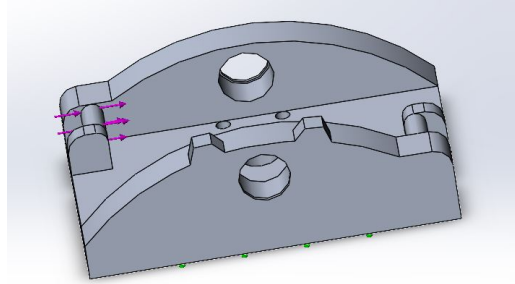


Figure 5.16: Parameters used for the simulation of the tension of the nylon rope on its support.

The result is presented in Figure 5.17, we can observe a maximum Von Mises stress value of 282.98 MPa located at the junction of the small shaft around which the cable is wound and the side of the part. Again, no plastic deformation should be observed since the computed stress remains under the yield strength.

Another important simulation which must be carried out is the impact of the combination of the torques in both the frontal and the sagittal planes. For this simulation, the underside in contact with the prosthesis it is attached to is fixed. We observe that the maximum Von Mises stress value occurs at the same location. In this case, the maximum Von Mises stress reaches 263.302 MPa.

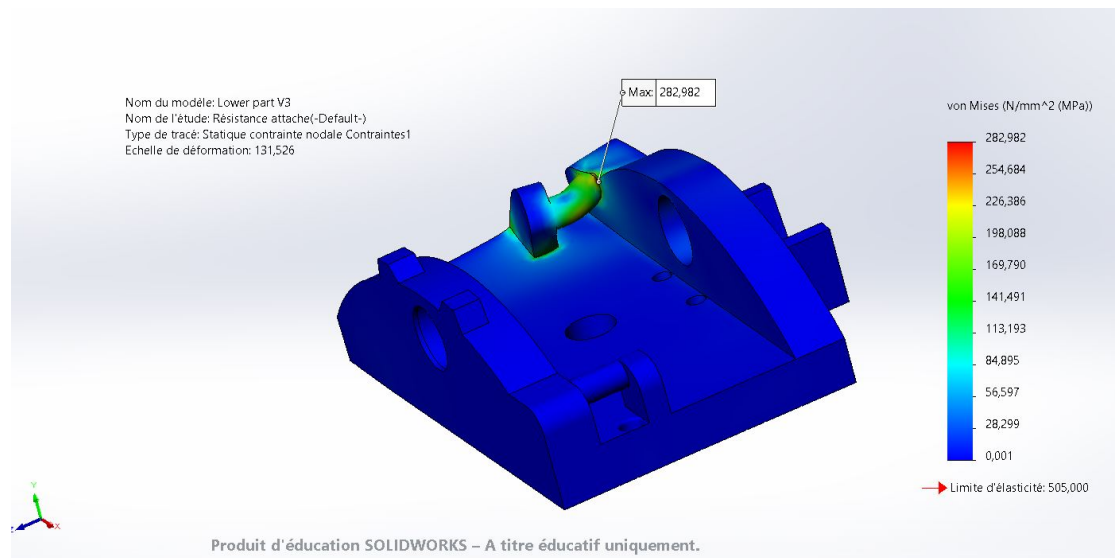


Figure 5.17: Von Mises stress on the support caused by the 4500 N tension in the nylon rope.

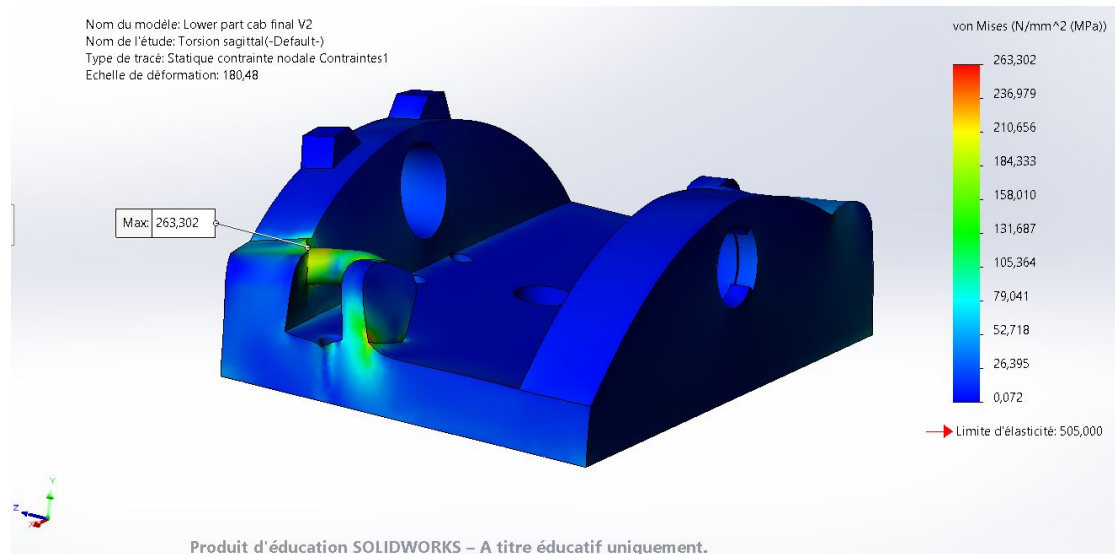


Figure 5.18: Von Mises stress on the foot part caused by the combination of the 27 Nm torque in the frontal plane and the 120 Nm torque in the sagittal plane.

As the inversion/eversion must be limited to 23° in both directions according to the specifications (Figure 8.1) mechanical stops are used. Both the shank part and the gear movement must be limited. Indeed, the rotation allowed by the deformation of the nylon rope is independent of the neutral position of the locking mechanism. The two simulated situations are illustrated in Figures 5.19 and 5.20.

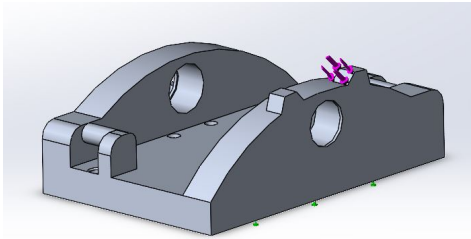


Figure 5.19: Parameters used for the simulation of the torque applied to the mechanical stop with the shank part.

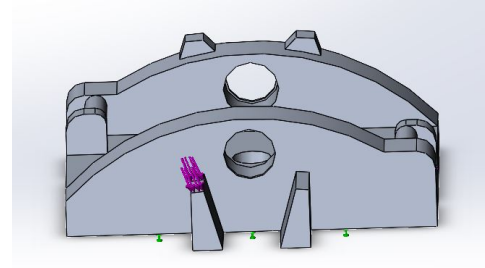


Figure 5.20: Parameters used for the simulation of the torque applied to the mechanical stop with the gear.

The results for both situations are shown respectively in Figures 5.21 and 5.22. For the mechanical stop limiting the shank part's movement, the maximum Von Mises stress is 64.898 MPa. This value is well below the yield strength of the material and ensures that there is no plastic deformation with a safety factor greater than 7.

For the mechanical stops limiting the gear's movement the maximum Von Mises stress value is much higher and reaches 166.27 MPa. Since it is less supported by the part, the mechanical stop is subjected to a much higher shear stress trying to tear it off. Nevertheless, the part is able to withstand the efforts.

Discussion

The simulations performed on the foot part show that the choice of the cast aluminium alloy 7075-T6 was relevant and permits to withstand the stresses applied to it. The biggest stress applied to it results from the tension in the nylon rope that pulls on its fixation. But a safety factor of approximately 1.78 is obtained for that specific case.

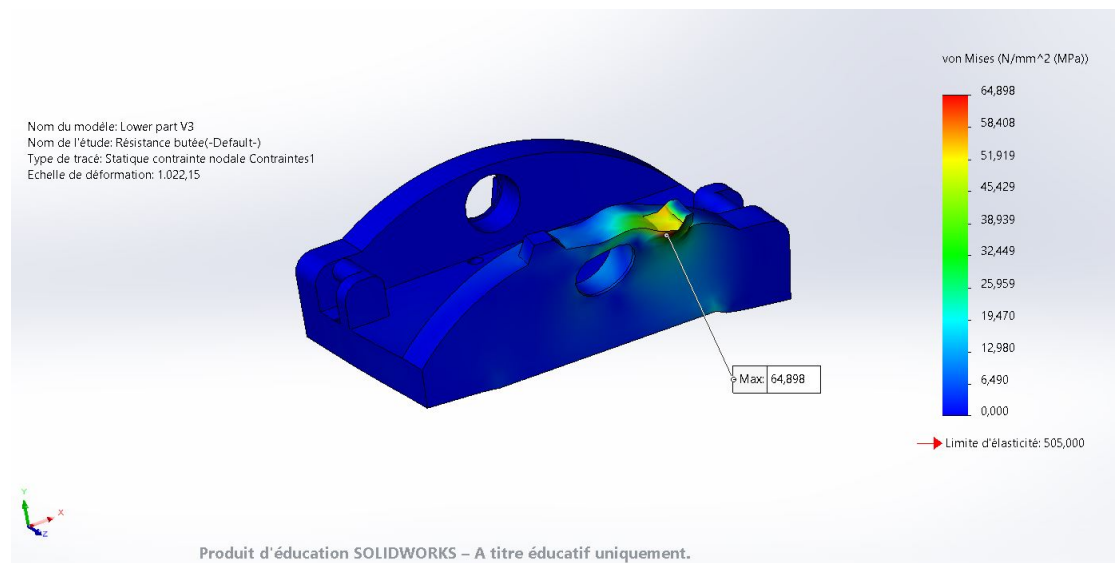


Figure 5.21: Von Mises stress on the mechanical stop caused by the 27 Nm torque in the frontal plane.

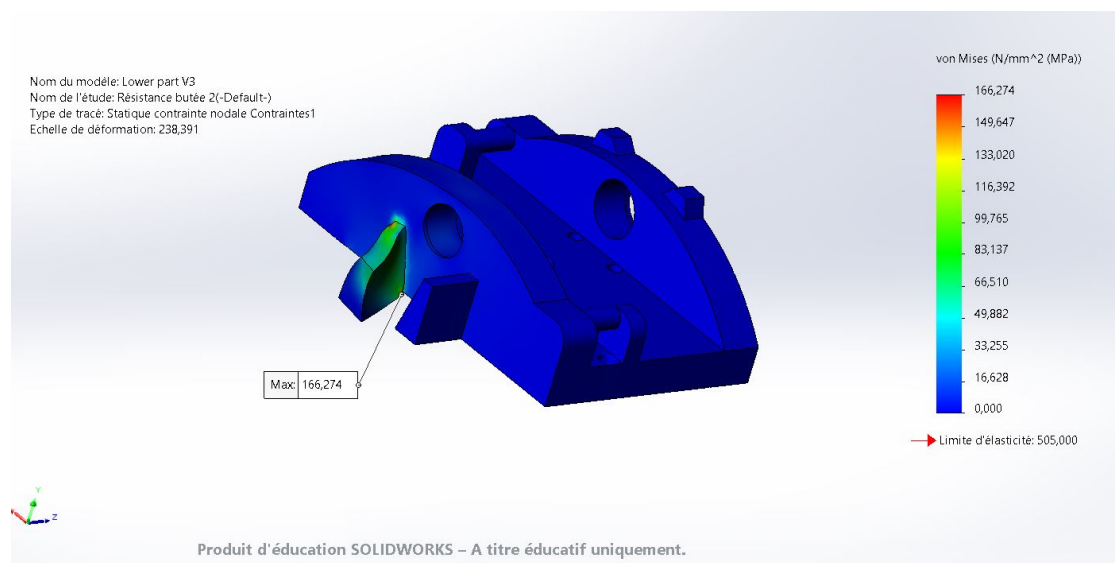


Figure 5.22: Von Mises stress on the mechanical stop caused by the 27 Nm torque in the frontal plane.

Shaft

Finally, the assumptions for the shaft material selection must be checked. The cast aluminium alloy 7075-T6 was selected. Its structural strength must be sufficient to sustain the torsion in the frontal plane and the bending in the sagittal plane.

Due to the nylon rope wrapped around it, the shaft undergoes a torsional stress. This is simulated by fixing the location where the rope is wound since it maintains the shaft in its neutral position and by applying the joint torque at the location where the gear is attached since it transmits the effort from the shank part to the shaft. It is illustrated in Figure 5.23.

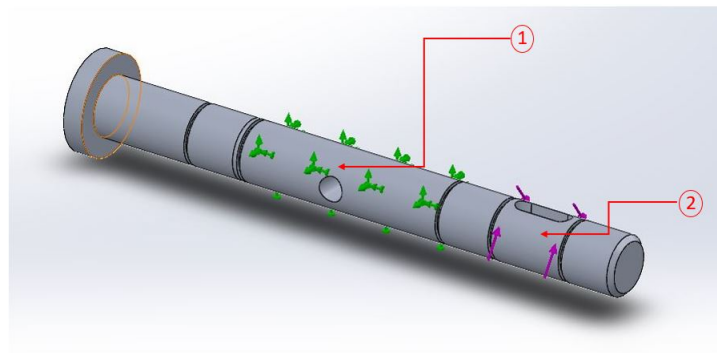


Figure 5.23: Parameters used for the simulation of the torsion of the shaft caused by the torque in the frontal plane. (1) Fixed geometry, (2) Torque.

The results are shown in Figure 5.24, a high maximum value of 389.79 MPa for the Von Mises stress is observed at the grooves intended for the circlip maintaining the bearing. This stress is dangerously close to the yield strength. In order to avoid it, another solution for the guiding of the bearing should be found. For example, flanged bearings could be used to avoid having to use circlips.

Finally, the combined efforts are simulated. The amputee's weight is distributed between the two support points and the bending stress caused by the torque in the sagittal plane is applied at the connection between the shaft and the foot part. A bearing-type support is used to simulate the bearings supporting the shaft. It is illustrated in Figure 5.25.

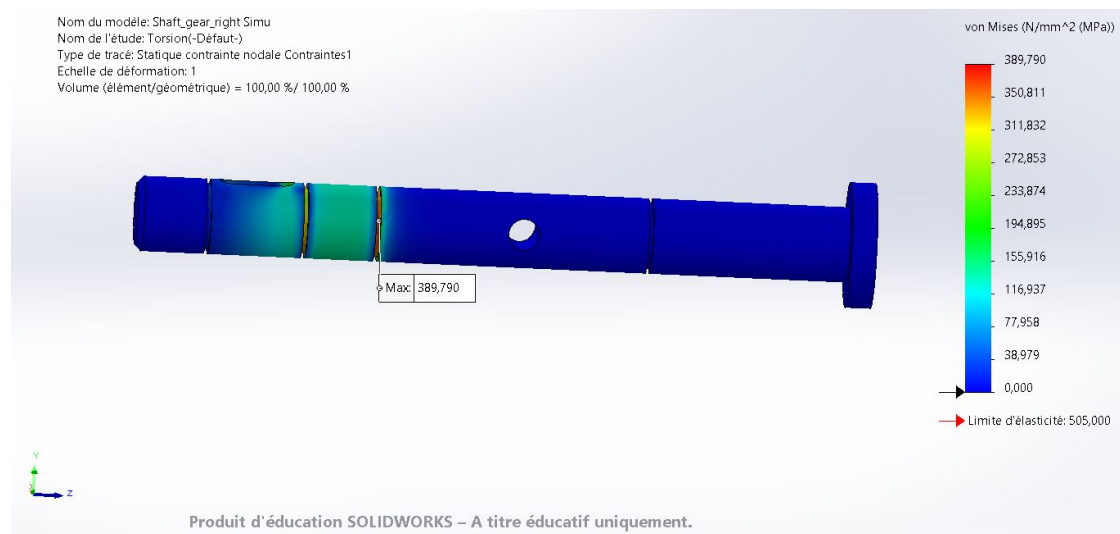


Figure 5.24: Von Mises stress on the shaft caused by the 27 Nm torque in the frontal plane.

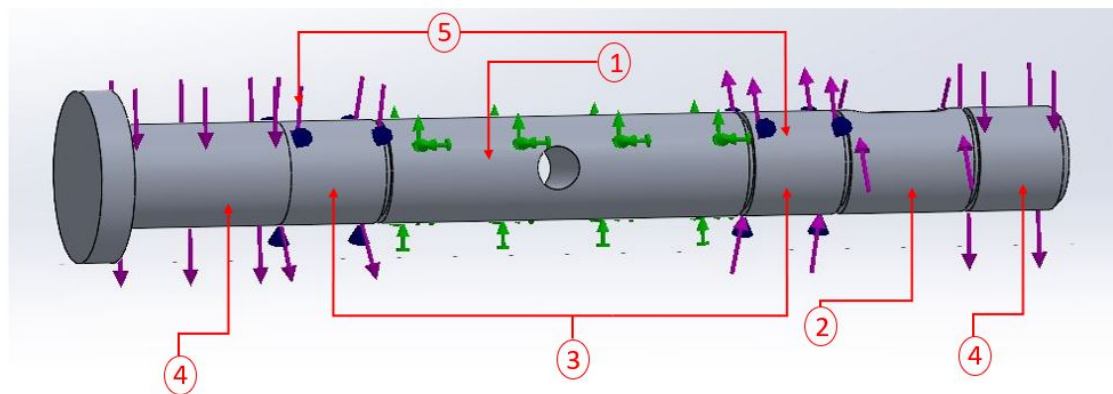


Figure 5.25: Parameters used for the simulation of the combined efforts applied on the shaft. (1) Fixed geometry, (2) Torque in the frontal plane, (3) Bearing-type supports, (4) Amputee's weight, (5) Torque in the sagittal plane.

The results are shown in Figure 5.26, again a stress concentration is observed in the grooves intended for the circlip. The Von Mises stress value is 465.796 MPa. It remains under the yield strength of the material but with a very small safety factor.

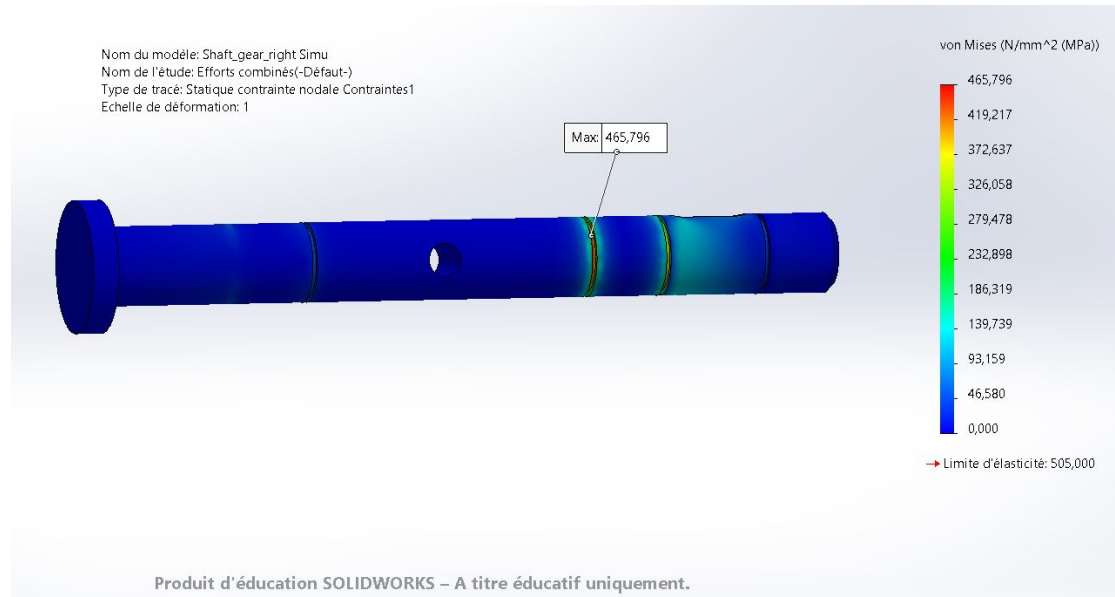


Figure 5.26: Von Mises stress on the shaft caused by the combination of the 27 Nm torque in the frontal plane, the 120 Nm torque in the sagittal plane and the 981 N amputee's weight.

Discussion

The simulations performed for the shaft proved that the cast aluminium alloy 7075-T6 is appropriate to sustain the efforts. However, the current guiding of parts using circlips causes stress concentrations in the grooves intended for their fixation. This results in stresses that are too close to the yield point and might cause plastic deformation due to low safety factors. To avoid this, alternatives for guiding the parts should be considered.

Chapter 6

Final choice

Now that the feasibility of the actuation principle as well as the structural strength has been demonstrated, a choice must be made between the three different options. As with previous choices made, this will be done on the basis of objective criteria to determine the best solution. First, the selection of the solution is made and then a brief description of its implementation will be given.

6.1 Comparison

A choice must be made between the three locking possibilities namely the hook and pawl, the clutch brake and the capstan. The following criteria will be used to assess and compare them:

- Lockable in any position: The device should be able to adapt its neutral position to any type of inclination provided by the specifications.
- Compact and lightweight: The device should be as compact and light as possible since the weight and the bulkiness are disincentives for the use of prosthesis.
- Wear resistance: The device must not wear out too quickly. In addition, it must not prevent the mechanism from functioning properly.
- Robustness: The device should withstand or overcome adverse conditions or rigorous testing.

In the Figure 6.1 is presented the comparative table. The solutions are compared in a relative way with the hook and pawl solution being the reference. The scores range from -2 to +2 with a positive score meaning the solution performs better for this criterion than the reference and vice-versa.

	Hook and pawl	Clutch brake	Capstan
Lockable in any position Weight = 3	0 0	2 6	2 6
Compact & Lightweight Weight = 2	0 0	0 0	2 4
Wear resistance Weight = 1	0 0	-2 -2	-1 -1
Robustness Weight = 1	0 0	-1 -1	-1 -1
Total	0 0	-1 3	2 8

Figure 6.1: Relative comparison of the three locking mechanisms.

For the first criterion, both the clutch brake and the capstan solutions perform better than the hook and pawl since they can be locked in any position belonging to the desired range of freedom. On the contrary, the hook and pawl solution's resolution is limited by the number of teeth on the gear which gives in the case of the MSGA1-36 spur gear a resolution of 10° . The value set by the specifications being 23° in each direction, it means only two possible locking positions in each direction in addition of the vertical position.

For the second criterion, the two first solutions are comparable in terms of weight and size. The capstan solution is lighter thanks to the aramid rope which has a very good weight to strength ratio.

For the third criterion, as presented in section 4.4.2 the brake shoe will be subjected to wear and will therefore probably require special attention during maintenance. For the capstan, the aramid rope in contact with the shaft will also experience friction. Regarding the hook and pawl, the poor resolution will result in impacts between the teeth during the meshing. This could have dramatic consequences on the behaviour of the mechanism.

Finally, the clutch brake solution is less robust than the hook and pawl because an uneven wear or a lack of grip for the brake shoe could lead to a malfunction. The capstan is also less robust as the cable winding could slip and overlap.

Given the small difference between the scores obtained, weights are associated to each criterion according to their importance. The main goal of this work being to add the inversion/eversion degree-of-freedom for ankle-foot prosthesis, the biggest weight was accorded to the capacity of the locking mechanism to adapt its neutral position. Just after that comes the compactness and weight that is the main constraint stated by amputees.

With the addition of the weights, the gap between the scores widens and the capstan solution gets the best final score. Its main advantages are the weight to strength ratio and the compactness. Therefore, it is the capstan that will be chosen as locking mechanism.

6.2 Description of the final design

This section will be devoted to the description of the final design chosen and the main elements that constitute it.

A global view of the mechanism is shown in Figure 6.2. The shaft (10) goes through the shank part (1) and the foot part (4) and is supported by four needle bearings. Two bearings are placed at the cantilevered end of the shank part and the two others in the foot part. A nylon rope (5) is glued inside the shaft and make three turns in each direction before making a last turn around its mounting support (6) to take advantage of the capstan effect and reduce the holding force needed to fix it. The end of the strand is glued inside a hole in the foot part. An aramid rope (8) also makes three turns around the shaft and is fixed inside the shank part (9) to form the locking mechanism which couples the shaft and the shank part when the amputee's weight is acting on it. The pyramidal adapter (2) which is typically fixed on top of prostheses is directly included in the shank part. Mechanical stops (3) are used to limit the range of motion. In addition, as the locking system can be engaged from any position the motion of the shaft must also be restricted to the expected range of motion. Otherwise, the deformation of the nylon cable would exceed that for which it was designed. It is done by using the head of a screw and mechanical stops (7) included in the foot part.

The figure 6.3 shows a view from below of the solution. On the foot part, holes (1) are included to attach the mechanism to a prosthesis in the place provided for the pyramidal adapter. The fixation of the nylon rope (2) is also visible.

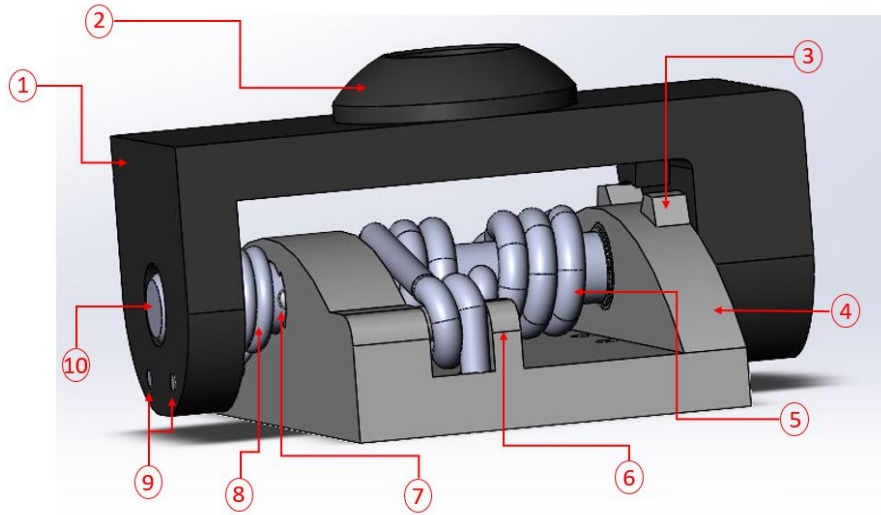


Figure 6.2: Global view of the CAD of the final solution. (1) shank part, (2) Pyramidal adapter, (3) Mechanical stop, (4) foot part, (5) Nylon rope, (6) Mounting bracket for the nylon rope, (7) Mechanical stop, (8) Aramid rope, (9) Fixation of the aramid rope, (10) Shaft.

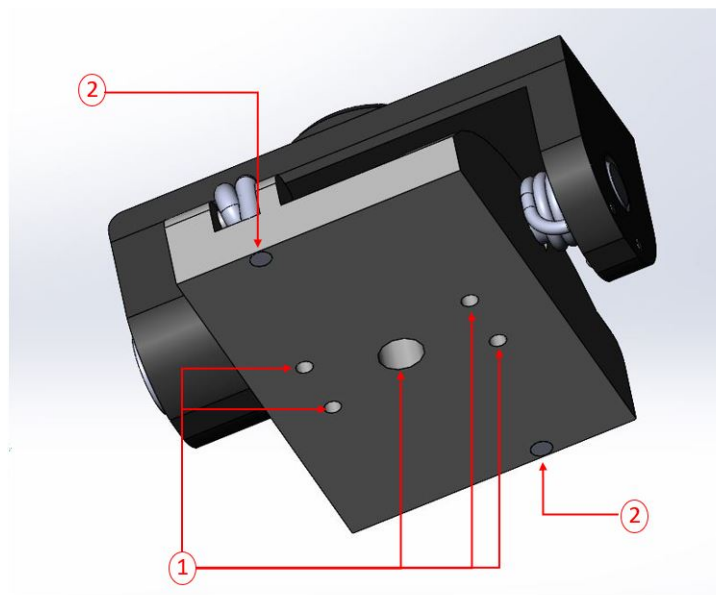
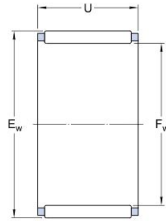


Figure 6.3: View from below of the CAD of the final solution. (1) Holes for mounting, (2) Fixation of the nylon rope.

The needle roller bearings selected are the K 12X15X10 TN available from SKF [60]. Their dimensions are the following:



DIMENSIONS

F_w	12 mm
E_w	15 mm
U	10 mm

Figure 6.4: Dimensions of the K 12X15X10 TN needle roller bearing. Screenshot from SKF website[60].

The bearing selection process described on the SKF website and the diameter of the shaft were used for the selection.

For the guiding of the bearings two circlips are used. They are placed against the inside of the foot part. The bores in the foot part contain a flange on the outside to hold the bearings in place. The circlips were selected on the basis of the axial effort produced by the amputee's weight when the foot is in maximal plantar flexion. ES-12 circlips from Smalley [61] were selected according to the diameter of the shaft and the axial efforts. Their dimensions are presented in Figures 6.5 and 6.6.

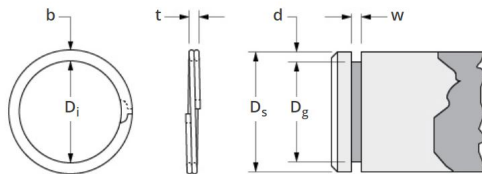


Figure 6.5: Dimensions of the ES-12 circlip. Screenshot from Smalley website [61].

	Value	Unit
D_s	12	mm
D_g	11,29	mm
W	0,7	mm
D_i	11,18	mm
t	0,6	mm
b	1,14	mm
Ring shear	7950	N

Figure 6.6: ES-12 circlip dimensions

A washer is used between the cantilevered shank part and the foot part to limit friction.

For the fixation of the nylon rope, thanks to the cabestan effect the holding force needed to fix after the turn made around the mounting support is only

$$F_{hold} = \frac{T_{load}}{e^{\mu\phi}} = \frac{4500}{e^{0.4*2\pi}} = 364.5 \text{ N} \quad (6.1)$$

	Weight [g]
Shank part	100,2
Foot part	291,8
Shaft	36,7
Bearings	12
Ropes	10
Total	450,7

An estimation of the mechanism's weight is made based on the information available on the elements' website or with the volume computed by Solidworks for the parts and the density of the material selected. As shown in Figure 6.7, the total estimated weight is 450.7 grams.

The final dimensions are presented in Figure 6.8. The width is 10 cm, the length is 11.3 cm and the height is 6.1 cm including the pyramidal adapter.

Figure 6.7: Weight break-down of the final solution

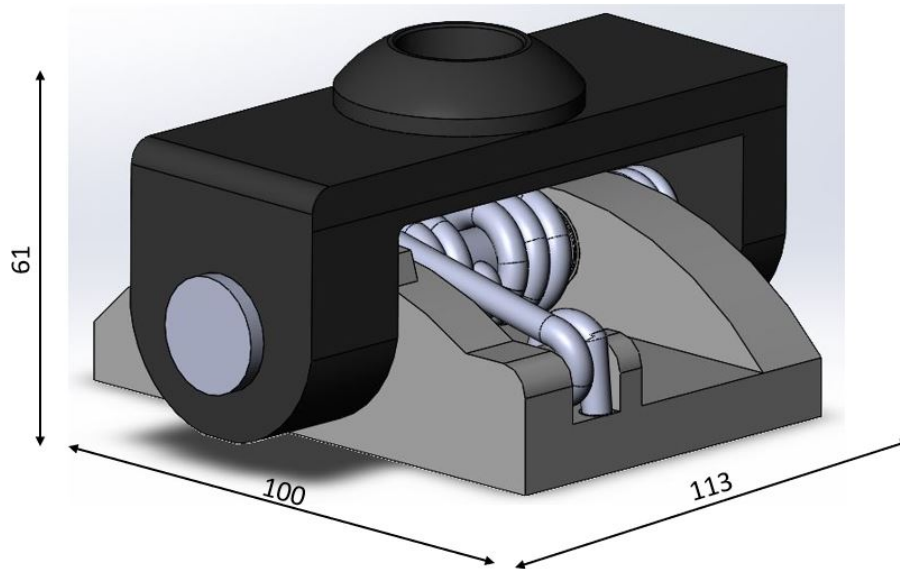


Figure 6.8: Final dimensions of the mechanism.

In order to see how the mechanism could be fixed to existing active prostheses, the active prosthesis ELSA [37] designed by François Heremans is used. It gives a better idea of what those dimensions actually represents for an ankle-foot prosthesis. The CAD view of the mechanism on top of ELSA is shown in Figures 6.9 and 6.10



Figure 6.9: Global CAD view of the mechanism fixed on top of ELSA, an existing active prosthesis[37].

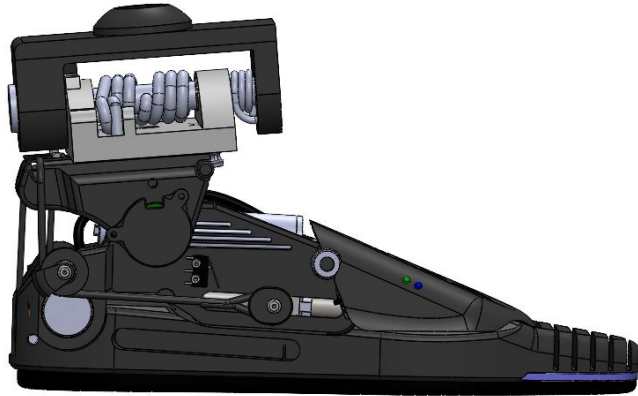


Figure 6.10: Side view of the CAD of the mechanism fixed on top of ELSA, an existing active prosthesis[37].

6.3 Assembly

This last section will describe how the mechanism can be assembled and how the different parts could be made.

To assemble the mechanism, the shaft is made to slide through the various bores, adding the guiding elements as we go along. As it goes through the cantilever, the two first bearings and the washer must be added. Then it goes through the foot part where the bearing is already placed in the bore and the circlips are prepared around the shaft. Before going through the other bore of the foot part, the last bearing is placed around the shaft and the screw is fixed. Then we can completely slide the shaft through the remaining bores and fix the circlips inside their respective grooves. Finally the nylon and the aramid ropes can be wound around the shaft and glued in their respective holes.

For the manufacturing of the different parts:

- The shank part could be 3D printed by classical methods of Selective Laser Sintering (SLS) since the polyamide 6 is a commonly used material in SLS.
- The foot part could be molded with the cast aluminium alloy 7075-T6.
- The shaft could be manufactured with a parallel lathe.

The rest of the elements, namely the bearings, the circlips and the ropes can be found on the website of the suppliers.

Chapter 7

Conclusion, Limitations and perspectives

The design procedure followed resulted in a first iteration of a solution. A feedback loop should be carried out to ensure that the solution best meets the specifications. In the next section, the limitations of the current device will be highlighted and perspectives for further iterations will be given. Finally, a conclusion for this work is provided.

7.1 Limitations and perspectives

If we refer to the specifications (Figure 8.1), we notice that some criteria and constraints are not respected.

The additional weight the mechanism could add to the existing prosthesis to which it is fixed was set to 200 grams. However, this first iteration ended up with a solution weighting approximately 450 grams. This is more than twice the level decided in the specifications. Future work should focus on finding materials that keep the weight to a minimum while resisting the efforts.

Furthermore, the prosthesis dimensions were supposed to be the one of a biological limb but the width of the mechanism is too large. It is well illustrated in Figures 6.9 and 6.10. This width was needed to have the right length for the deformation of the nylon rope which is about 4.8 cm. However, instead of having a general solution that can be attached to a wide range of active prostheses the choice of a solution specific to one prosthesis could be made. This way, the width could potentially be reduced by taking advantage of the geometry of the prosthesis and by using its lateral faces to fix the nylon rope for example. Furthermore, the

height could also be reduced by merging the foot part of the mechanism with the top of the selected prosthesis.

No experiment or simulation could be carried out to validate the performance of the prosthesis via the capstan effect. Indeed, to the best of the author's knowledge it is not possible to simulate the elastic deformation of a rope wound around a shaft in Solidworks. No data is therefore available to quantify the impedance obtained via the nylon rope. In future work, a proof of concept should be carried out which would provide initial impedance data and verify the proper functioning of the mechanism.

The mechanism does not currently contain an element to provide a minimum stiffness to the ankle when the weight is not applied to it, typically during the swing phase. A torsion spring with a low stiffness will probably have to be added between the shaft and the shank part to limit the unwanted oscillating movements that would occur when walking.

7.2 Conclusion

People suffering from a lower limb amputation face difficulties in getting around in everyday life. Ankle foot prostheses were developed to help these people regain their independence. Initially, the primary objective was walking and therefore the degree-of-freedom in the sagittal plane. Now that progress has been made concerning this DoF, attempts to implement the other DoF of the ankle are made to meet more complex and natural movement needs. For example, the DoF in the frontal plane called inversion/eversion is involved in many movements such as sidestep cutting or more globally in the stability when the ground is uneven.

To this end, this work focused on the implementation of a module adding the inversion/eversion DoF to existing active prostheses. For the design a classical design methodology was followed and for the project management agile methods were implemented with the help of the supervisors François Heremans and Renaud Ronsse.

The design resulted in a weight activated passive solution. It has two purposes: to allow the foot to adapt to the inclination of the ground when the weight is not applied to the mechanism and to provide stability via rendered mechanical impedance when weight is applied to it. To do so, a cantilever takes advantage of the deformation caused by the weight to activate a locking system that couples the shank part and the shaft thanks to a capstan using an aramid rope. To render the mechanical impedance, a nylon rope is wound around the shaft with the end of the strands fixed to the foot part. When the amputee's weight is acting on the mechanism, it is therefore the deformation of the nylon rope that allows the inversion/eversion movement.

To verify the feasibility of the concept and justify the material selection, numerical simulations were performed with the CAD software Solidworks. The simulations focusing on the deformation proved that the selected polyamide 6 was appropriate for the cantilevered part. The structural strength was also tested since the mechanism must sustain the efforts caused by both the joint torques in the frontal plane but also in the sagittal plane which is greater.

The solution obtained is a first step towards the integration of this degree-of-freedom in active prostheses. However a proof of concept could not be designed and further experiments need to be carried out.

Chapter 8

Appendix

Simon Vandergoeten		Specifications			Date: 04/03/2021 version : 2	
Objective: Design and control of the frontal DoF for ankle foot prosthesis						
Date			Origin	Criteria	Levels	
24/02	MF1 : Ambulation with inversion/eversion	MF1.1 : Level ground walking	Simon	CMF1.1.1 ROM of a sound ankle CMF1.1.2 Peak torque CMF1.1.3 Peak power CMF1.1.4 Time-varying impedance	35° (eversion : -12°, inversion : 23°) 0,27 [Nm/kg] 0,05 [W/kg] Stiffness : from 1,14 to 5,43 [Nm/rad kg] Damping : from 0,016 to 0,08 [Nm s/rad kg] 175 [°/s] 100 [steps/min]	
24/02				CMF1.1.5 Peak angular speed CMF1.1.6 Speed of gait		
26/02		MF1.2 : Sloped terrain (15°)	Simon	CMF1.2.1 ROM of a sound ankle CMF1.2.2 Peak torque CMF1.2.3 Peak power CMF1.2.4 Time-varying impedance	75° (eversion : -45°, inversion : 30°) 0,12 [Nm/kg] 0,58 [W/kg] (eversion mode : 1,26 [Nm/rad kg]) (inversion mode : 0,57 [Nm/rad kg]) 300 [°/s] 100 [steps/min]	
26/02				CMF1.2.5 Peak angular speed CMF1.2.6 Speed of gait		
24/02	MC1 : Safety		Simon	CMC1.1 Loading resistance CMC1.2 Life time	100 [kg] x safety factor > 3 years	
26-févr	MC2 : Transitions		Simon	CMC2.1 Transitions from ≠ modes	Sitting-standing-walking	
24-févr	MC3 : Cost		Amputee	CMC3.1 Additional cost	50%	
24/02	MC4 : Dimensions		Amputees	CMC4.1 Low build height CMC4.2 Additional weights CMC4.3 Prosthesis dimensions	< 15 [cm] < 200 [g] within volume of biological limb	
24-févr	MC5 : Noiseless		Simon	CMC5.1 Maximum value	35 dB	

Figure 8.1: List of requirements for the design procedure

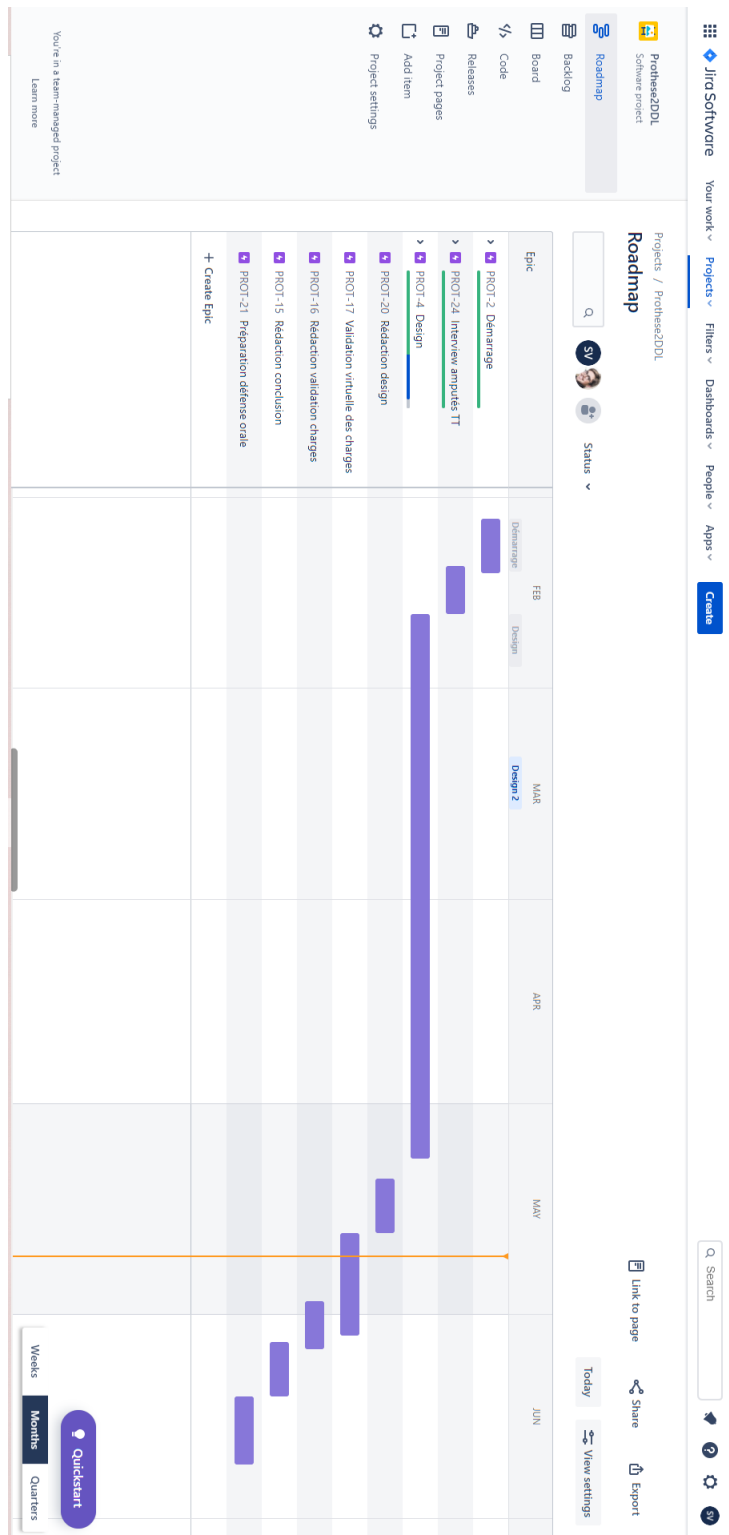


Figure 8.2: Roadmap for the second semester with the software *Jira*

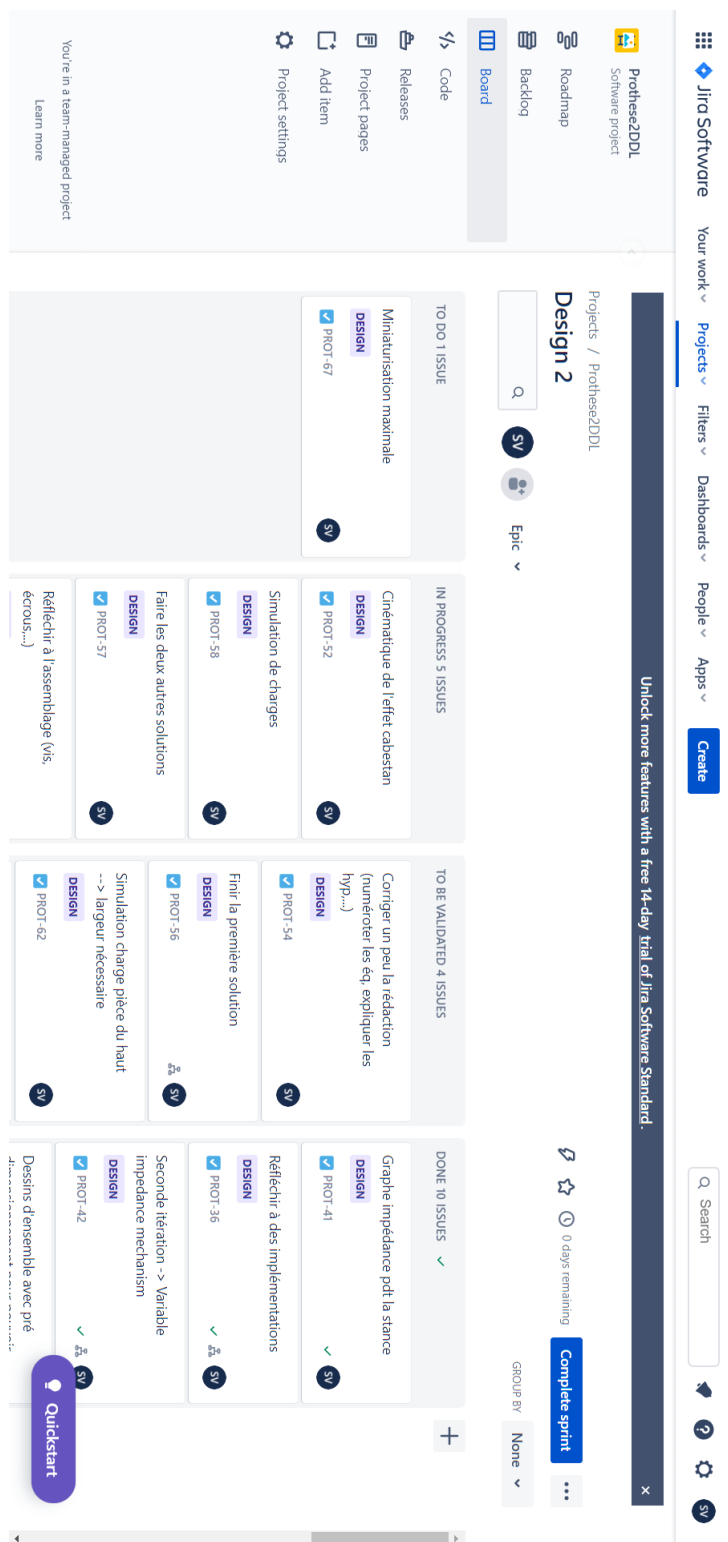


Figure 8.3: Example of a sprint with the different tasks

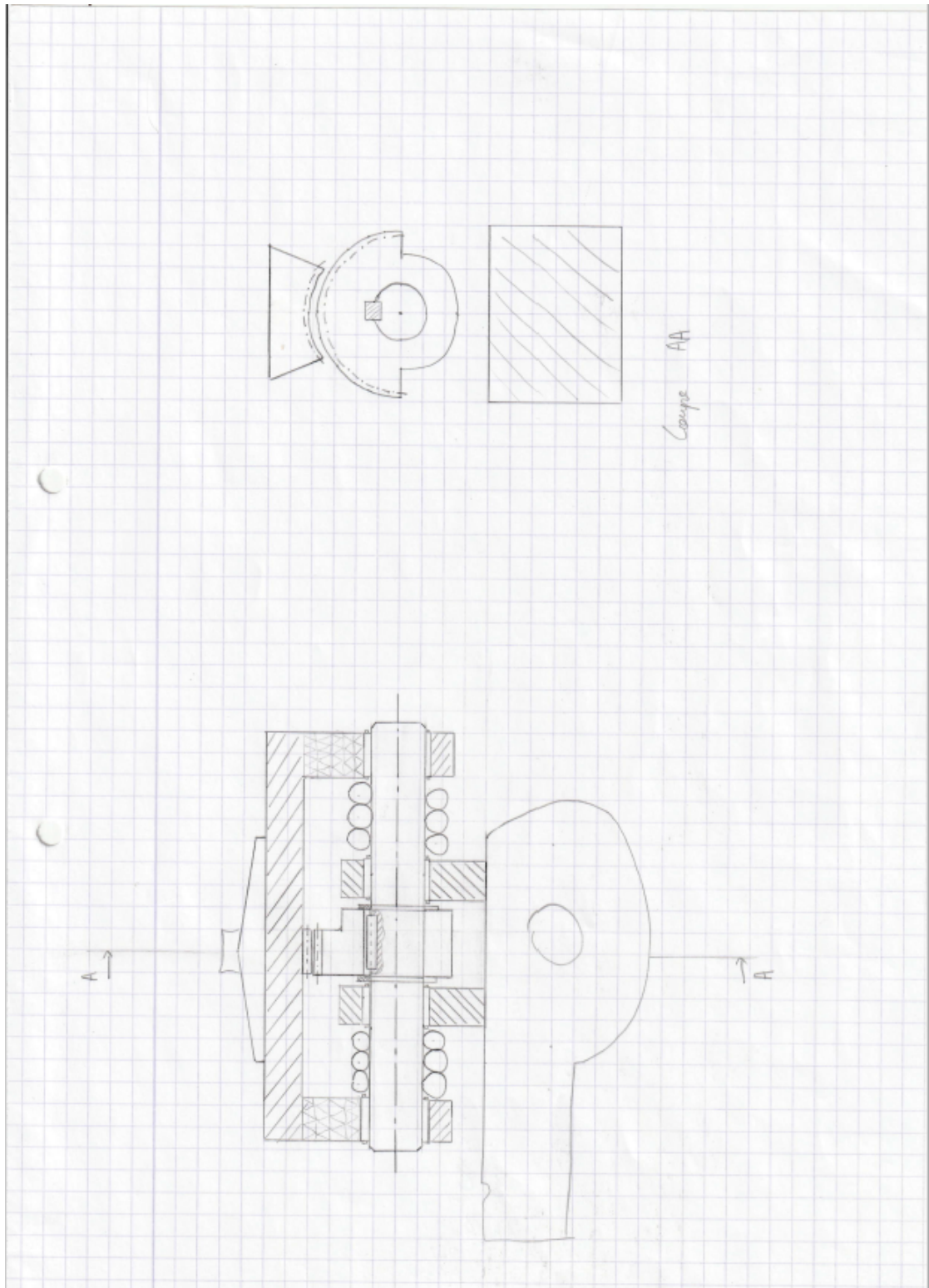


Figure 8.4: Sketch of the hook and pawl solution with the partial gear

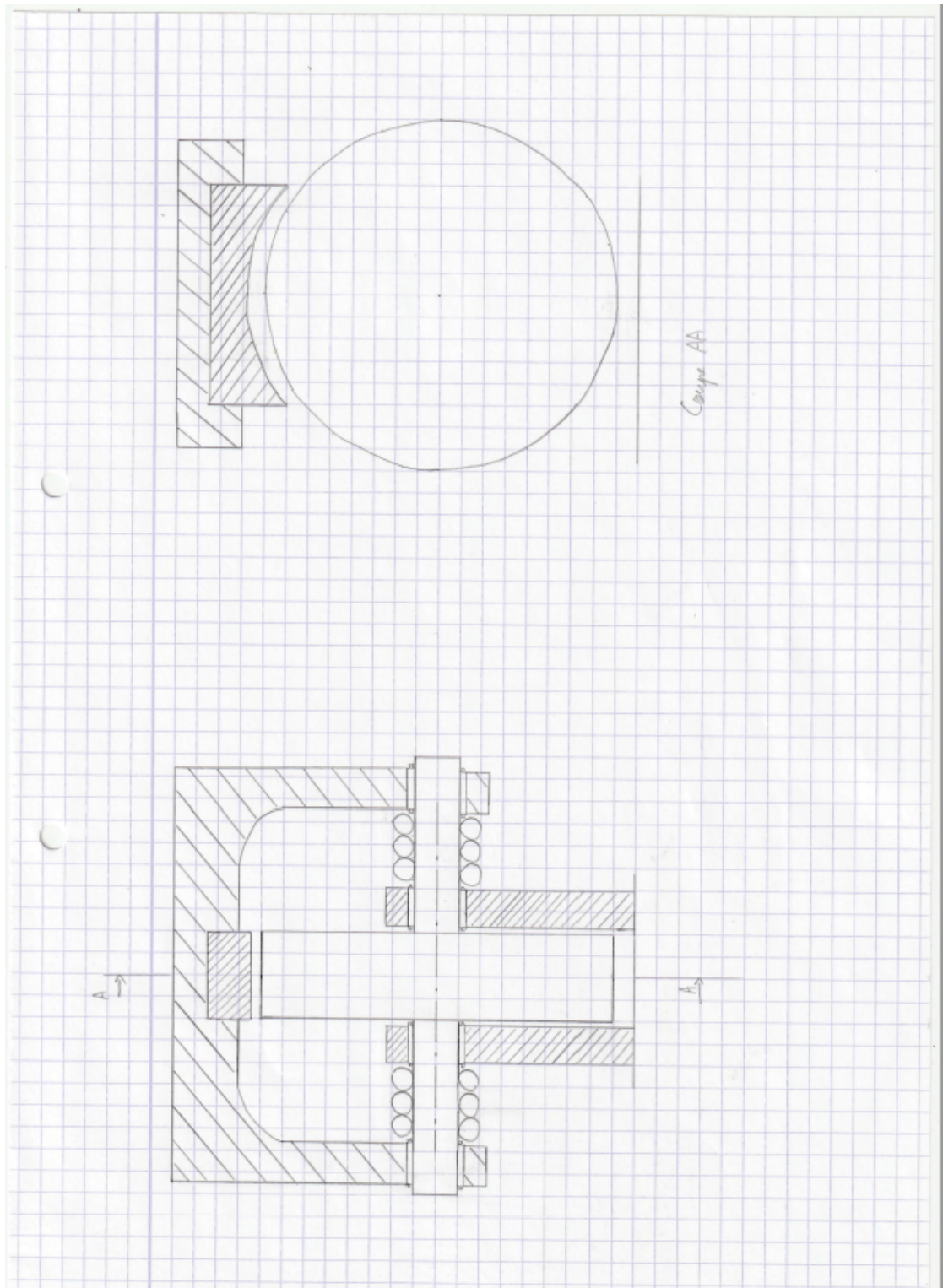


Figure 8.5: Sketch of the clutch brake solution

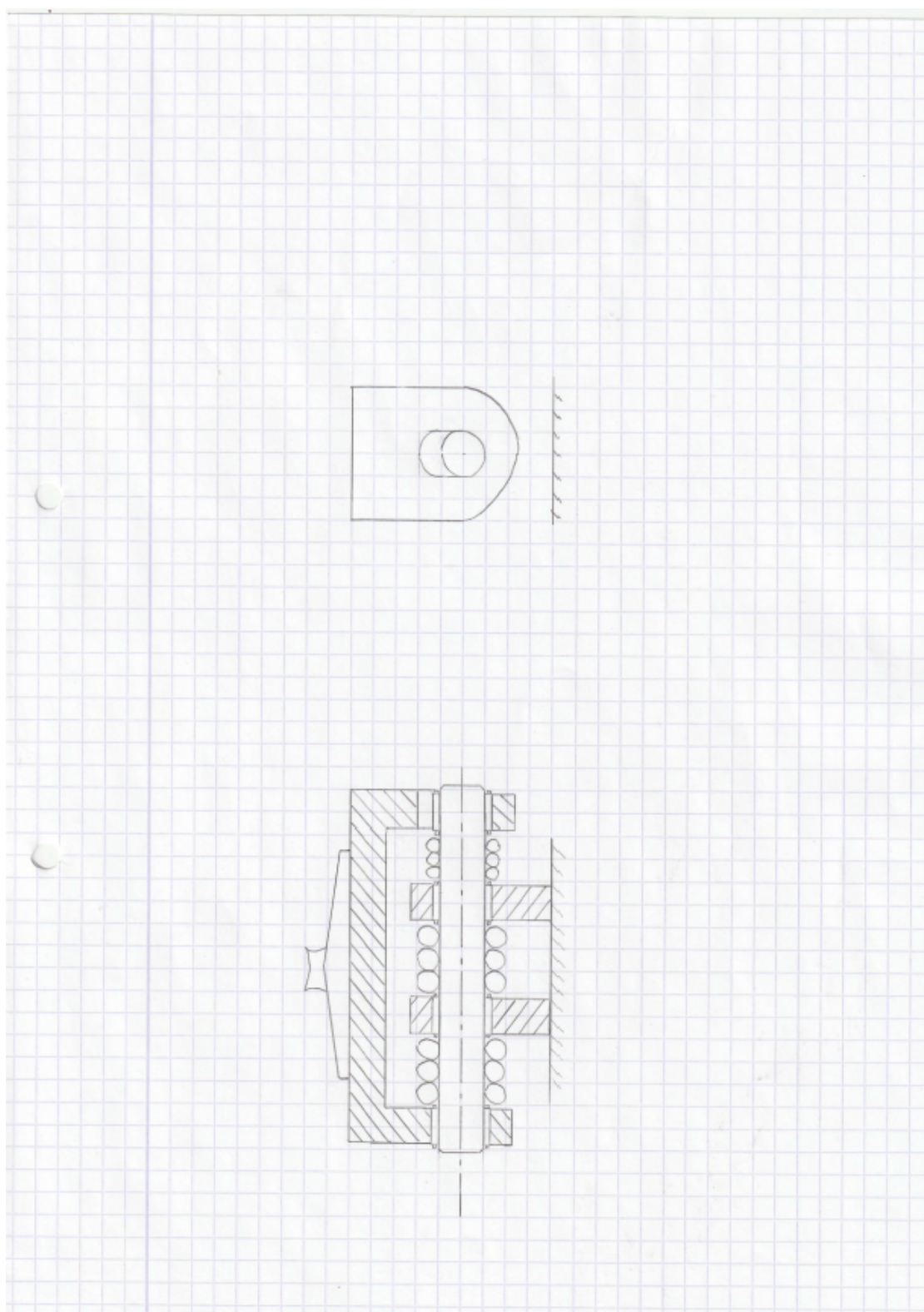


Figure 8.6: Sketch of the capstan solution

Bibliography

- [1] Rachael Miller, Graeme K. Ambler, Jozel Ramirez, Jonathan Rees, Robert Hinchliffe, Christopher Twine, Sarah Rudd, Jane Blazeby, and Kerry Avery. Patient Reported Outcome Measures for Major Lower Limb Amputation Caused by Peripheral Artery Disease or Diabetes: A Systematic Review. *European Journal of Vascular and Endovascular Surgery*, 61(3):491–501, March 2021.
- [2] T. R. Dillingham, L. E. Pezzin, and E. J. MacKenzie. Limb amputation and limb deficiency: Epidemiology and recent trends in the united states. *Southern medical journal*, 95(8):875–883, 2002.
- [3] Heiner Claessen, Herve Avalosse, Joeri Guillaume, Maria Narres, Tatjana Kvitkina, Werner Arend, Stephan Morbach, Patrick Lauwers, Frank Nobels, Jacques Boly, Chris Van Hul, Kris Doggen, Isabelle Dumont, Patricia Felix, Kristien Van Acker, and Andrea Icks. Decreasing rates of major lower-extremity amputation in people with diabetes but not in those without: a nationwide study in Belgium. *Diabetologia*, 61(9):1966–1977, September 2018.
- [4] Jason Daley. This 3,000-Year-Old Wooden Toe Shows Early Artistry of Prosthetics. <https://www.smithsonianmag.com/smart-news/study-reveals-secrets-ancient-cairo-toe-180963783/>. Accessed: 22-10-2020.
- [5] Mechanical Limbs Before The Machine Age. <https://beastrabban.wordpress.com/2013/10/19/mechanical-limbs-before-the-machine-age/>. Accessed: 22-10-2020.
- [6] Ambroise Paré Facts. <https://biography.yourdictionary.com/ambroise-pare>. Accessed: 23-10-2020.
- [7] S.G. Feuer O. Fliegel. Historical Development of Lower-Extremity Prostheses. http://www.oandplibrary.org/op/1966_04313.asp. Accessed : 24 – 10 – 2020.
- [8] History of The Prosthetic Leg Through The Ages. <https://www.amputee-coalition.org/history-prosthetic-leg/>. Accessed: 23-10-2020.

- [9] THE J.E. HANGER STORY. <http://www.hanger.com/history/Pages/The-J.E.-Hanger-Story.aspx>. Accessed: 23-10-2020.
- [10] A century of advances in prostheses. *Nature Materials*, 17(11):945–945, November 2018. Number: 11 Publisher: Nature Publishing Group.
- [11] American Orthotic and Prosthetic Association. <https://www.aopanet.org/>. Accessed: 26-10-2020.
- [12] Ashley Finn Christine Miller and Emily Delzell. Battle-field injuries: Saving lives and limbs throughout history. https://lrmagazine.com/cover_story/battlefield_injuries_saving_lives_and_limbs_throughout_history. Accessed : 09 – 03 – 2021.
- [13] D. Malgrange. Physiopathologie du pied diabétique. *La Revue de Médecine Interne*, 29:S231–S237, September 2008.
- [14] R. Versluys, A. Desomer, G. Lenaerts, P. Beyl, M. Van Damme, B. Vanderborght, I. Vanderniepen, G. Van der Perre, and D. Lefeber. From conventional prosthetic feet to bionic feet: A review study. In *2008 2nd IEEE RAS EMBS International Conference on Biomedical Robotics and Biomechatronics*, pages 49–54, October 2008. ISSN: 2155-1782.
- [15] Phases of the normal gait cycle. https://www.researchgate.net/figure/Phases-of-the-normal-gait-cycle_fig3_09362425. Accessed : 17 – 02 – 2021.
- [16] Movements of the ankle foot. https://www.researchgate.net/figure/Movements-of-the-ankle-foot_fig2_38540796. Accessed : 17 – 02 – 2021.
- [17] Evandro M. Ficanha, Mohammad Rastgaar, and Kenton R. Kaufman. Ankle mechanics during sidestep cutting implicates need for 2-degrees of freedom powered ankle-foot prostheses. *Journal of Rehabilitation Research and Development*, 52(1):97–112, 2015.
- [18] Inversion and eversion. https://commons.wikimedia.org/wiki/File:Eversion_and_inversion.jpg. Accessed : 17 – 02 – 2021.
- [19] Emile Moreau. Experimental investigation of the inversion/eversion motion for the design of a two degrees of freedom ankle prosthesis. Master’s thesis, Université catholique de Louvain-la-Neuve, <http://hdl.handle.net/2078.1/thesis:10619>, 2017.
- [20] Evandro Ficanha, Guilherme Ribeiro, Lauren Knop, and Mo Rastgaar. Estimation of the Two Degrees-of-Freedom Time-Varying Impedance of the Human Ankle. *Journal of Medical Devices*, 12(011010), January 2018.

- [21] Claire L. Brockett and Graham J. Chapman. Biomechanics of the ankle. *Orthopaedics and Trauma*, 30(3):232–238, June 2016.
- [22] Asbjørn Roaas and Gunnar B. J. Andersson. Normal Range of Motion of the Hip, Knee and Ankle Joints in Male Subjects, 30–40 Years of Age. *Acta Orthopaedica Scandinavica*, 53(2):205–208, January 1982. Publisher: Taylor & Francis _eprint: <https://doi.org/10.3109/17453678208992202>.
- [23] SACH foot. <https://www.pinterest.com/pin/73887250118199851/>. Accessed: 13-02-2021.
- [24] Brian J. Hafner, Joseph M. Czerniecki, Joan E. Sanders, and John Fergason. Transtibial energy-storage-and-return prosthetic devices: A review of energy concepts and a proposed nomenclature. *Journal of Rehabilitation Research & Development*, 39(1):1, February 2002. Publisher: VA Prosthetics Research & Development Center.
- [25] Steven H. Collins and Arthur D. Kuo. Recycling Energy to Restore Impaired Ankle Function during Human Walking. *PLOS ONE*, 5(2):e9307, February 2010. Publisher: Public Library of Science.
- [26] GK Klute, J Czerniecki, and B Hannaford. Development of powered prosthetic lower limb. 1998.
- [27] Hugh M. Herr and Alena M. Grabowski. Bionic ankle-foot prosthesis normalizes walking gait for persons with leg amputation. *Proceedings: Biological Sciences*, 279(1728):457–464, 2012. Publisher: Royal Society.
- [28] S. K. Au, J. Weber, and H. Herr. Biomechanical Design of a Powered Ankle-Foot Prosthesis. In *2007 IEEE 10th International Conference on Rehabilitation Robotics*, pages 298–303, June 2007. ISSN: 1945-7901.
- [29] Ottobock. Empower, Ottobock. <https://www.ottobock.fr/protheses/produits-a-z/membre-inferieur/pieds/empower.html>. Accessed: 01-11-2020.
- [30] International Conference on Intelligent Robots and Systems. State of the art in robotic leg prostheses: where we are and where we want to be. <https://belab.mech.utah.edu/iros2020/>. Accessed: 27-10-20.
- [31] T. Elery, S. Rezazadeh, C. Nesler, and R. D. Gregg. Design and Validation of a Powered Knee–Ankle Prosthesis With High-Torque, Low-Impedance Actuators. *IEEE Transactions on Robotics*, pages 1–20, 2020. Conference Name: IEEE Transactions on Robotics.

- [32] European Community’s H2020 Research and Innovation Programme. Cyber-legs++. <http://www.cyberlegs.eu/>. Accessed: 30-11-20.
- [33] B. Convens, D. Dong, R. Furnémont, T. Verstraten, P. Cherelle, D. Lefeber, and B. Vanderborght. Modeling, Design and Test-Bench Validation of a Semi-Active Propulsive Ankle Prosthesis With a Clutched Series Elastic Actuator. *IEEE Robotics and Automation Letters*, 4(2):1823–1830, April 2019. Conference Name: IEEE Robotics and Automation Letters.
- [34] L. Gabert, S. Hood, M. Tran, M. Cempini, and T. Lenzi. A Compact, Lightweight Robotic Ankle-Foot Prosthesis: Featuring a Powered Polycentric Design. *IEEE Robotics Automation Magazine*, 27(1):87–102, March 2020. Conference Name: IEEE Robotics Automation Magazine.
- [35] Alejandro F. Azocar, Luke M. Mooney, Jean-François Duval, Ann M. Simon, Levi J. Hargrove, and Elliott J. Rouse. Design and clinical implementation of an open-source bionic leg. *Nature Biomedical Engineering*, 4(10):941–953, October 2020. Number: 10 Publisher: Nature Publishing Group.
- [36] Martin Grimmer, Matthew Holgate, Robert Holgate, Alexander Boehler, Jeffrey Ward, Kevin Hollander, Thomas Sugar, and André Seyfarth. A powered prosthetic ankle joint for walking and running. *BioMedical Engineering OnLine*, 15(3):141, December 2016.
- [37] François Heremans, Sethu Vijayakumar, Mohamed Bouri, Bruno Dehez, and Renaud Ronsse. Bio-inspired design and validation of the Efficient Lockable Spring Ankle (ELSA) prosthesis. In *2019 IEEE 16th International Conference on Rehabilitation Robotics (ICORR)*, pages 411–416, Toronto, June 2019. ISSN: 1945-7901.
- [38] F. Heremans and R. Ronsse. Design of an energy efficient transfemoral prosthesis using lockable parallel springs and electrical energy transfer. In *2017 International Conference on Rehabilitation Robotics (ICORR)*, pages 1305–1312, July 2017. ISSN: 1945-7901.
- [39] H.-T. Pham, M.-N. Le, and V.-T. Mai. A Novel Multi-Axis Compliant Prosthetic Ankle Foot to Support the Rehabilitation of Amputees. pages 238–243, 2016.
- [40] E. A. Rogers, M. E. Carney, S. H. Yeon, T. R. Clites, D. Solav, and H. M. Herr. An Ankle-Foot Prosthesis for Rock Climbing Augmentation. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, pages 1–1, 2020. Conference Name: IEEE Transactions on Neural Systems and Rehabilitation Engineering.

- [41] Ryan D. Bellman, Matthew A. Holgate, and Thomas G. Sugar. SPARKy 3: Design of an active robotic ankle prosthesis with two actuated degrees of freedom using regenerative kinetics. In *2008 2nd IEEE RAS EMBS International Conference on Biomedical Robotics and Biomechatronics*, pages 511–516, October 2008. ISSN: 2155-1782.
- [42] Evandro Maicon Ficanha, Guilherme Aramizo Ribeiro, Houman Dallali, and Mohammad Rastgaar. Design and Preliminary Evaluation of a Two DOFs Cable-Driven Ankle–Foot Prosthesis with Active Dorsiflexion–Plantarflexion and Inversion–Eversion. *Frontiers in Bioengineering and Biotechnology*, 4, 2016. Publisher: Frontiers.
- [43] E. M. Ficanha, M. Aagaah, and K. Kaufman. Cable-Driven Two Degrees-of-Freedom Ankle–Foot Prosthesis. 2016.
- [44] Myunghee Kim, Tianjian Chen, Tianyao Chen, and Steven H. Collins. An Ankle–Foot Prosthesis Emulator With Control of Plantarflexion and Inversion–Eversion Torque. *IEEE Transactions on Robotics*, 34(5):1183–1194, October 2018. Conference Name: IEEE Transactions on Robotics.
- [45] Tsung-Han Hsieh. Design and control of a two-degree-of-freedom powered ankle-foot prosthesis. Master’s thesis, Massachusetts institute of technology, <https://hdl.handle.net/1721.1/124195>, 2019.
- [46] David Wynn. *Model-based approaches to support process improvement in complex product development*. PhD thesis, 10 2007.
- [47] A.Vankrunkelsven B.Herman, G.Kerckhofs. Lmeca2355 mechanical design in biomedical engineering.
- [48] Beitz W. Pahl G. *Engineering design: a systematic approach*. Springer, Berlin, Heidelberg, 1996.
- [49] M.J French. *Conceptual design for Engineers*. Springer, Berlin, Heidelberg, 1985.
- [50] Cross N. *Engineering design methods: strategies for product design*. Wiley, 1998.
- [51] Apoorva Srivastava, Sukriti Bhardwaj, and Shipra Saraswat. SCRUM model for agile methodology. In *2017 International Conference on Computing, Communication and Automation (ICCCA)*, pages 864–869, May 2017.

- [52] Shruti Sharma and Nitasha Hasteer. A comprehensive study on state of Scrum development. In *2016 International Conference on Computing, Communication and Automation (ICCCA)*, pages 867–872, April 2016.
- [53] US Food and Drug Administration. *Design control guidance for medical device manufacturers*. Rockville, USA, 1997.
- [54] M. Plooi, G. Mathijssen, P. Cherelle, D. Lefeber, and B. Vanderborght. Lock Your Robot: A Review of Locking Devices in Robotics. *IEEE Robotics Automation Magazine*, 22(1):106–117, March 2015. Conference Name: IEEE Robotics Automation Magazine.
- [55] KHK. KHK stock gears - MSGA1-36. <https://khkgears2.net/catalog2/MSGA1-36>. Accessed: 21-04-2021.
- [56] Robert C. Juvinall and Kurt M. Marshek. *Juvinall’s fundamentals of machine component design*. Wiley, Danvers, 2017.
- [57] Ming-xue Shen, Jin-peng Zheng, Xiang-kai Meng, Xiao Li, and Xu-dong Peng. Influence of Al₂O₃ particles on the friction and wear behaviors of nitrile rubber against 316L stainless steel. *Journal of Zhejiang University-SCIENCE A*, 16(2):151–160, February 2015.
- [58] J.W.S Hearle W.E Morton. *Physical properties of textile fibers, fourth edition*. Woodhead Publishing in Textiles, Cambridge, England, 2008.
- [59] Kanirope. Kanirope, professional ropes and equipment. <https://www.kanirope.fr/corde-polyamide-nylonbraid-5mm-16x-tressee-de-kanirope>. Accessed: 22-04-2021.
- [60] SKF. SKF, K 12X15X10 TN needle bearing. <https://www.skf.com/fr/products/rolling-bearings/roller-bearings/needle-roller-bearings/needle-roller-and-cage-assemblies/productid-K%2012X15X10%20TN>. Accessed: 17-03-2021.
- [61] Smalley. Smalley the engineer’s choice. <https://www.smalley.com/fr/ring/es-12>. Accessed: 17-03-2021.

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