MASTER THESIS

DEVELOPMENTS IN 2 DEGREE OF FREEDOM FINE FORCE DELIVERY BY AN ASSISTIVE ANKLE ORTHOSIS

Bart Hendriksen

DEPARTMENT OF BIOMECHANICAL ENGINEERING
FACULTY OF ENGINEERING TECHNOLOGY

EXAMINATION COMMITTEE
Prof. Dr. Ir. H. van der Kooij
Prof. Dr. Ir. M. Sartori
SUPERVISORS
Dr. A. Sadeghi (internal supervisor)
Dr. G.V. Durandau (external supervisor)

DOCUMENT NUMBER
BE – 970

05-12-2023

UNIVERSITY OF TWENTE.

Preface

Before you lies the master's thesis titled 'Developments in 2 degree of freedom fine force delivery by an assistive ankle orthosis', documenting design, and realisation. This thesis has been developed in the context of my graduation of the master's Mechanical Engineering track Personalised Health Technology at the University of Twente at the department of Biomechanical Engineering. I've worked on this thesis over the past year, from September 2022 until October 2023.

This thesis is meant as a proof of competence at a master's level in (Bio)mechanical Engineering and requires as such a rudimentary background in the field. Furthermore, this document can be used as a reference for further developments in the J-Lo project at the University of Twente, any other developments in two degree of freedom ankle-foot orthoses or further tests involving this design.

For the guidance of this thesis, I want to mainly thank Guillaume Durandau, who even when he left the university continued to give me weekly feedback on the process and design. I also want to thank Ali Sadeghi who replaced Guillaume as my official supervisor and who always was available for feedback and questions, and Donatella Simonetti for her indispensable help with the practical experiments in the last few weeks. Lastly, I want to thank Herman van der Kooij for the monthly meetings on the progress and his input in the process and anybody that was ready to answer questions that arose or helped in any other way.

Enschede, November 23
Bart Hendriksen

Nomenclature

Abbreviation	Full expression
AFO	Ankle-foot orthosis
DoF(‘s)	Degree(s) of freedom
RoM	Range of motion
PF	Plantarflexion
DF	Dorsiflexion
STD	Standard deviation
MS	Musculoskeletal
SEA	Series elastic actuator
SEE	Series elastic element
EMG	Electromyography
GRF(‘s)	Ground reaction force(s)
NMS	Neuromusculoskeletal

Abstract

This thesis describes the development of a 2 degree of freedom assistive ankle device. The aim was to support both plantarflexion, and inversion and eversion according to the design requirements made by the client. To do so, a methodical design approach was used, developing an assistive device that allows for full natural range of motion. The design is verified measuring the slippage of the connection to the shank as well as the EMG signals of the superficial muscles responsible for the three motions. The experiments show an average motion of the shank bracket of 2.1±1.1 mm for the different trials, which is considered relatively stable, but this comes at the cost of slight discomfort during wearing. Furthermore, for the EMG signals, a reduction up till 8.7% was seen for plantarflexion, 9.9% for eversion and the results for inversion need further research. In any case, this proves that with a Bowden cable actuated device, 2 degree of freedom actuation is possible, and that besides plantarflexors, also invertors and evertors can be supported.

Table of contents

PREFACE ..... **III**

NOMENCLATURE ..... **IV**

ABSTRACT ..... **V**

TABLE OF CONTENTS ..... **VI**

1 INTRODUCTION & BACKGROUND **1**

1.1 BACKGROUND 1
1.2 STATE OF THE ART 2
1.2.1 1DoF Bowden systems – Rigid connections *3*
1.2.2 1DoF Bowden systems – Flexible connections *3*
1.2.3 1DoF non-Bowden systems *4*
1.2.4 2DoF non-Bowden systems *4*
1.2.5 Quantitative measures of state of the art *5*
1.3 BIOMECHANICS AND DYNAMICS OF THE ANKLE 5
1.4 REPORT STRUCTURE 7

2 REQUIREMENTS & BOUNDARIES **8**

2.1 PROBLEM DEFINITION BY CLIENT 8
2.2 GOAL 8
2.3 DESIGN ASSIGNMENT 8
2.4 SYSTEM SUBDIVISION 9
2.5 USE CASES 9
2.6 REQUIREMENTS & BOUNDARY CONDITIONS 10

3 CONCEPT DEVELOPMENT **12**

3.1 THE FIRST STEP 12
3.2 JOINT TYPE 12
3.3 CONCEPTS 14
3.3.1 Concept 1: Joint A, no upper leg band *14*
3.3.2 Concept 2: Joint A, upper leg band *15*
3.3.3 Concept 3: Joint D, no upper leg band *15*
3.3.4 Concept 4: Joint E, no upper leg band *15*
3.4 CONCEPT CHOICE 17

4 CONCEPT DETAILING **18**

4.1 PRINCIPAL DESIGN 18
4.2 MECHANICAL DESIGN 20
4.2.1 Dimensioning of components *20*
4.2.2 Mechanical safety limit *22*
4.2.3 Fatigue analysis *22*
4.2.4 Mass estimation *24*

5 VERIFICATION & REALISATION **25**

5.1 DESIGN VERIFICATION 25
5.2 PROOF OF CONCEPT 27

5.2.1 Mechanical stability: shank bracket motion 27
5.2.2 Functionality: muscle activation levels 29

6 DISCUSSION & RECOMMENDATIONS 38

7 CONCLUSIONS 39

REFERENCES 40

APPENDIX A: LITERATURE REVIEW

APPENDIX B: METHODICAL PRODUCT DESIGN

APPENDIX C: EXPERIMENT 1 DETAILS | SHANK BRACKET STABILITY

APPENDIX D: EXPERIMENT 3 DETAILS | MUSCLE ACTIVATION LEVELS

APPENDIX E: COMPUTATION STRUCTURE

1 Introduction & background

Millions of people worldwide are suffering from paralysis in some form, the global estimate being 40 – 80 million people [1]. In the United States, this number was some 5.4 million in 2016 [2], on a population of around 250 million, equating to about 2% of the population of a developed country. This paralysis can have several causes, varying from for example musculotendinous to neural impairments. Patients with musculotendinous impairments (applied to lower extremities in this thesis) can find great benefit in physical therapy in multiple stages of rehabilitation [3]. To enable patients that do not have sufficient muscle power (yet), an assistive device can be used. While this is already applied in the rehabilitation process (or after if the insufficiencies become chronic), it is still very much in a development phase and the available devices still have their shortcomings. Often, orthoses are not perfectly aligned with the human joints and do not actuate in line with the complex bodily structures, exerting a stress on the already impaired joint structure. Furthermore, most lower extremity assist devices actuate one DoF in one or two directions and have no ability to target specific muscle groups, such as the plantarflexors, invertors, or evertors. With the field of multi DoF lower extremities assistive devices being relatively undiscovered, this thesis aims to add a Bowden cable actuated device to the state of the art. It could be utilized in a research context while also be made fit for rehabilitation or longer-term assistance.

The scope of the thesis is the development of a proof of principle for a 2 DoF Bowden cable actuated ankle orthosis according to the requirements specified by the project owner. Such a proof of principle is necessary to judge the feasibility of this type of external support. To do so while considering the proper alignment with the ankle tissues, a MS model of the patient is used to estimate the design parameters for the orthosis. Thus, the project will include the joint effects of utilising the orthosis, as well as a mechanical analysis of the setup. The specific use phase of the design, in which the device is made fit for either experiments, longer term use or a specific rehabilitation case is outside of the boundary conditions for this project.

1.1 Background

In general, it can be stated that patients suffering from tendinopathy (especially tendonitis), muscle atrophy, motor impairments and early stage cartilage damage can benefit from (assisted) rehabilitation (applied to the lower extremities in this thesis) [3]. Effectively, all these impairments boil down to the mechanical principle of stress. A force increase leads to a higher stress in the tissue, a misalignment (and thus contact area decrease) leads to a higher stress etc. This mechanical stress can then lead to tissue failure in a similar way as that mechanical stress can lead to other material failures. One of the main medical causes for these impairments are strokes (> 795.000 cases on the 330 million, in 2022, population) [4]. As is well-documented for the knee, an overall increase in load magnitude across the knee is associated with an '*increased incidence of osteoarthritis as accelerated progression of obesity' [5].* During normal walking, but especially during activities such as running or jumping, knee and ankle reaction loads increase significantly [6]. Detrimental effects of added weight or imbalances thus cause a magnified reaction on a joint level.

Those forces can be increased by one of the following causes:

1) Imbalance over the knee joint (i.e., muscles on either side of the mediolateral plane are activated asymmetrically continuously). This is problematic as the knee has very limited rotational freedom in that plane.
2) Overactivated muscles. Muscles that are continuously overactive and thus have higher forces over time cause higher joint reaction loads.
3) Orientation imbalance. Varus ankle deformity is associated with ankle osteoarthritis of the varus type, but Zhu, Li and Xu could not find definitive proof that this was caused by prolonged exposure to eccentric loading [7].
4) External forces, like a misaligned orthosis.

The complexity of the ankle (and knee) joint makes that any treatment is a delicate matter. Pharmaceutical and surgical solutions are available, but in general should not be a first step. In a healthy joint, a balance state is desired: tissue homeostasis.

Correct alignment of the joint in general, but also of the orthosis with the joint is vital for tissue homeostasis, which is mainly articulate cartilage homeostasis, defined as: *'the state at which degradation of extracellular matrix components is balanced by synthesis' [8].* In this case, the extracellular matrix material refers mainly to the collagen in cartilage. Note that also out-of-plane forces could induce stresses in the joint that hinder homeostasis, for which a correction force could be applied. Figure 1 [9] shows a cut view of the ankle. It can easily be seen that a misalignment of tibia and talus will lead to a decrease in force contact area and thus more pressure, and that out-of-plane forces can add either a stress on surrounding tissue, a friction force through the cartilage, or an added force by inducing misalignment.

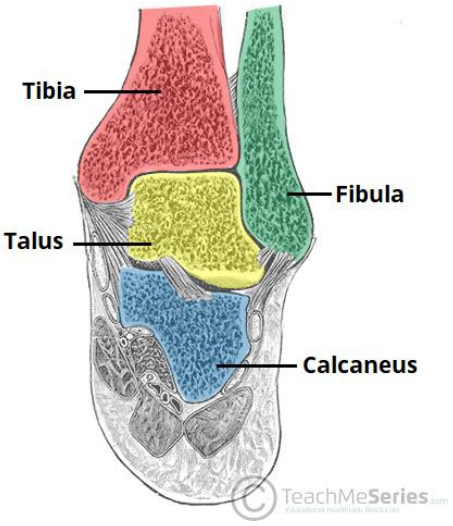


Figure 1: Posterior view of ankle joint and the bones it contains.

Effectively, this translates the problem from the medical to the mechanical domain and vice versa. This thesis will always work on the intersection of the two domains and the effects of a change in one domain must be measured or estimated in the other domain too.

1.2 State of the art

Plenty of assistive devices are on the market for the lower extremities. The use of these assistive devices varies from assisted lifting and military applications to long term assistance or rehabilitation assistance. They may be active or passive, and those active types vary quite significantly in their actuation systems. As the goal of this thesis is to develop a Bowden cable actuated system, only actuated ankle-foot orthoses are considered. This chapter will briefly discuss some possible applications for the device as well as highlight the relevant state of the art.

While there are many applications for orthoses, the main uses for an ankle-foot orthoses lie in three fields:

- Rehabilitation assistance.
- Research.
- Long-term support.

These applications all have different requirements, but in general they will fare well with light-weight devices that are easy to donn/doff and can be adapted relatively simple to individual patients. However, rehabilitation patients might benefit from stiffer devices providing them with support when their muscles are not able to fully do so yet, while more flexible devices may be beneficial for people that have muscle dystrophy or have tendon damage and need support in a specific movement. Both types thus have their advantages, but there are also clear downsides. The stiffer a construction becomes, the more precise the device should be aligned with the body, and with more flexibility comes the downside of lower support forces as the forces must be fully transferred to the body. With the context of multi DoF support, the need for more flexible devices becomes larger, as the design complexity greatly increases with extra DoF's. Regardless, most of these devices assist only in plantarflexion, while some newer devices can also provide dorsiflexion assistance. The actuation type closest to natural muscle actuation is a device using cables as actuators, as they provide pull action comparable to the muscles of the body.

There are of course many more devices than are discussed here, but for example rotary actuated exoskeletons are so far separated in functionality from linearly actuated ankle-foot orthoses that they are not explicitly discussed.

For the state of the art, the ankle-foot orthoses are divided into three separate categories:

- Bowden cable actuated one degree of freedom systems (1DoF Bowden systems).
- Non-Bowden cable actuated one degree of freedom systems (1DoF non-Bowden systems).

- Non-Bowden cable actuated two degree of freedom systems (2DoF non-Bowden systems).

The obvious missing category is Bowden cable actuated two degree of freedom systems (2DoF Bowden systems), which are currently non-existent. These categories are discussed briefly in the paragraphs below. These systems can all either contain a rigid connection between the shank and the foot or can be built without a rigid connection. Note that no distinction is made between tethered and untethered systems.

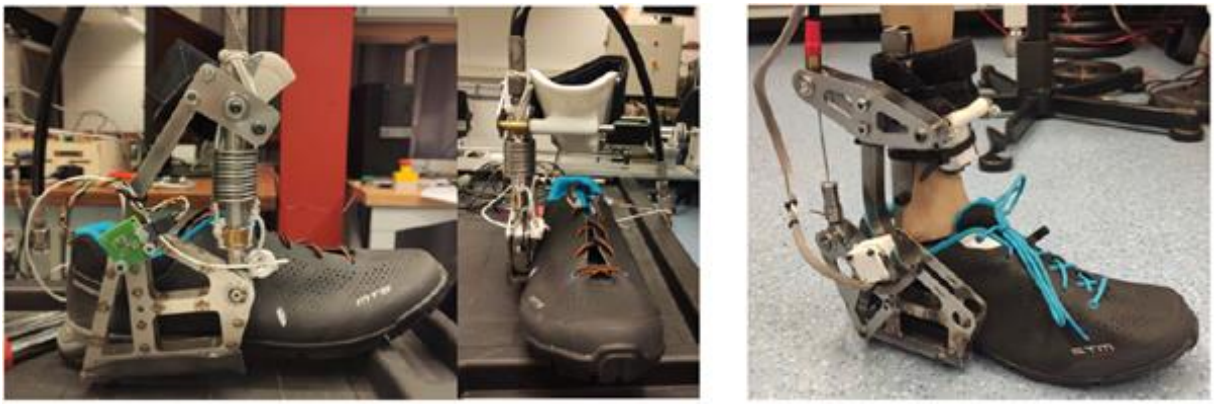


Figure 2: J-Lo with hydraulic and cable actuation. Left two pictures showing the hydraulic actuation, right picture the latest updated J-Lo with Bowden cable actuation.

1.2.1 1DoF Bowden systems – Rigid connections

The starting point of this design is the state of the art of the J-LO design at the University of Twente as described earlier, which is an example of a device with a rigid connection between shank and foot. The device is lightweight (especially distally), is mechanically robust as the weak link is the Bowden cable, which can be easily replaced if broken. However, there are some major downsides to this principle. As Jeroen Meijners already mentions in his thesis [12], there can be some definite improvements in comfort, both of the shoe and of the shin attachment. Furthermore, as the device is relatively stiff, there is a risk of joint misalignment. Also, the nature of the device requires that for both directions in a degree of freedom, two tethered actuators are needed. No recommendations are made on dorsiflexion support or correction forces. The biggest downside in the context of the current project is that the nature of the design does not allow for a large range of anchor point placement and thus does not allow proper joint alignment. The J-LO design can be seen in Figure 2 [12]. Note that the far-left image depicts an earlier iteration of the J-LO device, with a hydraulic actuation.

Another example of a device with a rigid connection is the design by Witte and their colleagues, with properties like the J-LO design, which can be seen in Figure 4 [13].

1.2.2 1DoF Bowden systems – Flexible connections

The second type of 1DoF Bowden systems is a prime example of an exosuit. It has no inherent stability, but fully utilizes the body's own MSS to induce plantarflexion support. It is therefore not suited for patients with severe muscular atrophy, but more suited towards patient with minor atrophy (either in rehabilitation or for permanent use). There have been some developments in this, but there is one device that is by far the most functional device on the market: the ReWalk Restore, which uses the apparatus developed by Conor Walsh and his team. The AFO can support both plantarflexion and dorsiflexion and uses just an insole in the patient's shoe to transfer forces to the foot. The device does however require quite an extensive apparatus on the shank, increasing distal weight and volume, as can be seen in Figure 3 [14].

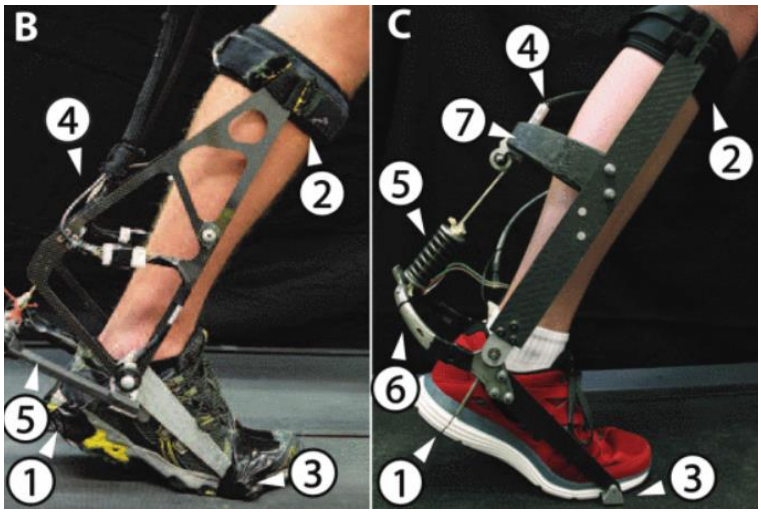


Figure 4: AFO design by Witte. Left type A and right showing type B.

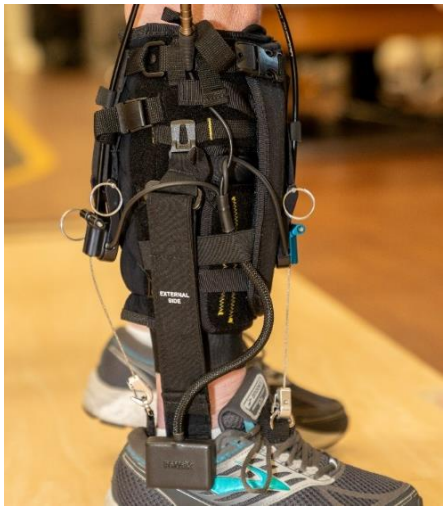


Figure 3: ReWalk Restore fully flexible AFO supporting PF and DF.

1.2.3 1DoF non-Bowden systems

The systems of the 1DoF non-Bowden type can be actuated by a linear motor, either using hydraulics, pneumatics, or electric motors. They all need rigid connections to actuate against. An example used to treat drop-foot in post-stroke rehabilitation is the device by Blaya [15], which can be seen in Figure 5, or the Achilles by the University of Twente in Figure 6 [16].

It must be noted that while these systems are often very sturdy, they do come with the downside of higher distal mass in comparison to Bowden systems, as the linear actuators add quite some mass.

1.2.4 2DoF non-Bowden systems

The field of 2DoF AFO's is relatively undiscovered. As control schemes for full gait support 1DoF systems has only been up to desirable standards in the last decade, not much research has expanded into 2DoF systems. At the University of Twente, one of the devices developed is the 2DoF, hydraulicly actuated AFO by Grootens [14], seen in Figure 7. This system fully allows for subtalar motion, but the relatively rigid design caused subjects to still report discomfort likely caused by misalignment.

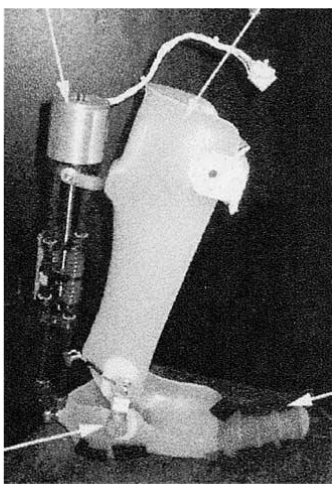


Figure 5: Blaya & Herr AFO.

Figure 6: Achilles AFO.

Figure 7: Grootens 2DoF AFO.

1.2.5 Quantitative measures of state of the art

The findings of the literature are quantified and summarised in the table in Appendix A, and the results are distilled to Figure 8. Note that the standard deviation is incredibly high. This can be explained by the fact that there is a large variation in AFO types.

	WEIGHT [G]	PF ROM [°]	DF ROM [°]	PF TORQUE [NM]
MEAN	1288	24.0	14.6	62.1
STD	879	8.9	4.6	44.1

Figure 8: Mean and standard deviation of all devices.

Correcting the data using only exosuits with Bowden cables yields very different results, with the design from Witte et. al [13] being the obvious standout, delivering well over 120 Nm of torque. The conclusion drawn here is that using stable steel structures and smart design the 120 – 150 Nm of torque is achievable, but using more delicate structures the torque range is under 20 Nm. This is summarized in Figure 9.

	WEIGHT [G]	PF ROM [°]	DF ROM [°]	PF TORQUE [NM]
MEAN	852	24.2	16.6	65.4
STD WITH WITTE	522	5.4	4.2	58.0
MEAN	851	20.3	14.3	19
STD WITHOUT WITTE	974	3.3	4.0	7.8

Figure 9: Mean and standard deviation of Bowden devices.

1.3 Biomechanics and dynamics of the ankle

The ankle is one of the most complex joints in the human body. In this chapter, the biomechanics of this joint during motion are briefly described, while also taking the variation over the gait into account. Finally, the effects of an assistive device on the kinematics and especially the dynamics are discussed.

Because of the geometry of the ankle the biomechanics is not as straightforward as for example the knee. Generally, motion of the body can be expressed in three axes and planes. Without losing too much accuracy, the ankle motion can be reduced to two axes, the PF/DF axis, and the subtalar axis. Rotary motion around the axis colinear with the lower leg is not a degree of freedom of the ankle, but of the lower leg. These motions are actuated by about 10 muscles. A contraction of one of these muscles in general causes motion in 2 planes. A cable-based actuation system will likely also have that effect, causing parasitic motion in nature. The orientation of these two axes is displayed in Figure 10 [10].

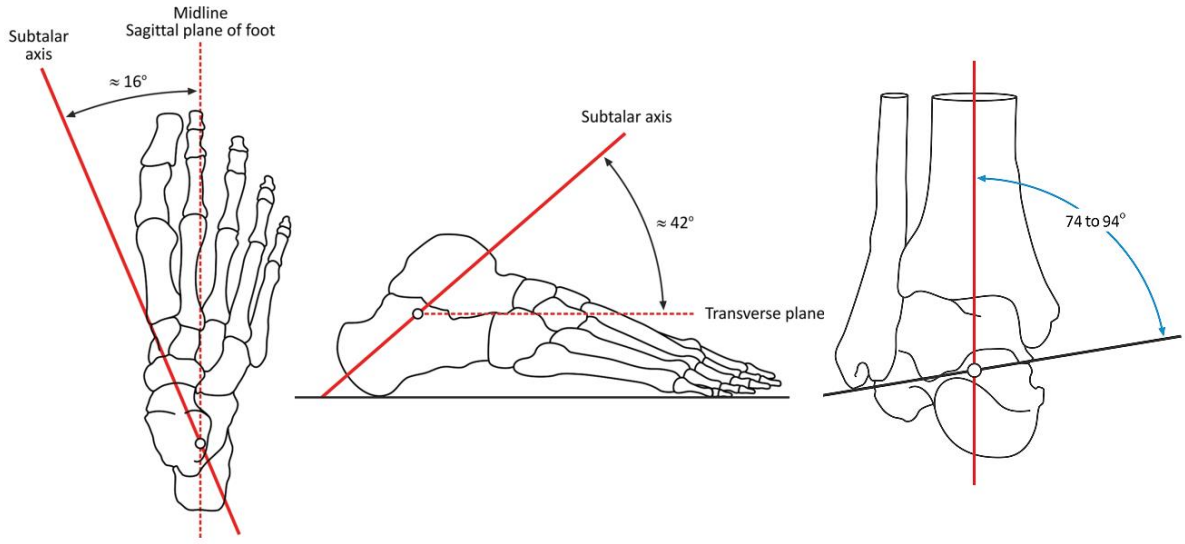


Figure 10: Axes of the ankle.

The gait cycle defined as the time from heel strike till heel strike of the same leg, also known as a stride. One stride consists of two steps, one for both legs. Recorded kinematics and dynamics data are usually normalised over the gait cycle, to accommodate easier comparison with other research. The gait cycle is displayed in Figure 11 [11].

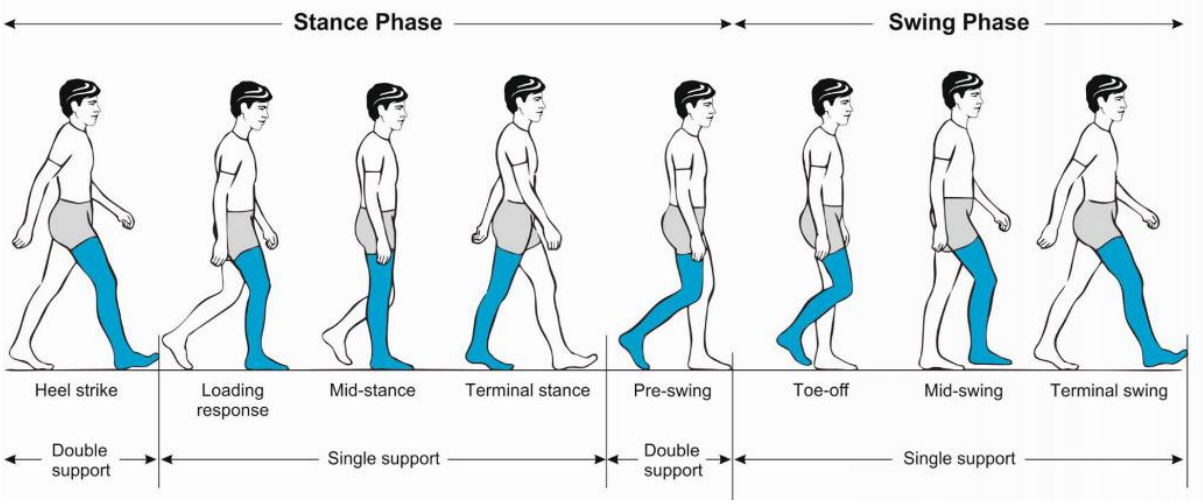


Figure 11: Gait cycle of the right leg.

For the purposes of this thesis, the gait normalised values of PF and DF are of interest, as well as subtalar motion. For personalised results, data must be recorded in a lab, which can then be displayed in a similar manner as Figure 12.

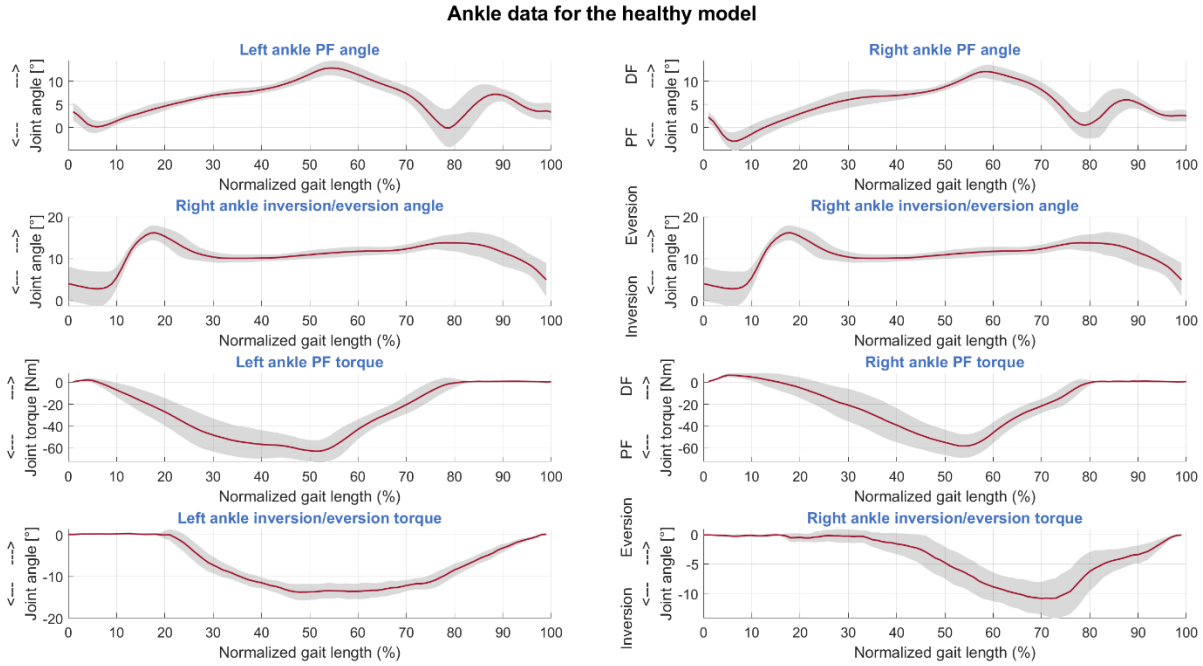


Figure 12: PF/DF and subtalar data over gait cycle.

For the second DoF, namely the subtalar motion, the interest is similar. However, the amount of research into this DoF is considerably smaller than into PF/DF motion.

1.4 Report structure

The thesis is divided in three parts. Part I of the thesis contains the problem definition, in which first the problem is defined and put into context in Chapter 1, where Chapter 2 works towards the requirements. Part I displays a biomechanical analysis, a review of available devices and the setup of the requirements.

The design process is discussed in Part II. This starts with the concept development in Chapter 3, and follows a standard synthesis procedure, working from a wide range of concepts to a final concept. Next, the mechanical analysis and concept detailing of this design is described in Chapter 4.

The thesis is concluded in Part III. Chapter 5 starts this off with the realisation and evaluation of the design, containing a design verification based on practical testing. This is followed by a discussion & recommendations section in Chapter 6, after which the final conclusions are drawn in Chapter 7.

The appendices contain more detailed documents supporting the thesis. Appendix A contains the full results of the literature review. The methodical product design can be found in Appendix B, and this document is more chronologically ordered than this main thesis document. Appendices C and D contain the results of the practical experiments, Finally, Appendix E contains a document on how to use and read the MATLAB script attached with this thesis.

2 Requirements & boundaries

From the background in chapter 1, the requirements can be defined. These are built up to from the problem definition, goals, and a system subdivision to better understand the boundaries of the project.

2.1 Problem definition by client

The client has defined this problem as follows. In the coming years, fine force delivery by an assistive device to humans will be necessary to guide precise and personalized ankle rehabilitation. The current solution for force transmissions in lower limb exoskeleton design relies heavily on rigid interfaces and force transmissions in parallel to the talocrural joint (allowing for plantar- and dorsiflexion), disregarding motions caused by for example the invertors and evertors (motion in the subtalar joint). Furthermore, for so much as the actuations are linear, they are limited to one line of action, or in the most recent cases both plantar- and dorsiflexion over one degree of freedom following a generic design. Isolated 1 DoF plantarflexion/dorsiflexion support and non-personalized targeting can lead to deterioration of cartilage, misuse of muscle, and tendon tissue and bone structures when using this support.

The one degree of freedom simplification furthermore limits the targeting of specific muscle groups. To do so in the end, a 2DoF device must be developed, starting with a proof of principle. To follow the natural actuation of the body as much as possible, the project must be carried out using a Bowden cable actuation. To provide better alignment, personalisation is desired, and thus this design process needs to include personalized biomechanical data in an early stage.

2.2 Goal

Therefore, what is needed is a new type of 2 degree of freedom interface, that can be easily personalised to a subject and the rehabilitation process. A device must be designed that can transmit loading in parallel to musculotendon structures to induce tissue homeostasis, which is the perfect environment for tissue healing. This must be done while providing a solution that is lightweight and precise; a new type of 2 DoF interface. With the ankle being a very complex joint and the PF/DF axis not being perpendicular to the anteroposterior plane, it is time to move the research from 1DoF systems to 2DoF and try to solve the misalignment problems still existing in a lot of systems on the one hand and apply precise forces to target specific muscle groups on the other. This project will aim to contribute to that, providing a proof of principle that can be used in either a research setting, a rehabilitation setting or possibly longer-term use, while also providing insight in the joint forces caused by the orthosis. This thesis ultimately will provide insight in the feasibility of 2DoF ankle-foot orthoses of this type.

The apparatus has the potential to improve functionality and free movement in the required settings. Also, using a cable actuated 2DoF device targeting specific muscles should be possible with further research. While the device can in principle contain structural elements between the shank and the foot, it is not primarily intended as a weight-bearing device and thus serves a different purpose than for example a full weight-bearing exoskeleton.

While there is a lot of research into the effectiveness of on-the-market devices, very little research is being done in the longer-term effects of (ankle) orthosis wear. Given the timeframe of the thesis, this is explicitly not researched.

2.3 Design assignment

The design assignment is to improve onto the state of the art in such a way that the total system allows for functional, personalized, precise plantarflexion and inversion/eversion support, allowing for small (force or orientation) corrections on the orthosis. This must be done within the boundaries of the requirements, where the joint torques are calculated based on model input, either through a MS model or alternative methods.

While there are alternative routes of solving this problem, the one described above is chosen. It was already explained that pharmaceutical or surgical solutions are possible but are not considered a decent first step. They are altogether discarded from here on out as this thesis is biomechanical in nature and tries to reduce the need for surgery or pharmaceutical intervention. Also, the solution of a passive device is discarded for multiple reasons:

1) While a passive device can be adapted to a patient and the environment, there are no adaptive possibilities once it is implemented apart from physically changing the device.
2) A passive device requires actuators (e.g.) springs to be installed distally on the body. This causes an increase in distal weight and does not satisfy the requirement that the Bowden cable actuation must be used.

However, in the future, the passive spring could potentially be used to assist in dorsiflexion. A passive dorsiflexion spring force could be overcome with the active plantarflexion force and when the active force is removed, the spring would dorsiflex the foot.

2.4 System subdivision

To make the total design assignment more explicit, the system is divided in four different interfaces. These interfaces link the musculoskeletal system to the actuation system, which is visualised in Figure 13, adapted from [14].
On the left, the normal functioning of a body is shown, with two principal muscles schematically depicted. On the right, a cable exosuit is shown. Note that the elements at the thigh are ignored for now, but in principle this reasoning would also apply to knee actuation. The purple patch at the shank is the musculoskeletal attachment where the actuation force (green arrow) connects to the Bowden cable actuation (red arrow). This force causes the purple patch and the heel to move towards each other, enabling plantarflexion. This results in two interfaces with the body and the force transmission in between them, yielding the three system elements referred to in this thesis. Analogous to this, inversion and eversion can be actuated (not depicted here).

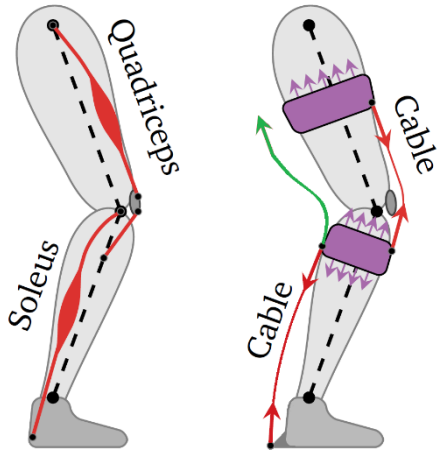


Figure 13: Schematic representation of system.

As the goal of this project is precise force delivery, which leads to multi degree of freedom actuation, the green force transmission cannot introduce any more constraints on the degrees of freedom. Since ankle and subtalar joints effectively constrain the three translational degrees of freedom in the joint, any added constraint will necessarily constrain rotation, which goes directly against the design goal. This eliminates the option to add joints on the interface and leaves the option for compliant materials. The connection can thus either be flexible or semi-stiff.

2.5 Use cases

For the final product a set of use cases must be defined to properly assess in what environments and under what conditions the device must operate. In first instance, the goal is to use the device in a lab setting for rehabilitation purposes. This means that the device can be made tethered for now and it can be assumed IMU's are available, a pressure sensitive treadmill can be used, and a motion capture system is present.

The device can therefore ultimately be used in rehabilitation, but probably first will be used in a research setting to test effectiveness. Some of the experiments that could be carried out include:

- Testing effectiveness in providing plantar-/dorsiflexion and subtalar flexion torque to correct for (mediolateral) perturbations during normal walking.
- Testing effectiveness in allowing natural subtalar motion during normal walking.
- Testing effectiveness in providing plantar-/dorsiflexion and subtalar flexion torque to correct for (mediolateral) perturbations during everyday activities such as stairs climbing, jumping, and running.
- Testing effectiveness in running a prescribed motion in the context of physical therapy.

- Finding the limit of reduction of the metabolic cost of walking of a (healthy) subject.

These experiments all are centred around a general set of use cases that range from donning and doffing to perturbation support. A list of use cases could be containing (but not limited to):

- Donning.
- Doffing.
- Standing.
- Stand to sit.
- Sit to stand.
- Normal walking (at certain speeds).
- Kneeling during stand.
- Bending during stand.
- Running (at certain speeds).
- Walking stairs.
- Walking terrains.
- Walking.

If the device were to work properly in all these use cases, a wide range of rehabilitations could be facilitated. If the actuation were to be made untethered in future versions of the device, it could even be used in daily life.

2.6 Requirements & boundary conditions

These goals must be realized within a certain framework. This framework is set by a list of requirements that relate to functionality (use) and design. The requirements are based on the stakeholders and on literature findings. They can be either fixed, meaning that whatever the design, they must be fulfilled, or variable, which means they are optimized in the design.

ID		Description
1. Use		
1.1		**Intended use**
1.1-010	f	The device shall be used in a revalidation environment
1.1-020	f	The device shall not be used for continuous wearing outside of rehabilitation facilities
1.1-030	f	The device shall actuate the human ankle (and subtalar) joint.
1.1-040	f	The device shall be usable on any patient going through the rehabilitation process.
1.1-050	f	The device shall be usable on any adult within the height range of [150, 190] cm.
1.1-060	f	If possible, the device shall be usable on paediatric patients.
1.2		**Personalization**
1.2-010	f	The total system (including actuation) must enable precise force delivery in personalized capacity.
1.4		**Users**
1.4-010	f	The device shall be applied by instructed personnel only.
2. Design requirements		
2.1		**General device design requirements**
2.1-010	v	The device shall be produced with available materials and methods at the UT facilities.
2.2		**General interface requirements**
2.2-010	v	The device shall emulate & enable natural motion of the body during normal walking as much as possible.
2.2-020	v	The device shall hinder natural tissue motion of the body during normal walking as little as possible.
2.3		**General system design**
2.3-010	v	The distal weight shall be minimized to reduce fatigue and extra needed muscle action.
2.3-020	v	The device shall be able to accurately measure applied forces (either directly from moments or forces, or any other sensor input).
2.3-030	f	The device shall be able to accurately measure position (expressed in terms the ankle DoF's).
2.3-040	v	The device shall be able to measure position with as little external sensors as possible.
2.3-050	f	The net joint reaction forces should be within acceptable ranges when using the device.
2.4		**MS - shank connection**
2.4-010	f	The device shall provide a stable connection to the shank such that precise force delivery is enabled (5 - 10% error max.)

ID		Description
2.4-020	v	The shear forces shall be kept within acceptable ranges such that ultimately the shank connection does not slip down during normal operation. This shall be achieved while still allowing for 2.2-020 and thus not with increased normal forces.
2.5		**MS - foot connection**
2.5-010	f	The device shall provide a stable connection to the foot such that precise force delivery is enabled (5 - 10% error max.)
2.6		**Foot - shank connection**
2.6-010	f	The device shall be able to support the full range of motion during normal walking for the ankle and subtalar joints.
2.6-020	f	The device shall be able to actuate under joint rotational speeds during normal walking.
2.6-030	v	The device shall provide an 'area of flexibility' around the joints to enable multi-DoF actuation.
2.7		**Actuation interface**
2.7-010	f	The device shall be connected to the current power supply and actuation chain in the lab (Bowden cable actuation).
2.7-020	v	The device interface with the current power supply and actuation chain in the lab shall be easily engaged and disengaged.
2.7-030	f	The device shall be able to accurately assist up to 50% of the bodyweight normalized ankle and subtalar torques during normal walking.
2.7-040	f	The device shall be able to deliver the assistance over multiple lines of action
2.7-050	f	The device shall not be able to provide more torque or force than the body could provide over that line of action.

3 Concept development

As could be seen in Figure 13, three main components must be developed. This was done systematically according to the methodology as set out by Prof. Dr. Ir. G.J. Verkerke. From the problem definition in part I of this thesis, a final design can be derived. The first step is to decide on the joint type used as the design of the shank and foot brackets largely depends on this, as well as the number of actuators used. These joint types are developed from the state of the art and this part of the design is mostly conceptual. After the joint type has been picked, the designs of the shank and foot bracket are chosen, as well as the number of actuators. From this, several concepts are defined and weighed against the requirements. After the funnelling to the final concept, this concept is detailed into a final design. All these results are worked out in the next part of the thesis, part III. Part III of the thesis can be seen as an answer to the first part of the design goal: the development of a novel type of 2DoF interface for ankle plantarflexion and subtalar motion. For a more extensive explanation of the design process, the reader is referred to Appendix B.

3.1 The first step

Synthesis I is the process of setting up as much as possible conceptual directions and distilling those into concepts. Motion of the foot can be reduced to two degrees of ankle freedom (plantarflexion and dorsiflexion, and inversion and eversion) and one degree of foot freedom (external and internal rotation of the foot), as can be seen in Figure 14 [14].

Since the goal of the design is to actuate the two degrees of ankle freedom, the starting point of the design is what joint must be used by the system at the ankle joint, what range of motion this joint must have and how the actuation forces are translated between device, actuation, and body. This means that the most important aspects of synthesis I are:

- Joint type.
- Range of motion.
- Force transmission.

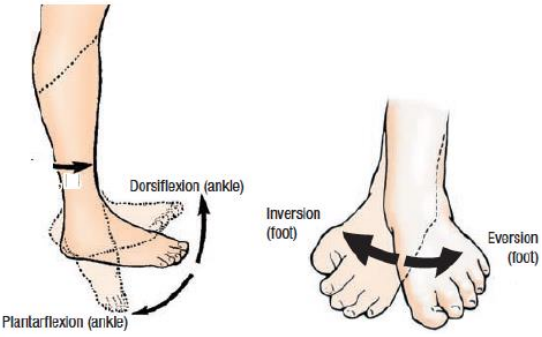


Figure 14: Ankle degrees of freedom.

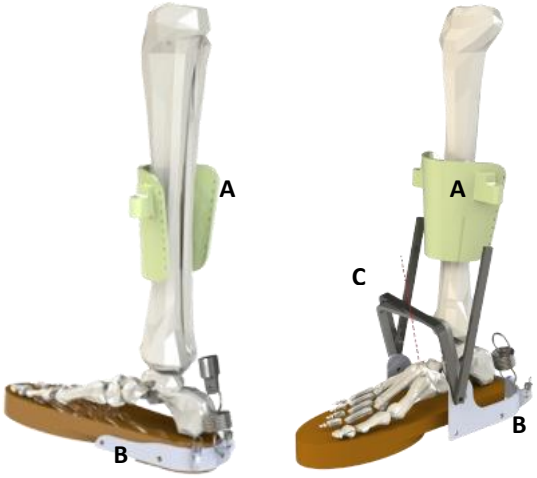


Figure 15: 2 joint types in the concept phase. A: Shank bracket, B: Foot bracket, C: Joint mechanism.

3.2 Joint type

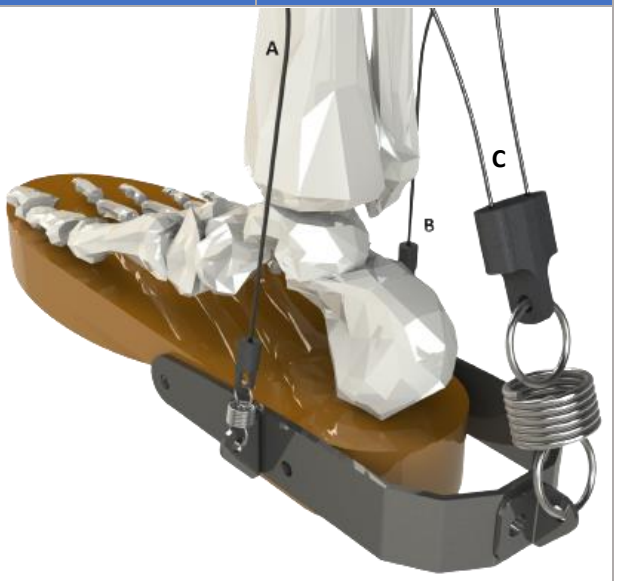
The joint types range from no joint to a mechanical connection like a U-type joint. If the goal is to actuate all degrees of freedom (3) at the location of the ankle, a spherical joint type is necessary since all DoF's are rotational. As there is an ankle present at the centre of the joint, this must then be realized with a 3-axis joint. Two of the axes must be at the ankle, while the third axis can be either at the ankle or higher up at the leg. Note that to realize this a rotation structure must be built around the leg, effectively creating a pin joint with the axial direction on the proximal/distal axis of the lower leg. The tibia and fibula structure of the leg rotates with respect to itself at the lower leg and thus the exosuit could actuate this DoF there. This will create such a complex design, while the requirements do not ask for 3 degrees of freedom. For further material on this exoskeleton joint, refer to chapter 6.3 of [14]. Removing that DoF gets a 2-axis joint, also known as a universal joint or U-joint, left side

of Figure 15. This is very similar to the design of Martijn Grootens with his pneumatic AFO Gradually removing material from the joint yields several different concepts ranging trough compliant materials, mechanical solutions and finally to the no-joint concept, which can be seen on the right side of Figure 15.

One of the primary aspects is the amount of force that can be transmitted to the shank without causing discomfort. After testing this, this is found to be in the range of 150 – 250 N (shear force). After weighing of the joint types based on three aspects, namely RoM, transmission of torques, forces and stresses and general bulkiness, three joints remain over the other joints, see also Figure 16. As the limitation on the force in the 150 – 250 N range would likely affect any concept based on the no-joint, this limitation is checked with the client, who had no objections. Therefore, all the three joint types are continued into the concept development.

~~~ Explanation starts below figure ~~~

<table>
<tr><th>Specifi-cation</th><th>Option 1</th><th>Option 2</th><th>Option 3</th></tr>
<tr><td>Actuation</td><td colspan="3">3 Bowden cables

A – Inversion cable
B – Eversion cable
C – Split plantarflexion cables
 
</td></tr>
<tr><td>Joint type</td><td>Joint type A: No connection
  </td><td>Joint type D: Compliant connection
  </td><td>Joint type E: Universal joint type
  </td></tr>
</table>

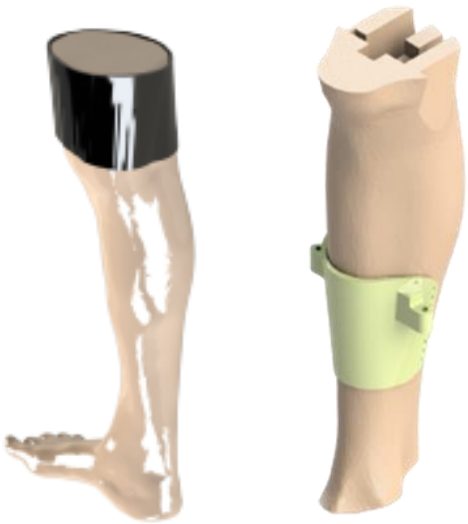
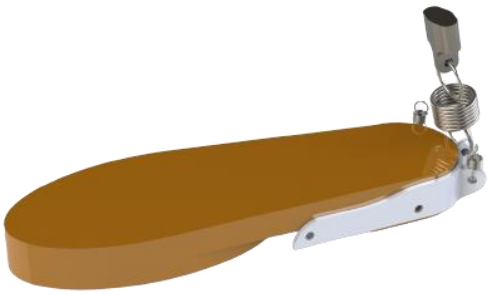
Specification	Option 1	Option 2	Option 3
Shank connection	Distal, anterior rigidity 	Distal, anterior rigidity + upper leg 	
Foot connection	Bracket connection 		

Figure 16: Morphological chart with the shank bracket in green, the proximal band in black, the foot bracket in silver metal and the shoe in brown.

3.3 Concepts

Based on these joint types, a morphological chart can be developed, seen in Figure 16. From this chart, 4 concepts are drawn up, seen in Figure 18. Note that all concepts utilise the 3 Bowden cable actuation. There are two main reasons for this.

- Firstly, the attachment points must be placed in line with the axes of the ankle to allow the best functionality (see further explanation in chapter 4.1). Using only 2 cables requires an optimum average in the placement but will ultimately reduce functionality.
- This also touches on the second reason, as the plantarflexion and inversion/eversion profiles do not match and thus will necessarily induce parasitic motion if not actuated individually. In theory this can be solved mechanically, but this will highly increase the design complexity.

3.3.1 Concept 1: Joint A, no upper leg band

This concept direction has no direct connection between the foot and the shank besides the existing ankle joint. The interfaces must thus be tightly connected to the leg and the foot, and all the force exerted by the system is fully translated to the musculoskeletal system as joint reaction forces. The torque in the system can be tracked by knowing the moment arm at a given position and measuring the force in the cable through a load cell. The advantage of this concept is its simplicity, causing low distal weight and no constraints on the degrees of freedom. The clear downsides include the higher force on the shank and the position tracking not being possible with an encoder. The concept is depicted in Figure 18a.

3.3.2 Concept 2: Joint A, upper leg band

his concept direction has no direct connection between the foot and the shank besides the existing ankle joint. The interfaces must thus be tightly connected to the leg and the foot, and all the force exerted by the system is fully translated to the musculoskeletal system as joint reaction forces. The shank connection is stabilized by a leg band. This leg band will be flexible as the upper leg does not have a low mechanical compliance. The torque in the system can be tracked by knowing the moment arm at a given position and measuring the force in the cable through a load cell. The advantages do not change with respect to concept 1, but it does distribute the forces better over the leg and thus could improve some aspects of concept 1. Figure 18b shows this concept.

3.3.3 Concept 3: Joint D, no upper leg band

An alternative to concepts 1 and 2, which do not use a connection between shank and foot, could be the use of compliant materials. Tuning material and geometrical properties provide a connection that allows for plantarflexion and dorsiflexion through flexion of the material. Subtalar flexion requires lengthening and shortening of the connections, which can be achieved through some sort of folding motion. The compliant connection can vary from traditional springs to compliant (synthetic) plastics. This concept allows for a wide range of position sensing possibilities, ranging from traditional encoders to using conductive properties of the material to sense rotation. It also limits the forces with respect to concepts 1 and 2, while it retains flexibility. However, compliant mechanisms could cause parasitic forces by their inherent spring-like characteristics and the proper balance on geometry and material might be difficult to achieve. This concept is visualized in Figure 18c.

3.3.4 Concept 4: Joint E, no upper leg band

Whereas concept 3 tries to add stiffness problem by compliant materials, thus allowing for both translation and rotation of the joint, this concept aims to reduce the degrees of freedom to two rotational ones. Two U-brackets, joined in their centres by a pin joint, attach to pin joints at the foot and the shank. This universal joint type of motion allows for plantarflexion, dorsiflexion, and subtalar motion. Utilizing three cables, plantarflexion and subtalar motion could be actuated. Dorsiflexion would require another actuator. A variety of sensors can be used, as the components only rotate with respect to each other and thus have little play. The rigid system causes little forces to be transferred to the body and allows for very stable motion. The downsides include mechanical complexity and a large(r) distal weight. Finally, Figure 18d shows this concept. Figure 17 shows the working principle of this concept. The three axes of rotation allow for 2DoF motion (the axes are not independent).

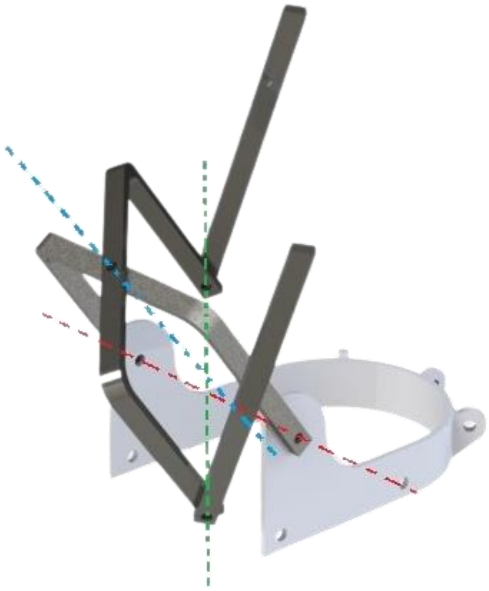


Figure 17: Working principle of concept E with the 3 axes of rotation.

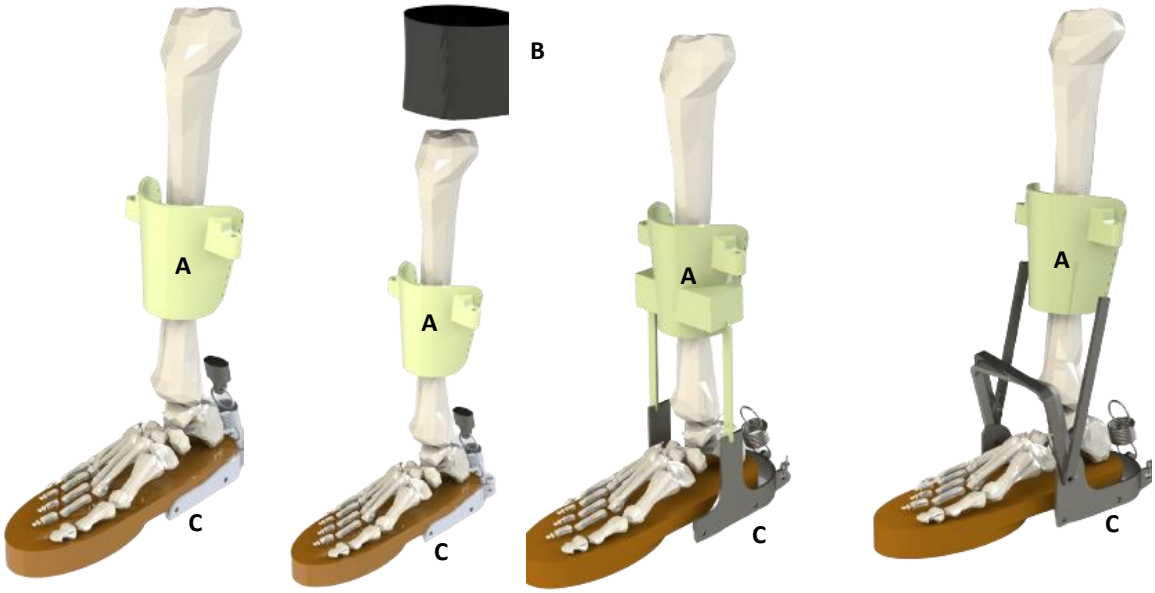


Figure 18: Concepts 1 through 4. A: shank bracket in different configurations. B: Proximal band for leg. C: Shoe and foot bracket in different configurations.

3.4 Concept choice

These concepts are weighed against a selection of the variable requirements. Not all the requirements are included, as some may be too specific in this phase. These requirements can be seen at the rankings in Figure 19. The concepts are then ranked on their ability to fulfil the requirement on a scale of 1 – 5 (very bad, bad, average, good, very good). This rating is then multiplied with a factor of 1 – 3 based on the relative importance of the criteria, and these results can be seen in their respective paths.

After ample discussion with the client, and his clear preference for a joint type A based concept, the decision was made to start with concept 1 and add the band that differentiates concept 2 from one at the instance that this is necessary for the functioning of the device.

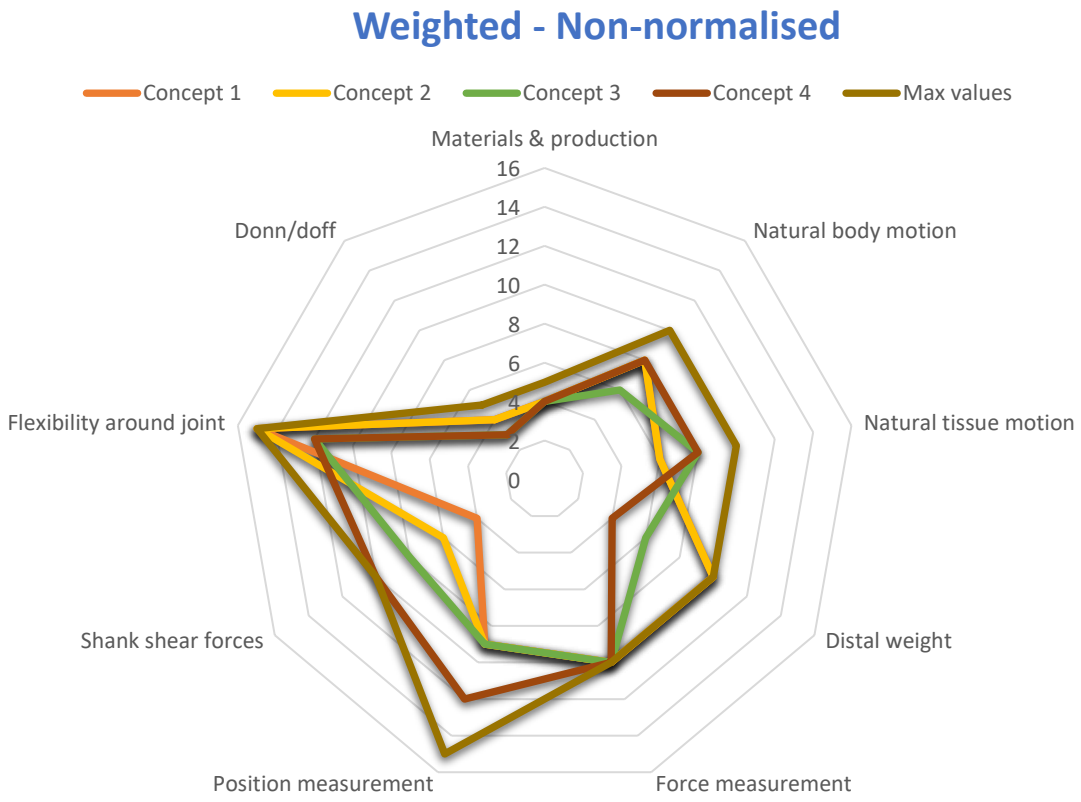


Figure 19: Weighed concept comparison.

4 Concept detailing

This chapter contains the detailing of the concept chosen at the end of Chapter 3. It will start from the required motions and forces and translate that to a detailed solution able to withstand those forces and provide that motion. This entails for example calculations on the mechanical properties of the brackets, but also looks at the necessary friction forces to transfer those forces to the musculoskeletal system. Note that all calculations in this chapter are based on the data of Huawei Wang and his colleagues [17]. For this chapter's figures, subject 4 is chosen as a base.

4.1 Principal design

The starting point for the final design is the required range of motion, during normal operation. To individually actuate plantarflexion, inversion and eversion, three actuation points are needed, as with less actuators, parasitic motion is necessarily induced on one of the non-actuated degrees of freedom. For the first iteration of the design, the location of these actuation points is chosen with respect to a superior point of view of the projection of the ankle axes on the transverse plane of the foot. The plantarflexion actuation point is located at perpendicular to the PF/DF axis at the calcaneus. The inversion/eversion actuation points are located on either side of the foot, perpendicular to the subtalar axis. In the first design iteration, the inversion/eversion actuation points were placed at the back of the bracket, but this proved to cause difficulty actuating the inversion motion as the moment arm is not maximalized, causing a larger force for the same torque.

The necessary actuation forces that are induced on the bracket can be determined from the inverse dynamics of the patients' motion analysis and the moment arm of the device over the gait cycle. It is assumed that the device is attached rigidly to the bike shoe and that the bike shoe is attached rigidly enough to the calcaneus to justify the assumption that any point on the bracket (behind the calcaneus) will follow a comparable path to the plantarflexors' moment arms in the patient's body. The moment arms of the three most important plantarflexors, the soleus, the gastrocnemius and the tibialis posterior are shown in Figure 20.

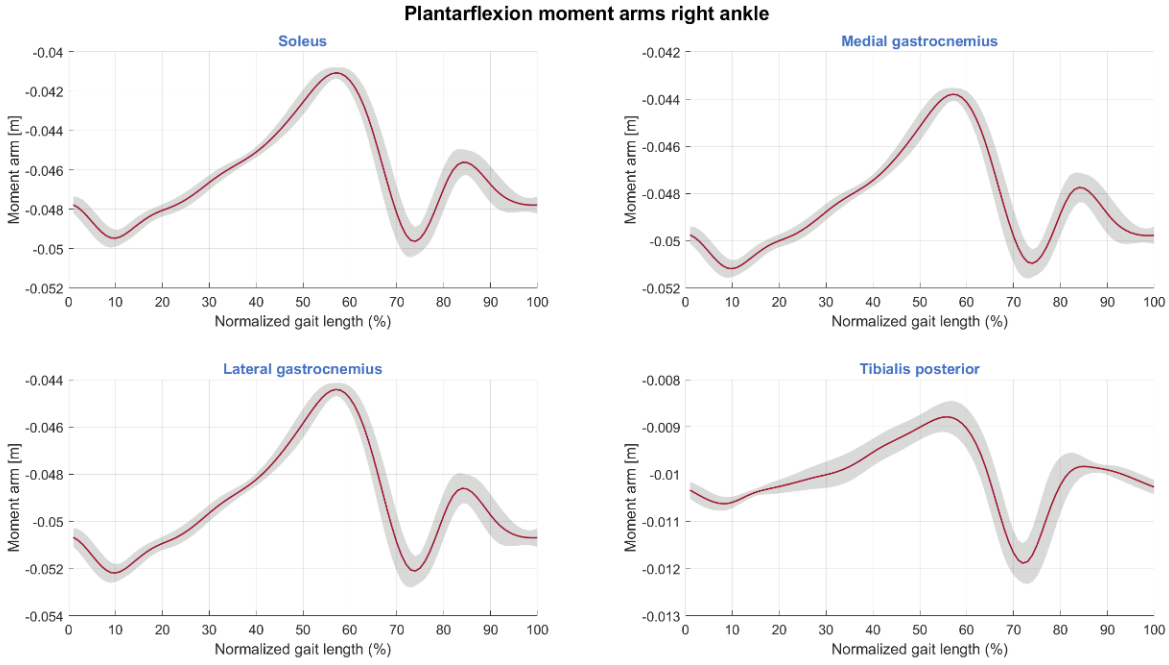


Figure 20: Muscle moment arms for plantarflexion. The red line indicates the mean value, with the shaded area being the $\pm\sigma$ values.

Averaging those out and taking a ratio of the device moment arm to a muscle moment arm at a known instance yields the curve in Figure 22 for the device moment arm. Note that this moment arm is taken at the attachment

point if the foot bracket was designed such that the plantarflexion actuation point was directly behind the calcaneus. This means that this is the smallest possible moment arm in the current configuration. The moment arm is of course approximately equal for both legs.

Now, the inverse dynamics results for the data reveal a moment profile which can be normalized over the gait cycle. If the goal is to apply 20% support, the required torque profile is shown in Figure 21.

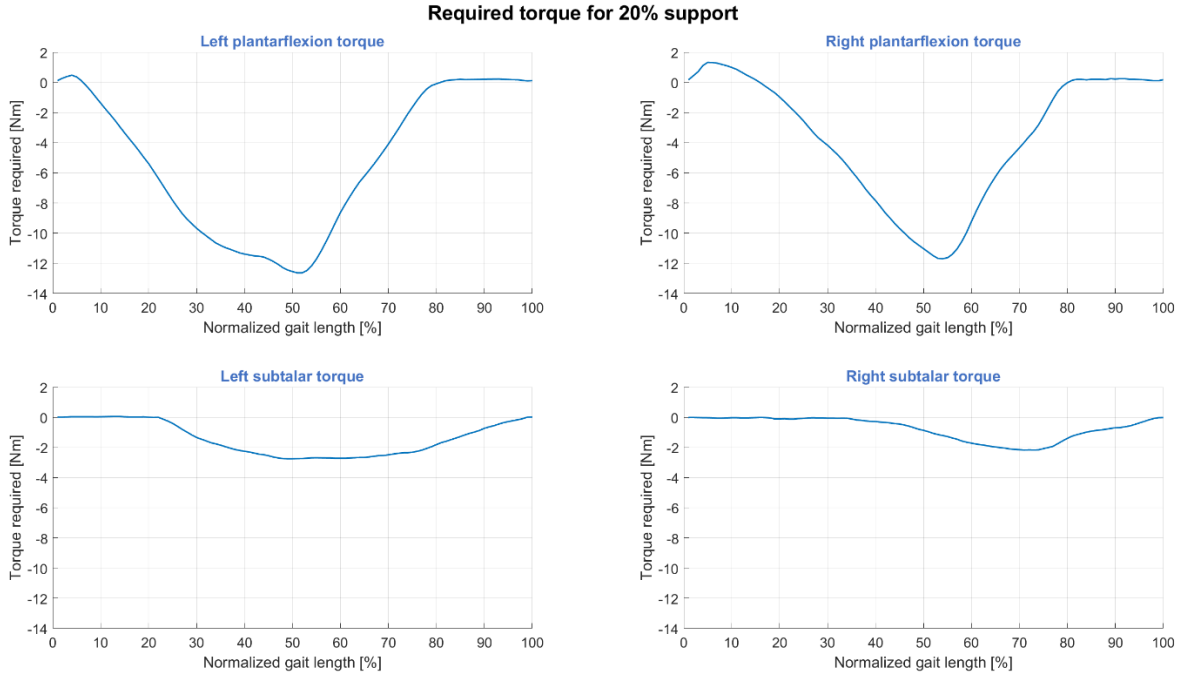


Figure 21: Required torque profiles at 20% support.

Dividing the necessary torques by the moment arms yields a force profile for both legs. Applying the shortest possible configuration of the device yields the blue force profiles for both legs in Figure 23. It can easily be seen that the peak force exceeds 300 N. From the force transmission test performed, the comfortable shear force limit on the shank bracket is for now in the range of 150 – 200 N. It's also visible that this is reached at around 25 mm of extension of the bracket from the shortest design. Thus, 30 mm is chosen as the base length for the design.
Given the design of the bracket, this yields fixed values for the inversion and eversion moment arms. Note that the inversion moment arm is shorter than the eversion moment arm, but they are assumed to be equal (to the shortest) for the purpose of the design.

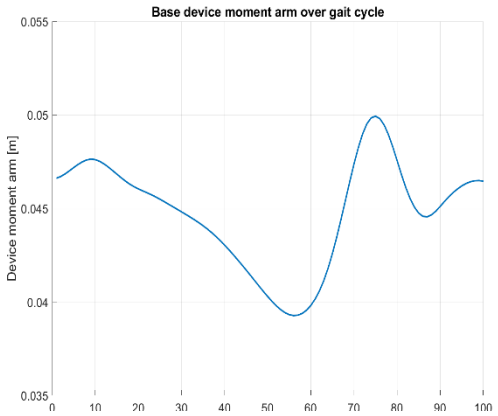


Figure 22: Device moment arm.

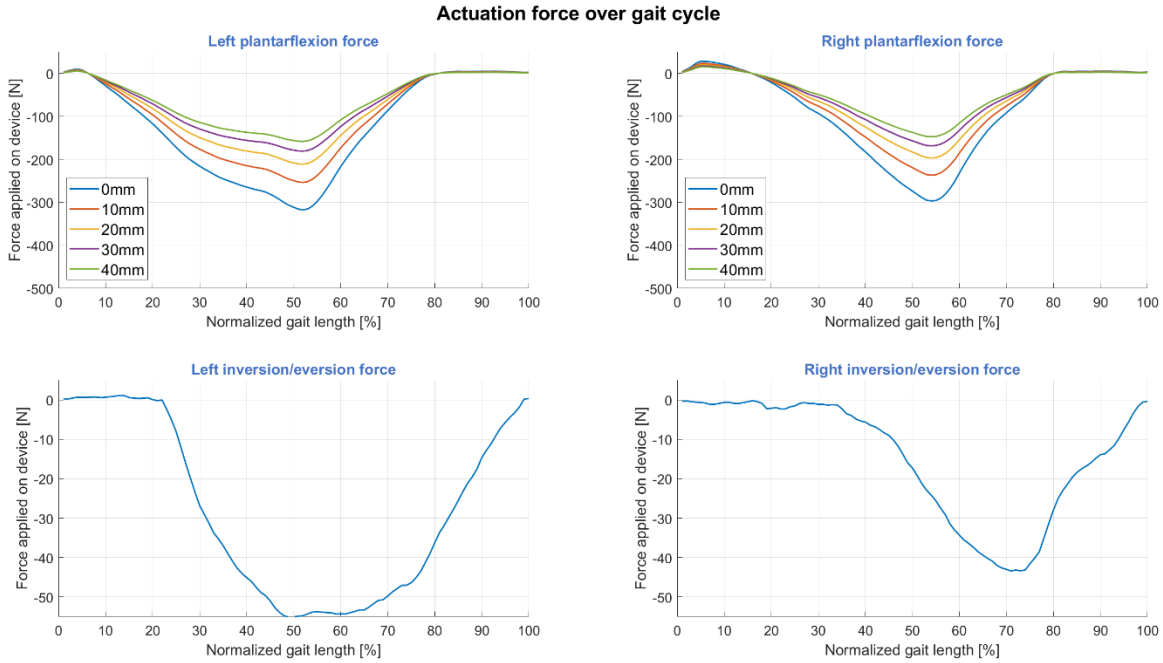


Figure 23: Required actuation forces.

4.2 Mechanical design

This section will deal with the mechanical design of the components needed for realising the principal design. This starts with the dimensioning of the foot bracket and the shank bracket, a discussion on the mechanical safety, a fatigue analysis and finally the mass estimation.

4.2.1 Dimensioning of components

As the foot is fitted in the bike shoe tightly, the transfer of the forces is not a problem in this case. This means that the potential weak link in the system is the mechanical design itself. Therefore, the connections and materials must be checked to be able to withstand the actuation forces. As calculated before, the actuation force likely will not exceed 200 N in normal operations. However, any other use case than normal walking will require larger applied torques and thus larger forces with the same design. This would require the friction slip problem described later to be solved, but in this case the bracket should not be the weak link. To make it operable under these conditions, the actuation force for these calculations is set to be 400 N, which should be more than enough for future improvements. Any standard bolt values are retrieved from the Applied Eurocode tables for standardized bolts [18].

It can be assumed that the pieces of metal that connect the actuator to the bracket are rigidly attached through a bolt connection, as the standard galvanised 8.8 M8 bolt has a clamping force (21.1 kN) and maximum shear force (14.1 kN) at least an order of magnitude larger than the applied forces (0.4 kN). Therefore, the bracket can be considered as one rigid piece.

For the dimensioning of the bracket, a final element analysis is run via SolidWorks at several geometries, all assuming S235 (standard construction steel) as a material. The parameters varied are the bracket thickness and the total bracket height, as can be seen in Figure 25.

These values are collected in a 3D grid and interpolated to yield respectively the maximum Von Mises stress in and the maximum displacement of the bracket in Figure 26. Also, the mass is plotted in Figure 24.

The connection to the bike shoe is made through 2 M6 bolts or alternatively through 2 pens with a diameter of 6 mm. As the centre distance between the 2 holes is approximately 1.15 times smaller than the distance from the middle hole to the actuation point, the maximum force exerted on the pins is around 460 N. This is again way under the standard clamp force and shear force values for a galvanised 8.8 M6 bolt or pin (respectively 11.6 kN and 7.72 kN).

With this, the final configuration of the bracket can be chosen. This is set at a bracket thickness of 3 mm, with an additional height of the bracket of 10mm (seen at the red dot in the figures).

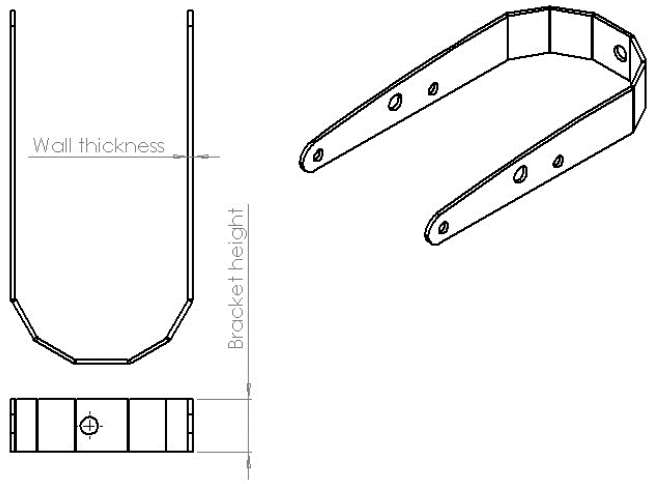


Figure 25: 2D drawing foot bracket.

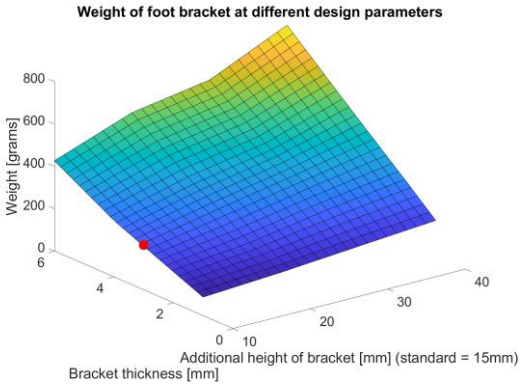


Figure 24: Mass of foot bracket. At the x-axis, the height, y-axis the thickness and the z-axis the weight.

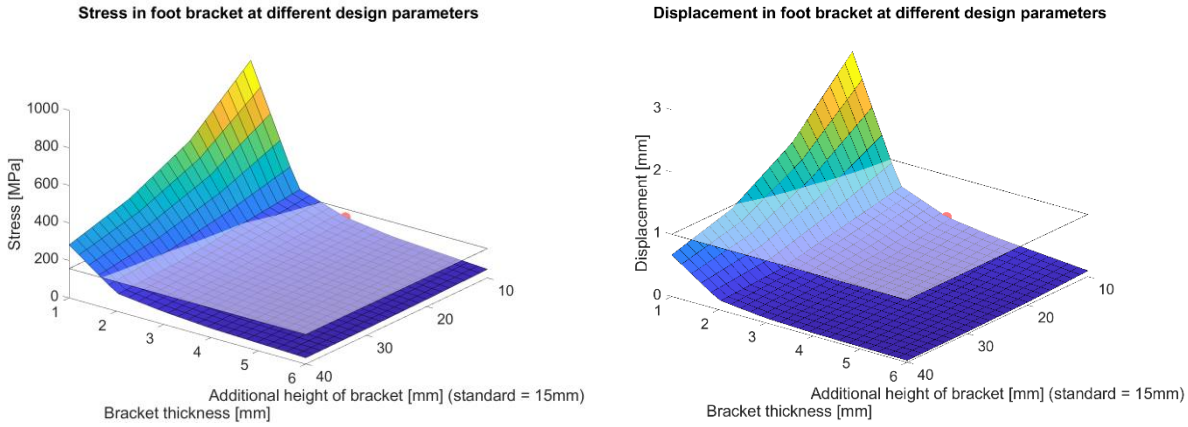


Figure 26: Stress and displacement in the foot bracket. At the x-axis, the height, y-axis the thickness and the z-axis the stress resp. displacement.

The shank bracket is lined with silicon and attached to the shank. The silicon liner could be padded, but as the primary concern is functionality, this will not happen at the first iteration. The same forces are applied as on the foot bracket.

The foot bracket only contains one attachment point at the calcaneus. To actuate this, ideally the connection point to the shank bracket is at the calf muscle. However, as the calf muscles bulge, the shank bracket is only solid at its anterior part. In theory, an attachment point could be made on the flexible posterior part, but with stability and functionality taking prime importance, the attachment point is placed at the rigid part. This does mean, however, that the plantarflexion actuation force will induce a parasitic torque. Therefore, the choice is made to split the Bowden cable around the leg, as can be seen in Figure 27.

Also, the orientation of the attachment points must be discussed. The holes of the plantarflexion supports are of course placed as close to the body of the bracket as possible, to reduce any bending moments in the frontal plane. They are also placed in line with the average orientation of the actuation line from the shank to the foot, to reduce shear torques. Now, the forces are of course not always in line with the bracket, which means that fatigue will come in as an important factor, with variations in both the force profile and the orientation of the vector over the gait cycle.

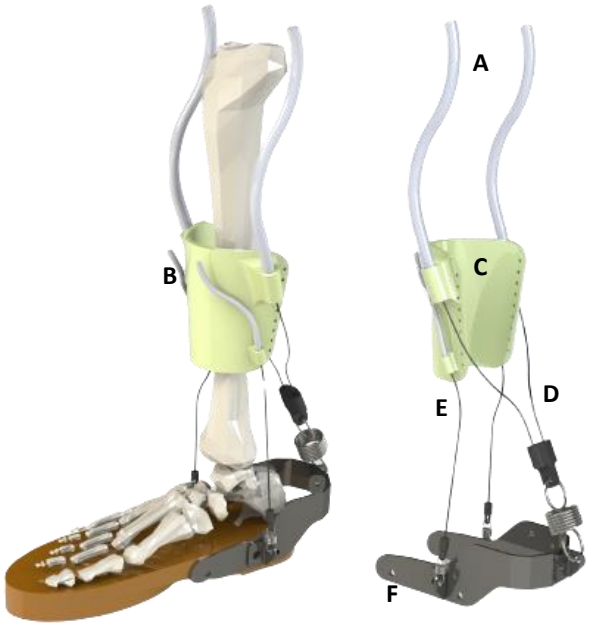


Figure 27: Overview of Bowden configuration and full render. A: 2 plantarflexion Bowden sleeves. B: Medial and lateral Bowden sleeve for resp. inversion and eversion. C: Shank bracket. D: Medial and lateral split of the plantarflexion Bowden cable, joined at the end to the SEE. E: Inversion and eversion Bowden cables, with at the end the connector and SEE. F: Foot bracket, connected to a bike shoe.

The added benefit of not placing the plantarflexion attachment points in line with the leg is that the forces are not solely transmitted as shear stress. However, if the forces are not completely perpendicular to the shank, the side effect of rotating the attachment point is that the bottom edge of the bracket will bury itself in the skin of the shank, which could be painful. The attachment points placement can be optimized in later iterations.

During the walking trials, the foot bracket was discovered to be unfit for walking. With the heel strike, the bracket design as set out in the first iteration would touch the ground before the heel, so the design was altered accordingly. The results have already been included in Figure 27 and as nothing changed functionally, this poses no problems in the design.

4.2.2 Mechanical safety limit

As there is no rigid connection between the shank and the foot apart from the ankle, in theory the Bowden cable could keep applying the force in case of a malfunctioning device, even if the ankle is in maximal plantarflexion. Naturally, this is not safe on a healthy ankle, let alone on a patients' ankle. Therefore, the Bowden cable must not be allowed more actuation length than necessary for plantarflexion actuation. This can be calculated by taking the distance between the attachment points at the foot and the shank at maximum dorsiflexion and maximum plantarflexion. This evaluates to a length of about 65 mm. The easiest way to ensure this safety is to either place a rigid sleeve or a clamp on the cable between the shank and the foot. For the first iteration, a clamp is chosen to suffice.

4.2.3 Fatigue analysis

The total system is subject to fatigue. This causes the stress resistance of the materials to reduce over time. To estimate the fatigue levels, the S-N curves of the materials are needed. For the foot bracket (S235), this is readily

available and well-established [19]. Using SolidWorks, both load events (plantarflexion and inversion curves) are added as a nondimensional factor on a 1N static load case and calculated for 1.000.000 cycles, which should amply cover the number of cycles the device will run. The load history curves (at the x-axis the gait cycle time) and the 1.000.000 cycle damage chart can be seen with most of the part remaining well under one percent of the estimated life in Figure 28. The force curves are applied at the points marked A and B respectively.

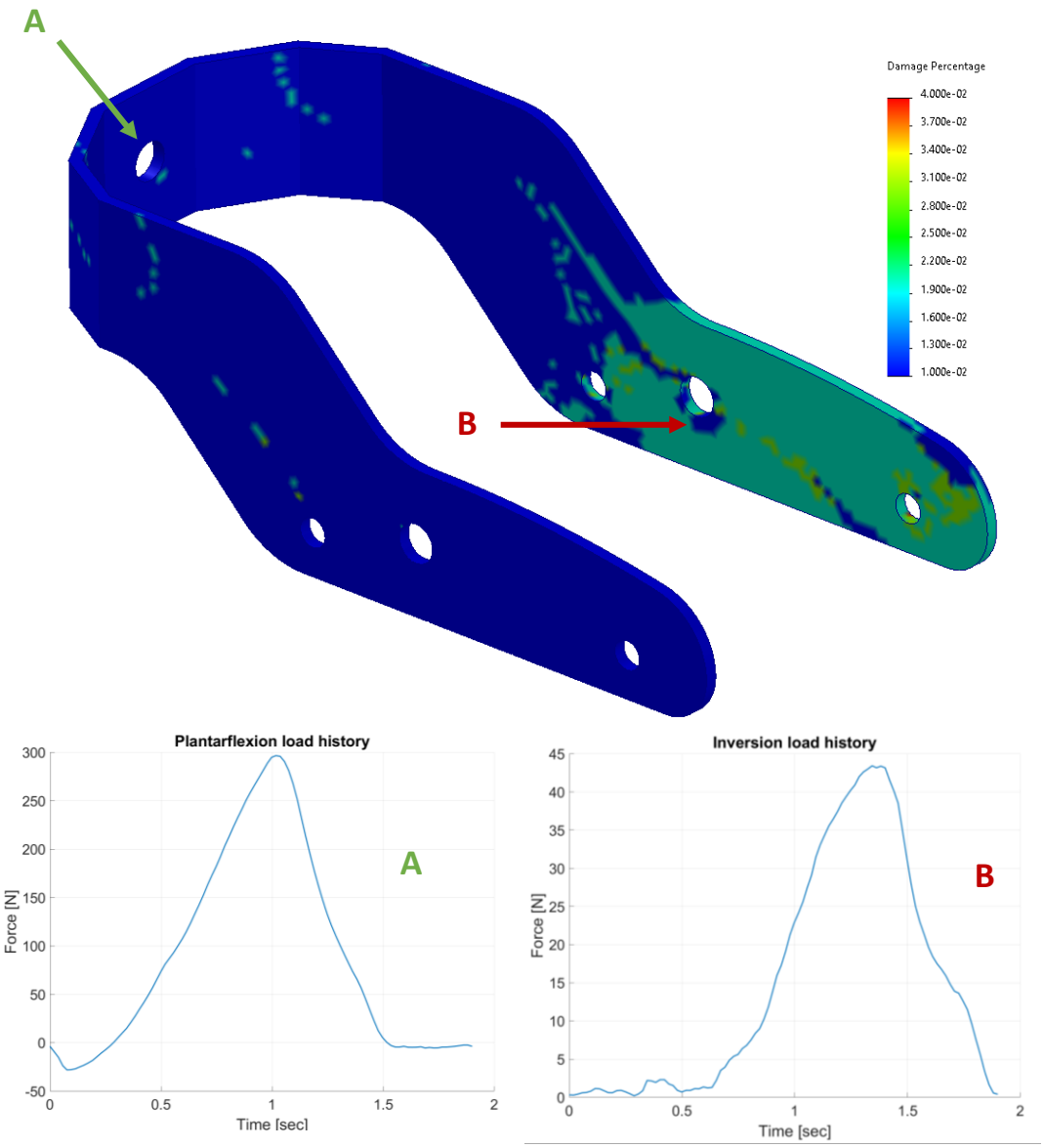


Figure 28: History curves and damage percentage.

The values for 3D-printed plastics are more in the experimental phase. As the shank bracket must be fairly strong, but ideally not brittle, a good first material to print would be PLA. For the SN-curves, the data from Safai et. al. is used [20]. The same history curves are applied on the shank bracket, and it is found that the bracket will likely fail between under 150.000 cycles. This is less than the foot bracket. However, as the bracket is 3D-printed with cheap plastics, the only problem that arises with this is that if it must be replaced, the Bowden cables and sleeves must be rewired. The results of this study can be found in Figure 29 at 100.000 cycles.

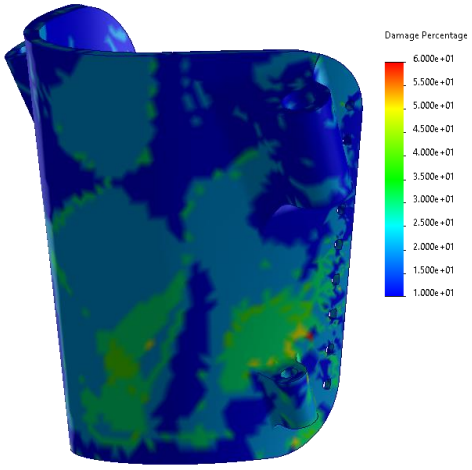


Figure 29: Shank bracket damage.

4.2.4 Mass estimation

The total mass of the system has quite some influence on the ease with which the device is handled. The total mass of the system can be derived from Figure 30. The weight of the Bowden cables is not entirely attributable to the system, so the mass is displayed with and without the cables & sleeves. Note that this mass can still be significantly reduced by for example taking an optimized bike shoe.

Element	Mass [grams]
Foot bracket including attachment points and nuts/bolts	287
Shank bracket anterior part	75
Shank bracket posterior part	75
Bike shoe (size 44) + size 6 pin and 4 screws	420
PF SEE + attach	80
Inversion/eversion SEE's + attach	20
PF Bowden sleeve (per meter)	55
Inversion/eversion Bowden sleeves (per meter)	110
PF Bowden cable	66
Inversion/eversion Bowden cables	58
Total with Bowden assembly	**1411**
Total without Bowden assembly	**1122**

Figure 30: Mass estimation of device.

5 Verification & realisation

Now that the concept is detailed theoretically, the design must be verified. This mainly entails a check against the requirements. Furthermore, it is important to deliver a proof of concept on the one hand and see the effects of the device on the MSS on the other.

5.1 Design verification

One of the prime verification elements is the weighing against the requirements. Figure 31 describes any requirements that have either not been met or have been met in an adapted manner. If the requirement is not in this figure, the requirement was fully met.

Requirements
1.1-050: The device shall be usable on any adult within the height range of [150, 190] cm.
While there were no specific trials done to see the limits of the device in terms of patient length, most patients should be able to benefit from the device. The device is mainly limited on torque, or in other words on heavier patients, which might correlate with the height of the patient. A body weight of under 80kg is ideal, which might not be realistic for patients over 180 cm.
1.1-060: If possible, the device shall be usable on paediatric patients.
With the previous requirement explanation comes the upside that lower body weights are more easily supported, meaning that paediatric patients could benefit. This should of course be researched separately, and certification should be gotten accordingly.
1.3-010: The total system (including actuation) must enable precise force delivery in personalized capacity.
The system has three actuation lines, which allow for much more personalized force delivery than with one actuation line. There is no research done yet in how to actuate the muscles specifically.
2.2-020: The device shall hinder natural tissue motion of the body during normal walking as little as possible.
The foot bracket causes no hindrance; however, the stiff bike shoe does limit the typical rolling of the foot and could cause slight deviations in the readings and natural walking. The shank bracket is causing a little discomfort with inversion of the foot, with pure plantarflexion there is no discomfort and free motion is enabled.
2.3-010: The distal weight shall be minimized to reduce fatigue and extra needed muscle action.
The weight is minimised and compared to the state of the art. For a detailed comparison, refer to Figure 32.
2.3-020: The device shall be able to accurately measure applied forces (either directly from moments or forces, or any other sensor input).
The use of the cables allows for precise force measurements in series with the cables. This can be done by either a force sensor or measuring the excitation and multiplying this with the spring constants.
2.3-030: The device shall be able to accurately measure position (expressed in terms the ankle DoF's).
The lack of stiff elements in the design prevents the use of an encoder. This means that the device must always be used with external motion capture systems or use an inertial measurement unit (IMU, like Xsens).
2.3-040: The device shall be able to measure position with as little external sensors as possible.
The inherent design as mentioned in the previous requirement requires external motion capture or IMU, but this is the only external sensing needed.

Requirements
2.3-050: The net joint reaction forces should stay in acceptable ranges when using the device.
Ideally the joint reaction forces do not change at all (as the primary goal is muscle support, not joint torque reduction). Too low joint reaction forces could also lead to deterioration because of under use. With the current experimentation setup, joint reaction forces cannot be estimated, as further experimentation is needed using force plates to compute the inverse dynamics and from that the static optimization of the MS model.
2.4-010: The device shall provide a stable connection to the shank such that precise force delivery is enabled (5 - 10% error max.)
While the results of the experiment testing the shank bracket stability were promising, with little movement detected, the final error cannot be calculated yet without a controller. The hypothesis is that by calculating the force displacement relationship in an experiment like experiment 1, this movement can be corrected for in the controller. However, the force displacement relationships are not showing a clear trendline as of now, so the final error might be too high.
2.4-020: The shear forces shall be kept within acceptable ranges such that ultimately the shank connection does not slip down during normal operation. This shall be achieved while still allowing for 2.2-020 and thus not with increased normal forces.
With the shear force test conducted in the concept detailing phase, the slip limit was calculated. The design does remain under this limit, so the connection will not slip. It does so while still allowing for 2.2-020, however, inversion is slightly uncomfortable.
2.6-030: The device shall provide an 'area of flexibility' around the joints to enable multi-DoF actuation.
With the choice of the joint, there is full flexibility around the joint.
2.7-020: The device interface with the current power supply and actuation chain in the lab shall be easily engaged and disengaged.
With the test setup currently used the engaging/disengaging is quite simple, but the definitive connection should be designed when the input stage and controller are integrated in the design.
2.7-030: The device shall be able to accurately assist up to 20% of the bodyweight normalized ankle and subtalar torques during normal walking.
Within the range of patients mentioned in 1.1-050, the maximum requirement would be around 80kg (24Nm support required). This is difficult to reach in the current configuration, the limit lies closer to 15% (16Nm) for heavier people.

Figure 31: Requirements weighing.

Compared to the state of the art in Appendix A, Figure 32 can be drawn up.

Name	Reference	Weight [g]	Notes
UT J-LO	[12]	1800	Rigid 1DoF
Witte type A	[13]	835	Rigid 1DoF, but higher torque
Witte type B	[13]	875	Rigid 1DoF, but higher torque
Martijn Grootens	[14]	3580	Rigid 2DoF, but higher torque
Exoboot	[21]	255	Soft 1DoF pneumatic boot, comparable torque
Malcolm	[22]	302	Flexible 1DoF Bowden, slightly higher torque
Collins	[23]	450	Passive 1DoF device
Mooney	[24]	1060	Rigid 1DoF system, higher torques
Thesis device	-	1122	Flexible 2DoF system, low torque

Figure 32: Weight comparison.

The weight is relatively comparable to the lower end of the state of the art, albeit a little higher. Naturally, soft pneumatic systems or passive systems will outperform this system in terms of weight. The developed system can produce much lower torques than the only other 2DoF system in this list (Grootens) but is much lighter and will likely allow for better precise actuation.

5.2 Proof of concept

For the final proof of concept, 2 aspects are of interest, namely functionality and stability. As this design is only the start of further research, there is a limit on what can be achieved. As the foot bracket is connected rigidly to the bike shoe, which is in turn tightly around the foot, the main concern for mechanical stability comes from the connection to the shank. To quantify this stability, a test was devised to look at the motion of the shank bracket with respect to the shank itself under a variety of loads applied. While judging the functionality in absolute terms is as simple as applying the device to a subject and then apply the support force in a certain direction while measuring the motion, this does not prove anything for loaded scenarios. Therefore, an experiment is devised to judge the functionality in terms of difference in muscle activation levels with and without support of the device.

5.2.1 Mechanical stability: shank bracket motion

The primary objective of this experiment is to judge the motion under load, but it also tries to get some insight into how tightly the bracket should be attached to the shank and how comfortable this is perceived. For more details on this experiment, the reader is referred to Appendix C. To properly judge the motion of the bracket under a variety of loads, the bracket position is measured over time using the motion capture setup in the Wearable Robotics Lab; the Qualisys system. As the motion is in theory only in line with the lower leg, one marker on the shank and one on the bracket should be enough. However, three markers are used on both elements to allow for postprocessing if necessary (apply rotations for example). Figure 33a shows a model of a leg with the thigh, shank, and ankle bodies. The joints are indicated with the large black dots. The force is applied on the shank bracket with the foot being firmly planted on the ground.

This force is applied by setting up a simple pulley system and a force sensor is placed in series with the tension cable, see also Figure 33. The pulley system has a force loss of around 4.5%, but this correction factor is already applied in the figures shown below. Of interest is the displacement in the y-axis. From groups of markers (1-3 and 4-6) the centroid position is calculated.

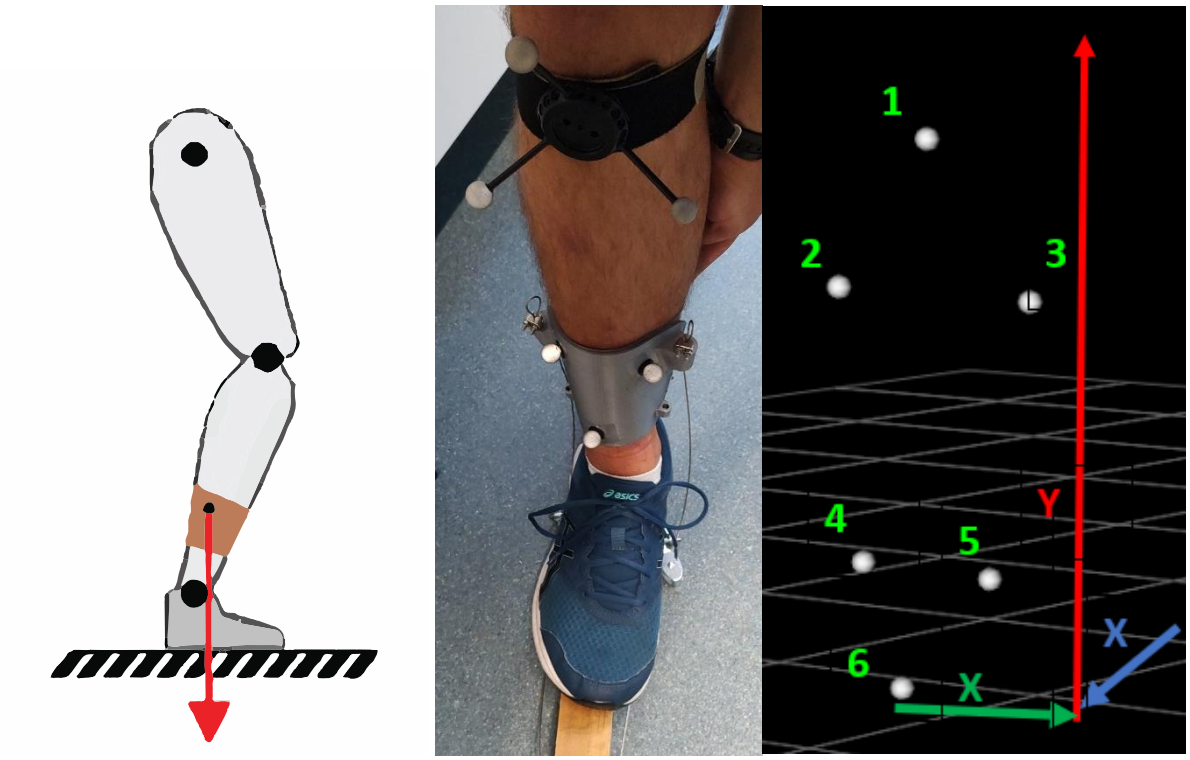


Figure 33: Experiment 1 setup. The model overview can be seen on the left. Three markers are placed on the shank, three on the bracket, as can be seen in the middle picture. The right picture shows the marker numbering and the axes, with the main interest in y.

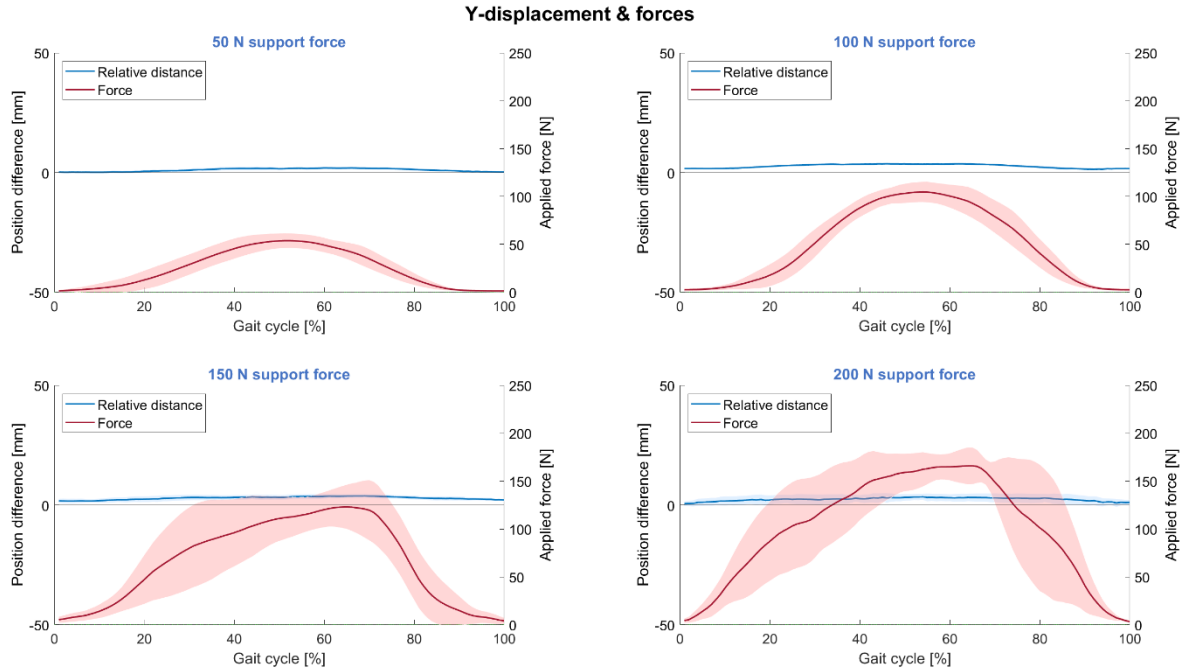


Figure 34: Displacements and force normalized over the gait cycle. The zero line is drawn to better judge the position difference, the line belongs to the left axis. The coloured lines indicate the means over several cycles, the grey band indicates $\pm\sigma$.

Any motion in the x and the z coordinate can be explained by balancing activities. Of particular interest is the motion in the y-axis. For the most part, the y coordinate of the bracket remains stable over time, with one more major jump in the 20kg trial. The bracket seems to remain relatively stable over time, even with larger force applied. If the left y-axis is set to around 100 mm in range (about the size of the bracket), the motions evaluate to Figure 34.

This means that all in all, the shank bracket is connected relatively stiffly to the shank. The remaining motion can be corrected for in control. The fact that the fit and slip characteristics are relatively good does not mean that the fit is also comfortable. The prime goal of this iteration of the bracket was functionality, and comfort was disregarded. However, after a full round of experimentations over the course of several hours, the bracket starts to feel quite uncomfortable, see Figure 35. Correction for this can be done with padding, but this will influence the stability of the bracket, as padding is relatively flexible. Also, as for another experiment, the muscle activation was of interest, there were EMG sensors under the bracket, which also contributed to the discomfort.

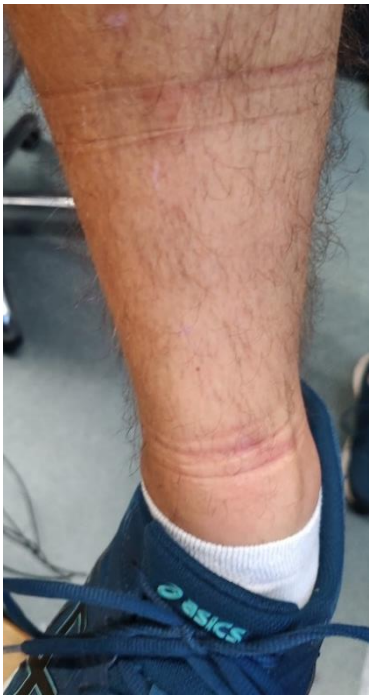


Figure 35: Imprints in leg after several hours of experiments

Subtracting the initial position from the whole array, calculating the absolute values and then respectively the mean and standard deviation gives a good summary of the movement of the bracket. These values can be found in Figure 36 along with the total average and standard deviation values for all trials.

Trial	Mean ± std
50 N	1.0954 ± 0.68748
100 N	2.5658 ± 0.96286
150 N	2.7479 ± 1.0492
200 N	2.1541 ± 1.6022
Total	2.1408 ± 1.0754

Figure 36: Mean & standard deviation values for experiment 1.

5.2.2 Functionality: muscle activation levels

In the same fashion as the previous paragraph, the functionality is assessed, in which the hypothesis is that with a given movement, the agonist muscle's activation will decrease. Also, in general, the antagonist muscle's activation should decrease. The exact hypotheses differ per motion and will be explained per experiment. The setup for this is a full motion capture system to capture the lower body, EMG sensors to capture muscle activations and the orthosis with a force sensor in the input stage. The markers are placed on the anatomical landmarks that are standard to the recording of the lower body, according to Figure 37. For plantarflexion, the muscles of interest are the soleus, and the lateral and medial heads of the gastrocnemius. The other plantarflexors are deeper muscles and can thus not be measured by EMG. For eversion, the peroneus longus and brevis can be measured as they are superficial too. Most invertors are not superficial, so that is more difficult.

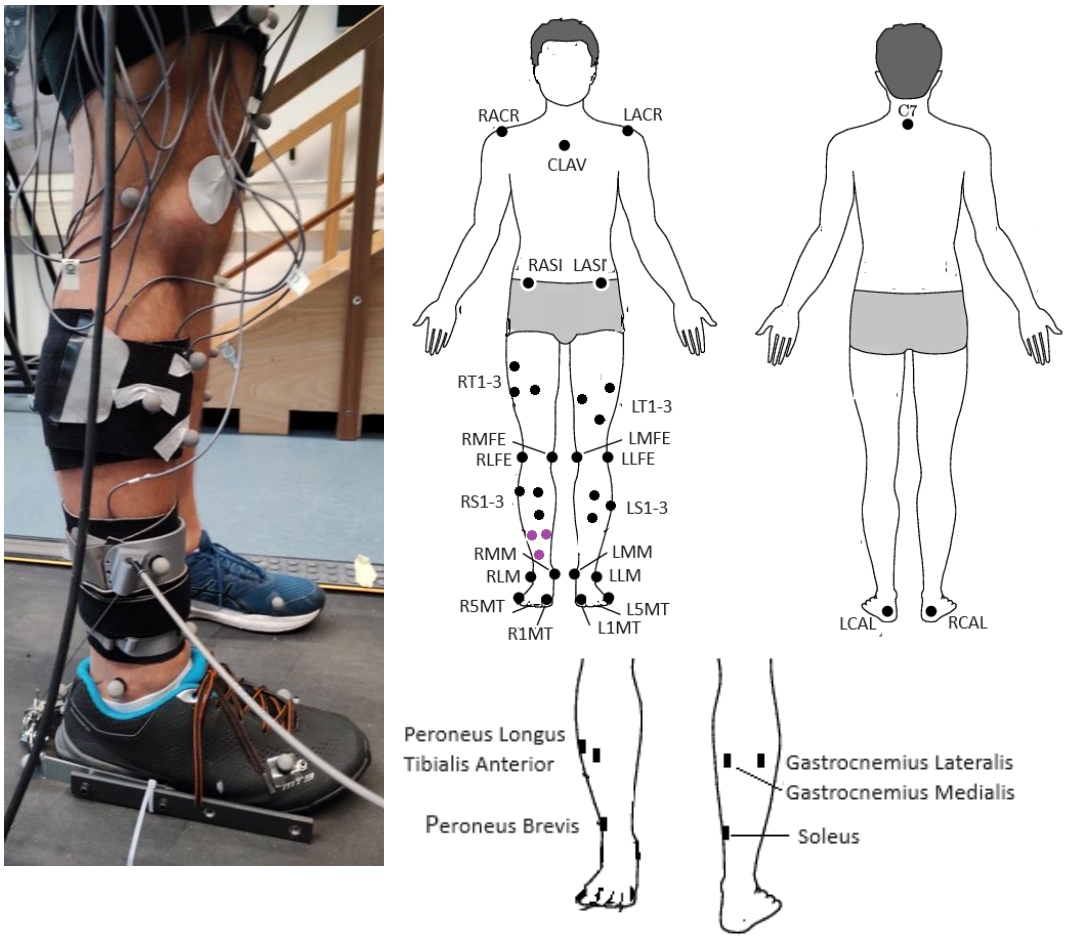


Figure 37: Experiment 3 setup. Left picture the right leg fitted with markers, EMG sensors and the device. Top right the marker placements with the names. Bottom right the EMG placements.

For this experiment, several movements were performed. As the muscle groups are of main interest, the motions discussed here are those that target those groups in an isolated fashion: calf raises for plantarflexion, and balance movements for inversion and eversion. All these motions were performed over several cycles and were gait normalized in the post-processing of the data. Also, the EMG activations were normalized to the maximum voluntary contraction of the muscles. As no actuator is yet generated, the force is generated by a constant weight. This is the simplest solution, but it has the downside that the antagonist muscles must get the weight back up. For the isolated movements, this is no problem, as the increase of antagonist activation with the antagonist motion is just as much a proof as the decrease of agonist activation during the agonist movement. However, the downside for this is that without an actuator, it is not quite possible to deliver 200 N of support yet.

Contrary to the experiment on the shank stability, this one does provide functional movement. One support type (plantarflexion, inversion, or eversion) is applied at a time to look at the individual effects. The general setup can be seen in Figure 38, where the shank and foot parts are depicted in gold, the actuation cable in light red and the sleeve in dark red.

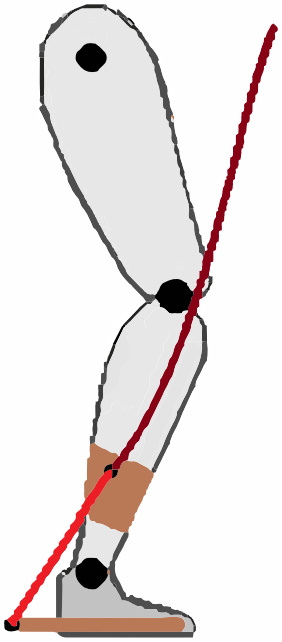


Figure 38: Experiment 3 setup. Shank and foot parts in gold, Bowden cable in light red, Bowden sleeve in dark red.

5.2.2.1 Plantarflexion results

The first motion of interest is plantarflexion, by means of calf raises. The main plantarflexors that are measured here are the soleus and gastrocnemius. However, the peroneus longus and brevis also support plantarflexion, although to a much lesser extent. The tibialis anterior is the only muscle that is measured that does not contribute to plantarflexion. The data from the calf raises is first filtered using in sequence a high-pass filter, a rectification and lastly a low-pass filter. These are performed using a Butterworth filter, using cutoff frequencies 25 Hz and 6 Hz respectively for the high- and low-pass filters. After this, data is segmented at the neutral stance, so the peak of the EMG data is at around 50% of the cycle. The results of this can be seen in Figure 39.

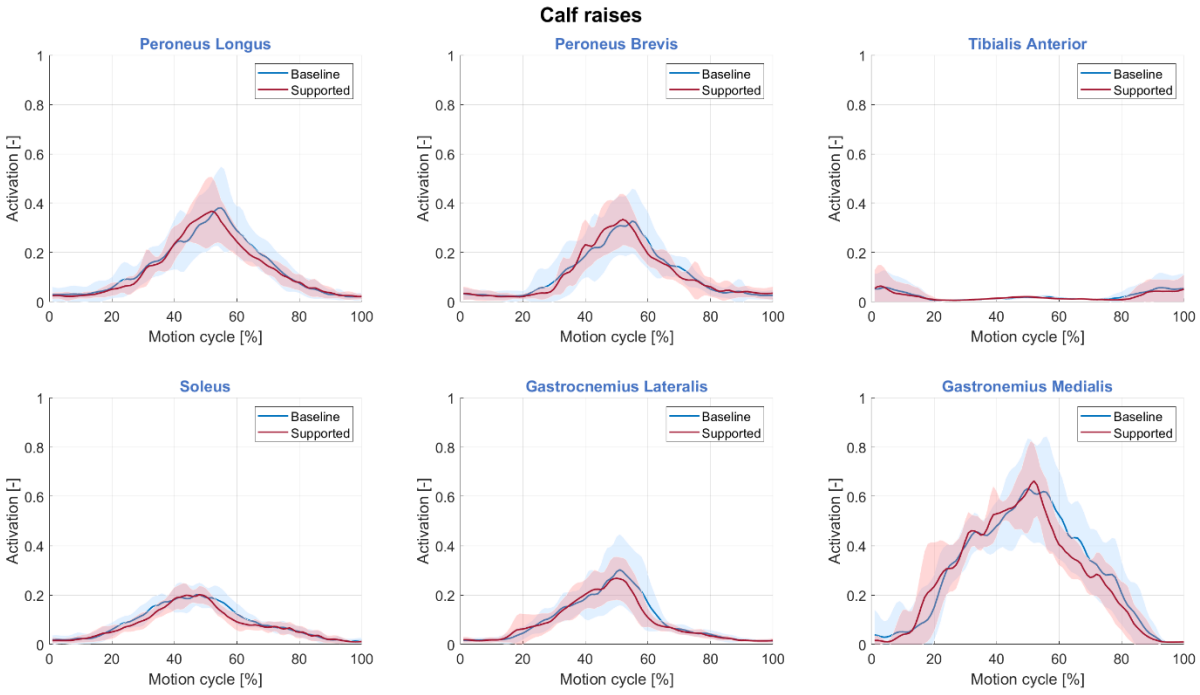


Figure 39: Experimental EMG results for calf raises. The coloured lines indicate the means over several cycles, the shaded area indicates $\pm\sigma$.

The tibialis anterior data can be regarded as noise or at least at irrelevant activation in this experiment as the only time it is activated is during the second half of the cycle, but that motion is in the same direction as gravity and thus there will be almost no activation. As expected, the other muscles do contribute to the motion. The profiles are quite similar with and without support, which seems plausible given that the support in this trial is about 5% of the total torque required for the motion. The expected maximum decrease is thus 5%. The true decrease is difficult to see from the plots, so to quantify this the root mean square of the signals is calculated (after filtering). These results, including the hypotheses per muscle can be found in Figure 40. The first column

after the hypothesis denotes the percentage change of the support trial versus the non-supported trial, the second column is the absolute change with respect to the maximum voluntary contraction. This is also expressed in a percentage as the maximum is 1. In this light, all the muscle activations do perfectly match the hypothesis, however the values are a bit on the lower side of expectations. For these values, the average decrease in plantarflexion activation is 8.7% (1.5% in absolute terms).

PF support			
Muscle	**Hypothesis**	**% with respect to no support**	**% absolute change**
Peroneus longus	Slight decrease	-6.6%	-1.2%
Peroneus brevis	Slight decrease	-5.2%	-0.8%
Tibialis anterior	Same	-7.5%	-0.3%
Soleus	Decrease	-9.5%	-1.0%
Gastrocnemius lateralis	Decrease	-11.2%	-1.6%
Gastrocnemius medialis	Decrease	-5.4%	-2.0%

Figure 40: Root mean squared values of the signals for calf raises.

5.2.2.2 Eversion support results

The second trial that is looked at is eversion support, by means of balancing experiments. The 2 most superficial evertors are the peroneus longus and brevis. However, the gastrocnemius also supports eversion, but to a lesser extent. The soleus and tibialis anterior both slightly contribute to inversion. For eversion support, two experiments were conducted, one leaning outward, and one leaning inward. The most important is the inward leaning, but the outward leaning is performed as a check. The data is filtered in the same fashion as before, using a Butterworth filter, and cutoff frequencies 25 Hz and 1.8 Hz respectively for the high- and low-pass filters. After this, data is segmented at the neutral stance, so the peak of the EMG data is at around 50% of the cycle. Leaning inward and outward are discussed separately below. The data for these trials is not as clear as with the calf raises, which can be caused by several factors, explained per trial. The fact that the peroneus longus and brevis' activation is much higher than the other muscles in the next four figures is because these are dedicated evertors, while the other muscles primarily activate inversion or eversion.

Inward leaning

In Figure 41, an inward leaning leg is shown to indicate the motion made during this trial. The results for leaning inward can be found in Figure 42.

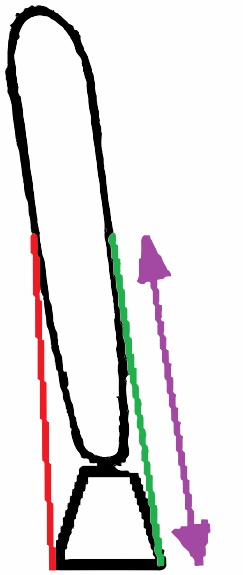


Figure 41: Inward leaning lower leg (posterior view). The red line indicates the invertors, the green line the evertors and the purple arrow the side of support.

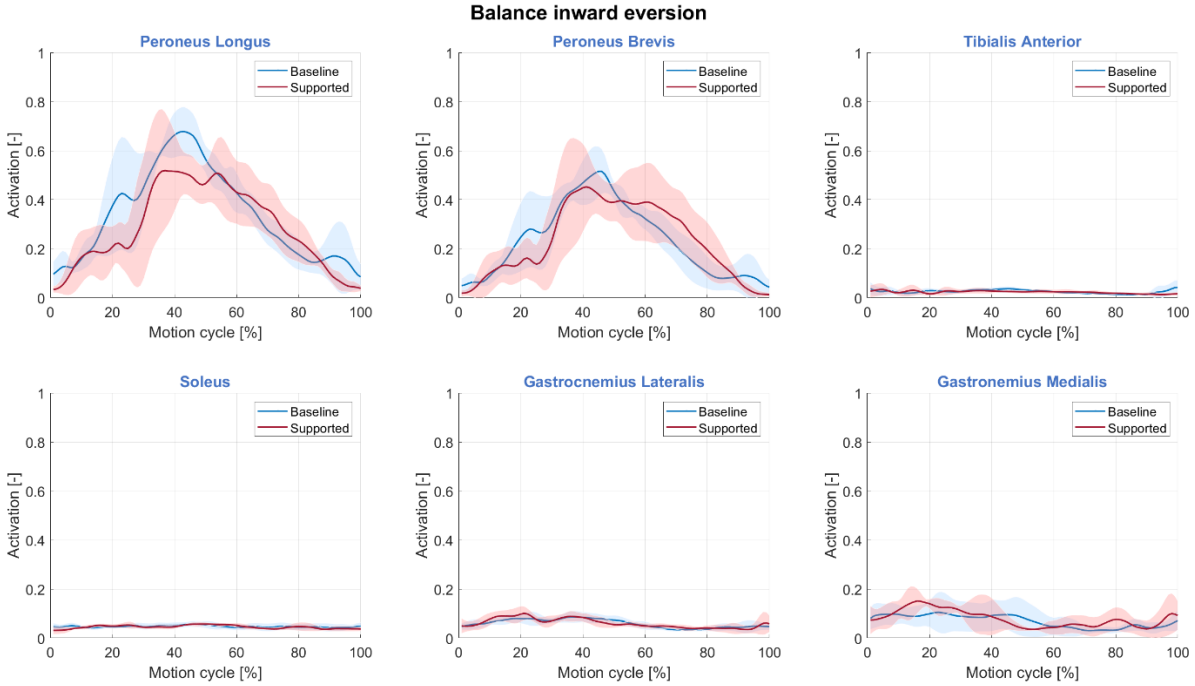


Figure 42: Experimental EMG results for inward leaning with eversion support. The coloured lines indicate the means over several cycles, the shaded area indicates ±σ.

Before the root mean squares are quantified, some behaviour of the plots must be explained.

1) **Behaviour at different segments in the cycle:** At the start of the motion, the invertors must initiate the movement. As there is eversion support, during the first part of the motion, a small decrease in the peroneus longus and brevis are expected. The activation of the gastrocnemii would probably not change notably. The tibialis anterior will increase slightly here, whereas the soleus' activation change would again probably be unnoticeable.
2) **Peak shift:** As there is support against the leaning, the peak in evertor activation is likely shifted to the end of the cycle. This behaviour is clearly observed.
3) **RMS values:** Since the activations do not change uniformly over the cycle, it is difficult to hypothesize the root mean square values over the total movement. The only muscles that should decrease overall are the peroneus longus and brevis.
4) **Stiffening:** A disturbance at either the lateral or medial side of the ankle causes the other side to stiffen up a little to counter. For walking, the body will adapt to this, but for these isolated movements this might not happen, no matter how long the subject is made comfortable with the support.

Looking at the root mean squared of the signals in Figure 43, number 3 is indeed observed. For the other four muscles, the results hover zero, but do not fully match the expectations set out in the hypotheses. This could be the effect of any combination of the list above, and before any final conclusions are drawn, the experiment should be repeated more to check if the results remain the same. If so, the cause of this must be found out through different experiments. However, the most important behaviour is the decrease in peroneus longus and brevis activation, which is clearly observed. For these values, the average decrease in evertor activation is 8.7% (1.5% in absolute terms).

Inward leaning, eversion support			
Muscle	**Hypothesis**	**% with respect to no support**	**% absolute change**
Peroneus longus	Decrease	-16.4%	-6.4%
Peroneus brevis	Decrease	-3.4%	-1.0%
Tibialis anterior	Slight increase	-5.8%	-0.2%
Soleus	Slight increase	-5.8%	-0.3%
Gastrocnemius lateralis	Slight decrease	3.4%	0.2%
Gastrocnemius medialis	Slight decrease	8.8%	0.8%

Figure 43: Root mean squared values of the signals for inward leaning with eversion support.

Figure 44: Outward leaning lower leg (posterior view). The red line indicates the invertors, the green line the evertors and the purple arrow the side of support.

Outward leaning

In Figure 44, an outward leaning leg is shown to indicate the motion made during this trial. The results for leaning inward can be found in Figure 45.

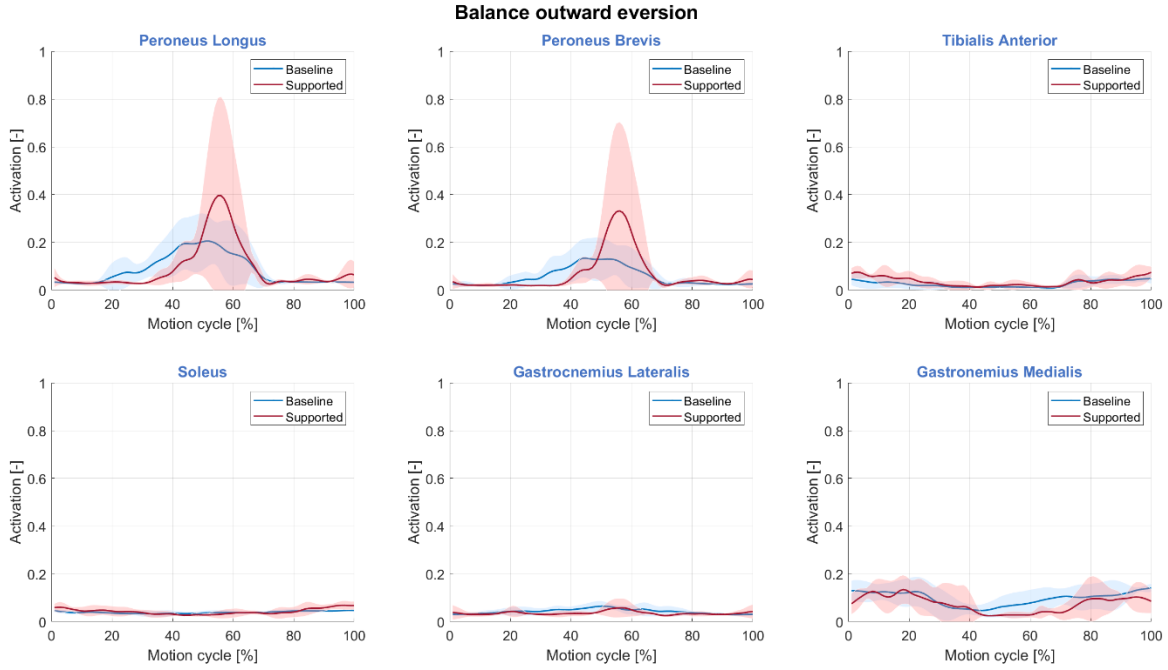


Figure 45: Experimental EMG results for outward leaning with eversion support. The coloured lines indicate the means over several cycles, the shaded area indicates ±σ.

The results for outward leaning are even more difficult to interpret. For leaning outward, the following things are of concern:

1) **Behaviour at different segments in the cycle:** At the start of the motion, the evertors initiate the movement. As there is eversion support, during the first part of the motion, a small decrease in the peroneus longus and brevis is expected, and a slight increase in the tibialis anterior as it must resist the eversion support. The activation of the gastrocnemius is likely less noticeable, but it should be lower at the start and likely remain lower. The soleus' activation will likely not change noticeably. The overall tibialis anterior should decrease if anything. All the phenomena are indeed observed.
2) **Peak shift:** As there is support in the direction of the leaning, the tibialis anterior starts activating earlier to counter the motion.

3) **RMS values:** Since the activations do not change uniformly over the cycle, it is difficult to hypothesize the root mean square values over the total movement. The only muscle that should increase overall is the tibialis anterior and to some extent the soleus.
4) **Stiffening:** This happens in the same manner as leaning inward.

Looking at the root mean squared of the signals in Figure 43, number 3 is indeed observed. The other four muscles do change quite significantly, more than with the other trial. Again, the most important conclusion is that the tibialis anterior and the soleus do increase, for the other behaviours again more experiments must be conducted.

Outward leaning, eversion support			
Muscle	**Hypothesis**	**% with respect to no support**	**% absolute change**
Peroneus longus	Slight decrease	31.5%	3.6%
Peroneus brevis	Slight decrease	64.1%	4.9%
Tibialis anterior	Increase	41.7%	1.4%
Soleus	Slight increase	16.0%	0.6%
Gastrocnemius lateralis	Slight decrease	-17.8%	-0.8%
Gastrocnemius medialis	Slight decrease	-19.1%	-2.1%

Figure 47: Root mean squared values of the signals for outward leaning with eversion support.

5.2.2.3 Inversion support results

The last trial of interest is inversion support, again by balancing experiments. The dedicated invertors are deep muscles, but the most superficial invertor is the tibialis anterior. To a much lesser extent, the soleus also supports inversion. For inversion support, again two experiments were conducted, one leaning outward, and one leaning inward. The most important is the outward leaning, but the inward leaning is performed as a check. The data is filtered in the same fashion as before, using a Butterworth filter, and cutoff frequencies 25 Hz and 1.8 Hz respectively for the high- and low-pass filters. After this, data is segmented at the neutral stance, so the peak of the EMG data is at around 50% of the cycle. Leaning inward and outward are discussed separately below. The data for these trials is not as clear as with the calf raises, which can be caused by several factors. There was quite some variation in the peak height of the muscles. Before the root mean squares are quantified, some behaviour of the plots must be explained.

Outward leaning

In Figure 46, an outward leaning leg is shown to indicate the motion made during this trial. The results for outward leaning can be seen in Figure 48.

Figure 46: Outward leaning lower leg (posterior view). The red line indicates the invertors, the green line the evertors and the purple arrow the side of support.

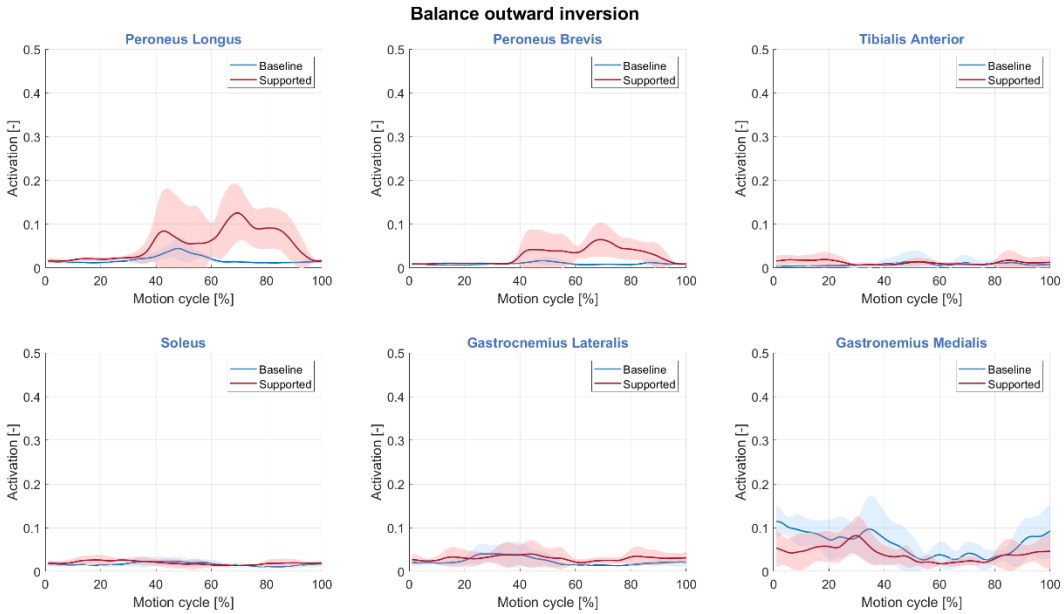


Figure 48: Experimental EMG results for outward leaning with inversion support. The coloured lines indicate the means over several cycles, the shaded area indicates ±σ.

Before the more general root mean square values, again a specific analysis of the behaviour over the cycle is conducted.

1) **Behaviour at different segments in the cycle:** At the start of the motion, the evertors initiate the movement. As there is inversion support, during the first part of the motion, a small increase in activation is expected for the peroneus longus and brevis and to a lesser extent the gastrocnemius. The only abnormality is the medial gastrocnemius in this reasoning. Towards the end, when the subject gets back up, if anything, a small decrease is expected in the tibialis anterior and the soleus. Not all these phenomena are observed.
2) **Peak shift:** As there is support countering the movement, if there is any peak, it is expected to shift a little towards the end of the cycle for the tibialis anterior and the soleus.
3) **RMS values:** Since the activations do not change uniformly over the cycle, it is difficult to hypothesize the root mean square values over the total movement.
4) **Stiffening:** This happens in the same manner as eversion support.

The results on the root mean squares can be seen in Figure 49. The peroneus longus and brevis are as expected. The other muscles again hover zero, but apart from the lateral gastrocnemius, the results tend away from the hypothesis. But, in absolute terms, the values are quite low.

Outward leaning, inversion support			
Muscle	**Hypothesis**	**% with respect to no support**	**% absolute change**
Peroneus longus	Increase	282.1%	5.9%
Peroneus brevis	Increase	287.1%	2.9%
Tibialis anterior	Decrease	17.5%	0.2%
Soleus	Decrease	17.0%	0.3%
Gastrocnemius lateralis	Increase	29.5%	0.8%
Gastrocnemius medialis	Increase	-33.5%	-2.5%

Figure 49: Root mean squared values of the signals for outward leaning with inversion support.

Inward leaning

In Figure 51, an outward leaning leg is shown to indicate the motion made during this trial. The results for outward leaning can be seen in Figure 50.

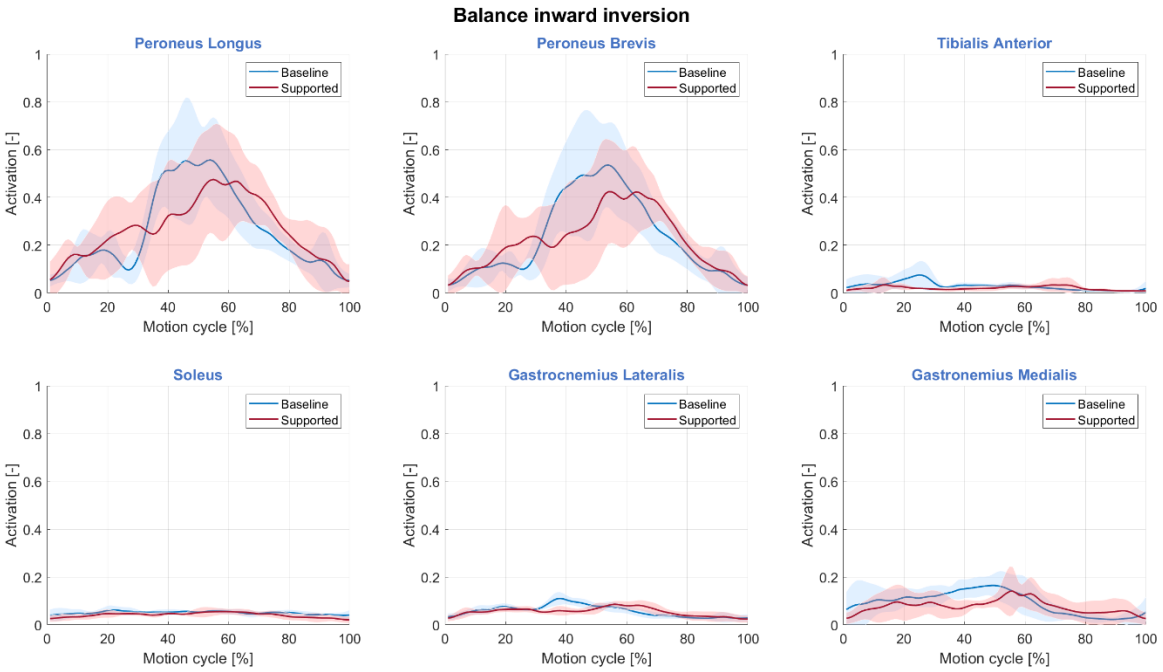


Figure 50: Experimental EMG results for inward leaning with eversion support. The coloured lines indicate the means over several cycles, the shaded area indicates ±σ.

As before, some notions about the plot behaviour include:

1) **Behaviour at different segments in the cycle:** At the start of the motion, the invertors initiate the movement. As there is inversion support, during the first part of the motion, a small decrease in the tibialis anterior is expected, and a slight increase in the peroneus longus and brevis as they must resist the eversion support. Their activation should also be a bit higher towards the end of the motion as they must resist the inversion support. The activation of the soleus is likely less noticeable, but it should be slightly lower at the start and likely remain lower. The gastrocnemius activation will likely be higher at the start and the end as that is when the motion is resisted most. All the phenomena are indeed observed, except for the gastrocnemius, which is only increased during the last part of the motion, at the beginning of the motion, a decrease is observed, which cannot be explained by this.
2) **Peak shift:** As there is support in the same direction as the leaning, the peroneus longus and brevis peaks would move a little forward to counter the support. A higher activation at the start is indeed observed, but as the maximum value is lower, the peak is not shifted.
3) **RMS values:** Since the activations do not change uniformly over the cycle, it is difficult to hypothesize the root mean square values over the total movement. A probable estimate is that the peroneus longus and brevis, and the gastrocnemius will increase slightly, and that the tibialis anterior and the soleus will decrease slightly.
4) **Stiffening:** This happens in the same manner as eversion support.

Figure 51: Inward leaning lower leg (posterior view). The red line indicates the invertors, the green line the evertors and the purple arrow the side of support.

The results of the root mean square values can be seen in Figure 52. The tibialis anterior and soleus behave as expected; the gastrocnemius hovers zero but does lean against the hypothesis. The fact that the peroneus longus and brevis do not increase has no explanation at this point. The decrease that is observed is too significant to be caused by the factors above.

Inward leaning, inversion support			
Muscle	**Hypothesis**	**% with respect to no support**	**% absolute change**
Peroneus longus	Increase	-7.5%	-2.5%
Peroneus brevis	Increase	-9.5%	-2.8%
Tibialis anterior	Decrease	-31.1%	-1.1%
Soleus	Decrease	-20.8%	-1.1%
Gastrocnemius lateralis	Increase	-10.2%	-0.6%
Gastrocnemius medialis	Increase	-18.5%	-2.0%

Figure 52: Root mean square values of the signals for inward leaning with inversion support.

5.2.2.4 Averages over trials and applied torque

For the final conclusions of the support, the difference between supported and non-supported is of interest. The percentage difference with respect to the maximum voluntary contraction is largely irrelevant as the motions do not necessarily need maximum voluntary contraction. The values for the primary motions of interest are summarised in Figure 53. If multiple muscles activate the motion, the value is the average of the muscles.

Trial	Muscle activation change
Plantarflexion (soleus, gastrocnemius)	- 8.7%
Eversion (peroneus longus, brevis)	- 9.9%
Inversion (tibialis anterior)	+ 17.5%

Figure 53: Average values of EMG activation decrease.

With the device setup used in the experiments, about 10Nm of peak plantarflexion torque, 6Nm inversion, and 6Nm eversion is applied. Increasing the moment arm (which can be done comfortably till about an extra 30mm) leads to a maximum of about 16Nm plantarflexion torque applied with the current setup. While the moment arm could in theory be extended more, the device would become bulkier.

6 Discussion & recommendations

It has already been shown that the device holds up against most of the requirements. On a more general level, it can also be said that most of the design goals from Chapter 2.2 are met. The primary goal was to develop a novel type of 2DoF ankle-foot orthosis and see the extent to which this type of device is feasible. The final tests show a reduction in muscle activation for the plantarflexion motion up to 8.7% and for inversion up to 9.9%. However, the activation of the tibialis anterior (evertor) increases (17%) rather than decreases. This is not expected and with the current state, more research is needed to check whether this was an anomaly in recording or if it is structural. It seems highly unlikely that it is structural, as simple mechanics disprove this. As the behaviour of the EMG signals over the gait cycle also is not entirely as expected, the hypothesis that this is an anomaly indeed seems substantiated. The specific behaviour for eversion and plantarflexion support is entirely as expected. A repetition of the experiment with more subjects seems in place, perhaps finding a way to look at more invertor muscles besides the tibialis anterior.

So, while the device in principle is functional, there are some clear paths to improve on, see the next paragraph. With the device setup used in the experiments, about 10Nm of plantarflexion torque, 6Nm inversion, and 6Nm eversion. Increasing the moment arm (which can be done comfortably to a total of about 80mm) leads to a maximum of about 16Nm plantarflexion torque applied with the current setup. After this comes the goal to venture into fine force delivery. As the device can actuate the three motions set out in the goals, precisely modulating the amount of force applied can target specific muscle groups. Targeting specific muscles within such a group is difficult, but any external device will target groups and not specific muscles. The goal of developing insight in the joint reactions of an ankle-foot orthosis is not attained to the extent set out from the start. While an effect can be calculated and the design can be adapted accordingly, the goal was to do this iteratively using the OpenSim's toolboxes, which would allow a more personalised design. This can therefore still be improved on. After an actuator has been developed and some final tunings in the design, the device can be used to start testing the use cases as set out in paragraph 2.5.

This is not the only improvement, as a wide variety of elements can still be optimised on the design. The amount of work that can be done in one thesis is limited and as such several recommendations are made. From the very start, in the literature review, it already became clear that there is little research in the longer-term effects of orthosis wear. In preparing the device for further research, this could be one of the elements that is measured. The force effects over longer periods and the medical effects of this are of special interest here. In this context the device can also be used to research the use effects on (a) specific muscle (groups). With measuring the muscle activations during use similarly to the experiments conducted above, the changes in muscle activation with and without the device can be tracked. While part of the thesis was the research into force effects, this was only done of patient data from a theoretical level. Ideally, such effects are computed real-time and so the control can be adjusted accordingly. Also, a practical point of attention is that with multiple lines of actuations comes a risk of entanglement. It might be useful to determine if one of the lines of action could be removed or whether it is possible to make sure entanglement cannot occur in any way. Finally, while the device is comfortable enough to be worn during the tests, wearing it for extended periods of time requires additional research into shank connection and the comfort of the bike shoe as the rigidity does limit natural rolling of the foot. Also, it would be useful to be able to provide about 40% of the normal required torque during walking, as that would enable the device to be used for running and jumping without slip. This should most likely happen through another iteration of the shank bracket, transferring the force at a larger area or other parts of the body to the musculoskeletal system. In the design process, the minimum required moment arm (30mm extension from the calcaneus) was chosen. Redesigning the foot bracket could give more torque delivery at the cost of a bulkier design. Once the controller is designed for the device, more control trials should be conducted to gather more data on the reduction in muscle activation.

7 Conclusions

In this thesis, the first developments on a novel type of 2DoF ankle-foot orthosis were made. The thesis demonstrates the feasibility of precise force delivery of muscle groups, though not individual muscles, with this orthosis design. This thesis has added to the research a lightweight and mechanically relatively simple orthosis which can be used in rehabilitation, further research or can even be adapted to be used over longer periods of time.

However, it is important to note some limitations. The orthosis, while promising, may not offer the level of comfort required for longer periods of use. Additionally, with the current setup, only up to 20% of the gait torque can be delivered during normal walking. To accommodate higher torques demands, further iterations of the shank bracket design or a longer posterior extension of the foot bracket's moment arm are necessary.

In conclusion, this thesis has explored the design and capabilities of a Bowden cable-actuated 2DoF ankle-foot orthosis, opening doors for further research and development in the field.

References

[1] WHO. "Spinal Cord Injury." https://www.who.int/news-room/fact-sheets/detail/spinal-cord-injury (accessed 13/02/2023.

[2] B. S. Armour, E. A. Courtney-Long, M. H. Fox, H. Fredine, and A. Cahill, "Prevalence and Causes of Paralysis-United States, 2013," (in eng), *Am J Public Health,* vol. 106, no. 10, pp. 1855-7, Oct 2016, doi: 10.2105/ajph.2016.303270.

[3] E.-E. Adrian, C. Jose, and I. C.-V. Antonio, "Load progression criteria in exercise programmes in lower limb tendinopathy: a systematic review," *BMJ Open,* vol. 10, no. 11, p. e041433, 2020, doi: 10.1136/bmjopen-2020-041433.

[4] CDC. "Stroke facts." https://www.cdc.gov/stroke/facts.htm (accessed 13/02/2023.

[5] D. D. D'Lima, B. J. Fregly, S. Patil, N. Steklov, and C. W. Colwell, Jr., "Knee joint forces: prediction, measurement, and significance," (in eng), *Proc Inst Mech Eng H,* vol. 226, no. 2, pp. 95-102, Feb 2012, doi: 10.1177/0954411911433372.

[6] P. Sadoghi, G. Roush, N. Kastner, A. Leithner, C. Sommitsch, and T. Goswami. *1-s2.0-S1751616114003890-main*.

[7] Y. Zhu, X. Li, and X. Xu, "Ankle joint pressure change in varus malalignment of the tibia," (in eng), *BMC Musculoskelet Disord,* vol. 21, no. 1, p. 148, Mar 4 2020, doi: 10.1186/s12891-020-3163-2.

[8] J. A. Bolduc, J. A. Collins, and R. F. Loeser, "Reactive oxygen species, aging and articular cartilage homeostasis," *Free Radical Biology and Medicine,* vol. 132, pp. 73-82, 2019/02/20/ 2019, doi: https://doi.org/10.1016/j.freeradbiomed.2018.08.038.

[9] Teach me anatomy. "The ankle joint." https://teachmeanatomy.info/lower-limb/joints/ankle-joint/ (accessed.

[10] F. Harrold and R. J. Abboud. "Chapter 2 - Biomechanics of the Foot and Ankle." https://musculoskeletalkey.com/chapter-2-biomechanics-of-the-foot-and-ankle/ (accessed 04/09/2023.

[11] W. Pirker and R. Katzenschlager, "Gait disorders in adults and the elderly," *Wiener klinische Wochenschrift,* vol. 129, no. 3, pp. 81-95, 2017/02/01 2017, doi: 10.1007/s00508-016-1096-4.

[12] J. Meijners, "The design, control and integration of a bowden cable actuation for an ankle exoskeleton," Master of Science, Biomechanical Engineering, University of Twente, 2022.

[13] K. A. Witte, J. Zhang, R. W. Jackson, and S. H. Collins, "Design of two lightweight, high-bandwidth torque-controlled ankle exoskeletons," in *2015 IEEE International Conference on Robotics and Automation (ICRA)*, 26-30 May 2015 2015, pp. 1223-1228, doi: 10.1109/ICRA.2015.7139347.

[14] M. E. Grootens, "Development of a 2-DoF Ankle Exoskeleton," EngD Thesis, University of Twente, Enschede, 2019.

[15] J. A. Blaya and H. Herr, "Adaptive control of a variable-impedance ankle-foot orthosis to assist drop-foot gait," (in eng), *IEEE Trans Neural Syst Rehabil Eng,* vol. 12, no. 1, pp. 24-31, Mar 2004, doi: 10.1109/tnsre.2003.823266.

[16] C. Meijneke, W. van Dijk, and H. van der Kooij, "Achilles: An autonomous lightweight ankle exoskeleton to provide push-off power," in *5th IEEE RAS & EMBS Interantional Conference on Biomedical Robotics and Biomechatronics, BioRob 2014*, ed. Sao Paulo, Brazil: IEEE, 2014, pp. 918-923.

[17] H. Wang, G. Durandau, and M. Sartori. "Comprehensive Kinetic and EMG Dataset of Daily Locomotion with 6 types of Sensor (1.0)." https://doi.org/10.5281/zenodo.6457662 (accessed.

[18] EurocodeApplied. "Table of design properties for metric hexagonal bolts M5 to M39 (stress area, shear strength, tensile strength, bearing strength)." https://eurocodeapplied.com/design/en1993/bolt-design-properties (accessed.

[19] G. İrsel, "Design of a precision planter chassis using computer-aided engineering and experimental validation," 03/22 2023, doi: 10.1177/09544062231159008.

[20] L. Safai, J. S. Cuellar, G. Smit, and A. A. Zadpoor, "A review of the fatigue behavior of 3D printed polymers," *Additive Manufacturing,* vol. 28, pp. 87-97, 2019/08/01/ 2019, doi: https://doi.org/10.1016/j.addma.2019.03.023.

[21] J. Chung, R. Heimgartner, C. T. O. Neill, N. S. Phipps, and C. J. Walsh, "ExoBoot, a Soft Inflatable Robotic Boot to Assist Ankle During Walking: Design, Characterization and Preliminary Tests," in *2018 7th IEEE*

References

International Conference on Biomedical Robotics and Biomechatronics (Biorob), 26-29 Aug. 2018 2018, pp. 509-516, doi: 10.1109/BIOROB.2018.8487903.

[22] P. Malcolm *et al.*, "Varying negative work assistance at the ankle with a soft exosuit during loaded walking," *Journal of NeuroEngineering and Rehabilitation,* vol. 14, no. 1, p. 62, 2017/06/26 2017, doi: 10.1186/s12984-017-0267-5.

[23] S. H. Collins, M. B. Wiggin, and G. S. Sawicki, "Reducing the energy cost of human walking using an unpowered exoskeleton," *Nature,* vol. 522, no. 7555, pp. 212-215, 2015/06/01 2015, doi: 10.1038/nature14288.

[24] L. M. Mooney, E. J. Rouse, and H. M. Herr, "Autonomous exoskeleton reduces metabolic cost of human walking," *Journal of NeuroEngineering and Rehabilitation,* vol. 11, no. 1, p. 151, 2014/11/03 2014, doi: 10.1186/1743-0003-11-151.

Appendix A:

Literature review

Developments in 2 degree of freedom fine force delivery by an assistive ankle orthosis

Bart Hendriksen, s2424576
Supervisor: Dr. G. V. Durandau
Draft date: September 12, 2022
Version: 2A

Table of Contents

TABLE OF CONTENTS **2**

INTRODUCTION **3**

FINDINGS **4**

PART I: EXOSUITS 4

History *4*

Applications *5*

Actuation types *6*

Materials *7*

Control *7*

State of the art *10*

PART II: TARGETED REHABILITATION 12

Biomechanics of the ankle *12*

Ankle anatomy & human gait *13*

Possible impairments and which to tackle *17*

Strategies for rehabilitation *19*

Effects on joints of (long-term) mechanical support *20*

APPLICATIONS & GAPS **21**

RESEARCH GOALS **22**

REFERENCES **23**

Introduction

In the context of developing a novel 2D assistive ankle orthosis to enable finer force delivery, a literature review has been conducted with the aim of finding the state of the art in ankle foot-orthoses, or one of several other names these devices go by. Also, a brief inquiry has been done into targeted rehabilitation, with the goal of setting out the biomechanics of the ankle during walking and the effects of an orthosis. From this, applications, a gap, and a research goal is formulated. This document contains the summary of the results of the literature review. The relevant sources have been gathered in an EndNote Library, this document provides the overview and determines applications for and gaps in the current state of the art.

This document has the following outline. It starts with chapter 2 on search methodology, followed by chapter 3 with the findings of said search. Chapter 4 sets out the applications and gaps, thus drawing a conclusion on the findings.

Findings

Chapter two is divided in two main parts, one revolving around exosuits and the other about targeted rehabilitation. Both these concepts are elemental to the state-of-the-art surrounding fine force delivery.

Part I: Exosuits

The first part of the findings will deal with exosuits. An exosuit is fundamentally different from an exoskeleton. For the rest of this literature review, an exoskeleton is defined as some type of mechanical structures designed to be worn by a human being for either assisting or enhancing limb movement. The exoskeleton can be passive or actuated by a variety of actuation chains. An exosuit is defined here as an exoskeleton made from soft, flexible materials. Exosuits are generally good for enhancing strength of the body, while a (rigid) exoskeleton is used in rehabilitation contexts, as it can stabilize impaired body parts.

The gravity of this review tends towards orthoses, which are essentially a subset of exosuits or exoskeletons, preventing movement or providing assistance in those movements to specific limbs or the spine. An orthosis is always extracorporeal and is placed over existing limbs, the latter in contrast with a prosthesis.

History

The history of the exosuit is quite a long one. It started quite elegantly with passive devices, then became bulky as the devices got actuated, and then slowly decreased in size again when actuations became more capable with smaller sizes. Recently, the switch from exoskeletons to exosuits was made, further decreasing weight and size. The timeline below, adapted from Kumar, Hote & Jain [1], shows some milestones characterising these four stages.

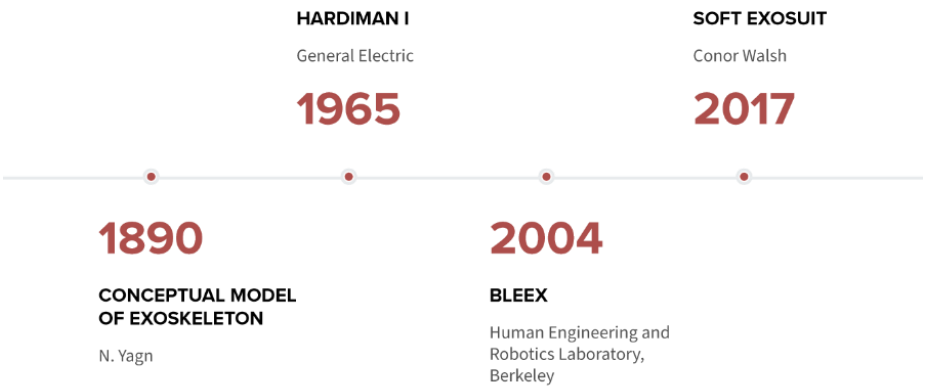


Figure 1: Timeline with milestones on exosuit history

The earliest conceptual model of an exoskeleton was developed by Nicholas Yagn in 1890 [2]. The passive device enhanced daily movements and laid the foundation for later for the Hardiman I, a device built by GE, augmenting human strength and endurance [3]. Later on, a fully functional device was developed by the Human Engineering and Robotics Laboratory at Berkeley, capable of supporting walking speeds carrying 75kg of payloads [4]. More recently, Conor Walsh and his team made developments in fully flexible exosuits, reducing metabolic cost [5]. This could not have been possible without major improvement in the controllers. In the last five to seven years great improvements have been made in first passive metabolic reduction, but then in active metabolic reduction [6].

Applications

Exosuits and exoskeletons have multiple applications. Below, three of them are mentioned, however one can see that some applications are more suited for exosuits and some for exoskeletons.

Rehabilitation

The application having the most common ground with the aim of this literature review is of course rehabilitation. Assisting people with impairments with orthoses or exoskeletons in their rehabilitation process might enable them to get on their feet quicker, as the device can enable them to walk more naturally when they can't yet bear full load. Where for example crutches lead to an abnormal gait, the use of external support that works over natural lines of action doesn't have that problem. Significant results can be achieved using an exosuit with plantarflexion and dorsiflexion support in post-stroke rehabilitation [7]. Even more so, a meta-study described that using an active exosuit, a more symmetrical body movement combined with reduced metabolic energy was detected in patients. In all the studies, passive exosuits also caused improvements in spatiotemporal parameters including velocity, cadence and stride length [8].

These exosuits are on the body, however that does not have to be the case. For example, the Lopes Exoskeleton Robot developed at the University of Twente was developed as a gait rehabilitation device. The device supports both a 'patient in charge' and 'device in charge' mode, which makes it highly adaptable to the situation of a specific patient [9].

Military

Militaries across the world are continuously trying to improve their combat forces. More enabled dismounted combatants are naturally an ambition here, both for fighting and humanitarian aid. As Mudie et al. [10] propose in their consensus paper on testing and evaluation of military exoskeletons for the dismounted combatant, military exoskeletons can be judged based on the five NATO key capability areas: *"mobility, lethality, survivability, sustainability and C4I (Command, Control, Communications, Computers and Intelligence)"*. Mobility and sustainability are two very important measures that can one-on-one be translated to a rehabilitation context.

Exoskeletons are clearly here to stay in a military context [11], however the main drawback is that they cannot be adapted to situations and people. The controller can adapt most easily and needs to be able to do that fast. Other adaptations will have to occur in the way sensors operate, in the way machine intelligence is used, but also in the way the exoskeleton is physically build. One needs modular adaptations to allow for multiple tasks, users and environments [12], as was also seen in a rehabilitation context.

Assistive exoskeletons

Humans have become more conscious about health-hazards related to heavy lifting, unnatural poses, repetitive tasks or a combination of those three, as many countries have had laws regarding this since the past decades. Sectors in which this is of grave importance include healthcare & construction. Workers in both of these sectors (and other sectors too) can benefit from supportive exoskeletons.

Unhealthy poses can be prevented by wearing an exoskeleton or exosuit, as is proven by Cho et. al. [13], where the researchers used an exoskeleton (or exosuit by the definitions of this literature review) to prevent workers to work in unhealthy bodily angles, and thus guiding them to corrections in their pose.

Exoskeleton support is not just limited to assistive lifting, but also to provide motion support in patients who are past rehabilitation as previously mentioned. These face the same challenges as are faced in rehabilitation contexts: light weight, compactness and comfort [14].

Actuation types

Several types of exosuits have been discussed already in the previous chapters. This chapter will deal with the different types of actuation that can be used with those different exosuits. An actuation system will likely consist of some combination of an energy carrier providing energy to an actuator which in turn drives a transmission. The transmission is directly or indirectly connected to the driven part and/or the human. This transmission not only transmits power, but can also act as a shock absorber on the side of either system or human. Actuation implies an active drive, but also passive exosuits are discussed below.

Passive

Exosuits can be passive, in which the system utilizes mostly springs to store gravitational energy through the stance phase in gait and release that during the swing phase. An application of this includes the recent work of Zhou et. al. on reducing interaction forces between the human body and the exoskeleton (thus effectively reducing plantar pressure by 14.7%) [15].

Active

There are many types of active support exosuits, but all use the general train described in the introduction of this paragraph and all add power to the coupled human-machine system. Human-powered exosuits are disregarded here, as those are classified under passive exosuits or are using human power to power electric devices [16].

Hydraulic actuations are mostly used on exoskeletons with load bearing applications [17]. General parameters of interest are weight and power density of the system. Hydraulics' heavy components make it difficult to apply on an exosuit, however for some orthoses there might be applications, which was why the original design of the device by J. Meijners in the current design of the Bowden cable ankle orthosis at the University of Twente [18] was based on hydraulic actuation. Linear electric actuators would work the same as hydraulic actuations, as they can very well replace cylinder based actuation systems.

Pneumatic actuations are used on (bio-inspired) exosuits and prosthetics, and are most often seen in soft exosuits [19]. Pneumatic (linear) actuators on exoskeletons could work, but are less powerful than hydraulic actuators. They are however a better choice when sterility is a measure in the design as they don't require an oil as pressure carrier.

Rotary electric actuation most often happens via a DC-motor with a transmission to get the correct torques and speeds on the joints. Mechanically this is relatively easy solution, at least on an exoskeleton, and it was for example applied on the earlier mentioned BLEEX [20]. However, rotary actuations have the disadvantage of being bulky on the side of the joints. This can be reduced with a transmission, but there is still considerable volume on the side of the limb.

Of special interest is cable actuation. A contracting cable resembles the workings of muscle very closely. A cable can only give tension in one direction, and thus always needs an agonist-antagonist pair to function on a two-way joint. A good example of this is the device by J. Meijners in the current design of the Bowden cable ankle orthosis at the University of Twente [18]. This device makes use of a series elastic element (SEE), which functions as a shock absorber, but also provides more stable and accurate force control and a lower reflected inertia. It does so by turning a force control problem into a position control problem, greatly improving the accuracy [21].

Magnetorheological (MR) fluids are also applied in these fields. The MR actuator usually consists of a DC motor and a MR clutch [22]. There is still a lot of research being done to get MR actuators to work and in the last two decades MR dampers, MR brakes and MR clutches have been widely applied [23]. A MR clutch changes direction, speed and other parameters under a changing magnetic field, while the DC motor that drives the system runs continuously. The main downsides appear in clumping of the particles, hard cake and particles oxidization [24]. Also, with the DC motor continuously running, heat becomes a serious issue to consider.

Materials

Rigid exoskeletons are necessarily made from rigid materials which are mostly metals and carbon fibres. Exosuits that are a little more compliant often still contain those components, while fully compliant exosuits will have less and less stiff components.

Material selection in general does not only take the rigidity into account, but must also deal with weight considerations, wearing comfort, thermal comfort and risk that are induced by (sharp) parts. With increasing inertia, one must also pay increasingly more attention to control as blunt trauma is a real risk with sudden unexpected movements.

Textiles

Teams like those at the Harvard Biodesign Lab are exploring opportunities in textiles. This has two main advantages over rigid structures: the worn part of the suit is extremely light and the wearer's joints are not constrained by any external rigid structures [25] as mentioned before.

Fully compliant or at least fully textile exosuits have the problem that as there is no internal stiffness in the system, the full force generated for actuation must be transmitted on the endoskeleton via the skin. The force transmitted is therefore limited by the friction that the skin-device interface can generate.

Control

Control of exoskeletons is arguably one of the most difficult parts of building and operating an exoskeleton. Patients can be unpredictable, human-machine interfaces (HMI) on the sensory side are easily disturbed and as the HMI on the actuation side must be stable enough to ensure no damage can be done by unexpected movements of the machine. These aspects are all discussed below, with the focus being on the sensory side.

Control paradigm

There are several high-level control types, which can for exosuits generally be summarized as model-based control versus human-in-the-loop (HITL) control. HITL uses for example respiratory data to optimize metabolic reduction given a control law, which is computationally intensive, but can definitely reduce the metabolic cost of walking [26]. A model based controller uses a dynamic (virtual) model of the (coupled) system to compensate for its own dynamics, after which support is provided [27].

This review is carried out in the context of a master's thesis on an ankle orthosis. For this ankle orthosis the high-level controller is already known to be a neuromusculoskeletal (NMS) model. Rather than starting to think about designing a (high- and low-level) controller once the design is relatively fixed, a design must be fitted to the high-level controller.

General control schemes

On an abstract level, exoskeleton control is simple. One uses input signal from the patient's body, processes those, and uses those signals to drive an actuator, which is essentially high-level control. One of the simplest ways to achieve this is using electromyographic sensors on the muscle whose line of action that must be enhanced and using only a gain applying that signal on an output. Of course, one does need some sort of more complex control scheme, for example like the architecture including low-level control that was used by J. Meijners in the current design of the Bowden cable ankle orthosis at the University of Twente [18], seen in Figure 2.

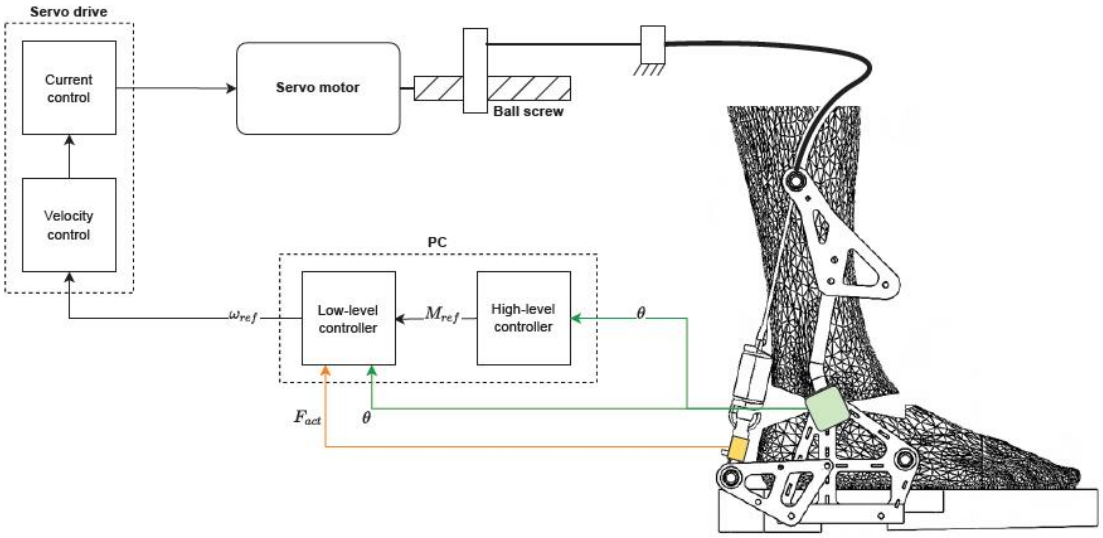


Figure 2: Control scheme of an ankle orthosis

Systems like these act as Artificial Movement Control Systems (AMCS). These have to work parallel to Human Movement Control Systems (HMCS) according to the schematic displayed in Figure 3 [28].

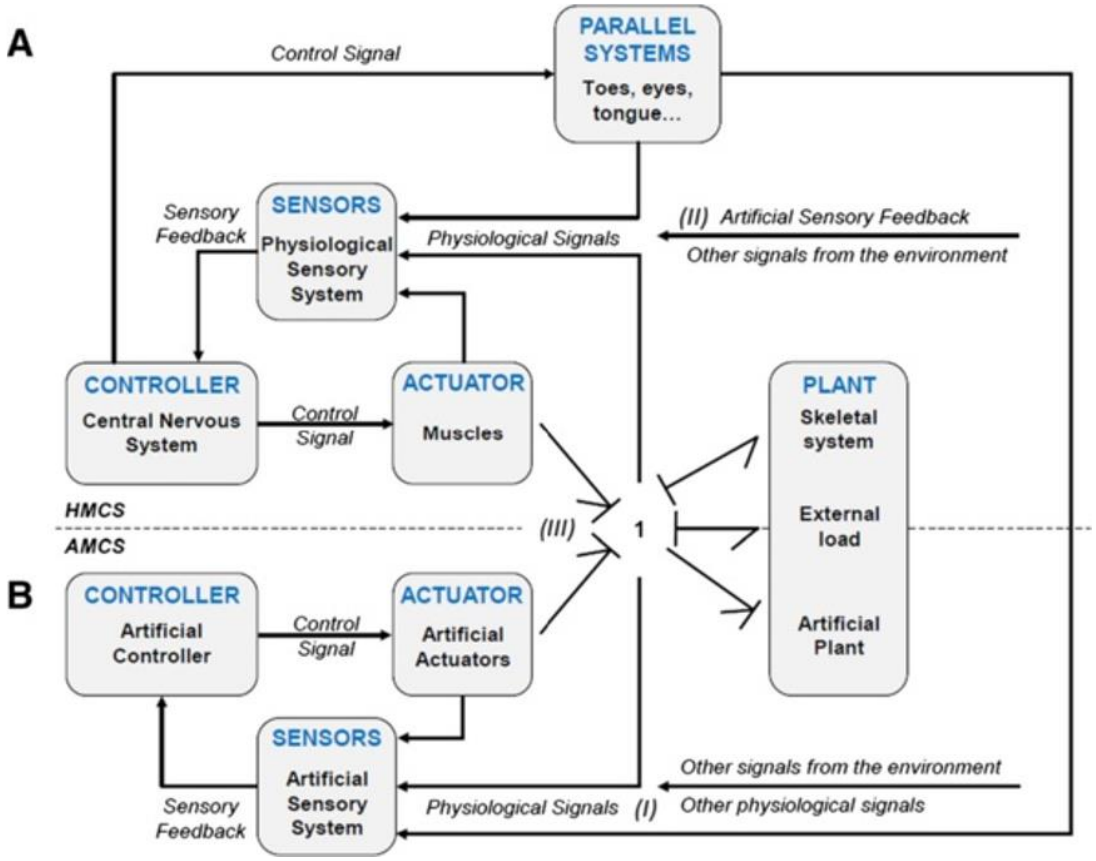


Figure 3: Schematic overview of biomechatronic interaction diagram

HMI sensory side

One of the aspects that makes control so challenging is getting a good, undisturbed signal from the human body. On a healthy subject this poses a variety of challenges already, let alone on impaired patients. The gold standard in this is using non-invasive electromyography (EMG), which is relatively simple to use. The downside is that as surface muscles are measured one can never fully map muscle activation over a cross-section of the limb, but this is corrected in the controller. Also, EMG signals are relatively prone to disturbances, like electromagnetic radiation, and have quite low accuracy for distal extremities [29], which is especially a major downside in prosthetics, but it also affects orthoses. As Suberbiola et. al. [30] mention, one can use multisensory integration to tackle this. Using tactile pressure sensors to create a hybrid control system they achieved better controllability of the system when EMG inputs are low.

Rather than relying on the signals a muscle sends, one can also look at the actual muscle contraction lengths and velocities. As an example of this mechanomyographic (MMG) signals can be used, which represent changes in muscle volume rather than electric signals in those muscles. An MMG signal is produced by a combination of an infrared (IR) sensor and force sensitive resistor (FSR) as is explained by Marques et. al. [29]. They showed that EMG sensors were slightly better at detecting gestures, however they were noisier than MMG sensors.

A challenge arises when patients have either lost part of the limb, the whole of the limb or have lost control of that limb. The problem is the same in any of those cases, as one can't rely on EMG or MMG anymore. One can also interface with the brain. For that, several teams have in the past tried to make electroencephalography (EEG) work, but that significantly underperforms EMG [31]. The EEG signal is extracted directly from the brain via metals electrodes on the skull. An alternative to this would be using brain haemodynamics, but that is still very much in early research stages [32].

So far, all these methods mentioned are non-invasive. Implanting small magnetic beads in the muscles and using their movement to control an exosuit or prosthetic could increase input accuracy, even for people who have lost their limbs [33]. Controlled removal of the limb in surgery does greatly improve the outlook of this process.

Signal processing

Once that signal is detected, the signal is processed. Most signals, especially EMG signals are prone to disturbances and need filters to produce useable data. A much applied filter here is the two-pass Butterworth [34]. In the model-based approach mentioned before, this processed signal is used as an input in a forward dynamics model to compute joint moments. These moments can be used in the virtual model estimating the joint kinematics. One can then also measure those joint kinematics and combine to an optimal estimate of the state using a Kalman filter. This process can be further optimized with an increasing number of sensors, integrating to the most likely state at the moment of measurement (multisensory integration)

HMI actuation side

The optimal state can be used to control an actuator (low-level control). One of the most important metrics here is passivity. Human-machine interactions can become unstable, even when both the HMCS and AMCS are stable. Calculating interaction stability of the coupled HMCS-AMCS with the outside world is very difficult for any unknown reaction. Therefore it is very difficult to guarantee interaction stability and one of the safest ways to guarantee that the interaction is stable, is by ensuring that the AMCS is energy passive, as we have empirically determined that the interaction of humans with passive objects is stable [35]. Energy passivity means that a power conjugated dynamic system is passive if and only if the energy that the system outputs can never be more than its input, or the equation below [35]:

$$\int_{-\infty}^{t} F(\tau)v(\tau)d\tau \geq 0$$

Here, F and v are the power conjugated force and velocity inputs or outputs of a system over time τ.

State of the art

The state of the art in the specific area of rehabilitation with at least some remaining muscle function is found to be one of two options. A wearable exosuit or orthosis made of softer materials or a more rigid exoskeleton. The device can be fully wearable or with an external power source, however, as the focus is on rehabilitation in clinical or research context, there is no immediate need for a system that is fully wearable. State of the art systems like the Lopes II are not meant to be fully wearable. Often, the fully wearable systems will be heavy, hindering the motion in rehabilitation, which makes tethered systems of great interest. The state of the art that is available at the UT is the tethered Bowden cable actuated orthoses for ankle plantarflexion. At this point in time this device only allows for plantarflexion support but doesn't take precise force delivery into account.

Precise force delivery

When talking about precise force delivery, there are two aspects relevant in the context of exosuits. First, the forces provided by the actuation system must be precise and secondly the way the force is transmitted to the body. The second aspect is of great interest here. While NMS controllers like the one used in the Symbitron exoskeleton [36] allow for muscle adaptations, none of the exoskeletons or exosuits allow for precise force delivery to for example readjust joint orientation during rehabilitation. For the force transmission, most of what is discussed so far deals with building an exoskeleton on the level of aiding or restoring body functions like walking & reaching. Studies do this by looking at the effects of the exoskeleton on for example plantarflexion force. But the human body has multiple plantarflexors and joints are often actuated by multiple muscles. Therefore, one can say that simply providing a plantarflexion force is not precise force delivery as there is no distinction between the different muscles that are actuated.

Not every actuation type is as useful for precise actuations as others while taking weight into account as an important constraint. Note that pneumatic or hydraulic actuators still require the actual actuators to be mounted at the distal part of the leg, greatly reducing metabolic efficiency of the device. Devices that can be made tethered are in that regard much more attractive. Also, as Meijners mentioned already [18], one must take into account the risks that come with leaks and custom build cylinders for these systems.

One versus two degree of freedom systems

A different way of categorising the systems, the one that will be used in the rest of this thesis, is the categorisation according to the degrees of freedom the system allows and actuates. Table 1categorizes the existing state of the art (this list isn't necessarily exhaustive). As the goal is precise force delivery while allowing for natural body motion, only systems that allow or can be made to allow multi degree of freedom movement are considered. Furthermore, these systems must actuate or must be adaptable to actuate multi degree of freedom movement. Under actuated degrees of freedom, the first digit indicates the degrees of freedom, while the term between brackets indicates whether that degree of freedom is actuated in one or two directions.

Name	Type	Actuation type	Actuation transmission	Actuated DoF's	Allowed DoF's	Weight per leg[grams]	PF RoM [°]	Torque available [Nm]
UT J-LO [18]	1DoF Bowden - Rigid	Bowden cable	SEA	1 (1D)	1	1800	-25	20
Witte AFO type A [37]	1DoF Bowden - Rigid	Bowden cable	Leaf spring SEE	1 (1D)	1	0.835	-30	120
Witte AFO type B [37]	1DoF Bowden - Rigid	Bowden cable	Leaf spring SEE	1 (1D)	1	0.875	-30	150

Name	Type	Actuation type	Actuation transmission	Actuated DoF's	Allowed DoF's	Weight per leg[grams]	PF RoM [°]	Torque available [Nm]
ReWalk Restore [7]	1DoF Bowden - Flexible	Bowden cable		1 (2D)	2			
Achilles [38]	1DoF non-Bowden	Linear Electric	Leaf spring SEE	1 (1D)	1	1500	-12	35
Hitt et al. [39]	1DoF non-Bowden	Linear Electric	Linear spring SEE	1 (1D)	1	1700	-25	41
Martijn Grootens [40]	2DoF non-Bowden	Linear Pneumatic	N/A	2 (2D)	2	3580	-45	74
ExoBoot [41]	1DoF non-Bowden	Soft pneumatic	N/A	1 (1D)	2	255	-20	23
Malcolm [42]	1Dof Bowden - Flexible	Bowden cable	N/A	1 (1D)	2	302	-18	28 @80kg (0.35 Nm/kg)
Gordon, Ferris [43]	1DoF non-Bowden	Artificial muscle (pneumatic)	N/A	1 (1D)	1	1200 ± 100	-17	50
Collins [44]	Unpowered	Spring actuated	N/A	1 (1D)	1	408 - 503	-18	9 @80kg (0.11 Nm/kg)
Mooney [45]	1DoF non-Bowden	Cable powered lever	Belt transmission	1 (1D)	1	1060		120
Symbitron [36]	2DoF exoskeleton	Rotary electric	SEA	2 (2D)	2	Self-supporting: N/A	-14	75

Figure 4: Table summarising the state of the art

Part II: Targeted rehabilitation

The second part of the review deals more with the medical rather than the technical side of rehabilitation. Starting with a general review of human gait and anatomy surrounding the ankle, it works towards possible impairments, strategies of rehabilitation, and it ends with a combination of part I and II in which the effects of (long-term) mechanical support on patients.

Biomechanics of the ankle

The ankle is one of the most complex joints in the human body. In this chapter, the biomechanics of this joint during motion are briefly described, while also taking the variation over the gait into account. Finally, the effects of an assistive device on the kinematics and especially the dynamics are discussed.

Because of the geometry of the ankle the biomechanics is not as straightforward as for example the knee. Generally, motion of the body can be expressed in three axes and planes. Without losing too much accuracy, the ankle motion can be reduced to two axes, the PF/DF axis and the subtalar axis. Rotary motion around the axis colinear with the lower leg is not a degree of freedom of the ankle, but of the lower leg. These motions are actuated by about 10 muscles. A contraction of one of these muscles in general causes motion in 2 planes. A cable-based actuation system will likely also have that effect, causing parasitic motion in nature. The orientation of these two axes is displayed in Figure 5[46].

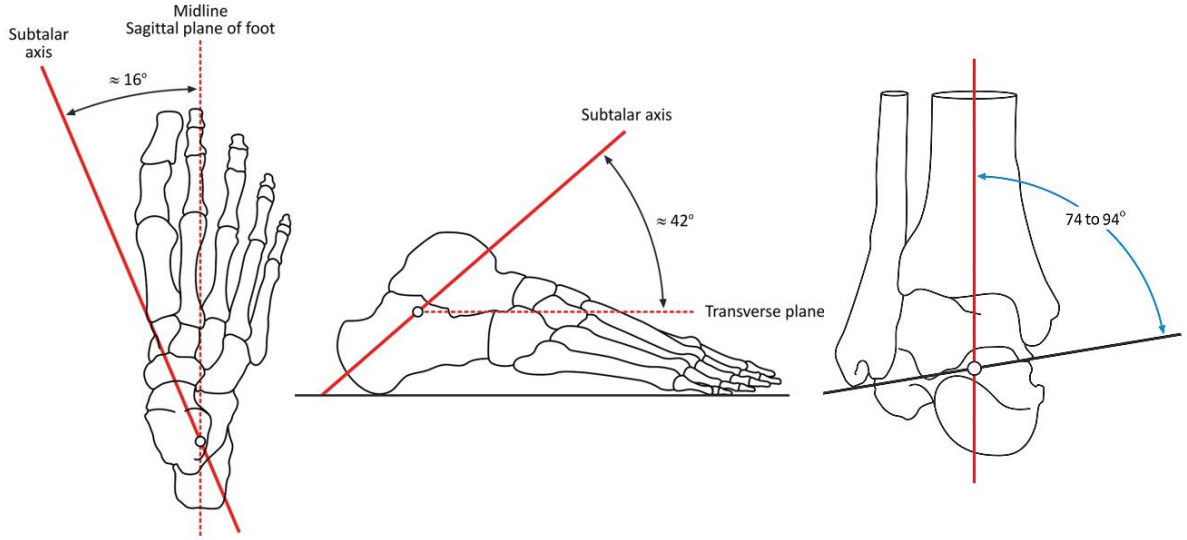


Figure 5: Axes of the ankle

Ankle anatomy & human gait

As the ankle is complex, it is often reduced in exoskeleton design. Figure 6 [47] defines terms on axes and planes. One can see 6 degrees of freedom (DoF's) are defined, and in reality, the ankle joint has that many DoF's. As was described in part I, most exosuits and/or orthoses will assist in plantarflexion and in some cases in dorsiflexion. In this, the joint is simplified to a hinge with one DoF and only plantar- and dorsiflexion action. The actuation happens parallel to the main plantarflexors (plantaris, lateral and medial gastrocnemius and soleus) and the main dorsiflexor (tibialis anterior), but neglects lines of action of for example the peroneus longus and peroneus brevis, which facilitate eversion. Table 1 describes the main muscles used in movements over the frontal and sagittal planes. Figure 7 provides a visual aid of those muscles [48].

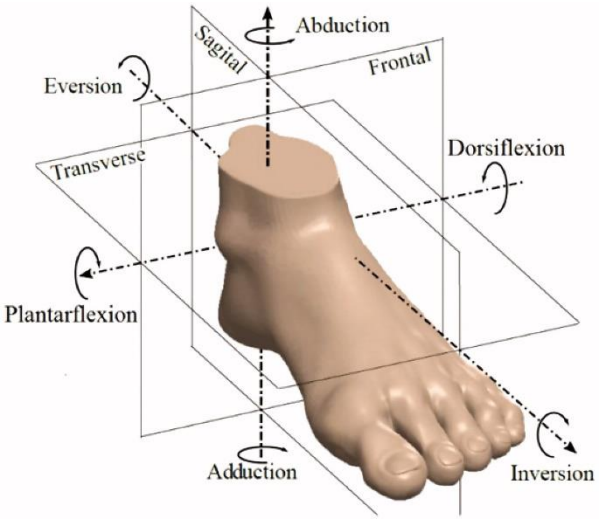


Figure 6: Planes and axes of the ankle

Table 1: Movement in the sagittal vs. the frontal plane seen from both perspectives

Action	Invertors	Evertors
Plantarflexion	Plantaris Soleus Tibialis posterior Flexor hallucis longus Flexor digitorum longus	Gastrocnemius Peroneus longus Peroneus brevis
Dorsiflexion	Tibialis anterior	Extensor hallucis longus Extensor digitorum longus Peroneus tertius
Action	**Plantarflexors**	**Dorsiflexors**
Inversion	Plantaris Soleus Tibialis posterior Flexor hallucis longus Flexor digitorum longus	Tibialis anterior Extensor hallucis longus
Eversion	Gastrocnemius Peroneus longus Peroneus brevis	Extensor digitorum longus Peroneus tertius

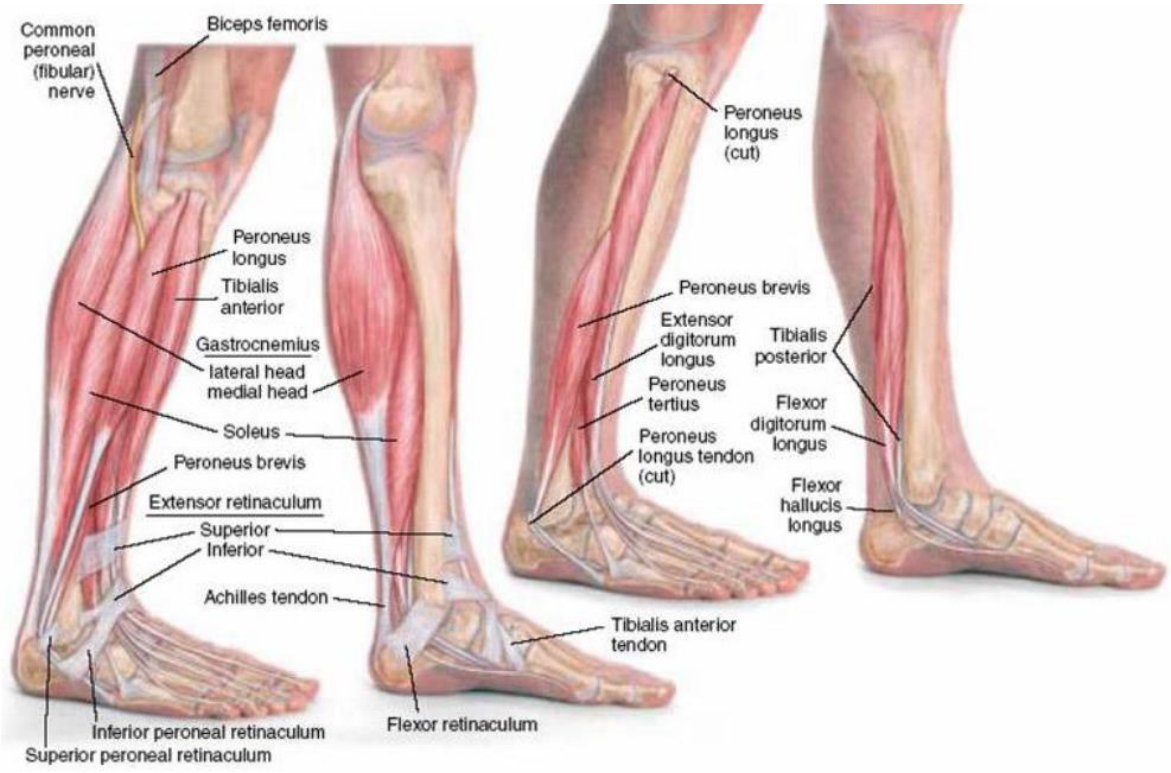


Figure 7: Overview of ankle actuating muscles

The human gait can be portrayed as in Figure 8 [49]. As one can see, the right leg starts providing peak plantarflexion force around mid-stance and does that through terminal stance and until toe-off. This is also when the body should provide most of its force, which is naturally what most exosuits or orthoses target.

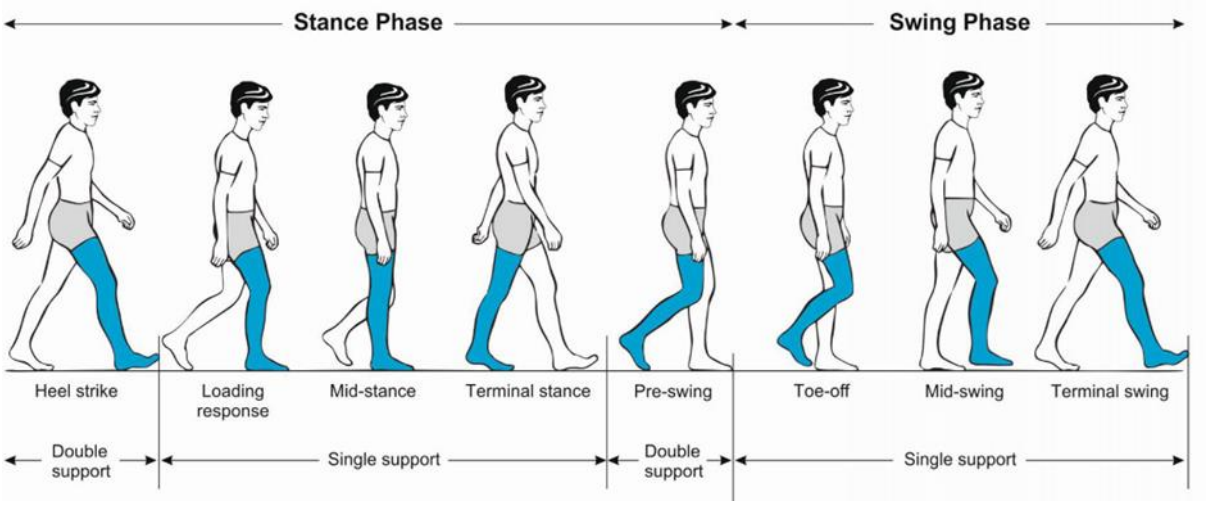


Figure 8: Human gait phases

Over that gait cycle, in the case of patient monitoring, researchers are often interested in the kinematics of and torques on the joints. For several datasets, J. Meijners summarized and averaged those parameters in Figure 9 [18].

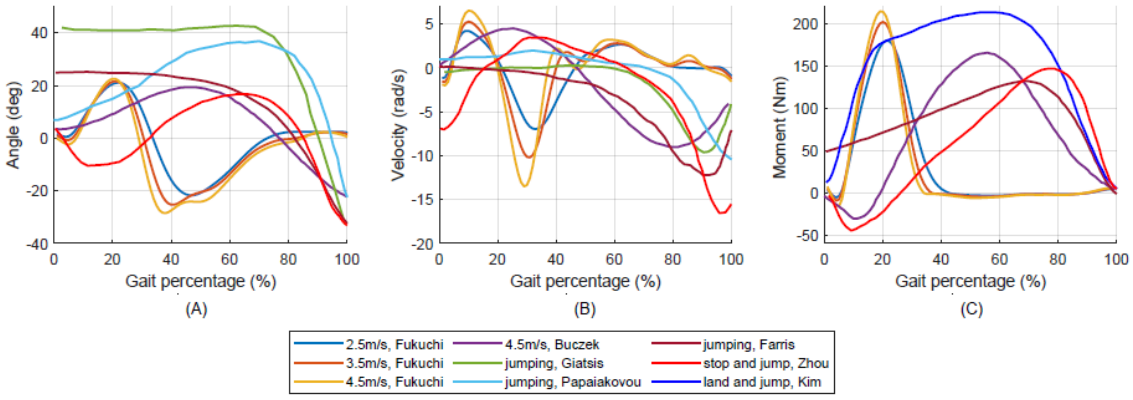


Figure 9: Ankle data over gait cycle

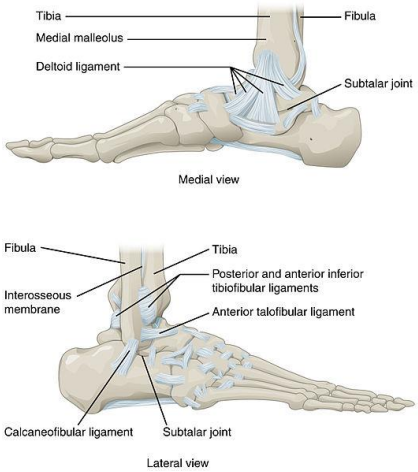


Figure 10: Ankle bones and ligaments

However, during this whole gait cycle, the muscles also provide stability to the body, and as was seen earlier in this paragraph, a lot of muscles are involved in this. Currently, no exosuits provide full support for this balance. Not only muscles are responsible for this stability, but stability is also provided passively. First, the joint is secured by the shape of the bones in the ankle, and secondly by the ligaments surrounding those bones [50], which can be seen clearly in Figure 10. Also, to reduce friction between those bones, cartilage is present, visualized in Figure 11 [51].

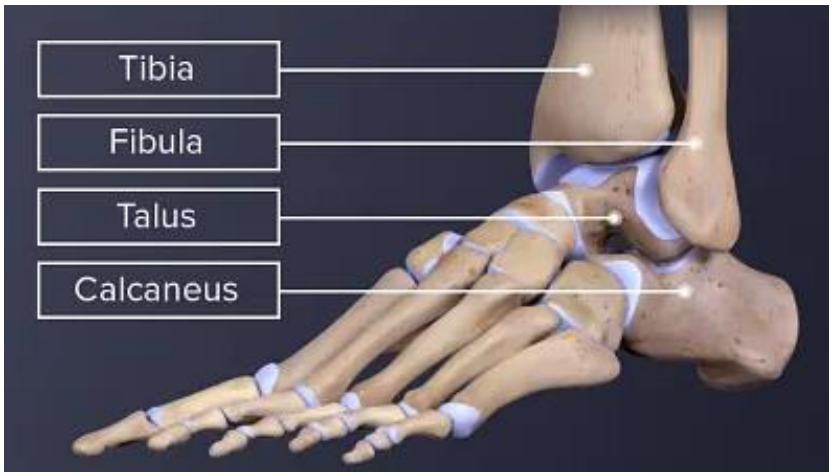


Figure 11: Cartilage in ankle joint

Possible impairments and which to tackle

As was found in the previous chapter, the ankle joint is relatively complex. Damage to either of the structures mentioned can result in reduced function in normal tasks like walking and running. Damage caused by a stroke or damage to the innervation of the muscles might also lead to reduced function. If one is to engage effectively in a rehabilitation process, one must not only understand the effects of certain impairments on gait, function and comfort, but also to what extent those effects can be slowed down or even reversed. Note that impairments discussed below comprise well over 80% of paralysis cases in the US [52], thus giving a good reference frame.

Muscle based impairments

Muscle impairments cause a disturbance in actuators in the HMCS in Figure 3. This can have several different causes, ranging from atrophy to tears.

Muscle atrophy can have several causes, among others trauma, dystrophy, inactivity, ageing or malnutrition. As long as the atrophy is not caused by dystrophy or ageing, patients can rehabilitate back to full function in many cases. For example, patients that suffer from loss of muscle mass caused by either prolonged periods of inactivity or mechanical unloading can regain full muscle mass with therapy, proven by the quadriceps muscle gaining an average of 6,2% muscle mass in 12 weeks of resistance training in young women (vs. 2,5% in older women) [53]. Muscular dystrophy (MD) can also cause muscle atrophy and is in turn caused by genetic mutations (not always inherited). Not all types of MD do necessarily cause trouble walking, but patients suffering from those that do benefit from exercise and physiotherapy [54]. As was already proven by Ziter et. al. in 1979, MD (Duchenne in this case) patients benefit from the use of orthoses [55]. More recently, Weichbrodt and their team also noted that degeneration of wrist muscles is at least slowed by the use of orthoses [56] and that the period of assisted movement and/or stance is prolonged, a conclusion that is shared by multiple researches, however there is little evidence that functional muscle use is regained [57].

Ruptures will first need surgery or rest after which rehabilitation starts. This is often a long and painful process, as patients can long after repair surgeries still suffer from weakness and pain with limited motion [58]. In this case, the torn muscle was the pectoralis major, but orthoses are widely applied on the lower extremities. A (stiff) orthosis can also be used to protect the muscle after surgery.

Starting exercises relatively quickly after surgery reduces muscle atrophy, however this is not always safe as it may damage structures and tissues. However, doing those exercises at decreased load with an anti-gravity treadmill does lead to better gait in patients [59]. At least at a passive level in people with calf muscle weakness, ankle-foot orthoses can have that same effect [60].

Cartilage based impairments

Damaged cartilage between any one of the bones depicted in Figure 11 can lead to serious pain during everyday activities as walking, but especially when the cartilage between the talus and the tibia or fibula is damaged. Rehabilitation cannot help to regrow cartilage, but it can be very effective in assisting in stabilizing the joint after total ankle arthroplasty or ankle arthrodesis. The cartilage damage often comes as a result of arthritis, for example rheumatoid arthritis, osteoarthritis, or post-traumatic arthritis.

Passive ankle-foot orthoses have only limited influence on joint stability and indeed show no reversive effects [61].

Tendon based impairments

Tendon injuries are fairly common, especially in sports (22% of total injuries [62]), and in the general population about 1.6% [63], mostly affecting the elderly part of population. That elderly part mostly suffers from tendinitis or tendinosis, while sports injuries are often physical trauma, like tears or ruptures. Both tendinosis and tendinitis are treated with rest first. After that, the treatment can be summarized as careful proper progressive exercise and for tendinosis it might be beneficial to use appropriate support, as collagen can take over 100 days to regrow [64].

If the tendinosis is not addressed properly, or in the case of trauma, a tendon can tear or even rupture. After surgery, which is only the beginning of the rehabilitation period, patients can start different treatment regimes. As Brumann et. al. [65] showed, full-weight bearing has the highest patient satisfaction and earliest return to pre-injury activities.

Signal based impairments

The human control system (HCS), which on a very general scale consists of the brain, the spinal cord and the neurons connecting those systems with each other and the muscles, is a delicate system. Damage anywhere in the HCS can cause patients to permanently lose function in one or more limbs. However, not all damage done is irreparable.

Motor neuron diseases, like amyotrophic lateral sclerosis (ALS) or primary lateral sclerosis (PLS) affect the neurons in the brain, which causes muscles to slowly loose volume. While ALS is fatal, exercise can help improve quality of life, which also holds for PLS (which isn't fatal) [66]. Of course, once nerves are irreparably damaged, rehabilitation has no effect anymore.

Multiple sclerosis (MS) patients, a demyelinating disease, benefit from rehabilitation not only on the motor symptoms and fatigue that is so typical for MS, but also have increased neuroplasticity and maybe a neuroprotective effect [67].

Stroke patients also greatly benefit from rehabilitation, with or without exosuits [7]. The same holds for any disease that damages the HCS: as long as the patient has some function left, rehabilitation exercise is still beneficial.

Other impairments

Broken bones or torn ligaments cannot directly be healed with rehabilitation, but need surgery and/or time. However, often motion of the affected limbs is impaired over longer periods of time. This is not a problem for healthy patients, but patients with other underlying diseases may helped by assisted rehabilitation as much as any of the patients mentioned above.

Strategies for rehabilitation

Rehabilitation is generally strategized in comparable ways across different fields, however there is not much standardization. The injury is first classified (examination), then diagnosed and lastly treatment is started based on the diagnosis [68]. In the context of assisted rehabilitation, one might opt to add modelling in there. A good musculoskeletal (MS) model is necessary for the NMS model explained in the section on control. From this point onward, the examination and diagnosis are assumed to be known. Modelling and treatment are discussed below.

Modelling

First, based on some (mostly external) markers, and an MS model is developed of the patients' limb(s). This is still very much in the early stages of developments and a lot of research is needed [69]. Software like SimTK's OpenSim can quite accurately model the joint dynamics and the model can thus be used as a virtual model in parallel with the real dynamics or in a stand-alone situation as a test model.

Treatment

For the treatment, the main question in this context is to what extent the use of an orthosis would be beneficial to the recovery of the patient. Not all patients would benefit from the same support, which means some degree of modularity is desirable.

Effects on joints of (long-term) mechanical support

As argued before on precise force delivery, it is important to take notion of the effect of direct actuation on muscles. This is not only true for effects while wearing the orthosis or exosuit, but just as important for the effects over time.

In the context of correcting impairments of stance, for example on children with paediatric flexible flat foot, the evidence is contracting and no real conclusions can be drawn. Some studies draw the conclusion that there is no strong evidence that the long term use has improved structural problems [70], while others claim that long-term use of (arch support) foot orthoses proves feasible to improve lower limb alignment [71]. Also, in children with cerebral palsy, it was found that using ankle-foot orthoses (AFO) either maintained or improved foot deformities or dysfunction [72]. In hemiplegic patients, also the effects on muscle activity is unclear [73]. The general conclusion is that passive AFOs have no significant influence on recovery or gait, but also don't have a negative influence.

Also, research is being done on the effects of aligning the axes of rotation of a human and the exoskeleton, and it is found that passive compensation for the alignment can lower interaction forces and torques at the HMI by 70% and 60% respectively [74]. However, this is also where the research seems to stop. There is no research on the influence of actuation without attention to precise force delivery versus actuation over multiple lines of action on a muscle or joint level over longer periods of time.

Applications & gaps

After the extensive review of the current and older literature, there are some clear gaps emerging. First of all, it was found that barely any studies deal with the longer-term consequences of wearing rehabilitation gear in general, but especially not on ankle orthoses. The results of such research would be very useful to any team developing assistive devices to the human body as it might provide new design insights.

One can see the most recent developments on stable ankle orthoses leaning towards softer materials, as for example the team of Conor Walsh proves. These orthoses first only assisted in plantarflexion but can as of very recently also provide dorsiflexion torque. However, they only provide that torque over one line of action and are as such essentially approaching a highly MDoF problem as a SDoF system. A (modular) device that can provide assistance over multiple lines of action is not yet on the market or in research.

There are some first careful steps in the pond of 2DoF devices, but the problems of misalignment do remain here. Martijn Grootens drew the conclusion that, using linear (pneumatic) actuators, a softer device doesn't remain stable.

Therefore, one can clearly state that there is a gap in the knowledge of 2DoF ankle orthoses. This thesis will aim to judge the effectiveness of a Bowden cable actuated ankle orthosis (given that the Bowden cable input stage is already available within the J-Lo project). Th

Research goals

Given the gaps in the research and the current state of the art, the following research goals can be set up. Note that the work of Jeroen Meijners is assumed as a base and the actuation designed there is assumed to be functional. Any optimizations that are being done on controller stability as assumed out of scope for this project.

The primary research goal of this project is to review the possibilities in novel 2DoF Bowden cable actuated ankle-foot orthosis design. A device will be designed that can transmit loading in parallel to the musculotendon structures to mimic natural actuation as much as possible. The aim is to solve the misalignment problem this way. The effectiveness must be judged in terms of mechanical functionality and the effect it has on the muscle(s) (activation).

References

[1] V. Kumar, Y. V. Hote, and S. Jain, "Review of Exoskeleton: History, Design and Control," *2019 3rd International Conference on Recent Developments in Control, Automation & Power Engineering (RDCAPE),* pp. 677-682, 2019.

[2] N. Yagn, "Apparatus for facilitating walking," United States of America Patent US420179A Patent Appl. US420179A, 1890.

[3] G. E. C. S. M. H. P. Operation and B. J. Makinson, *Research and Development Prototype for Machine Augmentation of Human Strength and Endurance: Hardiman I Project*. The Company, 1971.

[4] H. Kazerooni, "The Berkeley Lower Extremity Exoskeleton," in *Field and Service Robotics*, Berlin, Heidelberg, P. Corke and S. Sukkariah, Eds., 2006// 2006: Springer Berlin Heidelberg, pp. 9-15.

[5] L. Baker *et al.*, "Ankle optimization with a soft exosuit reduces metabolic cost of loaded walking," in *2017 International Symposium on Wearable Robotics and Rehabilitation (WeRob)*, 5-8 Nov. 2017 2017, pp. 1-1, doi: 10.1109/WEROB.2017.8383857.

[6] S. Galle, P. Malcolm, S. H. Collins, and D. De Clercq, "Reducing the metabolic cost of walking with an ankle exoskeleton: interaction between actuation timing and power," *Journal of NeuroEngineering and Rehabilitation,* vol. 14, no. 1, p. 35, 2017/04/27 2017, doi: 10.1186/s12984-017-0235-0.

[7] L. N. Awad, A. Esquenazi, G. E. Francisco, K. J. Nolan, and A. Jayaraman, "The ReWalk ReStore™ soft robotic exosuit: a multi-site clinical trial of the safety, reliability, and feasibility of exosuit-augmented post-stroke gait rehabilitation," *Journal of NeuroEngineering and Rehabilitation,* vol. 17, no. 1, p. 80, 2020/06/18 2020, doi: 10.1186/s12984-020-00702-5.

[8] M. A. Koch and J. M. Font-Llagunes, "Lower-Limb Exosuits for Rehabilitation or Assistance of Human Movement: A Systematic Review," *Applied Sciences,* vol. 11, no. 18, p. 8743, 2021. [Online]. Available: https://www.mdpi.com/2076-3417/11/18/8743.

[9] J. F. Veneman, R. Kruidhof, E. E. G. Hekman, R. Ekkelenkamp, E. H. F. V. Asseldonk, and H. v. d. Kooij, "Design and Evaluation of the LOPES Exoskeleton Robot for Interactive Gait Rehabilitation," *IEEE Transactions on Neural Systems and Rehabilitation Engineering,* vol. 15, no. 3, pp. 379-386, 2007, doi: 10.1109/TNSRE.2007.903919.

[10] K. L. Mudie *et al.*, "Consensus paper on testing and evaluation of military exoskeletons for the dismounted combatant," *Journal of Science and Medicine in Sport,* vol. 21, no. 11, pp. 1154-1161, 2018/11/01/ 2018, doi: https://doi.org/10.1016/j.jsams.2018.05.016.

[11] B. Murugan, "A Review on Exoskeleton for Military Purpose," (in eng), *i-Manager's Journal on Mechanical Engineering,* vol. 11, no. 2, p. 36, 2021.

[12] K. Mudie, D. Billing, A. Garofolini, T. Karakolis, and M. LaFiandra, "The need for a paradigm shift in the development of military exoskeletons," *European Journal of Sport Science,* vol. 22, no. 1, pp. 35-42, 2022/01/02 2022, doi: 10.1080/17461391.2021.1923813.

[13] Y. K. Cho, K. Kim, S. Ma, and J. Ueda, "A Robotic Wearable Exoskeleton for Construction Worker’s Safety and Health," in *Construction Research Congress 2018*, 2018, pp. 19-28.

[14] S. Bai and S. Christensen, "Biomechanical HRI Modeling and Mechatronic Design of Exoskeletons for Assistive Applications," in *Human Modeling for Bio-Inspired Robotics: Mechanical Engineering in Assistive Technologies*, 2017, pp. 251-272.

[15] N. Zhou, Y. Liu, Q. Song, and D. Wu, "A Compatible Design of a Passive Exoskeleton to Reduce the Body–Exoskeleton Interaction Force," *Machines,* vol. 10, no. 5, p. 371, 2022. [Online]. Available: https://www.mdpi.com/2075-1702/10/5/371.

[16] A. Singla, S. Dhand, A. Dhawad, and G. S. Virk, "Toward Human-Powered Lower Limb Exoskeletons: A Review," in *Harmony Search and Nature Inspired Optimization Algorithms*,

Singapore, N. Yadav, A. Yadav, J. C. Bansal, K. Deep, and J. H. Kim, Eds., 2019// 2019: Springer Singapore, pp. 783-795.
[17] S. Maowen, X. Ouyang, J. Mattila, H. Yang, and G. Hou, "One Novel Hydraulic Actuating System for the Lower-Body Exoskeleton," *Chinese Journal of Mechanical Engineering,* vol. 34, 12/01 2021, doi: 10.1186/s10033-021-00535-w.
[18] J. Meijners, "The design, control and integration of a bowden cable actuation for an ankle exoskeleton," Master of Science, Biomechanical Engineering, University of Twente, 2022.
[19] J. Nassour, G. Zhao, and M. Grimmer, "Soft pneumatic elbow exoskeleton reduces the muscle activity, metabolic cost and fatigue during holding and carrying of loads," *Scientific Reports,* vol. 11, no. 1, p. 12556, 2021/06/15 2021, doi: 10.1038/s41598-021-91702-5.
[20] A. Zoss and H. Kazerooni, "Design of an electrically actuated lower extremity exoskeleton," *Advanced Robotics,* vol. 20, pp. 967-988, 01/01 2006, doi: 10.1163/156855306778394030.
[21] G. A. Pratt and M. M. Williamson, "Series elastic actuators," in *Proceedings 1995 IEEE/RSJ International Conference on Intelligent Robots and Systems. Human Robot Interaction and Cooperative Robots*, 5-9 Aug. 1995 1995, vol. 1, pp. 399-406 vol.1, doi: 10.1109/IROS.1995.525827.
[22] J. Chen and W. H. Liao, "Design and control of a Magnetorheological actuator for leg exoskeleton," in *2007 IEEE International Conference on Robotics and Biomimetics (ROBIO)*, 15-18 Dec. 2007 2007, pp. 1388-1393, doi: 10.1109/ROBIO.2007.4522367.
[23] G. Liu, F. Gao, D. Wang, and W.-H. Liao, "Medical applications of magnetorheological fluid: a systematic review," *Smart Materials and Structures,* vol. 31, no. 4, p. 043002, 2022/03/01 2022, doi: 10.1088/1361-665x/ac54e7.
[24] S. A. Wahid, I. Ismail, S. Aid, and M. S. A. Rahim, "Magneto-rheological defects and failures: A review," *IOP Conference Series: Materials Science and Engineering,* vol. 114, p. 012101, 2016/02 2016, doi: 10.1088/1757-899x/114/1/012101.
[25] Harvard Biodesign Lab. "Soft Exosuits." https://biodesign.seas.harvard.edu/soft-exosuits (accessed.
[26] J. Zhang *et al.*, "Human-in-the-loop optimization of exoskeleton assistance during walking," *Science,* vol. 356, no. 6344, pp. 1280-1284, 2017, doi: doi:10.1126/science.aal5054.
[27] J. Vantilt *et al.*, "Model-based control for exoskeletons with series elastic actuators evaluated on sit-to-stand movements," *Journal of NeuroEngineering and Rehabilitation,* vol. 16, no. 1, p. 65, 2019/06/03 2019, doi: 10.1186/s12984-019-0526-8.
[28] P. H. Veltink, "Sensory feedback in artificial control of human mobility," *Technology and Health Care,* vol. 7, pp. 383-391, 1999, doi: 10.3233/THC-1999-7602.
[29] J. Marques, S. Ramos, M. Macedo, and H. Plácido da Silva, *Study of Mechanomyographic Alternatives to EMG Sensors for a Low-Cost Open Source Bionic Hand*. 2018.
[30] A. Suberbiola, E. Zulueta, J. M. Lopez-Guede, I. Etxeberria-Agiriano, and M. Graña, "Arm Orthosis/Prosthesis Movement Control Based on Surface EMG Signal Extraction," *International Journal of Neural Systems,* vol. 25, no. 03, p. 1550009, 2015, doi: 10.1142/s0129065715500094.
[31] J. Zhang, B. Wang, C. Zhang, Y. Xiao, and M. Wang, "An EEG/EMG/EOG-based multimodal human-machine interface to real-time control of a soft robot hand," *Frontiers in Neurorobotics,* vol. 13, 03/29 2019, doi: 10.3389/fnbot.2019.00007.
[32] J. Lobo-Prat, P. N. Kooren, A. H. A. Stienen, J. L. Herder, B. F. J. M. Koopman, and P. H. Veltink, "Non-invasive control interfaces forintention detection in active movement-assistive devices," *Journal of NeuroEngineering and Rehabilitation,* vol. 11, no. 1, p. 168, 2014/12/17 2014, doi: 10.1186/1743-0003-11-168.
[33] A. Trafton. "Magnets could offer better control of prosthetic limbs." https://news.mit.edu/2021/magnet-prosthetic-limb-control-0818 (accessed 12/10/2022.

[34] C. J. De Luca, L. Donald Gilmore, M. Kuznetsov, and S. H. Roy, "Filtering the surface EMG signal: Movement artifact and baseline noise contamination," *Journal of Biomechanics,* vol. 43, no. 8, pp. 1573-1579, 2010/05/28/ 2010, doi: https://doi.org/10.1016/j.jbiomech.2010.01.027.

[35] A. Q. Keemink, H. van der Kooij, and A. H. Stienen, "Admittance control for physical human–robot interaction," *The International Journal of Robotics Research,* vol. 37, no. 11, pp. 1421-1444, 2018, doi: 10.1177/0278364918768950.

[36] C. Meijneke *et al.*, "Symbitron Exoskeleton: Design, Control, and Evaluation of a Modular Exoskeleton for Incomplete and Complete Spinal Cord Injured Individuals," (in eng), *IEEE Trans Neural Syst Rehabil Eng,* vol. 29, pp. 330-339, 2021, doi: 10.1109/tnsre.2021.3049960.

[37] K. A. Witte, J. Zhang, R. W. Jackson, and S. H. Collins, "Design of two lightweight, high-bandwidth torque-controlled ankle exoskeletons," in *2015 IEEE International Conference on Robotics and Automation (ICRA)*, 26-30 May 2015 2015, pp. 1223-1228, doi: 10.1109/ICRA.2015.7139347.

[38] C. Meijneke, W. van Dijk, and H. van der Kooij, "Achilles: An autonomous lightweight ankle exoskeleton to provide push-off power," in *5th IEEE RAS & EMBS Interantional Conference on Biomedical Robotics and Biomechatronics, BioRob 2014*, ed. Sao Paulo, Brazil: IEEE, 2014, pp. 918-923.

[39] J. Hitt, A. M. Oymagil, T. Sugar, K. Hollander, A. Boehler, and J. Fleeger, "Dynamically Controlled Ankle-Foot Orthosis (DCO) with Regenerative Kinetics: Incrementally Attaining User Portability," in *Proceedings 2007 IEEE International Conference on Robotics and Automation*, 10-14 April 2007 2007, pp. 1541-1546, doi: 10.1109/ROBOT.2007.363543.

[40] M. E. Grootens, "Development of a 2-DoF Ankle Exoskeleton," EngD Thesis, University of Twente, Enschede, 2019.

[41] J. Chung, R. Heimgartner, C. T. O. Neill, N. S. Phipps, and C. J. Walsh, "ExoBoot, a Soft Inflatable Robotic Boot to Assist Ankle During Walking: Design, Characterization and Preliminary Tests," in *2018 7th IEEE International Conference on Biomedical Robotics and Biomechatronics (Biorob)*, 26-29 Aug. 2018 2018, pp. 509-516, doi: 10.1109/BIOROB.2018.8487903.

[42] P. Malcolm *et al.*, "Varying negative work assistance at the ankle with a soft exosuit during loaded walking," *Journal of NeuroEngineering and Rehabilitation,* vol. 14, no. 1, p. 62, 2017/06/26 2017, doi: 10.1186/s12984-017-0267-5.

[43] K. E. Gordon and D. P. Ferris, "Learning to walk with a robotic ankle exoskeleton," (in eng), *J Biomech,* vol. 40, no. 12, pp. 2636-44, 2007, doi: 10.1016/j.jbiomech.2006.12.006.

[44] S. H. Collins, M. B. Wiggin, and G. S. Sawicki, "Reducing the energy cost of human walking using an unpowered exoskeleton," *Nature,* vol. 522, no. 7555, pp. 212-215, 2015/06/01 2015, doi: 10.1038/nature14288.

[45] L. M. Mooney, E. J. Rouse, and H. M. Herr, "Autonomous exoskeleton reduces metabolic cost of human walking," *Journal of NeuroEngineering and Rehabilitation,* vol. 11, no. 1, p. 151, 2014/11/03 2014, doi: 10.1186/1743-0003-11-151.

[46] F. Harrold and R. J. Abboud. "Chapter 2 - Biomechanics of the Foot and Ankle." https://musculoskeletalkey.com/chapter-2-biomechanics-of-the-foot-and-ankle/ (accessed 04/09/2023.

[47] M. G. Alvarez-Perez, M. A. Garcia-Murillo, and J. J. Cervantes-Sánchez, "Robot-assisted ankle rehabilitation: a review," *Disability and Rehabilitation: Assistive Technology,* vol. 15, no. 4, pp. 394-408, 2020/05/18 2020, doi: 10.1080/17483107.2019.1578424.

[48] S. Carey. "Developing Strength & Stability in the Foot, Ankle and Lower Leg." https://www.mountainpeakfitness.com/blog/strength-stability-foot-ankle-lower-leg (accessed 04/10/2022.

[49] N. D. Nordin, A. G. Muthalif, and M. K. M Razali, "Control of transtibial prosthetic limb with magnetorheological fluid damper by using a fuzzy PID controller," *Journal of Low Frequency Noise, Vibration and Active Control,* vol. 37, no. 4, pp. 1067-1078, 2018, doi: 10.1177/1461348418766171.

[50] O. Rasmussen, "Stability of the Ankle Joint," *Acta Orthopaedica Scandinavica,* vol. 56, no. sup211, pp. 1-75, 1985/01/01 1985, doi: 10.3109/17453678509154152.
[51] D. Sorrento. "Ankle Joint Anatomy and Osteoarthritis." https://www.arthritis-health.com/types/osteoarthritis/ankle-joint-anatomy-and-osteoarthritis (accessed 04/10/2022.
[52] B. S. Armour, E. A. Courtney-Long, M. H. Fox, H. Fredine, and A. Cahill, "Prevalence and Causes of Paralysis-United States, 2013," (in eng), *Am J Public Health,* vol. 106, no. 10, pp. 1855-7, Oct 2016, doi: 10.2105/ajph.2016.303270.
[53] C. A. Greig *et al.*, "Blunting of adaptive responses to resistance exercise training in women over 75y," *Experimental Gerontology,* vol. 46, no. 11, pp. 884-890, 2011/11/01/ 2011, doi: https://doi.org/10.1016/j.exger.2011.07.010.
[54] National Health Service. "Muscular Dystrophy: Treatment." https://www.nhs.uk/conditions/muscular-dystrophy/treatment/ (accessed 05/10/2022.
[55] F. A. Ziter and K. G. Allsop, "The value of orthoses for patients with Duchenne muscular dystrophy," *Physical Therapy,* Article vol. 59, no. 11, pp. 1361-1365, 1979, doi: 10.1093/ptj/59.11.1361.
[56] J. Weichbrodt, B. M. Eriksson, and A. K. Kroksmark, "Evaluation of hand orthoses in Duchenne muscular dystrophy," *Disability and Rehabilitation,* Article vol. 40, no. 23, pp. 2824-2832, 2018, doi: 10.1080/09638288.2017.1347721.
[57] J. P. J. Bakker, I. J. M. de Groot, H. Beckerman, B. A. de Jong, and G. J. Lankhorst, "The effects of knee-ankle-foot orthoses in the treatment of Duchenne muscular dystrophy: Review of the literature," *Clinical Rehabilitation,* Article vol. 14, no. 4, pp. 343-359, 2000, doi: 10.1191/0269215500cr319oa.
[58] J. Moore, "Functional orthosis post pectoralis muscle rupture," *Journal of Hand Therapy,* vol. 28, no. 3, pp. 325-328, 2015/07/01/ 2015, doi: https://doi.org/10.1016/j.jht.2014.11.005.
[59] L. Palke *et al.*, "Anti-gravity treadmill rehabilitation improves gait and muscle atrophy in patients with surgically treated ankle and tibial plateau fractures after one year: A randomised clinical trial," *Clinical Rehabilitation,* vol. 36, no. 1, pp. 87-98, 2022, doi: 10.1177/02692155211037148.
[60] N. F. J. Waterval, F. Nollet, J. Harlaar, and M.-A. Brehm, "Modifying ankle foot orthosis stiffness in patients with calf muscle weakness: gait responses on group and individual level," *Journal of NeuroEngineering and Rehabilitation,* vol. 16, no. 1, p. 120, 2019/10/17 2019, doi: 10.1186/s12984-019-0600-2.
[61] Y.-C. Huang *et al.*, "Effects of Ankle-Foot Orthoses on Ankle and Foot Kinematics in Patients With Subtalar Osteoarthritis," *Archives of Physical Medicine and Rehabilitation,* vol. 87, no. 8, pp. 1131-1136, 2006/08/01/ 2006, doi: https://doi.org/10.1016/j.apmr.2006.03.021.
[62] D. Florit, C. Pedret, M. Casals, P. Malliaras, D. Sugimoto, and G. Rodas, "Incidence of Tendinopathy in Team Sports in a Multidisciplinary Sports Club Over 8 Seasons," (in eng), *J Sports Sci Med,* vol. 18, no. 4, pp. 780-788, Dec 2019.
[63] H. Riel, C. F. Lindstrøm, M. S. Rathleff, M. B. Jensen, and J. L. Olesen, "Prevalence and incidence rate of lower-extremity tendinopathies in a Danish general practice: a registry-based study," *BMC Musculoskeletal Disorders,* vol. 20, no. 1, p. 239, 2019/05/22 2019, doi: 10.1186/s12891-019-2629-6.
[64] E. Bass, "Tendinopathy: why the difference between tendinitis and tendinosis matters," (in eng), *Int J Ther Massage Bodywork,* vol. 5, no. 1, pp. 14-7, 2012, doi: 10.3822/ijtmb.v5i1.153.
[65] M. Brumann, S. F. Baumbach, W. Mutschler, and H. Polzer, "Accelerated rehabilitation following Achilles tendon repair after acute rupture - Development of an evidence-based treatment protocol," (in eng), *Injury,* vol. 45, no. 11, pp. 1782-90, Nov 2014, doi: 10.1016/j.injury.2014.06.022.
[66] S. Majmudar, J. Wu, and S. Paganoni, "Rehabilitation in amyotrophic lateral sclerosis: why it matters," (in eng), *Muscle Nerve,* vol. 50, no. 1, pp. 4-13, Jul 2014, doi: 10.1002/mus.24202.

[67] C. Donzé and C. Massot, "Rehabilitation in multiple sclerosis in 2021," *La Presse Médicale,* vol. 50, no. 2, p. 104066, 2021/06/01/ 2021, doi: https://doi.org/10.1016/j.lpm.2021.104066.

[68] N. Maffulli *et al.*, "ISMuLT Guidelines for muscle injuries," (in eng), *Muscles Ligaments Tendons J,* vol. 3, no. 4, pp. 241-9, Oct 2013.

[69] M. S. Shourijeh, N. Mehrabi, J. J. McPhee, and B. J. Fregly, "Editorial: Advances in Musculoskeletal Modeling and Their Application to Neurorehabilitation," *Frontiers in Neurorobotics,* Editorial vol. 14, 2020. [Online]. Available: https://www.frontiersin.org/articles/10.3389/fnbot.2020.00065.

[70] J. Y. Choi, W. H. Hong, J. S. Suh, J. H. Han, D. J. Lee, and Y. J. Lee, "The long-term structural effect of orthoses for pediatric flexible flat foot: A systematic review," *Foot and Ankle Surgery,* vol. 26, no. 2, pp. 181-188, 2020/02/01/ 2020, doi: https://doi.org/10.1016/j.fas.2019.01.007.

[71] A. Jafarnezhadgero, M. Madadi-Shad, S. M. Alavi-Mehr, and U. Granacher, "The long-term use of foot orthoses affects walking kinematics and kinetics of children with flexible flat feet: A randomized controlled trial," *PLOS ONE,* vol. 13, no. 10, p. e0205187, 2018, doi: 10.1371/journal.pone.0205187.

[72] X.-C. Liu *et al.*, "Long-Term Effects of Orthoses Use on the Changes of Foot and Ankle Joint Motions of Children With Spastic Cerebral Palsy," *PM&R,* vol. 10, no. 3, pp. 269-275, 2018, doi: https://doi.org/10.1016/j.pmrj.2017.08.438.

[73] J. Leung and A. Moseley, "Impact of Ankle-foot Orthoses on Gait and Leg Muscle Activity in Adults with Hemiplegia: Systematic literature review," *Physiotherapy,* vol. 89, no. 1, pp. 39-55, 2003/01/01/ 2003, doi: https://doi.org/10.1016/S0031-9406(05)60668-2.

[74] A. Schiele, "Ergonomics of exoskeletons: Objective performance metrics," in *World Haptics 2009 - Third Joint EuroHaptics conference and Symposium on Haptic Interfaces for Virtual Environment and Teleoperator Systems*, 18-20 March 2009 2009, pp. 103-108, doi: 10.1109/WHC.2009.4810871.

Appendix B:

Methodical Product Design

Developments in 2 degree of freedom fine force delivery by an assistive ankle orthosis

Bart Hendriksen, s2424576
Supervisor: Dr. G. V. Durandau

Draft date: November 28, 2022
Version: 1A

Table of Contents

TABLE OF CONTENTS **2**

INTRODUCTION **3**

1. ANALYSIS PHASE **4**

PROBLEM TRANSLATION FROM MEDICAL TO MECHANICAL DOMAIN 4
PROBLEM DEFINITION BY CLIENT 5
GOAL 5
DESIGN ASSIGNMENT 5
SYSTEM SUBDIVISION 6
STAKEHOLDERS 7
USE CASES 9
REQUIREMENTS & BOUNDARY CONDITIONS 10

2. SYNTHESIS PHASE **12**

STATE OF THE ART 13
1DoF Bowden systems – Rigid connections *13*
1DoF Bowden systems – Flexible connections *13*
1DoF non-Bowden systems *14*
2DoF non-Bowden systems *14*
Quantitative measures of state of the art *15*
SYNTHESIS I 16
Joint types *16*
Morphological chart I *16*
Force transmission test *18*
Comparison & weighing *18*
SYNTHESIS II 20
Morphological chart II *20*
Concepts description *22*
Requirements ranking and weighing criteria *26*
Comparison and weighing *26*
SYNTHESIS III 29
Actuation & general connection *29*
Foot bracket *32*
Shank bracket *33*
Mechanical safety limit *33*
Fatigue analysis *33*
Mass estimation *35*

REFERENCES **36**

APPENDIX: TABLE EXPLANATION MORPHOLOGICAL CHART 1 ..... **37**

Introduction

This document applies the methodology as set up in the course Integrative Design of Biomedical Products, by Prof. Dr. Ir. G.J. Verkerke. This integrative design method describes the whole project in three phases:

1) Analysis phase;
2) Synthesis phase;
3) Use phase.

These phases are all discussed in this document; however, its main purpose is describing the actual design process (synthesis phase). The analysis phase starts from a problem definition and research gaps and moves to a design goal. A literature review and a program of requirements are the final outcomes of this phase. In the synthesis phase, this is used as an input and concepts are developed. These concepts are funnelled into a final concept, which will meet the requirements at least on the headlines. This concept is then iteratively worked out into a design. The deliverables here are concept documentation and an iterative design report containing calculations and modelling. Also, the device will be validated for use experimentally, but also against the requirements. Lastly, the product design enters the use phase, in which the product is enrolled into the field of use. In this case, there will be no market to sell it on, as it will be used in an academic setting. In this context, the final deliverable will be a procedure to determine joint reaction forces using a neuromusculoskeletal model (NMS) in OpenSim and MATLAB based on the anchor points of the design.

1. Analysis Phase

This first section on the analysis phase aims to define the problem and the broad roadmap to a solution. Then, the requirements are defined, and this is all translated into a function analysis.

Problem translation from medical to mechanical domain

Millions of people worldwide are suffering from paralysis in some form. While there are no trustworthy numbers for the global numbers the global estimate is 40 – 80 million people [1]. In the United States, this number was some 5.4 million in 2016 [2], on a population of around 250 million, equating about 2% of the population of a developed country. As was found in the literature review, there is a vast range of possible causes, and one must also note that not every paralysis case would benefit from (assisted) rehabilitation.

In general, one can state that patients suffering from tendinopathy (especially tendonitis), muscle atrophy, motor impairments and early stage cartilage damage can benefit from (assisted) rehabilitation (applied to the lower extremities in this thesis) [3]. Effectively, all these impairments boil down to the mechanical principle of stress. A force increase leads to a higher stress in the tissue, a misalignment (and thus contact area decrease) leads to a higher stress etc. This mechanical stress can then lead to tissue failure in a similar way as that mechanical stress can lead to other material failures. One of the main medical causes for these impairments are strokes (> 795.000 cases on the 330 million, in 2022, population) [4]. As is well-documented for the knee, an overall increase in load magnitude across the knee is associated with an '*increased incidence of osteoarthritis as accelerated progression of obesity' [5].* During normal walking, but especially during activities such as running or jumping, knee and ankle reaction loads increase significantly [6]. Detrimental effects of added weight or imbalances thus cause a magnified reaction on a joint level.

Those forces can be increased by one of the following causes:

1) Imbalance over the knee joint (i.e. muscles on either side of the mediolateral plane are activated asymmetrically continuously). This is problematic as the knee has very limited rotational freedom in that plane.
2) Overactivated muscles. Muscles that are continuously overactive and thus have higher forces over time cause higher joint reaction loads.
3) Orientation imbalance. Varus ankle deformity is associated with ankle osteoarthritis of the varus type, but Zhu, Li and Xu couldn't find definitive proof that this was caused by prolonged exposure to eccentric loading [7].
4) External forces, like a misaligned orthosis.

The complexity of the ankle (and knee) joint makes that any treatment is a delicate matter. Pharmaceutical and surgical solutions are available, but in general should not be a first step. In a healthy joint, a balance state is desired: tissue homeostasis. Correct alignment of the joint in general, but also of the orthosis with the joint is vital for tissue homeostasis, which is mainly articulate cartilage homeostasis, defined as: *'the state at which degradation of extracellular matrix components is balanced by synthesis' [8].* In this case, the extracellular matrix material refers mainly to the collagen in cartilage. Note that also out-of-plane forces could induce stresses in the joint that hinder homeostasis, for which a correction force could be applied. Figure 1 [9] shows a cut view of the ankle. One can easily see that a misalignment of tibia and talus will lead to a decrease in force contact area and thus more pressure, and that out-of-plane forces can add either a stress on surrounding tissue, a friction force through the cartilage, or an added force by inducing misalignment.

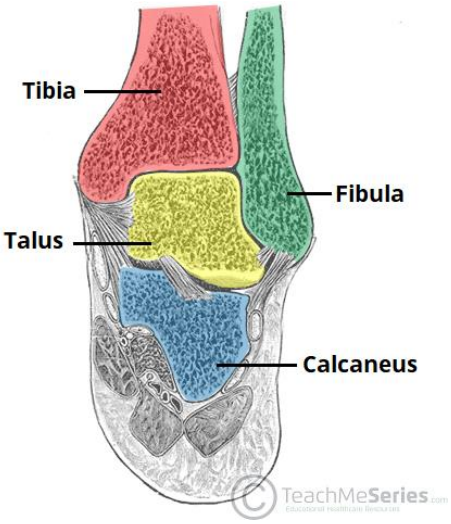


Figure 1: Posterior view of ankle joint

Effectively, this translates the problem from the medical to the mechanical domain and vice versa. This thesis will always work on the intersection of the two domains and the effects of a change in one domain must be measured or estimated in the other domain too.

Problem definition by client

The client has defined this problem as follows. In the coming years, fine force delivery by an assistive device to humans will be necessary to guide precise and personalized ankle rehabilitation. The current solution for force transmissions in lower limb exoskeleton design relies heavily on rigid interfaces and force transmissions in parallel to the talocrural joint (allowing for plantar- and dorsiflexion), disregarding motions caused by for example the invertors and evertors (motion in the subtalar joint). Furthermore, for so much as the actuations are linear, they are limited to one line of action, or in the most recent cases both plantar- and dorsiflexion over one degree of freedom following a generic design. Isolated 1 DoF plantarflexion/dorsiflexion support and non-personalized targeting can lead to deterioration of cartilage, misuse of muscle, and tendon tissue and bone structures when using this support.

The one degree of freedom simplification furthermore limits the targeting of specific muscle groups. To do so in the end, a 2DoF device must be developed, starting with a proof of principle. To follow the natural actuation of the body as much as possible, the project must be carried out using a Bowden cable actuation. To provide better alignment, personalisation is desired, and thus this design process needs to include personalized biomechanical data in an early stage.

Goal

Therefore, what is needed is a new type of 2 degree of freedom interface, that can be easily personalised to a subject and the rehabilitation process. A device must be designed that can transmit loading in parallel to musculotendon structures to induce tissue homeostasis, which is the perfect environment for tissue healing. This must be done while providing a solution that is lightweight and precise; a new type of 2 DoF interface. With the ankle being a very complex joint and the PF/DF axis not being perpendicular to the anteroposterior plane, it is time to move the research from 1DoF systems to 2DoF and try to solve the misalignment problems still existing in a lot of systems on the one hand and apply precise forces to target specific muscle groups on the other. This project will aim to contribute to that, providing a proof of principle that can be used in either a research setting, a rehabilitation setting or possibly longer-term use, while also providing insight in the joint forces caused by the orthosis. This thesis ultimately will provide insight in the feasibility of 2DoF ankle-foot orthoses of this type.

The apparatus has the potential to improve functionality and free movement in the required settings. Also, using a cable actuated 2DoF device targeting specific muscles should be possible with further research. While the device can in principle contain structural elements between the shank and the foot, it is not primarily intended as a weight-bearing device and thus serves a different purpose than for example a full weight-bearing exoskeleton.

While there is a lot of research into the effectiveness of on-the-market devices, very little research is being done in the longer-term effects of (ankle) orthosis wear. Given the timeframe of the thesis, this is explicitly not researched.

Design assignment

The design assignment is to improve onto the state of the art in such a way that the total system allows for functional, personalized, precise plantarflexion and inversion/eversion support, allowing for small (force or orientation) corrections on the orthosis. This must be done within the boundaries of the requirements, where the joint torques are calculated based on model input, either through a MS model or alternative methods.

While there are alternative routes of solving this problem, the one described above is chosen. It was already explained that pharmaceutical or surgical solutions are possible but are not considered a decent first step. They are altogether discarded from here on out as this thesis is biomechanical in nature and tries to reduce the need for surgery or pharmaceutical intervention. Also, the solution of a passive device is discarded for multiple reasons:

1) While a passive device can be adapted to a patient and the environment, there are no adaptive possibilities once it is implemented apart from physically changing the device.
2) A passive device requires actuators (e.g.) springs to be installed distally on the body. This causes an increase in distal weight and does not satisfy the requirement that the Bowden cable actuation must be used.

However, in the future, the passive spring could potentially be used to assist in dorsiflexion. A passive dorsiflexion spring force could be overcome with the active plantarflexion force and when the active force is removed, the spring would dorsiflex the foot.

System subdivision

To make the total design assignment more explicit, the system is divided in four different interfaces. These interfaces link the musculoskeletal system to the actuation system, which is visualised in **Error! Reference source not found.**, adapted from [14].

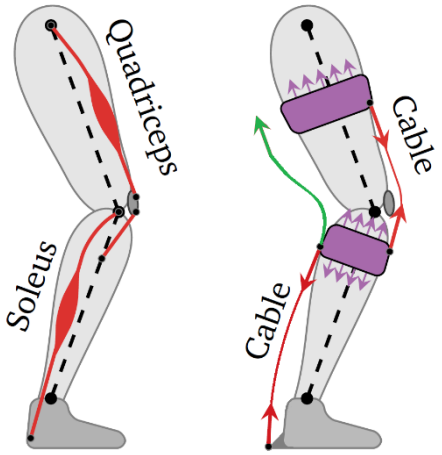


Figure 2: Schematic representation of the system.

On the left, the normal functioning of a body is shown, with two principal muscles schematically depicted. On the right, a cable exosuit is shown. Note that the elements at the thigh are ignored for now, but in principle this reasoning would also apply to knee actuation. The purple patch at the shank is the musculoskeletal attachment where the actuation force (green arrow) connects to the Bowden cable actuation (red arrow). This force causes the purple patch and the heel to move towards each other, enabling plantarflexion. This results in two interfaces with the body and the force transmission in between them, yielding the three system elements referred to in this thesis. Analogous to this, inversion and eversion can be actuated (not depicted here).

As the goal of this project is precise force delivery, which leads to multi degree of freedom actuation, the green force transmission cannot introduce any more constraints on the degrees of freedom. Since ankle and subtalar joints effectively constrain the three translational degrees of freedom in the joint, any added constraint will necessarily constrain rotation, which goes directly against the design goal. This eliminates the option to add joints on the interface and leaves the option for compliant materials. The connection can thus either be flexible or semi-stiff.

Stakeholders

There are several stakeholders with interest in this project. In the specifications above, the client was already mentioned. The analysis below provided a more exhaustive list of stakeholders, which is based on the lifecycle of the product. The different life phases are not necessarily connected to the project phases, nor are they necessarily chronologic. A stakeholder emerging in a certain phase necessarily has an interest in any of the succeeding phases. The stakeholders are scored on priority between 1 (low) and 3 (high) and influence between 1 (little) and 3 (much) to indicate the importance of that stakeholder. Priority is a measure of importance of the stakeholder from the designer's point of view, influence states to what extent the stakeholder can steer the course of the project.

Stakeholders					
Phase	**ID**	**Name**	**Function within project**	**Priority**	**Influence**
Idea	1	Guillaume Durandau	Client & user (in theory)	3	2
Design	2	Ali Sadeghi	UT supervisor	2	3
	3	Herman van der Kooij	Chair of thesis defence	3	3
	4	Biomechanical Engineering dpt.	Owner of the device, research group at UT.	2	1
	5	University of Twente	Employs client, technicians and researchers.	2	1
Development	6	Maura Eveld	Control designer	2	3
	7	Mahdi Nabipour	Control designer	2	3
Production	8	Technicians/ researchers	Practical work on the device	2	2
Testing	9	Test subjects	Enable product tuning	3	1
	10	Researchers	Carry out the tuning	2	2
Use	11	Ethics committee	Must approve any tests containing human subjects from here on out.	2	3
	12	Test subjects/patients	Depending on use case, these people are the final users of the device.	3	1
Maintenance	13	Technicians/ researchers	Practical work on the device.	2	2

Note that the phases design, development, production, and testing may iterate between different use cases.

ID	Stakeholder	Boundary conditions	Influence
0	Author	The author has one principal boundary, which is carrying out the project with a sufficient grade by making sure all stakeholder needs are met.	This stakeholder has by far the greatest influence on the project but must take all other stakeholders into account.
1	Guillaume Durandau	The client wants a workable system that can be used in a research environment.	The client steers the design in many ways and has thus a large influence.
2	Ali Sadeghi	The supervisor wants to guide author properly through the process and wants a technically workable solution.	The supervisor exerts large influence on the project, mainly on the design.
3	Herman van der Kooij	The chair wants to make sure the project adds to the state of the art such that the author's work is ample to graduate.	The chair has a lot of influence on the project, but mainly has influence on the process.
4, 5	University of Twente	Wants the device to be integrated properly in existing	There is some minor influence, but this is mainly exerted through stakeholders 1 – 3.

		systems and to be usable in the lab.	
6, 7	Control designers	The control designers define boundary conditions on the sensors to be used.	Through requirements on sensing, the control designers have a lot of influence on the design.
8, 10, 13	Researchers	The researchers (or whoever will ultimately work with the device) will want a device that can be tuned to a wide range of cases.	The researchers are considered, but ultimately, they will always make do with the given possibilities.
9, 12	Test subjects	If the subjects are patients, they want the device to help them in rehabilitation and care secondary about comfort. Test subjects will be more focussed on comfort.	The patients are the primary focus of the design, but they have no direct influence anywhere.
11	Ethics committee	All experiments with human test subjects must pass this stakeholder, whose main concern is safety of the hardware.	This hard boundary condition makes that this stakeholder has a lot of influence on the system.

Use cases

For the final product a set of use cases must be defined to properly assess in what environments and under what conditions the device must operate. In first instance, the goal is to use the device in a lab setting for rehabilitation purposes. This means that the device can be made tethered for now and one can assume IMU's are available, a pressure sensitive treadmill can be used, and a motion capture system is present.

The device can therefore ultimately be used in rehabilitation, but probably first will be used in a research setting to test effectiveness. Some of the experiments that could be carried out include:

- Testing effectiveness in providing plantar-/dorsiflexion and subtalar flexion torque to correct for (mediolateral) perturbations during normal walking.
- Testing effectiveness in allowing natural subtalar motion during normal walking.
- Testing effectiveness in providing plantar-/dorsiflexion and subtalar flexion torque to correct for (mediolateral) perturbations during everyday activities such as stairs climbing, jumping and running.
- Testing effectiveness in running a prescribed motion in the context of physical therapy.
- Finding the limit of reduction of the metabolic cost of walking of a (healthy) subject.

These experiments all are centred around a general set of use cases that range from donning and doffing to perturbation support. A list of use cases could be containing (but not limited to):

- Donning;
- Doffing;
- Standing;
- Stand to sit;
- Sit to stand;
- Normal walking (at certain speeds);
- Kneeling during stand;
- Bending during stand;
- Running (at certain speeds);
- Walking stairs;
- Walking terrains;
- Walking.

If the device were to work properly in all these use cases, a wide range of rehabilitations could be facilitated. If the actuation were to be made untethered in future versions of the device, it could even be used in daily life.

Requirements & boundary conditions

These goals must be realized within a certain framework. This framework is set by a list of requirements that relate to functionality (use) and design. The requirements are based on the stakeholders and on literature findings. They can be either fixed, meaning that whatever the design, they must be fulfilled, or variable, which means they are optimized in the design.

ID		Description
1. Use		
1.1		**Intended use**
1.1-010	f	The device shall be used in a revalidation environment
1.1-020	f	The device shall not be used for continuous wearing outside of rehabilitation facilities
1.1-030	f	The device shall actuate the human ankle (and subtalar) joint.
1.1-040	f	The device shall be usable on any patient going through the rehabilitation process.
1.1-050	f	The device shall be usable on any adult within the height range of [150, 190] cm.
1.1-060	f	If possible, the device shall be usable on paediatric patients.
1.2		**Personalization**
1.2-010	f	The total system (including actuation) must enable precise force delivery in personalized capacity.
1.4		**Users**
1.4-010	f	The device shall be applied by instructed personnel only.
2. Design requirements		
2.1		**General device design requirements**
2.1-010	v	The device shall be produced with available materials and methods at the UT facilities.
2.2		**General interface requirements**
2.2-010	v	The device shall emulate & enable natural motion of the body during normal walking as much as possible.
2.2-020	v	The device shall hinder natural tissue motion of the body during normal walking as little as possible.
2.3		**General system design**
2.3-010	v	The distal weight shall be minimized to reduce fatigue and extra needed muscle action.
2.3-020	v	The device shall be able to accurately measure applied forces (either directly from moments or forces, or any other sensor input).
2.3-030	f	The device shall be able to accurately measure position (expressed in terms the ankle DoF's).
2.3-040	v	The device shall be able to measure position with as little external sensors as possible.
2.3-050	f	The net joint reaction forces, muscle forces, tendon forces and any other bodily forces should be within acceptable ranges.
2.4		**MS - shank connection**
2.4-010	f	The device shall provide a stable connection to the shank such that precise force delivery is enabled (5 - 10% error max.)
2.4-020	v	The shear forces shall be kept within acceptable ranges such that ultimately the shank connection doesn't slip down during normal operation. This shall be achieved while still allowing for 2.2-020 and thus not with increased normal forces.
2.5		**MS - foot connection**
2.5-010	f	The device shall provide a stable connection to the foot such that precise force delivery is enabled (5 - 10% error max.)
2.6		**Foot - shank connection**
2.6-010	f	The device shall be able to support the full range of motion during normal walking for the ankle and subtalar joints.
2.6-020	f	The device shall be able to actuate under joint rotational speeds during normal walking.
2.6-030	v	The device shall provide an 'area of flexibility' around the joints to enable multi-DoF actuation.
2.7		**Actuation interface**
2.7-010	f	The device shall be connected to the current power supply and actuation chain in the lab (Bowden cable actuation).

ID		Description
2.7-020	v	The device interface with the current power supply and actuation chain in the lab shall be easily engaged and disengaged.
2.7-030	f	The device shall be able to accurately assist up to 50% of the bodyweight normalized ankle and subtalar torques during normal walking.
2.7-040	f	The device shall be able to deliver the assistance over multiple lines of action
2.7-050	f	The device shall not be able to provide more torque or force than the body could provide over that line of action.

2. Synthesis Phase

The synthesis phase aims to realize the previously defined functions. Normally, this synthesis phase consists of three sub-phases, which can be summarized as:

- Synthesis I: Wild variation of ideas
- Synthesis II: Concepts with i.e. size, materials, mode of operations etc.
- Synthesis III: Final, realistic, and workable solution. Considerations are made on production, cost, strength & materials. Finally, there might be a set of testing and experimentation.

The workflow of the system can be summarized as in Figure 3, a diagram representation, and Figure 4, the control workflow as developed by Jeroen Meijners [10].

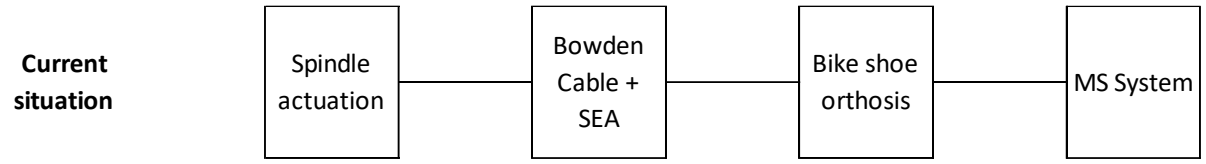


Figure 3: System workflow

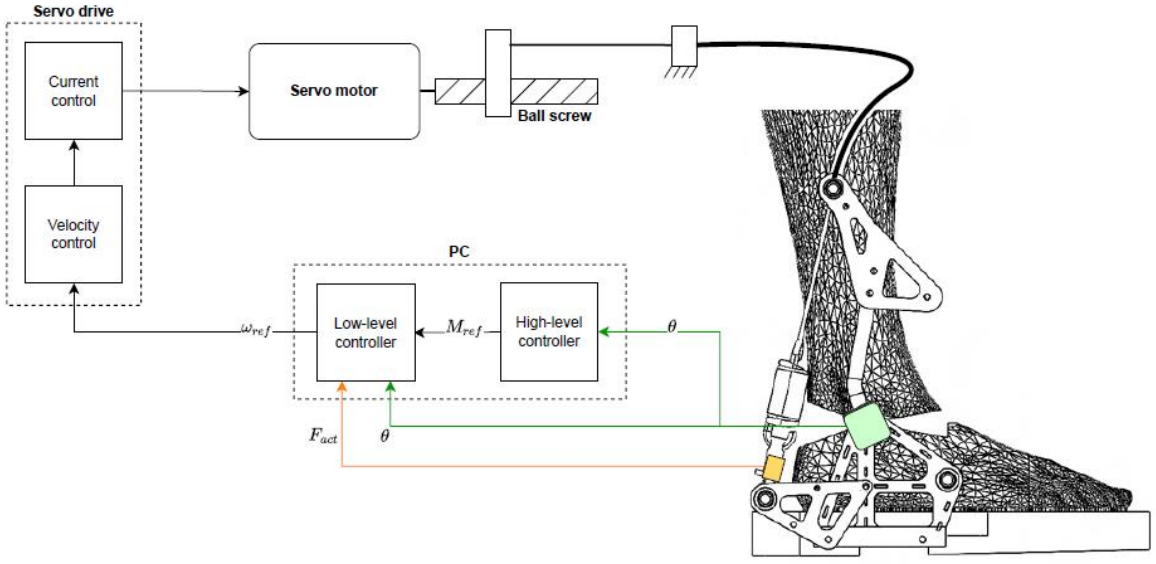


Figure 4: Control workflow

The device that is attached around the ankle must be adapted to be able to accommodate the subtalar correction force function. Furthermore, in a virtual model, the joint effects must be calculated, and the applied forces must be adapted based on this. For now, this can be done once per patient, but ideally this is done real time by the controller. This is summarized in the diagram in Figure 5.

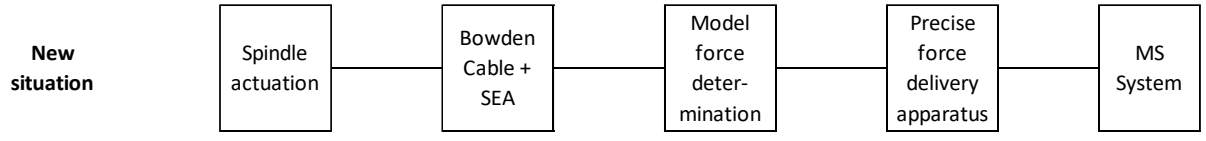


Figure 5: Desired workflow

State of the art

The state of the art was already briefly touched upon in the literature review. This section aims to go more in-depth on some of the prime examples of the subsets the state of the art of linearly actuated systems, which is defined now as:

- Bowden cable actuated one degree of freedom systems (1DoF Bowden systems).
- Non-Bowden cable actuated one degree of freedom systems (1DoF non-Bowden systems).
- Non-Bowden cable actuated two degree of freedom systems (2DoF non-Bowden systems).

The obvious missing category is Bowden cable actuated two degree of freedom systems (2DoF Bowden systems), which are currently non-existent. These categories are discussed briefly in the paragraphs below. These systems can all either contain a rigid connection between the shank and the foot or can be built without a rigid connection. Note that passive and unpowered systems are explicitly not considered, nor is a distinction made between tethered and untethered systems.

1DoF Bowden systems – Rigid connections

The starting point of this design is the state of the art of the J-LO design at the University of Twente as described earlier, which is an example of a device with a rigid connection between shank and foot. The device is lightweight (especially distally), is mechanically robust as the weak link is the Bowden cable, which can be easily replaced if broken. However, there are some major downsides to this principle. As Jeroen Meijners already mentions in his thesis [10], there can be some definite improvements in comfort, both of the shoe and of the shin attachment. Furthermore, as the device is relatively stiff, there is a risk of joint misalignment. Also, the nature of the device requires that for both directions in a degree of freedom, two tethered actuators are needed. No recommendations are made on dorsiflexion support or correction forces. The biggest downside in the context of the current project is that the nature of the design doesn't allow for a large range of anchor point placement and thus doesn't allow proper joint alignment. The J-LO design can be seen in Figure 6 [10].Note that the far-left image depicts an earlier iteration of the J-LO device, with a hydraulic actuation.

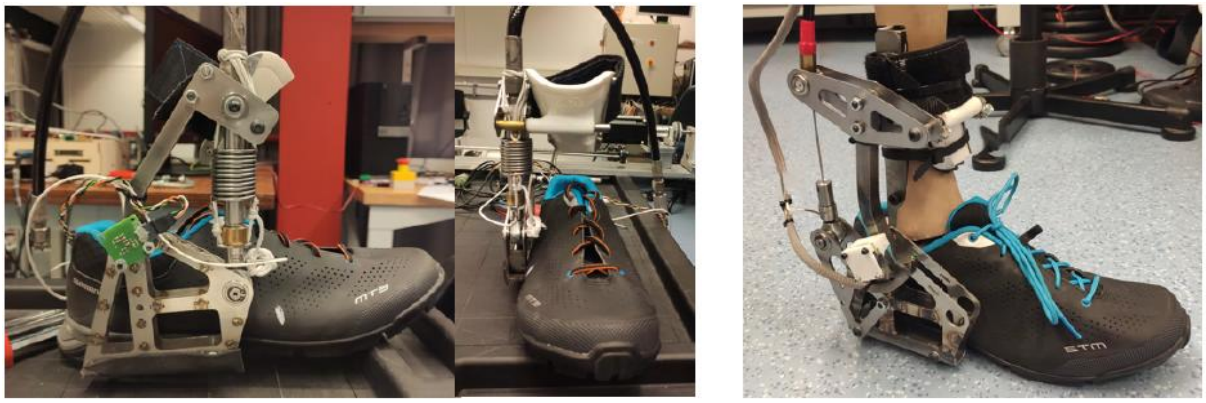


Figure 6: J-Lo with hydraulic and cable actuation

Another example of a device with a rigid connection is the design by Witte and their colleagues, with properties like the J-LO design, which can be seen in **Error! Reference source not found.** [11].

1DoF Bowden systems – Flexible connections

The second type of 1DoF Bowden systems is a prime example of an exosuit. It has no inherent stability, but fully utilizes the body's own MS system to induce plantarflexion support. It is therefore not suited for patients with severe muscular atrophy, but more suited towards patient with minor atrophy (either in rehabilitation or for permanent use). There have been some developments in this, but there is one device that is by far the most functional device on the market: the ReWalk Restore, which uses the apparatus developed by Conor Walsh and his team. The AFO can support both plantarflexion and dorsiflexion and uses just an insole in the patient's shoe to transfer forces to the foot. The device does however require quite an extensive apparatus on the shank, increasing distal weight and volume, as can be seen in Figure 8 [12].

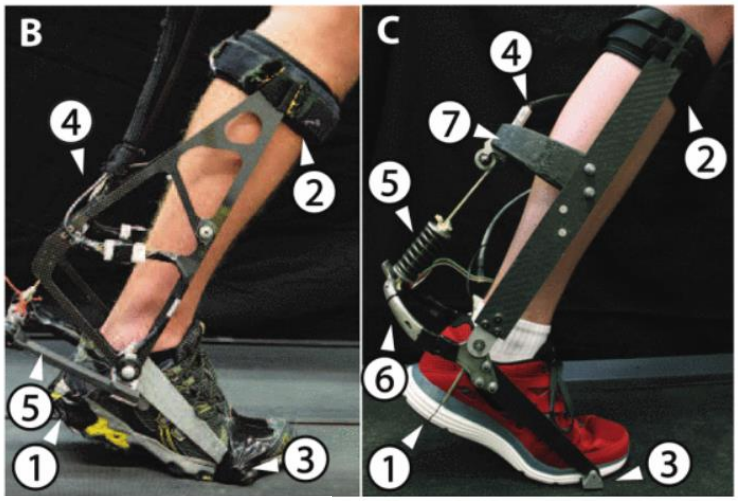


Figure 7: AFO design by Witte

Figure 8: ReWalk Restore AFO

1DoF non-Bowden systems

The systems of the 1DoF non-Bowden type can be actuated by a linear motor, either using hydraulics, pneumatics, or electric motors. They all need rigid connections to actuate against. An example used to treat drop-foot in post-stroke rehabilitation is the device by Blaya [13], which can be seen in Figure 11, or the Achilles by the University of Twente in Figure 10 [14].

It must be noted that while these systems are often very sturdy, they do come with the downside of higher distal mass in comparison to Bowden systems, as the linear actuators add quite some mass.

2DoF non-Bowden systems

The field of 2DoF AFO's is relatively undiscovered. As control schemes for full gait support 1DoF systems has only been up to desirable standards in the last decade, not much research has expanded into 2DoF systems. At the University of Twente, one of the devices developed is the 2DoF, hydraulicly actuated AFO by Grootens [15], seen in Figure 9. This system fully allows for subtalar motion, but the relatively rigid design caused subjects to still report discomfort likely caused by misalignment.

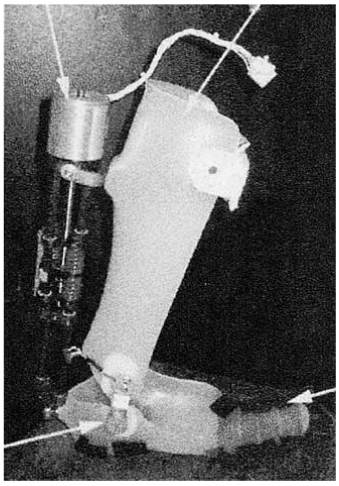


Figure 11: Blaya & Herr AFO

Figure 10: Achilles AFO

Figure 9: Grootens 2DoF AFO

Quantitative measures of state of the art

The results of the literature review can be found in the respective document (Appendix A of the main thesis), but the results are quantified in

	WEIGHT [G]	PF ROM [°]	DF ROM [°]	PF TORQUE [NM]
MEAN	1288	24.0	14.6	62.1
STD	879	8.9	4.6	44.1

Table 1. Note that the standard deviation is incredibly high. This can be explained by the fact that there is a large variation in AFO types.

	WEIGHT [G]	PF ROM [°]	DF ROM [°]	PF TORQUE [NM]
MEAN	1288	24.0	14.6	62.1
STD	879	8.9	4.6	44.1

Table 1: Mean and standard deviation of all devices

Correcting the data using only exosuits with Bowden cables yields very different results, with the design from Witte et. al [11] being the obvious standout, delivering well over 120 Nm of torque. The conclusion drawn here is that using stable steel structures and smart design the 120 – 150 Nm of torque is achievable, but using more delicate structures the torque range is under 20 Nm. This is summarized in

	WEIGHT [G]	PF ROM [°]	DF ROM [°]	PF TORQUE [NM]
MEAN	852	24.2	16.6	65.4
STD WITH WITTE	522	5.4	4.2	58.0
MEAN	851	20.3	14.3	19
STD WITHOUT WITTE	974	3.3	4.0	7.8

Table 2.

	WEIGHT [G]	PF ROM [°]	DF ROM [°]	PF TORQUE [NM]
MEAN	852	24.2	16.6	65.4
STD WITH WITTE	522	5.4	4.2	58.0
MEAN	851	20.3	14.3	19
STD WITHOUT WITTE	974	3.3	4.0	7.8

Table 2: Mean and standard deviation of Bowden devices

Synthesis I

Synthesis I is the process of setting up as much as possible conceptual directions and distilling those into concepts. Motion of the foot can be reduced to two degrees of ankle freedom (plantarflexion and dorsiflexion, and inversion and eversion) and one degree of foot freedom (external and internal rotation of the foot), as can be seen in Figure 12 [15].

As the goal is to actuate the two degrees of ankle freedom, the starting point of the design is what joint must be used by the system at the ankle joint, what range of motion this joint must have and how the actuation forces are translated between device, actuation, and body. This means that the most important aspects of synthesis I are:

- Joint type.
- Range of motion.
- Force transmission.

The joint type and actuation are summarized in a morphological chart. This chart will be weighed against each other on range of motion and force transmission. Furthermore, bulkiness is considered. Material use and bulkiness at the medial side of the ankle should be kept to a minimum as the ankles pass each other quite closely during walking. This does vary per person and this measure will be quantified in synthesis III.

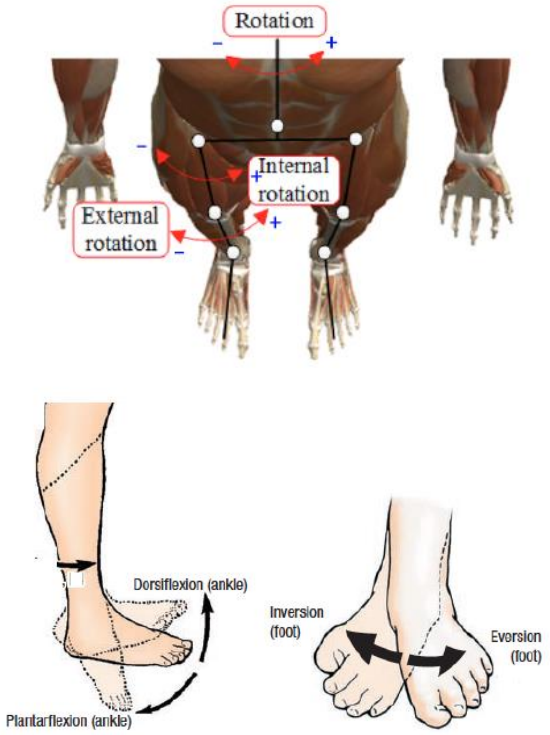


Figure 12: Ankle and foot DoF

Joint types

All joint types are summarized in the morphological chart. This paragraph briefly describes the logic behind the elements of the chart. If one wants to actuate all degrees of freedom (3) at the location of the ankle, a spherical joint type is necessary since all DoF's are rotational. As there is an ankle present at the centre of the joint, this must then be realized with a 3-axis joint (concept F). Two of the axes must be at the ankle, while the third axis can be either at the ankle or higher up at the leg. Note that to realize this a rotation structure must be built around the leg, effectively creating a pin joint with the axial direction on the proximal/distal axis of the lower leg. The tibia and fibula structure of the leg rotates with respect to itself at the lower leg and thus the exosuit could actuate this DoF there. This will create such a complex design, while the requirements don't ask for 3 degrees of freedom. For further material on this exoskeleton joint, refer to chapter 6.3 of [15]. Thus, this joint type is explicitly disregarded from here on out.

Removing that DoF gets a 2-axis joint, also known as a universal joint or U-joint (concept E). This is very similar to the design of Martijn Grootens with his pneumatic AFO. During normal locomotion, the subtalar range of motion is much smaller than the plantar- dorsiflexion motion. Therefore, other 2DoF options can be considered, such as compliance in the joint, or a mechanical solution. Compliance can be achieved at a larger scale (joint D), or a smaller scale (effectively making the axis suspension compliant, concept B). A mechanical solution would include the same motion, but no spring properties. The motion would instead be bound by a rail or something comparable (concept C). The last option is utilizing the ankle joint itself as the joint. This gives no restriction on RoM and requires no structural design (concept A).

Morphological chart I

The joint types described in the previous section are summarised in the first morphological chart, as can be seen in Figure 13.

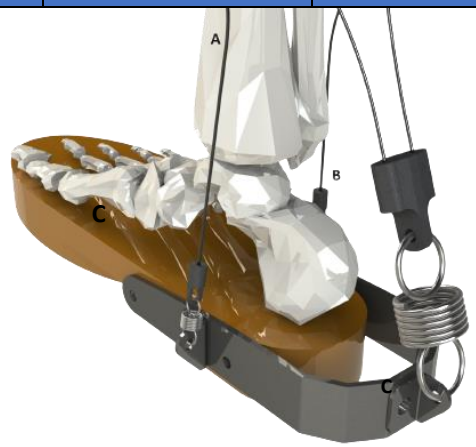
Specification	Option 1	Option 2	Option 3	Option 4	Option 5	Option 6
Actuation	3 Bowden cables A – Inversion cable B – Eversion cable C – Split plantarflexion cables					
Joint type	Joint type A: No connection 	Joint type B: Flexible hinge 	Joint type C: Sliding hinge 	Joint type D: Compliant connection 	Joint type E: Universal joint type 	Joint type F: 3-axis joint 

Figure 13: Morphological chart for Synthesis I

Joint type C, the sliding hinge, works with a slider over the shank bracket. The slider follows the black line in Figure 14a. But with inversion motion, the slider also slightly moves in the x (red) direction. Furthermore, the black line is not in the z (blue) direction, but in a plane combined of the z and y (green) direction. Joint types E and F only differ in that another axis of rotation is added in joint F. The purple axis is in line with the lower leg and allows for full 3D motion. Note that this has no functional advantage over type E.

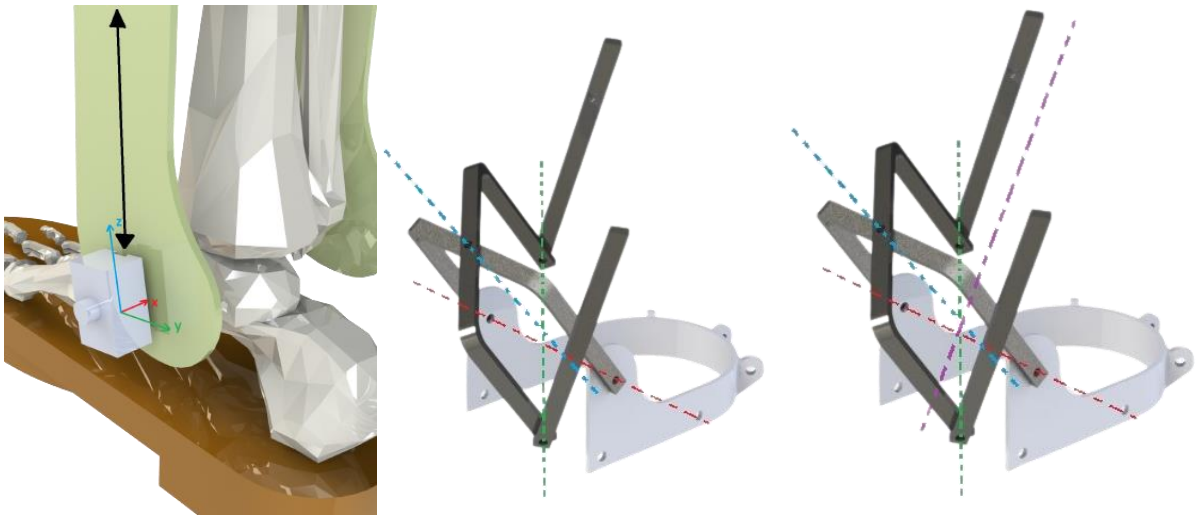


Figure 14: Extra details of the joint types. Left: joint type C, middle: joint type E, right: joint type F.

Force transmission test

The force transmission test conducted aimed to get insight in the limits of force that can be applied to the shank. The theoretical basis for this is Figure 15 [16]. Now, with the area of the shank connection being 17540mm^2, theoretically the force could be more than 500kg before the pressure threshold of 300kPa is reached. However, the connection is never flush, and the edge will bury in the skin, but the exact effects of this are difficult to determine in theory. Therefore, a test is conducted to see at what levels the bracket starts to slip and cause discomfort under several circumstances. The goal of this test is to determine what the force limit is that can be translated onto the bracket. The result of this is in the range of 150 – 250 N. This means that joint A would likely not be able to accommodate larger support torques without a way to decrease the transmission. Please note that this force test was set up in the context of synthesis I and a more extensive test will be conducted as a verification later.

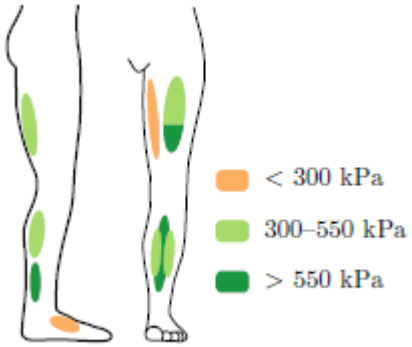


Figure 15: Pain pressure thresholds

Comparison & weighing

When weighing the joint types on a range of 1 – 5 (bad – good), on the three measures mentioned above, the results in Figure 16 and Table 3 are yielded. The explanation for the scoring can be found in the Appendix.

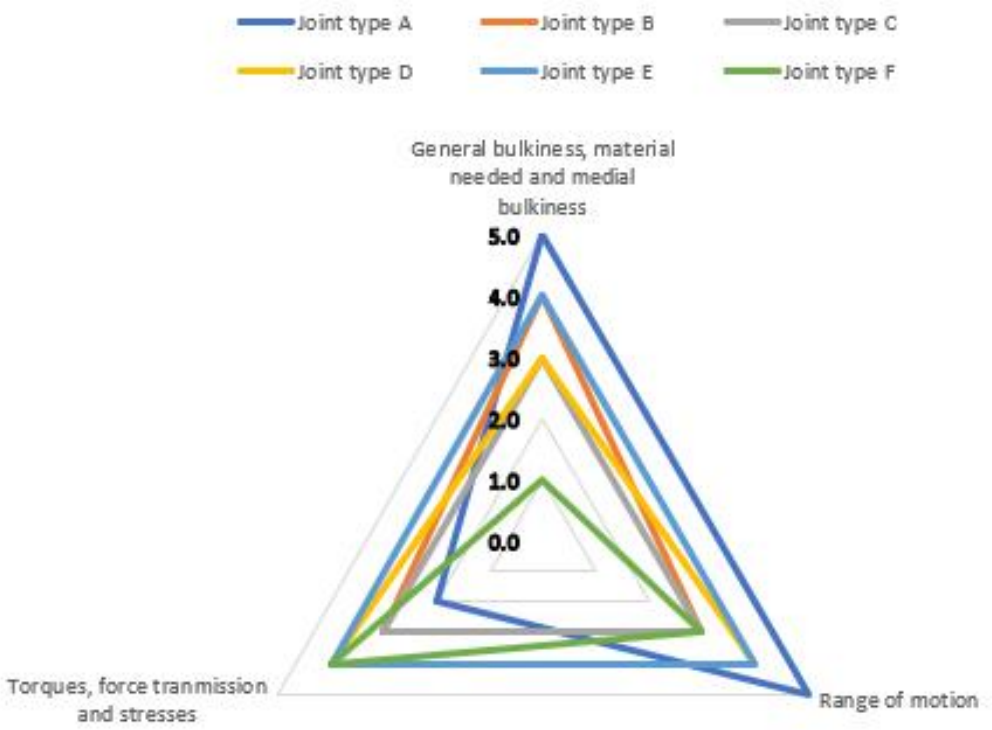


Figure 16: Radar graph morphological chart 1

	Joint A	Joint B	Joint C	Joint D	Joint E	Joint F
Total score	12	10	9	11	12	8

Table 3: Total scores morphological chart 1

From this, the conclusion can be drawn that joint types A, D and E score best. These are also the simples designs mechanically. After discussion with the client, the lower force transmissions inherent to joints A and B were no problem so, all three joints can be used in the concept development.

Synthesis II

Synthesis II entails the process of developing several workable concepts from the preconcepts emerging from synthesis I. The concepts are more detailed than in synthesis I and will be judged more extensively. The base for the comparison in synthesis II are the requirements, as developed earlier in this document. Some of these requirements are fixed and should be met by any of the concepts, while others are variable. The degree to which the concepts meet the variable requirements can be used as a base of concept comparison. The details of this are discussed later, as first the concepts must be defined through the morphological chart.

Morphological chart II

The first morphological chart yielded that joints A, D and E scored best (respectively 12, 11, and 12 points out of 15). These joints are thus considered in this morphological chart at specification 1. For actuation (specification 2), only a three-way actuation is considered. It is possible that one or more of these actuation lines is passive or semi-active.

The third specification deals with the shank connection. From the results of the force tests the conclusion must be drawn that one of the main challenges is going to be to properly transfer the forces needed for locomotion onto the musculoskeletal system without applying too much normal or shear pressures. Joint type A is influenced the most by this. Therefore, joint type A is evaluated at two different shank connection types:

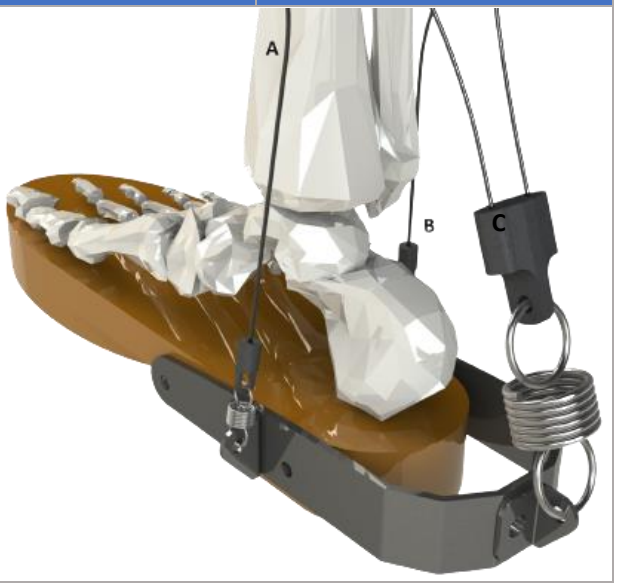
1. Shank connection at distal part of lower leg, anterior rigidity.
2. Shank connection at distal part of lower leg, anterior rigidity, with connection to upper leg to distribute more shear pressure.

At this stage it is assumed that joints D and E don't need the extra connection to the upper leg, so they are only evaluated at shank type 1. This can of course still be considered later in the design process.

The fourth specification is on the foot connection. All concepts use the bike shoe with the two pins at the sole as a base. The exact design of the bracket connecting the is part of synthesis III, so for now just one option is assumed.

Any other design choices are assumed to be part of synthesis III as all concepts can for example use a variety of sensors and fixating on this at this stage of the design would complicate the process unnecessarily.

All of this is summarised in morphological chart II, to be seen in Figure 17.

Specifi-cation	Option 1	Option 2	Option 3
Actuation	3 Bowden cables A – Inversion cable B – Eversion cable C – Split plantarflexion cables		

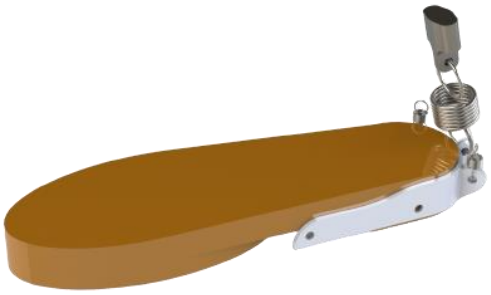
Specifi-cation	Option 1	Option 2	Option 3
Joint type	Joint type A: No connection 	Joint type D: Compliant connection 	Joint type E: Universal joint type 
Shank connection	Distal, anterior rigidity 	Distal, anterior rigidity + upper leg  	
Foot connection	Bracket connection 		

Figure 17: Morphological chart with the shank bracket in green, the proximal band in black, the foot bracket in silver metal and the shoe in brown.

Concepts description

As mentioned, there are 4 concepts to be defined.

Concept 1: Joint A, no upper leg band

This concept direction has no direct connection between the foot and the shank besides the existing ankle joint. The interfaces must thus be tightly connected to the leg and the foot, and all the force exerted by the system is fully translated to the musculoskeletal system as joint reaction forces. The torque in the system can be tracked by knowing the moment arm at a given position and measuring the force in the cable through a load cell.

The advantages include:

- Light distal weight: This concept has minimal parts and is relatively simple. The distal weight is thus minimal.
- No constraints on DoF: The lack of added joints means that the system imposes no constrains on the freedom of motion of the limbs and thus allows for any DoF.

The main disadvantages are:

- The lack of added stiffness makes it difficult to find two points whose position can be measured accurately enough to track the plantarflexion or dorsiflexion angle with an encoder (or other sensors). The subtalar flexion is thus also difficult to determine. The system would thus entirely rely on motion capture data or IMU's for position.
- The lack of stiffness means higher joint loads.
- The lack of stiffness means higher needed shear forces between the interfaces at mainly the leg.
- The lack of stiffness doesn't allow for mechanical limits to angular displacement, so the safety must be built in the actuation.

Figure 18: Concept 1

Concept 2: Joint A, upper leg band

This concept direction has no direct connection between the foot and the shank besides the existing ankle joint. The interfaces must thus be tightly connected to the leg and the foot, and all the force exerted by the system is fully translated to the musculoskeletal system as joint reaction forces. The shank connection is stabilized by a leg band. This leg band will be flexible as the upper leg does not have a low mechanical compliance. The torque in the system can be tracked by knowing the moment arm at a given position and measuring the force in the cable through a load cell.

The advantages include:

- Light distal weight: This concept has minimal parts and is relatively simple. The distal weight is thus minimal.
- No constraints on DoF: The lack of added joints means that the system imposes no constrains on the freedom of motion of the limbs and thus allows for any DoF.

The main disadvantages are:

- The lack of added stiffness makes it difficult to find two points whose position can be measured accurately enough to track the plantarflexion or dorsiflexion angle with an encoder (or other sensors). The subtalar flexion is thus also difficult to determine. The system would thus entirely rely on motion capture data or IMU's for position.
- The lack of stiffness means higher joint loads.
- The lack of stiffness means higher needed shear forces between the interfaces at mainly the leg. However, a part of this will be diminished by the leg band.
- The lack of stiffness doesn't allow for mechanical limits to angular displacement, so the safety has to be built in the actuation.

Figure 19: Concept 2

Concept 3: Joint D, no upper leg band

An alternative to concepts 1 and 2, which don't use a connection between shank and foot, could be the use of compliant materials. Tuning material and geometrical properties provide a connection that allows for plantarflexion and dorsiflexion through flexion of the material. Subtalar flexion requires lengthening and shortening of the connections, which can be achieved through some sort of folding motion. The compliant connection can vary from traditional springs to compliant (synthetic) plastics. This concept allows for a wide range of position sensing possibilities, ranging from traditional encoders to using conductive properties of the material to sense rotation.

The advantages include:

- Doesn't require traditional encoders for measuring position.
- Doesn't suffer from wear like concepts B and C.
- Mechanical simplicity & lack of bulkiness.
- Can limit shear forces on shank while still maintaining flexibility.
- Passive spring properties might prove practical in control and/or design.

The main disadvantages are:

- The proper balance on geometry and material properties might be difficult to achieve given all constraints, and for sure creates complexity in the design.
- Fatigue could impose a limit.
- Reaction forces caused by inherent flexibility of compliant material could be of negative influence on the MS system.
- Mechanical limits on motion might be difficult to impose.

Figure 20: Concept 3

Concept 4: Joint E, no upper leg band

Whereas concept 3 tries to add stiffness problem by compliant materials, thus allowing for both translation and rotation of the joint, this concept aims to reduce the degrees of freedom to two rotational ones. Two U-brackets, joined in their centres by a pin joint, attach to pin joints at the foot and the shank. This universal joint type motion allows for plantarflexion, dorsiflexion and subtalar motion. Utilizing three cables, one could actuate plantarflexion and subtalar motion. Dorsiflexion would require another actuator. A variety of sensors can be used, as the components only rotate with respect to each other and thus have little play.

The advantages include:

- Barely any forces translated to the musculoskeletal system due to the rigid system.
- Very stable motion and accurate measurements due to rigidity.

The main disadvantages are:

- Mechanical complexity.
- Large distal weight.

Figure 21: Concept 4

Requirements ranking and weighing criteria

Some of the variable requirements are used to compare the concepts. Not all variable requirements are suitable for this, and they are often too specific. Therefore, from the requirements a set of weighing criteria is devised. These criteria are weighed against each other to determine their relative importance. This criteria weighing can be seen in Figure 22. The ID refers to the requirement the criterium is linked to.

Ranking score matrix												
ID	Criteria	2.1-020	2.2-010	2.2-020	2.3-010	2.3-020	2.3-040	2.4-020	2.6-030	2.7-020	**Total**	**Weight**
2.1-020	Materials & production		0	0	0	0	0	0	0	1	1	**1**
2.2-010	Natural body motion	1		0	0	0	0	0	1	1	3	**2**
2.2-020	Natural tissue motion	1	1		1	0	0	1	0	1	5	**2**
2.3-010	Distal weight	1	1	0		0	0	0	0	1	3	**2**
2.3-020	Force measurement	1	1	1	1		0	0	0	1	5	**2**
2.3-040	Position measurement	1	1	1	1	1		1	1	1	8	**3**
2.4-020	Shank shear forces	1	1	0	1	1	0		0	1	5	**2**
2.6-030	Flexibility around joint	1	0	1	1	1	0	1		1	6	**3**
2.7-020	Donn/doff	0	0	0	0	0	0	0	0		0	**1**

Figure 22: Weighing criteria ranking matrix

Comparison and weighing

The concepts are weighed on the criteria on a scale of 1 – 5 (very bad, bad, average, good, very good). These results van be seen in Figure 23 and are visualized in the radar plot in Figure 24.

Non-weighted table					
Variable requirements		**Score**			
ID	**Criteria**	**Concept 1**	**Concept 2**	**Concept 3**	**Concept 4**
2.1-020	Materials & production	4	4	4	4
2.2-010	Natural body motion	4	4	3	4
2.2-020	Natural tissue motion	3	3	4	4
2.3-010	Distal weight	5	5	3	2
2.3-020	Force measurement	5	5	5	5
2.3-030	Position measurement	3	3	3	4
2.4-020	Shank shear forces	2	3	4	5
2.6-030	Flexibility around joint	5	5	4	4
2.7-020	Donn/doff	4	4	3	3
	Total	35	36	33	35

Figure 23: Non-weighed table comparison

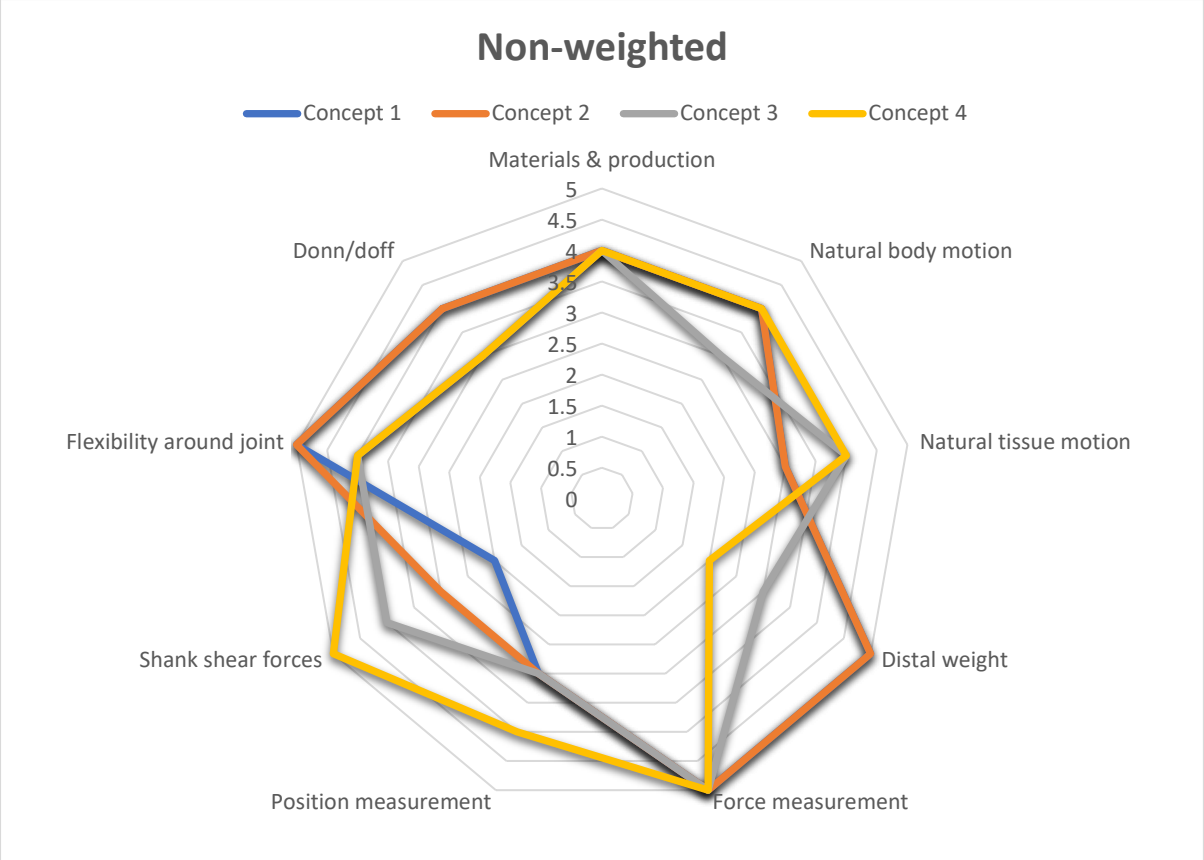


Figure 24: Spider plot of non-weighted comparison

Concept 2 leads here, with concept 3 being the worst off. If the weights are added, the results of Figure 25 and Figure 26 are yielded.

Weighted table						
Variable requirements		**Score**				
ID	**Criteria**	**Concept 1**	**Concept 2**	**Concept 3**	**Concept 4**	**Max values**
2.1-020	Materials & production	4	4	4	4	5
2.2-010	Natural body motion	8	8	6	8	10
2.2-020	Natural tissue motion	6	6	8	8	10
2.3-010	Distal weight	10	10	6	4	10
2.3-020	Force measurement	10	10	10	10	10
2.3-030	Position measurement	9	9	9	12	15
2.4-020	Shank shear forces	4	6	8	10	10
2.6-030	Flexibility around joint	15	15	12	12	15
2.7-020	Donn/doff	4	4	3	3	5
	Total	70	72	66	71	90

Figure 25: Weighted table comparison

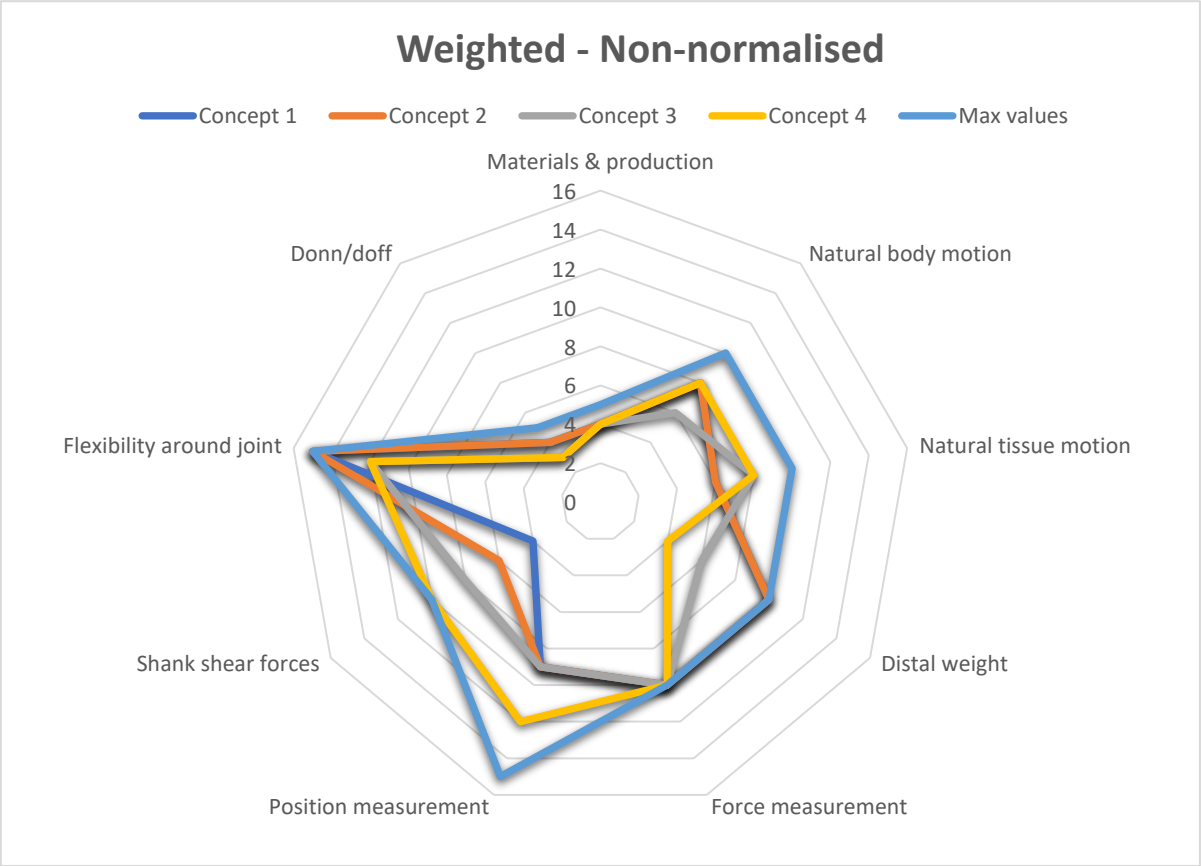


Figure 26: Spider plot of weighted comparison

Synthesis III

Synthesis III contains the detailing of the concept chosen at the end of synthesis II. It will start from the required motions and forces and translate that to a detailed solution able to withstand those forces and provide that motion. This entails for example calculations on the mechanical properties of the brackets, but also looks at the necessary friction forces to transfer those forces to the musculoskeletal system. The effects of these forces, both at the shank and the foot, are not discussed here, as they are not part of the factual design. For those, one is referred to the biomechanical verifications of the design. Note that any and all calculations in this chapter are based on the data of Huawei Wang {Wang, 2022 #108}. For this chapter's figures, subject 4 is chosen as a base. The calculations in this chapter must be seen as dimensioning calculations, the final verification for the detailed design is performed later.

Actuation & general connection

The starting point for the final design is the required range of motion, during normal operation. To individually actuate plantarflexion, inversion and eversion, three actuation points are needed, as with less actuators, parasitic motion is necessarily induced on one of the non-actuated degrees of freedom. For the first iteration of the design, the location of these actuation points is chosen with respect to a superior point of view of the projection of the ankle axes on the transverse plane of the foot. The plantarflexion actuation point is located at perpendicular to the PF/DF axis at the calcaneus. The inversion/eversion actuation points are located on either side of the foot, perpendicular to the subtalar axis. In the first design iteration, the inversion/eversion actuation points were placed at the back of the bracket, but this proved to cause difficulty actuating the inversion motion as the moment arm is not maximalized, causing a larger force for the same torque. The results of this can be seen in Figure 27.

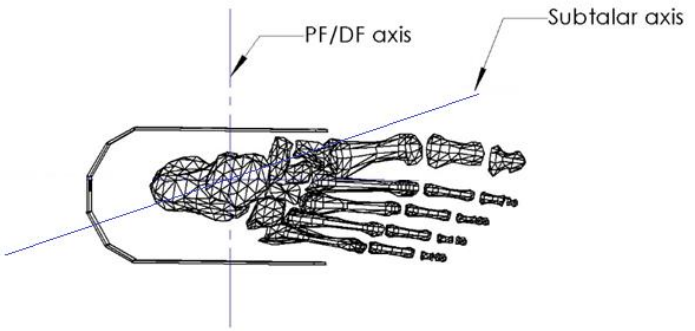


Figure 27: Approximate ankle axes mapping

The necessary actuation forces that are induced on the bracket can be determined from the inverse dynamics of the patients' motion analysis and the moment arm of the device over the gait cycle. It is assumed that the device is attached rigidly to the bike shoe and that the bike shoe is attached rigidly enough to the calcaneus to justify the assumption that any point on the bracket (behind the calcaneus) will follow a comparable path to the plantarflexors' moment arms in the patient's body. The moment arms of the three most important plantarflexors, the soleus, the gastrocnemius and the tibialis posterior are shown in Figure 28.

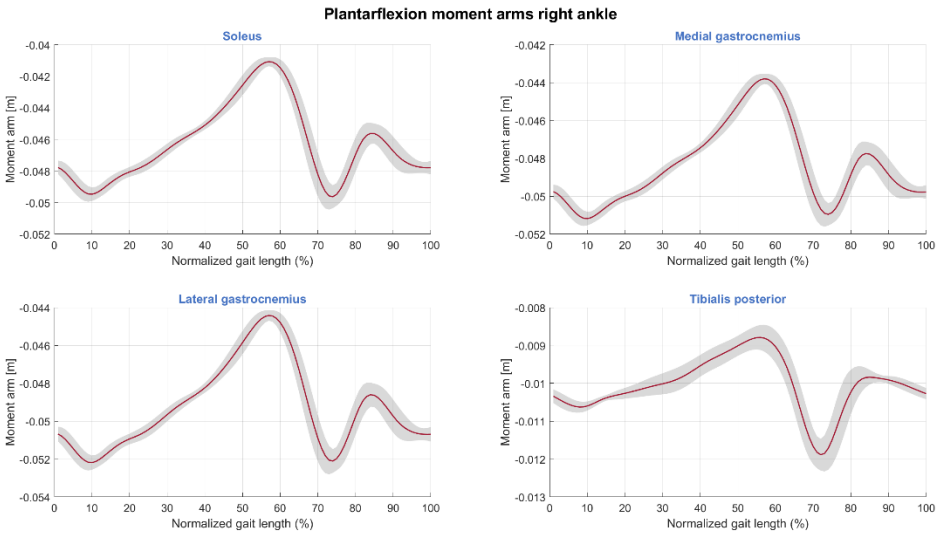


Figure 28: Moment arms over the gait cycle

Averaging those out and taking a ratio of the device moment arm to a muscle moment arm at a known instance yields the curve in Figure 29 for the device moment arm. Note that this moment arm is taken at the attachment point if the foot bracket was designed such that the plantarflexion actuation point was directly behind the calcaneus. This means that this is the smallest possible moment arm in the current configuration. The moment arm is of course approximately equal for both legs.

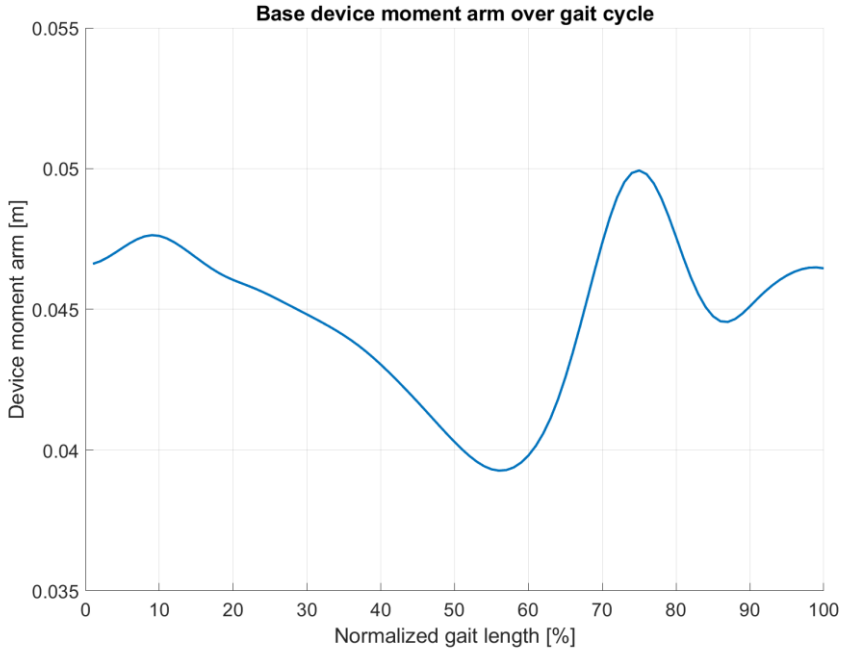


Figure 29: Device moment arm estimation from muscle moment arms

Now, the inverse dynamics results for the data reveal a moment profile which can be normalized over the gait cycle. The average moment profile is shown in Figure 30 for both plantarflexion and subtalar torques.

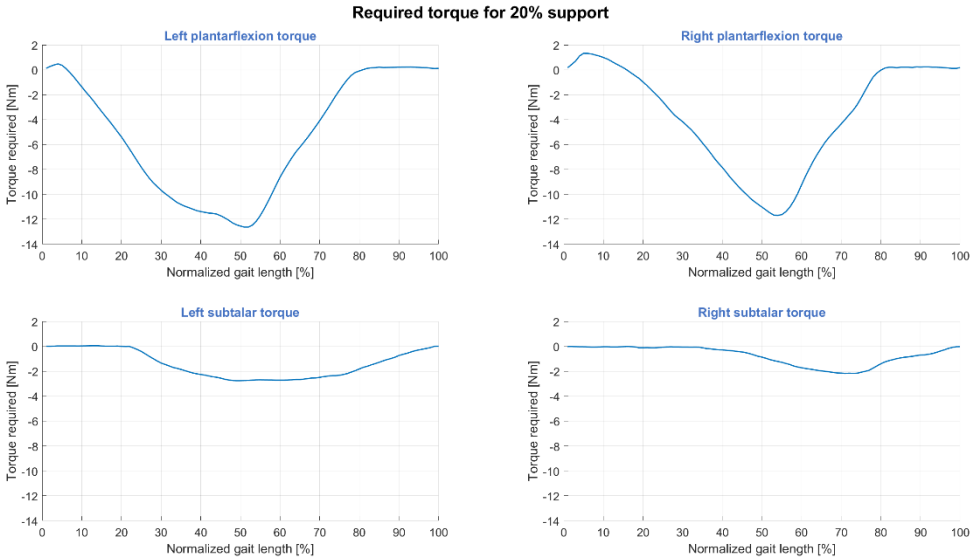


Figure 30: Moment profiles over gait

Dividing the necessary torques by the moment arms yields a force profile for both legs. Applying the shortest possible configuration of the device yields the blue force profiles for both legs in Figure 31. One can easily see that the peak force exceeds 300 N. From the force transmission test performed, the comfortable shear force limit on the shank bracket is for now in the range of 150 – 200 N. One can see that this is reached at around 30 mm of extension of the bracket from the shortest design. Thus, this is chosen as the base length for the design.

Given the design of the bracket, this yields fixed values for the inversion and eversion moment arms. Note that the inversion moment arm is shorter than the eversion moment arm, but they are assumed to be equal (to the shortest) for the purpose of the design.

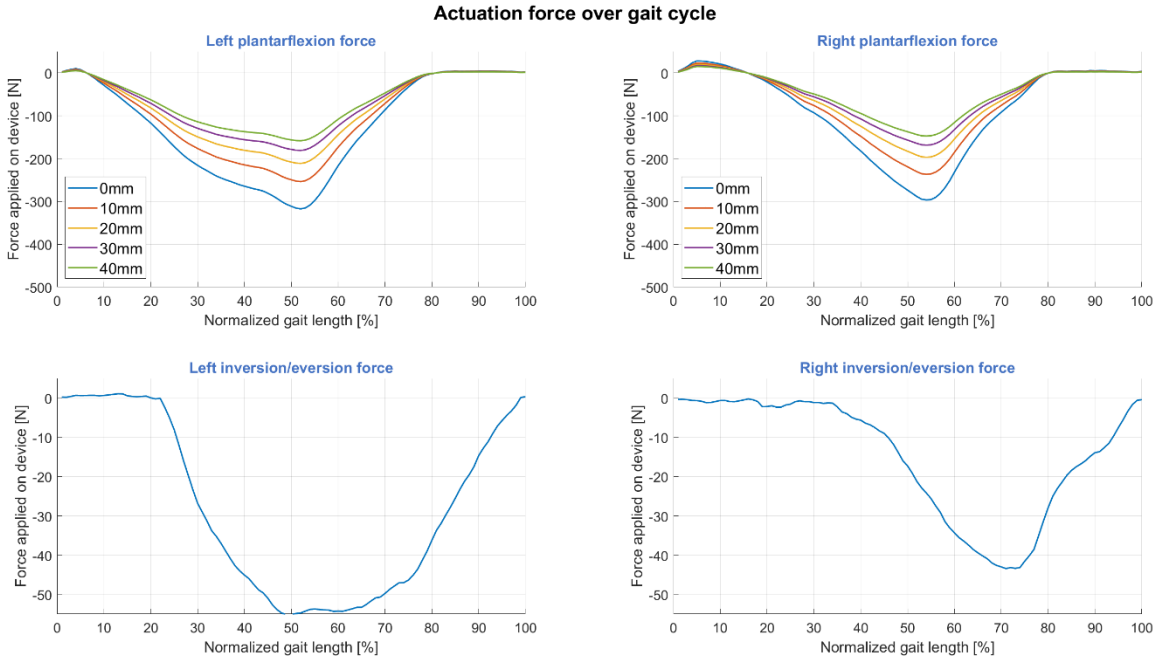


Figure 31: Force profiles for different moment arms

Foot bracket

As the foot is fitted in the bike shoe tightly, the transfer of the forces is not a problem in this case. This means that the potential weak link in the system is the mechanical design itself. Therefore, the connections and materials must be checked to be able to withstand the actuation forces. As calculated before, the actuation force likely won't exceed 200 N in normal operations. However, any other use case than normal walking will require larger applied torques and thus larger forces with the same design. This would require the friction slip problem described later to be solved, but in this case the bracket shouldn't be the weak link. To make it operable under these conditions, the actuation force for these calculations is set to be 400 N, which should be more than enough for future improvements. Any standard bolt values are retrieved from the Applied Eurocode tables for standardized bolts [17].

It can be assumed that the pieces of metal that connect the actuator to the bracket are rigidly attached through a bolt connection, as the standard galvanised 8.8 M8 bolt has a clamping force (21.1 kN) and maximum shear force (14.1 kN) at least an order of magnitude larger than the applied forces (0.4 kN). Therefore, the bracket can be considered as one rigid piece.

For the dimensioning of the bracket, a final element analysis is run via SolidWorks at several geometries, all assuming S235 (standard construction steel) as a material. The parameters varied are the bracket thickness and the total bracket height, as can be seen in Figure 33.

These values are collected in a 3D grid and interpolated to yield respectively the maximum Von Mises stress in and the maximum displacement of the bracket in Figure 32. Also, the mass is plotted in Figure 34.

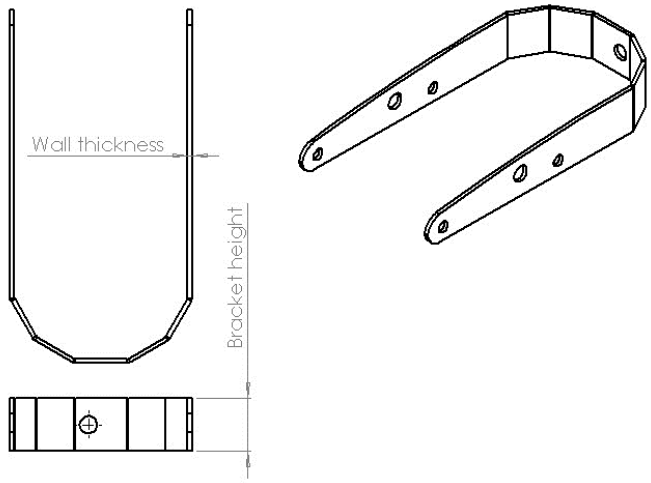


Figure 33: 2D drawing of foot bracket

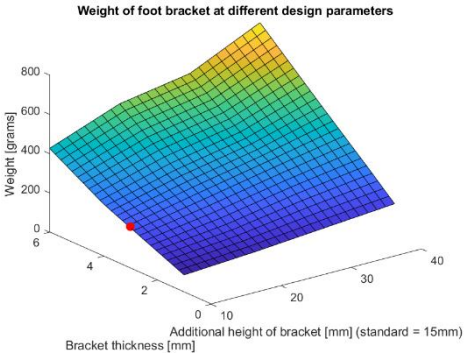


Figure 34: Mass of bracket

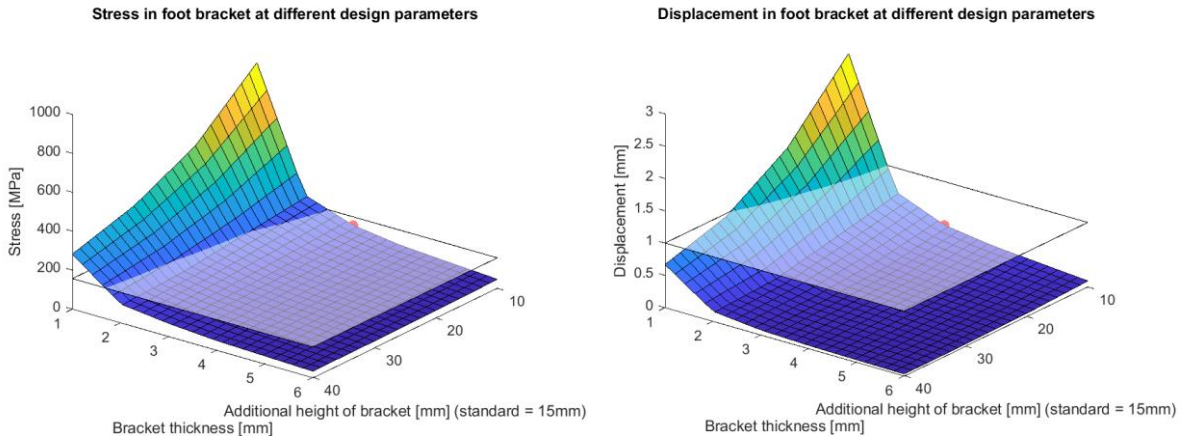


Figure 32: Stress and displacement in the foot bracket

The connection to the bike shoe is made through 2 M6 bolts or alternatively through 2 pens with a diameter of 6 mm. As the centre distance between the 2 holes is approximately 1.15 times smaller than the distance from the middle hole to the actuation point, the maximum force exerted on the pins is around 460 N. This is again way under the standard clamp force and shear force values for a galvanised 8.8 M6 bolt or pin (respectively 11.6 kN and 7.72 kN).

With this, the final configuration of the bracket can be chosen. This is set at a bracket thickness of 3 mm, with an additional height of the bracket of 10mm (seen at the red dot in the figures).

Shank bracket

The shank bracket is lined with silicon and attached to the shank. The silicon liner could be padded, but as the primary concern is functionality, this will not happen at the first iteration. The same forces are applied as on the foot bracket.

The foot bracket only contains one attachment point at the calcaneus. To actuate this, ideally the connection point to the shank bracket is at the calf muscle. However, as the calf muscles bulge, the shank bracket is only solid at its anterior part. In theory, an attachment point could be made on the flexible posterior part, but with stability and functionality taking prime importance, the attachment point is placed at the rigid part. This does mean, however, that the plantarflexion actuation force will induce a parasitic torque. Therefore, the choice is made to split the Bowden cable around the leg, as can be seen in Figure 35.

Also, the orientation of the attachment points must be discussed. The holes of the plantarflexion supports are of course placed as close to the body of the bracket as possible, to reduce any bending moments in the frontal plane. They are also placed in line with the average orientation of the actuation line from the shank to the foot, to reduce shear torques. Now, the forces are of course not always in line with the bracket, which means that fatigue will come in as an important factor, with variations in both the force profile and the orientation of the vector over the gait cycle.

The added benefit of not placing the plantarflexion attachment points in line with the leg is that the forces are not solely transmitted as shear stress. However, if the forces are not completely perpendicular to the shank, the side effect of rotating the attachment point is that the bottom edge of the bracket will bury itself in the skin of the shank, which could be painful. The attachment points placement can be optimized in later iterations.

Figure 35: Overview of Bowden configuration

Mechanical safety limit

As there is no rigid connection between the shank and the foot apart from the ankle, in theory the Bowden cable could keep applying the force in case of a malfunctioning device, even if the ankle is in maximal plantarflexion. Naturally, this is not safe on a healthy ankle, let alone on a patients' ankle. Therefore, the Bowden cable must not be allowed more actuation length than necessary for plantarflexion actuation. This can be calculated by taking the distance between the attachment points at the foot and the shank at maximum dorsiflexion and maximum plantarflexion. This evaluates to a length of 65 mm. The easiest way to ensure this safety is to either place a rigid sleeve or a clamp on the cable between the shank and the foot. For the first iteration, a clamp is chosen to suffice.

Fatigue analysis

The total system is subject to fatigue. This causes the stress resistance of the materials to reduce over time. To estimate the fatigue levels, one needs the S-N curves of the materials. For the foot bracket (S235), this is readily available and well-established [19]. Using SolidWorks, both load events (plantarflexion and inversion curves) are

added as a nondimensional factor on a 1N static load case and calculated for 1.000.000 cycles, which should amply cover the number of cycles the device will run. The load history curves (at the x-axis the gait cycle time) and the 1.000.000 cycle damage chart can be seen with most of the part remaining well under one percent of the estimated life in Figure 36. The force curves are applied at the points marked A and B respectively.

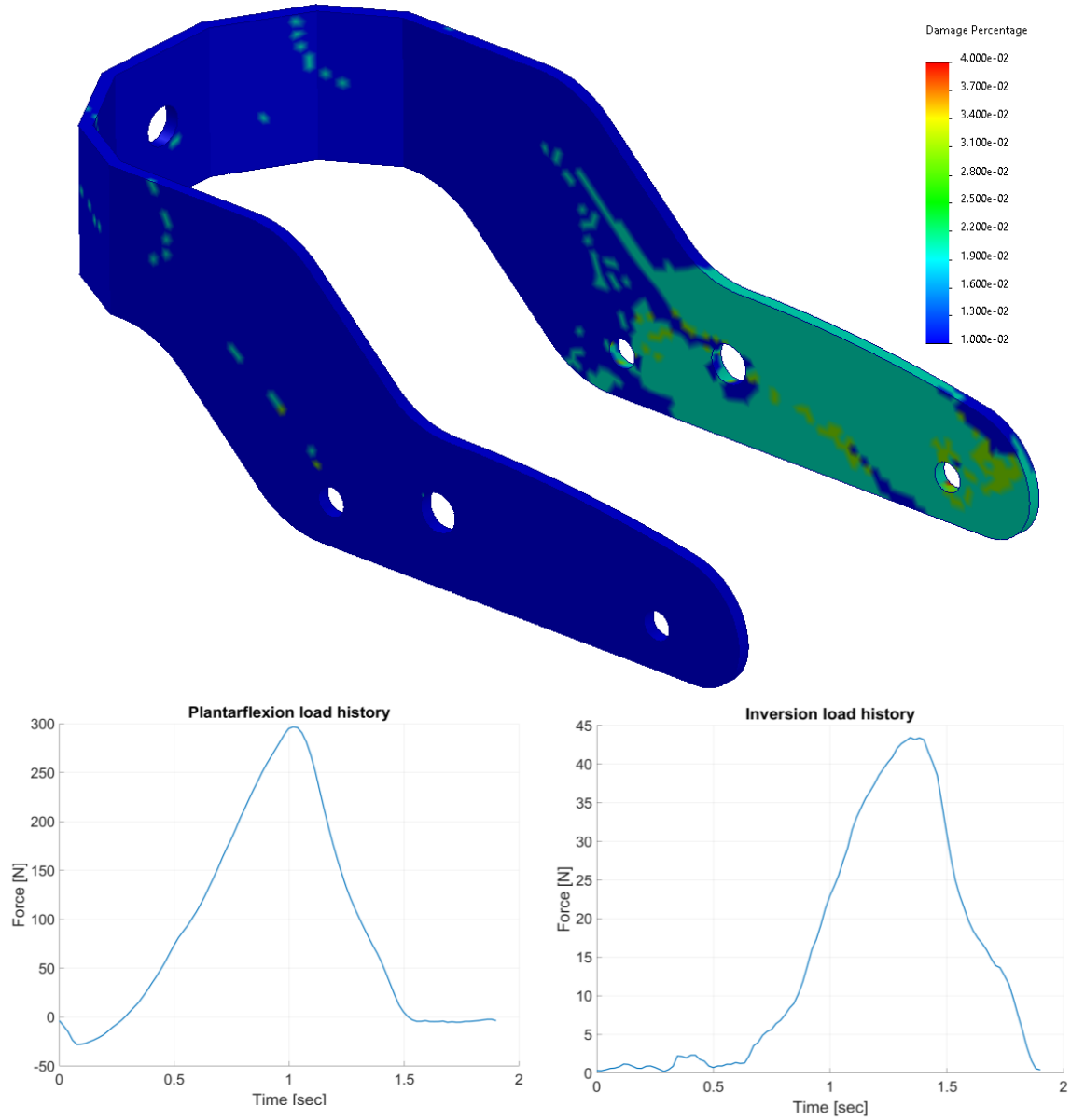


Figure 36: Load history curves and life of the bracket

The values for 3D-printed plastics are more in the experimental phase. As the shank bracket must be strong, but ideally not brittle, a good first material to print would be PLA. For the SN-curves, the data from Safai et. al. is used [20]. The same history curves are applied on the shank bracket, and it is found that the bracket will likely fail between under 150.000 cycles. This is less than the foot bracket. However, as the bracket is 3D-printed with cheap plastics, the only problem that arises with this is that if it must be replaced, the Bowden cables and sleeves must be rewired. The results of this study can be found in Figure 38 at 100.000 cycles.

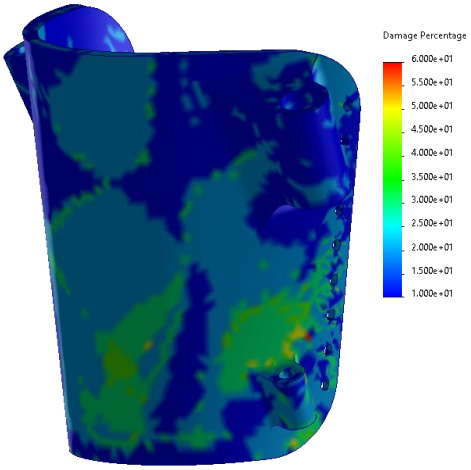


Figure 38: Shank bracket damage

Mass estimation

The total mass of the system has quite some influence on the ease with which the device is handled. The total mass of the system can be derived from Figure 39. The weight of the Bowden cables is not entirely attributable to the system, so the mass is displayed with and without the cables & sleeves. Note that this mass can still be significantly reduced by for example taking an optimized bike shoe.

Element	Mass [grams]
Foot bracket including attachment points and nuts/bolts	287
Shank bracket anterior part	75
Shank bracket posterior part	75
Bike shoe (size 44) + size 6 pin and 4 screws	420
PF SEE + attach	80
Inversion/eversion SEE's + attach	20
PF Bowden sleeve	220
Inversion/eversion Bowden sleeves (per meter)	110
PF Bowden cable	66
Inversion/eversion Bowden cables	58
Total with Bowden assembly	**1411**
Total without Bowden assembly	**1122**

Figure 39: Mass estimation of device

References

[1] WHO. "Spinal Cord Injury." https://www.who.int/news-room/fact-sheets/detail/spinal-cord-injury (accessed 13/02/2023.

[2] B. S. Armour, E. A. Courtney-Long, M. H. Fox, H. Fredine, and A. Cahill, "Prevalence and Causes of Paralysis-United States, 2013," (in eng), *Am J Public Health,* vol. 106, no. 10, pp. 1855-7, Oct 2016, doi: 10.2105/ajph.2016.303270.

[3] E.-E. Adrian, C. Jose, and I. C.-V. Antonio, "Load progression criteria in exercise programmes in lower limb tendinopathy: a systematic review," *BMJ Open,* vol. 10, no. 11, p. e041433, 2020, doi: 10.1136/bmjopen-2020-041433.

[4] CDC. "Stroke facts." https://www.cdc.gov/stroke/facts.htm (accessed 13/02/2023.

[5] D. D. D'Lima, B. J. Fregly, S. Patil, N. Steklov, and C. W. Colwell, Jr., "Knee joint forces: prediction, measurement, and significance," (in eng), *Proc Inst Mech Eng H,* vol. 226, no. 2, pp. 95-102, Feb 2012, doi: 10.1177/0954411911433372.

[6] P. Sadoghi, G. Roush, N. Kastner, A. Leithner, C. Sommitsch, and T. Goswami. *1-s2.0-S1751616114003890-main*.

[7] Y. Zhu, X. Li, and X. Xu, "Ankle joint pressure change in varus malalignment of the tibia," (in eng), *BMC Musculoskelet Disord,* vol. 21, no. 1, p. 148, Mar 4 2020, doi: 10.1186/s12891-020-3163-2.

[8] J. A. Bolduc, J. A. Collins, and R. F. Loeser, "Reactive oxygen species, aging and articular cartilage homeostasis," *Free Radical Biology and Medicine,* vol. 132, pp. 73-82, 2019/02/20/ 2019, doi: https://doi.org/10.1016/j.freeradbiomed.2018.08.038.

[9] Teach me anatomy. "The ankle joint." https://teachmeanatomy.info/lower-limb/joints/ankle-joint/ (accessed.

[10] J. Meijners, "The design, control and integration of a bowden cable actuation for an ankle exoskeleton," Master of Science, Biomechanical Engineering, University of Twente, 2022.

[11] K. A. Witte, J. Zhang, R. W. Jackson, and S. H. Collins, "Design of two lightweight, high-bandwidth torque-controlled ankle exoskeletons," in *2015 IEEE International Conference on Robotics and Automation (ICRA)*, 26-30 May 2015 2015, pp. 1223-1228, doi: 10.1109/ICRA.2015.7139347.

[12] Brooks Rehabilitation. "ReWalk Restore Soft Exo-Suit." https://brooksrehab.org/technology/rewalk-restore-soft-exo-suit/ (accessed.

[13] J. A. Blaya and H. Herr, "Adaptive control of a variable-impedance ankle-foot orthosis to assist drop-foot gait," (in eng), *IEEE Trans Neural Syst Rehabil Eng,* vol. 12, no. 1, pp. 24-31, Mar 2004, doi: 10.1109/tnsre.2003.823266.

[14] C. Meijneke, W. van Dijk, and H. van der Kooij, "Achilles: An autonomous lightweight ankle exoskeleton to provide push-off power," in *5th IEEE RAS & EMBS Interantional Conference on Biomedical Robotics and Biomechatronics, BioRob 2014*, ed. Sao Paulo, Brazil: IEEE, 2014, pp. 918-923.

[15] M. E. Grootens, "Development of a 2-DoF Ankle Exoskeleton," EngD Thesis, University of Twente, Enschede, 2019.

[16] J. Meuleman, "Design of a robot-assisted gait trainer," PhD Thesis - Research external, graduation UT, LOPES II, University of Twente, Enschede, 2015.

[17] EurocodeApplied. "Table of design properties for metric hexagonal bolts M5 to M39 (stress area, shear strength, tensile strength, bearing strength)." https://eurocodeapplied.com/design/en1993/bolt-design-properties (accessed.

Appendix: Table explanation morphological chart 1

	Joint type A	Joint type B	Joint type C	Joint type D	Joint type E	Joint type F
General bulkiness, material needed and medial bulkiness	This concept requires the least amount of material but might need extensions for PF/DF force reduction.	Requires more material. Also requires material at medial side, but just 2 plates, so medium medial bulge.	Comparable to B but requires guiding rail or comparable. Will probably result in larger medial bulge.	Stiffness required in the mediolateral plane, so might cause medial bulge.	Requires quite some material, no medial bulge. All material is in front of the foot.	Requires quite some material. Probably doesn't need a lot of material on medial side.
Range of motion	Widest range of motion. In principle only limited by the ankle.	Unlimited PF/DF motion, but subtalar is limited by the planar displacement cause by subtalar motion in the hinge.	Same as B, but limited to rail length (which is a function of subtalar angle)	Depending on the stiffness, can range from large RoM to small RoM	Mechanical design can be made to accommodate large RoM.	Provides one more DoF than necessary. Probably overcomplicating the design.
Torques, force tranmission and stresses	Requires all torque/force to be converted to stress on the human body.	Translates all force minus the compression times the spring constant in the hinge to the body.	Translates all force minus friction forces in the rail to the human body.	Translates all forces minus compression times spring constant of material to body. May have harmful side effect forces.	Most of the forces can be absorbed by the design. High potential for torque delivery.	Comparable to E
Tests conclusion	Force tests most important	Motion tests most important, force needed	Motion tests most important, force needed	Motion tests most important, force needed	Motion tests most important, force needed	Motion tests most important, force needed

Appendix C:

Experiment 1 details | Shank bracket stability

Developments in 2 degree of freedom fine force delivery by an assistive ankle orthosis

Bart Hendriksen, s2424576
Supervisor: Dr. G. V. Durandau
Draft date: October 6, 2023
Execution date: October 19, 2023
Version: 1A

Table of Contents

TABLE OF CONTENTS **2**

1. RATIONALE AND SCOPE **3**

2. OBJECTIVES **3**

3. STRATEGY **3**

SETUP 3
SCHEMATICS 4

4. RESOURCES **5**

5. RESULTS **5**

DATA LOCATIONS 5
GENERATED FIGURES 5

6. APPENDIX **6**

1. Rationale and scope

One of the limits of the device is the amount of force that can be transferred to the shank without causing pain or discomfort, while remaining stable on the shank. Of interest is the motion of the bracket relative to the shank as a function of the force input, all with varying tightness of the bracket connection.

2. Objectives

The objective is to measure the relative motion of the bracket and the shank and to plot this as a function of the force input.

3. Strategy

The subject will be fitted with the device to apply the forces, with several MOCAP sensors to measure position, while being fitted with the plantarflexion and inversion actuation cables.

Setup

The following setup will be used:

- Shank bracket + 3 actuation lines
  - 2 PF
  - 1 Inversion
- Force sensor to measure the applied forces.
- 6 Qualisys sensors to measure the shank bracket position. The numbering can be found in Figure 1.
  - 3 on the bracket
  - 3 on the shank

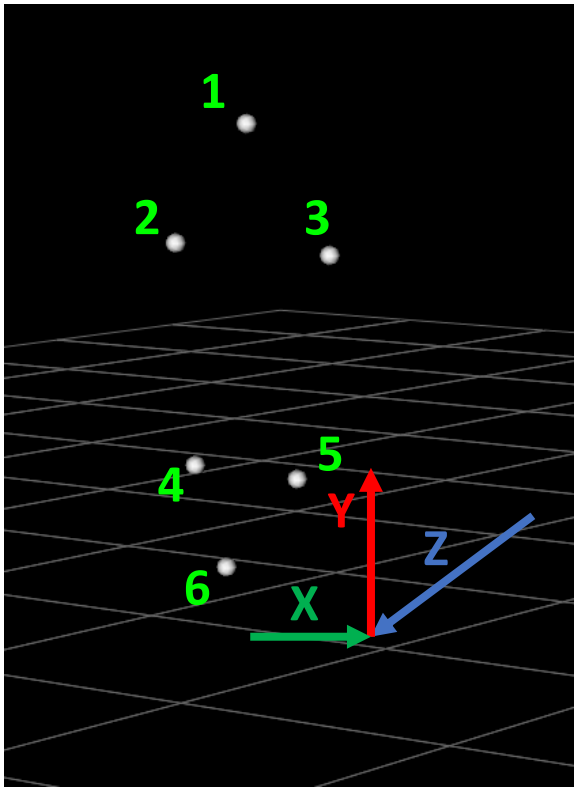


Figure 1: Marker numberings

For this experiment, the forces are applied manually as an exact input force profile is not necessary for the force-displacement relationship.

Note that the pulley setup has some inherent force losses which must be accounted for. These are calculated at the start by applying a known force on the calibrated sensor through the pulley system and seeing the percentage loss. This means that the input force is lowered by this percentage to be applied on the shank bracket. The

Schematics

On **Error! Reference source not found.**, a physical model of the experiment is depicted. The lower leg is represented by the blue line, the shank bracket by the golden bar and the tension line (Bowden cable) by the red line. Note that the force at the input ($F_{t|i}$) is not equal to the force that is measured ($F_{t|fs}$), or that is applied to the bracket ($F_{t|b}$). The pulley friction causes a force loss of about 4.5%, or:

$$F_{t|fs} = 0.955 \cdot F_{t|i}$$

The actual input force doesn't matter in this case, as the force is measured at $F_{t|fs}$. This is the value that the raw logs of the hx711 files will contain. To get the true value that is applied at the shank bracket, another force loss must be applied, or:

$$F_{t|b} = 0.955 \cdot F_{t|fs}$$

and:

$$F_{t|b} = 0.912 \cdot F_{t|i}$$

Now, as the bracket is fixed to the shank by means of a friction connection, the friction force must be equal to $F_{t|b}$. For the purposes of this experiment, the normal force by which this is achieved is not relevant and is thus not considered.

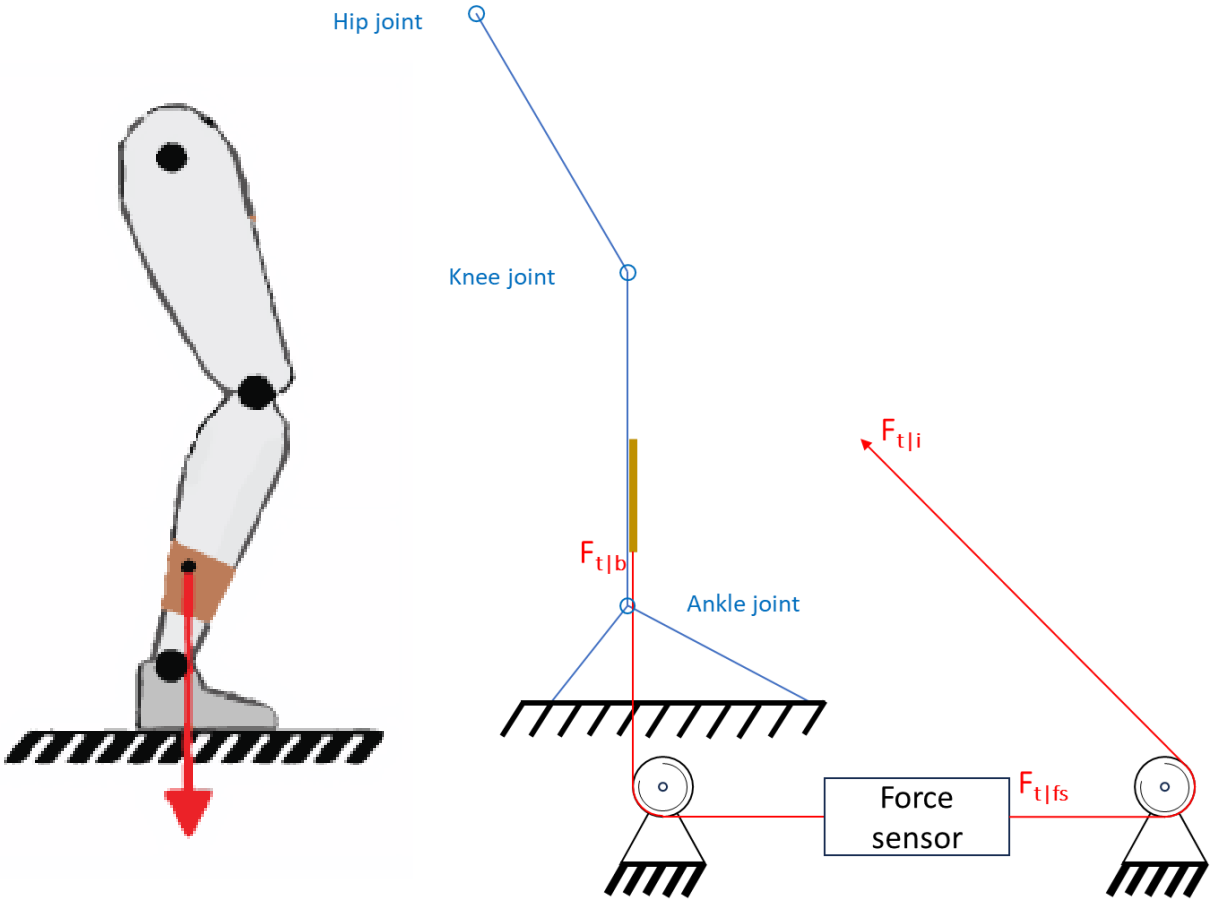


Figure 2: Physical model and free body diagram of the experiment.

4. Resources

- Orthosis + force measuring.
- Qualisys system for marker measurement

5. Results

Data locations

The table below contains the locations of the raw results files.

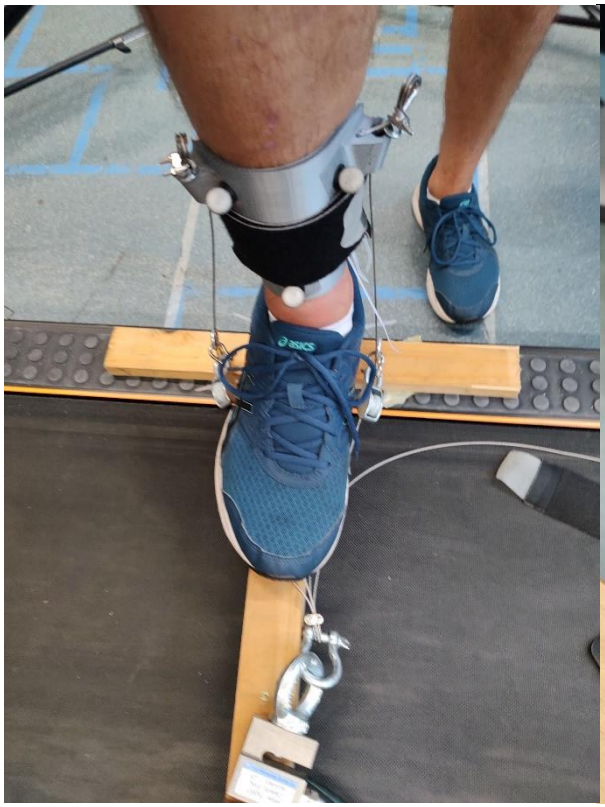
Applied magnitude	Perceived comfort	Position files	Force files
0 – 50 N	High	#Master's assignment\07_Code\computation\phase3\input\experiment1\qualisys\5kg.mat	#Master's assignment\07_Code\computation\phase3\input\experiment1\hx711\5kg.log
0 – 100 N	Medium	#Master's assignment\07_Code\computation\phase3\input\experiment1\qualisys\10kg.mat	#Master's assignment\07_Code\computation\phase3\input\experiment1\hx711\10kg.log
0 – 150 N	Low	#Master's assignment\07_Code\computation\phase3\input\experiment1\qualisys\15kg.mat	#Master's assignment\07_Code\computation\phase3\input\experiment1\hx711\15kg.log
0 – 200 N	Very low	#Master's assignment\07_Code\computation\phase3\input\experiment1\qualisys\20kg.mat	#Master's assignment\07_Code\computation\phase3\input\experiment1\hx711\20kg.log

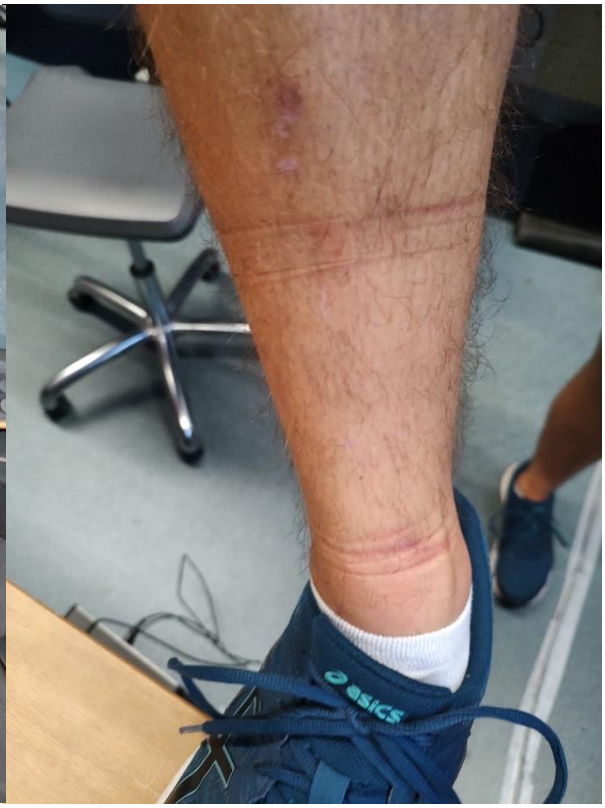
Generated figures

All the raw data figures that were generated for this experiment are stored in the figures folder of phase 3 in the code: #Master's assignment\07_Code\computation\phase3\figures\1\. Any relevant summary figures can be found in the main body of the thesis.

6. Appendix

Photos of the experiment. The imprints on the leg are after a full run of experiments, also with experiment 3, so a total of more than 3 hours. The connection was just silicon and the bracket, no padding was added.





Appendix D:

Experiment 3 details | Muscle activation levels

Developments in 2 degree of freedom fine force delivery by an assistive ankle orthosis

Bart Hendriksen, s2424576
Supervisor: Dr. G. V. Durandau
Draft date: October 6, 2023
Execution date: October 19, 2023
Version: 1A

Table of Contents

TABLE OF CONTENTS **2**

1. RATIONALE AND SCOPE **3**

2. OBJECTIVES **3**

3. STRATEGY **3**

SETUP 3
MARKER POSITIONS 3
EMG POSITIONS 4
SCHEMATICS 5

4. RESOURCES **7**

5. RESULTS **7**

DATA LOCATIONS 7
GENERATED FIGURES 7

APPENDIX: PHOTOS **8**

1. Rationale and scope

To prove that actuation of the device as designed does in fact provide subtalar motion, this test is performed. It should be performed in a lab setting, measuring the 3D position of the ankle relative to the shank. This is all done in the setting of providing a set of force combinations resembling forces applied during normal locomotion. Note that any and all locations of parts with respect to the body are meant for the right leg!

2. Objectives

The main goal is to retrieve the EMG data with and without support. This will ultimately provide insight in the working principle of the device at the one hand and will show if parasitic motion is induced on the other.

3. Strategy

The subject will be fitted with the device to apply the forces, with several MOCAP sensors to measure position and with EMG sensors to measure muscle activation.

Setup

The following setup will be used:

- Orthosis to provide forces.
- Qualisys to record body position over time.
- TMSI or Densys to measure the muscle activation of invertors and plantarflexors.

The position is plotted over the recording and ideally on the second y-axis the activation in separate plots for every motion, always in pairs with and without force support. It is probably difficult to provide both plantarflexion and inversion support at the same time, so the trials will likely be separate force applications.

This experiment weighs the effectivity of the device on movements that isolate the muscles with and without support. While the device can support up till 200 N in the current configuration, the motion is executed with 50 an 100 N of constant support as there is no actuator yet. Because the dorsiflexors must move the weight back up, it is not possible to go higher than 100 N (and 50 N is best). The normal dorsiflexion torques these muscles provide is much lower than the plantarflexion torque and 16 Nm of torque is too high. 4 Nm (which is about what is generated with 50 N of support), is their normal operating range and thus this should be doable. Figure 1 gives the summary of the tests performed.

Trials	Support variations [N]	Support type	Data used?
Walking trials	Constant 0, 50	None, plantarflexion, inversion, eversion.	No
Puppet motions	Peak matching 0, 50, 100, 150, 200	None, plantarflexion, inversion, eversion.	No
Calf raises	Constant 0, 50	None, plantarflexion	Yes
Inversion balance	Constant 0, 50	Inversion, eversion.	Yes
Eversion balance	Constant 0, 50	Inversion, eversion.	Yes

Figure 1: Experiments variations performed

Marker positions

The marker positions will be those necessary for the normal 2392 OpenSim model and they are placed according to Figure 2. The purple markers are the 3 markers on the shank bracket of the device.

EMG positions

For the plantarflexion, the muscles of interest are the soleus and the gastrocnemius. While the shank bracket has a strap at the calf, the EMG sensor can be placed under there for soleus measurement. The other plantarflexors are not superficial enough to measure.

For eversion, the peroneus longus and brevis are considered and tibialis anterior is measured for reference and as an invertor (it contributes for a small part). The placements can be seen in Figure 2.

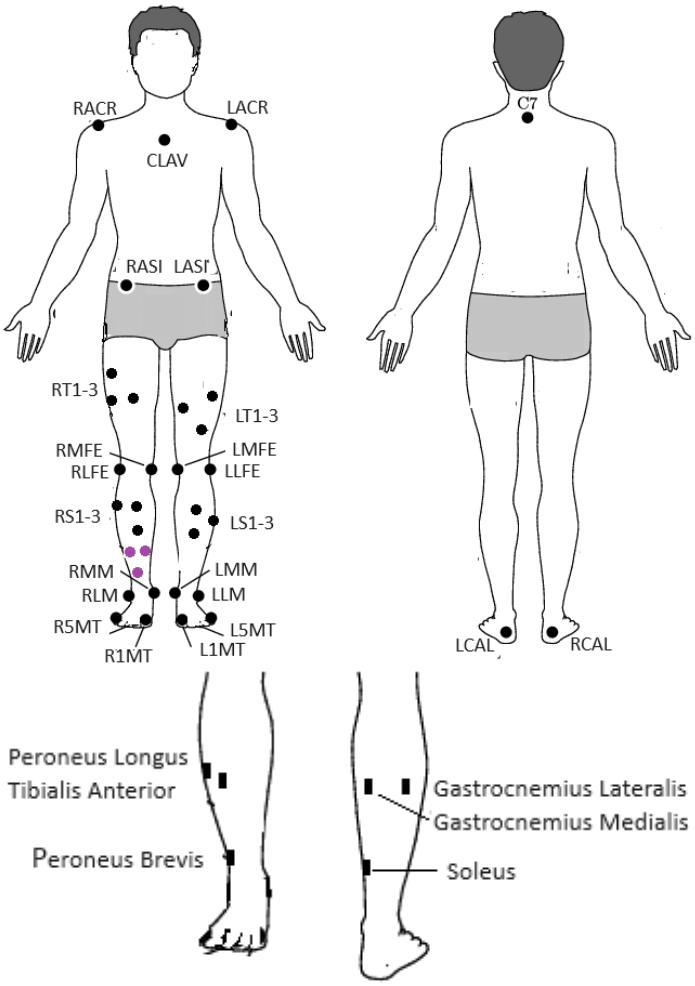


Figure 2: Marker and sensor positions on the human body.

Schematics

For this experiment Figure 3 represents the setup. The device is setup as it would be in the final application with the only exception that no actuator is used but the weighs are applied manually, either constant or by hand. The blue lines represent the lower leg, with the foot, shank, and thigh segments, along with the ankle, knee, and hip joints. The setup is a tensile force (red line) that is measured at the force sensor and guided by 2 pulleys and a sleeve (burgundy red line) from the input to the foot. There is a friction loss in the Bowden sleeve and in the pulleys of respectively 7% and 4.5%. In the final setup the lead screw at the input stage pulls the cable with respect to the sleeve causing a tension at the attachment point at the foot and a pressure at the attachment point at the shank (at the brown shank connector).

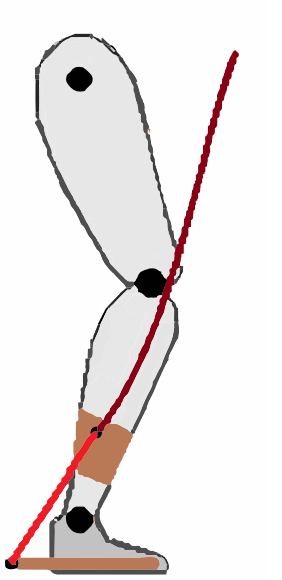


Figure 3: Physical model and free body diagram of the experiment setup, with a plantarflexion setup.

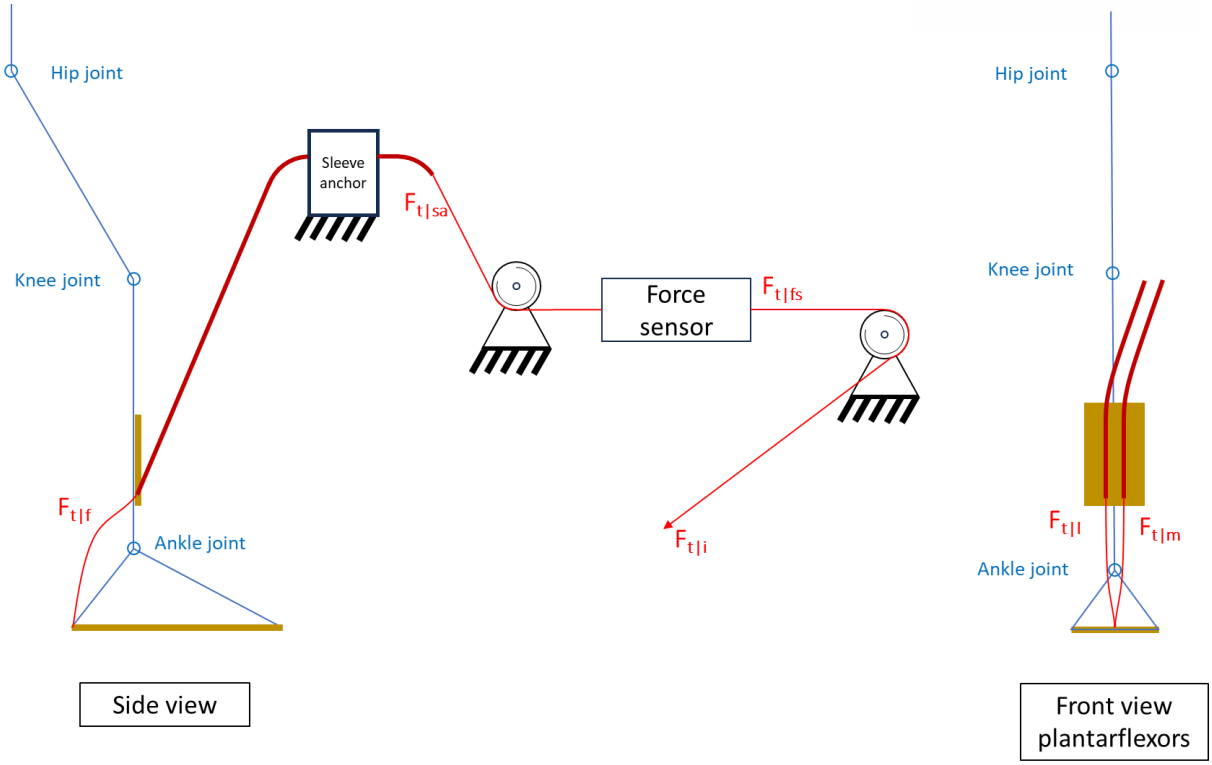


Figure 4: Free body diagram of the plantarflexion setup.

Figure 4 shows the configuration used for the experiment to when the plantarflexors are of interest. The Bowden cable is attached to the plantarflexion attachment point (A in Figure 6) at the foot bracket allowing for full force transfer in plantarflexion assistance. Note that the cable is split as it would be in the final configuration. The split is merged again after the sleeve anchor. The total force applied is not affected by this, but the pushing force on the shank bracket is distributed over the medial and lateral sides of the shank.

Figure 5 shows the same situation, but now when the invertors and evertors are of interest. The Bowden cable is attached to the inversion or eversion attachment points (respectively B and C in Figure 6) and thus the full force can be applied to help inversion or eversion.

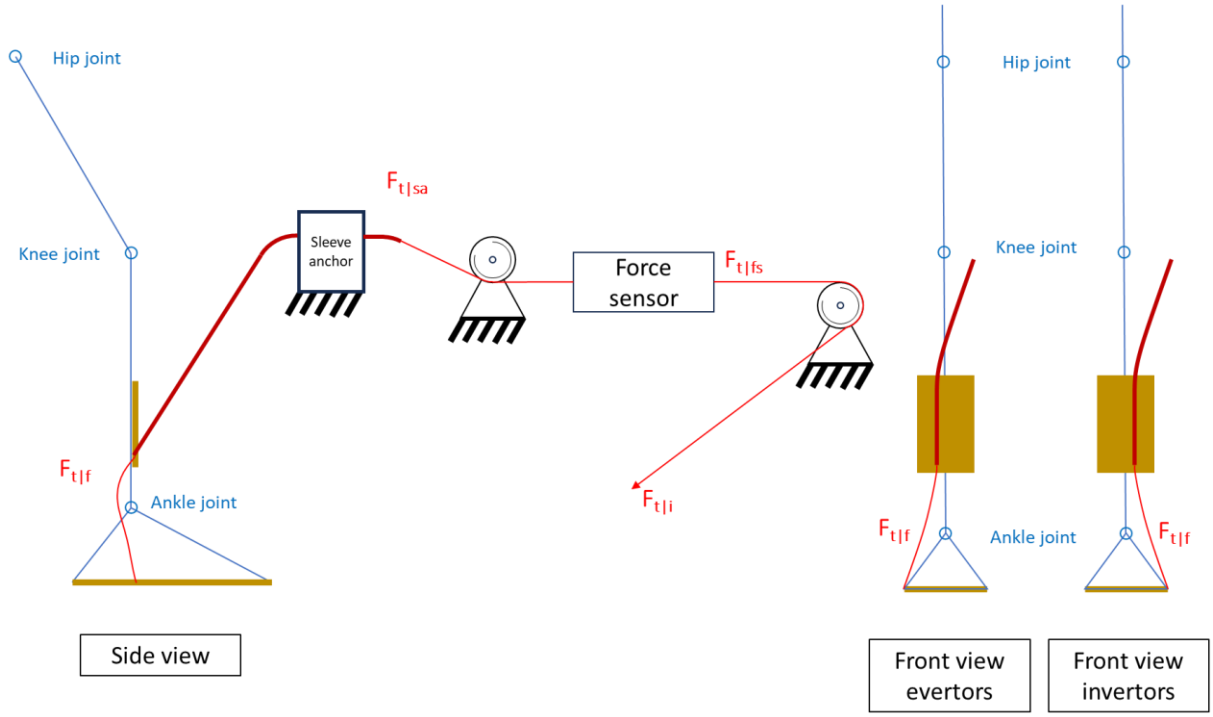


Figure 5: Free body diagram of the inversion/eversion setup.

Note that the force at the input ($F_{t|i}$) is not equal to the force that is measured ($F_{t|fs}$), or that is applied to the foot ($F_{t|f}$). The pulley friction causes a force loss of about 4.5%, or:

$$F_{t|fs} = 0.955 \cdot F_{t|i}$$

The actual input force doesn't matter in this case, as the force is measured at $F_{t|fs}$. This is the value that the raw logs of the hx711 files will contain. To get the true value that is applied at the shank bracket, another force loss must be applied, or:

$$F_{t|sa} = 0.955 \cdot F_{t|fs}$$

and:

$$F_{t|f} = 0.93 \cdot F_{t|sa}$$

Combining these two yields a final force reduction on the measured force to the applied force of:

$$F_{t|f} = 0.93 \cdot 0.955 \cdot F_{t|fs} = 0.888 \cdot F_{t|fs}$$

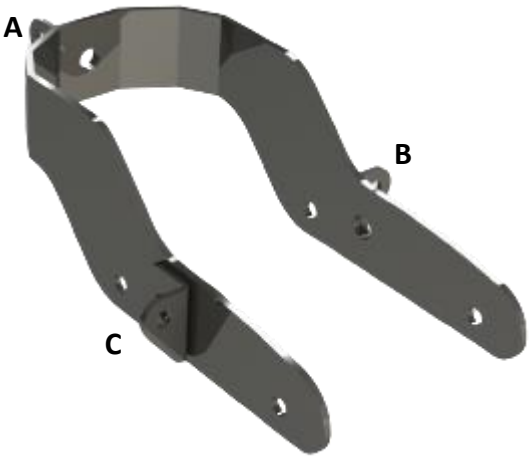


Figure 6: Isometric view of foot bracket with 3 attachment points. A: plantarflexion attachment, B: inversion attachment, C: eversion attachment

4. Resources

- Orthosis + force measuring.
- Qualisys with mocap and EMG through Densys

5. Results

Data locations

All the raw data are stored in the file location below, for all trials:

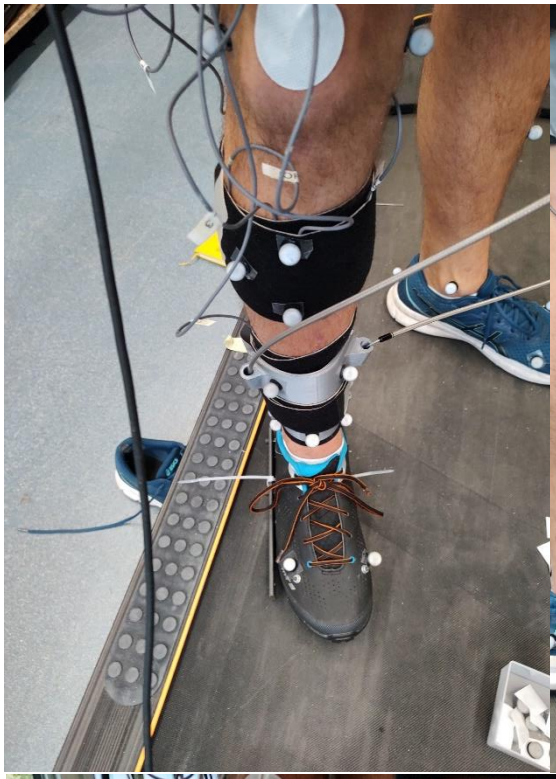
#Master's assignment\11 Experiments\labRecordings\bart_thesis\Data\exp3b

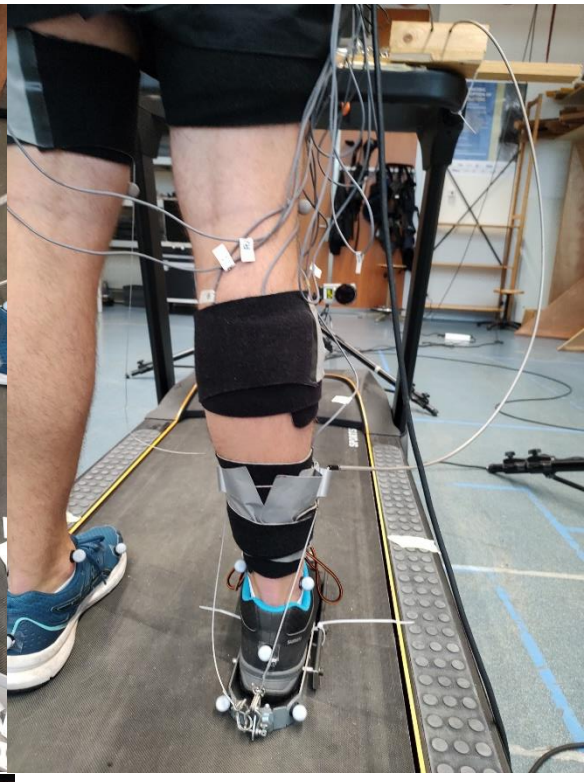
Generated figures

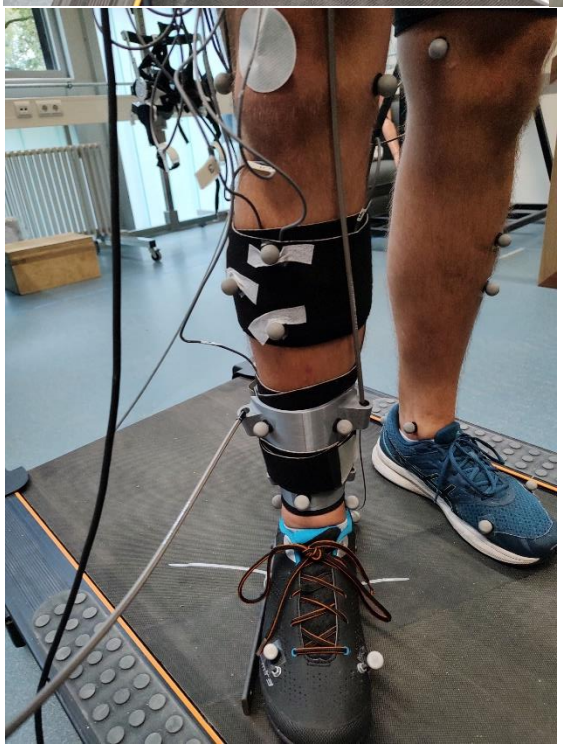
All the raw data figures that were generated for this experiment are stored in the figures folder of phase 3 in the code: #Master's assignment\07_Code\computation\phase3\figures\3b\. Any relevant summary figures can be found in the main body of the thesis.

Appendix: Photos

Photos of the marker placements during the experiment.







Appendix E:

Computation procedures

Developments in 2 degree of freedom fine force delivery by an assistive ankle orthosis

Bart Hendriksen, s2424576
Supervisor: Dr. G. V. Durandau
Draft date: January 26, 2023
Version: 1A

Table of Contents

TABLE OF CONTENTS....2

INTRODUCTION 3

1. GENERAL STRUCTURE....3
2. PHASE 1: ANALYSIS OF HEALTHY MODEL 5
3. PHASE 2: MECHANICAL DESIGN ESTIMATIONS & CALCULATIONS 6
4. PHASE 3: EXPERIMENTAL DATA ANALYSIS 7

Introduction

This document contains the logic, structure, and explanation of all the code that is written within the project and explains the structure that is setup to compute the optimal design for minimized joint loads. The document is divided in a general explanation in chapter one and a further specification of the phases in respectively chapters two through four.

1. General structure

The general structure of the computation done in the project can be summarized in the flow chart in **Error! R eference source not found.**.

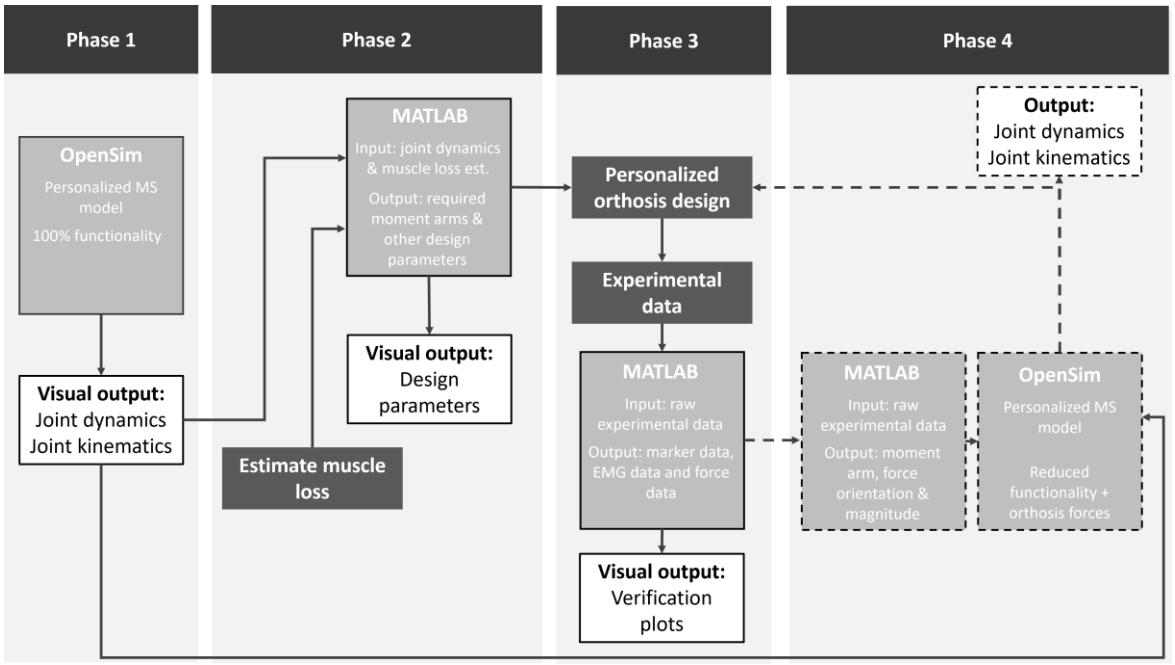


Figure 1: General flow chart of the code

Phases 1 through 3 have been completed within this thesis, phase 4 is merely a suggestion for a continuation of the project. In general, phase 1 is needed during the analysis of the thesis, phase 2 during the design and phase 3 during the verification. The process works as follows:

1) Start in phase 1 with visualisation of all the musculoskeletal data and saving those data in a structure that can be used as an input for phase 2.
2) Phase 2 is the estimation of the design parameters. Based on a muscle reduction factor (20% in this case, as the design requirements ask for 20% support), the script calculates data like the moment arm needed to provide the torques required or the effects of certain dimensions on the stress levels in components.
3) These inputs are used to generate a personalized design in phase 3. With this design, experiments are conducted, from which the design can be verified.
4) In the now non-existent phase 4, MATLAB and OpenSim are integrated to generate an iterative model that takes the orthosis forces and calculates the joint and muscle reactions from a reduced muscle model.

Furthermore, the general structure of the code in MATLAB can be summarized as in Figure 2. The functions are explained in the applicable chapters. For normal users, the code should only be edited inside the root.m file.

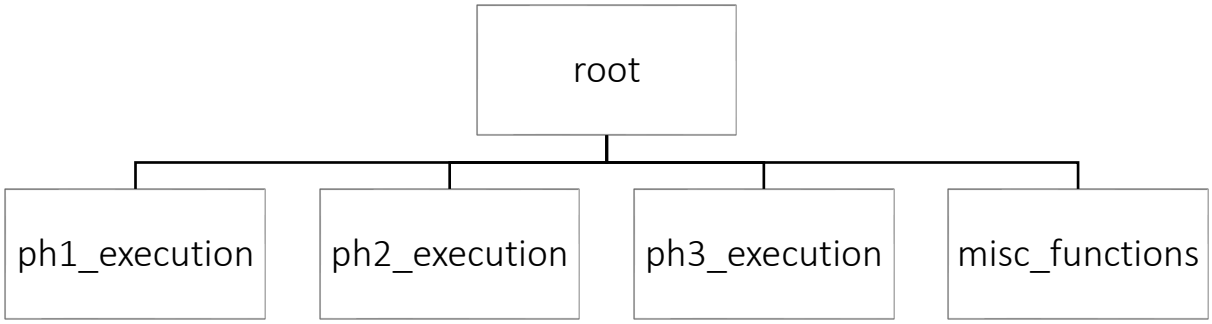


Figure 2: General code structure

Also, in Table 1, one can find a general overview of the options for the selectors in the root file.

Table 1: General selector overview

Root selector overview	
General selectors	
recalc	0/1 based on whether to use current segmented data or recalculate for another subject or new dataset.
subjectNr	4 – 12 based on which subject to use for the segmentation and plotting.
reduction	Set the value between 0 and 1 for the muscle reduction.

Any inputs used in the computation should be placed in the folder inputData and a proper reference should be defined in the function getPaths.m. Outputs will always be made in the same way; this should not be changed by the user. Any .mat structures get written to the folder output in the respective phase and any figures to the folder figures in the respective phase if the saveFigures selector is set to one.

Table 2: Overview of miscellaneous functions

Miscellaneous functions	
Custom functions often used in all code	
shade.m	A custom function, retrieved from the MATLAB discussion boards. This function takes a mean, and an upper and lower limit and plots it. Copyright (c) 2018 Javier Montalt Tordera.[1]
interparc.m	A custom function, retrieved from the MATLAB discussion boards. It is an advanced interpolation function. Copyright (c) 2010 John D'Errico.[2]
Functions determining the paths to the input data	
getPaths.m	Function that defines the paths to the input files.

[1] Javier Montalt Tordera (2023). Filled area plot (https://www.mathworks.com/matlabcentral/fileexchange/69652-filled-area-plot), MATLAB Central File Exchange. Retrieved October 23, 2023.

[2] John D'Errico (2023). interparc (https://www.mathworks.com/matlabcentral/fileexchange/34874-interparc), MATLAB Central File Exchange. Retrieved October 23, 2023.

2. Phase 1: Analysis of healthy model

Phase 1 consists of four main functions and subfunctions under those. The four main goals are:

1) Segmenting the kinematics and dynamics data.
2) Segmenting the muscles data
3) *Plotting the kinematics and dynamics data.*
4) *Plotting the muscles data.*

How this is done is explained in the table below. The numbers indicate a level (3 is the level of the main goals). Note that for the muscles analysis to work, the respective OpenSim analyses must be run first. As this is not needed later in the design, this is commented out by default. Define the paths in the getPaths().m function.

Ph1_execution functions		
Segmenting the kinematics and dynamics data.		
calc_kin_dyn.m	3	Takes the raw inverse kinematics and inverse dynamics OpenSim outputs, segments the data and normalises it over the gait cycle to be visualized. The output is the mean with a band of standard deviation.
calc_kin_dyn_peaks.m	4	A subfunction that looks at the peaks in the data to calculate the moments of heel strike upon which the data is segmented.
norm_kin_dyn.m	4	A subfunction that segments the data based on the peaks.
calc_kin_dyn_MSD.m	4	A subfunction that calculates the mean and standard deviation of the segmented data.
Segmenting the muscles data.		
calc_muscles.m	3	Takes the raw static optimization OpenSim outputs and other OpenSim outputs, segments the data and normalises it over the gait cycle to be visualized. The output is the mean with a band of standard deviation.
calc_muscle_peaks.m	4	A subfunction that looks at the peaks in the data to calculate the moments of heel strike upon which the data is segmented.
norm_muscles.m	4	A subfunction that segments the data based on the peaks.
calc_muscle_MSD.m	4	A subfunction that calculates the mean and standard deviation of the segmented data.
Plotting the kinematics and dynamics data		
ph1_kin_dyn_plot.m	3	Takes the segmented mean and standard deviations of the kinematics and dynamics and plots it.
Plotting the muscles data		
ph1_muscles_plot.m	3	Takes the segmented mean and standard deviations of the muscle data and plots it.
ph1_ma_plot.m	4	Plots the moment arms of the selected muscles.
ph1_act_lot.m*	4	Plots the activations of the selected muscles.
ph1_mf_plot.m	4	Plots the muscle forces of the selected muscles.
ph1_fl_plot.m	4	Plots the fibre lengths of the muscle forces.

* Please note that the EMG's are only available insofar as they were measured during the trails. The following muscles are available by default in the dataset:

EMG Channel	**Full muscle name**	**Abbreviation**
1	Rectus Femoris	rf
2	Vastus Medialis	vm
3	Vasus Lateralis	vl
4	Semitendinosus	sem
5	Biceps Femoris	bf
6	Tibialis Anterior	ta
7	Gastrocnemius Medialis	gm
8	Gastrocnemius Lateralis	gl
9	Soleus	sol

3. Phase 2: Mechanical design estimations & calculations

The main goal for phase 2 is generating data used for the mechanical design based on the subject data. It will also generate the figures necessary for interpretation.

Ph2_execution functions		
Function name		**Brief explanation**
calc_force.m	3	Takes the needed torques for walking and using the moment arm of the device, it calculates the force needed to provide the assistance. It does so based on the reduction parameter defined in the root.m file.
calc_MA_device.m	4	Estimates the moment arm based on the moment arms of the muscles.
genFatigueFigs.m	3	Generates the fatigue plots used as an input in SolidWorks for the materials specified.
genFigFoot.m	3	Plots all the data previously generated.

4. Phase 3: Experimental data analysis

Phase 3 deals with the interpretation of the experimental data generated in the lab.

Ph3_execution functions		
Function name		**Brief explanation**
calcEuclid.m	4	Loops over 2 sets of points and calculates the Euclidian norm between them.
calcGaitNormVals.m	3	Function that normalizes the data over the gait (for experiment 1).
calcMeans.m	4	Calculates the means and standard deviations of experiment 1.
calcRelDist.m	3	Takes the marker data for the bracket and the shank and calculates the distance between the centroids using calculateCentroid.m and calcEuclid.m
calculateCentroid.m	4	Calculates the centroid position of three markers.
emgFromQualisys.m	4	Extracts the EMG data, frequency, and total frames from the Qualisys output structure based on the input folder.
filterEmg.m	3	Filters the EMG signal using a HPF and LPF.
genFigExp1.m	3	Generates all the figures visualizing the interpreted experimental data results for experiment 1.
genFigExp3a.m	3	Generates all the figures visualizing the interpreted experimental data results for experiment 3a. This data is not used in the final thesis.
genFigExp3b.m	3	Generates all the figures visualizing the interpreted experimental data results for experiment 3b. This data is not used in the final thesis.
genRmsTable.m	3	Generates tables of the root mean squares of the signal.
getPeaks.m	4	Function to get the peaks/valley data for the trials, used as an input for segmentation.
interpolateMarkerData.m	4	If one of the markers of interest was not fully filled, this function does so by linearly interpolating the position of the marker based on the two others that are known.
loadEmg.m	3	Function to load the EMG data from Qualisys.
loadForce.m	3	Function to load the Force data from the HX711 sensor.
loadQualisys.m	3	Function to load the marker data from Qualisys.
markerFromQualisys.m	4	Function to extract the exact marker data from Qualisys for every trial.
normalizeEmg.m	3	Function to normalize the EMG data for experiment 3.
normalizeForces.m	3	Function to normalize the forces on the same time frame as the Qualisys data.
trimForce.m	4	Function that performs the actual trimming of the force data based on manual inputs.