Department of Precision and Microsystems Engineering

Conceptual design of a compliant hip orthosis for Trendelenburg gait

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Preface

This thesis concludes my master High-Tech Engineering at the faculty of Mechanical, Maritime and Materials Engineering at the Delft University of Technology. I am proud to graduate at one of the most technical disciplines here at Delft University of Technology.

I would like to thank my supervisors Jelle Rommers, Bram Sterke, and Just Herder for introducing me to this project, their constructive feedback, and guidance throughout the year. I would like to thank Rutger Osterthun for his insight into important factors in orthoses designs for patients. I would like to thank my fellow students for their feedback during the weekly meetings and in particular Boris for the countless hours of studying together, coffee breaks, and the way too fancy lunches. Finally, I would like to thank my friends and family for their support during my thesis and studies.

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Delft, December 2020

Contents

1 Introduction **1**

2 Literature Study **3**

3 Paper: Conceptual design of a compliant hip orthosis for Trendelenburg gait **17**

4 Discussion **29**

5 Conclusion **31**

Appendices: **33**

A Design process **33**

B Technical prototype design **37**

Chapter 1

Introduction

Trendelenburg gait is an abnormal gait that occurs when hip abduction muscles are weakened. These weakened muscles cannot deliver enough force to prevent the pelvis from drooping when standing on one leg. This results in an abnormal gait where the upper body sways from left to right to keep the center of gravity above the standing foot. Besides standing out from other people when walking, this causes problems at the hip joint, knees, and ankles. Existing orthoses for Trendelenburg gait help to support the upper body, but are uncomfortable or spacious external devices that obstruct other motions of everyday life. A piece of clothing that accommodates the orthosis such that it does not stand out without compromising the effectiveness does not exist. An attempt to design such an orthosis is started in this thesis with the conceptual design for a complaint hip orthosis for Trendelenburg gait.

1.1 Research objectives

The main objective of the orthosis is to prevent the upper body from drooping during gait. Besides this, the person should still be able to walk, the orthosis should be comfortable and may not protrude from the human body.

To investigate the performance of an orthosis, the adduction over flexion-extension stiffness ratio is investigated. Adduction is the motion that the pelvis makes when it droops and flexion-extension of the leg is the main motion of the leg during gait. If this ratio is high, the upper body can be supported without investing a lot of energy in deforming the orthosis to move the legs during gait.

The direction of the forces that the orthosis exerts on the human body are investigated. High shear forces are undesired because this enables creep to occur, which can be uncomfortable. Also, lower skin stiffness in tangential direction decreases the adduction stiffness of the orthosis, decreasing the above-mentioned performance of the orthosis.

The orthosis should be close to the human body during gait. This allows the person to wear clothes over the orthosis to hide it and makes it aesthetically more appealing than protruding orthoses.

1.2 Thesis outline

Chapter 2 presents a literature study, in which the terminology around the human gait, problems with Trendelenburg gait, and existing orthoses are investigated. Chapter 3 contains the main part of this thesis, a paper describing the conceptual design of a compliant hip orthosis for Trendelenburg gait. Chapter 4 discusses the advantages of this conceptual design and opportunities for future research. Chapter 5 contains the conclusion of this thesis. The appendices present the design process in chapter A and a more detailed overview of the prototype in chapter B.

Chapter 2

Literature Study

2.1 Introduction

Most people can walk without issues but for some this is not every day life. There are a lot of bones, muscles, nerves, and other parts active during the gait in order to make it happen. Failure of one of these can result in (partial) failure of the human gait. The introduction contains background information about the human gait, bones and muscles around the hip, hip weakness, and the Trendelenburg gait. The use of orthoses and flexures are introduced, and the objective of this research is stated.

2.1.1 Human gait

In order to understand the problem, we first take a look at the human gait. The human gait is split into two phases: the stance phase, when the foot is in contact with the ground, and the swing phase, when the foot is off the ground and swings forward to be put on the ground again. Figure 2.1 shows the gait cycle with its phases and sub phases. The stance phases takes up 60% and the swing phase 40% of the gait cycle. This means that both feet touch the ground at the same time at the beginning and end of the stance phase for about 20% of the total gait cycle. [1].

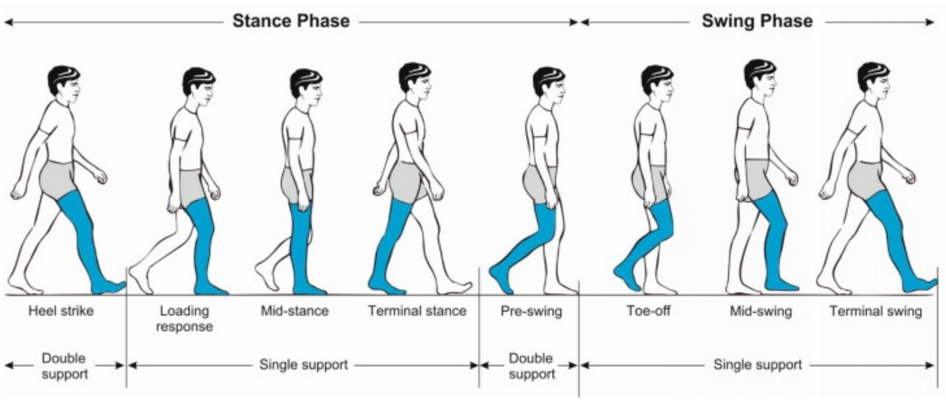
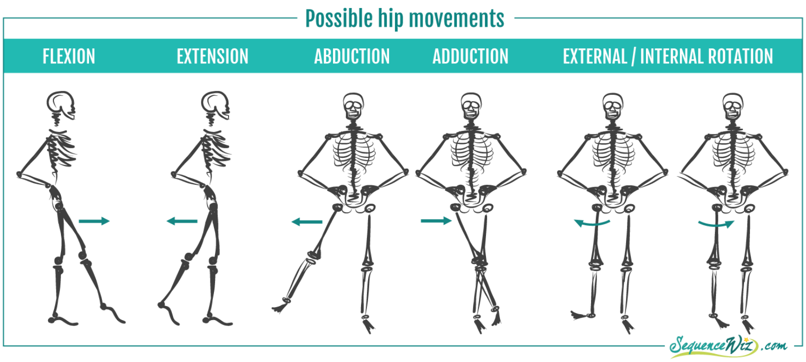
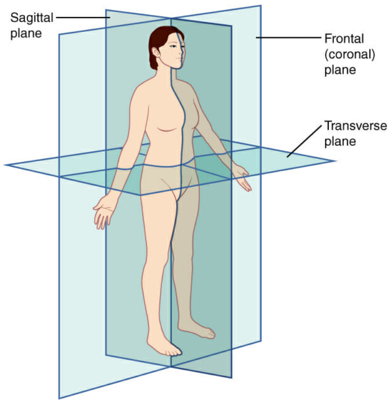


Figure 2.1: Human gait cycle [1]

During the gait cycle, the hip allows movement of the legs in order to make the steps needed to complete the cycle. The hip joint is a ball joint that allows for all three rotations of the upper leg with respect to the hip. Both directions of the three rotations are named as can be seen in figure 2.2a. Figure 2.2b shows the names for the planes of the human body. These terms will be used in this research.



(a) Names of the motions of the upper leg with respect to the hip [2]



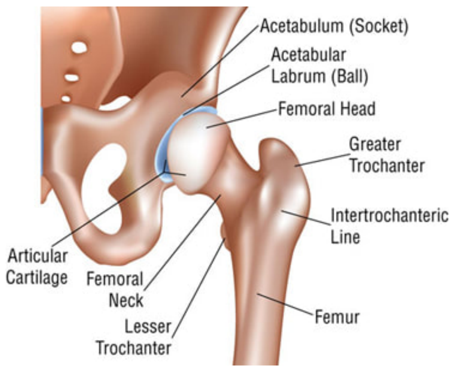
(b) Planes of the body [3]

Figure 2.2

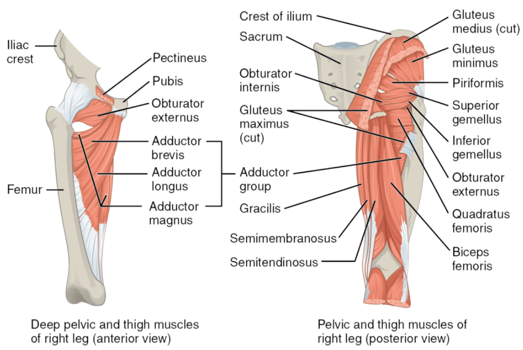
2.1.2 Bones and muscles of the hip

The hip joint has its socket in the pelvis (acetabulum) and the ball on the upper leg bone (femoral head) as can be seen in figure 2.3a. The attachment of the femoral head to the femur (femoral neck) is smaller in diameter to prevent clashing at large rotations. The muscles attach to the greater and lesser trochanter, which extends from the femur in order to create a larger moment arm.

Muscles actuate the bones to control the gait and maintain stability of the upper body. The muscles are visualized in figure 2.3b.



(a) Bones of the hip [4]



(b) Muscles of the hip [5]

Figure 2.3

2.1.3 Hip weakness

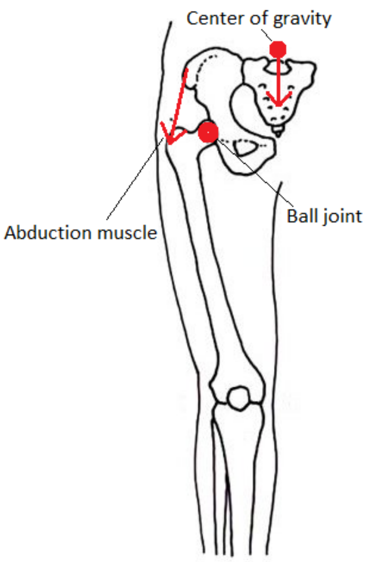
Pirker and Katzenschlager [1] investigated the hip disorders in adults and the elderly. They found that the prevalence of gait disorders increases from 10% in people aged 60–69 years to more than 60% in community dwelling subjects aged over 80 years. Besides elderly, also people who suffered a paraplegia can struggle with hip muscle weakness. A patient from Rijndam revalidatie that suffered a paraplegia was interviewed to get an insight in the requirements for the design. The interview is not appended to this research, but the conclusions from the interview are taken into account.

From all muscles used during gait, the hip abduction muscles are most sensitive to muscle weakness [6]. When all muscles are weakened gradually, the abduction muscles are the first to show negative impact on the gait. When one muscle is left out, gait is still possible except for the plantarflexors and gluteus medius. The first one is a muscle in the lower leg and the last one is one of the most important hip abduction muscles. When hip abduction muscles are weakened, the Trendelenburg gait is observed.

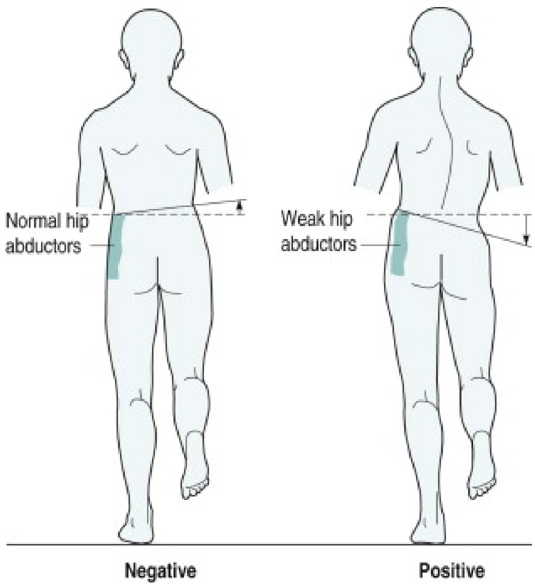
2.1.4 Trendelenburg gait

When the hip abductor muscles cannot deliver enough force in order to keep the pelvis in place, the pelvis drops to one side in the frontal plane, which is called Trendelenburg gait. This happens during the stance phase when only one foot is on the ground. The center of gravity stays approximately in the center (in frontal plane), but the support to the ground goes from two sides of the center to only one side. To keep the upper body upright, the hip abductor muscles pull between the greater trochanter and the pelvis to create a moment around the hip joint. This moment counters the moment generated by the gravitational force of the upper body. If the abduction muscles cannot deliver enough force, the moment of the upper body around the hip joint cannot be canceled, which results in the pelvis dropping to the non-supported side. Figure 2.4a shows this problem schematically with the direction of the forces.

When the pelvis drops, the upper body also drops which means that the person could fall. To prevent this, most people that suffer from Trendelenburg lean their upper body to the side of the weak hip abductor muscles. This way, the center of gravity moves towards the supporting leg, reducing the moment of the upper body around the foot. Figure 2.4b illustrates the stance with normal hip abductors and the Trendelenburg gait with the compensation of the upper body. On the long term, this compensation can result in other problems for example in the back.



(a) Forces of abductors and body weight



(b) Trendelenburg gait [7]

Figure 2.4

2.1.5 Orthoses

An orthosis is an "externally applied device used to modify the structural and functional characteristics of the neuromuscular and skeletal system" [8]. They can be attached to the human body to add stiffness where needed, for example to prevent a certain motion.

2.1.6 Flexures

Typically, systems that need to allow some motions contain parts that allow these motions, like sliders or bearings. The precision industry use flexures to allow these motions with the benefit of no wear and backlash that result in a more predictable motion. Another benefit of these flexures could be that they can be spatially designed to not stand out from an organic shape like the hip of a person. The shape of the flexure determines which motions it allows and which it obstructs. It also eliminates the need to (re)lubricate bearings.

2.1.7 Objective

The objective of this literature survey is to analyze the human body and gait to find important aspects in order to design the hip abduction orthosis. Solutions to similar problems are then investigated and evaluated.

2.2 Problem analysis

This section illustrates the challenges the design of the orthosis is expected to face. First, the functional analysis and the expected environment and use cases are explained. The range of motion and the expected forces on the orthosis are investigated, followed by the moment arm of the abductor muscles. Then the location of the joint and the tradeoff between stiffness and preload is touched upon. Finally, the requirements and wishes for the design of the orthosis are listed.

2.2.1 Functional analysis

The orthosis should support the abduction of the hip over the full range of motion, while allowing the other movements of the leg. There will be a tradeoff between these two, since one influences the other. The optimal solution is different per person since everyone differs in characteristics like muscle strength, length, etc. To make the orthosis useful , the ratio between the free stiffness and support stiffness should be as high as possible over the full range of motion.

2.2.2 Environment and use cases

The orthosis is worn around the hip during walking. This means it has to operate in the full range of outside temperatures (say between -10°C and 40°C) and could come into contact with water. It can also experience small impacts from walking into something or falling on the ground when accidentally dropping it.

Besides walking, for which the orthosis will be designed, there are other use cases that the orthosis needs to handle. These cases are listed below.

- Standing
- Sitting
- Standing up from and sitting down in a (wheel)chair
- Crouching
- Walking stairs
- Going to the toilet
- Putting the orthosis on and taking it off

Although support might be needed during these cases, this is not the scope of this research. What is important for the orthosis is that these cases are still possible to do without too much of a negative influence from the orthosis.

2.2.3 Range of motion of the leg

The orthosis should allow all other rotations (other than abduction/adduction) during the full range of motion. We therefor need to know how much a healthy person can rotate his/her leg in all directions. This is investigated for two cases: the maximum possible rotations that a person can move its leg, and the maximum rotations during walking. Both are listed in table 2.1.

Table 2.1: Maximum possible degrees of rotation between the hip and upper leg [7] and average degrees of rotation during gait [9]

Movement	Maximum possible degrees	Maximum walking degrees
Flexion	120° (lateral) - 150° (bilateral)	30°
Extension	10° - 15°	10° - 13°
Abduction	40°	5°
Adduction	25°	10°
External rotation	45°	4°
Internal rotation	35°	4°

2.2.4 Expected forces on the orthosis

The force that the orthosis needs to withstand follows directly from the body weight and moment arm. The gravitational force from the upper body gives a moment around the hip joint. The upper body weights approximately $\frac{2}{3}$ of the total body [10]. The moment arm is measured by Bardakos and Freeman which resulted in a mean value of 89.2 mm [11].
Say a person weights 75 kg, then the moment of his upper body around the hip joint is calculated from equation 2.1 to be 44 N m

$$M = \frac{2}{3} mgd \tag{2.1}$$

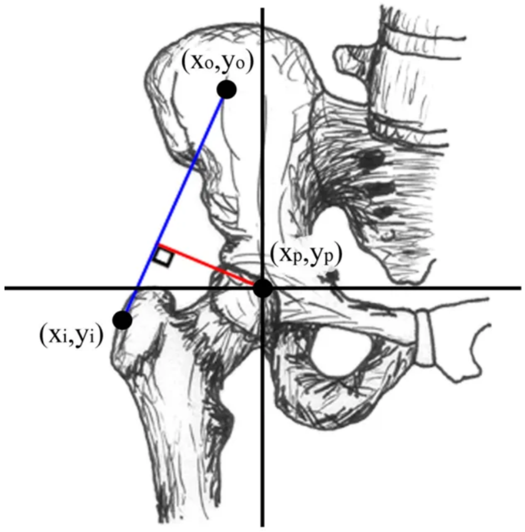
Where:

M = Moment of the upper body around the hip joint
m = Total mass of the person
g = Gravitational constant ($9.81\,\mathrm{m\,s^{-2}}$)
d = Moment arm from center of the upper body to the center of the hip joint

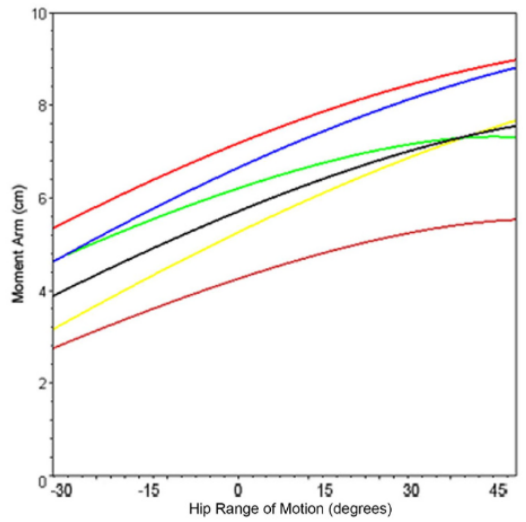
2.2.5 Abduction moment arm

The abduction muscles deliver force in order to deliver an opposite moment than the upper body around the hip joint. The moment arm of this force is not constant over the full range of motion. This is investigated by Henderson et al. [12] of which the results are given in figure 2.5. The moment arm increases when the leg abducts and decreases when the leg adducts. When the hip drops (observed in the Trendelenburg gait), this is the same motion as the adduction of the leg.

The muscle works optimal in its resting position. If there is strain in the muscle, the force that the muscle can deliver also decreases [13]. If the hips starts to drop because the abduction force is too low, two things happen. (1) The moment arm decreases, and (2) the muscle length increases. This means that even more force in the muscle is needed to keep the pelvis up right while the force that the muscle can deliver decreases. For the design of the orthosis this means that it should prevent even a bit of adduction. This way, the muscles can help out the most.



(a) "Coronal view of hip demonstrating hip abductor moment arm, defined as the length of a line originating at the joint center (red) which forms a 90° angle with the line of action (blue)." [12]



(b) "Line plot showing mean hip abductor moment arm through coronal plane motion for the normal femur (black) and femurs with greater trochanter displacements 2 cm medial (orange), 2 cm inferior (yellow), 2 cm (superior) green, 2 cm lateral (blue), 2 cm superior and 2 cm lateral (red)." [12]

Figure 2.5: Moment arm for abduction muscles over the full range of motion in frontal (coronal) plane.

2.2.6 Joint location

The leg rotates around the hip joint which is located inside the body. This makes it impossible (at least for the scope of this research) to place something close to the hip joint. If something, say a ball joint, is attached to the side of the hip, the center of rotation can only be matched for one of the three rotations. Any second rotation will never cross through the center of the hip joint and therefore, both other rotations will be obstructed by the part. Section 2.3.2 explains how this can be solved.

2.2.7 Stiffness versus preloaded force

Two working principles of the orthosis can be distinguished; stiffness and a preloaded force. With stiffness, it is meant that the resting position of the orthosis is the same as the leg. When the abduction/adduction occurs, the orthosis will support the leg due to the stiffness is that direction. With preloaded force, it is meant that the resting position of the orthosis lies somewhere in the abduction range of the leg. When the patient wears the orthosis, the adduction muscle (that is not weakened) has to deform the orthosis to its "resting position". This way, the adduction muscle can provide for both abduction and adduction movements. The downside is that the adduction muscle has to be stronger in order to overcome the extra force from the orthosis and is always active during walking. The two principles can also be combined in one design.

2.2.8 Requirements and wishes

There are a few requirements that the orthosis should fulfill in order to work. If these requirements cannot be met, the orthosis will not fulfill its purpose.

- The strength should be sufficient to not plastically deform for the whole range of motion.
- The lifetime should be X long. This is measured in amount of steps taken while wearing the orthosis and should be investigated further.
- The person wearing the orthosis should either be able to do daily activities (like sitting, walking stairs, going to the toilet) while wearing the orthosis or the orthosis should be easy to be taken off. Another option is to "flip a switch" that decreases the orthosis's support, but allows the daily activities without taking the orthosis off.
- The person wearing the orthosis should be able to put the orthosis on and take it of without help from someone else. The assumption here is that the person has only weakened hip abductors and nothing else.
- The orthosis should be printable.
- The orthosis should withstand splashes of water.
- The orthosis should withstand temperatures between $-10°C$ and $40°C$.

Besides the requirements, there are the wishes. The orthosis should be optimized towards these wishes to perform best.

- The free stiffness should be as low as possible.
- The support stiffness should be as close to the optimal value as possible over the full range of motion
- The forces on the body should be well distributed over the body.
- The mass of the orthosis should be minimized because the person wears the orthosis and has to carry the load.
- The orthosis should not stand out while wearing it.

The last wish: "the orthosis should not stand out while wearing it" is of high importance. This is concluded from the interview with the patient as explained in section 2.1.3 and will be one of the main focuses of the design of the orthosis.

2.3 State of the art

This chapter explains how similar problems as faced in the orthosis are tackled by literature. First, the support stiffness for compliant mechanisms is explained. Secondly, the influence of compliant mechanisms on the center of rotation is illustrated. A special group of compliant mechanisms that is potentially suitable for an orthosis is presented: compliant shell structures. Lastly, existing ortheses are shown and categorized.

2.3.1 Support stiffness

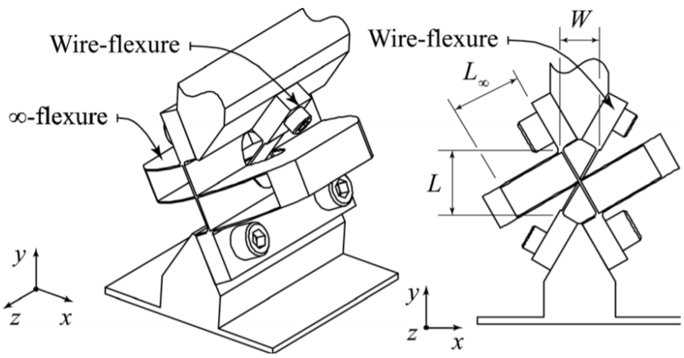
Compliant mechanisms lose their support stiffness when deflecting. Wiersma et al. [14] presented a method for optimizing the geometry of flexure hinges for support stiffness with reinforcement structures. The support stiffness is represented by the second eigenfrequency of the system. The second eigenfrequency is used to show stiffness in precision mechanisms because this second eigenfrequency typically limits the controllability of a system. At this frequency, the system is not stiff enough in the direction of the second eigen mode to control it within the set limits. Equation 2.2 shows the relation between the eigenfrequency and the stiffness.

$$\omega = \sqrt{\frac{k}{m}} \tag{2.2}$$

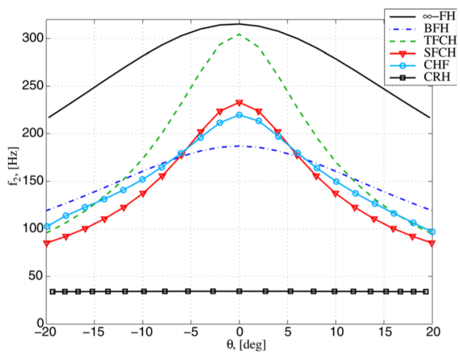
Where:

ω = Eigenfrequency
k = Stiffness
m = Mass

Wiersma et al. also present the ∞-flexure hinge; a hinge with torsional reinforcement structure that has a better support stiffness over its range of motion. Figure 2.6a and 2.6b show this ∞-flexure hinge and its second eigenfrequency as a function of the angle of deflection respectively. Although the flexure works better, it also takes up a lot more space. This is not desired for the orthosis.



(a) Isometric and front view of the ∞-Flexure hinge [14]



(b) Second eigenfrequency as a function of the angle of deflection [14]

Figure 2.6

2.3.2 Center of rotation

Section 2.2.6 showed that the joint location inside the human body can give some problems. Literature shows a clear solution to this: place flexures such that the instantaneous center of rotation is in the center of the ball joint.

The flexure design can be obtained using the Freedom and Constraint Topology (FACT) method [15]. The FACT-table shows all possibilities to constrain the system for the desired degrees of freedom. The mechanism can then be designed using building blocks like wire and leaf flexures. This is an useful method to quickly design a mechanism, but also often creates systems that take up a large space.

Another way to get the desired degrees of freedom is to combine different types of existing flexures. Although this does not solve the joint location problem, it is useful to see how parallel and series flexures behave. D. Farhadi Machekposhti et al. [16] added several flexures with one degree of freedom to obtain the desired system with three degrees of freedom. Each flexure adds one degree of freedom and they are connected in series. If the flexures were connected in parallel, the constraints of the flexures in one direction will obstruct the degrees of freedom of the others, resulting in a zero degree of freedom system. The opposite holds for constraints; adding constraints in series will not constrain the system, but adding them in parallel will.

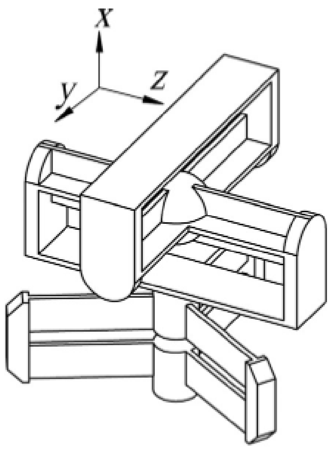


Figure 2.7: Several flexures is series [16]

Both methods have the disadvantage that they cannot take into account that the mechanism should fit the patient without standing out too far from the body. The methods will therefore result in big spatial flexures that are not desired.

Besides the method, a more general problem is that the instantaneous center of rotation does not stay at the same place when a flexure deforms. Figure 2.8 shows this problem.

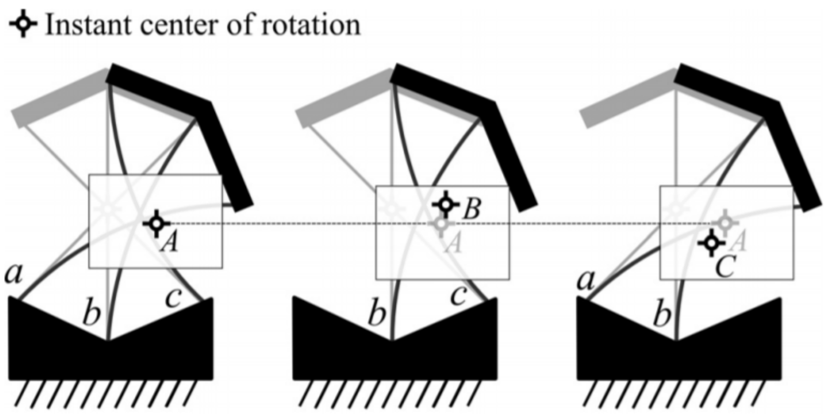
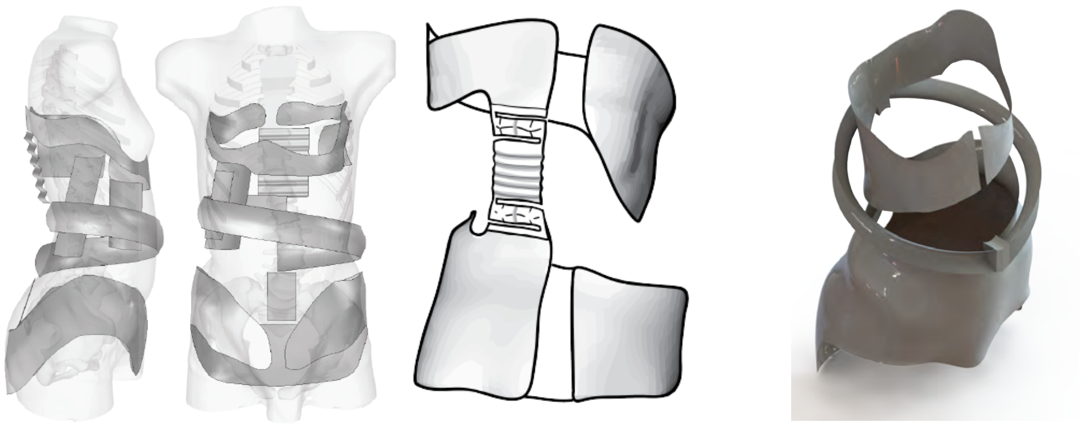


Figure 2.8: Displacement of the instantaneous center of rotation after deformation of the flexures [17]

2.3.3 Shell structures

Shell structures are spatially curved, thin-walled structures able to transfer or transmit, force, motion or energy through elastic deflection [18]. They have the advantage that they can be shaped around the body of the patient such that the mechanism does not stand out.

Joep Nijssen [19], Tim Dries [20], Joost Leemans [21], and Hylke Kooistra [22] researched compliant shell mechanisms with the design of a scoliosis orthosis as case study. Nijssen shows the use of ellipsoids to illustrate the compliant directions of a shell in the initial position. Dries and Leemans use a correction analysis to see where a orthosis should have stiffness in order to support the body. Kooistra chose a shell shape that has the desired degrees of freedom and changed the parameters for a more optimal shape. Figure 2.9 show the designs of the scoliosis orthoses.

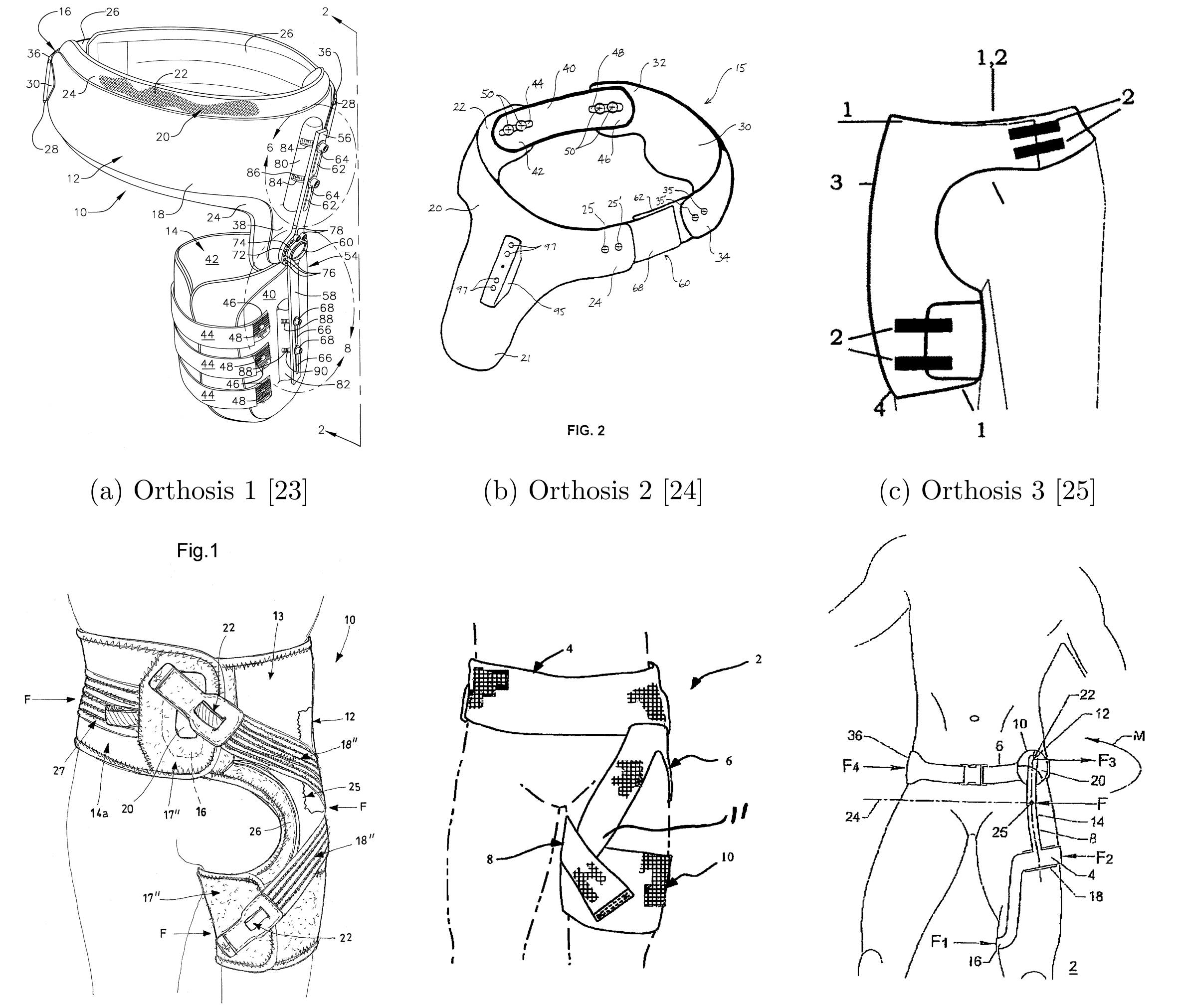


(a) Nijssen [19] (b) Dries [20] (c) Kooistra [22]

Figure 2.9

2.3.4 Orthoses

This section shows state of the art orthoses and how they overcome similar problems. Figure 2.10 shows several hip orthoses that will be explained.



(a) Orthosis 1 [23]

(b) Orthosis 2 [24]

(c) Orthosis 3 [25]

(d) Orthosis 4 [26]

(e) Orthosis 5 [27]

(f) Orthosis 6 [28]

Figure 2.10: State of the art hip orthoses

A notable difference between the orthoses is the stiffness and preload force as explained in section 2.2.7. Orthosis 1 and 2 (figures 2.10a and 2.10b) try to rigidly connect the leg to the pelvis, while orthoses 3, 4, 5, and 6 (figures 2.10d, 2.10c, 2.10e, and 2.10f) add flexible bands to support the muscle.

Orthosis 1 is the only one that uses a bearing to allow the motion of the hip, the others use flexibility. The joint location, as explained in section 2.2.6, makes it impossible to overlap more than one rotation. This results in constraining the internal/external rotation the besides the wanted abduction/adduction movements, making a normal gait impossible.

Figure 2.11 shows the same working principle in a knee orthosis. Since the knee is more of a revolute joint, this could be a working principle in the knee, but less in the hip.

(a) Bearing in a hip orthosis [29]

(b) Bearing in a knee orthosis [30]

Figure 2.11

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Chapter 3

Paper: Conceptual design of a compliant hip orthosis for Trendelenburg gait

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CONCEPTUAL DESIGN OF A COMPLIANT HIP ORTHOSIS FOR TRENDELENBURG GAIT

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ABSTRACT

Weakness of the hip abduction muscles can result in a gait disorder named Trendelenburg gait, which can lead to problems in the hip joint, knees, and ankles. In this paper, the conceptual design of a compliant hip orthosis to prevent Trendelenburg gait is presented. A theoretical analysis and measurements on a technical prototype show a high stiffness ratio between adduction and flexion-extension of the leg, and minimal shear forces from the orthosis on the human body while staying close to the human body.

1 INTRODUCTION

The human gait is most sensitive to weakness of the hip abduction muscles compared to other hip muscles [1]. During the single support phase, when only one leg is is on the ground, weakness of the abduction muscles causes the pelvis to droop. This abnormal gait is named Trendelenburg gait, which can lead to complications at the knees and ankles and accelerate wear of the hip joint [2].

To prevent Trendelenburg gait and to avoid surgery, an orthosis can constrain adduction of the leg, the motion that otherwise would be constrained by the abduction muscles. At the same time, the orthosis should allow the required motion between the leg and hip during walking. Because internal-external rotations are small during gait [3], and adduction needs to be prevented by the orthosis, flexion-extension of the leg is defined as the required walking motion in this paper.

Hip orthoses to aid Trendelenburg patients have not received much attention in the literature. A few hip orthoses exist [4–9], of which two main working principles can be distinguished. Lerman's Post-operative hip abduction orthosis [4] connects the upper body and leg with a revolute joint on the side of the hip. This joint allows flexion-extension of the leg while constraining adduction of the leg. The design of Weissleder et al. [6] connects the leg to the upper body using elastic bands to introduce a constant abduction moment around the hip joint.

Both designs have drawbacks. In Lerman's orthosis, the revolute joint constrains five degrees of freedom (DoF), while only one DoF needs to be constrained: adduction of the leg. Flexion-extension of the leg is only possible if the rotation axis of the revolute joint and hip joint perfectly align. Compliance in the orthosis and the human body make this possible, but this also decreases prevention of adduction from the orthosis. In Weissleder's orthosis, the constant force from the elastic bands requires other muscles to be active at all times when wearing the orthosis. Both Lerman's and Weissleder's orthosis introduce mainly shear forces on the body, which enables creep to occur. Also, because the human skin has a lower shear than pressure stiffness [10], the effectiveness of the orthosis is decreased.

A new field in orthosis designs are compliant mechanisms. Compliant mechanisms transfer or transmit, force, motion, or energy through elastic deflection [11]. Benefits are that they do not need lubrication and can be designed around the body to be aesthetically appealing. The free and constrained motions of compliant mechanisms can be found by interpreting the geometry as ideal flexures [12].

This paper introduces the conceptual design of a compliant hip orthosis for Trendelenburg gait, and presents proof of concept using a technical prototype. This concept only constrains one additional DoF next to adduction, while leaving flexion-extension of the leg free and does not need a constant force from other muscles. Also, shear forces from the orthosis on the human body are low compared to the pressure forces.

The structure of the paper is as follows. Section 2 starts with a problem analysis, followed by an overview of the conceptual design. The influence of the orthosis' dimensions on the stiffness is investigated, and the setup of a prototype is explained. The results are presented in section 3, showing both the adduction and flexion-extension stiffness, and the forces and moments experienced on the human body as measured on the prototype. Section 4 contains the discussion, and section 5 the conclusion.

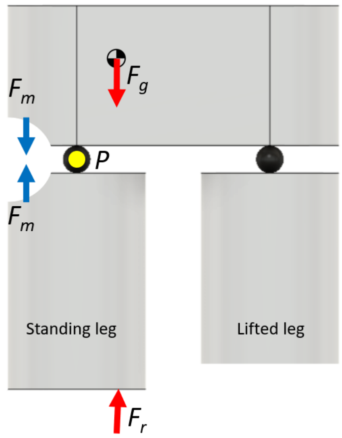

FIGURE 1: Free body diagram in frontal plane of a person with well working hip abduction muscles standing on one leg. The moment resulting from force F_g around hip joint P is cancelled by the force F_m from the abduction muscles.

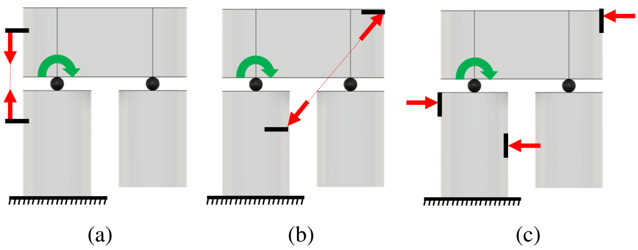

FIGURE 2: Three possible ways to add an abduction moment using an orthosis with the moment from the upper body in green and the forces from the orthosis on the body in red. The black lines represent attachments to the body.

2 METHOD

This section first defines objectives that the orthosis should be optimised for in section 2.1. Next, section 2.2 introduces the conceptual design of the compliant hip orthosis and explains the working principle. Section 2.3 shows a constraint analysis and section 2.4 a force analysis. A dimension analysis is done in section 2.5 to determine the influence of the orthosis' dimensions on the stiffness, and how these dimensions should be chosen for an effective compliant orthosis design. Finally, a technical prototype is presented in section 2.6.

2.1 Problem Analysis

When standing on one leg, the weight of the upper body and lifted leg introduce a moment around the hip joint. This moment is cancelled by contraction of the hip abduction muscles as shown in figure 1. Because the hip abduction muscles are weakened when Trendelenburg gait is observed, this moment cannot be cancelled anymore leading to drooping of the pelvis. Instead, the orthosis should provide the abduction moment to prevent the pelvis from drooping.

Next to this main requirement, three additional factors are defined that the orthosis should be optimised for. Firstly, flexion-extension of the leg should be obstructed as little as possible while maintaining the needed abduction moment to support the upper body. Because the orthosis designed in this paper is a compliant mechanism, all motions of the leg will deform the orthosis, and will therefore cost effort to the patient. To minimise the walking effort while keeping the needed supporting moment, the stiffness ratio of the adduction over flexion-extension stiffness should be maximised. Secondly, the orthosis should introduce minimal shear forces on the human body to minimize creep and reduce the effect of skin stiffness on the effectiveness of the or-

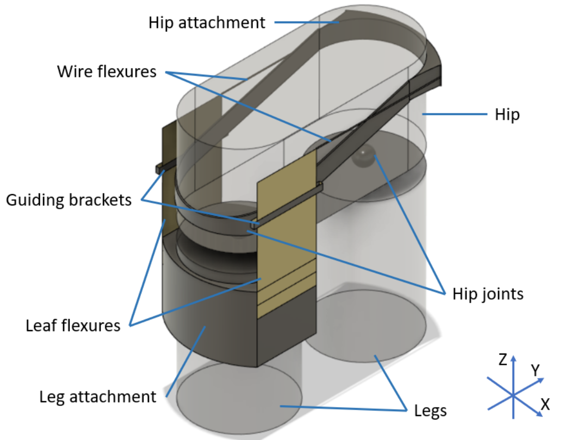

FIGURE 3: 3D Model of the conceptual design of the compliant hip orthosis. The hip and leg attachments are to visualize only and are not designed to perform as desired. The hip attachment is simplified in other pictures to prevent obstruction of view.

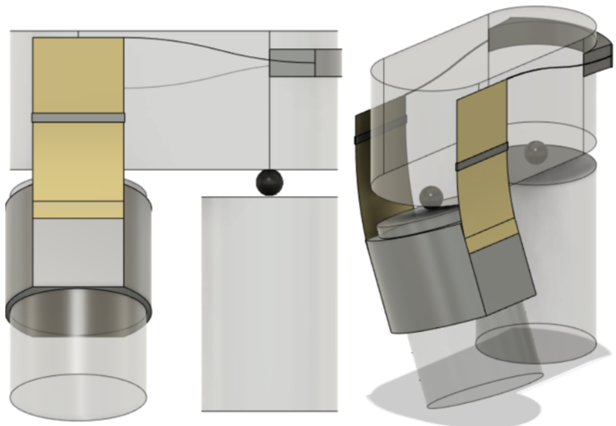

FIGURE 4: Front, and isometric view of the deformed orthosis with a 20° flexion angle of the leg. The attachment between guiding brackets and hip are hidden.

thosis. Lastly, the orthosis should not protrude from the human body. If the orthosis is close to the body, it can be hidden behind clothes, and the patient would not be distinguished by his/her disability to walk.

Figure 2 shows three ways to introduce an abduction moment to the human body. Figure 2a shows forces on the side of the hip, which is done in the concepts of Lerman and Weissleder. This shows how the direction of the supporting forces are tangential to the body, introducing high shear forces on the skin. Figure 2b introduces less tangential forces on the body, but also obstructs the motion of the other leg and is loaded in compression, which is undesired for compliant mechanisms. The last concept depicted in figure 2c exerts only pressure forces on the human body and is chosen for the compliant hip orthosis in this paper. A more in depth analysis of the forces in this concept is done in section 2.4.

2.2 Working Principle Of The Orthosis

Figure 3 shows a 3D model of the conceptual design with names of the parts and the coordinate system that will be used throughout this paper. The orthosis consists of a thin rectangular plate (leaf flexure) that connects to the thigh at the leg attachment and reaches to the upper body. From there, a wire flexure connects the top of the leaf flexure horizontally to the other side of the upper body to the hip attachment. A guiding bracket that can slide over the leaf flexure in YZ-plane guides the flexure with respect to the hip attachment. A leaf, wire, and guiding bracket are used on both the front and back of the person.

Figure 4 shows the deformation of the orthosis during flexion of the leg. The flexure is kept close to the human body by the guiding bracket and leg attachment. This ensures that the leaf flexure bends the same amount as the leg rotates. Because the leg rotates around the hip joint, the distance between the guiding bracket and leg attachment changes. The guiding bracket allows the vertical translation of the leaf, and the wire bends downwards in the front and upwards in the back to compensate for the vertical translation.

2.3 Constraint Analysis

A leaf flexure constrains three DoF; the translations in the direction of the length and width of the leaf, and the rotation around the axis perpendicular to the surface of the leaf. A wire flexure only constrains the translation in the direction of the wire. Because one translational constraint of the leaf flexure and the constraint of the wire flexure align in the design, the combination of the leaf and wire flexure connected in series constrains one DoF, the translation at, and in the direction of the wire. Figure 5 shows the constraints of the leaf, wire, and series connection of both as used in the orthosis. The hip joint is considered as a ball joint that constrains three DoF; all three translations through the joint. Therefore, the human body in combination with the orthosis constrains five DoF. The constraint lines are illustrated in figure 6. The remaining DoF is the rotation around the x-axis in the hip joint. This motion is the flexion-extension of the leg, and is the main required motion during walking as explained in section 1.

2.4 Force Analysis

As stated in section 2.1 , it is important for comfort and performance of the orthosis to minimize shear forces on the human body. The forces of the orthosis on the body in the mid-stance,

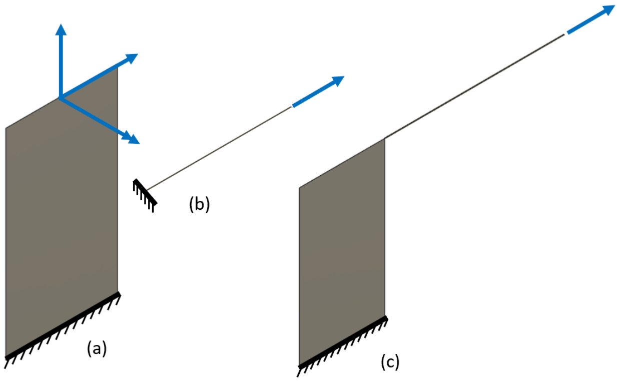

FIGURE 5: Constraints of a leaf flexure (a), wire flexure (b), and the series connection as used in the orthosis (c), showing the aligning constraint of the leaf and wire flexure in the series connection. Single arrows represent translational constraints, the double arrow represents a rotational constraint.

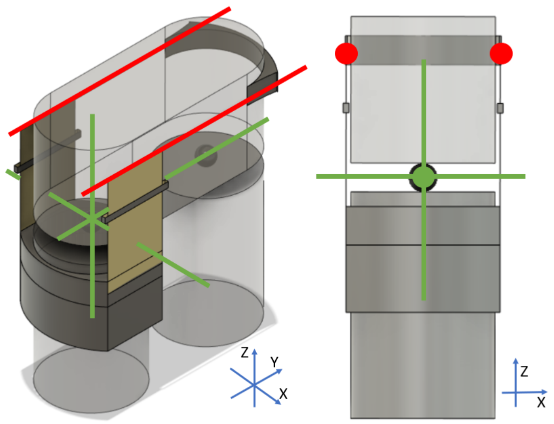

FIGURE 6: Isometric and side view of the design with three translational constraint lines of the hip in green and two translational constraint lines of the orthosis in red. The remaining DoF is the rotation around the y-axis through the hip joint, which is flexion-extension of the leg.

are modelled by the free body diagram of the leg and hip in figure 7. A moment around the hip joint originates from the gravitational force F_g from the upper body and lifted leg at a distance from the joint. Normally, the hip abduction muscle would create an opposing moment, but this muscle is weakened in the case of Trendelenburg gait. Instead, the orthosis creates this opposing moment by adding a force $F_{o,1}$ at a distance from the joint. The resulting forces $F_{h,y}$ and $F_{h,z}$ in the hip joint are equal but opposite to F_g and $F_{o,1}$. The gravitational force can be resolved by a reaction force at the ground, but the horizontal reaction force

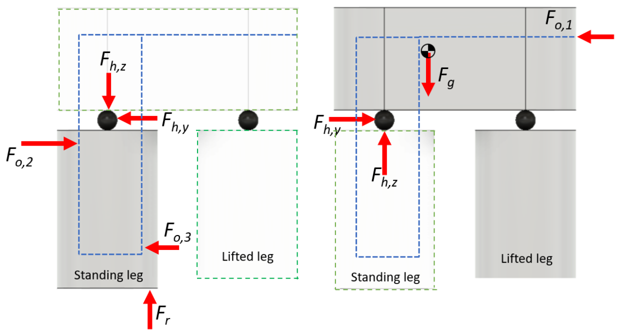

FIGURE 7: Free body diagram of the leg (left) and hip (right), showing no shear forces from the orthosis on the body during mid-stance. The mass of the orthosis is neglected.

$F_{h,y}$ cannot be resolved by the ground. Two opposing forces, $F_{o,2}$ and $F_{o,3}$, are needed on the supported leg to create a moment and force balance. During mid-stance, all forces of the orthosis on the body are horizontal forces that act in the frontal plane. Due to symmetry of the orthosis in the frontal plane, it is possible on both the upper body and leg to design an attachment of the orthosis such that the force on the body is mainly a pressure force.

2.5 Dimension Analysis

To reach a high adduction over flexion-extension stiffness ratio, the effect of the dimensions on the stiffness are investigated. The stiffness of the leaf and wire flexure in both directions are determined and expressed in the dimensions of the orthosis using linear beam theory. Based on this, the stiffness ratio is determined for a single leaf and wire flexure. This demonstrates which dimensions should be maximized or minimized to obtain a high adduction stiffness while keeping a low flexion-extension stiffness.

The stiffness of the leaf and wire flexure in both directions are determined separately and then added for the total stiffness. The flexion-extension stiffness is defined by the moment M_{fe} around the y-axis through the hip joint, needed to rotate the leg ϕ_y radians around the y-axis. The adduction stiffness is defined by the moment M_d around the x-axis through the hip joint, needed to rotate the leg ϕ_x radians around the x-axis. The dimensions used in the equations below are defined in figure 8. The leaf flexure has a length L_l, width b_l, and thickness t_l. The wire flexure has a length L_w and diameter D_w. a_x Is the distance along the x-axis between the hip joint and the orthosis, $a_{z,1}$ is the distance along the z-axis between the hip joint and the wire flexure, and $a_{z,2}$ is the distance along the z-axis between the hip joint and the guiding bracket. L_g Is defined as the distance from the bottom of the leaf up to the guiding bracket and can be written as $L_g = L_l - a_{z,1} + a_{z,2}$. Furthermore the Young's modulus E is used.

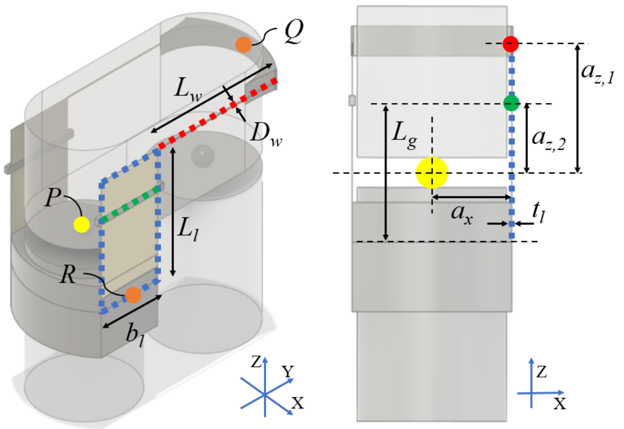

FIGURE 8: Definitions of orthosis' dimensions used in the stiffness calculations in isometric view and in xz-plane. Points Q and R show the position of the sensor measuring the forces and moments on the body in the prototype as explained in section 2.6.

2.5.1 Flexion-extension Stiffness For flexion-extension of the leg, a moment around the hip joint is modelled as a force on the leaf flexure at the guiding bracket. The bottom of the leaf flexure is clamped at the attachment point of the flexure to the leg. Linear beam theory shows that an angular rotation as a result of a force at a distance is determined by equation 1.

$$\phi_1 = \frac{F_1 L_1^2}{2EI_1} \tag{1}$$

Substituting $\phi_1 = \phi_y$, $F_1 = \frac{M_{fe}}{a_{z,2}}$, $L_1 = L_g$, and $I_1 = \frac{1}{12} b_l t_l^3$, and rewriting the equation results in the flexion-extension stiffness of the leaf flexure $k_{fe,l}$ as can be seen in equation 2.

$$k_{fe,l} = \frac{M_{fe}}{\phi_y} = \frac{E b_l t_l^3 a_{z,2}}{6L_g^2} \tag{2}$$

The wire flexure only bends vertically to allow the vertical extension due to the position of the center of rotation, as can be seen in figure 4. Both ends of the wire are assumed to be clamped and do not allow rotation. The displacement of the end of the wire flexure can be determined using equation 3.

$$\delta_1 = \frac{F_2 L_2^3}{12EI_2} \tag{3}$$

Substituting $\delta_1 = \frac{\phi_y a_x}{2}$, $F_2 = \frac{M_{fe}}{a_x}$, $L_2 = L_w$, and $I_2 = \frac{\pi}{64} D_w^4$ and rewriting the equation results in the flexion-extension stiffness of the wire flexure $k_{fe,w}$ as can be seen in equation 4.

$$k_{fe,w} = \frac{M_{fe}}{\phi_y} = \frac{3\pi D_w^4 E a_x^2}{32 L_w^3} \tag{4}$$

Because the leaf and wire flexures are connected in parallel for flexion-extension of the leg, the stiffness of both flexures can be added to determine the total flexion-extension stiffness of the leaf and wire flexure k_{fe}, as can be seen in equation 5.

$$k_{fe} = k_{fe,l} + k_{fe,w} \tag{5}$$

2.5.2 Adduction Stiffness For adduction, a moment around the hip joint is modelled as a force in the longitudinal direction of the wire flexure. The equation for an angular rotation as a result of a force at a distance is already given in equation 1. Substituting $F_1 = \frac{M_d}{a_{z,1}}$, $L_1 = L_l$, and $I_1 = \frac{1}{12} b_l^3 t_l$, and rewriting the equation results in the adduction stiffness of the leaf flexure $k_{d,l}$ as can be seen in equation 6.

$$k_{d,l} = \frac{M_d}{\phi_x} = \frac{E b_l^3 t_l a_{z,1}}{6L_l^2} \tag{6}$$

The wire flexure is loaded in its longitudinal direction. The equation for stretching of a wire is given in equation 7.

$$\delta_2 = \frac{F_3 L_3}{EA} \tag{7}$$

Substituting $\delta = \phi_x a_{z,1}$, $F_3 = \frac{M_d}{a_{z,1}}$, $A = \frac{1}{4}\pi D_w^2$, and $L = L_w$, and rewriting the equation results in the adduction stiffness of the wire flexure $k_{d,w}$ as can be seen in equation 8.

$$k_{d,w} = \frac{M_d}{\phi_x} = \frac{\pi D_w^2 E a_{z,1}^2}{4L_w} \tag{8}$$

Because the leaf and wire flexure are connected in series for adduction of the leg, the total adduction stiffness of the leaf and wire flexure k_d can be determined according to equation 9.

$$k_d = \frac{1}{\frac{1}{k_{d,l}} + \frac{1}{k_{d,w}}} \tag{9}$$

2.5.3 Stiffness Ratio For the design of the orthosis, it is important to have high adduction stiffness and low flexion-extension stiffness. The dimensions of the orthosis influence both stiffnesses but the amount in which they do is not the same. By defining the equation of the ratio of adduction over flexion-extension stiffness, a sensitivity of this ratio with respect to the dimensions of the orthosis can be determined. Equation 10 shows this ratio for the leaf flexure. If the ratio is maximized, the desired result is obtained. Therefore, the width of the leaf flexure (b_l), the distance from leg attachment to guiding bracket (L_g), and the vertical distance between the hip joint and the wire flexure ($a_{z,1}$) should be as high as possible, while the thickness of the leaf flexure (t_l), the length of the leaf flexure (L_l), and the vertical distance between the hip joint and the guiding bracket ($a_{z,2}$) should be as small as possible. Note that L_g is dependent on L_l, and needs to be smaller than L_l in order to make flexion-extension possible (because the guiding bracket slides over the leaf flexure).

$$\frac{k_{d,l}}{k_{fe,l}} = \frac{b_l^2 L_g^2 a_{z,1}}{t_l^2 L_l^2 a_{z,2}} \tag{10}$$

The same equation can be determined for the wire flexure, as can be seen in equation 11. The equation shows that the vertical distance between the hip joint and wire flexure ($a_{z,1}$), and the length of the wire flexure (L_w) should be maximized, while the diameter of the wire flexure (D_w), and the horizontal distance between the hip joint and wire flexure (a_x) should be minimized.

$$\frac{k_{d,w}}{k_{fe,w}} = \frac{8a_{z,1}^2 L_w^2}{3D_w^2 a_x^2} \tag{11}$$

2.6 Technical Prototype

A technical prototype is built to measure the stiffness of the orthosis and measure the forces on the body. Figure 9 shows the prototype. Aluminium extrusion profiles are used for the leg and upper body that are connected with a ball joint as the hip joint. The leaf and wire flexures are made out of spring steel and 3D printed PLA is used for the guiding bracket. The dimensions of the orthosis are chosen from available materials to be close to the average hip size of humans, and can be found in table 1. The names used for the dimensions are the same as explained in section 2.5 and illustrated in figure 8.

The flexion-extension stiffness of the prototype is measured using a force/torque sensor (ATI mini-40) to measure the input moment and an inclinometer (Seika NG4i) to measure the flexion-extension angle. The adduction stiffness is measured using the same sensor to measure the input moment. Because of a lower displacement due to higher adduction stiffness, a laser displacement sensors (Micro-epsilon optoNCDT 1750) is used to measure the adduction angle. Because this sensor measures a distance instead of an angle, the change in distance has to be transformed to a rotation around the hip joint. Figure 9 shows the prototype to measure the adduction stiffness. In this figure, the force/torque sensor is used to measure a force at a distance from the hip joint, that is translated to a moment around the hip joint. To measure the flexion-extension stiffness, the sensor is rotated 90° around the x-axis, such that a measuring axis aligns with the flexion-extension rotation axis. A moment is applied to the sensor, and the resulting rotation of the upper body is measured in the inclinometer.

To measure the forces and moments on the body, the force/torque sensor is first placed between the hip and the orthosis, and then between the leg and the orthosis. Figure 8 shows the position of the force/torque sensor with points Q and R.

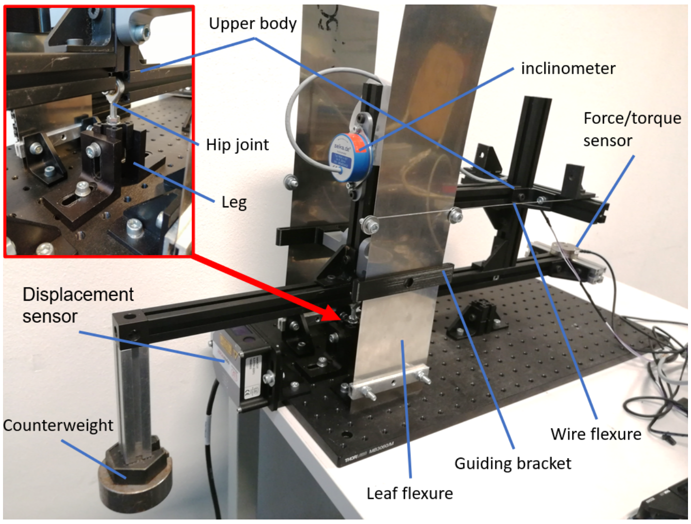

FIGURE 9: Prototype setup as used for measuring adduction stiffness.

TABLE 1: Prototype dimensions.

Dimension	Value
Leaf length (L_l)	0.18 m
Leaf width (b_l)	0.10 m
Leaf thickness (t_l)	0.50×10^{-3} m
Wire length (L_w)	0.21 m
Wire diameter (D_w)	1.0×10^{-3} m
Horizontal distance between leaf and joint (a_x)	0.080 m
Vertical distance between wire and joint ($a_{z,1}$)	0.13 m
Vertical distance between Guiding bracket and joint ($a_{z,2}$)	0.065 m

3 RESULTS

Section 2.1 explains three important factors for the design of the orthosis of which two are measured on the prototype . The results of the flexion-extension over adduction stiffness ratio measurement are shown in section 3.1. Section 3.2 shows the results of the forces and moments that the orthosis exerts on the body in the prototype. The amount of protrusion of the orthosis from the body is not measured on the prototype.

3.1 Stiffness Results

Figures 10 and 11 show the moment needed to rotate the hip in adduction and flexion-extension respectively. The blue dots show the data measured from the prototype. Figure 11 shows an increasing hysteresis at large angles of which the cause is discussed in section 4. The "Data fit" line is the average of the data points, except the data points that follow the vertical path at the highest and lowest angles of flexion-extension, because it is anticipated that this is not the stiffness but friction that is measured. The ratio between the measured average stiffnesses is $\frac{k_d}{k_{fe}} = 1.7 \times 10^3$. The "Linear beam theory" line represents the moment-angle behaviour of the orthosis according to linear beam theory as explained in section 2.5. "Finite element analysis" shows the behaviour of a non-ideal orthosis, simulated in COMSOL Multiphysics®. It is observed that the leaf flexures are not perfectly straight, but are pre-curved in the compliant direction. The pre-curve is measured by assembling the leaf flexures with their pre-curve in the same direction. The inclinometer is then used to measure how far the upper body rotates, which resulted in 5°. In this simulation, the orthosis is simulated to have this pre-curve in the leaf. Also, the stiffness of the test setup is simulated and added to this simulation.

3.2 Body Forces Results

Figures 12 and 13 show the forces and moments that the orthosis exerts on the upper body and leg. This is done by starting the measurement in mid stance without load, then adding an adduction moment, followed by a motion of the upper body representing the single support phase of the gait cycle. The introduction of the load and start of the cycle are indicated by the vertical lines. The three sub-phases of the single support phase are indicated with the vertical dashed lines. The adduction moment used in each case is different due to sensor limits.

Looking at the magnitude of the force on the body when the adduction moment is introduced, pressure forces $F_{Q,y}$ on the upper body and $F_{R,y}$ on the leg show the largest change. Shear forces $F_{Q,x}$ and $F_{Q,z}$ on the upper body show no change, and shear forces $F_{R,x}$ and $F_{R,z}$ on the leg show a change that is 22.4 and 19 times smaller than pressure force $F_{R,y}$ respectively. There is no magnitude change in the moments on the upper body when the load is introduced. On the leg, moment $M_{R,x}$ shows the

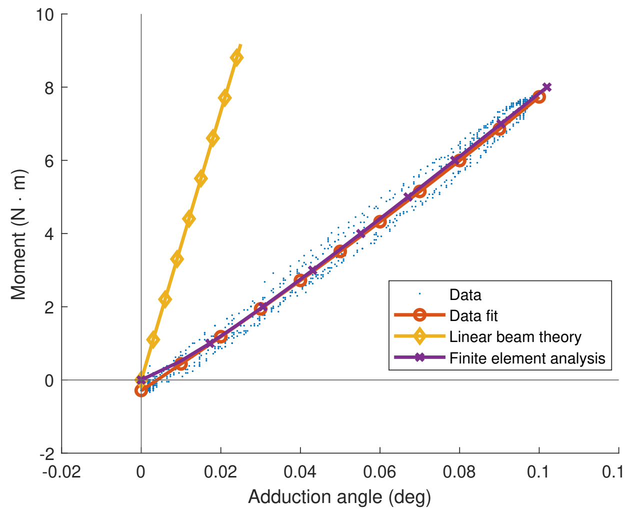

FIGURE 10: Moment-angle behaviour in adduction of measurements from the prototype, a fit through these data points, from Linear beam theory as predicted in section 2.5, and from finite element analysis using pre-curved leaf flexures of 5° and the stiffness of the prototype frame.

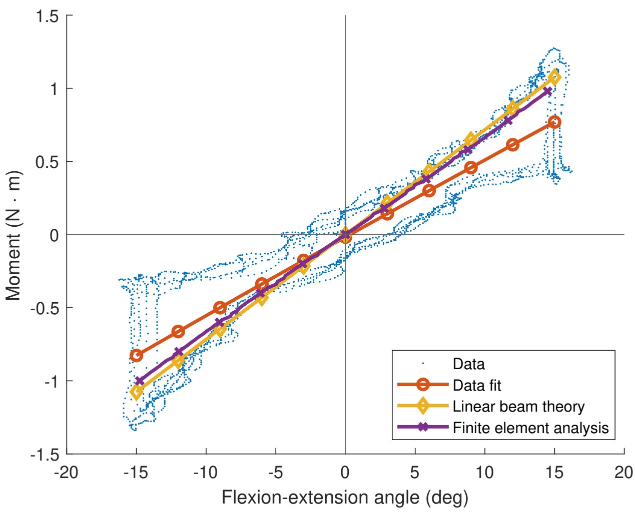

FIGURE 11: Moment-angle behaviour in flexion-extension of measurements from the prototype, a fit through these data points, from Linear beam theory as predicted in section 2.5, and from finite element analysis using pre-curved leaf flexures of 5° and the stiffness of the prototype frame.

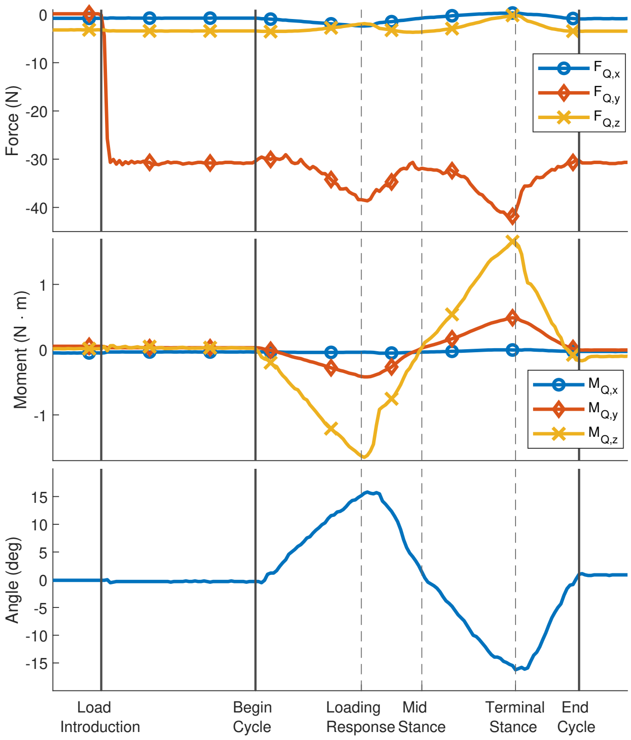

FIGURE 12: Forces and moments on the upper body during the single support phase of the gait cycle with an adduction moment of $6\,\mathrm{N\cdot m}$ measured at the side of the hip in point Q, as defined in figure 8

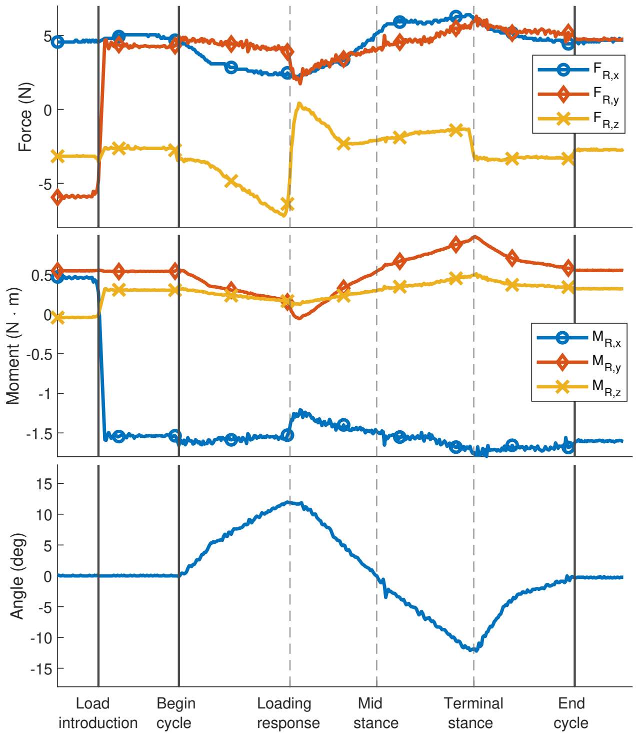

FIGURE 13: Forces and moments on the supported leg during the single support phase of the gait cycle with an adduction moment of $2\,\mathrm{N\cdot m}$ measured at the front of the leg in point R, as defined in figure 8

biggest change, $M_{R,y}$ shows no change, and $M_{R,z}$ shows a 5.8 times smaller change than $M_{R,x}$.

4 DISCUSSION

The orthosis in the prototype has an adduction stiffness that is three orders of magnitude higher than the flexion-extension stiffness, resulting in a low additional walking effort while preventing the upper body to droop. In addition, shear forces on the body are at least a factor 19 lower than the pressure forces, making the orthosis more comfortable and preventing creep of the orthosis to occur. Finally, the orthosis consists of thin flexures tangential to the body which do not protrude from the body during the gait, making it aesthetically appealing. Therefore, it is believed that this conceptual design could be developed into a well-working hip orthosis for Trendelenburg gait.

It is clear from figures 12 and 13 that the pressure forces $F_{Q,y}$ on the upper body and $F_{R,y}$ on the leg increase when the load is introduced. The shear forces $F_{Q,x}$ and $F_{Q,z}$ on the upper body and $F_{R,z}$ on the leg show very little change. The force $F_{R,x}$ on the leg is a pressure force but does not change as much when the load is introduced. This shows that the orthosis exerts mainly pressure forces on the body. The measured moments can always be cancelled by pressure forces on the body because they are measured outside the body. During the single support phase, the forces on the upper body change less than when the load is introduced. On the leg, the force $F_{R,z}$ shows a large change at the loading response. It is thought that the friction in the guiding bracket is the cause, because the friction force also acts along the z-axis and the direction that the bracket slides over the leaf flexure changes at this moment in the cycle. The same happens at the terminal stance, but because forces and moment on the leg are only measured on one side, the smaller curve of the flexure is not in line with the sensor, and this change is not as large as at the loading response. The moments on the upper body show a larger change during the cycle than when the load is introduced. This is because, during mid-stance, the forces act on both sides of the body, cancelling the moments. When the upper body rotates, these forces act in different directions, introducing moments.

One of the benefits of this design is that it largely avoids shear forces and exerts mainly pressure forces on the body. This increases comfort, but potentially also stiffness of the attachment, because the human skin has a higher stiffness in compression than shear. This makes it easier to design a stiff attachment of the orthosis to the human body.

The data in figure 11 of the flexion-extension stiffness shows a large hysteresis loop and a jerky behaviour when following a single cycle. It is thought that this is the effect of high friction in the guiding bracket. This bracket is made from 3D-printed PLA which acts as a sliding bearing, sliding over the leaf flexure during flexion-extension. Because the test is done in both directions, the hysteresis can be eliminated by averaging both directions, but the jerky behaviour can affect the precision of the measurement. This effect is minimized by measuring during multiple motions and averaging all data. Nevertheless, it is still an unwanted characteristic for the final orthosis. A new, lower friction, guiding bracket that still does not protrude far from the body could improve the functioning of the orthosis.

Buckling in the leaf flexure is observed when the adduction moment becomes too high. Buckling should be prevented since the adduction stiffness drastically decreases when his happens. This can be done in three ways: decreasing the length (L_l), increasing the width (b_l), and/or increasing the thickness (t_l). Decreasing the length of the leaf has no effect on the stiffness according to equation 10, but a minimum length is needed to make a certain flexion-extension angle. Increasing the width of the leaf flexure increases the adduction stiffness of the orthosis, but will result in the orthosis protruding from the body. Increasing the thickness increases the flexion-extension stiffness. These values need to be chosen such that the desired characteristics for a specific case are met while preventing buckling. The guiding bracket constrains the rotation around the z-axis, which can also help to prevent buckling.

Resulting forces from the orthosis are passed down through the hip joint. Since the muscle has a shorter moment arm to the hip joint than the orthosis, the resulting force due to the orthosis is expected to be lower than the force from the muscle. The direction of the resulting force in the hip joint will be different in both cases. Although both push the ball of the femur in the socket of the pelvis, the angle at which this happens is different. The effects of this need to be further investigated.

5 CONCLUSION

In this paper, the conceptual design of a new compliant hip orthosis for Trendelenburg gait is presented.

Three main design objectives for the orthosis are defined, (1) the adduction over flexion-extension stiffness ratio should be maximized, (2) shear forces on the body should be low compared to the pressure forces to the body, and (3) the orthosis should be close to the body while walking.

A dimension analysis using ideal flexures in linear beam theory shows the effect of the orthosis' dimensions on stiffness and how these dimensions should be chosen to maximize the adduction over flexion-extension stiffness ratio. A ratio of 1.7×10^3 is measured on the technical prototype.

A force analysis shows no shear forces from the orthosis on the human body during mid-stance phase. Measurements on the prototype at the hip attachment show no change in shear forces when an adduction moment of $6\,\mathrm{N \cdot m}$ is introduced. At the leg attachment, 19 times lower shear force than pressure force is measured when an adduction moment of $2\,\mathrm{N \cdot m}$ is introduced.

The use of thin flexures tangential to the human body and a guiding bracket that ensures the flexures to stay tangential during gait, results in the orthosis to stay close to the body while walking.

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Chapter 4

Discussion

This thesis presents the conceptual design of a compliant hip orthosis. First, the literature study analyses the human body and gait to find important aspects in order to design a compliant hip orthosis. The motions of the leg with respect to the hip during gait give a good indication for the deflection of the orthosis. An analysis of Trendelenburg gait from an engineering perspective is used to investigate where forces need to be added to support the upper body. This is compared to existing hip abduction orthoses which are also categorised by their working principle.

The three main design objectives are defined for the orthosis: to obtain a high adduction over flexion-extension ratio, low shear forces on the human body, and little protrusion of the orthosis from the human body. Measurements on the technical prototype show an adduction over flexion-extension stiffness ratio of 1.7×10^3 and at least 19 times lower shear forces than pressure forces on the body. These results are already discussed in the paper in chapter 3. The exact amount of protrusion of the orthosis from the human body is dependent on the attachment of the orthosis, which is not covered in this thesis. However, the technical prototype shows how the orthosis stays tangential to the body during gait which indicates that the protrusion will be minimal.

It can be concluded that the conceptual design works as desired. Future research has to be done to determine if the design also lives up to its potential when applied to humans. The next section gives an overview of future research that could be done.

4.1 Future research

This thesis presents the conceptual design of the compliant part of the orthosis but not the attachment of this part to the human body. The places where forces on the body are introduced and the direction of these forces are investigated in this thesis. This can be used in the attachment design. Also, the design exerts mainly pressure forces and little shear forces on the human body, making the design for an attachment to the body potentially easier. Besides the working of the orthosis, other factors have to be considered as well, for example putting the orthosis on and off and adjusting the position of the orthosis.

The proof of concept is performed using a technical prototype. The orthosis has not been worn by a person yet. Future research has to determine how human gait is affected when wearing this orthosis design.

The adduction over flexion-extension stiffness ratio is determined during mid-stance, meaning that the adduction stiffness is only measured at 0° flexion-extension angle. How this stiffness changes over the full flexion-extension range of the leg is not yet investigated. The guiding bracket could have a positive effect on this stiffness, because it constrains torsion of the leaf flexure.

The materials and dimensions used in the conceptual design are chosen to provide the proof of concept, not to perform optimally on a human body. The needed stiffness of the orthosis in both directions depends on the person wearing the orthosis. The desired values for this stiffness should be investigated. The paper presents a sensitivity analysis for the dimensions, showing the effect of the dimensions on the flexion-extension and adduction stiffness. This can be used to determine dimensions and materials for each case-specific.

Chapter 5

Conclusion

The objective for this thesis was to make a first step in the design of a compliant hip orthosis for Trendelenburg gait. A literature study investigates the cause of Trendelenburg gait from an engineering perspective, the requirements and design objectives, and looks at the state of the art of hip orthoses.

Chapter 3 presents the conceptual design of a compliant hip orthosis for Trendelenburg gait. This design consists of two leaf flexures, two wire flexures, and two guiding brackets where one of each is present in the front and back of a person. The leaf flexures are attached to the front and back of the thigh and go up to the upper body. From there a wire connects the top of a leaf flexure horizontally to the other side of the hip. The guiding bracket slides over the leaf flexure and keeps it tangential to the upper body. This design has a high adduction over flexion-extension stiffness ratio, little shear forces on the body, and small protrusions from the human body.

A prototype was designed and built to measure the stiffness ratio and forces from the orthosis on the body. This resulted in an adduction over flexion-extension stiffness ratio of 1.7×10^3 and at least 19 times lower shear than pressure forces on the body.

These results provide the proof of concept and the design is considered a success. Future research has to determine if the concept also works as desired on the human body.

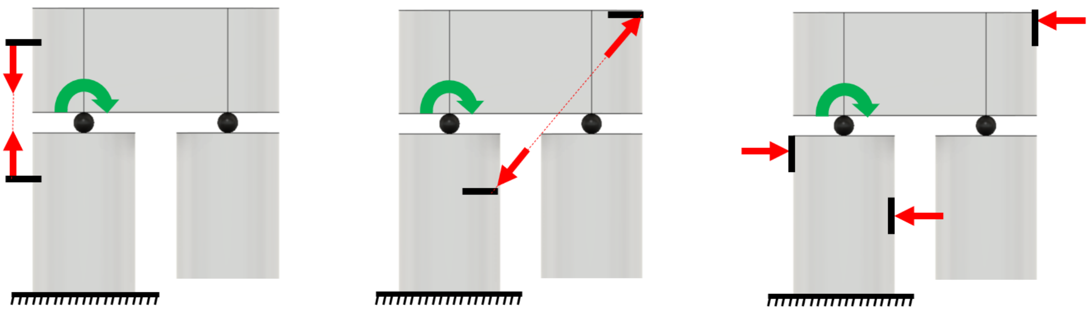
Appendix A

Design process

This section explains the process before reaching the final design presented in the paper. First, the way forces can be introduced in the body to create an abduction moment around the hip joint are investigated. Then, the designs are presented with their disadvantages and improvements in the next design.

A.1 Forces

The possible places to add a force to support the human body are investigated. The forces need to counter the moment from gravitational forces from the upper body and not-supported leg. Figure A.1 shows the possible ways to counter this moment with forces.



(a) (b) (c)

Figure A.1: Three possible ways to add an abduction moment using an orthosis with the moment from the upper body in green and the forces from the orthosis on the body in red. The black lines represent attachments to the body.

The concept in figure A.1a pulls in vertical direction on the side of the body, much like existing orthoses. Because the direction of the forces is tangential to the body, this will introduce mainly shear forces on the body, which are undesired. Figure A.1b also has two opposing forces, but on the other side of the joint and in opposite direction. Depending on the angle at which the forces act, this could introduce more pressure forces on the body. Although, there will always partially be a shear force, since the forces can never be horizontal. Also, adding a (compression) stiff connection in a straight line, will obstruct the movement of the other leg during the gait. Changing the shape of this orthosis to avoid it from obstructing the leg will introduce bending in the orthosis. This will require the design to be larger. The concept in figure A.1c has a third force, which omits the requirement for the forces to act in one line. The forces can be placed perpendicular to the body to create a (theoretically) 100% pressure force.

A.2 Non-feasible concepts

Though the forces on the body of the concept in figure A.1a are not as desired, possible solutions are investigated. A wire or leaf flexure on the side of the hip could be used. Both are stiff in tensions and allow flexion of the leg. A leaf flexure would stand out far from the body and is therefore left out from the beginning. The wire flexure is still an option and could be further investigated if other solutions do not work out.

The concept in figure A.1b is harder to realise, because the connection will be loaded in compression. Loading compliant mechanisms in compression could lead to buckling and is undesired. Complex geometries can minimise this effect but they would protrude from the body, which is also undesired. Therefore, this concept is not further investigated.

A.3 Concept development

It is harder to define all possible solutions for the concept in figure A.1c. At first, the use of a horizontal leaf flexure is investigated as can be seen in figure A.2.

Figure A.2: Horizontal leaf design

Here, only the horizontal leaf flexure is a compliant mechanism, and the vertical part is a stiff connection to the leg attachment. The rotation of the leaf along the horizontal axis allows flexion of the leg, while rotation around the axis perpendicular to the face of the leaf, is constrained by the leaf. The problem with this design is that the rotation axis of the leaf flexure and hip joint do not align. This results in the fact that flexion of the leg is only possible if the leaf flexure could also translate in vertical direction. This is the same motion that needs to be stiff to support the supported motion.

Instead of a leaf flexure, a wire flexure can be used. If the wire is placed at a vertical distance from the hip joint, the supported direction is constrained by the tension in the wire. The walking motion is still possible because of the lower stiffness of the bending of the wire. Though, if the wire is placed very high, it has to bend a lot to allow the walking motion. Also, the volume above the hip joint is obstructed by the upper body, making it impossible to connect a wire there. Adding the wire on one side of the body instead of straight above the hip joint allows the hip to drop in the direction perpendicular to the face of the wire and the hip joint.

Instead, several wire flexures can be used. Figure A.3 shows a prototype design of this concept. Note that the wires are above and below the hip joint. This means that the attachments of the wires need to flex in order to make the walking motion possible. Also, the wires above the hip joint are loaded in tension, while the ones below are loaded in compression, which is undesired as mentioned before. Only placing the wires above the hip joint results in a small distance between them. This small distance results in a low supporting stiffness, because the compliant planes as explained before are close to each other.

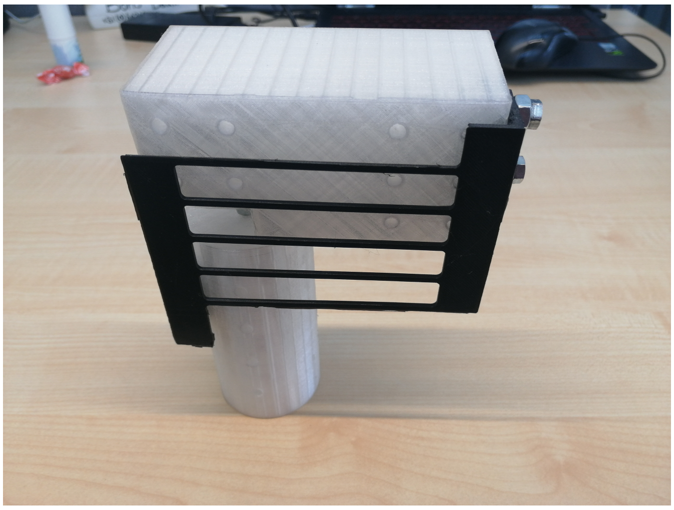


Figure A.3: Prototype design with five wires in front

A solution for this is to add a wire in the front and rear of the hip. Figure A.4 shows this design. This way, both wires are loaded in tension, while the compliant planes of both are not parallel, resulting in a higher total supporting stiffness. The connection to the leg still needs to be compliant because it crosses the hip joint, around which the body rotates during flexion of the leg. This connection needs to be stiff in the support direction as well, because it will be added in series to the wire flexure. A leaf flexure can work in this case. The only difference between this concept and the final design described in the paper is the guiding bracket. This is added to keep the orthosis close to the body at all times during the gait.

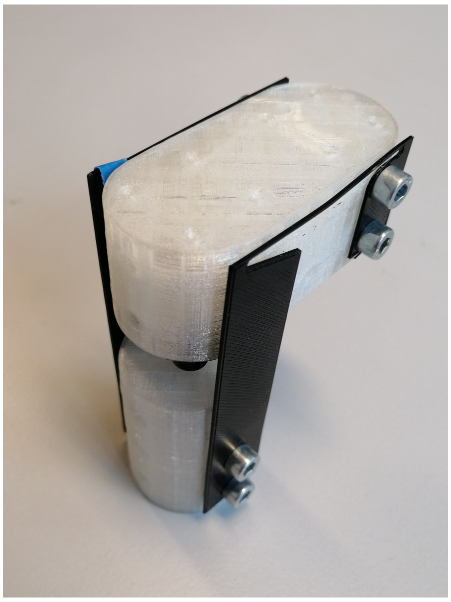


Figure A.4: Prototype design with a wire on the front and back

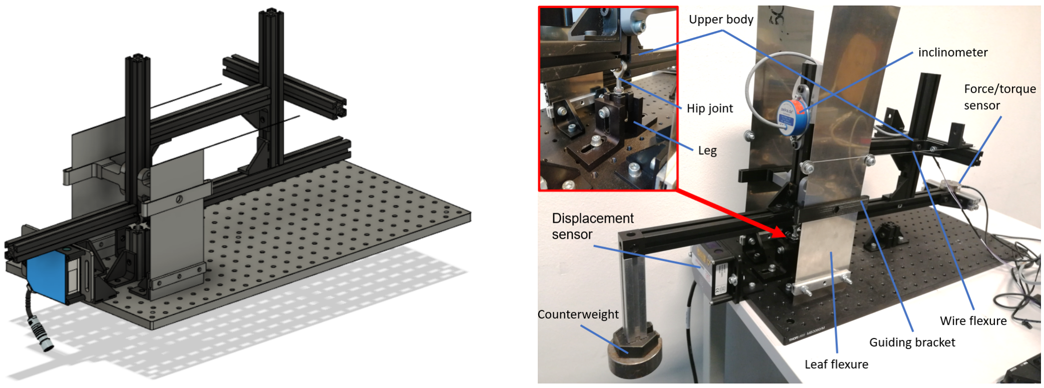
Also, the use of intermediate bodies between wire flexures is investigated. This decreases the walking stiffness because the effect of the horizontal distance between both attachments of the wires becoming shorter when bending can be mitigated by the displacement of the intermediate body. Besides this, the effective wire length could be longer, also decreasing the walking stiffness. The downside is that the wire flexures will be loaded in compression instead of tension. Because of this reason, the use of intermediate bodies is not further investigated.

Appendix B

Technical prototype design

The technical prototype is designed in Fusion 360 as can be seen in figure B.1a, and is built as can be seen in figure B.1b.

The design uses aluminium extrusion profiles for the leg and hip that are connected with a ball joint as the hip joint. The leaf and wire flexures are made out of spring steel and 3D printed PLA is used for the guiding bracket. The dimensions of the orthosis are chosen from available materials to be close to the average hip size of humans, and can be found in table B.1. The names used for the dimensions are illustrated in figure B.2.



(a) (b)

Figure B.1: Prototype in Fusion 360 (a) and as built in real life (b)

The flexion-extension stiffness of the prototype is measured using a force/torque sensor (ATI mini-40) to measure the input moment and an inclinometer (Seika NG4i) to measure the flexion-extension angle. The adduction stiffness is measured using the same sensor to measure the input moment. Because of a lower displacement due to higher adduction stiffness, a laser displacement sensors (Micro-epsilon optoNCDT 1750) is used to measure the adduction angle. Because this sensor measures a distance instead of an angle, the change in distance has to be transformed to a rotation around the hip joint. Figure B.1b shows the prototype to measure the adduction stiffness. In this figure, the force/torque sensor is used to measure a force at a distance from the

Table B.1: Prototype dimensions.

Dimension	Value
Leaf length (L_l)	0.18 m
Leaf width (b_l)	0.10 m
Leaf thickness (t_l)	0.50×10^{-3} m
Wire length (L_w)	0.21 m
Wire diameter (D_w)	1.0×10^{-3} m
Horizontal distance between leaf and joint (a_x)	0.080 m
Vertical distance between wire and joint ($a_{z,1}$)	0.13 m
Vertical distance between Guiding bracket and joint ($a_{z,2}$)	0.065 m

hip joint, that is translated to a moment around the hip joint. To measure the flexion-extension stiffness, the sensor is rotated 90° around the x-axis, such that a measuring axis aligns with the flexion-extension rotation axis. A moment is applied to the sensor, and the resulting rotation of the upper body is measured in the inclinometer. To measure the forces and moments on the body, the force/torque sensor is first placed between the hip and the orthosis, and then between the leg and the orthosis. Figure B.2 shows the position of the force/torque sensor with points Q and R.

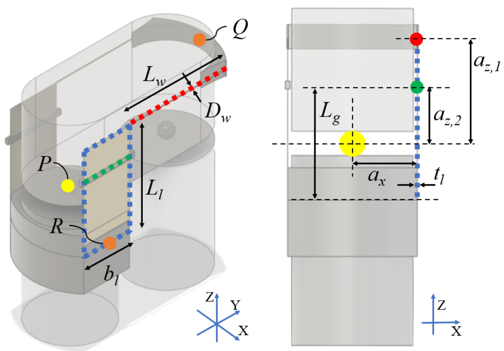


Figure B.2: Definitions of orthosis' dimensions used in the stiffness calculations in isometric view and in xz-plane. Points Q and R are show the position in the prototype of the sensor measuring the forces on the body.

B.1 Leaf flexure clamp

Specially designed brackets are used to clamp the leaf flexure on the bottom. Instead of just bolting the leaf flexure to the attachment points, this bracket simulates a line clamp. By clamping the leaf flexure this way, it is more similar to linear beam theory, making the comparison between both easier.

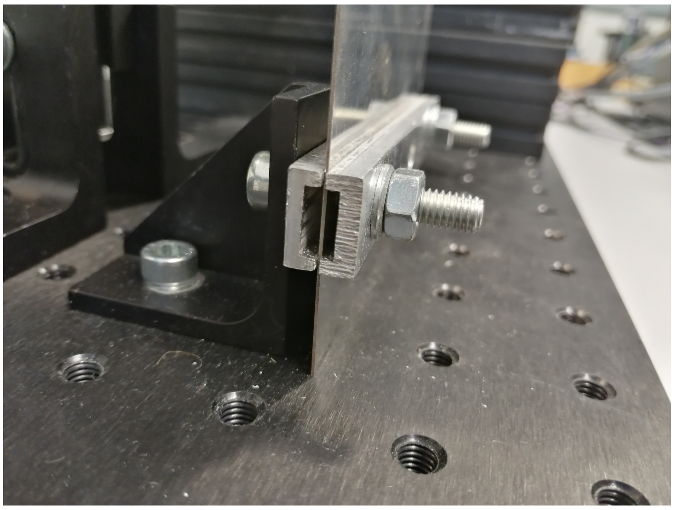


Figure B.3: Clamp bracket

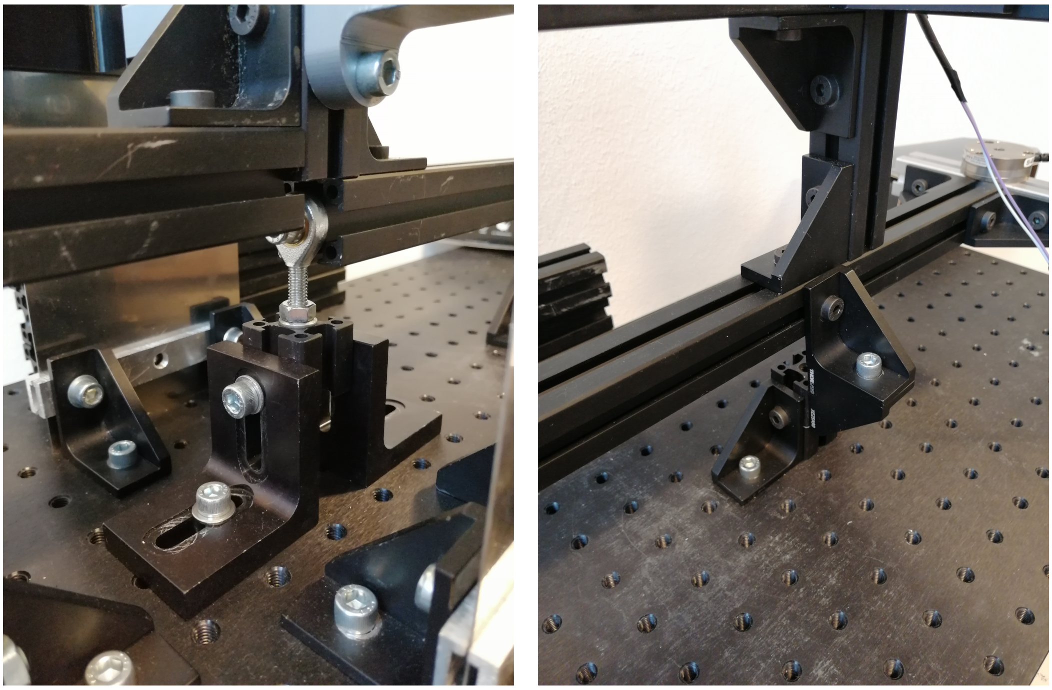
B.2 Adjustability

As can be seen in figure B.1b, the wire flexure attaches somewhere halfway to the leaf flexure. The top part is not attached and adds nothing to the test setup. This is done to easily adjust the length of the leaf flexure by attaching the wire flexure higher or lower. The attachment of the wire flexure to the body can also be adjusted in height.

The wire flexure is clamped on the extrusion profile on the body side. The wire flexure is also made too long to make it possible to increase or decrease the length by clamping at a different position.

B.3 Aligning

The ball joint is attached as figure B.4a shows. The height is chosen such that the horizontal extrusion profile that attaches to the ball joint is exactly 75 mm from the bottom plate. This way, that upper body can rest on three extrusion profiles, which have a width of 25 mm, to keep the upper body at its neutral position when attaching the orthosis. This guarantees that the orthosis is connected in the right way.



(a) Hip joint in prototype

(b) Parts used to align around the z-axis

Figure B.4

To align the rotation around the z-axis properly before attaching the orthosis to the body, the parts in figure B.4b are used. These parts lock the rotation around the z-axis by bolting the horizontal extrusion profile to the smaller extrusion profile connected to the bottom plate. Because of the large distance between the hip joint and this connection, the z-axis is constrained. When the orthosis is connected, both alignment mechanisms are removed.

B.4 Counterweight

A counterweight is used to balance the prototype around the ball joint. This weight balances the rotation around the x-axis and y-axis. The rotation around the z-axis does not need to be balanced because only gravitational forces act upon the system, and those do not actuate the rotation around the z-axis. The weight and position of the counterweight are determined by disconnecting the orthosis and creating balance for both rotations in all positions of the upper body.

B.5 Guiding bracket

The guiding bracket is designed as can be seen in figure B.5. Several designs are tested in the prototype to investigate the influence of the guiding bracket and in an effort to reduce the friction. Figure B.6 shows the different guiding bracket designs.

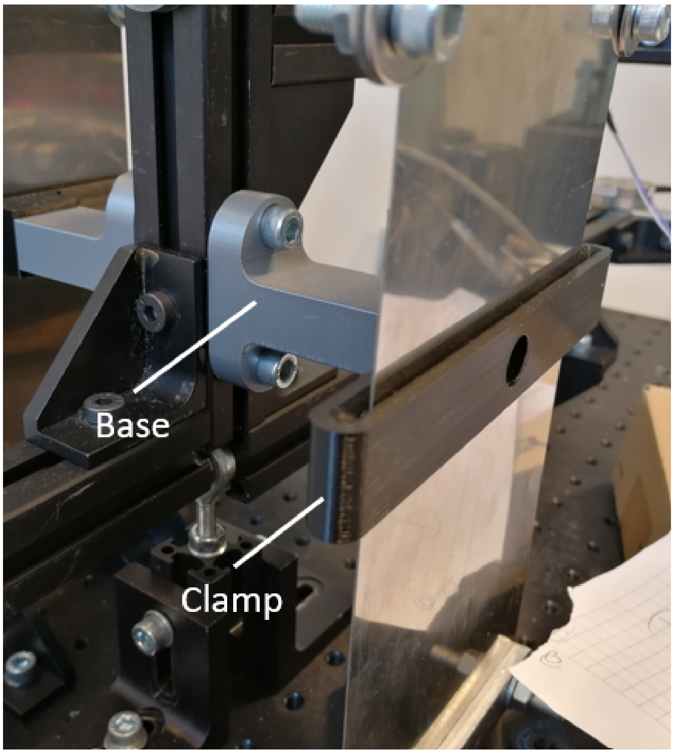


Figure B.5: Final guiding bracket design.

All designs use the guiding bracket base, a 3D-printed part that attaches to the extrusion profile of the body using two bolts. Because this extrusion profile is vertical, the guiding bracket can be attached at any height, while still aligning properly with the leaf flexure. The other part, the guiding bracket clamp, attaches to this base using a single bolt. The clamp has a hole all the way through the middle such that the bolt can be inserted from the front, while keeping a slim design. The different designs try to release different motions of the leaf flexure. The best working guiding bracket has small fillets on the corners and has a 1 mm slot for the 0.5 mm thick leaf flexure.

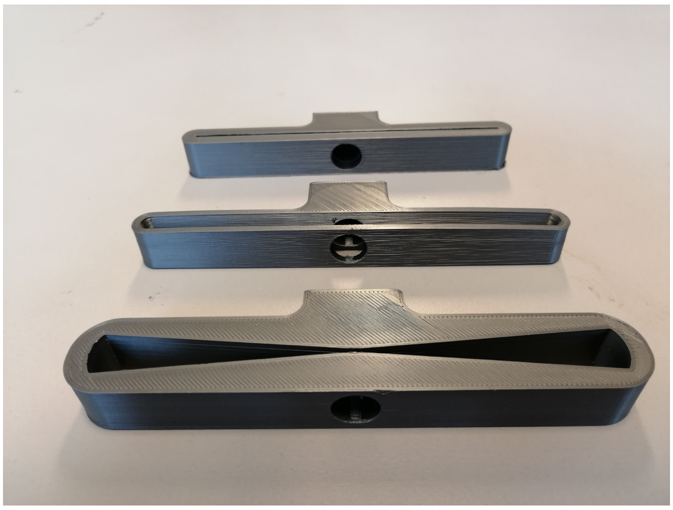


Figure B.6: Guiding brackets concepts.

B.6 Test setup stiffness

A finite element analysis (FEA) of the supporting stiffness of the prototype frame (representing the leg, hip joint, and upper body) is done using COMSOL Multiphysics®. Figure B.7 shows how this analysis is performed. The blue lines represent the frame of the prototype and have the cross-section and material properties of the aluminium extrusion profiles. The orthosis is assumed to be infinitely stiff by constraining the leg attachment is in all directions and the wire attachment in the direction of the wire, along the x-axis. The hip joint connects the translations of the leg and upper body in the joint, but allows rotations in all directions. A load is introduced at "Load" with the same magnitude as is used in the final test as performed in the paper.

Figure B.8 show the results of this analysis. In the left bottom, bending of the attachment of the ball joint is clearly visible. This bending allows the attachment to translate in the negative x-direction, which results in rotation of the upper body. This rotation as a result of a supporting moment shows a linear behaviour for these small deformations. This extra deformation as a result of the load is added in the "Finite element analysis" line of the support stiffness in the results section of the paper.

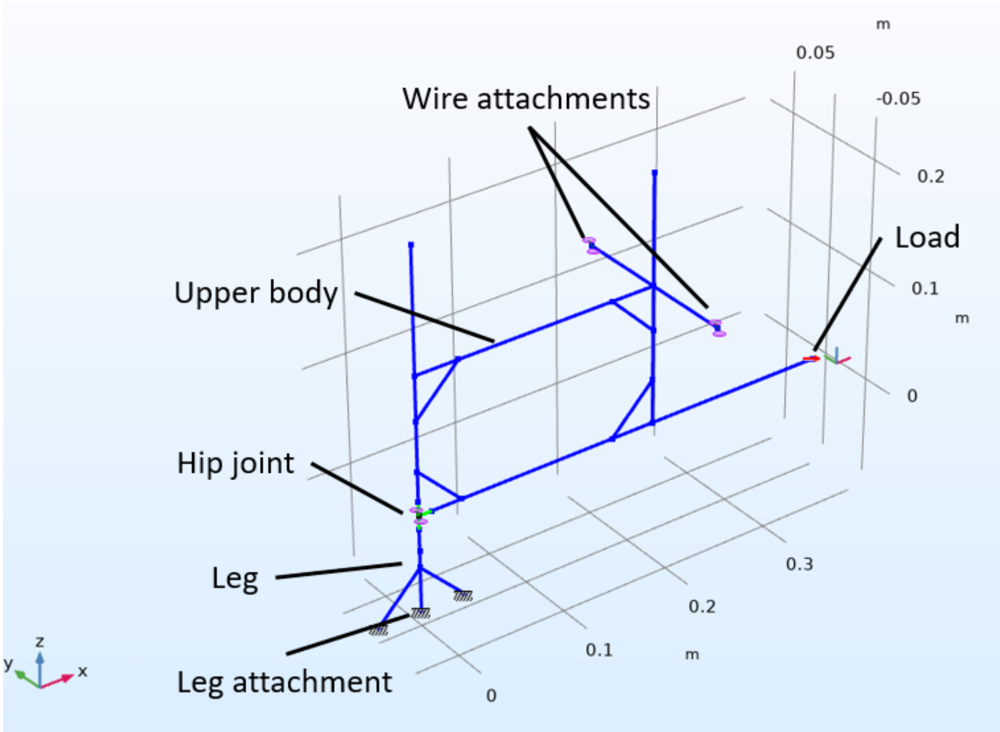


Figure B.7: FEA setup.

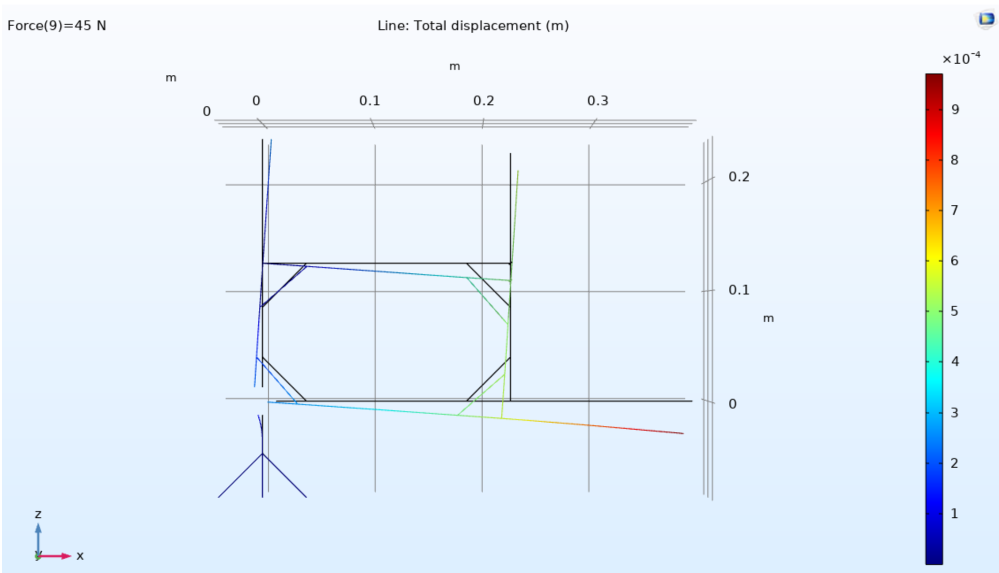


Figure B.8: Simulated deformation of the prototype when a supporting load of 45 N is added at the right most part.

B.7 Adduction stiffness FEM

For the FEM of adduction stiffness, only the orthosis is analysed. The model with definitions is illustrated in figure B.9 The stiffness of the frame is added at a later stage as is explained in section B.6. The Shell physics are used, making use of surface shapes and defining the thickness of those surfaces in the program. The flexure with a pre-curve of 5° is used as explained in the paper. These flexures are clamped at the bottom, constraining all motions. The end of the wire flexures are connected to the location of the hip joint, allowing all rotations, but no translations in the joint. The stiffness of the guiding brackets is added to prevent buckling of the leaf flexures. The value for the stiffness is again simulated in another FEM and is 150 N m rad^{-1}.

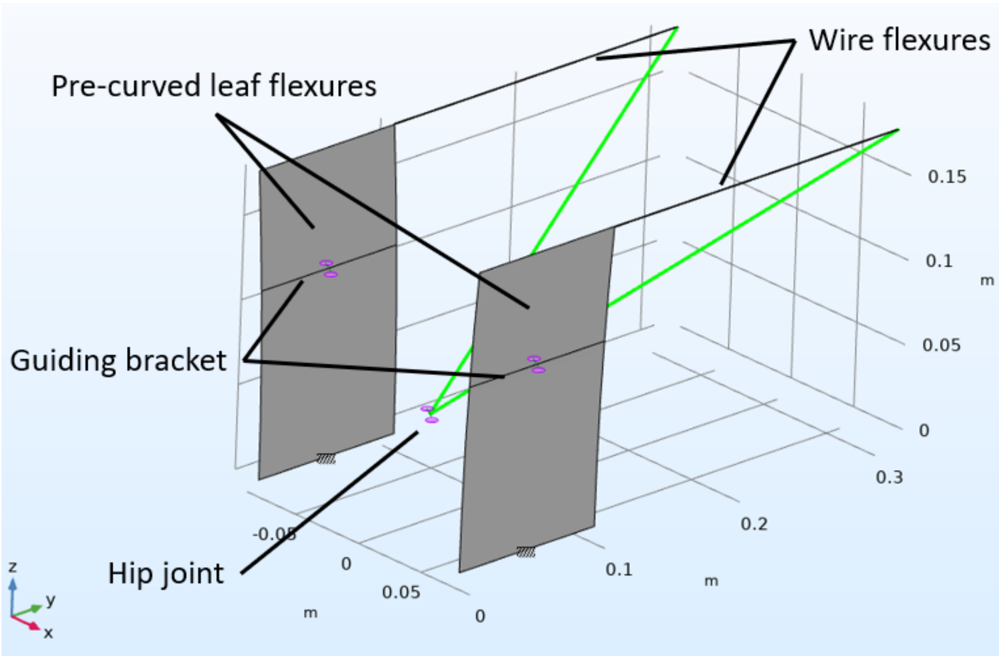


Figure B.9: Adduction stiffness FEM.

B.8 Flexion-extension stiffness FEM

For flexion-extension stiffness, again only the orthosis is analysed and the same orthosis as the support stiffness is used. The model with definitions is illustrated in figure B.10. Because this stiffness is so much lower than the support stiffness, the stiffness of the frame has minimal influence on the total stiffness and is neglected. The frame is used as an input rotation to push against the guiding bracket, as is done in the prototype. The Young's modulus of this frame, and the thickness are chosen to be very high to simulate the infinite stiffness of the frame. A rotation of the frame around the hip joint is introduced and the moment needed for this is measured. To prevent contact FEMs, which are complex and time-consuming, the guiding bracket and frame are defined to have the same translation along the x-axis, rotation around the y-axis and z-axis. The other motions are left free, meaning that the guiding bracket can slide over the leaf flexure along the y-axis and z-axis, and can rotate around the x-axis. The wires are also connected to the frame, because they have the same rotation as the upper body.

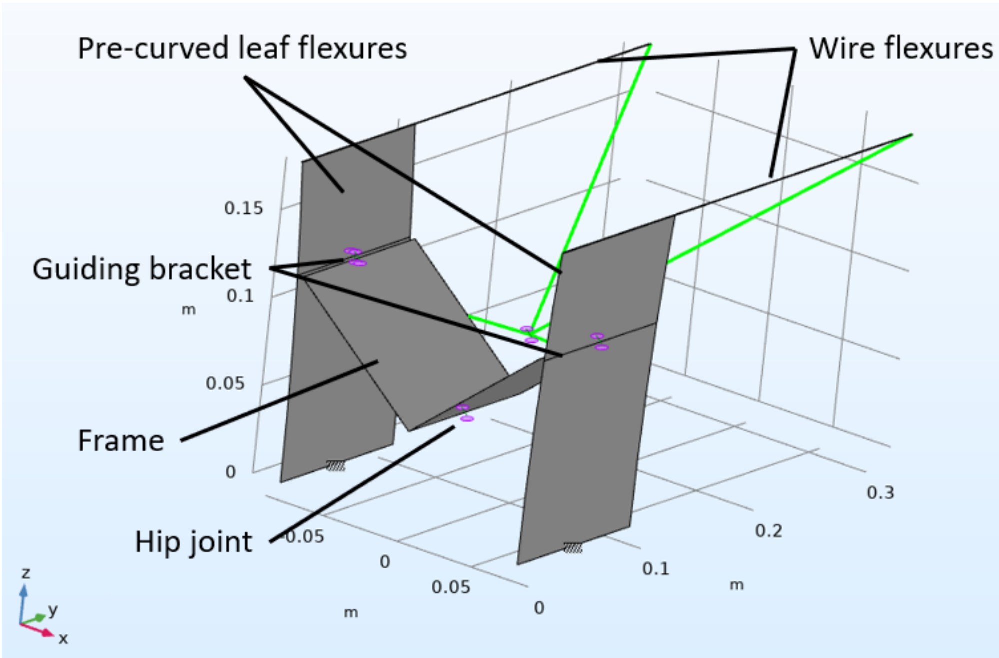


Figure B.10: Flexion-extension stiffness FEM.