

École polytechnique de Louvain

Bionic prosthesis for the lower limb Adapter for testing a complete leg prosthesis

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Abstract

Limb amputation is characterised in the surgical community as the last line of treatment for a patient. Amputation of a lower limb will affect the patient's mobility by increasing their metabolic cost and decreasing their walking speed that will limit their independence and quality of life. Today, prostheses are more and more all active (provided energy) with objective to be more pleasant to wear. For lower limbs, active prostheses exist for almost all amputations except hip disarticulation and hemiplevectomy linked with amputation of the hip joint. This is due to the low incidence rate compared to other types of amputation, limiting technological advances due to lack of investment. However, the Polytechnic School of Louvain la Neuve has invested in a PhD thesis with the objective of creating a fully active Canadian prosthesis (prosthesis specifically designed for hip disarticulation). In correlation with this doctoral thesis, this Master Thesis has been carried out in order to create a device whose objective is to create an adapter worn by an able-bodied person to which a Canadian prosthesis can be attached in order to carry out pre-tests without resorting to amputees. In this Master Thesis, the design of an adaptor allowing an able-bodied person to wear a Canadian prosthesis was designed, created and tested. This adapter was first designed to take into account a set of possible solutions that might exist. Then, a study of the different forces applying on the system was elaborated. Thanks to this, we were able to dimension the device and to elaborate its creation. Finally, the device was manufactured and tested in order to evaluate its reliability in relation to a study previously carried out in the literature by an amputee. This experimental part showed us that the device allows us to obtain encouraging results for its use in order to carry out the first modifications on the prototype of active Canadian prosthesis.

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Chapter 1

Introduction

This Master Thesis focuses on the design of an adapter to pre-test a Canadian prosthesis worn by a non-amputee.

The objective of this section is to establish a state of the art on the thesis statement: "Bionic prosthesis for the lower limb - Adapter for testing a complete leg prosthesis". To do so, we will first express the motivations for this work (why), then we will study the causes that lead to a complete amputation of the leg (up to the hip), the type of prosthesis to be used to help the amputee to move with such a pathology and the adapters that exist in the research community in order to pre-test different prostheses for all types of amputation (upper and lower limbs). Finally, we will give an overview of the different sections of this Master Thesis.

1.1 Motivation

The amputation of a limb or part of a limb is a highly disabling event for a person's locomotor aspect (movement, motion, energy, etc), physical appearance and psychological aspect. An amputation can be performed on an upper limb as well as on a lower limb (cf. Figure 1.1).

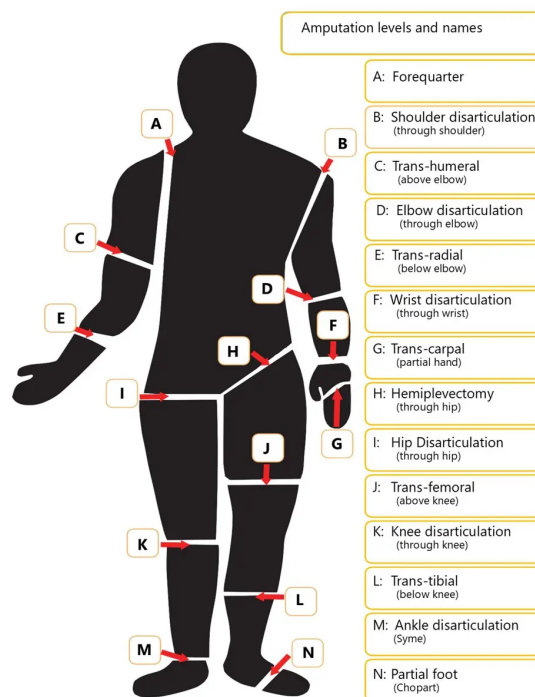


Figure 1.1: Enumeration of the names of the different amputations possible on the human body according to the location of the cut. [1]

This study will mainly focus on lower limb amputations. Among these, we find:

- Partial foot;
- Ankle disarticulation;
- Trans-tibial;
- Knee disarticulation;
- Trans-femoral;
- Hip disarticulation;
- Hemipelvectomy.

Each type of amputation faces different challenges. Indeed, the higher the amputation, the more complicated the prostheses and the longer the revalidation therapies for these patients are. Complexity is represented by the number of joints and muscles to be reconstructed. The greater the complexity, the more difficult it will be to stabilise and to propel the body. This will have a direct impact on the comfort of the prosthesis but also on the energy required during a walking cycle (cost of transport). A simple representation of the quantification of the energy expended can be represented by the rate of oxygen required to perform an effort. Figure 1.2 shows that the favorite speed of walking decreases with the severity of the amputation and that the O_2 cost increases. [2] [3]

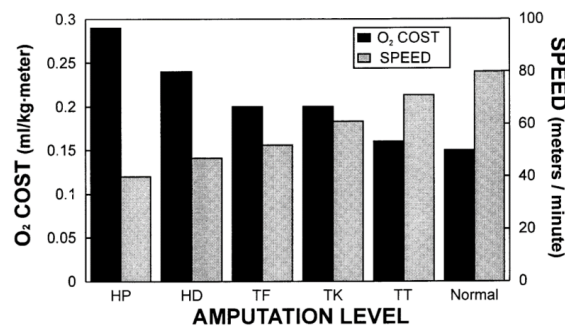


Figure 1.2: Variation in O_2 cost as a function of preferred walking speed as a function of amputation level and in comparison with data from an able-bodied person. HP = hemipelvectomy, HD = hip disarticulation, TF = trans-femoral, TK = knee amputation, TT = trans-tibial. [3]

A study conducted in the Netherlands in 2001 on 652 patients aged between 18 and 60 years (mean: 44.5 yrs) shows the different reasons that can lead to amputation and the proportion of people affected according to the level of amputation. These data are available in Table 1.1.

This table demonstrates a large disparity between the different levels of amputation. As a result of this disparity, it is only logical that the broad research carried out and the investment made in researching new prostheses should be proportional to the population affected by a type of amputation.

Hip disarticulation affects only 3.1% of lower limb amputees. As a result, very few studies have been conducted to improve the devices available on the market. This consequence is felt by patients affected by this condition as their mobility is limited by the lack of commercially available prostheses capable of faithfully reproducing the bio-mechanics of the leg [5].

	Mean (yr)	Median (yr)	Range (yr)
Age at amputation	25.1	22	0–57
Time since amputation	19.6	19	2–59
	<i>n</i>	<i>%</i>	
Reason for amputation			
Trauma	396		58.4
Cancer	101		14.9
Vascular	47		6.9
Diabetes	30		4.4
Other	100		14.7
Not given	4		0.6
Level of amputation			
Transtibial	316		46.6
Transfemoral	230		33.9
Knee	80		11.8
Hip	21		3.1
Ankle	16		2.4
Pelvis	12		1.8
Not given	3		0.4

NOTE. Data include 652 patients with 678 amputations.

Table 1.1: Information on the age of amputees, how long they have been amputees, and the proportion of different types of lower limb amputations and the causes of these amputations in a sample of 652 patients in the Netherlands. [4]

It has been shown that the prostheses available on the market for this type of amputation allow the patient to walk at half the walking speed of a non-amputee and that the metabolic cost is twice that of a healthy person [6]. The result is that after about 20 months of rehabilitation, 43% of those affected abandon their prosthesis due to the difficulty and lack of comfort it offers [7].

The motivation for this Master Thesis is to improve the quality of life of these people. Today, the Canadian prostheses on the market are active (provided energy) in the ankle and knee joints, but still passive (does not provide energy) in the hip joint. Making this type of prosthesis fully active would therefore reduce the metabolic cost and increase the comfort of use of these prostheses.

The objective of this study is the creation of an adapter in order to perform pre-tests on non-amputees of an active prosthesis design by Louis Devillez within the scope of his PhD thesis.

1.2 State of the art

In this section of the introduction, we will provide a state of the art on the subject of this thesis.

First, we will explain the causes that can lead to hip disarticulation.

Then, we will focus on the Canadian prosthesis as such: the problem, the difference with the anatomy of an able-bodied person, the description of the different parts that compose it and the consequence of this prosthesis on the gait cycle.

Finally, we will list and explore some types of upper and lower limb adapters that already exist in the research community.

1.2.1 Causes

The surgical procedure known as hip disarticulation was a procedure with dangers and disadvantages, both medical and prosthetic in the 19th and 20th centuries. Hip disarticulation is intended to prevent a fatal disease. Few people experienced this procedure and if they did, few survived during and after it.

The reasons of hip disarticulation are quite varied. The most common are traumas such as war injuries [8] followed by bone tumours [9] and cancer or, to a lesser extent, amputations due to vascular insufficiency, diabetes, osteomyelitis of the femur, failed hip replacements or congenital anomalies [10], etc.

We note that all of these causes must be taken to their extreme (before the patient dies) for complete disarticulation to be achieved. This is why the number of patients undergoing this type of amputation represents only 3.1% of the number of lower limb amputees, as mentioned above.

1.2.2 The Canadian prosthesis

Hip disarticulation is a mutilating procedure for the patient (cf. Figure 1.1). Following this operation, a prosthesis called a Canadian prosthesis was created so that the patient could regain mobility (cf. Figure 1.3). The purpose of this device is to replace the diseased limb by an artificial limb. The first prostheses designed had no joints in the hip, knee and ankle, which made movement very difficult. Subsequently, the devices have been improved and now include at least 2 to 3 anatomical joints, some of them motorised (active: provide energy). The problem with this type of device is that the metabolic cost to perform a movement is much higher than for a non-amputee and the walking speed is lower than the walking speed of an able-bodied person, which limits the patient's mobility (cf. Table 1.2).



Figure 1.3: Photo of a woman with a hip disarticulation wearing a Canadian prosthesis. [11]

From an anatomical point of view, the centre of rotation of the natural hip is not in the same place as the centre of rotation of the prosthetic hip. Indeed, the hip joint of the prosthesis is located below and in front of the normal axis of the hip (cf. Figure 1.4). [11]

In addition, a reference line can be drawn between the hip and knee joint of the prosthesis (cf. Figure 1.4). We notice that this does not pass through the ankle joint (normal position) but behind the ankle. This will ensure that the load transmitted between the foot and the hip joint is in front of the knee to avoid any risk of buckling at the knee. [11]

These two elements of symmetry should be taken into account when designing the device.

If we take a closer look at a Canadian prosthesis on the market today, it consists of several components. By analysing the figure 1.5 from the bottom upwards, we first see a prosthetic foot connected to a tubular adapter via a modular ankle joint (the articulation between these two elements is for the most part active). Then, this tubular adapter is connected to a second adapter via a modular knee joint. This knee consists of a piston system that acts as a shock absorbing spring in order to reproduce as well as possible the stresses applied by a real knee. Nowadays, most of the knees proposed on the market are motorised (active), which makes the system even more efficient. This is because the stiffness of the springs and the damping coefficient of the shock absorbers can be changed according to the use. Finally, the second adapter is connected to the pelvic socket (where the stump will be inserted) via a modular hip joint (helix) which is itself connected to the pelvic socket via a laminated cast plate. The proposed hip joints remain passive for the time being.

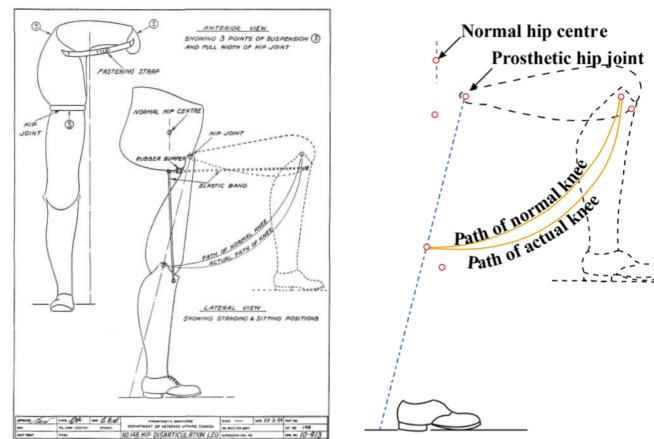


Figure 1.4: The figure on the left is an anatomical drawing by McLaurin showing the joints of a Canadian prosthesis and the hip joint of an able-bodied person. The figure on the right represents the bio-mechanics of a Canadian prosthesis showing the position of the hip joints. [11] [12]



Figure 1.5: Description of the different components of a new generation Canadian prosthesis sold by the firm Ottobock. [13]

Thanks to this type of prosthesis, the patient will be able to recover a slight part of his mobility. However, he will have to undergo a long rehabilitation in order to be able to reproduce a walking cycle. However, 43% of patients who underwent lower limb amputation abandoned their prosthesis in favour of a wheelchair after completing their rehabilitation. [14]

The main reasons for this are the discomfort of the prosthesis which creates pain in the stump (pressure, friction,...) and in the system of attachment of the prosthesis to the body. It is also due to the additional use of muscle and the involvement of certain parts of the body that do not have as their main function the realization of the walking cycle.

The walking cycle consists of two phases. The first is the stance phase, i.e. the movement in which the foot is in contact with the ground. And the second movement is the swing phase when the foot is no longer in contact with the ground.

The stance phase of a hip disarticulation amputee can be separated into 3 periods: the initial double-limb support, the single-limb stance and the second double-limb support. Similarly, the oscillation phase can be separated into 3 periods: the initial swing, the mid swing and the terminal swing (cf. Figure 1.6).

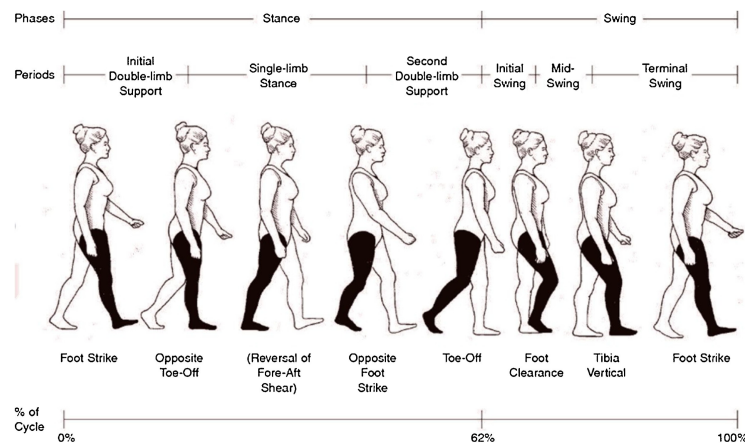


Figure 1.6: Description of the two segmented phases of the gait cycle. In the stance phase there are 3 periods: the initial double-limb support, the single-limb stance and the second double-limb support. In the swing phase there are also 3 periods: the initial swing, the mid-swing and the terminal swing. [15]

Firstly, regarding the stance phase of a person wearing a Canadian prosthesis, we notice that heel-to-floor contact, also known as foot strike (Figure 1.6), is created by swinging the prosthetic leg forward. The prosthesis' safety device allows a 15° movement at the hip joint. This movement, combined with the slight movement of the patient's pelvis, creates a natural backward tilt of the leg at the moment of contact between the heel and the ground.

Once contact with the heel is established, the amputee moves into the stance phase by rolling on the extended joints (the period between the opposite toe off and the opposite foot strike). For an amputee, this rolling is achieved through the momentum of the torso.

Finally, the toe-off is performed. This is what will make the transition between the stance phase and the swing phase. To do this, the prosthesis must be propelled forward. This movement is achieved by lifting and internally rotating the pelvis on the side of the missing leg.

Secondly, regarding the swing phase, it starts from toe off when the foot lifts off the ground.

In the past, the swing phase was performed by lifting and internally rotating the pelvis on the amputation side for an amputee as opposed to flexing the ankle, knee and hip joints. Today, this movement of the pelvis no longer needs to be performed to propel the prosthesis forward.

The swing phase starts at the toe-off. During this sub-phase, the prosthesis is tilted forward what will transmit the body weight to the front of the prosthetic foot.

Then, the inclination of the prosthesis will result in the angle of the hip decreasing. This decrease will cause the prosthesis to move forward. This moment at the hip level will compress the piston system present in the knee joint until we have a flexion of this one in order to arrive at the foot clearance sub-phase.

Finally, the transition from foot clearance to foot strike is achieved by relaxing the piston at the knee joint, which will extend the leg. The person will take support on the artificial leg in order to perform the stance phase again. [11]

1.2.3 Adapter

As part of our study, we will come and make an adapter so that a non-amputee can test a Canadian prosthesis.

This approach of testing prostheses on non-amputees is very often carried out especially in the framework of prototyping new prostheses. Indeed, testing prototypes directly on amputees is not easy to do due to the percentage of the population that corresponds to the needed profile but also from a financial and availability point of view in research laboratories. As a result, there are many different types of adapters (so that a non-amputee can wear a prosthesis) for both upper and lower limbs.

First of all, figures 1.7 and 1.8 represent 3 different adapter types for 3 different prostheses for ankle disarticulation but based on the same principle (each image represents 1 different adapter).

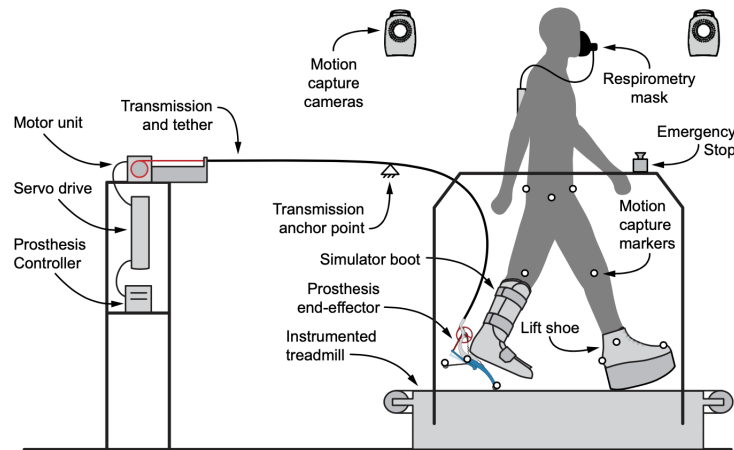


Figure 1.7: Adapter designed to test ankle disarticulation in an experimental approach on able-bodied people. [16]

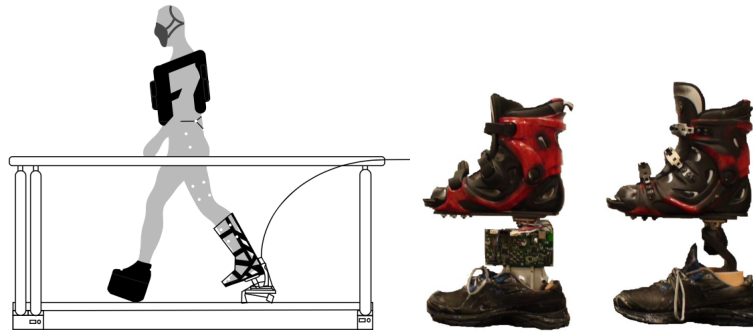


Figure 1.8: Adapters designed to test ankle disarticulation in an experimental approach on able-bodied people. [17][18]

It should be noted that the foot on the side where the prosthesis is tested is blocked in an immobilisation boot that blocks the movement of the ankle. For the "valid" foot (foot that functions normally without a prosthesis), it will wear a lifting shoe with a tilting sole so that the height of both feet is at the same level if it is not at the same level, the experimenter will limp and the results will not be correct.

Figures 1.9 and 1.10 represent 3 types of adapters used in cases of 3 different prostheses for a trans-femoral amputation but based on the same principle.

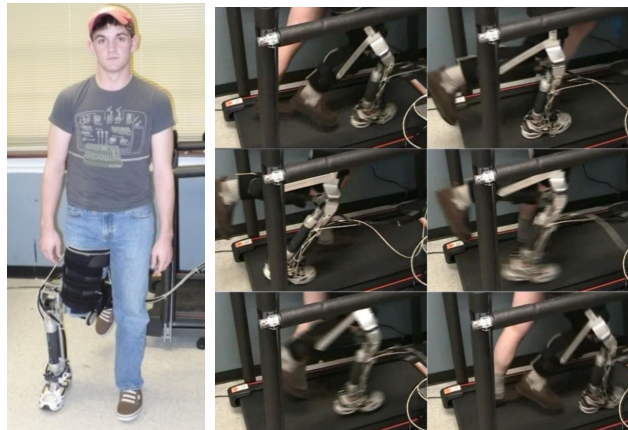


Figure 1.9: Adapter designed to test a trans-femoral amputation in an experimental approach on able-bodied people. [19]

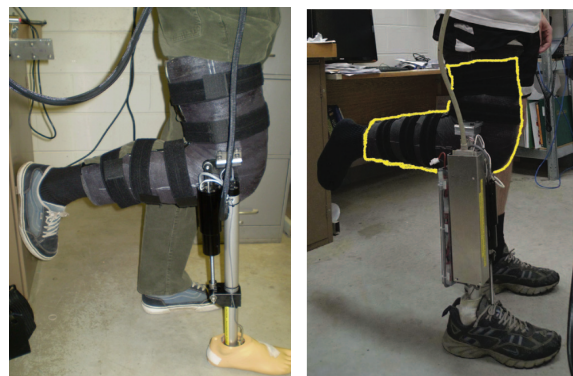


Figure 1.10: Adapters designed to test a trans-femoral amputation in an experimental approach on able-bodied people. [20] [21]

In order to artificial leg for trans-femoral amputations, the adapters worn by the subjects will lock the knee joint on the side of the leg wearing the prosthesis via a rigid brace. The prosthesis itself will be attached to the brace in such a way that the prosthesis and the biological pivot of the experimenter's knee are located in approximately the same place. In contrast to the adapters for disarticulated ankles, the experimenter will not have to wear compensated insoles.

In the case of upper limb amputations, the prototype prostheses are also tested on able-bodied people beforehand. Figure 1.11 represents 2 types of adapters for upper limb amputations in the hand and forearm.

For both types of adapter, we note that the artificial upper limb is not located where the prosthesis would be positioned on an amputee but either above (for the forearm) or below (for the hand) the natural position of the limb.

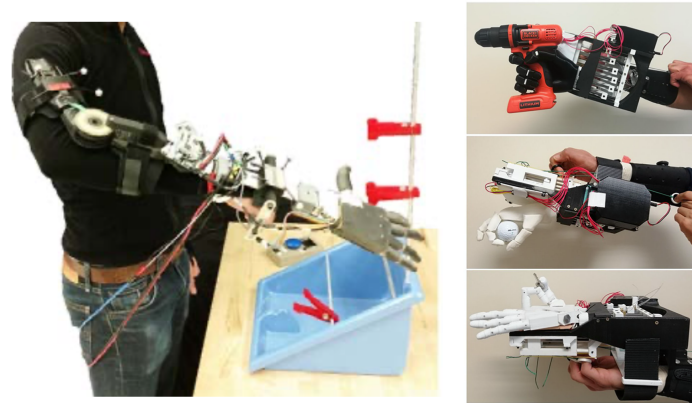


Figure 1.11: Adapters designed to test two prostheses for left forearm and right hand disarticulation in an experimental approach on able-bodied people.[22] [23]

Let's go back to the main topic of this report: hip disarticulation. For this kind of amputation, there are 2 types of adapters used for research.

The first model is shown in Figure 1.12. In this one, the artificial leg is positioned on the outside of the experimenter's body. The attachment of the prosthesis to the body is done by means of an angled metal part which is connected between the prosthesis and an orthosis consisting of a corset and a thigh attachment. In order to prevent the experimenter from helping himself with his blocked leg, the latter will not be in contact with the ground but will be raised via a compensated sole at the level of the leg considered as valid (not wearing the prosthesis). [24]

The second model (Figure 1.13) was created on a completely different principle from the first one. Indeed, the prosthesis takes the place of the blocked leg which is itself locked in anterior flexion in a socket. The fixation between the artificial leg and the body of the experimenter is done via a socket which takes the shape of the buttocks of the experimenter. During the experiments, the user will sit in the socket and perform the walking cycle.

In contrast to the first model, the able-bodied leg will not be elevated via a compensated sole because the locked leg does not naturally touch the ground when in the anterior flexion position.

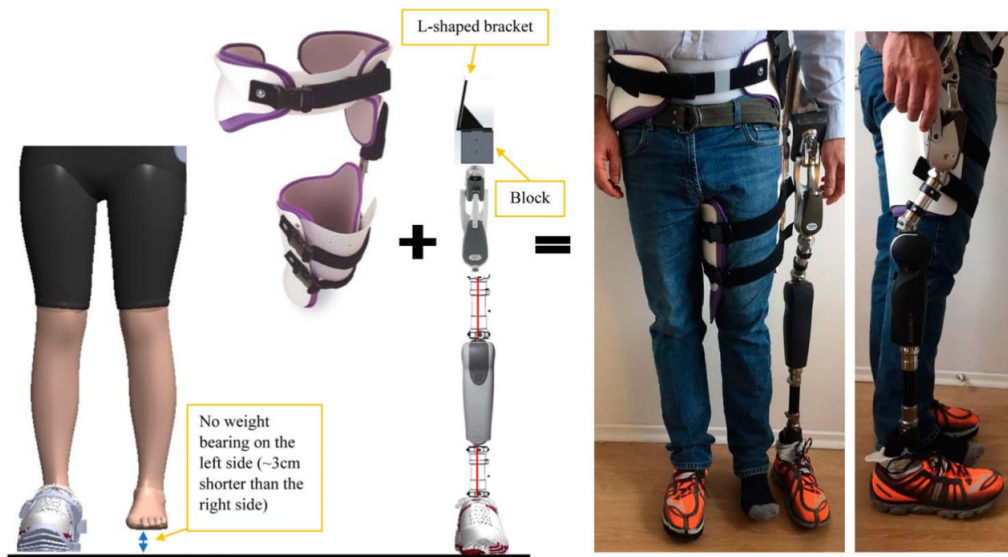


Figure 1.12: Model 1 of the adapter designed to test a Canadian prosthesis in an experimental approach on able-bodied people. [24]

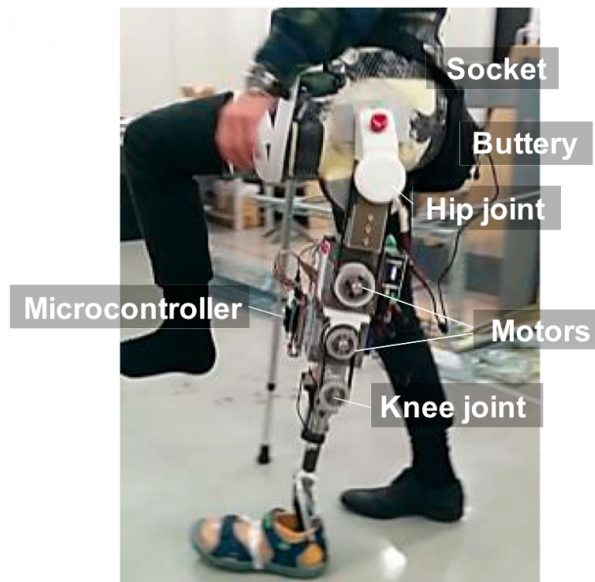


Figure 1.13: Model 2 of the adapter designed to test a Canadian prosthesis in an experimental approach on able-bodied people.[14]

1.3 Overview

This Master Thesis is composed of 4 sections:

- Design and development of the concept;
- Dimensioning relating to the calculation of load and moment acting on the experimenter and on the concept;
- Design, embodiment and assembly of the main elements of the prototype;
- Experiment to validate the reliability of the prototype.

Chapter 2

Conceptual design

2.1 Introduction

In this section, we will carry out the design phase of our device. This phase is a systematic and intelligent process in which designers generate, evaluate and specify designs for devices, systems or processes whose form(s) and function(s) meet customer objectives and user needs while satisfying a specified set of constraints.

First of all, we will draw up a list of specifications needed for our device. To do this, we will produce a graph of objectives and a table of specifications.

Then, we will gather this information to design a functional graph which will be used to make a morphological table. This is the result of a brainstorming of all the solution ideas for a given function according to the main specifications.

To finish, we will make a list of possible solutions which will be sorted from the best to the least good thanks to a decision matrix. Following this choice, we will describe the solution that emerges in more details.

In the remainder of this chapter, the different elements will be located in space according to the body planes (sagittal plane, frontal plane and transverse plane) listed in the figure 2.1.

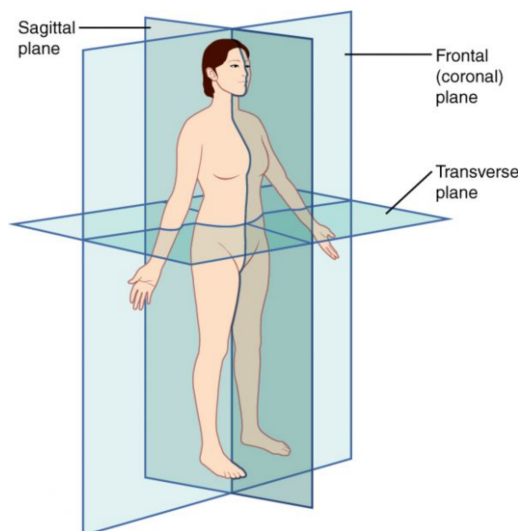


Figure 2.1: Description of the descriptive planes of the human body: sagittal plane, frontal plane and transverse plane. [25]

2.2 Functional specifications

The first step in a project is the translation of a request into functional specifications. In the broadest sense, describing the purpose of the device with the functions to be performed.

In this sub-chapter, we do not yet have any solutions as such, but we are asking ourselves what objectives the final solution should include.

To do this, we will go back and forth between the objectives that the device must achieve and that will describe the system itself.

In order to carry out this task, we need to ask ourselves a serie of questions:

- What objectives must the solution achieve?
- What features or attributes would you like the device to have?
- What do you want the device to do?
- ...
- Main Headings:
 - Geometry: Size, Height, space requirement, etc.
 - Kinematics: type of motion, velocity and acceleration, etc.
 - Safety: human, machine, etc.
 - Ergonomics: comfort, lighting, compatibility, etc.
 - etc.

Following this questioning, we will produce a graph of objectives and a summary table of the specifications.

2.2.1 Graph of objectives

A good way of achieving functional specifications is to use a graph of objectives.

By asking ourselves a serie of questions (as listed above), we can better visualise the main functions of our device.

We also need to sort these questions in order of importance and try to create links between the different answers. In order to get, at the end, a top-down tree structure.

Our main topic for our tree is: Adapter for testing a full leg prosthesis. This leaf will be the top of our tree.

From this main leaf, we will ask ourselves the questions:

- **What?:** What is this device?
- **Who?:** Who will carry the device?
- **Where?:** What needs to be replaced? Where will we position it?
- **Why?:** Why do we replace a whole leg? Why do we want to use able-bodied people? Why are there so many studies published on this subject?

- **What morphology?:** What is the difference between the ball-and-socket joint position (cf. Figure 3.1)) of an able-bodied person and an amputee? What are the important points from a bio-mechanical point of view? What are the criteria for adaptability with a Canadian prosthesis?
- **How safe is it?:** What are the risks of injury? What are the pain thresholds?
- **What ergonomics?:** How are we going to put the device on?
- **What weight?:** What is the weight of our system?

The figure 2.2 below represents the graph of objectives of our device. It represents the different answers to the questions classified by order of importance (from the most to the least important: from top to bottom and from left to right).

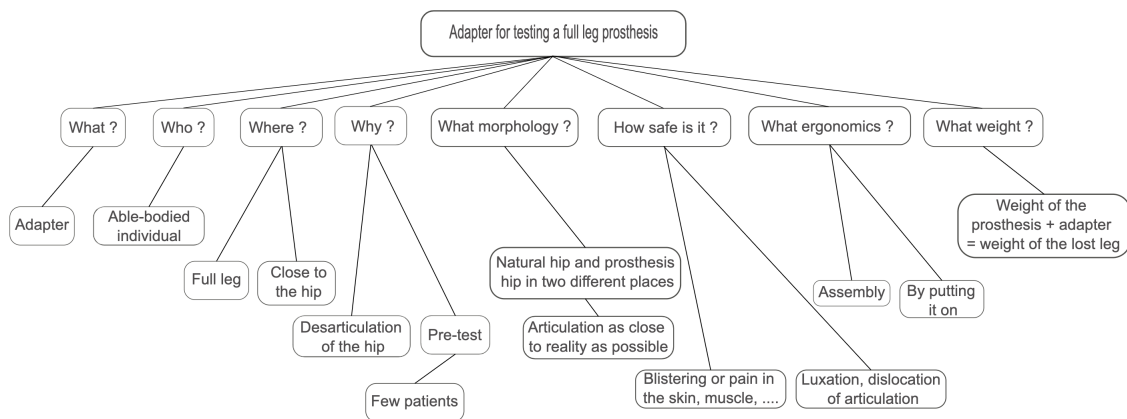


Figure 2.2: Graph of objectives applied on the adapter for testing a full leg prosthesis based on the questions asked in section 2.2.1.

2.2.2 Specifications

When designing a device, we need to define the different specifications that it must meet. We must therefore establish the main and constraint functions that represent the performance that our system must achieve.

Among these, we distinguish:

- **Objective:** expression of the attributes and behaviour that the customer or users expect from the device or system.
- **Main Function (MF):** an action verb that a successful design must accomplish. It always refers to an action performed by the product.
- **Constraint function (CF):** strict limits that a design must meet to be acceptable. Usually formulated in binary choices and imperatives. It mainly concerns the adaptation of the product to its environment.
- **Criterion or level:** expression of a quantification statement.

All this information will be collected in the table 2.1.

Lhotellerie Lucie		SPECIFICATIONS	Date : 25/02/2023 Version : 2
Context :		Creation of an adapter worn by a person who still has both legs in order to perform pre-tests on a Canadian prosthesis	
date	origin	Main Fonctions	
25/02	Lh.L	MF1: The device must be worn by an able-bodied person MF2: The device shall allow the manufacturer to attach any Canadian prosthesis MF3: The device allows the user to walk and perform stability tests with a prosthesis	
25/02	Lh.L	Functional Requirement (FR) criteria and levels FR 1.1: Must be suitable for an adult male or female of normal build FR 1.2: The adapter must be able to be put on and taken off several times without too much difficulty FR 2.1: A standard type prosthesis must be able to be attached FR 2.2: The Canadian prosthesis attached to the adapter must be capable of functioning under similar conditions as when worn by an amputee. FR 3.1: The user must be able to perform the experimental protocol FR 3.2: The experimenter shall not be able to use his or her leg replaced by the prosthesis under any circumstances	
25/02	Lh.L	Main constraints MC 1: Adaptable MC 2: Weight MC 3: Safety MC 4: Biomechanics MC 5: Dimension MC 6: Morphology MC 7: Ergonomy MC 8: Ease of use	
25/02	Lh.L	Constraint Requirement (CR) criteria and levels CR 1.1: Respect for the morphology of a woman and a man CR 1.2: Possibility of moving intermediate parts to fit the user's joints as closely as possible CR 2.1: The weight of the adapter + prosthesis must be within the weight range of the leg to be replaced CR 3.1: The device must be able to be used without harming the user CR 3.2: The adapter must not injure the skin or a muscle (pressure < 500 mmHg) CR 3.3: The adapter must not cause dislocation of a joint CR 4.1: The hip support system of the prosthesis on the adaptor should be close to the hip support system of an amputee wearing a prosthesis CR 4.2: The results of the simulations with the adapter should be similar to the result of the amputee wearing the prosthesis CR 5.1: The size and shape of the adapter should not interfere with the user's performance CR 6.1: The Canadian prosthesis shall not abut any part of the body of the experimenter CR 7.1: The adapter must be comfortable to wear CR 7.2: The device can be used for a period of 1 hour without major discomfort CR 8.1: The device must be able to be installed in less than 5 minutes CR 8.2: The device must be installed on the experimenter with a maximum of one external assistant	

Table 2.1: Table of specifications for creation of an adapter worn by a person who still has both legs order to perform pre-tests on a Canadian prosthesis.

2.3 Functional graph and morphological table

The design process and the design results are achieved in several steps.

- The first step is the creation of a **function graph** (Figure 2.3) that details the main technical functions of the device.
- The second step, based on the technical functions, is the creation of a **morphological table** (Table 2.2). To do this, we brainstormed and made sketches and drawings to represent the different ideas for the different sub-functions.

We will mainly focus on sub-functions related to the position of the user's body in its environment, blocking systems in order to block the device (especially at the level of the blocked leg), support systems in order to hold the body and comfort support in order to make the devices most comfortable to wear for the experimenter.

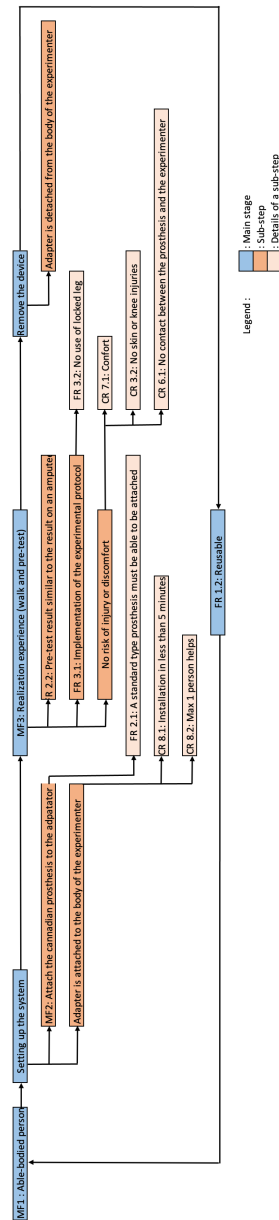


Figure 2.3: Functional graph detailing the main technical functions of an adapter worn by a person who still has both legs in order to pre-test a Canadian prosthesis.

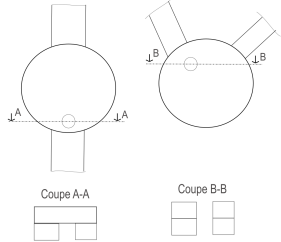
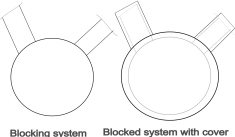
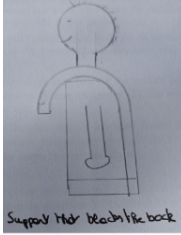
Position	 1.a	 1.b	 1.c
	 1.d	 1.e	
Locking system	 2.a	 2.b	 2.c
	 2.d	 2.e	
Support system	 3.a	 3.b	 3.c
	 3.d	 3.e	
Comfort support	 4.a	Rigid Support conforms to the shape of the body part in contact 4.b	

Table 2.2: Morphological table applied to the sub-function of an adapter worn by a person who still has both legs in order to perform pre-tests on a Canadian prosthesis.

In the table 2.2, the first part represents the position of the body. The position of the user's body in the space is the first thought during the design stage. It is the starting point for our concept thinking. Therefore, here are these figures in a larger dimension (in figure 2.4):



Figure 2.4: Position of the experimenter's body in space. These positions are the basis of our thinking when designing a solution.

The second part is the locking system. The locking system will be important in blocking the position of the healthy leg so that it does not hinder the movement of the prosthesis. Figure 2.a shows a first type of locking system based on a system of wedges (pins) that will be placed at the level of the kneecaps in order to limit their movements. Figure 2.b shows a belt that aims to maintain a stable geometry of the system. Figure 2.c shows a hand-tightening system. Figure 2.d shows screws. Figure 2.e shows a locking system via a system of parts that would fit over the ball joints when they are correctly positioned (see cover).

The third part represents the support system. Figure 3.a shows a rigid support system of the "gutter" type which will follow the shape of the part of the body to be supported. This trough is often made from a thermoformed plastic sheet. Figure 3.b shows a rope that connects two body parts. Figure 3.c shows a corset. Figure 3.d is a posture corrector for the back. Figure 3.e shows a rigid frame that is intended to press the upper body against the frame to correct the posture of the back.

The fourth part is the comfort system. In this part we find the protective cushions 4.a that can be placed between the support system and the user's body. And a rigid system that follows the shape of the user's body (4.b).

– The third step aims to design a **series of solutions** based on the ideas from the morphological graph. The objective is to mix the different ideas in themselves in order to take all possible solutions into account.

Finally, we made the schematization of 5 solutions varying essentially on the user's position.

In solution 1, the prosthesis is attached to the upper thigh with the non-amputee leg locked in posterior flexion with extension at the level of hip. A corset is used to maintain a natural back position and leg bracing is aided by a rope attached to the foot and linked to the back of the thigh.

In solution 2, the prosthesis is fixed below the buttock with the non-amputated leg locked in anterior flexion. A corset helps to maintain a natural back position. The brace system can be used to hold the leg in place but also as a support for the prosthesis attachment. A screw lock system can be used to keep the corset and the brace connected.

In solution 3, the prosthesis is fixed to the front of the thigh (in front of the body) with a non-amputee leg locked in posterior flexion with flexion at the level of hip. A corset is used to maintain a natural back position and leg bracing is aided by a rope attached to the foot and linked to the back of the thigh.

In solution 4, the prosthesis is fixed on the lateral side of the body with the non-amputated leg locked in posterior flexion. A corset is used to maintain a natural back position.

And in solution 5, the prosthesis is fixed below the thigh with an unamputated leg locked in lateral flexion. The brace system can be used to hold the leg in place but also as a support for the prosthesis attachment. A screw lock system can be used to keep the corset and the brace connected.

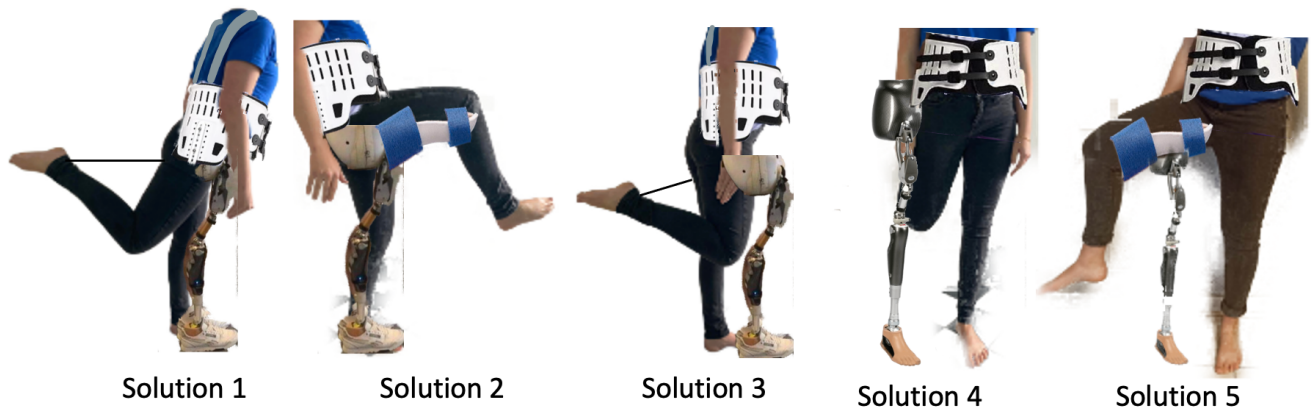


Figure 2.5: Proposal for a concept of an adapter worn by a person who still has both legs to pre-test a Canadian prosthesis.

2.4 Assessment of solution

In order to make an objective choice between the different concepts in figure 2.5, we will carry out a study based on a decision matrix.

A decision matrix is a tool for evaluating and determining the best option among different choices.

Our decision matrix is based on a relative comparison of solution 1 against solutions 2, 3, 4 and 5.

To do this, we will separate the different criteria into 3 categories:

- Category 1 based on the morphology of the adapter
- Category 2 based on the ergonomics of the adapter
- Category 3 based on the biomechanics of the adapter

Category 1 is based on morphological criteria. Among these, we find the bulkiness (allows the full movement of the walking gait without coming up against a part of the body), the ease of attaching the system to the body of the experimenter, the weight, the number of components and the respect of the body morphology.

Category 2 is based on criteria related to the ergonomics of the adapter. First of all, we have the attachment system between the body of the experimenter and the adapter. Secondly, we have the comfort criterion which is divided into two sub-categories: comfort in static position (the wearer remains in a static position for a defined time) and dynamic comfort (the wearer walks with the prosthesis attached to the adapter for a defined time and/or distance). The third criterion is adaptability (i.e. the ease to use the device on several people without modifying it). The fourth criterion takes into account how easy it is for the user to put on and remove the device.

Category 3 is based on criteria related to biomechanical compliance. We find the criterion of fidelity between the attachment point and the shell worn by an amputee, a Canadian prosthesis in comparison with the adapter and the Canadian prosthesis (comparison in the sagittal and frontal planes). In addition, we will also study the criterion based on the comparison between the gait cycle of an able-bodied person versus the cycle performed by a person wearing the device.

2.4.1 Morphological criteria

We place solution 1 at level 0 if a solution is better than solution 1, we give it a score of 1, if it is really better, we give it a score of 2 and vice versa for -1 and -2.

Table 2.3 below represents the desirability matrix for the category based on the morphology of the adapter.

	solution 2	solution 3	solution 4	solution 5
Bulkiness	2	0	2	2
Ease of attachment to the body	1	-2	-2	-2
Weight	0	0	0	0
Number of components	0	-1	-1	0
Respect for the body morphology	-1	-2	-2	0
Total for morphology	2	-5	-3	0

Table 2.3: Assessment of solution thanks to a decision matrix on the criterion based on morphology.

Justification:

- Bulkiness: In the case of a full gait cycle (Figure 6.12), solution 1 is likely to come into contact with the experimenter's leg if the experimenter does not maintain at least 20° of anterior flexion. Solutions 2, 4 and 5 come into contact with the leg (table score = 2).

For alternative 3, there is also a risk of it coming into contact with the leg (table score = 0).

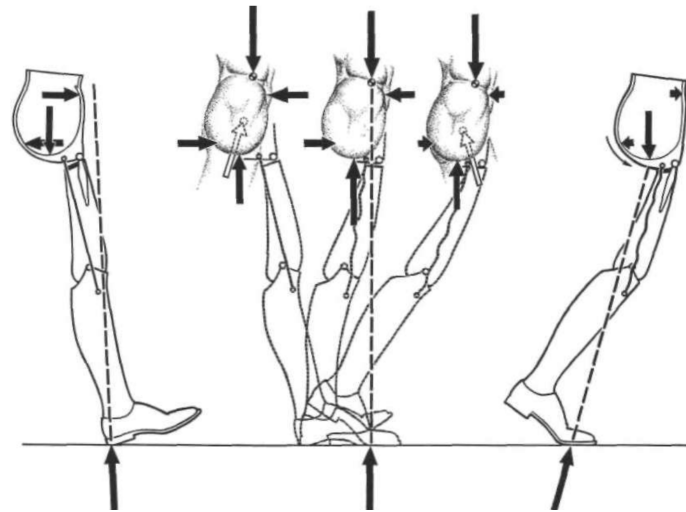


Figure 2.6: Anteroposterior force diagram of the Canadian-type hip-disarticulation prosthesis. A, Forces acting on prosthesis at heel contact; B, forces acting on stump at heel contact, mid-stance, and push-off; C, forces acting on prosthesis at push-off. [11]

- **Ease of attachment to the body:** We believe that it will be easier to attach the device to the experimenter's body via alternatives 1 and 2 because they are in direct contact with a part of the body. As a result, there will be less force applied to the attachment system on the upper body. In addition, for solution 2, all the weight will be transmitted directly to the buttock-adaptor contact. Therefore, there will be almost no upper body discomfort related to this factor (table score = 1). In contrast, for solutions 3 and 4, it will be quite complex to apply such large forces on the attachment system at the upper body level without disturbing the experimenter (table score = -2). For solution 5, we believe that the attachment system will be quite complicated to design and uncomfortable for the user (table score = -2).
- **Weight:** Regarding the weight of the device, it is rather complex to show at this stage that one solution is better than another one. Therefore, we give a score of 0 to all solutions.
- **Number of components:** we face the same problem as with the weight criterion. However, schematically, we can assume that solutions 3 and 4 include more parts via the attachment system (table score = -1) than solutions 1, 2 and 5 (table score = 0).
- **Respect of the body morphology:** in respect of the body morphology, it is above all the point of fixation and the position of the point of attachment between the prosthesis and the shell (in the sagittal and lateral planes) that are the two main criteria that should be respected. Thanks to figure 2.7, we can easily see that solution 1 and 5 respect almost identically the position of the attachment point. Solution 2, on the other hand, does not respect this position (table score = -1). But solutions 3 and 4 are even more disappointing because the attachment point is not in the plane of the body (table score = -2).

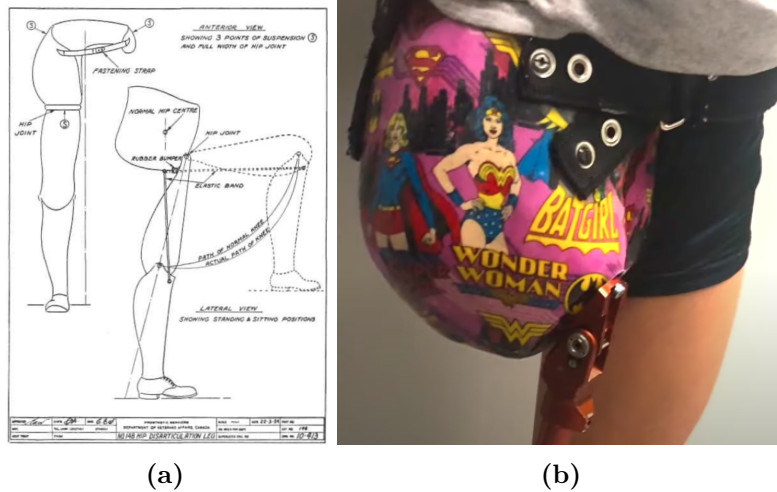


Figure 2.7: a: Original design of the Canadian-type hip-disarticulation prosthesis. From McLaurin [11] and b: Concrete vision of the attachment point. [26]

2.4.2 Criteria related to the ergonomics of the adapter

The table 2.4 below represents the decision matrix for the category based on the ergonomics of the adapter.

	solution 2	solution 3	solution 4	solution 5
Comfort in static position	0	2	2	-2
Comfort in case of dynamic movement	-2	-1	-1	-2
Adaptability	0	0	0	0
Easy to put on and take off	-2	-1	-1	0
Total regarding the ergonomics of the adapter	-4	0	0	-4

Table 2.4: Assessment of solution thanks to a decision matrix on the criterion based on ergonomics.

Justification:

- Comfort in a static position: the device does not really impose any constraints for solutions 3 and 4. The position is the same as when a person stands on one foot (table score = 2). Regarding solutions 1 and 2 which impose constraints on the experimenter's body (experimenter in unnatural position). We can judge that these are of the same type (table score = 0). Solution 5 is considered uncomfortable because it imposes great constraints on the experimenter's crotch (table score = -2).
- Comfort in case of dynamic movement: in case of dynamic movement, none of the solutions are comfortable for the user. Solutions 2 and 5 are considered the least comfortable (table score = -2), followed by solutions 3 and 4 (due to the difficulty of achieving the hip and knee movement) (table score = -1). Solution 1 is the one that best allows for a natural movement due to the position of the hip attachment point.
- Adaptability: The negative point is that the height of the Canadian prosthesis cannot be modified. However, we can imagine designing a slide system that would allow the height of the attachment point between the prosthesis and the adapter to be changed by a few centimetres. Such a system would have a similar implementation in the design of any solution. Also, a loose system similar to orthotics will also help to adjust the device to

the experimenter's body morphology. On this criterion, I therefore believe that all devices are equivalent (table score = 0).

- Easy to put on and take off: in the case of solution 2, the experimenter will need external help to put on and take off the device (table score = -2) unlike solutions 1, 3 and 4. For solutions 3 and 4, it will be necessary to ensure that either the parietal plane (for solution 3) or the frontal plane (for solution 4) includes the imaginary point of attachment between the prosthesis and the amputee's shell (table score = -1). For solution 5, the user will not have any real problem to put the device on, the negative side of this solution is the fixation of the blocked leg (table score = 0). For solution 1, as the adapter is in direct contact with the foreleg, it will only be necessary to calibrate the height of the attachment point via the slide, which should also be done for the other solutions.

2.4.3 Criteria related to the respect of biomechanic

The table 2.5 below represents the decision matrix for the category based on biomechanic compliance.

	solution 2	solution 3	solution 4	solution 5
Fidelity point of attachment	-1	-2	-2	0
Respect for the gait cycle	1	1	0	2
Total for biomechanic	0	-1	-2	2

Table 2.5: Assessment of solution thanks to a decision matrix on the criterion based on biomechanic.

Justification:

- Attachment point fidelity: as previously stated, solution 3 does not respect the position of the attachment point at all (see Figure 2.7) with respect to the frontal plane (table score = -2). The same is true for solution 4 with respect to the sagittal plane (table score = -2). For solution 2, the two points do not coincide as much as for solutions 1 and 5 (table score = -1 for solution 2 and table score = 0 for solution 5).
- Respect for the gait cycle: in the case of solutions 1, 4 and 5, there is a risk of contact between the leg and the prosthesis in dorsiflexion. This can create huge disturbances. In solution 2 and 3, there will be no obstacles when performing the movement. However, we believe that there will surely be the creation of a lateral misalignment (creation of an outward loop) when performing the movement. Therefore, we will give a score of 1 for this reason.

2.4.4 Decision matrix balance sheet

	solution 2	solution 3	solution 4	solution 5
Total	-0,6	-1,5	-0,4	0

Table 2.6: Balance decision matrix based on a fair weigh of the 3 categories according to the number of criteria.

Thanks to the above matrices, we notice that from a decision matrix point of view, solutions 1 and 5 get the same result (Table 2.6). Therefore, we will carry out the rest of our study on these two concepts until one of the two concepts will stand out.

2.5 Description of the selected solution

Thanks to the decision matrices, we have concluded that the most appropriate solutions for the realisation of our device are solutions 1 and 5 represented on figure 2.5. In the following, solution 1 will be referred to as concept 1 and solution 5 as concept 2 (see Figure 2.8).

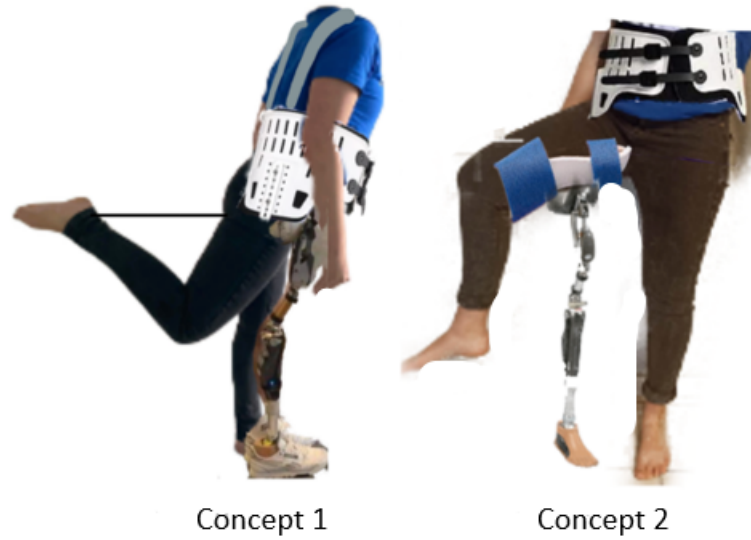


Figure 2.8: Design idea for concept 1 and 2 for the description of the selected solution.

2.5.1 Description of concept 1

The first step in creating the design for concept 1 is to determine the imposition of angles (α and β in Figure 2.9) on the locked leg to avoid any risk of contact between the device and the locked leg.

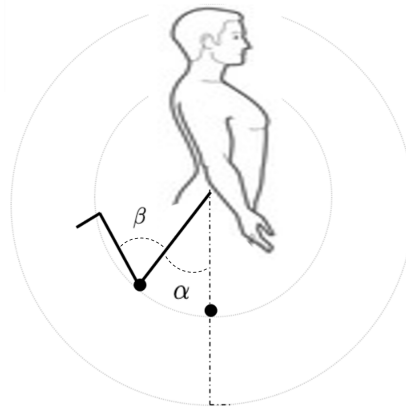


Figure 2.9: Descriptive diagram showing the angles between the blocked leg (solid line) and the leg representative of the Canadian prosthesis (dashed line).

From Figure 2.10 which represents the variation of the hip angle of a human and a passive dynamic walker in a walking cycle when the slope angle is 0.04 rad, we can see that the hip angle varies in the range $[-20^\circ; 25^\circ]$. [27]

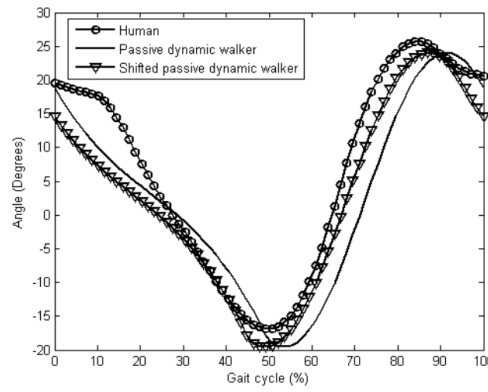


Figure 2.10: Variation of the angle at the hip during a walking cycle (slope angle=0.04 rad). [27]

Therefore, the Canadian prosthesis must enable the amputee to perform an angle movement between -20° and 25° without interacting with the leg. In order to avoid any risk, we will impose a α angle of 30° (see Table 3.4). Regarding the β angle, it is assigned a value of 90° . This value is representative of the possibility of movement of this anatomical part. However, the concept should also enable the amputee to achieve an angle α varying between 0° and 30° so that the user can increase the angle gradually according to his flexibility.

In order to help the user maintain this position over time, a cord will connect the locked foot to the lower part of the buttock (see Figure 2.8 concept 1). This rope will result in a zero torque at the locked knee.

Secondly, the Canadian prosthesis must be attached to the adapter at the level of the femoral triangle and the anterior region of the thigh. We will therefore make a support from a lower limb orthosis support (see Figure 2.11) which we will place on the front of the thigh. A laminated plate will be attached to this support and will form the link between the support and the prosthesis. The size of the support will be determined in section 3.4 of this report.

The laminate plate is a major component of a Canadian denture. It usually serves as a reinforcement between the pelvic socket and the representative hip joint. If we do not use a laminate plate, we risk damaging the socket to the point of breaking it. Therefore, in order to reinforce the device, we place a laminated plate that has superior mechanical properties to the socket.



Figure 2.11: Bracket to link the adapter to the Canadian prosthesis. This bracket will come from a lower limb orthosis from the source. [28]

Thirdly, the user will wear a back brace (see Figure 2.12) in order to distribute the forces and improve stability.



Figure 2.12: A corset that provides better support for the back. This corset will come from a lower limb orthosis from the source. [28]

Finally, the corset (Figure 2.12) and the rigid support (Figure 2.11) will be held together by a slide system (see Figure 2.13). In order to allow movement, the two slides will be connected to each other via a ball-and-socket joint. In addition, in order to limit any movement when the device is installed and to impose a constant α angle, a cover type system will be inserted at the ball joint taking up the geometry of the ball joint and the two slides.



Figure 2.13: Slide connecting the corset to the rigid support. These two guides are connected to each other via a ball and socket joint, which is locked by a cover system in order to keep the angle values constant.

2.5.2 Description of concept 2

The first stage of concept 2 is the production of the gutter (see Figure 2.2 3.a and the blue part in Figure 2.14). This gutter will be made from a thermoformed plate that is heat deformed to fit the shape of the user's thigh and calf. The dimensions of the gutter will be determined in section 3.4.

Secondly, the user will wear a back brace (see Figure 2.12) in order to distribute the forces and improve stability.

Thirdly, a rigid support will be fitted to the femoral triangle. This will be attached to both the corset and the gutter (see the burgundy piece in Figure 2.14). It is also on this piece that we will attach the Canadian prosthesis. Therefore, this piece will be made of a material that will contain the lamination plate.



Figure 2.14: Representative sketch of concept 2 without the Canadian prosthesis composed of the corset (green), the gutter (blue) and the piece connecting these two elements (burgundy).

Finally, the blocked leg will be locked into the gutter via a clamping strap system (see Figure 2.15). These will be made of fabric.



Figure 2.15: Clamping strap to maintain constant contact between the blocked leg and the gutter.

2.6 Conclusion

In conclusion, in this section we have shown that two concepts can meet our specification criteria.

However, for the time being, it is difficult for us to distinguish between them factually. That is why we will continue the rest of this report with the possibility of building both concepts (Figure 2.8). However, if one of the two appears to be a better fit to criteria, the other one will be left out.

In Chapter 1.2.3, we listed two types of adapter that already exist in the research community. However, despite the fact that they were mentioned in the solution proposals, the decision matrix showed that it was not on the criteria listed above, the best choice to meet our specifications. Indeed, the main problem of these two concepts is the fact that the prosthetic hip point is not located in an acceptable area (compared to concept 1 and 2) of the prosthetic hip point of an amputee.

Chapter 3

Load distribution on the human body

3.1 Introduction

The objective of this section is to observe the way loads and torques propagate from one body to another at the interfaces in order to distribute the loads as well as possible and so to dimension the different components of the adapter.[29]

To do this, we will dimension our two concepts using the Free Body Diagram (FBD) method applied to our multi-body system in static.

The FBD is a graphical representation of all the loads acting on a body such as, in static:

$$\sum_{i=1}^n F_n = 0$$

$$\sum_{i=1}^n M_n = 0$$

With F_n the load and M_n the torque associated with body n.

This chapter is divided into 3 sub-sections. A first subsection is devoted to the methodology. At the end of it, we will obtain the algebraic expression of the forces applied on the multi-body system of concepts 1 and 2. Then, we will transform these expressions in numerical values thanks to the data mentioned in the methodology section. Finally, we will interpret these data.

At the end of this section, we will have the results of the loads applied to the different segments and joints of the human body and the prosthesis. With these values, we can justify the geometry and requirements of the different components of both concepts.

3.2 Methodology

In this section, we will first define the different types of joints that exist in the mechanical world and that are present in the joints of our body. Secondly, we will schematise concepts 1 and 2 as a multi-body system. This will allow us to carry out the analysis of the external loads applied to each body via the FBD method in the last section.

3.2.1 Type of joint

The human body is a multi-body system composed of different types of joints.

In our study, we will focus on the following joints : pivot joint (neck), ball and socket joint (hip joint), hinge joint (knee joint) and the plane joint (between the foot and the ground) (see Figure 3.1).

When carrying out the FBD approach, it is important to think about the type of constraints(stress) that apply at the point where the cut is made. A cut can be defined as the place where one wishes to know the stress. In our analysis, an example of a cut can be located at the level of the kneecap. Depending on the type of stress, we can know the general form of the loads and torques of interaction between the two cut bodies (see Table 3.1).

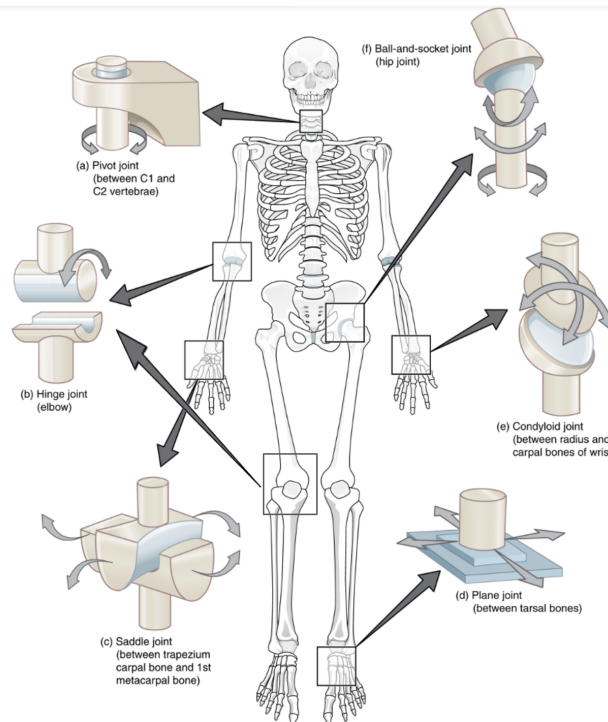


Figure 3.1: The human body is made up of six types of synovial joints. (a) Pivot joints allow rotation around an axis. (b) The hinge joint is the joint at the elbow and knee. (c) The saddle joint is the joint between the trapezoidal carpal bone and the first metacarpal bone at the base of the thumb. (d) The plane joint represents the connection between the tarsal bones of the foot and the ground. (e) The condyloid joint is representative of the radiocarpal joint of the wrist. (f) The ball and socket joints are located at the hip and shoulder joints.[30]

The X, Y and Z components represent the loads and the M_x , M_y and M_z components represent the torques relative to the direction of the coordinate system.

3.2.2 Schematic and given data

We will apply the FBD approach to concepts 1 and 2.

First of all, each concept was schematised in a diagram with the different parts of the human body (in black), the different parts of the prosthesis (in blue), the loads linked to the weight of each part of the body and the prosthesis (in red), the load reaction on the ground (in green) and the joints that are going to be cut off (large dots).

Type of joint	Transmitted load
Pivot joint	$\begin{Bmatrix} 0 & 0 \\ Y & M_y \\ Z & M_z \end{Bmatrix}$
Ball and socket joint	$\begin{Bmatrix} X & 0 \\ Y & 0 \\ Z & 0 \end{Bmatrix}$
Hinge joint	$\begin{Bmatrix} X & M_x \\ Y & M_y \\ Z & 0 \end{Bmatrix}$
Plane joint	$\begin{Bmatrix} 0 & M_x \\ 0 & M_y \\ Z & 0 \end{Bmatrix}$
Free joint	$\begin{Bmatrix} 0 & 0 \\ 0 & 0 \\ 0 & 0 \end{Bmatrix}$

Table 3.1: Each kinematic link has a transmitted force of a given form. This table shows the different forms of transmitted forces depending on the type of link.

The values, that we need for our further work, are the masses [31], the dimensions of the various elements [32] and the angles between the different elements. The angles were estimated on the basis of the morphology of the human body (in the chapter 2.5.1). The mass of the prosthesis is estimated in relation to the mass of the active prosthesis made by the PhD student Louis Devillez.

Concept 1 is schematised in Figure 3.2.

The data for concept 1 is shown in tables 3.2, 3.3 and 3.4 below.

Body		Mass	Part of the human body an prosthesis		Load
Body 1	M1	5.4 kg	Head + neck	F_1	-52.97 N
Body 2	M2	48.3 kg	Chest + pelvis + arms + hands	F_2	-473.82 N
Body 3 and 8	M3	6.5 kg	Thigh + knee	F_6	-63.77 N
Body 5	$M_{B'-E}$	1.41 kg	Mass between B' and E	F_3	-13.79 N
Body 6	$M3-M_{B'-E}$	5.09 kg	Mass between E and F	F_4	-49.98 N
Body 8 and 9	M4	4.5 kg	Passive prosthesis (1/2)	$F_8 = F_9$	-44.15 N
Body 4 and 7	M5	7.15 kg	1 Tibia + 1 calf + 1 ankle + 1 foot	$F_5 = F_7$	-70.14 N
Body 10	M_{rope}	0.16 kg	Rope	F_{12}	-1.57 N

Table 3.2: Table representing the different bodies of concept 1 (column 1), the different masses (column 2 and 3), an anatomical description of the different segments (column 4) and the external loads applied on these segments linked to the mass of the segments (column 5).

$F_{10} = \frac{F_1+F_2}{2} + F_3 + F_4 + F_5 + F_8 + F_9 + F_{12} = 487.16 \text{ N}$
$F_{11} = \frac{F_1+F_2}{2} + F_6 + F_7 = 397.31 \text{ N}$

Table 3.3: Reaction loads with the ground at point J and D in figure 3.2.

H	1.85 m	α	30°
L_1	0.2 m	β	90°
L_2	0.68 m		
L_3	0.46 m		
L_4	0.5 m		
$ B'E $	0.1 m		
L_{rope}	0.38 m		

Table 3.4: Length of the body segment and the value of the angles imposed on our device in order to install the device and limit the risk of contact between the device and the blocked leg.

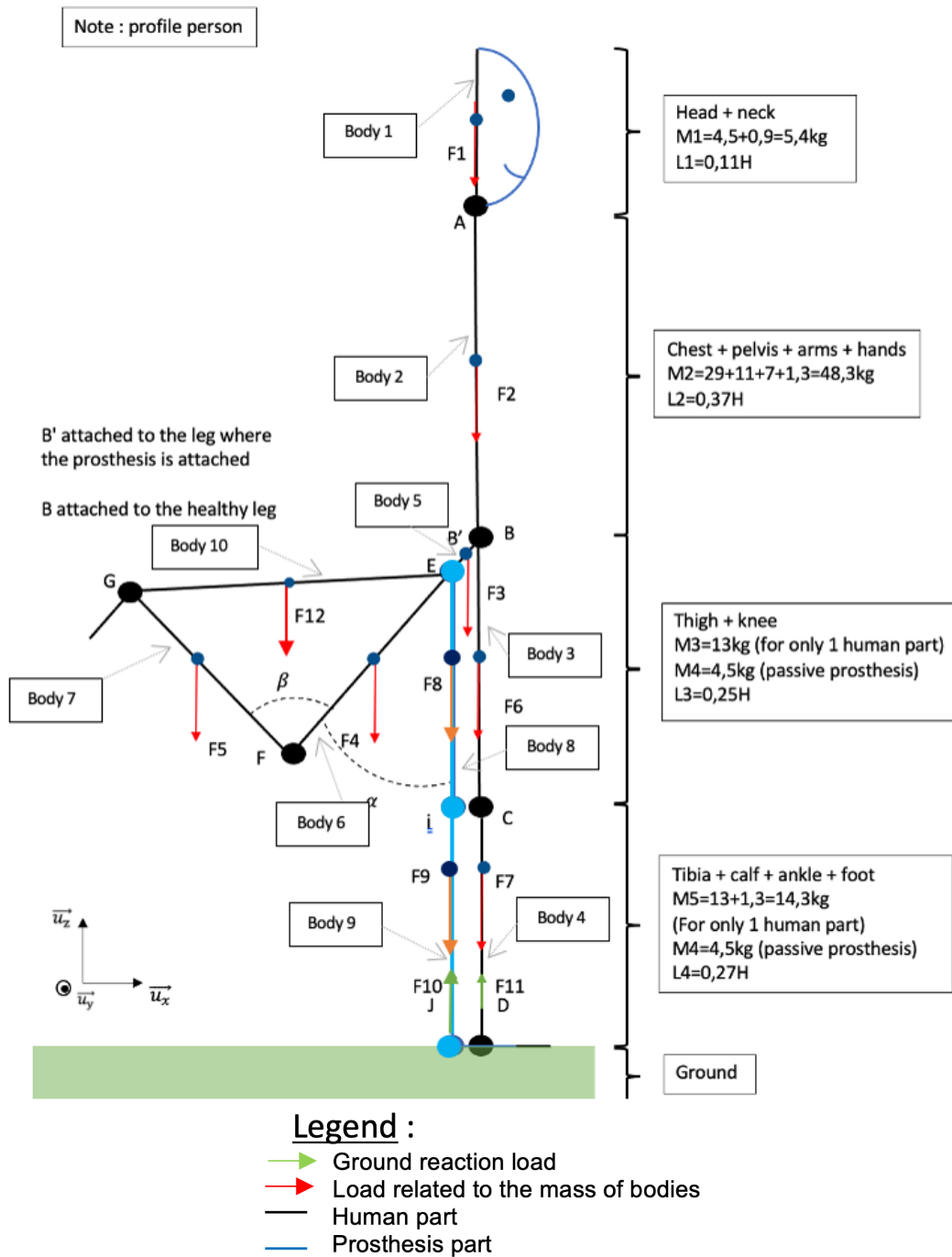


Figure 3.2: Schematic of concept 1 including external loads, joints and body and prosthesis based on the design of solution 1 in Figure 2.5.

The schematic of the system for concept 2 is depicted in Figure 3.3.

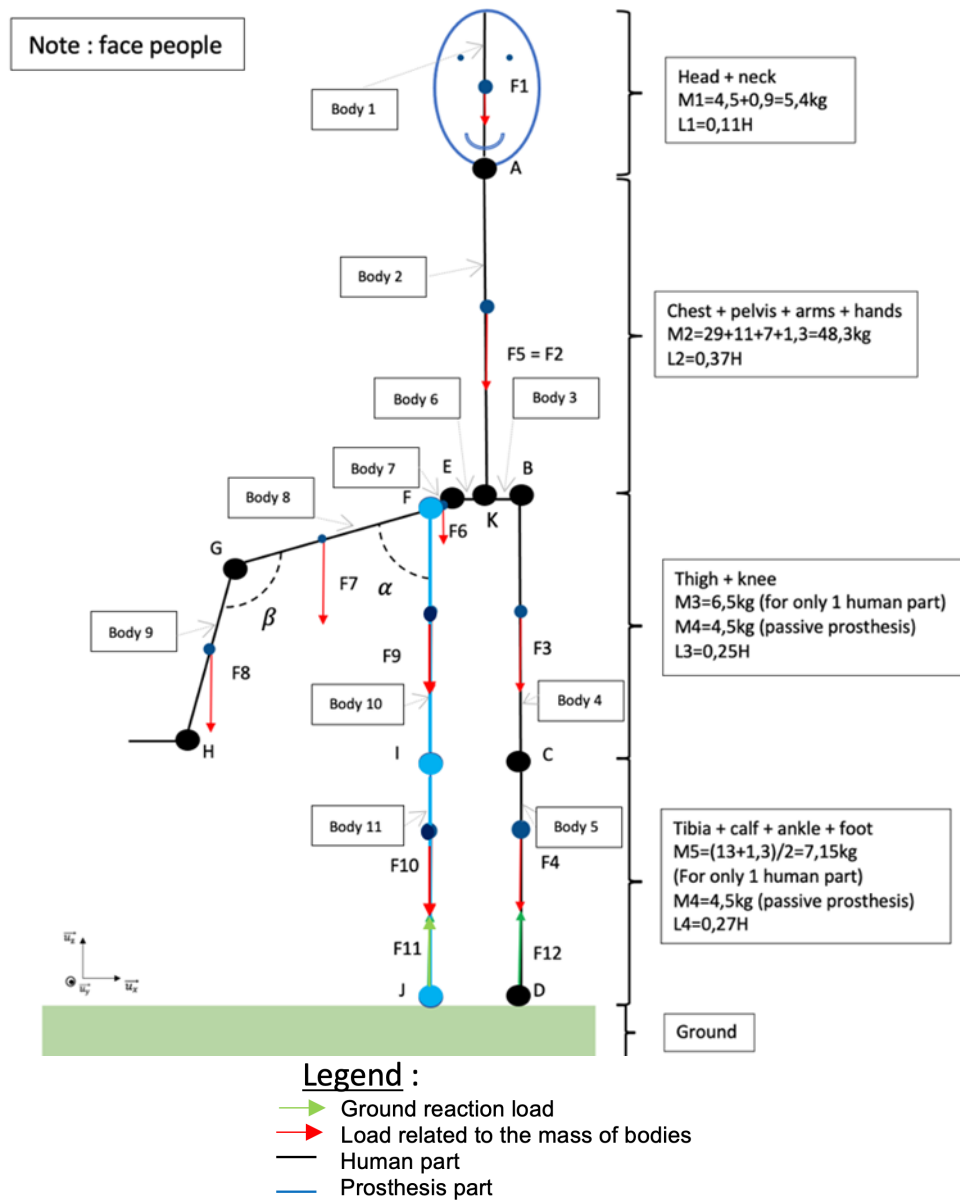


Figure 3.3: Schematic of concept 2 including external loads, joints and body and prosthesis based on the design of solution 5 in Figure 2.5.

The data for concept 2 is shown in tables 3.5, 3.6 and 3.7 below.

Body		Mass	Part of the human body		Load
Body 1	M1	5.4 kg	Head + neck	F_1	-52.97 N
Body 2	M2	48.3 kg	Chest + pelvis + arms + hands	F_2	-473.82 N
Body 4	M3	6.5 kg	Thigh + knee	F_3	-63.77 N
Body 10 and 11	M4	4.5 kg	Passive prosthesis (1/2)	$F_9 = F_{10}$	-44.15 N
Body 5 and 9	M5	7.15 kg	1 Tibia + 1 calf + 1 ankle + 1 foot	$F_4 = F_8$	-70.14 N
Body 7	M_{EF}	0.65 kg	Mass between E and F	F_6	-6.38 N
Body 8	$M3 - M_{EF}$	5.85 kg	Mass between F and G	F_7	-57.38 N

Table 3.5: Table representing the different bodies of concept 2 (column 1), the different masses (column 2 and 3), an anatomical description of the different segments (column 4) and the external loads applied on these segments linked to the mass of the segments (column 5).

$F_{11} = \frac{F_1+F_2}{2} + F_6 + F_7 + F_8 + F_9 + F_{10} = 485.6 \text{ N}$
$F_{12} = \frac{F_1+F_2}{2} + F_3 + F_4 = 397.31 \text{ N}$

Table 3.6: Reaction loads with the ground at point J and D in figure 3.3.

H	1.85 m	α	65°
L_1	0.2 m	β	130°
L_2	0.68 m		
L_3	0.46 m		
L_4	0.5 m		
$ FE $	0.05 m		
$ BK $	0.09 m		

Table 3.7: Length of the body segment and the value of the angles imposed on our device in order to install the device and limit the risk of contact between the device and the blocked leg.

The loads found in the above tables were calculated using the following formula:

$$\vec{F} = m\vec{a} \text{ with } \vec{a} = 9.81 \frac{m}{s^2} \text{ and } m = \text{mass of body}$$

3.2.3 Analysis of the external loads applying to each body

Firstly, we will focus on each body of the multi-body system (Figure 3.2 and 3.3).

An example of the FBD method applied to body 1 of concept 1 is presented below. This methodology will be applied on all the other bodies of concept 1 and 2.

- Example of FBD applied on the body 1 of concept 1:**

Let's take the example of body 1 of concept 1. This one is represented in figure 3.4 representing the external forces applied in red, the transmitted force in green, the center of the body mass as a blue circle, the extremities of the body as a black circle and the body as a black line segment. This color code will be used throughout the whole study on the FBD method.

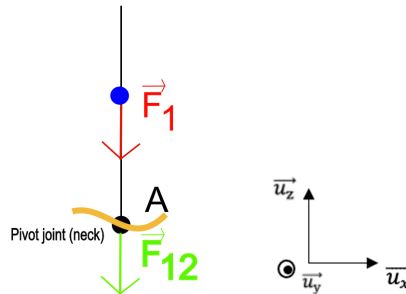
**Figure 3.4:** Example of FBD methods applied to body 1 of concept 1 representing the segment head + neck to explain the FBD method.

Figure 3.4 shows a cut at point A (cut between body 1 and 2 at the neck shown in orange in figure 3.4) represented as a joint pivot.

The external load acting on this body will be taken into account. In our case, we are only in the presence of the load linked to the weight M_1 .

At this stage of reflection, we are quite capable of calculating a balance of loads:

$$\vec{F}_{1,2} = \vec{F}_1$$

With $\vec{F}_{1,2}$ the load transmitted by body 1 on body 2.

We carried out the FBD study for body 1 of concept.

In the same way, we calculated the transmitted load equilibrium for the other set up bodies of concepts 1 and 2.

- **FBD applied on concept 1:**

The other segments of concept 1 will be first applied to this example.

To do this, we will represent all the segments with the external forces that apply to them as well as the resultant forces that they develop at their cut-off point in figure 3.5 below.

Thanks to figure 3.5, we realize the balance of forces in order to find the algebraic expressions of the loads at the end of the segments at the level of the cuts. These expressions are summarised in table 3.8 below.

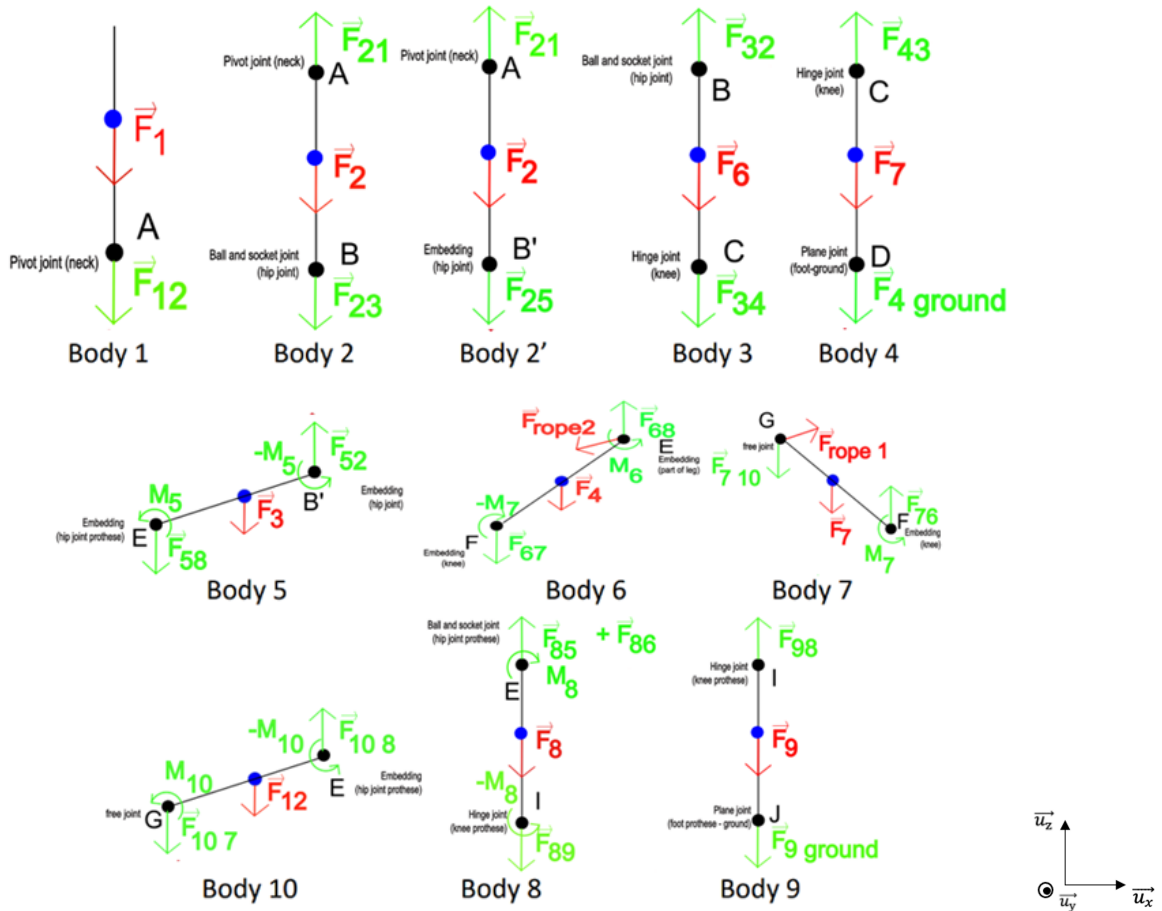


Figure 3.5: FBD applied to the segments of concept 1. On these diagrams, we can see the external forces in red and the forces transmitted to the cut-off point in green.

Body	Equilibrium of loads and torques
Body 1	$\vec{F}_{1,2} = \vec{F}_1$
Body 2	$\vec{F}_{2,1} = -\vec{F}_{1,2}$ $\vec{F}_{2,3} = \frac{(-\vec{F}_{2,1} + \vec{F}_2)}{2}$
Body 2'	$\vec{F}_{2,5} = \vec{F}_{2,3}$
Body 3	$\vec{F}_{3,2} = -\vec{F}_{2,3}$ $\vec{F}_{3,4} = -\vec{F}_{3,2} + \vec{F}_6$
Body 4	$\vec{F}_{4,3} = -\vec{F}_{3,4}$ $\vec{F}_{4,ground} = \vec{F}_{3,4} + \vec{F}_7$
Body 5	$\vec{F}_{5,2} = -\vec{F}_{2,5}$ $\vec{F}_{5,8} = -\vec{F}_{5,2} + \vec{F}_3$ $M_5 = (\vec{F}_{5,2} - \vec{F}_3/2) * \sin \alpha * B'E $
Body 6	$\vec{F}_{6,7} = -\vec{F}_{7,6}$ $\vec{F}_{6,8} = -\vec{F}_{6,7} - \vec{F}_4 - \vec{F}_{rope_z}$ $M_6 = (\frac{\vec{F}_4}{2} + \vec{F}_{6,7}) * \sin \alpha * (L_3 - EB')$
Body 10	$\vec{F}_{10,8} = 0$ $\vec{F}_{10,7} = -\vec{F}_{10,8} + \vec{F}_{12}$ $M_{10} = ((\vec{F}_{10,8} - \frac{\vec{F}_{1,2}}{2}) * \cos(15^\circ) * L_{rope}$
Body 7	$\vec{F}_{7,10} = \vec{F}_{10,7}$ $\vec{F}_{7,6} = -\vec{F}_{7,10} + \vec{F}_{rope_z} - \vec{F}_7$ $M_7 = (-\vec{F}_{rope1_z} + \vec{F}_{7,10} + \frac{\vec{F}_7}{2}) * \sin(\beta - \alpha) * L_4$
Body 8	$\vec{F}_{8,5} + \vec{F}_{8,6} = \vec{F}_{6,8} - \vec{F}_{5,8} + \vec{F}_{10,8}$ $\vec{F}_{8,9} = -(\vec{F}_{8,5} + \vec{F}_{8,6}) + \vec{F}_8$ $M_8 = M_{10} - M_5 - M_6$
Body 9	$\vec{F}_{9,8} = -\vec{F}_{8,9}$ $\vec{F}_{9,ground} = \vec{F}_{8,9} + \vec{F}_9$

Table 3.8: Algebraic expression of the forces transmitted to the cut-off point according to table data 3.2, 3.3 and 3.4.

So far, we have not yet calculated the expression of the tension load in body 10:

Expression of F_{rope1} : ($F_{rope1} = -F_{rope2}$)

Thanks to Body 10, we have:

$$\left\{ \begin{array}{l} \text{Along the x-axis: } \vec{F}_{7,10} + \vec{F}_7 - \vec{F}_{7,6} - \vec{F}_{rope1_z} = 0 \\ \text{Along the z-axis: } \vec{F}_{rope1_x} = 0 \\ M_7 = 0 = (-\vec{F}_{rope1_z} + \vec{F}_{7,10} + \frac{\vec{F}_7}{2}) * \sin(\beta - \alpha) * L_4 \end{array} \right.$$

$$\left\{ \begin{array}{l} \vec{F}_{rope1_x} = 0 \\ \vec{F}_{rope1_z} = \vec{F}_{7,10} + \frac{\vec{F}_7}{2} \\ \vec{F}_{rope1} = \vec{F}_{rope1_x} + \vec{F}_{rope1_z} \end{array} \right.$$

- FBD applied to concept 2:

The same methodology for the concept 2 will be used here at this stage.

To do this, we will represent all the segments with the external forces applied to them as well as the resultant loads they develop at their cut-off point in figure 3.6 below.

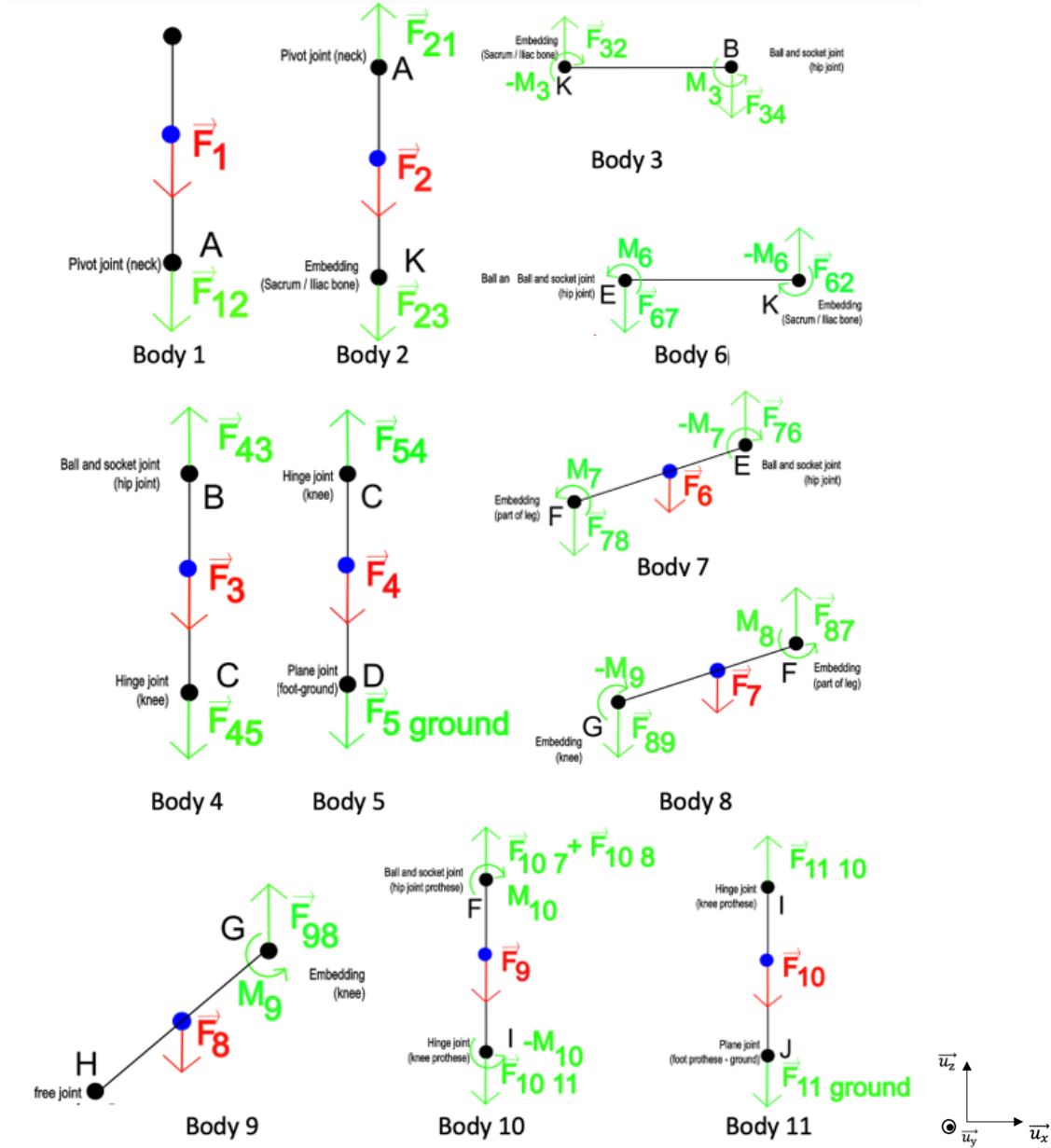


Figure 3.6: FBD applied to the segments of concept 2. These diagrams show the external forces in red and the forces transmitted to the cut-off point in green.

Figure 3.6 demonstrates that, for concept 1 the balance of loads in order to find the algebraic expressions of the loads at the end of the segments at the level of the cuts. These expressions are summarised in table 3.9 below.

Body	Equilibrium of loads and torques
Body 1	$\vec{F}_{1,2} = \vec{F}_1$
Body 2	$\vec{F}_{2,1} = -\vec{F}_{12}$ $\vec{F}_{2,3} = \vec{F}_{2,6} = \frac{-\vec{F}_{2,1} + \vec{F}_2}{2}$
Body 3	$\vec{F}_{3,2} = -\vec{F}_{2,3}$ $\vec{F}_{4,3} = -\vec{F}_{3,2}$ $M_3 = -\vec{F}_{3,2} * BK $
Body 6	$\vec{F}_{6,2} = -\vec{F}_{2,6}$ $\vec{F}_{6,7} = -\vec{F}_{6,2}$ $M_6 = \vec{F}_{6,2} * BK $
Body 4	$\vec{F}_{4,3} = -\vec{F}_{3,4}$ $\vec{F}_{4,5} = -\vec{F}_{4,3} + \vec{F}_3$
Body 5	$\vec{F}_{5,4} = -\vec{F}_{4,5}$ $\vec{F}_{5,ground} = -\vec{F}_{5,4} + \vec{F}_4$
Body 7	$\vec{F}_{7,6} = -\vec{F}_{6,7}$ $\vec{F}_{7,8} = \vec{F}_6 - \vec{F}_{7,6}$ $M_7 = (\vec{F}_{7,6} - \frac{\vec{F}_6}{2}) * \sin(\alpha) * FE $
Body 9	$\vec{F}_{9,8} = -\vec{F}_8$ $M_9 = \vec{F}_8 * \sin(\pi - \alpha - \beta) * L_4$
Body 8	$\vec{F}_{8,9} = -\vec{F}_{9,8}$ $\vec{F}_{8,7} = -\vec{F}_{8,9} - \vec{F}_7$ $M_8 = (\vec{F}_{8,9} + \frac{\vec{F}_7}{2}) * \sin \alpha * (L_3 - FE) + M_9$
Body 10	$\vec{F}_{10,7} + \vec{F}_{10,8} = \vec{F}_{8,7} - \vec{F}_{7,8}$ $\vec{F}_{10,11} = -(\vec{F}_{10,7} + \vec{F}_{10,8}) - \vec{F}_9$ $M_{10} = -M_8 - M_7$
Body 11	$\vec{F}_{11,10} = -\vec{F}_{10,11}$ $\vec{F}_{11,ground} = \vec{F}_{10,11} + \vec{F}_{10}$

Table 3.9: Algebraic expression of the forces transmitted to the cut-off point according to table data 3.5, 3.6 and 3.7.

In order to facilitate the reading and discussion of the results, we will express the numerical values of the algebraic expressions of tables 3.8 and 3.9 as a function of whether they belong to the human body or to the prosthesis. For the human body, we will still make a subdivision according to whether the body is in a neutral position (static standing) or in an abnormal position. Furthermore, we will compare similar body parts together, e.g. we will compare the leg of the prosthesis with the leg on the adapter side and the leg on the opposite side to the adapter.

For concept 1, the body in neutral position can be represented in figure by bodies 1, 2, 3 and 4. On the other hand, bodies 5, 6 and 7 represent the parts of the body in abnormal position. Finally, bodies 8, 9 and 10 represent the prosthesis and the adapter.

For concept 2, the body in neutral position can be represented in figure by bodies 1, 2, 3, 4 and 5. Bodies 6, 7, 8 and 9 represent the parts of the body in abnormal position. Bodies 10 and 11 represent the prosthesis.

Finally, for similar body parts, we can create 4 groups: head + neck; chest + pelvis + arms + hands; thigh + knee and tibia + calf + ankle + foot.

3.3 Results

In Section 3.2.3, we calculated the algebraic expression of the load applied to each cut of each segment for our two concepts, which represents the FBD equations. By solving the FBD equations for concept 1 (table 3.8) and concept 2 (table 3.9) from the data in tables 3.2, 3.3 and 3.4 for concept 1 and tables 3.5, 3.6 and 3.7 for concept 2, we obtain the numerical expression for the different loads and torques. tables 3.10 and 3.11 represent the expressions of these loads and torques as a function of the part of the body represented and as a function of whether the section is in a neutral, abnormal or prosthetic position. Comparable efforts are aligned in tables to facilitate the reading.

	Neutral position	Abnormal position	Prosthesis
<i>Head + neck</i>	$\vec{F}_{1,2} = -52.97N$		
<i>Chest + pelvis+ arms + hands</i>	$\vec{F}_{2,1} = 52.97N$ $\vec{F}_{2,3} = -263.4N$	$\vec{F}_{2,5} = -263.4N$	
<i>Thigh + knee</i>	$\vec{F}_{3,2} = 263.4N$ $\vec{F}_{3,4} = -327.16N$	$\vec{F}_{5,2} = 263.4N$ $\vec{F}_{6,7} = -35.07N$ $\vec{F}_{5,8} = -277.19N$ $\vec{F}_{6,8} = 85.05N$ $M_5 = 13.51Nm$ $M_6 = -7.7Nm$	$\vec{F}_{8,9} = -406.38N$ $F_{8,5+8,6} = 362.23N$. $M_8 = -5.51Nm$.
<i>Tibia + calf+ ankle + foot</i>	$\vec{F}_{4,3} = 327.16N$ $\vec{F}_{4,ground} = -397.31N$	$\vec{F}_{7,6} = 37.07N$ $\vec{F}_{7,10} = -1.57N$	$\vec{F}_{9,8} = 406.38N$ $\vec{F}_{9,ground} = -450.52N$

Table 3.10: Numerical expression of the forces transmitted at the cut-off point according to table 3.8. This table is separated into functions of the different representative sections of the body according to whether this section is in a neutral, abnormal or prosthetic position for concept 1.

	Neutral position	Abnormal position	Prosthesis
<i>Head + neck</i>	$\vec{F}_{1,2} = -52.97N$		
<i>Chest + pelvis+ arms + hands</i>	$\vec{F}_{2,1} = 52.97N$ $\vec{F}_{2,3} = \vec{F}_{2,6} = -263.4N$ $\vec{F}_{3,2} = 263.4N$ $\vec{F}_{3,4} = -263.4N$ $M_3 = -23.5Nm$	. . $\vec{F}_{6,2} = 263.4N$ $\vec{F}_{6,7} = -263.4N$ $M_6 = 23.5Nm$	
<i>Thigh + knee</i>	$\vec{F}_{4,3} = 263.4N$ $\vec{F}_{4,5} = -327.16N$	$\vec{F}_{7,6} = 263.4N$ $\vec{F}_{8,9} = -70.14N$ $\vec{F}_{7,8} = -269.78N$ $\vec{F}_{8,7} = 127.52N$ $M_7 = 12.08Nm$ $M_8 = -41.48Nm$	$\vec{F}_{10,11} = -441.45N$ $F_{10,7+10,8} = 397.31N$
<i>Tibia + calf+ ankle + foot</i>	$\vec{F}_{5,4} = 327.16N$ $\vec{F}_{5,ground} = -397.31N$.	$\vec{F}_{9,8} = 70.14N$. $M_9 = -4.53Nm$	$\vec{F}_{11,10} = 441.45N$ $\vec{F}_{11,ground} = -485.6N$.

Table 3.11: Numerical expression of the forces transmitted to the cut-off point according to table 3.9. This table is separated into functions of the different representative sections of the body according to whether this section is in a neutral, abnormal or prosthetic position for concept 2.

3.4 Discussion

In the previous subsection, we were able to obtain the numerical value of the loads and torques applied at different levels of the body. In this section, we will discuss these different values by comparing them to the loads applied to the human body in the static standing position. This position will be listed below as the neutral position of the body.

To do this, we will divide our discussion into two parts. First, we will study the parasitic loads, these are loads that have a value greater than the loads applied at the same point on the neutral body. Secondly, we will study the loads that apply to the interface between the body and the prosthesis (point E of concept 1 and F of concept 2).

This approach will be applied to both concepts.

- Applied discussion on concept 1:

In table 3.10, we can notice that the head + neck and chest + pelvis + arms + hands sections do not show any parasitic effort. The differences between the different bodies appear in the lower limbs.

First, let's focus on the thigh + knee section. Efforts at the level of the knees are not identical and we have the presence of a parasitic effort at the level of the knee of the prosthesis. This must be taken into account when designing the knee joint of the prosthesis. Concerning the moments, in the neutral position, this part of the body undergoes a zero moment in the configuration of figure 3.2. However, we notice the appearance of a moment M_5 applied to the hip of the experimenter (point B'). In order to improve our concept, we will install a corset the task of which is to distribute the M_5 torque in a series of loads on a much larger surface.

The moment M_5 is the moment resulting from the moments M_6 and M_8 located at point E (prosthesis-body interface). Therefore, the addition of this corset will also decrease the impact of these moments in the system.

Furthermore, the addition of the rope in concept 1 has a very beneficial effect for the user, as it induces a zero M_7 moment at the locked knee which will increase the comfort of the device.

At the level of the body-prosthesis interface (point E), 3 forces appear in table 3.10 but the force $F_{8,5+8,6}$ represents the resultant force of the forces $F_{5,8}$ and $F_{6,8}$. Therefore, the resultant force at point E is 362.23 N. In order to avoid any risk of injury to the skin and muscle, we must correctly size the bearing surface of the adapter (where the prosthesis will be attached) on the thigh.

It has been shown that muscle damage occurs when a pressure of 500 mmHg (66.6 kPa) is applied over a period of 4 hours, while skin destruction requires a pressure of 800 mmHg (106.6 kPa) over the same period of time.[33]

In order to make the device as comfortable as possible for the user, we will size the rigid support using a pressure limit of 500 mmHg with a safety factor of 10 [29]:

Literature pressure = P_{lit} [33]	66.6 kPa
Safety factor : SF	10
Limit pressure = P_{lim}	6.6 kPa
Load to the point E : $F_{8,5+8,6}$	362.23 N
Limit surface = $S_{lim} = \frac{F_{8,5+8,6}}{P_{lim}}$	543.4 cm^2

Table 3.12: Mathematical development to obtain the minimum surface on which the Canadian prosthesis support must be applied in order to avoid any risk of injury to the user for concept 1.

We find that the surface area of the rigid support, which is in contact with the skin, must be greater than or equal to 543.4 cm^2 (table 3.12). This means that if a width of 10 cm is imposed, the length of the support must be greater than or equal to 54.34 cm.

As a reminder, the leg segment on which the support rests is approximately 46 cm for a person measuring 185 cm [32]. This is why it is not possible to meet this requirement in order to avoid any risk of injury to the skin and the subcutaneous muscle. Therefore, the support between the prosthesis and the thigh will not only rest on the thigh, but will also extend over the knees (knee-hugging support) so that the forces acting at point E are not just supported by the frictional force between the skin and the support.

In order to improve the device and to keep the α angle constant, a metal support consisting of two rods will connect the corset to the rigid support which is located in the anterior thigh region.

In addition, the ball-and-socket connection between the two rods will be locked by a quick-release element so that the user can lock the angle α independently. This element will be of the cover type so that it does not pierce the connection, which could weaken it. This cover will also take up some of the M_5 torque reducing the stress on the hip joint.

In the Tibia + calf + ankle + foot section, we notice that the effort in the prosthetic leg is higher than the effort in the neutral leg. This factor should be taken into account when designing the prosthesis.

- Applied discussion on concept 2:

All these efforts are listed in table 3.11.

In the same way as for concept 1, the parasitic efforts do not appear on the upper body.

If we analyze the forces applied to the chest + pelvis + arms + hands, we notice, in the same way as for concept 1, a greater force at the knee of the prosthesis, which should be taken into account when designing the prosthesis.

At the body-prosthesis interface, the force $F_{10,7+10,8}$ is the resultant force of the forces $F_{7,8}$ and $F_{8,7}$. In order to avoid any risk of injury to the skin and muscle, the bearing surface of the adapter on the thigh must be correctly sized. Therefore, a trough will have to be designed (see Figure 3.a) to absorb these stresses.

Using the same methodology as for concept 1, we obtain :

Literature pressure = P_{lit} [33]	66.6 kPa
Safety factor : SF	10
Limit pressure = P_{lim}	6.6 kPa
Load at point E : $F_{10,7+10,8}$	397.31 N
Limit surface = $S_{lim} = \frac{F_{10,7+10,8}}{P_{lim}}$	596 cm^2

Table 3.13: Mathematical development to obtain the minimum surface on which the Canadian prosthesis support must be applied in order to avoid any risk of injury to the user for concept 2

The surface area of the rigid support, which is in contact with the skin, must be greater than or equal to 596 cm^2 (table 3.13). This means that if a gutter width of 10 cm is imposed, the length of the support must be greater than or equal to 59.6 cm.

This support will therefore have to extend over the entire body 8 and part of body 9.

Also, at the interface between the body and the prosthesis, we have a relatively high torque M_7 and M_8 compared to the torques present in concept 1. In order to limit this, we will need to design a part to connect the splint and the brace (see Figure 2). This part will be made of a composite material on which the laminated plate will be integrated in order to make the link between the adapter and the prosthesis. This part must also be able to absorb stress without injuring the user. Thus, it must be designed so that a man and a woman can put on the device and so that it can be placed in the buttock, at the top of the anterior and posterior thigh and in the triangle.

In order for all parts to be easily assembled and disassembled, we will use ball joint systems, which will be locked by a clamping system in order to keep the angles α and β constant (see Figure 3.3).

At the level of the tibia + calf + ankle + foot segments, effort at the level of the prosthesis segment are more important. These must also be taken into account when designing the prosthesis.

3.5 Conclusion

In this section, we have studied how loads propagate from one body to another at the interfaces between the different bodies in order to size the different components of the adapter. [29]

We have thus obtained the different critical points of the two concepts proposed in section 2 design. Indeed, we have calculated the different constraints that apply to the experimenter and the prosthesis adaptor. Thanks to this, we were able to dimension certain parts of the adapters in order to avoid any risk of injury and pain. In addition, these results can be used for the manufacture of the various components of the device.

However, following this study, we noticed that concept 2 has higher stresses than concept 1. This is especially true for the torques applied at the interface between the prosthesis and the adapter and at the hip (M_7 and M_8). Due to these higher stresses, the adapter will have to bear higher stresses and will therefore be less comfortable for the user.

We therefore believe that making the adaptor represented by concept 1 is more beneficial from a user comfort point of view but also from a stability and load bearing point of view.

Therefore, in the continuation of this work, we will only design the adapter for concept 1 (see Figure 3.2).

Thanks to the results mentioned in this chapter, we can visualise concept we will be working on. This will consist of two main parts: a corset and a rigid support extending from the top of the thigh to the knee, following the shape of the latter. The support and the corset will both be linked together by a system of adjustable metal rods via a revolute joint blocked by a clamping system. A rope will run from the locked foot to point E (Figure 3.2) in order to limit the stress on the leg blocked at height. Finally, the Canadian prosthesis will be fixed to the laminated plate attached to the rigid support at point E.

Chapter 4

Design, embodiment and assembly of the main elements of prototype

The objective of this chapter is to build the adapter prototype. In order to do this, we must first carry out a study of the different parts that make up the adapter. For each part, we have to choose the material(s) and the manufacturing process(es).

First, we will list on the main elements of prototype:

- Rigid support (positioned on the front of the thigh and must follow the shape of the knee to take a pressure limit of 6.7 kPa (see Table 3.12));
- Corset;
- Slides with articulated ball joint (the slides link the corset and the rigid support);
- Laminated plate (it connects the rigid support to the Canadian prosthesis);
- Locking cover (the cover will lock the slides in order to maintain the α angle at a value of 15°);
- Rope (the rope connects the blocked foot to the lower part of the buttock);
- Canadian prosthesis.

Secondly, we separate the parts that are going to be manufactured from raw materials and the parts that are going to be purchased already in final state.

Regarding the parts that will be purchased in their final (or almost final) state, we find the corset and the slide with articulated ball joint.

Finally, the parts that will be manufactured from raw materials are the laminated plate, the Canadian prosthesis, the rope, the rigid support and the locking cover.

4.1 Rigid support, corset and slides with articulated ball joint

The rigid support, corset and slides with articulated ball joint are grouped together in the purchase of a rehabilitation orthosis for a hip fracture available in the following reference [28].

4.1.1 Description and modification

Figure 4.1 represents the device.



Figure 4.1: Rehabilitation prosthesis for the postoperative lower limbs from which the rigid support, the corset and the slide with articulated ball joint will be drawn. [28]

This brace has a design having the following characteristics:

- Structure: pelvis frame, thigh hoop, abduction frame, ankle and foot support frame, etc;
- Material: polyethylene, breathable composite fabric, metal alloy support, etc;
- Applicable height: 150-190 cm;
- Adjustable to the waist, thigh and also to the height of the brace;
- Hip and knee joints, with limited position adjustment, can be fixed and moved to different positions;
- The angles of adduction, abduction, flexion and extension of the hip joint can be adjusted according to the patient's needs.

In our prototype, we will keep the corset, the ankle and foot support frame, the breathable fabric, a slide for the knee joint and the tightening straps.

The first step of transformation is the deformation of the ankle and foot support frame. The objective of this step is to widen the frame in order to be able to insert the front of the thigh and the knee which are wider than the calf and the foot.

To do this, we will use a heat gun to slightly melt the plastic to give it the desired shape (thermoforming).

The second step is the cutting of the two tops of the ankle and foot support frame along the dotted line in figure 4.2. Without this removal, the top of the frame comes into contact with the groin at the crotch of the experimenter which creates a strong discomfort when using the device.

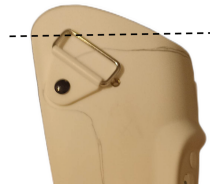


Figure 4.2: Line representing the place of cutting of the two tops of the support frame in order to avoid any pain during the use of the device.

There will be 2 holes in the sole of the foot and 2 holes on the side normally intended for the shin. These holes are used to attach the laminated plate to the frame. In order to make these holes, we will first drill the laminate plate and then place the laminate plate and the frame in their final position to deduce the position of the holes in the frame.

The part of the frame supporting the peg and the leg after these three operations obtains its final shape (see figure 4.3).



Figure 4.3: Final rigid support (without the drillings) that will fit at the front of the thigh and knee in the final device. The blue circles represent the location of the M6 holes.

The fourth and final step is the modification of the breathable composite fabrics that make the device comfortable. In our prototyping, we will not use the part that had the main purpose of maintaining the thigh. Therefore, we can recover the foam of this section as well as the tightening slack. The recovered foam will be cut in two parts and sewn to the foam that was mainly intended under the sole of the foot. Finally, the part of the adapter in contact with the knee and the anterior tibia will be protected with a triple layer of foam while the face of the thigh will be protected by a single layer of foam.

Figure 4.4 shows all the parts of the rehabilitation orthosis that will be used to create the prototype.



Figure 4.4: Parts of the rehabilitation orthosis used to make the prototype.

4.1.2 Canadian prosthesis

The Canadian prosthesis that we have for the realization of the tests on the adapter is a passive Canadian prosthesis of the left leg of the Ottobock brand represented in figure 4.5.



Figure 4.5: Canadian passive left leg prosthesis used to perform stabilization tests on the adapter.

In figure 4.5, there is also a pelvic socket made to be measured (for the person who provided us with the prosthesis after use). However, for the realization of our device, we will not use this pelvic socket (adapted to fit the stump and the pelvis of the donor). We will therefore detach the socket from the metal structure and we will just keep the metal structure and the foot of the prosthesis (see Figure 6.1).



Figure 4.6: Section of the Canadian prosthesis that will be used for the realization of prototype.

This one is composed of 6 main elements. Their technical sheets are available in the appendix 1. However, the prosthesis in figure 6.1 cannot be directly adjusted because its height is not adjusted to the height of the hip of the experimenter (described in section 5). Therefore, the tube that connects the hip joint to the knee joint will be replaced by a tube of the same external (ϕ_i 30mm) and internal (ϕ_e 25mm) diameter but with a length of 365mm.

4.2 Laminated plate

The laminate plate is an important element in a Canadian prosthesis. It is the link between the pelvic socket and the modular hip joint (see Figure 4.7).

Literature refers to laminated plate in figure 4.7 as casting anchor. In the Ottobock catalog we find it under the reference number 7Z53 (data sheet available in Appendix 1).

A lamination plate is also included in the base brace of the prosthesis in figure 4.5 but it was left out of our application because its does not match the desired geometry.

Therefore, we are going to create a plate specially dedicated for our application.

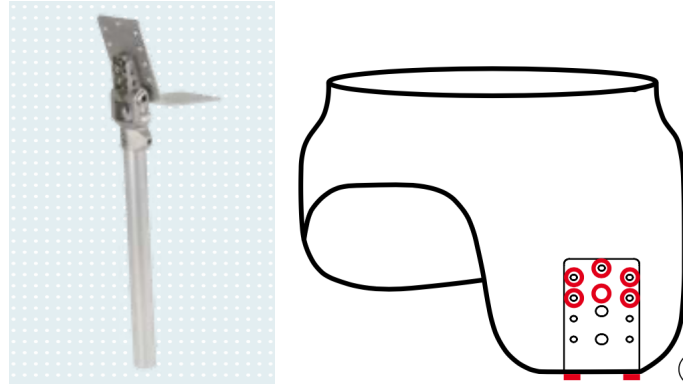


Figure 4.7: The upper part of the hip prosthesis is screwed to the laminated plate in the pelvic cavity (see on the right) and the lower part is connected to a clamp (see on the left).

To do this, we will first choose a material, explain its properties and then finish on the design of the part that will link the prosthesis to the rigid support.

4.2.1 Choice of materials

The flat lamination present in the Canadian prosthesis is made of either aluminum alloy or steel (see technical data sheet available in appendix 1).

We will assume that this part must have the following properties to suit our application:

- Have a low density ($< 3500 \frac{kg}{m^3}$);
- Be usable in a temperature range of $[-20^{\circ}; 40^{\circ}]$;
- Be easily machinable;
- Have a good hardness (Brinell hardness between 75 to 125);
- ect.

The main benefit of using rolled plates made from aluminum alloy is that the density of aluminum ($2,7 \frac{g}{cm^3}$) is lower than the density of steel ($8 \frac{g}{cm^3}$).

As a result of this characteristic, we will tend to go for a rolled plate made of aluminum alloy.

Also, in order to avoid any permanent deformation on our plate, the maximum pressure applied on the plate must be lower than the yield strength (S_y) of the chosen alloy. The maximum pressure will take place at point E (Figure 3.2) and the force applied at this point is 362 N. Assuming that the contact surface between the Canadian prosthesis and the laminated plate is 21 cm^2 ($7 \times 3 \text{ cm}$) and choosing a safety coefficient (ks) of 10, we can compute the pressure. The results are in table 4.1:

Force	90.16x9.81 N
Surface area	21 cm^2
ks	10
Pressure	4.21 MPa

Table 4.1: Development to obtain the pressure applied between the laminated plate and the Canadian prosthesis in order to know the limit S_y .

We notice that the applied pressure is low compared to the S_y of the aluminum alloys (between 34 and 505 MPa in table 4.9).

The aluminum alloys are classified in 9 series (see Figure 4.8) according to the components associated with aluminum:

Alloy Series	Principal Alloying Element
1xx.x	99.000% minimum Aluminum
2xx.x	Copper
3xx.x	Silicon Plus Copper and/or Magnesium
4xx.x	Silicon
5xx.x	Magnesium
6xx.x	Unused Series
7xx.x	Zinc
8xx.x	Tin
9xx.x	Other Elements

Figure 4.8: Series of aluminum alloys based on the molecular combination of aluminum with another element. [34]

Depending on the composition, an aluminum alloy has different properties. Table 4.9 represents the most common aluminum alloys used in the industry.

Appendix C-10 Mechanical Properties, Characteristics, and Typical Uses of Some Wrought Aluminum Alloys

Alloy	Brinell Hardness, H_B	Tensile Strength				Elongation in 2 in. (%)	Corrosion Resistance	Cold Work	Machine	Braz	Gas Weld	Arc Weld	Resistance Weld	Typical Uses
		Ultimate, S_u		Yield, S_y										
		ksi	MPa	ksi	MPa									
1100-0	23	13	90	5	34	45	A	A	E	A	A	A	B	Spinnings, drawn shapes, heat exchangers, cooking utensils, tanks
-H14	32	18	125	17	115	20	A	A	D	A	A	A	A	
-H18	44	24	165	22	150	15	A	B	D	A	A	A	A	
2011-T3	95	55	380	43	295	15	D	C	A	D	D	D	D	Screw machine parts
-T8	100	59	405	45	310	15	D	D	A	D	D	D	D	
2014-0	45	27	185	14	97	18	—	—	D	D	D	D	B	Heavy-duty forgings, aircraft structures and fittings, truck frames
-T4	105	62	425	42	290	20	D	C	B	D	D	B	B	
-T6	135	70	485	60	415	13	D	D	B	D	D	B	B	
2024-0	47	27	185	11	76	22	—	—	D	D	D	D	D	Aircraft structures, truck wheels, screw machine parts
-T4	120	68	470	47	325	19	D	C	B	D	C	B	B	
6061-0	30	18	125	8	55	30	B	A	D	A	A	A	B	Boats, rail cars, pipe, flanges, trailers
-T6	95	45	310	40	275	17	B	C	C	A	A	A	A	
6063-0	25	13	90	7	48	—	A	A	—	A	A	A	A	Furniture tube, doors, windows, pipe, fuel tanks
-T6	73	35	240	31	215	12	A	C	C	A	A	A	A	
7075-0	60	38	230	15	105	16	—	—	D	D	D	C	B	Aircraft structures and skins, skis, railings
-T6	150	83	570	73	505	11	C	D	B	D	D	C	B	

Note: Values are approximate median expectations for sizes about $\frac{1}{2}$ in. The H_B values were obtained from 500-kg load and 10-mm ball. Letters A, B, C, D indicate relative ratings in decreasing order of merit.

Source: ASM Metals Reference Book, American Society for Metals, Metals Park, Ohio, 1981.

Figure 4.9: Mechanical properties of the main aluminum alloys used in industry. [29]

Alloy 1100 is highly flexible, which could be unfavorable to our application. [35]

Alloy 2011 and alloy 2014 are known for their high strength and machinability due to the addition of lead. These are two criteria that we need for the production of the part. However, they are also known to make thick parts, which is not the characteristic of the part to be manufactured since it could cause problems when cold bending the part. [35]

Alloy 2024 is one of the most widely used high strength aluminum alloy. It has a good strength to weight ratio and high fatigue strength. It is intended to be in the form of thick sheet for machining corresponding to our demand. [35]

Alloy 6061 and 6063 are alloys specially designed to work in corrosive environment. We do not need such an alloy for our application. [35]

Finally, the 7075 alloy is known for its high strength. This one is used for the realization of extruded and forged parts, which does not meet our application either. [35]

Following these characteristics, we chose aluminum alloy of the 2024 series.

4.2.2 Property of the material

The aluminum alloy 2024 also called AlCu_4Mg_1 has the anatomical composition listed in table 4.2:

Si	Fe	Cu	Mn	Mg	Cr	Zn	Ti	Other	Al
0.5	0.5	3.8-4.9	0.3-0.9	1.2-1.8	0.1	0.25	0.15	0.15	rest

Table 4.2: Composition of aluminum alloy 2024 also referred to as AlCu_4Mg_1 . [36]

In the family of aluminum alloys there are different subcategories depending on the heat treatment applied (see Figure 4.10):

Appendix C-12 Temper Designations for Aluminum and Magnesium Alloys

Temper	Process
F	As cast
0	Annealed
Hxx	Strain-hardened. First digit indicates the specific combination of operations, second digit indicates the degree of strain hardening; thus H18 indicates a greater degree of hardening than does H14 or H24
T3	Solution-heat-treated, cold-worked, and naturally aged
T4	Solution-heat-treated and naturally aged
T5	Cooled from an elevated-temperature shaping process and artificially aged
T6	Solution-heat-treated and artificially aged
T8	Solution-heat-treated, cold-worked, and artificially aged

Figure 4.10: Name of heat treatments applied to aluminum and magnesium alloys. [29]

Alloy 2024 is often represented in the 0 and T4 form (see Figure 4.9). The T4 form presents mechanical properties 4 times superior to the 0 form. Therefore, we choose the 2024 T4 alloy for the realization of our rolled plate. This one will have the properties listed in table 4.3:

State	Specified thickness [mm]	S_u [MPa]	S_y [MPa]	Elongation %	Hardness HBS	Young's modulus [GPa]	Density $[\frac{kg}{m^3}]$
T4	1.5-6	425	275	14	120	73	2790

Table 4.3: Mechanical properties of aluminum alloy 2024 T4 for sheets (according to EN 485-2). [36]

As for the thickness of the plate, we will use a plate between 2.5 and 3 mm. [13]

4.2.3 Part design

The laminated plate is fixed between the orthosis (foot part, ankle part, deformed calf) and the prosthesis. In order to improve the rigidity of the laminated plate and to be able to adapt the device easily for users of different sizes without having to modify the device, we are going to integrate a Bosch-type profile between the laminated plate and the connecting part of the prosthesis (see Figure 4.11).

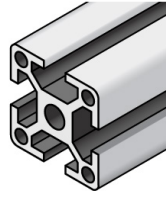


Figure 4.11: Bosch profile 30x30mm HFS. [37]

TO best size the laminated plate and the profile, we will base ourselves on the dimensions of the rigid support, the prosthesis connection piece and the size of the L3 section (Figure 3.2) which corresponds to the size of the femur.

The foot, ankle, and deformed calf orthosis will be used at the thigh and knee. It measures in the frontal plane 360 mm in height with a width that varies between 125 and 160 mm and in the transverse plane 265 mm in length and between 100 and 125 mm in width.

In order to the laminated plate to best fit the geometry of this support, we are going to manufacture our laminated plate in the shape of an "L". Its dimensions will be 365 mm long on the large side and 120 mm long on the small side.

The component part of the prosthesis which will be in contact with the profile which is itself in contact with the laminated plate as shown in figure 4.12. This part has a contact surface of 21 cm^2 ($70 \times 30\text{ mm}$) and the fixation with the laminated plate will be done via M6 drilling with a pre-existing distance of 30 mm.

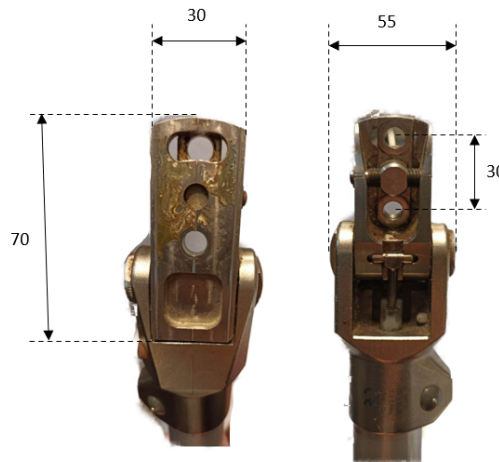


Figure 4.12: Pre-existing connection piece of the prosthesis with the laminated plate.

The section of the profile and the width of the laminated plate will be imposed on the low of the connecting part. Therefore, the profile will have a section of 30x30 mm in order to fit better with the connecting part but the choice of a profile with a larger section would not have been beneficial to our device due to the increase in mass.

Regarding the width the L-shape, we will based on both the communicating part of the prosthesis and the rigid support of the orthosis. The choice of a width of 30 mm would not be profitable for the device because the contact surface would not be large enough with the rigid support to ensure a good stability of the system. In order to increase this contact surface while trying to limit the mass of the part, we will make an arbitrary choice with a width of 55 mm.

Finally, the length of the profile will be based on the length of the femur $L_3 = 460$ mm. However, imposing the same length on the profile would not be correct because the thickness of the laminated plate, the protective foam and the rigid support must be taken into account as well as the head of the femur which is in contact with the pelvis. This is why we will impose a value of 540 mm length for the profile. In addition, this type of profile is specially used for its grooves, which allows the passage of special screws that can be tightened without drilling beforehand. This characteristic is going to be very interesting in our application. In fact, thanks to the sliding of the screw in its groove, we are going to be able to regulate easily the position of the various elements(Figure 4.13).

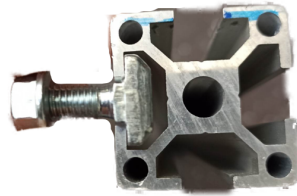


Figure 4.13: Bosch profile with special screw holes. The position of the screws in the profile is not fixed and can vary depending on the place where the clamping is done.

To summarize, the Bosch profile will have a section of 30x30 mm and a length of 540 mm and the laminated plate will be L-shaped with thickness of the plate will be 3 mm for a width of 55 mm and a length of the large section of 360 mm and a length of the small section of 120 mm (the plan of this piece (laminated plate) is available in appendix 3).

4.3 Design of the blocking at the level of blocked hip

In the design phase in Chapter II, we opted for a hip joint locked with a cover type device (Table 2.2 Case 2.e). However, the "rigid support, corset and slide with ball joint" device already includes a locking system based on a system of wedges (pins) (see Table 2.2 Case 2.a and Figure 4.14).

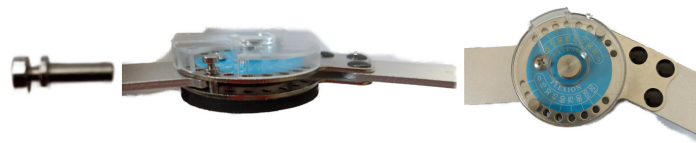


Figure 4.14: Wedges type locking system (pins) present in the "rigid support, corset and slide with ball joint" orthosis.

The slide that will be used to lock the hip joint at 30° is the slide used for locking the knee in the "rigid support, corset and slide with ball joint" device.

In the design of our prototype, it is interesting to study this wedge type locking already present on the device in order to know if it is strong enough to support the M_5 moment of 13.51 Nm (Table 3.10).

To do this, we will perform a test assuming that the sizing of the knee joint of the brace can sustain a torque without deforming the device, even when blocking the knee joint at 30° with the wedge system for a user of 100kg.

This torque is estimated in figure 4.15. A safety factor of 2 is applied because it is known that, while walking, the impact forces are more or less 2 times the weight forces due to gravity.

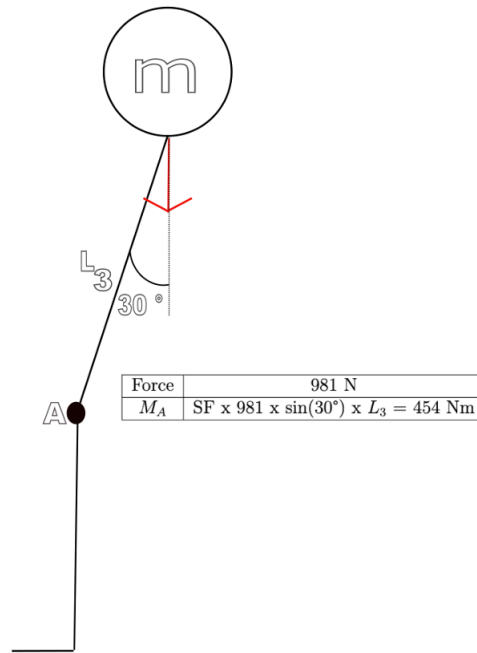


Figure 4.15: Diagram representing the hypothesis that the dimensioning of the knee joint of the orthosis can support a torque M_A such that if we come to block the knee joint at 30° thanks to the wedge system for a user of 100kg, the device should not deform.

Following the calculation of M_A , is the moment taken up by the wedge system of the slide at the knee, we find that M_A has an approximate value of 454 Nm. This value is much higher than M_5 of 13.51 Nm. As a result, the wedge-type locking system available in the rehabilitation brace layout is strong enough for our application. We do not need to design any additional locking system.

4.4 Rope analysis

In order to keep the β angle constant at 90° (figure 3.2), we thought we would block the leg in extension with a rope. This rope must be fixed at the ankle of the blocked leg and at the level of the corset (figure 4.16). Moreover, the system must be easy use so that this one must be simple of use so that the experimenter can put and withdraw the device alone. Therefore, we are going to use a slacker with a blocking type slide to adjust the length of the cord and the tightening at the level of the ankle. Moreover, in order to avoid injuries due to the tightening and the friction at the level of the ankle, we are going to sew a protective foam at the level of the section in contact with the ankle (see figure 4.17).



Figure 4.16: Blue dial on the corset represents the attachment point on one side of the rope.



Figure 4.17: Rope with zip lock and foam protection at the ankle.

From a dimensional point of view, this rope must be at least 380 mm long (figure 4.18 and table 3.4) (this dimension is determined by a trigonometric calculation).

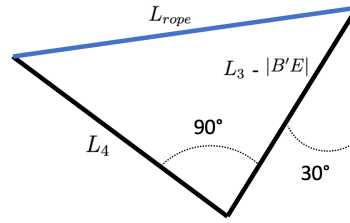


Figure 4.18: Representative diagram of the geometry of the leg blocked (black) by a string (blue).

4.5 Assembling the device

In the previous subsections, we have described all the components of the prototype. In this subsection, we will list the different stages to assemble in chronological order:

- 1: Attachment of the "L" shaped laminated plate with the rigid support from the orthosis;
- 2: Fixing of the laminated plate with the Bosch profile;
- 3: Fixing the slide with articulated ball at knee left at the level of corset;
- 4: Fixing the slide with articulated ball at knee left at the level of rigid support;
- 5: Change the tube representing the femur of the prosthesis so that the prosthesis is the same size as the experimenter's leg;
- 6: Attach the denture to the Bosch profile. Note: the profile and the denture should not be aligned, the profile should face away from the denture;
- 7: Fitting of protective foam and clamping loops;
- 8: Modifying the heights of the different elements at the Bosch profile and the slide so that all the joints of the prosthesis are at the right level with respect to the positions of the similar biological joints. To adjust the height of the laminated plate on the profile so that the knee and the face of the locked thigh are in good contact with the rigid support. To adjust the height of the corset with the slide.

The different stages of the assembly are shown in Figure 4.19. The number of each step is indicated above the image representing the step.



Figure 4.19: Steps constituting the assembly of the prototype. Each step is numbered above figure.

Figure 4.20 represents the assembled device worn by the experimenter.



Figure 4.20: Assembled prototype worn by the experimenter.

4.6 Conclusion

In this section, we have completed the design of the prototype design.

First, we noticed that the attachment point between the prosthesis and the adapter had a similar position to the attachment point between the hip joint and the pelvic socket. This was the most important criterion during the design stage.

Secondly, thanks to the use of the Bosch-type profile and the slide, we adjusted the device in such a way that it respected the body morphology of the experimenters without having to make permanent modifications to the prototype. The only element we could not modify was the dimensions could not the height of the Canadian prosthesis.

Thirdly, the hip joint present in the prosthesis that we was proposing were did not allow the realization of the extension of the leg. Therefore, the leg, which should normally be blocked at an angle of 30° (see figure 1) in order to avoid any risk of contact between the prosthesis and the adapter, was blocked at an angle of 15° chosen so that the trunk was aligned with the legs in a static standing position.

Chapter 5

Validation of the reliability of prototype during a gait cycle

In this section, the objective is to conduct the reliability study of the adapter designed to run pre-tests on a Canadian prosthesis for non-amputees.

In order to carry out our experiments, we have recruited a healthy subject who will take the reliability tests of the adapter. The subject is a right-handed 22-year old woman. She is 160cm tall and weighs 63 kg.

This section is organized in 4 sections:

- Methodology;
- Experimental results;
- Discussion of the results;
- Conclusion.

5.1 Methodology

The methodology section is divided into 4 parts.

First, we list the necessary equipment to perform the experimental part. Then we explain the experimental protocol followed by the measurement and data acquisition. Finally, we list the different steps of the data analysis.

5.1.1 Equipment

Firstly, the user will wear the adapter with a Canadian prosthesis to simulate the amputation of the left leg (see Figure 4.20).

To do this, he will rope the fully relaxed slack around his left foot and will ensure that the ball-and-socket connection between the brace and the rigid support is secured with 15° wedges. The user will then loosely put on the corset so that the experimenter has enough mobility to insert their left thigh into the rigid support and secure this part with the clamping device. Finally, the user will tighten the corset and will pull on the rope at his left foot to achieve a 90° angle between his thigh and calf.

The device has been designed to adapt to various user morphologies. Therefore, at the level of adjustments, there is the possibility of modifying the height of the slide linking the corset to

the rigid support, as well as the height of the attachment point between the prosthetic hip and the square section profile and the height of the laminated plate on this same profile.

Secondly, the user will need a crutch during the training phase.

Thirdly, during the training and experimentation phase, we will use a treadmill capable of being used at a speed of $0.5 \frac{m}{s}$.

Finally, the experimental data will be collected via XSENS^{MT} motion sensors (Movella) based on inertial measurement units (IMU). The layout of the sensors is available in the following link: Sensor-Tutorial.

5.1.2 Experimental protocol

Our goal in this experimental part is to (in)validate the reliability of the adapter. To do this, we will conduct an experiment and compare it to another experiment (in terms of shape, amplitude, position of extreme values, etc.) with an experiment that has been performed in the literature by an amputee. Our reference data comes from the article: "Design and optimisation of a hip disarticulation prosthesis using the remote centre of motion mechanism". [12].

This study shows the evolution of the angle of the hip, knee and pelvis as a function of the gait cycle. Three sets of data are available (see Figure 5.1), but only two of them will be used. Firstly, we will use the data relating to the amputated leg (curve 7E7) and secondly the data relating to the non-amputated leg (intact side) but which is impacted by the use of the prosthesis.

We chose this reference because this study is also carried out with a passive Canadian prosthesis (angle profile is different when using an active prosthesis). This prosthesis is very similar to the prosthesis we built (identical hip joint 7E7, similar knee joint and similar foot joint).

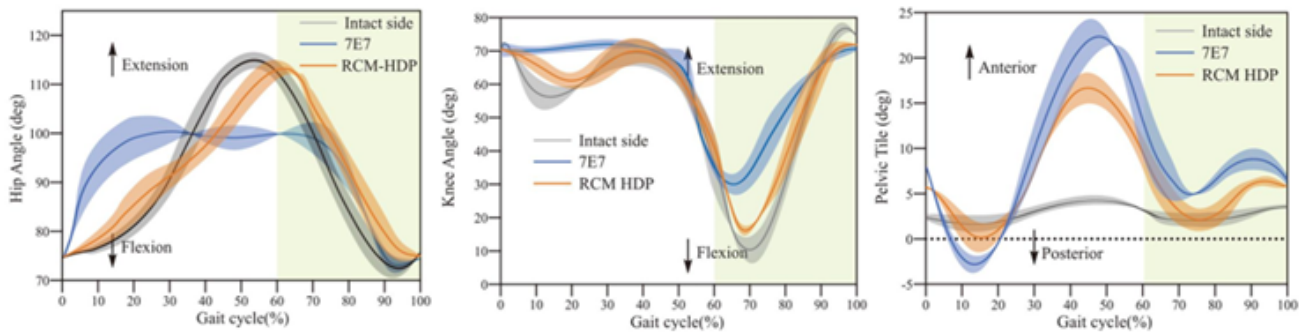


Figure 5.1: Graph representing the evolution of the hip, knee and pelvic angle as a function of gait cycle by an amputee wearing a Canadian prosthesis with an ootobock 7E7 hip joint, with a RCM-HDP hip joint and with measurements made on the amputee's valid leg.[12]

In order to obtain the most reliable data possible, we will carry out an experimental protocol similar to the one described in the article.

But before acquiring the data, the experimenter will first get a training period. The aim of this for the experimenter to get used to experimenter used to walking "naturally" with the device in order to try to remove the adaptation phase with the device form the evaluation. The training period consists of 7 days where he will wear the device for 1 hour 2x/day. During the first 5 days, the user will walk on crutches without the use of a treadmill to avoid any risk of falling. Then,

during the remaining 2 days, the experimenter will train on the treadmill by progressively increasing the walking speed until reaching a speed of $0.5 \frac{m}{s}$ on a zero slope. This speed is imposed on us by the article. However, figure 1.2 in section 1 showed that the natural speed of a person suffering from a hip disarticulation is $40 \frac{m}{min}$ which represents $0.6 \frac{m}{s}$ which is a value close to $0.5 \frac{m}{s}$.

Then, we are going to make the measurements. First of all, we are going to carry out control data in order to see if the sensors work well and to identify the differences between the real evolution given by the sensors and the theoretical curves found in scientific articles. To do this we will ask the user to walk on the treadmill without the prosthesis adaptor device at a speed of $0.5 \frac{m}{s}$ for 2 minutes (see video available in this link: normal walking). Then, after having installed the device and the XSENS^{MT} sensors on the experimenter (cf. section 5.1), we will ask the user to walk on a carpet at a speed of $0.5 \frac{m}{s}$ for 2 minutes in order to have a minimum of 10 walking cycles in the sample data (cf. video available in this link: walking with the device).

5.1.3 Measurement and data acquisition

To perform the measurements, we will use 8 XSENS^{MT} sensors: 1 on each foot, 1 on each inner side of the tibia, 1 on each outer side of the femur, 1 on the sternum and 1 on the sacrum. These sensors are connected to an application of the Robotran platform (a program originally from the MBsysWeb application which was then modified and adjusted in a Robotran program). This application shows the evolution of the body in real time as well as the evolution of the angles in an external terminal of the computer (see figure 5.2). This simulation makes it possible to observe the relevance of the values given in real time by the sensors.



Figure 5.2: Visualisation of the simulation in the MBsysWeb platform interface measured via the XSENS^{MT} sensors.[38]

Once the simulations (results) are finished, the data are directly transcribed into results files in a previously encoded folder. As this file is not visual, the last step is to transform this result file (bag file) into a txt file in order to analyse the recorded signals.

Appendix 4 explains the use, acquisition and transformation of the data provided by the sensors in more detail.

5.1.4 Data analysis

The data analysis is composed of 6 steps: opening of the txt file, visualization of the signals, identification of the beginnings and ends of the gait cycle, resampling of 10 periods, averaging of these 10 periods and comparison on the same graph of the resampled and averaged sample with the data of the article [12].

First of all, the opening of the two txt files is done with a python program. This step makes it possible to visualize and rework the different signals.

Secondly, the visualization of the signals will be represented. The sensors allow the acquisition of 47 signals every 0.02s. Those signals are represented as columns in the txt file. Among these we find the evolution of the flexion hip angle left and right, flexion knee angle left and right and the pitch torso angle as a function of time. The first step of the visualization is to transform these signals in degrees. Then, in order to average our signals over 10 periods, we want to display at least 10 periods. To do this, we will iterate our function for different starting and end times until we get the desired display. For example, in figure 5.3, the right knee angle signal for normal walking (without adapter) for a selected starting time of 70s and an end time of 102.2s and the left knee angle signal for walking with prosthesis for a selected starting time of 65.6s and an end time of 101.4s.

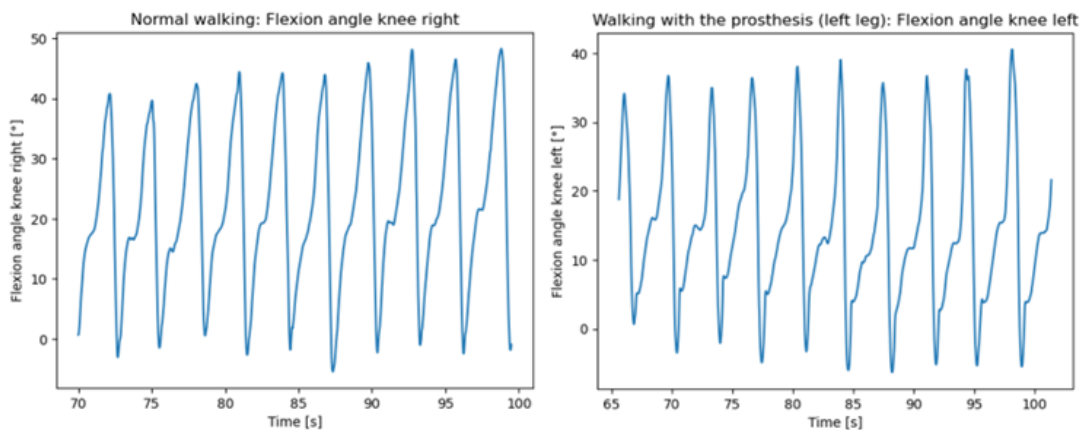


Figure 5.3: Display of the 10 gait cycles in the case of normal walking on the left and walking with prosthesis on the right.

Thirdly, we need to identify the walking cycle in these two samples. To do this, we will refer to the evolution of one of the signals we would like to display in the literature.

Concerning the example of the knee angle signal. Figure 5.4 represents this evolution during a cycle.

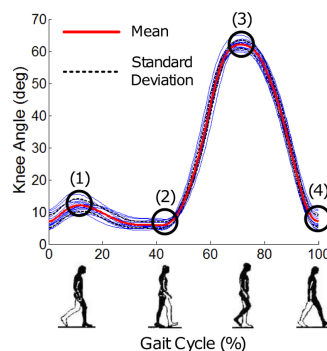


Figure 5.4: Evolution of the knee angle during a gait cycle.[39]

Figure 5.4 points out two main characteristics of the signal: a gait cycle begins with a minimum (local in figure 5.4 but global in our results) at approximately 0 % of the cycle. Towards 75 % of the cycle, we have the presence of a global maximum. Thanks to these characteristics, we will be able to identify each beginning of cycle for each period by imposing the beginning of the cycle as being the minimum of the signal.

It is important that all the signals coming from the same sample have their cycle start at the same time of the experiment in order to visualize the different phases of the cycle.

Fourth, we resample each cycle between 0 and 100% to represent a walking gait cycle'.

Fifthly, we have averaged the 10 periods to show just one average cycle.

Them, we display the reference data from the articles on the same graph [12], [40] and [41] available in figure 5.1 and in figure 5.5.

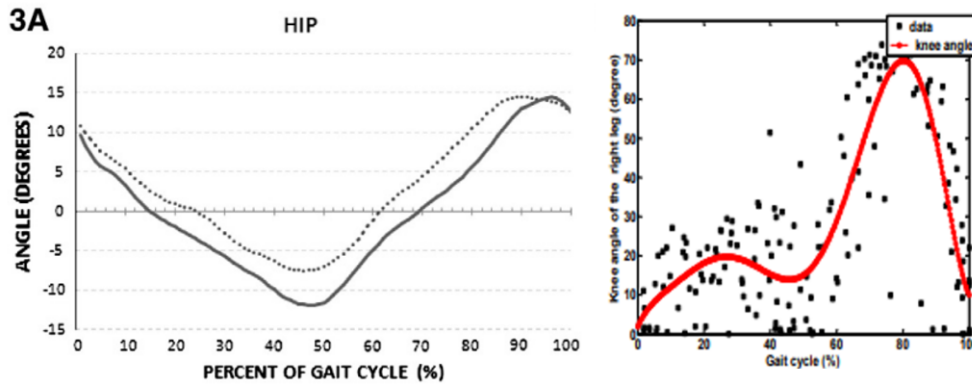


Figure 5.5: Graph showing the evolution of the knee and hip angle during a gait cycle in the case of control data (normal walking without disturbance).[40] [41]

A closer look at the links between the different signals shows that:

- The signal variation of the right hip angle will be compared with the signal variation of hip angle in figure [40];
- The signal variation of the right knee flexion angle will be compared with the signal variation of the knee flexion angle in figure [40].

These two comparisons are represented in figure 5.6.

Regarding the data of the walking with prosthesis file (the comparison will be made against the graphs 7E7 in figure 5.1):

- The experimental signal variation of the left hip angle signal will be compared with the signal variation of the hip angle signal;
- The experimental signal variation of the left knee angle signal will be compared with the signal variation of the knee angle signal;
- The experimental signal variation of the right hip angle signal will be compared with the signal variation of the intact hip angle side signal;
- The experimental signal variation of the right knee angle signal will be compared with the signal variation of the knee angle intact signal;
- The experimental signal variation of the pitch torso angle signal will be compared with the signal variation of the pelvic tilt signal.

These five comparisons are represented in figure 5.7 for the first two subtitles, in figure 5.9 for the two subtitles that follow and in figure 5.8 for the last subtitles.

5.1.5 Results

First, we analyzed the signals from the control data, i.e. the normal walk and identified 2 comparisons that can be seen in figure 5.6.

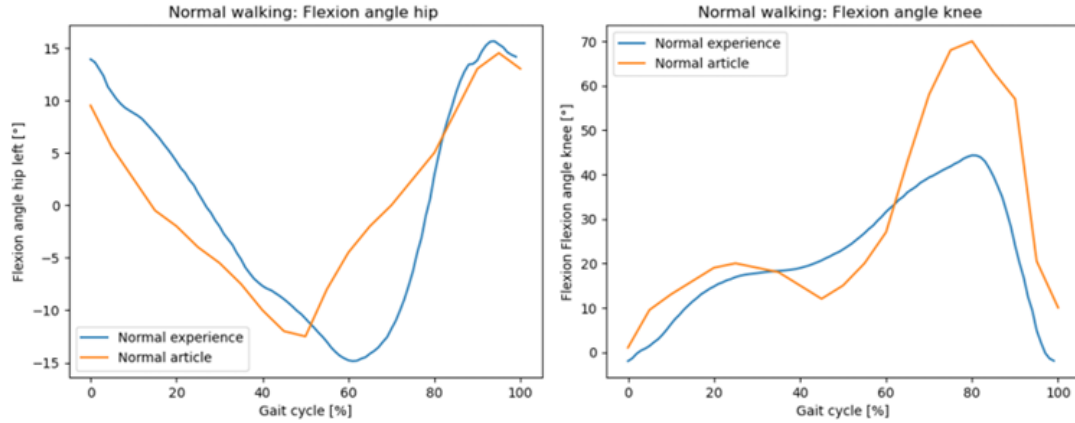


Figure 5.6: Comparison of the results for the normal walking hip and knee angle. The experimental data is represented by the blue curve and the data from the article is represented by the orange curve.[40]

Table 5.1 shows the values of the extreme values and amplitude of these 4 signals.

	Max	Min	Amplitude
Experiment hip angle	(94%,15°)	(61%,-14°)	29°
Article hip angle	(95%,14.5°)	(50%,-12.5°)	27°
Experiment knee angle	(80%,44°)	(0%,-2°)	53°
Article knee angle	(80%,70°)	(0%,1°)	69°

Table 5.1: Table showing all the values in amplitude and position of the extremities of the signals received for the control data.

Then we focused on the signals from the walking with prosthesis sample. In the same way as with the control data, table 5.2 represents the coordinates and amplitudes of the extremes of the different signals.

	Max	Min	Amplitude
Experiment hip angle prosthesis	(94%,35°)	(60%,-13°)	48°
Article hip angle prosthesis	(95%,34°)	(30%,6.5°)	27.5°
Experiment knee angle prosthesis	(80%,38°)	(0%,2°)	36°
Article knee angle prosthesis	(67%,40°)	(30%,-2°)	42°
Experiment hip angle intact side	(62%,31.5°)	(92%,2.5°)	29°
Article hip angle intact side	(55%,44.5°)	(95%,0.5°)	44°
Experiment knee angle intact side	(55%,53.5°)	(35%,0°)	53.5°
Article knee angle intact side	(70%,60°)	(95%,-8°)	68°
Experiment pelvic tilt angle	(57%,34.5°)	(91%,-14.5°)	49°
Article pelvic tilt angle	(48.5%,22.5°)	(12.5%,-3°)	25.5°

Table 5.2: Table showing all the values in amplitude and position of the extremities of the signals received for the experimental data with adapter.

Figure 5.7 and 5.8 represent the signals collected from the leg on the prosthesis side.

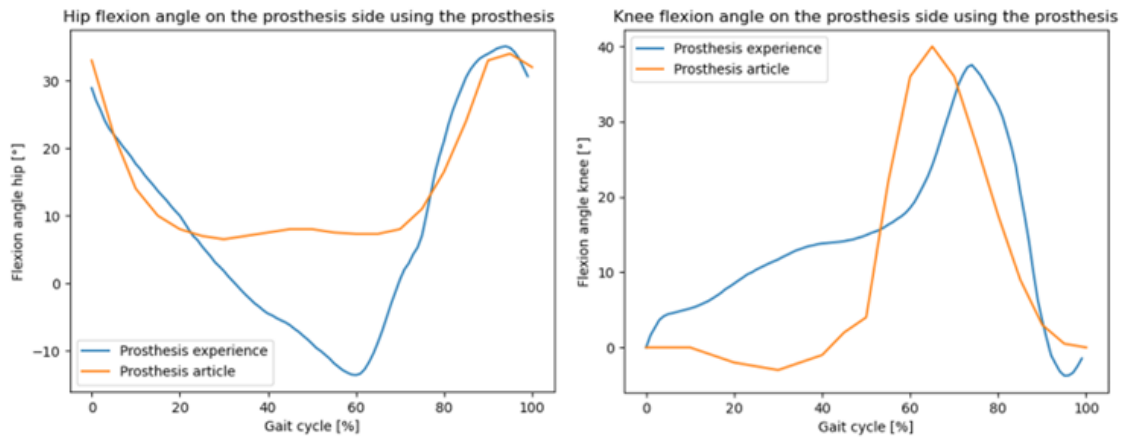


Figure 5.7: Comparison of the results for walking with prosthesis for the hip and knee flexion angle signals on the prosthesis side. The experimental data is represented by the blue curve and data from the article is represented by the orange curve.[12]

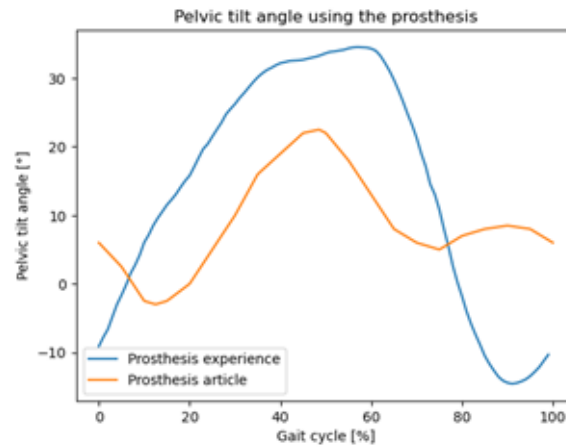


Figure 5.8: Comparison of the results for walking with prosthesis for the pelvic tilt angle signal by the prosthesis side. The experimental data is represented by the blue curve and data from the article is represented by the orange curve.[12]

Finally, figure 5.9 represents the signals received at the leg on the opposite side of the prosthesis.

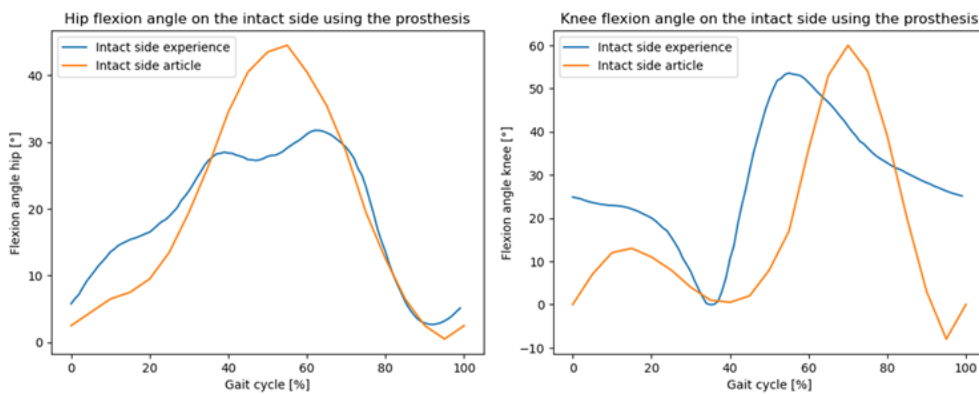


Figure 5.9: Comparison of the results for walking with prosthesis for the hip and knee angle flexion in the intact side. The experimental data is represented by the blue curve and data from the article is represented by the orange curve.[12]

The leg intact side will be normalized on the gait cycle of the prosthesis.

In order to improve the understanding of these signals, figure 5.10 represents the evolution of the walking cycle of the experimenter wearing the device. The percentage below each image represents a moment in the cycle in relation to figure 5.11.

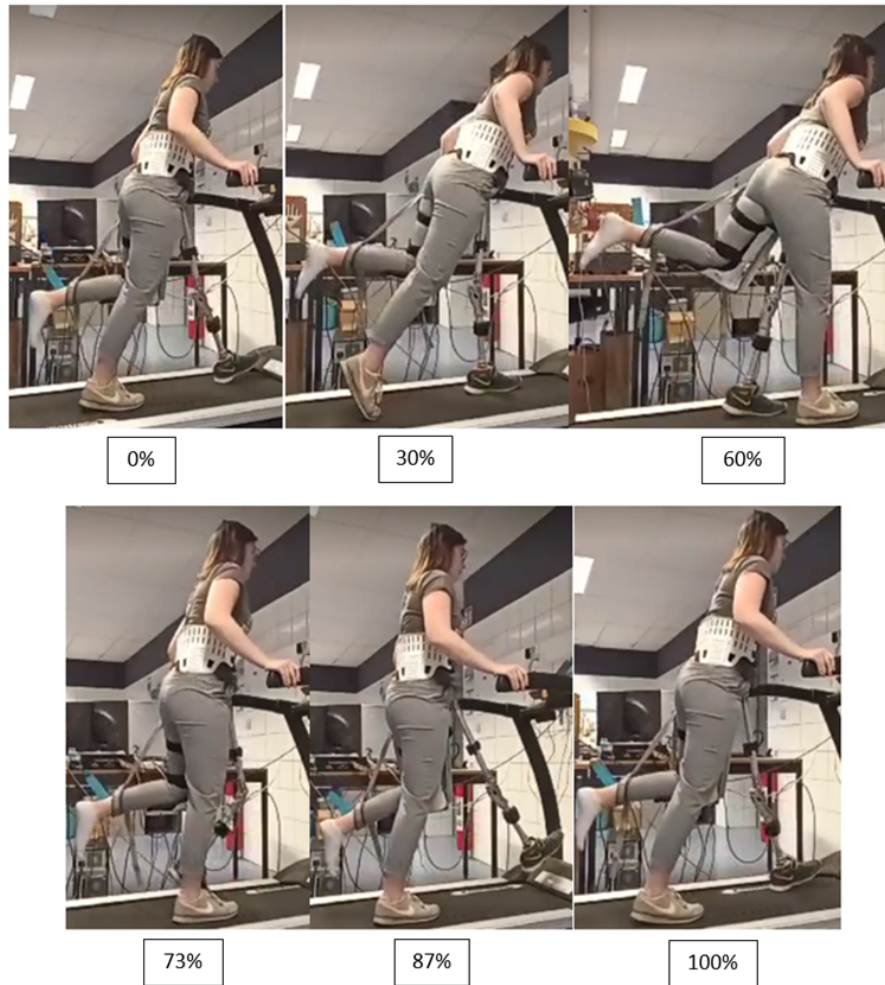


Figure 5.10: Evolution in images of the gait cycle performed by the experimenter wearing the prosthesis adaptor device.

5.2 Discussion

In the previous subsection, we displayed the measured signals processed. In this section, we will focus on the reliability of our device. To do this we will analyse the shape (overall shape of the signal), the amplitude, the position of the extremities and asymmetry of the different signals.

As reminder, figure 5.11 shows the different phases, tasks, periods and events of the cycle of the walk.

First, we focused on the signals from the control data, i.e. the normal walk. In the case of the control data, we have identified 2 comparisons which can be seen in figure 5.6.

Both measurements (control data and walking with prosthesis) were carried out using the same experimental protocol. However, if variations appear in the control data, they can be found in the signals obtained in the prosthesis experiment. This can be justified as an anomaly in the

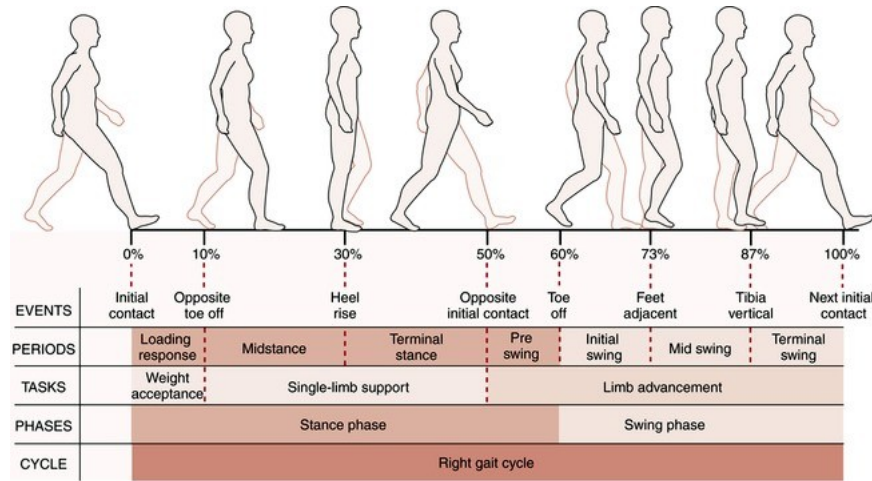


Figure 5.11: Detailed walk cycle showing the different phases, tasks, periods and events that will be referred to in the discussion related to the experiment.[42]

recording system.

Regarding the variation of the angle of hip flexion, we note that the two signals have the same shape (Figure 5.6 on the left). If we consider the article signal as the reference signal, the experimental signal has the same number of extreme values, but the amplitude of the movement varies to some degrees (2°) in favour of the experimental data. Equally, the reference minimum is 11% ahead (before) of the experimental minimum. As a result, the minimum hip angle appears for the reference at the opposite initial contact event and for the experiment at the toe off event.

Regarding the variation of the angle of knee flexion, we notice that the two signals follow the same pattern (Figure 5.6 on the right). However, we notice that the local minimum in the experimental signal does not appear. In terms of the positioning of the extreme values, we notice that the experimental maximum has a value 0.62x lower than the reference maximum. This could be explain by the fact that the reference was performed at a comfortable walking speed whereas the experimental data was performed at a low speed that forces the user to take shorter steps and thus decrease the knee angle.

In general, we note that the measured data does not exactly match the data from the literature. The three main variations are the variation in amplitude and the non-appearance of the second minimum at the knee angle and the shift of the minimum at the hip angle. This may be due to the gait style of the experimenter and the results provided by the sensors. Normally, the experimental and data from the article should be similar as there should not be any disturbance between the two data sets.

Secondly, we focus on the signals from the walking with prosthesis sample. In the same way as with the control data, table 5.2 represents the coordinates and amplitudes of the extremes of the different signals.

The signals collected from the leg on the prosthesis side are presented in figure 5.7 and 5.8.

If we first look at the evolution of the hip flexion angle (figure 5.7 on the left), we notice that the curves are similar. Indeed, if we consider the curve from the article as the reference curve, we notice that the curvature of the maximum follows the curvature of the maximum of the experiment and is located between the vertical tibia event and the initial contact. However, the

curvature of the reference minimum is represented as a saturation extending from approximately 20% of the cycle (between the opposite toe off and heel rise) to approximately 70% of the cycle against a peak at 60% (toe off) for the experimental maximum. The peak and saturation can be explain by the fact that saying that the movement of the prosthesis (explained in section 1.2.2) is largely governed by a movement of the pelvis and torso which will propel the prosthesis forward. In the experiment, a peak appears because the experimenter gives a large pelvic and torso thrust to propel the prosthesis forward (figure 5.10 picture 60, 73 and 87%) while the amputee tends to propel the prosthesis by a less brutal and softer movement which creates the saturation. The momentum given by the experimenter also increases the amplitude of the signal relative to the reference signal.

If we look at the evolution of the knee flexion angle (figure 5.7 on the right), we notice that the curves are similar. We notice that the experimental signal does not reach a minimum in the middle of the stance phase. However, this was also observed in the case of the control data, which makes this asymmetry "irrelevant". This lack of minimum might not be due to the low running speed of the data control).

From the point of view of the other extremes, the maxima are comparable in amplitude. However, the reference signal (between the toe off and the adjacent feet) is 13% of the cycle ahead of the experimental maximum (between the adjacent feet and the vertical shin). However, the maximum is of the experimental signal is at the same level in the gait cycle as the control signals.

In order to understand what follows on the evolution of the pelvic tilt angle, figure 5.12 gives an anatomical representation of analyzed movement.

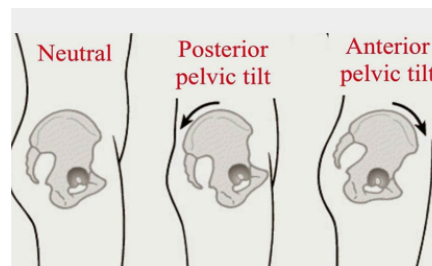


Figure 5.12: Explanation of the pelvic tilt movement.

If we look at the evolution of the pelvic tilt angle (figure 5.8 on the right), we notice that the curves look quite different. In the case of a "normal" cycle (without the use of a prosthesis), the amplitude of the signal is almost zero. However, the use of a Canadian prosthesis increases the amplitude (25.5°) due to the movement of the torso and the pelvis (almost non-existent during a "normal" cycle).

The curvature of the experimental signal can be explained in the same way as the evolution of the hip angle. Indeed, the user gives a momentum coming from the pelvis and the torso in order to move the prosthesis forward, which produces a peak at the level of the minimum of the hip flexion angle but a saturation at the level of the pelvic tilt angle (see figure 5.10). The saturation can be explained in terms of the time lag between the propagation of the signal from the upper limb to the lower limb.

Furthermore, the minimum point for hip flexion angle (figure 5.7 left) and the maximum point for pelvic tilt angle (figure 5.8) occur in the gait cycle at the same time as the toe off event. In the case of the experiment with the adapter, we notice that the amplitude is almost two

times (49° compared with 25.5°) higher than the amplitude of a cycle performed by an amputee because the propulsion movement is very important. We can see this visually in the video available in the following link: [walking with the device](#) and of the images available in figure 5.10. Moreover, the number of extreme values between the reference signal and the experimental signal is different. The reference signal gives 2 minima and 2 maxima whereas the experimental signal only gives one of each.

Finally, on the signals received at the leg on the opposite side of the prosthesis (figure 5.9). In the case of the experiment with the adapter, we analyzed the signals that the amplitude was almost two times higher than the amplitude of a cycle performed by an amputee because the propulsion movement is very important. We can see this visually in the video available in the following link: [walking with the device](#) and on figure 5.10. Moreover, the number of extreme values between the reference signal and the experimental signal is different. For the reference signal, we found two minima and two maxima whereas we could only find one of each in the experimental signal analysis.

If we look at the evolution of the hip flexion angle (figure 5.9 on the left), we notice that the signals are similar and centred in relation to each other. However, the experimental maximum is in the form of a saturation which is not the case for the maximum of the article which is in the form of a peak as for the control conditions. The saturation starts at 30% and ends at 70% of the cycle. During that time the leg on the opposite side goes. During this period of time the leg on the opposite side goes through the vertical tibia event, initial contact, opposite toe off and heel rise (figure 5.10 images 73 and 87%).

Regarding the minimum, they are similar in amplitude and position in the cycle.

Finally, if we look at the evolution of the knee flexion angle (figure 5.9 on the right), we notice that the signals look a little different. The experimental signal is 15% of a cycle ahead (before) of the article signal, considering that the local minimum of the article signal is the image of the global minimum of the experiment. This could be explained by an anticipation of the user to perform the movement.

We can also notice that the knee angle does not pass through two minima (one minimum towards 0% and one around -40 -50%). This may be due to the fact that the leg on the prosthesis side does not follow a "continuous" movement with respect to a normal gait cycle. Indeed, during the rolling of the prosthetic hip joint, the knee of the opposite side remains in extension for a longer time in order to stabilize the initial contact of the knee of the prosthesis to avoid its flexion during the transfer of the body weight on the prosthesis (figure 5.10 image 87, 100, 0 and 30%). This event can be the cause of the pressure saturation in the signal at approximately 25° .

5.3 Conclusion

In conclusion, in this section we conducted performed tests on the prosthetic adapter device. These tests have shown positive results and encourage the use of the adaptor for further testing on new Canadian prostheses in the laboratory.

However, the evolution of the joint angles does not follow the signals from the literature made by an amputee wearing a prosthesis similar to the prosthesis used for the experimental part. This is largely due to the fact that the experimenter performing the gait cycle is not used to the device, which explains his less smooth and natural movements. In addition, the use of a lateral upper body support at the safety handles of the treadmill also distorted the results. An

improvement to the experimental procedure would be to use the device without a crutch or handle on the treadmill. This would mean that the training period would have to be longer.

A disadvantage of the use of the adaptor is the disproportionate forward tilt of the trunk when using the adapter. In fact, during the experiment, we noticed that the user's trunk was strongly tilted forward due to the lifting of the locked leg (on the side of the device) towards the back. This tilt was already present when using the device in the static position (see Figure 4.20 left) but is even more obvious in the experimental video presented via a link at the end of section 5.2. Moreover, a discomfort linked to the dynamics of the blocked leg (unnatural posture) could also influence the results.

Finally, if we compare our results to the results obtained by the other two types of adapters available (see Figure 1.12 and 1.13) we can identify similarities and differences.

For the adapter in Figure 1.12, we do not have experimental data from a gait cycle experiment. However, experiments have been conducted for the adapter in figure 1.13. This device was also used to analyse the gait cycle of a person performing pre-tests on a passive and active Canadian prosthesis. In order to conduct these tests, the user also carried out a training phase of 1 month. This duration is much longer than the duration of our experiments (1 week). However, despite the longer training period, the use of their adapter also causes the body to lean forward when walking.

From a graphical point of view (figure 5.13), the signals compared to our measured signals, are the yellow reference signal and the red signal representing the signal of the passive prosthesis. We notice that with the adapter in figure 1.13, the step length is much shorter, which strongly decreases the amplitude of the angles compared to the reference. This is due to the fact that the momentum of the pelvis and the torso in their configuration is very limited due to the geometry imposed on the body. This is not the case with our device.

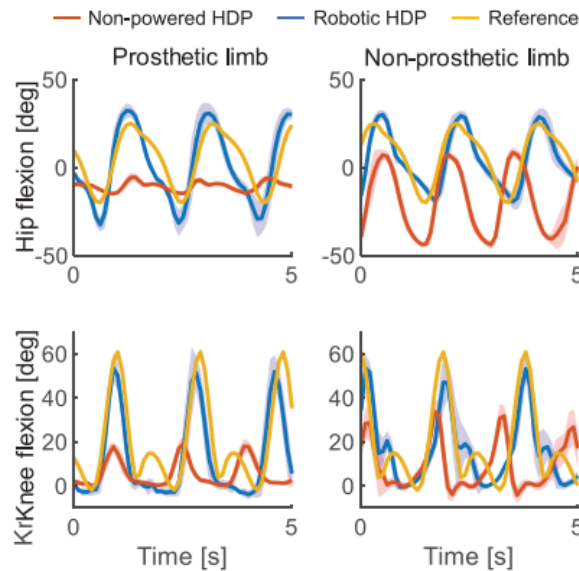


Figure 5.13: Graphical results obtained with the adapter of figure 1.13.[14]

Chapter 6

Conclusion

To conclude this master thesis, some points have to be reminded:

"Bionic prosthesis for the lower limb - Adapter for testing a complete leg prosthesis".

To start this thesis, we began by presenting the state of the art relating to adapters for testing a complete leg prosthesis. Chapter 1 gave the reader a better understanding of the different causes of amputation from the hip and on the Canadian prosthesis devices already available on the market (their function, points of attachment, impact on the gait cycle, energy expenditure, problems, etc.). Different prototype adapters existing for all types of amputation were also presented. Among these adapters, we discovered 2 devices already used in research to mimic the pathology of hip disarticulation.

Then we realized a complete design study in chapter 2 without taking into account the two existing adapters. To do this, we defined several important points that had to be respected by the device. Among these, we found the respect of the location of the attachment point between the prosthesis and the corset with respect to the attachment point between the prosthesis and the adapter. Following the realization of the functional specifications, taking into account the graph of goals and the specifications of the device. The realization of the function graph and the morphological table, we were able to propose 5 possibilities of concepts completely different from each other. Then, thanks to a decision matrix, we chose to keep two concepts that both met the goals of our device.

Subsequently, we conducted a study to identify the distribution of loads on the experimenter's body and on the device in chapter 3. Thanks to this study, we were able to distinguish the two concepts by keeping the one with the least forces and moments at the joints.

At this stage, the theoretical conceptualisation study was completed. The next logical step was the design, embodiment and assembly of the different elements of the prototype (chapter 4).

Once the prototype had been created, we carried out the experimental part of this thesis (chapter 5). The aim was to validate the reliability of the device in relation to the data available in the literature. This validation step allowed us to determine if it was possible to rely on the data collected by the adapter during the test on a non-amputee user. In the experimental section we were able to prove that the measured signals were encouraging for the realisation of the pre-test. Indeed, the signals measured were similar to the signals coming from the literature and in the case of unevenness, they could be easily interpreted thanks to the image and video showing the evolution of a gait cycle recorded during the experiment.

Regarding the ergonomics of the device, it is comfortable to wear, there are no injuries after use, it can be put on and taken off easily without external help and in less than 2 minutes. Moreover, the point of attachment between the prosthesis and the adapter respects the point of attachment between the prosthesis and the pelvic socket and there is no obstruction (no contact) during the gait cycle between the prosthesis-adapter and the user's leg. In conclusion, the elements identified in the graph of objectives, table of specification and in the functional graph presented in chapter 2 were all respected.

Regarding the improvements of the device and of this master thesis, the main thing to improve would be the experimental protocol. Indeed, the realization of the walking cycle remained rather jerky and not natural enough. An improvement would be to carry out a slightly longer training period and tests with several experimenters.

Also, the experiments were performed with the user supporting themselves on the treadmill's safety handle, which can influence the results. An improvement would therefore be to use the device with external support (no crutch...) which would probably allow us to obtain more results.

With regard to the design of the device, we noted that the user, even in a static position, had a forward leaning trunk. An improvement would be to limit this inclination (this remark was also made when studying the device already available on the market in figure 1.13).

Also, when performing the gait cycles, we believe that when propelling the leg to realize the swing phase, the user uses the momentum of their blocked leg which is not achievable by an amputee. Therefore, an improvement would be to limit this swing in favour of the torso and pelvis swing.

Finally, to further test the reliability of the adapter, it should be tested with another prosthesis and the data obtained should be compared with a cycle performed with the same prosthesis by an amputee. This second study would allow us to check if the variations in signals between an able-bodied person and an amputee are similar to the variations obtained in the experiment with the "7E7" prosthesis.

From a more personal point of view, this master thesis has enabled me to become more autonomous and less dependent on a group, as is the case in many of the projects on our courses. I am also pleased with my choice of subject, because I have remained motivated and passionate about the world of prostheses from the beginning to the end of this thesis. In fact, if I take a look back at my school career, I began with a Bachelor in mechanical engineering at the University of Mons. I have always had a passion for mechanics, but I couldn't see myself working on industrial machines. This is why I decided to register at another university for my master's degree to study mechanics dedicated to people. This master thesis developed my passion for external prostheses. This thesis enabled me to finish my studies on a high note and to begin my professional life. The next step I'd like to take is to work in humanitarian aid to help patients living in underdeveloped regions and countries to acquire prostheses made from locally available materials and technologies.

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6.1 Appendix

6.1.1 Appendix 1



Single-axis foot with toes

Reference number 1H38

In combination with the single-axis joint, the 1H38 single-axis foot with 10 mm heel height allows the user to achieve a secure stance quickly. It is only suitable for transfemoral prostheses.

Key features

- Natural shape with smooth surface and defined toes



Max. 100 kg

Information material

647G45=ALL_INT IFU 1H38 1H40

Scope of delivery

1H38	Single-axis foot with toes	1	Piece
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Technical data

Mobility grade	1
Max. body weight	100 kg
Side	left (L), right (R)
Sizes	21-28 cm
Weight without adapter*	ca. 365 g
Foot shape	Normal shape for a heel height of 10 +/- 5 mm
Colour	beige
System height with 2R51, 2R33/2R10*	50 mm, 46 mm
Build height with 2R51, 2R33/2R10*	68 mm, 64 mm

* Technical data refer to the size of 26 cm

• For the 1H38 in size 21 cm, please use the available options for size 22 cm.

Order example

Reference number	Side	Size
1H38	L	26

Figure 6.1: Single-axis foot with toes as part of the Canadian prosthesis. [43]



Single-axis foot adapter with screw connection

Reference number 2R10

The 2R10 single-axis foot adapter made of steel connects a single-axis prosthetic foot to the distal connector of a modular prosthesis.



Max. 100 kg

Information material

647G1630=ALL_INT IFU 2R10 2R33 2R51

Technical data

Article number	Size	Max. body weight	Weight
2R10=22-25	22 - 25 cm	100 kg	325 g
2R10=26-30	26 - 30 cm	100 kg	340 g

Figure 6.2: Single-axis foot adapter with screw connection as part of the Canadian prosthesis. [43]



Tube adapter

Reference number 2R50=AL/2R49=AL

The 2R50=AL and 2R49=AL tube adapters differ in length. They connect prosthetic components to each other. Adapter combinations allow for controlled angle and translational adaptation in the sagittal and frontal plane as well as adjustment of inward and outward rotation.



Max. 136 kg

Information material

647G902=ALL_INT IFU Tube adapters

Technical data

Article number	Diameter	Material	Min. system height	Max. system height	Min. build height	Overall length	Weight	Max. body weight
2R50=AL	30 mm	Aluminium	97 mm	232 mm	53 mm	214 mm	155 g	136 kg
2R49=AL	30 mm	Aluminium	97 mm	472 mm	53 mm	414 mm	255 g	136 kg

• For high loads on transtibial prostheses, a tube adapter with Ø 34 mm should be used (e.g. 2R57/2R76).

Figure 6.3: Adapter tube for the Canadian prosthesis. [43]



Knee joint, polycentric, with pneumatic swing phase control

Reference number 3R106

With the 3R106 polycentric knee joint, the stance phase is secured due to the four-axis joint design. Powerful dual-chamber pneumatics with integrated extension assist spring produce smooth pendulum movements of the prosthetic lower leg, even at higher walking speeds. The 4G70 lamination anchor is only included in the scope of delivery with the 3R106=KD variant.



Max. 100 kg

Information material

647G1640=ALL_INT	IFU 3R106
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Scope of delivery

3R106		1	Piece
2R49	Tube adapter	1	Piece
4G70	Lamination anchor only with 3R106=KD	1	Piece
613D83	Compression spring	1	Piece
613D83	Compression spring, heavy duty	1	Piece

Technical data

Article image



	3R106	3R106=KD	3R106=ST
Article number	3R106	3R106=KD	3R106=ST
Max. body weight	100 kg	100 kg	100 kg
Mobility grade	2, 3	2, 3	2, 3
Weight	760 g	755 g	765 g
Proximal connection	Pyramid	Lamination anchor	Threaded connector
Distal connection	Tube clamp Ø 30 mm	Tube clamp Ø 30 mm	Tube clamp Ø 30 mm
Knee flexion angle	170 °	170 °	170 °
System height	162 mm	184 mm	180 mm
Proximal system height to alignment reference point	-6 mm	16 mm	12 mm
Distal system height to alignment reference point	168 mm	168 mm	168 mm
Build height	212 mm	216 mm	212 mm
Proximal build height to alignment reference point	12 mm	16 mm	12 mm
Distal build height to alignment reference point	200 mm	200 mm	200 mm

Figure 6.4: Polycentric knee joint with pneumatic control of the oscillation phase of the Canadian prosthesis. [43]



Tube clamp adapter, angled

Reference number 4R56

The 4R56 tube clamp adapter is used in prostheses in combination with a hip joint. It is available with three different angles and, among other things, connects the 7E10 Helix 3D hip joint to the 2R30 tube, and this to the 4R57 rotation adapter or a knee joint.

Technical data

Article image



	4R56	4R56=1	4R56=2
Article number	4R56	4R56=1	4R56=2
Diameter	30 mm	30 mm	30 mm
Material	Titanium	Titanium	Titanium
System height	34 mm	34 mm	35 mm
Build height	54 mm	54 mm	55 mm
Weight	85 g	85 g	100 g
Angular offset	10 °	20 °	30 °
Max. body weight	100 kg	100 kg	100 kg

Figure 6.5: Tube clamp adapter, angled component of the Canadian prosthesis. [43]



Hip joint, monocentric, with inner extension assist

Reference number 7E7

The top of the prosthetic hip joint is screwed to the lamination plate laminated into the pelvic socket, and the bottom is connected with a tube clamp. The continuously variable extension assist limits the range of motion during walking. The prosthetic joint features a low structural height which minimises pelvic tilt while sitting.



Information material		
647G130=ALL_INT	IFU 7E7	
Scope of delivery		
7E7		1 Piece
7Z53	Lamination plate	1 Piece
7Z58	Lamination dummy	1 Piece
709Z11	Bit T40	1 Piece

Technical data	
Article number	7E7
Max. body weight	100 kg
Mobility grade	2, 3
Weight	620 g
Proximal connection	Lamination plate
Distal connection	Tube Ø 30 mm
Range	140 °
Min. system height	33 mm
Max. system height	360 mm
Min. build height	60 mm

Figure 6.6: Hip joint, monocentric, with inner extension assist constitutive of the Canadian prosthesis. [43]



Lamination plate

Reference number 7Z53

The lamination plate serves as the proximal connection for the 7E7, 7E9 and 7E10 hip joints.

Technical data		
Article number	Max. body weight	Material
7Z53	100 kg	Aluminium
7Z53=1-M10	125 kg	Steel

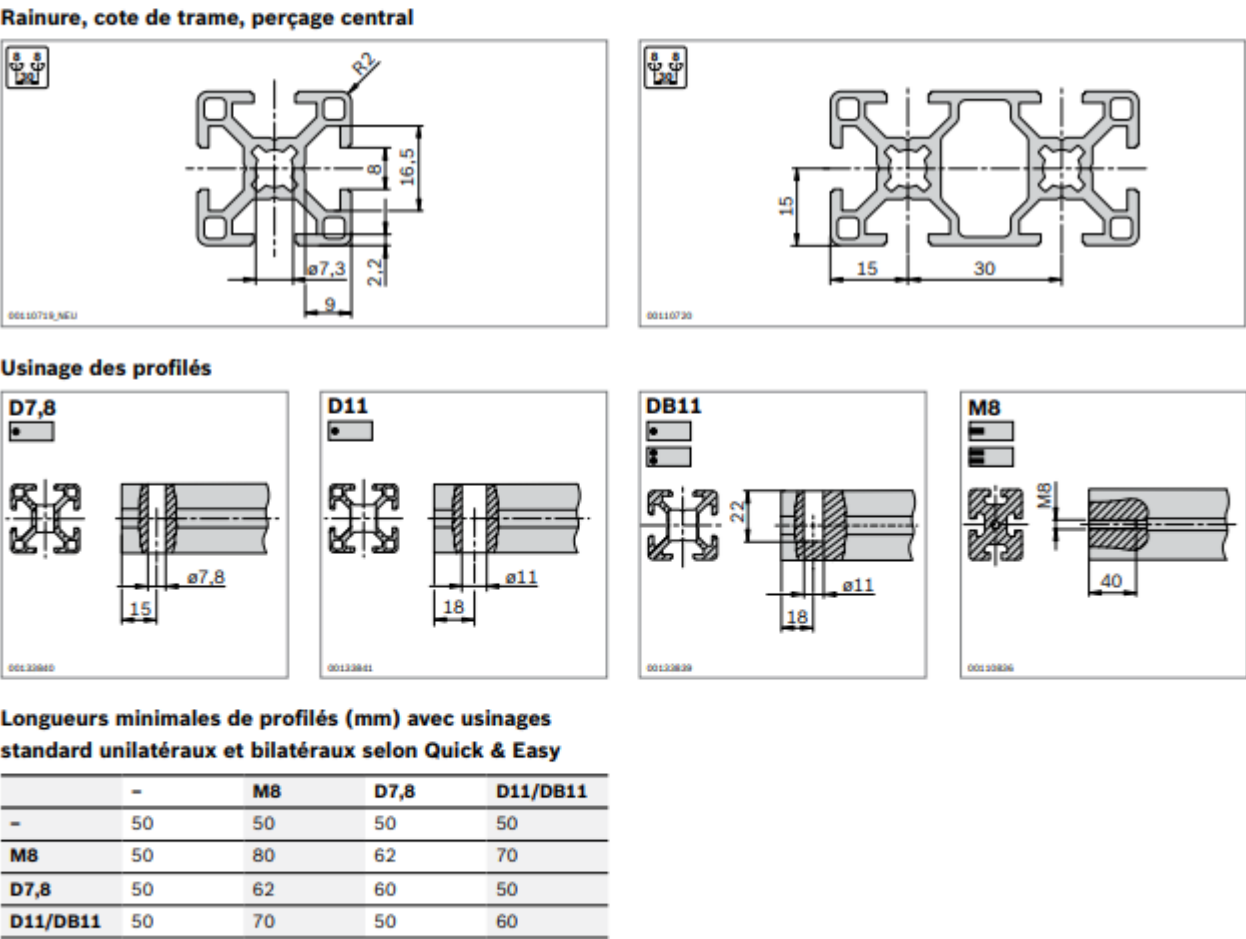
Figure 6.7: Product data sheet 7Z53 which is a flat lamination. This will reinforce the pelvic socket where it connects to the Canadian prosthesis. [43]

6.1.2 Appendix 2

Profilés d'étaiyage avec rainure de 8 mm

Profilés d'étaiyage avec rainure de 8 mm pour charges moyennes, comme les chariots, constructions légères, bâtis et cloisons de séparation.

Le profilé 30x60x60 pour le montage de vitrines et de rayonnages. Les profilés d'arrondissement permettent un montage élégant des couvercles et des capots.

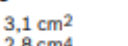


Bosch Rexroth AG, 3 842 540 393 (2017-08)

Figure 6.8: Data sheet Bosch profile with groove, grid dimension, centre hole. [44]

30x30

$A = 3,1 \text{ cm}^2$
 $I_x = 2,8 \text{ cm}^4$
 $I_y = 2,8 \text{ cm}^4$
 $W_x = 1,8 \text{ cm}^3$
 $W_y = 1,8 \text{ cm}^3$
 $m = 0,9 \text{ kg/m}$



18478

Cache	Couleur	ESD		N°
30x30 (1x)	Gris signalisation		100	3 842 548 744
30x30 (1x)	Noir		20	3 842 548 745
Matériaux :	PP			

Cache avec trou	Couleur	ESD		N°
30x30 (1x)	Gris signalisation		20	3 842 548 774
30x30 (1x)	Noir		20	3 842 548 775
Matériaux :	PP			

30x30 1N

$A = 3,7 \text{ cm}^2$
 $I_x = 3,1 \text{ cm}^4$
 $I_y = 3,5 \text{ cm}^4$
 $W_x = 2,0 \text{ cm}^3$
 $W_y = 2,3 \text{ cm}^3$
 $m = 1,0 \text{ kg/m}$



Technical drawing of the 30x30 1N L-shaped profile. The drawing shows the profile with its dimensions: total width 30, total height 30, flange thickness 3, and corner radii R3. The profile is shown in a perspective view with dimensions 30, 30, 3, and 10. The drawing also includes a small inset showing the profile's cross-section with dimensions 30, 3, 10, and 10.

Cache : Voir 30x30

30x30 2N

A = 3,5 cm²
I_x = 2,8 cm⁴
I_y = 3,5 cm⁴
W_x = 2,1 cm³
W_y = 2,7 cm³
m = 1,0 kg/m



The technical drawing shows a 30x30 2N angle. The overall dimensions are 30 mm by 30 mm. The drawing includes a cross-section view with a central hole and a side view showing the profile. The properties listed are: A = 3,5 cm², I_x = 2,8 cm⁴, I_y = 3,5 cm⁴, W_x = 2,1 cm³, W_y = 2,7 cm³, and m = 1,0 kg/m.

Cache : Voir 30x30

30x30		L (mm)	N°
1x		50 ... 5600	3 842 990 720 / L
1x	M8 	50 ... 5600	3 842 990 721 / L
1x	M8  M8	80 ... 5600	3 842 990 723 / L
1x	M8  D11	70 ... 5600	3 842 990 724 / L
1x	D7,8  D7,8	60 ... 5600	3 842 992 965 / L
1x	D11 	50 ... 5600	3 842 990 722 / L
1x	D11  D11	60 ... 5600	3 842 990 725 / L
1x	D11  D11V	60 ... 5600	3 842 990 726 / L
 20x		5600	3 842 509 178

Quick & Easy (p. 2-8, voir pages dépliantes)

30x30	3 842 993 703 / ...
Longueur L (mm)	50 ... 5600
Usinage standard des profilés (respecter la longueur minimale, p. 2-16)	M8 / D7,8 / D11 / DB11
Usinage individuel des profilés (L _{max} = 4900 mm)	DI / DIS / MT / MTS / MI / MIS / DG ¹⁾

¹⁾ DG_{max} = 45° ; L_{max} / L_{min} = 352 / 404 mm

30x30 1N		L (mm)	N°
1x		50 ... 5600	3 842 992 397 / L
20x		5600	3 842 506 948

Quick & Easy (p. 2-8, voir pages dépliantes)

30x30 1N	3 842 993 704 / ...
Longueur L (mm)	50 ... 5600
Usinage standard des profilés (respecter la longueur minimale, p. 2-16)	M8 / D7,8 / D11 / DB11
Usinage individuel des profilés (L _{max} = 4900 mm)	DI / DIS / MT / MTS / MI / MIS / DG ¹⁾

¹⁾ DG_{max} = 45° ; L_{max} / L_{min} = 352 / 404 mm

30x30 2N		L (mm)	N°
1x		50 ... 5600	3 842 992 399 / L
20x		5600	3 842 506 950

Quick & Easy (p. 2-8, voir pages dépliantes)

30x30 2N	3 842 993 705 / ...
Longueur L (mm)	50 ... 5600
Usinage standard des profilés (respecter la longueur minimale, p. 2-16)	M8 / D7,8 / D11 / DB11
Usinage individuel des profilés (L _{max} = 4900 mm)	DI / DIS / MT / MTS / MI / MIS / DG ¹⁾

¹⁾ DG_{max} = 45° ; L_{max1} / L_{max2} = 352 / 404 mm

Figure 6.9: Data sheet Bosch profile with groove, grid dimension, centre hole. [44]

6.1.3 Appendix 3

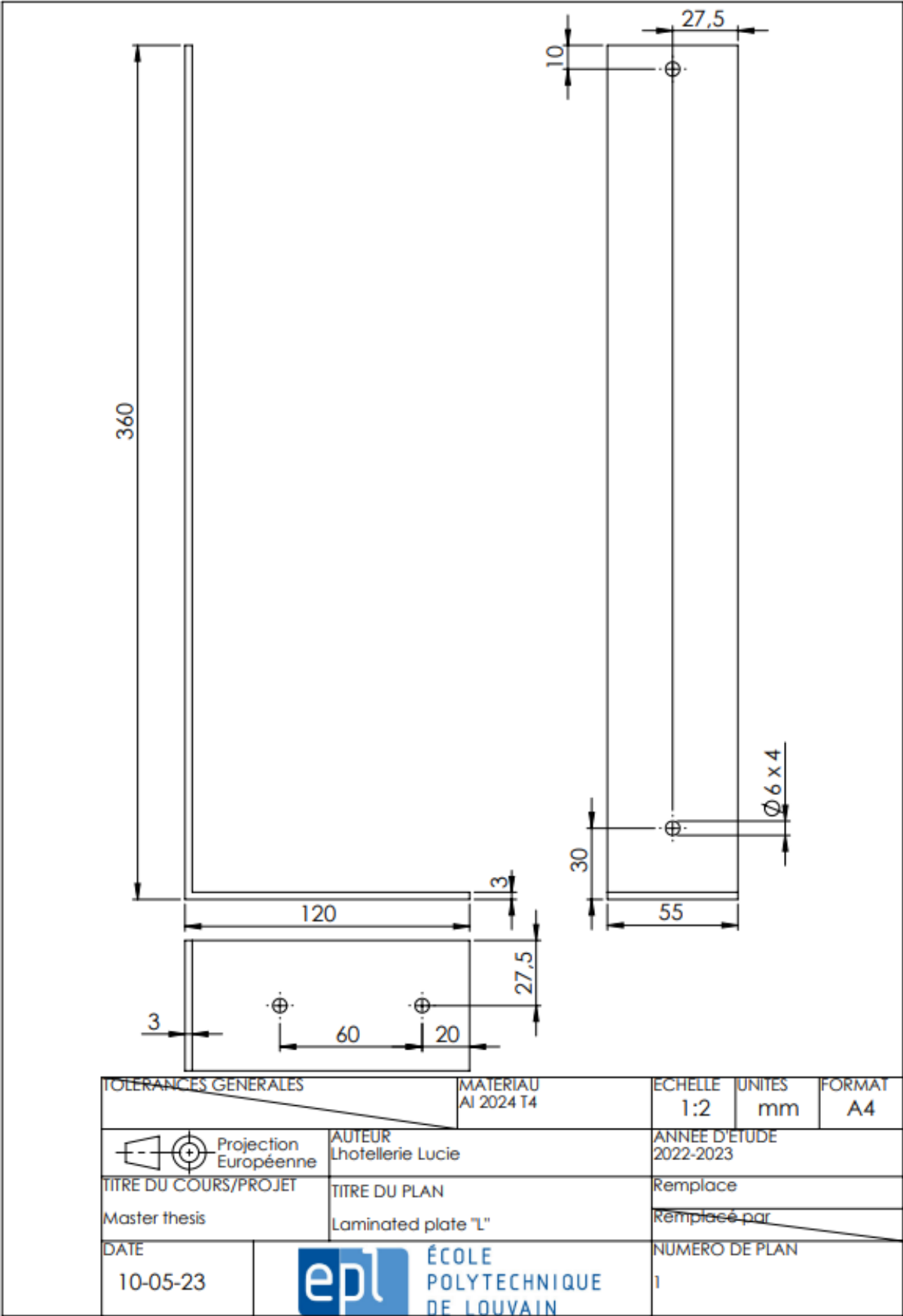


Figure 6.10: Plan of the "L" shaped laminated plate.

6.1.4 Appendix 4

Here is the document that our supervisor provided me in order to realize the acquisition of the signals via the XSENS^{MT} sensors.



Institute of Mechanics, Materials and
Civil Engineering (IMMC)

User guide: Xsens

User guide: Xsens

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1. Realtime visualization, calibration and data recording

- 1° Plug the Awinda station (with the antenna) into the Xsens computer using its USB cable.
- 2° Power on each tracker (the light should be blinking) you will use.
- 1° Open the terminal  on the Xsens computer. Enter one of the two following lines to launch the driver:
 - `./XSENS_1` if you use the trackers with their default position;
 - `./XSENS_2` if you use only the ≤ 8 trackers with the new names written with stickers.
- 2° This will automatically open your browser at the correct URL (localhost:5000), displaying the MBSysWeb platform. If possible, use Chrome as default browser (otherwise you might need to refresh the page).



- 3° In the upper left corner, click on the menu, then on "Connect" to load the skeleton model.

Figure 6.11: User guide: XSENS^{MT} page 1. [45]

- 4° In the terminal, if the trackers are correctly connected, the right number of trackers should be indicated after "Number of connected MTWs:". Press "y".
- 5° In the web interface, click on "Mag IMU's" on the bottom left corner. Align the trackers as on the image that appeared. Then click on "Calibrate".
- 3° Place the straps at each location on the subject (see [video](#)). They should be placed with the Xsens sign on top and the light below, and underneath the strap for better results.

Remarks:

- The pelvis tracker must always be placed!
- The pelvis tracker must be placed on top of the sacrum.
- The upper leg motion tracker must be placed on the flat surface of the thigh, above the knee. Pay attention that the extremity of the strap should not be between the legs, to avoid friction.
- The lower leg motion tracker must be placed on top of the flat surface of the shinbone.
- The foot motion tracker must be placed on the shoe tongue with the foot pad, and attached with the laces



- 6° Ask your subject to stand in the T (or N if the trackers from the upper-limb are not used) pose, and click on the calibration button in the web interface.



T pose



N pose

- 7° Even after calibration, there is still an unknown variable, the global yaw. This value should equal 0 when the pelvis IMU is oriented towards the east (so when the **subject looks towards the west** if this IMU is located on the back), thus we recommend to have this orientation when the calibration is performed. Otherwise, you need to manually tune the Yaw parameter (on the web platform) until the subject's movement looks right. It is advised to note the yaw parameter of this recording since it will be used during the data treatment.
- 8° In a new tab, type the following lines to record the messages from the trackers. The subject must stay in T (or N)-pose for 5 sec at the beginning of the recording.

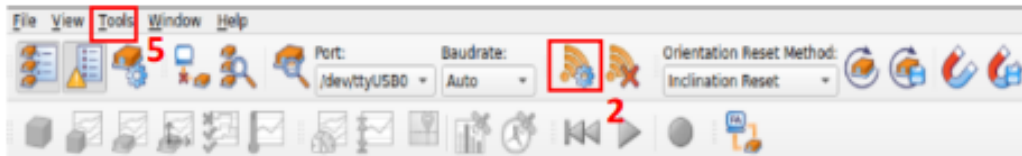

```
cd Folder/ → "Folder" is the name of the folder you want the data to be saved.
rosbag record --all
```
- 9° To end the recording, press CTRL+C. End also the driver the same way.
- 10° Turn off all trackers at the same time: see section 2.

Figure 6.12: User guide: XSENS^{MT} page 2. [45]

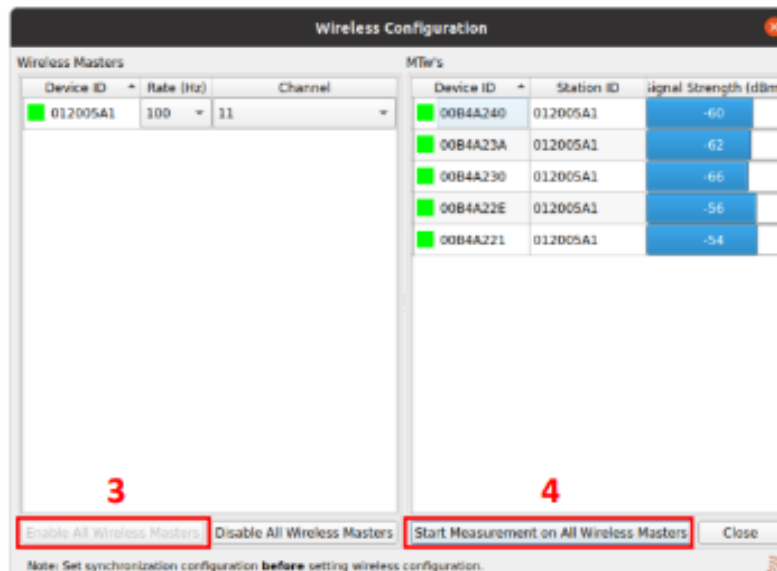
2. Powering off the trackers

You can power off all the trackers at the same time with the following procedure:

- 1° In the tab of the terminal where you launched the driver (which must be stopped), type `./XSENS_Disconnect`.
- 2° This should open an interface, where you have to click on "Wireless configuration".



- 3° A new window appears with only the antenna listed in the left column (with a grey square). Click on "Enable All Wireless Masters": the trackers will appear in the right column and all the squares become light green (you have to move them if they have been put on standby and do not appear).



- 4° Once all the trackers are listed, press "Start Measurement on All Wireless Masters" and the squares now turn dark green.
- 5° Close this window and select in the top left corner "Tools > Power Off > Wireless System" in the main interface. All the trackers should be now turned down. Close the interface.

Figure 6.13: User guide: XSENS^{MT} page 3. [45]

3. Data treatment and offline visualization

0° Adapt the sampling frequency depending on the number of trackers you are using.

In , open `catkin_ws/src/xsensdriver/src/mt_w_main.cpp` and change the value at line 94:

- a. For up to 4 trackers, use a frequency of 120 Hz.
- b. For up to 8 trackers (lower-limb), use a frequency of 100 Hz (**default option**).
- c. For all the 13 trackers, use a frequency of 60 Hz.

Once the frequency is changed, you must compile the code by entering the following lines in the terminal:

```
cd catkin_ws
catkin_make
```

1° In a new tab of the terminal, type the same line `./XSENS_#` as you used for the data recording.

2° This will again open your browser where you will have to connect once more. Adapt the "Yaw" slider to the value that was used during the recording. However, do not press anything in this terminal now, since you should not be connected to any trackers.

3° If you do this step just after the data acquisition, go to the tab where the record was done. Otherwise, type the following line in a new tab:

```
cd Folder/ → "Folder" is the name of the folder where the data were saved.
```

4° Type the following line:

```
rosbag play file.bag
```

where *file* is the name of the file you just recorded. It should show the recorded moves on the web interface. The goal of this step is to replay the file you just recorded and to apply calculations on it to obtain joint angles, step length and COM position.

5° The joints angles are saved in a `.txt` file that can be analyzed in Matlab. The file can be found in `catkin_ws/src/xsenspose/output`. Do not forget to change its name and to move it into another folder, otherwise it will be overwritten. The corresponding names of each column are given in the txt file in the same folder.

Figure 6.14: User guide: XSENS^{MT} page 4. [45]

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