Master's Thesis

MASTER
IN
BIOMEDICAL ENGINEERING

DEVELOPMENT OF A PREDICTIVE MODEL FOR CRUTCH-ASSISTED GAIT USING OPTIMAL CONTROL THEORY

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Summary

Crutches are an assistive device used all over the world, and have been around for thousands of years. However, not a lot of studies have been conducted in order to study the effects of its use from a biomechanical point of view. This project is focused on the development of a predictive model for crutch-assisted gait by using a predictive simulation based on an Optimal Control Problem (OCP). The generation of predictive simulations can be used as a tool to further understand how the musculoskeletal system works during crutch-assisted walking. The project is divided in two main objectives: determining if a muscle-driven predictive simulation is able to reliably predict crutch-assisted gait and studying the possibilities of reconstructing different gait patterns by tracking experimental ground reaction forces (GRFs) in the crutches.

In order to develop this project, the simulation framework PredSim was used. This framework, which proved reliable for non-assisted walking prediction, is modified to incorporate the use of crutches, and the settings are explored in order to obtain accurate predictions of crutch-assisted gait patterns. The use of different models, capable of moving in 2 or 3 dimensions, was also studied. OpenSim was used all throughout the project for the visualization and analysis of the resulting motions. For the tracking of GRFs, the OCP formulation was modified to add a tracking term, and experimental measures of the orientation and the forces applied to the tip of instrumentalized crutches were used as reference (GRFs) and as a way of validating results (orientations).

In general, the results from 2D simulations did not lead to consistent results and a good percentage of them were not natural motions. However, 3D simulations succeeded in predicting crutch-assisted gait patterns with relative consistency that coincide with some of the patterns defined by physiotherapists and physicians for crutches use. The number of trials is limited and some limitations were identified during the development of the project, reason why it is complicated to give definitive proof or draw definitive conclusions based on the work done. However, it was observed that prediction of crutch-assisted gait patterns was possible, despite all the limitations, using the PredSim simulation framework. It could also be observed that the OCP's cost function did not need to be modified from the original version, meaning that the choosing of an optimal crutch-assisted gait follows the same rules as non-assisted walking. Regarding the second objective of the project, GRFs were accurately tracked. Regardless of that, probably due to limitations of the project, the resulting gait patterns did not reliably reenact the experimentally measured patterns. In general, further exploration of the simulations' configuration should be done. However, this project is a valuable introductory exploration into predictive simulations for crutch-assisted gait.

Chapter 1

Introduction

Most people will need to use an assistive device for walking at some point in their lives: either for a few weeks or months until a lesion is healed enough, or for longer periods of time, if dealing with a chronic walking impairment. Crutches work by transferring the supporting of the body weight from the lower extremities to the upper trunk, meaning that sufficient upper body strength and coordination is needed in order for the aid to be effective. For spinal cord injury patients who, depending on the place of the injury, may still be capable of exerting force with the torso and arms, but have difficulties bearing their weight on their legs, the use of crutches, normally combined with lower limb exoskeletons, is a good option [1]. However, the use of crutches for long periods of time might lead to secondary injuries [2]. This is why understanding the implications of crutch-assisted gait on the musculoskeletal system is of such great importance in order to avoid, or at least minimize, the negative effects that might appear. These effects could be minimized if the patients had access to personalized gait analysis and training, however, before doing so, the crutch-assisted gait has to be further studied from a biomechanical point of view.

Assisted walking using crutches has not been widely studied from a biomechanical point of view. Most of the work related to the subject is developed by physiotherapists or physicians and is focused on explaining how to perform the different types of gait patterns and their benefits and disadvantages. There exist 4 main types of crutch-assisted gait patterns: 2-point, 4-point, 3-point and swing-through, each with multiple variations, depending on the way the crutches and feet touch the ground. However, when we try to take a deeper look into why are those the predetermined ones or to further understand the implications of using one or another in the joints and muscles, little information can be found. This is one of the reasons why it is so important to develop physics-based simulations that can help further understand why those are the chosen patterns and, furthermore, get more insight on how the patient's body behaves when crutches are being used.

1.1 Objectives

This project is divided into two sequential objectives, since the second one is completely dependent on the success of the first. The first objective is to determine if it is possible to reliably predict crutch assisted walking patterns. This is important to further understand how the movements are produced, and therefore improve personalized rehabilitation plans and develop novel treatments. However, it must also be taken into account that many people with chronic conditions will be using crutches for very long periods of time: in these cases, it would be extremely helpful to find the optimal way of using the aids, so that secondary issues or long-term negative impact on musculoskeletal health is reduced as much as possible. Once it is confirmed that this is possible, it is time to move on to the next objective, for which we are using experimental data collected with instrumentalized crutches. This second objective consists on studying whether or not it is possible to reconstruct the movements of the subject just by using the ground reaction forces (GRFs), in order to know what gait pattern they are doing without need to use sensors on their bodies, but just on the crutches. If this was possible, there would be a new and interesting possibility for lower limb orthosis control, since most people who need to use lower limb orthosis usually use crutches simultaneously in order to assist with balance.

1.2 Scope of the project

This project is mainly focused on studying the possibilities of the chosen line of research, rather than to give definitive proof or unequivocal conclusions about what crutch-assisted walking patterns work better for each individual patient or if they can be overall improved or are good as they are. Since the research on this field from a biomechanics point of view is not yet very advanced, this project is intended to also serve as a proof of concept and inspiration for future work.

1.3 Related work

Studies focused on crutch assisted-gait have been developed at different levels and using different strategies. An interesting example, which also has a biomechanical approach, using physics-based simulations, is the work published by the supervisor of this project, M. Febrer-Nafría *et al.* in 2021 "Prediction of three-dimensional crutch walking patterns using a torque-driven model". In this study, different cost functions were studied for the first time to predict 3D crutch-assisted gait with a full body model. In order to predict different gait patterns, swing duration and stride length were imposed to determine when ground reaction forces needed to be zero and to define spatial constraints. With the same optimal control problem (OCP) formulation, they were able to predict all the objective patterns, however, some showed better results when other formulations were used, and it was hypothesized that optimality criteria might be pattern-dependent [3]. In the project of this thesis, the objective is to determine if all patterns can be predicted with the same OCP formulation, without using any controls that directly influence the optimal solution's gait pattern.

In relation with the second objective of this work, it is important to note a project that performed the analysis of spatial and temporal parameters during crutch assisted gait (M. Narvaez, M. Salazar & J. Aranda, 2024), supervised by the co-supervisor of this thesis. Thanks to their work, we were able to get already analyzed experimental data that can be used both to validate the simulations of the first part of the project, and to be used as inputs for the prediction method in the second part. This data was collected using instrumentalized crutches and includes the GRFs for each crutch, measured with force sensors at the tips, and their orientations, evaluated using inertial measurement units (IMUs) [4].

Finally, it is very important to note that this project is based on a framework for predictive simulations of locomotion that was developed in the neuromechanics group of the Katholieke Universiteit Leuven under the name of *PredSim* (A. Falisse *et al.*, 2019 and L. D'Hondt *et al.*, 2024). The framework includes codes and data that allow to generate two- or three-dimensional muscle-driven predictive simulations of locomotion. This code has been proved able to reliably and consistently predict non-assisted walking [5, 6]. Our project is meant to determine if it is also reliable for predicting crutch-assisted gait and if any changes in the formulation or settings are necessary to do so. During the course of this project, we had the great opportunity of collaborating with the PredSim team, discussing results and getting feedback directly from developers of the framework.

Chapter 2

Theoretical background

In this chapter, key concepts about crutch-assisted gait will be presented. These are important in order to understand the methodology and the results obtained, and the consequent discussion. With the same objective, physics-based predictive simulations, which are a crucial part of the methodology of this project, are also briefly explained. For a better understanding of the concepts, both simulation and experimental projects have been consulted and key results and conclusions have been included in this chapter.

2.1 Crutch-assisted gait

2.1.1 Crutch-assisted gait patterns

Using crutches to assist walking implies a lot of changes in how the subject uses their body compared to non-assisted walking. Depending on the specific injury or disability the patient is dealing with, different types crutches may be used and different walking patterns must be adopted so that the subject is able to support their own weight and ambulate without further injuring themselves or feeling pain. This project focuses on forearm crutches, which, in addition to the handle, have an open cuff that surrounds the forearm and provides stability while still allowing freedom of movement. This type of crutches has been observed to be the most adequate for long-term users, such as spinal cord injury patients or exoskeleton users who have enough strength in the upper body [7, 8]. In most of the literature, for forearm crutches, there are four main defined walking patterns, each recommended for a different type of patient and with multiple variations. [9, 10]

2-point

The 2-point gait pattern is probably the simplest one, since it follows the natural movement of the arms to advance alternately with the feet, as done in non-assisted walking. This type of gait is also used in trekking when using poles and it has been observed to imply a reduction in ground reaction forces (GRFs) for the feet and suggested that it can improve walking efficiency [11, 12]. This pattern is adequate for partial weight bearing on both feet but does not provide as much support as other patterns, such as the 4-point, however, it allows the patients to advance faster, with experimental velocities of around 0.4 m/s [4].

To perform this gait pattern, starting from a standing still position, one of the crutches —let us take the right crutch as an example— swings forward together with the left foot. Once the foot and crutch touch the ground, the left crutch and the right foot replicate the same movement. These two sequential movements make up one full gait cycle.

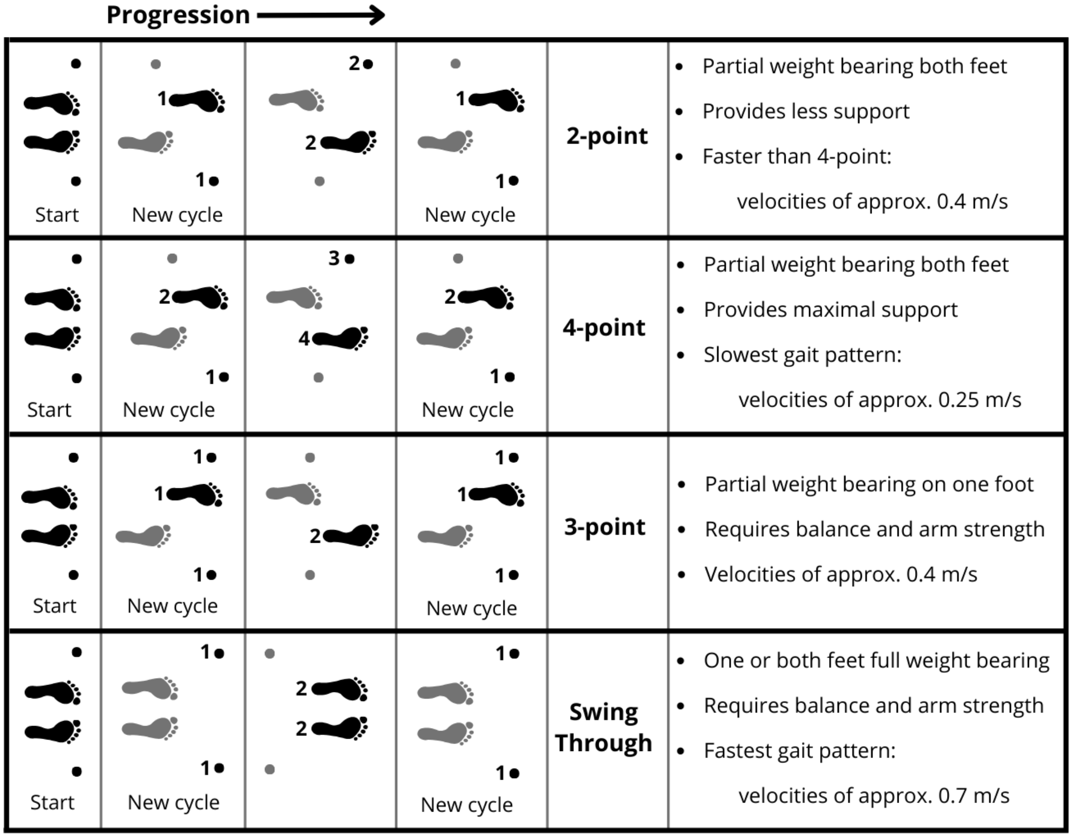

Figure 2.1: Diagrammatic representation of the 4 different main gait patterns, where shading indicates the elements that advance in a same frame and numbers indicate the order in which the movements happen.

4-point

For patients who require maximal support using their crutches, for example due to muscle weakness, and who have poor coordination and/or balance, the 4-point gait pattern is recommended. It also implies partial weight bearing for feet, as the 2-point pattern, but reduces even more the GRFs on the legs [13]. However, as commented before, this pattern does not allow velocities as high as the other ones, and experimental data suggests an approximate velocity of 0.25 m/s [9, 4].

In order to perform this pattern, first, one of the crutches, say the right one, swings forward. Once the right crutch is standing on the ground, the left foot advances. Then, the left crutch, which was left behind, advances, and finally the right foot swings forward. These steps make up one gait cycle.

3-point

The 3-point gait pattern is especially recommended for those patients who need to avoid partially or totally bearing weight on one of the feet. This pattern requires the patient to have moderate arm strength, but can allow for a higher forward velocity than the patterns previously described, with experimental velocities being around 0.4 m/s [14, 4].

The 3-point gait pattern consists in swinging forward both of the crutches together with the non-weight bearing foot and then, while supporting the weight more or less on the crutches depending on the requirements of the patient, advance the healthy foot.

Swing-through

This last pattern, the swing-through, requires moderate to high arm strength, coordination and balance, reasons why it is not accessible to a large group of patients who use crutches. However, if those requisites are met, it is the fastest crutch-assisted gait pattern, with experimental velocities surpassing 0.7 m/s [15, 4]. This pattern also requires one or both feet to be weight bearing.

To perform the swing-through, the two crutches advance simultaneously and, once they touch the ground, the patient fully carries their weight on the crutches and swings the whole body forward, landing in front of the crutches and then repeating the movement. This pattern can also be executed when one foot cannot bear any weight. Another variation of this pattern is the swing-to, which consists of the same movements but, when swinging the body forward, the patient's feet do not surpass the crutches.

2.1.2 Length of the crutches

This project is focused on crutch-assisted walking using forearm crutches, which have a handle that patients have to grab with their hand, and an armband around the forearm that helps transfer body weight and reduces pressure on the hands [7]. These kind of crutches have to be adjusted to the subject using them by changing the length of the pole under the handle. The recommended length is such that, when the patient is standing still with their arms relaxed and the crutch vertical on their side, the handle of the crutch is aligned with the head of the ulna bone, approximately at the wrist joint crease [16].

The length of the crutches has been observed to be a very important parameter to personalize to the patient. In a study with patients of hip replacement surgery using one crutch, it was found that shorter crutches led to more weight beared on the operated limb and less on the crutch, while longer crutches led to greater gait instability, both undesirable conditions [17]. Another study found that standard length crutches in a 3-point pattern were perceived by the subjects as more comfortable, even though the time integral of pressure in the hands was slightly higher for standard length [18]. These studies were performed with forearm crutches, but investigations with axillary crutches lead to similar results: one study determined that slightly shorter and longer crutches led to a significant increase in the range of motion of multiple articulations [19]. In another publication, which studied the swing-through pattern, it was observed that shorter crutches required more activation of the triceps, which would probably cause patients to feel fatigue if non-optimal length is used [20].

2.1.3 Ground Reaction Forces

There are multiple publications that study crutch-assisted gait GRFs, all of them working with experimental data. However, most of them focus on the forces on the feet, not on the tips of the crutches. Still, some of the publications contained interesting information about the reaction forces on the crutches: when performing a swing-through pattern, for example, total GRFs in the crutches can raise to about a 100% of the body weight of the subject, as they are swinging forward [21]. A pattern similar to a 3-point was studied in another project and it was observed that total GRFs in the crutches raise to be between 20 % and 40 % of the body weight, although this results might change in our simulations since the subject in that publication was using a complete lower limbs exoskeleton that allowed them to walk with complete spinal cord injury paraplegia [22].

The project developed by M. Narvaez, M. Salazar & J. Aranda in 2024 [4] included the recording and analysis of different gait patterns, and the data provided by them is extremely useful to visualize how crutch GRFs change depending on the pattern performed. However, it is important to take into account that participants, which were healthy subjects, were asked to emulate a foot injury, so right-left asymmetries are expected. Representative samples can be seen in Fig. 2.2. In the study of M. Febrer-Nafría (2021), simulated GRFs obtained with a torque-driven model for 4-point, 2-point and swing-through were observed to be similar to the experimental GRFs of the previous study (Fig. 2.2) in magnitude and shape [3].

2.1.4 Torques

In crutch-assisted walking, compared to unassisted gait, the arms must generate more force to deal with the weight supported by the crutches, so upper body torques are an important aspect

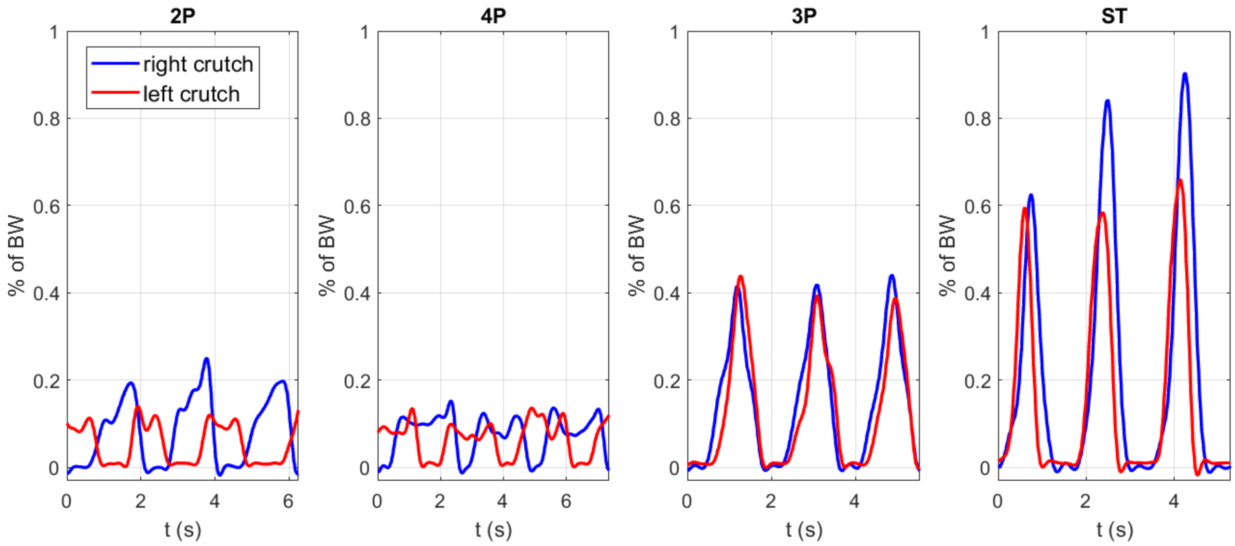

Figure 2.2: Representation of the percentage of body weight supported in the crutches during 3 gait cycles, extracted from the data recorded by M. Narvaez, M. Salazar & J. Aranda in 2024 [4], one plot for each principal crutch-assisted gait pattern. From left to right: 2-point (2P), 4-point (4P), 3-point (3P) and swing-through (ST).

to study. In contrast to ground reaction forces, torques can not be experimentally measured, so a biomechanical model is needed to estimate them. In general, torques produced in the joints during crutch-assisted gait have not been very studied in the past. In the review on biomechanical modeling applied to device-assisted locomotion developed by M Rodrigues *et al* (2022) , only 10 publications were featured that explicitly study the torques in the joints during crutch-assisted walking [23]. However, all but one of them used an inverse dynamics approach or tracking of experimental measures to study torques. The one different study is the one published in 2021 by M Febrer-Nafría *et al*, where torques in the shoulder are studied in simulations of different gait patterns using torque-driven predictive simulations [3]. In this study, the results are consistent with those from their other study (M Febrer-Nafría *et al*, 2020), where torques in all the joints of the model were computed by tracking experimental data for a 4-point gait pattern [24]. Although the experimental values for the tracking and development of the predictive simulations were taken with orthosis assistance of the subject, the resulting torques can be used as a reference in the analysis of results of crutch-assisted gait predictions. Another interesting study which is not included in the M Rodrigues review is the one developed by B Fité, where lumbar and shoulder extension torques were thoroughly studied, obtaining values of similar magnitude with those from M Febrer-Nafría [25]. Another study was found that focused on the purely mathematical modeling of a variation of a 3-point pattern, and provides a representation of the resulting torques in the shoulder, hip, knee, and ankle [26]. The torques are quite consistent in magnitude between the different studies, but it is important to take into account that shape, mainly, and magnitude, will depend on the kind of pattern that is being studied.

2.2 Physics-based simulations

Physics-based computer simulations are a very useful tool to understand how movement is produced and thus help define novel treatments or rehabilitation plans. These simulations, when performed with a model of the human body, can help generate movements by using only the mathematical description of the neuro-musculoskeletal system without making use of experimental data except for validation. This means that, by knowing how the underlying mechanisms of movement work from a mathematical point of view, and having a sufficiently accurate model of the subject, we are able to replicate which movement is optimal for each situation. By using this strategy, we are also able to examine the effects on movement of altering properties of the neuro-musculoskeletal model, such as decreasing muscle strength or changing the model's proportions. Walking motion, both healthy and pathological, has already been proved to be possible to predict using physics-based simulations, capturing both mechanic and energetic aspects of the gait. Pathological gait predictions help demonstrate how patients may adapt their gait to compensate specific conditions, like one-sided leg disorders [27] or the use of a prosthetic foot and ankle in amputees [28]. As previously commented, it can be observed in the M Rodrigues review on biomechanical modeling applied to device-assisted locomotion (2022) that most of the biomechanical studies on crutch gait use an inverse dynamics approach: only 7 out of 18 featured studies performed simulations and, of those, only three were not tracking experimental motion but predicting it [29, 30, 3]. When looking at the available studies performing predictive simulations of crutch-assisted gait patterns, it can be observed that most use a very simplified 2D model and stick to the prediction of the swing-though pattern, probably since it is the simplest to model [31, 32, 29, 30]. To our knowledge, the only study that uses physics-based simulations to predict 3-dimensional full-cycle crutch walking patterns with a full-body model is the one published by M Febrer Nafría in 2021. However, this study did not include muscles in the model, but worked with torque generators in the joints [3]. In our project, predictive simulations of different full-cycle crutch-assisted patterns have been obtained for the first time using a full-body 3D model which includes muscles in the lower limbs and lower half of the torso, while maintaining the torque generators at the upper limbs.

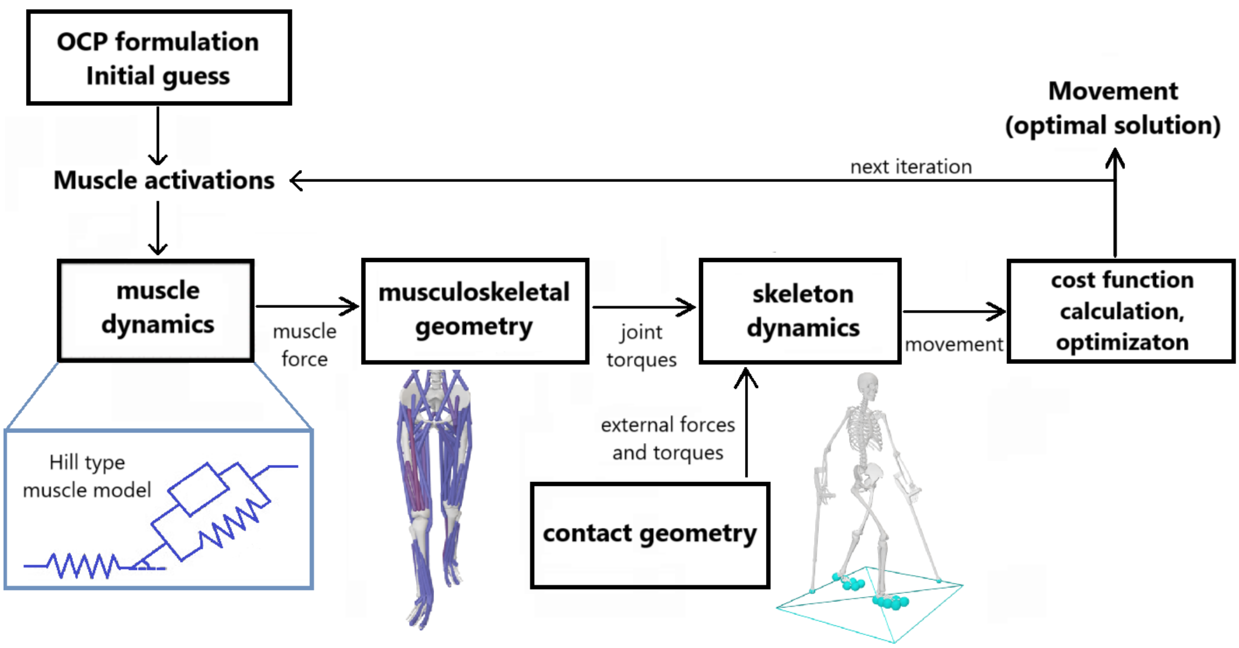

Figure 2.3: Schematic of the used muscle-based predictive simulations, which optimize a movement so that a cost function is minimized.

The simulation framework used in this project [5, 6] is based on a multi-body system of a human subject: it features different bodies, which usually correspond to a bone or a group of bones, and the joints that unite them and allow relative movement. The movement of the bodies and, in consequence, the behavior of the joints, is determined by the predefined muscle dynamics, which define the activations of the muscles and how they relate to force generation. In Fig 2.3, a schematic of the functioning of muscle-driven predictive simulations is represented. In order to define muscle behavior, the Hill type model is normally used. This model determines the relationship between a muscle's length, contraction velocity and activation, and the force that the muscle exerts [33]. The most simple Hill type muscle model consists of three elements: a contractile element that models the muscle's actin and myosin cross-bridges, a spring element in series representing the tendon, and another spring element in parallel with the other two modeling the connective tissues [34]. However, different configurations have been developed to better simulate physiological muscle behavior: the most common one introduces a pennation angle: the angle between the direction of the muscle fibers and the force-generating axis. This angle is located in the model between the series element, representing the tendon, and the contractile and parallel element, which represent the muscle [35]. Another possibility for these simulations is to use torque-driven models, which define a torque generator in each joint of the model and assume that the maximum torque exerted depends on the kinematics of that joint. This method allows for faster computation, but does not allow to study individual muscle and tendon properties [36].

In order to find the optimal solution, physics-based simulations typically define a multi-objective performance criterion that translates into a multi-term cost function that must be minimized. This cost function usually combines energy and effort considerations, including terms like metabolic energy consumption or muscle activation [5, 6]. Since this cost function is typically multi-dimensional and has too many variables to analytically solve the problem, numerical methods are used. Direct collocation is a powerful method to solve these problems. It works by iteratively approximating the objective function using polynomials in intervals around a given point and using that information to determine the following point at which the analysis should be done [37]. This second point should be such that it brings the solution closer to the problem's minimum, and the final solution should fall in the global minimum.

Chapter 3

Methods

3.1 Simulation Framework: PredSim

For the development of this project, simulation framework PredSim was used [5, 6]. This is an open source software and a repository, which includes everything needed to generate three-dimensional muscle-driven predictive simulations of locomotion, is available in GitHub [1]. This framework was developed to generate predictive simulations of locomotion and has been proved reliable to predict non-assisted gait of human subjects. In this project, even though some modifications to the code have been studied, all the simulations have been done using this software.

PredSim is OpenSim based, which means that it uses OpenSim's core libraries, models, and processing engine to generate the simulations. OpenSim is an open-source software system that allows users to develop models of musculoskeletal structures and create dynamic simulations of movement [38, 39]. PredSim makes use of OpenSim's defined musculoskeletal dynamics to generate the predictive simulations.

3.1.1 Musculoskeletal models

In order to perform the simulations, a musculoskeletal model is needed. These models are a combination of rigid bodies, which represent body segments. The rigid bodies can represent whole body parts, such as the hand or the torso, or can be more specific, representing just one bone, such as the ulna or the radius. In the models used in this project, more detail is observed in the feet than in the hands, since the feet behavior is more relevant than hand behavior when studying gait, even with crutches. Both a 2-dimensional and a 3-dimensional model were used to generate the simulations. For this project, the musculoskeletal models used were OpenSim models.

2D model

Using a 2D model can significantly reduce computing time due to its simplifications when compared to 3D models, which is why this model was used to perform most of the simulations in this project. However, it also has to be taken into account that 2D simulations might not be as representative of human biomechanics as 3D ones. This is due to the fact that the 2D model does not allow for movement out of the sagittal plane, so it can not capture movements that can be important to simulate crutch-assisted gait, such as lumbar bending or arm adduction.

The 2D model used in this project was mainly developed by A Seth, integrating definitions of different joints behavior [40, 41, 42, 43]. It weights 76.24 kg and measures 1.76 m. The model has 16 degrees of freedom (DoFs), three corresponding to pelvis translations, one for lumbar extension, four for each lower extremity (hip, knee, ankle and metatarsophalangeal joints, one for each), and two for each arm (one at the shoulder and one at the elbow). The model includes 18 muscles from the waist down. All joints in the legs are muscle-driven, while the joints in the arms and trunk (lumbar extension, and arm and elbow flexions) are driven by ideal torque actuators. The proportions of

[1] https://github.com/KULeuvenNeuromechanics/PredSim

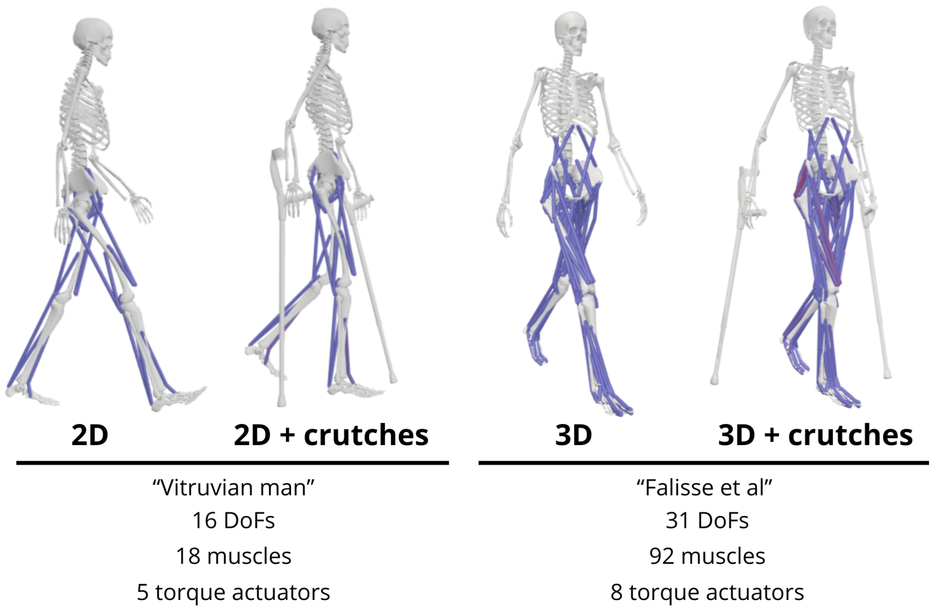

Figure 3.1: Different OpenSim models used, together with their basic characteristics.

this model are based on the Vitruvian Man [44].

3D model

The 3D model, even though it lengthens computing time, allows for more representative simulations of the human body, since it allows movement out of the sagittal plane by having more DoF and muscles than 2D models.

The 3D model used in this project was developed by A Falisse *et al* [45], using as a basis the model developed by A Seth to study muscle contributions during running, which was developed using experimental measures of a subject [46]. In A Falisse's work, the subtalar and metatarsophalangeal joints were added to the original model, and forearm rotation was eliminated. The result is a 3D model with 31 DoFs, with a weight of 63.14 kg and a height of 1.62 m. Each lower extremity has 7 DoFs, three at the hip, and one at each the knee, ankle, subtalar and metatarsophalangeal joints. Each arm has 4 DoFs, three at the shoulder and one at the elbow. 6 more DoFs correspond to pelvis rotations and translations with respect to the ground, and three more to the lumbar rotations. The model has 92 muscles from the chest down and 8 torque actuators that govern the behavior of all the arm joints.

3.1.2 Model with crutches

Crutches were added to the models of the human body by creating a joint between each hand of the model and a crutch. Since we work with forearm crutches, which are supposed to move together with the forearm, the joint was fixed, not allowing relative movement between the hand and the crutch, so the models with crutches had the same number of degrees of freedom as the ones without. The CAD model of the crutches was already available in the research group, as it had been previously developed and used in other projects [3, 24].

3.1.3 Terms of the Cost Function

PredSim uses a multi-objective cost function that includes ten different terms. However, it has been identified that only five of them have significant importance in finding the right solution: metabolic energy, muscle activation, actuators excitation, joint accelerations, and, to a lesser extent, joint limit torques. The rest of the terms end up not contributing at all to the objective function in our simulations, as they are of such a small order that they are approximated to zero.

The metabolic energy term is representative of the amount of energy that the body consumes to perform the movement, and during this project is estimated using the model developed by L.J. Bhargava *et al* in 2004 [47]. The muscle activation and actuators excitation terms are related to the effort the muscles and actuators are making in order to produce the movement. However, it has to be taken into account that the relationship between activation and excitation and exerted force is non-linear and depends on other variables such as the velocity of the movement or fatigue. The objective of the joint accelerations term is to minimize the rate at which joint velocities change in order to obtain smoother movements. Finally, the joint limit torques term refers to the passive joint torques, those that are generated in a joint without any active muscle contraction. These appear due to the mechanical properties of the surrounding tissues, such as tendons and ligaments.

3.1.4 Inputs and Outputs

To generate simulations, the only thing that needs to be modified is the main code, which is written in MATLAB. In order to perform a simulation, PredSim requires some inputs that have to be defined by the user before starting a simulation. An OpenSim model of the human body that includes forearm crutches is needed. In addition, to start the simulation, a motion file containing the relative angles of the joints in the model and the corresponding time is recommended to be used as an initial guess for the optimal control problem. In this project, the motion file used was the *IK_Guess_Full_GC* (Inverse Kinematics Guess Full Gait Cycle), provided by the PredSim GitHub repository, which featured a human model walking forward without making use of any assistive devices. Defining an initial guess helps reduce the time until convergence if compared to the also available 'quasi-random' configuration, which sets the initial guess to be the predeterminate position of the model statically moving forward. The paths to these files have to be added in the main code, together with the path to the folder where the results will be saved and a subject name.

Apart from these two required external files, it is also important to modify the settings of the simulations according to our objective in the main code. Any setting that is not specified in the main code will be set to the default configuration. One of the most important settings is the forward velocity, which defines the median velocity at which the subject has to advance. It is also important to set distance constraints between points in the model to avoid body parts and, in this case, crutches, from crossing through other body parts. It also has to be specified whether we want a half-gait cycle simulation, which imposes left-right symmetry during one gait cycle, or a full-gait cycle, which allows for asymmetries. Some characteristics of the musculoskeletal model can also be modified through the settings, such as muscle strength and tendon stiffness, which allows for gait predictions of simulated injury or weakness patients. Finally, it is also important to note that it is possible to modify the weights and exponents of each objective function term through the settings. This way, the user can change the importance of each term on finding the optimal solution: for example, if we were to perform a simulation that aims to minimize muscle effort over other parameters, the weight or exponent corresponding to the muscle activation term should be increased.

Once all important settings are defined, the code can be ran. First, information about the model is obtained using OpenSim, and a function is generated for the rigid-body dynamics of the model. Right after, external CasADi functions are defined that allows the calculation of exact derivatives. Then, it is time to solve the Optimal Control Problem (OCP), which incorporates implicit formulations of rigid-body dynamics, muscle contraction dynamics, and muscle activation dynamics. Using the previously created external functions, the cost function and constraints are defined, and the OCP is solved using CasADi [48] and Ipopt [49, 50]. [6]

Finally, post-processing is done on the data from the last step of the solver. At this point, variables that do not directly influence optimization, such as ground reaction forces or muscle fiber lengths, are computed. All the resulting variables, metadata, settings, and information about the model are saved as a MATLAB data file in the specified saving folder. Furthermore, kinematics and muscle activations are saved as an OpenSim motion file to allow visualization of the predictions. PredSim also saves a text file with MATLAB's command window output. [6]

3.2 Prediction of crutch-assisted gait patterns

3.2.1 Generation of Simulations

In order to study the predictions of crutch-assisted gait, different settings were tested in order to determine which were the optimal. To do so, one setting was changed at a time to study its consequences on the resulting motion.

Velocities

The forward velocity at which crutch users advance is much lower than non-assisted walkers, so it is important to study at which velocity the simulations should be performed to obtain the best results. In order to determine which velocities were optimal for simulating crutch-assisted gait, different trials were made with a healthy 2D model and all the predetermined settings, just changing the forward velocity. This was done both for full gait and half gait.

Full gait or Half gait

To study the difference between half gait and full gait, some simulations were performed with each configuration and then comparisons were made between the two groups. Knowing whether or not a half gait configuration is able to return good results is important, since it can save a lot of computation time due to needing to compute the variables only for half of the gait cycle.

Length of the Crutches

Finding optimal length of the crutches and seeing how changes in length can affect the resulting gait is also very important to ensure patients are not getting damage from using the crutches in a non-optimal way and, furthermore, it is one of the easiest parameters than can be changed in real life when dealing with crutch-assisted gait. Different trials were done scaling the crutches in the models. In order to determine the theoretical optimal length of the crutches, each model was measured to find the distance between the head of the ulna and the ground as the subject is in a standing still position with the arms relaxed at the sides. This distance should correspond to the length of the crutches from the handle to the tips [16]. However, the bodies corresponding to the crutch and the hand are joined such that the distance between the crutch handle and the position at which the handle would lie during gait, i.e., right under the carpal bones, is about 4 cm. To deal with this, we can directly subtract 4 cm to the length of the crutches in the simulation and, since it is a welded joint, the results should be the same as by changing the joint position.

Virtual Patients

In order to study the changes that having some kind of injury or condition that affects walking could cause on the results of the simulations, some trials were performed using virtual patients. These are healthy subject models that had some group of muscles debilitated in order to simulate a clinical condition. In the 2D trials performed, the groups of muscles that were debilitated were the ones involved with knee flexion and the ones involved with ankle flexion. The limitations on muscle force ranged from a 20% of the original value to an 80%. In the 3D simulations, the debilitated group of muscles was the same for the 4 trials: all the muscles included in the model were debilitated to half their default values, in order to simulate patients with general lower limb weakness.

Weights of the Cost Function Terms

Lastly, some very important settings to consider modifying are the weights assigned to the different terms in the cost function. These can be changed in order to give more or less weight to a given variable in the cost function, which will make the simulation give more or less priority to minimizing the term. Since we are studying assisted gait and the code was developed originally for non-assisted gait, it is possible that the body does not use the same criteria to choose the optimal motion, so it is important to investigate which weights are adequate for crutch-assisted walking.

3.2.2 Data Analysis

One of the most important tools used for data analysis during this project is the OpenSim GUI, since it allows for visualization of the resulting movements of the simulation. By uploading a model of the human body and a motion file containing time and relative angles of the joints, it recreates the movement of the full body in 3D. From OpenSim, the analysis tools Point Kinematics and Body Kinematics were also used on the results. These tools allow to obtain the position of any given point in the model and the orientation of any of the individual bodies.

With this information, together with the data that the simulation already provides us with, it was possible to develop some MATLAB functions that helped in the analysis. Multiple scripts have been created for different purposes, but some of them have been more significant in the study of the results:

- **Steps**: This code allows to easily see at which positions the subject is putting weight on the ground. It represents the position in the horizontal plane of both feet and crutches, while standing and swinging, but highlighting the difference between the two states and visually representing the amount of force exerted, to see where the subject was putting its weight.
- **GetFigures**: Developed to create plots of most of the variables stored in the simulation's output MATLAB data file: joint angles, torques, muscle activations, etc. It allows to compare the desired variable among as many simulations as wanted, plotting them together.
- **Torques**: This script analyzes the resulting torques of any simulation and returns important characteristics that can help analyze the result: maximum, minimum and standard deviation, for each joint.

In the beginning, most of the analysis was done from a very qualitative point of view. This is because, at first, the objective was to determine which settings resulted in a natural gait. By natural or physiological looking gait, we refer to those movements that do not imply very high or rapidly changing joint torques, and those which approximately follow any of the predefined gait patterns. For the first condition, comparison between the results was done, but the torques were also compared with external studies dealing with crutch-assisted gait. The second condition is checked using the visualizer tool in OpenSim and is mainly to avoid gait patterns that, even though might be optimal from a numerical point of view, they are not movements a person using crutches would normally do. These include making too many movements that seem unnecessary with the crutches, like raising them to be almost horizontal every time they swing forward, or even more unnatural gaits that can be easily identified, like jumping forward just propelled by their crutches and with flexed knees. As the project advances and more specific settings are being studied, resulting in slighter variations between results, the analysis of results focuses on comparison of variables using overlapping plots and includes quantitative analysis elements, such as comparing standard deviations of joint angles' evolution.

3.3 Ground Reaction Forces Tracking

3.3.1 Experimental Data Description

Access to experimental data from one subject was granted to us by the team of J. Aranda (co-supervisor of this thesis). These data included multiple measurements taken during a 10 m walk forward, turn around, and 10 m walk back to the starting point. The measurements were taken multiple times for each different gait pattern (2-point, 4-point, 3-point and swing-through) and, as previously commented, the subject was emulating a foot injury [4]. The most interesting variables for us were the crutches' orientations and the GRFs recorded at the tips of the crutches, since the angle of the crutches in the sagittal plane and at what times and with which magnitude the crutches touch the ground are good indicators of which gait pattern is being followed. The GRFs were recorded using a pressure sensor in the tips of the crutches, which means that the forces were measured in absolute value and no differentiation was done between horizontal and vertical forces. The orientations were measured in the crutch's reference system, so the angles provided by the measurements were not exactly the Euler angles in the ground reference system. However, since the crutches' did not experience a lot of axial rotation, the "pitch" angle that was recorded (corresponding to the sagittal plane of the crutch) should be representative of the crutches' angle in the sagittal plane of the musculoskeletal model. However, to avoid errors and, at the same time, to try to find the least needed inputs in order to reliably reconstruct the motions, it was decided that GRFs would be tracked and orientations of the crutches would be used for validation of results, since no videos nor more significant measurements from the experimental setup were available. From these measurements, one representative full gait cycle for each of the trials was selected to be used as a reference in the simulations.

3.3.2 Modifications to PredSim code

In order to do the tracking of the forces, the PredSim code was modified to add a new term to the objective function: one that was minimized as the resulting vertical GRFs from the simulations resembled more the reference values. Only the vertical GRFs of the simulations were used for comparison due to simplicity in the coding, assuming that the horizontal component would not be as relevant to finding the correct solution. By defining a new term in the cost function, we are not imposing the resulting GRFs to be exactly the same as the reference ones, or only allowing some error, but the simulation will try to make them as similar as possible without significantly sacrificing other important terms of the objective function, such as metabolic energy or muscle activation. By adding this term, there is a need to find its optimal weight, for which multiple simulations are ran with different force tracking term weights. In order to choose the optimal weight for the tracking forces term, different values were tried using the 2D model, to reduce computational time. The objective was to find for which weights the simulation was able to better replicate the reference GRFs.

3.3.3 Data Analysis

In order to analyze the results from the tracking of ground reaction forces, a script was also developed to compare the experimental and simulation values. To check if the GRFs were being correctly tracked, they are plotted together with the ones used as reference, and their cross correlation coefficient is computed, returning a value between 0 and 1 that can be interpreted as similarity: 0 being completely different and 1 being the same signal, even if shifted in time. Since we had access to the experimental orientations of the crutches in the gait patterns that we tried to reconstruct, these are also compared both qualitatively, by plotting them together, and quantitatively, by computing again the cross correlation coefficient.

Chapter 4

Results and Discussion

4.1 Predicting crutch-assisted gait patterns

4.1.1 Velocities

Most simulations with velocities higher than 0.6 m/s converged to patterns that seemed variations of the swing-through, where the subject was bearing almost its full weight in only one of the crutches while the feet motion resembled more that of running than walking, or where the swing-through was very asymmetrical, when it is supposed to be one of the most symmetrical patterns. These, unless the patient is a very athletic person, are very complicated gait patterns to follow, as they would require a significant amount of balance, coordination, and arm strength. For medium velocities, from 0.3 m/s to 0.6 m/s, most of the resulting motions looked natural, replicating more or less the predefined gait patterns. However, for velocities of 0.3 m/s and lower, the simulations took significantly longer to converge to an optimal solution: if we take into account all trials that converged with the 2D model (including those with modified settings, but not those with changed term weights), the mean iterations needed to converge for a velocity of 0.3 m/s or lower are 3241, while for 0.4 m/s this value goes down to 1748. For higher velocities, the number of iterations goes up again: for velocities of 0.5 m/s, it is of 2170, and for velocities of 0.6 m/s or higher, it is of 2070. Moreover, simulations with velocities of 0.3 m/s or lower tend to converge to solutions where the subject walks tip-toeing, swings the crutches more times than needed in any predefined pattern, or takes very short steps without significantly reducing joint velocities: characteristics that are not considered physiological. Regarding the final values of the cost function, no significant differences were observed between trials at different forward velocities.

Another interesting concept to study is the torque generated in the joints, and an interesting way of comparing results is to look at the standard deviation of the joint torques over time. If we compare all trials done with different velocities and the predetermined settings, including both full gait and half gait simulations, it can be observed that the mean standard deviation of the torques among all the joints raises slightly as the forward velocity increases (Fig. 4.1).

Just by qualitatively looking at the resulting motions, it was reasonable to determine that velocities greater than 0.6 m/s did not reliably lead to optimal results for crutch users. Furthermore, when comparing to the bibliography and experimental data, we see that focusing on the 0.3 to 0.6 m/s range of velocities is reasonable. Moreover, it is reasonable to try to minimize the standard deviation of the torques, since this would translate to a more stable and constant gait, without sudden movements or big changes. By integrating the two perspectives, and taking into account that time is limited and faster simulations would allow a better insight by providing more results, it was decided that the bulk of the simulations moving forward in the project would be done at a forward velocity of 0.4 m/s, with only a few trials with velocities of 0.3, 0.5 and 0.6 m/s. For the 3D simulations, almost all of the trials were performed using a forward velocity of 0.4 m/s.

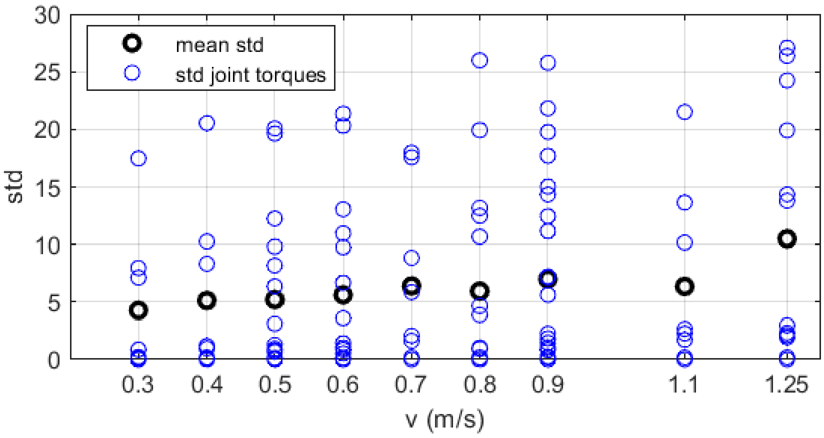

Figure 4.1: Representation of the standard deviations of the torques over time for the 16 different joints of the model (in blue) in simulations for different forward velocities. Means of all standard deviations are computed for each different velocity and represented in black.

4.1.2 Full gait or Half gait

2D model

As previously explained, simulations can be done using a "halfgait" or a "fullgait" configuration. For half gait simulations, it would be expected that only patterns with right-left symmetry would appear: 2-point, 4-point and swing-through. However, among the results that resembled any of the predefined crutch-assisted gait patterns or slight variations, only the 2-point and the swing-through could be observed using the 2D model, both for full gait and half gait. Although the same model with the same settings, just changing between half gait and full gait, did not reach a similar solution in most of the cases, no general tendency could be deducted from the results. However, non-physiological (with abnormally high cost function terms) "optimal" solutions with cost function values that are more than one order of magnitude greater than most other trials appear only for full gait configuration, if no other settings are modified.

One thing that does change when comparing the two configurations while only changing forward velocity and keeping the rest of the settings at their default values, is the mean number of iterations before convergence: for half gait it is 1183 and, for full gait, 1967. The value of the cost function for the optimal solution also changes depending on the setting: the mean unscaled objective function for full gaits is 1219.4 with a standard deviation (std) of 450.7, while it is 1517.4 with a std of 181.3 for half gaits (without taking into account outliers).

These values confirm that the half gait simulations, with its symmetry restrictions, take less time to be computed. This makes sense since the simulation does not need to compute all the variables for one full gait cycle, but it computes only half of it and assumes that, in the second half, the behavior of the right side will be identical to the behavior of the left side in the first half and vice versa. When taking a look at the objective function values, it seems that full gait is slightly more effective in finding the optimal solution, however, it is also more susceptible to simulating a non-physiological gait. With this information, it was determined that if the musculoskeletal model was symmetric, and if getting a 3-point gait pattern was not a priority, half gait simulations would be conducted since they are more cost-effective than full gait ones.

3D model

Regarding simulations done with the 3D model, only 4 out of 16 were performed using a full gait configuration. This is because the 3D simulations, even with a half gait configuration, require a long computing time, of multiple hours or even days, so changing to full gait implies a significant amount of time will be added to the simulations. Almost all 3D simulations but one converged to either a

2-point pattern or a delayed 2-point, where the crutches touch the ground somewhat earlier than the opposite foot. However, one of the simulations did converge to a variation of a 3-point pattern, where both crutches advance in unison each time a foot starts swinging. This only simulation that resulted in a different pattern was performed using a full gait configuration.

The different pattern is, as the 2-point, a symmetric movement that could, theoretically, have been obtained using a half gait configuration. Since not many trials were performed it is complicated to draw conclusions from this results. However, it seems possible that little asymmetries may influence the simulated gait pattern.

4.1.3 Length of the Crutches

2D model

For the 2D model, the theoretical optimal length of the crutches is 0.923 m. However, taking into account the distance from the crutches' origin point and the place at which the crutches' handles would actually lie, it is hypothesized that the optimal length to which the crutches in the simulation should be set is 0.883 m. This is very close to the predetermined length of the crutches, which is 0.875 m, so we decided to generate simulations using this value and scaling the length of the crutches by 0.95 and 1.05 to compare. With the 0.95 scaling, the resulting length of crutches is 0.831 m from handle to tip, and with the 1.05 scaling, it is 0.919 m.

All simulations led to a similar quality of results (understanding quality as more natural movements, without sudden changes or peaks in torques, and similarity to predefined gait patterns). The number of iterations was higher in the shorter crutches simulations: it was 1651 for the original length results, 2349 for the 0.831 m crutches, and 1455 for the 0.919 m ones. Moreover, variability can also be observed if looking at the objective function. The use of longer crutches led to a significant increase in the unscaled cost function: 2539.7 versus 1358.5 for the original crutches and 1129.2 for the shorter ones.

Since no big difference was observed between the original length, which is very similar to the theoretical optimal length, and a scaling of 0.95, it was decided to use the default length for the rest of simulations in 2D. The results seem to be consistent with the recommendations from bibliography.

3D model

When using the 3D model to perform the simulations, the length of the crutches has to be scaled accordingly. In the 3D model we use, the distance between the ground and the head of the ulna bone is of 0.824 m. When subtracting the 4 cm of correction to this value, we observe that the distance that should be set in the simulations between the handle and the tip of the crutches is of approximately 0.784 m. Since by looking at 2D results it seemed that this theoretical length or slightly shorted led to better results, most of the simulations were performed using lengths of 0.78 m (almost the theoretical length) and 0.75 m, which corresponds to a scaling of approximately 0.96. A couple of simulations were performed using 0.805 m crutches (which would correspond to a scaling of about 1.03), however, the simulations led to motions where the tips of the crutches were afar from the body in a seemingly unnatural way, as can be seen in Fig. 4.2. Furthermore, even if the difference was not not very high, when looking at the mean through all joints of the standard deviation of torques through time, it could be observed that longer crutches led to higher deviations: 4.35 for 0.805 m crutches, 3.82 for 0.78 m, and 3.54 for 0.75 m.

Both lengths led to very similar values of the objective function (257 for the 0.75 m crutches and 251.1 for the 0.78 m ones). However, using shorter crutches ultimately led to longer computing time, with a mean of 7582 iterations compared to the 4952 iterations of the trials with the longer crutches. All the resulting motions looked natural and all except for one converged to a 2-point or delayed 2-point gait pattern, so we thought it would be interesting to compare the GRFs generated by the movements with the different crutch lengths, but no significant difference could be observed. The mean weight, as a percentage of body weight, beared by the crutches as a whole was of 11.8 % for the 0.75 m crutches and 10.7 % for the 0.78 m ones, with the weight evenly distributed between

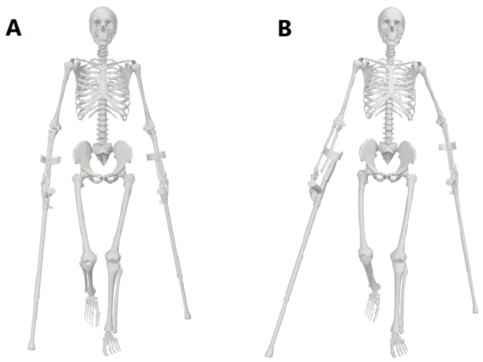

Figure 4.2: Screenshots of the model performing the simulated motion from approximately the same moment in the gait cycle for a 2-point gait pattern for the 0.75m crutches at the left (A) and the 0.805 crutches at the right (B).

the two crutches in the full gait trials (in the half gait trials, it is implicit that the same weight will be beared at both crutches).

Since it was not possible to do a large number of simulations using the 3D model, it is not reasonable to draw definitive conclusions from these trials. However, it can be observed that both 0.78 m, the theoretical optimal length, and 0.75 m, slightly shorter crutches, lead to physiological results. On the other hand, longer crutches were discarded for more simulations due to its rise in the torques standard deviation and the unnatural appearance of the resulting motions.

4.1.4 Virtual Patients

2D model

When working with virtual patients, some of the results seemed to correctly represent what would be expected when a group of muscles is debilitated: there were some trials (4 out of 25) where the debilitation of either the knee or the ankle led to significantly lower GRFs in the affected leg than in the healthy one (differences of more than 10% of the body weight between feet). Some trials debilitating one side resulted in higher contact forces in the injured leg, however, the difference of GRFs in the feet did not reach 10 % in any trial. In general, no difference was observed on the total weight beared by the feet of the subject between virtual patients and healthy subjects: it was of 76 % of the body weight for both groups.

Overall, there was not a consistency of results, as more debilitated subjects would sometimes lead to motions that almost do not bear weight on the crutches, but less debilitated subjects would sometimes lead to motions where the crutches were much more used for support.

3D model

In the case of 3D virtual patients, which had the lower limbs debilitated to 50% of their capacity, it can be observed that the GRFs produced in the feet decrease when weakness is simulated: while for healthy subject the mean GRFs at the feet are approximately 91 % of the body weight, for the virtual patients this value goes down to 83 %. All simulations led to a 2-point or a delayed 2-point gait pattern but one, which is part of the virtual patients group. This different pattern corresponds to a simulation using a full gait configuration and debilitating the muscles as described.

It is important to note that most of the simulations were generated using a healthy subject model, so it is natural to think that the optimal gait for them would not imply a heavy use of the crutches, since it has been studied that energy consumption and, in a smaller measure, muscle activation, increase when healthy subjects use crutches [7]. However, carrying the crutches without them touching the ground would probably imply a significant increase in muscle work done by the arms, so it would make sense that simulations with a healthy subject converged to a 2-point pattern, since it is the one that, in general, offers less support, while allowing a wider range of velocities, as can be seen in trekking with poles [11, 12]. Although the 2D simulations do not quite back-up this assumption, it seems to be correct for the 3D simulations. Furthermore, a different pattern than the 2-point only appeared when debilitating the model. The appearance of this different pattern could be favored by the use of the full gait configuration, however, it is interesting to take into account the influence of debilitating muscles, since no full gait trials with healthy subjects led to different patterns.

4.1.5 Weights of the Cost Function Terms

2D model

At first, the comparison was done between the results in which the subject was walking normally and laying almost no weight on the crutches, but the terms of the cost function did not change much: the metabolic energy and limit torques terms stayed at almost the same value, while actuators excitation showed a decrease of approximately 13 %, joint accelerations decreased around 15 %, and muscle activations rose a 14 %.

Then, nine selected results which satisfied the criteria to be considered good (natural, relatively smooth movements and similarity to one of the predefined patterns) were selected from all the trials. When comparing the values of the cost function terms between the selected group and the all the trials, more variability could be observed. First of all, all terms had lower values for the "good" trials, with an overall decrease of the objective function of 20 %. Metabolic energy and joints accelerations terms only decreased in approximately 15 %, however, the muscle activations and limit torques terms decreased around 25 %, and the actuators excitation term decreased slightly more than 30 %. To check if forcing the actuators excitation term to be lower would overall improve the results, some trials were done using higher weights for the term. While the default weight is of 10^6, a value of 10^8 was tried. The simulations, however, did not lead no more better results. While it is true that in most of the trials the value of the actuators excitation term decreased, it increased for all other terms. Moreover, when visualizing the resulting motions they appeared non-physiologic, displaying behaviors like tip-toeing or stomping the feet on the ground instead of landing them smoothly.

When comparing "normal" walking patterns with the ones that use the crutches, reduction in actuators excitation makes sense since the arms are not as important if no weight is beared in the crutches. The increase in muscle activation might be explained with the same reason, the muscles are located mainly at the legs, so if the arms, which are actuated with torque generators, do not contribute as much to the movement, the muscles in the legs would need to exert more work. When the "good" results are compared with all the trials together, it can be seen that low values of the terms muscle activations, limit torques, and actuator torques, are good indicators of a physiological result. However, it could also be seen that it does not make much sense to try to impose a reduction in one of the terms, since it seems to make the others rise, together with the total value of the objective function.

3D model

Just by changing the model from 2D to 3D, the simulations started returning only physiological results that resembled the predefined crutch-assisted gait patterns and had smooth and natural looking movements. When looking at the terms of the cost function, it could be observed that, with the default weights, the simulations replicated somehow the the behavior of the selected "good" 2D results. 3D trials showed a reduction of approximately 90 % in the metabolic energy, joints accelerations and actuators excitation terms, of almost 70 % in the limit torques term, and of about 15 % for the muscle activations.

Once the first simulations in 3D were performed, we could observe that, actually, the weights did not need to be modified. Although the muscle activations did not decrease as much as the other terms, the new results seemed to replicate human behavior with a much better accuracy than 2D results, even the selected "good" ones. This brings us to the conclusion that the default weights, which were designed thinking about non-assisted walking, can also lead to quality walking predictions when crutches are used without any major changes needed.

4.1.6 General Discussion

Multiple simulations were performed and studied during the development of this first part of the project, for which a basic classification can be seen in Table 4.1. In order to achieve good results for the simulation, different parameters have been observed to be important. It is also important to take into account that the majority of the tests were done using a 2D model, and it is not possible to affirm that the conclusions can be extrapolated to 3D simulations. However, the length of the crutches could be studied both in 2D and 3D, resulting in better results, in both cases, for the theoretical optimal length and slightly shorter crutches. Moreover, we observed that the optimal range of velocities determined with the 2D trials (0.4 to 0.6 m/s) was consistent with bibliography and, besides, it led to physiological results when applied to 3D simulations. Changes between half gait and full gait configurations should be further studied to determine exactly how it influences the optimal solution. Also, more simulations should be generated using specifically the 3D model, studying both changes in forward velocity and different virtual patients, in order to see if the simulations are able to predict different gait patterns, since the only ones that were predicted were the 2-point and delayed 2-point, and a variation of the 3-point, where both crutches advance each time the subject raises any foot from the ground, instead of swinging them only with one foot.

In general, the results were much better when the simulations were performed using a 3D model than when using a 2D one. This can be quantitatively seen with the huge decrease in the mean of the solutions' objective function, of slightly more than 85 %: from 1764.2 in the 2D trials to 251.4 for the 3D solutions. This hints that crutch-assisted walking has an important part of the movement happening outside the sagittal plane. This makes special sense when thinking about patterns where the crutches do not swing in unison, since the subject is supporting its weight on one crutch while swinging the other. Since the length of the crutches does not allow the arm to be extended down at the side of the body without the crutch touching the ground, it would make sense for the subject to slightly separate the crutch from the body as it swings, so that the arm does not need to flex as much to keep it from touching the ground. This separation can be achieved allowing arm adduction or lumbar bending, movements that are not allowed in the 2D model, but probably help lead to the good results obtained in 3D.

However, one big limitation was identified in these results. The PredSim simulation framework allows to set a lower and upper bound for the time it takes the subject to perform one gait cycle. By using the default settings, the upper bound is 2 s. When studying the results, it can be seen that most simulations reach this upper bound, meaning that, probably, the optimal solution would increase even more the time for one gait cycle. This is not important for faster gait patters, such as the swing-through, but is specially important for slower gait patterns, like the 4-point. By limiting the gait time in patterns that require more movements in one cycle, like in the 4-point pattern, the results might display shorter steps than natural, or directly not converge to this patterns and instead find that another one is more optimal to fit within this temporal bounds. In the M Narvaez Dorado *et al* study, it was found from experimental data that, while 2-point and swing-through patterns have a mean stride time lower than 2 s (1.79 s and 1.59 s, respectively), the same can not be said about 4-point (2.53 s) and 3-point (2.05 s) [4]. By the end of the project, the code was adapted accordingly, adding the temporal bound as an input in the main code, and provisionally setting it to 5 s, since the four simulations (not included in the analysis) generated with higher bounds converged to have a shorter gait time. However, setting the bound higher also lead, in one case, to the simulation identifying 6 steps (instead of the normal 2) as one full gait cycle, so it is important to further study the relation between this time bound and the results, in order to determine the optimal setting value. Even though the code was eventually modified, among the all the analyzed results, 58 out of 98 reached this maximum value for the full gait cycle time, so further exploration must be done. With the four trials, it could be observed that lower velocities (0.2 and

0.3 m/s) do seem to lead to better results when combined with a greater upper gait cycle temporal bound: the steps get longer, and the movements seem more natural than in the simulations with identical settings but a lower temporal bound.

	2D model							
	Converged	**Not converged**	**Non physio-logical**	**No use of crutches**	**∼2P**	**∼4P**	**∼3P**	**∼ST**
Total	84	28	22	31	26	0	3	9
Without changing weights	53	21	14	18	8	0	3	9
Changing weights	31	7	8	13	18	0	0	0
Vel <0.4	7	1	2	4	1	0	1	0
Vel 0.4 - 0.5	50	20	13	23	23	0	1	6
Vel >0.5	27	6	7	6	2	0	1	4
Half gait	49	9	12	18	23	0	0	0
Full gait	35	19	10	14	3	0	3	10
Virtual patients	26	16	7	13	3	0	3	3
Healthy subjects	58	12	15	18	23	0	0	6
Shorter crutches	6	0	1	1	4	0	0	0
Normal crutches	75	26	19	29	22	0	3	9
Longer crutches	3	2	2	1	0	0	0	0

	3D model							
	Converged	**Not converged**	**Non physio-logical**	**No use of crutches**	**∼2P**	**∼4P**	**∼3P**	**∼ST**
Total	14	2	1	1	8	4	1	0
Without changing weights	10	2	0	1	5	4	1	0
Changing weights	4	0	1	0	3	0	0	0
Half gait	11	1	1	1	7	3	0	0
Full gait	3	1	0	0	1	1	1	0
Virtual patients	5	1	0	0	3	1	1	0
Healthy subjects	9	1	1	1	5	3	0	0
Shorter crutches	5	1	0	0	3	1	1	0
Normal crutches	6	0	1	0	4	1	0	0
Longer crutches	3	1	0	1	1	2	0	0

Table 4.1: Tables containing the number of simulation results generated with the 2D model (upper) and 3D model (lower), classified in different groups (columns) for different settings (rows). The classification groups that are featured are: converged (simulations that converged within a reasonable number of iterations), not converged, non-physiological (in which are classified very asymmetric movements, unnecessarily exaggerated movements, tip-toeing, etc.), no use of crutches (meaning that the mean force supported by the crutches does not arrive to 5% of the body weight of the subject), ∼2P (similar to 2-point pattern), ∼4P (similar to 4-point), ∼3P (similar to 3-point) and ∼ST (similar to swing-through).

4.2 Reconstructing gait by tracking Ground Reaction Forces

Our second objective for this project, as commented before, is to determine whether or not it is possible to reconstruct crutch-assisted gait movements just by tracking experimental GRFs. To generate the predictions, the 2D model with the theoretical optimal crutch length (equal to the distance between ground and head of the ulna minus the correction) and the 3D model with slightly shorter crutches (scale of 0.96) were used.

4.2.1 Weight of the Tracking Forces Term

Different weights were tried for the tracking forces term, as it can be seen in Fig. 4.3. In order to determine which weight led to a better tracking of the GRFs, the cross correlation coefficient was calculated for each trial. The weights that led to better results were 0.5 and 0.6, as they display very high correlation values for both crutches: for a weight of 0.5, the combined cross correlation taking into account both crutches is 0.981, while for a weight of 0.6 it is 0.986. For lower weights, the similarity between simulation and reference decreases, and so does the cross correlation coefficient, returning values under 0.9 (0.847 for a weight of 0.1 and 0.533 for 0.01). When the weight of the tracking term is increased to more than 0.6, the similarity between signals and therefore, the cross correlation, decreases again, reaching a value of 0.646 for a weight of 0.7 and 0.668 for a weight of 1.

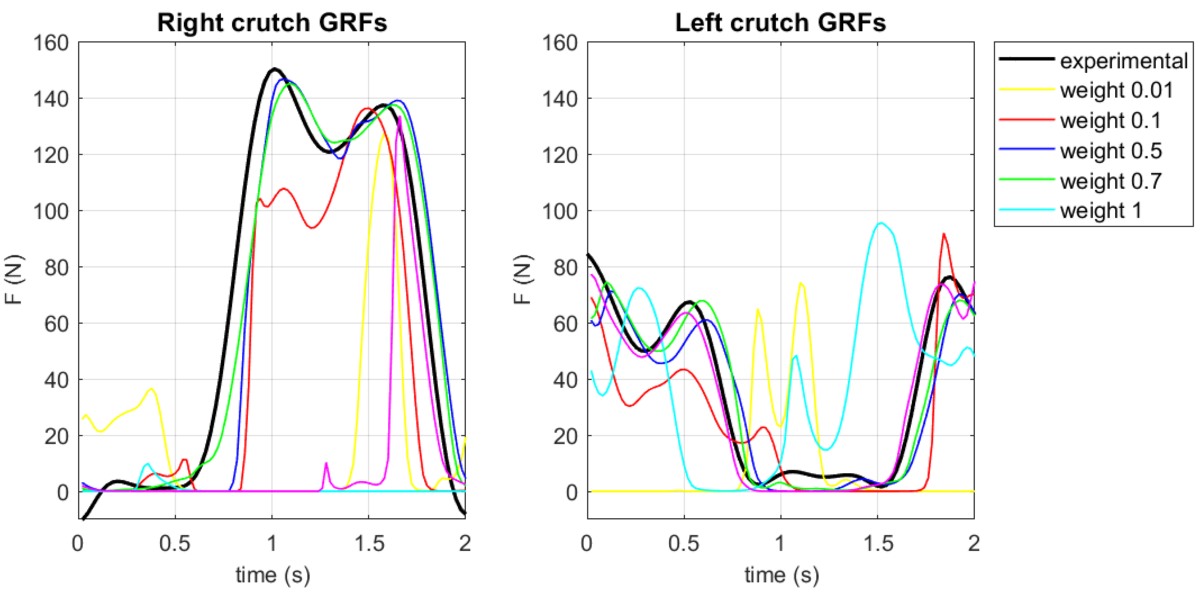

Figure 4.3: Representation of the reference crutch GRFs (in black, labeled experimental) together with the simulated GRFs (colored) for different weights of the tracking forces term.

4.2.2 2D model

Since 0.5 and 0.6 were observed to be the best values for the force tracking term for the 2-point experimental trial, we decided to compare their results generating one trial for each for the remaining gait patterns: 4-point, 3-point and swing-through. When comparing the resulting GRFs, it could be observed that, although for the 2-point pattern the weight of 0.6 led to a slightly higher correlation, for the other patterns that were tried the best results were obtained with the 0.5 weight, as can be seen in Table 4.2 and Fig. 4.4.

By looking at the resulting motions for the 0.5 weight in OpenSim, it can be seen that, even though the GRFs might be very similar, it does not mean that the movement is the one that we would expect. For the 3-point trial, for example, the subject keeps one crutch ahead of them all the time, while the other taps the ground ahead of the body when the opposite foot touches the ground but immediately goes back to behind the subject, where the force is exerted. On the other hand,

Patterns	2-point	4-point	3-point	swing-through
Cross-correlation (Weight 0.5)	0.981	0.991	0.976	0.941
Cross-correlation (Weight 0.6)	0.986	0.809	0.974	0.834

Table 4.2: Values of the total cross correlation between the reference GRFs and the results from the simulation tracking forces, for weights of the term of 0.5 and 0.6.

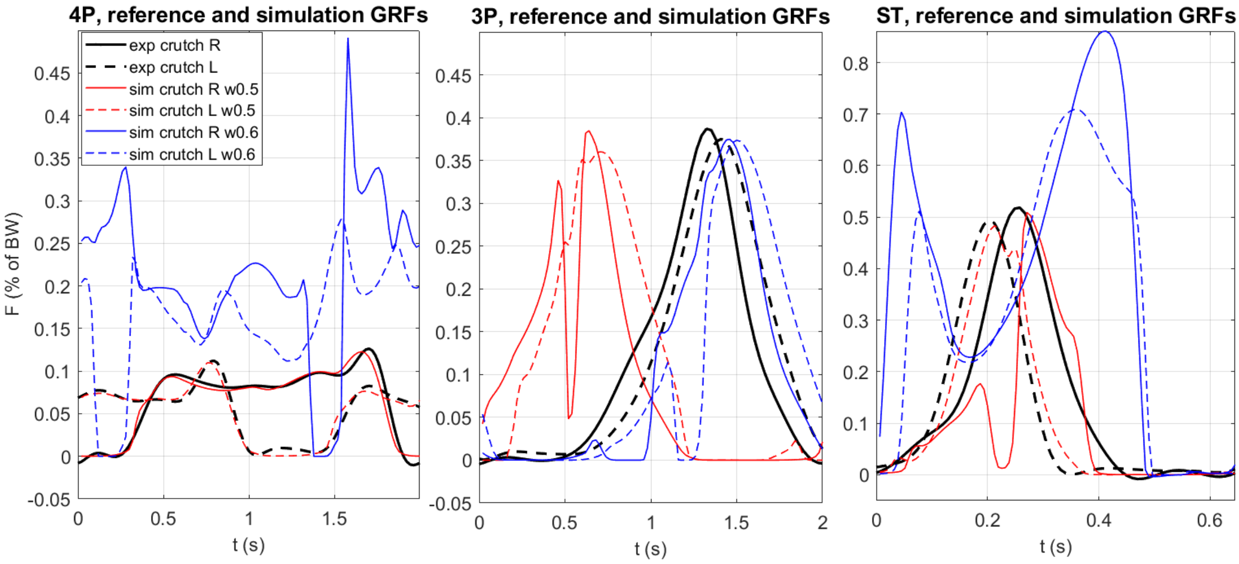

Figure 4.4: Representation of the reference crutch GRFs (in black, labeled exp) together with the simulated GRFs, for weights of the tracking term of 0.5 (blue) and 0.6 (red) and different gait patterns: 4-point (4P), 3-point (3P) and swing-through (ST), from left to right.

the swing-through trial clearly converged to a 3-point pattern. For the 4-point trial, the subject can be seen performing a pattern similar to the defined 4-point, however, it advances the opposite crutch before swinging one foot, instead of swinging the one on the same side, resembling more a delayed 2-point. Finally, when looking at the 2-point pattern, we see that the subject does perform the expected pattern, but with a slight variation, since the right crutch advances simultaneously with the opposite foot, but the left crutch advances and touches the ground slightly before than the right foot. This pattern actually resembles the movements a person would do when not a lot of weight can be carried on the left foot, since it reduces the time where only the left foot and one crutch are support points. Despite that, when looking at the simulations' feet GRFs, not a lot of difference can be observed: only about a 6 % less of the body weight is carried in the left foot compared to the right. However, when all the patterns are observed, it can be seen that, for a weight of 0.5, there is consistently less weight carried in the left foot, especially in the 3-point and swing-through, where the difference is of about 15 % of the body weight, and less obviously in the 4-point, where the difference is of only about 3 %.

However, some of the differences between the simulated motions and the predefined patterns might be due to the experimental trial, since no video reference is available to confirm exactly how the gait was performed. In order to check if the simulations are actually correctly predicting the experimental gait, it is interesting to compare the orientations of the crutches between the simulations and the experimental values. By doing so, we see that, even though the experimental and simulated orientations follow similar patterns, they are quite different, as can be appreciated in Fig. 4.5. When cross correlation coefficients are calculated for each pattern, we can see that the most similar orientations correspond to the 4-point and swing-through patterns, with values of 0.542 and 0.570, respectively. On the other hand, 3-point got a score of 0.382 and 2-point got 0.278.

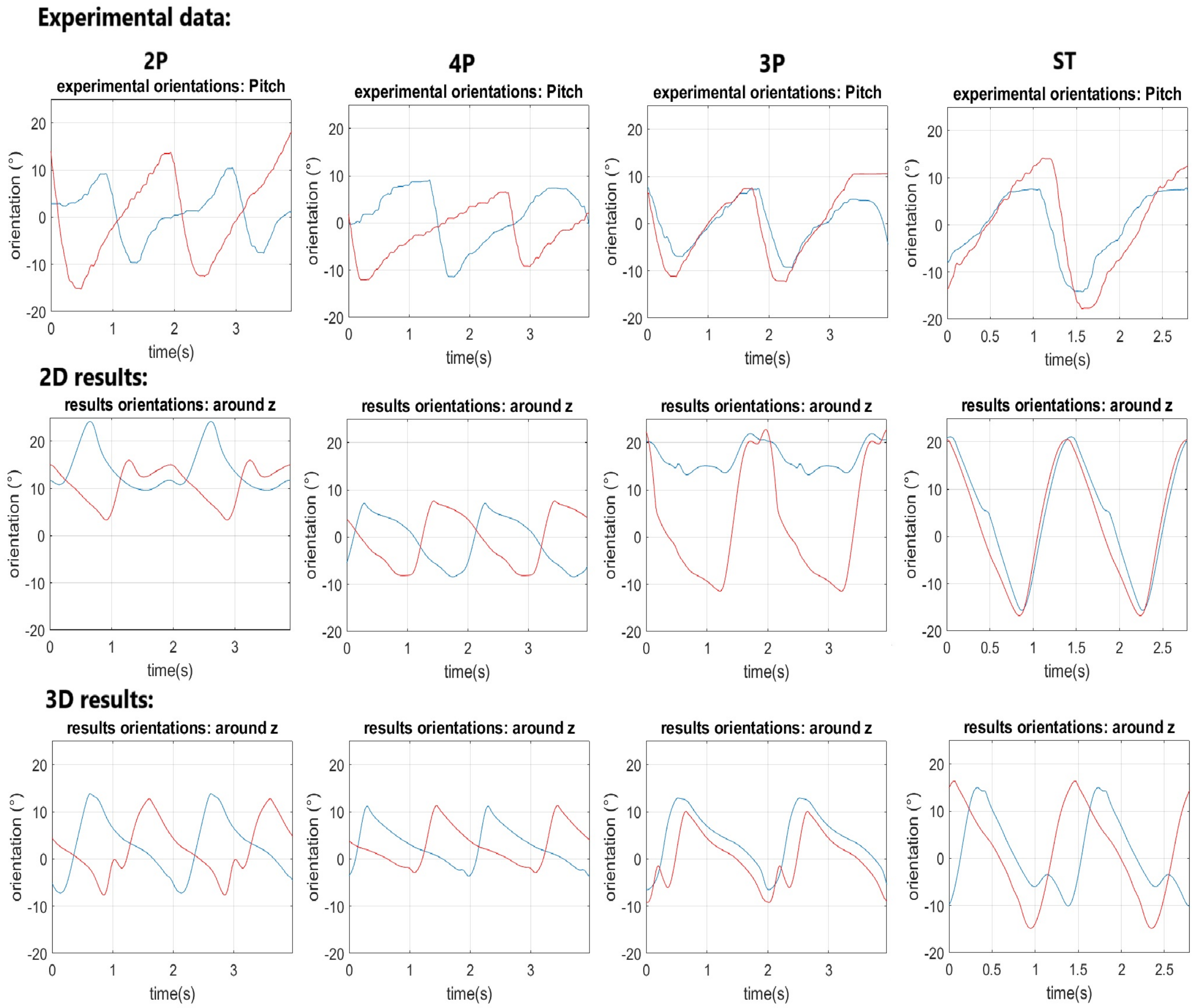

Figure 4.5: Representation of the measured orientations in the sagittal plane of the crutches in the experiment (top row) with the simulated orientations in the sagittal plane of the subject for 2D (2nd row) and 3D (3rd row) models. For a weight of the tracking term of 0.5 and featuring the different gait patterns: 2-point, 4-point, 3-point and swing-through, from left to right.

4.2.3 3D model

When using a 3D model, the similarity between the reference GRFs and the simulated ones improves. When looking at cross correlation values, we see that the lowest is for the swing-through pattern, just like in 2D, with a value of 0.962. 2-point and 4-point patterns got a score of 0.999 and 3-point got a score of approximately 1, meaning that the tracking was almost perfect. When the maximum GRFs of the feet are compared, it can also be observed that, in general, the left foot bears less weight than the right one. This is correct for 2-point (4 % less of weight on the left foot), for 4-point (12 % less) and specially for 3-point (32 % less), but for the swing-through, a different behavior is observed: the left foot actually bears up to 25 % more weight than the right foot.

When looking at the resulting patterns, despite the accuracy of the GRF tracking, we see that they are not as accurate as it might be expected: the 4-point pattern converged to a delayed 2 point and the swing-through converged to a mix between swing-through and 3-point, where the subject was lifting both feet from the ground but more in a running motion than in the predefined swing-through. However, the 2-point trial did converge to a perfect 2-point pattern, and the 3-point trial converged to an almost theoretical 3-point with the left as the injured leg, with the exception that, in the simulation, the left crutch discretely taps the ground once during the swing forward. These observations are consistent with the results of comparing the orientations: for the best predicted patterns, the cross correlations between simulated and experimental orientations were higher (0.546 for 2-point and 0.537 for 3-point), while they decreased for the swing-through (0.477) and even more for the 4-point (0.384).

4.2.4 General Discussion

By looking at the results, one can hypothesize that the leg in which the subject was faking the injury is the left, however, more measurements of the subject performing the pattern that could help determine the movement they were doing would be very useful for further testing. Also, due to the big difference between experimental and simulated crutch orientations compared to GRFs, it is suspected that tracking also experimental orientations could lead to much more accurate reconstructions of gait patterns. In terms of replicating a the correct gait pattern, the 2-point simulations led to the more accurate results, however, when looking at the similarity of orientations, the most similar overall was for the swing-through in 2D, which actually converged to a 3-point. This can also mean that orientations might not be enough either to correctly reconstruct all gait patterns, since orientations between patterns where the crutches follow the same behavior might be extremely similar: for example, swing through might get confused with 3-point, since the crutches advance simultaneously in both. Both in 2-point and 4-point the crutches alternate, however, since the separate GRFs of the crutches of the 4-point pattern superpose much more than in the 2-point (offering extra support), it is reasonable to think that the simulations can differentiate between the two when reconstructing the pattern. The reason why the predictions seem to confuse 4-point and delayed 2-point, however, might be that delayed 2-point lengthens the time of the crutches touching the ground, resembling more the 4-point GRFs.

Some low cross correlations for the orientations, such as the one for 3-point in 2D, can be easily understood, since the pattern simulated did not replicate a 3-point, however, it is strange that some trials whose patterns actually replicated the experimental trials also display very low values, such as the 2-point trial for 2D. An important thing to take into account, as we can also see that some GRFs are not tracked perfectly, is that the simulation is trying to replicate the experimental GRFs obtained with force sensors, which include forces in all directions, by using only the vertical GRFs of the simulation. This means that the vertical component of the GRFs, which is the only one used for tracking, should actually be smaller than the reference ones. To deal with this issue, the best option is to take into account all GRFs of the simulation in order to do the tracking. However, since the cost function terms are calculated using external Casadi functions, it is more complicated than just tracking one component. Adding the horizontal forces would probably, however, lead to better reconstructions of gait patterns. Another very important limitation to take into account, as commented in the discussion of the first part of the project, is that the gait cycles are limited to last less than 2 s each, which might avoid the simulations from predicting the correct gaits. As previously discussed, this might affect specially slow patterns like the 4-point. However, as can be seen in Fig. 4.5, all patterns but the swing-through end up with the maximum time for a gait cycle (each plot displays 2 full gait cycles, and 2-point, 4-point, and 3-point all show a finish time of almost 4 s). In order to see how this affects the tracking of forces, the trials should be repeated allowing more freedom for the gait cycle time.

Conclusions and recommendations

During this project, it could be seen that simulating crutch-assisted gait was possible without imposing the pattern that the subject would be doing, just by having an accurate musculoskeletal model of the subject and setting the forward velocity to the desired value. This opens the door to further personalization of both rehabilitation plans, for people who suffered an injury and must use crutches for a definite period of time, and assisted locomotion, for people with a locomotive impairment who will use crutches for, maybe, their whole life. This last group could be extremely benefited from studies in this subject, since they are going to be using crutches for long periods of time (years), meaning that an understanding of crutch-assisted walking is as important for them as the study of non-assisted gait for people who do not need assistive devices. In order to avoid possible secondary effects from using crutches for a long period on time, the biomechanics need to be thoroughly studied. By having a deeper understanding of how different patterns affect the musculoskeletal system of a patient, problems like joint or muscle overload, which can ultimately lead to structural damage of the tissues [51, 52], could be avoided or, at least, mitigated.

Multiple aspects that ultimately narrow the path to optimal crutch-assisted gait prediction could be observed in this project. First of all, and probably most importantly, performing the simulations in 3D, i.e. allowing movement of the joints outside the sagittal or parasagittal plane, is extremely important for the simulations to be physiological. When comparing the resulting motions of 2D and 3D simulations, it can be clearly seen that, although most of the 2D results did not resemble any predefined pattern, or unexpected movements were observed (like jumping with the feet without using the crutches, or tip-toeing), all the 3D simulations converged to a physiological gait pattern that, besides, resembled the predefined crutch-assisted gait patterns.

The fact that almost all simulations in 3D converged to a 2-point gait pattern could be partially explained by the fact that the model used was, in the majority of the cases, healthy. Since the use of crutches has been demonstrated to increase the effort done by a healthy subject to walk [7], it is reasonable to assume that the simulation will "choose" the gait pattern which differs the least from non-assisted walking. In the rest of the trials, the maximum force of all the muscles was set to be half of the original, meaning that the debilitation is symmetrical. 3-point and swing-through patterns are more recommended for people with an asymmetric condition. However, it should be further studied how to accurately represent a real injury in the musculoskeletal model, since in this project the virtual patients were obtained by simply debilitating full groups of muscles. The simulation framework allows to also modify tendon stiffnesses, which would probably help to better represent the real condition.

Furthermore, it has to be taken into account that all but one of the 3D simulations were performed setting a forward velocity of 0.4 m/s, due to the results obtained in 2D. This velocity would be reasonable for 2-point and 3-point patterns, but is significantly higher than the velocities reported for a 4-point pattern (0.25 m/s) and significantly lower than the ones for swing-through (0.7 m/s). In order to get more gait patterns, performing multiple trials with different velocities could be useful. Finally, although the theoretical optimal crutch length seems to work correctly, and so does a slightly shorter length, it would be interesting to perform more trials with different crutch lengths, in order to see if this is consistent for all gait patterns or if the different patterns have different optimal lengths.

The bounds in gait cycle time must be taken into account and further studied for future work. A 2 s upper bound for gait cycle makes sense for normal walking, however, in crutch-assisted walking, this value may fall short. By allowing more freedom to this variable, it is possible that gait patterns that barely appear in the simulations made (like the 4-point) would also be predicted. This might also be a reason why, for lower velocities, the simulations took significantly longer to converge and did not usually lead to good results: by setting a low velocity while having this time constraint for the gait cycle, the subject would try to advance slowly, while still performing at least one gait cycle every 2 seconds. In order to fulfill this condition, the simulation would make the steps abnormally short. When looking at the few simulations performed with a higher gait cycle time upper bound and low forward velocities it can be observed that, indeed, steps get longer.

In conclusion, in relation with the first objective of the project, it is safe to say that, even with the identified limitations, predictions of crutch-assisted walking can be performed using the PredSim simulation framework, leading to results that seem physiological and depict predefined crutch-assisted gait patterns or slight variations. Furthermore, it was observed that the cost function used to solve the simulation (through the optimal control problem formulation), did not need to be changed from the original, which was designed and observed effective for predicting non-assisted gait. This shows that the human body probably uses the same criteria to choose which is the optimal gait in assisted and non-assisted gait, opening the possibility to use this same simulation framework for predicting gait with assistive devices other than crutches.

Regarding the second objective of the project, it was seen that tracking the ground reaction forces of an experimental trial was possible by adding a tracking term to the cost function of the PredSim simulation framework. Although the simulations tracked the GRFs of the experimental trials with considerable accuracy, significant error could be observed in some of the trials. The tracking could be improved by comparing the norm of all the GRFs of the simulation, instead of just the vertical forces, with the reference GRFs. However, to do so, the formulation of the Optimal Control Problem should be again modified. Since the terms of the objective function are computed in external function and not in MATLAB itself, this is not so direct, which is why it was not included in this project yet.

It could also be observed that, even when the GRFs were accurately tracked, it did not mean that the orientations of the crutches were correctly replicated. This could be due to the restriction in gait cycle time, which would avoid some patterns from being chosen as the optimal solution. In order to either confirm or discard this hypothesis, more trials should be performed with the same tracking parameters, just changing the upper bound on the time of one gait cycle. Another possibility would be that the predictive simulations are not able to distinguish between GRFs of different patterns, however, even though this could somehow explain confusion between 3-point and swing-through, it would be less probable for 2-point and 4-point.

In conclusion, even though it was possible to track GRFs with significant accuracy, the results did not reliably replicate the experimental gait pattern. In order to improve the results, few modifications should probably be done to the code. First of all, the upper bound on the time for one gait cycle should be increased. Then, horizontal GRFs should be added to the cost function so that the experimental values, which do not differentiate between vertical and horizontal forces, are compared with the simulations' GRFs norm instead of only the vertical component. The last proposal would be to also add a term for tracking the orientations of the crutches in order to better predict the gait patterns. However, to do so, new experimental data would probably be needed in order to be able to validate our results.

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