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Λαμία, 2025

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UNIVERSITY OF THESSALY

SCHOOL OF SCIENCE

DEPARTMENT OF COMPUTER SCIENCE AND BIOMEDICAL INFORMATICS

DEVELOPMENT OF A SIMULATION MODEL OF A SEMI-ACTIVE ANKLE-FOOT ORTHOSIS SYSTEM

Iliana Platona

FINAL THESIS

ASSISTANT PROFESSOR

Vartholomeos Panagiotis

Professor

Lamia, 2025

«Με ατομική μου ευθύνη και γνωρίζοντας τις κυρώσεις [(1)], που προβλέπονται από της διατάξεις της παρ. 6 του άρθρου 22 του Ν. 1599/1986, δηλώνω ότι:

*1. Δεν παραθέτω κομμάτια βιβλίων ή άρθρων ή εργασιών άλλων αυτολεξεί **χωρίς να τα περικλείω σε εισαγωγικά** και χωρίς να αναφέρω το συγγραφέα, τη χρονολογία, τη σελίδα. Η αυτολεξεί παράθεση χωρίς εισαγωγικά χωρίς αναφορά στην πηγή, είναι λογοκλοπή. Πέραν της αυτολεξεί παράθεσης, λογοκλοπή θεωρείται και η παράφραση εδαφίων από έργα άλλων, συμπεριλαμβανομένων και έργων συμφοιτητών μου, καθώς και η παράθεση στοιχείων που άλλοι συνέλεξαν ή επεξεργάσθηκαν, χωρίς αναφορά στην πηγή. Αναφέρω πάντοτε με πληρότητα την πηγή κάτω από τον πίνακα ή σχέδιο, όπως στα παραθέματα.*

*2. Δέχομαι ότι η αυτολεξεί **παράθεση χωρίς εισαγωγικά**, ακόμα κι αν συνοδεύεται από αναφορά στην πηγή σε κάποιο άλλο σημείο του κειμένου ή στο τέλος του, είναι αντιγραφή. Η αναφορά στην πηγή στο τέλος π.χ. μιας παραγράφου ή μιας σελίδας, δεν δικαιολογεί συρραφή εδαφίων έργου άλλου συγγραφέα, έστω και παραφρασμένων, και παρουσίασή τους ως δική μου εργασία.*

3. Δέχομαι ότι υπάρχει επίσης περιορισμός στο μέγεθος και στη συχνότητα των παραθεμάτων που μπορώ να εντάξω στην εργασία μου εντός εισαγωγικών. Κάθε μεγάλο παράθεμα (π.χ. σε πίνακα ή πλαίσιο, κλπ), προϋποθέτει ειδικές ρυθμίσεις, και όταν δημοσιεύεται προϋποθέτει την άδεια του συγγραφέα ή του εκδότη. Το ίδιο και οι πίνακες και τα σχέδια.

4. Δέχομαι όλες τις συνέπειες σε περίπτωση λογοκλοπής ή αντιγραφής.

Ημερομηνία: 24/02/2025

Η Δηλ.

Πλάτωνα Πιλάνα

(1) «Όποιος εν γνώσει του δηλώνει ψευδή γεγονότα ή αρνείται ή αποκρύπτει τα αληθινά με έγγραφη υπεύθυνη δήλωση
του άρθρου 8 παρ. 4 Ν. 1599/1986 τιμωρείται με φυλάκιση τουλάχιστον τριών μηνών. Εάν ο υπαίτιος αυτών των πράξεων
σκόπευε να προσπορίσει στον εαυτόν του ή σε άλλον περιουσιακό όφελος βλάπτοντας τρίτον ή σκόπευε να βλάψει άλλον, τιμωρείται με κάθειρξη μέχρι 10 ετών.»

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ΗΛΙΑΝΑ ΠΛΑΤΩΝΑ

Τριμελής Επιτροπή:

Βαρθολομαίος Παναγιώτης, Επίκουρος Καθηγητής

Ιακωβίδης Δημήτριος, Καθηγητής

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ABSTRACT

Computational biomechanics models play a key role in understanding the complexity of the human musculoskeletal system and its response to various pathologies. This study utilizes the OpenSim computational environment to simulate and analyze the biomechanical effects of altered muscle stimulation and activation. These tweekings of muscle functionality are made to render abnormal function, such as those due to neurological disorders (e.g. foot drop). Specifically, the research examines how muscle activation is affected as a function of the input signal (muscle excitation) and how deviations in hip flexion, knee angle and ankle angle affect the gait cycle. Special emphasis is put on conditions similar to those affecting foot drop. Both swing and stance phases of gait are simulated. For this purpose, a simplified model (an existing OpenSim model) was used, which represents the right lower-limb and the key muscles affecting the hip, knee and ankle movements. Furthermore, a lumped parameter, dynamic muscle model was studied and its input excitation signals were modified in order to simulate biomechanical deviations of gait cycle that resemble those of pathological conditions.

These simulation studies offer a better understanding of the lower-limb biomechanics and their association with the muscle operations. Most importantly, the corresponding simulation results provide guidelines on how to alter the muscle operation in a lower-limb computational model for generating simulation conditions that represent an abnormality. This modeling approach (of creating pathological conditions), can be used as a tool in the design of orthotic devices and exoskeletons, and can facilitate engineers and scientists in designing assistive devices (passive or robotic), such as Ankle Foot Orthosis (AFO). For example, the engineers can generate abnormal functions and test whether their devices compensate for the abnormality (such a study is not included in the Thesis). The knowledge gained from this study aims to improve the accessibility and effectiveness of biomechanical devices and can support the creation of customized therapeutic solutions for gait rehabilitation.

Chapter 1: INTRODUCTION

Biomedical engineering is a new area of research in medicine and biology, providing new concepts and designs for the diagnosis, treatment and prevention of various diseases [1]. The discipline of biomedical engineering has emerged as an integrating medium for two dynamic professions, medicine and engineering, and has assisted in the struggle against illness and disease by providing tools in fields such as biomechanics, biosensors, biomaterials, image processing, and artificial intelligence that can be utilized for research and diagnosis by health care professionals.

Biomechanics is responsible for the movement of a living body and expresses how muscles, tendons, bones and ligaments work together so that the human body moves. It also applies principles like mass, heat transfer, fluid mechanics, kinetics, dynamics, mechanics of materials, among others to analyze biological systems and together with motor control, they use laws of nature to simulate the movements of biological objects. The field of biomechanics first appeared in the Renaissance, with Borelli being the first to study and make observations on the mode of operation (from a physicist's point of view) and the basic principles of muscular and skeletal dynamics. Borelli calculated the forces required for equilibrium in various joints of the human body well before Newton published The Laws of Motion. He was, thus, the first to understand that the levers of the musculoskeletal system magnify motion rather than force, so that muscles must produce much larger forces than those resisting the motion [2]. Since then, the field of biomechanics has blossomed, but there are still many questions that need to be answered. One of the most recent advancements in the field includes the implementation of biomechanical modeling of the human body, which allows scientists to simulate and present selected areas of the body and study it by analyzing features like COM, DoF. This allows many applications to thrive in biomedical engineering. There are significant areas like control of artificial organs and rehabilitation engineering, which combine biomechanical modeling, signals and systems and the basics of robotics to offer a variety of powered prostheses and other products that enhance people’s lives [3],[4]. However, the design and production of such objects is still a major challenge and for this reason it is very beneficial to use simulation environments (e.g. OpenSim, AnyBody) that return the design and development of the human body as well as many pathologies and complications.

1.1. Medical Background

The medical field encompasses a vast array of knowledge and practices aimed at diagnosing, treating, and preventing diseases to improve patient health and well-being. One critical aspect of medical science is the rehabilitation of neurological diseases or pathologies which deteriorate the quality of life. An example of such a pathology is Foot Drop, which is a general term that describes the loss of ankle dorsiflexion and it is an easily recognized sign of a more complex systemic or localized neurologic process [5]. This condition leads to an unsafe gait and might result in falls for the person.

There are various reasons that might cause this pathology, such as muscular, neurological, spinal, autoimmune and musculoskeletal disorders. The most common cause of foot drop is compression of the peroneal nerve in the leg that controls the muscles involved in lifting the foot. A serious knee injury can lead to the nerve being compressed or it could also be injured during hip or knee replacement surgery, which may cause foot drop. It is also associated with a nerve root injury in the spine and it is more common to notice this pattern in people with diabetes, as they are more susceptible to nerve disorders [6]. The disorders which affect the spinal cord or brain - such as stroke, multiple sclerosis (MS) or amyotrophic lateral sclerosis (ALS) - might as well cause Foot Drop. To further examine the diseases due to which foot drop can occur, a table below is listed:

Disease/Condition	Symptoms	Percentage of Foot Drop Occurrence
Lumbosacral Spine Surgery Complications	Post-surgical complications affecting nerve function.	13.9%
Lumbosacral Spine Disorders (Non-surgical)	Nerve compression or damage due to spinal issues.	12.8%
Hip Replacement Surgery	Nerve injury during surgery leading to muscle weakness.	7.8%
Peripheral Neuropathy	Numbness, tingling, and weakness in the lower limbs.	Not specified.
Muscular Dystrophy	Progressive muscle weakness and degeneration.	Not specified.
Stroke	Sudden weakness or paralysis on one side of the body, including potential foot drop.	Not specified.
Multiple Sclerosis	Muscle weakness, spasticity, and coordination issues.	Not specified.
Cerebral Palsy	Muscle stiffness and weakness affecting movement.	Not specified.
Charcot-Marie-Tooth Disease	Progressive loss of muscle tissue and touch sensation, particularly in the feet and legs.	Not specified.

Table 1. Diseases that can cause Foot Drop.

There are different impairment levels of Foot Drop characterized by the joint's range of motion. The condition starts with a mild level and is extended to moderate or severe impairment. For these levels, it should be mentioned that the mild impairment minimally affects gait, during the moderate impairment there is a significant reduction that impacts functional tasks (such as walking or climbing stairs) and the severe impairment is characterized by the requirement of assistive devices for mobility [7]. During Foot Drop, the ankle and foot dorsiflexors, namely the tibialis anterior, extensor digitorum longus, and extensor hallucis longus, are affected and cannot help clear the foot during the swing phase of walking and control plantar flexion of the foot on heel strike. It is widely known that the main muscle, the weakness of which results in Drop Foot, is the tibialis anterior and therefore the simulations that follow the pathological gait pattern focus on the central muscle.

1.2. Study Contribution

The contributions of this study include:

Demonstrate the use of OpenSim for simulating Pathological Gait Patterns: The study focuses on replicating pathological conditions, such as drop foot, within OpenSim to analyze their effects on movement and stability, which will help to better understand the biomechanical challenges faced by individuals with impaired gait.

Demonstrate the use of OpenSim as a tool for analysis of gait dynamics subject to pathological conditions: The work highlights the use of OpenSim as a research tool for analyzing musculoskeletal dynamics under pathological conditions in order to foster advancements in orthotic and rehabilitation technologies.

Demonstrate the use of OpenSim as a tool for Orthotic System Design: By providing detailed simulation data and insights into pathological gait mechanics, it serves as a tool for engineers and designers to develop and refine ankle-foot orthotic (AFO) systems. The approach is not limited to AFO.

1.3. Study goal

Comparison of Pathological Simulations Across Gait Parameters:
The aim is to simulate realistic conditions and provide a robust foundation for evaluating potential orthotic solutions. The insights of these simulations enable a better understanding of the biomechanical challenges associated with impaired movement and they provide biomechanical data from pathological simulations.

Use of OpenSim as a Biomechanical Tool:
The study demonstrates the use of many algorithms and methods that contribute to the development of a digital twin of a musculoskeletal human body. Through these methods, one can study the mathematical and mechanical background and gain more knowledge on how to use OpenSim for simulations and then include the data to solve a real-world biomechanics problem.

1.4. Structure

This work is divided into 5 sections. More specifically, Chapter 1 includes the introduction which describes the goals of this work, the definition of Foot Drop as well as the structure of this work. Chapter 2 is the literature review where kinematics of lower limb are presented and insights are provided on musculoskeletal bodies' simulations using an open-source software environment like OpenSim. Additionally, Chapter 2 includes background of biomechanics simulations by analyzing how systems and controllers work together with the mathematical equations used. Chapter 3 presents examples of simulations with healthy musculoskeletal bodies and a detailed explanation on how the muscle features are used to achieve the desired excitation and dynamic activation of the muscle, after studying it as a dynamic mechanical system. It also includes methods and algorithms that were used to modify the model and make physiological gait simulations. Chapter 4 is devoted to the methods of the final simulations and results, which provide a modified model that could be used as a prototype - and a tool - to simulate muscle pathologies during gait cycle. Lastly, Chapter 5 includes the conclusion of this work, as well as future ideas.

Figure 2. Foot Drop representation

Chapter 2: LITERATURE REVIEW

2.1. Biomechanics simulations

Biological advances have been made in several medical fields that aim to find new therapies, such as tissue engineering. Among these, there is a strong emphasis on bone and cartilage research that aims to enhance fracture healing in individuals with traumatic injuries and to mitigate early osteoarthritic changes in athletes. In the realm of biomechanics, many new insights have deepened our understanding of joint kinematics. With these new advancements, they have paved the way for improved ligament reconstruction techniques and more effective rehabilitation strategies. For instance, the ankle is a highly intricate structure, relying on the interplay of stabilizing ligaments and tendons that generate muscle forces. These elements work together to support locomotion and maintain both static and dynamic stability. The resulting motions involve complex combinations of 3D translations and rotations, influenced by ligament forces, joint contact forces, external forces, and forces generated by the musculoskeletal system.

The field of computational biomechanics offers a range of tools designed to analyze and simulate biomechanical systems. This offers researchers the chance to explore various aspects of human movement and musculoskeletal function. Notable tools include finite element modeling platforms (e.g., FEBio, Abaqus), musculoskeletal modeling software (e.g., OpenSim, AnyBody), and joint modeling and simulation tools (e.g., LifeMOD, SIMM). Among these, OpenSim stands out as a versatile platform for modeling, simulating, and analyzing the dynamics of musculoskeletal systems [8].

2.2. The ankle joint

The ankle joint, also known as the talocrural joint, is a pivotal structure in the human body that facilitates movement and stability. It is formed by the interaction of three bones: the tibia, fibula, and talus. These bones create a hinge joint that allows for two primary motions—dorsiflexion (raising the foot upward) and plantarflexion (pointing the foot downward). The ankle is further supported by a network of ligaments, including the deltoid ligament on the medial side and the lateral ligaments, which provide stability by preventing excessive movement that could lead to injuries such as sprains [9]. The ankle joint possesses three degrees of freedom, allowing movements (dorsiflexion/plantarflexion, inversion/eversion and abduction/adduction) in multiple planes. The axis of rotation of the ankle joint complex in the sagittal plane occurs around the line passing through the medial and lateral malleoli (Figure 4). The coronal plane axis of rotation occurs around the intersecting point between the malleoli and the long axis of the tibia in the frontal plane. The transverse plane axis of rotation occurs around the long axis of the tibia intersecting the midline of foot.

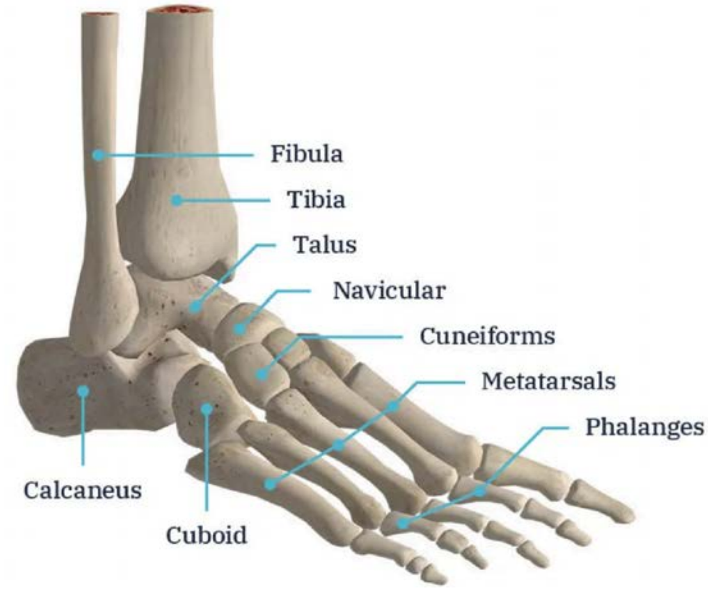

Figure 3. Foot anatomy

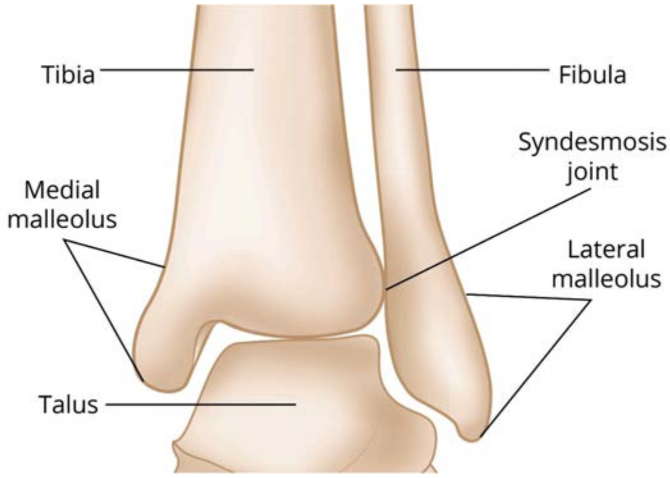

Figure 4. Ankle features

In addition to these static stabilizers (the network of ligaments), the ankle relies on dynamic support from muscles and tendons. Key contributors include the gastrocnemius and soleus muscles (Figure 6), which drive plantarflexion, and the tibialis anterior (Figure 7), which facilitates dorsiflexion. The joint also plays a crucial role in absorbing and distributing forces during weight-bearing activities like walking, running, and jumping. Its intricate balance of stability and mobility enables smooth transitions between movements, making it essential for daily function and athletic performance[10]. It is of paramount importance to understand the anatomy and biomechanics of the ankle both for diagnosing and treating injuries and for designing rehabilitation protocols to restore optimal function.

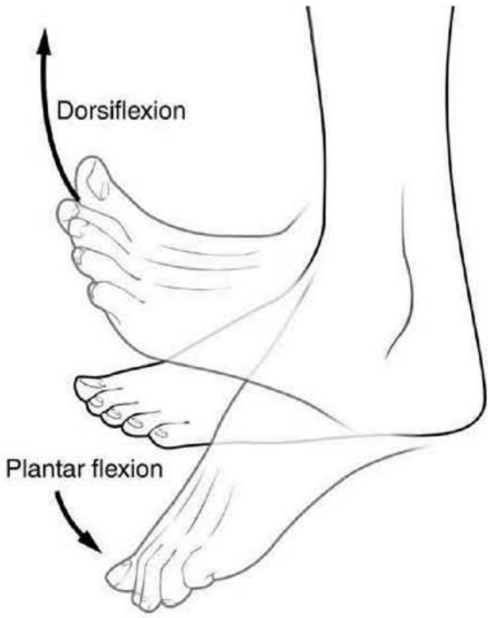

Figure 5. Representation of dorsiflexion and plantar flexion

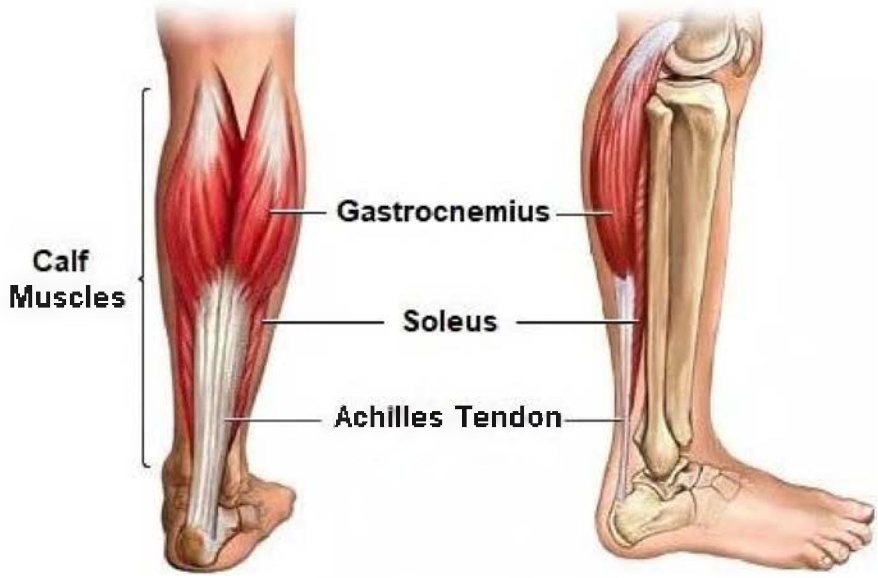

Figure 6. Gastrocnemius and Soleus - the main muscles responsible for plantar flexion

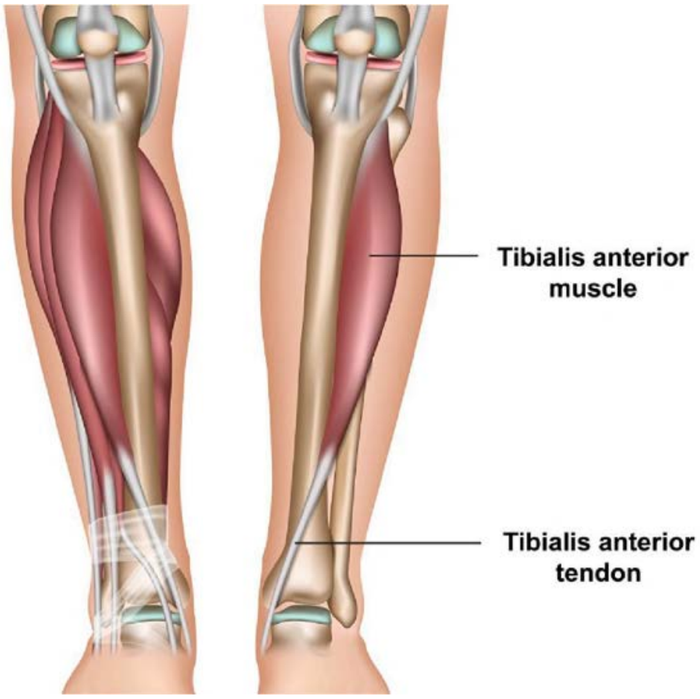

Figure 7. Tibialis anterior muscle and tendon

2.2.1. Biomechanics of the ankle

The ankle joint can be analyzed as a biomechanical system, functioning as a hinge joint governed by the principles of classical mechanics. Structurally, it is formed by the tibia and fibula, which create a mortise that articulates with the talus. This configuration allows for two primary movements: dorsiflexion and plantarflexion, while also enabling secondary motions like inversion, eversion, and slight rotation.

Mechanically, the ankle operates as a lever system, in which the bones act as rigid links, and the muscles and tendons serve as force generators. Muscles like the tibialis anterior and gastrocnemius produce the forces that maintain the structural integrity of the joint while allowing smooth transitions between movements and are required for angular motion, while ligaments such as the deltoid and lateral ligaments stabilize the joint by counteracting excessive movements [11].

The ankle also plays a crucial role in energy management. Tendons like the Achilles function as elastic components, storing mechanical energy during dorsiflexion and releasing it during plantarflexion. This energy storage and release mechanism enhances efficiency in movements such as walking, running, and jumping. Additionally, the talus's articulation within the mortise allows for subtle sliding and rotational components, influenced by ligament constraints and external forces.

To better understand foot and ankle biomechanics and the use of any ankle foot orthotic system, it is useful to delve into concepts such as foot drop, inversion/eversion of the foot, pronation/supination, and the associated angles and axes that govern these movements.

Foot drop is characterized by the inability to dorsiflex the foot (the backward bending and contracting of your hand or foot) and can result from various factors including nerve damage or muscle weakness. This leads to challenges in lifting the foot during the swing phase of gait and it impacts mobility and stability [12].

In a patient with drop foot, the first movements to assess for weakness are ankle eversion and inversion, which are movements of the foot around its sagittal axis. Inversion involves tilting the sole inward, while eversion tilts it outward and they are both crucial movements for maintaining balance and adapting to uneven terrain.

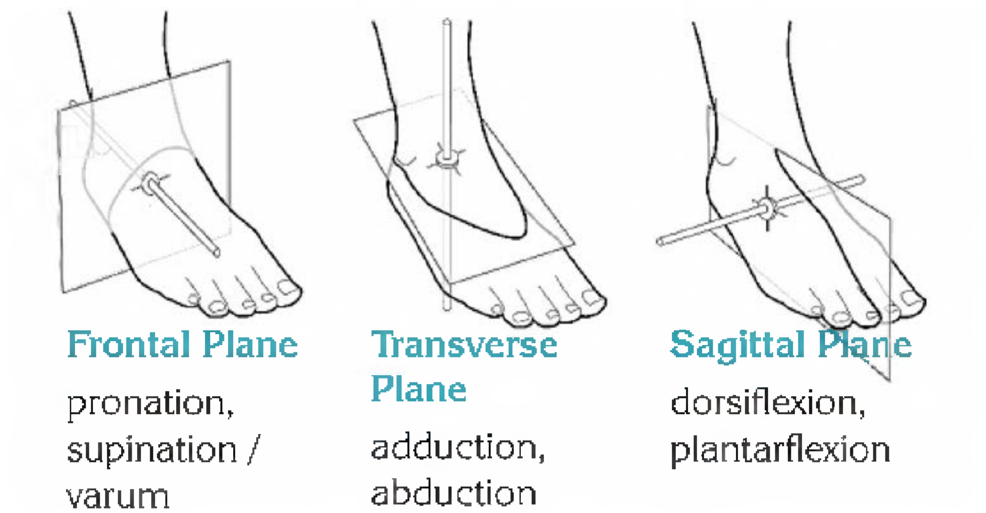

Figure 8. Ankle planes

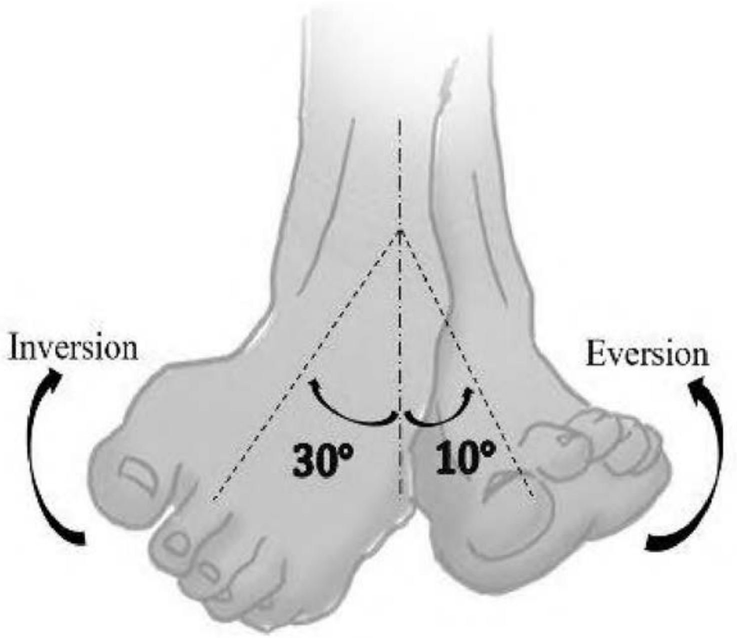

Figure 9. Inversion and Eversion of the ankle

On the other hand, pronation and supination (Figure 10) are movements around the subtalar joint that affect the orientation of the foot in relation to the ground. Pronation involves a combination of dorsiflexion, eversion, and abduction (the movement of a limb or other part away from the midline of the body, or from another), while supination entails plantarflexion, inversion, and adduction (a motion that pulls a structure or part towards the midline of the body, or towards the midline of limb, carried out by one more adductor muscles), both of which facilitate weight distribution and shock absorption during walking and running[13].

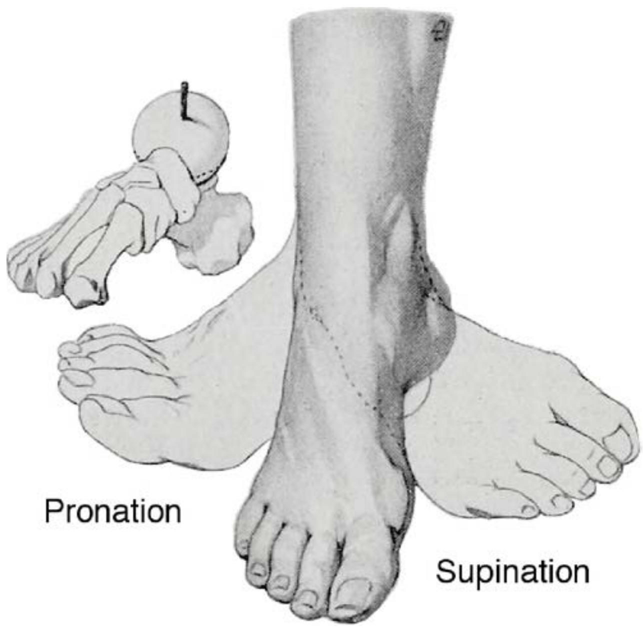

Figure 10. Pronation and Supination of the ankle

As for the deviation and inclination angles, the former denote the deviation from a neutral position, while the latter angles measure the tilt of the foot relative to a horizontal plane. Understanding these angles is crucial for diagnosing and treating conditions such as pes planus (flat foot) or pes cavus (high arch).

The ankle joint possesses three degrees of freedom, allowing movements (dorsiflexion/plantarflexion, inversion/eversion and abduction/adduction) in multiple planes and the ankle axis serves as the rotational center for these movements.

2.3. Simulation platforms for musculoskeletal systems

Simulation platforms play a crucial role in biomechanics and more specifically in the study of musculoskeletal modeling and design of assistive devices. A variety of simulation environments cater to different aspects of human movement analysis, ranging from general-purpose physics-based simulators to specialized biomechanical modeling software. A widely used framework is OpenSim, which offers musculoskeletal simulations, detailed modeling of muscle dynamics, joint kinematics, and neuromuscular control. Another example is AnyBody, a Modeling System that focuses on inverse dynamics and optimization to estimate muscle forces and joint loads. Finite element analysis (FEA), which is the process of predicting an object's behavior based on calculations made with the finite element method (FEM) [14], tools such as Abaqus and ANSYS allow for detailed stress-strain analysis of biological tissues and assistive devices, making them particularly useful for prosthetic and orthotic design. When it comes to active ankle-foot orthoses (AFOs), simulation platforms must incorporate both biomechanical and control system modeling to accurately replicate human-device interaction. MATLAB and Simulink are frequently employed for developing control algorithms, while OpenSim or AnyBody can integrate such controllers into human movement simulations.

2.4. Physiological Signal Modeling and Control in Foot Drop Rehabilitation

Physiological systems can be analyzed through the lens of signals and systems, where inputs and outputs are studied to understand system behavior. In the context of foot drop, the system includes the neuromuscular pathways, skeletal muscles and it may also contain some external assistive devices that interact to restore or compensate for normal movement. Signal-oriented approaches focus on capturing, processing and modeling the signals driving these physiological systems, such as neural commands and muscle activations.

Foot drop, as aforementioned, is characterized by the inability to dorsiflex the foot during the swing phase of gait. It arises from weakened or paralyzed tibialis anterior muscles or disrupted neural control pathways. The signals from the central nervous system (CNS) that control these muscles are critical for proper foot clearance during walking. By examining these signals as input to the system and the resulting foot motion as the output, it becomes possible to model the dynamics of the impairment and design interventions [15]. Functional Electrical Stimulation (FES) and active Ankle-Foot Orthoses (AFOs) are examples of assistive technologies that interface with these physiological signals to compensate for the loss of neural control [16].

FES relies on delivering electrical pulses to stimulate motor neurons and generate muscle contractions. The stimulus signal serves as the input and the resulting muscle force or joint movement becomes the output. The relationship between these can be characterized by transfer functions that capture the system's behavior under different stimulation parameters. For example, changes in frequency or amplitude of the input pulse train can modulate the muscle force output, allowing for precise control of foot dorsiflexion. Noise, primarily from external sources of variability in tissue conductivity, can interfere with these signals and make the control process a bit complicated, so there are specific techniques used to eliminate this (e.g. low pass filtering) [17].

In foot drop caused by neural deficits, the CNS's ability to produce consistent excitation signals to the tibialis anterior is impaired. As a result, external assistive systems must replicate or augment these signals and active AFOs are equipped with sensors and actuators that can dynamically respond to gait phases by detecting signals (such as ground reaction forces or joint angles). These sensor-derived signals are processed in real time to generate control inputs for actuators that assist in foot dorsiflexion. For instance, inertial measurement units (IMUs) can capture the orientation and acceleration of the leg, so that they provide input data for control algorithms [18].

The tibialis anterior muscle, central to foot dorsiflexion, can be modeled as part of a neuromuscular system driven by electrical signals. The muscle's force output depends on the excitation signal and muscle properties and when simulating foot drop, the system's input-output relationship is disrupted. This highlights the need for external interventions, such as signal-based models, which help analyze how input deficiencies—such as reduced or absent neural excitation—translate into diminished output [19].

Ground reaction forces (GRFs) play a significant role in signal-based modeling of foot drop. GRFs, measured using force plates, provide insights into the interaction between the foot and the ground. These forces are inputs to biomechanical models that calculate joint torques and muscle forces during gait and analyzing these signals enables the fine-tuning of assistive devices to restore more natural gait patterns [20].

In summary, signal-oriented approaches to physiological systems provide a framework for understanding and addressing foot drop. By modeling the input-output dynamics of neural, muscular and mechanical systems, it becomes possible to design interventions that restore functionality and techniques such as FES, active AFOs and advanced signal processing tools allow for real-time interaction with these systems and offer promising solutions for individuals with foot drop.

Excitation-contraction coupling describes the physiological process of converting an electrical stimulus (action potential) to a mechanical response (muscle contraction) [21]. Therefore, one can realize that the cycle of actin-myosin is the equivalent of having an input signal (neural excitation) which refers to the action potential and produces the needed amount of tension for the muscle to excite and finally be activated. There are two fundamentally different approaches to studying the biomechanics of human movement: forward dynamics and inverse dynamics. Forward Dynamics is the procedure in which the (internal) forces and/or torques of a system are known and used to calculate the motion (parameters such as angular velocity, angular acceleration) that later results in the reaction forces [22] and the input for the system is a neural signal, which plays an important role to the overall muscle activation. This process (Figure 16) is called muscle activation dynamics and the output, a_i, will be mathematically represented as a time varying value with a magnitude between 0 and 1.

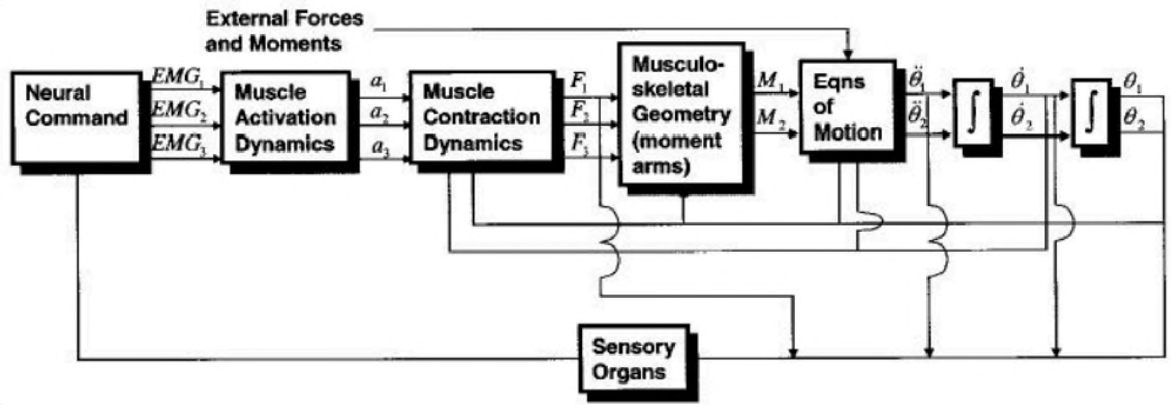

Figure 11. Forward dynamics approach to simulate muscle activation dynamics [22]

Inverse dynamics approaches the problem by measuring the external forces acting on the body. The input for the system is the angular velocity and acceleration (derivatives of the theta angle) and the output contains the external forces that are applied on the system [23].

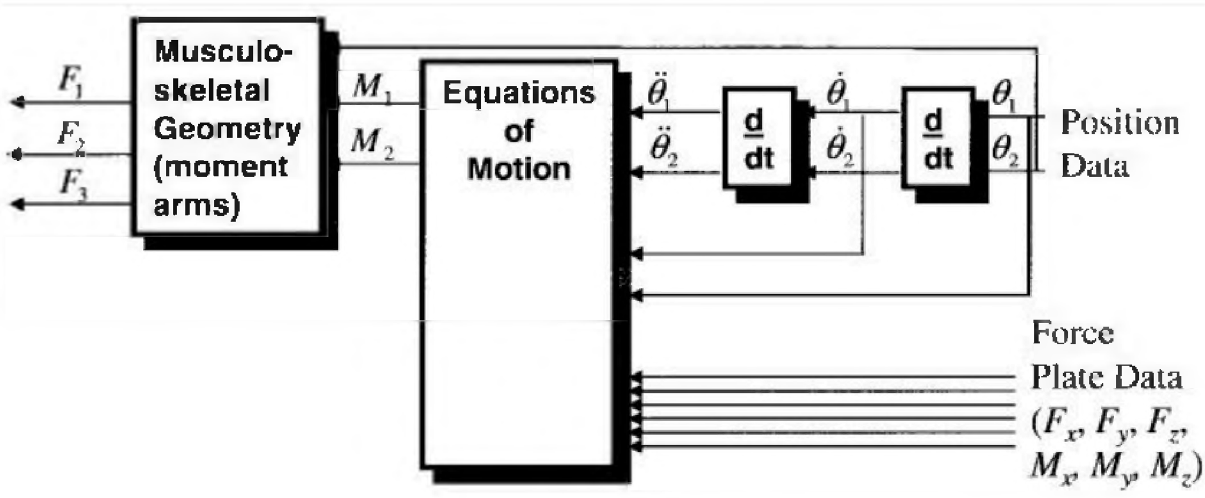

Figure 12. Inverse dynamics depiction. Data processing flows from right to left [23]

The activation dynamics of muscle can be modeled with a first-order differential equation (equation 2) that originates from the equation 1 and is depicted in a simplified form. This equation relates the rate of change of muscle activation (i.e., the concentration of calcium ions within the muscle) to the muscle excitation (i.e., the firing of motor units). Biological systems like muscle activation can be approximated as linear systems, and this first-order differential equation captures the delay between excitation and activation and the smooth, exponential-like rise and fall of activation in response to changes in excitation.

$$\frac{du(t)}{dt} = [\frac{1}{\tau act} \cdot (\beta + (1 - \beta)e(t))] \cdot u(t) \frac{1}{\tau act} \cdot e(t) \quad (1)$$

$$\frac{da}{dt} = \frac{u-a}{\tau(a,u)} \quad (2)$$

The equation (1) describes the relationship between neural excitation and muscle activation and is the origin of the differential equation (2). This equation models how the neural input is transformed into muscle activation, which accounts for the dynamics of muscle fiber recruitment and relaxation and includes the dependence of force generation on muscle length and contraction velocity.

More thoroughly, muscle activation dynamics describe the transformation of electromyographic (EMG) signals into neural activation, capturing the time delay between electrical excitation and force production. This delay is represented by the time constant τ_{act}, which governs the rate at which a muscle transitions between activation and deactivation states. The activation dynamics model proposed by Zajac (1989) uses a first-order equation to express how neural activation u(t) evolves based on the rectified and filtered EMG signal e(t). A key feature of this model is the presence of the scaling factor β, which determines how quickly the muscle responds to activation and deactivation. When e(t)=1, the muscle reaches full activation with a time constant of τ_{act}, whereas when e(t)=0, the time constant increases to τ_{act}/β. While the first-order differential model effectively captures these dynamics, numerical methods such as Runge-Kutta integration are typically required for solving it in discrete-time applications. In practice, higher-order models may offer improved performance when working with discretized data and that gives a more precise representation of muscle activation behavior.

To better understand the equation (1), the plots of each variable of the equation are provided below. The input was the sinusoidal wave: u = sin(2*pi*0.5*t). In the following figure, the output y(t) depicts the muscle contraction in time, after getting the input signal u(t) and transforming it step-by-step to the e(t), which activates the muscle [24].

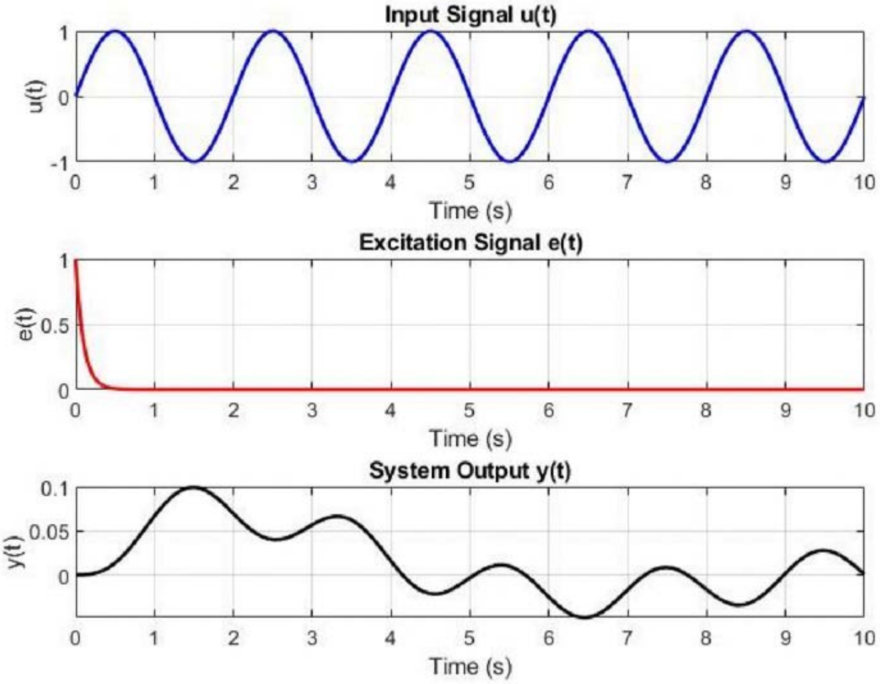

Figure 13. System's response with the input signal u(t), excitation function e(t), and the system output y(t). The MATLAB code used to compute these results can be found in the Appendix.

When the Forward Dynamics Tool is executed, it uses the excitation signals provided by the controls file (the values that have been produced for each specific moment in time) and the first-order dynamics model to compute how muscles produce force and how this force drives movement in the skeletal system. The OpenSim API handles the simulation process by integrating the equations of motion and updating the state of the model over time. The relationship between excitation and activation is implemented by the MuscleFirstOrderActivationDynamicModel class [25] that manages the first-order differential equation and ensures that the muscle's activation evolves over time in response to the excitation input.

2.5. Related Work

Biomechanical gait models hold significant potential in the advancement of assistive devices with examples of exosuits, AFOs, exoskeletons. The utilization of biomechanical models enables researchers and engineers to gain deeper insights into the dynamics and control architecture of human movement, as well as the interactions among muscles, bones, and tendons generating locomotion [26]. There are several recent studies and examples of such devices, which are mentioned below:

G-Exos: A Wearable Gait Exoskeleton for Walk Assistance

The G-Exos is a low-cost ankle orthosis designed to aid individuals with gait impairments, such as foot drop. This wearable exoskeleton assists in dorsiflexion, plantarflexion, and ankle stability through a hybrid system combining active motor assistance and passive elastic bands. It was tested on 10 volunteers with foot drop resulting from varied conditions (e.g. stroke, incomplete spinal cord injury, and acute inflammatory transverse myelitis) and the results indicated significant improvements in dorsiflexion amplitude and ankle eversion and inversion during gait with the use of G-Exos compared to walking without it[27].

Modeling the Interaction Between Wearable Assistive Devices and Digital Human Models

This systematic review identifies interaction modeling approaches between wearable assistive devices and digital human models and focuses on the softness of biological tissue [28]. An approach that integrates the viscoelastic (the property of materials that exhibit both viscous and elastic characteristics when undergoing deformation [29]) behavior of soft biological tissue in digital human models could improve the design of wearable assistive devices and their efficiency and efficacy.

Active Ankle–Foot Orthosis Design and Computer Simulation

This paper introduces an innovative ankle–foot orthosis specifically engineered to assist individuals with foot-drop by providing support exclusively during the dorsiflexion phase of the gait cycle, which occurs during the swing phase. The device utilizes a single actuator coupled with a straightforward transmission system and a suspension block and in order to optimize its performance across varying initial conditions, the design incorporates multi-objective optimization techniques through simulations [30].

Soft Ankle Exoskeleton to Counteract Dropfoot and Excessive Inversion

This study develops a wearable exoskeleton designed to assist individuals with dropfoot, which provides simultaneous assistance in dorsiflexion and eversion through its actuation and transmission systems. Proof-of-concept experiments demonstrated the device's potential to improve gait patterns by enhancing foot clearance and promoting proper heel contact during walking [31].

Development and Preliminary Evaluation of a Lower Body Exosuit to Assist Gait in Individuals with Neurological Impairments

A preliminary evaluation of a lower body exosuit designed to assist gait in individuals with neurological impairments by improving mobility for users and providing active support during the swing phase of the gait cycle. This soft, lightweight device employs Bowden cables to transmit mechanical assistance to the ankle joint, facilitating dorsiflexion and improving foot clearance [32].

Feasibility of an optimal EMG-driven adaptive impedance control applied to an active knee orthosis

An interaction model based on the gait2392 neuromusculoskeletal model from OpenSim and a simulation environment developed on MATLAB were used by [33] in order to estimate the torque provided by a user wearing a lower limbs exoskeleton. This study presents the development of a wearable ankle exoskeleton that enhances plantar flexion during walking. and has been designed to augment the push-off phase of the gait cycle, which improves walking efficiency and reduces the metabolic cost of locomotion.

Chapter 3: SIMULATION AND ANALYSIS OF MUSCULOSKELETAL MODELS USING OPENSIM

3.1. OpenSim models

OpenSim is an open-source software toolkit widely used in biomechanics to build musculoskeletal models, simulate movement, and analyze resulting behaviors after simulating and studying the dynamics of human and animal movement. OpenSim allows users to build detailed models of the musculoskeletal system, incorporating bones, muscles, tendons, and ligaments, and then simulate how these components interact during various movements. It supports forward dynamics, inverse dynamics, and optimization techniques, making it ideal for applications such as gait analysis, sports performance improvement, and rehabilitation research [34]. Its extensibility and active user community ensure continuous development and innovation, solidifying its role as a cornerstone in the field of computational biomechanics.

There are some basic things that a person needs to know in order to use OpenSim, which include a modest level of C++ programming skills and knowledge of the OpenSim API. An application programming interface (API) is a set of rules followed by software programs so that they can communicate with each other [35]. In the case of OpenSim, when there is interaction with its API, several classes are available for use. It is also simple to use Matlab or Python to run methods and OpenSim objects.

3.1.1. Tug-of-War Simulations

The **TugOfWar.osim** model in OpenSim is designed to simulate a simplified musculoskeletal system engaging in a tug-of-war scenario, providing an illustrative example of muscle force generation and coordination during dynamic activities. This model includes multiple actuators that mimic muscle-tendon units, offering a controlled environment to study the biomechanics of force application and resistance under specific loading conditions [36].

As a biomechanical system, the TugOfWar model emphasizes the interaction between muscle forces, joint movements, and external loads and that makes it a useful tool for exploring principles of neuromuscular control and mechanical advantage. By simulating opposing forces, the model provides insights into how muscles coordinate to maintain balance and achieve maximum output in competitive or resistive tasks [37].

The TugOfWar.osim model is particularly valuable in understanding how muscle properties such as activation, force-length, and force-velocity relationships influence performance. It also serves as a platform for testing optimization strategies, evaluating energy efficiency, and assessing the impact of various design modifications on musculoskeletal performance. This foundational example is often used as a stepping stone for developing more complex simulations tailored to specific research objectives.

Through this model, the relationship between neural excitation signals and resulting biomechanical movements can be examined. In this context, the Forward Dynamics Tool in OpenSim was used to simulate muscle activation dynamics. Neural input signals (muscle excitation), were provided through control files created using MATLAB scripts. These control files included time-dependent excitation values for specific muscles and were used as primary input for the simulation process.

Muscle excitation signals range from 0 (no excitation) to 1 (full excitation), representing the neural drive sent to the muscle. However, excitation does not immediately result in activation, as the process is governed by a first-order differential equation. This equation models the dynamic relationship between calcium ion concentration and muscle fiber activation. OpenSim's **MuscleFirstOrderActivationDynamicModel** class manages this transition, ensuring that muscle activation evolves over time in response to neural input. Using the OpenSim API, these dynamics are integrated with the equations of motion to simulate the forces and movements of the musculoskeletal system.

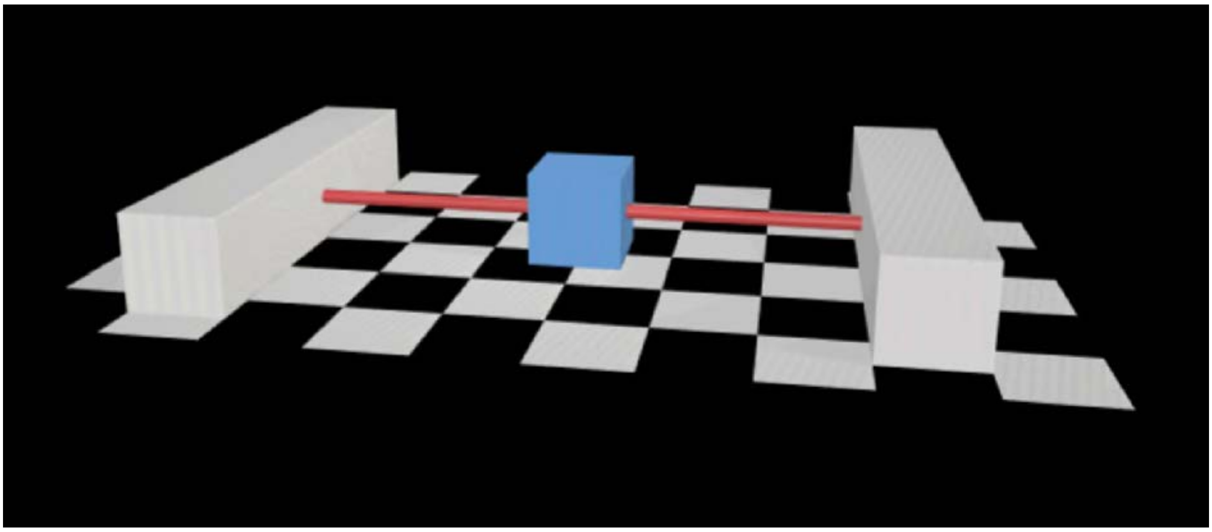

Figure 14. Tug-of-War model

Musculoskeletal modeling and simulation offer great potential for neurorehabilitation applications and research. There are many impairments that can be both observed and simulated through an environment, such as OpenSim, which is a freely available application that provides well developed musculoskeletal modeling elements and simulation tools. The challenge of better understanding how to use the control of musculoskeletal systems in order to produce movement is what makes most movement scientists interested in simulating such systems. Another significant achievement and tool is features like the electrical state of the muscle (activation) and its neural input (excitation) that sources from the central nervous system (CNS), which can be simulated successfully through OpenSim and help engineers, physiotherapists and designers [38]. The force-producing properties of muscle are complex, highly nonlinear, and can have substantial effects on movement, but in order to make the movement simulation simpler, there are Hill-type muscle models. Hill's muscle model refers to the 3-element model consisting of a contractile element (CE)) in parallel with lightly-damped elastic parallel element (PE) [39]. Another important parameter is the differential equation that describes muscle activation and muscle–tendon contraction dynamics when using a Hill-type muscle model (Figure 15).

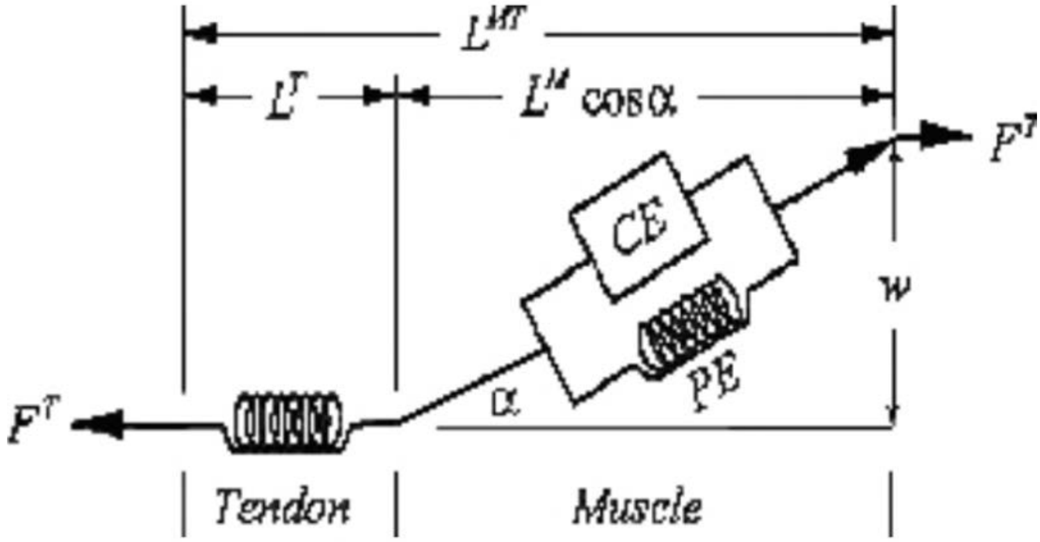

Figure 15. Hill's muscle model [39].

The initial target of the Tug-of-War tutorial was to design an optimal muscle for a tug-of-war competition that pulls the center of the block onto a specific (chosen) side of the arena within a 1.0-second simulation using the OpenSim platform. To achieve this, there were specific values given to several muscle parameters with the condition that they fall within specified physiological constraints while maximizing performance, as analyzed below in 3.1.1. In paragraph 3.1.2., there is an input excitation and more specifically a sinusoidal wave applied on the right model muscle that is used to activate and contract the muscle and as a result to move the block towards its direction. This simulation highlights the influence of an excitation signal on muscle behavior and system dynamics within the Tug-of-War model. This study together with the Forward Dynamics Tool and OpenSim API establishes a foundation for exploring excitation signals and their implications in neuromuscular research and orthopedic applications.

3.1.2. Muscle properties in OpenSim for Tug-of-War simulation

Some of the parameters include:

- The maximum isometric (muscle contraction without motion) muscle fiber force (Fiso) is the maximum force a muscle can generate isometrically, constrained between 500 N and 2000 N.
- The optimal muscle fiber length (Lopt), where the muscle generates maximum force, should be between 0.1 m and 0.3 m.
- Tendon slack length (LT_slack) is the tendon length when not under tension, constrained between 0.1 m and 0.3 m.
- The muscle fiber pennation angle at optimal fiber length (θopt) describes the angle between muscle fibers and the line of action at optimal length, constrained between 0.0 rad and 0.5 rad (pi/6 in the simulation).

- The activation time constant (τact) and deactivation time constant (τdeact) describe the muscle's activation and deactivation dynamics, constrained between 0.01 s and 0.06 s, and 0.03 s and 0.09 s, respectively.
- The maximum contraction velocity (the speed at which the muscle fibers shorten or lengthen) normalized by optimal fiber length (Vmax) should be between 2 and 10. According to the force-velocity relationship, muscle force production decreases as the speed of contraction increases.
- The muscle excitation time history is the time-varying input signal to activate the muscle, constrained between 0 and 1. Initial activation will be 0.01 (muscle activation ranges from 0 (completely inactive) to 1 (fully activated)), solved for equilibrium states (this involves adjusting the muscle fiber length and tendon length so that the muscle-tendon unit is in a balanced state where forces are properly distributed and the system is not experiencing any undue tension or slack).

3.2. First Order Activation Dynamics

The development of force for a muscle is not a simple procedure during which the muscle either relaxes or produces force instantly. This transformation is complex, not only because the properties of muscles are complex, but because the tendon affects the transmission of muscle force to the skeleton[38]. It is a series of events that begins with the firing of motor units and culminates in the formation of actin–myosin cross-bridges within the muscle. When the motor units of a muscle depolarize, action potentials are elicited in the fibers of the muscle and cause calcium ions to be released from the sarcoplasmic reticulum. The increase in calcium ion concentrations then initiates the cross-bridge formation between the actin and myosin filaments. The relaxation of muscle depends on the re-uptake of calcium ions into the sarcoplasmic reticulum. This re-uptake is a slower process than the calcium ion release, so the time required for muscle force to fall can be considerably longer than the time for it to develop[39].

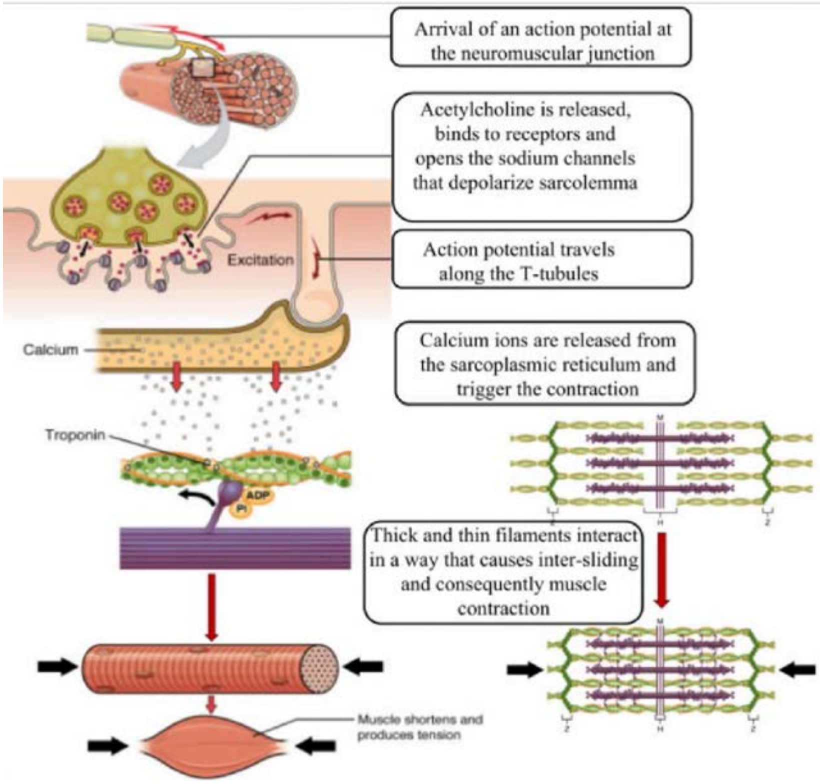

Figure 16. Actin-Myosin muscle contraction

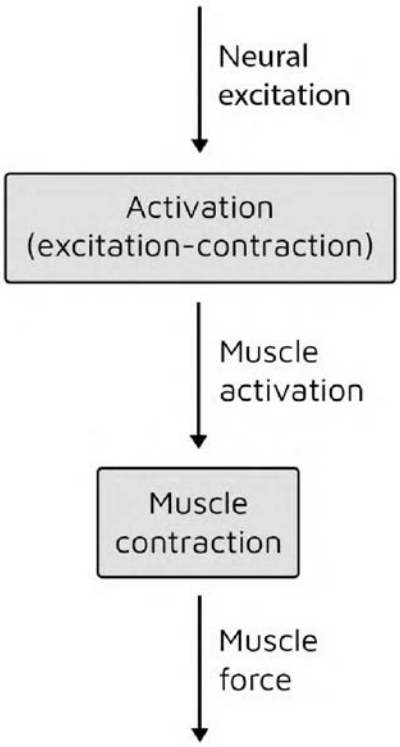

Figure 17. Excitation of the muscle as a block diagram

3.2.1. Muscle Excitation and Block Displacement

In order for the muscle to activate, it needs to be excited first. Muscle excitation occurs when there is a neurochemical signal that gives to muscle fibers the opportunity to generate electrical impulses as a response to these stimuli [40]. The aim of the Tug-of-War experiment was to design an optimal muscle in order to win the competition. In other words, several parameters are applied to one of the muscles so that the desired outcome (block displacement to the direction of the chosen muscle) would be achieved. In the case that is analyzed below, there is a sinusoidal wave as an input using the Excitation Editor (allows you to visually inspect and edit muscle excitation patterns) of OpenSim. This signal enters the differential equation describing muscle activation and muscle–tendon contraction dynamics (Figure 17) and then the needed force is produced. Below you can notice the displacement of the block due to the Right Muscle excitation with the input signal being a sinusoidal signal, as aforementioned.

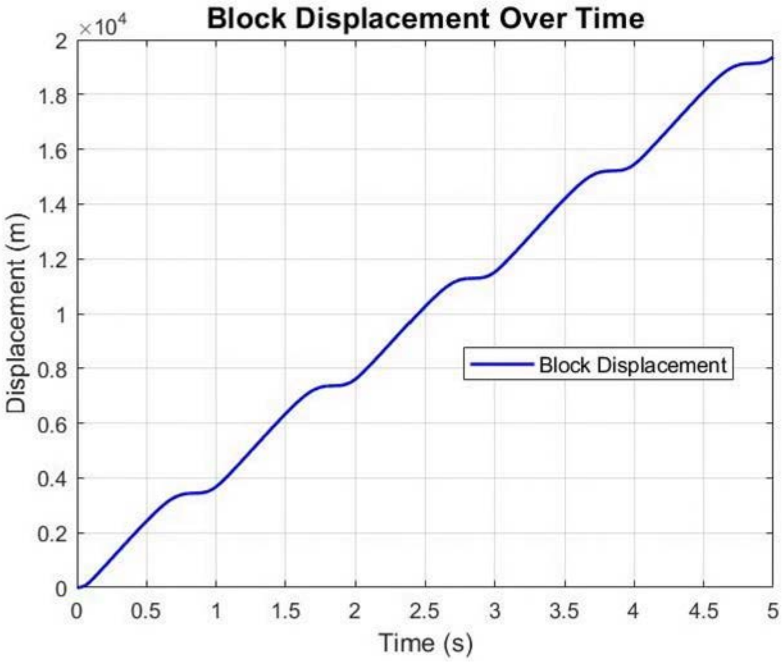

Figure 18. Block displacement due to Right Muscle excitation

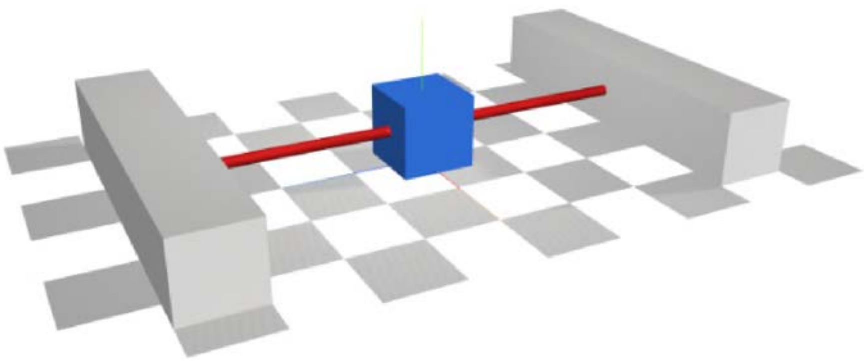

Figure 19. Tug of War model

The OpenSim API allows users to have an interaction between OpenSim classes and MATLAB. In this experiment, there is a sinusoidal wave $A \cdot sin(\frac{2\pi \cdot t}{T} + \pi) + b$, where A=0.5m , b=0.5) that is used as a neural signal (input) for the muscle. The script also includes some rules that make a new controls file in XML. The controllers file needs to have some values (extracted from the signal), kp and kv gains and the time all imported and calculated specifically for the Right Muscle and for the duration of the simulation (5 seconds in total).

3.3. Physiologic Gait Simulations With Leg6dof9musc OpenSim Model

The gait cycle can be divided in two primary phases: Swing and Stance. The stance phase (approximately 60% of the gait cycle) is defined when the foot is in contact to the ground and the swing phas*e* (approximately 40% of the gait cycle), period of time that the foot is not in contact with the ground, the ankle is dorsiflexed and the leg is moving forward and preparing for the next contact[41].

The **Leg6Dof9Musc.osim** model in OpenSim serves as a robust tool for studying human lower-limb biomechanics as it provides valuable insights into the dynamics of muscle forces and joint kinematics. This model, which includes six degrees of freedom at the hip, knee, and ankle joints, as well as nine muscle-tendon actuators, is designed to simulate complex lower-limb movements such as walking, running, and jumping [42]. The Leg6Dof9Musc model enables researchers to explore various aspects of human movement, including muscle activation patterns, joint interactions, and force generation during dynamic tasks.

The model is a flexible and accurate representation and for this reason it is a critical tool in biomechanical studies and allows for the simulation of muscle forces, joint movements, and the effects of different mechanical conditions on human locomotion. It is a thorough starting point for understanding the biomechanics of healthy movement as well as for studying pathological conditions that affect muscle function and joint performance and also getting insights into the role of muscles, tendons, and joints in maintaining balance and optimizing performance.

In the context of this study, the **Leg6Dof9Musc.osim** model is used to investigate the effects of muscle pathologies and offers the opportunity to notice how altered muscle activation patterns influence joint kinematics and overall movement. The modified version of this model, which incorporates specific muscle pathologies, will be discussed in detail in the methodology section, where the model's adjustments and the resulting simulations are explained. The aim of this study is to simulate these pathological conditions in order to gain a deeper understanding of how musculoskeletal disorders impact lower-limb function and contribute to abnormal gait

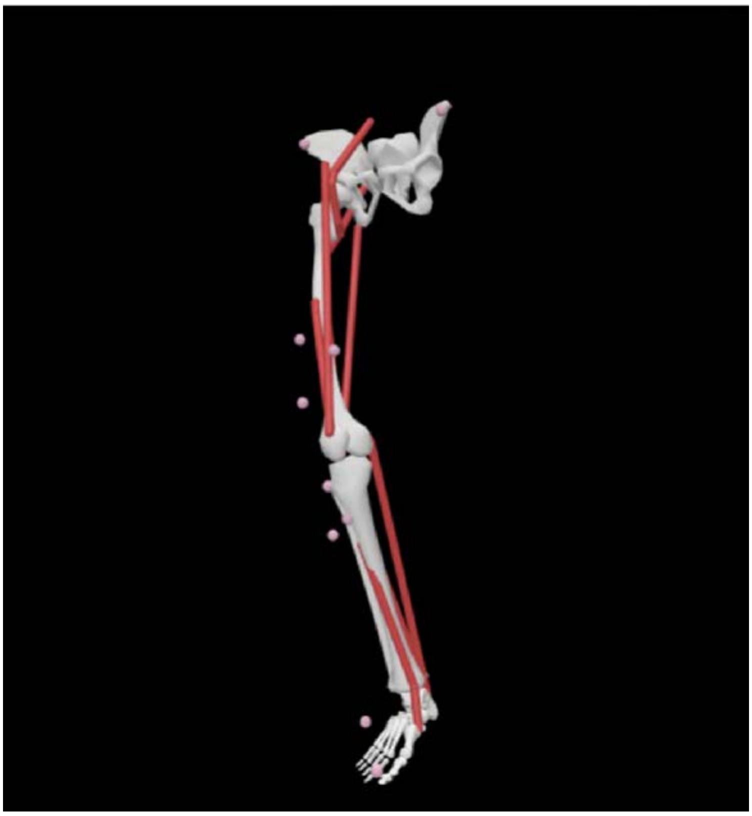

Figure 20. Leg6dof9musc model

3.3.1. Swing Phase

This phase begins when the toes leave the ground at toe-off and ends when the heel strikes the ground to transition into the stance phase. During walking, the swing phase typically constitutes approximately 40% of the entire gait cycle and can be divided into three distinct stages: initial swing, mid-swing, and terminal swing. In the first stage (initial swing), the motion is primarily driven by the hip flexors, particularly the iliopsoas, with additional contributions from the rectus femoris and happens until the maximum knee flexion to reduce the leg's moment of inertia and to ensure that the foot clears the ground. Simultaneously, the ankle dorsiflexors, such as the tibialis anterior, maintain dorsiflexion to prevent the foot from dragging. In the mid-swing stage, the knee begins to extend as the quadriceps contract to stabilize the limb, while the ankle remains dorsiflexed to maintain foot clearance and there is minimum muscular effort during that time of the cycle. The terminal swing, or deceleration phase, prepares the foot for contact with the ground and so the hip extensors try to decelerate the forward motion of the thigh while stabilizing the pelvis. Concurrently, the quadriceps fully extend the knee and the ankle dorsiflexors maintain dorsiflexion to ensure proper heel strike[43]. Leg6dof9musc is a simplified OpenSim model used to study specific forces generated by some of the muscles. A Forward Dynamics simulation was conducted in order to analyze how key leg muscles (e.g hip flexors, knee extensors, and ankle dorsiflexors) work and notice the swing phase of the gait cycle after applying algorithms such as Static

Optimization and Computed Muscle Control, both of which shall be explained in detail in the next paragraphs. The results revealed insights into how these muscles coordinate to ensure foot clearance and prepare the leg for the subsequent stance phase. More specifically, hip flexors were identified as primary contributors to forward propulsion in early swing and the dorsiflexors played a crucial role in preventing foot drop throughout the phase. The ability to estimate muscle forces provides a deeper understanding of the biomechanical demands of a musculoskeletal system and serves as a foundation for further studies in human movement science.

3.3.1.1. Static Optimization (SO)

Static Optimization is a method used to estimate muscle activations (Figure 21) and forces that align with the positions, velocities, accelerations, and external forces (such as Ground Reaction Forces - GRF) observed during a motion. This technique is referred to as "static" because the calculations are performed independently at each time frame, without solving the equations of motion across time steps. This lack of integration makes Static Optimization highly efficient and computationally fast; however, it does not account for activation dynamics or tendon compliance[44].

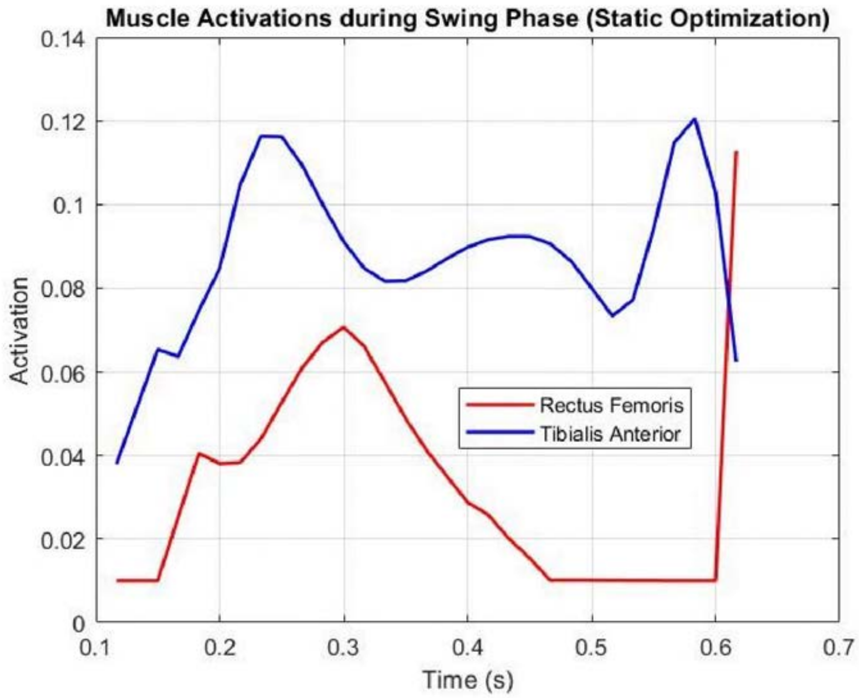

Figure 21. Muscle activations when the Static Optimization method is used during swing phase.

The activation plot illustrates the muscle activations during the swing phase of gait for the rectus femoris and tibialis anterior and represents the neural input required to stimulate each muscle and generate movement. Using the Static Optimization (SO) method, which provides a snapshot of muscle control at each time frame, the activations are computed by minimizing a cost function while satisfying the constraints of joint positions, velocities, and external forces. An important note on the plot is that the peak angles of each activation show when the muscle contributes most to the foot assistance.

3.3.1.2 Computed Muscle Control (CMC)

Computed Muscle Control (CMC) is a method used to compute muscle activations and forces required to reproduce a motion. Unlike Static Optimization, CMC uses discretization methods, and more specifically, zero-order reservation, which is a mathematical model of practical signal reconstruction that describes the result of converting a discrete-time signal into a continuous-time signal while preserving the value of each sample for a sample interval. It does so by minimizing a cost function over time using feedback control while making sure that muscles produce forces in a coordinated manner to achieve the target movement[45].

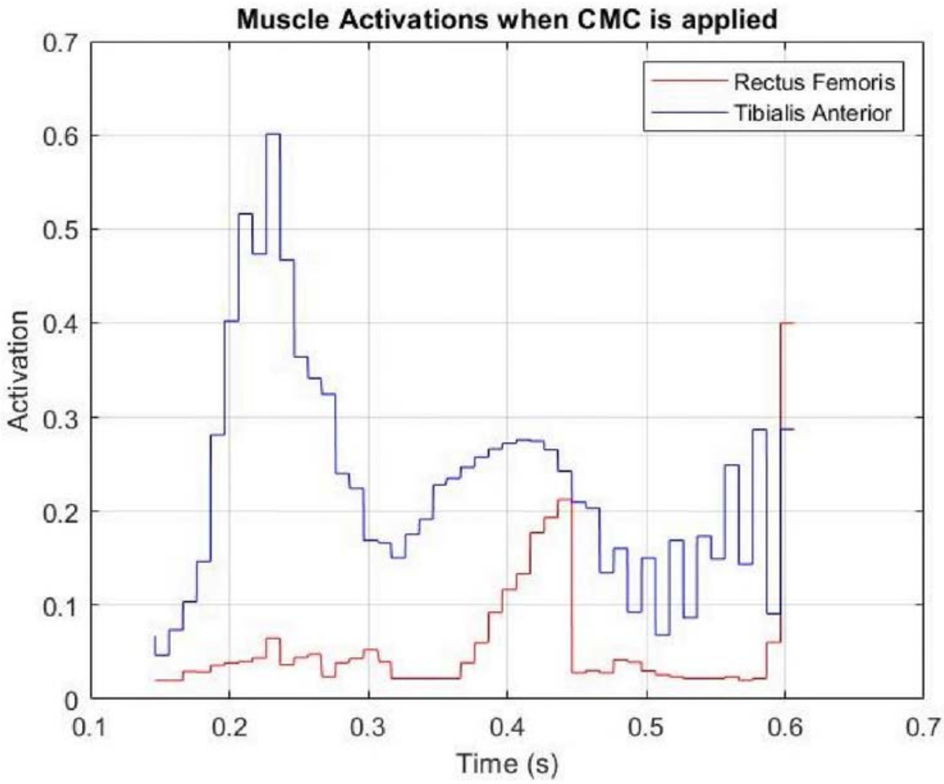

Figure 22. Muscle activations when the Computed Muscle Control method is used during swing phase.

This method adjusts muscle activations based on the body's biomechanical needs, such as stabilizing joints and generating force when a person does several activities, such as walking. Muscles like the rectus femoris and tibialis anterior have specific activation patterns depending on the phase of movement and are responsible for procedures like hip, knee and ankle movement (rotation or flexion).

3.3.1.3. Comparison of the two methods (SO - CMC)

The Forward Simulation using Static Optimization and Forward Simulation using Computed Muscle Control methods are two widely used techniques in musculoskeletal modeling for simulating muscle activations. The primary difference between these methods lies in their approach to controlling muscle forces during simulations. Static Optimization uses optimization algorithms to determine muscle excitations that minimize the difference between model output and experimental data that only focus on each time frame, as aforementioned. In contrast to that, Computed Muscle Control employs dynamic optimization to control muscle forces through a more complex, time-dependent model of muscle dynamics and studies features of the muscle that affect muscle behavior, such as joints, velocities and thus it provides a more realistic result due to the fact that it takes into consideration more dynamically oriented parameters.

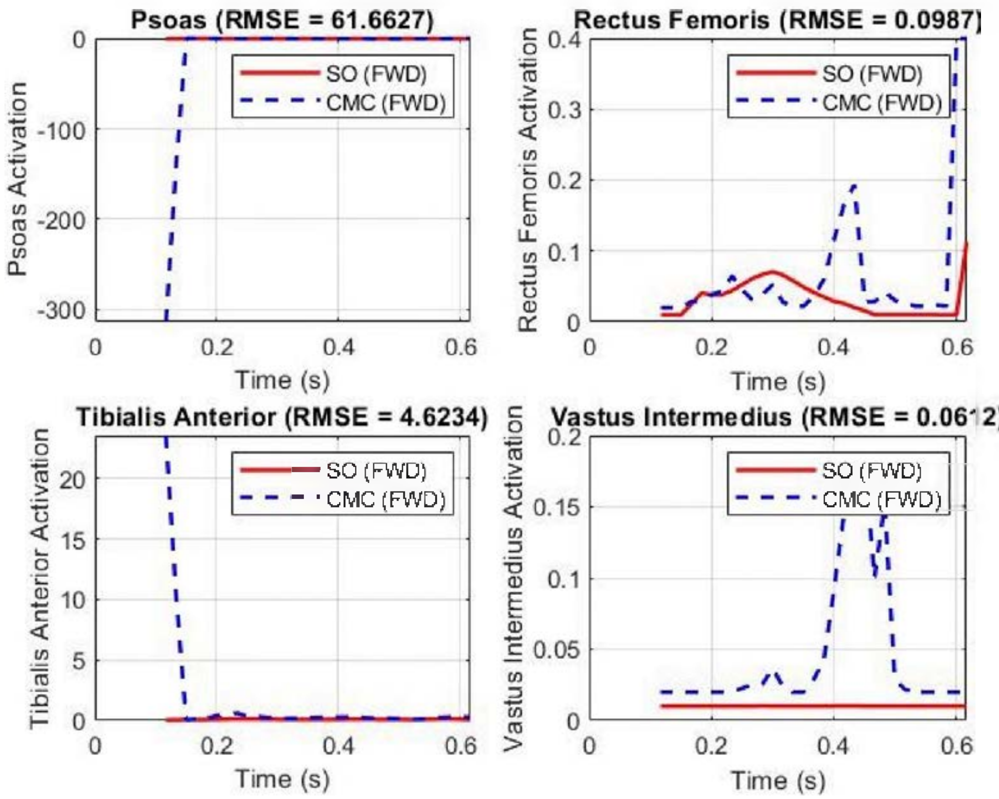

Figure 23. Comparison of muscle activations between Static Optimization and Computed Muscle Control

3.3.2. Stance Phase

The stance phase of the gait cycle begins when the foot makes contact with the ground and ends as the foot prepares to leave the surface. This phase, which makes up about 60% of the gait cycle, is divided into three stages: initial contact, mid-stance and terminal stance. In stance, muscles like the hip extensors (gluteus maximus and the hamstrings), quadriceps (the rectus femoris, the vastus lateralis, the vastus intermedius, and the vastus medialis) and calf muscles (gastrocnemius, soleus and plantaris) stabilize the body and generate the necessary force for forward movement.

3.3.2.1. Inverse Dynamics (ID)

Inverse dynamics is a computational technique that can be defined as follows: given a set of trajectories of the end-effector and actuated joints in function of time, find the set of forces/torques that produce this motion. Starting with kinematic data, such as joint angles, velocities, and accelerations, and combining it with external forces (e.g., ground reaction forces), inverse dynamics applies Newton's laws to deduce the net forces and moments at each joint[46]. The motion and force data are integrated with known segment properties, such as mass and inertia of a rigid-body musculoskeletal system, to estimate the internal efforts (e.g. contact forces, ligaments) required to produce observed movements.

Inverse dynamics is widely used in gait analysis and other biomechanical studies to understand how the body responds to external forces and generate the necessary joint actions[47]. However, due to the fact that it calculates only the net forces and moments, it is often paired with other tools like static optimization or muscle-driven simulations.

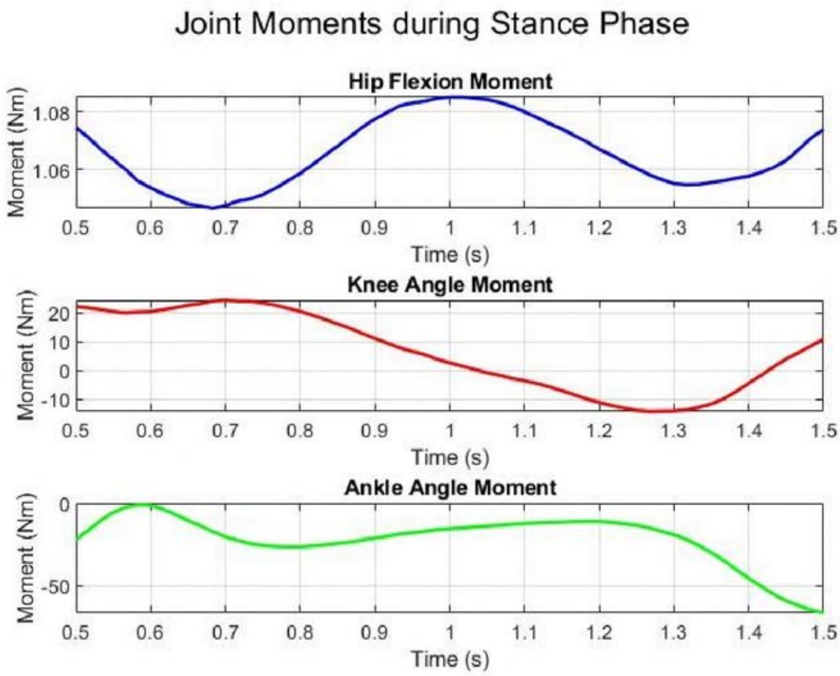

Figure 24. Joint moments after using Inverse Dynamics in the stance phase.

3.3.2.2. Residual Reduction Algorithm (RRA)

Residual Reduction Algorithm (RRA) is designed to improve dynamic consistency of kinematics and ground reaction forces in movement simulations of musculoskeletal models[48]. RRA works by making minor adjustments to a model's mass properties and kinematics to reduce these errors and result in more realistic simulations, which will obey and verify Newton's laws of motion.

It relates to Forward Dynamics by ensuring a simulation's inputs (e.g., joint moments and external forces) align with observed kinematics, but it more often follows Inverse Dynamics[49].

In the stance phase of the gait cycle, RRA plays a critical role in improving the accuracy of simulations because it makes sure that joint torques and forces are properly aligned with the measured ground reaction forces and it also offers insight in the joint moments (hip flexion, knee and ankle angle) in critical phases of walking.

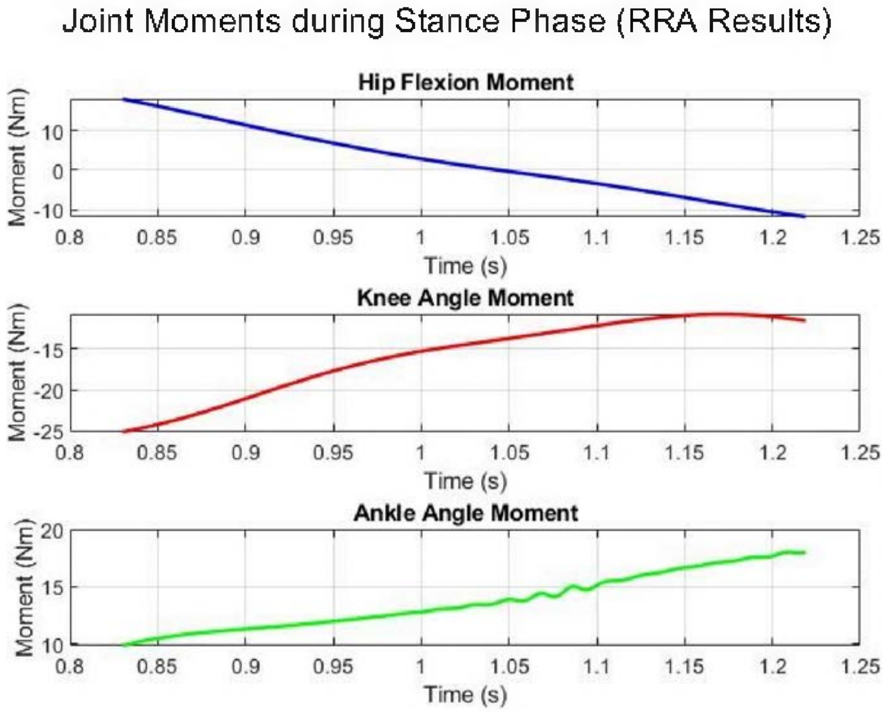

Figure 25. Joint moments after the final RRA run in the stance phase.

Chapter 4: SIMULATION EXPERIMENTS AND RESULTS

As mentioned, the aim of this study is to create a musculoskeletal system with modifications aimed to model pathological conditions. This enables the simulation of an abnormal gait cycle and allows a better approach to the design of assistive devices. To achieve the above, the Leg6dof9musc and ToyDropLanding models provided by OpenSim were used.

4.1. Simulations of the modified model Leg6dof9musc in the swing phase

There are several muscles that are responsible for abnormal gait and exacerbate some injuries or complications. In the model used, 9 muscles are represented and the following simulations relate to the 4 that affect gait. Among them, the one that will be analyzed extensively is the Tibialis Anterior, which is the central muscle of the Foot Drop. All are observed to bring about complications in the swing phase of gait and the influences, as they occur, on the activation of each muscle, on the joint moments for hip flexion, knee and ankle angle, and on the overall effect of the simplified model will be described in detail.

4.1.1. Rectus Femoris Muscle Pathology

Muscle injuries affect mainly the lower limbs and specifically superficial and bi-articulated muscles, such as the hamstrings and rectus femoris. The rectus femoris (RF) is the most commonly injured quadriceps muscle, representing up to 68 % of quadriceps injuries [50], [51] and it is most commonly observed in young athletes [59]. The RF actively contributes to the knee extension in the swing phase and it also helps to obtain a proper step length and to prepare the foot for initial heel contact, because the ability of RF to control the distal segments seems plays a relevant role in controlling the movement amplitude and the position of knee and foot at the end of the swing phase.

In the following simulations, some pathologies have been created in the RF. There are reduced excitations of the muscle by decreasing the factor as shown in each case, starting at 75% of normal, continuing at 50%, decreasing to 25% and finally to a completely dead muscle. All pathological conditions are compared to the healthy model developed using Static Optimization and Computed Muscle Control, which were discussed in the previous chapter.

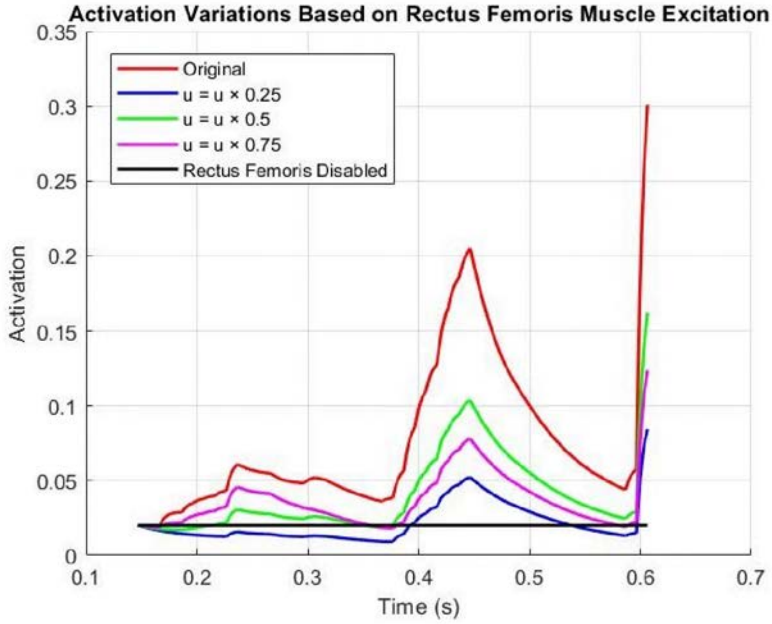

Figure 26. Rectus Femoris muscle activations for different excitation values.

Along with the reduced muscle excitation, the moments of the directly affected joints, such as hip flexion and knee angle were also studied. At the different values that u takes and thus the whole musculoskeletal system is affected, the simulations are shown compared to the initial condition. The rectus femoris is a powerful hip flexor, but it is largely dependent on the position of the knee and hip to assert its influence as it played an important role in regulating knee flexion; removal of the rectus femoris actuator from the model resulted in hyperflexion of the knee, whereas an increase in the excitation input to the rectus femoris actuator reduced knee flexion [52], [53].

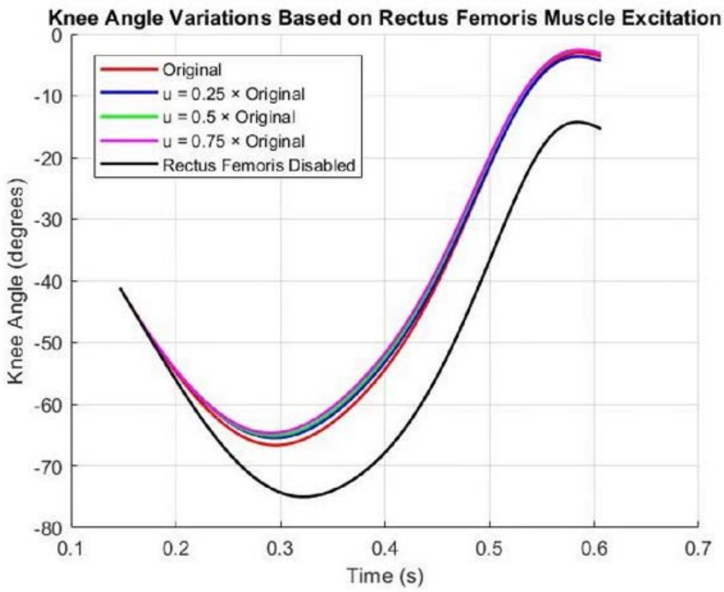

Figure 27. The effect of Rectus Femoris muscle in knee angle for different excitation values of the muscle.

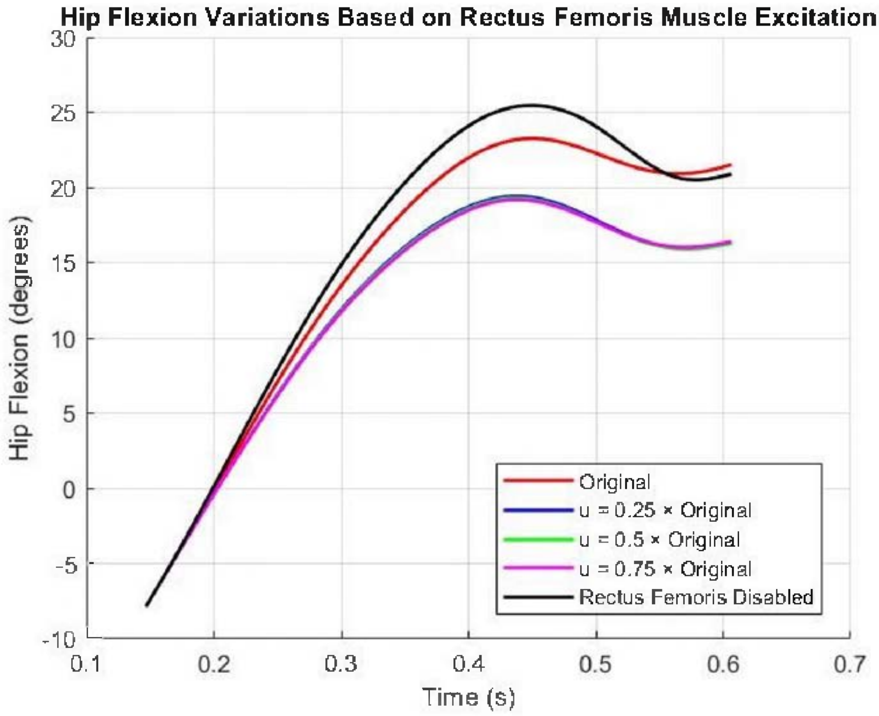

Figure 28. The effect of Rectus Femoris muscle in hip flexion for different excitation values of the muscle.

This particular muscle with its fluctuations in excitations has little effect on the ankle angle, as it has no direct connection to the ankle joint, as shown below (Figure 29). Only in the case of complete inactivation does the torque of this joint seem to change, because it is cooperatively covered by other muscle groups.

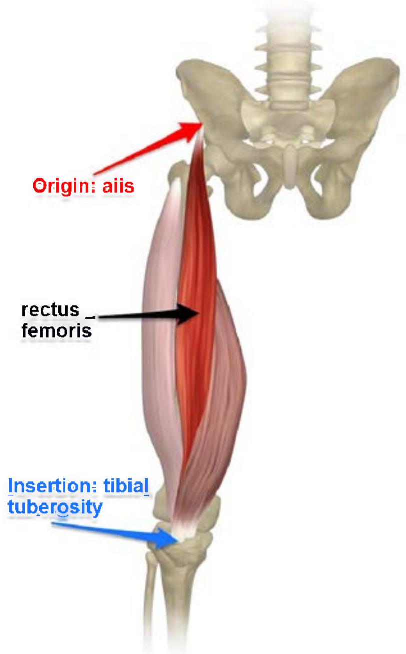

Figure 29. The Rectus Femoris muscle.

4.1.2. Tibialis Anterior Muscle Pathology - Drop Foot

The tibialis anterior muscle, also known as the tibialis anticus, is the largest of four muscles in the anterior compartment of the leg and the strongest dorsiflexor and a primary inverter of the foot. Dorsiflexion is critical to gait because this movement clears the foot off the ground during the swing phase and the early stance phase. The primary function of the tibialis anterior is dorsiflexion and thus, paralysis of this muscle results in “foot drop,” or an inability to dorsiflex. This paralysis can be caused by nerve injury, like direct damage to the deep peroneal nerve, or a muscle disorder, like ALS and it is often most obvious during gait when the patient cannot clear their foot during the swing phase[54].

Any pathology in the tibialis anterior causes severe alteration in the joint torque of the ankle joint and leaves the remaining accessory muscles that are part of the dorsiflexors to support the limb. The corresponding pathological appearances are listed below in the diagram with the different activations (Figure 30), as well as in those with the joint moments, including hip flexion (Figure 31), knee angle (Figure 32) and ankle angle (Figure 33) .

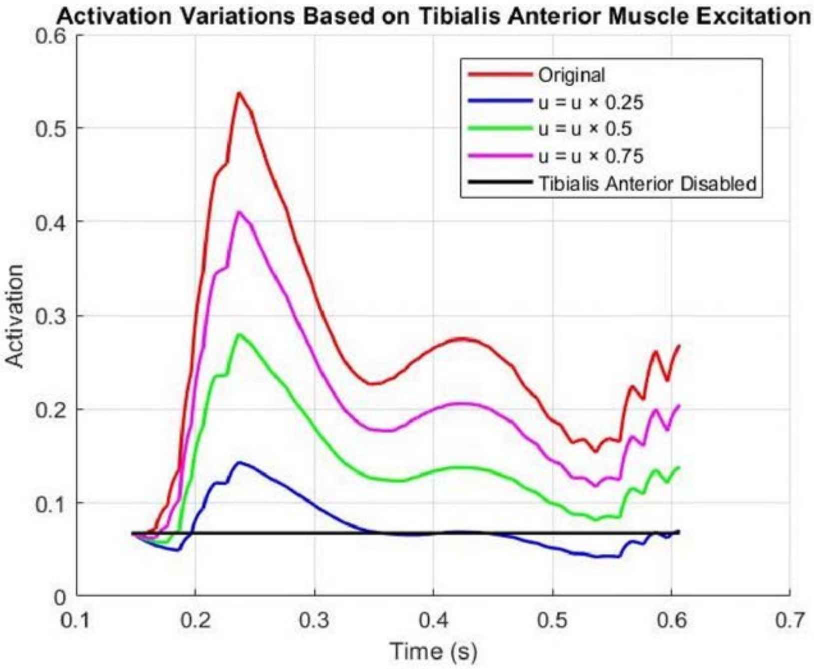

Figure 30. Tibialis Anterior muscle activations for different excitation values.

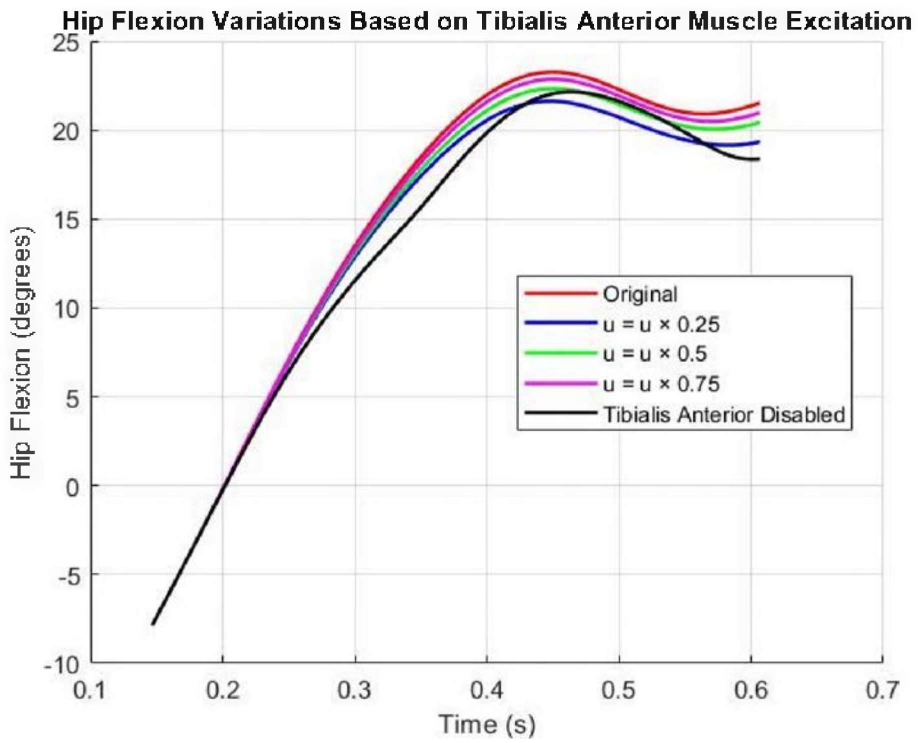

Figure 31. Tibialis Anterior muscle in hip flexion for different excitation values of the muscle.

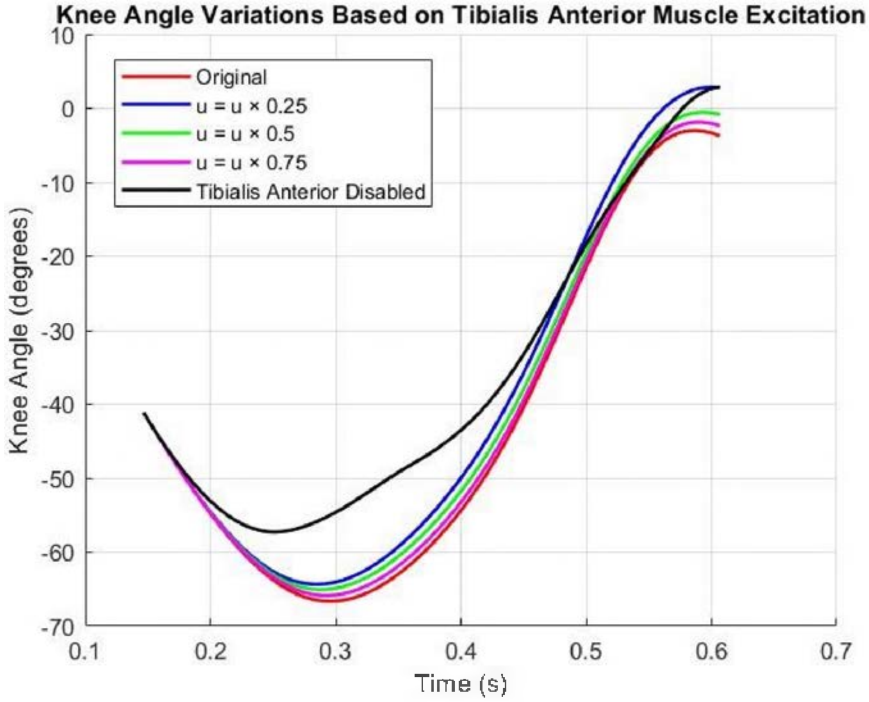

Figure 32. Tibialis Anterior muscle in knee angle for different excitation values of the muscle.

The Figure 31 and Figure 32 do not show much influence due to pathology in the tibialis anterior, as it was heavily involved in the rotational movement of the ankle joint. In contrast, Figure 33 clearly shows how the normal situation changes and in each case how the simulation differs.

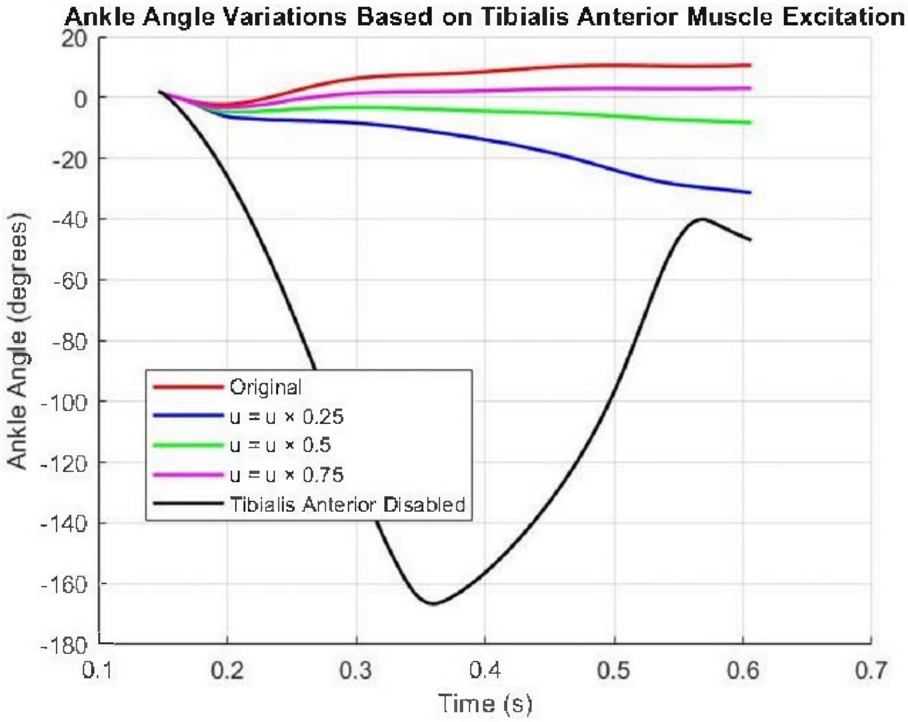

Figure 33. Tibialis Anterior muscle in ankle angle for different excitation values of the muscle.

4.2. Simulations of the modified model Leg6dof9musc in the stance phase

4.2.1. Hip Flexion and Knee Angle: How the Psoas, Rectus Femoris, and Vastus Intermedius affect the joint moments

The psoas muscle is the main flexor and is actively associated with hip flexion, which is something one can figure out by noticing its position in the body (Figure 34). This muscle plays a crucial role in overall lower limb movement and a pathology might be either a reduced activation or an overactivation of it, both of which are mostly caused by lower back pain[55]. The activation of psoas muscle might also differ due to alternative positions and exercises. Below there is a representation (Figure 35) of the range of psoas muscle activations for different excitation values, which indicates a possible pathology to the specific area.

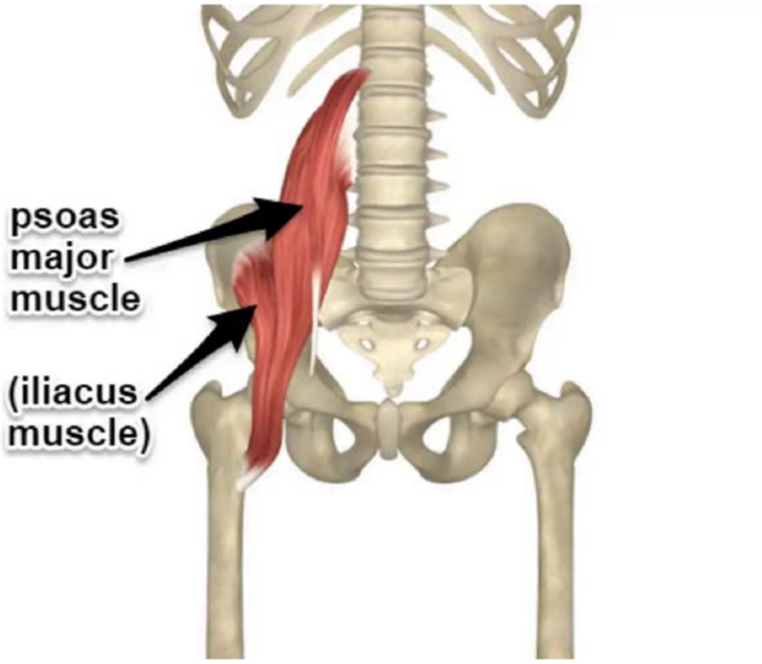

Figure 34. Psoas muscle.

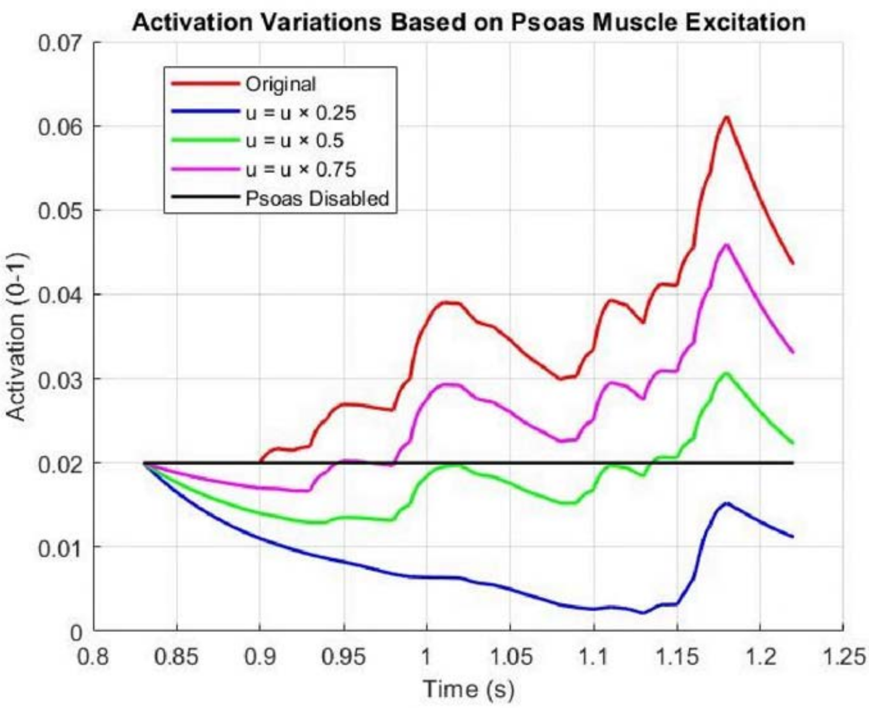

Figure 35. Psoas muscle activations for different excitation values.

Psoas might not affect each joint moment separately, but it does impact the overall body posture and the activity of walking. The following graphs express how the joint moments and more specifically the hip flexion and knee angle are altered due to reduced psoas excitation and as a result different activation dynamics for the muscle. The simulations refer to the stance phase of gait cycle and have been produced through multiple simulations and with the use of the Residual Reduction Algorithm and Computed Muscle Control methods.

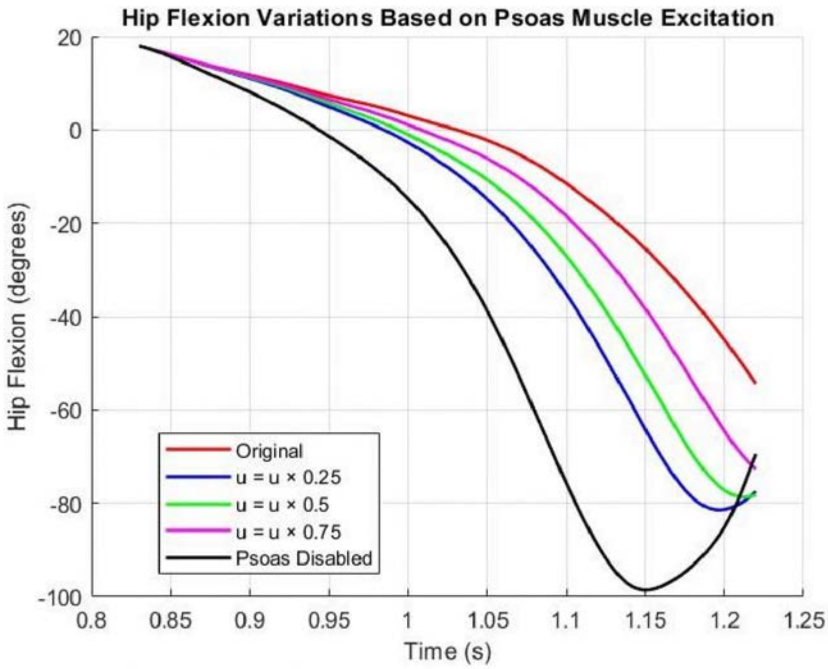

Figure 36. Psoas muscle in hip flexion for different excitation values of the muscle.

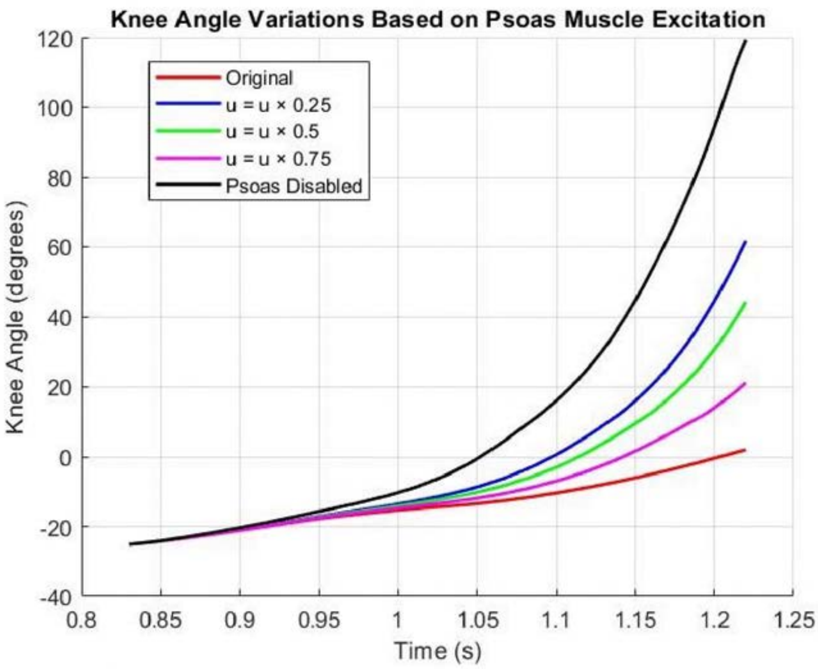

Figure 37. Psoas muscle in knee angle for different excitation values of the muscle.

The activity of Rectus femoris (RF) located during the stance had a minimal effect, but in early to mid-stance (Figure 38) the lengthening RF accelerated the knee (the RF is important during the transition from stance phase to swing phase to control knee flexion) and hip into extension, which opposed swing initiation[56]. In the following simulations, during the stance gait phase, some pathologies appear in the RF. There are reduced excitations of the muscle by decreasing the factor as shown in each case, starting at 75% of normal, continuing at 50%, decreasing to 25% and finally to a completely dead muscle. All pathological conditions are compared to the healthy model developed using the Residual Reduction Algorithm and Computed Muscle Control.

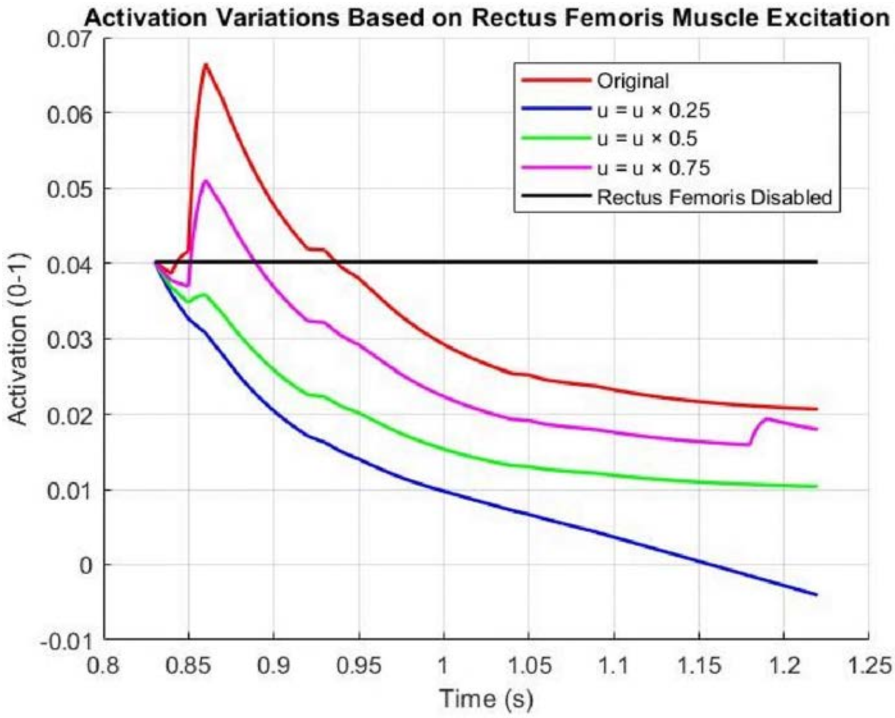

Figure 38. Rectus Femoris muscle activations for different excitation values.

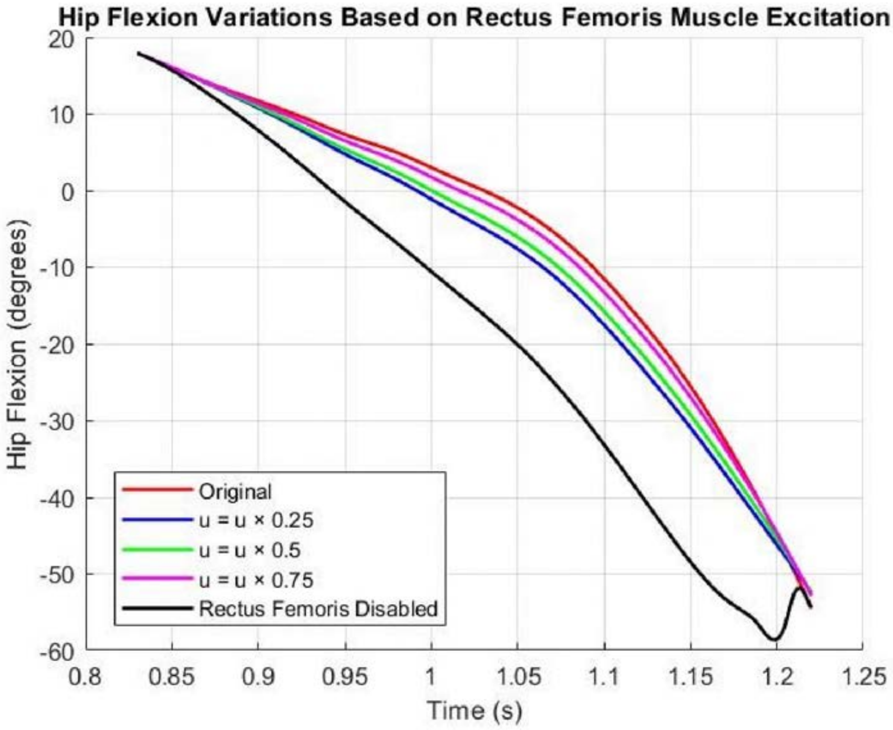

Figure 39. Rectus Femoris muscle in hip flexion for different excitation values of the muscle.

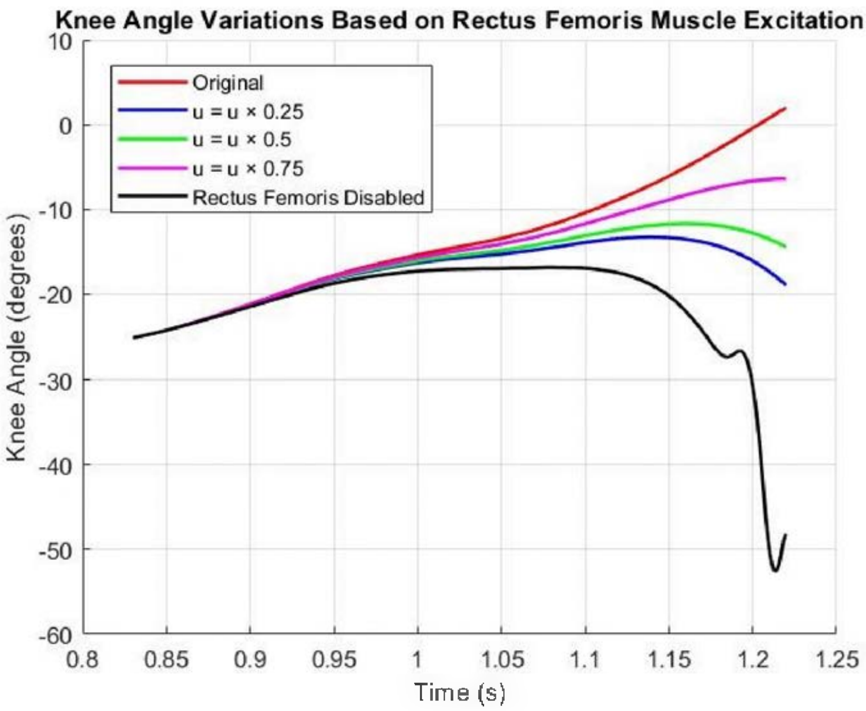

Figure 40. Rectus Femoris muscle in knee angle for different excitation values of the muscle.

Activity of the Vastus Intermedius (VI) during hip flexion might contribute to stabilizing the knee joint as an antagonist and might help to smooth knee joint motion. During the middle of the stance phase of running, when the hip and knee are flexed underneath the trunk, EMG activity of the Vastus Intermedius reached a specific percent of the activity observed in maximal knee extension[57]. This suggests that flexion of the hip joint evokes VI activity and thus, it is of paramount importance to analyze the specific joint moments to see how a pathological vastus intermedius simulation affects activities such as walking.

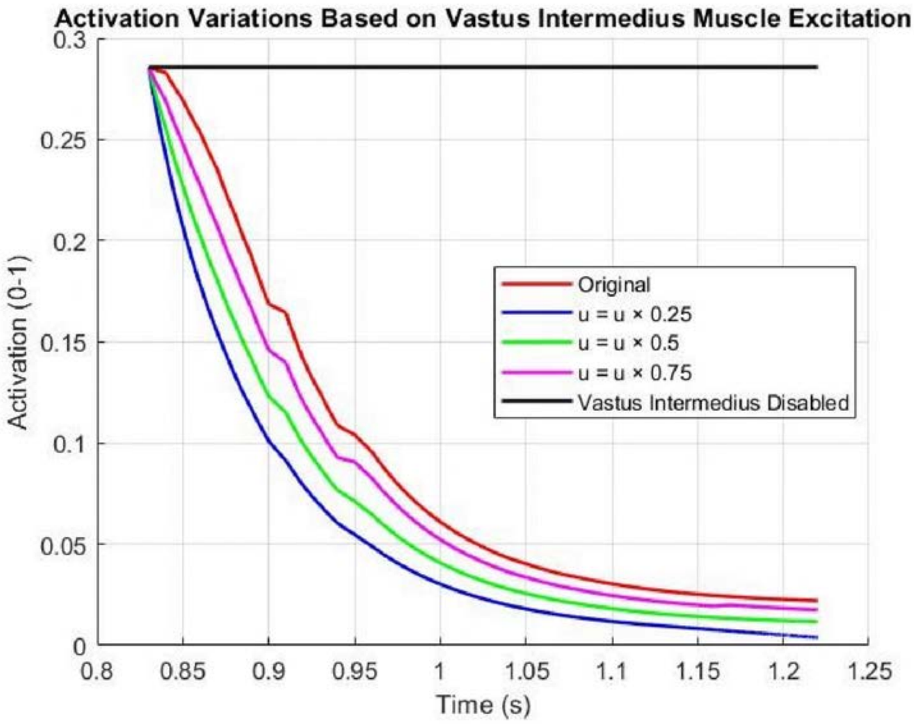

Figure 41. Vastus Intermedius muscle activations for different excitation values.

Vastus Intermedius generates energy during the early stance phase of the gait cycle[58]. For this reason, the muscle activation in each case decreases after the early stage of stance, as the muscle, together with the other quadriceps, flexes the knee to stabilize the leg and prepare for the swing phase.

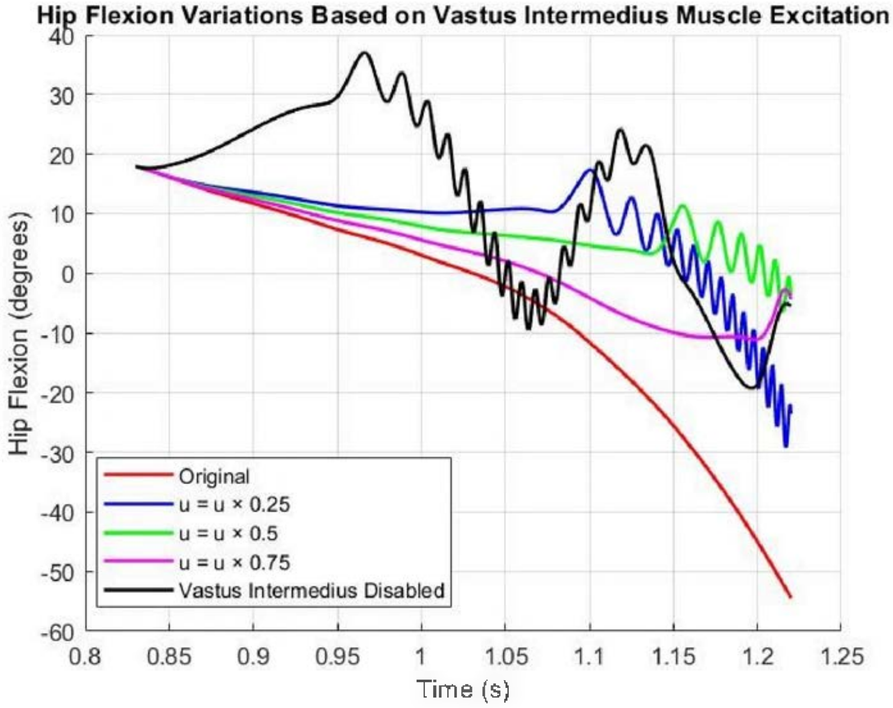

Figure 42. Vastus Intermedius muscle in hip flexion for different excitation values.

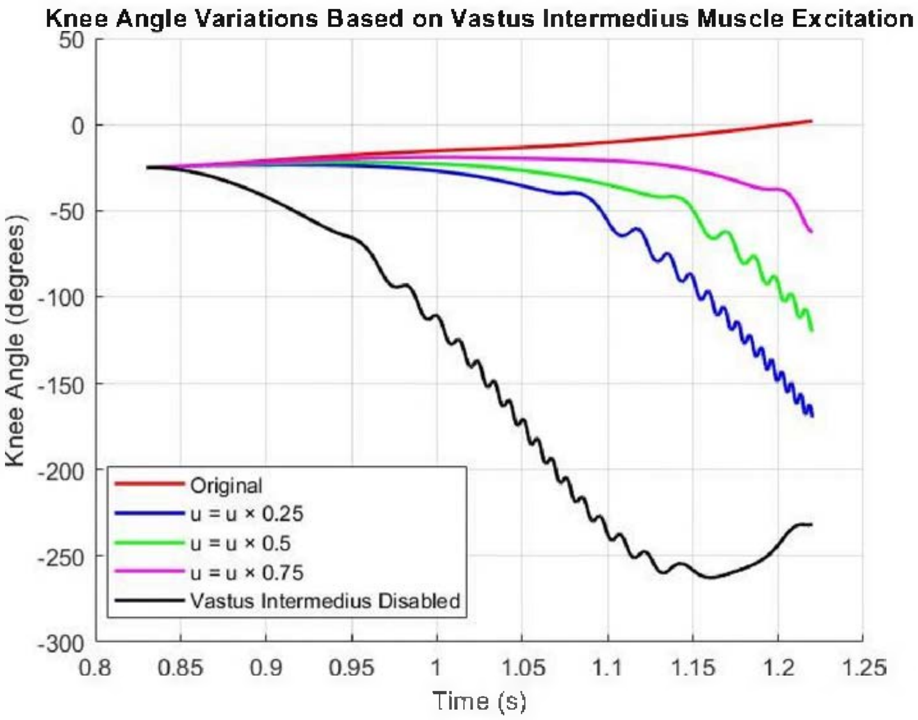

Figure 43. Vastus Intermedius muscle in knee angle for different excitation values.

In order to stabilize the knee joint, VI needs to contribute to the hip flexion as an antagonist. It is helpful when the knee motion needs to smooth and it happens again during early stance. This explains why the curves, when having a much lower excitation value than the original (healthy) musculoskeletal model, appear to have these oscillations both in hip flexion and in knee angle figures, right after the VI has finished assisting the leg.

4.2.2. Tibialis Anterior Muscle Pathology During The Stance Phase OF Gait Cycle

Tibialis anterior functions to stabilize the ankle as the foot hits the ground during the contact phase of walking (eccentric contraction) and acts later to pull the foot clear of the ground during the swing phase (concentric contraction). Therefore, tibialis anterior muscle activity can influence rearfoot motion during the stance phase of walking and contribute to the control of gait through its activation[59]. For this reason, the activations according to the pathological excitation values, alongside with the joint moments, are presented below.

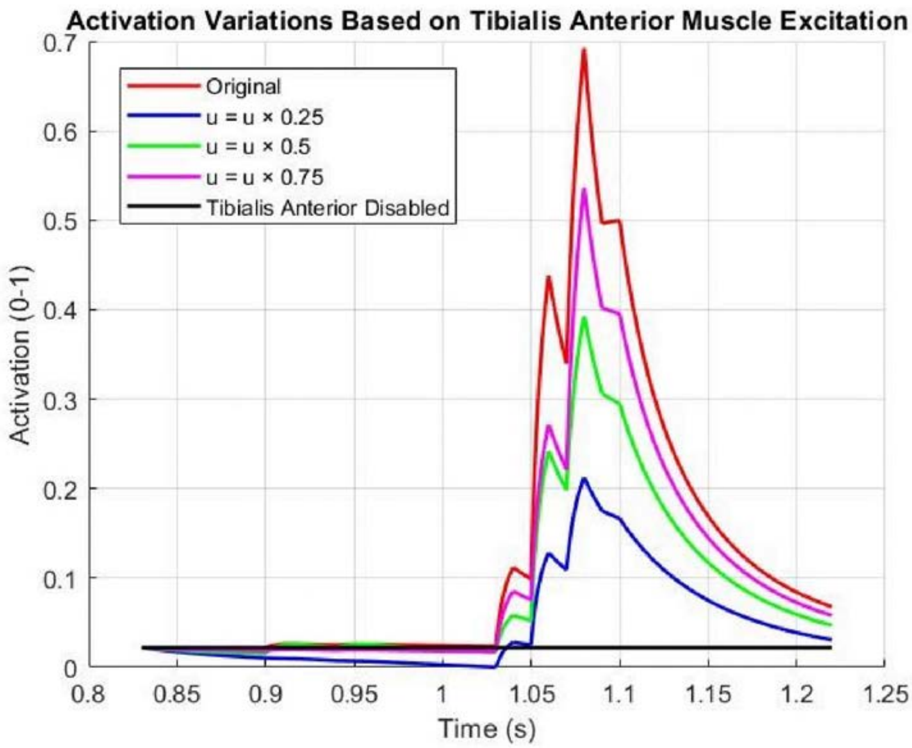

Figure 44. Tibialis Anterior muscle activations for different excitation values.

During the stance phase, the tibialis anterior primarily acts to eccentrically control plantarflexion right after heel strike (initial contact), which prevents the foot from slapping the ground too forcefully. However, due to the fact that the tibialis anterior has minimal contribution, the activation of the muscle decreases gradually and that may as well be observed in Figure 44. The plantarflexors then take over to support weight-bearing and propulsion and the tibialis anterior quits supporting the foot and therefore there are visible oscillations in Figure 45.

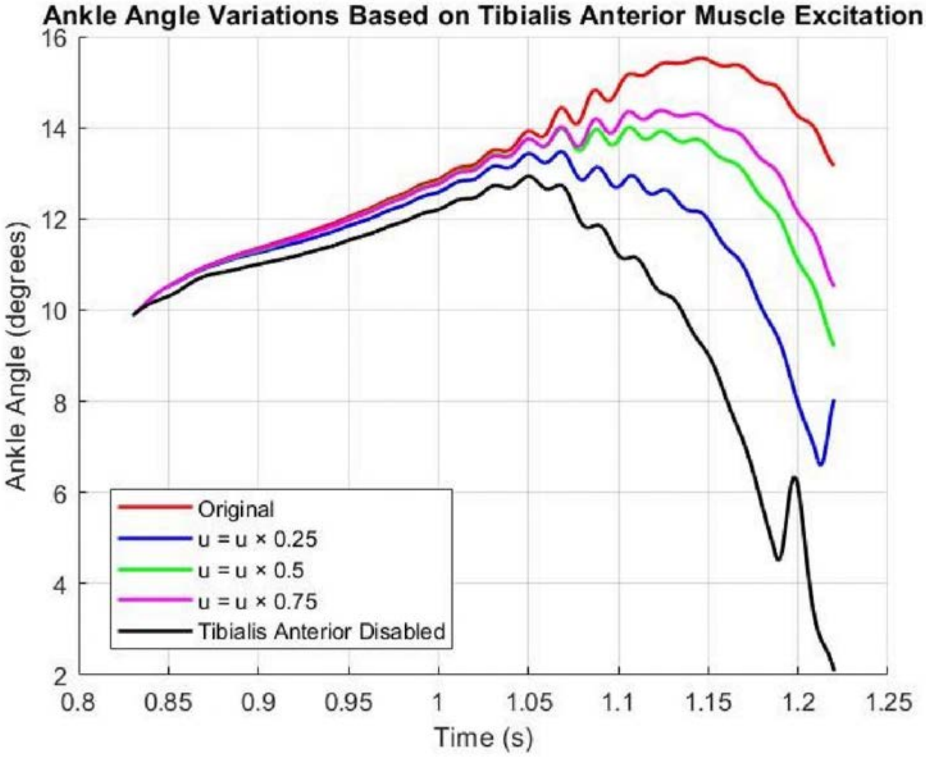

Figure 45. Tibialis Anterior muscle in ankle angle for different excitation values of the muscle.

4.3. ToyDropLanding Simulations with an AFO

AFOs or "Ankle Foot Orthoses" are plastic splints which are made to keep feet and ankles in a good position for standing and walking and can be worn on one foot or both feet[60]. Ankle-foot orthosis (AFO) has been used as a representative solution to improve gait quality by decreasing the ankle kinematic asymmetry in patients with foot drop. There are three main kinds of AFOs: flexible (it is worn inside your shoe and provides foot support, but not medial and lateral support), rigid (Rigid orthotic devices are designed to control function and are used primarily for walking or dress shoes. They are often composed of a firm material, such as plastic or carbon fiber[61]), and jointed (it is a type of Ankle-Foot Orthoses that is characterised by one or more joint mechanisms[62]). A new addition to the already existing AFOs is the hybrid ankle-foot orthosis that uses a pneumatic artificial muscle[63], which not only assists the user, but it also helps the muscle to maintain its functionality, unlike Functional Electrical Stimulation, which gradually weakens the muscle.

The **ToyDropLanding.osim** model is an example provided in OpenSim to demonstrate the simulation of dynamic activities, such as landing from a jump. This model represents a simplified musculoskeletal system and is used for illustrating fundamental biomechanical concepts and testing simulation tools. It refers to a drop-landing scenario that provides insights into joint kinetics, muscle forces, and the coordination required for shock absorption and stabilization upon impact. In this study, the model was used to evaluate the likelihood of an ankle inversion injury during a drop landing on a sloped surface. It features 23 degrees of freedom and 70 muscle-tendon actuators, offering a simplified yet effective framework to analyze the biomechanics of impact and stabilization.

There were different examples to evaluate the landing of the model. The first one was a musculoskeletal body with no assistive device, contradictory to simulation two, where the model had an Ankle-Foot Orthosis (AFO).

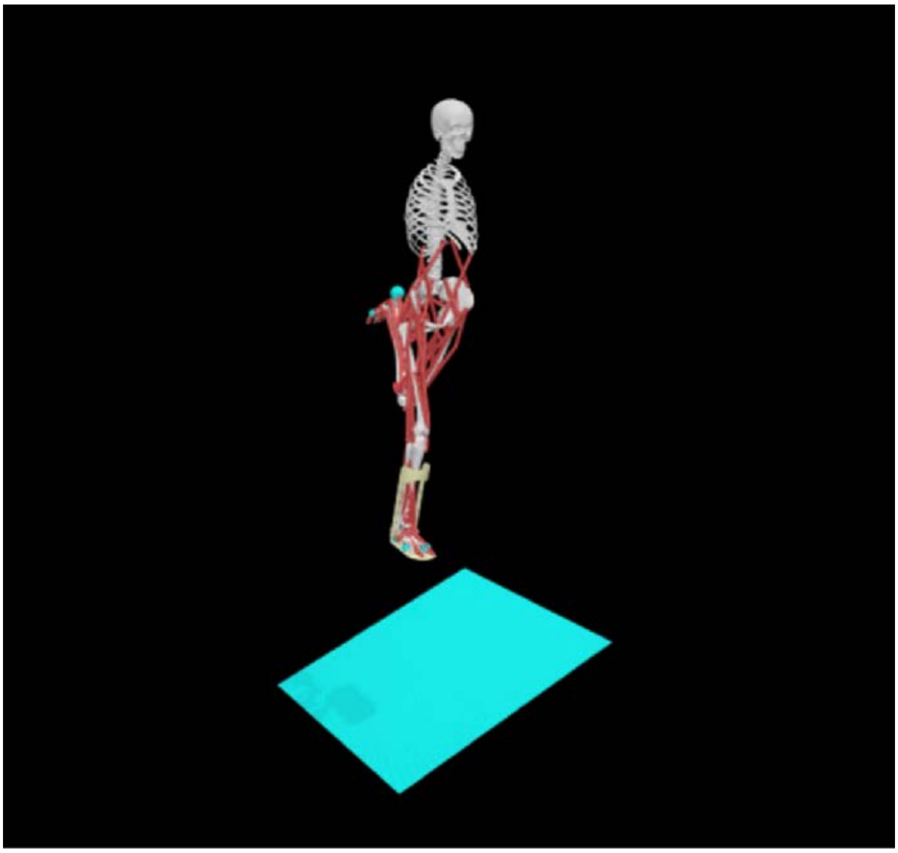

Figure 46. ToyDropLandingAFO model

4.3.1. Unassisted Model Simulation

In the first simulation experiment, the ToyDropLanding.osim model was used to analyze subtalar angles during a drop landing without any orthotic intervention. The results, as depicted in Figure 47, showed that the peak subtalar angle reached 41.058 degrees at 0.363 seconds. Since this angle significantly exceeds the threshold of 25 degrees (0.436 radians), it indicates a high likelihood of ankle inversion injury during this unassisted drop landing. This outcome highlights the inherent instability of the ankle joint under such dynamic conditions, emphasizing the need for external support mechanisms to mitigate injury risks. There is also the curve that depicts the ankle angle during the simulation, so that it can later be used for comparison.

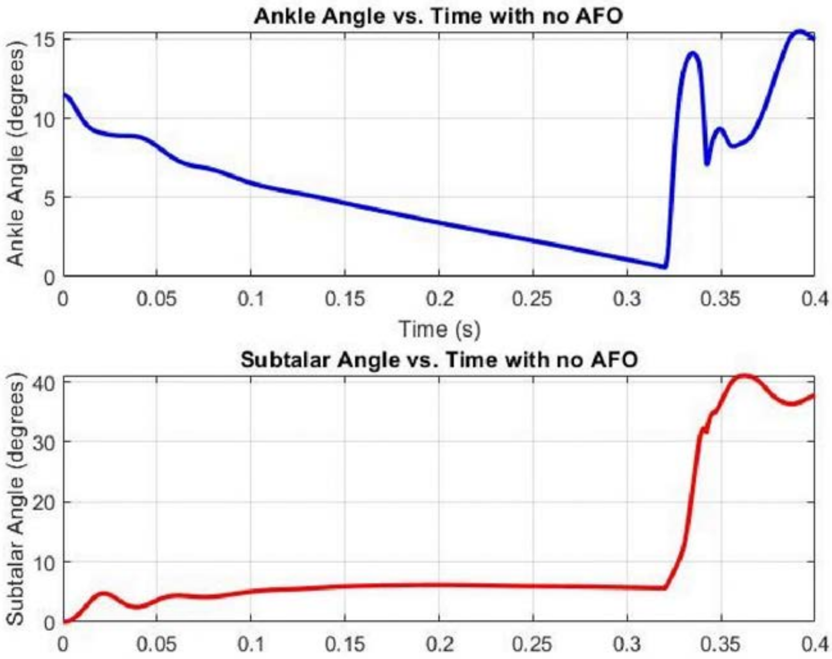

Figure 47. Drop landing on a sloped surface with no orthotic (unassisted)

4.3.2. Simulation with the use of SoftAFO

The second experiment introduced a modified model, **ToyDropLandingAFO.osim**, which incorporates an ankle-foot orthosis (AFO) on the right foot. This version of the model includes additional components such as the AFO footplate and cuff, along with corresponding joints. When the soft AFO was applied, the subtalar angle during landing decreased compared to the unassisted model, as illustrated in Figure 48. This reduction in peak angle suggests that wearing a soft AFO enhances ankle stability, but it still indicates a high likelihood of ankle inversion injury.

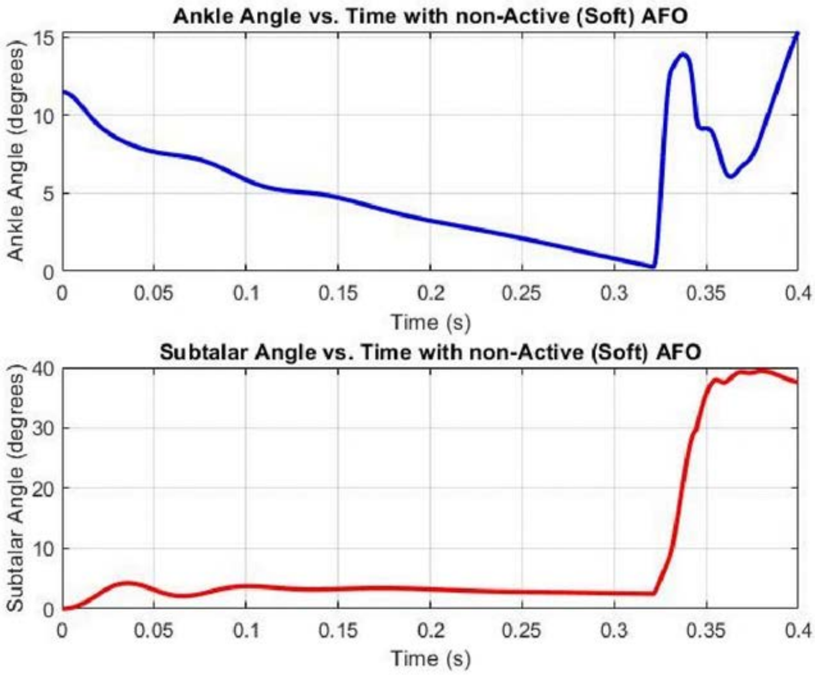

Figure 48. Drop landing on a sloped surface with a passive AFO

In order to better understand the difference between the two simulations, there are plots with the results combined and compared below. As it is shown, the peak angle when using a soft AFO is less than 40 degrees, which indicates that there is a less likely chance for someone to be injured.

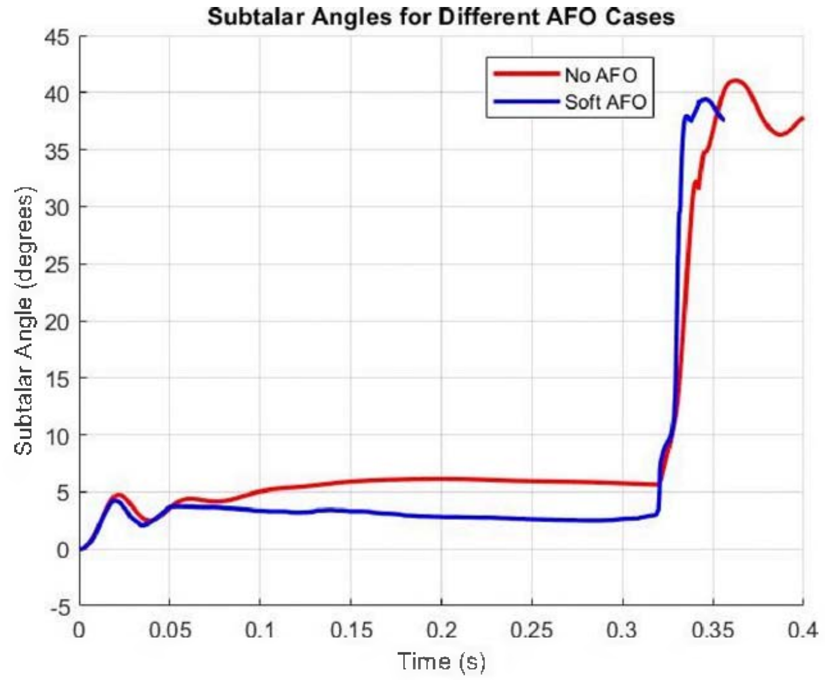

Figure 49. Comparison between the two simulations for the subtalar angle

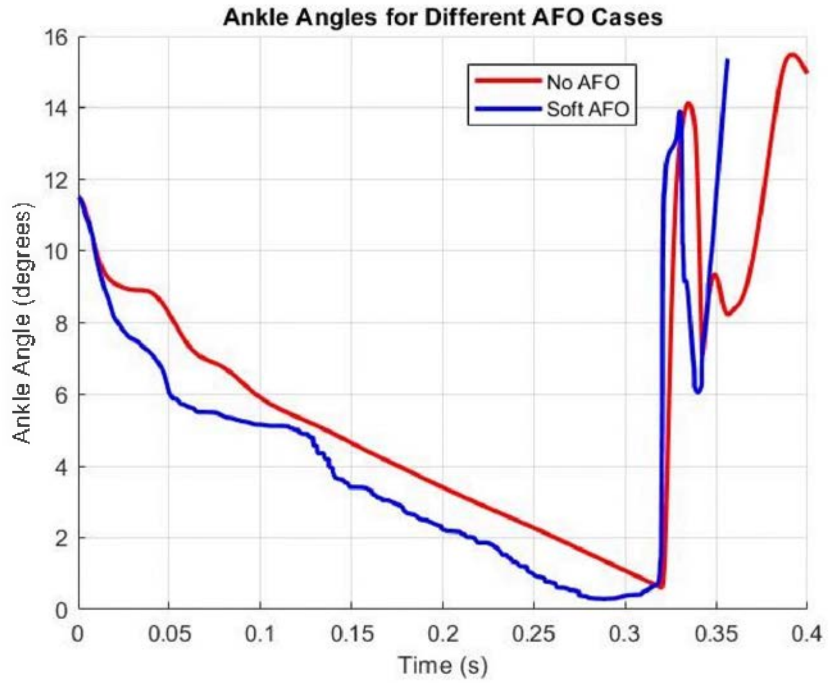

Figure 50. Comparison between the two simulations for the ankle angle

Chapter 5: CONCLUSIONS

5.1. Study Evaluation

In conclusion, this study presents a comprehensive evaluation and comparison of several methods and tools that are used to simulate muscle pathologies. Each simulation case offers new insight into the development of the particular muscle pathology and the complication it causes in the specific phase of gait. This is an extremely useful tool for quality upgrading in the design of assistive devices and the general enhancement of biomechanics.

In this study a very reliable and helpful software, OpenSim, was used, which in combination with all the documentation it offers, made it possible to simulate several examples of pathologies, thus demonstrating how valuable a tool it is for its use as a basis for the design of musculoskeletal systems and AFOs.

Key findings from this research highlight how a digital twin of a musculoskeletal with Drop Foot (or any other condition caused by muscle pathologies) can be fully used to accelerate the development of a suitable, personalised and effective lower limb orthotic. This is achieved by proposing a design method that focuses more on the swing phase of the gait cycle during which Foot Drop is observed and saves time and energy because it virtually simulates the AFO, which can then be matched with real data and systems more accurately.

5.2. Research Outlook

There are several directions for future research based on the findings of this work:

- Extensive clinical trials to validate the performance of these methods in real clinical settings are necessary. This will involve working with healthcare professionals and institutes to test the methods in real-life data during procedures such as rehabilitation treatments following injury or health complications.

- Another path that can be followed involves the development, initially in a virtual environment and then in the real world, of an active orthotic, which will be adapted, based on measurements from pressure sensors, to the user's needs for better support of the lower limb. This can be combined with the integration of pneumatic muscles, thus creating a hybrid model that avoids the use of FES, which until now has been considered a novelty because, according to recent research, it seems to weaken the muscle more than it strengthens it.

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APPENDIX: MATLAB Code for Simulating the Differential Equation

This appendix presents a MATLAB script used to simulate and analyze a differential equation that models a dynamic system's response to an input signal. The equation describes how the system's state evolves over time based on an excitation function e(t) and an input function u(t). The general form of the equation is:

$$\frac{du(t)}{dt} = [\frac{1}{\tau act} \cdot (\beta + (1 - \beta)e(t))] \cdot u(t)\frac{1}{\tau act} \cdot e(t)$$

where:

- u(t) represents the system's state variable.
- e(t) is an excitation function modeled as an exponential decay.
- τact is a time constant affecting the system's response speed.
- β is a parameter controlling the influence of e(t).

```
% MATLAB script to simulate and analyze the differential equation: du(t)/dt=
[(1/τact) * (β + (1 -β)e(t))]*  u(t)*( 1/τact)* e(t)


% Parameters - they can be modified so that one can observe their effects

a1 = 1; % Coefficient for first-order term

a2 = 2; % Coefficient for second-order term

b = 0.5; % Coefficient for input u(t)

tau = 0.1; % Time constant for e(t)


% Time vector

t = 0:0.01:10; % Time from 0 to 10 seconds with 0.01s intervals


% Input u(t) (e.g. a sinusoidal function)

u = sin(2*pi*0.5*t); % Frequency = 0.5 Hz


% Definition of e(t) as a simple exponential decay

e = exp(-t/tau);
```

```
% Differential equation: a2*y''(t) + a1*y'(t) + y(t) = b*u(t)
% Convert to system of first-order equations for numerical solution
differential_eq = @(t, y) [y(2);  % y(1) = y(t), y(2) = y'(t)
                           -a1*y(2)/a2 - y(1)/a2 + b*u(round(t/0.01)+1)/a2];

% Initial conditions [y(0), y'(0)]
y0 = [0; 0];

% Use of ode45 to solve the system
[t_out, y_out] = ode45(differential_eq, t, y0);

% Extract the solution
y = y_out(:, 1); % y(t)
dy = y_out(:, 2); % y'(t)

% Plot the results
figure;

% Plot input u(t)
subplot(3,1,1);
plot(t, u, 'b', 'LineWidth', 1.5);
title('Input Signal u(t)');
xlabel('Time (s)');
ylabel('u(t)');
grid on;

% Plot e(t)
subplot(3,1,2);
plot(t, e, 'r', 'LineWidth', 1.5);
title('Excitation Signal e(t)');
```

```
xlabel('Time (s)');
ylabel('e(t)');
grid on;

% Plot system output y(t)
subplot(3,1,3);
plot(t, y, 'k', 'LineWidth', 1.5);
title('System Output y(t)');
xlabel('Time (s)');
ylabel('y(t)');
grid on;
```