BACHELOR'S THESIS

Bachelor of Mechanical Engineering

ESTIMATION OF MUSCLE FORCES FOR TREADMILL GAIT TRIALS

Report and Annexes

Author: Deepak Prabakar

Director: Gil Serrancolí Masferrer

Date: June 2017

Abstract

Subject-specific musculoskeletal models and computational walking models in general have come a long way in improving clinical treatment of walking disorders. But more accurate predictions of such forces and their locations can help in designing knee replacements to recover normal walking.

This project aims to predict muscle forces for treadmill gait trials when varying speeds and analyse the differences in those muscle forces. Three walking trials with different speeds were studied in this project. Two approaches were used to predict forces. In Approach A, knee contact force information was used as input of the algorithm, and in Approach B, these data was used only to validate the results. An OpenSim musculoskeletal model of the right leg was used to obtain inverse kinematics and inverse dynamics data, and muscle length and moment arms.

The algorithm to estimate muscle forces consisted of a two-level nested optimization. The outer level optimizes the time-independent parameters and the inner level optimizes the time-dependent parameters. Kinematics and ground reaction force data used in this project were obtained from the fourth grand challenge competition to predict in vivo knee loads.

Muscle force estimation values obtained in Approach B (usual case) were significantly different from Approach A (unique case) for most muscles. The results from this study reinforce results of previous studies. Medial and lateral force distribution was also analysed. The muscles with the maximum and minimum differences in mean forces for the three different speeds were identified and possible reasons for these differences were discussed.

Acknowledgements

First of all, I would like to thank my parents for all the amazing chances that they have encouraged me to utilize. The support that they have provided is what has brought me anywhere in life.

Next, I have to thank Gil Serrancolí Masferrer, my mentor for having agreed to guiding me on my bachelor thesis even after my late request. I could not have asked for a friendlier, more helpful mentor. He took time off his weekends to answer the many questions I had.

I would like to express my gratitude towards Sílvia Urban and Mati Torné from the International Relations Office in EEBE for being so warm and entertaining my request for an extension of deadline to submit the documentation.

I also need to thank the staff from Amrita Center for International Programs; Anandi Rao, Sujatha Ram, Reshmi Ravi, Dr. Maneesha V. Ramesh and others for egging me on and encouraging me to take up this student exchange program.

I cannot finish without thanking all the staff at the Mechanical Engineering Department, Amrita School of Engineering, Bangalore for the knowledge that they have imparted on me. Special thanks to the head of the department, Dr. Nagaraja S.R. for recommending me for this student exchange program and facilitating this opportunity.

Glossary

addbrev – adductor brevis
addlong – adductor longus
addmagProx – adductor magnus proximal
addmagMid – adductor magnus middle
addmagDist – adductor magnus distal
addmagIsch – adductor magnus ischial
bflh – biceps femoris long head
bfsh – biceps femoris short head
edl – extensor digitorus longus
ehl – extensor hallucis longus
fdl – flexor digitorum longus
fhl – flexor hallucis longus
gaslat – gastrocnemius lateralis
gasmed – gastrocnemius medialis
gem - gemeli
glmax1 – gluteus maximus superior
glmax2 – gluteus maximus middle
glmax3 – gluteus maximus inferior
glmed1 – gluteus medius anterior
glmed2 – gluteus medius middle
glmed3 – gluteus medius posterior
glmin1 – gluteus minimus anterior
glmin2 – gluteus minimus middle
glmin3 – gluteus minimus posterior
grac - gracilis
pect - pectineus
perbrev – peroneus brevis
perlong – peroneus longus
pertert – peroneus tertius
piri – periformis
quadfem – quadratus femoris
recfem – rectus femoris
sart – sartorius
semimem – semimembranosus
semiten – semitendinosus
tfl – tensor fascia latae
tibant – tibialis anterior
tibpost – tibialis posterior
vasint – vastus interior
vaslat – vastus lateralis
vasmed – vastus medialis
J – muscle moment

f – muscle force
l_m^0 – optimal muscle fiber length
l_t^s – slack length of tendon
l_M – length of muscle
l_T – length of tendon
l_{MT} – length of muscle-tendon
α – pennation angle
a – activation
ma, r – moment arms

Index

1. Preface

1.1. Origin of the Study

A study about predicting muscle forces for a subject on different over ground gait trials at self-selected speeds was previously carried out [1]. This study used inverse dynamics, optimisation techniques and EMG data to estimate forces and validated them using experimental data. With a method to estimate forces established, I was intrigued by how muscle forces change for different speeds. The most basic action apart from walking at a constant speed over the ground would be walking at varying speeds. I felt like it would be interesting to see which muscles see a hike or dip in magnitude of force when the walking speed changes.

1.2. Motivation

Biomechanics is a field that I had never been exposed to as a student in my home university. I found it hard to accept that in a world where we are sending reusable rockets into space, we are not able to predict the forces that are right inside our own body. There are millions of people suffering from arthritis and other knee disorders who would be directly benefitted by improved clinical treatment resulting from a more comprehensive knowledge of knee contact forces. I consider this project as my contribution to improving a science that has the potential to improve millions of lives.

1.3. Previous Requirements

To carry out this project, knowledge regarding the basics of the human anatomy and biomechanics was required. OpenSim, a software platform for modeling humans, animals, robots, and simulating their interaction and movement, was used to compute data regarding inverse kinematics and inverse dynamics. Then, this data was loaded onto a MATLAB script to run an optimization algorithm. A learning period to acquire the knowledge to use these two software was also required.

2. Introduction

Millions of people around the world are affected by osteoarthritis and other knee related disorders every year. The current clinical treatments in place are largely subjective and this causes ambiguity in the treatments prescribed by clinicians. Treatment of knee disorders means alleviation of joint pain and in extreme cases, restoration of walking. Medical science is crucial for clinical treatment, but engineering has the means to make the process more effective and accurate.

2.1. Scope of the study

The knowledge of in-vivo muscle and joint force values would help doctors in improving rehabilitation treatments, designing knee replacements and other prosthesis, or orthosis for people who suffer from walking disabilities. We cannot directly measure in-vivo muscle forces as it would be invasive. For this reason, computational methods have to be used to estimate these forces.

Subject specific musculoskeletal models play an important role in muscle force prediction. By inputting the ground reaction forces and kinematics data from gait trials, we can calculate resultant joint loads using inverse dynamics. But, there is indeterminacy when calculating muscle forces since the human body has more muscle actuators than degrees of freedom. This problem can be overcome by assuming that the body activates its muscles following an optimal criterion. Inverse dynamics forces can be loaded onto a static optimization program that minimizes the cost function frame by frame to estimate in-vivo muscle forces.

2.2. Objectives of study

Three heel strike to heel strike gait trials for varying treadmill speeds were identified and static optimization were used to estimate forces, moments and actions in the lower limb. The original code [1] was also modified to include simultaneous computation of all three gait trials. Two methods of study were used to estimate the contact forces. The first method makes use of the experimental knee contact forces to calibrate model parameters within the outer level (Approach A). The second method does not take into consideration the experimental knee contact forces (Approach B). The results of the optimization are expected to answer questions regarding the behaviour of muscles when varying walking speeds. The goal would be to identify the muscles which are affected more by a change in speed.
The following aspects of treadmill gait trials are also discussed and analysed:

- Solving the muscle force sharing problem in treadmill gait trials when varying speed with a two-level optimization

- Analyze the difference in results when knee contact force information is tracked or not at the outer level

- Analyze if muscle forces are different among different walking speed trials, and which muscles have the main differences

3. Anatomy of the lower limb

3.1. Introduction

The lower limb is the most important element in the human body for locomotion. Pelvis, thigh, shank and foot along with the muscles that actuate them work together to carry out movement. The lower limb comprises a significant portion of body mass. Leg bones are also quite strong because these bones need to be able to support the whole body's weight. The lower extremity, containing the ankle, knee, and hip joints, has been of great interest to biomechanics researchers.

3.2. Skeletal structure

In total, the lower limb contains 62 bones. These include 10 pelvis and leg bones, 14 at the ankle and 38 bones in the feet. The longest bone in the body, the femur is located in the leg. Some of the largest joints in the body are also located in the lower extremity. The hip joint, knee joint and the centre of the ankle joint, all lie in a straight line when the body is upright. This represents the mechanical longitudinal axis of the leg, the Mikulicz line. The following contains brief descriptions for some of the most important bones in the lower extremity.

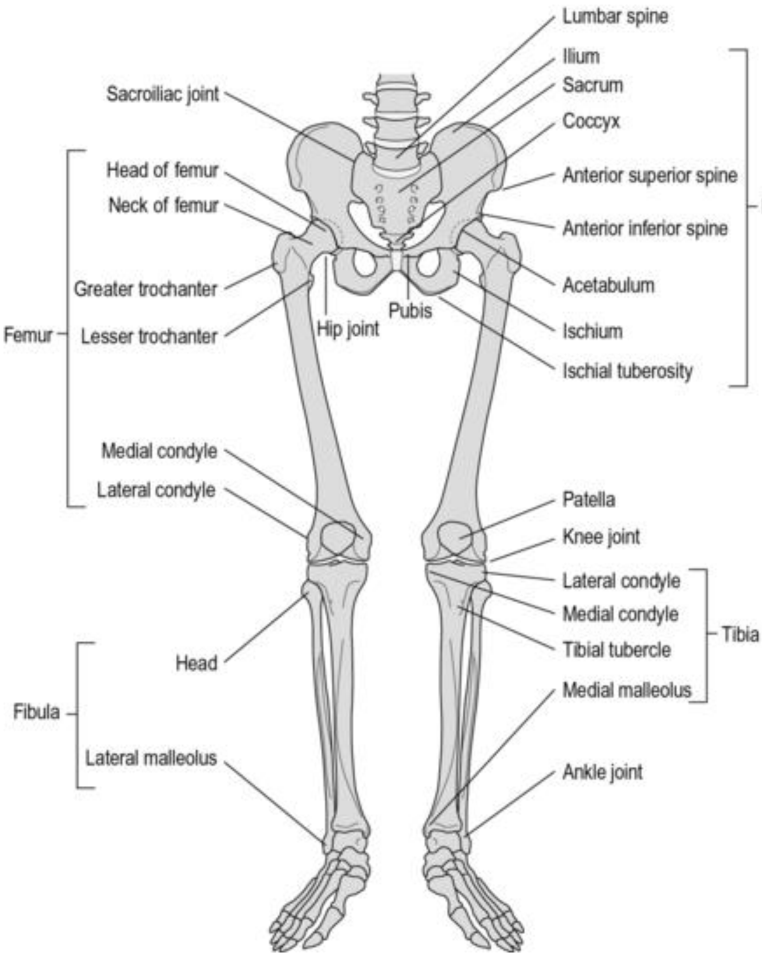

Figure 3.1. A representation of the bones in the lower limb [2]

3.2.1. Femur

The femur is the main bone of the thigh connecting the hip to the knee. This is the longest and strongest bone in the human body. The rounded top end of the femur articulates in the acetabulum of the pelvic bone, the lower end articulates with the patella. Since the femur is the only bone in the thigh, it serves as an important attachment point for all the muscles that exert their forces over the hip and knee joints.

3.2.2. Tibia

The tibia, commonly known as the shinbone is one of the two bones connecting the knee to the ankle. The tibia is the strongest and largest of the two bones in the lower leg. It is also the second longest bone in the human body. The tibia is part of four joints: the knee, ankle, superior and inferior tibiofibular joints.

3.2.3. Fibula

The fibula is the smallest bone in the lower leg. In proportion of length to width, the fibula is the slenderest of all the longest bones. The fibula supports a very little portion of the body weight. Its main purpose is to provide space for attachment of muscles. It has grooves for certain ligaments which gives them leverage.

3.2.4. Patella

The patella, commonly known as the knee cap is a thick circular-triangular bone that articulates in the femur and covers and protects the anterior surface of the knee joint. It is the largest bone in the body that is covered completely by muscle or ligament. The patella increases the leverage that the tendon can exert on the femur by increasing the angle at which it acts.

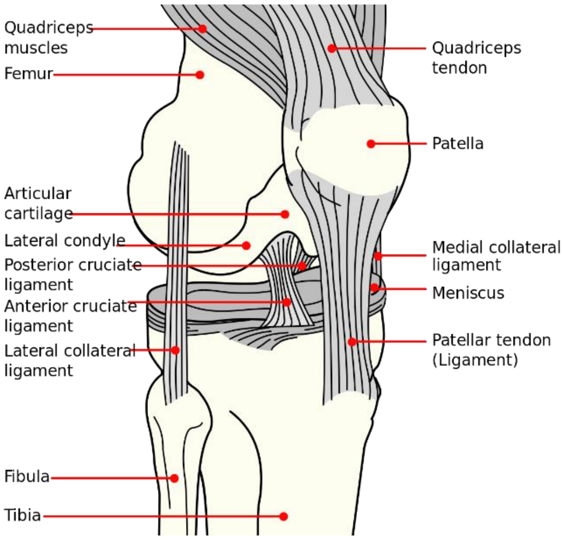

Figure 3.2. Representation of the patella and other components of the knee joint [3]

3.3. Muscles

Most of the muscles in the leg span long distances. The muscles contract and relax to exert force on the bones and create movement. However, there are smaller muscles whose roles are helping larger muscles, stabilize joints, help rotate joints and facilitate other fine-tuned motion. There are several groups of muscles working at the pelvis, thigh, calf and foot..

3.4. Anatomical Terms of Motion

Anatomical terms of motion are describing motion of organs, joints, limbs and other sections of the body using specific anatomical terms. Anatomical motion can also be classified based on the type of movement

- Rectilinear motion
- Rotational motion

Many general movements also have an opposite movement, also known as an antagonistic movement. The term antagonistic is also used in this study to describe muscles that perform antagonistic movements. All these motions are defined in anatomical planes they occur in.

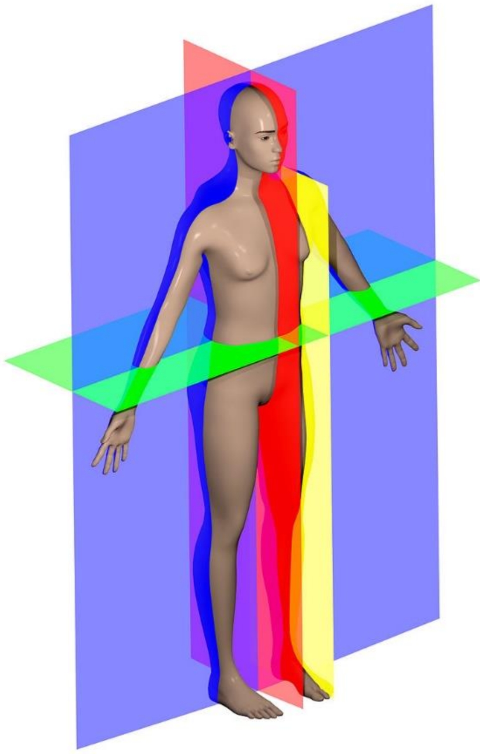

Figure 3.3. Planes on the basis of anatomy [10]

- Sagittal plane is the one that divides the body from left to right (medial and lateral) (in red)
- Parasagittal planes are those that are in parallel to the sagittal plane (in yellow)
- Coronal plane is the one that divides the body into the front and the back (anterior and posterior) (in blue)
- Transverse plane is the one that divides the body into the top and bottom (superior and inferior) (in green)

Some of these anatomical terms of motion (degrees of freedom of the joints) that are used in this study are described briefly in the coming sections.

3.4.1. Flexion and Extension

Flexion and extension refers to increasing and decreasing the angle between two body parts:

Flexion decreases the angle between two body parts. For example, when the knee flexes, the lower leg goes backward towards the thigh.

Extension increases the angle between two body parts. E.g., when the knee extends, the leg straightens out.

3.4.2. Abduction and Adduction

These terms are used to describe movements towards or away from the midline of the body.

Abduction is moving a body part away from the midline of the body. E.g., abduction of the hip would be moving the legs away from the body like when doing a split.

Adduction is moving a body part towards the midline of the body. E.g., adduction of the hip would bring the legs closer to each other.

3.4.3. Rotation

Rotation may be internal or external, internal being rotation towards the body's midline and external being rotation away from the body's axis.

3.4.4. Medial-Lateral Translation

Medial is used to refer to structures closer to the center (closer to the sagittal plane). Lateral is used to refer to structures away from the center. Medial-lateral translation happens perpendicular to the sagittal plane.

3.4.5. Superior-Inferior Translation

This is in reference to the vertical axis of the body. Superior is used to refer to something relatively higher and inferior is used to refer to something relatively lower. Superior-inferior translation happens perpendicular to the transverse plane.

3.4.6. Anterior-Posterior Translation

Anterior refers to anything that is directed towards or situated at the front and posterior is used to refer anything that is directed towards or situated at the back. Anterior-posterior translation occurs perpendicular to the coronal plane. Medial-lateral, superior-inferior and anterior-posterior translations occur in many joints but they can approximated to zero because they are in the order of a few millimetres and are hence negligible. This is also true for some joints with internal rotations.

3.5. Osteoarthritis and the necessity for clinical treatment

Arthritis is a general term that is used to call the inflammation of joints. Osteoarthritis is one of the most common degenerative joint disease. It is associated with a breakdown of cartilage in joints and can occur in any joint in the body. It occurs in important joints such as the knees, hips and spine because these joints support most of the body weight and transmit high forces. It could also affect other smaller joints such as fingers, neck and toes.

The cartilage is a tough, elastic, fibrous tissue that covers the end of normal bones. Cartilage acts as a solid lubricant and a shock absorber. These qualities come from the rubbery nature of cartilage. Osteoarthritis causes the cartilage to lose its elastic nature, thus also making it more susceptible to damage. With application of high stresses in the affected joint, the condition of the cartilage worsens. After a period of time, the cartilage may completely wear away, reducing the joint’s ability to absorb shocks and move smoothly. When this condition deteriorates further, the cartilage might completely wear away and bones may rub. The pain in the joint also worsens as the condition of the cartilage deteriorates.

Osteoarthritis can be treated by exercise and weight loss and if needed, medications and physical therapy. As the disease deteriorates, the pain worsens and surgery might end up as the final unavoidable option. There are several types of surgery that can treat osteoarthritis, two of them are:

- Joint replacement surgery, also known as arthroplasty replaces a damaged joint with an artificial one that has been created for the patient. This surgery is considered when a person’s quality of life is significantly affected by the disease.
- Joint fusion is taken up when joint replacement is not applicable. The bones making up a joint are fused or welded together which could relieve pain. This type of surgery is applicable only to a few joints.

4. State of the Art

Osteoarthritis affects 9.8% of males and 18% of females over 60 years of age [4]. The knee is one of the joints most affected by the degenerative disease. The maximum forces that are transmitted through the knee are around 1-3 times the body weight in cases of normal walking [5]. Due to this fact, even small malalignments between the femur and tibia could cause major changes in the forces transmitted through the knee. These small malalignments tend to further progress the disease.

As the disease progresses, arthroplasty becomes the most effective way to treat it. Arthroplasty is the surgical procedure through which the surface of a musculoskeletal joint is replaced, remodelled or realigned. This procedure does not come without its own inconveniences. The material that is used to replace bone does not regenerate itself like biological tissue does. Hence, the knowledge of knee contact force values would be a benefit to monitor a certain rehabilitation treatment.

4.1. Measurement of forces

4.1.1. Experimental Measurement

In-vivo joint forces can be measured directly only through specially designed prosthesis. Prosthesis can be designed to have an array of force sensors than can provide data about joint forces.

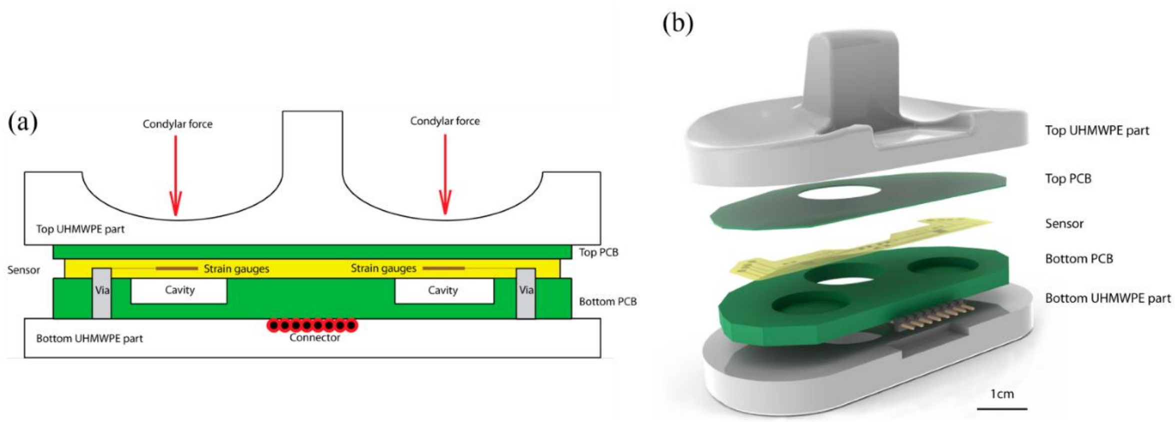

Figure 4.1. A force measuring prosthesis [6]

The disadvantage to this is the fact that the prosthesis needs to be surgically implanted. Since not everyone can be implanted with an instrumented prosthesis, direct measurement of knee contact forces is not always feasible. This necessitates the computation of knee contact forces.

4.1.2. Computer Based Prediction of Knee Contact Forces

Clinical treatment of knee disorders has been subjective in nature for a long time. This means that several treatments can be prescribed to treat one problem. Using objective subject-specific models is a way to address this problem. This allows several iterations to be tested and the optimal design to be identified. Applying the same approach on treating orthopaedic patients is much more complicated because the model has to be unique to every single patient [7]. Muscle anatomy vary from patient to patient.

Subject-specific models have been used while attempting to predict in vivo forces. Inverse and forward dynamics are two approaches to estimate forces involved in a certain movement. On the one hand, inverse dynamics uses experimental motion analysis (kinematics) and external forces from trials to obtain the forces at the human joints. Forward dynamics on the other hand uses muscle activations and forces to predict the kinematics integrating the equations of motion.

One issue faced by computer model based prediction of muscle forces is the indeterminacy. The human body has more muscle actuators than degrees of freedom. In essence, this means that there are an infinite number of combinations of muscle activations that would result in the same resultant joint loads, there is indeterminacy in muscle force calculation.. This problem can be tackled by assuming that the brain activates muscles following an optimal criterion. The problem of indeterminacy can also be overcome by using experimental electromyography data that would show the muscle excitations coming from the central nervous system. This is a top down approach to obtaining the joint forces.

Current computational models are unable to predict knee contact forces with a high accuracy. Hence, experimental data is required to validate the predictions of a computer model. Knowledge of knee contact forces can further predict outcomes like stress, wear, damage and others which cannot be measured in-vivo. Overall, improvements in measurement or computation of knee forces would significantly improve clinical treatment.

5. Methodology

5.1. Experimental Data

All experimental data used in this study has been obtained from the “Grand Challenge Competition to Predict In-Vivo Knee Loads” [7]. The unique purpose of this competition was to provide researchers with a comprehensive data set that allows them to validate their prediction of knee contact forces. The data released included motion capture, ground reaction forces, femur-tibia contact forces from a subject implanted with an instrumented knee prosthesis. These data allows researchers to estimate and validate knee contact forces varying degrees of accuracy.

The subject of the fourth grand challenge competition was an 88 years old male (height: 166 cm, weight: 66.7 kg) with a knee instrumented prosthesis (femoral tray and femoral component). This study involves the estimation of forces for a varying speed treadmill gait trial. In this treadmill trial, in which the subject walked normally, the treadmill accelerated and then decelerated, between 0.8 m/s and 1.4 m/s at 1.0 m/s^2.

Trajectories of 53 surface markers, ground reaction forces from three force plates, knee contact force from the instrumented knee prosthesis and fluoroscopy data of the knee are the data available with this grand challenge competition used in this study.

5.2. Musculoskeletal Model

Inverse dynamics analysis in OpenSim was carried out to obtain resultant forces from input data (kinematics and ground reaction forces). OpenSim is a freely available, open-source software that allows users developing models of musculoskeletal systems and create dynamic simulations of a wide variety of movements. It is one of the main applications from Simbios, a NIH Center for Biomedical Computation at Stanford University. Some of the most important features available with OpenSim are:

- Scaling of musculoskeletal models
- Inverse Kinematics
- Inverse Dynamics
- Forward Dynamics
- Other analyses and simulations

In this study, a musculoskeletal model of the lower limbs was used. Such a model was constructed in OpenSim using parameters measured from the subject's anatomy [7]. This model has 23 degrees of freedom and 44 muscle actuators.

The 23 degrees of freedom include: 3 rotations (adduction, flexion and rotation) and 3 translations at the pelvis, 3 rotations (adduction, flexion and rotation) at the hip, 3 rotations (adduction, flexion and rotation) and 3 translations (superior-inferior, anterior-posterior, medial-lateral) at the knee, 3 rotations (adduction, flexion and rotation) and 3 translations (superior-inferior, anterior-posterior, medial-lateral) for the patella with respect to the femur and 2 rotations (flexion, eversion) at the ankle.

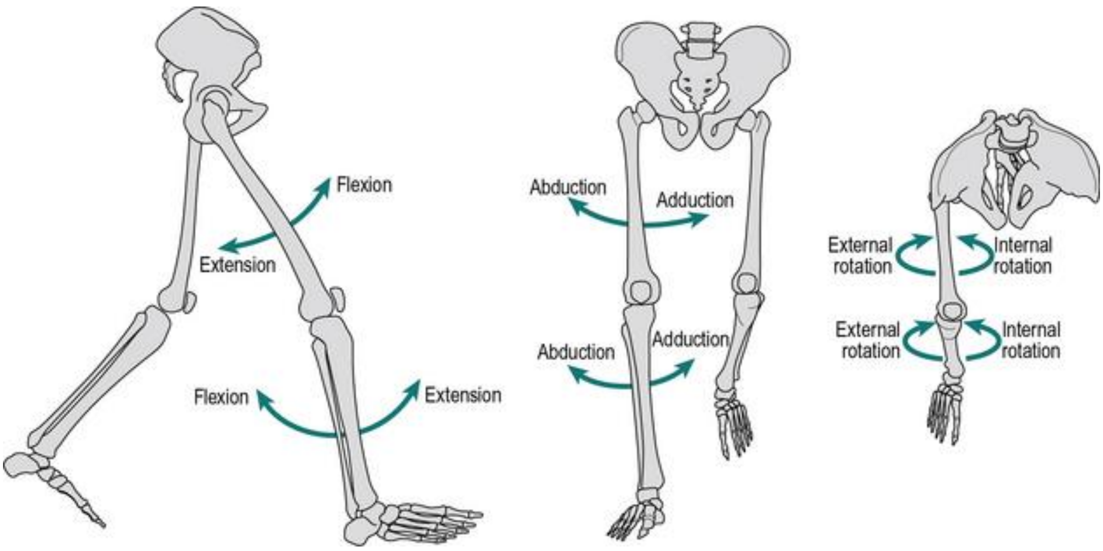

Figure 5.1. Adduction, Flexion and Rotation of the hip and knee [2]

Even though the musculoskeletal model has 23 degrees of freedom, the number of degrees of freedom was reduced for simplicity. The patellar flexion was coupled to the knee flexion so that they act as if they were welded. All other patellar degrees of freedom were locked at a constant value of 0. Knee flexion was obtained from the inverse kinematics analysis and all other degrees of freedom at the knee were fixed to a certain position/orientation as these relative translations and rotations can be neglected.

A rigid tendon muscle model was used for each muscle-tendon unit. This model contains the tendon and the muscle fibers (Figure 5.2.). In this case the length of the tendon (l_T) was considered to be constant, and the length of the muscle-tendon unit was obtain from OpenSim. Angle α is the orientation between the tendon and muscle fibers (pennation angle). Muscle fibers has a length (l_M) and it contains two parts in parallel: a contractile element (CE), representing the active part of the muscle, and the passive element (PE), representing the passive elements of the parallel structures of the muscles. From this muscle-tendon model, the muscle force become a function of the muscle activation, normalized length of the muscle and normalized velocity of the muscle:

$$F^M = F_0^M \left[a f_{act}\left(\tilde{l}^M\right) f_v\left(\tilde{v}^M\right) + f_{pass}\left(\tilde{l}^M\right) \right] \quad \textbf{Eq.5.1.}$$

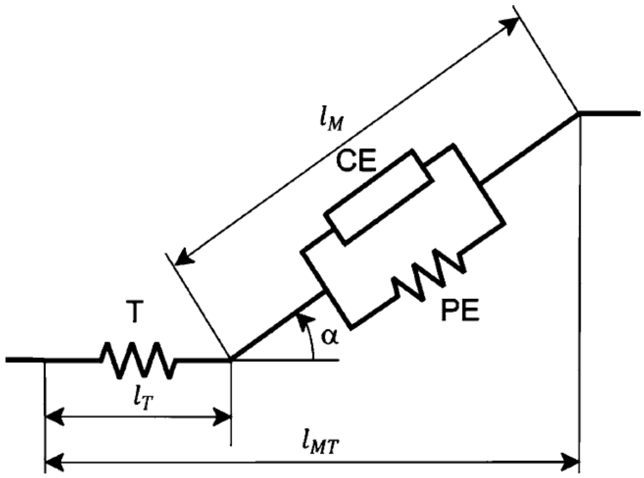

Figure 5.2. Hill's muscle tendon model

For this project, the muscles in the leg were classified into the medial, lateral and central muscles as shown in the figure below.

Medial	Lateral	Central
adductor brevis	biceps femoris long head	extensor digitorus longus
adductor longus	biceps femoris short head	extensor hallucis longus
adductor magnus distal	gastrocnemius lateralis	flexor digitorum longus
adductor magnus middle	gemeli	flexor hallucis longus
adductor magnus ischial	gluteus medius anterior	rectus femoris
adductor magnus proximal	gluteus medius middle	soleus
gastrocnemius medialis	gluteus medius posterior	vastus interior
gluteus maximus superior	gluteus minimus anterior	
gluteus maximus middle	gluteus minimus middle	
gluteus maximus inferior	gluteus minimus posterior	
gracilis	peroneus brevis	
iliacus	peroneus longus	
pectineus	peroneus tertius	
psoas	periformis	
quadratus femoris	tensor fascia latae	
sartorius	vastus lateralis	
semimembranosus		
semitendinosus		
tibialis anterior		
tibialis posterior		
vastus medialis		

Table 5.1. Muscles classified based on their anatomical location [1]

In OpenSim, inverse kinematics analysis is a tool that steps through each time frame of experimental data and positions the model in a pose that best matches experimental marker data and coordinate data for that time frame. The sum of weighted squared errors of markers is minimized in order to find the best match for each time frame. A subject-specific Opensim model, experimental marker trajectories for the trial and a settings file containing the information for the IK tool, including the marker weights are the basic inputs required to run an Inverse Kinematics tool in OpenSim.. Once Inverse Kinematics tool has finished, a file containing generalized coordinate trajectories is obtained as output.

After the generalized coordinate trajectories are obtained, the inverse dynamics tool can be run to calculate the inverse dynamics loads of the model for that movement. It determines generalized forces at each joint responsible for its moment. The inverse dynamics tool solves the classical force equations, Newton's second law (F=ma) in an inverse dynamics sense to obtain the forces and torques that are responsible for body movement. The generalized coordinate trajectories, the OpenSim model and the external load data are provided as inputs to the inverse dynamics tool, and a file containing the time histories of the net joint torques and forces is obtained. Further, an analysis tool is used to obtain the data regarding muscle-tendon lengths and velocities, and moment arms for the different muscles in the lower leg.

5.3. Optimization Algorithm

In this study, a two level optimization algorithm in MATLAB was used to calibrate neuromusculoskeletal model parameter values to data from the three selected gait trials. The nested optimization has two levels (outer and inner levels).

5.3.1. Outer Level Optimization

The outer level optimization uses a nonlinear least squares algorithm to correct the time-independent model parameter values such as optimal muscle fiber length scale factors and tendon slack length scale factors. The outer level function minimizes a weighted sum of squares of terms including three terms:

- Minimize undesirable quantities (reserve activations)
- Track quantities (contact forces)
- Constrain penalty terms (to constrain the normalized length of the muscles into a physiological operating muscle range)

The following graph represents the normalized length versus muscle force.

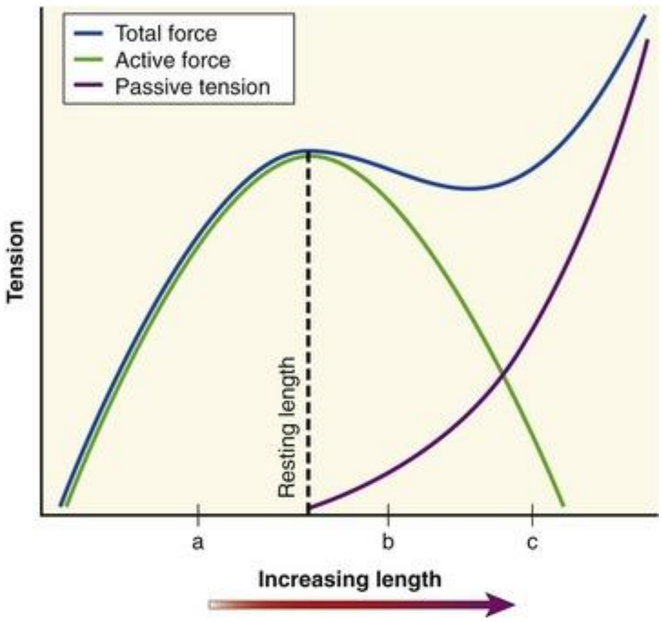

Figure 5.3. Muscle length versus muscle force[8]

From this graph, we can see that as the muscle length increases beyond a certain point, the passive force is higher than the active force. The maximum active force is obtained when the length of the muscle reaches its optimal value (l_0^M). The natural range of muscle operation is around the optimal length of the muscle, according to previous studies [9].The passive muscle forces are minimized in this study in an effort to keep the active muscle forces maximum. A penalty term in the cost function will be introduced to keep normalized length of the muscle (length of the muscle divided by optimal length of the muscle) close to one.

The optimization is mainly used to solve the muscle force sharing problem, since the human has more muscles than degrees of freedom. Six resultant inverse dynamics loads are balanced with the moments exerted by the muscles. Furthermore, residual moments (one for each joint) are introduced to help the optimization to balance the loads.. Residual moments are used only for correction and they have no physical meaning. However, these residual moments are minimized in the inner-level cost function.

$$M_{jk} = \sum_i^{44} f_i \, r_{ijk} + \sum_k^6 a_{resjk} T_o \qquad \textbf{Eq.5.2.}$$

The a_{resjk} term in this equation is the reserve activation (one for each joint load), which when multiplied by a constant moment T_0, becomes the value of a residual actuator.

As mentioned earlier, two approaches were used. In Approach A, experimental knee contact forces were tracked by the optimization in the outer level and in Approach B, experimental knee contact forces were not tracked. Both approaches used the same inner-level optimization. In the approach where the knee contact forces were tracked, model and experimental knee medial and lateral contact forces were tracked by the outer-level optimization.

This code also uses penalty terms to keep certain parameters in check.

5.3.2. Inner Level Optimization

The inner level of optimization used a fast quadratic programming algorithm to optimize the design variables for the time dependent muscle activations using the current guess (of the outer-level optimization) for model parameter values. It minimizes the sum of squares of muscle and reserve activations. The inner level cost function is common to both the approaches used in this study. The inner level optimization tracks only 6 inverse dynamics loads because these six loads are not affected by the knee contact forces.

$$J = \sum_i^{44} a_i^2 + \sum_k^6 a_{res,k}^2 \qquad \textbf{Eq.5.3.}$$

6. Results

6.1. Inverse Dynamics

Inverse dynamics loads obtained from OpenSim were used as one of the inputs in the optimization code. Hip flexion, adduction and rotation, knee flexion, ankle and subtalar moments were the six degrees of freedom tracked in Approach B.

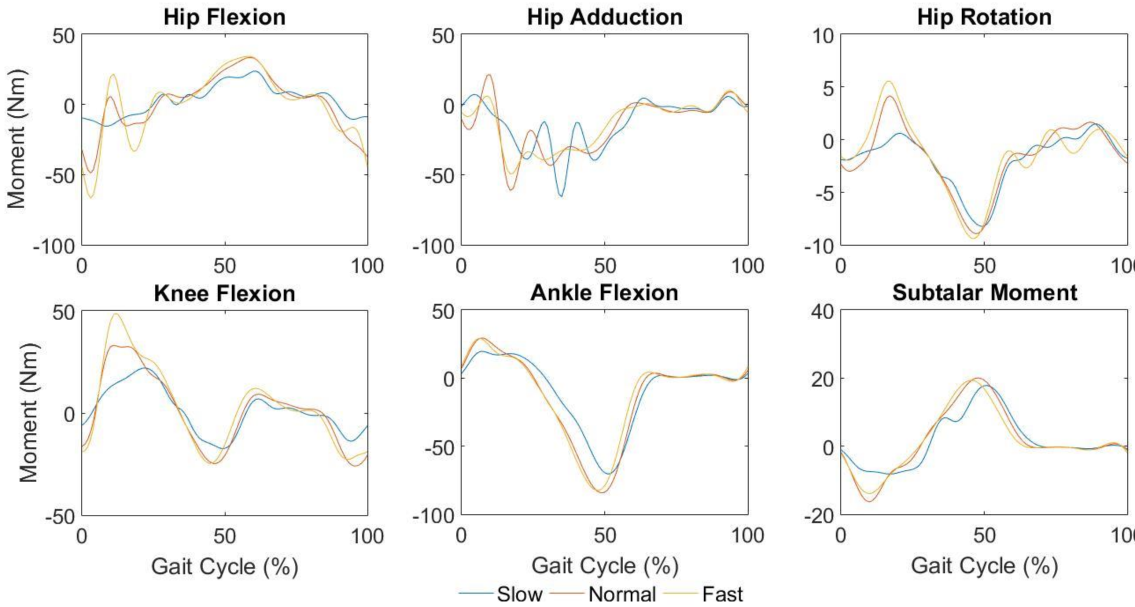

Figure 6.1. Inverse Dynamics data of the 6 tracked loads for Approach B: Hip Flexion (Positive: Flexion, Negative: Extension), Hip Adduction (Positive: Adduction, Negative: Abduction), Hip Rotation (Positive: Internal, Negative:), Knee Flexion Moment (Positive: Extension, Negative: Flexion), Ankle Flexion (Positive: Dorsiflexion, Negative: Plantarflexion), Subtalar Moment (Positive: Internal Rotation, Negative: External Rotation)

6.2. Differences in Approaches A and B

6.2.1. Medial-Lateral Forces

The medial-lateral knee contact forces for the trial at normal speeds are shown in Figure 6.2. This figure shows medial-lateral experimental knee contact forces and those obtained from the optimization with approach A and approach B. We can observe that medial forces contribute to most of the total forces. We can also notice that predicted forces in Approach A are closer to the experimental values than in Approach B. It can be seen that the lateral knee contact forces are tracked much better by Approach A than by Approach B.

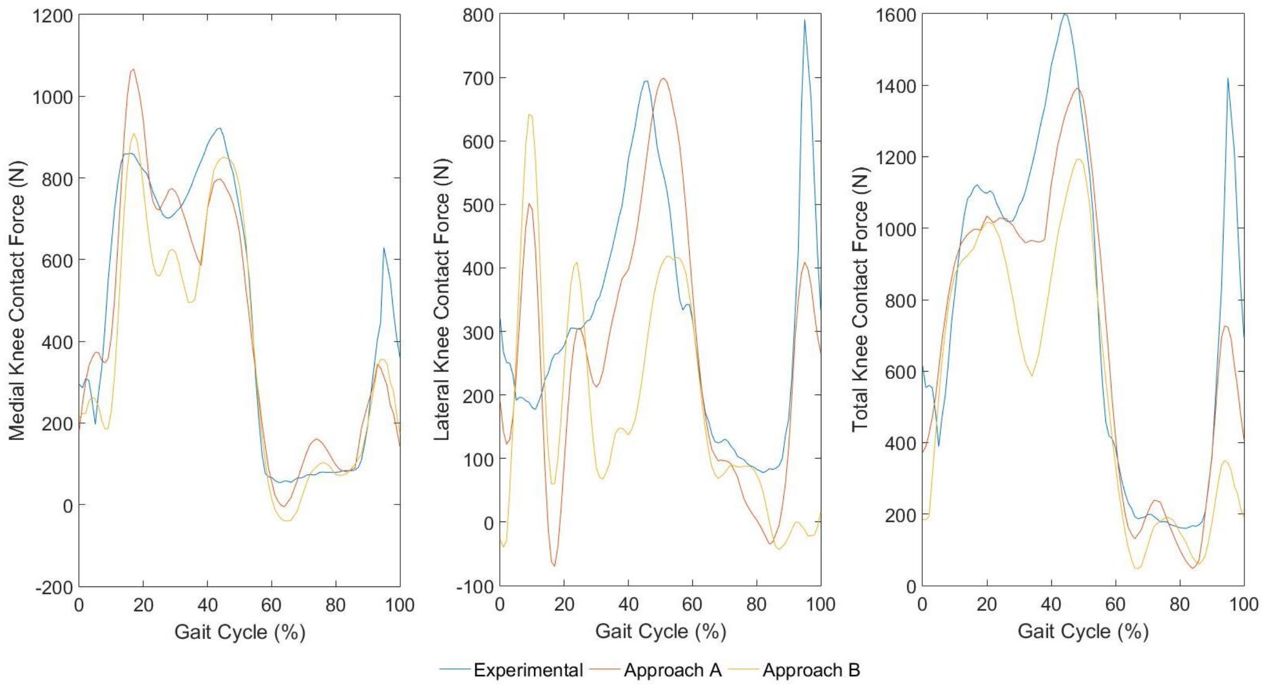

Figure 6.2. Difference in medial lateral forces between different approaches

6.2.2. Optimal Muscle Fiber Lengths

The optimal muscle fibre lengths were plotted for all the 44 muscles in the leg. We can see that the values are quite low for almost all of the muscles with just the sartorius having an optimal muscle fiber length of 1.

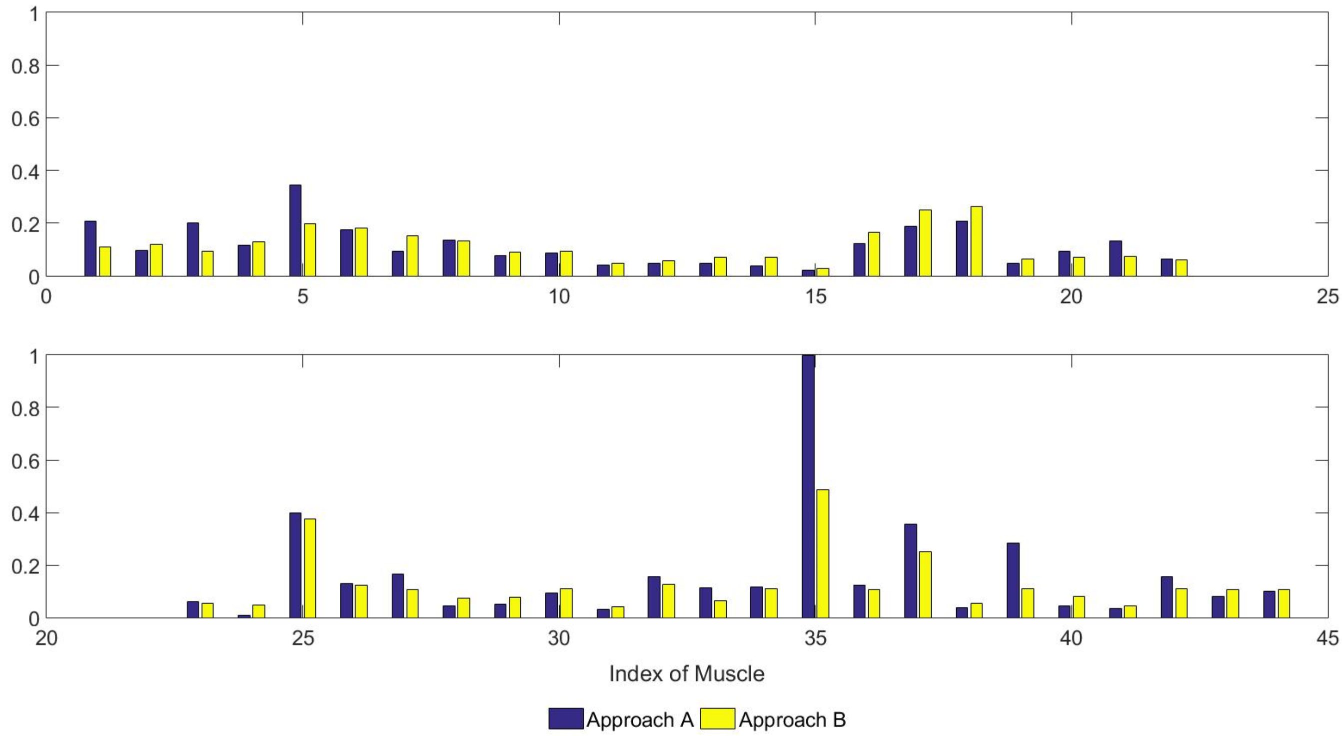

Figure 6.3. Optimal muscle fiber lengths for all muscles (see appendix for index of muscle)

6.2.3. Slack Length of Tendons

The muscles were also considered to be tendons here as the model used in this study does not have any tendons. The slack length for all 44 muscles were plotted and the results are presented below.

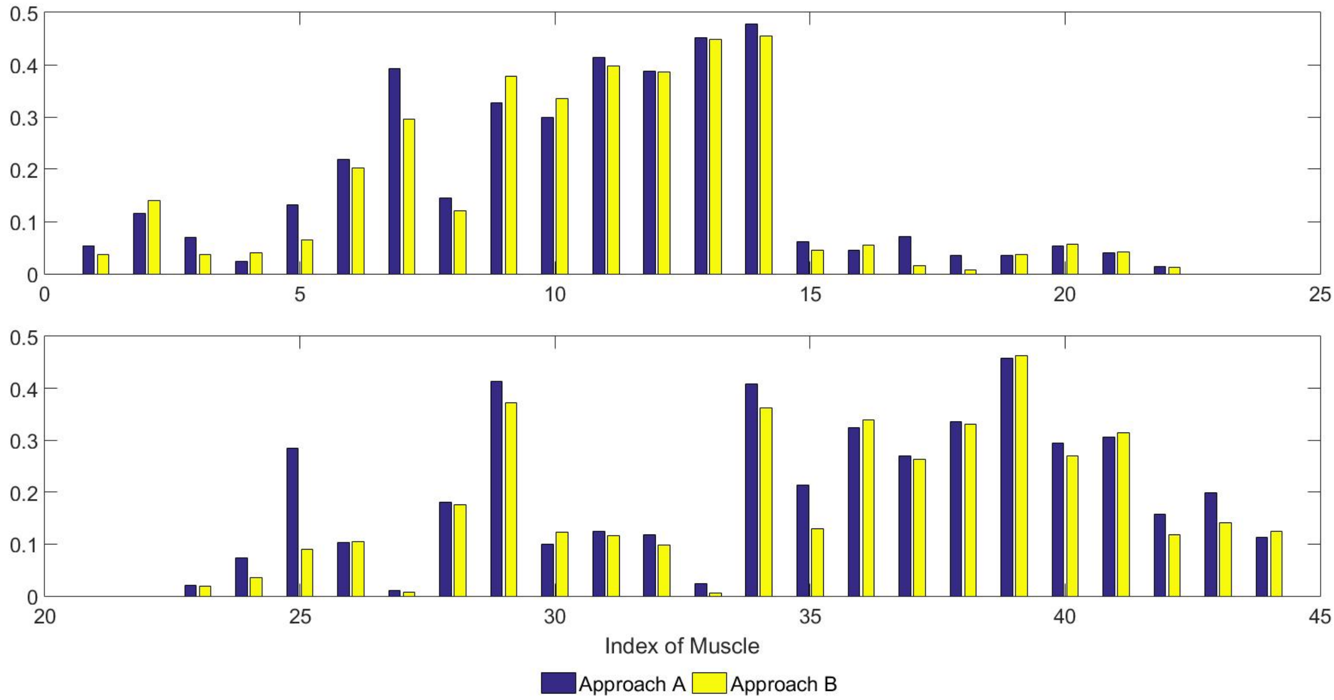

Figure 6.4. Slack length of all muscles (see appendix for index of muscle)

6.3. Statistical Analysis

A Student's t-test is used to determine whether two sets of data are significantly different from each other. Here, the muscle forces for all 44 muscles for two approaches for all three trials were processed to obtain the muscle force differences approaches B and A. The mean force along the trial was calculated for each muscle, trial and approach. Then differences between both approaches were calculated for each muscle and trial. With these values a Student's t-test was carried out for each muscle and it was found that the results for 36 of the 44 muscles were significantly different. This comes to show that there was a significant difference in estimation depending on whether experimental knee contact forces were tracked or not. The muscle forces have been plotted for eight muscles with the lowest p values.

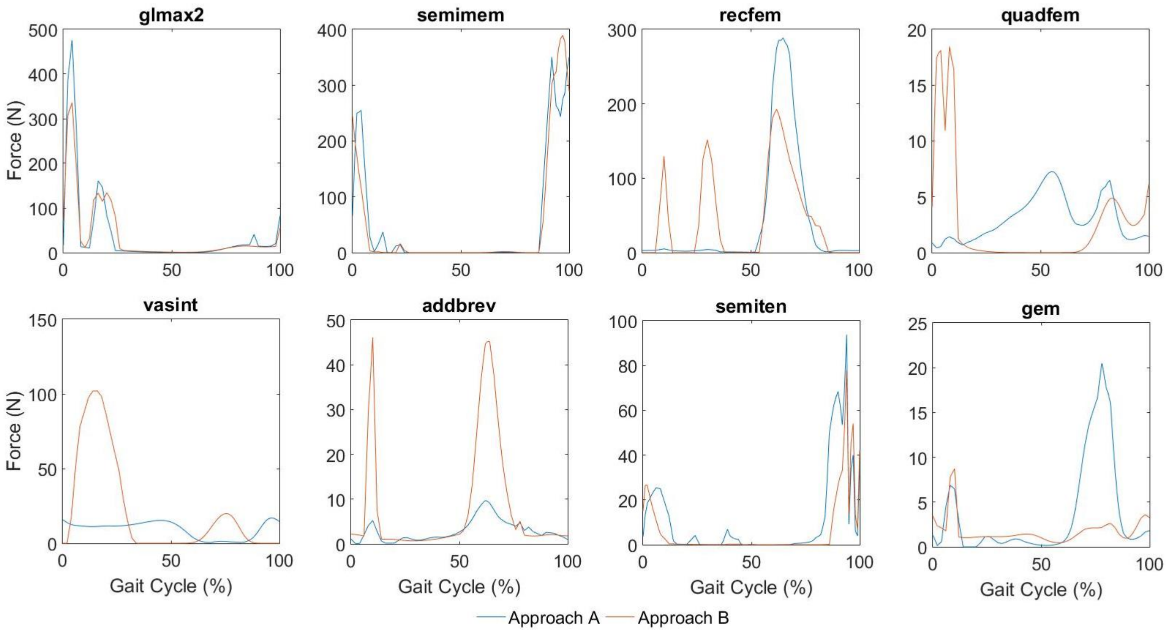

Figure 6.4. Muscle forces for both approaches for the 8 muscles with lowest p-values

6.4. Muscle Behaviour during variation in speed

The mean difference in muscle forces between the slow and fast trials were calculated and the plots for the muscles with the maximum difference in force were plotted.

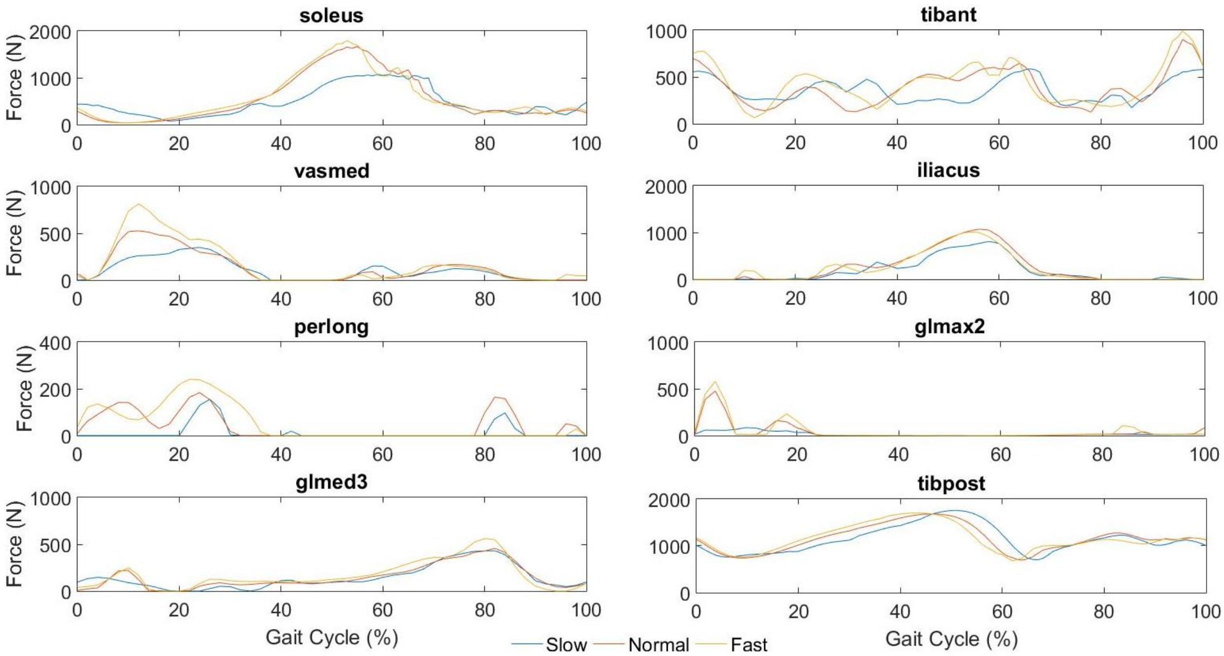

Figure 6.5. Differences in the muscle forces for variation in the walking speed

7. Environmental impact and Economic Analysis

7.1. Environmental Impact

The objective of this study is to contribute technology that improves clinical treatment of people with osteoarthritis or other knee disorders. The environmental impacts of this study and the technology is very little to none. The only way in which this technology could be impacting the environment would be by making the treatment process more efficient and avoiding spending resources unnecessarily on such treatment.

7.2. Economic Analysis

The economic cost of this project is represented by

- The depreciation of the instrumentation used during the project
- The cost of time of all the people working on it

The number of hours a student has worked on this project is around 600 hours. It can be assumed that the 600 hours were spent on the laptop because the study is computer based.

Assuming cost of depreciation to be 0.08 €,

Total cost of depreciation = 0.08 € * 600 = 48 €

Assuming the salary for an engineering student to be 8 €/ hour,

Total cost of work for the night = 8 € * 600 = 4800 €

Total cost of project = 4848 €

7.3. Social Impact

All over the world, there are millions of people who live with knee disorders due to ineffective clinical treatments. As this technology improves, the better treatment that comes along is going to contribute to a collective improvement in the quality of life of these patients. These improvements could also improve the cost of such treatment and make it accessible to the poorer section of society.

Conclusions

This study aimed to understand how different muscles behaved for different speeds of walking. It used two approaches: one tracking experimental knee contact forces and one that did not track these forces. These two approaches were expected to give results of varying accuracies. Approach A should give more accurate predictions. Then, the muscles which had largest variations were identified.

The soleus, tibialis anterior, vastus medialis, iliacus, peroneus longus, gluteus maximus middle, gluteus medialis posterior and tibialis posterior are the eight muscles with maximum differences in mean forces of the fast and slow trials. These differences can be explained by the differences in the muscle action and the difference in their contribution to the movement among speeds.

First of all, soleus is the muscle responsible for plantarflexion. Plantarflexion happens faster when the gait speed is faster. This direct correlation means that as the gait speed increases, we can see an increase in the soleus force. The tibialis anterior and its antagonist peroneus longus are seen in this list of muscles with highest differences among speeds. Tibialis anterior is responsible of dorsiflexion, as opposed to plantarflexion. It is also important for stabilizing the ankle as the heel contacts the ground. We can see that the forces for these muscles are directly related to the speed: muscle forces increase with an increase in speed.

Vastus medialis is also interesting to analyze. The knee flexion moment (Figure 6.1.) increases with an increase of the speed. Vastus medialis is involved in the knee extension. Accordingly, we can see that the force of this muscle increases as the speed of walking increases.

On the other hand, there are several muscles including the gracilis, sartorius, gluteus minimus anterior and semitendinosus whose forces had no differences based on the speed of walking. The sartorius is responsible for flexion, abduction and lateral rotation of the hip. However, since this is a weak muscle, neither does the muscle affect the speed nor does the speed affect the muscle. The gluteus minimus, which is responsible for hip abduction, does not have a high variation, since hip abduction does not play an important role in walking on a treadmill, which is mainly contained at the sagittal plane. None of the three bundles of gluteus minimus showed any considerable differences in forces for different speeds.

A previous study [1] showed that the biceps femoris long head, biceps femoris short head, gastrocnemius lateralis, gluteus maximus superior, gluteus medius anterior, gluteus medius middle, gluteus medius posterior, psoas, rectus femoris, semimembranosus, soleus, tensor facia latae and vastus lateralis had the maximum differences in forces between approaches A and B. But that study was carried out for six overground gait trials of the same speed (self-selected speed). However, only one of those muscles, semimembranosus, had large differences in predicted forces in this study. This can be indicative of the fact that accuracy of the two different approaches might change with respect to the speed of trial. It can also be noted that 36 of the 44 muscles turned out to be statistically significantly different according to a Student's t-test. This shows that there is a large difference in forces predicted by these two approaches.

This code gives us accurate answers when we track the experimental knee contact forces. But we would not be able to accurately predict knee contact forces in all humans since just a few people in the world have instrumental prosthesis implanted. However, even while tracking the knee contact forces, there is room for improvement. More research is needed to design a better, more accurate outer level optimization. The results can be calibrated better by making use of the EMG data available.

In conclusion, this study found the muscles which were most affected by a change in the speed of walking. It was also demonstrated that the predictions are closer to the experimental values when knee contact forces are tracked. These results can be improved by using the electromyography data to follow the forward dynamics approach to reach the muscle forces. Muscle forces for other activities could also be studied to design better cost functions that predict contact forces better.

References

[1] G. Serrancolí, A. L. Kinney, B. J. Fregly, and J. M. Font-Llagunes, "Neuromusculoskeletal Model Calibration Significantly Affects Predicted Knee Contact Forces for Walking," *J. Biomech. Eng.*, vol. 138, no. 8, p. 81001, 2016.

[2] "Basic Sciences." [Online]. Available: https://musculoskeletalkey.com/basic-sciences/.

[3] "Patella Diagram." [Online]. Available: http://en.wikipedia.org/w/index.php?title=Image%3AKnee_diagram.png.

[4] A. D. Woolf and B. Pfleger, "Burden of major musculoskeletal conditions," *Bull. World Health Organ.*, vol. 81, no. 9, pp. 646–656, 2003.

[5] D. D. Lima, D. D. D'Lima, B. J. Fregly, S. Patil, N. Steklov, and C. W. Colwell, "Knee joint forces : prediction , measurement , and significance," *Proc. Inst. Mech. Eng. H.*, vol. 226, no. 2, pp. 95–102, 2013.

[6] D. Forchelet *et al.*, "Enclosed electronic system for force measurements in knee implants," *Sensors (Switzerland)*, vol. 14, no. 8, pp. 15009–15021, 2014.

[7] B. J. Fregly *et al.*, "Grand challenge competition to predict in vivo knee loads," *J. Orthop. Res.*, vol. 30, no. 4, pp. 503–513, 2012.

[8] S. K. Hunter and D. A. Brown, "Muscle: the primary stabilizer and mover of the skeletal system," in *Kinesiology of the musculoskeletal system*, 2010.

[9] E. M. Arnold, S. R. Hamner, A. Seth, M. Millard, and S. L. Delp, "How muscle fiber lengths and velocities affect muscle force generation as humans walk and run at different speeds," *J. Exp. Biol.*, vol. 216, no. 11, pp. 2150–2160, 2013.

[10] D. Richfield, "Medical gallery of Blausen Medical 2014," *WikiJournal Med.*, vol. 1, no. 2, pp. 9–11, 2014.

Appendix A

Index of Muscles

1. addbrev	2. addlong	3. addmagProx	4. addmagMid	5. addmagDist	6. addmagIsch	7. bflh
8. bfsh	9. edl	10. ehl	11. fdl	12. fhl	13. gaslat	14. gasmed
15. gem	16. glmax1	17. glmax2	18. glmax3	19. glmed1	20. glmed2	21. glmed3
22. glmin1	23. glmin2	24. glmin3	25. grac	26. iliacus	27. pect	28. perbrev
29. perlong	30. pertert	31. piri	32. psoas	33. quadfem	34. recfem	35. sart
36. semimem	37. semiten	38. soleus	39. tfl	40. tibant	41. tibpost	42. vasint
43. vaslat	44. vasmed					

Differences in forces based on speed for all muscles

The differences in muscles forces for the three different speeds were plotted in the results section for 8 muscles with the highest mean differences. Here, you can see the differences in muscle forces of all 44 muscles for different speeds in the order of highest to lowest mean difference between the fast and slow trials.

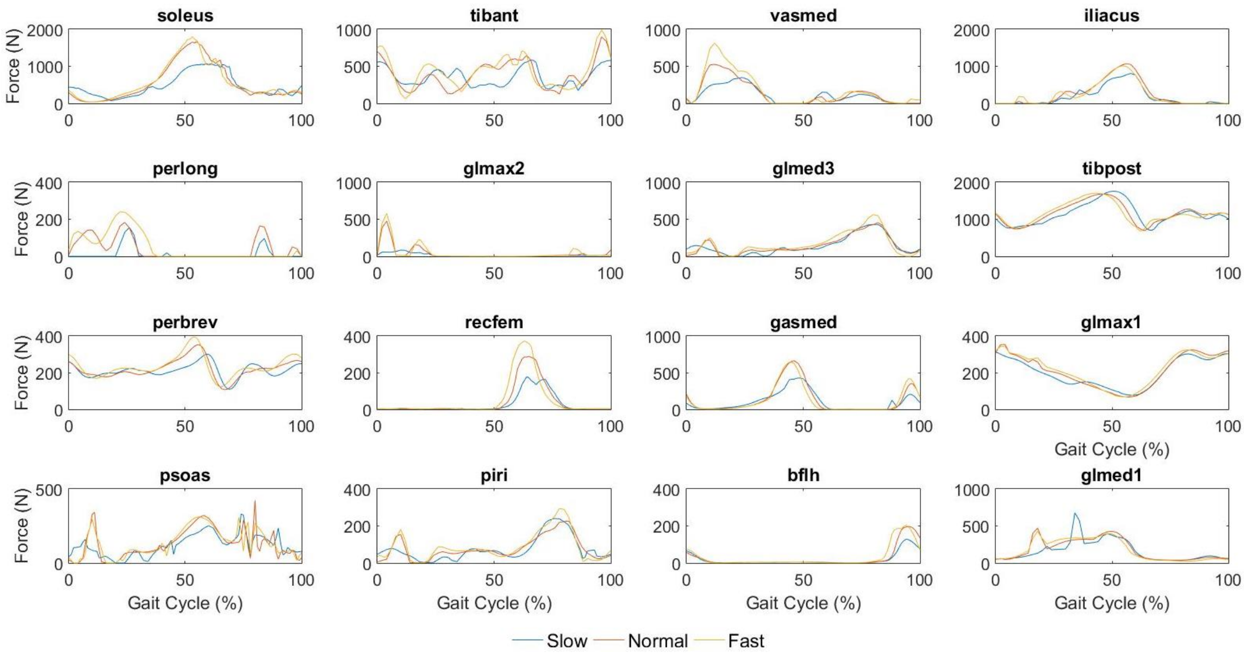

Figure A.1.

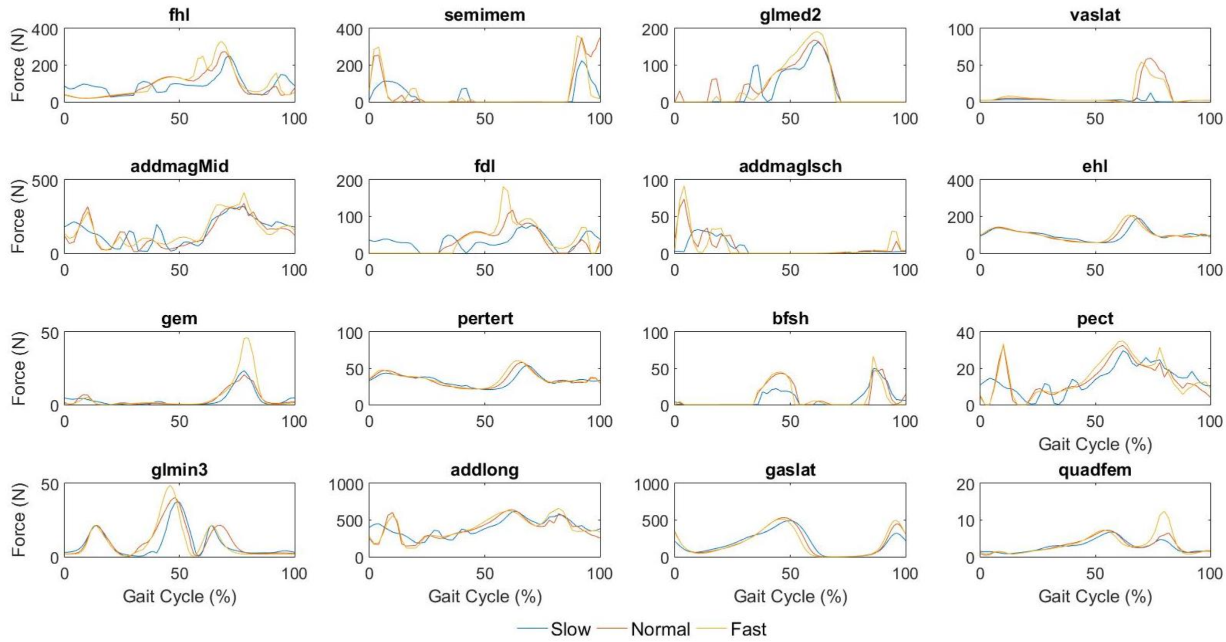

Figure A.2.

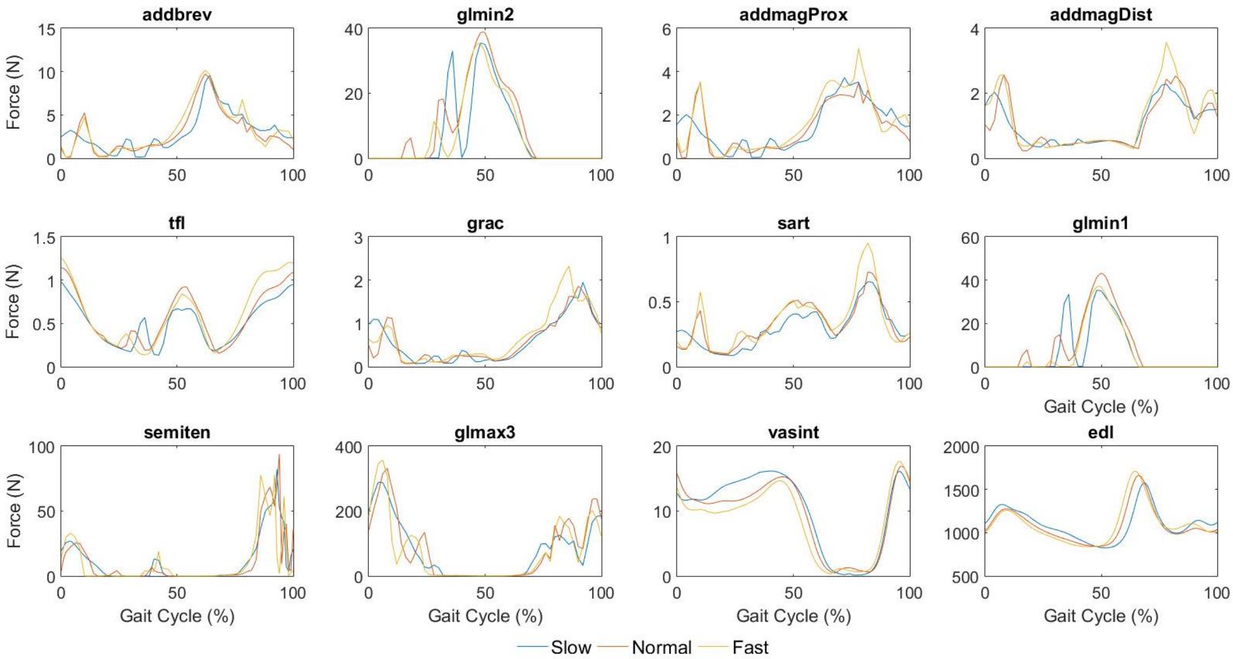

Figure A.3.