STUDY OF PAIN, MOTION, AND MUSCLE ACTIVITY FOR LOWER LIMB AMPUTEES

By

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ABSTRACT

This thesis investigates the interplay between pain and gait biomechanics in unilateral transtibial amputees (TTAs) to understand compensatory mechanisms and adaptations in the musculoskeletal system. The aim is to explore how pain influences gait performance by analysing spatial and temporal parameters, kinematics/kinetics, muscle activity, and pain outcomes using advanced tools such as motion capture, force plates, electromyography (EMG), and musculoskeletal modelling.

The study first developed a rigorous experimental protocol to acquire multi-modal biomechanics gait data (i.e., marker trajectories, ground reaction force, EMG, and self-reported pain measures). This protocol was then successfully applied to capture gait data from eight able-bodied controls and six individuals with transtibial amputation (TTA). The data analysis and modelling identify significant gait deviations in TTAs, including reduced walking speed, altered stride length, and asymmetrical ground reaction forces, correlating these with pain levels. Advanced musculoskeletal simulations provide further insights into joint kinematics, kinetics, and muscle forces, highlighting the compensatory strategies adopted by amputees.

The findings emphasise the need for multidisciplinary approaches in rehabilitation, integrating biomechanical analysis and pain management to enhance mobility and quality of life for TTAs. This research bridges key knowledge gaps, offering practical recommendations for improved prosthetic development and rehabilitation strategies.

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LIST OF ABBREVIATIONS

TTA	--Unilateral transtibial amputation
AB	--Able-bodied (non-amputees)
MSK	--Musculoskeletal
GRF	--Ground reaction force
EMG	--Electromyograph
ROM	--Range of motion
IK	--Inverse kinematic
ID	--Inverse dynamic
SO	--Static optimisation
BMI	--Body weight index
LBP	--Low back pain
PLP	--Phantom Limb Pain
PIS	--Participant information sheet
BPI	--Brief pain inventory test
RMS	--Root mean square

STATEMENT OF ORIGINAL AUTHORSHIP

The work contained in this thesis has not been previously submitted to meet requirements for an award at this or any other higher education institution. To the best of my knowledge and belief, the thesis contains no material previously published or written by another person except where due reference is made.

Signature:

Date: 02/09/2024

Chapter 1: Introduction

Lower limb amputation, especially transtibial amputation (TTA), can make it difficult to walk and has an effect on the quality of life. This is mostly because of the pain and changes in how the person walks. Pain, such as phantom limb pain (PLP) and low back pain (LBP), can make rehabilitation more difficult and can lead to biomechanical imbalances and compensatory strategies in amputees. Because of these problems, we need to do a full study on the biomechanical and pain-related effects of TTA to make clinical outcomes and strategies for prosthetic rehabilitation better.

This thesis aims to explore the complex relationship between pain and altered gait biomechanics in individuals with TTA. It seeks to address existing gaps in the literature by employing rigorous experimental protocols, advanced biomechanical models, and multidisciplinary approaches. The objectives include acquiring high-quality, multi-modal gait biomechanics data from both able-bodied individuals and those with TTA; evaluating gait dynamics and muscle activity; and analysing the influence of pain on movement patterns and compensatory mechanisms. By utilising advanced tools, such as OpenSim, for musculoskeletal modelling and validated pain assessment measures, this research aims to provide new insights into the dynamic interplay between pain and biomechanics, ultimately contributing to advancements in prosthetics, rehabilitation, and clinical practices.

Through its detailed analyses, this thesis makes significant contributions to the fields of prosthetics and orthotics, biomechanics, and pain management. It helps us learn more about how amputees compensate and how their bodies have changed to deal with these issues. It also shows how important it is to use approaches from different fields to solve these problems and gives useful suggestions for making people with TTA more mobile and improving their quality of life.

The subsequent sections of this chapter outline the aim and objectives of the thesis, its significance and contributions to the field, and the overall structure of the document.

1.1 Aim and Objectives

This thesis aims to comprehensively explore the interplay between pain and altered gait biomechanics in TTAs.

The objectives are:

- To develop a rigorous experimental protocol to acquire multi-modal biomechanics gait data (i.e., marker trajectories, ground reaction force, EMG, and self-reported pain measures) using equipment and tools, such as the motion capture system; force platforms; wearable EMG sensors and validated pain questionnaire.
- To acquire multi-modal biomechanics gait data from able-bodied controls and individuals with transtibial amputation (TTA), with high quality for data processing and biomechanics modelling.
- To evaluate gait biomechanics by examining spatial, temporal, and kinetic gait parameters, such as stride length, cadence, ground reaction forces, and joint kinematics, in amputees and comparing them to able-bodied. Secondly, it seeks to assess muscle activity by investigating electromyography in the intact and amputated limbs to identify compensatory strategies and asymmetries. Another objective is to analyse the influence of pain by exploring how pain levels, as

reported through validated pain questionnaires, correlate with gait abnormalities and compensatory mechanisms in amputees.

- To utilise advanced musculoskeletal modelling tools, including OpenSim, to simulate gait dynamics and quantify joint kinematics, kinetics, and muscle forces. Finally, it aims to derive clinical implications from these findings, offering recommendations to address existing knowledge gaps and advance our comprehension of the interplay between pain and gait biomechanics in amputees.

1.2 Contribution

By providing detailed analyses of gait dynamics and their relationship with pain, the thesis has significant contributions:

- It deepens the understanding of compensatory mechanisms adopted by amputees and their biomechanical adaptations compared to able-bodied individuals.

- It highlights the necessity of integrating multidisciplinary approaches in prosthetics and rehabilitation.
- It offers practical recommendations for enhancing the quality of life and functional mobility for individuals with unilateral lower limb amputation.

1.3 Structure of the thesis

This thesis starts by providing an overview of the aim and objectives of the thesis (Chapter I). Then it provides a general literature review illustrating the background of the lower limb amputation (anatomical, physiological associated with lower limb amputation, followed by the pain accompanied with lower limb amputation) in Chapter II. The systematic review, Study protocol, Physical Instrumental Measurement Study, OpenSim Modelling and Simulation study, and conclusion are in Chapter III-VII, respectively.

Table 1.1 Thesis Structure and Chapters Overview

Chapter I	**Introduction** Aim and Objectives
Chapter II	**Literature Review** Illustrating the medical background of lower limb amputation and pain
Chapter III	**Systematic Review** Exploring tools for assessing functional movement and pain in amputees
Chapter IV	**Study Protocol** Detailed experimental setup and methodology
Chapter V	**Physical Instrumental Measurement Study** Data collection and preliminary analysis
Chapter VI	**OpenSim Modelling and Simulation Study** Biomechanical modelling and advanced analysis
Chapter VII	**Conclusion and Future Directions** Summarizing findings, discussing limitations, and proposing future research

Chapter 2: Literature Review

This chapter examines the main aspects of gait analysis, focusing on how they help explain the challenges faced by individuals with lower limb amputation. It emphasizes the importance of spatial parameters, such as stride length and width, and temporal parameters, including cadence and step duration, which are essential for understanding walking patterns. Ground reaction forces are discussed to show how the body interacts with the ground during movement, providing insights into balance and load distribution. The analysis of muscle activation patterns further explains the roles of various muscles during walking and how they adjust to compensate for the missing limb. The chapter also explores the use of modelling and simulation tools, such as OpenSim, to study joint movements, muscle forces, and overall biomechanics in amputees. These tools provide valuable insights into how prosthetic limbs affect gait and identify opportunities for improvement in rehabilitation and prosthetic design.

Additionally, the chapter highlights gaps in the current research, particularly regarding the relationship between pain and compensatory gait mechanisms. It underscores the need for further studies to better understand how pain impacts movement and to develop more effective clinical practices for individuals with lower limb amputation.

Gait analysis is a multifaceted method used to evaluate human walking patterns, providing insights into biomechanical and physiological aspects of movement. For the general population, gait analysis assesses gait, focusing on joint kinematics, muscle activity, and GRF. It is crucial in diagnosing, treating, and managing various gait disorders. In the context of TTAs, gait analysis becomes even more critical. It helps in understanding the compensatory mechanisms and altered biomechanics due to the amputation, such as asymmetrical gait patterns, joint range of motion (ROM), and muscle activation disparities to AB. This difference is pivotal for improved understanding of the differences in amputee gait compared to the AB. This difference is pivotal for better improved understanding of the differences in amputees gait compared to the AB.

2.1 Background

Unilateral transtibial amputation (TTA) is a disabling condition that significantly influences several body systems, with the most significant impact on musculoskeletal (MSK) function (Roberts, 1990; Wasser et al., 2021). Worldwide, 1.2 million people each year experience a significant limb amputation because of vascular diseases (54%), including diabetes and peripheral arterial disease, trauma (45%), and cancer (less than 2%; Ahmad et al., 2014). According to the Nationwide Inpatient Sample, there are around 115,000 people in the United States (US) that undergo lower extremity amputation annually, of which 50,000-60,000 are above the ankle joint; 65% are diabetics, and 75% are due to peripheral artery disease (Nowygrod et al., 2006). In the UK, the number of amputations in 2014 was 25,312. The prevalence rate per 100,000 (95% confidence intervals) for amputation was 26.3 (26.0–26.6), with rates significantly higher in Northern England (North: 31.7; 31.0–32.3, Midlands: 26.0; 25.3–26.7, South: 23.1; 22.6–23.5; Ahmad et al., 2014). As a result of the increasing rate of diabetes and the population ageing, the growth of amputation is expected to increase, with the prediction that the amputee population will double by 2050. Although amputation affects many activities of daily living, it is the effect on the capacity for locomotion that is the most apparent (Shumway-Cook & Woollacott, 2017).

Human movement is complicated, involving complete body coordination and the concurrent coordination of limb segments. Muscles, both small and large, excite the MSK system. The MSK system is a descriptive term for human muscle, bone, and joints that create movement from one plane to another, from frontal to sagittal. For humans, it is most often achieved by walking or running and is one of the most critical functions of an independent lifestyle(Roberts, 1990; Shumway-Cook & Woollacott, 2017). Following amputation, missing a segment with muscle, bone, and joint will affect the MSK harmony and extend to affect functional performance (Allami et al., 2016).

Lower limb amputation presents a profound physical and psychological challenge, affecting both the quality of life and the functional mobility of amputees (Jensen et al., 2001). The anatomical structure of the lower half of the body including bones, joints, muscles and fibrous connective tissue that attaches bone to bone (ligament), physiological and pathological human motion pattern, lower limb prosthesis, and MSK analysis on movement science.

This chapter will discuss the importance of gait analysis for unilateral lower limb amputees from prosthetic fitting, motion analysis, functional assessment, quality of life and long-term effect.

TTAs often experience residual limb pain and phantom limb pain, significantly impacting their rehabilitation outcomes (Davis, 1993). The understanding of pain perception could have a relationship with the biomechanical imbalances, compensatory movements, and subsequent MSK disorders among TTAs (Allami et al., 2016). The study of MSK modelling can provide key insights into these alterations, helping to optimize prosthesis design and rehabilitation approaches (Delp et al., 2007; Samuelsson et al., 2012). However, the complex interplay between pain and MSK changes among TTAs remains unexplored. There is a pressing need to investigate deeper into this area, which could hold significant implications for enhancing the quality of patient care and improving clinical outcomes among this population (Diebal-Lee et al., 2017). This chapter will explore the effect of pain on the gait kinematic and kinetic changes.

2.1.1 Anatomical structure of the pelvis and lower limb

The lower limb and pelvis are important for daily activities like walking. This lower half of the body encompasses a complex network of bones and muscles that work to produce movement. Particularly in gait, the harmonious function of the pelvis and lower limbs is crucial for stability, force distribution, and efficient energy use during locomotion. The anatomical changes combined with lower limb amputation play a pivotal role in the abnormal gait for TTAs. Understanding the specific contributions of these structures is essential for appreciating the full scope of their significance in gait dynamics (Kemp, 2010).

The pelvis (Figure 2.1) is a bony structure that connects the trunk to the lower limbs. It is composed of several bones, including the ilium, ischium, and pubis, which are joined together by ligaments and cartilage. The physiological movement of the pelvis includes sagittal plane motion (tilting), frontal plane motion (listing) and transverse plane motion (rotation). Anterior pelvis tilt appears during sitting down, and posterior pelvis tilt appears standing up from a sitting position. Lateral tilt of the pelvis involves tilting the pelvis to one side, such as standing on one leg. Rotation of the pelvis involves twisting, such as when turning the body to one side. The muscles of the pelvis, such as the gluteal muscles and hip flexors, provide stability and support to the lower limbs, allowing for efficient transfer of force from the legs to the upper body. This is important for generating propulsion and maintaining a steady gait (Radin et al., 1991; Roberts, 1990).

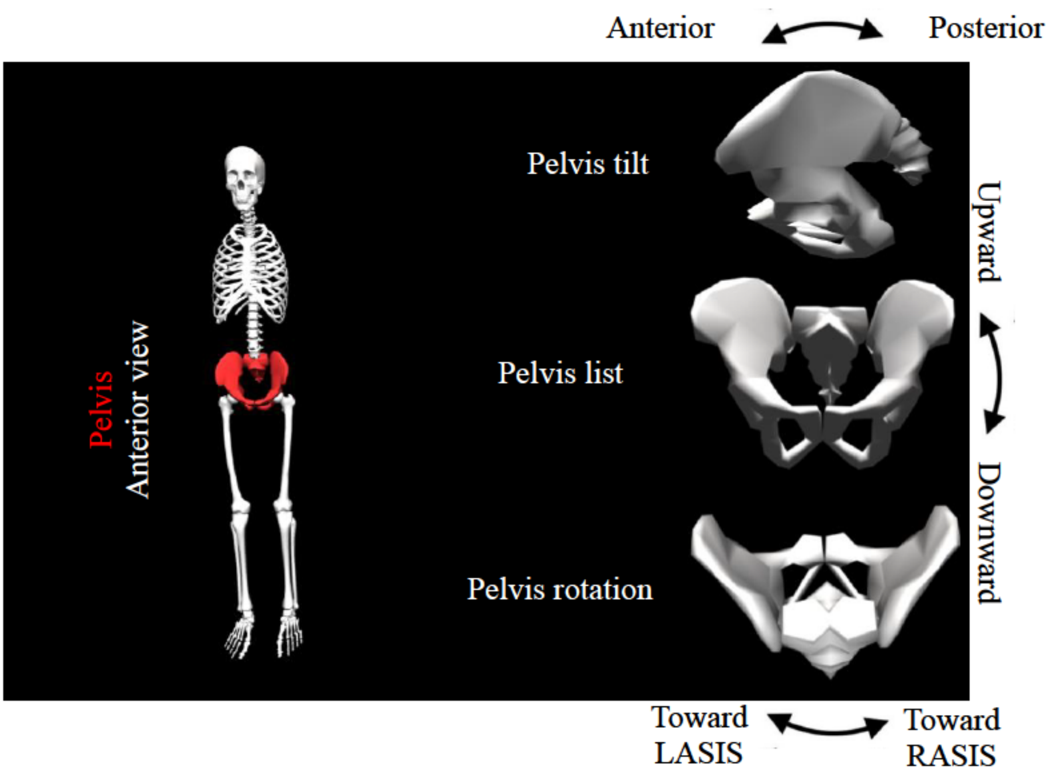


Figure 2.1 The anatomical illustration of the pelvis highlighting the pelvis location on a human skeleton, with arrows representing the directions of physiological movements.

Further, the lower limbs have the largest bone in the body, the femur. The femur is a long bone that is supported by the pelvis, and it articulates with the tibia, fibula and patella to form the knee joint. The tibia is a bone that supports most of the body's weight and forms the shank. The fibula is a slender bone that runs parallel to the tibia and provides stability to the ankle joint. The bones of the foot include the tarsals, metatarsals, and phalanges, which are responsible for supporting the weight of the body and enabling movement (Culham et al., 1986). The muscles of the lower half of the body are responsible for movement and stability and are attached to the bones through tendons. The major muscles of the lower half of the body include the quadriceps, hamstrings, gluteal, and calf muscles. The quadriceps are a group of four muscles located in the front of the thigh that are responsible for extending the knee joint. The hamstrings are a group of three muscles located in the back of the thigh that are responsible for flexing the knee joint. The gluteal muscles are a group of three muscles located in the buttocks that are responsible for stabilizing the pelvis and extending the hip joint. The calf muscles are located at the back of the lower leg and are responsible for plantarflexion of the ankle joint. The muscles of the lower half of the body work in conjunction with each other to enable movement and maintain stability. For example, during walking, the glutes and quadriceps work to extend the hip and knee joints, while the hamstrings work to flex the knee joint and the calf muscles work to plantarflex the ankle joint. In conclusion, the anatomical structure of the lower half of the body is complex and diverse and is comprised of bones and muscles that work together to enable movement and maintain stability. Understanding the anatomy of the lower half of the body is important for healthcare professionals, athletes, and individuals looking to improve their physical fitness (Figure 2.2) (Paulsen & J, 2015).

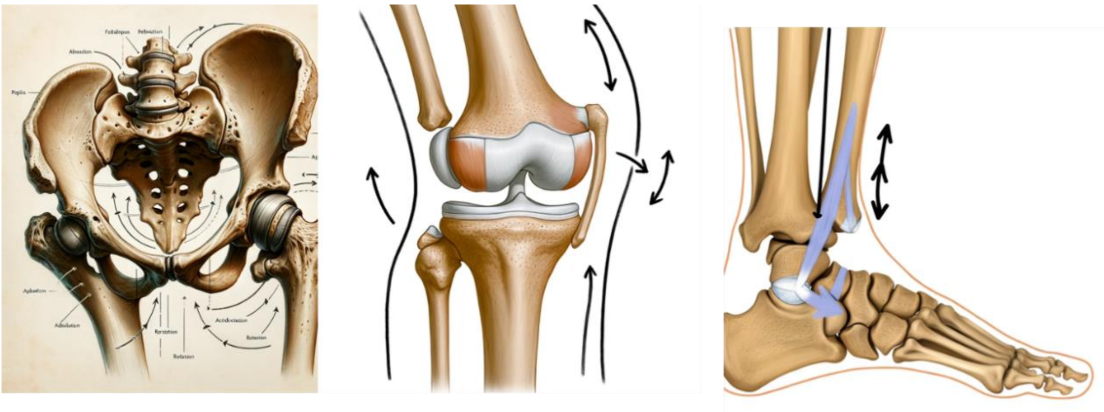


Figure 2.2 This composite anatomical illustration showcases the pelvis, hip, knee, and ankle joints along with their respective bony structures and possible movements. The hip joint is represented by the articulation of the femur and pelvic bones, showing flexion, extension, abduction, and adduction. The knee joint illustration emphasizes the femur, tibia, and fibula, with arrows depicting the flexion and extension movements. The ankle section illustrates the tibia, fibula, and foot bones, with arrows indicating dorsiflexion and plantarflexion.

2.1.2 Physiological and pathological human motion

Human motion is a complex process brought out by the interplay between the nervous and MSK systems. From a neural perspective, the nervous system controls and coordinates human motion through the central nervous system, which includes the brain and spinal cord. The central nervous system processes sensory information and sends out motor commands. Motor neurons in the peripheral nervous system transmit these commands to the muscles, initiating contraction. This process involves intricate feedback loops where sensory information from muscles and joints is relayed back to the central nervous system to fine-tune movements, ensuring balance and coordination. From the other side, the MSK system, comprising bones, muscles, tendons, ligaments, and joints, provides the structural framework and mechanical forces necessary for movement. Bones act as levers, muscles generate force, and tendons transmit this force to the bones to create movement at the joints. Ligaments and cartilage provide stability and facilitate smooth motion at the joints (Hamilton et al., 2008). Lower limb amputation significantly impacts the MSK system, leading to numerous adaptations and challenges. Post-amputation, individuals often experience changes in gait biomechanics, which can result in altered load distribution and increased reliance on the remaining limb and visual feedback for balance. This adaptation is necessary due to the loss of proprioceptive feedback from the amputated limb (Walton et al., 2024).

Thus, human motion represents a complex network of various physiological systems. A thorough understanding of the physiological and pathological aspects of human motion can substantially enhance our capability to understand how amputation could influence their gait. As we continue to decipher the intricacies of human motion, this knowledge will help predict amputees gait performance; therefore, the next section will discuss the lower limb amputation and the main elements of prostheses.

2.1.3 Lower limb amputation and prostheses

Lower limb amputation, a critical surgical intervention, can span several levels, each with its distinct surgical protocol. These levels can range from minor procedures, such as toe amputation, to more severe measures like hip disarticulation. Between the years 2008 to 2022, at the King Abdul-Aziz University Hospital, located in the western region of Saudi Arabia, Jeddah, conducted a total of 645 lower-limb amputation procedures. Of these, 2 were hip disarticulations, constituting a mere 0.5% of the cases. The majority were transfemoral amputations, amounting to 271 cases or 42% of the total. TTA were also frequent, accounting for 230 cases or 35.7%. The rest comprised of 21-foot amputations 3.3%, 33 forefoot amputations 5.1% and 88 toe amputations 13.6% (AlMehman et al., 2022).

When it comes to the surgical technique for transfemoral amputations and TTA, the procedure begins by dividing the anterior muscles of the femur or tibia. This division is in line with the skin flap that is designed to cover the residual limb or 'stump.' The quadriceps group in transfemoral amputations or the tibialis anterior in TTA is purposefully left longer than the bone to provide a soft tissue flap. This extended flap acts as a cushion, reducing discomfort when using a prosthetic limb. Next, the posterior muscles (hamstrings and calf muscles) are severed to match the end of the bone. In transfemoral amputations, the adductors are also partially or fully surgically cut. These muscles are then reattached to the bone at novel sites utilizing tension myodesis techniques, which involve suturing the muscle to the bone to stimulate scar formation and enhance stability. Following this, the anterior flap, crafted from the quadriceps or tibialis-anterior muscles, is draped over the end of the bone. Its fascia, the layer of connective tissue enveloping the muscle, is then sutured to the fascia of the posterior muscles. This meticulous process of muscle reattachment and careful balancing of adductor and abductor muscle groups, especially crucial in transfemoral amputations, plays a vital role in determining the amputee's potential for successful rehabilitation and quality of life post-amputation (Al-Shuka et al., 2019; Gottschalk, 1999; Figure 2.3).

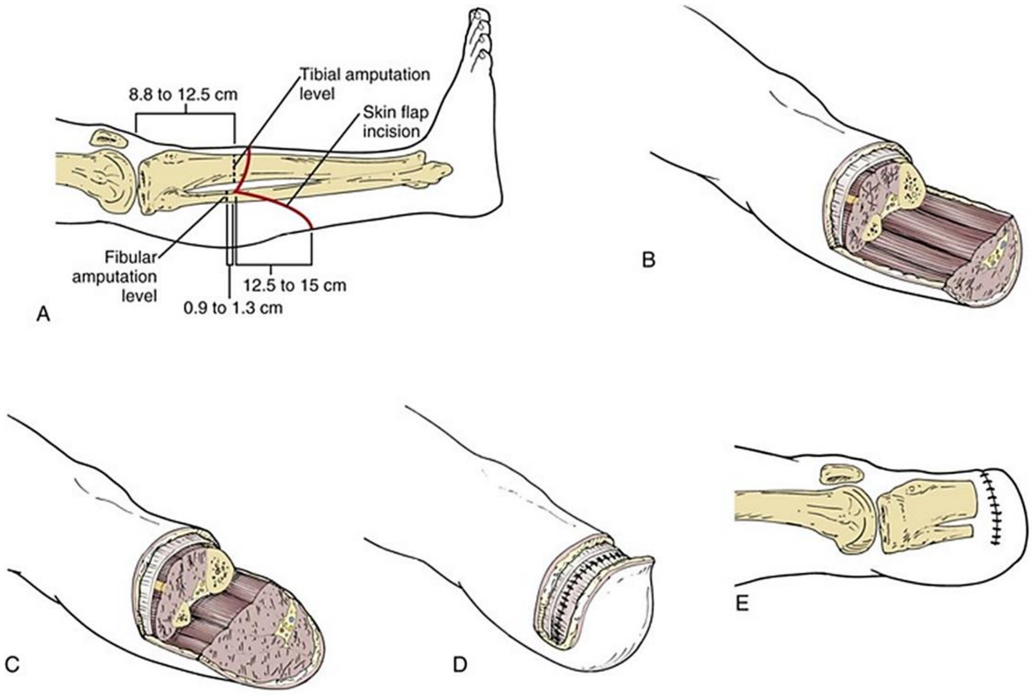


Figure 2.3 Below-knee amputation. A, Marking the skin incisions. B, Fashioning the flaps after bone transection. C, the soleus muscle is tailored to create a proper flap. D, the deep posterior fascia is sutured to the anterior deep fascia and periosteum. E, Closure (Gottschalk, 1999).

A lower limb prosthesis is carefully designed to serve as a replacement for a lost portion of the body (Figure 2.4). It connects to what remains of the limb through a socket, which is customized to fit the unique contours and requirements dictated by the nature of the amputation. This adaptable device is engineered to replicate different parts of the lower limb, such as the foot, ankle, shank, knee, and thigh. Its purpose is to bring back a measure of independence and movement to individuals who have experienced limb loss (Boone et al., 2012).

Figure 2.4 Components of the lower limb prosthesis (Selvam et al., 2021).

2.1.4 Pain associated with unilateral lower limb amputation

This section will focus on the pain associated with amputation. Low back pain and phantom limb pain are the most reported pain among TTAs. Low back pain and phantom limb pain are the most reported pain among TTAs (Ehde et al., 2000; Smith et al., 1999). Low back pain (LBP) is usually described as pain or discomfort in the lower part of the spine, which carries most of the body's weight. This pain, whether it is LBP or phantom limb pain (PLP), can be caused by various factors such as poor posture, muscle strains, nerve compression, and conditions like osteoarthritis or herniated discs (Clauw & Chroitsos, 1997). The spine, also known as the vertebral column or backbone, is a complex structure running from the base of the skull to the pelvis. It comprises 33 bones called vertebrae, separated by soft discs. The spine also contains the spinal cord and a network of nerves that send signals to and from the brain. The spine is divided into several regions: the cervical spine (neck), thoracic spine (mid-back), lumbar spine (low back), sacrum (pelvis), and coccyx (tailbone). Each region has its own characteristics and functions. For example, the cervical spine is the most flexible part, supporting the head and allowing head and neck movement. The thoracic spine supports the rib cage and protects the organs in the chest, while the lumbar spine is the most stable and supports the upper body, allowing lower body movement (Yeager, 1986).

LBP is often seen in people who have had a lower limb amputation. This pain can arise from various causes, such as standing posture, leg-length differences, and the way they walk (Trendelenburg gait; AlMehman et al., 2022; Butowicz, Acasio, et al., 2019). There isn't much research on the exact link between low back pain and lower limb amputation. However, it is known that amputation changes the way amputees walk and can increase the weight the intact limb has to carry, which leads to biomechanical changes and might lead to the pain. A study by Sadowski et al. (2022) found that 37.8% of amputees experienced low back pain. The study identified risk factors such as older age, higher body weight index (BMI), and a history of LBP before the amputation. It also noted that LBP significantly affected the quality of life, impacting their ability to work and enjoy leisure activities. Although this study only looked at one point in time and cannot prove cause and effect, it provides valuable information on how common LBP is and its impact on amputees (Sadowski et al., 2022). Another study by Devan et al. (2017) surveyed a large group of people with non-dysvascular lower limb amputations and found that 45.8% had low back pain. The study identified risk factors like age, amputation level, BMI, previous low back pain, and psychiatric issues. It suggested that targeting these risk factors might help reduce the prevalence of low back pain in these individuals (Devan et al., 2017). These studies have limitations, such as being cross-sectional, meaning they cannot establish causality, and having samples from one country, which may limit the generalization of the results. However, they provide important insights into the prevalence and factors associated with pain in amputees and highlight the impact of LBP on physical function and overall well-being. The pain influenced gait compensation, and the majority of amputees reported pain, but the relationship between pain and TTAs gait compensation is not well studied. Thus, the next chapter will review the literature for further investigation.

2.2 Gait spatial and temporal parameters

Gait analysis focusses on assessing human walking patterns, particularly spatial and temporal parameters that provide quantitative insights into the human gait:

- Spatial parameters include step length (the distance between the heel strikes of opposite feet); stride length (the distance between successive heel strikes of the same foot); and stride width (the lateral distance between the heels) during consecutive footfalls under spatial parameters (Figure 2.5).
- Temporal parameters, relate to gait's timing aspects as stance time (the duration of foot contact with the ground).
- Gait speed, and cadence (the number of steps per time unit) integrates spatial and temporal data, representing the distance covered over time.

These parameters are essential in clinical contexts for diagnosing and monitoring conditions like Parkinson's disease, stroke, and orthopaedic issues, where deviations from normal gait patterns can signal underlying problems. Various tools, such as motion capture systems and wearable sensors, are employed to measure these parameters accurately, aiding in the assessment and intervention planning for gait abnormalities (Bertoli et al., 2018; Bytyçi & Henein, 2021).

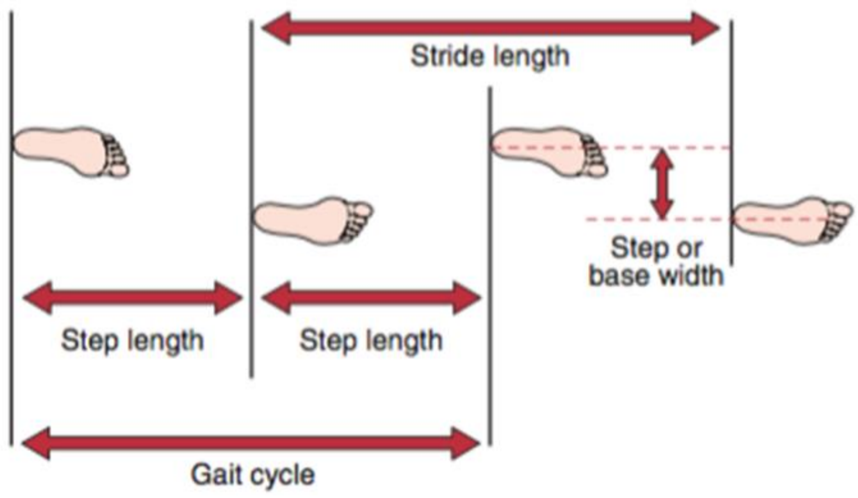


Figure 2.5 Spatial parameter of gait (Bytyçi & Henein, 2021).

2.3 Ground reaction force

GRF for gait analysis, especially for TTA, is important for understanding the dynamics of TTAs both feet (amputated and intact). The GRF is a critical component in assessing how individuals foot interacts with the ground during walking, providing valuable insights into the biomechanics of gait. Among TTAs, the analysis of GRF is even more significant as it helps in identifying asymmetries between the prosthetic and intact sides. These asymmetries can contribute into understanding three two main points:

1. Is there a difference between both feet (intact and amputated) for TTAs?
2. Is the presence of pain and pain reporting influencing the difference?

For TTAs, understanding GRF is crucial in several ways. GRF helps in evaluating the load distribution between the prosthetic and intact sides. This is important as amputees tend to overload the intact side, which can contribute to secondary complications such as osteoarthritis. In a study published in 2018, researchers focused on understanding the loading symmetry in lower-limb amputees, highlighting the complexity of these measurements and their importance in refocusing rehabilitation targets (Highsmith et al., 2018). Further, TTAs often exhibit asymmetric gait patterns, which can lead to increased load on the intact limb, contributing to pain (chronic pain or low back pain) and other joints. This asymmetry is typically observed in both kinetic and kinematic parameters. Studies have shown that the intact limb often bears more weight, leading to overuse injuries and chronic pain conditions (Eshraghi et al., 2014).

Overall, GRF measurements play a critical role in the comprehensive assessment of gait, particularly for TTAs, enabling the importance of including GRF in the gait analysis.

2.4 Three-dimensional kinematics and kinetics of human motion

The field of three-dimensional kinematics and kinetics of human motion is a critical area of study that provides insights into the complexities of human movement. It involves the measurement and analysis of motion in three-dimensional space, considering the forces and moments that cause each movement. The use of the VICON system, a motion capture system is widely recognized in biomechanics research. VICON systems utilize high-speed cameras to track reflective markers placed on an individual's bony landmarks, which corresponds to specific anatomical landmarks. The placement of these markers is important for accurate motion analysis and is typically guided by standardized models such as the Plug-In Gait model. The position of the markers is registered in real-time, allowing for the capture of dynamic movements across various tasks, whether it is simple walking or complex athletic manoeuvres. In combination with motion capture, force plates are embedded within the walkway to measure the GRF generated with no interference with the gait. These forces are integral to the understanding of the loads transmitted through the joints and the subsequent kinetic analysis. EMG, another key component in gait analysis, involves the use of sensors to record muscle electrical activity during motion. These data are essential for understanding the muscle activations that drive the movements, providing a comprehensive view of the muscular activity during the gait cycle.

The integration of kinematic data from motion capture, kinetic data from force plates, and muscle activity data from EMG allows for a complete analysis of human gait. This is particularly important in gait analysis for individuals with MSK deficit, such as lower limb amputation and in-depth analysis of the compensatory gait movement.

2.5 Musculoskeletal modelling and gait simulation

MSK modelling and gait simulation represent a convergence of biomechanics, neural control, physiology, and computer science to enhance our understanding of human movement. The application of these models, such as those created using OpenSim, has been instrumental in various fields, including rehabilitation, sports science, and the optimization of assistive devices. OpenSim, a comprehensive open source software for simulating MSK dynamics and neuromuscular control, enables researchers to simulate, predict and analyse human movement through computational models. The process begins by scaling an OpenSim MSK model to fit individual kinematic data, captured using motion capture systems like VICON. This personalization ensures that the model reflects the anatomical and functional realities of the subject, especially important in the context of gait analysis for individuals with limb amputations. By using inverse kinematic (IK) tools, the model can simulate gait patterns, which are then enhanced by incorporating GRF obtained from force plates. The integration of these forces allows for the estimation of joint torques through inverse dynamics (ID) analysis. Further refinement comes with the estimation of muscle activations, where static optimization techniques leverage the torque estimations to compute the muscle forces required for movement.

This holistic approach not only simulates gait, but provides insights into the muscular contributions to locomotion. The significance of MSK modelling and gait simulation extends to gait analysis, especially for TTAs. These simulations can offer a detailed understanding of the amputees' gait, revealing the amputated and intact side joints ROM and joints torque. This information is important not only to find the differences within TTAs, but also to compare them with the AB. The research and development of OpenSim models for individuals with transtibial amputation have further highlighted the benefits and challenges of adapting traditional models to reflect the unique requirements of amputees. These adapted models help in visualizing and analysing the gait of amputees. Thus, using a validated model will be effective in the gait simulation (Figure 2.6). The sequence begins with the initial data gathering from a laboratory setup. This data is then used to scale the OpenSim model, which is instrumental in deriving the kinematic parameters of gait, specifically the ROM at the joints during the walking cycle. Subsequently, the force plate data is collected, providing raw measurements that lead to the estimation of joint torque.

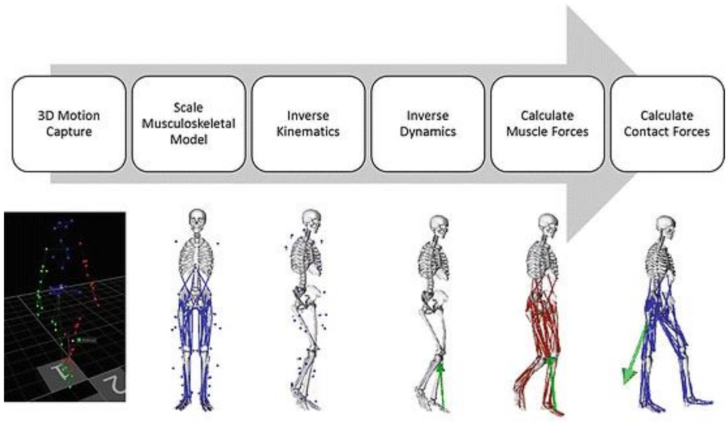


Figure 2.6 Illustration of the process of data collection and OpenSim modelling gait simulation (Wesseling et al., 2018).

2.6 Summary

In this chapter, an extensive overview of lower limb amputation, touching on its global incidence, causes, and the accompanying changes in pain and mobility has been provided. The chapter discussed the lower body's anatomy, physiological gait aspects, and the impact of pathological conditions on movement. Through examining amputation levels, surgical processes, and prosthetic components, the groundwork has been set for a detailed gait analysis discussion.

Chapter 3: Is abnormal movement in lower limb amputees a predictor of chronic pain? A systematic review

This chapter will discuss the relationship between altered movement patterns following lower limb amputation and the development of chronic pain. It will explore how changes in gait mechanics, muscle activation, and load distribution contribute to pain experiences in amputees. Furthermore, the chapter will get into the various tools and methodologies commonly used for assessing pain for amputees, including self-report scales, functional assessments, and advanced imaging techniques. Additionally, the chapter will investigate the modelling systems employed to measure amputees' kinematics and kinetics. These systems, such as motion capture technology and computational MSK models, provide detailed insights into the biomechanical alterations and compensatory strategies adopted by amputees. Understanding these changes is important for developing targeted interventions aimed at improving gait efficiency and reducing pain. By synthesizing current research and clinical practices, we will search about the effect of pain on the amputees’ compensatory gait biomechanics.

3.1 Introduction

Amputation is a life-change disability and globally, 1.2 million people each year experience a significant limb amputation due to vascular diseases (54%), trauma (45%) and cancer (less than 2%). According to the Nationwide Inpatient Sample, around 115,000 people in the United States undergo a lower limb amputation annually, of which 50,000-60,000 occur above the ankle joint. Additionally, 65% of these cases involve diabetics and 75% are due to peripheral artery disease (Nowygrod et al., 2006). In the UK, there were 25,312 amputations in 2014 with a prevalence rate per 100,000 (95% confidence intervals) of 26.3 (26.0–26.6) (Ahmad et al., 2014). Moreover, with the increasing incidence of diabetes and an ageing population, the number of amputations is expected to rise. It is predicted that the amputee population will double by 2050 (Ahmad et al., 2014). Individuals with lower-limb loss typically exhibit abnormal movement patterns. For example, abnormalities in gait include increased truck-pelvis movement and asymmetry in the lower limb kinematics between the intact and amputated limbs. In addition, compensatory mechanisms may develop in terms of performance criteria and muscle recruitment strategies (van der Kruk et al., 2021) to restore similar movement to before the amputation. These abnormal movements, along with their contributing factors, could increase the risk of musculoskeletal conditions secondary to the amputation, such as chronic pain, osteoarthritis in the intact limb (Ding et al., 2020) and osteoporosis in the amputated limb (Gailey et al., 2008). The prevalence of chronic pain in individuals with lower-limb amputation is high, with half experiencing phantom limb pain (pain in the limb that is no longer present), 55% to 76% experiencing residual limb pain, and 52% to 71% experiencing back pain (Davis, 1993; Fergason & Smith, 1999). While amputation-specific pain (phantom limb and residual limb pain) decreases or remains constant, back pain increases over time. In addition, pain in this body site has been reported to be 'more bothersome,' as it can interfere with daily activities and detrimentally affect functional outcomes and quality of life (Jensen et al., 2001). Understanding pain and its predictive factors is essential for improving functional outcomes post-amputation. Pain and functional outcome measures have been recommended for assessing the efficacy of physical rehabilitation treatments (Jensen et al., 2001). However, pain is complex. Its multidimensional and subjective nature make it challenging to measure accurately. A valid and reliable pain measure should consider the multifaceted complexity of chronic pain experience, such as location, severity, frequency, duration, interference with daily activities and response to pain relief treatments. While multidimensional pain assessment measures have been developed, such as the McGill Pain Questionnaire (MPQ) and Brief Pain Inventory (BPI), only a few are used in the amputee population to enhance the understanding of pain mechanisms and quantify outcome measures. The review provides a brief overview of the importance of biomechanics factors in the initiation and progression of chronic pain in people with lower limb amputation. The research questions to be addressed were 1) if there is evidence of the altered movement patterns following the lower limb amputation correlates with chronic pain; 2) which tools have been used to assess the functional movement and pain; 3) modelling system utilized and results; 4) which tools will be used in clinical practice to improve the management of the pain conditions secondary to the amputation.

3.2 Methods

Electronic databases, PubMed, CINAHL, MEDLINE, and SPORT Discus, were searched from the earliest records until March 2024, following the PRISMA Statement (Page et al., 2021).

3.2.1 Eligibility criteria

Studies were included in the review if they 1) focused on the biomechanics studies of activities of daily living in people with lower limb amputation; 2) used spatial-temporal data (speed, distance, step and stride characteristics); kinematic data (excursion, range of motion, coordinates, angles at or between segments/joints) or kinetic data (force, moment, power, impulse or electromyography) as an outcome measure; 3) used a questionnaire or tools to assess pain or discomfort; 4) peer-review and 5) publications written in English. Studies were excluded if they 1) were a case study or review design; 2) reported only imaging data;3) used bone-anchored prostheses.

3.2.2 Review process

After eliminating duplicates, two independent authors (Alsayed K and Ding Z) conducted a double screening of the titles and abstracts using the predefined inclusion criteria to identify eligible articles. Subsequently, full text manuscripts of the remaining articles were obtained and subjected to an additional screening process. To ensure a comprehensive search, reference lists from retrieved articles and previous systematic reviews were meticulously examined to identify additional relevant articles that might have been missed from the electronic searches. Consensus discussions resolved disagreements between the authors, ensuring a rigorous and reliable selection of articles for the review.

3.2.3 Quality assessment and data extraction

The methodological quality of the studies included in the review was assessed to determine the potential for bias. A checklist, adapted from similar studies, was utilised (Wilde et al., 2011). Modifications were made to account for the unique characteristics of the amputee population and chronic pain. This enabled an assessment on population description, experimental methodology and the reporting of results. Each item was assigned a sore of either positive (1) if the criterion was met or negative (0) if it was not. The Ratcliffe et al. rating score system was employed to rate the quality of the reviewed articles. Studies were categorised as high quality if they achieve a score of > 67%, medium quality for scores between 33% – 67%, and low quality for scores < 33%. Two authors (Alsayed K and Ding Z) independently evaluated the included studies using this checklist and compared the findings. Any disagreements between the authors were discussed and resolved by consensus. The following study details were extracted from included studies using a customised data extraction form: study aims, design, sample size, participant demographics, level of amputation, pain assessment tool/questionnaire used, task conducted, equipment used, body segments/joints analysed, kinematic and kinetic variables evaluated, statistical analysis technique, statistically significant outcomes. For biomechanics modelling studies, the modelling techniques/tools used to quantify the kinematic/kinetic variables that cannot be directly measured were also extracted.

3.3 Results

Figure 3.1 shows the PRISMA flowchart that illustrates the review process. Initially, a search identified 263 articles. After removing duplicates, 223 articles remained for further evaluation. Out of these, 64 articles were excluded based on the irrelevance to the review topic as determined by screening the titles and abstracts. Inclusion criteria were then applied to the full texts of 159 articles, of which 18 met the inclusion criteria.

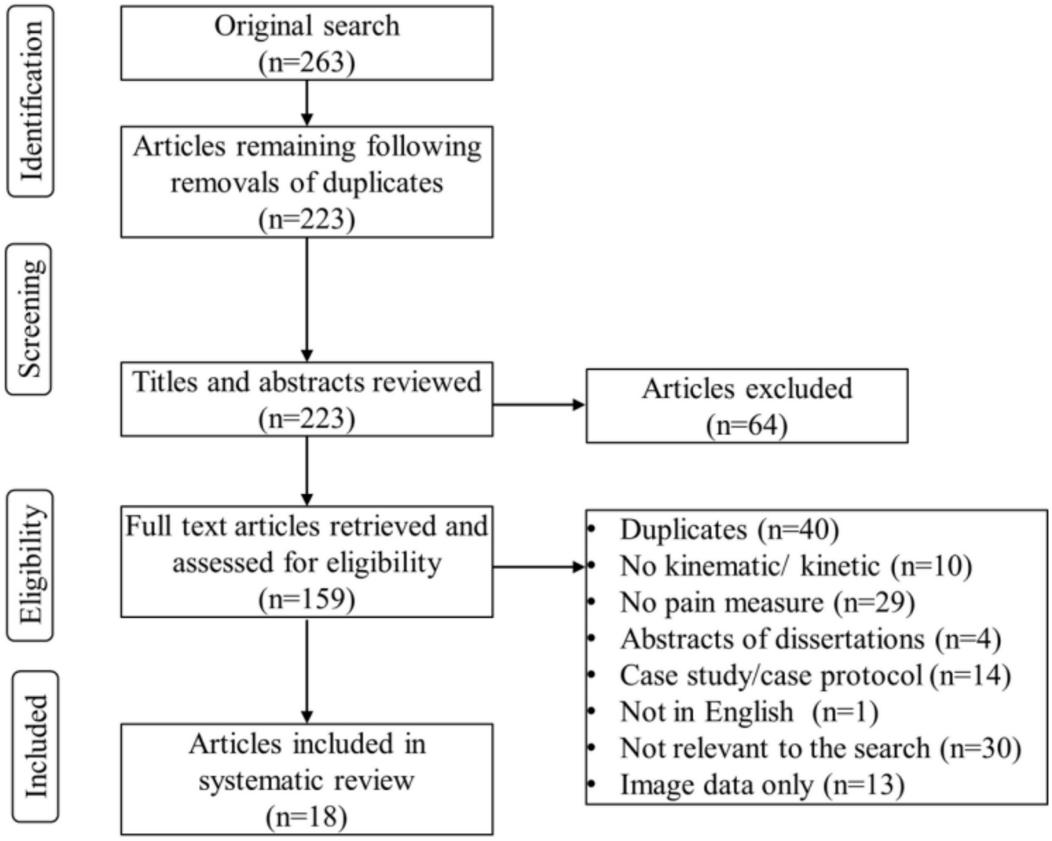


Figure 3.1 PRISMA flowchart of the study selection.

3.3.1 Quality of review studies

Sixteen studies received a high-quality score of over 67% (Acasio et al., 2022; Actis et al., 2018; Berge et al., 2005; Butowicz, Acasio, et al., 2019; Butowicz et al., 2018; Dillingham et al., 2019; Fatone et al., 2016; Honegger et al., 2021; Kulkarni et al., 2005; Mahon et al., 2020; Morgenroth et al., 2010; Petrini et al., 2019; Russell Esposito & Wilken, 2014; Segal et al., 2014; Talbot et al., 2017; Wolf et al., 2012), while the remaining studies were rated as medium-quality with scores ranging from 33–67% (Postema et al., 1997; Sjödahl et al., 2001). Table2.1 provides a summary of quality assessment for all included studies highlighting potential sources of bias. In particular, 14 studies adequately described the population, and three studies provides justification for the chosen sample size (TableA.1). All studies provided a detailed definition of the outcome measures with eight of them using a valid and reliable tool to assess the multidimensional feature of chronic pain. However, only four studies employed blinding of the assessors to group status and only two studies reported estimates of the random variability related to prosthetic components for the outcome measures.

3.3.2 Sample Size

The participant sample sizes ranged from 2 to 50 (with a median of 10). The majority of the studies (62%) of focused on transtibial amputees (Figure 3.2). Out of a total of 335 amputees, 52% were transtibial amputees with low back pain, while only 9% were transtibial amputees without low back pain; 12% of the participants were transfemoral amputees without low back pain. Six studies included both amputees and non-amputees as participants (Russell Esposito & Wilken, 2014); however, among these, four studies had incomparable groups in terms of age and BMI (Actis et al., 2018; Honegger et al., 2021; Morgenroth et al., 2010; Russell Esposito & Wilken, 2014). Furthermore, six studies categorised amputee participants into pain and pain-free groups (Butowicz et al., 2018; Fatone et al., 2016; Kulkarni et al., 2005; Morgenroth et al., 2010; Russell Esposito & Wilken, 2014; Wasser et al., 2021).

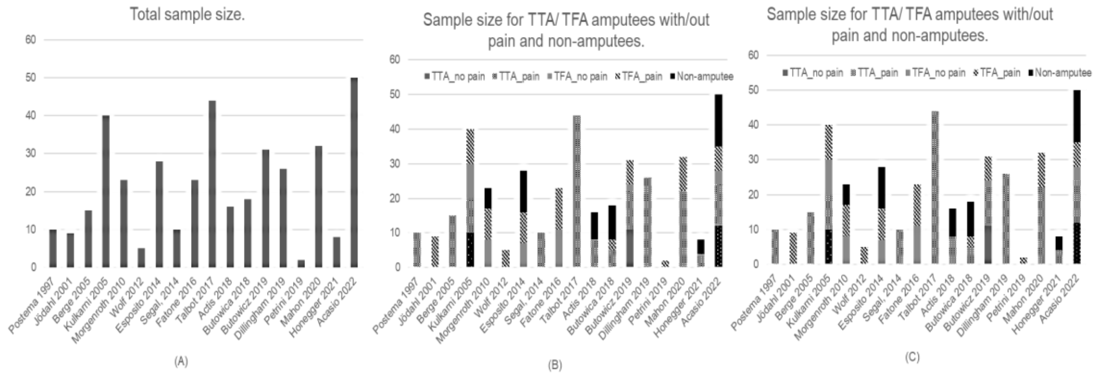


Figure 3.2 A) the total sample size in each study, B) transtibial amputees (TTA), transfemoral amputees (TFA) and non-amputee sample size in each study, C) the percentage of transtibial amputees (TTA), transfemoral amputees (TFA) and non-amputees in each study.

3.3.3 Pain questionnaire

In the pain questionnaires examined across 18 studies (Table A.2), pain severity emerged as the most frequently measured factor (16 out of 18 studies). Two of these studies used two pain-related questions to assess pain severity. Following pain severity, the location of pain was measured in 7 out of 18 studies, while pain interference with daily activities was assessed in 5 out of 18 studies, and pain frequency was considered in 4 out of 18 studies. Among the 18 studies analysed, the Visual Analog Scale (VAS) was the most commonly used pain assessment tool, employed in 7 of the studies. Notably, the Prosthetic Evaluation Questionnaire (PEQ) emerged as the preferred tool for assessing various dimensions of pain, encompassing severity, location, frequency, and duration (Table A.2).

3.3.4 Experimental measures

Walking speed was the primary spatiotemporal measure in walking gait. Trunk and lower limb kinematics were measured in 12 studies using either an infrared camera-based motion capture system or a goniometer. The difference between the intact and amputated limb kinetics was measured using force plates. Additionally, muscle activity during functional tasks was investigated in three studies (Table A.3).

3.3.5 Computational modelling tools

For kinematics modelling, three studies utilised the Vicon software tool (Oxford, UK) with two of them providing a detailed description of the kinematic model in Vicon. Additionally, nine studies employed Visual3D software tool (Washington DC, USA), but all of them treated the software tool as a black box, resulting in no available kinematic model information. Musculoskeletal modelling and finite element modelling techniques have been used to estimate kinetic data that are difficult or impossible to measure experimentally, such as muscle forces and tissue-level loads in sit-to-stand and level walking. During the sit-to-stand, greater abdominal muscle activity and higher tissue loading were observed compared to AB subjects ($p < 0.001$). During level walking, it was observed that unilateral lower limb amputees with chronic low back pain exhibited higher peak trunk muscle forces ($p < 0.005$) and spinal loads ($p < 0.013$; Table 3.2)

Table 3.1 Assessment of the methodological quality

Doman and item number	Description	% of studies scoring positive
Study population		
1	Was the study population adequately described?	100%
2*	Were both groups recruited from the same population?	78%
3*	Were both groups comparable for age, BMI/weight?	67%
4	Was the type of prosthesis investigated?	61%
5	Did the authors include a sample size justification?	17%
6	Were the eligibility criteria specified?	78%
Measurement and outcome		
7	Were the outcomes defined in detail?	100%
8$	Did the study used a valid and reliable tool to measure the multidimensional feature of chronic pain?	44%
9	Was a system for standalised movement measurement reported?	94%
10	Were the measures clearly defined, valid, reliable, and implemented consistently across groups?	89%
11	Did the method description enable accurate replication of the measurement procedures?	83%
12	Were the people assessing the outcomes blinded to the participants' group assignments?	22%
Data presentation		
13	Are the main findings of the study clearly described?	100%
14	Were the statistical tests appropriate?	83%
15	Were the results of between-group statistical comparisons reported for at least one key outcome?	94%
16	Did the study provide estimates of the random variability in the data for the main outcomes (prosthetic prescription)?	11%
17&	Did the study have sufficient power to detect a clinically important effect where the probability value for a difference being due to chance is less than 5%?	22%
18	Was the reliability and/or validity of the outcomes commented upon?	72%

Note:

*Positive if there is no difference between the pain and pain-free groups with lower limb amputations or the amputee and non-amputee groups.

$ If it is a valid and reliable tool to measure pain was determined by the pain literature review ().

&A clinically important effect was determined by the effect size (ES) > 0.5 (Seay et al., 2011).

Table 3.2 Summary of studies used modelling software tools and respective models.

Author, year	Software tool	Kinematic/kinetic model
Postema et al., 1997	Vicon	--
Sjödahl et al., 2001	Vicon	Vicon PlugInGait model
Berge et al., 2005	Vicon	In-house kinematics model based on Euler decomposition method
Kulkarni et al., 2005	Vicon, Visual3D	--
Morgenroth et al., 2010	EvART, Visual3D	--
Wolf et al., 2012	OrthoTrak, Visual3D	--
Russell Esposito & Wilken, 2014	Qualisys, Visual3D	--
Segal et al., 2014	Motion Analysis, Visual3D	--
Fatone et al., 2016	Motion Analysis, Visual3D	--
Talbot et al., 2017	Vicon, Visual3D	--
Actis et al., 2018	Motion Analysis, Visual3D, OpenSim	--
Butowicz et al., 2018	Qualisys, Visual3D	--
Butowicz, Dearth, et al., 2019	Vicon	--
Dillingham et al., 2019	Vicon	--
Petrini et al., 2019	Vicon	--
Mahon et al., 2020	Vicon, Visual3D	--
Honegger et al., 2021	EvART, Visual3D	A whole-body musculoskeletal model with lumbar spine; Finite element model of the lumbar spine.
Acasio et al., 2022	OrthoTrak, Visual3D	A non-linear finite element model of the spine at the T12 and the S1 vertebrae

-- indicates not applicable

3.4 Discussion

The review aimed to identify evidence supporting a potential link between altered movement patterns following lower limb amputation and the occurrence of chronic pain. While altered movement patterns have been widely reported, only one study provided evidence indicating an increase in low back pain among individuals with transtibial amputation (Mahon et al., 2020). Furthermore, in the studies examining new prostheses and rehabilitation regimens, a reduction in pain was observed, however, no significant differences in movement patterns were found (Dillingham et al., 2019; Postema et al., 1997; Sjödahl et al., 2001; Talbot et al., 2017).

Regarding the second question, the Visual Analog Scale was found to be the most frequently used pain assessment tool in amputee biomechanics studies. This popularity can be attributed to its ease of administration and simplicity, making it widely applicable in both clinical and research settings. However, pain in the amputee population is complex, with variations in its cause, location, duration, and impact on daily living. In this regard, the PEQ emerged as the most comprehensive tool for assessing the multidimensional pain experience (Shojaei et al., 2016). By incorporating multiple scales and domains, PEQ ensures that pain evaluation goes beyond a single dimension, offering a more nuanced perspective on the pain experiences of individuals with amputations. Regarding the third question, the majority of studies focused on investigating the whole lower limb (Berge et al., 2005; Dillingham et al., 2019; Petrini et al., 2019; Postema et al., 1997; Segal et al., 2014; Sjödahl et al., 2001; Talbot et al., 2017; Wolf et al., 2012), with seven studies solely examining the trunk (Butowicz, Acasio, et al., 2019; Butowicz et al., 2018; Fatone et al., 2016; Kulkarni et al., 2005; Mahon et al., 2020; Morgenroth et al., 2010; Russell Esposito & Wilken, 2014) and only three studies investigating both the trunk and the entire lower limb (Acasio et al., 2022; Actis et al., 2018; Honegger et al., 2021). However, due to the limitation of pain investigation (i.e., the location of pain was not investigated in most studies), there is currently no direct evidence showing a correlation between pain and the altered motion pattern in specific body segments. To comprehensively understand the altered movement patterns in relation to the onset of pain following lower limb amputation, a thorough examination of the trunk and the entire lower limb is essential. This examination should include investigating the lumbar-pelvic coordination, as well as the joints of the hip, knee, ankle, and the stump. As pain in lower limb amputees often occurs in the lower back, residual limb, and stump, a comprehensive analysis of these segments together is necessary to obtain a more complete understanding of the biomechanical changes and their potential impact on pain development. Additionally, researchers have demonstrated a relationship between low back pain and lumbar and hip joint movements (Papi et al., 2019). Therefore, current studies on back pain in amputees are limited to quantifying lumbar spine movements only, and the quantification of lower limb joint movements is lacking.

Regarding the fourth question, out of all the studies investigating amputee motion patterns, nine were conducted in a clinical setting, typically involving a multidisciplinary team of healthcare professionals, such as orthopaedic surgeons and physical therapists (Berge et al., 2005; Butowicz, Acasio, et al., 2019; Butowicz et al., 2018; Mahon et al., 2020; Morgenroth et al., 2010; Petrini et al., 2019; Segal et al., 2014; Sjödahl et al., 2001; Wolf et al., 2012). These studies performed clinical gait analyses with the primary goal of assessing and diagnosing gait abnormalities and functional limitations in individuals with limb loss, using sophisticated equipment, such as motion capture systems, force plates, and EMG. However, when dealing with variables that could not be directly measured, they frequently treated the software tools as a black box, making it challenging to compare their results. In particular, several studies (Berge et al., 2005; Butowicz, Acasio, et al., 2019; Butowicz et al., 2018; Mahon et al., 2020; Morgenroth et al., 2010; Petrini et al., 2019; Segal et al., 2014; Sjödahl et al., 2001; Wolf et al., 2012) analysed three-dimensional trunk and pelvis motions. For example: Butowicz et al., 2018 modelled the trunk as a single rigid segment, defined proximally by the sternal notch and C7 vertebrae, and distally at the T8 spinal level using Visual3D. On the other hand, Morgenroth et al., 2010 only focused on the lower thoracic spine when modelling the trunk. The different definitions of rigid body segments, joint coordinate systems, and relative motion between segments have rendered their findings incomparable. It is important to note that none of the mentioned studies utilized the ISB recommendations for defining the spine coordinate system in the analysis of spinal motion. To ensure more standardized and comparable results, it is recommended to follow the ISB recommendations for defining the spine coordinate system when modelling spinal motion, regardless of the models and software tools used. These recommendations define spine motion as the motion that occurs between the pelvis and the thorax. By adhering to these guidelines, future studies can enhance the consistency and reliability of their findings in evaluating amputee motion patterns. This, in turn, will provide valuable evidence to better understand the onset of back pain following limb amputation.

3.5 Conclusion

In conclusion, by addressing these aspects and utilising more comprehensive assessment tools, researchers can contribute to a better understanding of the complex relationship between altered movement patterns and chronic pain following lower limb amputation. This knowledge can lead to improved management strategies and improved patient outcomes in the amputee population.

This chapter offers a thorough examination of articles that explore the unusual movements of amputees and the resulting secondary complications, such as lower back pain. Also, discussed the abnormal kinematic and kinetic evaluation methods that have been mentioned in various literature for both unilateral lower limb amputees and non-amputees, along with the implications of these in clinical rehabilitation. The following chapter will illustrate how this tool can assist us in quantifying the kinematic and kinetic results for not only non-amputees, but also those with a unilateral lower limb amputation.

Chapter 4: Physical experimental protocol for amputee gait

Building on the systematic review from the previous chapter, which explored whether abnormal movement in lower limb amputees predicts pain, this chapter outlines a detailed research protocol. This protocol examines the relationship between various gait biomechanics variables—specifically gait spatial, temporal characteristics, GRF, and muscle activity of TTAs and compares them with the AB. The chapter focuses on detailing the laboratory equipment and setup, including the instrumented gait analysis including motion capture system (VICON), force plates, and EMG. TTAs will also complete a pain questionnaire to report their pain levels, which will be analysed in relation to their gait performance. This comprehensive approach aims to provide a thorough understanding of the biomechanics and pain-related outcomes in TTAs, offering valuable insights for improving gait analysis for TTAs.

4.1 Background

Lower limb amputation is an MSK condition that requires long-term rehabilitation (Willey et al., 2018). Common gait abnormality identified in the literature alterations in the walking speed, stride width, length, and stance time, as well as changes in the prosthetic and intact limb GRF; further, the effect extends to the imbalance in the muscle activation (Amma et al., 2021; De Marchis et al., 2022; Wagner et al., 2020). Key muscles contributing to gait include the tibialis anterior, gastrocnemius, vastus medialis, vastus lateralis, gluteus medius, and gluteus maximus (Ellis et al., 2014). In amputees, particularly those with transtibial amputation, some of these muscles may be partially or completely absent (e.g., tibialis anterior), and, the effect will reach residual muscles of the amputated limb as well as intact limb muscles (Wagner et al., 2020). Consequently, the gait biomechanics of AB can serve as a reference. Additionally, the difference between TTAs and AB gait biomechanics could be influenced by the pain frequency, duration, and intensity. To compensate for these changes and maintain balance, the body often adopts new movement patterns, like those without amputation (Wagner et al., 2020). Further, such compensatory movements can lead to secondary MSK disorders over time, including an increased risk of LBP, weakened muscles in the amputated limb, and osteoarthritis in the intact knee (Ding et al., 2020). This comprehensive approach in examining amputees' gait aims to provide a deeper understanding of the biomechanical changes associated with amputation and the potential use of pain reporting in predicting gait biomechanical changes. The Brief Pain Inventory (BPI) is a highly regarded tool in pain assessment because it offers a dual evaluation of both pain intensity and the degree to which pain interferes with an amputee's daily activities (e.g., walking). Unlike other pain questionnaires such as Prosthetic Evaluation Questionnaire (PEQ) and/or Visual

Analog Scale (VAS), which may focus solely on the severity of pain, the BPI discusses how pain affects various aspects, such as daily activities like walking, providing a more holistic view of the patient's condition.

This chapter protocol focuses on investigating gait biomechanics in two groups: experimental TTAs and a control group of AB individuals. Additionally, this chapter aims to investigate the effect of the pain reporting score on influencing gait abnormality among amputees.

4.2 Methodology

This study employed an experimental design to analyse the gait biomechanics of TTA and AB individuals. We used a motion capture system with two force plates and EMG sensors to measure gait spatiotemporal parameters, GRF, and muscle activation patterns. TTAs also completed a pain reporting questionnaire (BPI) to assess the amputees' pain levels. The method was meant to learn more about the mechanics of TTAs' walking by comparing it to the normal walking pattern seen in AB people. It was also meant to find out how pain affects the biomechanics of compensatory walking.

4.2.1 Research Design

The study was reviewed and approved by the University of Birmingham Research Ethics Committee (Approval Reference: ERN_21-0166). Written informed consent was obtained from all participants before participation. The participant information sheet (PIS) was provided to participants prior to consent, detailing the experimental procedures, the use of the motion capture system, force plates, and EMG sensors, and the purpose of the study. Participants were also informed about the presence of the force plates during the walking tests.

A quantitative approach was used to systematically compare gait biomechanics between TTAs and AB individuals. The independent variables included the participant group (TTA vs. AB) and the side of the limb (intact vs. amputated). The dependent variables included spatial and temporal gait parameters (e.g., cadence, step length, stride width, stride length, stance time), GRF, and muscle activity. Additionally, the BPI questionnaire for TTAs was used to assess the relationship between pain and gait biomechanics.

4.2.2 Participants

The study included a control group of AB individuals and an amputee group of TTAs. All participants were capable of walking independently without assistive devices, as per the criteria (Andrews et al., 2017).

Participants were eligible if they were aged between 18 and 65, had normal or corrected-to-normal vision, could walk independently for at least six minutes, and were proficient in English. For TTAs, eligibility required a minimum of six months of prosthetic limb use.

Exclusion criteria included:

- History of neurological diseases (e.g., Multiple Sclerosis, Stroke, Parkinson's disease).
- Severe MSK deformities in the lower limbs or spine.
- History of spinal or lower limb fractures or previous back surgery.
- Significant health conditions (e.g., heart, renal, or respiratory dysfunction).
- Current pregnancy.
- Skin conditions interfering with marker or electrode placement.

- COVID-19 symptoms or inability to speak or read English.

4.2.3 Instruments

For the gait analysis, a range of specialised instruments was utilised to ensure precise and comprehensive data collection. The VICON motion capture system (VICON, UK), equipped with eight cameras and reflective markers, was used to record detailed kinematic data. Markers were strategically placed on key anatomical landmarks and clusters as detailed in Figure 4.1 and Table 4.1, enabling accurate tracking of body movements in three dimensions. Ground reaction force (GRF) data were obtained using two force plates (AMTI, USA), which operated at a sampling frequency of 2,000 Hz. These force plates provided critical insights into the forces exerted by the participants during gait cycles. For muscle activity, ten Delsys Trigno wireless electromyography (EMG) sensors were employed. The sensors were positioned according to the protocols outlined in Table 4.2 and recorded at a frequency of 2,000 Hz. Four muscles (Gluteus maximums, Gluteus medius, Vastus medialis and Vastus lateralis) recorded bilaterally, and two muscles (Tibialis anterior and Gastrocnemius) recorded from the non-amputated side for TTAs and dominant side from AB group.

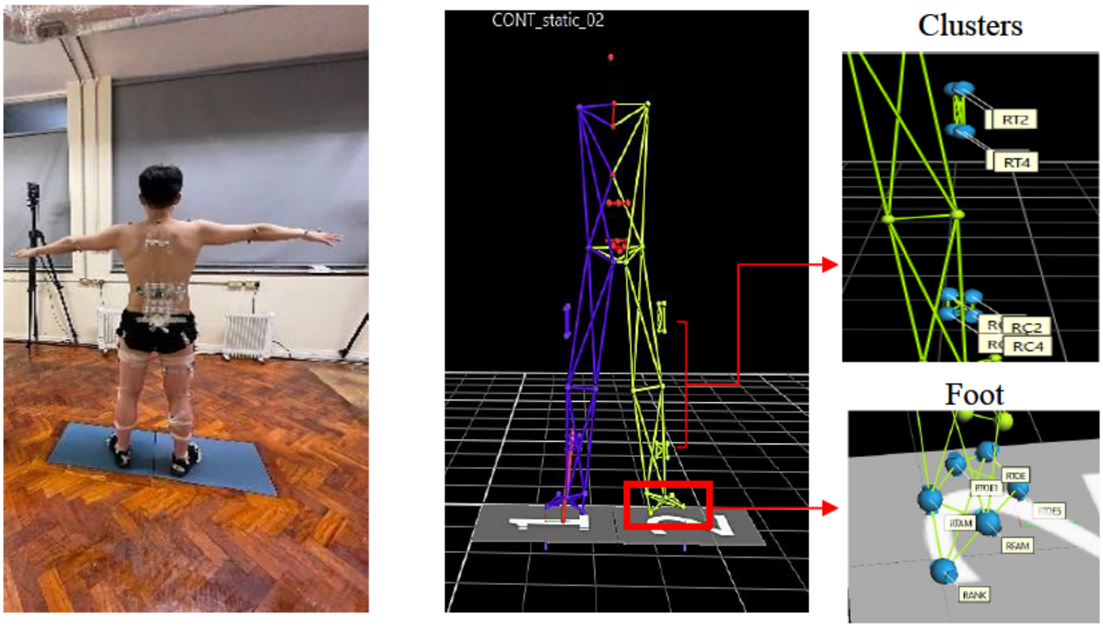


Figure 4.1 The markers placement

Table 4.1 Anatomical location of reflective markers on the lower back and lower limb segments (Ding et al., 2020; Ferraris et al., 2019).

Segment	Marker	Marker Label	Location	Prosthesis Specification
FOOT	Lateral & medial Malleolus	L/R FAM & TAM	Fibula apex malleolus & Tibia apex malleolus	Lateral / medial lower prosthesis rubber
	Calcaneus	L/R ANK	Heel	Same as intact foot
	First toe & Third toe & Fifth toe	L/R TOE1 & TOE & TOE5	On the first, third and firth toe over shoe	
LEG	Lateral and medial Knee Epicondyle	L/R FLE & FME	Lateral side of upper tibia & medial side of upper fibula	Lateral/ medial side of the prosthesis socket
	Shank and thigh cluster plate	L/R C1-4 & T1-4	Upper 1/3 of medial shank and middle medial thigh.	Matching location to the intact limb
PELVIS	Anterior and posterior Superior Iliac Spine	L/R ASIS & PSIS	Anterior and posterior aspect of the pelvis iliac crest	NA
Trunk	L5-S1 cluster	1-3 LSC	Middle between L/R	NA

			lumbosacral joint L5-S1	
	L1-L2 cluster	1-3 LLC	upper lumber area L1-L2	
	CTJ	C7	Cervicothoracic junction area C7-T1	
	Sternum Manubrium	STM	Upper part of the sternum	
	Xiphoid process	PX	Lower part of the sternum	
	Coracoid process	L/R ACR	inferior to the lateral aspect of the clavicle	

Table 4.2 Descriptive and figure of the anatomical location of EMG and muscle function on the lower back and lower limb segments (seniam.org).

EMG	Muscle	Function	Placement
1	**Gluteus maximus**	Extends, laterally rotates and lower fibres assist in adduction of the hip joint. The upper fibres assist in adduction. Through its insertion into the iliotibial tract, helps to stabilise the knee in extension.	The electrodes need to be placed at 50% on the line between the sacral vertebrae and the greater trochanter. This position corresponds with the greatest prominence of the middle of the buttocks well above the visible bulge of the greater trochanter.
2	**Gluteus medius**	Abduction of the hip joint. The anterior fibres medially rotate and may assist in flexion of the hip joint; the posterior fibres laterally rotate and may assist in extension.	Electrodes need to be placed at 50% on the line from the crista iliaca to the trochanter.
3	**Vastus medialis**	Extension of the knee joint.	Electrodes need to be placed at 80% on the line between the anterior spina iliaca superior and the joint space in front of the

			anterior border of the medial ligament.
4	**Vastus lateralis**		Electrodes need to be placed at 2/3 on the line from the anterior spina iliaca superior to the lateral side of the patella.
5	**Tibialis anterior**	Dorsiflexion of the ankle joint and assistance in inversion of the foot.	The electrodes need to be placed at 1/3 on the line between the tip of the fibula and the tip of the medial malleolus.
6	**Gastrocnemius**	plantarflexion of the ankle joint and assist in flexion of the knee joint.	Electrodes need to be placed on the most prominent bulge of the muscle.

4.2.4 Procedure

Before data collection, participants underwent a comprehensive preparation process. This began with obtaining informed consent, following their thorough review of the PIS, which detailed the study's purpose, procedures, and potential risks. We informed the participants about the experiment's setup, which involved using a 7-meter-long walkway with two force plates to study human walking patterns. We made the participants aware of the force plates during the walking tests.

After providing consent, participants were guided to the setup area, where reflective markers and EMG sensors necessary for motion analysis were carefully attached to their skin. During the walking tests, participants were instructed to walk naturally along the 7-meter walkway without focusing on the force plates to avoid interference with their gait. To ensure accurate and reliable data, the best three walking trials were selected where the participant's foot landed perfectly on the force plates. Before the real data collection started, each participant went on two to three practice walks. After that, they walked the route three times to get accurate gait data (Figure 4.2).

Additionally, amputee participants were required to complete the BPI questionnaire to report their pain levels, providing critical context for understanding gait abnormalities (Figure 4.3).

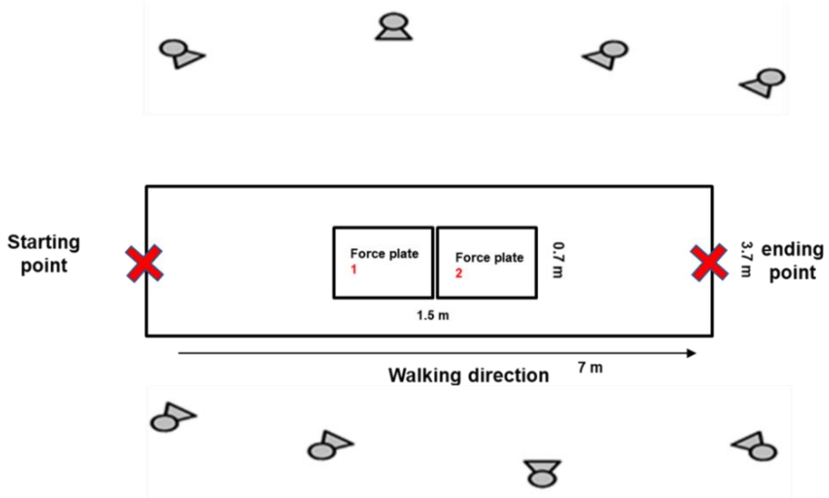
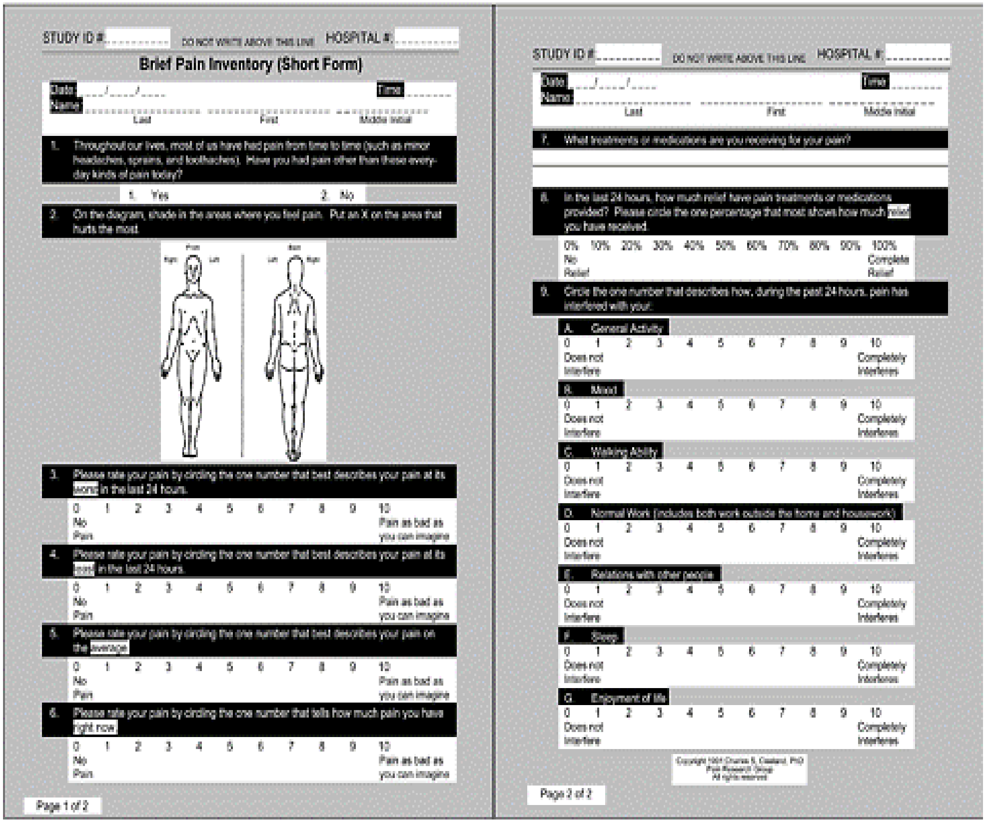


Figure 4.2 Illustration of the walking trial starting to the ending point and dimensions.

Data collection was divided into two phases: a general participant information phase followed by the experimental phase. In the general participant phase, participants' basic demographic information of age, mass, and height were recorded, and only amputees were asked to complete a pain assessment questionnaire. This preliminary stage was essential for establishing baseline data for each participant.

For the experimental phase, the preparation was carried out. The eight-camera VICON motion capture system and force plates were calibrated at the beginning of each data collection session. Once the equipment was ready, participants had reflective markers and EMG sensors strategically placed, as illustrated in Table 4.1 and Table 4.2. These EMG sensors were then calibrated in conjunction with the VICON system and force plates. This setup allowed for the precise capture of motion, force and muscle activity to be collected at the same time and stored in a secured file (.C3D) for further processing and analysis.

The study was conducted in the motion laboratory, located in Room G60 of the Old School of Engineering Building at the University of Birmingham (Figure 4.4).



STUDY ID #__________ DO NOT WRITE ABOVE THIS LINE HOSPITAL #__________

Brief Pain Inventory (Short Form)

Date: ___/___/___ Time: ________

Name: __________ __________ __________
Last First Middle Initial

1. Throughout our lives, most of us have had pain from time to time (such as minor headaches, sprains, and toothaches). Have you had pain other than these every-day kinds of pain today?

1. Yes 2. No

2. On the diagram, shade in the areas where you feel pain. Put an X on the area that hurts the most.

3. Please rate your pain by circling the one number that best describes your pain at its worst in the last 24 hours.

0 1 2 3 4 5 6 7 8 9 10
No Pain — Pain as bad as you can imagine

4. Please rate your pain by circling the one number that best describes your pain at its least in the last 24 hours.

0 1 2 3 4 5 6 7 8 9 10
No Pain — Pain as bad as you can imagine

5. Please rate your pain by circling the one number that best describes your pain on the average.

0 1 2 3 4 5 6 7 8 9 10
No Pain — Pain as bad as you can imagine

6. Please rate your pain by circling the one number that tells how much pain you have right now.

0 1 2 3 4 5 6 7 8 9 10
No Pain — Pain as bad as you can imagine

Page 1 of 2

STUDY ID #__________ DO NOT WRITE ABOVE THIS LINE HOSPITAL #__________

Date: ___/___/___ Time: ________

Name: __________ __________ __________
Last First Middle Initial

7. What treatments or medications are you receiving for your pain?

8. In the last 24 hours, how much relief have pain treatments or medications provided? Please circle the one percentage that most shows how much relief you have received.

0% 10% 20% 30% 40% 50% 60% 70% 80% 90% 100%
No Relief — Complete Relief

9. Circle the one number that describes how, during the past 24 hours, pain has interfered with your:

A. General Activity
0 1 2 3 4 5 6 7 8 9 10
Does not interfere — Completely Interferes

B. Mood
0 1 2 3 4 5 6 7 8 9 10
Does not interfere — Completely Interferes

C. Walking Ability
0 1 2 3 4 5 6 7 8 9 10
Does not interfere — Completely Interferes

D. Normal Work (includes both work outside the home and housework)
0 1 2 3 4 5 6 7 8 9 10
Does not interfere — Completely Interferes

E. Relations with other people
0 1 2 3 4 5 6 7 8 9 10
Does not interfere — Completely Interferes

F. Sleep
0 1 2 3 4 5 6 7 8 9 10
Does not interfere — Completely Interferes

G. Enjoyment of life
0 1 2 3 4 5 6 7 8 9 10
Does not interfere — Completely Interferes

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Figure 4.3 Brief pain inventory-short form (Bendinger & Plunkett, 2016).

Figure 4.4 Motion analysis lab located at G60, old School of Engineering Building.

4.3 Data Processing and Analysis

To process and analyse the data acquired from our gait analysis laboratory involved the integration of data from three primary sources: reflective markers, force plates, and EMG signals. Our data processing framework was designed to dissect gait spatial, gait temporal and integrated special and temporal parameters meticulously (Chapter 2). For the GRF data processing, force plates were used to extract mediolateral, vertical and anteroposterior direction, allowing for a precise understanding of the body forces acting on the force plate during gait. We analysed the EMG data by extracting the raw muscle EMG during the gait cycle, normalising the data to 101 data points, and finally rescaling the data between zero and one (Beauchet et al., 2017; Bukowski, 2006; Orekhov et al., 2019).

The methodology played an important role in identifying the TTAs gait biomechanical alteration and then compare to AB gait. Through this comparative analysis, valuable insights were gained into the compensatory gait mechanisms adopted by TTAs.

The statistical analysis of the data was conducted using two main approaches. Initially, unpaired t-tests were used to examine differences in demographic variables such as age, weight, height, and BMI between the two groups: TTAs and AB. Following this, a one-way ANOVA was used to assess if there were any statistically significant differences in gait biomechanics between the TTAs and AB individuals. If significant differences were found, Tukey's post hoc test was employed to further analyse the significance between the intact and amputated limbs of TTAs. The significance threshold was set at $p<0.05$ for all tests.

To assess the relationship between pain and gait outcomes, the statistical correlation between pain scores, as measured by the BPI questionnaire, and various gait parameters, including spatial and temporal characteristics, and GRF in amputees. The strength of these correlations will be represented by the coefficient of determination (R).

For the data analysis, SPSS (IBM Corp., 2020, IBM SPSS Statistics for Windows, Version 27.0, Armonk, NY: IBM Corp) was used.

4.4 Ethics and Limitations

The risk from the procedures proposed within this project is very low. All participants were advised that they can stop the experiment at any time. Non-invasive attaching procedures of the surface sensors and reflective markers include slight discomfort from minor abrasion of the skin area. A hypoallergenic light scrubbing paste was used, which is distributed by a medical supplier for this purpose. Grids were attached to the participant using a medical-grade adhesive foam and secured in place using Hypofix tape. This is commonly used in physiotherapy clinics. As all equipment was designed to be used clinically, it was not expected for there to be an allergic reaction; should this occur, all adhesive adhesives would be removed, the area washed with water and a moisturiser will be applied. Prior to electrode placement, the skin needed to be shaved to remove any hair. However, single-use disposable razors were used thus there was no expected risk from this procedure.

The motor tasks accomplished by the participants did not lead to anxiety or stress higher than that already expected from everyday life. To deal with this, appropriate rest time was provided throughout the experimental trials. Additionally, extra rest periods were given to the participants at any time if they needed them.

Participants were informed that if they notice any irritation during or immediately following the study, then they should inform the research team. A first aider from the Department would be called to assist them. The area would be cleaned with soap and water. Participants would be informed that if the irritation persists for more than 24-hours, they should inform the investigators of this study so that they may advise them as to whether to consult their local medical clinic.

Moving heavy equipment, such as the force plate and desks, in the laboratory may lead to injury. All devices in the laboratory are CE marked. Safety checks were made at the start and end of each session. Faults reported to Dr Ding for follow-up with the manufacturer. The laboratory space was kept clear, and the equipment was tidied between experiments.

Risk of exposure to COVID-19 for individuals including participants and researchers involved in the motion study in G60 has been mitigated. Study arrangements comply with site policies (Y3, School of Engineering Building) in respect of COVID-19. The number of individuals in the motion laboratory will satisfy the capacity (< 4 individuals in G60). Prior to starting any work in the laboratory, the COVID-19 return to work induction was completed. Each researcher also received a new laboratory induction, covering topics such as social distancing and zones. PPE like face coverings, gloves, disinfectant gels, and sanitisers were provided in the laboratory. Face coverings and gloves should be worn by any individuals in the laboratory.

4.5 Summary

This chapter offers an in-depth exploration of the study protocol, starting with the foundational background that informs the research. It progresses to detail the study's methodology, including the research design, characteristics of the participants, the instruments employed, and the procedures followed, as well as the statistical approaches for analysis. The chapter then described how the collected gait data will be synthesized to enhance the understanding of gait mechanics. The subsequent chapter will discuss the instrumented gait outcomes for both participant groups, comparing our findings with existing literature to contextualize the results within the broader field of study.

Chapter 5: Gait Biomechanics in Unilateral Transtibial Amputees and Able-Bodied Individuals: A Physical Instrumental Measurement Study

In the previous chapter, the study protocol was outlined, detailing the experimental design, data collection methods, instruments used for gait analysis, and laboratory setup. Additionally, data processing, analysis, and synthesis for examining gait biomechanics were discussed. Building on this, the current chapter will present the demographic information of the study participants, followed by the data of instrumented gait biomechanics. The chapter will focus on spatiotemporal parameters, GRF, and EMG outcomes for TTAs, comparing these with AB individuals. The spatiotemporal data will provide insights into variations in walking speed, stride width, cadence, step length, stride length, and stance time between TTAs and AB participants. GRF data will reveal differences in force distribution and peak forces during walking. EMG results will show muscle activation patterns, highlighting compensatory mechanisms in TTAs. Finally, the influence of the pain reporting on the amputees instrumented gait biomechanics will be investigated.

5.1 Background

Instrumented gait analysis, which includes spatiotemporal, ground reaction force (GRF), and muscle activity, is crucial for understanding the functional mobility of TTAs. This analysis offers insights into the biomechanical changes in gait for amputees compared to AB. Key spatiotemporal parameters such as walking speed, stride width, cadence, step length, stride length, and stance time, along with GRF and electromyography (EMG), are frequently analysed to assess the biomechanical impact of lower-limb amputation on their gait. Recent studies have highlighted significant biomechanical differences in gait between transtibial amputees and AB. Notable variations include changes in step length and cadence which are affected by the amputees' compensatory gait. GRF, especially at the first peak, reveal how amputees compensate during gait, with adjustments in knee and hip flexion to maintain balance and stability. (Eshraghi et al., 2014; Fukuchi et al., 2019).

Pain can lead to changes in gait mechanics as individuals adapt their movements to minimize discomfort. These adaptations often include altered stride length, cadence, and walking speed, which can be accurately assessed only when pain levels are measured and investigated with these changes (Walton et al., 2024), particularly through standardized pain reporting questionnaires like the Brief Pain Inventory (BPI), plays a significant role in unifying pain reporting outcomes among amputees. There is a need to measure pain and investigate the pain scoring relationship to the gait biomechanics for amputees. Despite the advancements in gait analysis technology and pain assessment methodologies, there remains a lack of comprehensive studies that integrate these two aspects. Most current research focuses either on biomechanical evaluations or on pain management independently, with less attention given to how subjective pain assessments can predict and influence specific gait parameters in transtibial amputees.

This study aims to fill the existing research gaps by:

- Comparing spatial, temporal gait variable, GRFs, and muscle activity between TTAs and AB to highlight key compensatory gait patterns for TTAs.
- Investigate the effect of pain levels using BPI on gait spatiotemporal parameters and GRF among TTAs.

This chapter aims to determine how increases in pain scores influence gait characteristics and compare these findings with AB individuals to identify any significant deviations or compensatory mechanisms employed by amputees.

5.2 Methods

This study employs a comparative design to evaluate gait parameters between TTA and AB. Participants, aged 18 to 65 years. The participants were divided into two groups: a control group composed of AB volunteers, and an amputee group consisting of individuals with unilateral lower limb transtibial amputations (TTA). The amputee group used a prosthesis for more than six months and was capable of walking without using assistive devices. This study has been approved by the University of Birmingham Ethical Committee (ERN_21-0166). An informed written consent form was obtained from each participant (Chapter 3).

A full gait cycle is defined as the interval between the heel strike of one foot and the subsequent heel strike of the same foot, starting at 0% with the first heel strike and ending at 100% with the second heel strike using maker trajectory placed on the heel. The heel strike is determined by identifying the lowest point in the vertical trajectory of the marker on the heel (Orekhov et al., 2019), as shown in Figure 5.1.

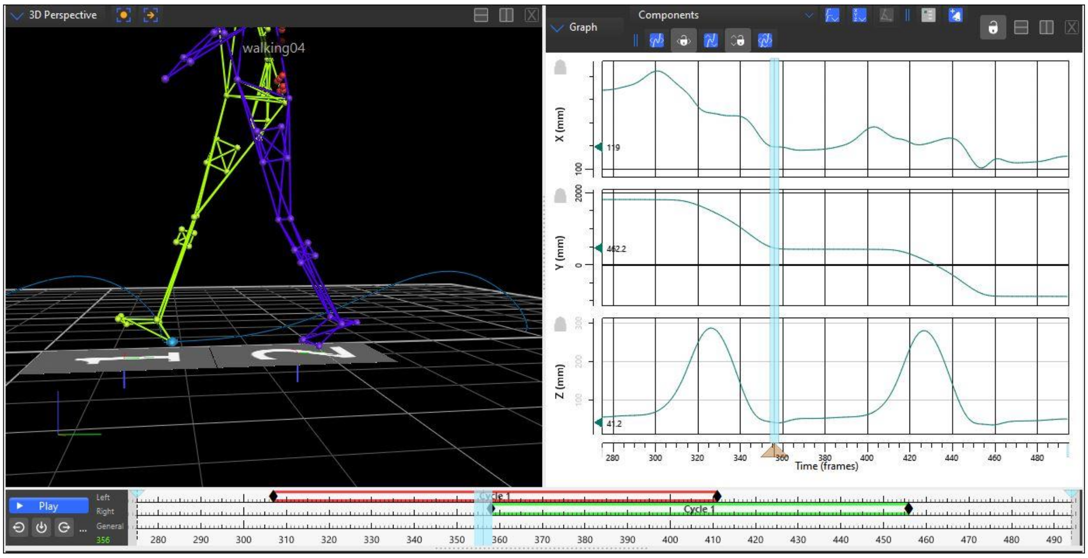

Figure 5.1 Illustration of the placement of trajectory markers within the VICON system, including the method used to label the gait cycle phases.

GRF and motion data were analysed using MATLAB software (The MathWorks, Inc., Natick, MA, USA). Before calculating the gait spatial, temporal, GRF, and EMG, data were sampled to 101 data points and in addition, GRF normalized to the body weight of each participant at the time of measurement. The parameters calculated included stance time, step length, stride width, walking speed, and cadence (Beauchet et al., 2017; Bukowski, 2006; Figure 5.2). Homogeneity of walking was assessed based on the absolute difference between the intact and prosthetic limbs for amputees, and between the left and right legs for AB individuals.

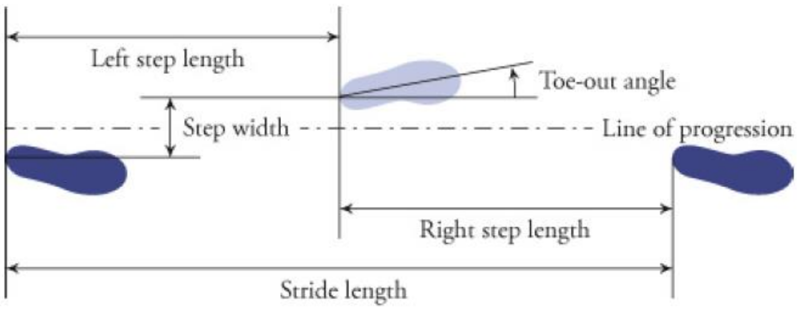

Figure 5.2 Illustration of gait spatial in the horizontal plane (Bytyçi & Henein, 2021)

5.3 Statistical Analysis

The normality of the collected data was tested before performing ANOVAs. All data are presented as the mean and standard deviation (mean ± SD). The spatial, temporal and GRF were analysed using a one-way ANOVA, with Tukey's post hoc test employed to compare sub-groups, with significance levels set at $p<0.05$.

The correlation between pain scores and gait outcomes has been conducted using Pearson's correlation coefficient (R). The R-value indicates the strength and direction of the linear relationship between two variables, with correlations categorised as strong, moderate, or weak. Specifically, R values between ±0.70 and ±1.00 were considered to represent a strong linear relationship, suggesting a close connection between the variables. Moderate correlations were identified with R values between ±0.40 and ±0.69, indicating a relationship with some variability. Weak correlations were defined by R values between ±0.10 and ±0.39, where the association between variables was present but not strong. Correlations with R values close to 0, typically between -0.09 and 0.09, were interpreted as indicating little to no linear relationship (J. F. Pallant, 2007). The statistical significance of these correlations was evaluated using the p-value, with a p-value less than 0.05 considered statistically significant. This threshold suggests that the observed correlation is unlikely to have occurred by chance, thereby highlighting the meaningful relationships between pain scores and gait instrumented outcomes (J. Pallant, 2020; J. F. Pallant, 2007).

The data were analysed using SPSS (IBM Corp. (2020), IBM SPSS Statistics for Windows, Version 27.0. Armonk, NY: IBM Corp).

5.4 Results

In this study, we included a total of 14 participants, divided into two groups: eight AB individuals in the control group, and six amputees in the TTA group. Table 5.1 details the demographics for each group.

The TTA group consisted of 4 males and 2 females with an average age of 45 years (±13 years), which was significantly greater than the AB group, whose average age was 29 years (±4 years; $p = 0.02$). Despite this age difference, there were no significant differences in height and mass between the groups. The TTA group had an average height of 1.77 m (±0.11 m) and an average mass of 82 kg (±21 kg), compared to the AB group, which had an average height of 1.70 m (±0.04 m) and an average mass of 75 kg (±12 kg; $p = 0.13$ for height and $p = 0.27$ for mass). Consequently, there was no significant difference in Body Mass Index (BMI) between the groups, with the TTA group having a BMI of 26.16 kg/m^2 (±6.57 kg/m^2) and the AB group a BMI of 25.97 kg/m^2 (±4.54 kg/m^2; $p = 0.36$).

Pain levels were assessed using the Brief Pain Inventory (BPI), focusing on worst pain, least pain, current pain, and pain interference with walking. The worst pain score averaged 4.33 (±2.05), the least pain was 1.17 (±0.69), and the current pain was 1.33 (±0.75). Pain interference with walking was rated at 2.33 (±1.11; Table 5.2).

Table 5.3 presents the correlation coefficients (R) between time since amputation, pain presence, pain reason, and various pain measures. The time since amputation showed weak correlations with pain measures, including worst pain ($R = 0.10$), least pain ($R = 0.20$), average pain ($R = 0.21$), current pain ($R = 0.24$), and pain interference with walking ($R = -0.30$), with no significant p-values observed. In contrast, pain presence showed strong correlations with worst pain ($R = 0.94$, $p = 0.01$), average pain ($R = 0.84$, $p = 0.03$), and pain interference with walking ($R = 0.94$, $p = 0.01$), indicating a significant relationship between these variables.

Table 5.1 Demographic characteristics of AB and TTAs.

Demographics	Age (y)	Gender	Height (m)	Mass (kg)	BMI (kg/m^2)
AB	29 ± 4	6M,2F	1.70±0.04	75±12	25.97±4.54
TTA	45±13* *p*=0.02	4M,2F	1.77±0.11 *p*=0.13	82±21 *p*=0.27	26.16±6.57 *p*=0.36

Note
BMI: body mass index
* $p<0.05$

Table 5.2 BPI outcome for TTAs

	Time since amputation (Years)	Pain presence	Pain reason	Worst	Least	Current pain	Pain interferes with walk
TTA 1	2.75	Y	PLP	4	1	1	3
TTA 2	3.58	N	NA	0	0	0	0
TTA 3	1.08	Y	LBP & PLP	6	2	2	3
TTA4	4.17	Y	LBP	5	1	2	3
TTA5	7.83	Y	PLP	6	2	2	2
TTA6	2.08	Y	LBP & PLP	5	1	1	3

Note
PLP: phantom limb pain
LBP: low back pain

Table 5.3 Correlation between pain measures and pain reporting among amputees.

Correlation (R)	Worst pain	Least pain	Current pain	Pain interferes with walk
Time since amputation	R=0.10 *p*=0.85	R=0.20 *p*=0.71	R=0.24 *p*=0.64	R=-0.30 *p*=0.57
Pain presence	R=0.94 *p*=0.01**	R=0.76 *p*=0.08	R=0.80 *p*=0.06	R=0.94 *p*=0.01**

Note:
*. Correlation is significant at the 0.05 level
**. Correlation is significant at the 0.01 level

5.4.1 Gait spatial and temporal parameters

We compared gait temporal, spatial, and spatiotemporal parameters between TTA and AB groups. We assessed both the intact and amputated limbs of the TTA group and the dominant and non-dominant limbs of the AB group.

Gait Temporal Parameter: Stance time, which indicates the duration each foot remains on the ground while walking, showed no significant differences between groups. The intact limb had a stance time of 0.75 ± 0.07 seconds (p=0.31), while the amputated limb had a shorter stance time of 0.69 ± 0.04 seconds (p=0.13). The AB group's dominant limb had a stance time of 0.73 ± 0.05 seconds, with the non-dominant limb showed lower value of 0.71 ± 0.10 seconds (p=0.60).

Gait Spatial Parameters: Stride width was greater in the TTA group with a value of 0.14 ± 0.03 meters compared to the AB group's stride width of 0.10 ± 0.02 meters (p=0.01). Stride length was consistent across both groups and conditions, with the TTA group showing 1.41 ± 0.18 meters for the intact limb and 1.54 ±0.18 meters for the amputated limb, while the AB group had stride lengths of 1.43 ± 0.11 meters for the dominant limb and 1.41 ± 0.10 meters for the non-dominant limb (p=0.80). Step length followed a similar pattern.

Integrated Spatial and Temporal Parameters: The speed of amputees' gait was 1.31 ± 0.01m/s, which is faster than the AB group's speed of 1.16 ± 0.05 m/s (p=0.02). The cadence, measured in steps per second, also differed, with the AB group had higher cadence of 1.80 ± 0.12 steps/sec compared to the TTA group's 1.65 ± 0.07 steps/sec (p=0.02; Table 5.4).

Table 5.4 Gait temporal, spatial, and spatiotemporal comparison for TTA and AB

Gait temporal parameter	TTA		AB	
	Intact	**Amputated**	**Dominant**	**Non-dominant**
Stance Time (s) *p =0.53*	0.75 ± 0.07 (p=0.31)	0.69 ± 0.04 (p=0.13)	0.73 ± 0.05 (0.60)	0.071 ± 0.10
Gait spatial parameter				
Stride Width (m) ***p =0.01***	0.14 ± 0.03 * (p=0.01)		0.10 ± 0.02	
Stride Length (m) *p =0.80*	1.41 ± 0.18 (p=0.36)	1.54 ± 0.18 (p=0.31)	1.43 ± 0.11 (p=0.84)	1.41 ± 0.10
Step Length (m) *p =0.62*	0.69 ± 0.08 (p=0.36)	0.75 ± 0.07 (p=0.31)	0.72 ± 0.05 (p=0.69)	0.70 ± 0.05
Integrated special and temporal parameters				
Speed (m/s) ***p =0.02***	1.31 ± 0.01 * (p=0.02)		1.16 ± 0.05	
Cadence (steps/sec) ***p = 0.02***	1.65 ± 0.07 (p=0.09)	1.61 ± 0.05* (p=0.03)	1.80 ± 0.12	

Note:
The p value on the first column represents the ANOVA significance level between TTA (intact and amputated) and AB (dominant and nondominant).
* Results for the post-hoc comparison: statistical difference between TTA and AB p<0.05.
† Results for the post-hoc comparison: statistical difference between prosthetic and the intact limb p<0.05.

5.4.2 Ground reaction force (GRF)

In both TTA and AB, the analysis of GRF during gait provided significant insights into the differences in biomechanical adaptations between/within groups. The GRF was analysed across three planes: mediolateral, vertical, and anteroposterior.

Mediolateral GRF: the first peak (lateral direction) occurred at different magnitudes and times between the groups. The amputated limb in the TTA group showed a higher peak force (-0.01 ± 0.00 BW) compared to the intact limb (-0.03 ± 0.0 BW) and the AB group (-0.02 ± 0.01 BW; p=0.01), with the peak occurring earlier in the gait cycle for the amputated limb (3% ± 2%). The second peaks (medial direction) were higher in the intact limb (0.08 ± 0.02 BW) compared to the AB group (0.06 ± 0.02 BW). Also, the second peak occurred significantly at 26%±6% (p=0.02; p=0.03) late for the amputated limb compared to intact and AB group.

Vertical GRF: the first peak was slightly lower for the amputated limb (1.06 ± 0.17 BW) compared to the AB group (1.08 ± 0.06 BW), with a significant difference in the time of peak occurrence, where the amputated limb showed a later peak (24% ± 4% of the gait cycle, p=0.02) compared to the AB group. The second vertical peak was also reduced in the amputated limb (0.97 ± 0.08 BW) compared to both the intact limb (1.00 ± 0.06 BW) and the AB group (1.10 ± 0.06 BW), with the peak occurring at 48% ± 3% in the gait cycle for the amputated limb.

Anteroposterior GRF: In the anteroposterior plane, the first peak (posterior direction) was lower in the amputated limb (0.05 ± 0.04 BW) compared to both the intact limb (0.14 ± 0.05 BW) and the AB group (0.14 ± 0.02 BW; p=0.02). This peak occurred earlier in the gait cycle for the amputated limb (9% ± 3%). The second peak (anterior direction) was also lower in the amputated limb (0.09 ± 0.05 BW) compared to the AB group (0.20 ± 0.02 BW), occurring at 54% ± 5% of the gait cycle. (Table 5.5 and Figure 5.3).

Table 5.5 Ground reaction force for AB and TTA.

GRF	TTA				AB	
	Intact peak (BW)	Intact peak occurs (%)	Amputated peak (BW)	Amputated peak occurs (%)	Peak (BW)	AB peak occurs (%)
Mediolateral						
1st peak (lateral)	-0.03±0.0 (p=0.13)	5%±1% (p=0.25)	-0.01±0.00*† (p=0.01; p≤0.01)	3%±2% (p=0.06; p=0.09)	-0.02±0.01	5%±2%
2nd peak (Medial)	0.08±0.02* (p=0.01)	18%±5% (p=0.37)	0.07±0.02 (p=0.08; p=0.18)	26%±6%*† (p=0.02; p=0.03)	0.06±0.02	19%±2%
3rd peak (Medial)	0.06±0.01* (p=0.01)	47%±4% (p=0.37)	0.06±0.02 (p=0.09; p=0.43)	44%±4% (p=0.11; p=0.17)	0.05±0.01	47%±1%
Vertical						
1st peak	1.09±0.11 (p=0.46)	18%±3% (p=0.28)	1.06±0.17 (p=0.38; p=0.31)	24%±4%*† (p=0.03; p=0.05)	1.08±0.06	19%±2%
2nd peak	1.00±0.06* (p=0.00)	52%±4% (p=0.17)	0.97±0.08* (p=0.01; p=0.28)	48%±3% (p=0.14; p=0.08	1.10±0.06	50%±2%
Anteroposterior						
1st peak (Posterior)	0.14±0.05 (p=0.45)	11%±1% (p=0.41	0.05±0.04*† (p≤0.01; p=0.02)	9%±3% (p=0.07; p=0.11)	0.14±0.02	11%±1%
2nd peak (Anterior)	0.16±0.05 (p=0.08)	59%±4% (p=0.06)	0.09±0.05*† (p≤0.01; p≤0.01)	54%±5%† (p=0.17; p=0.05)	0.20±0.02	56%±2%

Note:
BW: N/(kg*9.8).
Gait cycle (%): represents the peak value of the occurrence during the normalized gait cycle in percentage.
P value for the intact VS the able-bodied, and the amputated VS AB then VS intact limb respectively.
* Results for the post-hoc comparison: the statistical difference between TTA and AB $p<0.05$.
† Results for the post-hoc comparison: the statistical difference between amputated and intact limb $p<0.05$.
p-value under intact and amputated limbs represents the difference to the AB.

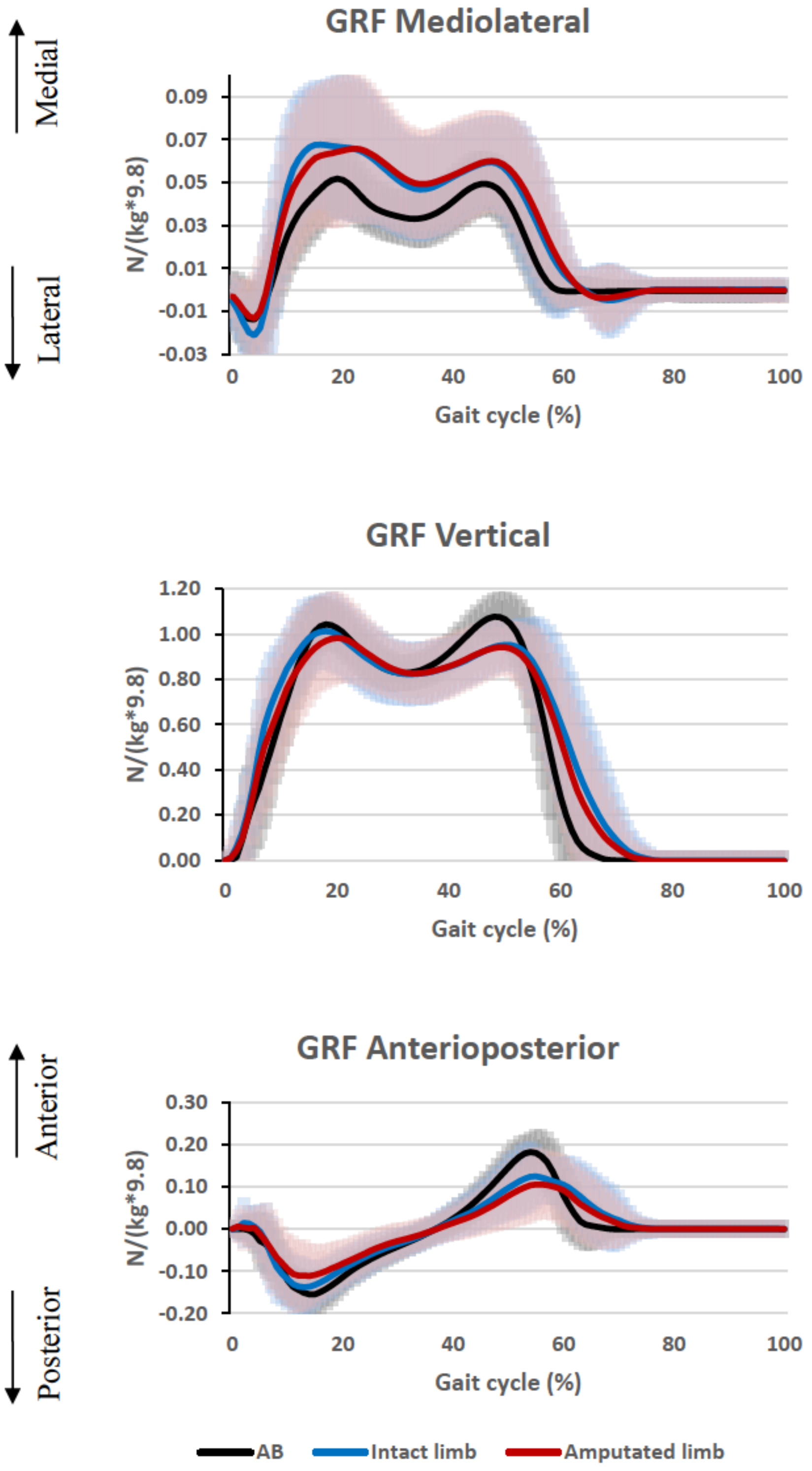

Figure 5.3 Mean and standard deviation of the ground reaction force of three limbs. Amputated (red colour) and intact (blue colour) represent the average of six TTAs; the AB (black colour) represents the average of eight AB. The y-axis represents the force N/(kg*9.8), and the x-axis represents the gait cycle (%).

5.4.3 Muscle activation

Gluteus Maximus: The control group exhibited an average muscle activity of 0.32 ± 0.11, higher than the intact side, which showed an average of 0.25 ± 0.06. The amputated side had a slightly lower activity compared to AB, at 0.29 ± 0.10.

Gluteus Medius: Muscle activity in the AB group was 0.31 ± 0.13. Interestingly, the intact side showed higher activity at 0.36 ± 0.05, while the amputated side had an activity level of 0.32 ± 0.10.

Vastus Medialis: In the AB group, the average activity for the vastus medialis was 0.26 ± 0.07. This was lower than the intact side, which had an activity of 0.33 ± 0.10, and the amputated side, which showed an activity of 0.32 ± 0.08.

Vastus Lateralis: The AB group showed an average activity of 0.30 ± 0.08, which is comparable to the amputated side's activity of 0.29 ± 0.13. The intact side had a slightly lower activity level of 0.27 ± 0.10.

Rectus Femoris: In the AB group, the rectus femoris had an average activity of 0.26 ±0.08. The intact side exhibited a higher activity of 0.31 ± 0.16, and the amputated side showed the highest activity at 0.38 ±0.17.

Gastrocnemius: The AB group demonstrated an average activity of 0.23 ±0.06, which was lower than the intact side's 0.30 ±0.12.

Tibialis Anterior: The AB group had an average muscle activity of 0.33 ±0.08. The intact side showed a slightly higher activity at 0.38 ±0.16 (Figure 5.4; Figure 5.5).

In summary, muscle activation levels varied among the groups, with the intact side often showing higher activity levels when compared to the AB group and the amputated side demonstrating significant variability across different muscles. This indicates distinct adaptations in muscle recruitment during the gait cycle among amputees and those with intact residual limbs.

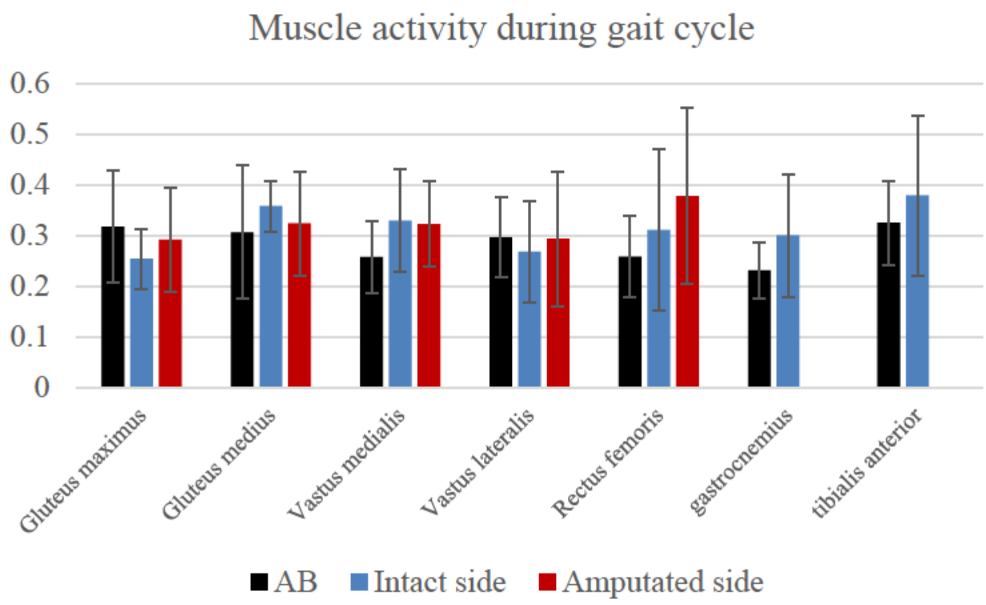

Figure 5.4 Muscle activity between limbs during the gait cycle.

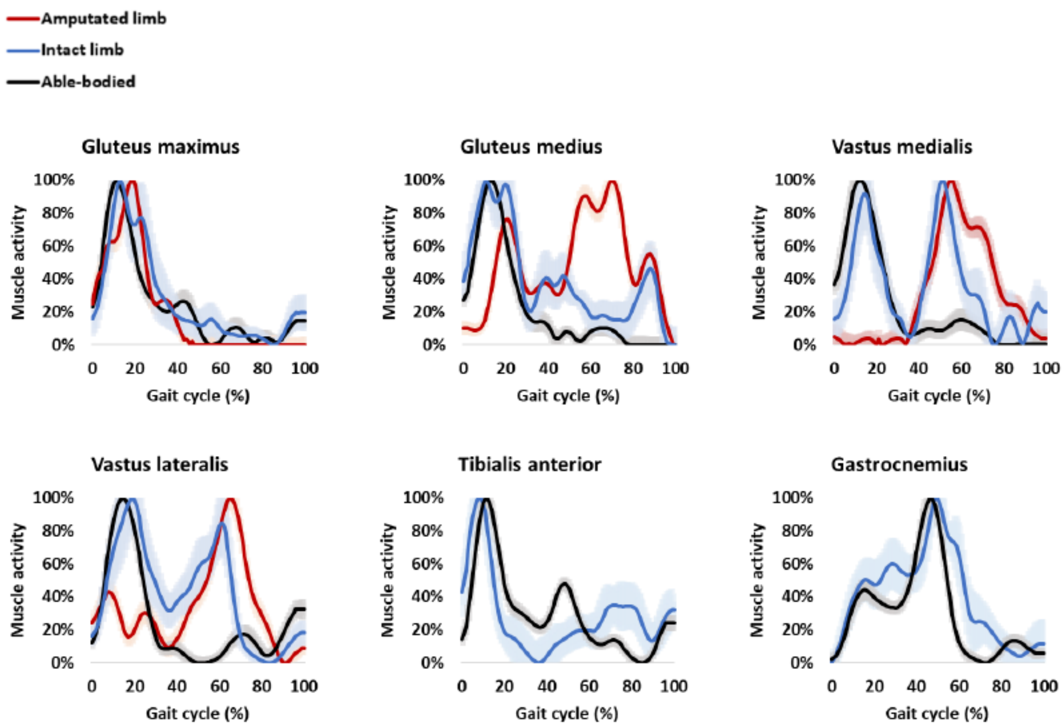

Figure 5.5 Muscle activity during the gait cycle, the name of each muscle presented on the left side. The red is the amputated side, blue intact side and black is the AB group.

5.4.4 Pain and instrumented gait

In this study, we examined the relationship between pain scores and various gait outcomes in participants with transtibial amputations. We conducted the analysis on both intact and amputated limbs, categorizing gait outcomes into temporal, spatial, spatiotemporal, and GRF parameters. The pain scores that were considered included worst pain, least pain, current pain (now), and pain related to walking ability.

Table 5.6 demonstrates significant correlations between gait outcomes and pain scores, particularly for TTA amputated limb. Strong positive correlations were observed between pain and several gait parameters, especially stride width, stride length, and step length. Stride width showed consistent significant correlations with pain, with the highest correlation when pain interfered with walking ($R = 0.96$, $p < 0.01$). Similarly, stride length demonstrated strong correlations with worst pain ($R = 0.93$, $p = 0.01$) and when pain interfered with walking ($R = 0.98$, $p < 0.01$), indicating that pain has a substantial impact on stride length reduction in amputees. In terms of speed, significant correlations were also found for amputees, especially when pain interfered with walking ($R = 0.92$, $p = 0.01$). The correlation between speed and pain was also significant for TTA intact limb under this condition ($R = 0.94$, $p < 0.01$), suggesting that both groups experience a reduction in speed when pain interferes with walking. Cadence showed weak and non-significant correlations, indicating that cadence is less influenced by pain.

Table 5.7 demonstrates the correlation between GRF and pain scores for TTA amputated and intact limb. In the medial-lateral (ML) 1st peak, significant negative correlations were observed for the TTA intact limb, particularly when pain interfered with walking ($R = -0.95$, $p < 0.01$) and for the least pain ($R = -0.94$, $p = 0.01$), suggesting a decrease in GRF with increasing pain. However, amputees showed weak and non-significant correlations in the ML_1st peak. In the ML_2nd peak, both limbs exhibited non-significant correlations across pain scores. In the

ML_3rd plane, strong correlations were found in intact and amputated limb, with the TTA intact limb showing significant correlations, particularly for pain interfering with walking (R = -0.88, p = 0.06), and amputated limb showing significant correlations for worst pain (R = -0.86, p = 0.03).

For the Vertical_1st plane, weak and non-significant correlations were observed in intact and amputated limb. However, in the Vertical_2nd plane, amputated limb demonstrated strong negative correlations with pain, particularly for worst pain (R = -0.87, p = 0.03) and least pain (R = -0.91, p = 0.01), indicating a reduction in vertical forces with increased pain.

In the anterior-posterior (AP) planes, both groups exhibited weak and non-significant correlations, with minimal impact of pain on GRF. Overall, amputated limb showed significant correlations between pain and GRF primarily in the ML_3rd and Vertical_2nd planes, while the TTA intact limb exhibited stronger correlations across multiple planes, particularly in the ML_1st and ML_3rd planes.

Table 5.6 Linear correlation between gait outcomes and pain reporting scores among unilateral lower limb amputees.

Gait outcome VS Pain score		Worst	Least	Current	Pain interferes with walk
Stance time	INT	R=0.68; p= 0.14	R=0.72; p= 0.11	R=0.67 p= 0.15	R=0.92**; p= 0.01
	AMP	R=0.81; p= 0.05	R=0.66; p= 0.15	R=0.68; p= 0.14	R=0.87*; p= 0.02
Stride width		R=0.94**; p= 0.01	R=0.85*; p= 0.03	R=0.85*; p= 0.03	R=0.96**; p≤ 0.01
Stride length	INT	R=0.60; p= 0.21	R=0.40; p= 0.43	R=0.49; p= 0.32	R=0.79; p= 0.06
	AMP	R=0.93**; p= 0.01	R=0.89*; p= 0.02	R=0.91*; p= 0.01	R=0.98**; p≤ 0.01
Step length	INT	R=0.56; p= 0.25	R=0.35; p= 0.49	R=0.46; p= 0.36	R=0.75; p= 0.09
	AMP	R=0.97**; p≤ 0.01	R=0.92*; p= 0.01	R=0.97**; p≤ 0.01	R=0.92**; p= 0.01
Speed		R=0.81; p= 0.05	R=0.78; p= 0.07	R=0.84*; p= 0.04	R=0.92*; p= 0.01
Cadence	INT	R=-0.74; p= 0.09	R=-0.73; p= 0.10	R=-0.72; p= 0.11	R=-0.94**; p= 0.00
	AMP	R=-0.32; p= 0.54	R=-0.48; p= 0.34	R=-0.36; p= 0.48	R=-0.65; p= 0.16

Note:
INT: intact limb
AMP: amputated limb
The R represented the correlation of the instrumented gait outcome with pain scoring.
*. Correlation is significant at the 0.05 level
**. Correlation is significant at the 0.01 level

Table 5.7 Linear correlation between gait GRF and pain reporting scores among unilateral lower limb amputees

GRF VS Pain score		Worst	Least	Current	Pain interferes with walk
ML_1st	INT	R=-0.88*; p= 0.02	R=-0.94**; p= 0.01	R=-.89*; p= 0.02	R=-0.95**; P≤ 0.01
	AMP	R=-0.49; p= 0.33	R=-0.24; p= 0.64	R=-0.30; p= 0.56	R=-0.22; p= 0.67
ML_2nd	INT	R=-0.76; p= 0.08	R=-0.64; p= 0.17	R=-0.79; p= 0.06	R=-0.67; p= 0.15
	AMP	R=-0.57; p= 0.24	R=-0.60; p= 0.21	R=-0.65; p= 0.16	R=-0.25; p= 0.64
ML_3rd	INT	R=0.96**; p= 0.00	R=0.79; p= 0.06	R=0.84*; p= 0.04	R=0.88*; p= 0.02
	AMP	R=0.79; p= 0.06	R=0.86*; p= 0.03	R=0.77; p= 0.08	R=0.87*; p= 0.02
Vertical_ 1st	INT	R=0.28; p= 0.58	R=0.28; p= 0.59	R=0.36; p= 0.48	R=0.52; p= 0.29
	AMP	R=0.28; p= 0.59	R=0.41; p= 0.42	R=0.42; p= 0.41	R=0.62; p= 0.19
Vertical_ 2nd	INT	R=-0.34; p= 0.51	R=0.00; p= 1.00	R=-0.15; p= 0.78	R=-0.20; p= 0.71
	AMP	R=-0.87*; p= 0.03	R=-0.91*; p= 0.01	R=-0.83*; p= 0.04	R=-0.81; p= 0.05
AP_1st	INT	R=-0.72; p= 0.10	R=-0.87*; p= 0.03	R=-0.88*; p= 0.02	R=-0.78; p= 0.06
	AMP	R=-0.39; p= 0.45	R=-0.38; p= 0.45	R=-0.42; p= 0.41	R=-0.06; p= 0.91
AP_2nd	INT	R=-0.75; p= 0.09	R=-0.69; p= 0.13	R=-0.56; p= 0.25	R=-0.71; p= 0.11
	AMP	R=0.05; p= 0.92	R=0.13; p= 0.80	R=0.24; p= 0.65	R=-0.23; p= 0.66

Note:
INT: intact limb
AMP: amputated limb
The R represented the correlation of the instrumented gait outcome with pain scoring.
*. Correlation is significant at the 0.05 level
**. Correlation is significant at the 0.01 level

5.5 Discussion

The primary goal of this chapter was to analyse the gait performance of individuals with TTAs within a controlled laboratory setting and compare it to that of AB individuals. This analysis focused on identifying key differences in stance time, step length, and stride length between the intact and prosthetic limbs of TTAs. We aimed to determine whether these differences exhibited a trend or were statistically significant when compared to the AB group. Additionally, variations in stride width, cadence, and speed between the TTA group and the control group were explored. Further investigation was conducted on the peak values of the GRF in the medio-lateral, vertical, and anteroposterior directions, comparing TTAs' intact and prosthetic limbs, as well as against those of AB individuals, to quantify how the TTA GRF measurements diverge from those of AB controls.

The secondary objective was to assess the impact of pain on gait performance among TTAs. Utilizing a standardized pain measurement tool BPI, the individual pain scores with instrumented gait metrics were used to determine if and how pain could influence measured gait parameters. This approach aimed to provide insights into the potential modulation of gait by pain experiences in amputees.

5.5.1 Gait spatial and temporal parameters

Our results are similar to those of other studies, like those by Jarvis et al. (2017) and Schmid-Zalaudek et al. (2022), which found that people who have lost a lower limb can walk comparatively normally in terms of time and space, especially when using advanced prosthetic limbs and after completing extensive rehabilitation programs. This supports the notion that modern prosthetic technology and rehabilitation strategies are effective in restoring gait characteristics close to those of non-amputees (Jarvis et al., 2017; Schmid-Zalaudek et al., 2022).

In our study, TTA demonstrated a significant increase in walking speed and stride width compared to the AB group. Specifically, the TTA group exhibited a walking speed of 1.31 m/s, which was significantly higher than the 1.16 m/s observed in the AB group ($p = 0.02$). Additionally, the stride width was notably greater in the TTA group (0.14 meters) compared to the AB group (0.10 meters, $p = 0.01$). These findings align with those of Jarvis et al. (2017), who also found that TTAs tend to have a faster walking speed and wider stride width compared to AB individuals. The increased stride width may reflect an adaptive strategy to enhance walking stability, compensating for the altered biomechanics associated with prosthetic limb use (Jarvis et al., 2017). In addition, our findings showed that the cadence of TTAs was significantly lower than that of AB individuals ($p = 0.02$) and that the prosthetic limb had a significantly lower cadence than both the intact limb and the AB group ($p<0.05$). This reduction in cadence is consistent with the literature (Jarvis et al., 2017; Schmid-Zalaudek et al., 2022) and suggests that TTAs may adopt a slower, more deliberate step frequency as part of their gait adaptation. Interestingly, there is limited research specifically comparing the cadence of the amputated and intact limbs, making our study's investigation into these variations an important contribution. Our results show that people who have had a unilateral transtibial amputation have a lot of different walking patterns, especially when it comes to cadence. The adaptive mechanisms required to fit the prosthetic limb likely contribute to the amputees' cadence (Ichimura et al., 2022). Additionally, we observed a shorter stance duration on the prosthetic side compared to the intact side, a finding that aligns with Schmid-Zalaudek et al. (2022), who suggested that this may be a compensatory mechanism to maintain balance and walking efficiency. The faster swing phase on the prosthetic side may help to minimize the time spent on a limb with reduced sensory feedback and stability, thereby reducing the potential for instability during gait (Schmid-Zalaudek et al., 2022). Even though these things are different, TTAs' overall walking patterns are not that different from those of AB people. This suggests that current prosthetic designs and rehabilitation programs are effective at not only restoring basic mobility but also improving gait symmetry and temporal dynamics (Kaufman et al., 2008). However, while our results are promising, they also highlight subtle but potentially impactful differences in gait mechanics that may not reach statistical significance but could have important functional implications for amputees. These findings underscore the ongoing need for advancements in prosthetic technology and personalised rehabilitation approaches, particularly those that enhance the integration and sensory feedback of prosthetic limbs. Our comparison of gait parameters between TTAs and AB individuals provides valuable insights into the effectiveness of investigating the gait temporal, and spatial for amputees and investigate how are they different from the control group (AB).

5.5.2 Ground reaction force

The study shows that there are big differences in GRF during the gait cycle between people with TTA and AB controls, mainly in the mediolateral, vertical, and anteroposterior planes. These differences are critical for understanding the biomechanical adaptations and compensatory mechanisms that amputees employ during walking.

In the mediolateral direction, our findings revealed significant differences between the intact and amputated limbs in the TTA group, as well as between these limbs and those of AB controls. Specifically, the first peak in the lateral direction occurred earlier and at a higher force in the amputated limb compared to both the intact limb and the AB group. In the intact limb, the second peak, which is in the medial direction, was significantly higher than in the AB group and occurred later in the amputated limb. These findings suggest that individuals with TTAs experience shifts in mediolateral forces, likely as a compensatory mechanism to maintain lateral stability during gait. This fits with earlier research by Kobayashi et al. (2023), who found that people who have had one limb amputated often have asymmetrical walking patterns that cause big changes in GRF, especially in the mediolateral direction, to help keep the body stable while walking (Kobayashi et al., 2023).

In the TTA group, vertical GRFs also demonstrated notable differences between the amputated and intact limbs, as well as when compared to AB individuals. Compared to the AB group, the first vertical peak was lower in the amputated limb and occurred later in the gait cycle, reflecting an adaptive strategy to reduce loading on the affected limb. This delay and reduction in peak force suggest that amputees may unconsciously modulate vertical loading to minimise discomfort or instability during the stance phase. Similarly, the reduced second vertical peak in the amputated limb reinforces the idea of a load reduction strategy. These results are in line with earlier research, like those by McCrory et al. (2001), which suggests that these changes may help protect the affected limb from too much stress, but they could also cause imbalances that change the way the person walks overall (McCrory et al., 2001).

In the anteroposterior plane, the first peak (posterior direction) was significantly lower in the amputated limb compared to both the intact limb and the AB group. This peak also occurred earlier in the gait cycle for the amputated limb, which could indicate a reduced ability to generate backward propulsive forces. The amputated limb similarly reduced the second peak (anterior direction), suggesting a compromise in forward propulsion during gait. These findings highlight a key challenge in prosthetic limb use: while prosthetics can restore some level of functionality, they do not fully replicate the natural kinetics of an intact limb. This discrepancy in anteroposterior forces may contribute to altered gait mechanics, potentially leading to long-term issues such as joint wear or discomfort. McCrory et al. (2001) discussed similar altered loading patterns in patients with hip arthroplasties, emphasising the need for improved prosthetic designs that better mimic natural limb function (McCrory et al., 2001).

The data from this study suggest that despite advances in prosthetic technology, individuals with transtibial amputations continue to face significant biomechanical challenges during walking. These problems manifest as altered GRFs in all three planes of movement. Compensatory mechanisms, essential for stability and mobility, trigger these changes, potentially increasing the risk of developing secondary musculoskeletal problems over time. These findings underscore the importance of ongoing research to investigate the muscle activity of both sides of amputees during gait.

5.5.3 Electromyograph (EMG)

We learnt a lot about the adaptive muscle activity patterns of people with amputations compared to people who are not. Our findings reveal how the loss of a lower limb necessitates compensatory mechanisms and altered motor control strategies, which impact various muscles differently throughout the gait cycle.

The gluteus maximus showed distinct patterns of muscle activity across the groups. The AB group exhibited the highest overall muscle activity, with an average of 0.32 (±0.11), while the amputated side of the TTA group showed slightly lower activity at 0.29 (±0.10). Interestingly, the intact side of the TTA group had lower activity than the AB group, averaging 0.25 (±0.06). These results suggest that while the amputated side compensates for the loss of the limb, the intact side also undergoes significant adaptation, potentially supporting and stabilising the body during walking. The lower activity on the intact side compared to the AB group may indicate a redistribution of muscular effort across the body to manage the altered biomechanics caused by amputation.

Muscle activity in the gluteus medius showed a different pattern, with the intact side of the TTA group demonstrating higher activity (0.36 ± 0.05) compared to both the AB group (0.31 ± 0.13) and the amputated side (0.32 ± 0.10). This elevated activity on the intact side likely reflects its role in providing additional stabilisation and balance, compensating for the reduced functionality of the amputated limb. The amputated side also showed significant muscle activity, suggesting ongoing compensatory efforts, although these were slightly less than those observed on the intact side.

The vastus medialis muscle was more active on both the intact and amputated sides of the TTA group, with activation levels of 0.33 ± 0.10 and 0.32 ± 0.08, respectively, compared to the AB group (activation levels of 0.26 ± 0.07). This suggests that the TTA group relies more heavily on the vastus medialis for stability and movement on both sides, potentially due to compensatory mechanisms following amputation. This indicates that the vastus medialis on both sides of the body in TTA individuals is highly active during gait, likely to stabilise the knee joint and assist in propulsion. The increased activation on the amputated side, in particular, suggests a compensatory response to maintain knee stability and support during the stance phase.

In the vastus lateralis, the AB group and the amputated side of the TTA group exhibited comparable muscle activity levels, with averages of 0.30 (± 0.08) and 0.29 (± 0.13), respectively. The intact side showed slightly lower activity (0.27 ± 0.10), which might indicate a redistribution of effort as the amputated side takes on a greater role in maintaining gait dynamics. The fact that the AB group and the amputated side had similar levels of activity suggests that, despite the difficulties of using a prosthetic limb, the vastus lateralis can still work approximately the same way it does in people who are not amputees.

The rectus femoris in the TTA group showed notably higher activity on the amputated side (0.38 ± 0.17) compared to both the intact side (0.31 ± 0.16) and the AB group (0.26 ± 0.08). This increased activity on the amputated side reflects the additional demand placed on the quadriceps muscle to compensate for the loss of natural limb function. The elevated activity on the intact side also suggests that both limbs are working harder than in AB individuals to maintain effective gait mechanics, with the amputated side taking on a particularly significant role in this compensatory process.

The gastrocnemius muscle showed increased activity on the intact side of the TTA group (0.30 ± 0.12) compared to the AB group (0.23 ± 0.06). This increased activity is likely due to the fact that the gastrocnemius muscle of the intact limb has to work harder to provide propulsion during the push-off phase of gait to make up for the lost contribution from the amputated side. This adaptation is critical for maintaining forward momentum and stability, given the limited capacity of the prosthetic limb to generate similar forces.

The tibialis anterior demonstrated higher activity on the intact side of the TTA group (0.38 ± 0.16) compared to the AB group (0.33 ± 0.08). This suggests that the intact side is actively compensating for the reduced dorsiflexion capability of the amputated limb, ensuring proper foot clearance during the swing phase, and preparing for a stable heel strike.

Overall, the EMG results show that muscle activation is very different in people with TTA compared to people with AB. This is because people with TTA have a lot more adaptive and compensatory mechanisms in place. People with TTA have to make big adjustments to their spatiotemporal parameters and ground reaction forces while they walk. Their increased muscular demand and altered activation patterns demonstrate this. Various factors, including pain, could influence these adaptations, suggesting significant biomechanical adjustments. Therefore, it is critical to consider the potential role of pain in these compensatory mechanisms and develop more effective rehabilitation strategies that address both the physical and sensory challenges faced by amputees. By enhancing our understanding of these muscle activation patterns, we can better tailor interventions to improve gait efficiency and reduce the risk of long-term complications for individuals with transtibial amputations.

5.5.4 Pain and instrumented gait

While biomechanics has received extensive attention in amputee research, the impact of pain on gait patterns, particularly in relation to amputation, has received less attention. This chapter provides a detailed exploration of the relationship between pain and various gait outcomes, revealing significant insights into how pain influences the walking patterns of individuals with transtibial amputations.

The results of this study underscore the profound impact of pain on gait dynamics, with specific emphasis on the strong correlations observed between pain scores and several key gait parameters. Our findings align with previous research, such as Orekhov et al. (2019), who reported that different activities, like walking or cycling, can influence knee movements and the risk of knee pain in amputees. We observed significant variations in gait parameters such as walking speed, stride width, and cadence in our study, closely linked to pain levels (Orekhov et al., 2019).

For instance, we found that stride width exhibited strong negative correlations with pain scores, particularly with worst pain and current pain on the amputated limb ($R = -0.94$, $p = 0.01$; $R = -0.93$, $p = 0.01$). This suggests that as pain intensifies, individuals tend to narrow their stride, possibly as a strategy to reduce discomfort and maintain balance. This is further supported by the correlation between stride width and mood, indicating that emotional factors influenced by pain can also affect gait stability.

Pain also significantly impacted gait speed. The strong negative correlation between gait speed and worst pain on the amputated limb ($R = -0.87$, $p = 0.02$) highlights how increased pain can lead to slower walking speeds. This finding emphasises the dual impact of physical and emotional pain on overall mobility, further underscoring the need for comprehensive pain management strategies that address both aspects to improve gait outcomes for TTAs.

In the analysis of GRF, the ML and vertical peaks showed significant correlations with pain. For the intact limb, there was a strong negative correlation between the first ML peak and various pain measures, including the worst pain and pain interference during walking. This suggests that as pain levels rise, the lateral stability during gait decreases, which may reflect a compensatory mechanism to minimise instability. The third ML peak on the intact limb also demonstrated strong correlations with pain, indicating that pain significantly impacts lateral forces during the latter stages of the gait cycle.

The second peak on the amputated limb, particularly the vertical GRF peaks, showed strong negative correlations with pain scores. These findings suggest that higher pain levels are associated with decreased vertical loading during gait, likely due to a reduced push-off force or an adaptive strategy to minimize discomfort on the affected limb. This reduced vertical force could contribute to altered gait mechanics and increased reliance on compensatory movements.

While the AP GRF peaks generally showed weaker correlations with pain compared to the ML and vertical peaks, they still provided insight into how pain may influence forward and backward forces during gait. The less pronounced correlations suggest that the compensatory mechanisms in the anteroposterior plane may be subtler or less directly influenced by pain levels.

Overall, these findings highlight the critical role of pain in altering gait dynamics in individuals with transtibial amputations. The significant correlations between increased pain and reduced stride width, slower gait speed, and altered GRF patterns emphasize the need for targeted interventions to manage pain effectively. By improving pain management strategies, we can potentially enhance gait stability and mobility, thereby improving the overall quality of life for individuals with amputations.

These findings provide important insights into the relationship between pain and gait outcomes in TTAs. The results underscore the complex interplay between physical and psychological pain and gait mechanics, highlighting the importance of comprehensive rehabilitation strategies that address both the biomechanical and sensory aspects of pain. Understanding these relationships allows for better prediction of compensatory gait patterns and supports the development of more effective interventions to mitigate the impact of pain on walking ability.

5.6 Conclusion

This chapter offers an instrumented gait analysis for individuals with amputations and compared to controls (non-amputees), highlighting key differences in spatial, temporal, GRF, EMG, and the influence of pain on gait dynamics. The findings highlight the amputees' compensatory gait in terms of walking speed, stride width, and cadence. However, subtle yet significant variations in gait mechanics, particularly in cadence, stance time, and GRF, indicate the ongoing challenges faced by amputees in achieving biomechanical symmetry and stability during walking.

Furthermore, the investigation into the impact of pain on gait performance highlights the critical role of pain level on amputee's gait variable. The strong correlations between pain levels and alterations in stride width, gait speed, and GRF patterns suggest that pain significantly influences gait mechanics, necessitating targeted interventions to address both physical and psychological aspects of pain.

Overall, this study provides valuable insights into the complex interplay between compensatory gait adaptations, and pain reporting questionnaire (BPI) among amputees. The findings support the need for further advanced gait analysis.

Thus, the next chapter will illustrate using the MSK model using OpenSim to simulate and calculate the gait kinematics and kinetics for amputees and compare their findings with AB for advancing our understanding about amputees' gait alteration.

Chapter 6: Gait Biomechanics in Unilateral Transtibial Amputees and Able-Bodied Individuals: OpenSim Modelling and Simulation

The previous chapter presented and analysed the demographic data for TTAs and AB participants for their instrumented gait biomechanics. Additionally, it investigated the impact of pain reporting on gait outcomes. This chapter will focus on employing the musculoskeletal (MSK) model OpenSim for gait simulation. This chapter will highlight OpenSim application in gait simulation, estimating critical biomechanical parameters such as joint angles, joint torques, and muscle forces.

The chapter will begin by outlining the steps required to set up and execute gait simulations in OpenSim. This process includes preparing the input data, scaling the MSK model, and conducting the simulations to capture the dynamic movements of the participants. Subsequently, the chapter will discuss how to utilize the results from these simulations to compute key biomechanical metrics.

6.1 Background

The investigation of the gait spatiotemporal, GRF, EMG and pain measurement with TTAs underscores the need for advanced modelling techniques to understand gait kinematics and kinetics of TTAs and to compare them to AB individuals. This chapter will illustrate how OpenSim, an open-source software for biomechanical modelling and simulation, provides a robust platform for these advancements. By utilizing OpenSim, it is possible to gain deeper insights into the biomechanical adaptations and compensatory mechanisms essential for comprehending joint kinematics (joint range of motion), and joint kinetics (joint torque and muscle force) in TTAs. The chapter will then compare these findings with those of AB individuals, regardless of the prosthesis type used by TTAs.

OpenSim is a comprehensive software tool designed for creating and analysing dynamic simulations of human movement (Delp et al., 2007). It offers a suite of features to build accurate MSK models, perform simulations, and analyse results. The platform supports the study of various human motions by providing detailed anatomical and functional representations of the MSK system. A pivotal development in OpenSim modelling is the full-body musculoskeletal model created by Apoorva Rajagopal and colleagues. This high-fidelity model represents lower limb musculature and overall body dynamics (Rajagopal et al., 2016). Building on the Rajagopal model, Andrea Willson thesis introduced a transtibial amputee OpenSim model, enhancing gait dynamics simulation for individuals with below-knee amputations by incorporating a modified lower leg with elements like a socket, pylon, and foot, and adjusting mass properties to reflect anatomical differences post-amputation while maintaining muscle-driven simulation fidelity (Willson et al., 2017).

The primary aim of this chapter is to simulate the gait of TTAs using the OpenSim modelling system and compare these simulations with the gait patterns of an AB control group. By leveraging the capabilities of Rajagopal's full-body model and using Andrea's Transtibial (below knee) amputation model (Willson et al., 2017), this chapter seeks to provide a detailed analysis of the kinematics and kinetics involved in TTA gait. The secondary aims to identify significant differences in kinematic and kinetic movement patterns during gait between TTAs and AB participants.

The ultimate goal of this chapter is to enhance the understanding of TTA gait as simulated by MSK models. By comparing the simulated gait patterns of amputees to the AB individuals, the chapter's goal is to find the biomechanical differences that could enhance our understanding of the TTAs gait dynamics.

Thus, the use of the OpenSim platform, specifically the Rajagopal full-body MSK model, represents a significant advancement in the study of human gait. This research aims not only to deepen our biomechanical understanding of amputee gait but also to effectively utilize laboratory-collected data. By integrating these advanced modelling techniques with precise experimental data, it is possible to achieve a more comprehensive analysis of gait dynamics in individuals with unilateral transtibial amputations.

6.2 OpenSim models

This study utilized advanced OpenSim models to analyse gait dynamics in both TTAs and AB individuals. For AB subjects, the comprehensive Rajagopal 2016 full-body musculoskeletal model, known for its high-fidelity representation of muscle-driven human movement (Rajagopal et al., 2016) was used. For transtibial amputee subjects, the transtibial (below-knee) amputee OpenSim model developed by Andrea Willson (Willson et al., 2017) was used. This specialized model modifies the Rajagopal framework to accurately reflect the anatomical and biomechanical characteristics of individuals with below-knee amputations.

6.2.1 Rajagopal 2016 Full-Body Musculoskeletal Model

The Rajagopal 2016 full-body musculoskeletal model represents a pivotal development in the field of biomechanical modelling, specifically designed to simulate muscle-driven human gait. This model features a highly detailed representation of the human MSK system, including 37 degrees of freedom to accurately capture joint kinematics across the entire body. This extensive ROM covers all major joints, such as the spine, hips, knees, ankles, shoulders, elbows, and wrists, allowing for comprehensive simulations of both upper and lower limb movements.

One of the standout features of the Rajagopal model is its inclusion of Hill-type models for 80 muscle-tendon units. These units simulate the force-generating capacities of muscles based on their physiological properties, including optimal fiber length, pennation angle, and maximum isometric force. The data for these parameters were meticulously derived from a combination of experimental measurements and MRI data from cadaver studies and healthy subjects. This robust dataset ensures that the model can produce realistic and accurate simulations of various activities, including walking and running.

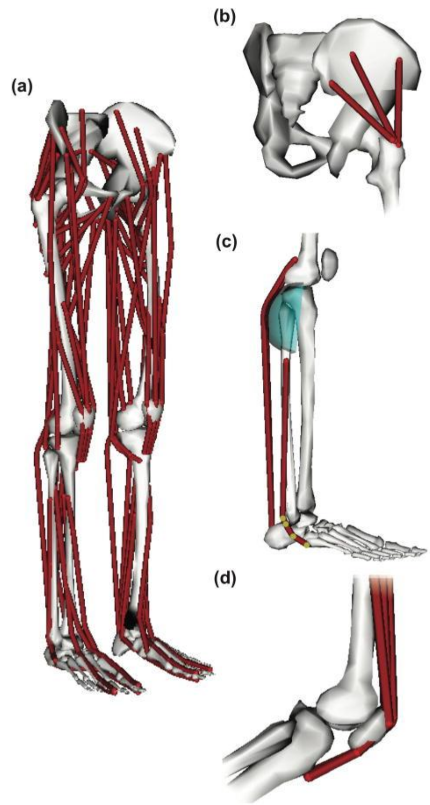

Figure 6.1 Muscles were represented as massless linear actuators. (a) The model has 80 muscle-tendon units (40 per leg) that control the lower limbs. (b) The muscle is modelled with large attachment regions, such as the gluteus medius, employing many separate muscle-tendon units. (c) Muscle geometry was represented as a collection of body-fixed points (highlighted) and wrapping surfaces. (d) The force transmission mechanism between the quadriceps and the patellar ligament was implicitly simulated by wrapping the quadriceps muscles around the patella and inserting them straight into the tibia.(Rajagopal et al., 2016).

The Rajagopal model is particularly valuable for its application in gait simulation. Its detailed representation of lower limb musculature and joint dynamics provides researchers with the tools needed to study the intricacies of human walking and running with high fidelity. The model's accuracy and computational efficiency make it a valuable resource for both clinical and research settings. It has been extensively validated against experimental data, demonstrating its ability to replicate the biomechanics of human movement reliably. This validation ensures that simulations based on the Rajagopal model are both accurate and relevant for real-world applications, making it an indispensable tool for studying pathological gait, designing rehabilitation protocols, and developing new prosthetic and orthotic devices (Rajagopal et al., 2016; Figure 6.1).

6.2.2 Transtibial (Below-Knee) Amputee OpenSim Model

The transtibial (below-knee) amputee OpenSim model developed by Andrea Willson marks a significant advancement in the simulation of gait dynamics for individuals with lower limb amputations. This model, detailed in Willson's 2017 thesis, builds upon the robust framework of the Rajagopal model to specifically address the unique needs of transtibial amputees. By modifying the lower leg to reflect the anatomical and biomechanical characteristics of an amputee's limb, this model provides a highly accurate tool for studying amputee gait.

Key modifications in the transtibial amputee model include the integration of SolidWorks CAD models for a generic socket, pylon, and foot. These components are meticulously designed to replicate the structure and function of a transtibial prosthesis. Adjustments to the mass properties and moments of inertia are made to reflect the differences between intact and amputated limbs. For instance, the residual limb mass is set to approximately 50% of the original tibia segment mass, and the centre of mass is adjusted to be 30% closer to the knee joint. These adjustments ensure that the model accurately simulates the biomechanical behaviour of an amputee's limb.

The model retains 72 Hill-type muscles to simulate the force-generating capacity of both the residual limb and the intact musculature. These muscle properties are based on data from cadaver and MRI studies, ensuring that the model's muscle dynamics are realistic. Additionally, the socket and pylon are modelled as separate bodies, allowing for greater flexibility and generalization in representing different prosthetic designs. The ankle joint includes three degrees of freedom: plantar flexion-dorsiflexion, internal-external rotation, and inversion-eversion, providing a comprehensive range of motion for gait analysis (Willson et al., 2017).

One of the primary applications of the transtibial amputee model is in gait analysis for individuals with below-knee amputations. It provides critical insights into the biomechanical impacts of amputation and the effectiveness of various prosthetic designs. This model is particularly useful for optimizing prosthetic components and rehabilitation strategies, with the ultimate goal of improving mobility and quality of life for amputees. However, it is important to note that while the model is highly detailed, it does not account for the effectiveness of below-knee muscles in creating joint torque post-amputation, as these muscles are typically rendered ineffective.

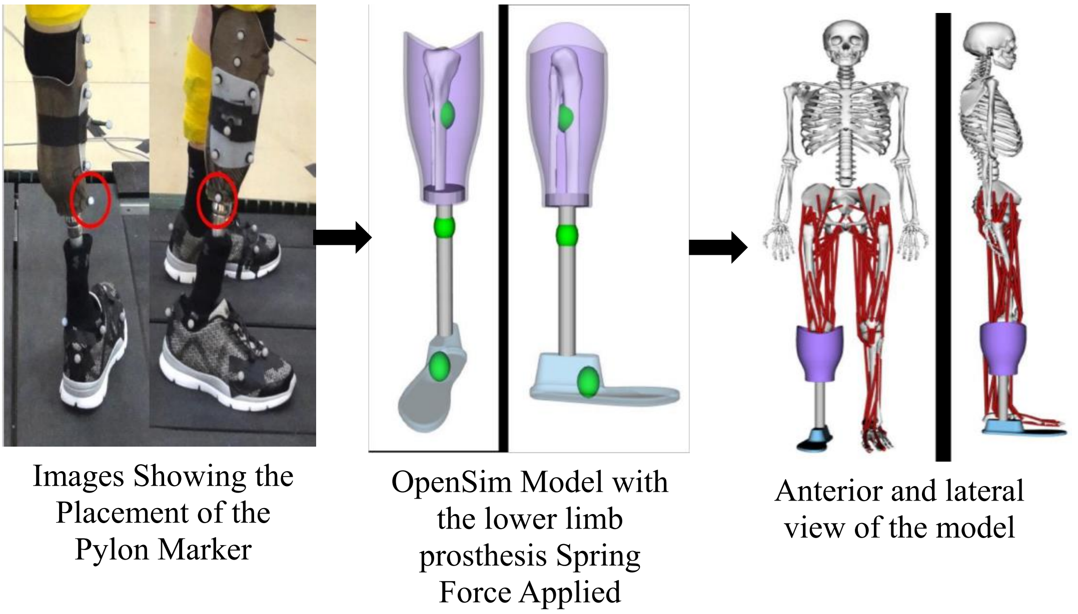

Figure 6.2 Sequence of developing Andrea of transtibial amputation model: (left) Initial prototype with highlighted components for data collection and gait analysis; (middle) schematic of the prosthetic design including the socket, pylon, and foot; (right) musculoskeletal model used for simulating and analysing gait dynamics in transtibial amputees (Willson et al., 2017).

By utilizing these two models, it is possible to comprehensively analyse and compare gait patterns between TTA and AB individuals. This dual-model approach facilitates a deeper understanding of the biomechanical impacts of transtibial amputation, allowing us to gather nuanced insights into the differences between TTA and AB gait. Ultimately, this approach aims to enhance our understanding of TTA gait in a controlled laboratory setup, providing a solid foundation for improving gait performance and mobility for individuals with transtibial amputations (Figure 6.2).

6.3 Experimental data processing

In our study, the experimental data preparation is a critical phase to ensure the accuracy and reliability of the muscle force estimations using the OpenSim pipeline. The process involves several key steps: participant preparation, motion capture, data processing, and integration into the OpenSim framework.

6.3.1 Participant

Participants were recruited based on specific inclusion criteria to ensure homogeneity in the study sample. Prior to data collection, each participant underwent a detailed briefing about the study's procedures, and written informed consent was obtained. Details can be found in Chapter 4.

6.3.2 Motion capture

The motion capture setup involved an instrumented walkway embedded with two force plates to capture GRFs and an eight-camera motion capture system to record the three-dimensional trajectories of the reflective markers. The walkway was calibrated before each session to ensure the accuracy of the force data. Participants were instructed to walk at self-selected speeds along the walkway, and several trials were recorded to capture a comprehensive dataset. Details can be found in Chapter 4.

6.3.3 Data processing

The raw motion capture data processed using Vicon Nexus software. This involved filtering the marker trajectories and GRF data to remove noise and ensure smooth signals. The marker trajectories were smoothed using a low-pass Butterworth filter. GRF data were synchronized with the motion capture data, and any gaps in the marker trajectories were filled using interpolation techniques provided by the Vicon Nexus software (Nexus, 2008).

6.3.4 Integration into OpenSim

We exported the gait data from Vicon Nexus in OpenSim-compatible formats, specifically as motion reflector marker data in the (.trc) format and force plate data in the (.mot) format. These formats are essential for running a model in OpenSim. Then imported the marker trajectory and GRF data into OpenSim. Scaling the Raj 2016 MSK model for each AB participant to match their anthropometry and the Andrea 2017 MSK model for each TTA participant. This scaling process involved adjusting

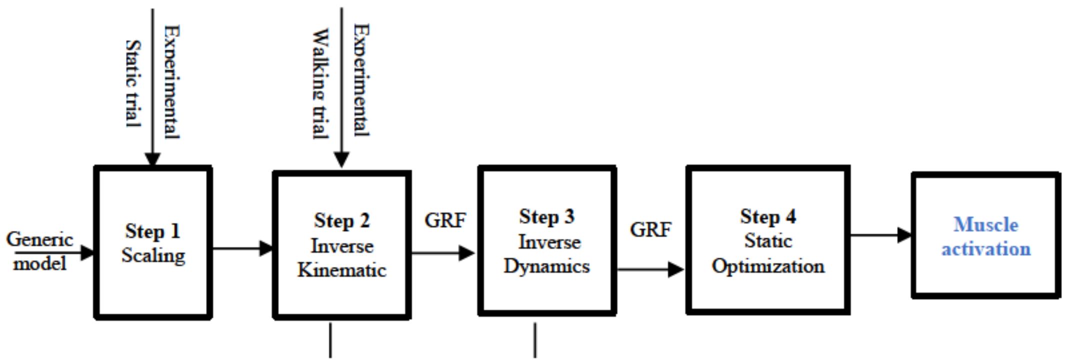

Figure 6.3 OpenSim Gait Analysis Pipeline. The process includes four primary steps: (1) Scaling of the musculoskeletal model to match participant dimensions, (2) Inverse Kinematics to calculate joint angles from marker trajectories, (3) Inverse Dynamics to compute joint torques, and (4) Static Optimization to estimate muscle activations. These steps collectively enable detailed biomechanical analysis of gait performance.

the model's segment lengths and mass properties to align with the participant's dimensions based on static pose data. After scaling, the used of the inverse kinematics (IK) tool in OpenSim to calculate joint angles from the marker trajectory data. The inverse dynamics (ID) tool then used these joint angles, along with the GRF data, to compute the net joint moments. The static optimisation (SO) tool used these moments and the angles of the joints as inputs to figure out the muscle forces and activations (Figure 6.3).

6.3.5 Quality Assurance

During the data preparation process, visual inspections of the marker trajectories were made to verify the accurate tracking of all markers during the motion capture sessions. Furthermore, thorough assessments were conducted of the force plate data by doing meticulous alignment checks in addition to marker trajectory inspections. Verifying the accuracy and compatibility of the recorded motions with the GRF readings was a critical step. In order to conduct the next analysis, it was critical to accurately synchronise the force plate data with the marker data. Additionally, validation of the muscle activation model with EMG data was performed to assess the strength of the muscle activation prediction.

As a result, the primary goal was to improve the quality of data in order to provide meaningful and reliable insights in both motion analysis and MSK modelling.

6.4 OpenSim Pipeline

In biomechanical gait analysis, the accurate estimation of muscle forces and joint dynamics is important for understanding human movement and then identifying gait abnormalities. OpenSim offers robust tools for modelling, simulating, and analysing the MSK system. This pipeline outlines the comprehensive steps involved in using OpenSim to transform raw motion capture data into meaningful biomechanical insights.

The process begins by scaling a generic MSK model to match the physical dimensions and mass properties of the individual subject. This ensures that the virtual model accurately represents the subject's anatomy, providing a solid foundation for subsequent analyses. The next step, IK, involves computing joint angles that best align the model with the experimental marker data. This alignment is crucial for accurately simulating the subject's movements.

After IK, ID calculates the generalised forces and moments at the joints that produce the observed movements. This step is essential for understanding internal joint loads and the contributions of different muscle groups. Finally, SO resolves these joint moments into individual muscle forces, distributing the required forces across the muscles.

Each step of this pipeline is designed to ensure accuracy and reliability, from initial data collection to the final muscle force estimations. By integrating these advanced computational techniques, the OpenSim pipeline provides a powerful framework for gait biomechanical analysis, enhancing the ability to study human movement and improve clinical outcomes for individuals with movement disorders. The following sections will discuss each step-in detail.

6.4.1 Scaling the Model

In OpenSim, scaling the MSK model entails adjusting the model's geometry and mass properties to match the subject's physical dimensions and mass. This process begins with attaching reflective markers to the subject's anatomical landmarks and capturing their positions using a motion capture system during a static trial. Then calculated the distances between pairs of experimental extracted markers to compared them to the corresponding distances on the virtual model's markers to determine the scale factors and checking the scaling error by dividing the distance of the experimental marker pair (e) by the distance of the virtual, unscaled marker pair (m) (Equation 1).

$$s1 = e1/m1 \qquad \text{Equation 1}$$

Further, If the segment dimension scaled by more than one marker pair, the marker pairs average calculated and while summing the scaling factors and dividing by the number of scaling pairs (Equation 2):

$$s = (s1 + s2 + \cdots + sn)/n \qquad \text{Equation 2}$$

Next, apply these scale factors to adjust the dimensions of the body segments, ensuring accurate alignment of the virtual markers with the experimental markers (Table 6.1). Additionally, we scale the locations of joints and other critical anatomical landmarks. To ensure a consistent mass distribution, we update the model's mass, and inertial properties based on the subject's mass. We adjust muscle and ligament properties, such as optimal fibre length and tendon slack length, accordingly, to maintain the model's integrity. Lastly, we check the scaled model by showing a preview of the static pose, checking for marker errors, and making small changes to the scale factors and marker positions repeatedly until we get the best alignment (Figure 6.4).

Table 6.1 The OpenSim scaling segment and makers used to scale

Scaling segment	markers	
femur x	RFEL-RFME	LFEL-LFME
femur y	RPSIS-RFME	LPSIS-LFME
femur z	RPSIS-RASIS	LPSIS-LASIS
shank x	RTAM-RFAM	LTAM-LFAM
shank y	RFME-RTAM	LFME-LTAM
foot x	RTOE1-RTOE5	LTOE1-LTOE5
foot y	RTAM-RANK	LTAM-LANK
foot z	RANK-RTOE	LANK-LTOE

Note: L=left, R=right, PSIS=posterior superior iliac spine, ASIS=anterior superior iliac spine, FEL=femur lateral epicondyle, FME=femur medial epicondyle, TAM=medial lower tibial tuberosity, FAM=lateral lower fibula touristy, ANK=heel, TOE1=big toe, TOE=third toe, TOE5=fifth toe.

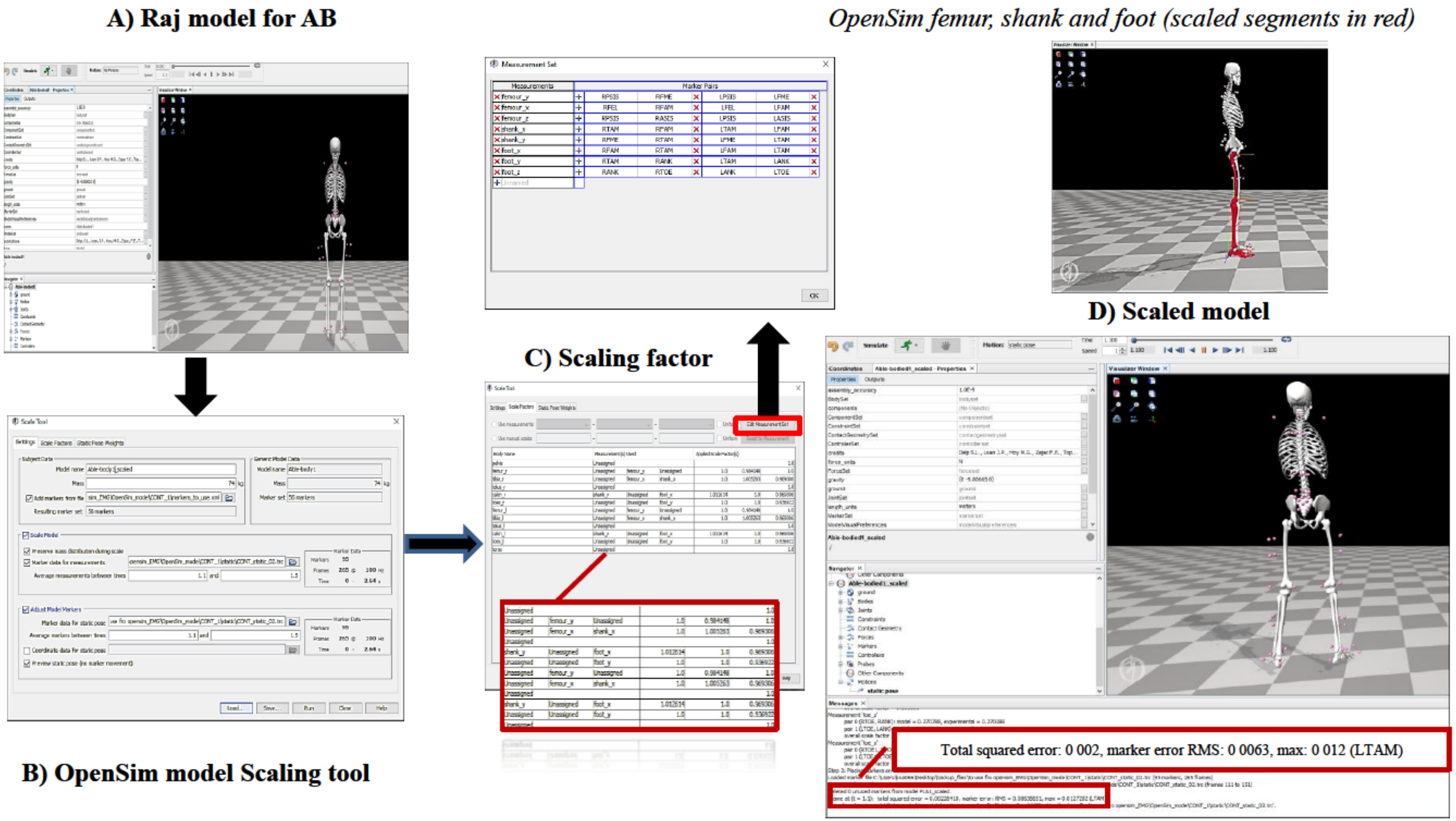

Figure 6.4 The figure illustrates the process of scaling an OpenSim MSK model, including initializing the generic model (A), placing markers (B), configuring the measurement set (C), and adjusting the Scale Tool settings (D) to match the participant's anthropometric data.

6.4.2 Inverse Kinematics (IK)

Inverse Kinematics (IK) in OpenSim is employed to compute joint angles that best match the experimental marker trajectory data collected from the motion capture system. Initially, the experimental data is prepared by ensuring markers are placed on anatomical landmarks with minimal skin movement. The generic MSK model is scaled to match the subject's anthropometry, ensuring accurate model dimensions. The IK Tool in OpenSim is then configured with the scaled model and the experimental marker trajectory file. Marker weights are adjusted to prioritize the accuracy of specific markers, and the IK analysis is run to compute the joint angles for each frame of the motion capture data. The IK calculate the differences between the measured marker locations and the model's virtual marker locations, subject to joint constraints (Klous, 2010) (Equation 3).

$$Square\ Error = \sum_{i=1}^{markers} w\left(xi^{subject} - \ xi^{model}\right)^2 \qquad \text{Equation 3}$$

The results, including joint angles and translations, are evaluated for accuracy by comparing the positions of the virtual markers with the experimental markers. Any discrepancies are addressed by iteratively refining the marker weights and placements (Figure 6.5).

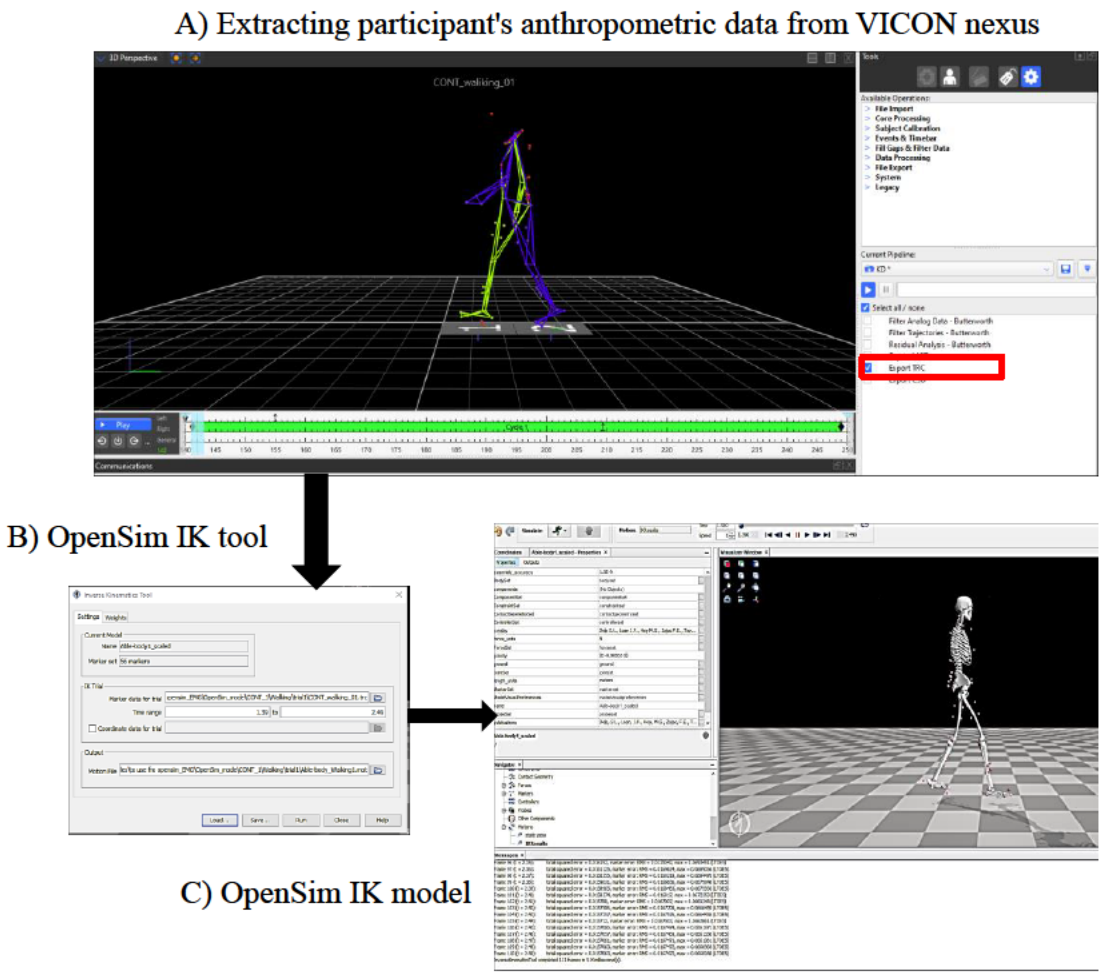

Figure 6.5 Steps for OpenSim IK joints angle simulation: A) extracting participant's anthropometric, B) OpenSim IK tool and C) the walking simulation.

6.4.3 Inverse Dynamics

Inverse Dynamics (ID) in OpenSim is used to compute the generalized forces and moments (net joint torques) responsible for producing the observed movement. This step utilizes the joint angles obtained from the IK analysis and external forces measured using force plates. The ID Tool in OpenSim is configured with the scaled model, the motion file from the IK analysis, and the external load file containing the ground reaction forces. The analysis is run to solve the equations of motion, yielding the net joint torques and forces. The fundamental principles of inverse dynamics are rooted in Newton's second law of motion which states that the sum of all forces acting on a body is equal to its mass times its acceleration.

These ID results are reviewed to ensure consistency with the measured external loads and expected physical behaviour. Visualization tools in OpenSim are used to animate the model with the computed forces and moments, providing a clear understanding of the movement dynamics (Figure 6.6).

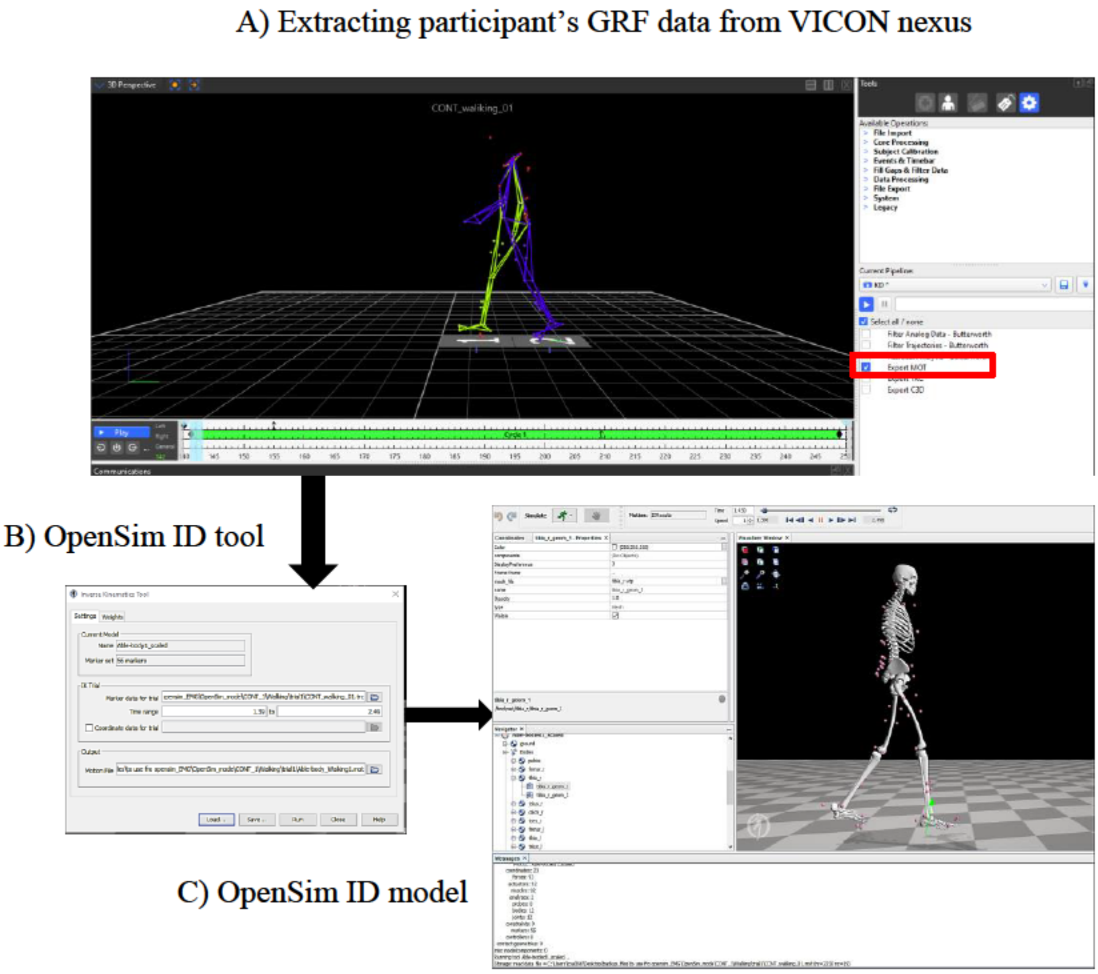

Figure 6.6 Steps for OpenSim ID joints torque estimation: A) extracting participant's GRF, B) OpenSim ID tool and C) joints torque estimation.

6.4.4 Static optimization

In OpenSim, muscle activations are computed through a series of steps involving inverse kinematics, inverse dynamics, and static optimization. IK initiates the process by using motion capture data to fit a musculoskeletal model to the observed movement data, thereby calculating joint angles and positions. Next, ID uses these joint angles, along with external forces (e.g., GRF), to compute the net joint moments required to produce the observed motions. SO, which resolves these net joint moments into individual muscle forces, is the critical step. This is done by minimising an objective function, typically formulated as the sum of squared muscle activations, ensuring a realistic and physiologically plausible distribution of forces among the muscles. The optimisation process uses the following objective function (Equation 4):

$$\text{Objective function} = \sum_{m=1}^{n} (a_m)^p \qquad \text{Equation 4}$$

where n is the number of muscles, a_m represents the activation level of muscle m, and p is a user-defined constant, often set to 2.

The output from SO includes time histories of muscle activations and forces, which are critical for understanding the dynamics of muscle function during movement. By ensuring that the computed muscle activations are both efficient and physiologically realistic, SO provides a comprehensive view of how muscles contribute to motion, particularly under different loading and kinematic conditions. In biomechanics research, this optimisation method plays a crucial role in analysing how individuals, particularly those with musculoskeletal impairments, compensate during movement and recruit different muscles to maintain stability and function (Figure 6.7).

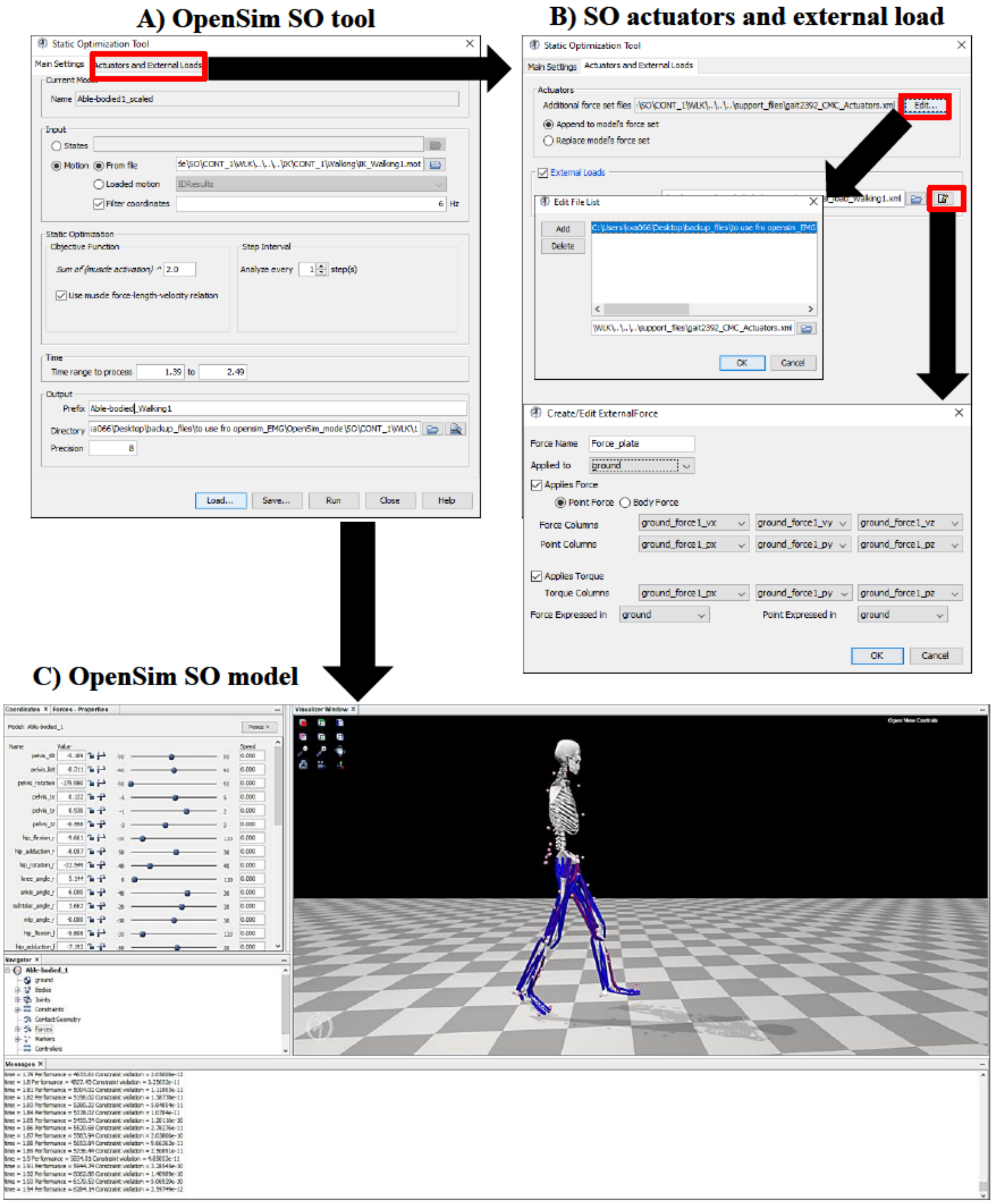

Figure 6.7 Steps for OpenSim SO muscle estimation: A) setup the static optimisation SO tool, B) inserting SO actuators and GRF C) muscle estimation.

6.5 Statistical Analysis

The normality of the collected data was tested before performing ANOVAs. All data are presented as the mean and standard deviation (mean ± SD). The joint angle, joint torque and muscle force were analysed using a one-way ANOVA, with Tukey's post hoc test employed to compare sub-groups, with significance levels set at $p<0.05$.

The data were analysed using SPSS (IBM Corp. (2020), IBM SPSS Statistics for Windows, Version 27.0. Armonk, NY: IBM Corp).

6.6 Results

The data will be presented sequentially, beginning with the IK analysis, which details the computed joint angles. Subsequently, the joint torque derived from the ID analysis will be discussed. Finally, the muscle force obtained from SO.

6.6.1 Inverse kinematic

The comparison of joint angles between the amputee (TTA) group and the control (AB) group reveals the following results:

Pelvis Tilt: The TTA group exhibited lower pelvis tilt on the amputated side (6.99 ± 2.60°) compared to the AB group (15.86 ± 7.41°; $p<0.01$). This lower tilt of the pelvis in the TTA group suggests a way for them to keep their balance and stability while they walk, which could be because the prosthesis changed their biomechanics.

Pelvis List: There was no significant difference in pelvis list between the TTA group's intact side (5.82 ± 2.91°) and the AB group (4.24 ± 0.81°; $p = 0.43$), indicating that the intact side of the TTA group maintains a similar level of lateral pelvic tilt as the AB group.

Pelvis Rotation: The TTA group showed lower pelvis rotation on the amputated side (8.55 ± 3.15°) compared to the AB group (14.98 ± 4.98°; $p = 0.02$). This decrease in rotation may reflect an adaptation to reduce the rotational forces on the prosthetic limb while walking.

Hip Flexion: Both the intact (40.70 ± 7.58°) and amputated (40.14 ± 8.53°) sides of the TTA group demonstrated slightly lower hip flexion angles compared to the AB group (42.97 ±

3.85°), although these differences (p = 0.46). This suggests that hip flexion remains relatively consistent between the groups, despite the presence of a prosthesis.

Hip Adduction: The TTA group showed differences in hip adduction, with the intact side exhibiting greater adduction (13.98 ± 3.31°; p = 0.03) compared to the amputated side (12.35 ± 3.25°; p = 0.01) and the AB group (22.79 ± 7.31°). The reduced adduction in the amputated limb may serve as a compensatory adjustment to avoid excessive loading on the prosthesis.

Hip Rotation: Hip rotation was higher on the amputated side of the TTA group (14.69 ± 5.59°) compared to the AB group (11.44 ± 2.92°), though this difference was not statistically significant (p = 0.22). This suggests a slight increase in rotational movement as a compensatory strategy, possibly to assist with the stability and propulsion phases of gait.

Knee Flexion: The knee flexion angle was lower on the amputated side of the TTA group (48.48 ± 12.09°) compared to both the intact side (51.07 ± 7.63°; p=0.06) and the AB group (58.73 ± 5.14°; p=0.03). This reduction in knee flexion on the amputated side suggests limitations in the range of motion due to the prosthesis, which may impact the overall gait efficiency.

Ankle Dorsiflexion: Ankle dorsiflexion was reduced on the amputated side of the TTA group (15.65 ± 4.15°) compared to the AB group (27.97 ± 5.72°; p<0.01), indicating limited ankle mobility due to the prosthetic limb. The intact side of the TTA group showed comparable dorsiflexion to the AB group (27.69 ± 6.14°; p = 0.94), highlighting that the limitations are specific to the prosthetic side (Table 6.2).

The TTA group exhibits significant adaptations in joint angles across the pelvis, hip, knee, and ankle when compared to the AB group. These adaptations are particularly pronounced on the amputated side, reflecting the biomechanical challenges posed by the use of a prosthetic limb. These findings underscore the importance of targeted rehabilitation strategies to address these joint angle discrepancies and improve gait efficiency and stability in individuals with transtibial amputations.

Table6.2 Joints ROM for TTA (intact and amputated) limb and AB group.

Parameter	TTAs		AB
	Intact	**Amputated**	
	Pelvis		
Pelvis tilt (degree)	6.99 ± 2.60* ($p \leq 0.01$)		15.86 ± 7.41
Pelvis list (degree)	5.82 ± 2.91 ($p=0.43$)		4.24 ± 0.81
Pelvis rotation (degree)	8.55 ± 3.15 * ($p=0.02$)		14.98 ± 4.98
	Hip		
Hip flexion (degree)	40.70 ± 7.58 ($p=0.51$)	40.14 ± 8.53 ($p=0.46$)	42.97 ± 3.85
Hip adduction (degree)	13.98 ± 3.31* ($p=0.03$)	12.35 ± 3.25* ($p=0.01$)	22.79 ± 7.31
Hip rotation (degree)	13.51 ± 5.21 ($p=0.40$)	14.69 ± 5.59 ($p=0.22$)	11.44 ± 2.92
	Knee and ankle		
Knee flexion (degree)	51.07 ± 7.63 ($p=0.06$)	48.48 ± 12.09* ($p=0.03$)	58.73 ± 5.14
Ankle dorsiflexion (degree)	27.69 ± 6.14 ($p=0.94$)	15.65 ± 4.15*† ($p \leq 0.01$; $p \leq 0.01$)	27.97 ± 5.72

Note:
* Results: statistical difference between TTA and AB $p<0.05$.
† Results: statistical difference between amputated and the intact limb/side $p<0.05$.

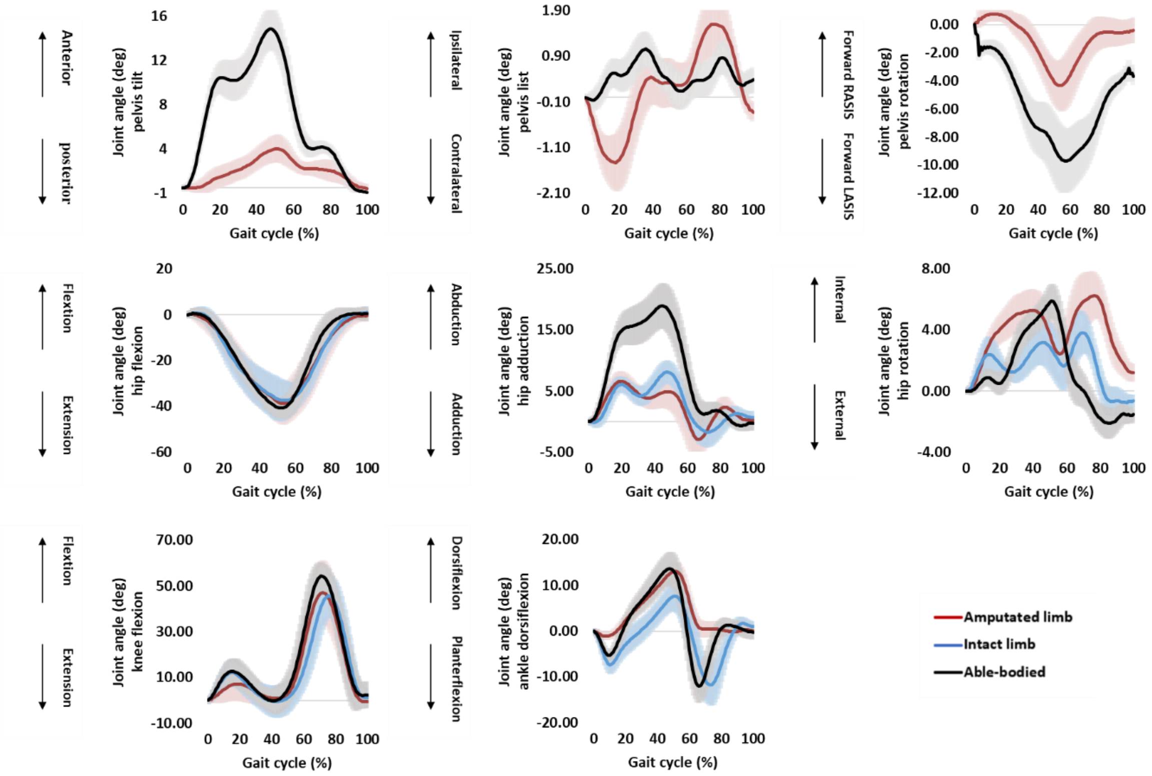

Figure 6.8 Hip, knee and ankle ROM pattern for the TTAs (intact, amputated limb) and AB

6.6.2 Inverse dynamics

The comparison of joint torque between the TTA and the AB group reveals the following results:

Hip Flexion: The torque generated during hip flexion was notably different between the TTA group's intact side and the amputated side. The intact limb of the TTA group exhibited a significantly higher hip flexion torque 2.92 ± 0.65 Nm/(kg*9.8) compared to the amputated side (2.15 ± 0.65 Nm/(kg*9.8), though both were higher than the AB group 2.09 ± 0.49 Nm/(kg*9.8). This suggests that in TTA individuals, the intact limb compensates for the reduced torque on the amputated side, likely due to the limitations imposed by the prosthesis. The reduced torque on the amputated side may indicate a need for targeted rehabilitation to enhance the muscular strength and functionality of the hip flexors on the prosthetic side.

Hip Adduction: The torque for hip adduction was generally lower in the TTA group compared to the AB group. The intact side of the TTA group showed a torque of 1.24 ± 0.42 Nm/(kg*9.8), while the amputated side exhibited a lower torque of 0.89 ± 0.40 Nm/(kg*9.8), compared to the AB group's 1.41 ± 0.48 Nm/(kg*9.8).

Hip Rotation: The torque associated with hip rotation did not show significant differences between the groups. The TTA group's intact side had a torque of 0.34 ± 0.16 Nm/(kg*9.8), and the amputated side had a slightly lower torque of 0.21 ± 0.12 Nm/(kg*9.8), compared to the AB group’s 0.27 ± 0.14 Nm/(kg*9.8). This implies that both groups maintain a relatively intact hip rotation torque, suggesting that amputation has less of an impact on rotational stability.

Knee Flexion: A significant reduction in knee flexion torque was observed on the amputated side of the TTA group 0.94 ± 0.52 Nm/(kg*9.8) compared to both the intact side 1.81 ± 0.64 Nm/(kg*9.8) and the AB group 1.64 ± 0.16 Nm/(kg*9.8). This marked decrease in torque on the amputated side suggests limited knee flexor strength, which may impact the ability to control the knee during the swing phase and absorb shock during the stance phase. The intact side also exhibited reduced torque compared to the AB group, indicating overall bilateral adaptations in response to the amputation.

Ankle Dorsiflexion: The torque generated during ankle dorsiflexion was significantly lower in the TTA group compared to the AB group, particularly on the amputated side 1.36 ± 1.03 Nm/(kg*9.8) and the intact side 1.55 ± 1.06 Nm/(kg*9.8), versus the AB group's 3.14 ± 0.33 Nm/(kg*9.8). This drop-in ankle dorsiflexion torque on both sides of the TTA group shows that ankle function is severely limited. This could make it harder to control foot positioning during the gait cycle and make walking less efficient overall. The lower torque on the amputated side is particularly concerning, as it underscores the challenges of achieving adequate foot clearance and push-off power with a prosthetic limb (Table 6.3).

The joint torque analysis reveals significant disparities between the TTA and AB groups, particularly in the hip, knee, and ankle joints. The amputated side of the TTA group consistently shows reduced torque across all measured joints, indicating the biomechanical limitations imposed by the prosthesis and the need for compensatory strategies from the intact limb. These findings highlight the importance of targeted rehabilitation to address these torque deficiencies, improve muscle strength, and enhance functional outcomes for individuals with transtibial amputations.

Table6.3 Joints torque for TTA (intact and amputated) limb and AB group

Joint torque Nm/(kg*9.8)	Intact	Amputated	AB
	Hip		
Hip flexion	2.92 ± 0.65* (p=0.03)	2.15 ± 0.65† (p=0.20; p=0.04)	2.09 ± 0.49
Hip adduction	1.24 ± 0.42 (p=0.54)	0.89 ± 0.40 (p=0.07; p=0.15)	1.41 ± 0.48
Hip rotation	0.34 ± 0.16 (p=0.43)	0.21 ± 0.12 (p=0.46; p=0.15)	0.27 ± 0.14
	Knee		
Knee flexion	1.81 ± 0.64 (p=0.53)	0.94 ± 0.52 *† (p=0.01; p≤0.01)	1.64 ± 0.16
	Ankle		
Ankle dorsiflexion	1.55 ± 1.06 * (p≤0.01)	1.36 ± 1.03 * (p≤0.01; p=0.5)	3.14 ± 0.33 (2.71-3.68)

Note:
* Results: statistical difference between TTA and AB p<0.05.
† Results: statistical difference between amputated and the intact limb p<0.05.

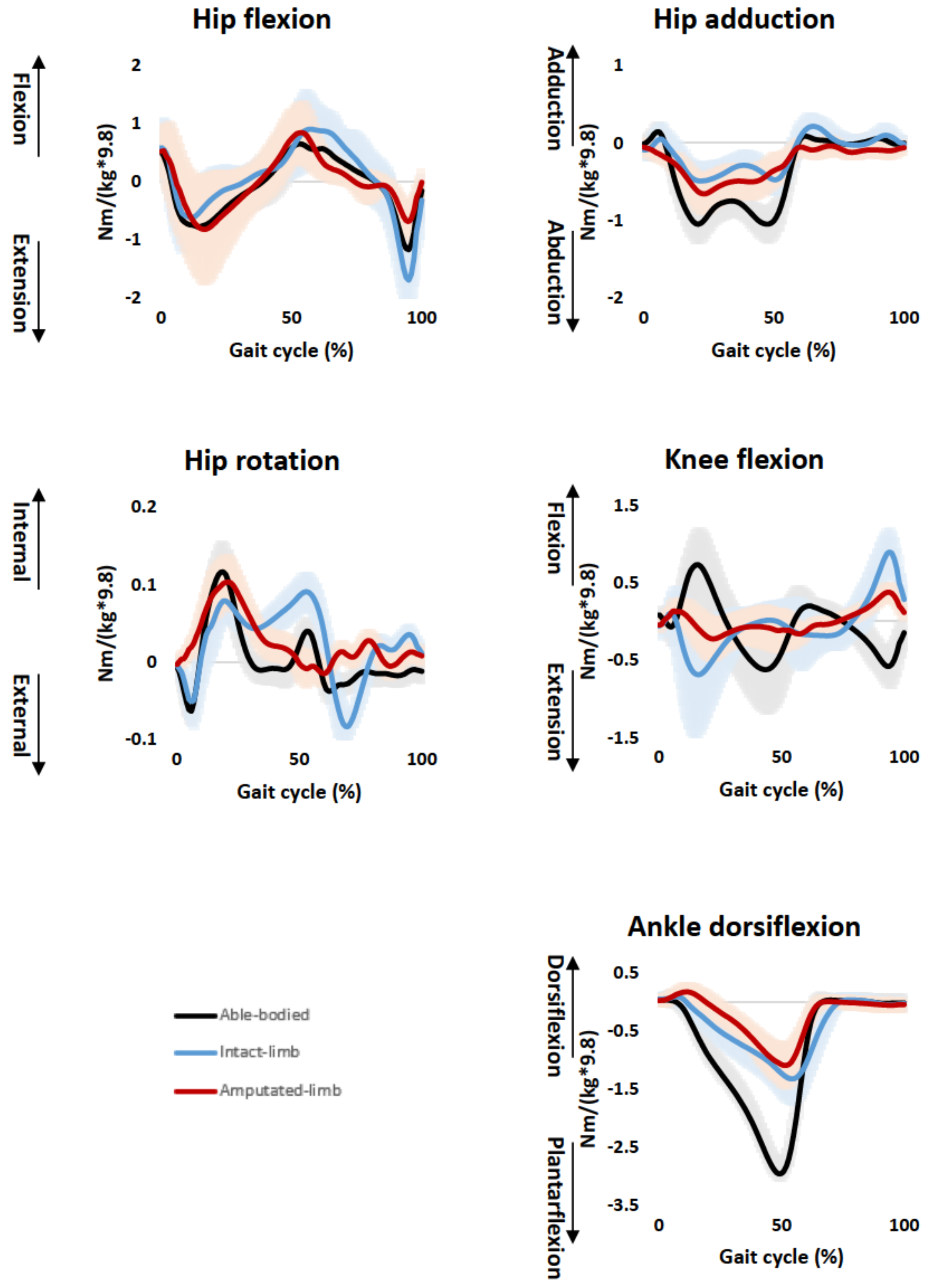

Figure 6.9 The joint torque for amputee intact side (blue), amputated side (red) and AB (black) during gait cycle where the x axis represents the gait cycle and y axis represent the joint torque in Nm/(kg*9.8) and the arrow showing the direction of the movement.

6.6.3 Analysis of Muscle Forces and Their Influence on Joint Torques During Gait

The analysis of muscle forces and joint torques during gait reveals several significant differences between AB and amputees both with the intact and the amputated limb.

For hip flexion, the iliopsoas muscle activation was higher in amputees on the amputated limb (0.70 ± 0.21 BW, $p < 0.01$) compared to both the intact limb (0.36 ± 0.08 BW, $p \leq 0.01$) and AB individuals (0.30 ± 0.10 BW). The hip flexion joint torque also showed significant differences, with the highest values observed in the intact limb of amputees (2.92 ± 0.65 BW, $p = 0.03$) compared to their prosthetic limb (2.15 ± 0.65 BW, $p = 0.20$) and AB individuals (2.09 ± 0.49 BW).

In terms of hip adduction, the adductor magnus muscle activation in amputees was low across both limbs but higher on the intact limb (0.02 ± 0.01 BW, $p = 0.04$) and the amputated limb (0.02 ± 0.00 BW, $p \leq 0.01$) compared to AB individuals (0.01 ± 0.00 BW). The joint torque for hip adduction was not significantly different between groups, although it was lower in amputees (0.89 ± 0.40 BW, $p = 0.07$) compared to AB individuals (1.41 ± 0.48 BW).

For hip rotation, the sartorius muscle activation was higher in amputees, particularly on the prosthetic limb (0.12 ± 0.03 BW, $p = 0.01$) compared to AB individuals (0.06 ± 0.04 BW). Joint torque for hip rotation did not show significant differences, although it was lower in amputees (0.21 ± 0.12 BW, $p = 0.15$) compared to AB individuals (0.27 ± 0.14 BW).

Knee flexion analysis showed significant increases in biceps femoris activation for amputees on the amputated limb (0.35 ± 0.22 BW, $p = 0.03$) compared to both their intact limb (0.11 ± 0.01 BW, $p = 0.03$) and AB individuals (0.12 ± 0.04 BW). Knee flexion torque was lower in amputees (0.94 ± 0.52 BW, $p < 0.01$) compared to AB individuals (1.64 ± 0.16 BW).

Finally, in the ankle dorsiflexion analysis, the Gastrocnemius muscle activation was not significantly lower in amputees (0.95 ± 0.19 BW, $p = 0.08$) compared to AB individuals (1.35

± 0.64 BW). Ankle dorsiflexion torque was significantly lower in amputees (1.36 ± 1.03 BW, $p \leq 0.01$) compared to AB individuals (3.14 ± 0.33 BW; Table 6.4 and Figure 6.10)

Overall, these results highlight the significant compensatory mechanisms and increased demands on specific muscle groups for amputees, particularly on the amputated limb.

Table 6.4 Muscle force of AB and Amputees

Muscle force (BW)	**Amputee**		**AB**
	Intact	Amputated	
Iliopsoas	0.36 ± 0.08* (p≤0.01)	0.70 ± 0.21*† (p≤0.01; p≤0.00)	0.30 ± 0.10
Gluteus maximus	0.08 ± 0.03 (p= 0.40)	0.07 ± 0.04 (p= 0.40; p=0.17)	0.08 ± 0.06
Adductor magnus	0.02 ± 0.01 (p= 0.04)	0.02 ± 0.00 (p= 0.04; p≤0.01)	0.01 ± 0.00
Gluteus medius	0.34 ± 0.17 (p= 0.42)	0.41 ± 0.16 (p= 0.42; p =0.32)	0.39 ± 0.21
Sartorius	0.09 ± 0.04* (p≤ 0.01)	0.12 ± 0.03* (p≤ 0.01; p =0.01)	0.06 ± 0.04
Biceps femoris	0.11 ± 0.01 (p= 0.30)	0.35 ± 0.22*† (p= 0.03; p =0.03)	0.12 ± 0.04
Vastus medialis	0.12 ± 0.04 (p= 0.08)	0.07 ± 0.01† (p= 0.08; p ≤0.01)	0.09 ± 0.05
Vastus lateralis	0.35 ± 0.16 (p= 0.14)	0.14 ± 0.04† (p= 0.14; p ≤0.01)	0.19 ± 0.10
Gastrocnemius	0.95 ± 0.16		1.35 ± 0.64

	(p= 0.08)	
Tibialis anterior	0.26 ± 0.06 (p= 0.45)	0.25 ± 0.13

Note:
* Results: statistical difference between TTA and AB $p<0.05$.
† Results: statistical difference between amputated and the intact limb $p<0.05$.

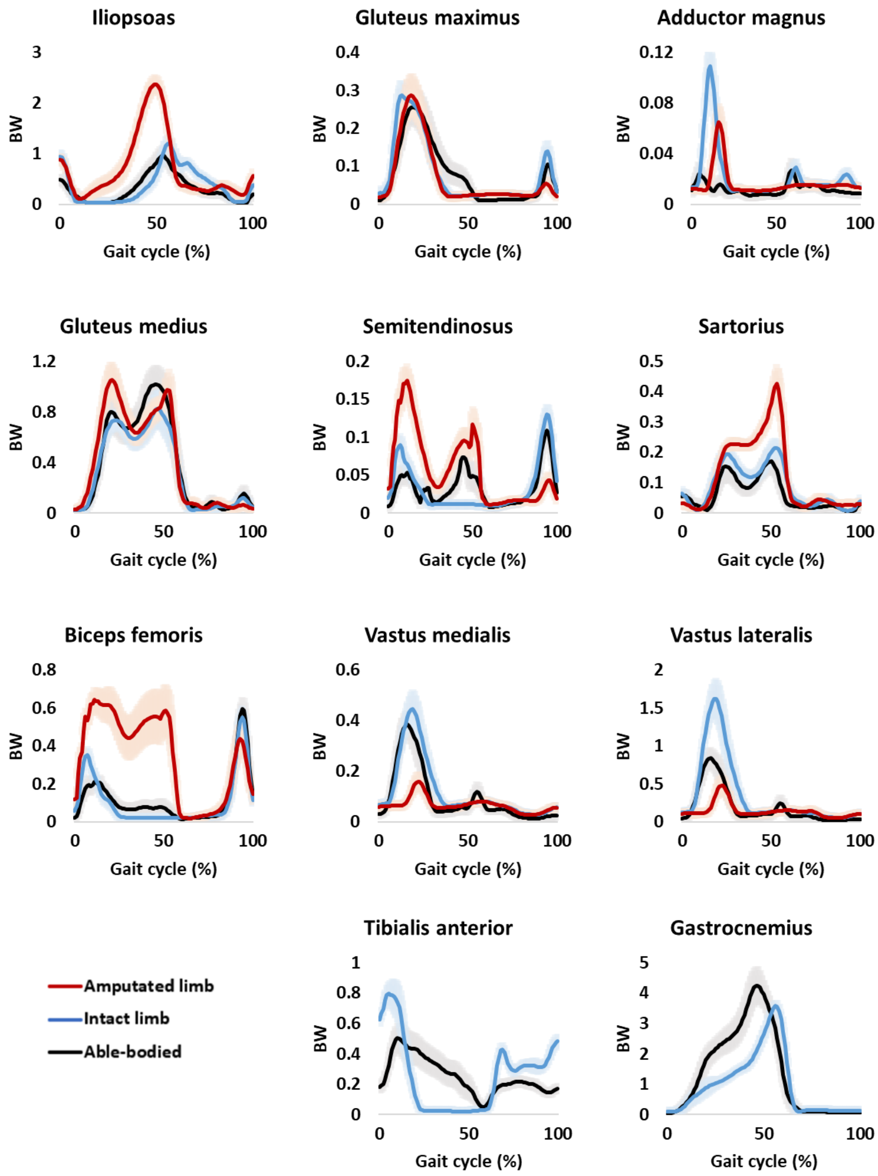

Figure 6.10 The muscle force for amputee intact side (blue), amputated side (red) and AB (black) during gait cycle where the x axis represents the gait cycle and y axis represent the joint torque in BW and the arrow showing the direction of the movement.

6.7 Discussion

The successful execution of the scaling pipeline for both TTA and AB participants indicates a robust method for adjusting model dimensions to match physical anthropometry. The total square error for the AB model and TTA model was 0.01 ± 0.01 and 0.01 ± 0.00, respectively, maintaining a consistent overall error of 0.01 ± 0.01. This low total square error suggests a high accuracy level in the scaling process, as it ensures that virtual markers closely align with experimental markers. Similarly, the RMS error was low for both models (0.01 ± 0.01 for AB and 0.01 ± 0.00 for TTA), reinforcing the precision of the model scaling. The maximum error, slightly higher for the AB model (0.04 ± 0.02) compared to the TTA model (0.03 ± 0.01), remained minimal, indicating that even the largest deviations between the model and experimental data were small.

These outcomes are consistent with findings in the literature where accurate scaling is crucial for realistic musculoskeletal modelling. Studies have shown that maintaining low RMS and total square errors is critical for the fidelity of simulations and subsequent biomechanical analyses (Harandi et al., 2020; Raabe & Chaudhari, 2016).

6.7.1 Range of Motion and Joint Dynamics

The ROM analysis between the amputees (TTA) and the control group (AB) reveals distinct biomechanical adaptations in individuals with amputations, particularly on the amputated side.

Pelvis Tilt: The TTA group demonstrated a significantly lower pelvis tilt on the amputated side (6.99 ± 2.60°) compared to the AB group (15.86 ± 7.41°, $p<0.01$). This reduced tilt suggests that TTA individuals may adopt a strategy to maintain balance and stability while walking, possibly as a compensatory adjustment due to the altered biomechanics introduced by the prosthesis. This finding aligns with previous research that indicates amputees often exhibit altered pelvis mechanics as part of their compensatory strategies to manage the demands of ambulation with a prosthesis (Wasser et al., 2017).

Pelvis List: No significant differences were found in the pelvis list between the intact side of the TTA group (5.82 ± 2.91°) and the AB group (4.24 ± 0.81°, $p = 0.43$). This suggests that lateral pelvic tilt remains relatively consistent between the groups, particularly on the intact side of the TTA individuals.

Pelvis Rotation: A significant reduction in pelvis rotation was observed on the amputated side of the TTA group (8.55 ± 3.15°) compared to the AB group (14.98 ± 4.98°, $p = 0.02$). The decrease in pelvis rotation on the amputated side may be an adaptive mechanism to minimise the rotational forces on the prosthetic limb during gait, which is critical for maintaining gait stability and preventing excessive strain on the prosthetic interface.

Hip Flexion: Both the intact and amputated sides of the TTA group exhibited slightly lower hip flexion angles compared to the AB group, although these differences were not statistically significant. The consistency in hip flexion across groups suggests that the presence of a prosthesis does not significantly alter hip flexion, implying a degree of hip movement preservation in TTA individuals.

Hip Adduction: Significant differences were observed in hip adduction, with the TTA group's intact side showing greater adduction compared to the amputated side and the AB group. The reduced adduction on the amputated side likely reflects a compensatory adjustment to avoid excessive loading on the prosthetic limb, which could destabilise gait or increase the risk of discomfort.

Hip Rotation: Hip rotation was slightly higher on the amputated side of the TTA group, though the difference was not statistically significant. The slight increase in rotational movement may be a compensatory strategy to assist with the stability and propulsion phases of gait, which is essential for maintaining forward momentum and reducing the risk of falls.

Knee Flexion: The TTA group's amputated side exhibited significantly lower knee flexion compared to both the intact side and the AB group. This reduction in knee flexion suggests limitations in the range of motion due to the prosthesis, which could impact overall gait efficiency by restricting the ability to absorb shock during the stance phase and generate sufficient power during the swing phase.

Ankle Dorsiflexion: Ankle dorsiflexion was significantly reduced on the amputated side of the TTA group compared to the AB group, indicating limited ankle mobility due to the prosthetic limb. This restriction in dorsiflexion is concerning, as it affects the ability to achieve proper foot clearance during the swing phase and may contribute to a higher risk of tripping or inefficient gait patterns.

The TTA group demonstrates significant adaptations in joint angles across the pelvis, hip, knee, and ankle when compared to the AB group. These adaptations are particularly pronounced on the amputated side, reflecting the biomechanical challenges posed using a prosthetic limb. The findings underscore the importance of using OpenSim IK to measure amputees ROM. This is consistent with the literature, which emphasises the need for customised rehabilitation programs that consider the unique biomechanical demands of prosthetic gait (Modenese et al., 2011).

6.7.2 Joint Torque Analysis

The joint torque analysis between the amputees (TTA) and the control group (AB) reveals distinct biomechanical adaptations in individuals with amputations, particularly on the amputated side.

Hip Flexion: The torque generated during hip flexion showed significant differences within the TTA group and when compared to AB controls. The intact limb of the TTA group exhibited a higher hip flexion torque 2.92 ± 0.65 Nm/(kg*9.8) compared to the amputated limb 2.15 ± 0.65 Nm/(kg*9.8), with slightly higher than the AB group 2.09 ± 0.49 Nm/(kg*9.8). This disparity suggests that the intact limb compensates for the reduced torque on the amputated side, likely due to the mechanical limitations imposed by the prosthesis. The reduced torque on the amputated side indicates a need for targeted rehabilitation to enhance the muscular strength and functionality of the hip flexors, which are crucial for maintaining effective gait dynamics (Rajagopal et al., 2016).

Hip Adduction: The torque for hip adduction was generally lower in the TTA group compared to the AB group, with the intact limb showing a torque of 1.24 ± 0.42 Nm/(kg*9.8) and the amputated limb showing 0.89 ± 0.40 Nm/(kg*9.8), compared to the AB group's 1.41 ± 0.48 Nm/(kg*9.8). While these differences were not statistically significant, they highlight the challenges in maintaining adduction strength, which is essential for lateral stability during gait. This decrease in adduction torque on the amputated side may contribute to difficulties stabilizing the pelvis during the stance phase, necessitating compensatory strategies that may not fully restore normal gait patterns.

Hip Rotation: Hip rotation torque did not show significant differences between the TTA and AB groups. The intact limb of the TTA group demonstrated a torque of 0.34 ± 0.16

Nm/(kg*9.8), while the amputated limb showed 0.21 ± 0.12 Nm/(kg*9.8), like the AB group's 0.27 ± 0.14 Nm/(kg*9.8). This finding implies that the prosthesis and residual limb musculature in TTA individuals can maintain rotational control during gait, thereby preserving hip rotational stability.

Knee Flexion: A significant reduction in knee flexion torque was observed on the amputated limb of the TTA group 0.94 ± 0.52 Nm/(kg*9.8) compared to both the intact limb 1.81 ± 0.64 Nm/(kg*9.8) and the AB group 1.64 ± 0.16 Nm/(kg*9.8). This marked decrease in knee flexion torque on the amputated side suggests limited knee flexor strength, which is critical for controlling knee motion during the swing phase and absorbing impact during the stance phase. The lower torque on the intact side compared to AB controls also shows that both sides have changed because of the amputation. This could be because of compensatory overuse or changes in neuromuscular control (Rajagopal et al., 2016)

Ankle Dorsiflexion: The torque generated during ankle dorsiflexion was significantly lower in the TTA group compared to the AB group, particularly on the amputated side 1.36 ± 1.03 Nm/(kg*9.8) and the intact side 1.55 ± 1.06 Nm/(kg9.8), versus the AB group's 3.14 ± 0.33 Nm/(kg*9.8). The fact that the TTA group had a big drop in dorsiflexion torque on both sides shows that their ankle function is severely limited, which is important for keeping their feet in the right place and pushing off properly during the gait cycle. The particularly low torque on the amputated side highlights the difficulties in achieving adequate foot clearance and propulsion, which are essential for efficient and safe ambulation (Rajagopal et al., 2016).

The joint torque analysis underscores the significant biomechanical challenges faced by individuals with transtibial amputations, particularly on the amputated side, where reduced torque in the hip, knee, and ankle joints reflects the limitations imposed by the prosthesis. These findings emphasize the importance of using OpenSim ID to measure joint torque for amputees.

6.7.3 Analysis of Muscle Forces and Their Influence on Joint Torques During Gait

The analysis of muscle forces and their impact on joint torques during gait for TTAs' and AB individuals reveals significant biomechanical distinctions, particularly in the activation of muscle groups to compensate for limb loss. These findings shed light on amputees' adaptive strategies, particularly in the context of their prosthetic limb, and highlight the increased demands on specific muscles to maintain functional gait.

Hip Flexion: The study found that the iliopsoas muscle activation was significantly higher on the amputated limb of amputees (0.70 ± 0.21 BW) compared to both the intact limb (0.36 ± 0.08 BW) and AB individuals (0.30 ± 0.10 BW). This elevated activation suggests that the iliopsoas compensates for the lack of natural limb mechanics, working harder to initiate hip flexion during gait. The hip flexion joint torque also showed a notable disparity, with amputees' intact limb generating the highest torque 2.92 ± 0.65 Nm/kg*9.8), which reflects a compensatory mechanism where the intact limb takes on more of the workload. This increased demand on the intact limb highlights the necessity for targeted strengthening and rehabilitation efforts to mitigate potential overuse injuries.

Hip Adduction: In the analysis of hip adduction, the adductor magnus muscle activation was relatively low across all groups, though it was slightly higher in the intact and amputated limbs of amputees compared to AB individuals. This suggests that while the adductor magnus contributes to maintaining stability, its role is less pronounced, likely due to altered gait mechanics that reduce the need for significant adduction. The joint torque for hip adduction did not show significant differences between the groups, indicating that despite the altered muscle activation, the torque generated remains somewhat consistent, albeit lower in amputees. This lower torque may impact lateral stability during gait, necessitating compensatory strategies in other muscle groups.

Hip Rotation: The sartorius muscle exhibited significantly higher activation on the amputated limb of amputees (0.12 ± 0.03 BW) compared to AB individuals (0.06 ± 0.04 BW). This increased activation is indicative of the sartorius muscle's role in compensating for the rotational stability typically provided by the natural limb. Although the joint torque associated with hip rotation did not differ significantly between the groups, the higher muscle activation in amputees suggests that additional muscular effort is required to maintain rotational control, which is crucial for stabilising the pelvis and lower limb during gait.

Knee Flexion: Knee flexion presented one of the most significant disparities, with the biceps femoris muscle showing markedly higher activation ($p= 0.03$) on the amputated limb of amputees (0.35 ± 0.22 BW) compared to both their intact limb (0.11 ± 0.01 BW) and AB individuals (0.12 ± 0.04 BW). This increased activation likely compensates for the reduced mechanical advantage provided by the prosthesis, particularly during the swing phase of gait. So, knee flexion torque was much lower on the prosthetic limb 0.94 ± 0.52 Nm/(kg*9.8) than in AB individuals 1.64 ± 0.16 Nm/(kg*9.8). This shows that the prosthesis has mechanical limitations and highlights the need for targeted rehabilitation to improve knee function.

Ankle Dorsiflexion: The gastrocnemius muscle activation, while not significantly lower in amputees, showed a trend towards reduced activity, particularly on the amputated limb (0.95 ± 0.19 BW) compared to AB individuals (1.35 ± 0.64 BW). A significantly lower ankle dorsiflexion torque was found in amputees 1.36 ± 1.03 Nm/(kg*9.8) compared to AB individuals 3.14 ± 0.33 Nm/(kg*9.8). This shows how difficult it is for amputees to push off and clear their feet during gait. This limitation is critical because it affects the overall efficiency and safety of walking with amputees.

The comparative analysis of muscle forces and joint torques underscores the complex compensatory mechanisms employed by transtibial amputees to maintain gait. The increased demands on specific muscle groups, particularly on the prosthetic limb, reflect the biomechanical adaptations necessary to overcome the challenges posed by the absence of a limb. These findings emphasise the importance of targeted rehabilitation strategies that focus on strengthening the affected muscle groups and improving joint function to enhance gait efficiency and reduce the risk of secondary complications in individuals with transtibial amputations. These adaptations are crucial for maintaining mobility and quality of life in this population.

6.8 Conclusion

This chapter offers the OpenSim modelling for both amputees (TTAs) and non-amputees (AB) using OpenSim MSK modelling system. Both TTA and AB individuals successfully finished the scaling step, which offered a dependable approach to align model dimensions with participants' physical characteristics, such as weight and height. The quality of the model is confirmed by the low total square and RMS errors. Additionally, the small maximum error emphasises the precision of the model, guaranteeing exceptional realism in MSK modelling.

The gait biomechanics of TTAs are shown to adjust through significant variations in joint kinematics, particularly in pelvic tilt and hip adduction. Prior studies corroborate these results and indicate the importance of measuring joints angle and calculating joint ROM.

The analysis of joint torque and muscle forces during gait reveals the significant impact of amputation on the biomechanics of movement, particularly highlighting the differences in forces during hip flexion and knee extension between amputees' (intact and amputated limb) and non-amputees. Specifically, iliopsoas and biceps femoris show increased activation in the amputated limb, reflecting the greater effort required to move and stabilise the amputated limb. This heightened activation suggests a compensatory mechanism aimed at maintaining balance and gait stability. However, despite these adaptive responses, the amputated limb demonstrates reduced muscle activation (Vastus medialis and Vastus lateralis) and joint torque (hip adduction, knee flexion, and ankle dorsiflexion) when compared to able-bodied individuals.

In the next chapter, we will conclude the thesis main findings, the limitation we faced, and the future research direction.

Chapter 7: Conclusion, Limitations, and Future Directions

The primary aim of this thesis was to investigate the interplay between pain and changes in gait biomechanics in TTA using a variety of advanced methods. The findings provide critical insights into biomechanical adaptations, compensatory strategies, and the influence of pain on gait dynamics in amputees, particularly in comparison to AB.

7.1 Conclusion

A rigorous experiment protocol was developed, using advanced equipment and software tools. This protocol was reviewed and approved by the University of Birmingham research ethics committee.

Using the experimental protocol, the project successfully acquired biomechanics gait data from AB and TTA. The recruitment met our target, and data were demonstrated in good quality.

Gait biomechanics were evaluated, including spatial, temporal, and kinetic gait parameters, such as stride length, cadence, ground reaction forces, and joint kinematics, in amputees and comparing them to AB. Key findings in gait analysis are:

Temporal measures like stance time did not show any significant differences between the groups in terms of gait parameters. However, TTAs had wider strides, which suggests that they had changed how they balanced. Spatiotemporal parameters highlighted that amputees walked faster but with lower cadence, reflecting adaptations to maintain functional mobility. Furthermore, the limb that had been amputated had significant decreases in the GRFs in the mediolateral, vertical, and anteroposterior planes. This showed that the prosthesis put biomechanical limits on the body.

The musculoskeletal model of unilateral transtibial amputation was applied to provide information on joint kinematics, kinetics and muscle forces. The key findings in musculoskeletal modelling are:

Joint kinematics and torque analyses revealed notable adaptations. TTAs displayed a reduced pelvic tilt and rotation on the amputated side, likely reflecting efforts to stabilise the prosthetic limb. The amputated side showed significant reductions in knee flexion and ankle dorsiflexion angles, indicating limitations in the range of motion. It was seen that joint torques increased on the limb that wasn't amputated, but hip flexion, knee flexion, and ankle dorsiflexion torques decreased significantly on the side that wasn't amputated. This shows the functional imbalance that happens when a prosthesis is used.

Muscle activation patterns further highlighted these compensatory strategies. Amputees had very different muscle forces, with more activation of hip flexors like the iliopsoas and knee extensors like the vastus lateralis. This suggests that they depend on certain muscle groups to fix biomechanical problems. The intact limb also showed elevated muscle activity, reflecting its critical role in compensating for limitations on the prosthetic side.

Researchers found that pain profoundly affects gait biomechanics. Pain interference with walking strongly correlated with reductions in stride width, stride length, and speed, particularly on the amputated limb. GRF parameters exhibited significant negative correlations with pain scores, indicating that increased pain is associated with reduced force generation during gait.

In conclusion, this thesis has provided valuable insights into the biomechanical adaptations and challenges faced by individuals with lower limb amputations. By highlighting the critical role of pain in influencing gait mechanics and the compensatory strategies employed by amputees, this work underscores the need for personalized and innovative rehabilitation strategies. Future research should focus on integrating advanced technologies, such as feedback training and FES, to further enhance the quality of life for amputees. Through continued interdisciplinary collaboration and research, there is potential to significantly improve the outcomes and experiences of this population.

7.2 Limitations

This thesis has made significant contributions, but it should acknowledge some limitations:

Sample size: The small sample size limited the study, yet it accurately represented the population.

Pain Assessment: While pain was identified as a critical factor influencing gait, the study relied on self-reported measures of pain. Objective measures, such as neuroimaging or biomarkers, could provide a more comprehensive understanding of pain's impact on gait mechanics.

Cross-sectional Design: The study's cross-sectional design limits the ability to infer causality between amputation, pain, and altered gait mechanics. Longitudinal studies are needed to explore the progression of these factors over time.

7.3 Future Directions

Building on the findings of this thesis, several avenues for future research are recommended:

Feedback Training: Future studies should explore the use of feedback training as a rehabilitation strategy for amputees. By providing real-time visual or auditory feedback on gait mechanics, amputees could be guided to adopt more symmetrical and efficient walking patterns, potentially reducing compensatory strategies and associated pain.

Intervention Testing: Studies assessing specific interventions, such as targeted rehabilitation protocols or novel prosthetic designs, would provide actionable insights for improving mobility and reducing pain in TTAs.

Functional Electrical Stimulation (FES): FES has shown promise in improving muscle function and reducing pain in individuals with various neurological impairments. Research should investigate the efficacy of FES in enhancing gait mechanics and reducing compensatory movements in amputees. This approach could offer a novel intervention to address the challenges identified in this thesis, particularly in muscle activation patterns and joint torque distribution.

Longitudinal Studies: To better understand the long-term impact of amputation and pain on gait mechanics, longitudinal studies are essential. These studies could track changes in gait over time and assess the effectiveness of different rehabilitation strategies, including those mentioned above.

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Appendices

Appendix A

Systematic review

This section includes a series of detailed tables that summarize the key components of the selected studies. (Table A.1) presents the sample size, study design, and demographics of each study. (Table A.2) provides an overview of the pain assessment tools utilized in each study. Finally, (Table A.3) outlines the experimental measures and key findings from the selected studies.

Table A.0.1 Summary of the studies sample size, study design, and demographics.

Author (year)	Sample size	Age (years)	BMI (kg/m^2)	Study design	Study aims
(Postema et al., 1997)	TTA=10	49	--	Cohort study	Assess a prosthesis or its components
(Sjödahl et al., 2001)	TFA=9	33	23	Cohort study	Assess a rehabilitation regiment
(Berge et al., 2005)	TTA=15	51	29	Cohort study	Assess a prosthesis or its components
(Kulkarni et al., 2005)	TTA=20; TFA=20		27	Case series	Determine the prevalence of back pain
(Morgenroth et al., 2010)	TFA=17; non-amputee=5	51*	26	Case control	Quantify kinematics/kinetics differences
(Wolf et al., 2012)	TFA=5		25	Cross sectional study	Assess a prosthesis or its components
(Russell Esposito & Wilken, 2014)	TFA=16; non-amputee=12	30*	27	Case control	Quantify kinematics/kinetics differences
(Segal et al., 2014)	TTA=10	56	27	Cross sectional study	Assess a prosthesis or its components

(Fatone et al., 2016)	TTF=23	48	28	Case control	Quantify kinematics/kinetics differences
(Talbot et al., 2017)	TTA=44	27		Randomized controlled trial	Assess a rehabilitation regiment
(Actis et al., 2018)	TTA=8; non-amputee =8	44*	29	Case control	Quantify kinematics/kinetics differences
(Butowicz et al., 2018)	TTA=5; TFA=3; non-amputee =10	38	28*	Case control	Quantify kinematics/kinetics differences
(Butowicz, Dearth, et al., 2019)	TTA=22; TFA=9	34	28	Case control	Quantify kinematics/kinetics differences
(Dillingham et al., 2019)	TTA=26	50		Cohort study	Assess a prosthesis or its components
(Petrini et al., 2019)	TFA=2	42		Cohort study	Assess a prosthesis or its components
(Mahon et al., 2020)	TTA=22; TFA=10	27	26	Case control	Quantify kinematics/kinetics differences
(Honegger et al., 2021)	TTA=4; non-amputee =4	46	29	Case control	Quantify kinematics/kinetics differences
(Acasio et al., 2022)	TTA=24; TFA=11; non-amputee=15	35	28	Case control	Quantify kinematics/kinetics differences

Note:

*indicates significant difference from the non-amputee group

-- indicates not applicable

TTA – transtibial amputees

TFA – transfemoral amputee

Table A0.2 Summary of selected studies pain assessment tools.

Author,year	Pain assessment tool	Measure of chronic pain experiences	Applications and findings
(Postema et al., 1997)	-VAS -Customised questionnaire	-Severity: a scale of 0 “no pain” to 10 “extreme pain”. -Address the specific preferences and order of importance of different aspects of the prosthesis	-Amputees who utilized energystoring prosthetic feet experienced lower pain severity (7.0) compared to those using conventional feet (7.3; $p < 0.05$), as evaluated by the Visual Analog Scale.
(Sjödahl et al., 2001)	-Self-reporting pain	-Types of reporting pain: low back pain, phantom limb pain, stump pain.	-Before gait training, 71% of transfemoral amputees reported low back pain compared to other types of pain ($p < 0.05$). However, after gait training, the selected transtibial amputees did not report low back pain.
(Berge et al., 2005)	-Chronic Pain Grade Questionnaire - Multidimensional Fatigue Inventory	-Severity: a scale of 0 “no pain” to 10 “pain as bad as could be”. -Pain interference with daily activities: a 10-point discrete scale of 0 “no interference/not bothersome” to 10 “unable to carry on any activities/as bothersome as could be.” -Outcome measure: performance, residual limb pain grade, and fatigue.	-The only significant difference between the populations, based on the type of prosthesis pylon used, was observed in the questionnaire for performance, where amputees using shock-absorbing pylon found it 1.7 times easier to perform normal day-to-day activities compared to those

			using other pylons (p = 0.04).
(Kulkarni et al., 2005)	-Visual Analog Scale -Customised, semi-structured questionnaire	-Severity: a scale of 0 “no pain” to 10 “extreme pain”. -Location: back pain, stump pain or phantom limb pain. -Frequency: occasionally, often, or constant.	-Transfemoral amputees were more likely to experience back pain (81%) compared to transtibial amputees (62%; $p < 0.05$), and among those with severe back pain, 89% and 81% also reported severe pain in the phantom limb and severe stump pain, respectively. -Amputees with pain had a signi cantly lower average selfselected walking speed (1.06 ± 0.25m/s) compared to amputees without pain (1.26 ± 0.18m/s) ($p < 0.05$). -Center of pressure displacement was 114% higher among amputees with pain than the painfree group.
(Morgenroth et al., 2010)	-24-item Roland-Morris Disability Questionnaire (RMQD) -Chronic Pain Grade Questionnaire	-Severity: indirectly provide information about pain severity based on the RMQD scores / a scale of 0 “no pain” to 10 “pain as bad as could be”. -Location: lower back pain.	-Amputees who self-reported lower back pain had an average pain level of 4.8 (SD 1.8) over the previous 3 months. Despite their pain, they were a population with relatively little associated disability, reporting an average

			interference with work and social engagements of 2.2 (SD 2.9). -Amputees with pain exhibited a signi cantly higher trunk range of motion (ROM) in the transverse direction (15.4 ± 4.2 degrees) compared to amputees without pain (11.1 ± 3.8 degrees) with a pvalue of 0.029.
(Wolf et al., 2012)	-A customised questionnaire	-No detailed information is available	-No detailed information is available
(Russell Esposito & Wilken, 2014)	-Prosthetic Evaluation Questionnaire (PEQ)	-Severity: ranged between extremely intense to extremely mild. -Location: phantom limb pain, residual limb pain or back pain. -Frequency: selecting from "never", "only once or twice", "a few times (about once/week)", "fairly often (2/3 times/week)", "very often (4–6 times/week)", "several time every day and all the time" or "almost all the time". -Duration: selecting from "I have none"; "a few seconds", "a few minutes", "several minutes to an hour", "several hours", "a day or two" and "more than two days".	-The group of transfemoral amputees with low back pain reported experiencing low back pain with a frequency ranging from one time per week to all the time or almost all the time. During walking, the trunk anglewas signi cantly greater among transfemoral amputees with pain (7.44 ± 3.65 degrees) compared to transfemoral amputees without pain (4.34 ± 4.4 degrees) with a pvalue of 0.003.
(Segal et al., 2014)	-Residual limb pain grade	-Severity: present, average, worst/least	- Among patients with transtibial

		residual limb pain ranged from 0 “no pain” to 10 “severe pain”. -Pain interference with daily activities: ranged from 0 “no interference” to 10 “severe interference.”	amputations, the overall pain severity was 2.5 ± 1.5 for the torsion adapter and 3.2 ± 1.6 for the rigid adapter, showing no statistically signi cant difference ($p = 0.14$). However, wearing the torsion adapter resulted in approximately a 50% reduction in pain interference with daily activities compared to wearing the rigid adapter, and this difference was found to be statistically significant ($p = 0.026$).
(Fatone et al., 2016)	-Visual Analog Scale	-Severity: a scale of 0 “no pain” to 10 “extreme pain”.	-Although participants in the no back pain group had higher comfort scores, these differences were not statistically significant ($p = 0.02$) as assessed by the Visual Analog Scale.
(Talbot et al., 2017)	-Brief Pain Inventory	-Severity: “worst”, “least”, “average” and “current” levels, with scores ranging from 0 “no pain”, to 10 “pain, as bad as one can imagine.” -Pain interference with daily activities including general activity, walking, work, mood, enjoyment of	-Pain severity improved from baseline to 12 weeks in both transtibial amputee groups, with the group that had a home-based neuromuscular

		life, relations with others, and sleep.	electrical stimulation rehabilitation program in addition to the traditional military amputee rehabilitation program (TMARP) showing greater improvement compared to the group that had only the TMARP alone.
(Actis et al., 2018)	-Modified Oswestry Low Back Pain Questionnaire	-Severity: range from 0 “the pain is mild and comes and goes” to 5 “the pain is severe and does not very much.” -Pain interference with daily activities including personal care, lifting, walking, sitting, standing, sleeping, social life, traveling and employment/homemaking.	-Out of the six participants with transtibial amputation, only one reported experiencing more than minimal low back pain.
(Butowicz et al., 2018)	-Visual Analog Scale	-Severity: a scale of 0 “no pain” to 10 “extreme pain”.	-Participants with lower limb amputations were excluded if they reported experiencing pain or discomfort rated at more than 3 out of 10 on a visual analog scale, regardless of the cause.
(Butowicz, Dearth, et al., 2019)	- National Institutes of Health (NIH) minimal dataset for chronic low back pain - Visual Analog Scale	-Location; Low back pain. -Duration: pain that persisted for at least three months and with pain on at least half the days in the past six months.	-Low back pain status was determined using the NIHrecommended minimal data set, where chronicity is de ned as pain that persisted for at least three

		-Severity: a scale of 0 “no pain” to 10 “extreme pain”.	months and with pain on at least half the days in the past six months. Additionally, individuals with phantom limb pain and/or discomfort, regardless of the cause (> 4/10 on a visual analog scale for pain), were excluded from the study.
(Dillingham et al., 2019)	-Prosthetic Evaluation Questionnaire (PEQ)	-Severity: ranged between extremely intense to extremely mild. -Location: phantom limb pain, residual limb pain or back pain. -Frequency: selecting from “never”, “only once or twice”, “a few times (about once/week)”, “fairly often (2/3 times/week)”, “very often (4–6 times/week)”, “several time every day and all the time” or “almost all the time”. -Duration: selecting from “I have none”; “a few seconds”, “a few minutes”, “several minutes to an hour”, “several hours”, “a day or two” and “more than two days”.	-Participants reported signi cantly higher levels of satisfaction, comfort, and function with the immediate t modular prosthetic device compared to the conventional device, as indicated by PEQ ratings (30.9 vs 24.8, p = 0.002).
(Petrini et al., 2019)	-Neuropathic Pain Symptom Inventory -Visual Analog Scale	-Severity: a scale of 0 “no pain” to 10 “worst pain imaginable”/ a scale of 0 “no pain” to 40 “extreme pain”. -Duration: the duration of pain experienced during the past 24 hours	- Participants showed reduced phantom limb pain with the use of neural sensory feedback.

		was de ned by selecting from “less than 1 hour”, “between 1 and 3 hours”, “between 4 and 7 hours”, “between 8 and 12 hours” and “permanently.” -Frequency: the frequency of pain attacks during the past 24 hours was de ned by selecting from “no pain attack”, “between 1and 5”, “between 6 and 10”, “between 11” and “20 to more than 20”.	
(Mahon et al., 2020)	- Customized, self-reported medical questionnaire - Visual Analog Scale	-Location: low back pain. -Severity: a scale of 0 “no pain” to 100 “extreme pain”.	- The percentage of individuals with transtibial limb loss experiencing low back pain and pain while walking decreased from 0 months to 2 months and 4 months. In contrast, the percentage of individuals with transfemoral limb loss experiencing low back pain and pain while walking initially increased from 0 months to 2 months and 4 months but then generally decreased at 6 months and 12 months. Low back pain severity while walking was highest for individuals with both transtibial and transfemoral limb loss at 2 months and 4 months. However,

			for individuals with transtibial limb loss, low back pain intensity while walking was less at 0 months, 6 months, and 12 months. Notably, individuals with transfemoral limb loss did not report low back pain while walking at 0 months, 6 months, or 12 months.
(Honegger et al., 2021)	-Modi ed Oswestry Low Back Pain Questionnaire	-Severity: ranged from 0 "the pain is mild and comes and goes" to 5 "the pain is severe and does not very much." -Pain interference with daily activities including personal care, lifting, walking, sitting, standing, sleeping, social life, travelling and employment/homemaking.	-None of the participants, including both AB individuals (0 ± 0%) and those with transtibial amputation (2.5 ± 3%), reported more than minimal low back pain.
(Acasio et al., 2022)	- Customized, self-reported medical questionnaire - Visual Analog Scale	-Location: low back pain. -Severity: a scale of 0 "no pain" to 100 "extreme pain".	- The percentage of individuals with transtibial limb loss experiencing low back pain and pain while walking decreased from 0 months to 2 months and 4 months. In contrast, the percentage of individuals with transfemoral limb loss experiencing low back pain and pain while walking initially increased from 0 months to 2 months and 4

			months but then generally decreased at 6 months and 12 months. Low back pain severity while walking was highest for individuals with both transtibial and transfemoral limb loss at 2 months and 4 months. However, for individuals with transtibial limb loss, low back pain intensity while walking was less at 0 months, 6 months, and 12 months. Notably, individuals with transfemoral limb loss did not report low back pain while walking at 0 months, 6 months, or 12 months.

Table A.0.3 Summary of selected studies experimental measure and key findings

Author, year	Experimental measure	Key findings
(Postema et al., 1997)	-Spatiotemporal measure: walking speed; cadence -Kinematic measure: whole lower limb range of motion -Kinetic measure: impulse of deceleration and acceleration phase of the anteroposterior component of the ground reaction force; ankle power.	The impulses of the anteroposterior component of the ground force showed small, statistically non-significant differences. The power storage and release phases as well as the net results also showed small differences.
(Sjödahl et al., 2001)	-Spaciotemporal measure: walking speed	Self-selected comfortable walking speeds increased in unilateral transfemoral amputees after a novel therapeutic treatment.
(Berge et al., 2005)	-Spaciotemporal measure: walking speed; step length -Kinematic measure: knee motion -Kinetic measure: loading rate and decelerative peak of ground reaction force	No differences were found across pylons for self-selected walking speed, prosthetic sidestep length, prosthetic side loading rate and decelerative peak of the vertical ground reaction force, peak pylon acceleration. The only statistically significant was for the prosthetic-side knee angle where subjects displayed an average of 2.6 degrees more extension with the rigid pylon than the shock absorbing pylons.
(Kulkarni et al., 2005)	-Spaciotemporal measure: walking speed -Kinematic measure: lumbar spine range of motion. -Kinetic measure: impact ground reaction force; centre of pressure	Impact ground reaction forces during walking, irrespective of limb, were significantly greater ($p < 0.05$) in the pain-free group than in the pain group, as was walking speed. Gait asymmetry measures were similar in both groups. Centre of pressure displacement

		measures during standing were greater in the pain group than in the pain-free group.
(Morgenroth et al., 2010)	-Spaciotemporal measure: walking speed -Kinematic measure: lumbar spine range of motion	Transfemoral amputees with low-back pain showed greater transverse plane rotational excursion in their lumbar spine during walking when compared with transfemoral amputees without low back pain ($p = 0.029$; effect size = 1.03). There were no significant differences in sagittal or coronal plane lumbar spine excursions during walking between these two groups.
(Wolf et al., 2012)	-Spaciotemporal measure: walking speed; step length -Kinetic measure: vertical ground reaction force; ankle, knee and hip power	Knee power generated by the nondisabled limb during stair ascent for subjects wearing the C-Leg was significant greater than for those wearing the Power Knee. Knee power generated by prosthetic knee units was significant greater for subjects while wearing the Power Knee.
(Russell Esposito & Wilken, 2014)	-Spaciotemporal measure: walking speed - Kinematic measure: continuous relative phase, calculated using three-dimensional pelvis and trunk motion	The patient groups maintained transverse plane CRP consistent with able- bodied participants ($p = 0.966$), but not sagittal ($p < 0.001$) and frontal plane CRP ($p = 0.001$).
(Segal et al., 2014)	-Spaciotemporal measure: step	Amputees wearing a torsion adapter tended to take more low- and medium-intensity steps per day (331 ± 365 and 437 ± 511 difference in steps; effect size = 0.44 and 0.17; convedence interval [CI], 70–592 and 71–802;

		p = 0.019 and 0.024, respectively)
(Fatone et al., 2016)	-Spaciotemporal measure: walking speed -Kinematic measure: pelvic, lumbar, and thoracic motion	Opposite patterns of motion were observed between groups (persons with transfemoral amputation with and without low back pain) in sagittal and transverse lumbar kinematics
(Talbot et al., 2017)	-Spaciotemporal measure: walking speed	There was no group difference in the residual limb
(Actis et al., 2018)	-Kinematic measure: trunk and whole lower-limb motion. -Kinetic measure: EMG of eight muscles and ground reaction forces	Participants with transtibial amputation had greater peak angles during the sit-to-stand motion, including trunk extension, trunk lateral bending toward the intact leg, trunk axial rotation toward the prosthetic leg, and trunk-pelvis axial rotation toward the intact leg. Participants with transtibial amputation were more asymmetric than control participants with greater vertical force generation in the intact leg and greater posterior and medial force generation in the prosthetic leg shortly after lift-off.
(Butowicz et al., 2018)	-Spaciotemporal measure: walking speed -Kinematic measure: lumbar spine range of motion -Kinetic measure: EMG activity from the trunk muscles (bilateral thoracic and lumbar erector spinae)	Lower limb amputees demonstrated a second thoracic erector spinae onset during mid-to-terminal swing, and activation for a larger percentage of the gait cycle. Lower limb amputees demonstrated an earlier onset of lumbar erector spinae and activation for a larger percentage of the gait cycle at most speeds.

		Lower limb amputees walked with increased frontal plane trunk range of motion.
(Butowicz, Dearth, et al., 2019)	-Kinematic measure: trunk and pelvis motion -Kinetic measure: EMG activity; centre of pressure	Centre-of-pressure measures and trunk muscle activation ratios were similar between groups, while participants with chronic low back pain demonstrated greater trunk motion and reduced local dynamic stability.
(Dillingham et al., 2019)	-Spaciotemporal measure: walking speed; step length, stance characteristics	There was no evidence of significant difference in gait characteristics as described by stance phase, double support, or any decrement in gait speed with the immediate t modular prosthetic system.
(Petrini et al., 2019)	-Spaciotemporal measure: walking speed	Walking speed increased while mental and physical fatigue decreased for both participants during neural sensory feedback.
(Mahon et al., 2020)	-Spaciotemporal measure: walking speed -Kinematic measure: trunk pelvis range of motion and intersegmental coordination	An interaction effect between time and group existed for sagittal ($p = .039$) and transverse ($p = .009$) continuous relative phase at self-selected walking velocity and transverse trunk range of motion ($p = .013$) and sagittal continuous relative phase ($p = .005$) at controlled walking velocity. Trunk range of motion generally decreased, and trunk-pelvis coordination generally increased with increasing time after initial ambulation. Sagittal trunk and pelvis range of motion were always less and frontal trunk-pelvis coordination

		was always greater for persons with more distal limb loss.
(Honegger et al., 2021)	-Kinematic measure: trunk and whole lower limb motion -Kinetic measure: ground reaction force	Participants with transtibial amputation had greater axial rotation toward their intact limb ($p = 0.029$), greater abdominal muscle activity ($p < 0.001$)
(Acasio et al., 2022)	-Spaciotemporal measure: walking speed -Kinematic measure: trunk and whole lower limb motion -Kinetic measure: ground reaction force	In the frontal and transverse planes, thorax range of motion was up to 66.6% smaller in lower limb amputees without chronic low back pain compared to amputees with chronic low back pain ($p < 0.001$) and injured individuals without low back pain ($p < 0.001$). In the sagittal plane, pelvis range of motion was 50.4% smaller in amputees without chronic low back pain compared to amputees with chronic low back pain ($p = 0.014$).

Appendix B

Experimental study protocol

In this section will provide the risk passement form (Figure B.1), laboratory setup (Figure B.2), consent form (Figure B.3), and ethical approval (Figure B.4).

THE UNIVERSITY
OF BIRMINGHAM
SCHOOL OF ENGINEERING
MEMORANDUM

Risk Assessment No.

The Risk Assessment number will be allocated by the local H&S Coordinator

The RA must be signed by the PI or Supervisor to approve it being suitable and sufficient.
Each person named below must have their own copy of the RA, any subsequent changes or reviews must received by each person named below.

Persons covered by this assessment

Dr Ziyun Ding, Miss Christine Singleton, Mr Weida Wang, Mr Khalid Alsayed,

All personnel will refuse to carry out procedures they deem unsafe or do not feel suitably trained to carry out.
All University of Birmingham staff and visitors are covered under the University of Birmingham's insurance policy.
All personnel should make themselves aware of the School of Engineering's Health and Safety Code of Practice
Brief Description of task

Figure 0.1 Risk assessment for the motion analysis laboratory

Lab setup

Name: ----------------- Contact: ---------------------------------- Date: -----------------

Calibration:

☐ VICON

☐ Chair height

☐ EMG (two EMG systems)

☐ Force Plate1

☐ Force Plate2

☐ System (VICON+EMG+Foce-plate)

☐ Consent sign

☐ Brief Pain Inventory

☐ Skin preparation for markers/sensors

☐ Particpants' outfit

☐ Markers placement

☐ Sensors placement

General Information:

Non-invasive attaching procedures of the surface sensors and reflective markers include slight discomfort from minor abrasion of the skin area. A hypoallergenic light scrubbing paste will be used, which is distributed by a medical supplier for this purpose. Grids will be attached to the participant using a medical grade adhesive foam, and secured in place using Hypofix tape. This is commonly used in physiotherapy clinics. As all equipment is designed to be used clinically, we do not expect an allergic reaction, but should this occur, all adhesives will be removed, the area washed with water and a moisturiser will be applied. Prior electrode placement, the skin will need to be shaved to remove any hair. However, single use disposable razors will be used thus there is no expected risk from this procedure.

The motor tasks accomplished by the participants will not lead to anxiety or stress higher than that already expected from everyday life. To deal with the above, appropriate rest time will be provided throughout the experimental trials. Additionally, extra rest periods will be given to the participants at any time, if they need it.

Figure 0.2 laboratory setup

UNIVERSITY OF BIRMINGHAM

Dr Ziyun Ding
School of Engineering
University of Birmingham
Birmingham B15 2TT
Email: [redacted]

Version 2.0, 31/01/2022

Ethics Code: ERN_21-0166

Study of Pain, Motion, and Muscle activity for Amputees.

CONSENT FORM

Name (capital letters):

Date of Birth:

Participation Code:

Contact (Phone/E-mail):

Please initial the boxes

1. I confirm that I have read and understood the Participant Information Sheet dated 31/01/2022 for the above study and have had the opportunity to ask questions. ☐
2. I understand that my participation is voluntary and that I am free to withdraw at any time without giving any reason, and without my medical care, education or legal rights being affected. ☐
3. I agree to my anonymised data being stored on password-protected University of Birmingham computer systems. ☐
4. I agree to take part in the research project. ☐
5. I agree to be contact regarding participation in future studies. ☐
6. I would like to be provided with a summary report of our findings at the end of the study, at my request (optional - only initial the box if you wish to receive this) ☐
7. I understand that when I withdraw my consent, no additional data will be collected; the data collected up to the point of withdrawal will be retained and used in the study. ☐

Name of subject	Date	Signature
______________	______________	______________
Name of person taking consent	Date	Signature
______________	______________	______________

Consent Form v2.0 31/01/2022

Figure 0.3 consent form

aResearch Passport Application Form – Version 5.1 08/Jul/2020

Please refer to the guidance notes before completing the form.

Section 1 - Details of Researcher *To be completed by Researcher*

1.	Surname: Alsayed	Prof☐ Dr☐ Mr☑ Mrs☐
	Forename(s): Khalid	Miss ☐ Ms☐ Other☐
	Contact Tel: Contact Email:	
2.	Date of birth:	
3.	Professional registration details, if applicable (Doctors undertaking any form of medical practice should confirm they have a licence to practise). N/A ☑	
4.	Employer: University of Birmingham or place of study: University of Birmingham	
	Work Address/Place of Study: School of Engineering, University of Birmingham, B15 2TT	
	Post or status held: PhD Mechanical Engineering Student	

Section 2 - Details of Research *To be completed by Researcher*

5. What type of Research Passport do you need? Project-specific ☑ Multi-project ☐

If you will be conducting one project only please complete the details below. If you anticipate that you will be undertaking more than one project at any one time, please give details in the Appendix.

Project Title: A study of functional electrical stimulation on lower limb amputees

Project Start Date: 01/01/2022 End Date: 01/06/2024

Proposed start and end-date of 3-year Research Passport:

Start Date: End Date:

NHS organisation(s):	Dept(s):	Proposed research activities:	Manager in NHS organisation:
West Midlands Rehabilitation Centre, Birmingham Community Healthcare NHS Foundation Trust	Amputee service	Taking consent on site	Poornashree Ramamurthy

Section 3 – Declaration by Researcher *To be completed by Researcher*

6.	Have you ever been refused an honorary research contract?	Yes ☐ No ☑
	Have you ever had an honorary research contract revoked?	Yes ☐ No ☑
	If yes to either question, please give details:	

I understand that my data will be used for the purposes described in section 6 of the Information for researchers, R&D and HR staff in Higher Education Institutions and the NHS guidance document. I warrant that the information provided as part of this Research Passport application and the attached documents are a true and accurate reflection.

Signed*: Date: 13/10/2021

When Sections 1-3 have been completed, the researcher should forward the form to the appropriate person to complete Section 4.

** It is recommended that the researcher prints, signs and scans the form. Where this is not possible, the researcher should state 'authorised by email', in place of a wet-ink and scanned signature. Where authorisation occurs by email, the full email trail should be presented as evidence with the document for*

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Section 4 - Suitability of Researcher *To be completed by researcher's substantive employer, e.g. line manager, or academic supervisor*		
7.a	Will this person's research activity mean that they may be undertaking regulated activity with children and/or adults as defined in the Safeguarding Vulnerable Groups Act 2006, as amended (in particular by the Protection of Freedoms Act 2012)? (please use the Research Passport algorithm to make this judgement)	Yes ☐ No ☒
7. b	I am satisfied that the above named individual is suitably trained and experienced to undertake the duties associated with the research activities outlined in this Research Passport form.	
	Signed*:	Date:
	Name: Prof. Karl Dearn	Job Title: Head of Mechanical Engineering, Deputy Head of School of Engineering
	Department and Organisation: School of Engineering, University of Birmingham	
	Address: School of Engineering, University of Birmingham, B15 2TT	
	Tel No: +44 (0) 121 414 4190	Email: k.d.dearn@bham.ac.uk
	Managerial responsibility for the applicant: Line manager	
When Section 4 has been completed, the researcher should forward the form to the appropriate person to complete Section 5. ** It is recommended that the person authorising Section 4 prints, signs and scans the form. Where this is not possible, they should state 'authorised by email', in place of a wet-ink and scanned signature. Where authorisation occurs by email, the full email trail should be presented as evidence with the document for further authorisations and/or in application for an HRC/LoA. Identity and/or other checks may be made using videoconferencing, or other technology, where appropriate.*		

Section 5 - Pre-engagement checks *To be completed by the HR department of the researcher's substantive employer or registry at place of study*		
8.	Does the above named individual's research involve Regulated Activity with children and/or adults as defined in the Safeguarding Vulnerable Groups Act 2006, as amended (in particular by the Protection of Freedoms Act 2012)?	☐ Yes ☐ No
	If yes to the above, has the above named individual been checked against ISA barred lists for adults and/or children, as appropriate and have you received confirmation via the criminal record disclosure that the person is not barred from working with adults and/or children? (*NB individuals who are barred from working with adults or children must not undertake a regulated activity in the NHS with the vulnerable group from which they are barred, and you must not submit a Research Passport form in such cases).*	Checked against: ISA Adults List? Yes ☐ No ☐ N/A ☐ ISA Children's List? Yes ☐ No ☐ N/A ☐
	Can you confirm that a clear criminal record disclosure has been obtained for the above-named individual, with no subsequent reports from the individual of changes to this record? *NB for Regulated Activity this must be an enhanced level criminal record check. For non-regulated activity, ensure the criminal record check is at the mandated level.*	Yes ☐ No ☐ N/A ☐
	If yes, please provide details of the clear disclosure:	
	Date of disclosure:	Type of disclosure:
	Disclosure No.:	Organisation that requested disclosure:
9.	Have the pre-engagement checks described below been carried out with regard to the above-named individual and is confirmation of the necessary checks, including any required satisfactory documentary evidence, available in the employing organisation's/place of study's records?	
	▪ Employment/student screening:	
	○ ID with photograph	Yes ☐ No ☐
	○ two references	Yes ☐ No ☐
	○ verification of permission to work/study in the UK	Yes ☐ No ☐
	○ exploration of any gaps in employment	Yes ☐ No ☐
	▪ Evidence of current professional registration	Yes ☐ No ☐ N/A ☐
	▪ Evidence of qualifications	Yes ☐ No ☐
	▪ Occupational health screening / clearance	Yes ☐ No ☐
	Is the named individual on a fixed term contract or is the contract end imminent? Yes ☐ No ☐	
	Please indicate current contract end-date Date:	

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Figure 0.4 Ethical approval

Appendix C

Instrumental gait analysis

In this section will provide the data for each participant spatiotemporal findings: For the TTA individual, speed remains relatively consistent across six trials, ranging from 1.29 to 1.34 m/sec. Stride width varies slightly between 0.08 and 0.18 meters. Cadence, step length, stride length, and stance time are reported separately for the intact and amputated limbs. The intact limb shows a cadence range of 1.52 to 1.68 steps per second, while the amputated limb ranges from 1.50 to 1.65 steps per second. Step length for the intact limb ranges from 0.59 to 0.80 meters and for the amputated limb from 0.61 to 0.83 meters. Stride length for the intact limb ranges from 1.16 to 1.65 meters, and for the amputated limb from 1.23 to 1.79 meters. Stance time for the intact limb ranges from 0.66 to 0.89 seconds, while for the amputated limb, it ranges from 0.64 to 0.73 seconds. For the eight AB individuals, speed ranges from 1.05 to 1.25 m/sec. Stride width varies from 0.07 to 0.13 meters. Cadence for the dominant limb ranges from 1.62 to 1.93 steps per second, while for the non-dominant limb, it ranges from 1.60 to 1.90 steps per second. Step length for the dominant limb ranges from 0.66 to 0.81 meters and for the non-dominant limb from 0.64 to 0.81 meters. Stride length for the dominant limb ranges from 1.31 to 1.61 meters, and for the non-dominant limb from 1.29 to 1.61 meters. Stance time for the dominant limb ranges from 0.67 to 0.80 seconds, while for the non-dominant limb, it ranges from 0.63 to 0.79 seconds (Table C.4&5), the p value of the amputees spatiotemporal (Table C.6).

The GRF For the TTAs, the medio-lateral peaks show slight variations between the intact and prosthetic limbs. The first peak ranges from -0.05 to 0.00 for the intact limb and -0.01 to 0.00 for the prosthetic limb. The second peak ranges from 0.06 to 0.12 for the intact limb and 0.03 to 0.09 for the prosthetic limb. The third peak ranges from 0.05 to 0.07 for the intact limb and 0.04 to 0.09 for the prosthetic limb. In the vertical direction, the first peak ranges from 0.95 to 1.23 for the intact limb and 0.91 to 1.41 for the prosthetic limb, indicating higher forces on the prosthetic side in some trials. The second peak ranges from 0.92 to 1.08 for the intact limb and 0.86 to 1.09 for the prosthetic limb. In the antero-posterior direction, the first peak ranges from -0.21 to -0.08 for the intact limb and -0.12 to 0.00 for the prosthetic limb, showing a greater negative force for the intact limb. The second peak ranges from 0.01 to 0.09 for the intact limb and 0.02 to 0.07 for the prosthetic limb. For the AB individuals, the medio-lateral peaks are more consistent across individuals. The first peak ranges from -0.05 to -0.01, the second peak from 0.04 to 0.09, and the third peak from 0.04 to 0.06. In the vertical direction, the first peak ranges from 1.03 to 1.23, and the second peak from 1.02 to 1.18, showing less variability than the TTA data. In the antero-posterior direction, the first peak ranges from -0.16 to -0.10, and the second peak from 0.17 to 0.23, indicating a more uniform force distribution compared to the TTAs. (Table C.7&8), and the p value amputees GRF (Table C.9). For the BPI findings to the instrumented gait findings the mean scores for the worst, least, and average pain were 4.33 (±2.05), 1.17 (±0.69), and 2.58 (±1.37), respectively. Current pain (Now) had a mean score of 1.33 (±0.75). Pain affecting activity levels averaged 2.67 (±1.49), while pain impact on mood and walking ability averaged 1.83 (±0.90) and 2.33 (±1.11), respectively. The mean scores for pain impact on work and sleep were 2.00 (±1.00) and 1.50 (±0.96), respectively. Lifestyle disruption due to pain had a mean score of 1.17 (±0.69).

The spatiotemporal parameters showed varied results. The mean speed for the TTAs was -0.16 m/s (±0.02). Stride width had a mean of -0.04 m (±0.03). Cadence for the intact limb was 0.14 steps/sec (±0.07), while for the amputated limb, it was 0.19 steps/sec (±0.05). The mean step length for the intact and amputated limbs were 0.02 m (±0.08) and -0.04 m (±0.08), respectively. Stride length for the intact limb was 0.02 m (±0.18), and for the amputated limb, it was -0.11 m (±0.18). Stance time for the intact limb was -0.02 s (±0.07), and for the amputated limb, it was 0.03 s (±0.04).

The GRF parameters also displayed variability among the participants. The mean ML_intact_Peak_1 was 0.01 (±0.01), while ML_intact_Peak_2 and ML_intact_Peak_3 had means of -0.10 (±0.02) and -0.01 (±0.01), respectively. For the amputated limb, ML_amputated_Peak_1 had a mean of 0.06 (±0.01), ML_amputated_Peak_2 had a mean of -0.02 (±0.02), and ML_amputated_Peak_3 had a mean of -0.01 (±0.02). Vertical force peaks for the intact limb were -0.01 (±0.10) for Peak_1 and 0.08 (±0.06) for Peak_2. For the amputated limb, Vertical_amputated_Peak_1 and Vertical_amputated_Peak_2 were 0.05 (±0.17) and 0.13 (±0.08), respectively. The anterior-posterior (AP) force peaks for the intact limb were 0.00 (±0.05) for Peak_1 and -0.19 (±0.03) for Peak_2, while for the amputated limb, they were 0.00 (±0.00) for both Peak_1 and Peak_2.

The data indicate that TTAs experience various levels of pain, which appear to correlate with changes in spatiotemporal gait parameters and GRF. The mean values and standard deviations highlight the variability among individuals, which is crucial for understanding the impact of pain on gait mechanics in TTAs. Further analysis is necessary to quantify these relationships and explore potential interventions to mitigate pain-related gait deviations (Table C.10).

Table C.0.4 Spaciotemporal data for the six TTAs

Parameter	TTA intact/amputated limb					
	1	2	3	4	5	6
Speed (m/sec)	1.30	1.29	1.34	1.33	1.32	1.31
Stride width (m)	0.14	0.08	0.18	0.14	0.15	0.16
Cadence (step/sec)	1.67/1.62	1.76/1.63	1.52/1.50	1.66/1.65	1.68/1.64	1.64/1.61
Step length (m)	0.70/0.72	0.59/0.61	0.80/0.83	0.76/0.80	0.60/0.82	0.73/0.75
Stride length (m)	1.40/1.42	1.16/1.23	1.65/1.79	1.55/1.61	1.21/1.64	1.49/1.57
Stance time (sec)	0.69/0.65	0.66/0.64	0.89/0.73	0.74/0.72	0.73/0.70	0.76/0.72

Note:
The following variables cadence, step length, stride length, and stance time the data presented as intact limb /prosthetic limb.

Table C.0.5 Spaciotemporal data for the eight AB

Parameter	AB dominant/non-dominant limb							
	1	2	3	4	5	6	7	8
Speed (m/sec)	1.15	1.18	1.18	1.15	1.15	1.25	1.15	1.05
Stride width (m)	0.08	0.11	0.13	0.11	0.09	0.07	0.13	0.08
Cadence (step/sec)	1.93 / 1.90	1.62 / 1.60	1.91 / 1.82	1.63 / 1.60	1.84 / 1.80	1.92 / 1.86	1.73 / 1.75	1.79 / 1.80
Step length (m)	0.68 / 0.67	0.71 / 0.69	0.81 / 0.76	0.68 / 0.67	0.72 / 0.71	0.66 / 0.64	0.68 / 0.70	0.80 / 0.81
Stride length (m)	1.36 / 1.35	1.42 / 1.40	1.61 / 1.56	1.35 / 1.34	1.44 / 1.43	1.31 / 1.29	1.36 / 1.38	1.60 / 1.61
Stance time (sec)	0.67 / 0.65	0.80 / 0.79	0.68 / 0.63	0.79 / 0.78	0.71 / 0.69	0.68 / 0.65	0.75 / 0.76	0.73 / 0.73

Note:
The following variables cadence, step length, stride length, and stance time the data presented as dominant limb /non-dominant limb.

Table C.0.6 Spatiotemporal p value

Group	Speed	Stride width	Cadence	Step length	Stride length	Stance time
TTA VS AB	0.02	0.43	NA	NA	NA	NA
Dominant VS Non-dominant	NA	NA	0.64	0.69	0.84	0.60
Dominant VS Intact	NA	NA	0.01	0.36	0.36	0.31
Dominant VS Amputated	NA	NA	0.00	0.31	0.31	0.13
Non-dominant VS Intact	NA	NA	0.03	0.43	0.40	0.20
Non-dominant VS Amputated	NA	NA	0.00	0.22	0.26	0.34
Intact VS amputated	NA	NA	0.11	0.27	0.27	0.12
ANOVA_p	**≤0.01**	**≤0.01**	**0.02**	0.62	0.80	0.53
ANOVA_f	65.61	11.05	4.47	0.61	0.34	0.77

Table C.0.7 GRF data for the six TTAs

Parameter		**TTA intact/amputated limb**					
		1	2	3	4	5	6
Medio-lateral	1st peak	-0.02/0.00	-0.01/0.00	-0.05/0.00	-0.03/-0.01	-0.04/-0.01	-0.03/-0.01
	2nd peak	0.06/0.07	0.12/0.09	0.07/0.08	0.07/0.06	0.08/0.03	0.09/0.08
	3rd peak	0.06/0.04	0.05/0.04	0.07/0.09	0.07/0.06	0.07/0.08	0.07/0.07
Vertical	1st peak	0.95/1.03	1.07/0.99	1.22/1.41	1.23/1.08	1.06/0.92	1.03/0.91
	2nd peak	0.92/1.05	1.08/1.09	1.03/0.90	0.97/0.98	1.05/0.86	0.95/0.94
Antero-posterior	1st peak	-0.10/-0.12	-0.08/0.00	-0.21/-0.02	-0.16/-0.03	-0.19/-0.10	-0.09/-0.04
	2nd peak	0.07/0.07	0.09/0.03	0.03/0.02	0.07/0.04	0.03/0.06	0.01/0.03

Note:
The following variables represent the GRF peaks in three direction for six TTA and the data presented as intact limb/prosthetic limb.

Table C.0.8 AB GRF

Parameter		AB							
		1	2	3	4	5	6	7	8
Medio-lateral	1st peak	-0.05	-0.03	-0.02	-0.01	-0.01	-0.01	-0.02	-0.01
	2nd peak	0.04	0.04	0.09	0.06	0.05	0.05	0.04	0.07
	3rd peak	0.05	0.06	0.06	0.06	0.06	0.04	0.04	0.04
Vertical	1st peak	1.09	1.07	1.23	1.04	1.05	1.03	1.08	1.07
	2nd peak	1.04	1.04	1.12	1.13	1.02	1.15	1.15	1.18
Antero-posterior	1st peak	-0.12	-0.15	-0.15	-0.16	-0.16	-0.10	-0.15	-0.13
	2nd peak	0.19	0.17	0.21	0.20	0.18	0.21	0.19	0.23

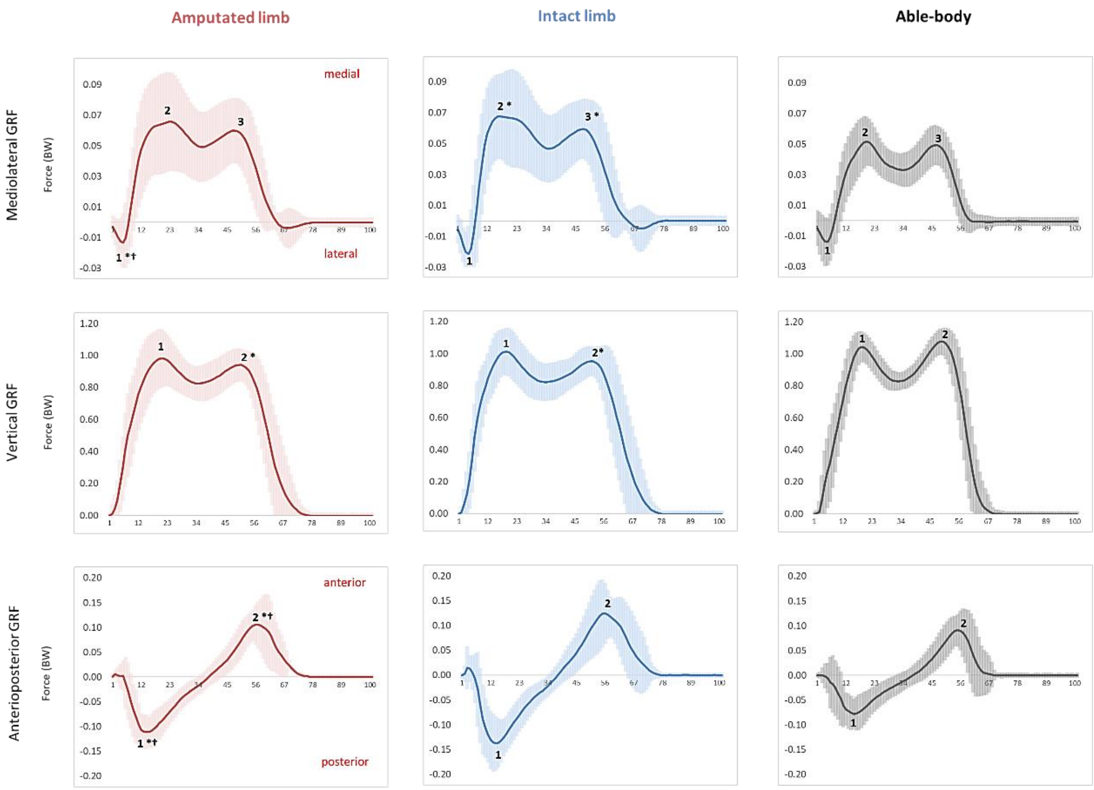

Figure 0.5 GRF for each group.

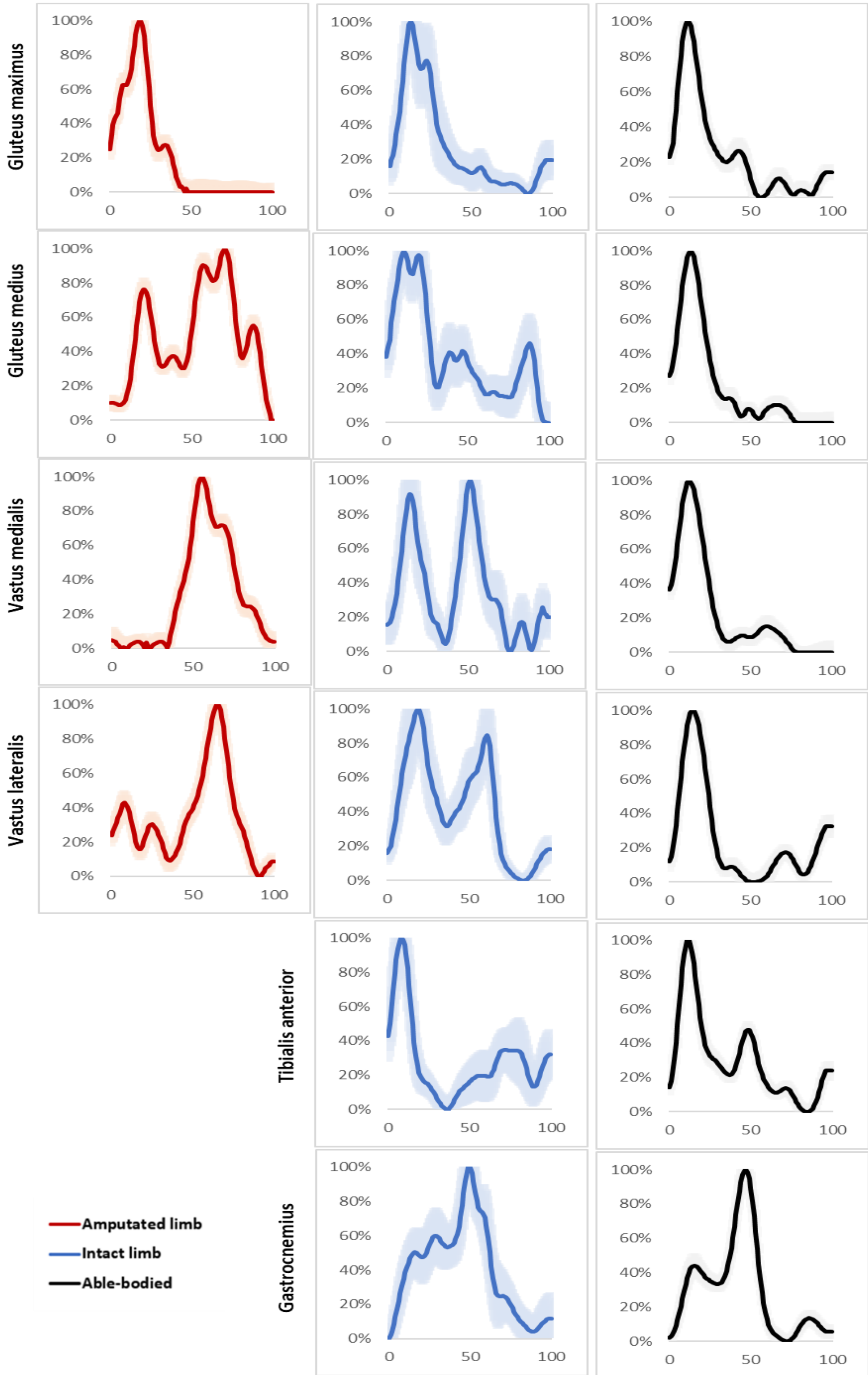

Figure 0.6 Muscle activity for each group.

Table C.0.9 GRF p-value

GRF_peaks	Control VS intact (p_value)		Control VS amputated (p_value)		Intact VS amputated (p_value)	
Mediolateral						
1st_peak	0.13	0.25	0.01	0.06	0.00	0.09
2nd_peak	0.01	0.37	0.08	0.02	0.13	0.03
3rd_peak	0.01	0.37	0.09	0.11	0.43	0.17
Vertical						
1st_peak	0.46	0.28	0.38	0.03	0.31	0.05
2nd_peak	0.00	0.17	0.01	0.14	0.28	0.08
Anteroposterior						
1st_peak	0.45	0.41	0.00	0.07	0.02	0.11
2nd_peak	0.08	0.06	0.00	0.17	0.00	0.05

Table C.0.10 BPI outcomes to the Gait spatiotemporal and GRF for each amputee

Pain outcome measure	TTA_1	TTA_2	TTA_3	TTA_4	TTA_5	TTA_6
Time since amputation	33	43	13	50	94	25
Pain presence	Y	N	Y	Y	Y	Y
Pain reason	PLP	NA	LBP&PLP	PLP	LBP	PLP
Worst	4	0	6	5	6	5
Least	1	0	2	1	2	1
Average	2.5	0	4	3	4	2
Now	1	0	2	2	2	1
Activity	2	0	5	3	3	3
Mood	2	0	2	2	3	2
Walking ability	3	0	3	3	2	3
Work	2	0	3	3	2	2
Sleep	3	0	1	2	2	1
Lifestyle	2	0	1	1	1	2
Spatiotemporal						
Speed (m/s)	0.14	0.13	0.18	0.17	0.16	0.15
Stride Width (m)	0.04	-0.02	0.08	0.04	0.05	0.06
Cadence (steps/sec)	-0.13 / -0.18	-0.04 / -0.17	-0.28 / -0.30	-0.14 / -0.15	-0.12 / -0.16	-0.16 / -0.19
Step Length (m)	-0.02 / 0.00	-0.13 / -0.11	0.08 / 0.11	0.04 / 0.08	-0.12 / 0.10	0.01 / 0.03
Stride Length (m)	-0.03 / -0.01	-0.27 / -0.20	0.22 / 0.36	0.12 / 0.18	-0.22 / 0.21	0.06 / 0.14

Stance Time (s)	-0.04 / -0.08	-0.07 / -0.09	0.16 / 0.00	0.01 / -0.01	0.00 / -0.03	0.03 / -0.01
GRF						
ML _Peak_1	-0.13 / -0.18	-0.04 / -0.17	-0.28 / -0.30	-0.14 / -0.15	-0.12 / -0.16	-0.16 / -0.19
ML Peak_2	-0.02 / 0.00	-0.13 / -0.11	0.08 / 0.11	0.04 / 0.08	-0.12 / 0.10	0.01 / 0.03
ML Peak_3	-0.03 / -0.01	-0.27 / -0.20	0.22 / 0.36	0.12 / 0.18	-0.22 / 0.21	0.06 / 0.14
Vertical Peak 1	-0.04 / -0.08	-0.07 / -0.09	0.16 / 0.00	0.01 / -0.01	0.00 / -0.03	0.03 / -0.01
Vertical Peak 2	0.00 / 0.02	0.01 / 0.02	-0.03 / 0.02	-0.01 / 0.01	-0.02 / 0.01	-0.01 / 0.01
AP Peak 1	0.01 / 0.02	0.07 / 0.04	0.02 / 0.03	0.02 / 0.01	0.03 / -0.03	0.04 / 0.03
AP Peak 2	0.01 / -0.01	0.00 / -0.01	0.02 / 0.04	0.02 / 0.01	0.02 / 0.03	0.02 / 0.02

Note:
(-): TTA smaller than AB
A/B: intact side/amputated side
PLP: phantom limb pain
LBP: low back pain
ML: mediolateral direction.
AP: anteroposterior direction.

Appendix D

Modelling supplementary data

Scaling

In OpenSim, evaluating the accuracy of a scaled musculoskeletal model involves assessing various error metrics, including total error, root mean square (RMS) error, and maximum error. These metrics help ensure that the model accurately represents the subject's anatomy and kinematics. Here are some best practices and recommendations for managing these errors:

Total Error

Total error in OpenSim scaling refers to the cumulative discrepancy between the positions of experimental markers on the subject and the corresponding virtual markers on the model. A low total error indicates that the model closely matches the subject’s physical dimensions, which is essential for accurate biomechanical analysis. The total error is minimized by fine-tuning marker placements and scale factors during the scaling process.

Root Mean Square (RMS) Error

RMS error is a critical metric used to quantify the average deviation between the model and experimental data. It is calculated as the square root of the mean of the squared differences between the positions of corresponding markers. OpenSim recommends keeping the RMS error as low as possible to ensure high fidelity in the model's kinematic predictions. Typically, an RMS error below 2 cm is considered acceptable for high-quality models. This threshold ensures that the discrepancies between the virtual and experimental markers are minimal, allowing for accurate simulation of movement dynamics.

Maximum Error

Maximum error represents the largest single deviation between any pair of corresponding markers on the model and the subject. This metric is crucial because it highlights the worst-case discrepancy, which can indicate potential issues with specific markers or segments. OpenSim users are advised to keep the maximum error as low as possible, ideally below 4 cm, to ensure that even the largest deviations do not significantly impact the overall accuracy of the model.

The static pose scaling trajectory to reflective markers error for each model (eight AB and six TTA) in centimetre presented in the (TableD.10).

Table D.0.11 Static scaling trajectory to reflector markers error during model scaling

Scale outcome/Model	Total square error	Root mean square	Max error	Marker
Able_bodied 1	0.02	0.02	0.04	LANK
Able_bodied 2	0.03	0.02	0.04	LANK
Able_bodied 3	0.00	0.00	0.02	LT2
Able_bodied 4	0.01	0.01	0.02	RANK
Able_bodied 5	0.02	0.01	0.04	RANK
Able_bodied 6	0.01	0.01	0.04	RFEL
Able_bodied 7	0.02	0.02	0.04	RANK
Able_bodied 8	0.01	0.02	0.02	LANK
TTA_1	0.00	0.02	0.02	RFME
TTA_2	0.00	0.01	0.03	RT1
TTA_3	0.01	0.01	0.04	RTAM
TTA_4	0.01	0.02	0.04	RFME
TTA_5	0.01	0.01	0.03	RANK
TTA_6	0.01	0.01	0.03	RANK

Inverse kinematics

Each participant joint angle during gait cycle in degrees (TableD.11). The detailed statistical analysis p value in the (Table D.12).

Table D.0.12 Joint ROM

Parameter		Pelvis			Hip			Knee	Ankle
	#	list	tilt	rotation	flexion	adduction	rotation	flexion	dorsiflexion
Intact limb	1	2 02	2 42	12 49	40 40	16 05	19 04	56 57	23 76
	2	3 93	6 43	10 17	34 30	9 32	9 88	58 35	26 35
	3	9 89	9 87	4 24	46 28	12 43	18 99	49 15	38 11
	4	3 52	5 25	7 85	37 12	19 59	7 57	54 13	23 59
	5	3 78	3 13	11 62	54 19	14 81	17 93	52 91	33 60
	6	7 64	7 37	4 94	31 88	11 67	7 65	35 28	20 72
Amputated limb	1	[redacted]			37 56	9 88	12 72	66 62	14 33
	2				35 39	9 72	9 55	54 31	21 19
	3				37 89	10 69	18 85	42 08	20 24
	4				46 75	19 23	10 40	31 96	11 59
	5				54 93	12 79	25 21	57 97	16 60
	6				28 34	11 81	11 40	37 97	9 97
AB	1	4 10	6 05	10 59	39 79	15 66	8 84	58 19	24 07
	2	4 95	19 06	15 21	44 34	26 33	7 72	50 33	34 37
	3	3 59	7 38	18 59	38 53	17 39	10 15	61 59	23 69
	4	2 93	21 88	15 03	41 75	27 35	9 74	55 97	36 14
	5	4 11	6 57	20 65	43 77	9 91	16 18	57 97	19 32
	6	3 81	21 85	11 00	46 38	26 55	9 96	66 99	30 42
	7	4 68	18 38	22 05	38 75	24 96	14 76	54 15	23 43
	8	5 71	25 68	6 74	50 41	34 17	14 17	64 61	32 33

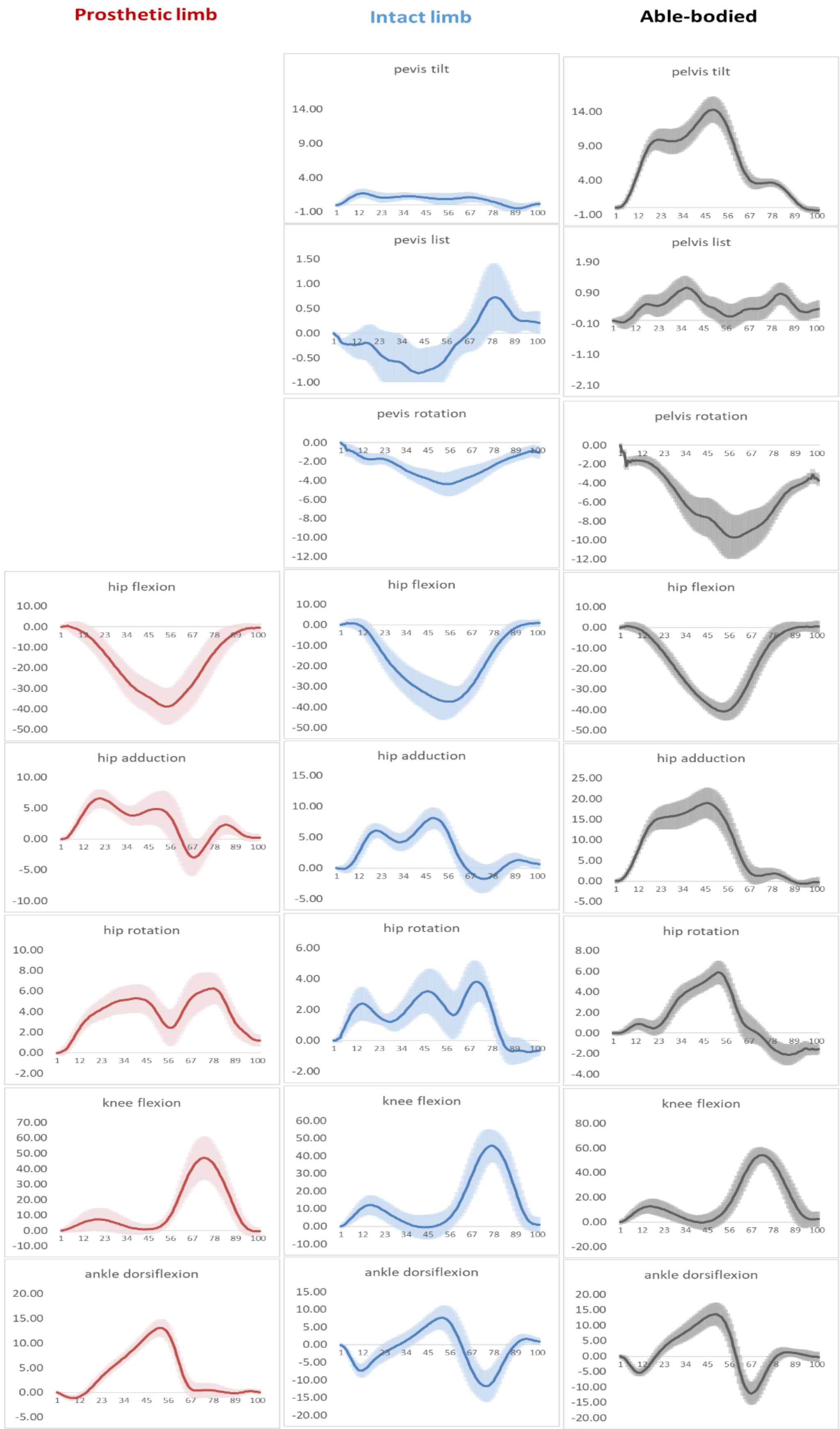

Figure 0.7 Joint ROM for each group.

Table D.0.13 P value

Group		Intact VS AB	Amputated VS AB	Intact VS Amputated
Pelvis	Pelvis_list	0.43		0.53
	Pelvis_tilt	0.01		0.35
	Pelvis_rotation	0.02		0.45
Hip	Hip flexion	0.51	0.46	0.83
	Hip_adduction	0.03	0.01	0.16
	Hip_rotation	0.40	0.22	0.56
Knee &ankle	Knee_flexion	0.06	0.03	0.60
	Ankle dosiflexion	0.94	0.00	0.00

Inverse Dynamics

Each participant joint angle during gait cycle in Nm/(kg*9.8) (Figure D.8; Table D.13). The detailed p value in the (Table D.14).

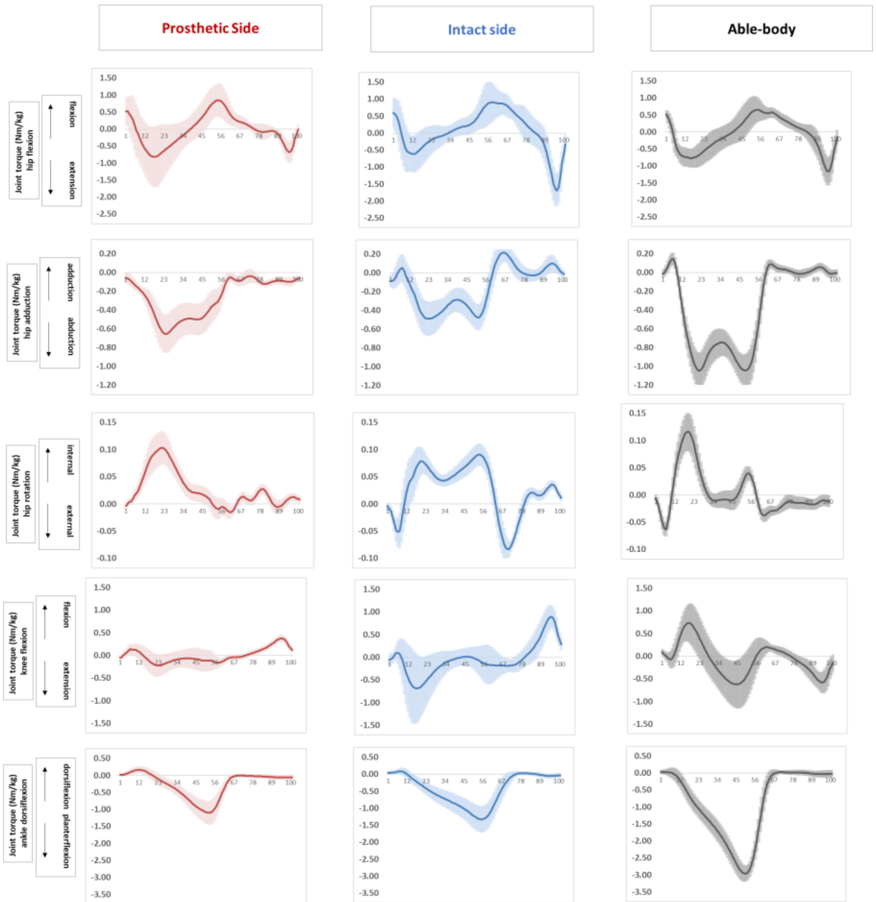

Figure 0.8 Joint torque for each group

Table D.0.14 Joint torque for each participant

Parameter		**Hip**			**Knee**	**Ankle**
	#	flexion	adduction	rotation	flexion	dorsiflexion
Intact limb	1	3.16	1.66	0.24	1.82	2.14
	2	2.80	1.24	0.27	1.88	2.18
	3	3.41	1.31	0.59	2.38	2.84
	4	2.16	1.43	0.48	2.71	1.96
	5	3.88	1.47	0.37	1.25	0.14
	6	2.08	0.36	0.10	0.80	0.06
Amputated limb	1	2.00	0.72	0.18	0.91	2.72
	2	3.06	1.47	0.35	0.85	2.32
	3	1.97	0.83	0.13	0.87	1.49
	4	2.02	1.37	0.40	2.01	1.57
	5	2.81	0.53	0.14	0.76	0.04
	6	1.06	0.42	0.05	0.26	0.01
AB	1	3.21	1.39	0.21	1.70	2.86
	2	1.88	1.20	0.37	1.82	2.71
	3	2.52	2.01	0.38	1.81	3.07
	4	1.78	1.37	0.16	1.42	2.82
	5	2.10	0.82	0.20	1.43	3.68
	6	1.90	1.01	0.13	1.53	3.24
	7	1.73	1.13	0.16	1.59	3.60
	8	1.63	2.35	0.56	1.85	3.13

Table D.0.15 Joint torque p value

Parameter	Intact VS AB	Amputated VS AB	Intact VS amputated
Hip flexion (Nm/weight)	**0.03**	0.20	0.04
Hip adduction (Nm/weight)	0.54	0.07	0.15
Hip rotation (Nm/weight)	0.43	0.46	0.15
Knee flexion (Nm/weight)	0.53	**0.01**	**0.00**
Ankle dorsiflexion (Nm/weight)	**0.00**	**0.00**	0.50

Muscle Force

The table D.16 presents the muscle forces for various muscles in six amputees (Amp_1 to Amp_6) compared to able-bodied (AB) individuals and intact limbs of amputees (Int). The muscle forces are represented as a percentage of body weight (Nm/kg*9.8).

Amp_1: This participant exhibited the highest muscle forces among the amputees, particularly in the iliopsoas (1.01), biceps femoris (0.67), and gluteus maximus (0.04). The high activation of these muscles suggests that Amp_1 relies heavily on these muscle groups for hip flexion and knee stability during gait. However, the activation of muscles like the semitendinosus and tibialis anterior was relatively low, indicating a possible compensation mechanism or muscle imbalance.

Amp_2: Similar to Amp_1, Amp_2 showed elevated forces in the iliopsoas (0.69) and biceps femoris (0.29), with slightly lower values than Amp_1. The gluteus medius (0.39) was more active in this participant, possibly contributing to better hip abduction stability. However, the overall lower force in the gastrocnemius (0.05) may suggest reduced push-off strength during the stance phase of gait.

Amp_3: Amp_3 exhibited muscle force patterns similar to Amp_1, with high activation in the iliopsoas (0.96) and biceps femoris (0.64). The semitendinosus and gluteus maximus forces were consistent with those of Amp_1, indicating a similar strategy for hip and knee movement. The lower activation of the tibialis anterior (0.16) might reflect a reduced ability to control dorsiflexion during the swing phase.

Amp_4: This participant had relatively balanced muscle forces across most muscles, with iliopsoas (0.50) and gluteus medius (0.52) showing notable activation. The forces in the biceps femoris and semitendinosus were lower compared to Amp_1 and Amp_3, suggesting a different compensatory strategy or possibly a better adaptation to the prosthesis, requiring less reliance on these muscles.

Amp_5: The muscle force profile for Amp_5 was similar to Amp_4, with moderate activation of the iliopsoas (0.50) and biceps femoris (0.16). Notably, the gluteus medius showed higher activation (0.58), which might indicate a strong compensatory mechanism for hip stability. The gastrocnemius (0.19) and tibialis anterior (0.08) forces were low, reflecting limited push-off and dorsiflexion control.

Amp_6: Amp_6 exhibited the most balanced muscle forces among the amputees, with the iliopsoas (0.53) and gluteus medius (0.58) showing significant activation. The forces in the biceps femoris and semitendinosus were similar to those in Amp_4 and Amp_5, suggesting a consistent strategy for knee and hip control. The tibialis anterior (0.08) had the lowest force, indicating potential challenges in controlling foot drop during gait.

Table D.0.16 Muscle force for each amputee

BW (Nm/kg*9.8)	Iliopsoas	Biceps femoris	Semitendinosus	Gluteus maximus	Gluteus medius	Adductor magnus	sartorius	Vastus medialis	Vastus lateralis	Soleus	Gastrocnemius	Tibialis anterior
AB_1	0.26	0.10	0.03	0.02	0.07	0.01	0.02	0.03	0.04	0.08	0.04	0.20
AB_2	0.10	0.08	0.01	0.11	0.26	0.01	0.05	0.17	0.37	1.38	0.85	0.18
AB_3	0.41	0.10	0.04	0.02	0.60	0.02	0.15	0.15	0.32	1.62	0.22	0.10
AB_4	0.32	0.06	0.03	0.03	0.58	0.01	0.08	0.11	0.23	1.27	1.20	0.47
AB_5	0.37	0.15	0.04	0.06	0.26	0.01	0.04	0.06	0.13	2.42	0.98	0.41
AB_6	0.29	0.18	0.04	0.12	0.29	0.01	0.03	0.07	0.13	1.49	0.81	0.34
AB_7	0.24	0.11	0.02	0.09	0.29	0.01	0.04	0.09	0.19	1.71	0.78	0.18
AB_8	0.41	0.15	0.04	0.20	0.76	0.01	0.10	0.07	0.12	0.82	0.12	0.14
Int_1	0.40	0.13	0.04	0.07	0.67	0.01	0.15	0.07	0.18	1.23	0.04	0.17
Int_2	0.52	0.12	0.03	0.03	0.36	0.02	0.11	0.09	0.24	0.87	0.04	0.24
Int_3	0.39	0.09	0.03	0.06	0.42	0.02	0.10	0.06	0.14	1.13	0.04	0.21
Int_4	0.30	0.10	0.03	0.12	0.21	0.03	0.05	0.17	0.52	0.89	0.06	0.30
Int_5	0.29	0.10	0.03	0.10	0.19	0.03	0.05	0.17	0.53	0.82	0.06	0.34
Int_6	0.29	0.10	0.03	0.12	0.18	0.03	0.05	0.16	0.48	0.78	0.05	0.29
Amp_1	1.01	0.67	0.04	0.02	0.19	0.01	0.14	0.05	0.11			
Amp_2	0.69	0.29	0.03	0.05	0.39	0.01	0.17	0.05	0.11			
Amp_3	0.96	0.64	0.04	0.02	0.21	0.01	0.12	0.05	0.10			
Amp_4	0.50	0.16	0.07	0.10	0.52	0.02	0.10	0.07	0.16			
Amp_5	0.50	0.16	0.08	0.11	0.58	0.02	0.10	0.08	0.19			
Amp_6	0.53	0.16	0.08	0.12	0.58	0.02	0.10	0.08	0.20			