Biomechanical quantification and analysis of the human foot and ankle joint complex

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"No one who achieves success does so without acknowledging the help of others. The wise and confident acknowledge this help with gratitude." – Alfred North Whitehead

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ABSTRACT

Title: Biomechanical quantification and analysis of the human foot and ankle joint complex

Keywords: Biomechanics, gait analysis, sensor technologies, wearable device, 3D motion analysis

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The human foot and ankle joint complex are essential lower limb components responsible for facilitating smooth gait and other activities related to everyday living. Unfortunately, this natural movement is lost in amputees and difficult to replicate even with the aid of prosthetic devices. Thus, to contribute to the understanding of the fundamental characteristics relative to natural lower limb movement, this study aims to quantify and analyse the biomechanical features of the human foot and ankle joint complex in relation to the sagittal and frontal planes of the foot.

The research design and methodology comprise an experimental process wherein 15 healthy young to middle-aged individuals were recruited. The participants were required to navigate a set of five low-level activities, particularly 1) level walk, 2) step up and down, 3) 20° slope incline and decline walk, 4) standing and sitting, and 5) lifting an object. The participants were equipped with a sensor platform device consisting of three ADXL 345 sensors, strategically positioned to restrict kinematic data to the foot and ankle. Notably, data from the sensors were collected in conjunction with reference data from a motion capture camera system to ensure the integrity of the sensors. In terms of kinetics, force plates were used to obtain the reaction forces acting on participants during the experimental activities.

It was observed that a loss of data, which primarily occurred due to poorly optimised code of the interface application, significantly decreased the accuracy of the data collected from the sensor platform. Nonetheless, a comparative analysis of the findings showed similar patterns between the experimental results and the reference data, thus validating the research design and method used. Therefore, suggestions could be made to enhance data accuracy and generalisability in future research. For the kinetic aspect of this study, it was suggested that a modular force plate system be used to investigate the reaction forces of supplementary experimental activities. In terms of the kinematic data, accuracy could be enhanced by recruiting a larger sample size, optimising the code behind the application interface, and considering the transverse plane of the foot in addition to the sagittal and frontal plane.

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LIST OF ABBREVIATIONS

AMMA ~ Ankle motion monitoring application

CoP ~ Centre of pressure

FS ~ Full-scale

GRF ~ Ground reaction force

HHMM ~ Hierarchal hidden Markov model

HMM ~ Hidden Markov model

HREC ~ Health research ethics committee

IMU ~ Inertial measurement unit

MEMS ~ Micro-machined electromechanical system

3D ~ Three-dimensional

ROM ~ Range of motion

CHAPTER 1

1. INTRODUCTION

1.1 Background

The human lower limb's ability to interact with the ground is made possible by the kinetic linkage formed by the foot and ankle joint complex. These critical components facilitate a natural gait and various activities integral to daily living (Brockett & Chapman, 2016). Unfortunately, this natural movement is frequently lost in lower limb amputees, who must rely on alternate muscle groups to enable a smooth gait pattern. Moreover, this compensation may give rise to further psychological complications and physical injuries (Van Tuan, 2017).

With the above in mind, this study acknowledges the importance of a healthy and functional foot and ankle joint complex. As such, it intends to employ various technologies and systems to quantify and analyse the fundamental biomechanical characteristics of these components during gait and other activities related to everyday living. Through this, the study aims to obtain relevant and invaluable information that may further be utilised to develop an ankle-foot prosthesis to facilitate a smoother gait pattern for lower limb amputees. Although, it should be noted that the actual design of a prosthesis is outside the scope of the present study. Instead, the sole purpose of this research project is to identify the key biomechanical principles and functions of the human gait pattern using appropriate gait quantification systems, sensor technologies, and analysis techniques.

1.1.1 The human gait cycle

Human gait is a complex biomechanical process that involves a sequence of alternating movements of the lower limbs in a periodic motion that results in the forward progression of the human body (Price *et al.*, 2021). To analyse an individual's walking pattern, it is crucial to identify the shortest, most distinct, and reproducible movement in gait. This movement is referred to as the gait cycle (ProtoKinetics, 2018). Furthermore, it can be suggested that a comprehensive understanding of the human gait cycle is vital in the development of interventions to enhance mobility, such as an ankle-foot prosthesis device. The ensuing sections examine the particulars of a natural gait pattern, as well as variations observed in lower limb amputees during gait.

1.1.1.1 Natural gait cycle

According to Price *et al.* (2021), a complete gait cycle comprises two distinct phases: the stance phase, which accounts for approximately 60% of one cycle, and the swing phase, which makes up the remaining 40%. To further differentiate between the specific movements within the cycle, these two phases can be subdivided into eight sub-phases, as defined by Carollo and Matthews (2015). Figure 1.1 illustrates the gait cycle concept by demonstrating the human anatomical structure throughout the eight sub-phases of gait.

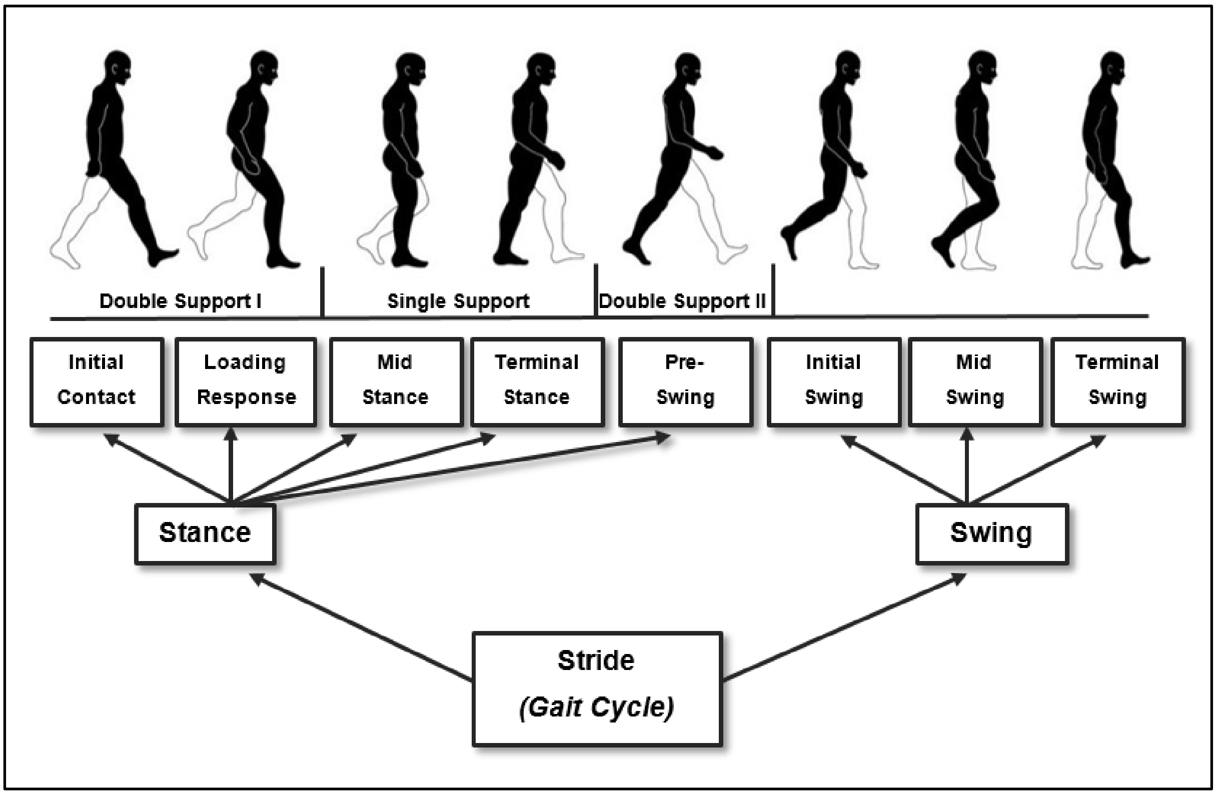


Figure 1.1: Visual representation of the sub-phases of the human gait cycle (Stöckel *et al.*, 2015)

To elaborate on the above illustrated sub-phases, Table 1.1 provides a detailed description of the gait cycle concept by characterising the eight sub-phases of gait according to Carollo and Matthews (2015).

Table 1.1: Sub-phases of the human gait cycle

	Sub-phase	Description
Stance phase	Initial contact	This phase represents the instant the foot strikes the ground.
	Loading response	This phase represents the initial double support period when the limb accepts the body weight.
	Mid-stance	This marks the initial stage of single support, during which the body progresses beyond the stance phase and ends in front of the stance limb as body weight is conveyed.
	Terminal stance	This is the final phase of single support and ends with opposite initial contact.
	Pre-swing	This phase represents the final double support period when the knee flexes in preparation for the swing phase. In this phase, body weight is shifted to the opposite limb.
Swing phase	Initial swing	This phase represents the first swing period where maximum knee flexion occurs.
	Mid-swing	This phase represents the swing period where maximum hip flexion occurs and ends with a vertical tibia.
	Terminal swing	This is the final phase in the swing period, where the extended knee achieves maximum step length, and the limb is properly positioned for weight acceptance.

Gait patterns may differ among individuals due to factors such as age, weight, height, strength, sex, flexibility, or walking speed (Grujicic, 2022). By studying a diverse sample group, more generalised gait pattern findings may be obtained, thereby providing a more extensive understanding of gait movements.

1.1.1.2 Gait deviations in lower limb amputees

The application of prosthetic devices in the restoration of mobility for lower limb amputees is a commonplace practice. However, despite advancements in the design of lower limb prosthetics, a natural gait is challenging to replicate and may prompt amputees to engage alternative muscle groups (Di Gregorio and Vocenas, 2021). This adaptive strategy often results in an increase in metabolic oxygen consumption and up to 100% more energy requirement, depending on the type of lower limb amputation (Kirshner, 2023). Furthermore, gait deviations and abnormalities may lead to secondary complications such as muscle weakness, contracture, pain, diminished confidence in the prosthetic device, or persistent irregular gait behaviours (Van Tuan, 2017).

In the case of transtibial (below-knee) amputations, it is observed that commonly used ankle prostheses display a limited range of motion (ROM) when compared to their physiological counterparts. Consequently, this results in prolonged heel strike and weight bearing on the residual limb during the swing phase. Therefore, to attain optimal stride length with the non-prosthetic limb, heel rise occurs earlier in the cycle and is greater than that of a standard gait pattern. This leads to a greater elevation in the amputee’s body and a consequent surge in loading force borne by the non-prosthetic limb, thus, requiring increased quadriceps contraction to absorb the impact (Smith *et al.*, 2004). Additionally, during the stance phase, the energy generated by the prosthetic limb is reduced by half compared to that of the typical limb, necessitating a greater energy consumption in the muscles above the residual limb (Kirshner, 2023).

When considering both transtibial and transfemoral (above-knee) amputations, Varrecchia *et al*. (2019) states that amputees exhibit an irregular gait pattern compared to individuals of the same age without amputation. Moreover, this observation highlights that both levels of amputation display a common yet specific gait pattern, with transfemoral amputees exhibiting a more asymmetric pattern. Notably, the gait performance and adaption of these amputees are highly dependent on the level of amputation and the type of prosthesis employed. It was also found that lower limb amputees may demonstrate an abnormal trunk (torso) motion during functional tasks, however the underlying mechanisms of this are not well understood (Yoder *et al*., 2019). This noticeable increase in trunk motion is closely associated with an elevated risk of falling and may lead to additional physical harm.

In conclusion, gait deviations or abnormalities in lower limb amputees may significantly impact their mobility and increase the risk of both physical and psychological complications. Further research on the cause-effect relationship of these deviations and the development of improved prosthetic devices may help to improve the quality of life for individuals with lower limb amputations.

1.1.2 Biomechanics of anatomical components during gait

The complex structure of the lower limb includes multiple muscles and articulations responsible for providing stability, absorbing force, and propelling motion during the gait cycle (McKeon & Hoch, 2019). Understanding the biomechanics of these components is crucial for evaluating and treating gait abnormalities and lower limb pathologies. The ensuing sections examine the components (more specifically, the articulations and muscles) responsible for gait movement, as well as kinetic and kinematic attributes of the foot and ankle complex throughout the gait cycle.

1.1.2.1 Ankle-foot bone structure and articulations

The anatomical structure of a functional human ankle joint complex is composed of three key articulations that enable the mobility of the foot (Brockett & Chapman, 2016), and are essential for force absorption and propulsion throughout the ankle complex (McKeon & Hoch, 2019). These triform articulations include the tibiotalar, subtalar and talonavicular joints as highlighted in Figure 1.2 below.

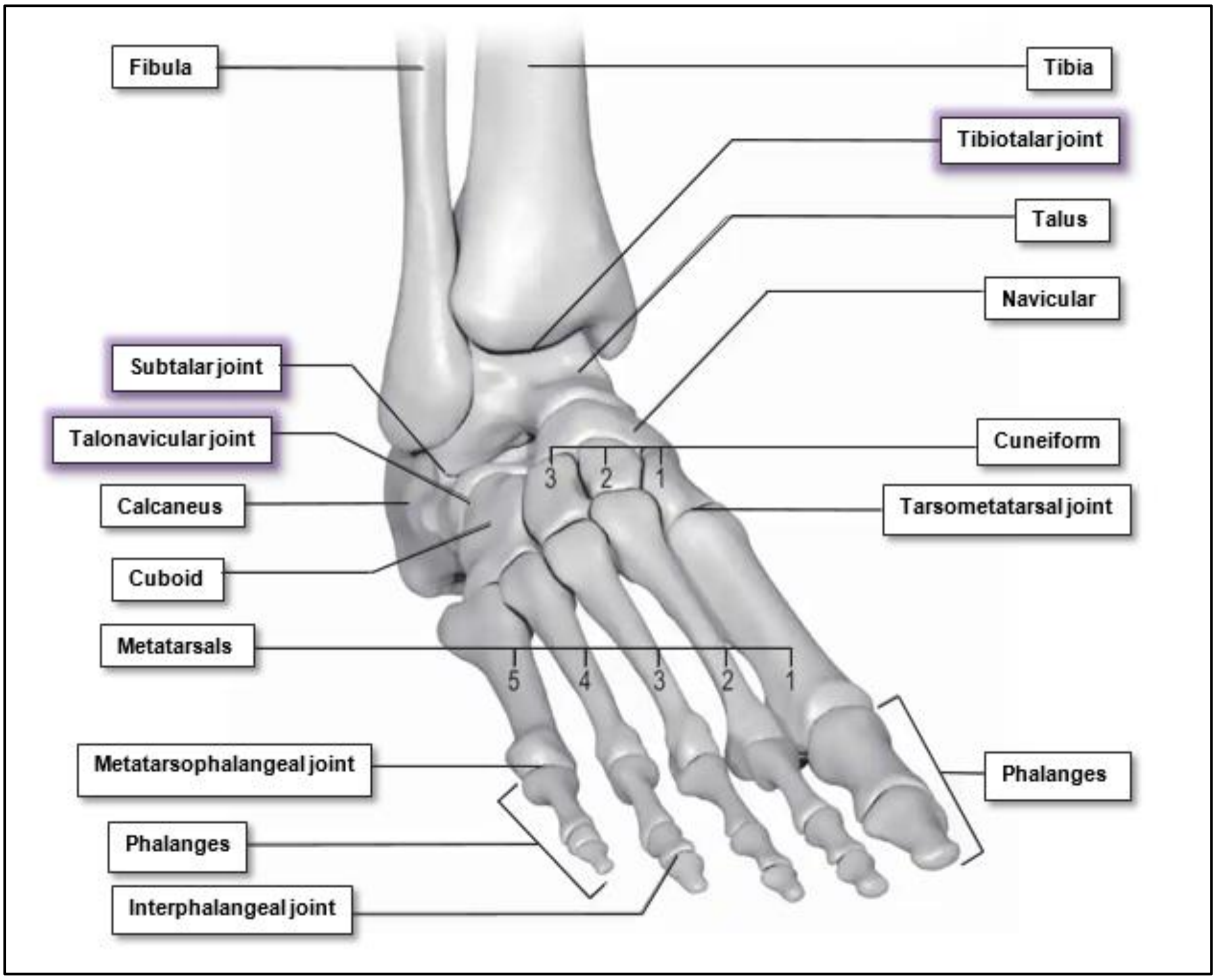


Figure 1.2: Ankle-foot bone structure and articulations (Musculoskeletal Key, 2018)

The subsequent statements expound on the articulations highlighted in the above diagram.

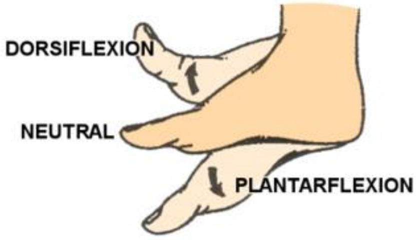
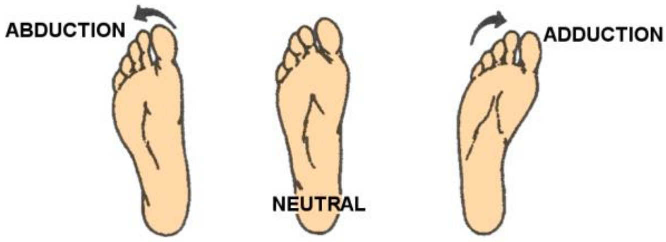
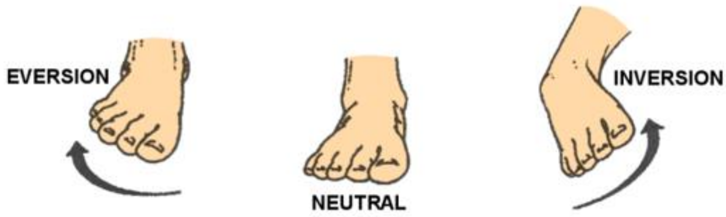
- The **tibiotalar joint** is a synovial joint that primarily serves as a hinge and enables the flexion movement of the foot (Brockett & Chapman, 2016). In looking at the bony structure of the human foot and ankle, it is seen that this joint superiorly includes the distal tibia and fibula, and the talus inferiorly, with the ends of the tibia and fibula forming an ankle mortise. The mortise shapes the proximal segment of the tibiotalar joint, providing the tibiotalar joint and lower leg with significant stability (McKeon & Hoch, 2019).
- The intricate structure of the **subtalar joint** provides the basis for foot and ankle inversion-eversion motions (Brockett & Chapman, 2016). This joint is superiorly formed from the calcaneus, and inferiorly from the talus, with the latter resting on the frontal portion of the calcaneus (Angin & Simsek, 2020).
- The **talonavicular joint** is a functional unit that contributes to the inversion and eversion movements of the foot and ankle and shares a common axis with the subtalar joint. This joint links the talus and navicular bony structure, where the head of the talus articulates with the posterior part of the navicular (Brockett & Chapman, 2016).

Overall, the complex network of articulations between the bones of the foot and ankle plays a crucial role in the kinematics and kinetics of gait. Furthermore, each joint's unique structure and ROM allows for the necessary movements and stability required for efficient locomotion.

1.1.2.2 Ankle range of motion

The human foot and ankle joint complex, enabled by the triform articulations previously outlined, is capable of a range of six fundamental motions. These include sagittal plane movements of plantarflexion and dorsiflexion, transverse plane movements of abduction and adduction, and frontal plane movements of inversion and eversion (Alvarez-Perez *et al.*, 2019). The movements and planes are elucidated in Table 1.2 below.

Table 1.2: Planes of motion

Plane	*Movement*	*Representation*
Sagittal	Plantarflexion & Dorsiflexion	
Transverse	Abduction & Adduction	
Frontal	Inversion & Eversion	

Figures from Orthquake (2020)

These motions are vital to the ankle's ROM, which is the term used to describe a joint's ability to progress through its complete spectrum of movements (Sears, 2023). In the context of gait, the foot and ankle joint complex is primarily responsible for sagittal plane motion, specifically the plantar- and dorsiflexion movements at the tibiotalar joint. The overall ROM for this plane is between 65-75°, with dorsiflexion accounting for about 10-20° and plantarflexion accounting for about 40-55°. This ROM is reduced to 30° during walking and increases to 37° and 56° during ascending and descending stairs, respectively. Additionally, minor motion in the frontal plane is observed, with an overall ROM of approximately 35° (23° inversion – 12° eversion) (Brockett & Chapman, 2016).

In conclusion, understanding the ROM of the ankle is crucial in assessing and addressing gait deviations in lower limb amputees. Furthermore, by considering the ROM of the ankle joint during gait, healthcare professionals can better identify and treat gait irregularities, leading to improved functional outcomes.

1.1.2.3 Muscles of the lower limb

The lower limb muscles are vital in providing stability, support, and movement to the human body. These muscles work together in a complex manner to facilitate various lower limb movements, including walking, running, and jumping. Additionally, lower limb muscles help to maintain posture and balance, especially during weight-bearing activities. These functions are made possible by the elaborate interplay between the muscles, bones, and joints of the human lower limb. According to a study by Ivanenko *et al.* (2004), the coordination and activation of these muscles are dependent on factors including type of movement, intensity, and speed.

In Table 1.3 the lower limb musculature is systematically classified based on their respective functions, specifically regarding plantarflexion and dorsiflexion, and further delineated by their corresponding anatomic points of origin (Walden, 2018).

Table 1.3: Muscles of the leg

	Muscle	*Origin*
Plantarflexion	Gastrocnemius	Originates from the lower posterior surface of the femur above the medial condyle.
	Soleus	Originates from the upper half of the posterior tibia surface.
	Flexor hallucis longus	Originates from the lower two-thirds of the posterior fibula surface.
	Flexor digitorum longus	Originates from the lower two-thirds of the posterior tibia.
	Tibialis posterior	Originates from the posterior surfaces of the tibia and fibula.
	Peroneus Brevis	Originates from the lower two-thirds of the fibula's lateral surface.
	Peroneus longus	Originates from the head of the fibula and the upper two-thirds of this bony structure.
Dorsiflexion	Tibialis anterior	Originates from the upper half of the lateral and anterior surfaces of the tibia.
	Extensor hallucis longus	Originates from two-thirds of the inner surface of the frontal fibula.
	Extensor digitorum longus	Originates from the lateral condyle and anterior surface of the tibia and fibula, respectively.

Although the above muscles are categorised as either plantar- or dorsiflexors, it is imperative to recognise that their functional roles may incorporate additional movements beyond the specified categories. The following section builds on the knowledge of muscular activity by examining the reaction forces on the lower limb during gait.

1.1.2.4 Lower limb reaction forces

According to an investigation by Brunner and Rutz (2013), a pivotal function of muscle activity during gait is to regulate the external moment, which encompasses the impact of gravity or reaction forces throughout movement. Hence, a connection exists between muscular activity and reaction forces, where gait muscles produce a counteractive force against the ground, known as the ground reaction force (GRF). By analysing the timing and magnitude of the GRF, it is possible to gain insight into the activation patterns of the lower limb muscles and their contribution to the overall gait pattern (Winter, 2009). To commence this analysis, an initial examination of the reaction forces at play is required.

When evaluating the leverage ratios of the body, it becomes apparent that the muscles within the lower limb create the maximum forces acting on the bones (Hastings, n.d.).

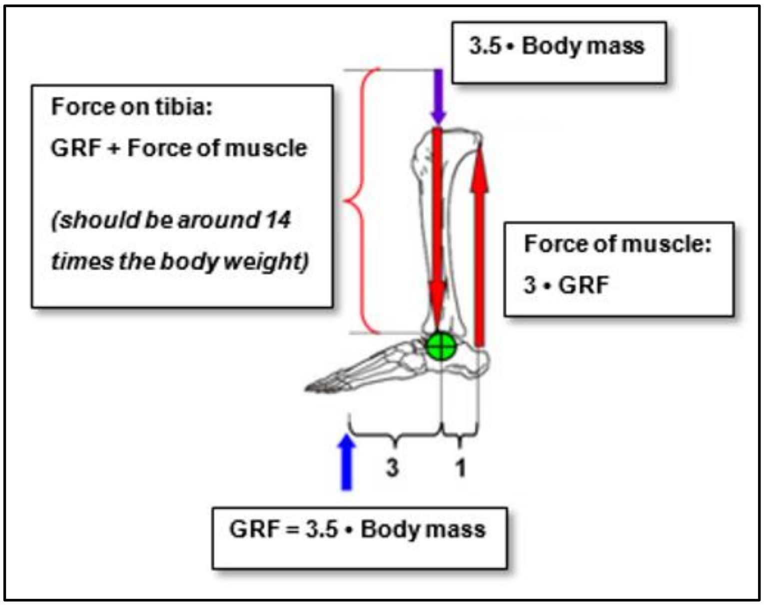


Figure 1.3: Free body diagram of stationary lower limb (Hastings, n.d.)

The illustration in Figure 1.3 reveals the ankle-to-Achilles tendon distance ratio to be roughly 3:1. Moreover, a force equivalent to 3.5 times the person's weight is exerted at the forefoot against the ground. Based on this ratio, it can be inferred that the force generated by the muscles is three times the GRF. As per Hastings (n.d.), the total force exerted on the ankle complex by the tibia bone is estimated to be approximately 14 times the individual's total body weight. While these principles do not establish the reaction forces of the ankle joint complex during all sub-phases of gait, they provide valuable insight into the potentially extreme forces that may be exerted on the ankle and foot.

A study by Bonnefoy-Mazure and Armand (2015) found that the reaction forces imposed on the foot and ankle complex during gait are inconsistent and instead exhibit a dynamic pattern contingent upon the limb’s weight distribution. This fluctuation is depicted in Figure 1.4, which encapsulates the changes in reaction forces experienced by the foot on the ground during the stance phase.

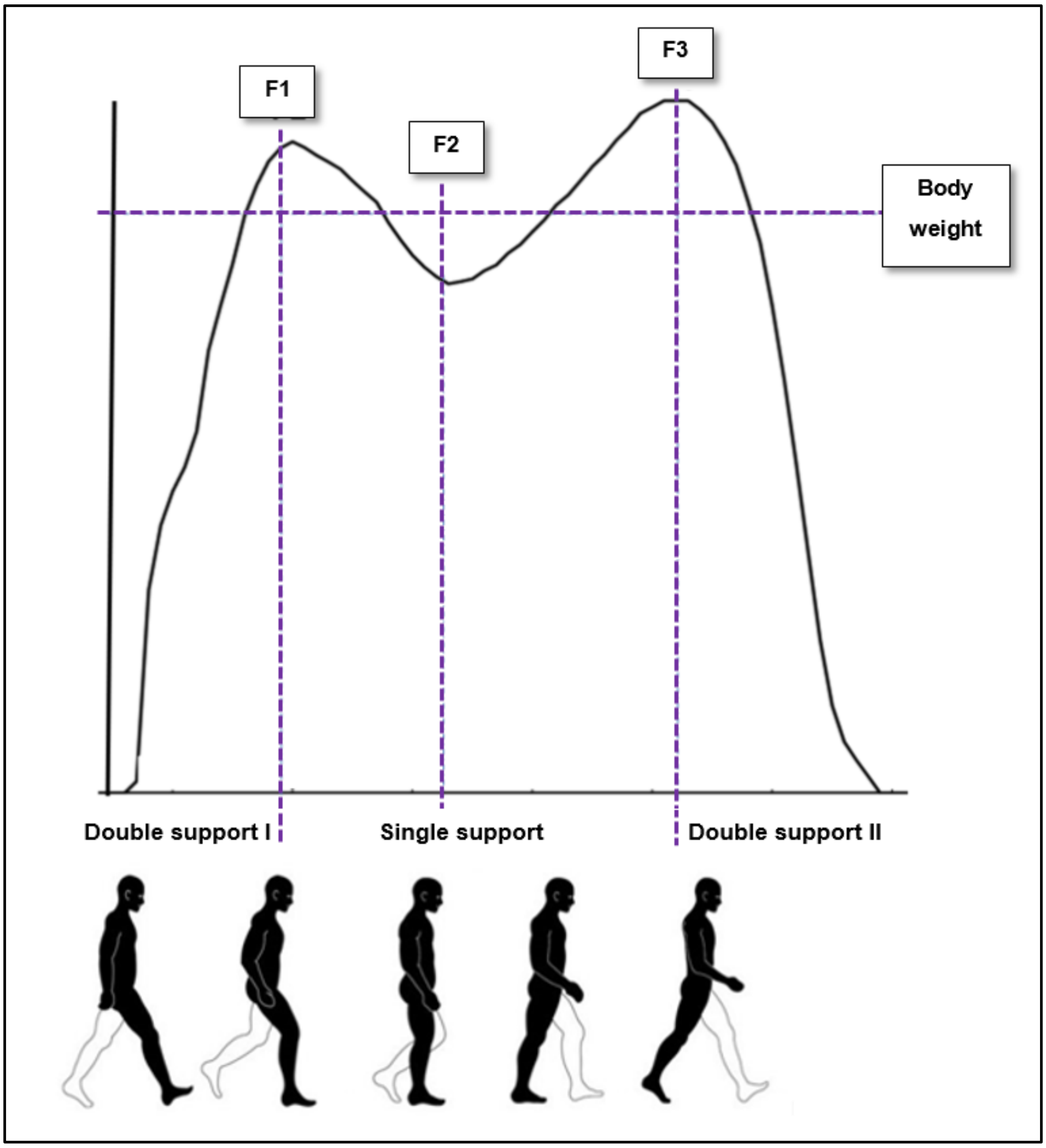


Figure 1.4: Dynamic reaction forces during the stance phase (Bonnefoy-Mazure & Armand, 2015)

Further investigation yielded a study by Brockett and Chapman (2016), which revealed that the ankle joint complex experiences a substantial amount of force, amounting to nearly five times the individual's body weight during the stance phase. This force escalates up to thirteen times the body weight during more rigorous gait activities, such as running. The researchers assert that empirical investigations indicate that the tibiotalar joint is responsible for approximately 83% of the load transmission, which is a function of both ligamentous forces and positional effects of the ankle.

From the preceding discussion, it can be inferred that the ankle complex is exposed to substantial external and internal reaction forces during gait. Moreover, this force is a dynamic entity that varies as body weight is distributed between the lower extremities. Additionally, it would be prudent to assess the reaction forces during a broader range of gait activities, including but not limited to inclined walking. Such evaluations would aid in comprehending the impact of reaction forces on the foot and ankle complex during daily activities.

1.2 Problem statement

The previous discourse establishes the crucial role of the human foot and ankle joint complex as integral lower limb constituents that can withstand considerable dynamic forces and accommodate movement in various directions. The biomechanical functions of these components facilitate natural gait and enable other related activities encountered in daily life, such as running, ascending stairs, and walking up an incline among others.

Despite the advancements in ankle-foot prosthetic devices, lower limb amputees still face challenges in attaining a natural and comfortable gait pattern. The limitations of these devices lead to uneven weight and force distribution throughout the limb, resulting in physical and psychological complications such as muscle strain, contracture, and decreased confidence. These gait abnormalities are largely due to the reduced ROM in the prosthesis, compared to the physiological functions of the human foot and ankle complex. To address this problem, it is imperative to quantify and analyse the specific biomechanical characteristics of the human foot and ankle complex during gait and other daily activities. By doing so, it may be possible to contribute knowledge guided towards the development of a prosthesis that closely replicates a natural gait pattern among amputees. However, the design of such a prosthesis is beyond the scope of the present study.

1.3 Research purpose and objectives

With reference to the problem statement, the primary aim of this study is to quantify and analyse the fundamental biomechanical properties of the foot and ankle complex that are likely to impact the development of a more effective ankle-foot prosthesis design. Therefore, the purpose of this study can be concisely defined as:

- To quantify and analyse the biomechanical functions of the human foot and ankle complex during various gait-related activities.

In achieving this purpose, the secondary objectives are defined as follows:

- To determine the dynamic reaction forces between the lower limb and the ground for different activities related to everyday living.
- To design and develop a wearable sensor platform device and MATLAB® application interface for monitoring the foot and ankle joint complex movements.
- To evaluate the accuracy and precision of the experimental findings by comparing the results to reference data.

1.4 Research design and methodology

To achieve the above stated research purpose and objectives, the study comprises a literary and empirical investigation.

The literature review will examine diverse forms of published research and theoretical texts pertinent to gait analysis. This will include an examination of various technologies commonly used for gait quantification, along with the fundamental principles and techniques associated with the collection and analysis of gait data. The primary aim of this inquiry will be to garner a comprehensive understanding of the subject matter and establish a theoretical basis to facilitate the design and execution of an empirical investigation.

The empirical element of this study will employ an experimental procedure for raw data gathering. The data collection process will be conducted at the PhASRec facility, located at the North-West University Potchefstroom Campus, under the supervision of a qualified biokineticist. To ensure the highest quality of data, a wearable sensor platform will be used in conjunction with laboratory equipment, comprising a Qualysis motion capture camera system and AMTI force plates. Following the setup of the equipment, participants will be required to perform a series of low-level activities experienced in everyday life. Through these activities, a set of defined variables will be obtained for further processing and analysis.

1.5 Delineations

The following points serve to elaborate on the specific delineations made for the present study.

- This study is delimited to the analysis of low-level activities that are commonly encountered in everyday life. This study will therefore not consider high-intensity movements such as running, hiking, or swimming, amongst others.
- To obtain natural gait data, this study will exclusively recruit healthy young to middle-aged individuals. Elderly or injured/diseased participants may exhibit an irregular gait pattern and will not be included in this study.
- This study will focus on the biomechanical principles of the foot and ankle complex. However, it is important to understand that knee and hip movements also play a significant role in facilitating a natural gait.
- The study is delimited to the sagittal and frontal planes of the foot, i.e., only data relating to dorsiflexion/plantarflexion and inversion/eversion will be collected and analysed.

1.6 Significance of research

The research holds significance from both theoretical and practical perspectives.

Theoretically, this research aims to expand on the knowledge of the fundamental biomechanical functions of the human foot and ankle complex that facilitate natural movement. This will be done through background research and a literature review in which various technologies and techniques are investigated. Additionally, from this research, theoretical suggestions guided towards the implementation of a gait study may be made.

In terms of practical significance, this research aims to provide valuable insights and data that could facilitate the development of an ankle-foot prosthesis. Moreover, given that the study employs a practical experimental procedure utilising advanced sensor technologies, it has the potential to offer further significance in the realm of gait analysis research.

1.7 Dissertation outline

The present dissertation is structured in accordance with the research design and methodology. The framework for this research is outlined and elaborated on below:

❖ *Chapter 1: Introduction*

The introductory chapter offers a comprehensive explanation of the study's contextual underpinnings, which is aimed at defining a rational problem statement. Through an investigation of the terminologies and concepts relating to the human ankle and foot biomechanics, this chapter furnishes the reader with essential background information and rationale that forms the groundwork of the study. Additionally, this chapter outlines the research objectives set to ensure the systematic completion of the study. Finally, this chapter incorporates a brief discussion of the research methodology to be utilised in this study, along with an overview of the forthcoming chapters.

❖ *Chapter 2: Literature review*

The literature review chapter provides a detailed exploration of relevant theoretical information necessary for the study by investigating previously published works in the form of articles, book chapters, related websites, or general discussions. Research presented in this chapter includes gait features, quantification systems and technologies (for both kinetic and kinematic quantification), and real-time data analysis techniques. This chapter intends to utilise existing research to identify appropriate technologies, theories, mathematical models, assumptions, limitations, and potential challenges that may be relevant to the current study.

❖ *Chapter 3: Research methodology*

The third chapter builds upon the preceding literature review and describes the research design and methodology reflected upon throughout the study. Subsequently, this chapter serves as a blueprint for defining the selected research design of the empirical study, and includes a discussion on sampling criteria, equipment and materials, experimental protocols, responsibilities, and selection of variables among other things.

❖ *Chapter 4: Results and discussion*

This chapter presents the details of the empirical investigation, which entails the utilisation of a various quantification technologies aimed at acquiring kinetic and kinematic data related to the human foot and ankle complex. Furthermore, the chapter provides the details of the data analysis techniques necessary for inferring interpretations and drawing conclusive outcomes.

❖ *Chapter 5: Conclusion*

The final chapter is used to link the findings of the study to the initially outline purpose and objectives of the research. This is done by providing a concise overview of the research study. Additionally, this chapter highlights challenges encountered and reflects on the research process.

1.8 Chapter summary

The purpose of this chapter was to establish the contextual background and rationalisation for the research problem and method. From the background research it was found that the human foot and ankle complex are vital lower limb components necessitated for a natural gait and other related movements. Nevertheless, those with ankle-foot prostheses face considerable difficulties in replicating smooth gait, which may result in further complications. Therefore, the overall purpose of this study is to quantify and analyse the fundamental biomechanical functions of the human foot and ankle during various activities related to daily life. Through this, it may be possible to contribute relevant and invaluable information guided towards developing an ankle-foot prosthesis that supports smoother gait movement. Finally, this chapter provides a brief overview of the research method, delineations, significance, and framework for the upcoming chapters.

The following chapter furthers the literary information of the background investigation by delving into the intricate details of advanced gait analysis technologies and techniques.

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CHAPTER 2

2. LITERATURE REVIEW

2.1 Introduction

The previous chapter introduced the study by examining relevant background information to formulate a research problem and purpose. To further the introductory concepts and create a theoretical foundation for the research, this chapter expands on the key notions of the study by investigating various published works and theoretical texts related to sensor-based gait analysis. The main considerations of this chapter are highlighted in Figure 2.1

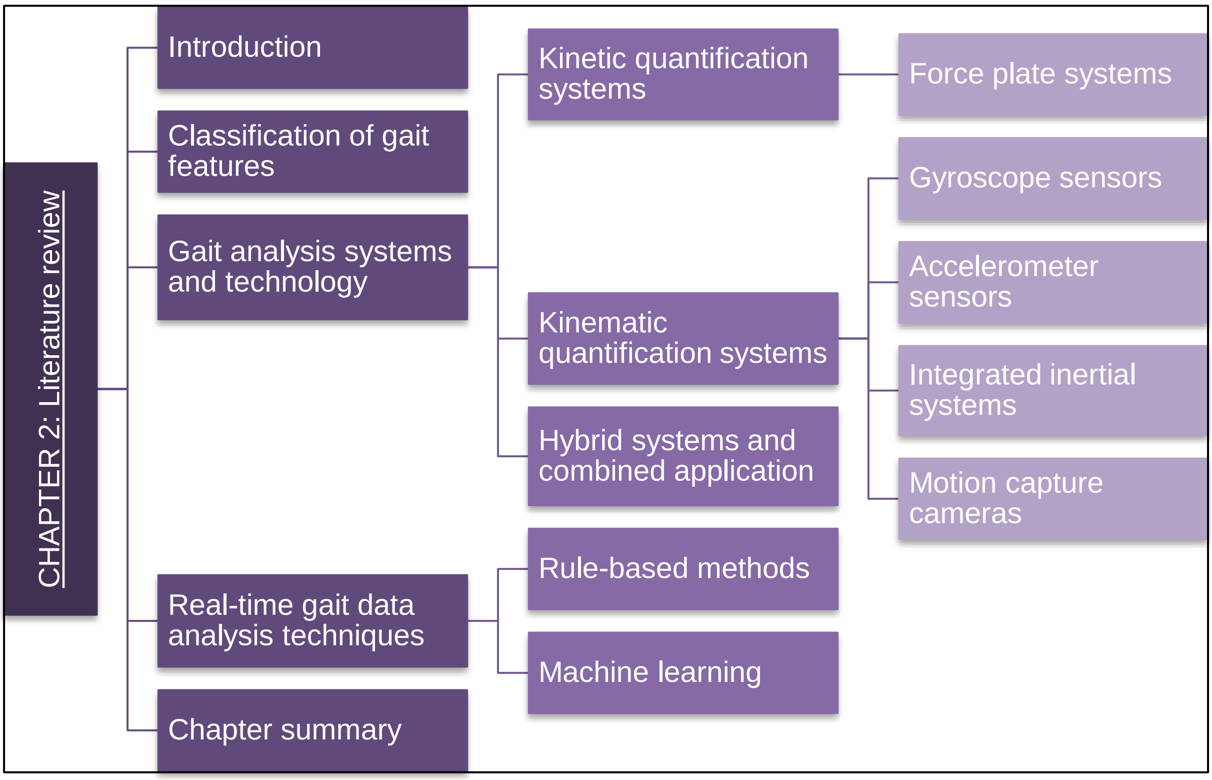

Figure 2.1: Literature review breakdown

In accordance with Figure 2.1, this chapter precedes with an investigation of the key features to be considered in a gait study and identifies prospective systems and technologies for kinetic and kinematic quantification. In terms of kinetic quantification systems, the literature focuses on the use of force plate systems in gait studies. On the other hand, kinematic quantification systems include a range of technologies including gyroscope sensors, accelerometer sensors, integrated inertial units, and motion capture cameras. Subsequently, the review examines the use of hybrid systems and a combined application of multiple technologies. Ultimately, this chapter culminates in the identification of commonly employed real-time data analysis techniques, particularly rule-based methods, and machine learning algorithms.

2.2 Classification of gait features

The intricate interplay of the nervous, cardiorespiratory, and musculoskeletal systems underlies the complex nature of human gait, which is further influenced by an individual's age, personality, mood, and sociocultural factors (Pirker & Katzenschlager, 2016). Characteristics and features within a gait pattern can be divided into three distinct categories, namely: temporal (time-related), spatial (space-related), and spatiotemporal (space-time-related) (Prasanth *et al*., 2021). In continuation of the concepts discussed in the previous chapter, this section examines the gait parameters within these categories, aiming to identify appropriate parameters for a gait analysis study.

Prasnath *et* al. (2021) outlines the common intra- and inter-stride parameters of a gait analysis study, as listed in Table 2.1 below.

Table 2.1: Common features in gait studies

Parameter	*Intra-stride*	*Inter-stride*
Temporal	○ Gait event ○ Gait phase ○ Step duration ○ Swing/stance time frame	○ Stride time frame ○ Cadence
Spatial	○ Step length	○ Stride length
Spatiotemporal	○ Joint angle ○ Joint torque ○ Ground reaction force (GRF) ○ Centre of pressure (CoP)	-

The intention of this table is to identify and categorise prevalent gait features into temporal, spatial, or spatiotemporal parameters, thereby enabling their inclusion in the study’s methodological and empirical components. Notably, features within these categories exhibit both kinetic and kinematic properties and are therefore not restricted to a specific group. Thus, it is imperative to carefully consider the gait analysis systems and technologies capable of kinetic and kinematic quantification.

2.3 Gait analysis systems and technologies

A clinical gait analysis involves the study of the well-balanced distribution of loads in stance and symmetric motion, cadence, and dynamic force sharing and distribution during gait activities. Consequently, gait analysis systems and technologies are proficient in determining the trajectories of joint and limb motion, characterising various levels of gait-related activities, or monitoring athletic performances (Akhtaruzzaman *et al.*, 2016). These systems and technologies are divided into kinetic and kinematic quantification systems, which respectively deal with limb reaction forces and motion.

2.3.1 Kinetic quantification systems

Kinetics is a branch of physics that deals with the analysis of static and dynamic forces or torques with rotational motion, often using free-body diagrams and vector mathematics (Dusto, 2020). In the context of gait analysis, kinetic quantification systems are used to determine dynamic reaction forces between the ground and joints. As such, the primary focus of this section centres on force plate technology.

2.3.1.1 Force plate systems

Force plate mechanisms generally comprise a metallic plate equipped with a load cell positioned at each corner and are frequently used in gait analysis studies to measure the forces exerted by a user during gait activities (Lamkin-Kennard & Popovic, 2019). As a kinetic quantification system, force plates provide the facilities to determine induced forces as well as their directions. However, Akhtaruzzaman *et al.* (2016) emphasise the importance of the combined use of kinematic information to provide a comprehensive understanding of the gait principles. Nonetheless, the integration of force plate systems into gait analysis studies offers a more holistic understanding of gait dynamics.

2.3.1.1.1 Related works

Force plates serve as valuable tools for illustrating motion dynamics and analysing ground forces involved in human movement. However, these systems are under constant improvement due to their widespread applicability and reliability. An innovation by Patel *et al.* (2022) introduced a force plate system featuring a plate for receiving force, along with numerous single-axis load cell sensors arranged to support the plate. Positioned at an angle to the plate's plane and oriented towards the centre, these sensors record data regarding force vectors on the angled load cell whenever a force is exerted on the plate. The presented designs included a rectangular plate with a load cell at each corner, and a hexagonal plate with load cells supporting each of the six sides. The shape of the plate determines the distribution of forces and moments that can be detected, while a higher sensor count may allow for finer spatial resolution, enabling the detection of subtle variations across the surface.

A similar endeavour by Wardoyo *et al.* (2016) aimed to construct a cost-effect force plate, designed to distinguish and evaluate the biomechanics involved in both standing and walking activities. Using a flexible force transducer positioned within a rubber mat consisting of square blocks (250 mm x 150 mm x 10 mm), their prototype exhibited a peak load tolerance of 60kg. Despite its comparatively smaller dimensions than those of commercial alternatives, the plate was still able to comfortably accommodate both feet. Furthermore, the prototype boasted other advantages including portability, flexibility, and outdoor applicability. In terms of the walking trial, two distinct peaks in the waveform were observed, as depicted in Figure 2.2(a). However, a faster walk trial yielded a single sharp peak with similar amplitude, as shown in Figure 2.2 (b). Notably, these graphs define the magnitude along the y-direction.

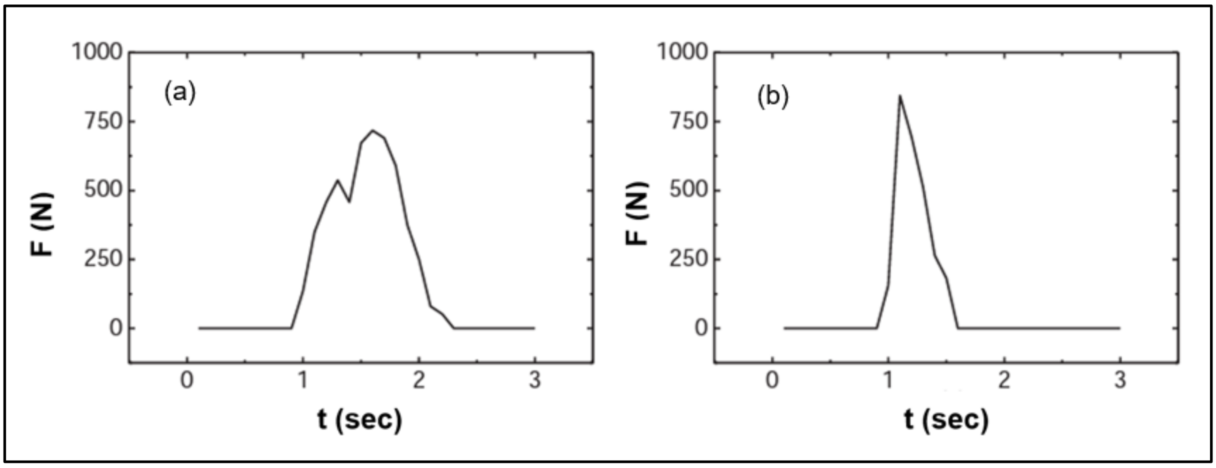

Figure 2.2: Force plate waveforms in different walk trials (Wardoyo *et al.*, 2016)

Based on this observation, Wardoyo *et al.* (2016) proposes that this dissimilarity stems from insufficient sensitivity and sensor configuration, particularly during brief intervals.

In terms of implementation, a study by Beckham *et al.* (2014) delves into the utilisation of force plates for monitoring sports performances, aiming to provide insights for future researchers on the core principles of force plate application. These principles encompass the theoretical basis, force plate design and function, key aspects of data acquisition, and other noteworthy technical details. With regards to the theoretical basis, the researchers underscore two key facts: 1) all ground-based movements hinge upon the forces exerted on the ground, even during horizontally oriented actions like running, and 2) differences in forces relate to differences in executed movements. Furthermore, Beckham *et al.* (2014) offer recommendations for selecting an appropriate sampling frequency, indicating that 100 – 200 Hz suffices for low-impact gait activities such as walking. Additionally, they noted that calibration protocols could encompass load ranges from 0 to 500kg, which is notably different than the prototype of Wardoyo *et al.* (2016). The researchers conclude that due to the complex nature of force plate technology and its ability to yield diverse data, force plate systems can facilitate a wide array of analysis projects. This emphasises the importance of meticulously selecting an appropriate force plate system tailored to the study's specific objectives.

Through their ability to precisely measure GRFs, force plates provide invaluable insights into the dynamics of human movement, prompting continuous efforts to enhance these systems. This literature review not only offers recommendations to guide the present study in selecting an appropriate system but also highlights recent innovations in this field. For instance, Patel *et al.* (2022) introduced a force plate system that incorporates single-axis load cell sensors positioned at an angle to the plate's plane. These sensors, arranged in various configurations, enable the recording of force vectors when external forces are exerted on the plate. This approach highlights the ongoing commitment to refining force plate technology. As technology specifications can considerably vary between systems, further discussion regarding these aspects is presented in subsequent chapters.

2.3.1.1.2 Working principles and basic theoretical concepts

Newton's third law of motion declares that when an object exerts a force on another object, the second object simultaneously exerts an equal and opposite force on the first object. This foundational principle is present in force plate systems, where the GRF can be determined from the force a user exerts onto the plate. This mechanism is depicted in Figure 2.3(a), which illustrates the reference frame of a rectangular force plate system capable of detecting forces along three axes, i.e., the x-, y-, and z-axis. In Figure 2.3(b), localised reaction force vectors counteracting a foot are depicted (Kwon, 1998).

The functionality of force plate systems hinges on the utilisation of either load cells, housing piezoelectric elements, or strain gauges, to determine forces acting on the plate. Consequently, when a force is applied, the load cell sensors undergo distortion, leading to quantifiable voltage fluctuations that are proportionate to the applied force (Lamkin-Kennard

& Popovic, 2019). This concept is portrayed in Figure 2.3(c), where the four load cells emit distinct reaction force vectors. The summation of these force vectors equates to the cumulative effect of all ground reactions, as represented in Figure 2.3(b). It can therefore be affirmed that system (b) corresponds to system (c) (Kwon, 1998).

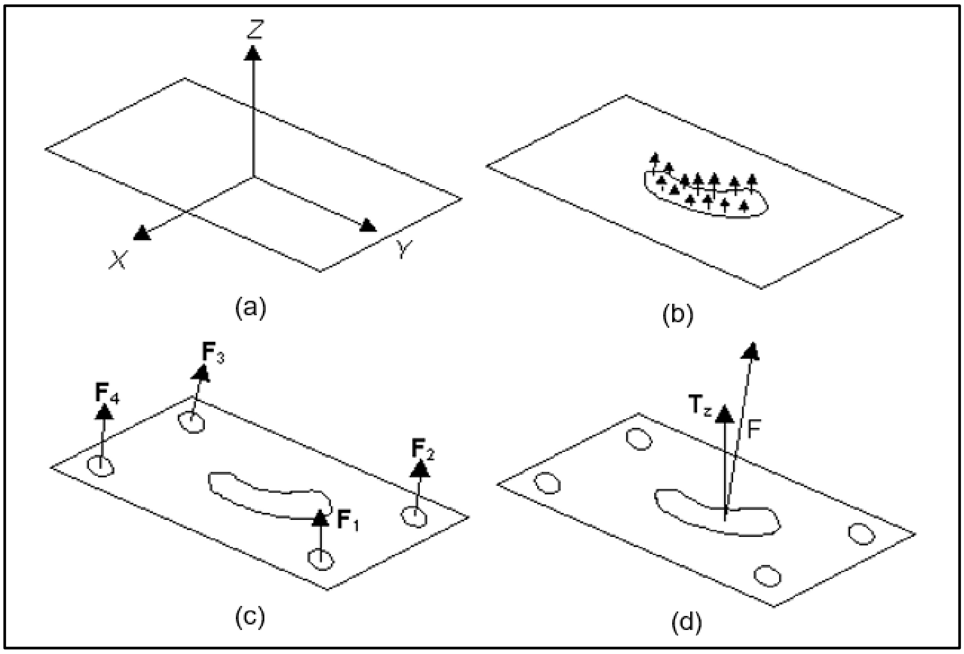

Figure 2.3: Force plate reaction forces (Kwon, 1998)

The total single reaction force $\mathbf{F}$ and torque $\mathrm{T_z}$ can be determined from the four reaction force vectors, as illustrated by Figure 2.3(d). To express this numerically:

$$\boldsymbol{F} = \boldsymbol{F_1} + \boldsymbol{F_2} + \boldsymbol{F_3} + \boldsymbol{F_4} \qquad (2.1)$$

However, since the forces can act in three axes, the single reaction force vector can also be expressed as:

$$\boldsymbol{F} = F_x + F_y + F_z \qquad (2.2)$$

The system represented by (d) is equivalent to system (c), and thus is also equivalent to system (b). The torque component, also called the free torque, arises from the interplay of forces around the vertical axis. This torque is therefore only present on the z-axis (Kwon, 1998).

In terms of the computation of the CoP, assume the true origin O' is at position (a,b,c), as depicted in Figure 2.4. The z-component of the CoP is always at position 0 (Kwon, 1998).

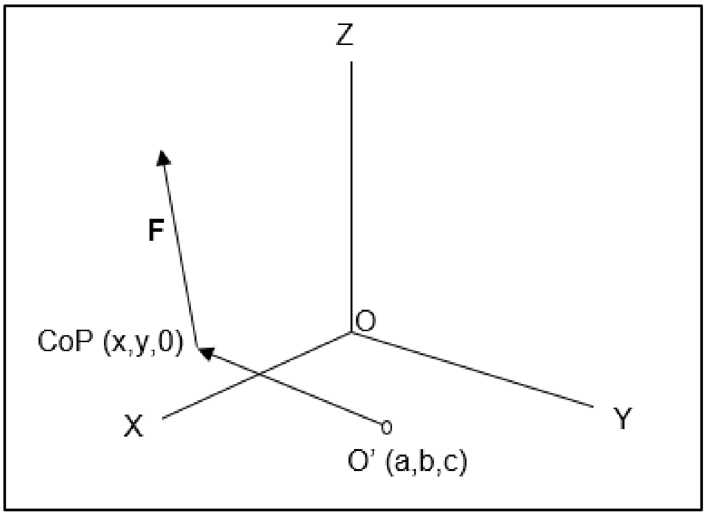

Figure 2.4: Force plate centre of pressure (Kwon, 1998)

The moment determined from the plate is equivalent to the moment generated by the reaction force vector $\mathbf{F}$ about the true origin plus the free torque $\mathrm{T_z}$. Therefore:

$$\boldsymbol{M} = [x - a, y - b, -c] \times [F_x, F_y, F_z] + [0,0,T_z] \tag{2.3}$$

Or, when expressed as a matrix:

$$\begin{bmatrix} M_x \\ M_y \\ M_z \end{bmatrix} = \begin{bmatrix} 0 & c & y-b \\ -c & 0 & -(x-a) \\ -(y-b) & x-a & 0 \end{bmatrix} \begin{bmatrix} F_x \\ F_y \\ F_z \end{bmatrix} + \begin{bmatrix} 0 \\ 0 \\ T_z \end{bmatrix} \Rightarrow$$

$$\begin{bmatrix} M_x \\ M_y \\ M_z \end{bmatrix} = \begin{bmatrix} (y-b)F_z + cF_y \\ -cF_x - (x-a)F_z \\ (x-a)F_y - (y-b)F_x + T_z \end{bmatrix} \tag{2.4}$$

From this matrix equation, expressions for positions x, y, and free torque value T_z can be simplified to:

$$x = -\frac{M_y + cF_x}{F_z} + a \tag{2.5}$$

$$y = -\frac{M_x + cF_y}{F_z} + b \tag{2.6}$$

$$T_z = M_z - (x-a)F_y + (y-b)F_x \tag{2.7}$$

Depending on the type of force plates used, M_x, M_y, M_z, F_x, F_y and F_z may be able to be measured directly from the channels, whereas the position of the true origin should be displayed on the calibration data sheet provided by the manufacturer (Kwon, 1998). With these fundamental principles, force plate systems may offer a robust means to capture and quantify GRF during various gait-related activities.

2.3.1.2.2 Force plate specifications in gait studies

Force plate specifications are contingent upon the specific application at hand. Therefore, judicious selection of these systems necessitates a thorough consideration of key requirements relevant to gait analysis. For example, in the context of walking trials, the length of the force plate system should be substantial to ensure adequate data acquisition. To provide an overview of commercially available force plate system specifications, Table 2.2 presents a comparison between the Zebris FDM1.5, FDM 2, and AMTI HPS464508 force plates.

Table 2.2: Force plate manufacturer specifications comparison

Specification	*FDM 1.5*	*FDM 2*	*AMTI HPS464508*
Dimensions (WxLxH) (cm)	158 x 60.5 x 2.1	212 x 60.5 x 2.1	46.4 x 50.8 x 8.25
Number of sensors	11 264	15 360	-
Sampling rate (Hz)	100, 200, 300	100	-
Accuracy (% of FS)	< 3	< 3	± 0.1
COP accuracy (mm)	-	-	0.2

Aligning with the study of Beckham *et al*. (2014), the sampling frequencies outlined in this table are sufficient for low-impact activities such as walking. However, the FDM 1.5 system equipped with a 300 Hz sampling frequency may yield an improved data accuracy. Force plate systems therefore need to be tailored to the unique demands of the application. Pertinent specifications to consider include overall dimensions, number of sensors, sampling rate, and CoP accuracy.

2.3.2 Kinematic quantification systems

Kinematics is a field that deals with mathematical descriptions and various physical principles used to describe the movement of real-world objects (Dusto, 2020). Within the current study, systems of this nature predominantly include inertial sensors, which are based on Newton's law of inertia. For gait analysis studies, these inertial measurement units (IMUs) consist of gyroscope, accelerometer, and optionally magnetometer sensors (Alexis, n.d.; Tau *et al.*, 2012). In addition to these technologies, motion capture camera is frequently used in conjunction to facilitate in determining of joint position, motion trajectory, and joint angle variations (Akhtaruzzaman *et al*., 2016).

2.3.2.1 Gyroscope sensors

Gyroscope sensors utilise the principle of conservation of angular momentum and have frequent application in the determination of angular position and velocity (WatElectronics, 2020), which play a crucial role in assessing user motion and posture (Tao *et al*., 2012). This sensor technology is therefore a valuable component in gait analysis, capable of capturing kinematic changes in the foot and ankle complex throughout the gait cycle.

2.3.2.1.1 Related works

A study conducted by Shih *et al.* (2013) highlighted that gyroscope sensors have consistently demonstrated their capability in effectively quantifying foot pronation motions during gait activities in previous research. To further this statement, their study made use of a tri-axial gyroscope sensor to measure the kinematic changes of the foot during intense level running. In an experiment consisting of 15 male participants, they were able to observe kinematic variations within 30 minutes of intense running protocols. To validate the reliability of the signal data and the accuracy of the sensor, their findings were compared to reference data obtained from a motion capture camera, with positive outcomes. Their results exhibited a high correlation between ankle ROM and peak angular velocity in the frontal plane. It was also found that the angular velocity significantly correlated with participant heart rate and perceived exertion. The researchers therefore concluded that kinematic changes in the foot could be related to running injuries, particularly based on the peak angular velocity of the foot in the frontal plane.

In addition to the above-mentioned research, a relevant study by McGrath *et al*. (2012) assessed temporal gait parameters during walking and running trials using a gyroscope sensor. Their experimental configuration consisted of five healthy participants with an age range of 26 to 32 years, and a treadmill platform for participants to execute trials. Their experimental configuration consisted of five healthy participants with an age range of 26 to 32 years, and a treadmill platform for participants to execute trials. These trials encompassed four distinct speeds, with 2 and 4 km/h categorised as walking speed, and 8 and 12km/h categorised as jogging/running speed. Trials transpired in 20 second intervals conducted twice for each participant. Data collection involved the attachment of two inertial sensors to the participants' left and right shanks, while reference data was concurrently gathered through a

CODA motion capture system. To delineate the heel-strike and toe-off sub-phases within the gait cycle, McGrath *et al.* (2012) adapted two gait analysis algorithms previously proposed by Hreljac and Marshal (2000), and Zeni *et al.* (2008). In the subsequent analysis, the researchers compared the temporal parameters derived from the adapted algorithm to the reference motion capture camera data. The results of this comparison demonstrated a high accuracy percentage error across all trials, thus signifying the accuracy of both the algorithm and gyroscope. Their ensuing discourse revealed that accuracy was enhanced when the sensor was positioned in closer proximity to the reflective markers on the participant's foot. However, it was noted that due to the specific placement of the sensors on the shanks, a certain degree of time lag was expected between the peaks of the angular velocity signals.

The outcomes derived from the studies establish the feasibility and reliability of gyroscope sensors in capturing the intricate kinematics of the foot and ankle. The validation of this is corroborated by their comparison against the benchmark of a motion capture analysis system. Furthermore, it is evident that sensor placement should align with the reflective markers for enhanced accuracy of experimental outcomes. Therefore, relating this information to the present study substantiates the use of a gyroscope technology in a sensor platform device to quantify and analyse kinematics of the human foot and ankle joint complex. Moreover, a notable observation in the studies is the limited diversity evident in both the experimental procedures and participant demographics. Addressing this gap through a diverse participant group and additional experimental activities could yield valuable insights.

2.3.2.1.2 Working principles and basic theoretical concepts

Conceptually, gyroscopes encompass a rotating wheel with an axis capable of assuming various orientations, drawing upon the principle of conserving angular momentum. In simplified terms, this principle states that the total angular momentum of a system remains constant in both magnitude and direction when the net external torque acting upon the system is zero (Alexis, n.d.). As a micro-machined electromechanical system (MEMS), gyroscopes exploit the phenomenon of the Coriolis effect, as illustrated in Figure 2.5, to determine angular velocity.

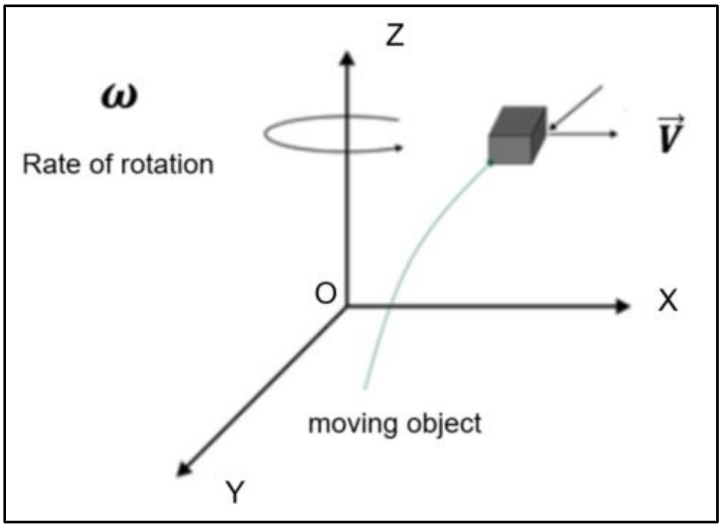

Figure 2.5: The Coriolis effect (Alexis, n.d.)

The Coriolis effect, which relates to Newton's second law of motion, refers to the motion that occurs when an external force is exerted, leading to the gyroscope's movement. This effect manifests as an observable inertial force that is proportionate to the angular rate of rotation within a coordinate system (Alexis, n.d.; Tao *et al.*, 2012). Rotation in the MEMS induces a Coriolis acceleration in the mass, which is obtained from the formula:

$$\vec{a}_{cor} = 2\vec{V} \times \vec{\omega} \tag{2.8}$$

Where

- V is the velocity observed from the rotating frame, and
- ω is the rate of rotation.

Angular velocity is thus determined from the Coriolis effect by measuring the induced displacement in the spinning mass due to motion along a perpendicular axis, with respect to time (Alexis, n.d.). Mathematically, this can be expressed as:

$$\omega = \frac{\Delta\theta}{\Delta t} \tag{2.9}$$

Where

- ω is the angular velocity,
- $\Delta\theta$ is the change in angular displacement, and
- Δt is the change in time in sec.

However, akin to any sensor, the potential for error or bias in reading exists, potentially compromising the precision of outcomes (SparkFun, 2013). Bias in the MEMS can be determined from the following expression, as delineated by Alexis (n.d.).

$$\delta\omega_{BIAS} = \delta\omega_{const} + \delta\omega_{BS} + \delta\omega_{BD}, \quad (2.10)$$

$$\frac{d}{dt}\omega_{BD} = \omega(t);\ \omega \sim N(O, Q) \quad (2.11)$$

Where Q represents a recognised quantity with the units $(°/\mathrm{h})/\mathrm{sqrt}(\mathrm{h})$, and the bias effect is present in three forms (Alexis, n.d.), namely:

- Fixed bias (ω_{const}), which is represented by a constant (Alexis, n.d.),
- Bias stability (ω_{BS}), which is the variation from one revolution to the next (Alexis, n.d.), and
- Bias drift (ω_{BD}), which can mathematically be modelled as a combination of time and temperature-dependent behaviours (Gulmammadov, 2009).

To mitigate the effect of bias drift, most gyroscope sensors incorporate an integrated temperature sensor. This addition enables the monitoring of temperature-dependent variations, facilitating their correction through sensor calibration processes (SparkFun, 2013). Therefore, despite the potential challenges of bias and drift, gyroscope sensors, through their intricate working mechanisms, stand as valuable tools for measuring angular velocity.

2.3.2.1.3 Gyroscope specifications in gait studies

In the process of selecting an appropriate gyroscope sensor for a specific application, several crucial specifications demand consideration, including full-scale (FS) range, sensitivity, and bias (SparkFun, 2013). Research by Zhao *et al.* (2019) examines the application of MEMS gyroscopes for human gait analysis, with a focus on two specific inertial sensors for their study: the Nano IMU (nIMU) from MEMSense Inc., USA, and the ADIS16448 *i*Sensor® from Analog Devices Inc., USA. The detailed specifications of these two sensors, as outlined by Zhao *et al.* (2019) are listed below in Table 2.3.

Table 2.3: Gyroscope manufacturer specifications comparison

Specification		nIMU	ADIS16448 iSensor®
Mass (g)		20	15
Size (mm)		45 x 23 x 13	24.1 x 37.7 x 10.8
Operating temperature (°C)		0 to +70	-40 to +85
Sampling rate (Hz)		150	400
Gyroscope	Range (°/s)	±600	±1000
	Nonlinearity (% of FS)	±0.1	±0.1
	Noise (°/s)	0.56	0.27
	Bandwidth (Hz)	50	330

This table indicates that the nIMU device possesses a lower bandwidth, measurement range, and operational temperature scale. Nonetheless, Zhao *et al.* (2019) noted that the bandwidth and range variability of the gyroscope within the nIMU prove inadequate due to the pronounced dynamics encountered in the foot and ankle, particularly during heel-strike events. According to Bancroft and Lachapelle (2012), the maximum angular velocity experienced by a toe-mounted gyroscope can reach up to 1500°/s during running and 2000°/s during sprinting trials. In light of this, the ADIS16448 *i*Sensor® is more adept for walking trials, given its dynamic range of ±1000°/s (Zhao *et al*., 2019). The ADIS16448 *i*Sensor® further excels due to its considerably higher sampling rate, as well as its marginally smaller size and mass.

It is important to note that the table and ensuing discussion are intended to serve as a guiding framework for sensor selection within gait analysis applications. This nuanced decision-making process, guided by scrupulous technical considerations, emphasises the critical role of sensor selection in ensuring the accuracy and reliability of kinematic data extraction.

2.3.2.2 Accelerometer sensors

Accelerometers are a type of inertial sensor designed primarily for the measurement of acceleration forces in a linear motion, offering insights into an object's velocity over time. In biomedical applications, accelerometer sensors are often employed for step counting, monitoring physical activities, or enabling motion-based creations and suppression techniques (Kumar *et al*., 2021). Tao *et al.* (2012) further assert that accelerometer sensors hold relevance for the analysis of human gait. By affixing these sensors to the foot or leg, they enable the determination of acceleration and velocity, pivotal components for conducting comprehensive gait analyses. Therefore, the integration of accelerometer technology empowers the precise execution of gait analysis, thereby enriching the understanding of human movement dynamics.

2.3.2.2.1 Related works

A study by Patterson and Caulfied (2013) utilised a foot-mounted accelerometer for the purpose of discerning changes in user gait patterns. The experiment involved eight healthy participants, with kinematic data collection encompassing 10 trials for two distinct walking conditions: normal and stiff ankle walking. Subsequent MATLAB® analysis revealed that the peak total acceleration during the initial swing was considerably greater in stiff ankle walking, accompanied by a shortened duration of time to reach peak acceleration. From these findings it was inferred that the data derived from tri-axial accelerometer measurements is simple to process and hold potential in detecting alterations within ankle gait patterns. However, it is important to note that the stiff ankle walking condition was artificially induced and should be regarded as a preliminary outcome.

Additional literature encompasses a study conducted by Godfrey *et al.* (2015), which examined the utilisation of an accelerometer-based system and algorithm for the purpose of conducting a comprehensive gait analysis. In their study, a body-worn sensor system was used and compared to data garnered from reference equipment. The experimental protocol entailed the participation of healthy individuals aged 20 to 40 years in the young group, and 50 to 70 years in the older group. From normal walking trials, collected data was inputted into and analysed using a developed MATLAB® algorithm, yielding estimations of crucial parameters like initial contact, final contact, and step length. Through comparison of these results to the reference data, Godfrey *et al.* (2015) were able to confirm the validity of an accelerometer-based body-worn sensor system, however, found a lack of consistency and poor asymmetry results between the sensor system and reference equipment. Through further experimentation, it was found that this was due to inherent differences between the systems, rather than the sensor system's ability to quantify gait characteristics.

The outlined studies highlight the value of accelerometer sensors in gait quantification and analysis. However, it is noteworthy that discrepancies in variability and asymmetry between accelerometer and reference systems were observed. To circumvent this concern in the current study, an integrated sensor device will be used as opposed to a standalone accelerometer-based system. In summation, the evidence suggests that accelerometer sensors serve as valuable tools capable of detecting changes in gait patterns, thereby warranting their inclusion this study. Notably, although a more diverse participant demographic was used for both studies, the experimental procedure remained constrained and only considered walking trials.

2.3.2.2.2 Working principles and basic theoretical concepts

Accelerometers function as electromechanical sensors adept at measuring static and dynamic forces or acceleration along the three principal axes – x, y, and z. This is achieved through a mechanical sensing element that includes a mass suspended within a framework. When the sensor experiences an acceleration, this mass is displaced from its equilibrium position, inducing a mechanical strain on the sensing element. This strain generates a measurable voltage fluctuation that can be translated into acceleration values along different axes (Tao *et al.*, 2012).

As previously specified, the human foot is capable of movements along the three axes and planes - sagittal, transverse, and frontal. Hence, a triple-axis tilt should be considered, as illustrated in Figure 2.6 (Fisher, n.d.).

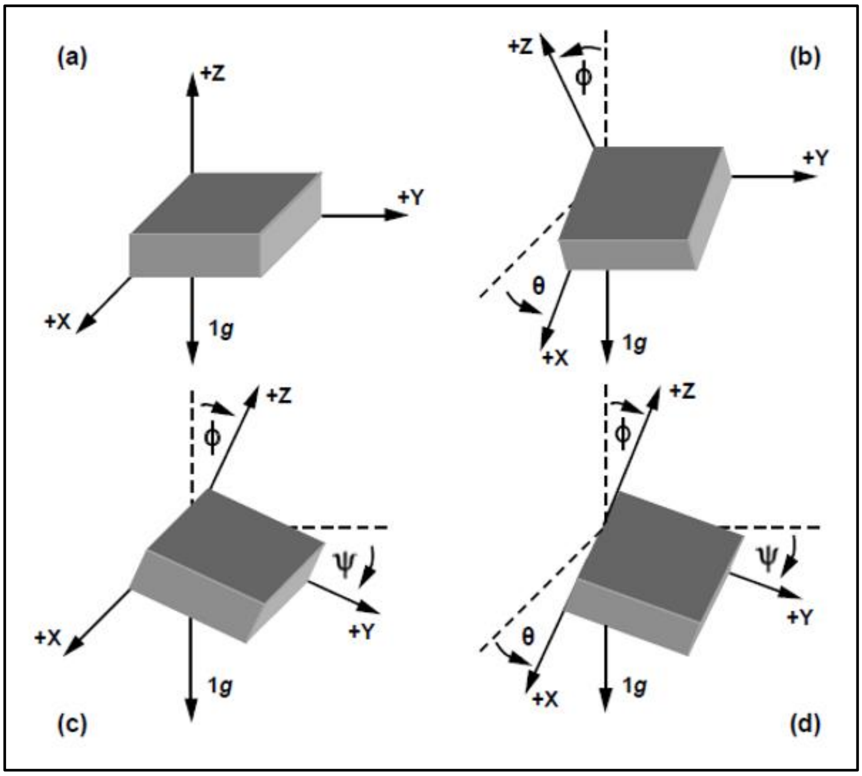

Figure 2.6: Triple axis tilt (Fisher, n.d.)

Basic trigonometry yields the following equations for angles θ, ψ, and ϕ, which denote the angles between the horizon and the x-axis, the horizon and the y-axis, and the gravity vector and the z-axis, respectively (Fisher, n.d.).

$$\theta = \tan^{-1}\left(\frac{A_x}{\sqrt{A_y^2 + A_z^2}}\right) \tag{2.12}$$

$$\psi = \tan^{-1}\left(\frac{A_y}{\sqrt{A_x^2 + A_z^2}}\right) \tag{2.13}$$

$$\phi = \tan^{-1}\left(\frac{\sqrt{A_x^2 + A_y^2}}{A_z}\right) \tag{2.14}$$

Additionally, to minimise errors arising from offset and sensitivity, it is crucial to calibrate these values. The calibrated output acceleration can then be utilised to accurately determine the angle of inclination. Accounting for these issues, the accelerometer output can be expressed as (Fisher, n.d.):

$$A = A_{OFF} + (Gain \times A_{ACTUAL}) \tag{2.15}$$

Where:

- A_{OFF} is the offset error,
- $Gain$ is the accelerometer gain, and
- A_{ACTUAL} is the actual acceleration of the accelerometer.

Like gyroscope sensors, this theoretical framework outlines the principles underlying the accelerometer component within the IMU device. Consequently, these principles should be comprehended in conjunction with the theoretical concepts of gyroscopes.

2.3.2.2.3 Accelerometer specifications in gait studies

Much like the process of gyroscope selection, accelerometer specifications should be selected based on the particular application, i.e., a gait study. Key parameters to consider when opting for an accelerometer encompass measurement range, frequency response, grounding, high and low-frequency limit, noise, sensitivity, and temperature range (Omega, n.d.).

Table 2.4 outlines manufacturer specifications of a tri-axial accelerometer utilised in prior gait studies by Godfrey *et al.* (2015) and Patterson and Caulfield (2013). Respectively, the IMU devices used in these studies are Axivity AX3 and Xsens MTx, respectively.

Table 2.4: Accelerometer manufacturer specifications comparison

Specification		***Axivity AX3***	***Xsens MTx***
Mass (g)		11	30
Size (mm)		23 x 32.5 x 7.6	38 x 53 x 21
Operating temperature (°C)		0 to +65	-20 to +55
Sampling rate (Hz)		25 to 400	100
Accelerometer	Range (g-force)	±2/4/8/16	±5
	Linearity (% of FS)	unspecified	0.2
	Noise (units /√Hz)	unspecified	0.002
	Bandwidth (Hz)	unspecified	30

While some specifications of the Axivity AX3 remain unstipulated, it is evident that this IMU offers the advantage of a configurable sampling rate and measurement range. In the study of Godfrey *et al.* (2015), the sensor was programmed to operate at a sampling frequency of 100Hz and a full-scale range of ±8g, surpassing the specifications of the Xsens MTx. Considering that both studies focused on standard walking trials, a higher measurement range may result in more dependable data. Furthermore, as with the theoretical principles, these specifications should be integrated with the gyroscope particulars of the selected IMU device.

2.3.2.3 Integrated inertial systems

Integrated inertial systems merge the functionalities of gyroscope and accelerometer sensors into a singular IMU. This combination brings together the theoretical underpinnings of both these sensors, particularly relevant when applied to a gait analysis study, and has gained widespread recognition within gait analysis procedures, as affirmed by Tao *et al.* (2012).

2.3.2.3.1 Related works

In a recent study by Homan *et al.* (2022), a wearable sensor platform, comprising of accelerometer and gyroscope sensors, was used to estimate various spatiotemporal gait parameters. These parameters are delineated in Table 2.5 below, which builds onto the spatiotemporal parameters introduced earlier in Table 2.1.

Table 2.5: Gait parameters for wearable sensor platform

Parameter	***Unit***	***Description***
Cycle duration	sec	The total time of one full gait cycle.
Cadence	steps/min	The frequency of cycles executed within a minute.
Stance phase	% of cycle duration	The percentage of time the foot remains in contact with the ground throughout a cycle.
Swing phase	% of cycle duration	The percentage of time the foot is in the air during a single cycle.
Loading phase	% of stance	The percentage of time between the heel strike and foot flat sub-phases.
Foot-flat phase	% of stance	The percentage of time where the foot is completely in contact with the ground during the stance phase.
Pushing phase	% of stance	The percentage of time between the foot flat toe-off sub-phases.
Double support phase	% of cycle duration	The percentage of time both feet are in contact with the ground during a single cycle.
Stride length	m	The total distance covered between successive footprints on the ground.
Stride velocity	m/sec	The forward speed of a single cycle.
Peak angle velocity	°/sec	The maximum angular velocity between maximum heel clearance and minimum toe clearance.
Max swing speed	m/sec	The peak forward velocity of the foot during the swing phase.
Strike angle	°	The angle of the foot in relation to the ground at the moment of heel contact.
Lift-off angle	°	The angle of the foot in relation to the ground at the end of the push phase.
Swing width	m	The maximum lateral displacement between the projected trajectory and the actual path of the foot in the swing phase.
3D path length	% stride length	The trajectory length of the foot throughout a complete cycle in a three-dimensional space.

The experimental procedure conducted by Homan *et al.* (2022) involved 10 healthy male participants executing three sets of normal walking trials. The data collected from this process underwent analysis through gait analysis software called GaitUp. This data was then subjected to a comparative analysis against reference data from a three-dimensional (3D) motion capture camera and force plate system. In total, the wearable sensor platform and reference motion capture camera yielded 238 gait cycles. However, among these, 214 cycles were cast-off due to improper foot contact with the force plate systems and other errors. Nevertheless, the sensor platform demonstrated high validity across several parameters, as well as an excellent repeatability when measured between sessions. A limitation of the review was the relatively small number of participants, which constrained the generalisability of the findings. Although, the researchers suggest that the acquired findings could serve as a practical foundation for conducting validation studies with larger participant samples. Additionally, the study's sole focus on male participants prompted a recommendation for future investigations to incorporate a more diverse range of participants.

Another relevant study, conducted by Xie *et al.* (2022), considers 3D skeletal tracking for comprehensive gait analysis and relies on data acquired from an integrated inertial sensor system. In resonance with the previously examined studies of the current literature review, Xie, and colleagues (2022) utilise reflective markers affixed to the sensors, and a 3D motion capture camera system to obtain reference data. Correspondingly, the study outlines gait parameters that bear resemblance to those elucidated in Homan *et al.* (2022), including stride length, step length, step width, cycle duration, and swing and stance phase duration. Through numerous normal walk trials, the sensor-derived data underwent collection and transmission to a feature detection module, enabling the extraction of gait features from the raw data. The ensuing analysis showed the sensor system's ability to accurately determine most gait parameters with a 3% error and displacement with an error of 2.3%. Notably, the investigation revealed that discrepancies emerged particularly in participant step width, where a 10% error was observed in comparison to the reference data, due to overly narrow step width.

The study of Homan *et al.* (2022) underscores the significance of diversifying the participant pool to encompass both male and female subjects, thus enabling more generalised findings. In that same study, it is further substantiated that numerous trials of data collection may prove beneficial, since errors in data collection may render several sets of data unusable. Furthermore, both reviewed studies expanded on the foundations laid by Prasnath *et al.* (2021) by providing supplementary spatiotemporal parameters for a gait study.

In conclusion, the use of IMUs in gait applications showcase a commendable capacity to yield reasonably accurate kinematic data. Nevertheless, it is imperative to implement strategies to bolster accuracy. These strategies include the use of reference equipment (e.g., motion capture cameras), appropriate placement of reflective markers and sensors, repeated trials, and the involvement of a qualified biokineticist to oversee and ensure the precise execution of gait activities. These combined measured collectively contribute to enhancing the fidelity of the acquired data, and thus fortify the reliability of the subsequent analysis.

2.3.2.4 Motion capture cameras

Motion capture cameras are common gait quantification systems that rely on a network of multiple cameras and diverse camera angles to record real life movement of a subject. This intricate process translates the captured movement into sequences of Cartesian coordinates within a 3D spatial framework (Guerra-Filhol, 2005). Targeted features for this system include joint positions, motion trajectories, and angle variations throughout the gait cycle (Akhtaruzzaman *et al.*, 2016). Therefore, as a reference system, motion capture camera data may serve as a benchmark for assessing the performance of other gait quantification technologies.

2.3.4.4.1 Motion capture camera specifications in gait studies

With regards to the working principles and fundamental concepts, motion capture systems do not rely on specific equations or laws that are relevant to the study. However, their effective operation requires adherence to essential resources and calibration protocols. Guerra-Filhol (2005) outlines these prerequisites, which encompass a designated capture room, reflective body markers, camera equipment, and acquisition software. Furthermore, it is important to ensure that the dimensions of the capture room are adequate to enable recording from multiple viewpoints at a sufficient distance. Additionally, uniform lighting within the room is equally important to mitigate the occurrence of sharp shadows and unwanted highlights on the experimental scene. The integration of these resources ensures the accurate capturing of the subject’s movement, providing the reference data necessary for subsequent analysis.

Guerra-Filhol (2005) identifies suitable motion capture camera systems for typical gait studies, including the Kodak™ ES-310, Sony™ XCD-X700, and Sony™ DFW-V500. Notably, in the study of McGrath *et al.* (2012), a CODA motion capture camera system was used to acquire reference data. To provide a comprehensive overview, Table 2.6 outlines and compares the manufacturer specifications of these systems.

Table 2.6: Motion capture camera manufacturer specification comparison

Specification	*ES-310*	*XCD-X700*	*DFW-V500*	*CODA*
Frame rate (FPS)	85	15	30	-
Operating temperature (°C)	0 to 40	-5 to +45	-10 to +50	-
Mass per unit (g)	680	250	305	500

It is worth highlighting that frame rate is among the most important specification considerations in gait studies, as it directly impacts the data quality acquired. This attribute, as indicated by the table, exhibits a considerable spectrum of variations across systems.

2.3.3 Hybrid systems and combined application

As noted by Akhtaruzzaman *et al.* (2016), contemporary gait studies are increasingly turning their attention towards hybrid systems, amalgamating various quantification technologies into a singular integrated framework. This combination holds the potential of mitigating the limitations inherent in individual systems. For the quantification systems considered in this study, the following benefits and limitations can be mentioned, as outlined by Akhtaruzzaman *et al.* (2016).

Table 2.7: Quantification systems and technology benefits and limitations

	Benefits	*Limitations*
Force plate systems	○ No direct attachment to the user/patient's body. ○ CoP can be easily determined. ○ Useful in calculating GRF.	○ Needs to be combined with kinematic quantification systems for gait analysis. ○ Foot contact may be misplaced and lead to errors in findings. ○ Only suitable for laboratory setting. ○ Cannot capture movements that extend beyond the confined area of the plates.
Integrated inertial system	○ Generally high sampling rate. ○ Tri-axial data measurement enables 3D analysis. ○ Small, lightweight, and inexpensive. ○ Output signals can be immediately recorded. ○ Daily activities can be monitored. ○ No need for a controlled environment.	○ Sensor attachments may be uncomfortable for users. ○ Skin movement artefacts may result in interrupted data measurement. ○ Complex approach is required for gait parameter estimation. ○ Varying sensor attachment positions may show dissimilarities in acceleration sensing.
Motion capture camera	○ Easier to use than other systems. ○ Offers robust and precise capture of physical motions. ○ Allows for the reuse of captured gait images. ○ Participant geometry and textures are accurate. ○ Little skill required to use.	○ Needs appropriate setup with active line-of-sight. ○ Faster movement requires a faster sampling rate. ○ Reflective marker attachments may be uncomfortable. ○ Skin movement artefacts over the skeletal structure may display unexpected disturbances.

Table 2.7 highlights the general benefits and limitations of the previously examined quantification systems and technologies. The benefits listed in this table support the selection of the designated systems and technologies, whereas the limitations may provide suggestions towards the research design and method component of the study, however, this is examined in the following chapter.

According to Weizman *et al*. (2021), wearable sensor systems prove particularly advantageous in real-world gait and fall studies owing to their compact size, lightweight design, and cost effectiveness. This point is further substantiated by the merits of inertial sensors outlined in Table 2.7. A study conducted by Feng *et al*. (2023) emphasized the value of gait analysis studies when combined with wearable sensors, due to their ability to provide a convenient, efficient, and economical means for data collection. Their study demonstrated that capability of wearable sensors to facilitate high-precision gait feature extraction for subsequent analysis. Moreover, both Feng *et al*. (2023) and Weizman *et al.* (2021) noted that wearable sensors are highly effective to measure gait levels both within and beyond laboratory setting. These findings serve as rationale for incorporating and integrating a wearable sensor device with additional quantification systems into the present study.

2.4 Real-time gait data analysis techniques

Real-time gait data analysis techniques aim to identify and differentiate gait events within the acquired raw gait data. As per Table 2.1, gait features fall under three primary classifications: time domain, frequency domain, and time-frequency domain, with frequency referring to an analytical space. Within these three categories, multiple real-time analysis algorithms can be employed for gait studies. there are several real-time analysis algorithm techniques that can be utilised in gait studies. These algorithm types are identified by Prasnath *et al.* (2021) and listed below in Table 2.8.

Table 2.8: Real-time gait analysis techniques

Domain	***Algorithm type***
Time domain	○ Rule-based methods ○ Fuzzy interference system ○ Machine learning ○ Phase portrait
Frequency domain	○ Adaptive oscillator ○ Spectral analysis
Time-frequency domain	○ Wavelet transform ○ Empirical mode decomposition

Among these analysis algorithms, Prasnath *et al.* (2021) identify the most popular method in gait studies as the rule-based method, a preference possibly attributed to its simplicity and intuitive nature in contrast to other techniques. The second most prevalent approach in gait studies, are machine learning techniques. The subsequent sections provide succinct insights into the underlying principles of these analysis techniques.

2.4.1 Rule-based methods

Rule-based methods are commonly employed in time-domain gait event detection techniques due to their simplicity, intuitive application, and minimal complex computational nature. This technique relies on predefined rules and conditional statements (if-else logic), interconnected by AND/OR operators or inequality constraints (Prasnath *et al.*, 2021).

A case in point is seen in a study by Zhao *et al.* (2019), where a rule-based approach was employed for gait event identification. This approach comprised three rules in the event detection procedure, namely, peak detection, flat zone detection, and zero-crossing detection. To illustrate these rules, Zhao *et al.* (2019) portray the stance phase as a characteristic of walking or running, wherein the foot enters a stance phase marked by zero velocity before transitioning into the subsequent phase. The researchers suggest that by meticulously designing rules and selecting appropriate parameters, this information can be adeptly used in the flat-zone detection procedure to accurately classify consecutive stance phases in gait trials.

Referring to the present study, the use of rule-based methods for feature selection and extraction appears to be a judicious choice. Given the intricate nature of the data acquired from inertial sensors and the inherent complexities of human motion, and simplicity of rule-based methods, this approach aligns well with the overall purpose of the study.

2.4.2 Machine learning

Machine learning is another method that is commonly used in gait studies, and relates to the time domain (Prasnath *et al.*, 2021). Alfayeed and Saini (2021) characterise machine learning in gait analysis as a biomechanical simulation that establishes the connection between multi-dimensional input data and an output model.

This method is evident in a study by Ghassemi *et al.* (2018), where a hierarchal hidden Markov model (HHMM) machine learning method was used to evaluate gait detection in individuals with Parkinson's disease. The HHMM, extending on the capabilities of the conventional hidden Markov model (HMM), offers certain advantages in capturing complex patterns and hierarchal structures in sequential gait data. In the study, the use of the HHMM was complemented by rule-based methods and two variants of dynamic time warping, contributing to a comprehensive analytical approach. The study highlighted that while the HHMM incurred higher computational demands compared to the other methods, its learned model could continue to be applied to further gait detection studies both offline and online. Notably, the study found the HHMM to significantly outperform the other methods in terms of precision. The conclusions drawn by Ghassemi *et al.* (2018) underline the effectiveness of simpler methods for homogenous data; however, when confronted with more heterogeneous assessments, the HHMM demonstrated a significant performance advantage over other methods.

In the current study, the anticipated nature of the data is likely to be homogenous, characterised by a certain degree of uniformity amongst collected samples. Given this context, the application of machine learning methods might be deemed unnecessary.

2.5 Chapter summary

The purpose of this chapter was to review and analyse prior research related to sensor-based gait analysis. This research was in the form of journal articles, relevant websites, and book chapters.

In pursuit of this, this chapter commenced with an investigation of key feature categories essential to a gait analysis study, which were identified as temporal, spatial, and spatiotemporal. Within this framework, the literature then examined previously published texts dedicated to sensor-based gait analysis. This inquiry branched into two distinct categories: kinetic gait quantification technologies, represented by force plate systems, and kinematic gait quantification technologies, consisting of inertial sensors (gyroscopes, accelerometers, and integrated units) and motion capture cameras. Furthermore, the literature was able to effectively provide insights into the overall reliability and applicability of various quantification technologies within the context of a gait study. Moreover, this exploration shed light on pertinent theoretical concepts, working principles, and manufacturer specifications. The advantages and constraints of these technologies were carefully evaluated, particularly in relation to the benefits offered by their combined application. The focus of this chapter then shifted towards real-time gait data analysis techniques, revealing rule-based methods and machine learning as predominant approaches for analysing gait data. It was concluded that due to the homogenous nature of the expected data, rule-based methods may offer certain benefits over machine learning methods.

The subsequent chapter expands on this literature by incorporating the research and theoretical concepts to formulate a research design and method for sensor-based gait analysis.

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CHAPTER 3

3. RESEARCH METHODOLOGY

3.1 Introduction

The previous chapter established a theoretical foundation for the study by examining previously published works related to sensor-based gait analysis. To build upon this foundation, this chapter applies fundamental concepts to design the research and describe the features of the utilised method. Figure 3.1 below breaks down this chapter by highlighting the key discussion points.

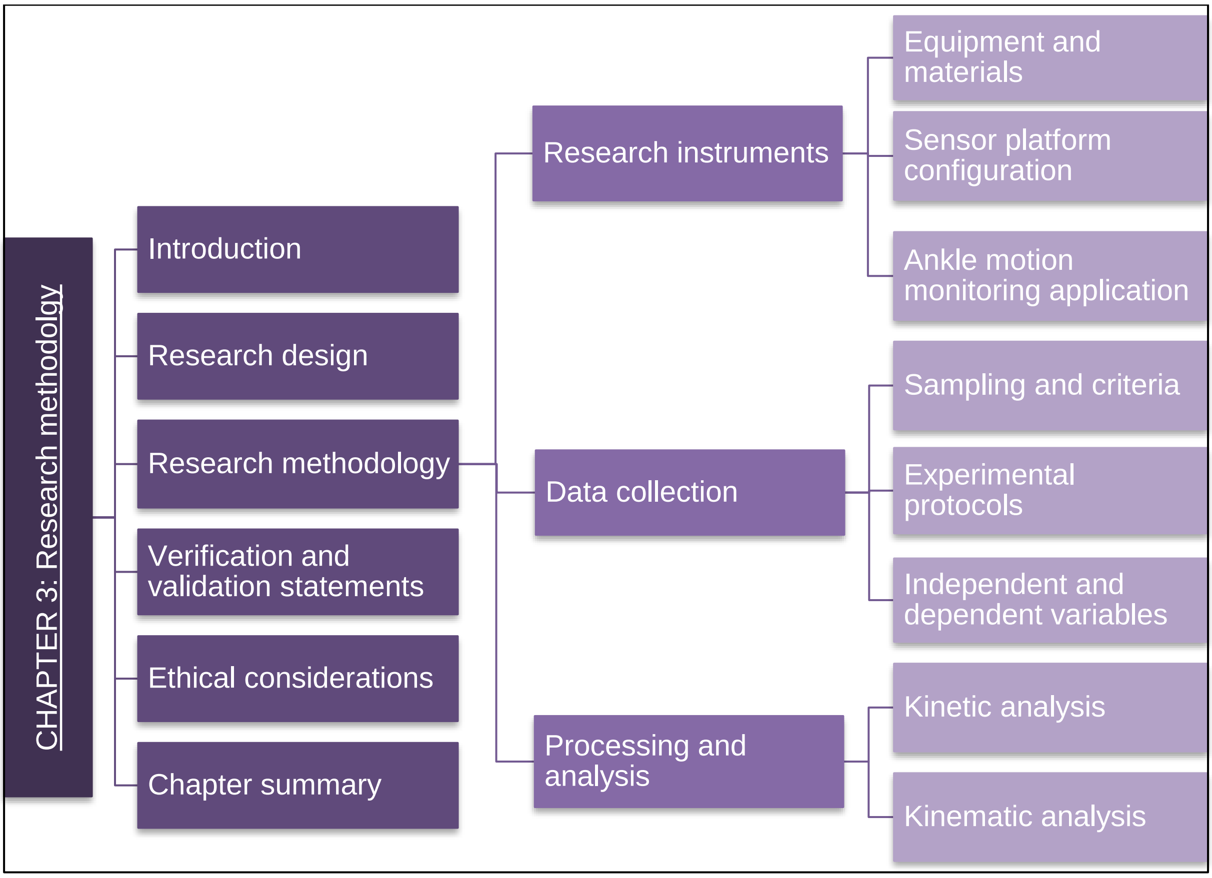

Figure 3.1: Research methodology breakdown

The chapter commences by describing the research design which was carefully selected to establish a solid framework for the study. Following this, the details of the research methodology are presented and deals with the practical implementation of the chosen design. Notably, the research design and methodology work together to ensure the acquisition of reliable, accurate, and contextually relevant data. This methodology includes three key sections that focus on the research instruments, data collection considerations, and processing and analysis techniques. Subsequently, the chapter reflects on the verification and validation aspects of the study, which is used to ensure that certain specifications have been met and that the results of the study fulfil its intended purpose. Finally, due to the nature of the research, the chapter reflects on the ethical guidelines that were adhered to.

3.2 Research design

The research design refers to a conceptual blueprint within which research is conducted and is used to form the outline of data collection and analysis procedures (Leedy & Omrod, 2015). To further this, Hofstee (2006) states that the research design plays a pivotal role in identifying and deliberating on the research approach of the study. In this context, Hofstee (2006) suggests that the research design be based on the benefits and limitations, with supporting discussions from similar literary work.

An **experimental** research design is a deductive approach characterised by its controlled methodology and ability to maximise the precision of findings to draw specific conclusions. As such, this research design is often used to ascertain the effect of an independent variable on a controlled dependent variable (Sirisilla, 2023). It was found that this design was substantially relevant in several of the previously examined gait studies of the literature review, with the following distinct benefits and limitations.

❖ *Benefits*

A high level of control in this research design allows the researcher to select and isolate specific variables for outcome determination. This feature was demonstrated in the studies by Homan *et al.* (2022) and Xie *et al.* (2022), where both groups were able to determine and isolate the specific variables that would be most appropriate to their research. Furthermore, through a high level of control, this research design is able to simplify the establishment of cause-and-effect relationships to draw specific conclusions.

Another benefit of this research design is its compatibility with additional methods in an attempt to ensure the accuracy of findings, often through verification and validation studies. Several of the previously reviewed works, including the studies conducted by Shih *et al.* (2013) and Godfrey *et al.* (2015), illustrate the use of comparative analysis techniques to compare experimental data with reference data, thereby enhancing the reliability of results.

- *Limitations*

  A limitation of an experimental research design is that it is susceptible to human error of participants and researchers. As a result, this may negatively impact the accuracy of collected data. This drawback is demonstrated in the study by Homan *et al.* (2022), where 214 of 238 gait cycles were discarded due to improper foot contact with the force plate system.

  An additional limitation of this research design is its vulnerability to real-time events that may introduce external factors into the data collection process. For instance, motion capture cameras may detect a glare from sunlight, leading to potential interference with measurement accuracy, as was noted in the study by Guerra-Filhol (2005).

As highlighted, the experimental research design's flexibility in manipulating variables and methods offer distinct advantages in determining outcomes and enhancing the accuracy of data. Applied to the present study, this research design provided the means to isolate specific variables and parameters. To mitigate the potential drawbacks associated with an experimental design, meticulous measures were taken to establish a controlled experimental environment with minimal external influences. This was through the presence of a qualified biokineticist to reduce human error and ensure that laboratory equipment was set up in a suitable location.

3.3 Research methodology

The research methodology refers to the general approach used by the researcher to conduct a study and guides the selection of specific tools (Leedy and Omrod, 2015). Therefore, this section offers a detailed insight into how the research design is applied within the study, with particular attention to three vital subcategories, as identified by Hofstee (2006). These encompass the research instruments, data collection considerations, and processing and analysis techniques.

3.3.1 Research instruments

Hofstee (2006) defines the research instruments as any test or tool employed for data acquisition. In the context of this study, a specialised sensor platform was developed to capture angular data relating to a user's foot and ankle joint. The ensuing sections provide an exploration of the instrumentation specifics in terms of equipment and materials, sensor platform configuration, and application interface developed to facilitate the data collection process.

3.3.1.1 Equipment and materials

The experimental equipment used in this study consisted of the sensor platform equipment and laboratory equipment available at the PhASRec facility. Detailed information regarding these components is made available in Table 3.1 below.

Table 3.1: Experimental equipment

	Equipment	*Quantity*	*Purpose*
Sensor platform equipment	ADXL 345 sensor	3	Three inertial sensors were used for collecting joint angle data during the various trials.
	Motherboard	1	This component served as the central platform to which the sensors connected, facilitating data transfer between the sensor and application interface via a Bluetooth connection.
	Power source	1	A power bank was used to power the sensor platform.
	Connecting wire	3 x ±1.2 m	Connecting wire was required to connect the sensors to the motherboard.
	Laptop	1	A laptop was required to operate the MATLAB® application.
Laboratory equipment	AMTI BP400600 force plates	3	Force plates were used to gather force data during trials (sampling frequency = 2 kHz).
	Oqus 300+ camera	8	The cameras were used to track motion during trials (sampling frequency = 200Hz).
	Reflective markers	22	These were placed on various locations on the participant to assist the cameras in accurately tracking motion.
	Laptop	1	A secondary laptop was required to operate the Qualisys motion analysis application.

Electronics housing components were designed on Siemens NX to safeguard the delicate electronics of the sensor platform, i.e., the sensors, the motherboard, and the power source (Annexures A to D). Additional materials included an elastic belt to fit around the participant's waist, and bandage wrap and adhesive electrodes for attaching the sensors to various locations on the lower limb.

3.3.1.2 Sensor platform configuration and alignment

The configuration of the sensor platform is portrayed in the Figure 3.2.

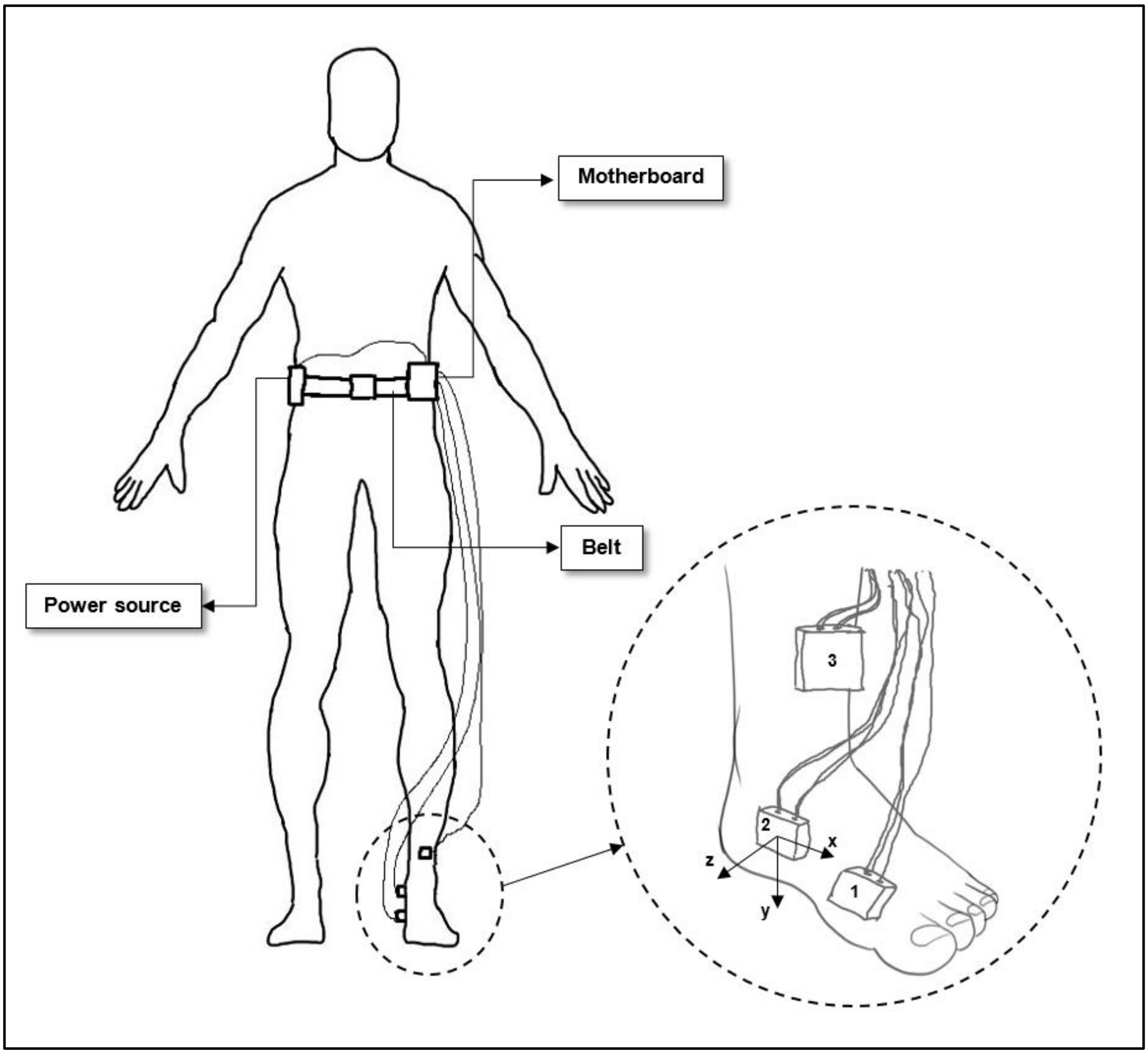

Figure 3.2: Sensor platform configuration

The motherboard and power source housings were designed to easily attach to the elastic belt fitted around the participant's waist, as depicted in Figure 3.2. This figure also shows the three ADXL 345 sensors and their placement on the left foot and lower left shin, with wired connections to the motherboard. This can be more easily observed in the provided close-up view. Furthermore, the sensors were meticulously labelled from one to three, with sensor one and two in the medial position of the foot, and sensor three on the distal shin to complete a comprehensive gait analysis. Sensors one and three specifically captured sagittal movements of the tibiotalar join, while sensors two and three focused on the frontal movements of the subtalar and talonavicular joints. Notably, despite the sensors maintaining a consistent of axis of orientation and static pose calibration protocols, potential errors may arise from misalignment due to human error.

3.3.1.3 Ankle motion monitoring application

The ankle motion monitoring application (AMMA) was developed using MATLAB® R2022b, to serve as bridge between the researcher and the sensor platform. This application facilitated the control of the data flow from the sensor, including the initiation and termination of data generation, as well as data storage. The layout of the AMMA interface showcases an intuitive arrangement, incorporating easily identifiable and well-marked components. Figure 3.3 presents the AMMA interface design, complete with numbered labels for each component.

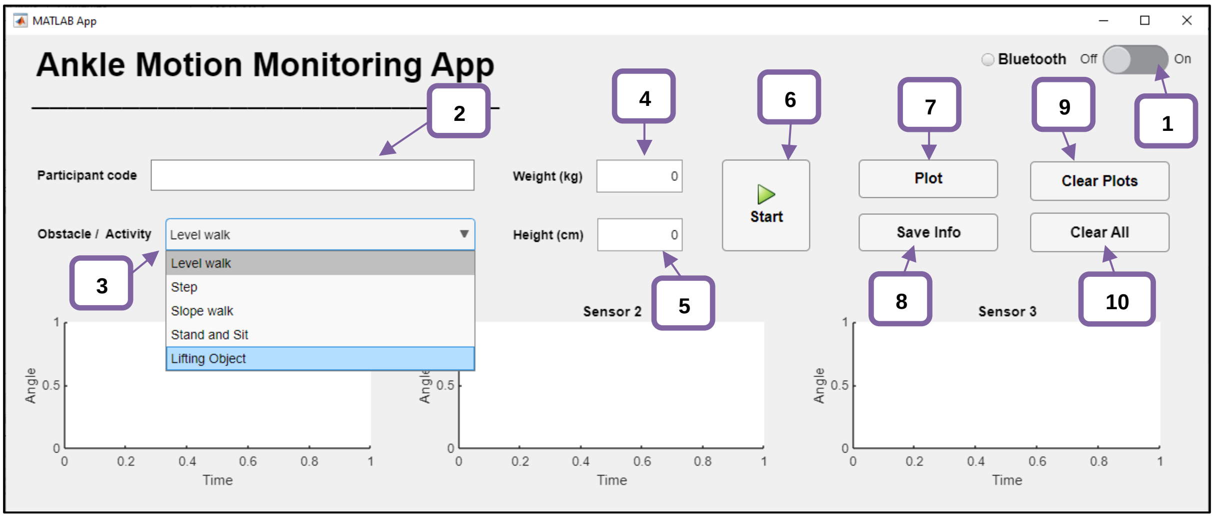

Figure 3.3: Ankle motion monitoring application interface

Table 3.2 expands on the above figure by providing the details of each labelled component.

Table 3.2: Ankle motion monitoring application component list

Label	Component	Function
1	Switch	When activated, the application establishes a Bluetooth connection with the motherboard, and when deactivated, the connection is severed.
2	Edit box (text)	The code and trial number are inputted into this field.
3	Drop down	This drop-down menu is used to select the activity about to commence.
4	Edit box (numeric)	The participant weight is inputted into this field.
5	Edit box (numeric)	The participant height is inputted into this field.
6	Button	This button becomes accessible upon establishing the Bluetooth connection. When activated, it initiates data collection, with the application recording real-time data into a text file. When deactivated, this process terminates.
7	Button	When pressed, this button retrieves and plots the data from the corresponding text file.
8	Button	When pressed, this button writes the participant code, weight, and height to a separate text file.
9	Button	When pressed, this button clears the plots.
10	Button	When pressed, this button clears all data from the interface including plots, participant code, weight, height, and resets the activity menu to the initial level walk trial.

3.3.2 Data collection

This section serves as a crucial component to the research methodology by providing the details of the sampling and criteria, experimental protocols, and key variables of the study. By attending to these foundational elements, the data may be gathered in an optimal format for further analysis, thereby enhancing the overall validity of the study.

3.3.2.1 Sampling and criteria

To inform the design of an improved ankle joint prosthesis, it was imperative to garner knowledge of natural human foot and ankle biomechanics. The sample size was therefore determined based on insights from prior gait studies, specifically those of Shih *et al*. (2013)

and Oliveira *et al.* (2021), which successfully obtained accurate gait data from a sample of 15 participants. Furthermore, given the comparison of gait data to a Qualisys motion capture system, a paired analysis was required. To achieve a type 2 error rate of 20% and statistical power of 80%, it was recommended that no less than 15 steps be taken per activity for studies with 15 to 20 participants (Oliveira *et al*., 2021). In summary, the sample size rationale was as follows:

- A total of 15 participants were recruited for the experiment.
- A total of three clean trials were collected for each activity, with no less than 15 steps per activity when applicable.

For the population group, the following inclusion criteria were defined:

- Male or female participants between the age of 18 and 45.
    - Participants older than 18 can participant in the experimental process without the permission of a guardian. By 7-8 years of age, a natural gait pattern has developed, however, lower limb length continues to increase until 13-15 years (Froehle *et al.*, 2013). Therefore, a minimum age of 18 ensures that a natural and fully developed gait cycle can be measured.
    - From the age of 58 onwards, gait speed decreases due to deterioration of ankle function, which begins at middle-age (32-57 years) (Ko *et al*., 2012). The maximum age considered for this study therefore lies directly in between these ages at 45 to ensure that a typical gait speed can be measured.
- Participants should be free of any injury or recent surgery (less than six months) that may impair gait movements and daily activities.
    - A natural gait needs to be observed to understand the complexities of the human foot and ankle.
- Participants must weigh less than 150kg.
    - This is due to the load cell of the force plate only being able to measure a maximum of 200kg and is more accurate when measuring below this limit.

Furthermore, the following exclusion criterion was defined:

- Any individual that suffers from disorders or diseases that may impair gait activities.

3.3.2.2 Experiment protocols

The experimental protocols were explained to each participant in the presence of a biokineticist and independent person. Participants were ensured that should they not feel comfortable, they were free to withdraw from the study at any time. The general outline of the experimental protocols for each participant was as follows:

1. Introductions and brief explanation of what was expected of the participant, as well as an explanation of the measuring equipment.
2. Signing of the informed consent document.
3. Participant’s height and weight was measured and inputted into the developed MATLAB® application along with their anonymous reference code.
4. Sensors were fitted to the participant’s left foot and the reflective markers were placed in their respective locations.
5. Static pose calibration of the equipment was done.
    i. Participant was required to stand in position with feet shoulder width apart for ±5 seconds.
    ii. The motion capture cameras only focused on the lower part of the body, i.e., from the waist downwards.
6. **Level walk** trials commenced.
    i. This trial made use of the inertial sensors, force plate system and motion capture cameras.
    ii. Participant was required to begin trials with the left foot.
    iii. Real-time data was collected.
    iv. Sensor data as plotted and checked to ensure that a complete trial had been collected.
    v. Data was stored to a .txt file and cleared from the laptop screen.
    vi. Trial was repeated until at least three clean trials were collected.
7. **Step** trials commenced.
    i. This trial made use of the inertial sensors and motion capture cameras.
    ii. Participant stepped up with left foot, paused, and stepped down with left foot.
    iii. Real-time data was collected.
    iv. Sensor data as plotted and checked to ensure that a complete trial had been collected.
    v. Data was stored to a .txt file and cleared from the laptop screen.
    vi. Trial was repeated until at least three clean trials were collected.
8. **Slope walk** trials commenced.
    i. This trial made use of the inertial sensors and motion capture cameras.
    ii. Participant walked up a 20° slope starting with the left foot, paused, turned around, and walked down the slope again starting with left foot.

   iii. Real-time data was collected.
   iv. Sensor data as plotted and checked to ensure that a complete trial had been collected.
   v. Data was stored to a .txt file and cleared from the laptop screen.
   vi. Trial was repeated until at least three clean trials were collected.
9. **Stand and sit** trials commenced.
   i. This trial made use of the inertial sensors and motion capture cameras.
   ii. Participant started at a standing position, took a seat on provided chair, paused, and rose to end the trial in a standing position.
   iii. Real-time data was collected.
   iv. Sensor data as plotted and checked to ensure that a complete trial had been collected.
   v. Data was stored to a .txt file and cleared from the laptop screen.
   vi. Trial was repeated until at least three clean trials were collected.
10. **Lift object** trials commenced.
   i. This trial made use of the inertial sensors, force plate system and motion capture cameras.
   ii. Participant stepped onto the force plate, squatted down to pick up a box, lifted and briefly held position, squatted again to put box down, and ended the trial in a standing position.
   iii. Real-time data was collected.
   iv. Sensor data as plotted and checked to ensure that a complete trial had been collected.
   v. Data was stored to a .txt file and cleared from the laptop screen.
   vi. Trial was repeated until at least three clean trials were collected.
11. Researcher and biokineticist removed sensors and reflective markers from the participant.
12. Reimbursement was provided.
13. Closing remarks in which the participant was thanked.

3.3.2.3 Independent and dependent variables

To distinguish between the independent and dependent variables of the study, the following definitions are outlined by Bhandari (2020):

- The **independent variable** of a study is manipulated or varies during an experimental study to examine its effects on a certain phenomenon and is therefore classified as the cause of an event.
- The **dependent variable**, on the other hand, changes due to manipulation of the independent variable and is therefore classified as the outcome of an examination.

As per these definitions, Table 3.3 below identifies and describe the selected independent and dependent variables of the kinetic and kinematic aspects of the current study.

Table 3.3: Selected variables

	Variable	Unit	Description
Independent variables	Activity	-	This variable represents one of the five activities defined for the experimental protocols.
	Mass	kg	This variable represents the participant's mass.
	Height	m	This variable represents the participant's height.
	t	s	This variable represents the duration of the trial.
Dependent variables	W_p	N	This variable represents the weight of the participant.
	F	N	This variable represents the reaction force experienced by the participant's foot.
	Cadence	steps/min	This variable represents the number of cycles performed in a minute.
	$\theta_{sagittal}$	°	This variable represents the joint angle in the sagittal plane.
	$\theta_{frontal}$	°	This variable represents the joint angle in the frontal plane.
	$\omega_{sagittal}$	°/sec	This variable represents the angular velocity in the sagittal plane.
	$\omega_{frontal}$	°/sec	This variable represents the angular velocity in the frontal plane.

The outcomes of the experimental study are presented in a quantitative format, largely in the form of numerical results and statistical figures.

3.3.3 Processing and analysis

3.3.3.1 Kinetics

To obtain the weight of each participant, the obtained participant mass values were multiplied with the gravitational constant, i.e.,

$$W_p = \text{Mass} \times 9.81 \quad (3.1)$$

This value represents the reaction force acting on a participant in a stationary upright position. Moreover, this weight value was used to indicate a criterion among participants so that force fluctuations during activities could be better identified and understood.

To allow for easier analysis, the three trial force graphs of each participant was synchronised and averaged so that a mean dataset could be obtained. From this dataset, various characteristic peaks and dips could be identified and labelled $F_1, F_2, F_3, \ldots, F_{N-1}, F_N$, where N is the total number of characteristic features present in an activity.

The level walk force graph was further used to determine the cadence of each participant, which was obtained from:

$$\text{Cadence} = \text{Cycle duration} \div 60 \tag{3.2}$$

Notably, cadence is not a kinetic parameter. However, cadence and phase durations were used in a kinetic setting to observe the weight-bearing and weight-shift throughout the gait cycle. Furthermore, to ensure data validity among the trials, four key aspects were deliberated. These were reproducibility, expert review, quality of equipment, and comparison to previous works.

3.3.3.2 Kinematics

The kinematic processing and analysis phase of the study followed a modified framework, based on a model by Wang *et al*. (2016), adapted to align with the current study. This framework, as presented in Figure 3.4, acted as a blueprint for conducting a quantitative gait analysis.

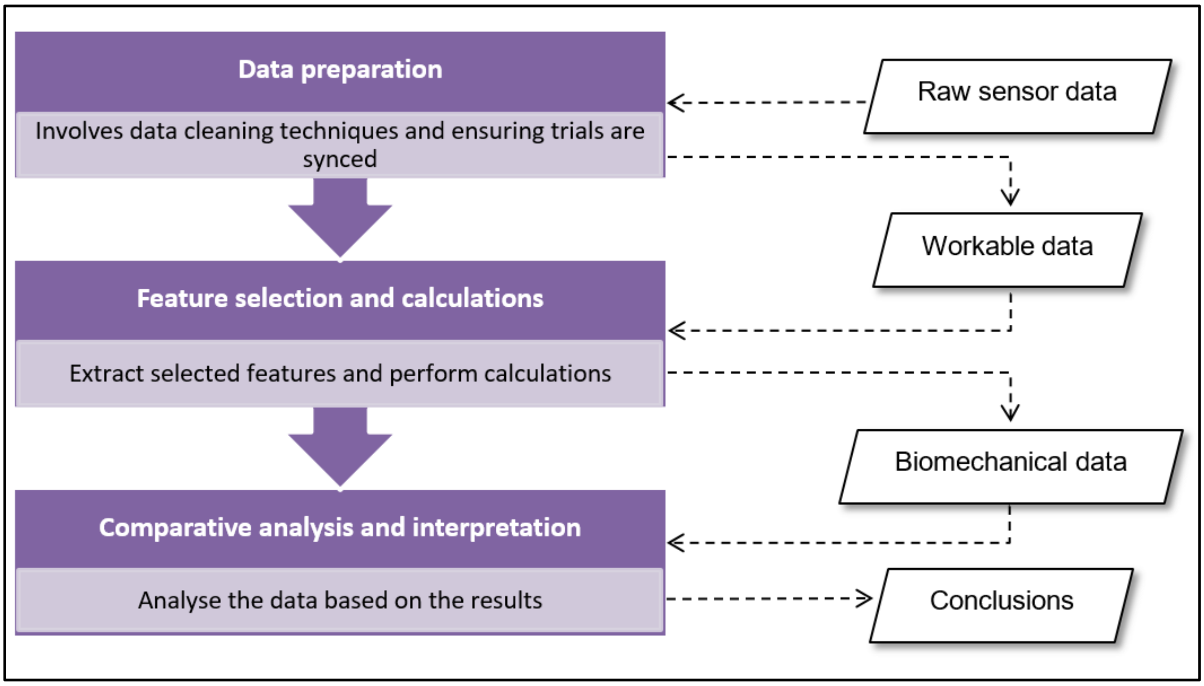

Figure 3.4: Overview of gait analysis framework (based on Wang *et al.,* 2016)

Three key stages could be identified from this figure, namely: data preparation, feature selection and calculations, and analysis and interpretation.

3.3.3.2.1 Data preparation

Data from the sensors were generated at a baud rate of 115 200 bps and stored to a .txt file. As part of the data preparation stage of the analysis, MATLAB® code was written to identify and remove outliers and other irregularities within the data set. Outliers were identified as values lying beyond three standard deviations of the column mean and were replaced by the average of the neighbouring rows' values. A similar corrective approach was implemented for 'not a number' (NaN) values and row irregularities, which were identified as rows of zeros or incomplete data rows. Following this, further refinement procedures were implemented to the acquired data. These procedures included the synchronisation of the clean trials and was followed by the derivation of a mean dataset, which was plotted against time. Notably, this procedure was consistently applied to all datasets.

3.3.3.2.2 Feature selection and calculation

Upon completion of the data preparation process, the sensor data was ready for further calculation and analysis. The calculations encompass the joint angle and angular velocity in the sagittal and frontal planes of the foot, as per the set delineations of the study.

Therefore, the equations for the joint angle were derived from the orientation of the sensor axes, as previously illustrated in Figure 3.3:

$$\theta_{sagittal} = \theta_{sensor\ 1_x} - \theta_{sensor\ 3_z} \tag{3.3}$$

$$\theta_{frontal} = \theta_{sensor\ 3_x} - \theta_{sensor\ 2_z} \tag{3.4}$$

Notably, all joint angle measurements were initially expressed in degrees (°), unless otherwise specified. Furthermore, the equations for angular velocity were derived from equation 2.9, and are as follows:

$$\omega_{sagittal} = \frac{\Delta\theta_{sagittal}}{\Delta t} \tag{3.5}$$

$$\omega_{frontal} = \frac{\Delta\theta_{frontal}}{\Delta t} \tag{3.6}$$

The results obtained from these equations were initially expressed in degrees per second ($°/\mathrm{sec}$), unless otherwise specified.

3.3.3.2.3 Comparative analysis and interpretation

The results were evaluated using the Shapiro-Wilks normality test with an accepted alpha (α) level of less than 0.05. Furthermore, all statistical analyses were conducted using the statistical package of social software (SPSS) and MATLAB® available through the North-West University network.

Furthermore, the sensor results were compared to the reference data obtained from the motion capture camera system. Differences between the datasets were evaluated using the paired t-test of the statistical parameter mapping (SPM{t}) package on MATLAB®, with a set critical threshold (α = 0.05). If the SPM{t} curve exceeded this set threshold, the angular data was considered as significantly dissimilar at the specific time nodes. Data validity was then evaluated using a Bland-Altman analysis with an agreement level of 5% being accepted as valid.

3.4 Verification and validation statements

Verification and validation are processes used in research, development, and testing to ensure the reliability, functionality, and compliance of a solution. The following definitions are provided to distinguish the two (Hamilton, 2019):

- **Verification** is performed to check whether a solution meets set specifications and ensures that the implementation of the solution is consistent with the design.
- **Validation** aims to ensure that a solution fulfils its intended purpose, as well as assess the solution’s performance under real-life conditions.

In accordance with these definitions, the design statements of the study served to verify and validate the research by ensuring certain conditions and concepts were effectively translated into tangible solutions. Furthermore, these design statements are closely linked to the set research purpose and objectives of the study.

The following **verification** design statements were outlined:

1. The sensor platform configuration isolates foot and ankle joint complex movements during various low-level activities related to everyday living.
2. The developed MATLAB® application allows for effective control of the wearable sensor platform.

Validation of the kinetic data was done by examining figures of previous studies and checking for reproducibility. The overall quality of processed kinematic data was established through comparison of sensor data with reference data acquired from the motion capture camera system. Thus, in terms of the **validation** design statements, the following were specified:

1. Force data could be validated against findings of previous studies.
2. Reproducibility could be observed in the collected force data.
3. Comparative analysis could be performed on the level walk activity data.
4. Comparative analysis could be performed on the step activity data.
5. Comparative analysis could be performed on the slope activity data.
6. Comparative analysis could be performed on the stand and sit activity data.
7. Comparative analysis could be performed on the lift object activity data.

3.5 Ethical considerations

This study was approved by the health research ethics committee (HREC), with the following ethical considerations ensured throughout the study to protect the interests of all stakeholders.

❖ *Autonomy*

This study ensured the autonomy of the participants through the equal, fair, and respectful treatment, irrespective of race, cultural values, background, or personal views. Prior to the study, participants provided with an informed consent document (Annexure E), which was available in English, Afrikaans, and Setswana. This document provided information on the study's purpose, criteria, risks, benefits, reimbursement, and details on data storage. Furthermore, participants were notified that should they not feel comfortable, they would be allowed to withdraw from the study at any time.

❖ *Privacy and confidentiality*

This study ensured the privacy and confidentiality of participants by guaranteeing no form of identification appearing in published works upon completion of the study. Each participant was therefore assigned a unique reference code in the place of their name. All data is stored on a password-protected folder and can only be accessed by the researcher and study supervisor. Furthermore, all hard copies of data (signed informed consent documents) are stored in locked cupboards located in the study leader's office at the North-West University, Potchefstroom campus. All data will be stored for seven years, after which it will be destroyed.

❖ *Non-harmfulness*

This study was conducted at the PhASRec facility between the dates of 01/06/2023 to 15/06/2023. Participants were not expected to undergo any sort of severe physical or psychological stress, beyond those of the risks encountered in everyday life. In addition, to ensure that no social pressure was experienced, participants were tested individually. In the case of injury, participants were to be immediately withdrawn from the study and allowed to leave. However, to mitigate this risk the experimental process was headed by the researcher and qualified biokineticist.

- *Justice*

  The identity of each participant was kept confidential using an anonymous reference code in place of their name. Full cooperation and honesty were required from the participants; however, participants were not required to provide any information they wished to keep private. All participant data will be stored in a password protected folder for seven years, after which it will be destroyed.

3.6 Chapter summary

The overarching aim of this chapter was to elaborate on the research design and methodology used to achieve the set purpose and objectives of the study.

In terms of the research design, an experimental research design was selected due to its high level of control and compatibility with additional methods. The research methodology built upon this design and considered three key aspects, specifically the research instruments of the study, data collection considerations, and the procedures followed for the processing and analysis phase. In terms of the research instruments, the chapter elaborated on the equipment and materials used, configuration of the sensor platform, and development of an application to interface to facilitate with data collection. The data collection considerations provided the details of the required participant sample and criteria, experimental protocols, and independent and dependent variables of the study. It was established that the experimental procedure recruited 15 participants, who were required to navigate five low-level activities so that angular data relating to the foot and ankle joint complex could be collected. These activities included a level walk, step up and down, slope walk, stand and sit, and lifting an object. The following section of the research methodology, i.e., the processing and analysis phase, described the techniques and calculations required for drawing meaningful results and conclusions from the data. Finally, the chapter provided the details of the verification and validation aspects to ensure that the study meets set specifications and that data fulfils its intended purpose.

The following chapter presents the kinetic and kinematic results of the experimental procedure by employing the delineated processing and analysis techniques.

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CHAPTER 4

4. RESULTS AND DISCUSSION

4.1 Introduction

The previous chapter served as a bridge between the foundational theories of the literature review, and the practical implementation of the research method. To further this, this chapter transitions into the analysis phase of the study wherein the results of the experiment are discussed. This chapter is broken down in Figure 4.1 below.

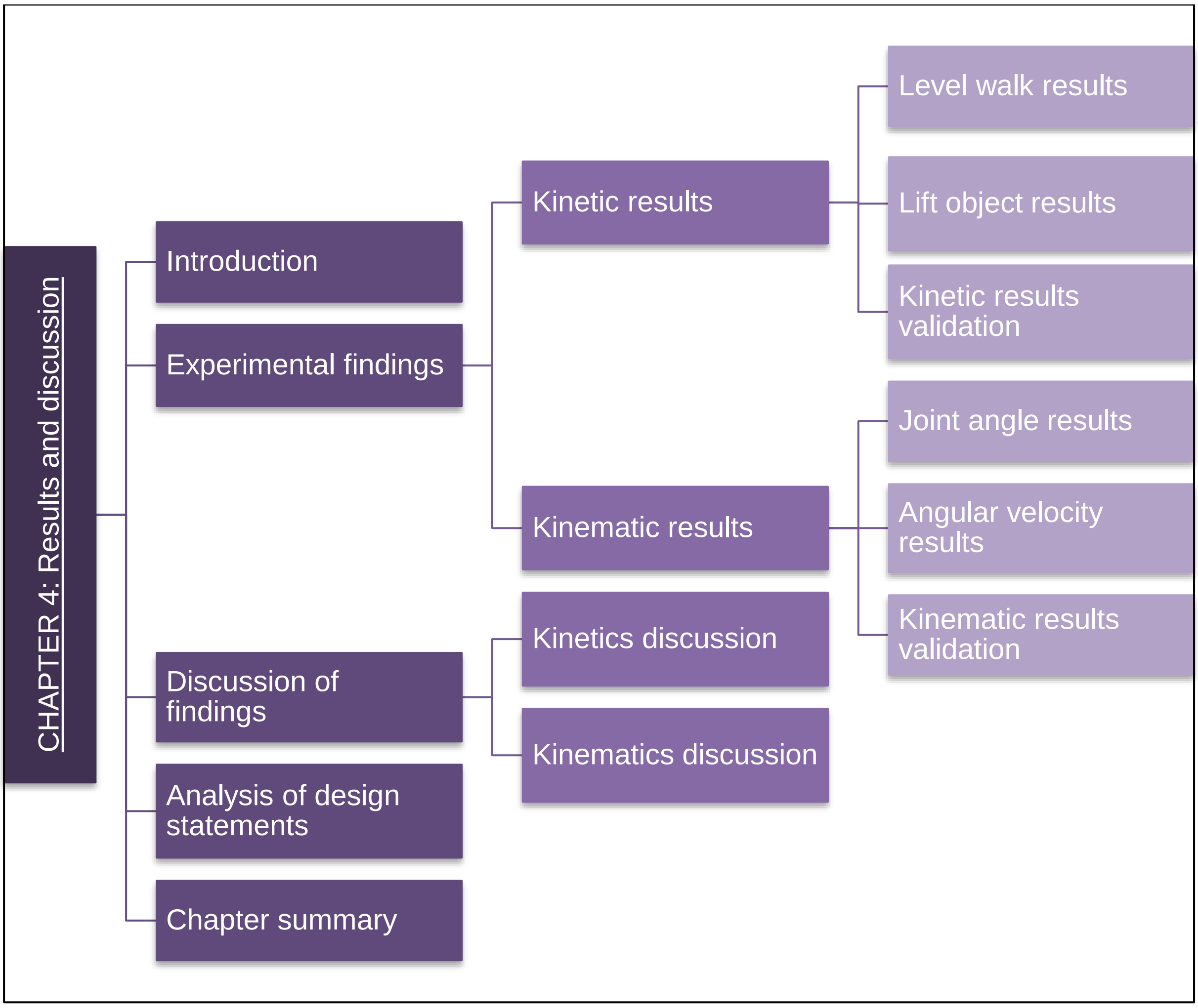

Figure 4.1: Empirical study breakdown

The upcoming data analysis section embarks on a comprehensive examination of the acquired data set. The main body of this chapter is divided into the kinetic and kinematic results, wherein the specifics of the data processing and analysis procedures are presented, and the validation procedures of the experimental data is discussed. Subsequently, to derive meaning from this data, the following section is dedicated to the interpretation and discussion of the research findings. Finally, to ensure that the research aligns with the outlined objectives, the design statements of the study are analysed.

4.2 Experimental findings

4.2.1 Kinetic results

Force plate data could not be collected for all experimental activities, specifically the step, slope, and stand and sit trials. This was due to the force plates' horizontal design and fixed attachment to the ground. Therefore, force plate data is exclusively presented for the level walk and lift object activities. Furthermore, technological faults resulted in the unavailability of force plate data for participants 14 and 15 (P14 and P15), while loss of data issues rendered an additional four participants' data sets unusable. Specifically, these were for participants one, three, four, and 13 (P01, P03, P04, and P13).

Table 4.1 below presents the weight values acquired for the remaining participants. These values were calculated using equation 3.1.

Table 4.1: Participant weight values

Participant code	W_p [N]	Participant code	W_p [N]	Participant code	W_p [N]
P02	539.55	P07	1255.68	P10	814.23
P05	814.23	P08	519.93	P11	814.23
P06	598.41	P09	745.56	P12	539.55

The reaction forces and corresponding participant weight data are crucial components in the analysis of gait kinetics. The ensuing sections provide the detailed findings and of the reaction force data for the two experimental activities.

4.2.1.1 Level walk results

Although three AMTI force plates were available for the level walk activity, the data presented in this section correlates specifically to the first force plate. This plate was selected for analysis since not all plates were consistently utilised throughout the trials, which was attributed to varying stride lengths and human error. The following force data represents the resultant reaction force, i.e., the resultant force experienced in the x, y, and z axes.

Figure 4.2 depicts the force plate data of P02, which illustrates that a constant pattern was obtained for all trials.

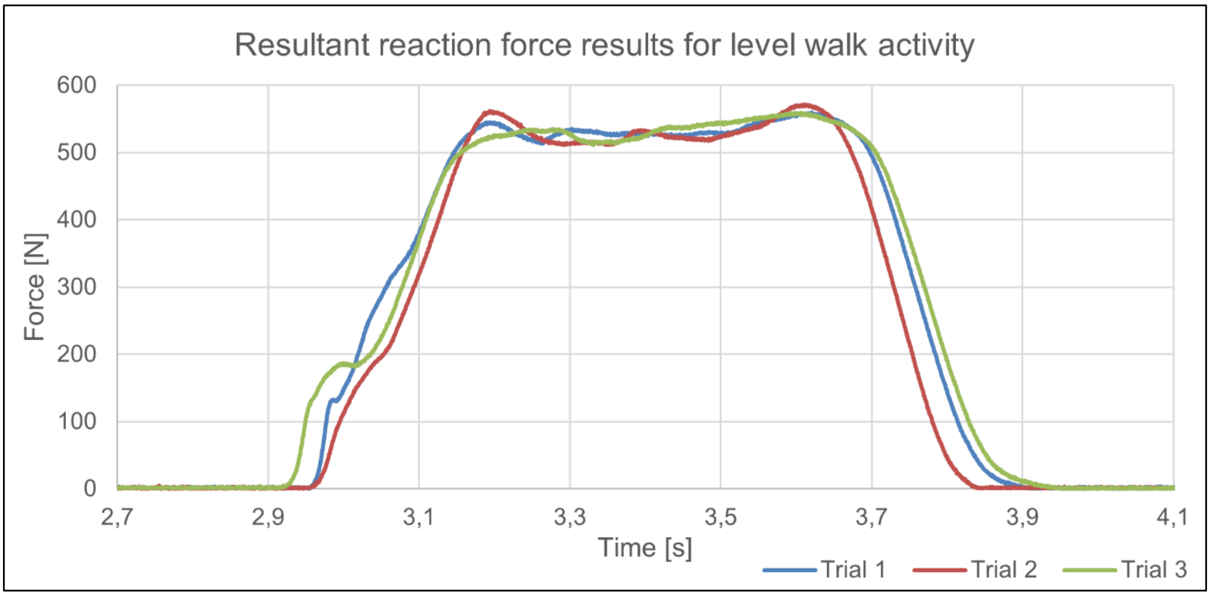

Figure 4.2: P02 – Resultant reaction force results for level walk activity

Figure 4.3 below provides a visual representation of the mean force data for P02's level walk activity. Additionally, this figure highlights the key points that are extracted from the data.

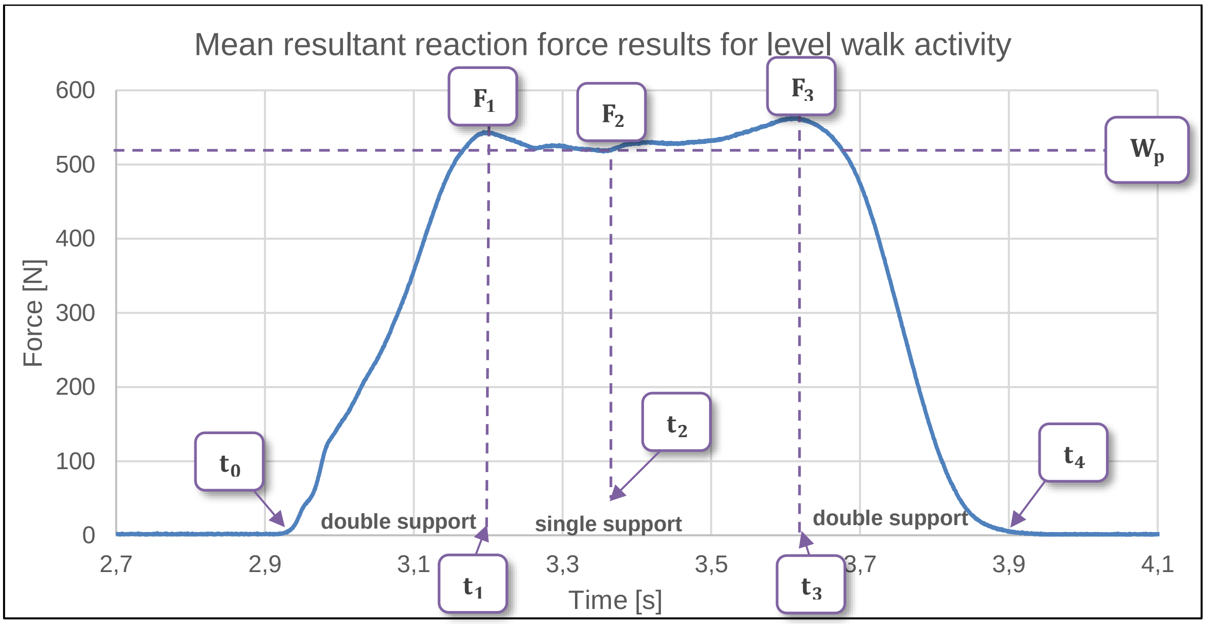

Figure 4.3: P02 – Mean resultant reaction force results for level walk activity

Figure 4.3 offers a clear depiction of the double support and single support phases, aligning with the findings of Bonnefoy-Mazure and Armand (2015). Extracted from this graph are three significant force variables: F_1 and F_3, which characterise the two peaks, and F_2 which indicates the dip between the peaks. Furthermore, these peaks represent the end of the double support I and the start of double support II phases within the stance phase of the gait cycle (Bonnefoy-Mazure and Armand, 2015). Additionally, the timeline of the graph is divided into intervals t_0, t_1, t_2, t_3, and t_4, which further distinguish between the sub-phases within the gait cycle. The variable W_p denotes the weight of the participant, as detailed in Table 4.1. Notably, the values relating to time are expressed to four decimal places to align with the 2 kHz sampling frequency of the force plates.

The data extracted from Figure 4.3 indicates the following values for force:

$F_1 = 543.68$ N, $F_2 = 517.22$ N, and $F_3 = 562.41$ N

and for time:

$t_0 = 2.9320$ s, $t_1 = 3.1935$ s, $t_2 = 3.3550$ s, $t_3 = 3.6065$ s, and $t_4 = 3.8975$ s.

Following a similar rule-based method, the subsequent Table 4.2 displays the values derived for the remaining participants.

Table 4.2: Level walk activity force data

	F_1 [N]	F_2 [N]	F_3 [N]	t_0 [s]	t_1 [s]	t_2 [s]	t_3 [s]	t_4 [s]
P02	543.68	517.22	562.41	2.9320	3.1935	3.3550	3.6065	3.8975
P05	850.82	783.76	851.06	2.7725	3.0385	3.1700	3.4285	3.7420
P06	588.31	564.75	641.91	1.8790	2.1325	2.2465	2.5470	2.8150
P07	1249.24	1114.40	1337.93	2.1425	2.3310	2.4140	2.7195	2.9335
P08	565.55	476.06	559.28	3.0355	3.3080	3.4130	3.6400	3.9545
P09	554.80	552.45	773.13	2.3635	2.7475	2.7565	2.9995	3.2465
P10	834.63	790.38	804.52	3.3055	3.5890	3.6865	3.7565	4.3410
P11	839.12	796.11	843.61	2.8100	3.1625	3.2805	3.5995	4.7700
P12	560.53	547.04	561.38	3.0940	3.6765	3.8660	4.2520	4.7685

From Table 4.2, it is evident that $F_1 > F_2$ and $F_3 > F_2$, indicating that all graphs have similar shape.

Table 4.3 presents the percentage of the double support and single support sub-phases within the gait cycle, as well as the overall duration and cadence, which is the cycle duration divided by the number of seconds in a minute.

Table 4.3: Level walk sub-phases and cadence

	Double support I [%]	*Single support [%]*	*Double support II [%]*	*Cycle duration [s]*	*Cadence [steps/min]*
P02	27.0844	42.7758	30.1398	0.9655	62.1440
P05	27.4369	40.2269	32.3363	0.9695	61.8876
P06	27.0833	44.2842	28.6325	0.9360	64.1026
P07	23.8306	49.1150	27.0544	0.7910	75.8534
P08	29.6518	36.1262	34.2220	0.9190	65.2884
P09	43.4481	28.5391	27.9728	0.8830	67.9502
P10	27.3781	16.1758	56.4462	1.0355	57.9430
P11	17.9847	22.2959	59.7194	1.9600	30.6122
P12	34.7865	34.3685	30.8450	1.6745	35.8316

The findings of Table 4.3 exhibit the diverse cadence range among the participants and varies from ±30 steps/min to ±75 steps/min, with the average cadence equating to approximately 58 steps/min. Furthermore, in some cases the double support I phase is predominant, while double support II is more pronounced in others.

4.2.1.2 Lift object results

This activity required participants to step onto the force plate, squat down to pick up an object, pause, and then perform a second squat to place the object back down. As with the earlier analysis, this section examines the resultant of the reaction forces experienced in the x, y, and z axes.

Figure 4.4 depicts the force plate data recorded during P02's lifting object activity. This figure reveals a consistent pattern across all trials.

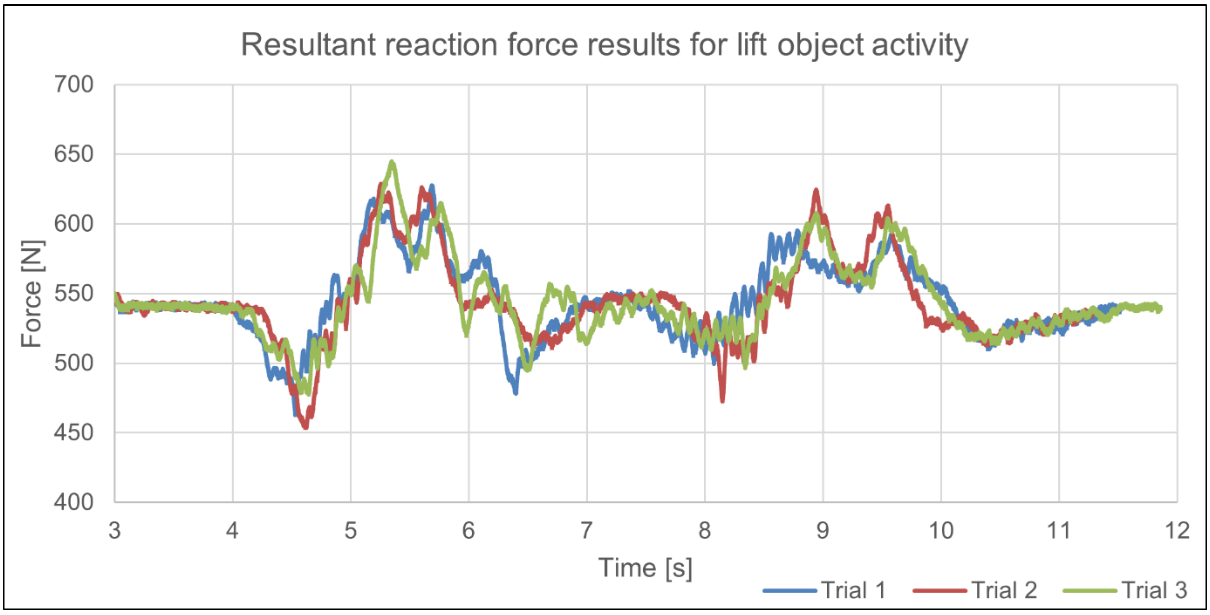

Figure 4.4: P02 – Resultant reaction force results for lift object activity

Figure 4.5 portrays the mean plot for P02's lifting object activity. Additionally, the figure highlights the key points that are extracted from the data.

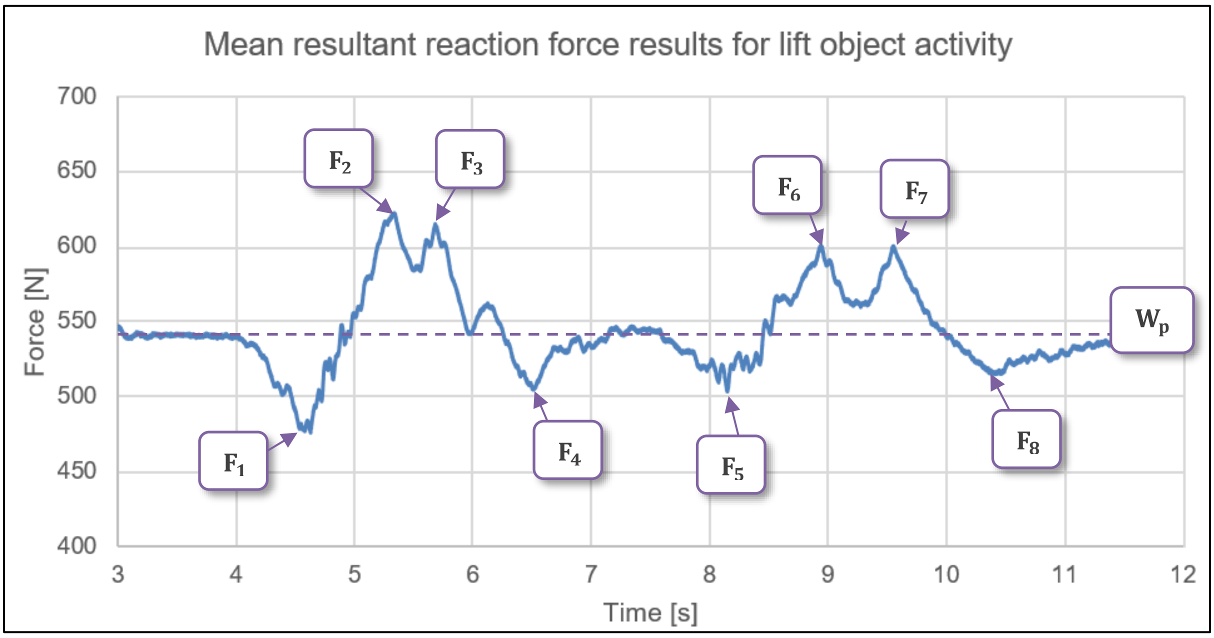

Figure 4.5: P02 – Mean resultant reaction force results for lift object activity

Extracted from this graph are eight values that represent the characteristic force peaks and dips within the lift object activity, these are:

$F_1 = 475.95$ N, $F_2 = 620.81$ N, $F_3 = 611.19$ N, $F_4 = 506.36$ N, $F_5 = 503.34$ N, $F_6 = 597.78$ N, $F_7 = 596.50$ N, and $F_8 = 515.26$ N.

These characteristic features were observed across all participant trials and are summarised in Table 4.4 below.

Table 4.4: Lift object activity force data

	F_1 [N]	F_2 [N]	F_3 [N]	F_4 [N]	F_5 [N]	F_6 [N]	F_7 [N]	F_8 [N]
P02	475.95	620.81	611.19	506.36	503.34	597.78	596.50	515.26
P05	646.52	983.56	1036.49	681.76	642.76	1016.70	1000.34	706.69
P06	555.01	672.28	761.40	498.55	553.68	691.34	716.76	526.63
P07	1009.49	1500.33	1505.16	1088.03	986.11	1490.64	1438.33	1118.51
P08	441.50	594.75	634.33	477.46	458.23	487.15	587.05	479.91
P09	473.59	955.03	991.11	531.63	548.86	906.95	941.57	556.75
P10	688.78	958.43	1027.04	697.01	714.48	946.82	1001.50	712.30
P11	474.26	612.63	642.20	514.22	494.35	615.99	621.90	509.66
P12	722.26	917.42	912.07	745.39	757.37	870.51	948.12	722.26

The values in Table 4.4 reveal that all datasets have similar shape by providing the values at the specific marker. It can be observed that all force values drop below the participant weight as the activity is initiated and continues to fluctuate similarly throughout.

4.2.1.3 Kinetic results validation

In terms of result validation, the following statements can be made on the four key aspects previously specified.

- *Reproducibility*

  The credibility of the kinetic results is ensured through its reproducibility. This was apparent in the constant pattern obtained across trials, as well as in the mean datasets.

- *Expert review*

  The raw kinetic data was collected and prepared for further analysis by a qualified biokineticist, thus ensuring a high standard of data.

- *Quality equipment*

For the data collection, an AMTI BP400600 force plate was used, which is a well-established model with a high sampling frequency of 2 kHz.

❖ *Comparison to previous works*

The data obtained from the level walk activity resembled the findings of Bonnefoy-Mazure and Armand (2015) presented in the background research.

4.2.2 Kinematic results

To process and analyse the kinematic results obtained from the sensor, the study makes use of the adapted gait analysis framework, as presented again in Figure 4.6. Notably, as with the kinetic results, data loss issues rendered four participants' data sets unusable, specifically: P01, P03, P04, and P13.

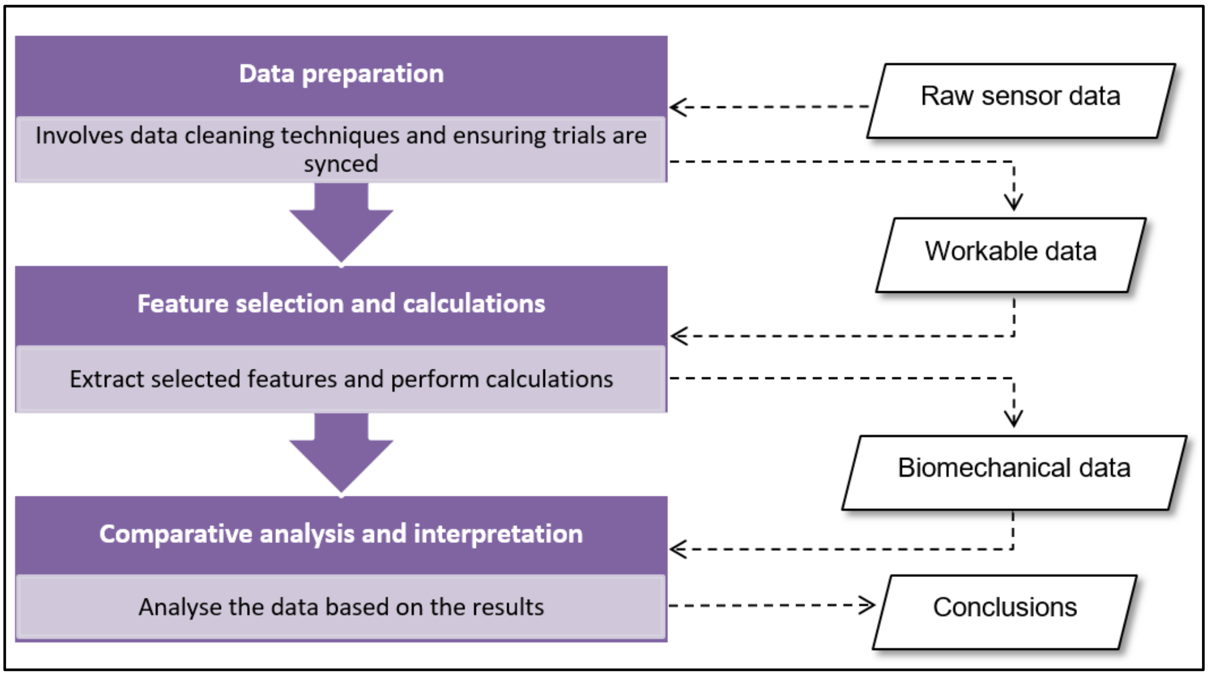

Figure 4.6: Adapted gait analysis framework

Following this framework, the data preparation stage involves the implementation of the data refinement procedures previously outlined. For a visual representation of these procedures, the ensuing figures showcase the raw data obtained during P02's lifting object trials.

Figure 4.7 presents the plotted z-component data recorded from sensor three which is located on the participant's distal shin. From this, it is evident that while all trials are complete, they have not been synchronised.

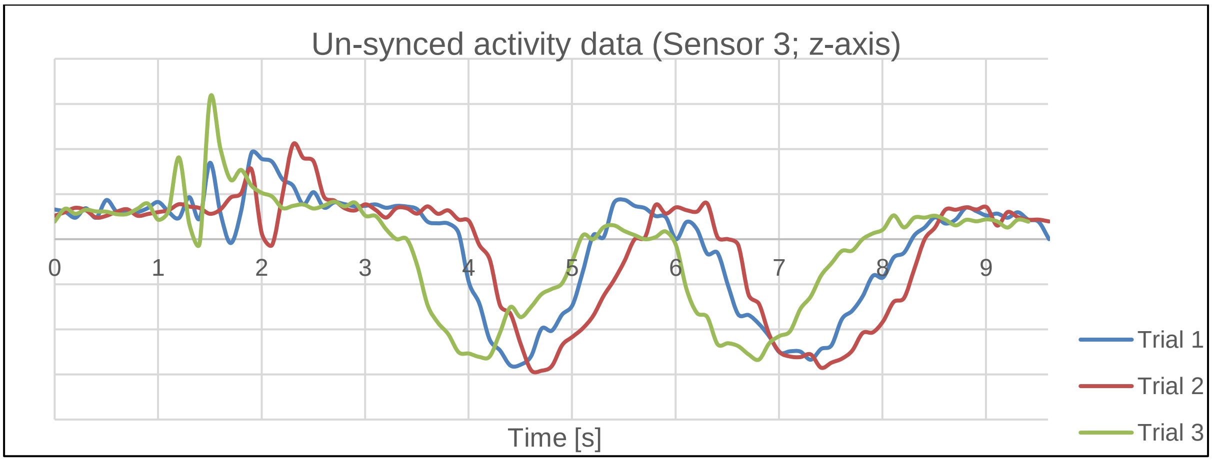

Figure 4.7: Un-synced trial data

In Figure 4.8, the data is presented in a time-synchronised format which shows a significantly improved overlap between the three trials when compared to the previous figure.

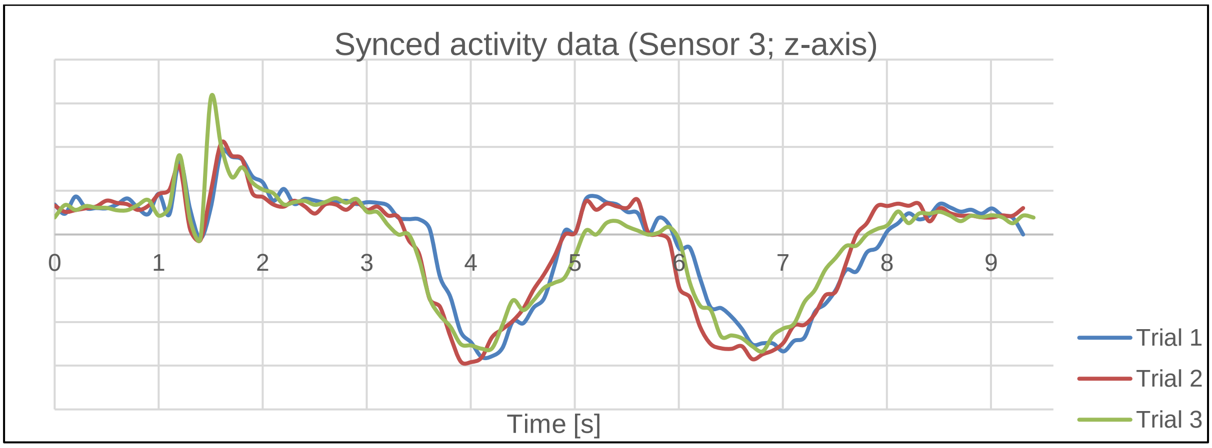

Figure 4.8: Synced trial data

Subsequently, Figure 4.9 displays the mean dataset for P02's lift object activity.

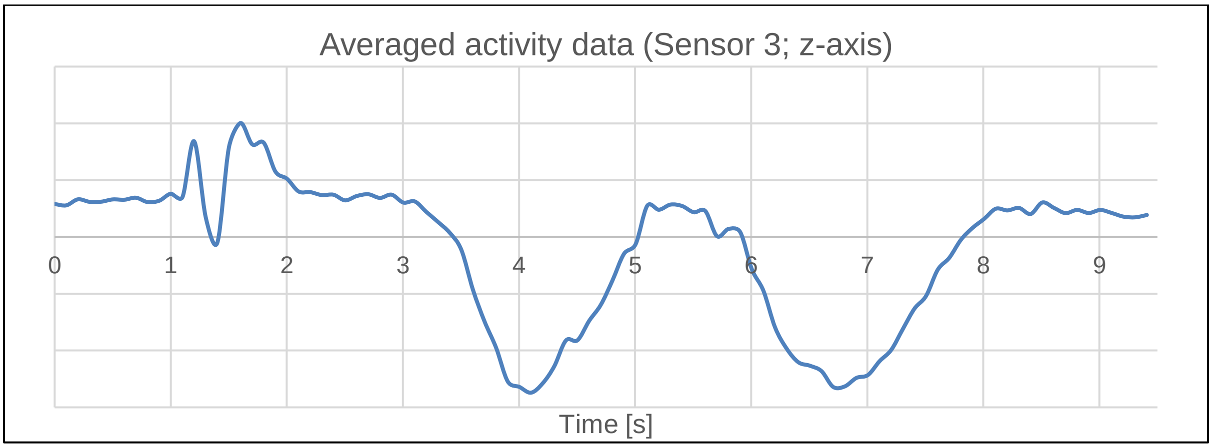

Figure 4.9: Mean trial data

This mean data provides a foundation for conducting further calculations with increased accuracy. It should be reiterated that the figures only provide the data relating to the z-axis of sensor three, however, the refinement process was applied to all the collected data.

4.2.2.1 Joint angle results

The results presented here describe the joint angles obtained in the sagittal plane, which comprise dorsiflexion and plantarflexion movements, and the frontal plane, which comprise inversion and eversion movements. These results were obtained from equations 3.3 and 3.4, respectively, where the required inputs could be retrieved from the refined data.

In continuation with the previous data preparation, Figure 4.10 presents an extract of the Excel calculations and results for P02's lift object activity.

	A	B	C	D	E	F	G
1		P02					
2		Sagittal			Frontal		
3	Time	Calculation	Offset	Smooth	Calculation	Offset	Smooth
4	0.00	1.450	0.000	0.000	-3.070	0.000	0.000
5	0.10	-0.147	1.417	0.992	-2.553	-0.073	-0.051
6	0.20	-1.203	3.460	2.720	-3.720	-0.470	-0.344
7	0.30	0.140	4.113	3.695	-3.557	0.857	0.496
8	0.40	-0.320	5.880	5.225	-3.543	0.455	0.467
9	0.50	-0.063	11.302	9.479	-2.207	2.493	1.886
10	0.60	0.137	18.403	15.726	-3.160	4.830	3.947
11	0.70	-0.893	23.530	21.189	-2.120	4.597	4.402
12	0.80	-1.037	28.593	26.372	-4.277	6.037	5.546
13	0.90	0.197	31.100	29.682	-3.670	5.047	5.197
14	1.00	-0.237	34.627	33.143	-5.423	5.653	5.516

Figure 4.10: Excel extract for joint angle calculations

This figure displays the calculated results for the sagittal and frontal planes. The results for all participants were again time-synchronised to ensure that all trials began and ended at the same instance, and the offset could be removed. Moreover, to eliminate oscillations in the data, an exponential smoothing filter with a smoothness factor of 0.7 was applied. From there, the activity average could be calculated, which is the average of all participant datasets.

The following figures illustrate the mean data plots for the level walk, step, slope, stand and sit, and lift object activities. Notably, a positive value indicates dorsiflexion or inversion depending on the plane considered. Similarly, a negative value indicates either plantarflexion or eversion.

Figure 4.11 shows the mean data for the **level walk** activity. This figure shows that on average, sagittal plane movement encompassed ±10° dorsiflexion and ±8° plantarflexion. Frontal plane movement mainly involved inversion movement with a maximum of ±5°, however minor eversion may be expected.

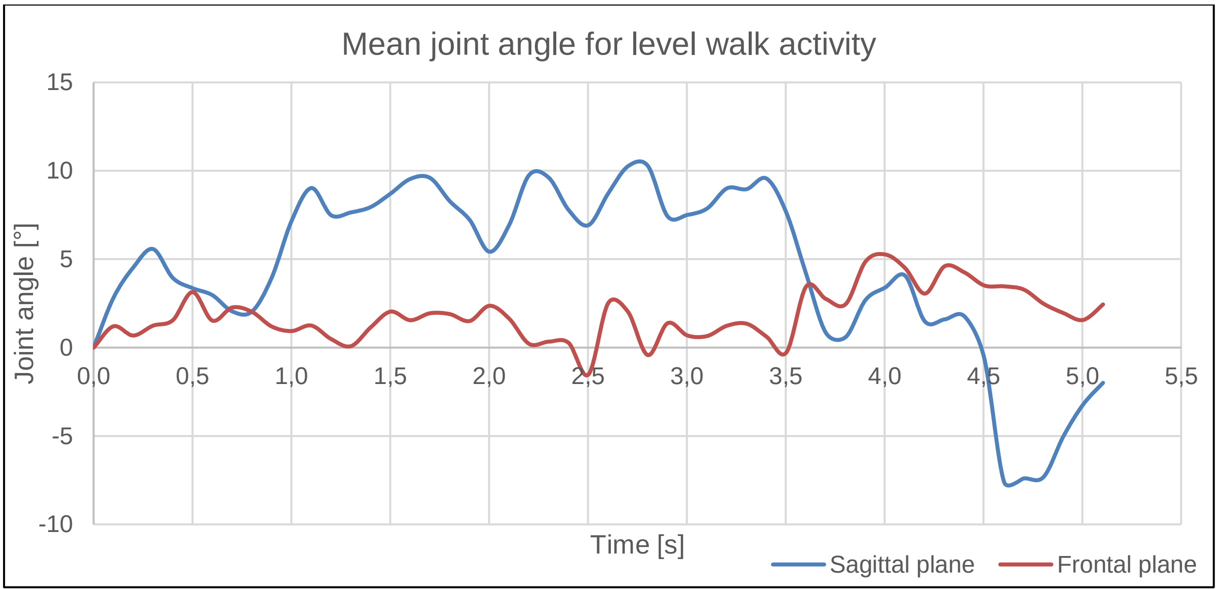

Figure 4.11: Mean joint angles for level walk activity

Figure 4.12 shows the mean data for the **step** activity. This figure shows that average sagittal plane movement for this activity only comprised dorsiflexion with two distinct peaks at ±23° and ±41°, representing the step-up and step-down phases respectively. The frontal plane primarily included inversion movement with an mean maximum of ±5°.

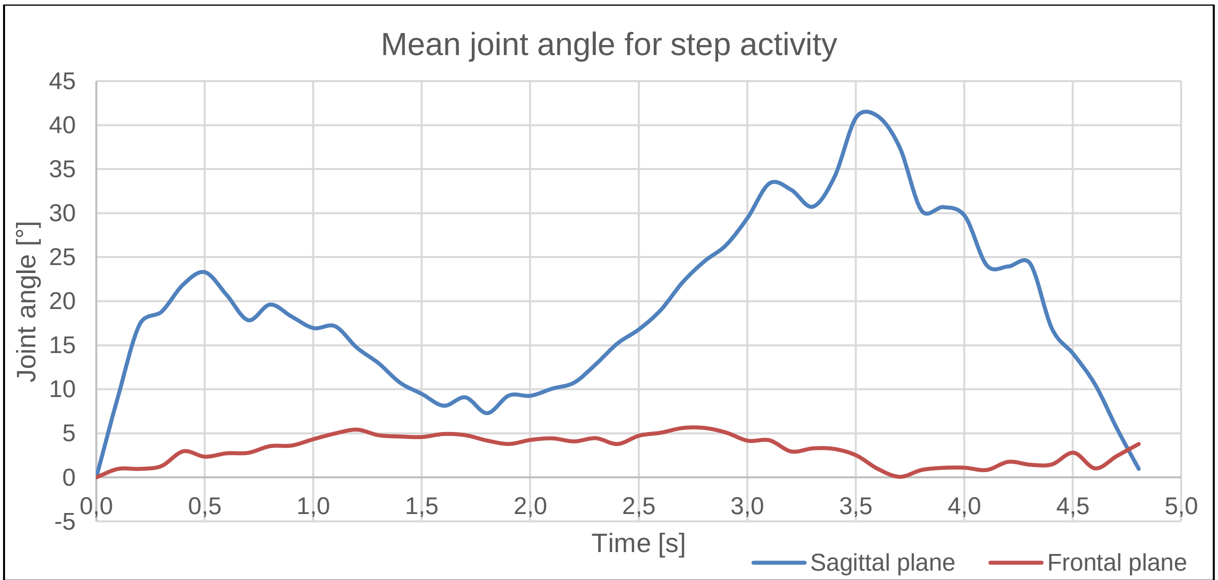

Figure 4.12: Mean joint angles for step activity

Figure 4.13 shows the mean data for the **slope** activity. The figure shows that sagittal plane movement typically lies between ±14° dorsiflexion and ±17° plantarflexion; where dorsiflexion is present as the participant walks up the slope and plantarflexion is present as the participant walks down. Movement in the frontal plane comprises ±4° eversion and ±6° inversion.

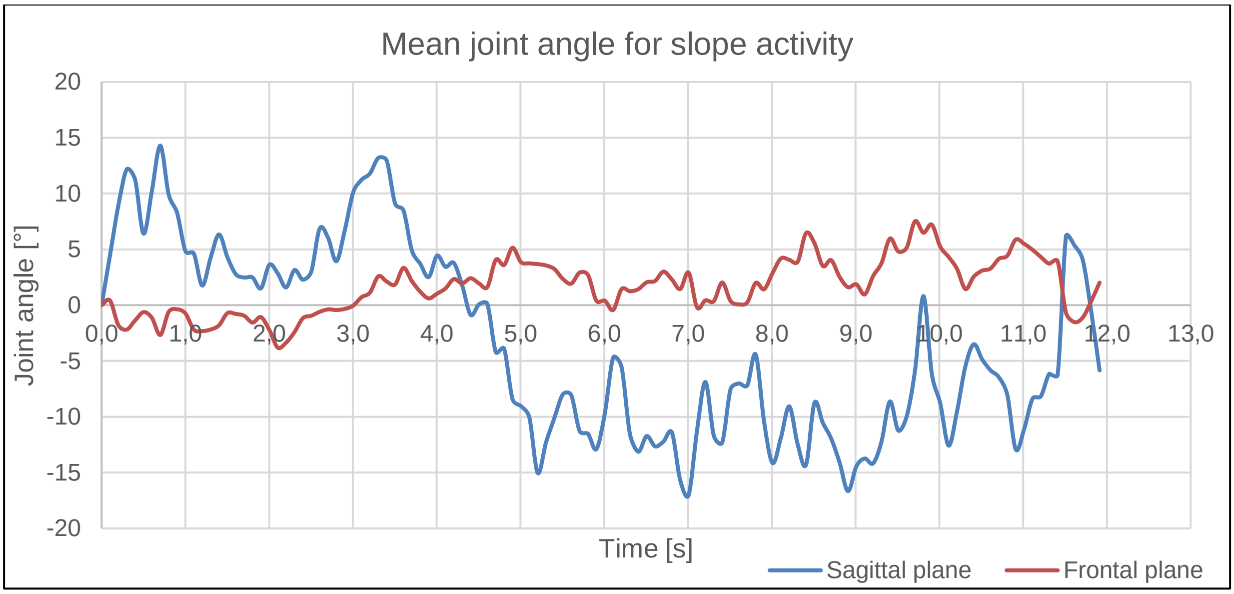

Figure 4.13: Mean joint angles for slope activity

Figure 4.14 shows the mean data for the **stand and sit** activity. Sagittal plane movement for this activity primarily consists of dorsiflexion, with an average maximum of ±14°. For the frontal plane movement, eversion was indicated to be ±2° while the overall inversion was ±3°.

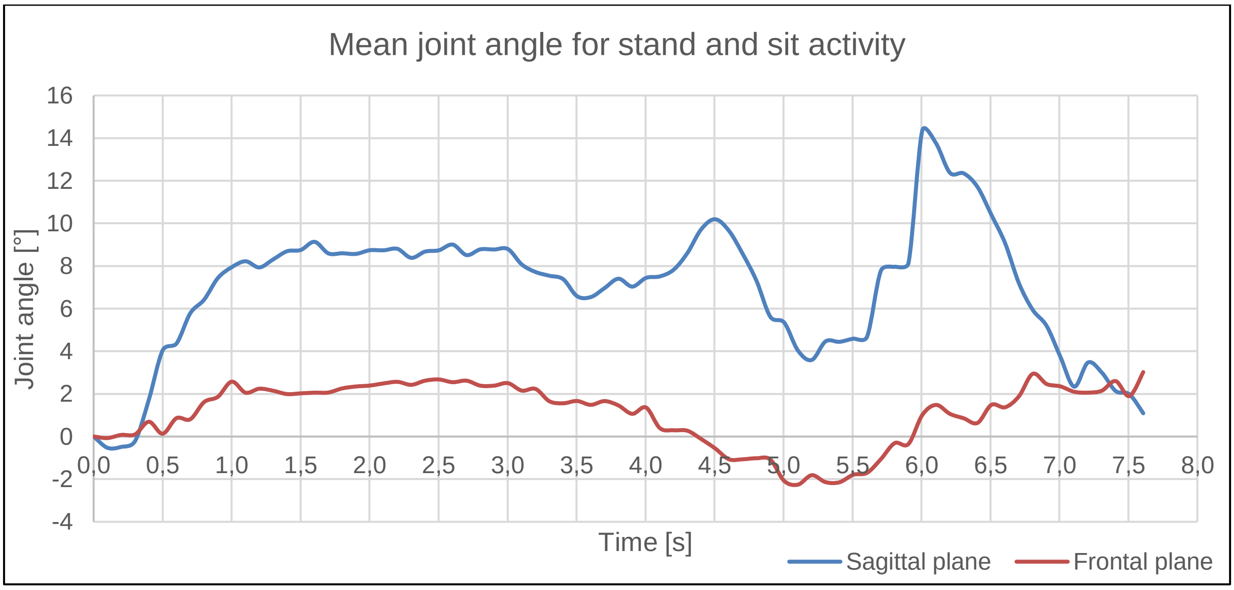

Figure 4.14: Mean joint angles for stand and sit activity

Figure 4.15 shows the mean data for the **lift object** activity. The figure illustrates two distinct peaks in the sagittal plane, which represent the two squat movements the participant performs to lift and put down an object. On average, movement for the downward motion indicates a maximum of ±30°, while the second downward motion has a lower average of ±25°. The frontal plane movement somewhat resembles the pattern of the sagittal plane data and indicates an mean maximum of ±5° inversion.

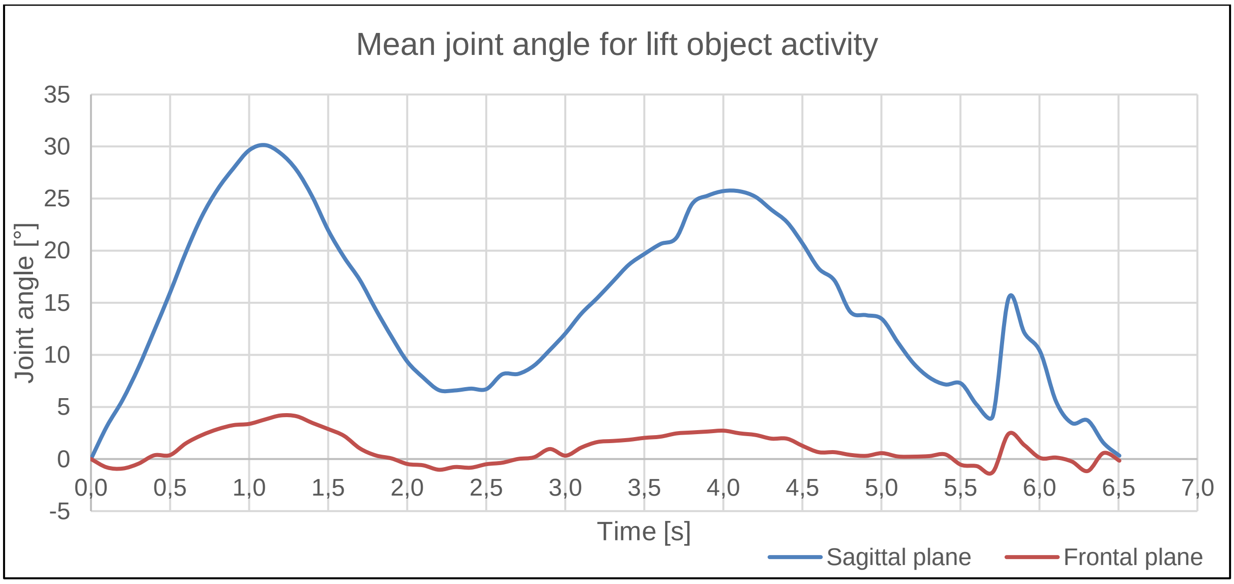

Figure 4.15: Mean joint angles for lift object activity

The data presented in these figures denote the mean data determined across all trials. Consequently, variations in overall trial duration and execution of trials by the participant affect the outcomes of mean data and thus the patterns presented in the above figures. Therefore, these illustrations should only be used to provide a general indication of the individual movement patterns. Furthermore, to ensure accuracy, the findings should be compared to reference data obtained from the motion capture system.

4.2.2.2 Angular velocity results

The other kinematic consideration of this study entails the angular velocities present in the foot and ankle joint complex during the delineated activities. Like the previous discussion, this section describes the values obtained in the sagittal and frontal planes, which were determined using equations 3.5 and 3.6, respectively. The required inputs for these equations were obtained from the refined data.

Extending on P02's lift object activity, Figure 4.16 offers an excerpt of the Excel calculations and results obtained.

	A	B	C	D	E	F	G
1		P02					
2		Sagittal			Frontal		
3	Time	Calculation	Offset	Smooth	Calculation	Offset	Smooth
4	**0.00**	-15.951	0.000	0.000	5.162	0.000	0.000
5	**0.10**	-10.556	6.260	4.382	-11.655	-3.230	-2.261
6	**0.20**	13.420	-7.626	-4.023	1.632	13.986	9.112
7	**0.30**	-4.595	3.497	1.241	0.133	-3.280	0.438
8	**0.40**	2.564	40.010	28.379	13.353	21.096	14.898
9	**0.50**	1.998	56.793	48.269	-9.524	24.076	21.323
10	**0.60**	-10.290	37.063	40.425	10.390	-1.598	5.278
11	**0.70**	-1.432	36.430	37.629	-21.545	15.118	12.166
12	**0.80**	12.321	10.889	18.911	6.061	-9.158	-2.760
13	**0.90**	-4.329	21.079	20.429	-17.516	6.793	3.927
14	**1.00**	-8.616	-2.098	4.660	14.985	12.388	9.849

Figure 4.16: Excel extract for angular velocity calculations

Following a similar processing method as that of the joint angles, the average data plots could be determined for the angular velocity results. The subsequent figures depict the mean data for the level walk, step, slope, stand and sit, and lift object activities. A positive angular velocity indicates that a joint is undergoing extension or moving in an open angle. Likewise, a negative angular velocity signifies flexion or movement in a closing angle.

Figure 4.17 shows the mean data for the **level walk** activity. The figure shows that the average sagittal plane angular velocity lies between ±80°/s flexion and ±40°/s extension. In terms of the frontal plane, the angular velocity lies between ±50°/s flexion and ±30°/s extension.

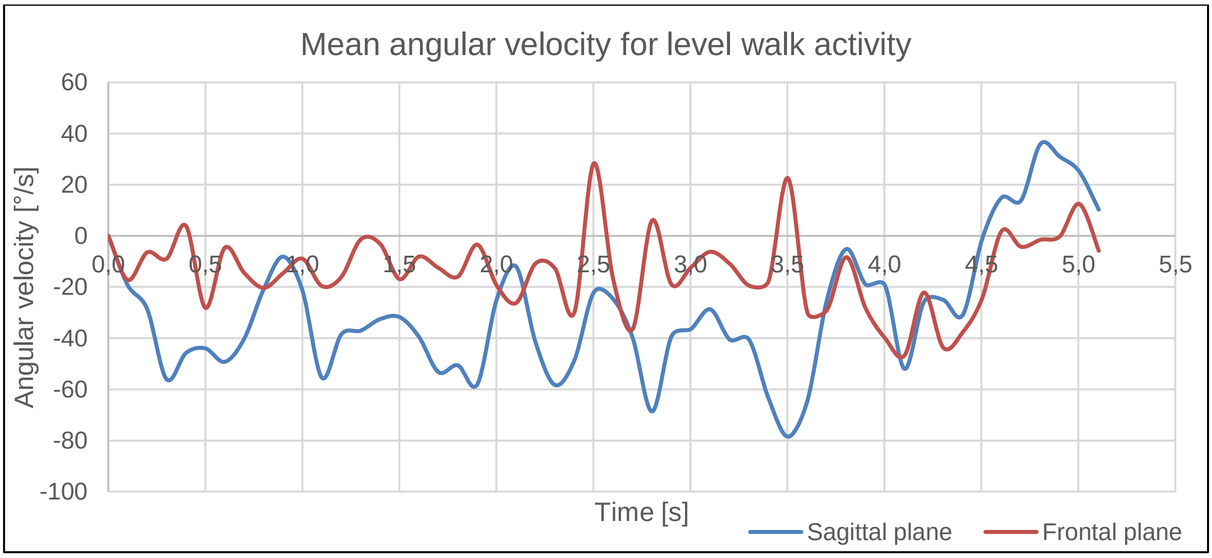

Figure 4.17: Mean angular velocity for level walk activity

Figure 4.18 shows the mean data for the **step** activity. In terms of the sagittal plane, the step activity typically demonstrates average flexion at ±210°/s. On average, angular velocity in the frontal planes is characteristically constant throughout the activity, not exceeding ±25°/s extension.

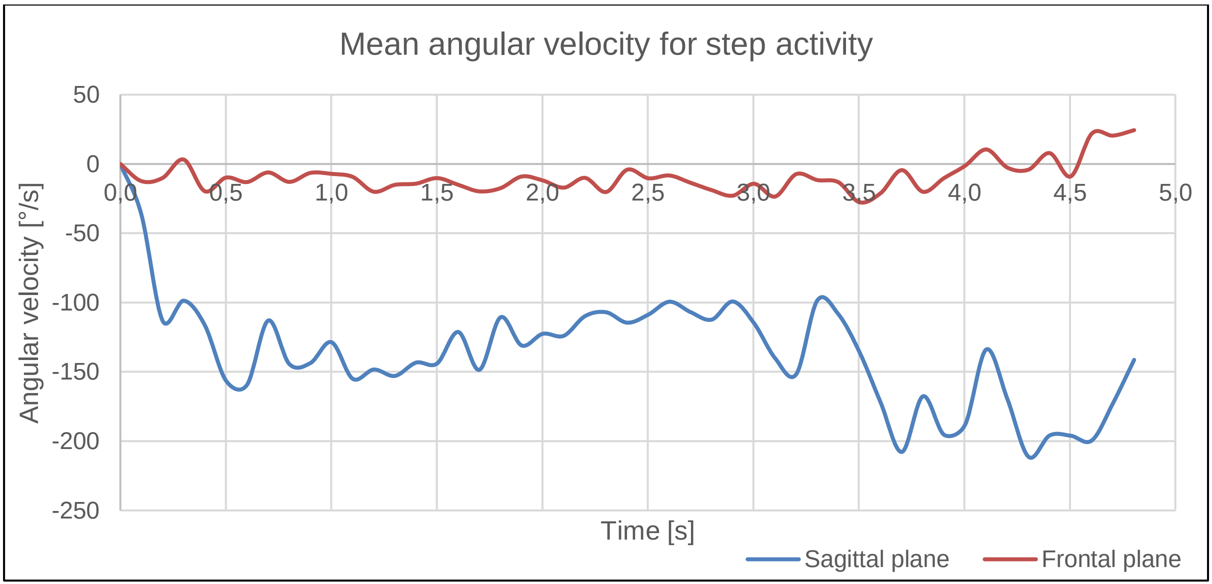

Figure 4.18: Mean angular velocity for step activity

Figure 4.19 shows the mean data for the **slope** activity. This figure shows that the average angular velocity in the sagittal plane comprises a mean maximum extension at ±82°/s and flexion at ±116°/s. In terms of the frontal plane, angular velocity is reasonably constant and within the bounds of ±20°/s and ±30°/s for flexion and extension, respectively.

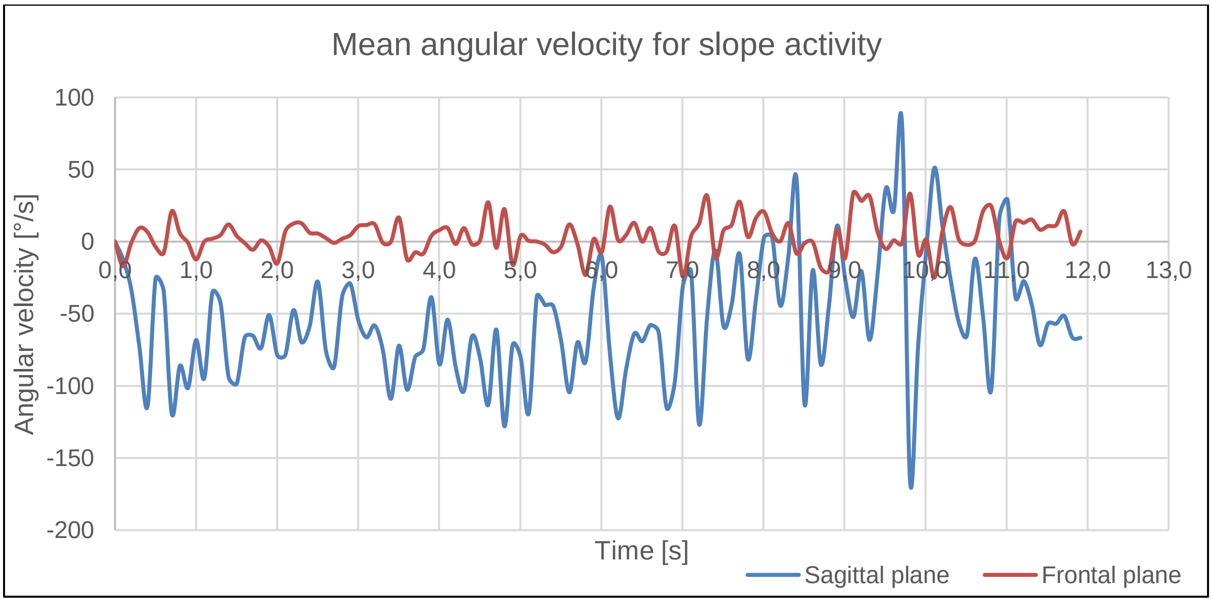

Figure 4.19: Mean angular velocity for slope activity

Figure 4.20 shows the mean data for the **stand and sit** activity. This figure shows that in terms of the sagittal plane, the average angular velocity lies between ±17°/s and ±27°/s for flexion and extension, correspondingly. For the frontal plane, the average data lies between ±10°/s flexion and ±11°/s extension.

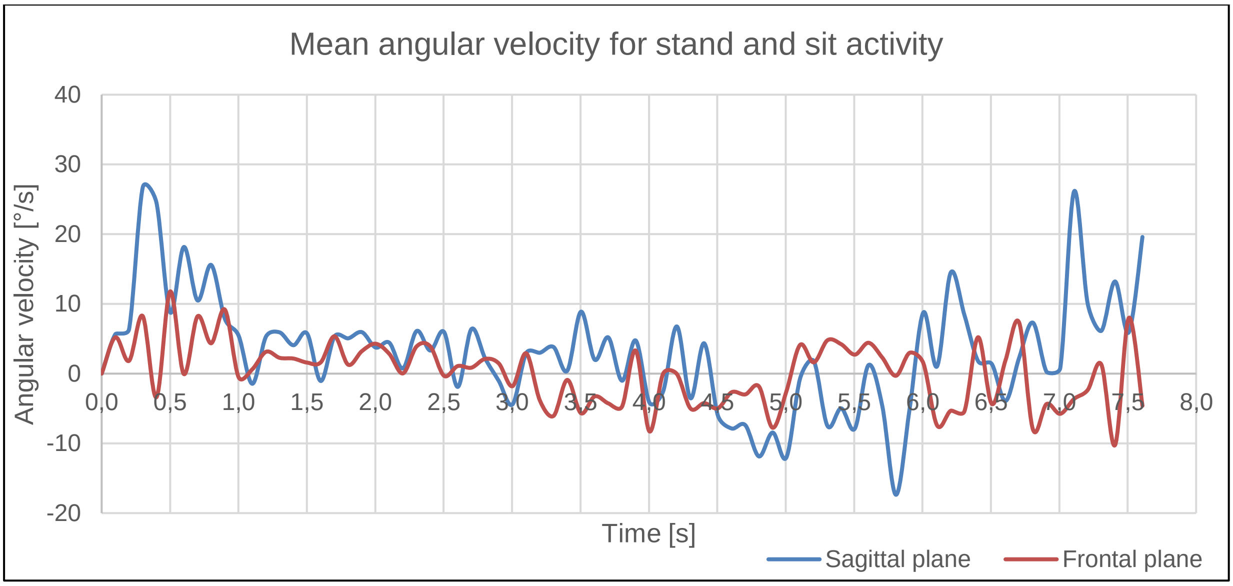

Figure 4.20: Mean angular velocity for stand and sit activity

Figure 4.21 shows the mean data for the **lift object** activity. In terms of the sagittal plane, the lift object activity typically exhibits flexion at ±76°/s. In terms of the frontal plane, the angular velocity is primarily related to extension at an average maximum of ±25°/s, however flexion may be experienced.

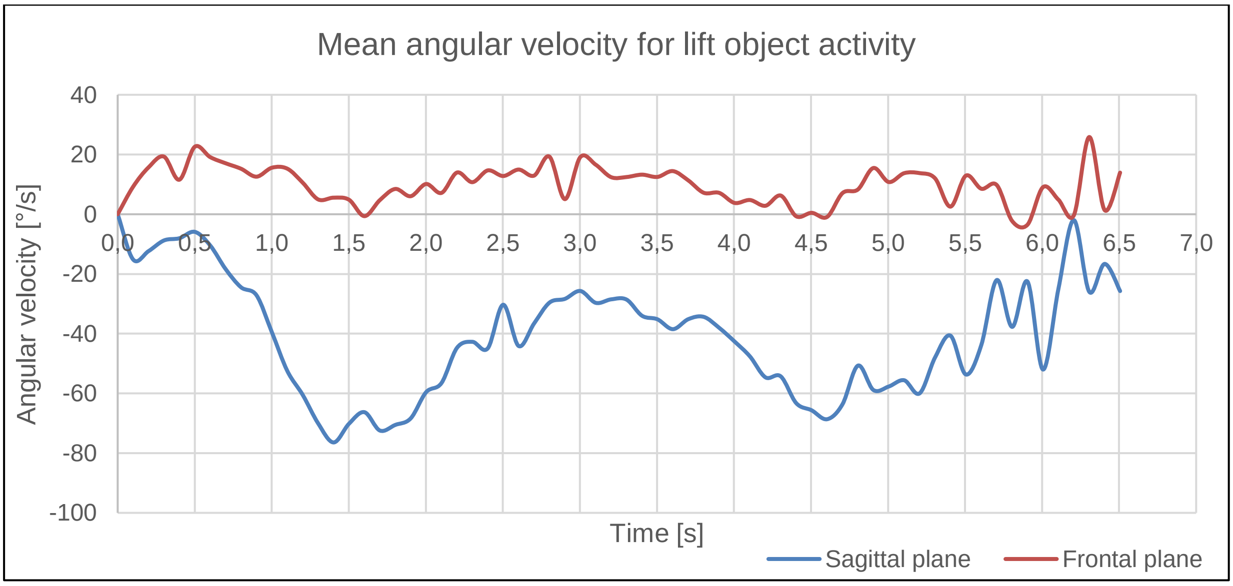

Figure 4.21: Mean angular velocity for lift object activity

As with the mean joint angles obtained, variations in trial duration and execution of trials by the participant affect the outcomes of the mean data and thus the patterns presented in the above figures. Thus, the figures should only be used to provide a general indication of joint angular velocity. To determine the accuracy of the results, a comparative analysis should be conducted, as performed in the subsequent section.

4.2.2.3 Kinematic results validation

To determine the accuracy of the kinematic results, a comparative analysis was conducted in which the results of the mean joint angles and angular velocities are compared to reference data from a Qualisys motion capture camera system. The subsequent figures present the results of this analysis. The calculated findings are represented by the red (for joint angle) and blue (for angular velocity) lines, whereas the lighter areas around the solid lines represent the standard deviation.

Figure 4.22 portrays the statistical parameter map for the lift object activity relative to the sagittal plane.

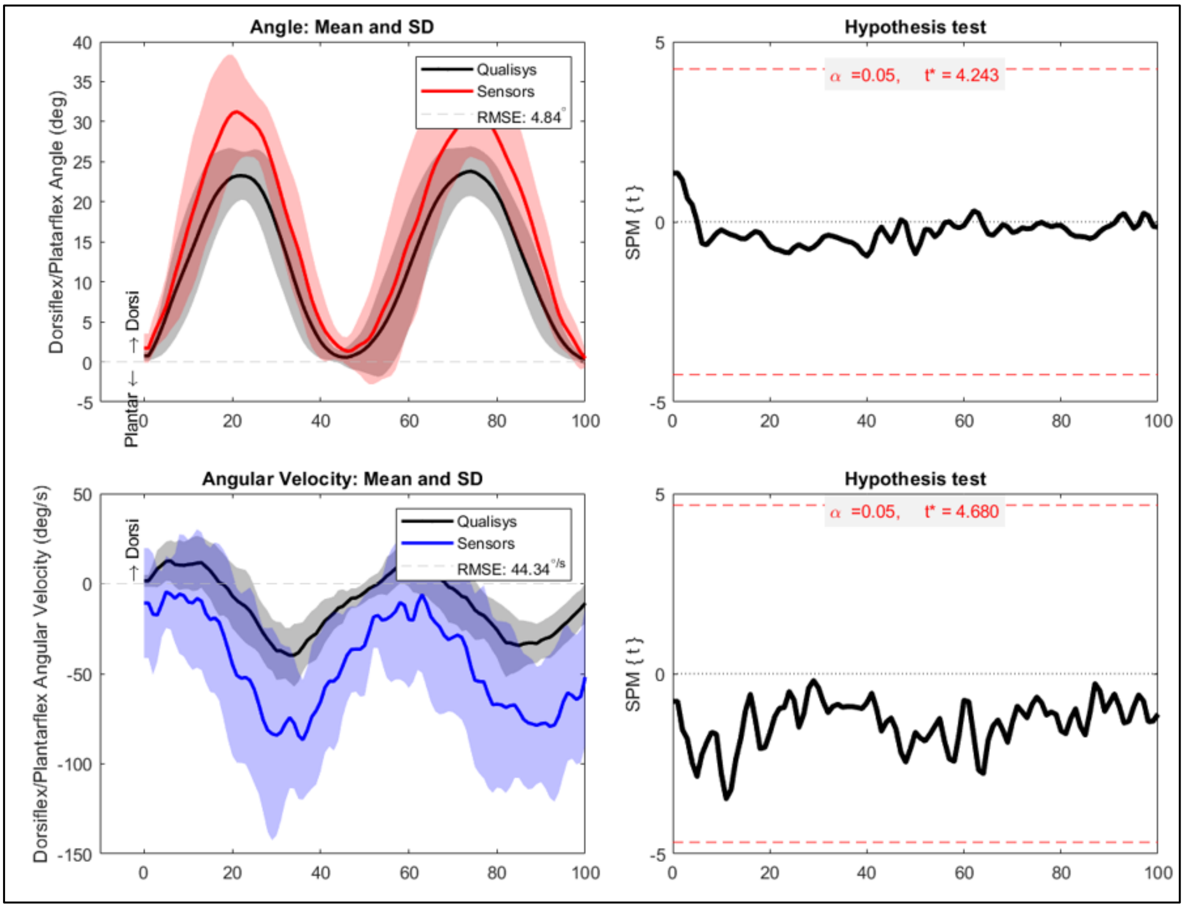

Figure 4.22: Statistical parameter mapping for lift object activity (Sagittal plane)

Figure 4.23 presents the statistical parameter map for the lift object activity relative to the frontal plane.

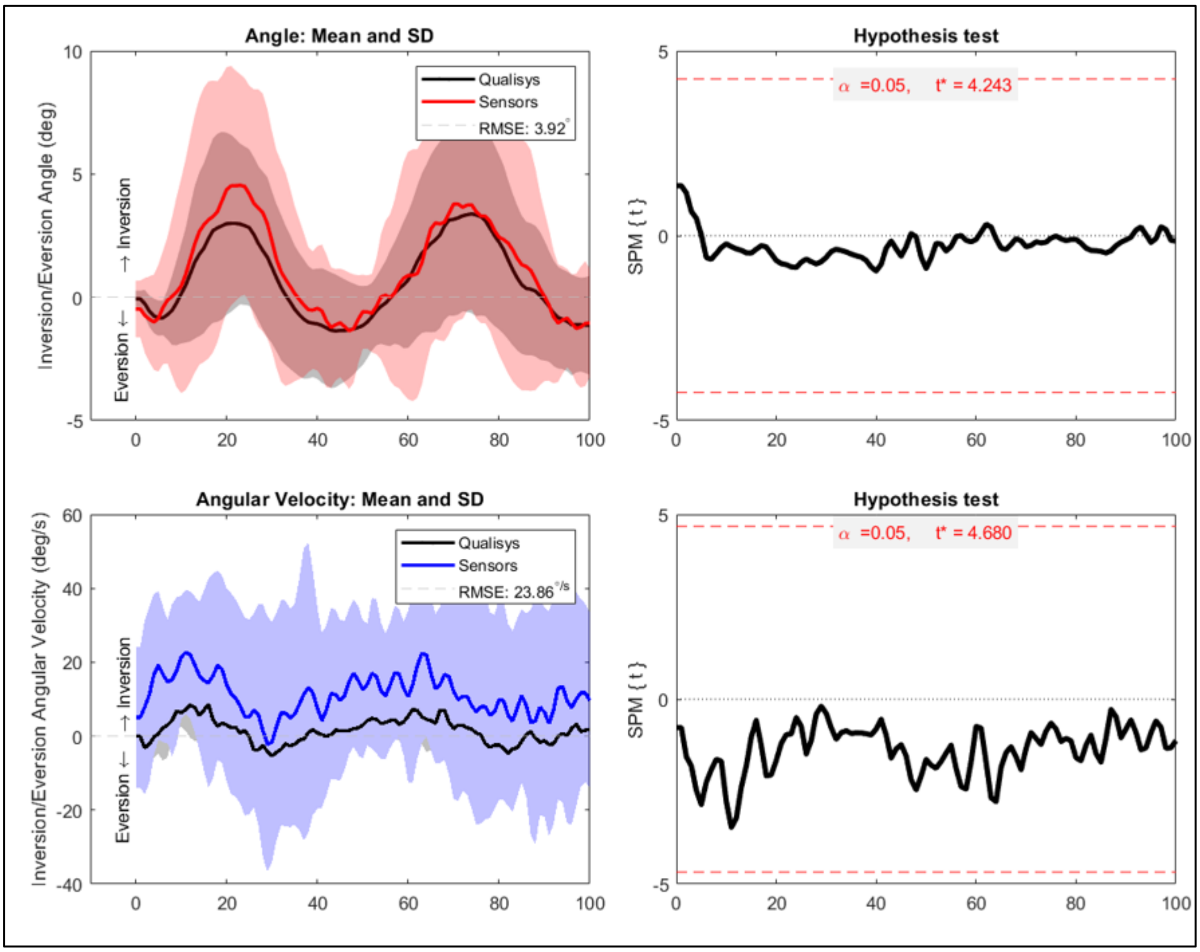

Figure 4.23: Statistical parameter mapping for lift object activity (Frontal plane)

Subsequently, a Bland-Altman analysis of the lift object activity in Figure 4.24 shows that there is a mean difference of -4.39° between the joint angle results, and that the true average is likely to fall within the range of -5.83° and -2.96°. Similarly, there is a mean difference of 33.26°/s between the angular velocity results and that the true mean falls within the range of 12.61°/s and 53.92°/s. This information simplifies to an average error of -4.39° (95% CI [-5.83°, -2.96°]) for the joint angle data and an average error of 33.26°/s (95% CI [12.61°/s, 53.92°/s]).

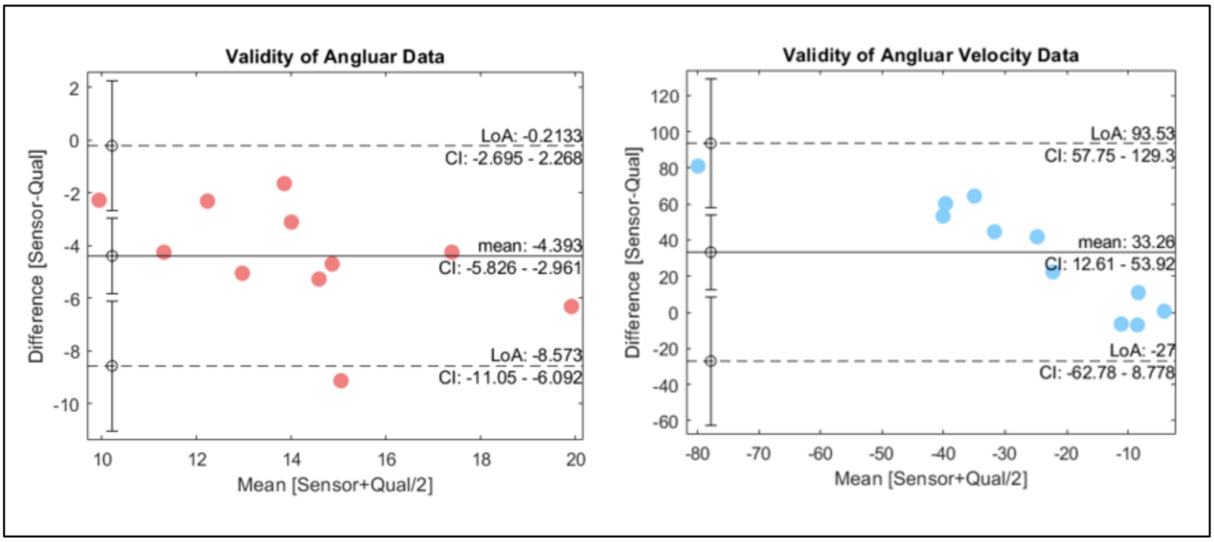

Figure 4.24: Bland-Altman analysis for lift object activity (Sagittal plane)

The Bland-Altman analysis in Figure 4.25 shows that the average error within the frontal plane was -0.44° (95% CI [-3.18°, 2.31°]) for the joint angle, and -9.72°/s (95% CI [-25.07°/s, 5.64°/s]) for angular velocity.

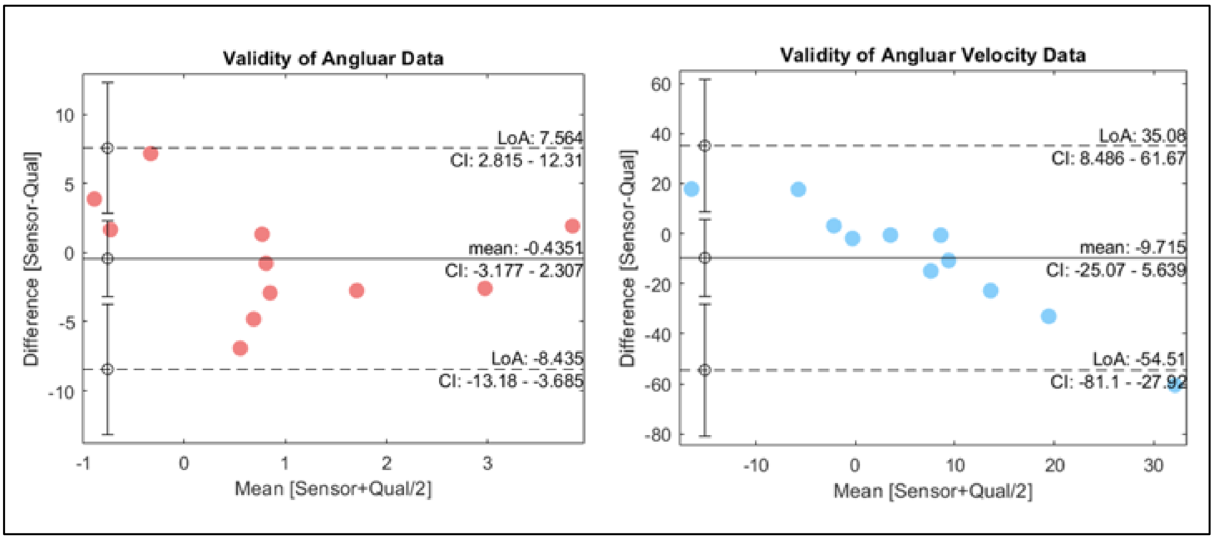

Figure 4.25: Bland-Altman analysis for lift object activity (Frontal plane)

Annexures F to I provide the Bland-Altman analysis figures of the remaining activities. To summarise the results of the joint angle analysis Table 4.5 below provide the average errors and boundaries of the confidence interval across all activities.

Table 4.5: Bland-Altman analysis of joint angle results

Activity	Plane	Average error [°]	CI lower bound [°]	CI upper bound [°]
Level walk	Sagittal	-7.27	-8.90	-5.65
	Frontal	-1.76	-4.72	1.20
Step	Sagittal	-11.89	-14.71	-9.07
	Frontal	-2.81	-5.94	0.32
Slope	Sagittal	5.38	0.74	10.02
	Frontal	0.14	-2.17	2.45
Stand and sit	Sagittal	-1.78	-3.17	-0.38
	Frontal	-0.74	-2.66	1.17
Lift object	Sagittal	-4.39	-5.83	-2.96
	Frontal	-0.44	-3.18	2.31

Similarly, Table 4.6 presents the Bland-Altman analysis results relative to the angular velocity in the sagittal and frontal planes.

Table 4.6: Bland-Altman analysis of angular velocity results

Activity	Plane	Average error [°/s]	CI lower bound [°/s]	CI upper bound [°/s]
Level walk	Sagittal	43.17	-2.03	88.36
	Frontal	15.28	-24.56	55.11
Step	Sagittal	129.70	94.08	165.20
	Frontal	10.44	-19.05	39.93
Slope	Sagittal	62.22	15.71	108.7
	Frontal	1.85	-11.40	15.11
Stand and sit	Sagittal	-3.05	-16.43	10.34
	Frontal	-1.83	-12.92	9.26
Lift object	Sagittal	33.26	12.61	53.92
	Frontal	-9.72	-25.07	5.64

4.3 Discussion of findings

The discussion of findings is based on interpretations made from the results previously provided. As with the results section, this discussion is divided into the kinetic and kinematic considerations.

4.3.1 Kinetic discussion

The values presented in Table 4.3 exhibit the diverse range of cadence and sub-phase durations among participants. This cadence variation is attributed to the differences in walking speed and step frequency, which may be affected by individual characteristics such as age or gender as affirmed by Pirker and Katzenschlager (2016). Furthermore, variations in dominant double support phases suggests differences in weight-bearing patterns. Relating the percentage sub-phase to the resultant reaction force, it can be stated that the individual experiences a maximum force of $\mathrm{F_1}$ in double support I. Similarly, a maximum force of $\mathrm{F_3}$ is experienced in double support II. In the single support phase, weight-bearing shifts and the force experienced by the limb $\mathrm{F_2}$ is less than $\mathrm{F_1}$ and $\mathrm{F_3}$, i.e., $\mathrm{F_1} > \mathrm{F_2}$ and $\mathrm{F_3} > \mathrm{F_2}$.

In terms of the lift object activity, eight characteristic features are identified in Figure 4.5. These are labelled from $\mathrm{F_1}$ to $\mathrm{F_8}$, where the initial dip from rest to $\mathrm{F_1}$ is attributed to the downward movement of the participant that acts in the direction of the gravitational force. The increase in force from $\mathrm{F_1}$ to $\mathrm{F_2}$ occurs when the participant rises from the squat and acts against gravity. The peaks at $\mathrm{F_2}$ and $\mathrm{F_3}$ indicate a weight-shift from the forefoot, which is where force typically acts in a rising squat motion, to an even distribution in the foot. Finally, the decrease in force from $\mathrm{F_3}$ to $\mathrm{F_4}$ represents the participant's transition to rest. This explanation is valid for the second half of the force plot where the participant performs an additional squat to place the object down. The findings in Table 4.4 indicate that $\mathrm{F_2}$ is not necessarily greater than $\mathrm{F_3}$, and that individual characteristics of the participants may result in the maximum force being experienced at different stages of the activity. Furthermore, this case may also be made for $\mathrm{F_6}$ and $\mathrm{F_7}$.

4.3.2 Kinematic discussion

With regards to the kinematic results, it was found that the average joint angle values derived from the sensor tended to exhibit higher values compared to those recorded by the reference system, except for the slope activity. In contrast, the average angular velocity values from the sensor were generally lower than that of the reference system. This was primarily observed in Figure 4.22, however can be further witnessed in Annexures F to I. This divergence is quantified by the average error, where a negative value indicates that the reference values are lower than the calculated values and vice versa. This is presented in Tables 4.5 and 4.6.

Moreover, these differences may largely be attributed to the data loss due to the ankle motion monitoring application’s (AMMA) low sampling frequency, Bluetooth connectivity issues, and a reduced sample size. Further contributing factors encompass the presence of outliers, human error, and errors in data processing, all of which might have resulted in a high standard deviation. Despite static pose calibration, achieving consistency in sensor positioning poses a challenge, making it reasonable to anticipate inevitable errors arising from sensor misalignment due to human error. Notwithstanding these challenges, the observed patterns in the kinematic data from the sensor and the reference system remained constant, thus, suggesting that the method used to determine the kinematic features in both the sagittal and frontal plane is valid.

Furthermore, the average errors and confidence interval bounds provided a statistical framework for understanding the quantifying differences. This is also shown in Tables 4.5 and 4.6, where the interpretation is not solely based on the average error, but also considers the range within which the true values are likely to fall. Additionally, the consistent approach across all activities further reinforces the validity of the kinematic data, where the observed patterns hold true across difference scenarios. This indicates the generalisability of the findings.

4.4 Analysis of design statements

Table 4.7 analyses the design statements set to verify and validate the study as well as ensure that the research objectives were addressed.

Table 4.7: Outcome analysis

Design statement	Outcome
Verification	
The sensor platform configuration isolates foot and ankle joint complex movements during gait-related activities.	The placement of the ADXL 345 sensors effectively allowed for the collection of data related to the foot and ankle joint complex.
The MATLAB® application allows for effective control of the wearable sensor platform.	The MATLAB® application effectively enabled the control of data collection.
Validation	
Force data could be validated against findings of previous studies.	Force data was successfully validated against findings of the background research.
Reproducibility could be observed in the collected force data.	The force results displayed a consistent pattern among the trials, demonstrating reproducibility.
Comparative analysis could be performed on the level walk activity data.	The results of the level walk activity were successfully compared against the reference data.
Comparative analysis could be performed on the step activity data.	The results of the step activity were successfully compared against the reference data.
Comparative analysis could be performed on the slope activity data.	The results of the slope activity were successfully compared against the reference data.
Comparative analysis could be performed on the stand and sit activity data.	The results of the stand and sit activity were successfully compared against the reference data.
Comparative analysis could be performed on the lift object activity data.	The results of the lift object activity were successfully compared against the reference data.

This table clearly demonstrates that the verification and validation design statements were fulfilled, thereby confirming and validating the research.

4.5 Chapter summary

With regards to the key kinetic findings, characteristic features, including peaks and dips, could be identified for the level walk and lift object activities. Additionally, it was observed that differences in age, height, weight, and sex may be contributing factors to variations in gait patterns, thus, resulting in variations in weight-bearing. Furthermore, to validate the kinetic findings, four crucial aspects were considered, specifically: reproducibility, expert review, quality equipment, and comparison to previous works. The validation discussion yielded positive outcomes, therefore ensuring the reliability of the results and interpretations.

Concerning the kinematic findings, the chapter first provided the details of the data preparation phase, in which all clean trials were time-synchronised and averaged so that a mean dataset could be obtained for each participant's activity. The data could then be used to determine the joint angles and angular velocity in the sagittal and frontal planes. To validate these findings, a comparative analysis was conducted wherein the experimental results were compared to reference data obtained from a motion capture camera system. It was found that although data loss issues resulted in a high standard deviation in the findings, similar patterns were observed in comparison. Moreover, the consistent approach across activities reinforces the validity of the kinematic data. The aspects suggest both validity of the method itself, as well as a generalisability of the findings.

The ensuing chapter concludes the study by providing a summative discussion on the research process.

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CHAPTER 5

5. CONCLUSION

5.1 Introduction

The final chapter presents a summative analysis and discussion of the entire research study. Figure 5.1 below provides a clear framework of the concluding discussion.

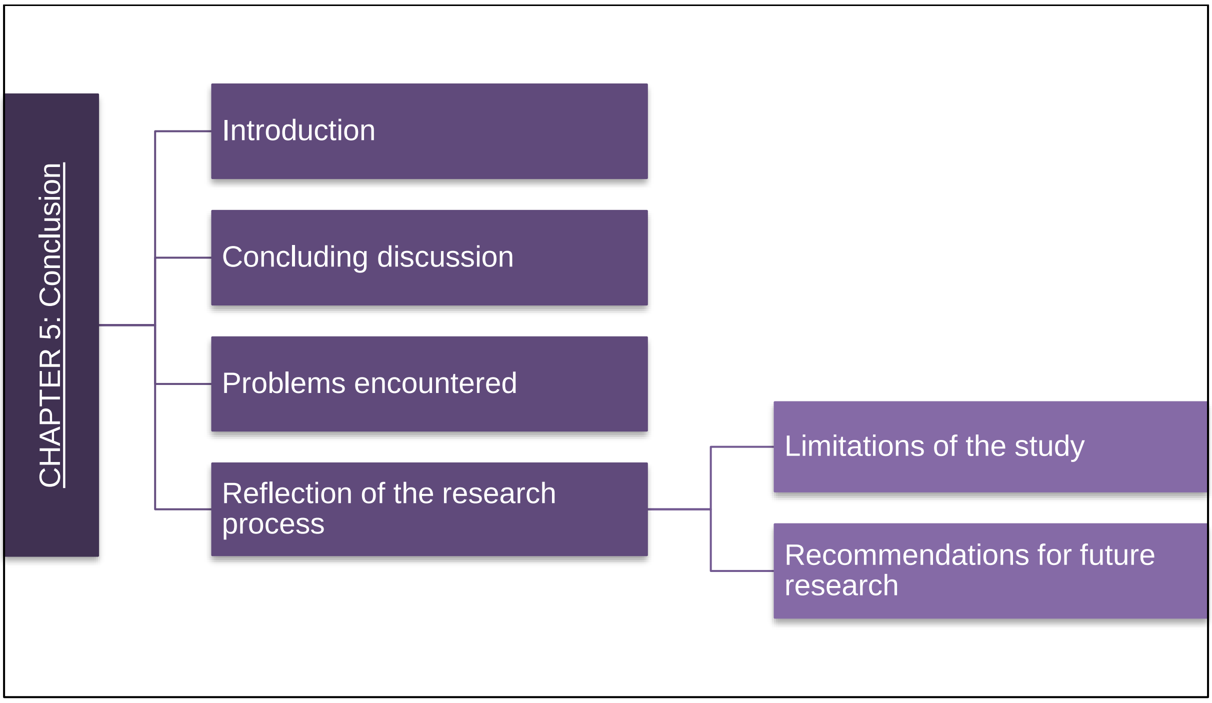

Figure 5.1: Conclusion breakdown

Per the above figure, this chapter provides a concluding discussion to the study by reiterating the key findings and relating them to the defined research purpose and objectives. The chapter then provides a concise examination of the problems encountered throughout the study, which is followed by a reflection of the research process. This reflection involves scrutinising the study's limitations and proposes recommendations for future research.

5.2 Concluding discussion

The primary purpose of this research study was to quantify and analyse the biomechanical functions of the human foot and ankle joint complex during various activities encountered in everyday life. This purpose was set to further the understanding of natural human movement and provide relevant data to be utilised in developing an ankle-foot prosthesis. Furthermore, three research objectives were defined in support of the overarching purpose of the study, these were:

- To determine the dynamic reaction forces between the lower limb and the ground for different activities related to everyday living.
- To develop a wearable sensor platform device and MATLAB® application interface for monitoring the foot and ankle joint complex movements.
- To evaluate the accuracy and precision of the experimental findings by comparing the results to reference data.

To achieve these objectives, an experimental research design was selected for the study based on its high level of control that allows researchers to select and isolate specific variables, as well as its compatibility with other methods. The experiment utilised a force plate system, a wearable sensor platform, and motion capture cameras to collect kinetic and kinematic data. In terms of sample size, 15 participants were recruited for the study; however, due to the low sampling rate of the developed ankle motion monitoring application (AMMA), Bluetooth connectivity issues, and technical difficulties, the effective sample size was reduced to 11 participants with usable kinematic data and nine with usable kinetic data. Once the wearable sensor platform and reflective markers for the motion capture camera were equipped, participants were required to navigate a set of five low-level activities. These were 1) level walk, 2) step up and down, 3) 20° slope incline and decline walk, 4) standing and sitting, and 5) lifting an object. From these activities, dynamic reaction force data could be collected from the force plates, and angular data could be collected from the sensor platform and reference motion capture camera system.

In terms of findings, the kinetic results displayed a constant pattern among trials, contributing to the credibility of the findings. Furthermore, the force data demonstrated the characteristic features of the reaction forces experienced by an individual during certain activities, which furthers the understanding of joint kinetics during natural movement. Regarding kinematics, angular data from the sensors could be used to determine the joint angles and angular velocity relative to the sagittal and frontal planes of the foot. These findings showed a high standard deviation compared to the reference data obtained from the motion capture cameras, which was attributed to the low sampling frequency of the AMMA, Bluetooth connectivity issues, reduced sample size, human error, and outliers in the data set. Nonetheless, similar patterns between the sensor results and reference data were observed, hence indicating the validity of the method itself.

Therefore, through the experimental investigation, the three research objectives were successfully addressed with the following outcomes.

- The dynamic reaction forces between the human lower limb and ground were determined for two activities related to daily living. These activities were a normal walk and lifting an object.
- A wearable sensor platform and MATLAB® application interface (i.e., the AMMA) was developed and successfully allowed for angular data, relating to the human foot and ankle, to be collected.
- The accuracy and precision of the experimental findings was evaluated by comparing the results to reference data obtained from a motion capture camera system.

This was further confirmed by the study's verification and validation section, wherein several design statements, linked to the research purpose and objectives, were addressed.

5.3 Problems encountered

The AMMA was coded to capture data at 200 Hz to match the reference data; although, due to the complexity and length of the code, the effective sampling frequency was reduced to 10 Hz. This sampling frequency resulted in the AMMA writing data every 0.1 seconds instead of the initially intended 0.005 seconds, thus resulting in significant data loss. However, from the comparative analysis, it was clear that similar patterns between the sensor results, and reference data could be observed across trials. This signified the validity of the method.

In addition to this problem, several minor challenges were encountered throughout the study. The following points describe the challenge, the mitigation techniques, and the outcome of these:

- **Challenge:** An unstable Bluetooth connection resulted in intermittent communication issues between the AMMA and the motherboard. Consequently, the collected data exhibited irregularities including rows of zeros, incomplete data rows, not a number (NaN) values, and extreme outliers. Notably, outliers were identified as values lying beyond three standard deviations of the column mean.

  Mitigation: Regarding row irregularities, the entire row was substituted with the mean of the following and former rows. In cases where the anomaly was observed at the beginning or end of the dataset, the data was replaced with the following or former row, respectively. A similar procedure was applied to NaN values and outliers, whereby inconsistent values were replaced with the mean of the neighbouring row values.

Outcome: This approach effectively stabilised the values, thereby enhancing the accuracy of the collected data. Nevertheless, it is worth noting that the restored values may still have presented a degree of inaccuracy, indicating that the AMMA's accuracy is partially definitive.

- **Challenge:** Substandard placement of the sensors and wearable equipment on the feet and waist may have contributed to an irregular gait pattern amongst certain participants.

  Mitigation: To ensure maximum comfort, thin connecting wires were utilised, and housing components were designed to be as compact as possible. This minimised the overall weight of the wearable equipment. Moreover, the motherboard and power source components could easily be relocated to positions that offered greater comfort.

  Outcome: Although measures were taken to ensure a natural gait pattern, it is still possible that participants may have exhibited irregular walking patterns due to physical discomfort. In addition to this, placement of the sensors on the limb may not have been consistent throughout trials.

- **Challenge:** Inadequate fastening of the sensors to the limb resulted in undesired vibration and instability during experimental trials. These vibrations introduced noise and disturbances into the measurements, leading to inconsistencies in the collected data. As a result, the precision and reliability of the data was negatively affected.

  Mitigation: To reduce vibrations in the sensors, bandage wraps and adhesive electrodes were used to securely fasten the sensor housings to the participant's limb. These components functioned as dampeners and substantially reduced the presence of vibrations during experimental trials.

  Outcome: The use of bandage wraps and adhesive electrodes significantly reduced the occurrence of vibrations and external noise, therefore enhancing the overall accuracy and precision of the recorded data. However, it is necessary to acknowledge that minor vibrations may still have been present in the sensors.

5.4 Reflection of the research process

A reflection of the research process may contribute to individual learning and development, as well as motivate the need for future research. Consequently, the following sections discuss the study's limitations and offer suggestions for future works.

5.4.1 Limitations of the study

The following potential limitations follow from the defined delineations or were noted during the study:

- This study only considered five low-level activities, thus restricting data and conclusions to these activities.
- This study focused on the properties of the foot and ankle joint complex; however, knee and hip movements also play a role in facilitating natural gait. Although this simplified the study, the extent to which the results could be generalised was limited.
- The study only considered movement in the sagittal and frontal planes of the foot; therefore, no conclusions can be made regarding the transverse plane.
- During the experimental trials, there was potential error arising from misalignment of the sensors due to human error.

5.4.2 Recommendations for further research

It is essential to consider potential opportunities for future research and developments in the context of a similar gait analysis study. In terms of the **kinetic** data quantification, the following suggestion could be made:

- Future research could investigate the reaction forces of additional activities related to everyday living, though a modular force plate system may be required for this.

For the **kinematic** aspect of the study, the following may be recommended:

- Future research could consider a larger sample size to enhance the generalisability of the collected data in the case of human error and other data loss issues.
- Future research could optimise the coding behind the application interface to increase its sampling frequency. Alternatively, future researchers should consider the hardware requirements, specifically the computer RAM and CPU.
- Future research could take transverse plane motions (adduction and abduction movements) into account to determine their role in gait and other related activities.

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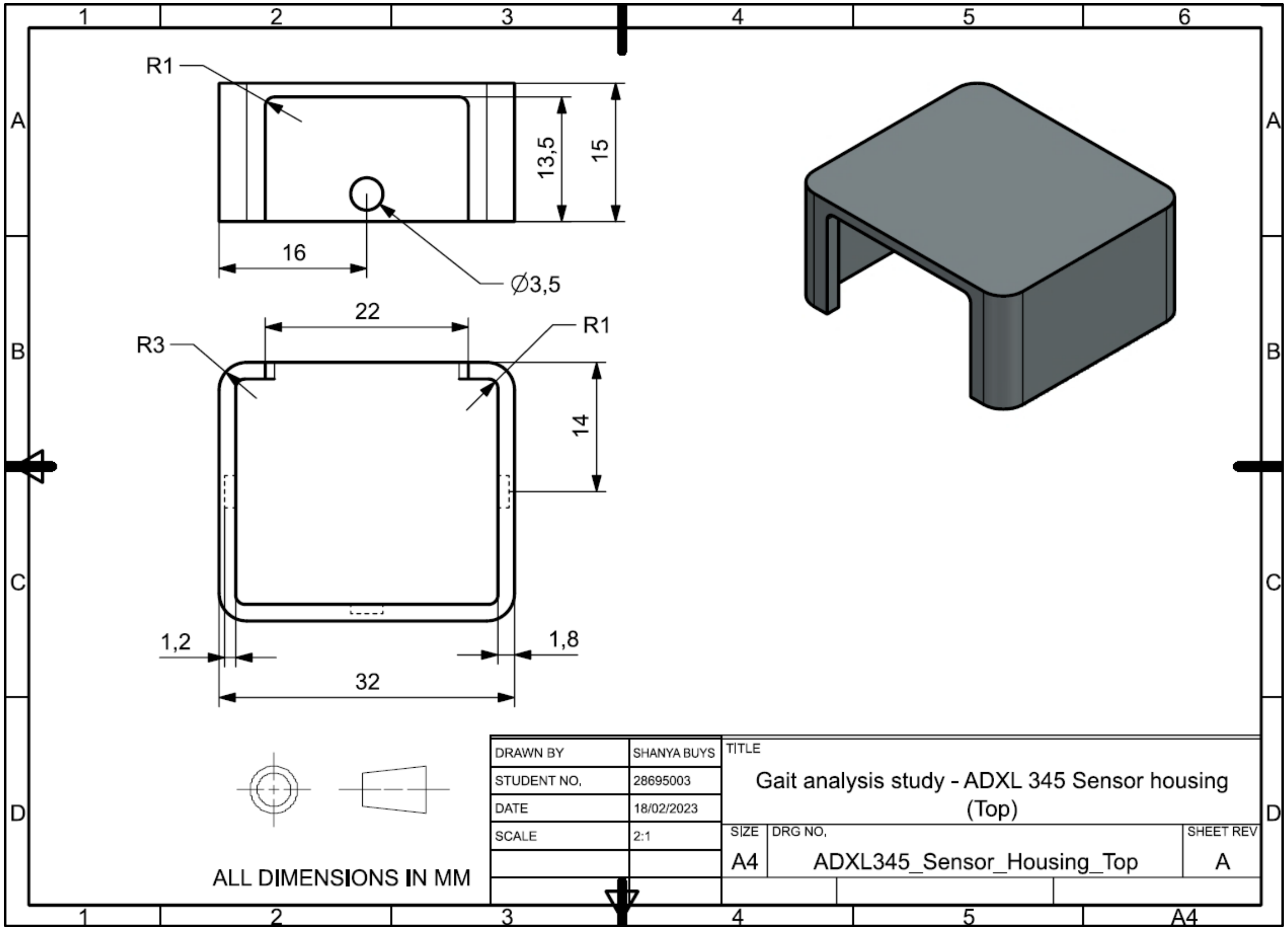
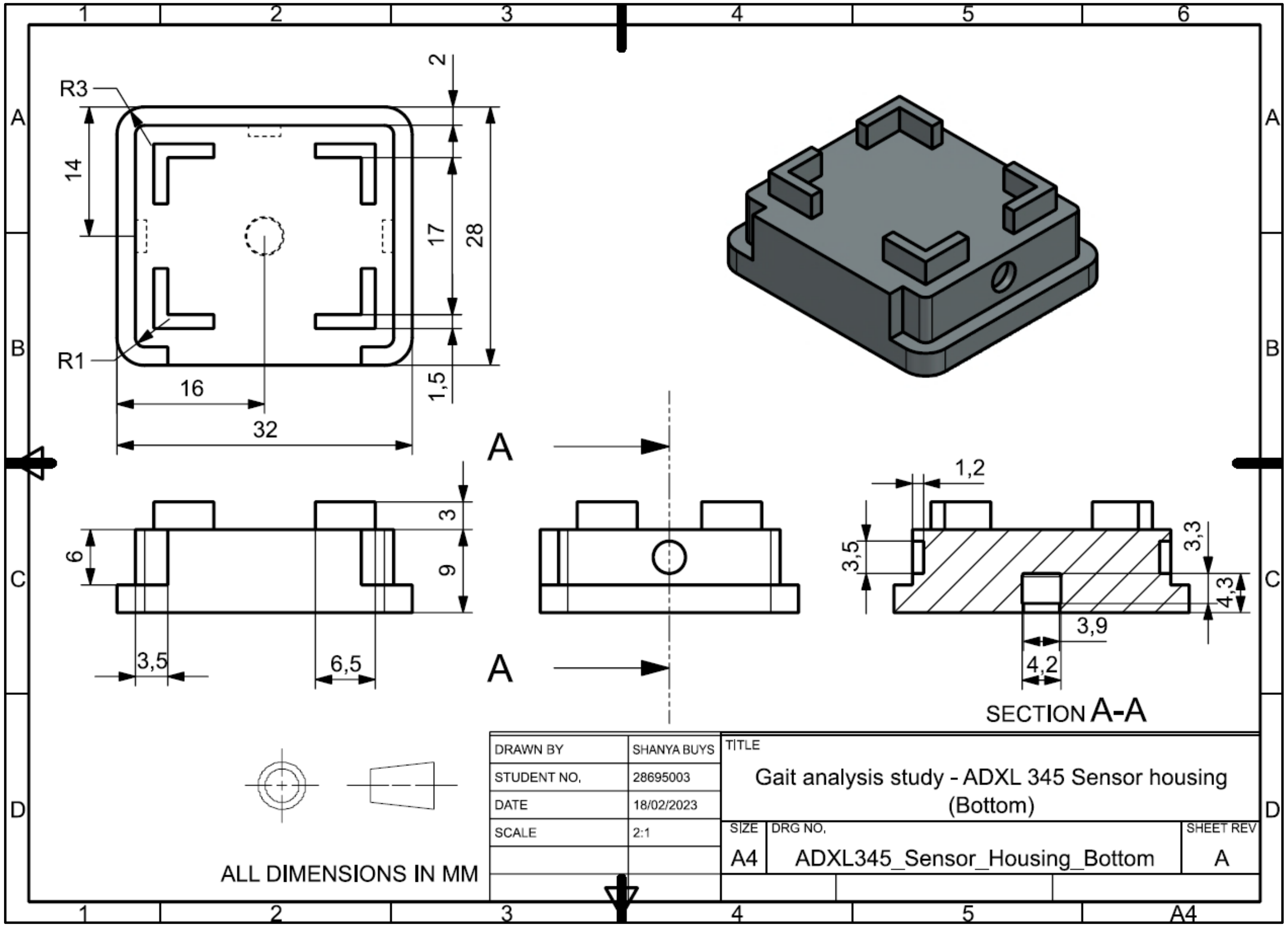
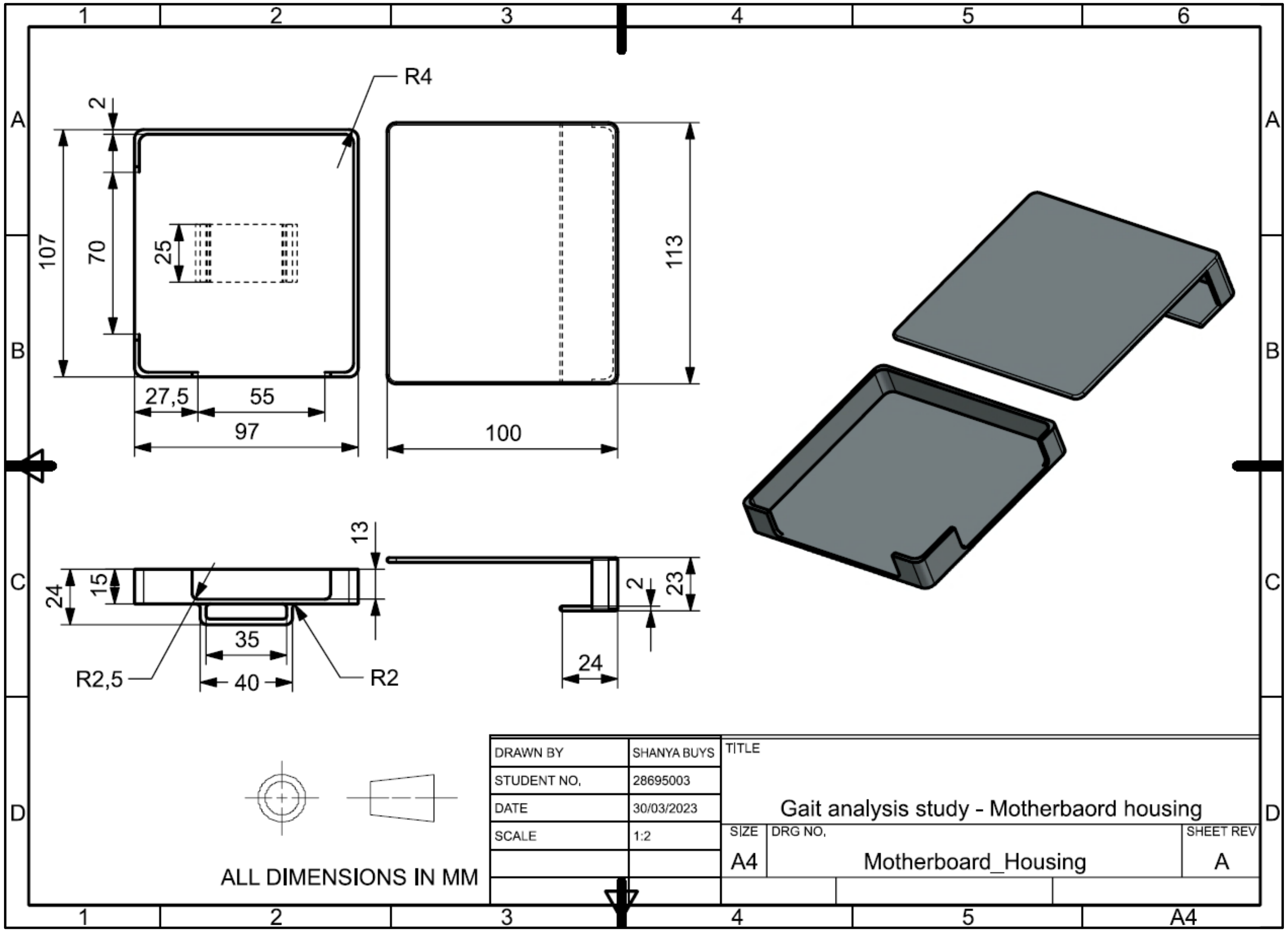
ANNEXURE A – SENSOR HOUSING (TOP)



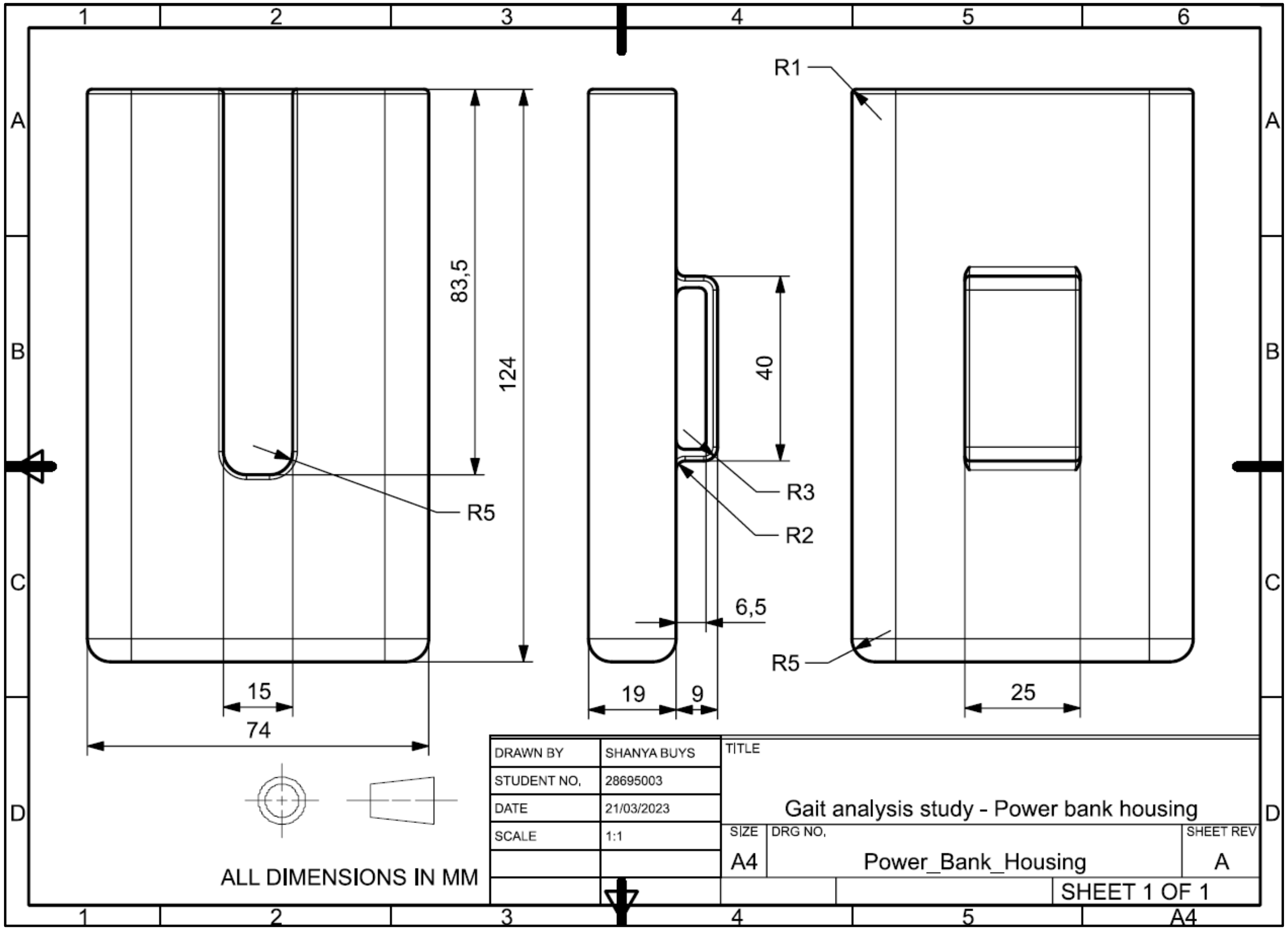
ANNEXURE B – SENSOR HOUSING (BOTTOM)



ANNEXURE C – MOTHERBOARD HOUSING



ANNEXURE D – POWER BANK HOUSING



ANNEXURE E – INFORMED CONSENT DOCUMENT (ENGLISH)

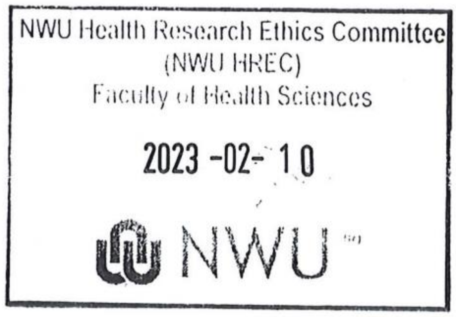
Private Bag X1290, Potchefstroom

South Africa 2520

Tel: +2718 299-1111/2222

Fax: +2718 299-4910

Web: http://www.nwu.ac.za



NWU-HREC Stamp

INFORMED CONSENT DOCUMENTATION FOR BIOMECHANICAL QUANTIFICATION AND ANALYSIS OF THE HUMAN FOOT AND ANKLE JOINT COMPLEX

TITLE OF THE RESEARCH STUDY:	Biomechanical quantification and analysis of the human foot and ankle joint complex
ETHICS REFERENCE NUMBERS:	NWU-00182-22-A1
PRINCIPAL INVESTIGATOR:	Prof. MJ Grobler
POST GRADUATE STUDENT:	Shanya Buys
ADDRESS:	Office 213, Building N1, NWU Engineering Campus 18 Calderbank Ave, Potchefstroom, 2520
CONTACT NUMBER:	071 304 5439

You are being invited to take part in a **research study** that forms part of a master's study. Please take some time to read the information presented here, which will explain the details of this study. Please ask the researcher or person explaining the research to you any questions about any part of this study that you do not fully understand. It is very important that you are fully satisfied that you clearly understand what this research is about and how you might be involved. Also, your participation is **entirely voluntary,** and you are free to say no to participate. If you say no, this will not affect you negatively in any way whatsoever. You are also free to withdraw from the study at any point, even if you do agree to take part now.

This study has been approved by the NWU-Health Research Ethics Committee of the Faculty of Health Sciences of the North-West University (NWU-00182-22-A1) and will be conducted according to the ethical guidelines and principles of Ethics in Health Research: Principles, Processes and Structures (DoH, 2015) and other international ethical guidelines applicable to this study. It might be necessary for the research ethics committee members or other relevant people to inspect the research records.

What is this research study all about?

- This study plans to measure the mechanical functions of the ankle joint complex during various everyday activities. This will be done using accelerometer and gyroscope sensors attached to the lower leg to measure foot-ankle movements, and EMG sensors to measure muscle responses. Force plate and motion capture systems will also be used to determine ankle and foot motions. This study will determine how the functions of the ankle complex enables natural movement and several forms of gait activities.
- This study will be conducted at the PhASRec facility, supervised by experienced health researchers trained in engineering and biokinetics. A total of 15 participants are required for this study.

Why have you been invited to participate?

- To be included in the present study the following criteria have to be met:
    - *Male or female between the ages of 18 and 45.*
        - *This is the age group most likely to have a fully developed and non-deteriorating gait pattern.*
    - *Must be free of any injuries or recent surgery (< 6 months) that might impair gait or activities of daily living.*

- *Must live within the Potchefstroom area. This is due to budgetary and time constraints on the study.*
- *Must weigh less than 150 kg.*
  - *The load cell that shall be used can measure a maximum of 200kg but is more accurate when measuring below the threshold.*

- You have therefore been invited to be part of this research study because you meet all relevant inclusion criteria and are asked to please assist in the analysis of the mechanical functions of the ankle joint complex, by allowing us to collect and use your data (after it has been completely anonymised).
- You will unfortunately not be able to take part in this research if you are:
  - *Suffering from diseases and/or disorders that impair gait or activities of daily living (e.g., osteoarthritis, rickets, etc.).*

What will be expected of you?

- The sample size of this study includes both the number of participants, as well as the number of trials each participant will need to complete. We require a total of 15 participants to complete five (5) trials of nine (9) different activities, wherein no less than 15 steps will be taken for each activity.
- Your height and weight will also be measured on a scale – this will be linked to an anonymous code.
- You may be expected to wear short pants and remove shoes for the experiment. This is because sensors will need to be placed on the foot and lower leg area and may provide more accurate data if placed as close to the limbs as possible. Furthermore, if preferred, participants may request to have sensors placed on them by a researcher of similar gender.
- The data collection process will take place at the PhASRec facility on the North-West University Potchefstroom Campus. This process will last for approximately 45 minutes, wherein you will be expected to complete five (5) trials across nine (9) different activities of daily living. These activities include level walking, ascending five steps, descending five steps, walking up an inclined (20°) surface, walking down a declined (20°) surface, standing from a sit position, lifting an object, full ankle/foot plantarflexion and full ankle/foot dorsiflexion.
- All activities will be completed while wearing a sensor platform device on your right limb, and reflective markers to track movements across the activities. Please note that

none of the tracking equipment is invasive; all markers are externally fitted onto clothing and/or skin using double-sided tape.

- You will be required to sign this form at the laboratory in the presence of independent person (Ian Thompson). A trusted witness will be arranged if requested.

Will you gain anything from taking part in this research?

- Direct gains from this study will include a functional report that will provide an overview of kinetic and kinematic data related to activities of daily living. However, since the primary researcher has no background in the health sciences this report will have no clinical implications.
- Other gains of the study will include the measurement of the mechanical functions of the ankle joint complex, and how this ankle enables natural gait movement and gait activities.
- This study may also provide relevant valuable information to the design and development of an ankle and foot prostheses, to allow for more natural gait movements.

Are there risks involved in you taking part in this research and what will be done to prevent them?

Please note that all precautions have been taken to minimise any potential risks of injury to you and/or the research staff. All testing will be facilitated by a qualified engineer and biokineticist to ensure maximum participant safety throughout the data collection processes.

- <u>Possible Covid-19 infection</u>

  The risk of Covid-19 infection would not exceed that of going to a shop/mall. All Covid-19 regulations of the time will be followed and may include: (i) all equipment being adequately sanitised prior to use, (ii) researchers wearing a face mask, sanitising hands, and wearing neoprene gloves, and (iii) participants may be required to wear a face mask and sanitise their hands upon entry to the testing venue.
- <u>Ascending and descending stairs</u>

  Measurements are required while you, as the participant, ascend and descend stairs as you would in everyday life. The activity shall be done in the presence of a qualified biokineticist, and the stairs shall conform to all safety regulations (e.g., railings will be present to prevent a risk of falling, and step heights will not exceed 20cm).
- <u>Ascending and descending slopes</u>

Measurements are required while participants ascend and descend a sloped surface (20˚). The activity shall be done in the presence of a qualified biokineticist, and the slope shall meet all safety regulations to prevent any potential for falling (e.g. anti-slip surfacing will be used).

- Anthropometric measurements

  To ensure protection of participants, no visual capturing of photos or videos will be permitted, other than that of the motion capture system, which will only be used to determine the location of the reflective markers in the form of numerical data. Participants will not be required to remove any clothing items other than their shoes, and participants may choose to have the sensors and reflective markers placed on by researcher of same gender.

How will we protect your confidentiality and who will see your findings?

- The confidentiality of your findings will be protected by only working with one participant at a time, and any identifying information will be coded such that no one except the primary researcher and study supervisor shall know your identity, or that you are taking part in the study. Your privacy will be respected by not forcing you to reveal any information you wish to keep private. All results will be kept confidential by keeping all findings in password-protected files and USB drives. Only the researchers and study supervisors will be able to look at your findings. Findings will be kept safe by locking hard copies in locked cupboards in the researcher's office and electronic data will be stored in password-protected folders.
- Participants will be evaluated one at a time during their specified period, in the presence of the primary researcher and biokineticist. Participants may be required to wear short pants and remove their shoes during the experimental process, so that sensors and reflective markers may be placed as close to the limb as possible. The participant may request that a researcher of the same gender perform this process.
- No visual data (photos or videos) will be recorded during the experiment. Only numerical data will be collected from the motion capture cameras.
- Findings will be kept safe by scanning hard copies of data and storing them in a password-protected external hard drive in the study/project leader’s office at the North-West University (NWU), Potchefstroom Campus, for 7 years where after it will be destroyed. This data will be backed up on RedCap, a cloud-based system at the North-West University.

What will happen with the findings or samples?

- The findings of this study will only be used to analyse the mechanical functions of the human ankle joint complex, and how it facilitates gait movement. Any personal identifying information obtained during data collection will be coded such that your confidentiality will be retained. The results of the study will be published in a peer-reviewed journal such that the findings can be made public to facilitate further research and aid practitioners involved in evaluating knee function. No personal information will be made public and will not be made available to anyone outside of the study; therefore, your privacy and confidentiality will be always retained.

How will you know about the results of this research?

- We will provide you with your results as well as the results of this research when the study has been completed by emailing the completed study to you.
- Results will be provided in the form of a summary of findings via email, with no clinical implications since the primary research has no background in the health sciences.
- You will also be informed of any new relevant findings by email.

Will you be paid to take part in this study and are there any costs for you?

Participants shall be reimbursed using the time-inconvenience-expenses principle. A maximum duration of 45 minutes is expected for each participant (a schedule will be made available closer to the time). The TIE principle recommends that participants are offered a meal and refreshment to the value of R50. Participants will also be reimbursed at R3/km travelled; however, transportation will be arranged if requested.

Is there anything else that you should know or do?

- If you have any questions and/or queries, you can contact the primary researcher Ms Shanya Buys at 28695003@student365.msfed.nwu.ac.za, or mediator Dr Mark Kramer at Mark.Kramer@nwu.ac.za.
- You can also contact the NWU-Health Research Ethics Committee via Mrs Carolien van Zyl at 018 299 1206 or carolien.vanzyl@nwu.ac.za, if you have any concerns or complaints about the research.
- You will receive a copy of this information and consent form for your own purposes.

Declaration by participant

By signing below, I …………………………………….…………. agree to take part in the research study titled: Biomechanical quantification and analysis of the human foot and ankle joint complex.

I declare that:

- I have read this information/it was explained to me by a trusted person in a language with which I am fluent and comfortable.
- The research was clearly explained to me.
- I have had a chance to ask questions to both the person getting the consent from me, as well as the researcher, and all my questions have been answered.
- I understand that taking part in this study is **voluntary** and I have not been pressurised to take part.
- I may choose to leave the study at any time and will not be handled negatively if I do so.
- I may be asked to leave the study before it has finished if the researcher feels it is in the best interest, or if I do not follow the study plan, as agreed to.

Signed at (*place*) ....…..........…………………. on (*date*) …………....………. 20......

....

Signature of participant Signature of witness

Declaration by person obtaining consent

I *(name)* ……………………………………………..……… declare that:

- I clearly and in detail explained the information in this document to

  ………………………………………………….
- I did/did not use an interpreter.
- I encouraged him/her to ask questions and took adequate time to answer them.
- I am satisfied that he/she adequately understand all aspects of the research, as discussed above.
- I gave him/her time to discuss it with others if he/she wished to do so.

Signed at (*place*) ....…........…………………. on (*date*) …………....………. 20......

....

Signature of person obtaining consent

Declaration by researcher

I *(name)* …………………………………………….……… declare that:

- I explained the information in this document to ensure that everything is understood by the participant.
- I did not use an interpreter.
- I encouraged him/her to ask questions and took adequate time to answer them.
- The informed consent was obtained by an independent person.
- I am satisfied that he/she adequately understand all aspects of the research, as described above.
- I am satisfied that he/she had time to discuss it with others if he/she wished to do so.

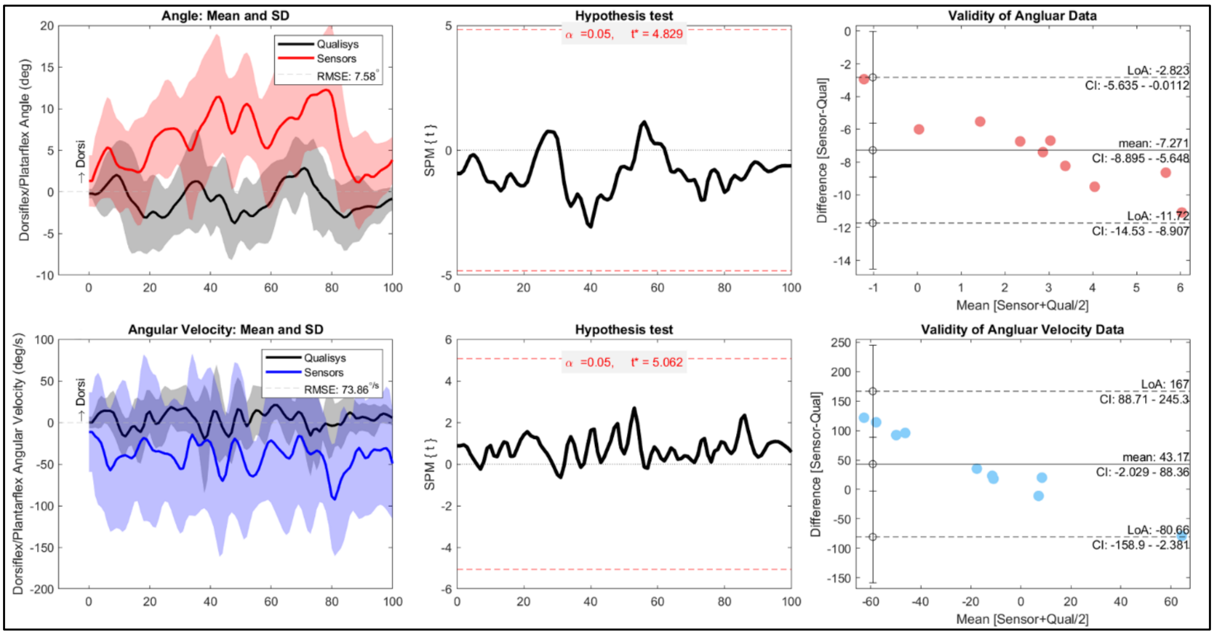
Signed at (*place*) ....….......…………………. on (*date*) …………....………. 20......

....

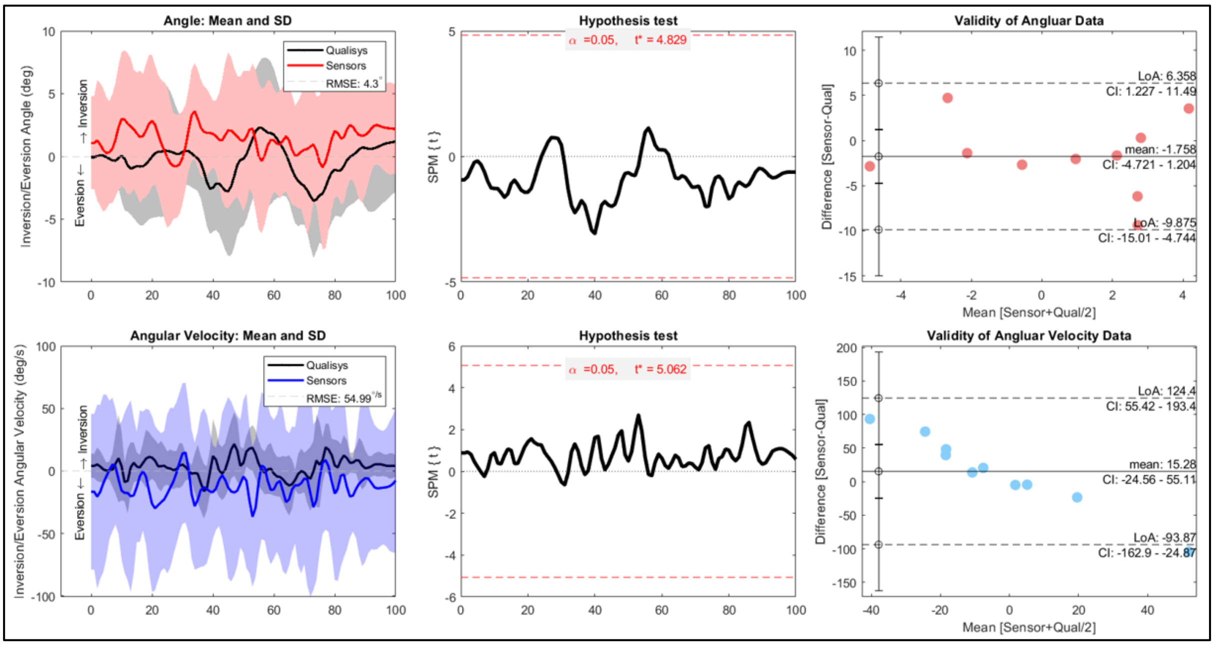
Signature of researcher

ANNEXURE F – LEVEL WALK ACTIVITY COMPARATIVE ANALYSIS

Sagittal plane

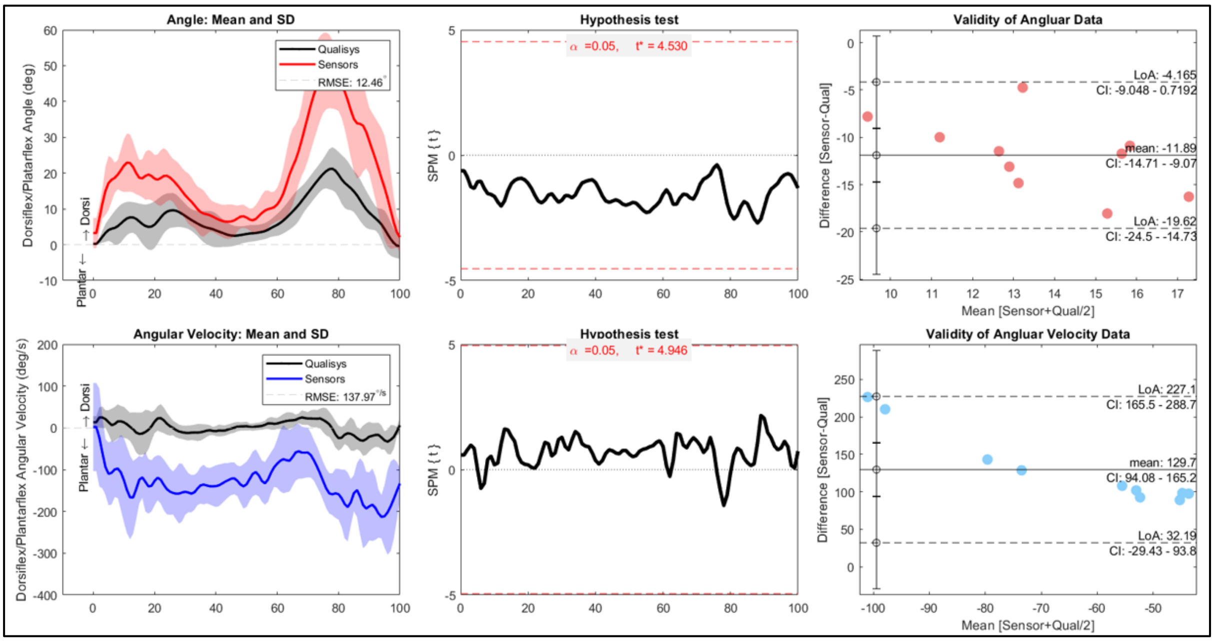


Frontal plane

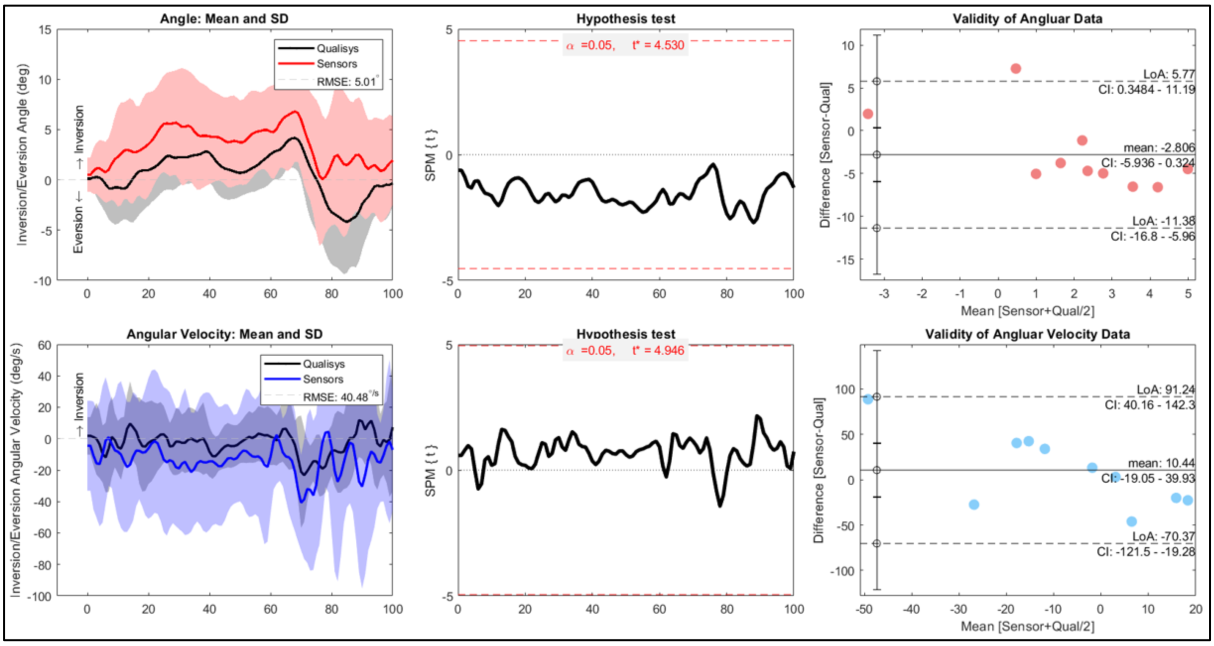


ANNEXURE G – STEP ACTIVITY COMPARATIVE ANALYSIS

Sagittal plane

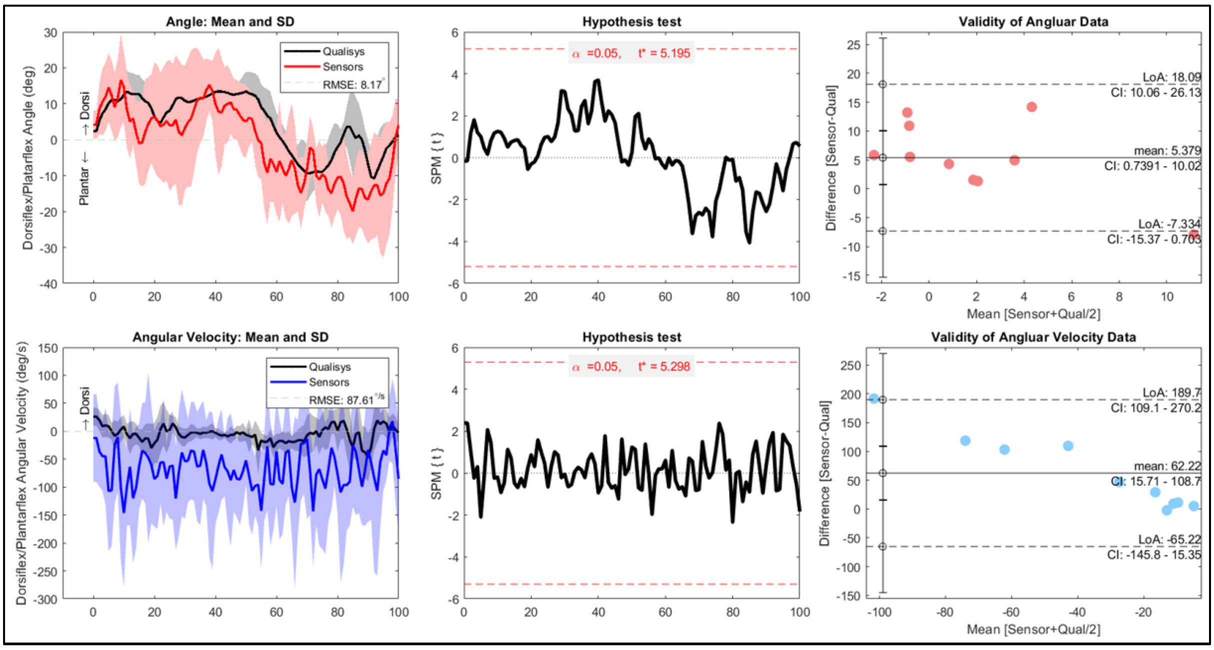


Frontal plane

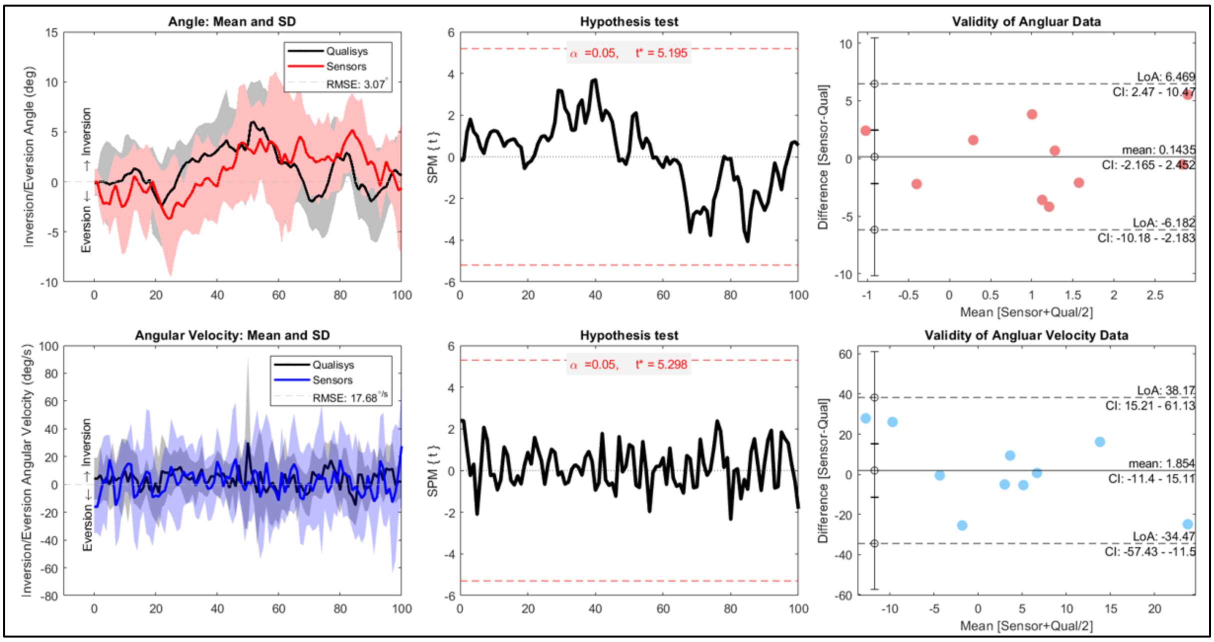


ANNEXURE H – SLOPE ACTIVITY COMPARATIVE ANALYSIS

Sagittal plane

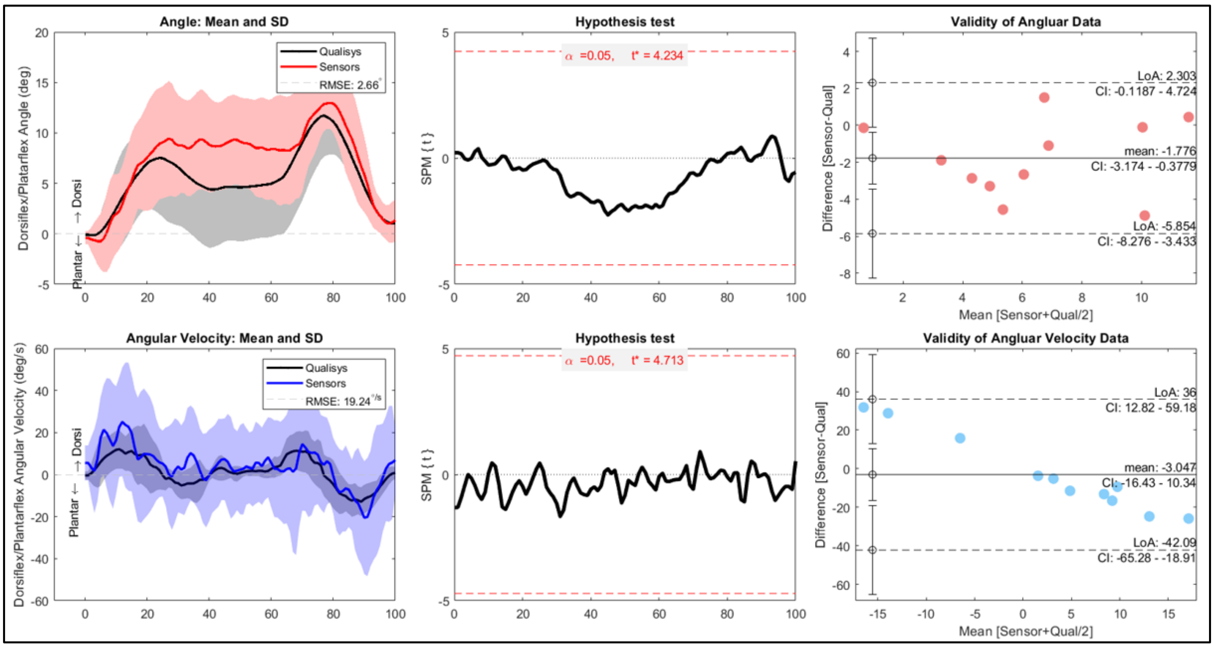


Frontal plane



ANNEXURE I – STAND AND SIT ACTIVITY COMPARATIVE ANALYSIS

Sagittal plane



Frontal plane