UNIVERSITY OF STRATHCLYDE

Faculty of Engineering

Department of Biomedical Engineering

MASTERS DISSERTATION

MSc Biomedical Engineering

A Comparison of Human Walking Using a Conventional Gait Laboratory and the Motek Medical Computer Assisted Rehabilitation Environment (CAREN)

By

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A dissertation submitted in partial fulfilment of the requirements of the award of a

Master of Science of the University of Strathclyde

August 2014

UNIVERSITY OF STRATHCLYDE

DEPARTMENT OF BIOMEDICAL ENGINEERING

DECLARATION

This thesis is the result of the author's original research. It has been composed by the author and has not been previously submitted for examination, which has led to the award of a degree.

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Signature of Student

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___15th August 2014____

Date

ACKNOWLEDGEMENTS

There is no possible way to say that I accomplished this thesis on my own. This work was possible thanks to the support of a numerous amount of people who helped me throughout this project.

I would primarily like to thank my father, Manuel, and my mother, Rita. Without their support and guidance, I would not have arrived where I am today. They have been there for me in all aspects of my life, both personal and academic. They always inspired me to follow my dreams, and have helped me in every step of my career. I would also like to thank my sister, Analisa, and my two brothers, Karl and Samuel, for being a constant support.

My greatest recognition goes to Dr. Andrew James Murphy who has supervised me throughout this project. He patiently guided me to understand and appreciate the importance of this project, and was of great assistance when he helped me tackle the problems encountered during the length of this project. I would like to thank him for sharing his extensive knowledge in the field of motion analysis, which was priceless and appreciated.

I would also like to thank other members of the Department of Biomedical Engineering including Prof. Phil Rowe who has gave me invaluable insight into the field of Biomechanics, Mr. Craig Childs for his assistance in the laboratories and Mr. Stephen Murray for his guidance when using the equipment in the machining laboratory. The participants that took part in the testing phase of this project, who without them this project would have not been possible.

Finally, I would like to extend my gratitude to my many fellow classmates and friends, namely Socratis Dimitris Komaris, who were there in time of need throughout this scholastic year and who have contributed in some way or another towards the completion of this project.

ABSTRACT

The primary aim of this study was to investigate the variation of human gait biomechanics during different walking conditions, with and without optic flow. The secondary aim was to investigate the accuracy of the two motion-analysis systems used. Following ethical approval of this investigation, testing was initiated. Participants walked under four different walking conditions, over ground, treadmill walking, treadmill walking with a virtually generated grass pathway, and treadmill walking with a grass pathway and a simplified interactive avatar. In addition to fully instrument biomechanical analysis of gait, an unstructured interview was used to gain insight to participants' sensations experienced during the different environments. Data was then subjected to an analysis that determined the effect of the optic flow environment on the six chosen gait parameters. A static and dynamic accuracy analysis was performed in both laboratories to determine the accuracy of the motion analysis systems. The study reported a variation in the gait cycle parameters between the four walking conditions evaluated. The results obtained in this study were negatively influenced by the repercussive effects of fixed treadmill speeds and the level of immersion in the virtual environment due to optic flow. The negative effects of fixed treadmill speed had a distinct effect on the cadence and stride length values, while the negative effect of optic flow echoed similar cadences and stride lengths from the treadmill walking environment to the treadmill walking with optic flow (grass pathway only) environment, due to the participants' similar sensations during the two walks. The knee kinematics measured showed that they are not affected by the change in environment, while the ankle kinematics showed significantly different results for the various walking conditions, such that it was speculated that this gait parameter varies in relation to the level of optic flow immersion. Furthermore, the accuracy analysis performed in this study showed both motion analysis systems analysed showed excellent accuracies.

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LIST OF ABBREVIATIONS AND NOTATIONS

HJC	Hip Joint Centre
KJC	Knee Joint Centre
AJC	Ankle Joint Centre
CAREN	Computer Assisted Rehabilitation Environment
CWS	Customary Walking Speed
TSP	Temporal-Spatial Parameters
WC1	Walking Condition 1 – Over ground walking
WC2	Walking Condition 2 – Treadmill walking
WC3	Walking Condition 3 – Treadmill walking with grass path way
WC4	Walking Condition 4 – Treadmill walking with grass pathway and a simplified interactive avatar

NOMENCLATURE

Temporal	Changing in relation with time
Spatial	Changing in relation with space
Stride	The distance as measured from one foot strike to the consequent foot strike for the ipsilateral foot
Ipsilateral	Of the same side
contralateral	Of the opposite side
Kinesiology	The study of Human Movement
Anthropometry	The measurement of the size and proportions of the human body.

ANATOMICAL PLANES

Coronal	Divides the body into anterior and posterior portions
Frontal	Synonymous to the coronal plane
Horizontal	Divides the body into equivalent superior and inferior portions
Median	Divides the body into equivalent left and right portions
Midsagittal	Synonymous to the Median plane
Sagittal	Divides the body into left and right portions, which are not necessarily equivalent.
Transverse	Synonymous to the Horizontal plane.

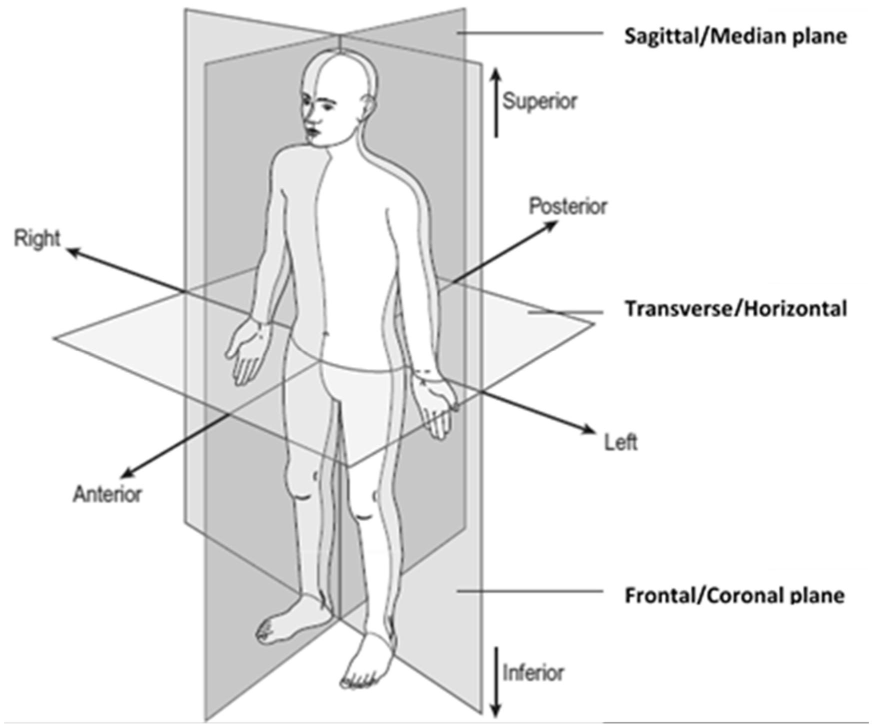


Figure 1: Anatomical Planes
The human body in the anatomical position, with the three reference planes and six fundamental directions. (Levine, Richards and Whittle, 2012)

1 INTRODUCTION

Normal human walking, otherwise referred to as gait, is the method of locomotion involving the use of both legs, alternately, to provide both support and propulsion. Gait has been systematically studied since the renaissance by historical figures such as Da Vinci, Galileo and Newton (Levine et al., 2012). Owing to today's increasingly sophisticated methods of measurement, normal gait may be intensely analysed and interpreted. This provides a standard against which pathological gaits can be compared, in order to be categorised and rehabilitated accordingly. In order to realise patients' gait deviation, a measured understanding of the patients' gait should be performed:

> "I often say that when you can measure what you are speaking about, and express it in numbers, (then) you know something about it"
>
> (Kelvin, 1883)

Clinical gait analysis is the process of making these measurements, to understand patients' gait in order to help the clinician provide the most efficient rehabilitation possible. With the help of the state-of-the-art optoelectronic motion analysis systems, patients' gait may be analysed in real-time, giving immediate and accurate.

The primary objective of this project is to compare and evaluate the variation in the gait of healthy individuals when walking in four different environments:

1. Over ground normal walking (in a conventional gait laboratory),
2. Treadmill walking,
3. Treadmill walking with virtual reality based optical flow of a moving grass path
4. Treadmill walking with virtual reality based optical flow of a moving grass path and an interactive simplified avatar.

The testing is aimed at providing knowledge on the benefits and ill effects of over ground walking and treadmill walking with and without virtual reality assisted optic flow during rehabilitation testing. Treadmills are nowadays considered as an important tool for clinical and human movement biomechanics research. They offer a convenient and controlled environment which promotes ease of data collection due to the lesser spatial volume used, in relation to over ground walking, when operating it. Still, the treadmill imposes drawbacks that are related to the lack of visual feedback from the surrounding environment, which impose gait deviations on the users when using such a device. In this regard, Sloot et al. remarked that when subjects are imposed treadmill speeds along with lack of optic flow from their surroundings, their gait parameters would be altered when compared to subjects who perform normal over ground walking (Sloot, van der Krogt and Harlaar, 2014).

The secondary objective of this project challenges the accuracy of state-of-the-art motion capture systems in order to provide an increased understanding of the inter-systems accuracy along with an intra-system accuracy. This objective is expected to recognise any improvements (or flaws) in the accuracy of in-house optoelectronic motion analysis systems utilised in the primary objective. Obtaining a quantitative measure of the systems' accuracy would enhance the fidelity in the results obtained when using such systems.

Following ethical approval of this investigation by the University Of Strathclyde Department Of Biomedical Engineering Research Ethics Committee, testing was performed on healthy participants using an established biomechanical model for the lower extremity.

2 THEORY AND LITERATURE REVIEW

2.1 HUMAN GAIT BIOMECHANICS

Biomechanics is recognised as one of the sub-disciplines of kinesiology, and encompasses the precise mechanical description of human movement. Human gait biomechanics can be considered as a sub-discipline of biomechanics, which studies the way humans walk, within this field one of the primary goals of many research teams is to increase understanding about rehabilitation strategies for pathological gait.

2.1.1 THE GAIT CYCLE

The gait cycle is defined as “any method of locomotion characterised by periods of loading and unloading of the limbs.” (Kirtley, 2006)

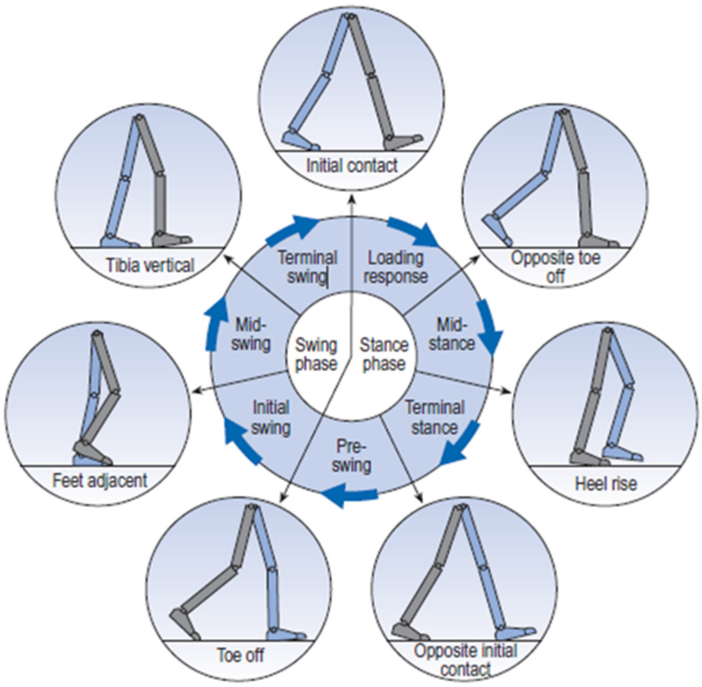


Figure 2: The Gait Cycle
Leg positions during a single gait cycle. (Levine et al., 2012)

The gait cycle incorporates a sequence of predictable events, which take place in a precise order over a specific period. It is divided into two major phases, the *stance phase*, which refers to the instants when the foot is in contact with the ground, and the *swing phase*, which refers to the instant when the foot is swinging forward in preparation for the next step.

2.1.1.1 THE STANCE PHASE

This phase is the most important and complex phase of the gait cycle since the lower limb has to provide a firm support to balance the body weight on one leg while allowing for forward progression. This phase can be subdivided into a further five events, as follows:

- *Loading Response (0-10%):* This event initiates with the moment the foot touches the floor and continues until the other foot lifts off the ground. This event allows the legs to absorb the initial shock during weight bearing while allowing for stability.

- *Mid-Stance (10-30%)*: This event takes place during the first half of the single limb support interval of the gait cycle. It initiates from the moment the contralateral foot lifts from the ground to the point when the body weight aligns directly over the fore foot. This event aids the musculoskeletal system to provide limb and trunk stability while it progresses over the stationary foot.

- *Terminal Stance (30-50%)*: This event takes place during the last half of the single limb support interval of the gait cycle. It initiates from the moment the heel raises from the ground to the instant when the other foot strikes the ground. This event allows the body weight to transfer in front of the fore foot.

- *Pre-Swing (50-60%)*: This is the final event on the stance phase. During this interval double-stance occurs (both feet are in contact with the ground). It initiates from the moment the contralateral foot touches the ground to the instant when the primary foot leaves the ground. This event allows the primary limb to transfer its weight onto the secondary (contralateral) limb while it prepares for the rapid demands of the swing phase.

2.1.1.2 THE SWING PHASE

This phase allows for the presently unloaded limb to progress and swing in front of the stance limb in order for forward progression to occur. This phase is subdivided into a further three events, as follows:

- *Initial swing (60–73%)*: This event initiates when the foot lifts from the ground and ends at the instant when the swinging foot is opposite the stance foot. This event allows time for the foot to rise from the ground enough to have clearance to move forward.
- *Mid-Swing (73-87%)*: This event brings the swinging foot further in front of the stance foot until the tibia is vertical. This allows the hip to flex, to bring the thigh anterior to the body weight line of action, for forward progression to occur.
- *Terminal Swing (87-100%)*: This is the final event of the gait cycle and initiates with the tibia in the vertical position and ends with the foot striking the ground in preparation for another cycle. This event allows the knee to extend the tibia in order to prepare the foot for contact with the ground.

2.1.2 GAIT PARAMETERS

Once the concept of the gait cycle is grasped, the next step is to understand the different types of parameters, which can be measured during the gait cycle. Literally hundreds of discrete parameters may be distilled from a fully instrumented analysis of human gait for quantitative comparative reasons. For clarity and brevity, only those parameters that are later analysed in full are described here.

2.1.2.1 TEMPORAL SPATIAL PARAMETERS

Temporal spatial parameters (TSPs) of gait are also known as general gait parameters. These parameters are mainly comprised of *walking speed*, *cadence* and *stride length* (Robinson and Smidt, 1981). TSPs give an idea of the subjects' walking ability, but with little specific information.

- *Walking speed*

This parameter is most commonly expressed in meters per minute (m/min) in order to simplify compatibility with other TSPs (such as cadence, which is expressed in steps per minute). Walking speed can be simply calculated using the following equation:

Equation 1: Walking Speed

$$Walking\ speed = \frac{Distance}{Time}$$

Although each individual person has a customary walking speed (CWS), the actual speed is adjusted continuously according to the conditions in the surrounding environment (Waters et al., 1988). Two studies by Murray et al. (1964) and Murray et al. (1970) showed that the CWS on a smooth level ground averages 82 m/min for healthy adults, where men were found to be 5% faster and women 6% slower than the group mean. The standard deviation reported for CWS is approximately 10% of the mean value (Waters et al., 1988 and Murray et al. 1964), where 4% of this deviation is related to leg length. For individuals below the age of 60 years, age was found to have no impact on this deviation from the. Thus, the greater factor of deviation seemed to be spontaneous variability, i.e. variability due to momentary sub-conscious walking speed deviation.

- *Cadence*

This gait parameter is normally dependant on walking speed and is used to specify the number of steps taken per minute. Cadence can be simply calculated using the following formula:

Equation 2: Cadence

$$Cadence = \frac{Number\ of\ steps\ (in\ one\ minute) \times 60}{Time\ (seconds)}$$

All healthy adults have a natural cadence, which is normally a little less than 120 steps/minute on normal ground at normal walking speed. This varies in relation to the leg length of the individual (among other variants), since longer legs will produce a slower cadence. This can be appreciated by noticing the average cadence for healthy adult women (Figure 3), whose cadence is 117 steps/min during their own CWS, while the average cadence for healthy adult men is 111 steps/min *(Perry, 1992)*. The mean adult cadence for both men and women is 113 steps/min.

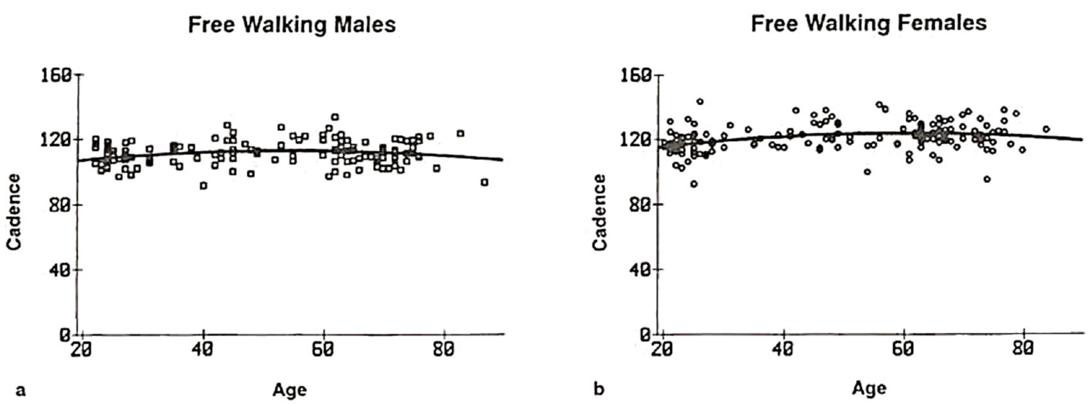


Figure 3: Normal cadence during CWS for (a) Males, and (b) Females.
It can be noted that the average for women is slightly lower than that for men throughout their lifespan, which is normally due to women being shorter than men are on average. (Perry, 1992)

- *Stride Length*

A stride is the distance between two consecutive steps, one for each foot (Figure 4). Stride length is constantly altered along with walking speed and cadence, during normal day-to-day activities to adjust to the surrounding environment. It is observed that stride length, walking speed and cadence can be related to each other by the following equation:

Equation 3: Stride Length

$$Stride\ length = \frac{120\ \times walking\ speed}{Cadence}$$

Generally, stride length is measured as the length between two successive foot strikes. Stride length for a healthy adult performing a walk at CWS, averages at 1.41 metres. Men are reported to have a 14% longer stride length than women (Perry, 1992).

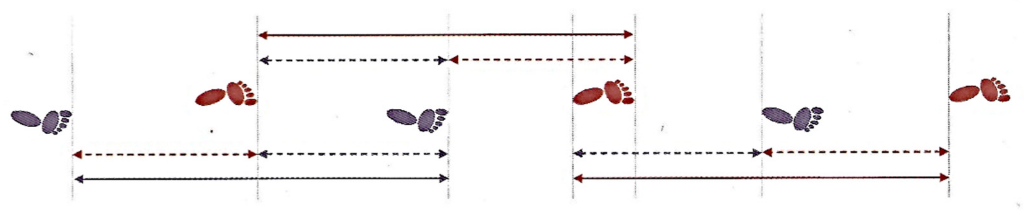


Figure 4: Stride Length
A schematic diagram explaining how stride length is measured. (Baker, 2013)

2.1.2.2 KINEMATIC PARAMETERS

Kinematic parameters describe the gait cycle in terms of the *angles*, *displacements*, *velocities* and *accelerations*. In this study only anatomical sagittal plane kinematics (flexion/extension of joints) are investigated. In the following description of human walking kinematics, discrete values are indicative only:

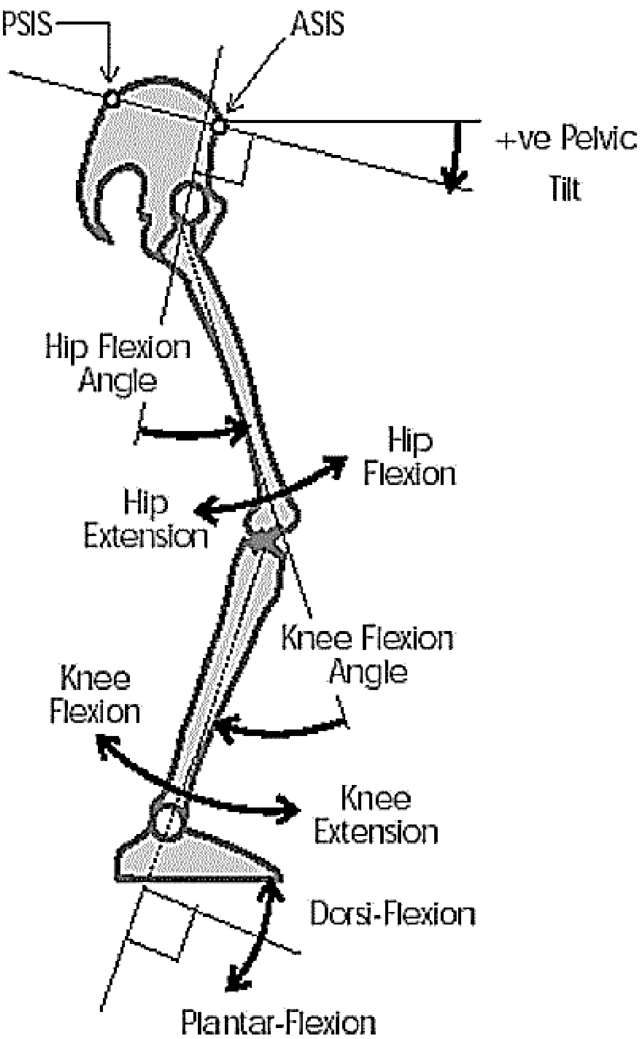


Figure 5: Anatomical sagittal plane kinematics.
An anatomical diagram showing all the sagittal plane kinematics of the lower extremity.

- *Hip Flexion-Extension Angle (Perry, 1992)*

During one gait cycle, the hip moves through two major phases: increasing extension during stance phase and increasing flexion during swing phase (Figure 6). At initial contact, the hip flexes at 30° and remains at approximately this position until the end of the loading response. At mid stance, the hip gradually extends until the hip joint reaches its neutral alignment at approximately 38% of the gait cycle. The thigh then adopts a posterior alignment with respect to the hip, reaching peak extension of 10°, during which the contralateral foot contacts the ground at 50% of the gait cycle. During pre-swing, the hip reverses its direction and begins to flex back to its neutral position at 0°, which occurs close to the end of the stance. The hip joint continues flexing during initial swing, when it reaches 15° of flexion, until the end of the mid-swing phase, when it reaches approximately 35° of flexion. The flexion angle remains within ±5° throughout the terminal swing phase.

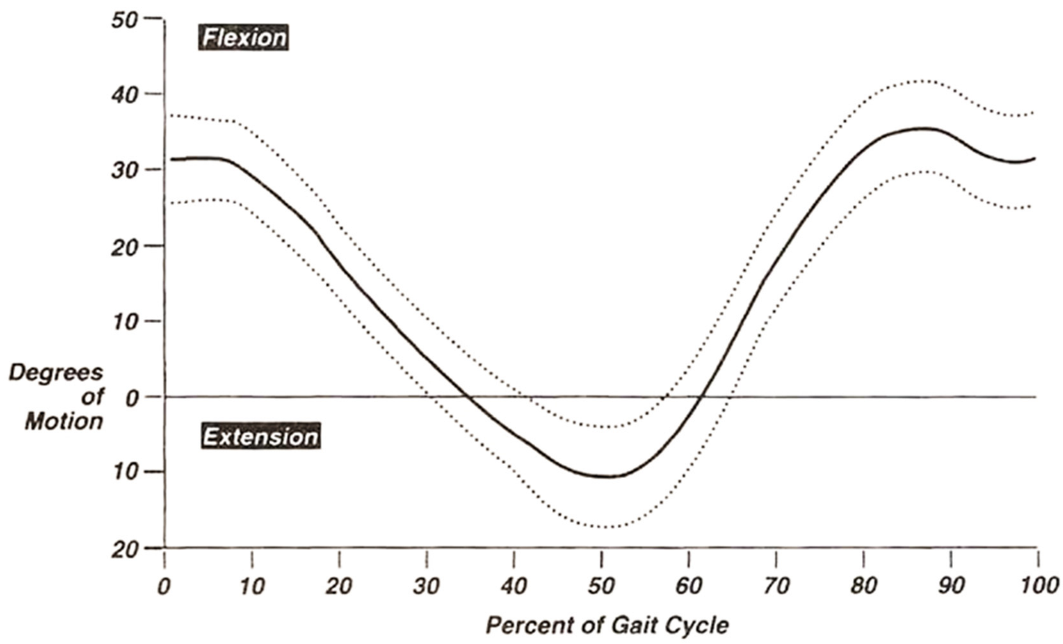


Figure 6: Hip range of motion during one gait cycle.
Black line shows the mean values, and dotted lines show one standard deviation from the mean. (Perry, 1992)

- *Knee Flexion-Extension Angle (Perry, 1992)*

During one gait cycle, the knee flexes and extends in an alternating fashion, through a range of motion of 0° to 70° (Figure 7). At initial contact the knee is located at 5° of flexion, although Inman *et al.* (1981) stated that this value may range from slight hyperextension (-2°) through flexion (5°). Throughout the loading phase, the knee flexes to 18°, and subsequently stops following the start of limb support (at approximately 15% of the gait cycle). The knee then extends from mid stance until it reaches a minimum of about 3° flexion at 40% of the gait. The second period of flexion begins towards the end of the terminal stance phase, and rapidly increases with the initiation of double-limb support. At the end of the pre-swing phase (62% of the gait cycle), the knee is at 40° of flexion and keeps flexing until the end of the initial swing phase. A maximum knee angle of 60° is reached at the beginning of the mid-swing phase, although it should be noted that Murray et al. (1964) reported a maximum flexion angle of 70° by healthy individuals during a normal walk. Following a brief pause in mid-swing the knee changes direction and extends through the remaining portion of the mid swing and throughout terminal swing phase. Full extension is achieved at approximately 97% of the gait cycle with a knee joint angle of between 5° of flexion to 3° of hyperextension. Finally, the knee goes back to 5° of flexion in preparation for the initial contact of the next gait cycle.

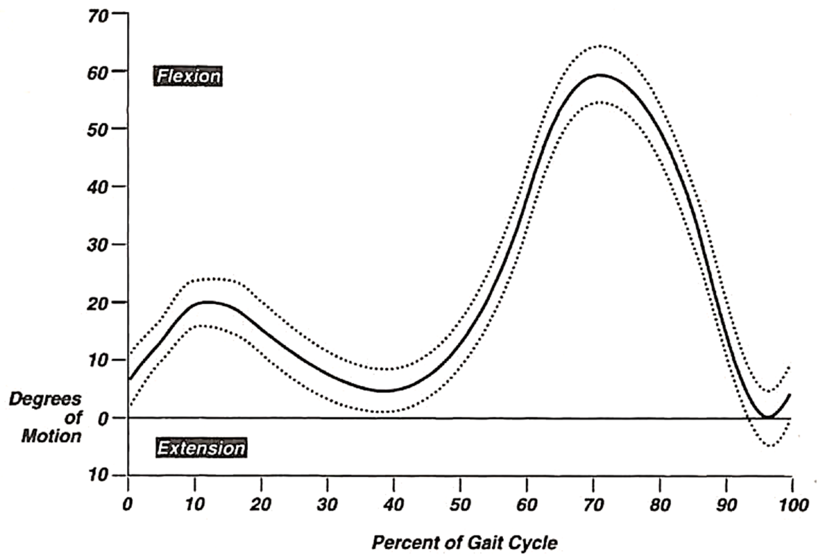


Figure 7: Knee range of motion during one gait cycle.
Solid line shows the mean values and dotted lines shows one standard deviation from the mean. (Perry, 1992)

- *Ankle Dorsi-Plantar Flexion Angle (Perry, 1992)*

During one gait cycle, the ankle plantar flexes and then dorsi flexes twice in an alternating fashion, through an ankle motion range of 30° (Figure 8). At initial contact, the ankle is at its neutral position, although plantar flexion of 3° - 5° is common in some individuals. During the loading response, the ankle starts plantar flexing, and as soon as forefoot contact occurs (foot flat), the ankle changes direction and initiates to dorsi flex. At this instant, the foot is stationary and the tibia does all the motion. Dorsi flexion continues through mid-stance and the first half of the terminal stance, reaching a maximum angle of 10° by 48% of the gait cycle. During the pre-swing phase, the ankle changes direction and starts to slowly plantar flex, until a rapid ankle plantar flexion occurs towards the terminal double support phase, reaching a maximum angle of 20° of plantar flexion. At the end of stance phase, toe-off initiates the final dorsiflexion movement, which gradually moves the ankle joint towards its neutral position. This neutral position is reached by mid-swing, and maintained for the rest of the swing phase, except for a small drop of 3° - 5° of plantar flexion during terminal swing phase.

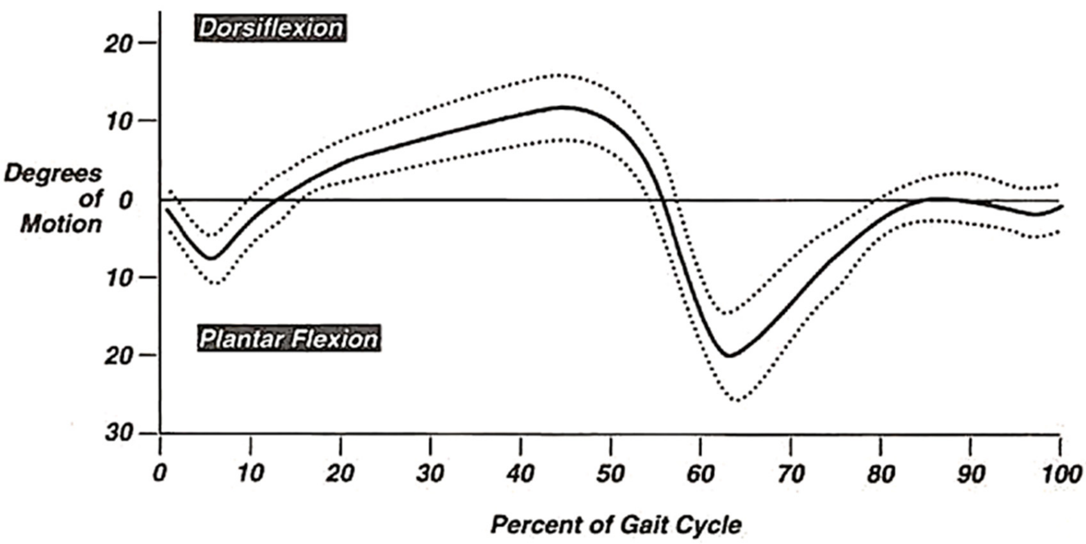


Figure 8: Ankle range of motion during one gait cycle.
Solid line shows the mean values and dotted lines show one standard deviation from the mean. (Perry, 1992)

2.1.2.3 KINETIC PARAMETERS

Kinetic parameters describe the gait cycle in terms of the *forces* and *moments* acting on and within the lower extremity of the human body. Due to constraints dictated by the scope of this thesis, the kinetic parameters of the gait cycle will not be investigated.

2.1.2.4 GAIT GRAPHS

The aforementioned gait parameters are visualised using gait graphs. These type of graphs are used to analyse and compare the measured gait parameter with reference data (Figure 9).

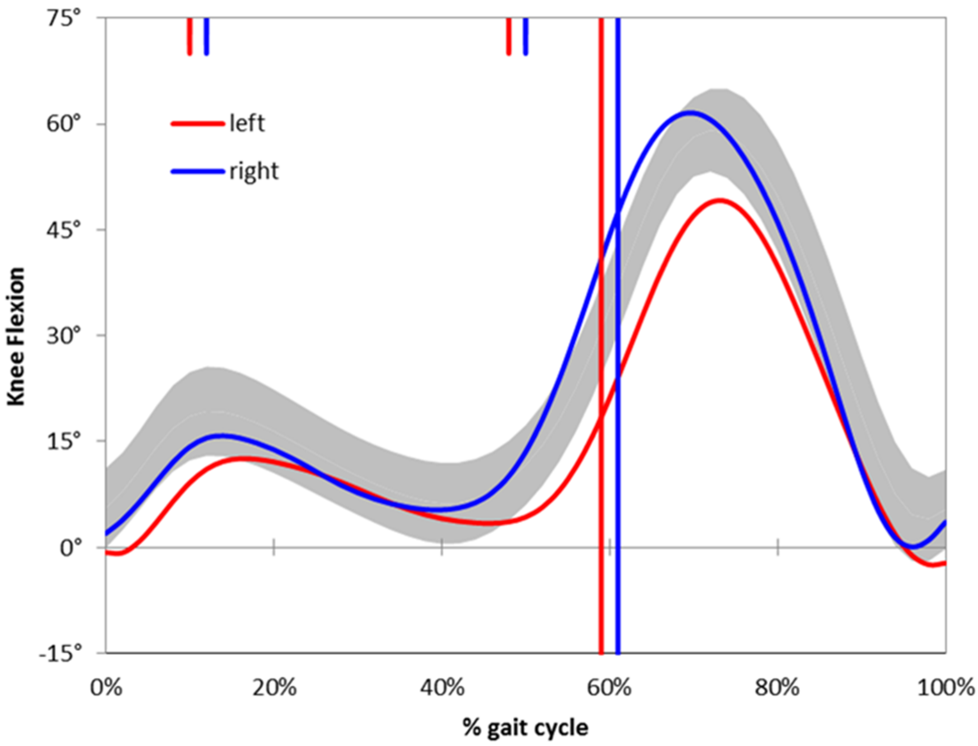


Figure 9: A typical gait graph.
This graph shows the knee flexion angle during one gait cycle. Reference data is given for one standard deviation (grey). (Baker, 2013)

2.1.3 MOTION ANALYSIS

Motion analysis (motion capture), consists of the acquisition of gait parameters and the anthropometric data, in order to study participants' movement. Motion analysis allows the gait analyst to assess participants' movements repeatedly under different conditions depending on the undertaken study.

Research into human movement using motion analysis has matured to its present level in the last few decades, and is now considered as an essential tool for biomechanical research, clinical diagnosis of pathological gait and other physiological abnormalities (Gage, 1993, and Cook et al., 2003). Motion analysis may be conducted using a variety of tools and hardware configurations, including (but not limited to) video based systems, optical and infra-red based systems, electromagnetic systems, inertial based, and (electro)goniometry based systems.

The motion capture hardware used in this project was optical infra-red based hardware produced by Vicon (Vicon Motion Systems, Oxford, UK), which allows optical tracking of small reflective markers within a 3-dimensional capture volume. The reflective markers are fixed on palpable anatomical locations on the individual being recorded, and the system records the trajectories of these markers, thus being able to track the individual's motion in 3D.

In the following subsections, an overview of the motion analysis laboratories used, and the biomechanical model chosen for this study is given.

2.1.3.1 3D MOTION ANALYSIS LABORATORIES USED IN THIS STUDY

All testing sessions took place within the University Of Strathclyde Department Of Biomedical Engineering Biomechanics Suite. Specifically, a state-of-the-art Conventional Human Movement Analysis Laboratory (Figure 10), and an installation of Motek Medical's Computer Assisted Rehabilitation Environment (CAREN) Extended system (Figure 12). A full description of the relevant hardware and configuration is provided in the Methodology (Section 3.1).

The Conventional Human Movement Analysis Laboratory comprises:

- Vicon T-Series system, using six model T-160 cameras and six model T-40S cameras,
- Four Kistler piezoelectric based force platforms, and
- A 10-meter walk way

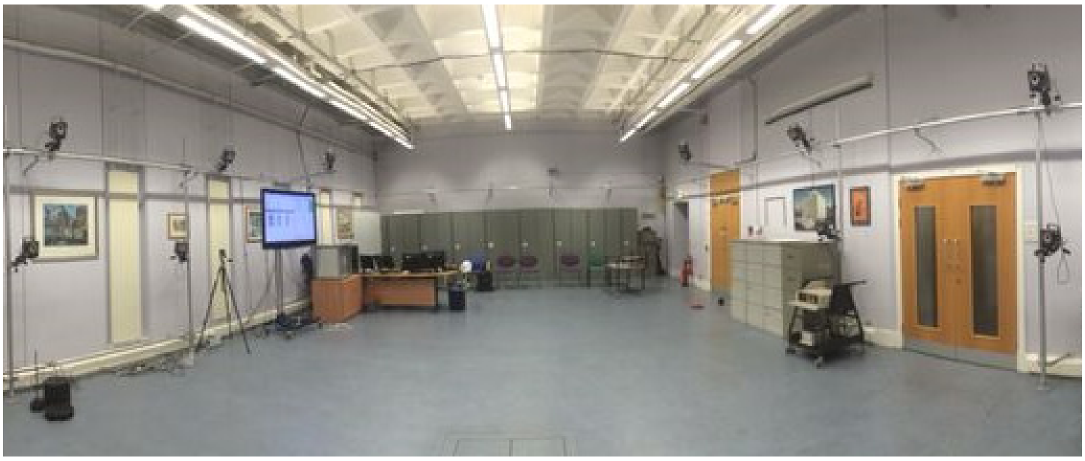


Figure 10: The Conventional Human Movement Analysis Laboratory used in this study.

Vicon T-160 cameras have a 16-megapixel resolution at a frame rate of 120fps (Figure 11), allowing the user to accurately capture finer details of the motion of the reflective markers. The frame rate can be increased (while diminishing the resolution) up to 2000fps in order to capture fast moving objects. Vicon T40S cameras have a 4-megapixel resolution at a frame rate of 515fps (Figure 11), which means that this type of camera is capable of capturing clearer quality images of faster moving objects. Moreover, both camera models also include a "Full Marker Grayscale" feature, which allows the cameras locate the reflective markers in a way that not only their edge shape is obtained. Thus, the system is able to calculate the radius, centre and diameter of the marker in 2-D more accurately using every pixel of grayscale information, thus improving system accuracy and precision.

Figure 11: The Vicon(R) Optoelectronic Motion Capture T-Series Cameras used in the Conventional Gait Laboratory.

The CAREN extended system comprises:

- Vicon Bonita System , using 12 Model B-10 cameras,
- A 3-meter diameter MOOG E-6DOF (MOOG, Inc.) electrically actuated platform,
- A Bertec (Bertec Corporation) 1x2m dual belt treadmill instrumented with two force plates, and
- A 180° Projection screen, along with four image generators and projectors.

Figure 12: The Motek Medical CAREN Extended system used in this study.

Vicon B-10 cameras have a 1-megapixel resolution at a frame rate of 250fps (Figure 13). These type of cameras are documented to be able to capture markers with a precision of 0.5mm in a 4m x 4m volume. In addition to this, they have the advantage of having a wider field of view when compared with the Vicon T series cameras, which allows the cameras to capture markers in a wider volume. The wide field-of-view feature, allows the cameras to be used closer to the subject, which is advantageous when using the relatively compact CAREN configuration.

Figure 13: The Vicon(R) Optoelectronic Motion Capture B-10 Series Cameras used in the Motek Medical CAREN extended laboratory.

The CAREN Extended system generates a virtual environment, which has the ability of complete user immersion. This system is designed for clinicians to utilise innovative rehabilitation techniques, and to obtain measurements of comprehensive studies and evaluations. The virtual environment moves at the same velocity as that of the treadmills, thus providing appropriate optical flow that mimics real-life over ground walking in an outdoor simulated pedestrian environment. The system also makes use of two-oversized Bertec force platform, which minimises changes that might occur in the gait patterns of the subject due to any width constraints of the treadmill.

The MOOG motion platform contains six hydraulic actuators, which are controlled independently to enable motion in six degrees of freedom: medio-lateral translation, anterior-posterior translation, superior-inferior translation, pitch, yaw and roll (Sinitski, Lemaire and Baddour, 2013).

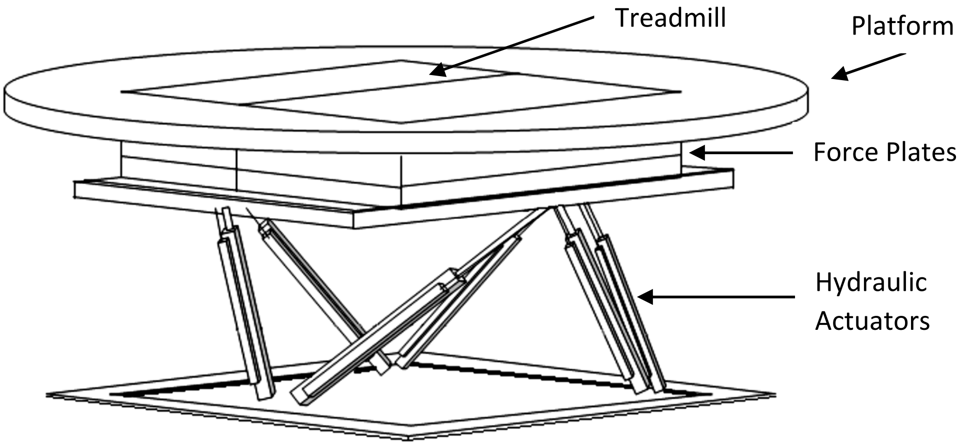


Figure 18: MOOG E-6DOF motion platform (showing labelled components) which is installed in the Motek Medical CAREN Laboratory. (Sinitski, Lemaire and Baddour, 2013)

2.1.3.2 PLUG-IN GAIT BIOMECHANICAL MODEL

Due to the human body's complexity, a biomechanical model is necessary to achieve a useful explanation of how the body parts move in relation to each other. The Vicon Plug-in Gait (PiG) model portfolio is the most widely verified and used model in the field of clinical motion analysis (Figure 19). PiG divides the lower extremity of the human body into seven segments; the pelvis, two femurs, two tibias, and two feet. These segments are connected by joints, which all have 3-degrees of freedom.

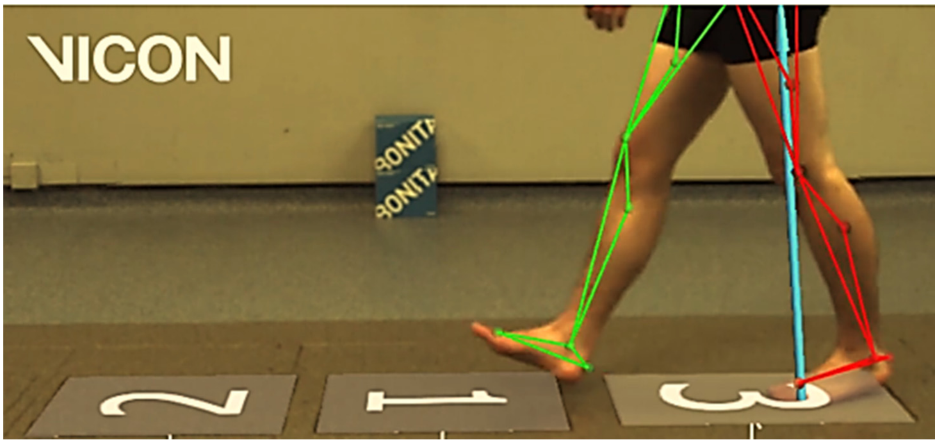


Figure 19: The Vicon PiG model showing the virtual model over-and-above the real markers. (Vicon, 2013)

2.1.3.3 MARKER PLACEMENT

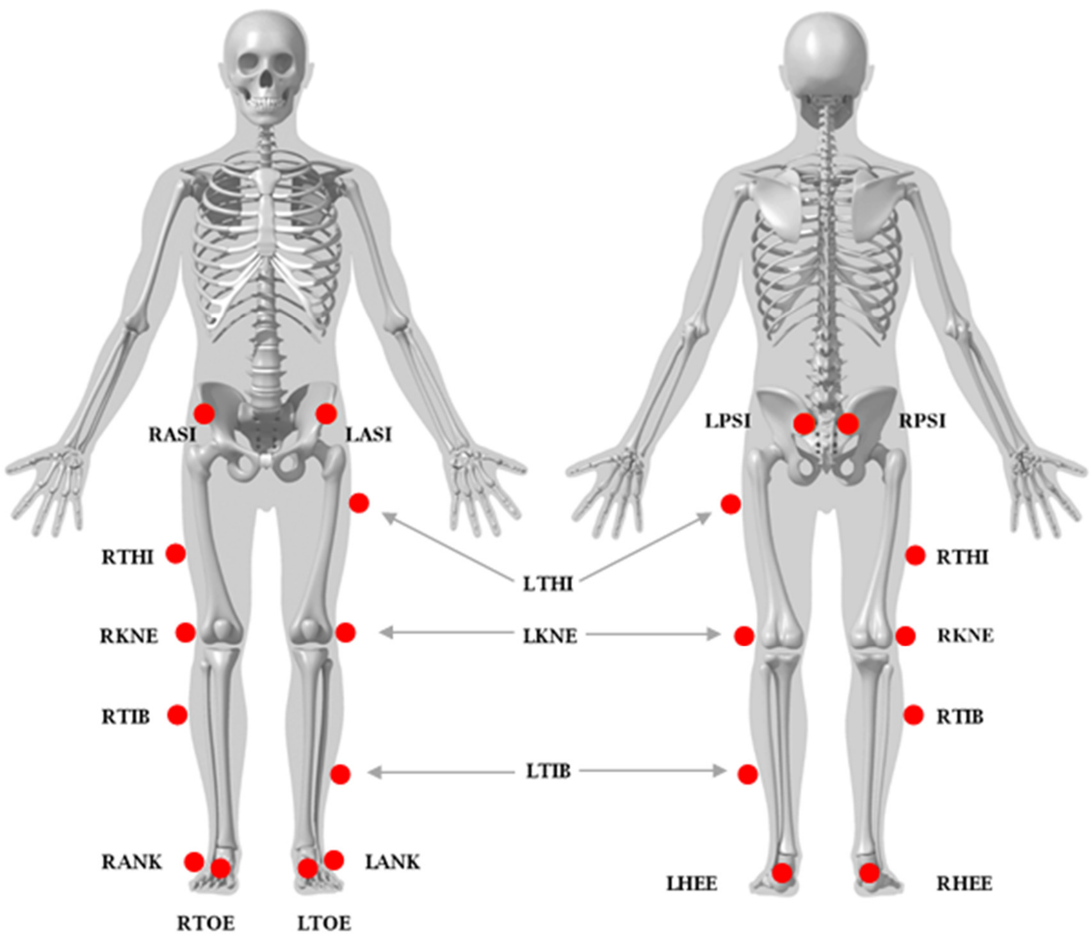


Figure 20: A model that can be used as an aid when applying the lower body PiG reflective marker set to the correct anatomical locations.

For the lower limb version of the PiG model that is going to be used in this study (Figure 20), 16 reflective markers are required (Table 1).

Table 1: A summary of all the markers used in the lower limb version of the PiG model, showing their label, location and anatomical location.

Marker Label	Marker Location	Description
LASI	Left ASIS	Placed directly over the left anterior superior iliac spine.
RASI	Right ASIS	Placed directly over the right anterior superior iliac spine.
LPSI	Left PSIS	Placed directly over the left posterior superior iliac spine.
RPSI	Right PSIS	Placed directly over the right posterior superior iliac spine.
LKNE	Left Knee	Placed on the lateral epicondyle of the left knee.
RKNE	Right Knee	Place on the lateral epicondyle of the right knee.
LTHI	Left Thigh	The greater trochanter of the left femur is located, and the marker is placed 1/3 of the way, on the superior part of an imaginary line between this location and RKNE marker.
RTHI	Right Thigh	The greater trochanter of the right femur is located, and the marker is placed 1/3 of the way, on the inferior part of an imaginary line between this location and LKNE marker.
LANK	Left Ankle	Placed on the lateral malleolus of the left ankle.
RANK	Right Ankle	Placed on the lateral malleolus of the right ankle.
LTIB	Left Tibia	Placed on the inferior 1/3 of an imaginary line between the LKNE and the LANK markers.
RTIB	Right Tibia	Placed on the superior 1/3 of an imaginary line between the RKNE and the RANK markers.
LTOE	Left Toe	Placed over the second metatarsal head, on the mid-foot side of the equinus break between forefoot and mid-foot.
RTOE	Right TOE	Placed over the second metatarsal head, on the mid-foot side of the equinus break between forefoot and mid-foot.
LHEE	Left Heel	Placed on the calcaneous at the same height of the LTOE marker.
RHEE	Right Heel	Placed on the calcaneous at the same height of the RTOE marker.

2.1.3.4 MARKER TRACKING AND RECONSTRUCTION

In order to track the markers during motion capture accurately, the motion analysis system needs to be calibrated accordingly. For this reason, static and dynamic calibrations are performed prior to capturing any data.

Static calibration is performed by placing a calibration wand (Figure 21) which contains reflective markers with known inter-marker distance, in the origin of the laboratory. The cameras capture the calibration wand and then computer software calculates the relationship between the known 3-dimensional positions, and the 2-dimensional positions captured in the field-of-view of the cameras. These relationships are used along with sophisticated algorithms to calibrate the cameras with respect to the global laboratory coordinate system. A dynamic calibration is performed by waving the calibration wand around the capture volume in the laboratory, in sight of all the cameras. This generates a number of simultaneous equations, which are solved, via algorithms by the specialised software, in order to determine the precise relationship of each camera to the calibrated volume. Once this is done, any point within this calibrated space can be tracked with high accuracy, as long as two cameras have the marker in their field-of-view.

Figure 21: The Vicon Calibration Wand. (One Measurement Group Ltd., n.d.)

Following calibration, PiG now requires the motion analyst to measure and input specific anthropometric dimensions into the motion analysis software, specifically: body mass (in kilograms), height (in mm), leg lengths (in mm), knee widths (in mm) and ankle widths (in mm). The participant is then calibrated to the system, by statically taking, what is referred to as, the T-pose (Figure 22), during which the motion analysis system captures the locations on the markers on the subject being analysed. The marker locations are then assigned their respective marker label names, which are tabulated in Table 1.

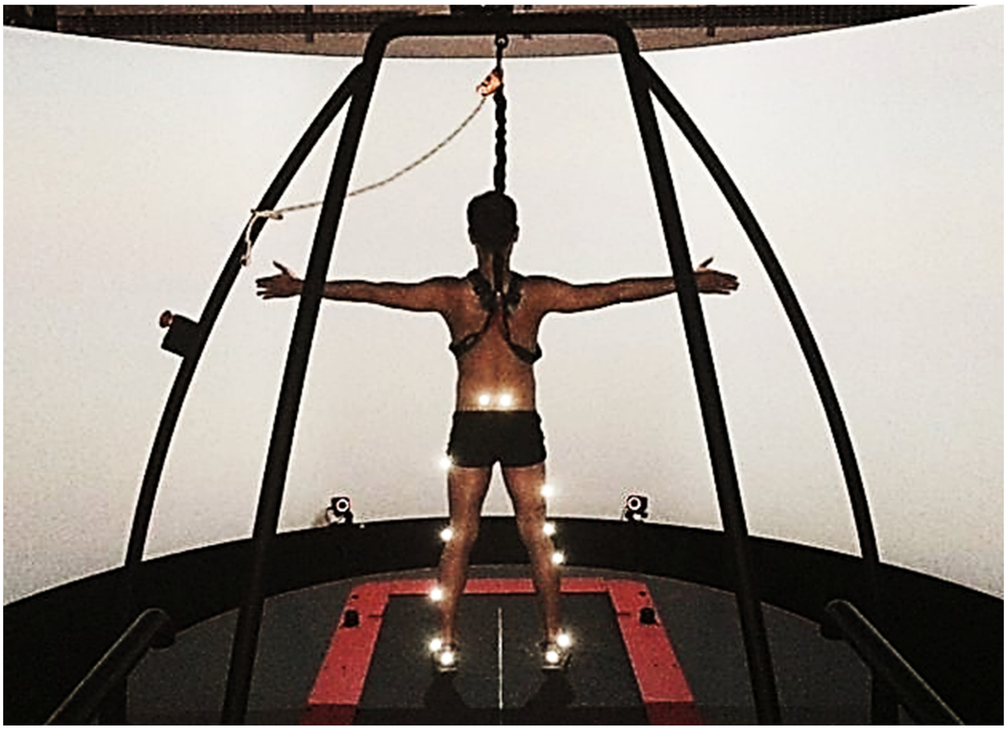


Figure 22: Calibration T-Pose from a posterior view.

The system further uses the labelled markers to define the seven aforementioned lower body segments. Each segment is assigned a three dimensional coordinate system which is embedded within it, which allows its orientation to be described. A segment is defined by 3-points, which are formed by a line representing the principal axis (such as the femur axis, having the hip joint centre and the knee joint centre as its edges), and a point. Together, these 3 points form a plane, which lies in one of the anatomical planes. The principal axis gives the alignment of the segment, while the point measures how much rotation exists about that axis. Using this information obtained from then sixteen marker locations, and the anthropometric data for the participant being analysed, then the motion analysis software is capable of building the PiG model of the subject. This PiG model will be later used, along with the marker trajectories of the motion performed, to measure the TSPs and joint kinematics, as required.

2.1.3.5 CONSEQUENCE OF MARKER MISPLACEMENT

Correct marker placement is of paramount importance if the clinician is to have confidence in the quality of the data that is captured. "Marker misplacement is one of the most common sources of variability in clinical gait analysis" (Baker, 2013). Misplacing a marker can lead to incorrect output of gait parameters, depending on which of the markers might have been misplaced. A concise description of the gait parameters (used in this project) affected by certain marker misplacements is given in the following sub-sections:

- *Pelvic Markers*

The pelvic markers are the LASI, RASI, LPSI and the RPSI. If one of the ASIS markers is too high, this will raise the mid-point between the ASIS markers by half the offset distance, which will in turn tilt the pelvis in a way that a mild increase in hip extension could be recorded.

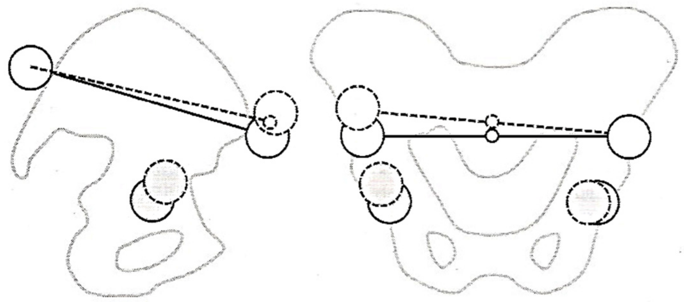


Figure 23: The effect of misplacing the ASIS marker, on the estimated location of the HJC (Hip Joint Centre). Solid lines show original positions, and dashed lines show altered positions. (Baker, 2013)

A change in the orientation of the pelvis (that is, a misplacement of any of the pelvic markers) will change the estimated position of the hip joint centre in the biomechanical model. If an ASIS marker is placed higher than it is supposed to be, then the ipsilateral hip joint centre will be higher by the same amount, which could have an effect on the knee and ankle joint centres, and thus their kinematics. Finally, if the PSIS markers are placed too high, then the pelvis will appear to be tilted more anteriorly, and the hip will appear to be more flexed than it really is.

- *Knee and Thigh markers*

The knee and thigh markers are the LTHI, RTHI, LKNE and the RKNE. If the thigh marker is placed anterior to its theoretical location, then the sagittal plane of the femur will be internally rotated, and thus knee flexion and extension will no longer occur in line with the thigh. This marker offset will lead to the phenomenon called cross talk where knee flexion will be mistaken for knee adduction by the system. This can be easily understood by a person standing in front of another person walking with an externally rotated thigh towards the former person. When the knee of the walking person flexes, the observer will appear to see knee adduction, although in reality the knee is not adducting at all. In the same way, a LTHI or RTHI marker, which is offset anteriorly, will replicate the same effect and vice-versa is offset posteriorly.

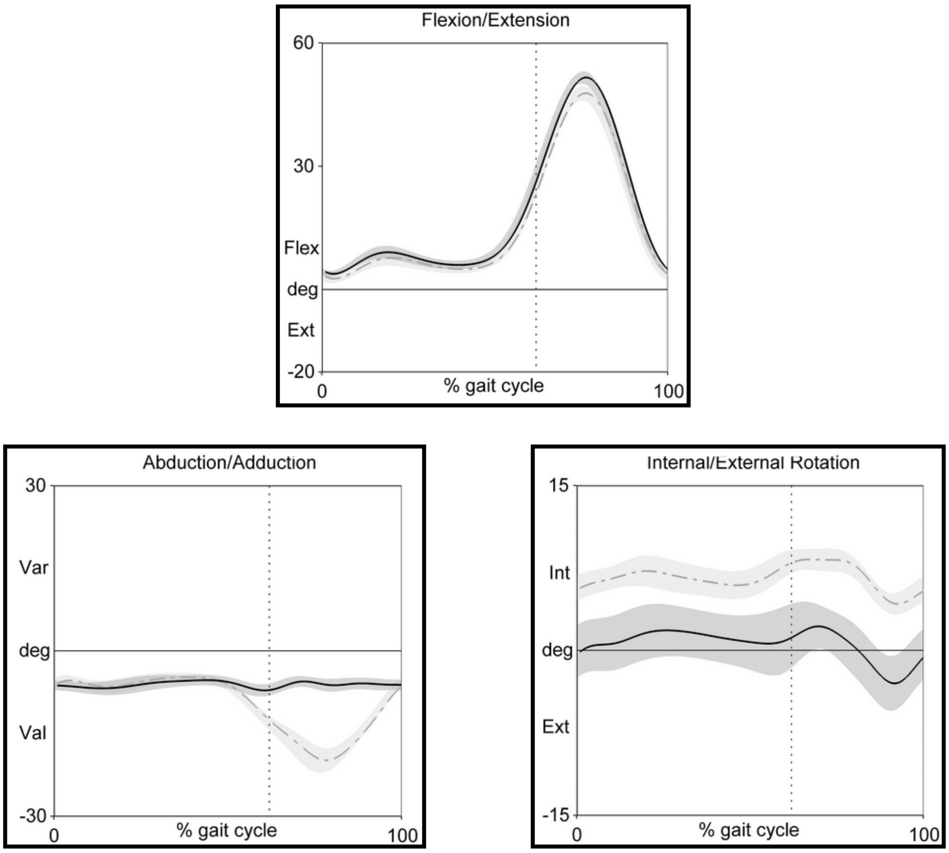


Figure 24: Knee Joint Kinematics Gait Graphs.
These graphs show the effect of cross talk, due to an anteriorly misplaced thigh marker. The solid line shows the knee joint kinematics obtained for a correctly oriented thigh marker, while the dotted line shows the knee joint kinematics obtained for an incorrectly placed marker (anteriorly). (Baudet *et al.*, 2014)

Furthermore, if the thigh marker is placed anterior, then the coronal plane of the thigh would be internally rotated, such that the system will estimate the knee joint centre to be more posterior (Figure 25). This will lead for a slight underestimation of knee flexion along with a small effect on ankle dorsiflexion. The opposite will occur, if the knee markers are placed anterior, since the knee joint centre will now be estimated to be more anterior than its true anatomical location, which will lead to a slight over estimation of hip and knee flexion along with ankle dorsiflexion.

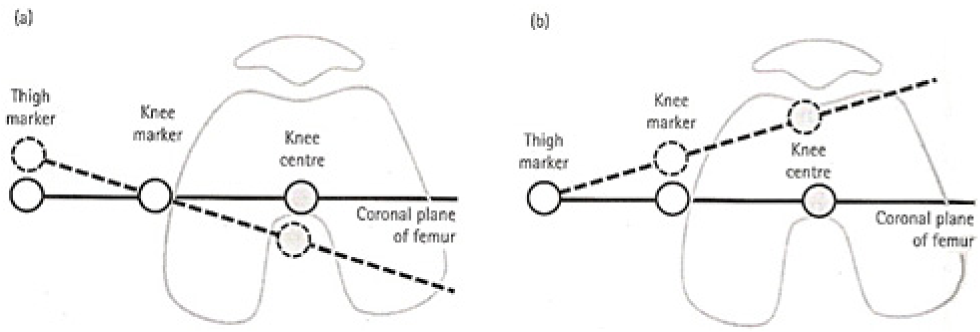


Figure 25: The effect of misplacing the (a) thigh markers, and (b) knee markers on the estimated location of the KJC (Knee Joint Centre).
Solid lines represent accurate marker placement, and dashed line represent misplaced markers and the effect on the estimated KJC. (Baker, 2013)

- *Tibial and Ankle markers*

The tibial and ankle markers are the LTIB, RTIB, LANK and RANK. For the gait parameters of interest in this project, this marker set does not largely affect any of the parameters, if there is a slight offset in the marker position. Nonetheless, it is good to point out that similar occurrences will ensue if misplacement of the thigh and knee markers takes place. If the LTIB and RTIB markers are placed anterior to their theoretical position, the shank's sagittal plane would be internally rotated. Since the range of motion of the ankle is much smaller than that of the knee, the affect that this internal rotation will have of the sagittal plane kinematics of the knee and ankle is nominal. Moreover, anterior placement of the tibial markers would lead to a slightly offset ankle joint centre estimation in the coronal plane, but since the ankle and tibial markers in the coronal plane are much closer to each other, with respect to the thigh and knee markers, the offset will again create negligible effects. On the other hand, if the ankle markers are placed anteriorly, then the estimated AJC will be anterior, thus the system makes a minor underestimation of knee flexion and ankle dorsiflexion.

- *Foot Markers*

The foot markers are the LHEE, RHEE, LTOE and RTOE. If the imaginary line between the heel markers and toe markers is not parallel to long axis of the foot, this will affect the sagittal plane kinematics of the foot according to how tilted the imaginary line is to the foot axis. Attention should be given to locate the heel markers to be in level with the toe markers, as the system will consider the line between these two markers to be parallel to the long axis of the foot during the static calibration. Misplacement of these makers would therefore offset the ankle angle by a few degrees throughout the trial captures.

2.1.3.6 INVERSE KINEMATICS

When the motion capture is completed, the trajectories of all the physical markers and other computed locations (such as HJC, KJC and AJC) are saved by the system. The biomechanical model then uses these trajectories along with the anthropometric data of the subject to perform what is referred to as inverse kinematics to calculate the kinematic data that is requested by the user (Figure 26). In order to measure the sagittal plane kinematics required for this project the biomechanical model uses Euler angles. The model compares the embedded coordinate systems of each segment to derive the relative orientation of two adjacent segments.

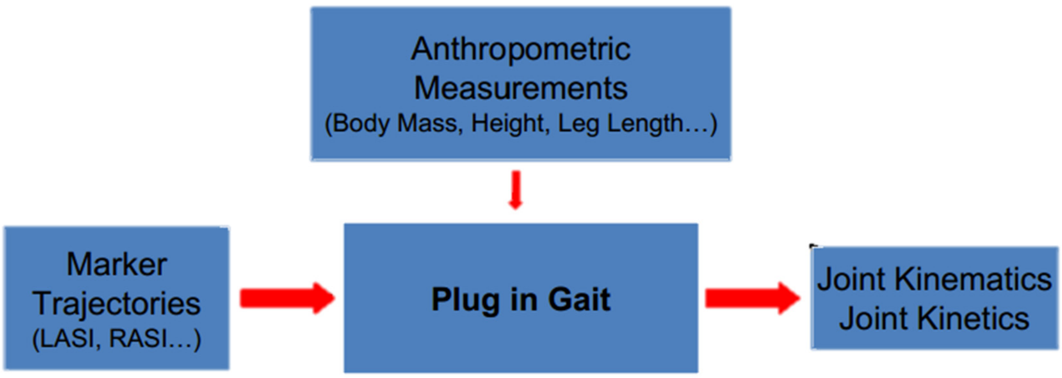


Figure 26: A process chart showing the inputs needed by the Plug-in Gait model to output the Joint Kinematic. (Paolini, n.d.)

2.1.4 INFLUENCE OF THE ENVIRONMENT ON GAIT PARAMETERS

Many factors can influence the production and execution of human movement. These factors could be classified into two categories, *intrinsic* and *extrinsic* factors. Intrinsic factors consist of the physiological processes within the body and the body's psychological situation. Conversely, extrinsic factors can also be referred to as environmental factors. These can be due to surrounding objects, which the brain considers as obstacles, surrounding physical circumstances and surrounding psychosocial circumstances. In this project, the effect of a treadmill-walking environment and the effect of different types of optic flow during treadmill walking are going to be investigated, in relation to the gait parameters discussed in section 2.1.2. A short introduction to the influence of these environments on human movement is given in this section, while a more thorough literature review on previous studies performed concerning this matter is reported in the subsequent section.

It is often required to study gait while the subject walks on a treadmill rather than over ground. This allows the researcher or clinician to control the walking speed of the subject in a convenient environment, which requires a smaller capture volume. When utilising a treadmill, subjects can be conveniently connected to breathing tubes, or wired for tailored testing equipment. Moreover, a treadmill environment offers a safer environment for the subject, since an overhead weight-bearing structure with an accompanying harness can be used to provide assistance in case the subject loses balance or falls over (this is appreciated more in subjects with pathological disorders).

However, treadmill gait imposes subtle changes in the subjects' natural gait, particularly concerning sagittal joint kinematics, stride length and cadence (discussed further in section 2.1.5). The reason for these alterations in the subjects' gait are thought to result from several extrinsic environmental factors, such as the subjects' awareness of the limited length of the treadmill belt, which may cause the subjects to shorten their stride. Other factors may be, the feeling of walking on an artificial surface, which would alter the neuromuscular feedback and sensation, and the small changes in speed which the subject experiences due to the treadmill belt decelerating during foot strike and accelerating during foot off, effectively storing energy in the treadmill motor. These minor variations in treadmill speeds are reduced when using a large treadmill with a powerful motor (Savelberg et al., 1998), which is the case for the treadmill installed in the CAREN extended system in the laboratory used in this study.

Finally, it should be noted that when walking on a treadmill the subjects experience a lack of optic flow that creates a phenomenon known as perceptual conflict. Perceptual conflict occurs because humans are used to have relative visual movement in relation to their surrounding environment when walking. Therefore, when walking on a treadmill an eccentric sensation occurs, because of the subject walking at certain velocity while the environment remains static, which was found to have an effect on the subjects' gait (Lee and Hidler, 2008).

These extrinsic factors from the environment are sensed by exteroceptive receptors in the sensory system of the human body, and the information obtained from these receptors is combined with information obtained from interoceptive receptors. This information is used by the brain to plan the next series of movements in order to reach the final destination. Now, if we take the case were no optic flow exists, in theory the brain will notice a change in the external environment and will therefore alter the gait accordingly.

2.1.5 LITERATURE REVIEW OF THE EFFECT OF TREADMILL AND OPTIC FLOW ON GAIT PARAMETERS

Although the treadmill is thought to be an essential tool for studies incorporating clinical and sports biomechanics research, the question remains whether the advantages of using the treadmill outweigh the disadvantages. Several researchers have previously studied whether or not treadmill walking (TW) does resemble over ground (ordinary conventional) walking, with respect to participants' gait TSPs and kinematics. In 1983 Strathy *et al.* compared TW to over ground walking (OW) (Strathy, Chao and Laughman, 1983). The researchers compared the kinematics of the knee with patterns of foot-to-floor contact, and identified that during TW toe contact time and lower heel contact time were higher, while a higher cadence was required on the treadmill in order to keep the same velocity as that of OW. Furthermore, the knee's range of motion during TW was smaller in the sagittal plane, while a smaller angle of extension during heel strike was noticed. Murray *et al.* (Murray et al., 1985) investigated further and found that hip angles were smaller in extension during stance phase during TW, and suggested this was because of the reduced step length. They went further and reported that a smaller maximum ankle dorsiflexion angle during stance phase occurred because of the knee tending to be less extended during TW. The study also showed that subjects showed that the cadence was higher with a resulting shorter stride length during TW. These two studies above, although they gave in-depth reasoning for their results, several important variables in the methodology weren't explained in detail, such as the surface landmarks used to calculate joint centres and methods of data filtering. Thus, the data reported by these researchers will only be adapted to this project qualitatively (i.e. when comparing the results of this project with those reviewed in literature studies, the results from these two studies will not be considered quantitatively since the source of the data and its processing is not clear)

Alton *et al.* (1998) performed a comparative kinematic analysis of over ground and treadmill walking. 17 healthy participants were marked with 5 markers on their right hand side, and a 3D Kinematrix® motion analysis system was used to track them. The participants walked over ground at their preferred walking speed, while the treadmill speed was worked out for every participant by rounding their average velocity over ground to the nearest 0.2 km/hr. The study's results are shown in Table 2.

Table 2: Results obtained by Alton et al. (Alton et. al., 1998)

Variable	Over ground walking		Treadmill Walking		P-Value
	Mean	St. Dev.	Mean	St. Dev.	
Cadence (Step/min)	117	6	122	4	0.0003
Stride length (m)	1.32	0.14	1.37	0.13	0.0027
Max. Hip Flexion Angle (°)	28	4	32	4	0.0001
Max. Knee Flexion Angle (°)	69.9	3.43	70.59	3.48	0.3
Max Ankle Dorsiflexion Angle (°)	39	6	39	7	0.94

Alton *et al.* were in agreement with both Strathy *et al.* and Murray *et al.* concerning cadence, and reported that the cadence on the treadmill had increased with the means increasing by five steps per minute from OW to TW. It was suggested that the higher cadence was due to the sense of urgency the subject feels to place the foot of the swinging limb onto the treadmill, as the contralateral limb is being dragged backwards behind by the treadmill. The stride length in this study increased during TW, which disagreed with all previous studies reported, except for a study by Wall and Charteris (Wall and Charteris, 1981), who only identified a longer stride length in the initial ten minutes of treadmill walking and attributed it to inexperience treadmill users. Wall and Charteris go on and suggest that this longer stride length leads to an increased hip range of motion, which explains the increase in the maximum hip flexion angle, which was obtained in this study only. Finally, no significant differences were noted in knee and ankle maximum angles, which was alike to that reported by other previous studies. Keeping in mind that this study is more than 15 years old and the technology available had its limitations, the methodology taken is considered to have taken the necessary precautions. The author points out that in future studies, the participants should be asked to report any sensations experienced during the TW in comparison to the over ground walk, the participants should be allowed to get used to TW environment, and recording during the first two minutes of the TW are not recommended. This was reflected in the methodology followed in this study, since the participants recruited were asked to be experienced in treadmill walking in order to avoid confounding factors in our study. In addition, the participants were allowed time to accustom themselves to the treadmill walking environment. Following each motion capture on the treadmill, the participants in this study will be asked to comment on the sensations they felt when walking in the different walking conditions, in accordance with the recommendation suggested by these researchers. By taking this approach, any outlying data could be checked for any relation with a specific sensation, which the participants would observe.

Riley *et al.* (Riley et al., 2007) performed a similar analysis between over ground and TW to that undertaken by Alton *et al.* This study was the first of its kind to investigate the kinetics of TW, but since this goes beyond the scope of this project, only the kinematic comparison of the study will be reviewed. 33 healthy subject between the ages of 18 and 35 years were recruited, 4 of which were eliminated due to technical problems during testing, and another 3 eliminated due to a body mass index (BMI) exceeding 30. The researchers eliminated these subjects since they felt that a BMI larger than 30 would affect the gait of the individual. The remaining 26 participants (equivalent numbers of males and females), were fixed with a PiG model marker set, and a Vicon 624 motion analysis system was used to track their movement. The subjects were asked to walk at a self-selected CWS along the 15m walkway and the same velocity was recorded and applied on the treadmill for the TW trials. The subjects were allowed to familiarise themselves to the TW prior to capture, although the time made available for this familiarisation process was not documented. Results from Riley *et al.* are shown in Table 3.

Table 3: Results for the TSP variables obtained by Riley *et. al.* (Riley *et al.*, 2007)

TSP Variable	Over ground walking		Treadmill Walking	
	Mean	St. Dev.	Mean	St. Dev.
Cadence (Step/min)	114.13	8.3	113.7	8.11
Stride length (m)	1.55	0.13	1.48	0.12
Walking Speed (m/s)	1.48	0.18	1.41	0.16

Table 4: Results for the sagittal plane kinematics obtained by Riley *et. al.* The negative sign shows that the TW values were larger than the over ground values. (Riley *et al.*, 2007).

Variable	Mean Difference	St. Dev
Max. Hip Flexion Angle (°)	0.64	1.31
Max. Knee Flexion Angle (°)	0.68	1.74
Max Ankle Dorsiflexion Angle (°)	-1.69	5.92

Riley *et al.* offered that the TSPs were smaller during TW than for over ground walking since the velocity used for TW was kept constant throughout the experiment, and it was noted that the treadmill velocities were slower than the over ground velocities. Although in previous studies (Strathy *et al.*, Murray *et al.* and Alton *et al.*) the cadence values were reported to increase in TW, it was argued by the researchers that according to Matsas *et al.* (Matsas, Taylor and McBurney, 2000) if the subject is allowed *only* 6 minutes of practice on the treadmill, the cadence difference will vanish. A lot of debate was noticed in all the papers reviewed up to this point, regarding the time allowed for the subjects to accommodate themselves to the treadmill environment. With hindsight to the Equation 3 mentioned in section 2.1.2.1, which related cadence with walking speed and stride length, the researchers should have pointed out that the slight decrease in cadence in this study is attributed to the fact that the walking speed on the treadmill was slower than that over ground. The relationship states that the cadence and the walking speed are proportional to each other; therefore, it makes sense that the cadence would be smaller if the walking speed was slower. Riley *et al.* noted statistically significant decreases in the peak hip, knee and ankle flexion angles of TW. This does not agree with Alton *et al.* and Wall and Charteris's reports, although both of them reported that this could have been due to errors in their methodology. In contrast, Murray *et al.*, Strathy *et al.* and Matsas *et al.* all similarly reported reduced knee range motions with treadmill walking. The authors finally pointed out that the kinematic differences that they found were all within the range of repeatability of the gait parameters' variability. The decision taken by the researchers to control the participants taking part in this study will be utilised in this study in order to avoid lurking variables in the output data. Therefore, participants will only be recruited if they fall in a specific range of height and BMI. The data obtained during the testing phase of this study, will also be check for any statistical significance between the peak hip, knee and ankle dorsi flexion angles, for the different walking conditions, which will be evaluated. This data could then be quantitatively compared to the data reported in the above studies.

Lee *et al.* (Lee and Hidler, 2008) undertook a similar study to the one done by Riley *et al.*, were they did a comprehensive analysis of the temporal-spatial gait parameters, joint kinematics, joint kinetics and muscle activation patterns during the two walking modalities. In this literature review, only the TSPs and kinematic comparisons of this study will be reviewed since the other parameters go beyond the scope of this project. 19 healthy participants were recruited, where 8 of the participants (4 men, 4 women) were in the age range of 50 – 70 years old, while 11 participants were between 18-30 years old. This wide range of ages was done on purpose to determine any age-related difference in the two modes of walking, but the researchers commented that no age-related differences were found in any of the gait parameters investigate, although it was pointed out that older participants were extremely fit. The motion analysis system used an active marker system and consisted of a single CodoMotion CX1048 infrared camera station to capture the lower body active markers. The participants were asked to walk over ground on an approximately 5 meter long walkway at their CWS. The average speed was obtained from the first 3 over ground trials, and this value was used as the treadmill speed. About 3 minutes of treadmill walking was allowed for the participants to accommodate themselves to the TW environment, after which 30-second trials were captured. After the trials, Visual 3D was used to create subject-specific link segment models and to obtain the gait parameters. The results obtained for the TSPs and the joint kinematics are reported in Table 5.

Table 5: Results obtained by Lee et al. The TSPs are dimensionless because the researchers normalised the data to account for individuals various heights. (Lee and Hidler, 2008)

Variable	Over ground walking		Treadmill Walking		P-Value
	Mean	St. Dev.	Mean	St. Dev.	
Cadence (dimensionless)[1]	45.1	4.0	46.6	4.5	0.28
Stride length (dimensionless)[1]	0.73	0.09	0.71	0.08	0.86
Walking Speed (dimensionless)[1]	0.27	0.04	0.28	0.04	0.95
Max. Hip Flexion Angle (°)	31.5	4.0	31.4	4.1	0.97
Max. Knee Flexion Angle (°)	69.1	4.3	67.7	4.7	0.48
Max Ankle Dorsiflexion Angle (°)	13.9	4.2	12.2	4.4	0.15

A repeated-measure ANOVA statistical analysis was performed between three random trials for over ground walking and three random trials for TW, for all the gait parameters listed in Table 5. All the parameters showed no statistical difference between the two modalities since the P-value reported was greater than the significance level (α) of 0.05. Lee *et al.*, questioned the aim of their own study, when they referenced a study performed by Van Ingen Schenau (van Ingen, 1979), in which he states that if the treadmill speed remains constant, in theory, there should not be any change in the dynamic behaviour between the two different environments. They go on to mention that they believe that a number of factors could (obviously) violate this theoretical statement, amongst which are the change in velocity of the treadmill during foot strike and foot off. In fact, Lee *et al.* measured a drop of 2.5% of the treadmills' velocity during these instants, but using the kinetic data obtained they noticed that this only affected ankle moments, and could not affect the changes in knee and hip moments which were observed later in the gait cycle. Therefore, they state that perhaps the most likely reason for the differences between these two modalities is the lack of optic flow that the participants come across while performing TW. It was noted that in the statistical analysis performed by these researchers a repeated-measures ANOVA was performed without checking for normality of the data being analysed. According to the central limit theorem, if a sample size is smaller than 30, as is the case in these kind of studies (refer to any participant group size in aforementioned studies), the data does not reflect the population (which is assumed normally distributed). Thus, if the sample is assumed to be normally distributed, as done by Lee *et al.* without checking for normality, the resulting data would have a larger possibility of obtaining a false positive (Type I error) or a false negative (Type II error). (Dixon, 2008)

It has been shown in several studies, that vision plays an important role in gait. Warren *et al.* (Warren et al., 2001), wrote a paper on how "Optic flow is used to control human walking", and suggests that optic flow does in fact alter locomotion control strategies in our mind. Warren *et al.* suggest that a lack of optic flow may alter balance and stability, along with the participants' perception of the relative location with respect to the treadmill. These factors are supported by other researchers, such as Regnaux *et al.* (Regnaux et al., 2006), who remarked that since the two modalities offer different optic flow patterns, the brain would attempt to preserve the kinematic patters of the over ground ambulation and reflect it in the TW ambulation. Consequently, minor changes would be put into effect in the TW gait pattern, which would create a cascade of changes in the gait cycle. Moreover, this hypothesis is supported in three other papers by two different groups of researchers (Carollo *et al.*, 2002, Ivanenko *et al.*, 2004 and Ivanenko *et al.*, 2006), who observed that during human locomotion, the kinematics appear to be the desired control variable. Another study by Stoffregen *et al.* (Stoffregen et al., 2002) documented in the "Handbook of Virtual Environments: Design, Implementation, and Applications", that when walking over ground, a combination of optical flow and vestibular inputs[1] affect the postural control. In an environment where treadmill walking is performed without any optic flow, this combination is not available, since the mind is only receiving the vestibular inputs, while the optic flow is missing. This leads to what is known as perceptual cue conflict, when the mind receives vestibular cues but does not receive kinaesthetic cues.

[1] The vestibular sensory system is located within the ear, and contributes to balance and spatial orientation. The brain uses information from the vestibular system in the head and from proprioceptive systems throughout the body to understand the body's dynamics and kinematics (including its position and acceleration) from one moment to the next.

Consequently, it is assumed that the brain follows the same aforementioned hypothesis, which ultimately varies the gait cycle in a way to preserve the kinematic pattern of over ground ambulation. Thus, in the results of this study it is expected that a trend would be found showing a statistical difference in the TSP data obtained for the over ground and treadmill walking, while the kinematic data would not be expected to be statistically different for the same 2 walking conditions, which would reflect the hypothesis mentioned above.

A study by Sheik-Nainar *et al.* (Sheik-Nainar and Kaber, 2007) reflected the interest raised in the subject on the effect of optic flow on gait ambulation during treadmill walking. The study "investigated the effect of optic flow on gait behaviour during treadmill walking using an immersive virtual reality (VR) setup and compared it with conventional treadmill walking (TW) and over ground walking (OW)." 19 healthy participants (5 women, 14 men) were recruited for this study, with an age range of 21 to 36 years old (24.6 ± 3.7 years) and having 20/20 vision (with or without correction). The recruited participants were experienced in training with a treadmill. The system used to generate the virtually realistic environment, utilised an in-house fabricated Virtual Reality Locomotion Interface (VRLI) setup, developed to provide realistic visual cues to represent real world environments. The VRLI setup consisted of a Biodex RTM 400 rehabilitation treadmill, an Ascension Technologies motion tracking system, a Virtual Research VR8 head mounted display (HMD) to generate a high-fidelity 3D graphical simulations of the real locomotion environment with a 60° field of view and a Silicon Graphics Zx10 Visualize workstation to merge all the equipment feedback together. Four MotionStar sensors were used to track the velocity of the participants (one on each ankle, one on the lower back and one on the HMD), in order to alter the velocity of the VR environment according to the participants velocity. A further six markers were attached to the participants to track their motion. The participants were asked to walk under the three locomotion conditions mentioned previously using their own CWS. To accommodate the participants in the different environments they were asked to stretch and then walk two times along the 35-meter long walkway prior to the OW, walk for 10 minutes on the treadmill before capturing the TW data and finally walk for 15 minutes using the VRLI setup prior to capturing the VR data. The results reported are tabulated in Table 6.

Table 6: Results obtained by Sheik-Nainar et al. (Sheik-Nainar and Kaber, 2007)

Variable	OW		TW		TW with VR	
	Mean	St. Dev.	Mean	St. Dev.	Mean	St. Dev.
Cadence (steps/min)	112.26	13.12	105.45	14.21	110.10	14.07
Stride length (m)	1.38	0.2	1.22	0.18	1.22	0.16
Walking Speed (m/s)	1.29	0.2	1.07	0.17	1.11	0.18
Knee Angle at Foot Strike (°)	4.62	5.12	2.59	4.53	4.63	4.62
Ankle Angle at Foot Strike (°)	6.53	4.68	1.77	3.27	2.28	3.55

Over ground cadences and stride lengths were greater than TW ones, which reinforces the fact that there exists a difference among these conditions, including VR. The results for these two parameters showed partial agreement with Murray *et al.* and Lee *et al.*'s work, although all previous studies were performed at a fixed treadmill speed in contrast to the CWS used in this study. Permitting the participants to use their own CWS allowed the researchers to observe how the perception of walking speed changes for the three different modalities, and hence effects the gait behaviour. This way, the researchers determined that the decrease in stride length and cadence for TW, in relation to OW is primarily attributed to the effect of the lack of optic flow during TW. With reference to Alton *et al.*'s study, it was mentioned that the majority of the participants in their study felt that the treadmill walking speed in the absence of optic flow was faster than their CWS (over ground walking speed), although it was still the same walking speed the participants used in the over ground walking environment. This argument supports the fact that walking in a TW environment with a lack of optic flow degrades the perception of self-motion such that the participants feels the treadmill is moving faster than their CWS. This was reflected in the results obtained for the walking speeds, since the over ground walk revealed the participants' CWS, while the TW showed a drastic decrease in walking speed, which then increased in the VR environment. The knee and ankle angles reported in this study were at heel strike, which do not reflect previous studies (which typically reported the maximum angles for the knee and ankle). Nonetheless, it was noted that the knee angles during over ground ambulation were very close to those during the VR ambulation, while the TW ambulation knee angles at foot strike were much lower than the other two modalities. These values show that optic flow could have affected the participants' kinematics, but the Sheik-Nainar *et al.* commented that this could be due to potential participant anxiety and cautiousness during the TW conditions. Yet, several other factors could have been the reason behind the difference between over ground walking and VR, such as discomfort and disorientation when using the HMD. The HMD device completely covered the subjects' peripheral vision along with any visual contact with their environment including the ground or the treadmill itself. The author stated that this might have created a sense of instability leading to spatiotemporal and kinematic responses, which might have caused the subjects to demonstrate a more cautious behaviour, therefore masking the effect of optic flow on the gait behaviour. In conclusion, this study showed that although the TW and VR modalities produced different results than over ground walking, still the VR gait parameters were, in some situations, closer to over ground conditions. Therefore, it is reasonable to say that the presence of optic flow during VR did influence gait behaviour, resulting in significantly higher cadence and higher speed than in TW conditions, and knee angles for VR approximated those in over ground walking. It should be noted that the limitation used to control the recruited participants based on their vision would be implemented in the recruitment phase of this study to avoid lurking variables. It is expected that the kinematic data, especially the knee kinematics, obtained during this study, during TW would show a tendency to be statistically different from both over ground walking and TW with VR. Conversely, the kinematic data between over ground and TW with VR would not be expected to be statistically different, or less statistically different from the former pair-wise comparison.

The literature reviewed above allows us to understand that whilst it is clear that much work has been done in this sector of gait biomechanics, questions remain, and agreement between studies is not universal. It is clear that there remains room for improvement in the methodologies used, and more studies in line with the ones discussed would enhance the understanding of gait deviations in different walking environments. Thus, the primary objective of this project is justified and it is expected that this study would enrich the insight of certain relationships found in the studies reviewed.

In light of Sheik-Nainar *et al.*'s study, it was noticed that the participants used a self-paced treadmill in contrast with the formerly reviewed studies which always imposed a treadmill speed which approximated the participants CWS. Therefore, two further studies were investigated, which studied the effect of a "Preferred walking speed in a virtual environment" (Bartlett and Sessoms, 2012) and "the effect of a virtual reality environment (VRE) on gait parameters during fixed speed (FS) and self-paced (SP) walking" (Sloot, van der Krogt and Harlaar, 2014).

Bartlett *et al.* recruited 19 participants (12 male, 7 female) having a mean age of 29.2 ± 5 years. The equipment used in this study consisted of the same setup that will be used in this study (i.e. the Motek Medical CAREN extended system). The participants were allowed 6 minutes of familiarisation to the VRE, followed by a 3-minute walk in SP mode and another 3-minute walk in FS mode, both with and without VR. The SP treadmill speed was controlled by the participants' position and speed, while the FS speed was based on an SP trial performed to determine the participants walking speed. It was found that during FS walking the VRE seemed to improve the walking pattern of the participant, approximating it to over ground walking. At SP walking it was noticed that the gait becomes more cautious, upon which the researchers commented that this might be due to a decrease in positional awareness because of the real-time response of the treadmill, which continuously increases and decreases the velocity of the treadmill according to the participants' reactions. The researchers concluded that VRE may be used for clinical gait analysis, but further studies on the control of SP in VRE are required. This study strengthened the author's decision on choosing FS over SP walking during the testing phase of this study, but still a problem arises when it comes to choosing the treadmill speed during FS walking. Thus, the study aforementioned by Sloot *et al.* was reviewed, which gave the author a potential method of approaching this problem.

Sloot *et al.* recruited 20 participants (15 male, 5 female) with an age mean of 28 ± 6.5 years for men and 24 ± 3.5 years for women. The equipment used in this study is the Motek Medical CAREN extended system. The participants' over ground CWS was determined using a GAITRite walkway system (CIE Systems, Inc.), then the participant was relocated in the CAREN system to measure their CWS on the treadmill with and without optic flow. The approach taken to measure the participants' CWS was, to allow the participant to hold a wireless hand held device, which controlled the treadmill walking speed while walking in the VRE. The participants were allowed as much time as necessary to adjust the treadmill speed. The results obtained for this study were as follows:

Table 7: Results obtained by Sloot *et al.* (Sloot, van der Krogt and Harlaar, 2014)

Variable	OW		TW		TW with VR	
	Mean	St. Dev.	Mean	St. Dev.	Mean	St. Dev.
Walking Speed (m/s)	1.27	0.17	0.93	0.14	1.03	0.13

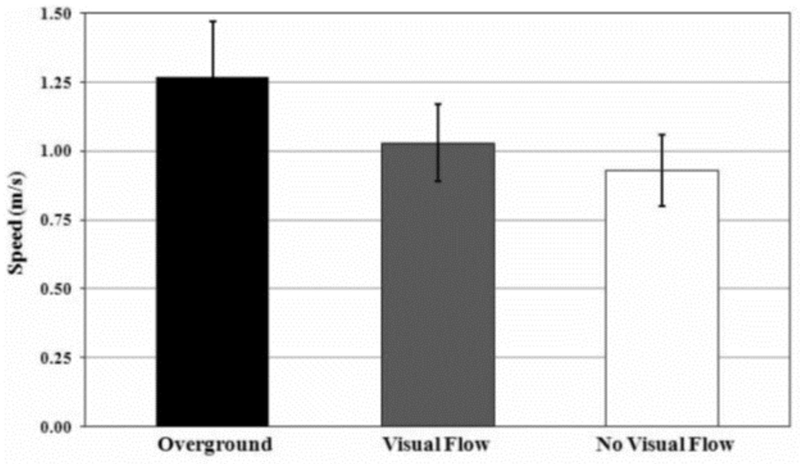


Figure 27: A bar chart showing the mean speed of the 20 participants at each walking condition. Error bars indicate one standard deviation from the mean. (Sloot, van der Krogt and Harlaar, 2014)

The over ground CWS was found to be statistically different from both other modalities, with a P-value smaller than 0.001, while both treadmill conditions were also significantly different from each other, with a P-value equal to 0.002. The researchers commented that although the participants were given unlimited time to choose their walking speed they still selected their walking speed within 30 to 60 seconds. The researchers went on to state that a longer acclimation period while walking in a VRE may have further increased the treadmill walking speeds, but still the VRE produced closer walking speeds to over ground walking speeds. This strengthens the concept that the mind does use optic flow to plan the consecutive gait cycles, and that an increased sense of realism occurs during VRE. The reasoning behind the treadmill speed choice, based on the participants own feedback, used in the study by Sloot *et al.*, will be reflected to a certain extent in this project. Although the CAREN laboratory being used does not have the wireless hand held device used in the aforementioned study, another similar approach will be taken based on the methodology undertaken during this study.

In this literature review, an evaluation of previous studies performed on the effect of walking in different environments (over ground walking, treadmill walking without any optic flow and treadmill walking with optic flow) has been performed. The author focused on the gait parameters that will be investigated further in this study, that is the main TSPs (cadence, walking speed and stride length) and the sagittal plane kinematics (hip joint angles, knee joint angles and ankle joint angles). The latter two studies investigated, focused on the outcome of choosing different modes of walking (fixed speed and self-paced treadmill modes) on the treadmill embedded within the Motek Medical CAREN extended system. These two studies guided the researchers in this study to define their approach on which treadmill mode to use, along with which method to utilise when choosing a treadmill speed in the VRE for the participants.

2.2 ACCURACY OF 3D MOTION ANALYSIS SYSTEMS

All motion capture and analysis systems, including the systems being used in this project to evaluate the gait of the participants, experience measurement errors. Measurement accuracy depends on several factors, primarily the field of view of the cameras, calibration algorithms used by the system manufacturer and the cameras' resolutions. Earlier system had measurement errors of 2-3 mm in discrepancies between the 3-dimensional measured and real locations of the markers, in a capture volume larger enough to capture one gait cycle (Whittle, 1982). Nowadays, thanks to design and calibration improvements, typical errors have been reduced to less than 1mm.

In this study, in line with the previously stated secondary objective, the *accuracy* of the in-house optoelectronic motion analysis systems will be investigated. For the reader's understanding, it is good practice to point out the difference between the technical descriptions of the terms used to define errors in motion analysis systems (Levine et al., 2012):

Resolution	The ability of the system to measure small changes in marker position.
Precision	A measure of system 'noise', based on the amount of variability there is between one frame of data and the next.
Accuracy	The relationship between the real location of the markers and the virtual (simulated) location of the markers.

Although most systems are accurate enough to measure the trajectories of the limbs and the kinematic angles of the joints, a problem concerning accuracy errors in these specific type of kinematic systems emerges. When it is necessary to derive the velocities and accelerations of the markers from their trajectories, a second-order differentiation is necessary. This type of mathematical calculation amplifies any measurement 'noise' in the original data such that wildly erratic and often unusable results are obtained. This is normally solved by applying a low-pass filter to the original data in order to smoothen it out, and subsequent differentiation is performed. However, this process eliminates any genuine high accelerations within the original data, such as those obtained during foot strike.

The motion capture system being used in the conventional gait laboratory in this study utilises a Vicon T-Series camera system, which improves system accuracy by applying a mathematical algorithm known as Full-Marker Greyscale (section 2.1.3.1). This uses every pixel of grayscale information to locate the centroid (geometric centre) of each marker within the cameras' field of view. This algorithm uses the optical grey-scale density of all pixels in the markers' area to fit a virtual 2D circle closely around the marker. The closer the virtual circle is, in relation to the real circumference size and location of the marker, the higher is the accuracy of the system.

Although these methods are implemented to improve the system's accuracy when locating a marker in 3D, error will always remain to some extent. When investigating accuracy of a system, it should be understood which kind of error the investigator would be looking for. There exists two types of errors:

- *Random* errors: these type of errors lead to imprecise systems due to inaccuracies in repeatability measurements, and

- *Systematic* errors: these type of error lead to inaccurate systems due to incorrect calibration of the equipment, incorrect use of system by experimenter, and/or any other error that lead to an incorrect reading of the true value.

In this study, we are interested in systematic errors, since the accuracy of the systems being measured is affected by these type of errors. These errors are generally noticed by taking repeated measurements of the data of interest and subsequently plotting the resulting data on a bar chart. The chart obtained would be expected to follow a Gaussian normal distribution (refer to), which is defined by its *spread* and *mean*. The mean shows the true value which is being investigated, while the spread indicates how much confidence should be placed in the mean value measured by the system, which is represented by *standard deviation*. A systematic error is detected by locating the mean of the resulting data and if it is not equal to the true value, then the system is said to be inaccurate.

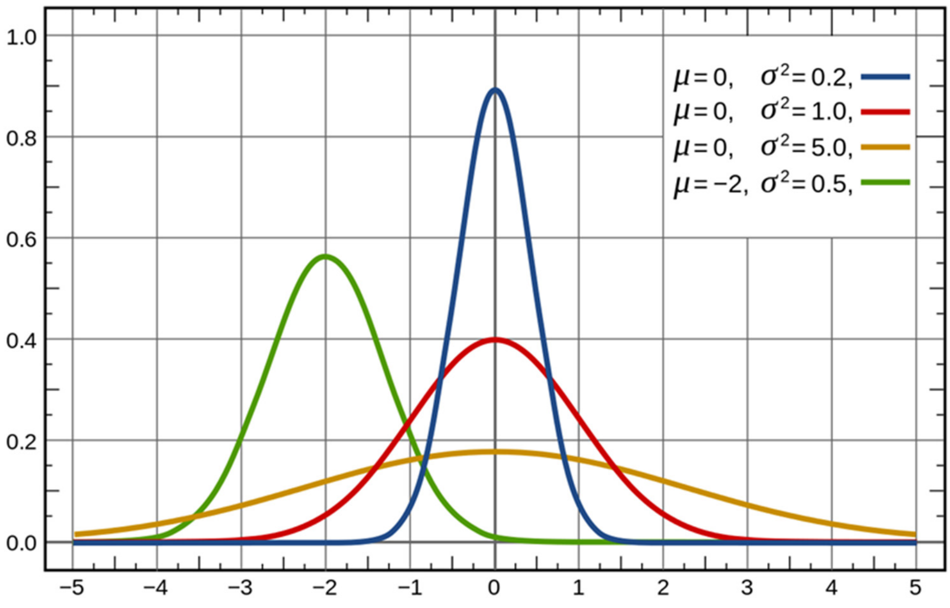


Figure 28: A graph showing the variation that a Gaussian distribution could display. Considering the red distribution as the standard one, then the blue distribution displays an increase in precision, while the orange one displays a decrease in precision. The Green distribution displays a loss of accuracy while displaying a slight increase in precision. (Inductiveload - licensed under Public domain via Wikimedia Commons), 2014)

2.2.1 LITERATURE REVIEW OF STUDIES PERFORMED ON ACCURACY OF 3D MOTION ANALYSIS SYSTEMS

Accuracy testing is considered by some as the ultimate test for any measuring system. In preparation for this literature review, studies that performed an accuracy analysis of motion analysis systems were researched, but although several papers exist regarding systematic accuracy studies of motion analysis systems, most of them go beyond the scope of this project. This literature review will therefore evaluate papers that were considered influential on the accuracy testing performed during this project.

The 'Clinical Gait Analysis Forum' of Japan held a Comparison Meeting of Motion Analysis Systems to compare measurement accuracy of a number of motion analysis systems used in rehabilitation, for which results have been published, by Ehara *et al.* (Ehara, Ebitani and Fujii, 2002). Each company was allowed to choose any biomechanical model of their choice, and a 7-meter-by-7-meter space was allocated for cameras to be set up in any configuration necessary. Each company was required to perform three tasks as follows:

- *Task 1*: A participant walked along a 3-meter long walkway, with a stick, consisting of two markers on each edge, in the participants' hand (Figure 29). The accuracy of the distance between the two markers was then evaluated.
- *Task 2*: A participant walked along the same walkway as in Task 1, but this time the participant was asked to hold an L-shaped device. The 3D coordinates of the three markers were measured by the motion analyses system, and the accuracy of the angle was later evaluated.
- *Task 3*: A participant walked along the same walkway as the two previous tasks, but this time using a set of markers as illustrated in Figure 30. The virtual point A was calculated using the three real markers B, and virtual point B was similarly calculated from the three real markers C. The accuracy of the distance between the 3D coordinates of A and D were later evaluated.

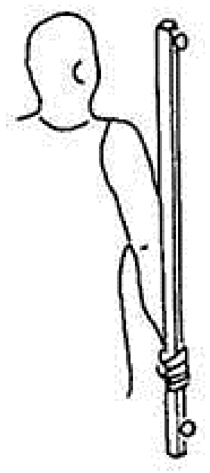


Figure 29: A drawing depicting the placement of the markers during Task 1. (Ehara, Ebitani and Fujii, 2002)

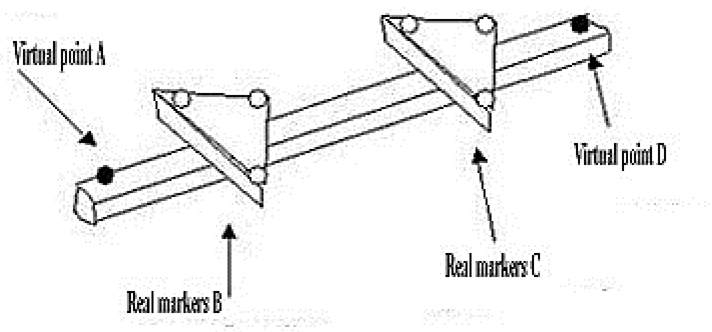


Figure 30: A drawing depicting the placement of the real markers in relation to the virtual markers used in Task 3. (Ehara, Ebitani and Fujii, 2002)

The results reported for the accuracy of all the motion analyses systems in each task are shown in Table 8, Table 9 and Table 10 respectively.

Table 8: A table showing the results obtained for all the systems in Task 1, for the measurement accuracy of distance between the two points (units in mm). (Ehara, Ebitani and Fujii, 2002)

System	True Value	Mean Value	St. Dev.	Average Absolute[1]	Max. Error +[2]	Max. Error - [3]
VICON	899.5	899.92	0.24	0.42	1.04	-0.23
Frame-DIAS II	900.0	896.61	3.81	3.59	1.87	-12.54
Visualeyez	899.0	890.31	1.54	8.69	-6.34	-16.54
PhaseSpace	899.0	929.79	58.69	42.49	137.47	-24.34
Peak Motus Real Time	900.5	905.18	1.02	4.68	7.90	0.11
Peak Motus Video	900.5	897.40	3.11	3.54	3.62	-9.84
Eagle Digital System	902.0	899.23	0.36	2.77	-1.74	-3.65
Pro Reflex	899.0	901.32	0.75	2.32	4.21	0.71

[1] The Average Absolute is the mean of absolute error.
[2] Maximum Error + is the maximum value measured by the system minus the true value.
[3] Minimum Error – is the minimum value measured by the system minus the true value.

Table 9: A table showing the results obtained for all the systems in Task 2, for the measurement accuracy of angle between the three points (units in degrees). (Ehara, Ebitani and Fujii, 2002)

System	True Value	Mean Value	St. Dev.	Average Absolute	Max. Error +	Max. Error -
VICON	90	89.86	0.15	0.16	0.18	-0.57
Frame-DIAS II	90	90.65	0.48	0.68	1.90	-0.44
Visualeyez	90	90.24	0.48	0.41	2.05	-1.57
PhaseSpace	90	87.87	9.26	6.44	11.62	-23.05
Peak Motus Real Time	90	89.48	0.40	0.56	0.92	-1.97
Peak Motus Video	90	89.73	0.67	0.58	2.13	-2.36
Eagle Digital System	90	90.52	0.13	0.52	0.85	0.18
Pro Reflex	90	89.68	0.24	0.33	0.22	-1.18

Table 10: A table showing the results obtained for the systems that performed Task 3, for the measurement accuracy of the distance between the two virtual points (units in mm). (Ehara, Ebitani and Fujii, 2002)

System	True Value	Mean Value	St. Dev.	Average Absolute	Max. Error +	Max. Error -
VICON	899.5	900.00	0.15	0.50	1.09	0.12
Visualeyez	899.0	888.89	3.28	10.11	-0.84	-26.10
Peak Motus Real Time	900.0	901.83	1.00	1.83	4.90	-0.33
Peak Motus Video	900.0	899.30	2.06	1.68	5.02	-8.57
Eagle Digital System	900.0	899.23	0.35	0.77	0.28	-1.68

The tasks, which were used to measure the accuracy of the systems, were considered rigorous enough since the marker configurations were moved throughout a standard capture volume. This ensures that the cameras' accuracy is consistent throughout the whole volume of capture and not fluctuating all through the entire volume. From the three result tables shown above, the most accurate system was considered as one that generated the closest mean to the true value along with the smallest standard deviation. The standard deviation was noted down and considered in the accuracy of the systems since some systems were reported to generate data larger than and smaller than the true value at random, which coincidently resulted in a mean value that was close to the true value. An accurate system should be one that generates mean values closer to the true value, but in this case precision (or repeatability) of the system had to be taken into consideration, to avoid misinterpreting the data in the case that data was coincidently averaging towards the true value during the testing phase. From an analysis of the results tabulated above, the most accurate system was considered as the Vicon system, since it showed the greatest accuracy (and precision) throughout all the tests. This study was relevant with regards to the project being developed in many aspects. The methodology used in the tasks of the above study will be adapted for this project during the dynamic accuracy-testing phase, and the data measured will be similar to the data represented above. Representing the data in this manner will give enough information to evaluate the accuracy (and to a certain extent the precision) of the systems being investigated.

Three studies with similar methodologies, were performed to evaluate the accuracy of three different systems, namely a five-camera Vicon system (Kidder et al., 1996), a 15-camera Vicon 524 system (Myers et al., 2004) and an eight-camera Optitrack (V100:R2) motion capture system (Kertis, 2012). The study by Kertis will be discussed in this literature review since it is the most recent study, nonetheless results will be compared amongst all three studies. The motion capture system utilised 15.9 mm diameter markers, and captured 3D marker data at a sampling frequency of 100Hz. ARENA motion capture software was used to obtain the 3D marker data. The accuracy for this system was determined both statically and dynamically. For static linear testing, a reference cone resembling foot marker locations with three markers attached to it at predetermined distances was used (Figure 31). The reference cone was placed in five locations in the capture volume along the Cartesian coordinate axes (Figure 32). A 3-second trial was recorded at each of the five designated location.

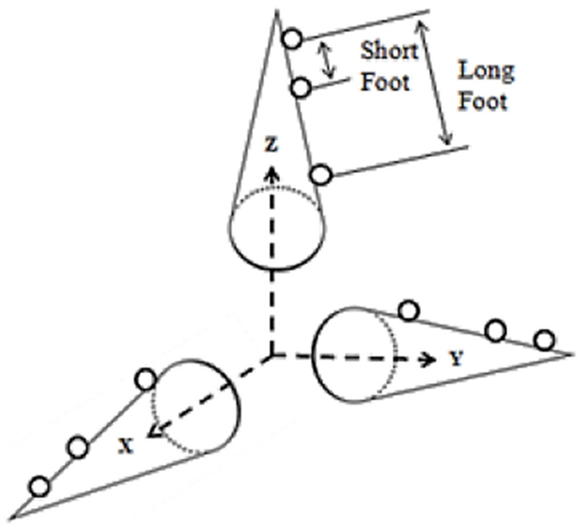


Figure 31: A schematic drawing of the tri-axial calibration cone used for static testing.
The short foot distance represents distance between heel and ankle markers, while long foot distance represents heel to toe markers. (Kertis, 2012)

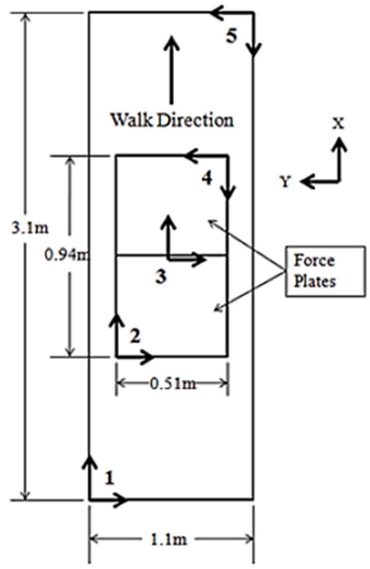


Figure 32: A schematic drawing showing the location of the calibration cone during the stating testing. (Kertis, 2012)

For linear dynamic testing, the reference cone was attached to thigh/leg bar to represent a leg with typical marker placement when using a lower body biomechanical model (Figure 33). The lower extremity setup was then translated across the capture volume of the laboratory at CWS in both the positive and negative walking X-direction. Angular dynamic testing employed the use of a Biodex System III (Figure 34), which is able to rotate through a range of 305° at 90°/second. Data was recorded for a 180° windows for five consecutive trials in all the three planes of motion (XY, XZ and YZ planes), for both clockwise and counter clockwise rotation, of which only a 2-second portion of constant angular velocity was used for each trial.

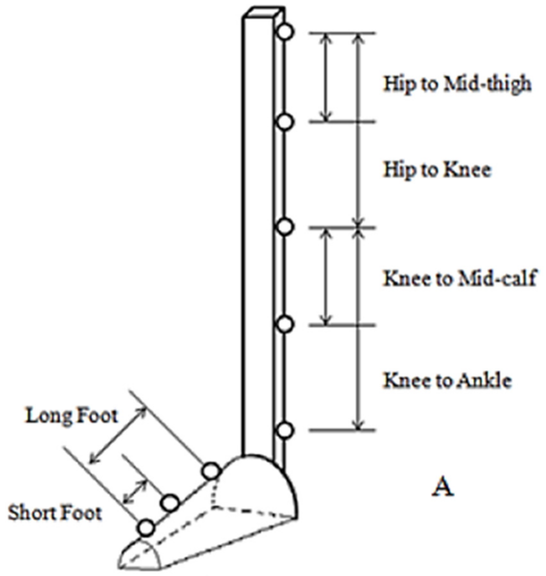


Figure 33: A schematic diagram of the linear dynamic frame used to represent a leg with typical marker placement. (Kertis, 2012)

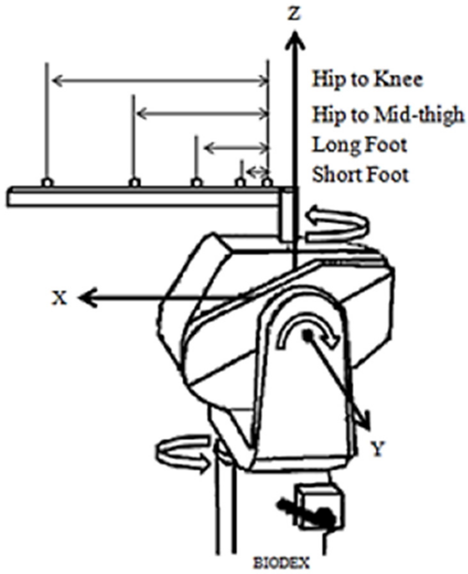


Figure 34: The Biodex setup used during rotational dynamic testing.
The marker locations are representing the ones used on a leg in a lower body biomechanical model. (Kertis, 2012)

The results reported for the static, dynamic linear and rotational testing are shown in Table 11, Table 12 and Table 13 respectively.

Table 11: The accuracies obtained for linear static testing, along with the variation on the data computed at α=0.05. (Kertis, 2012)

Marker Distance	Orientation (Axis)	Accuracy (%)	Variation (mm)
Short Foot (57.5 mm)	X	99.31	0.17 ± 0.15
	Y	99.37	0.31 ± 0.15
	Z	99.64	0.04 ± 0.15
Long Foot (140.6mm)	X	99.76	0.52 ± 0.15
	Y	99.81	0.63 ± 0.15
	Z	99.90	0.35 ± 0.15

Table 12: The accuracies obtained for linear dynamic testing, along with the variation on the data computed at α=0.05. (Kertis, 2012)

Marker Distance	Orientation (Axis)	Accuracy (%)	Variation (mm)
Short Foot (57.5 mm)	Forward (+X)	95.59	0.05 ± 0.21
	Backward (-X)	96.41	0.18 ± 0.20
Long Foot (140.6 mm)	Forward (+X)	96.89	0.25 ± 0.23
	Backward (-X)	97.08	0.37 ± 0.23
Hip to Mid-thigh (205.3 mm)	Forward (+X)	99.46	0.31 ± 0.23
	Backward (-X)	99.54	0.33 ± 0.21
Hip to Knee (417.8 mm)	Forward (+X)	99.70	0.18 ± 0.27
	Backward (-X)	99.77	0.22 ± 0.25
Knee to Mid-Calf (140.6mm)	Forward (+X)	99.37	0.25 ± 0.21
	Backward (-X)	99.27	0.13 ± 0.21
Knee to Ankle (140.6mm)	Forward (+X)	99.61	0.13 ± 0.26
	Backward (-X)	99.60	0.09 ± 0.26

Table 13: The accuracies obtained for angular dynamic testing, along with the variation of the data computed at α =0.05. (Kertis, 2012)

Marker Distance	Orientation (Axis)	Accuracy (%)	Variation (mm)
Short Foot (57.5 mm)	XY	94.82	0.38 ± 0.21
	XZ	98.21	0.27 ± 0.18
	YZ	97.17	0.27 ± 0.20
Long Foot (140.6mm)	XY	97.89	0.18 ± 0.24
	XZ	99.04	0.10 ± 0.19
	YZ	98.43	0.28 ± 0.21
Hip to Mid-Thigh (140.6mm)	XY	98.93	0.29 ± 0.25
	XZ	99.41	0.28 ±0.22
	YZ	99.52	0.24 ± 0.23
Hip to Knee (140.6mm)	XY	99.54	0.27 ± 0.27
	XZ	99.68	0.53 ± 0.29
	YZ	99.42	0.61 ± 0.31

The static calibration testing results obtained by Kertis are comparable to those reported in the studies performed on the alternative Vicon systems by Kidder *et al.* and Myers *et al.* Results for Kidder showed static accuracy and variation in data with a minimum of 99.4% accuracy and 0.6 ± 0.83 mm respectively, at the same significance level used in Kertis' study. Myers reported minimum accuracy of 99.88% and a variation in the data of 0.60 ± 0.14 mm at the same significance level. Linear dynamic trials were also comparative to Myers' study, but Kidder *et al.* did not perform the linear dynamic testing. Myers *et al.* reported that their Vicon system had a minimum accuracy of 99.81% with a variation in the data of 0.53 ± 0.18 mm at a significance level of 0.05. The minimum accuracy obtained by the Optitrack system evaluated by Kertis was of 95.59% and a variation in the data of 0.37 ± 0.23 mm, which shows that the system has room for improvement, although accuracy is still considerably high. Finally, the angular dynamic trials for Kidder *et al.* reported a minimum accuracy of 98.3% and variation in the data of 1.49 ± 0.10 mm at the same significance level, while Myers *et al.* reported a minimum accuracy of 99.81% and a data variation of 2.96 ± 3.53 mm. These values were better than the ones obtained for the Optitrack system, showing that comparatively this motion analysis system has further room for improvement were accuracy is concerned. Kertis commented that although the Optitrack system had lower accuracy values than the Vicon systems he suggested that this could be due to more cameras being used in the Myers study, although this is not applicable for the Kidder *et al.* study since they use less cameras that the Optitrack system, while still obtaining better accuracies. The linear and dynamic testing used in this study would be adopted, along with knowledge obtained from the previously reviewed comparative study, as will be explained in the section 3. Angular dynamic testing would not be performed in this study due to unavailable necessary equipment, and the time-constraints of this project. Finally, the author expect a tendency to show the accuracies during static testing to be close to the accuracies obtained for linear dynamic testing, although a slight decrease in precision due to an increase in variation should be noticed during the dynamic testing.

For the readers' knowledge, it should be noted that the SAMSA (Standard Assessment of Motion System Accuracy) protocol is being devised by the GCMAS (Gait and Clinical Movement Analysis Society) to create a standard for accuracy testing of motion analysis systems worldwide. In the proceedings of the 12th Annual GCMAS Conference, Piazza *et al.* commented on the protocol proposed for quantifying the accuracy of a motion analysis system (Piazza, Chou and Denniston, 2007). The authors of these proceedings commented that a SAMSA protocol is intended to test reflective-marker based motion analysis systems by using a device consisting of a bar fitted with markers rotating at 60RPM by a motor (Figure 35). The protocol is designed to test the system's ability at three specific tasks:

- Task 1: Tracking the motion of moving markers,
- Task 2: Resolving markers using a subset of the cameras available, and
- Task 3: Resolving markers that pass close to each other during the trial.

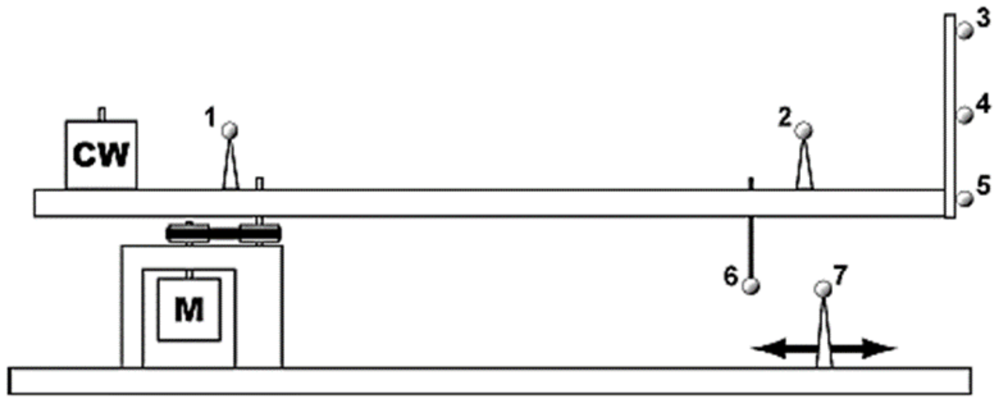


Figure 35: Device used by the SAMSA protocol to assess accuracy of motion analysis systems. (GCMAS SAMSA Device Plans, 2013)

This accuracy standard has the acceptable threshold limits detailed in the protocol, which can be found in the paper by Piazza *et al.* (Piazza, Chou and Denniston, 2007). Although such rigorous testing of a system is encouraged by the author, this was not performed in this project due to the time-constraints, thus a simplified protocol was followed and is detailed in the following section.

3 METHODOLOGY

3.1 HARDWARE AND CONFIGURATION

3.1.1 COMPARISON OF HUMAN GAIT BIOMECHANICS

Motion capture for this part of the study was carried out using the two Vicon camera systems discussed in section 2.1.3.1. The sample rate was kept throughout the study at 100 Hz. The Vicon T-series system was calibrated prior to each session to within 0.5 mm, while the Vicon Bonita system was calibrated to within 0.3mm.

3.1.2 ACCURACY ANALYSIS

The same arrangement of hardware and configuration used in the primary objective of this study was utilised for this secondary objective. The accuracy of each one of the cameras was recorded for this study for documentation purposes. Moreover, a tri-axial milling machine was used to measure the real 3D locations on the markers on the Vicon Calibration Wand, to obtain the inter-marker distance of the markers on the calibration wand. Full description is given in section 3.3.2.

3.2 PARTICIPANTS

Ethical approval was granted by the departmental ethics committee at the Department of Biomedical Engineering at the University of Strathclyde. Seven participants (3 male and 4 female) agreed to volunteer for the primary objective of this study. All the participants were healthy able-bodied adults between the age of 22 and 33. In order to qualify for the study the participants were required to have normal lower limb function, thus being able to walk at a normal daily walking speed (CWS) on their own, without excess physical exertion by the participant. Furthermore, the participant's height had to be between 1.5 and 1.9 meters, they were expected to have a basic knowledge of walking on a treadmill, have a 20/20 vision (with or without correction) and finally having a Body Mass Index (BMI) smaller than 30. The participants who fulfilled the inclusion criteria, were given a Participant Information Sheet (Appendix A) which they were asked to read, understand and keep for future reference. They were subsequently given a Consent Form (Appendix A) which they were asked to read and sign, if they comply with the specifics mentioned in the form, prior to taking part in the study. Table 14, summarises the participants' age, anthropometric data and BMI.

Table 14: A table summarising data of the participants involved in this study. (Data in shown as Mean (S.D.))

N	Age (Years)	Mass (Kg)	Height (mm)	Inter ASIS Distance (mm)	Leg Length (mm)	Knee Width (mm)	Ankle Width (mm)	BMI
7	27.6 (3.8)	72.8 (17.2)	1721.2 (97)	233.9 (40.7)	895.3 (25.5)	97.4 (8.6)	69.7 (6.6)	24.4 (4.5)

3.3 TESTING PROTOCOL

3.3.1 COMPARISON OF HUMAN GAIT BIOMECHANICS

The participants were asked to wear a pair of day-to-day walking shoes and tight fitting clothes, in order to avoid skin marker movement, which is an artefact that can be easily over looked, and may lead to erratic data if it is not addressed properly. Male participants were asked if they were comfortable to stay topless, while female participants were asked to wear a sports bra or any kind of tight clothing that allowed the researcher to access the abdomen for marker placement. If the participants complied with the above-mentioned terms, the anthropometric data was collected. Height (mm) was collected using a stadiometer that was accurate to within 0.25cm; Weight (N) was measured using the Kistler force plates located in the Conventional Human Movement Analysis Laboratory and the mass (kg) was calculated by dividing the weight by the acceleration due to gravity ($g = 9.81$ m/s^2). An anthropometer that was accurate to within 0.1cm was used to measure the inter-ASIS distance (mm), knee width (mm), and ankle width (mm). Finally, a measuring tape that was accurate to within 0.1cm was used to measure the leg length.

Subsequently, sixteen 0.014m diameter retro-reflective markers were attached to the participants (Figure 36) according to the anatomical locations defined by the lower limb PiG model (Table 1). The markers were placed meticulously, in order to avoid marker misplacement. Following marker placement, the participants were asked to stand in the centre of the capture volume while taking the calibration T-pose (Figure 22). Approximately a capture of 200 frames was performed, and the system was then allowed a short time to build the PiG model, which would be later used during the rest of the session.

(a) (b)

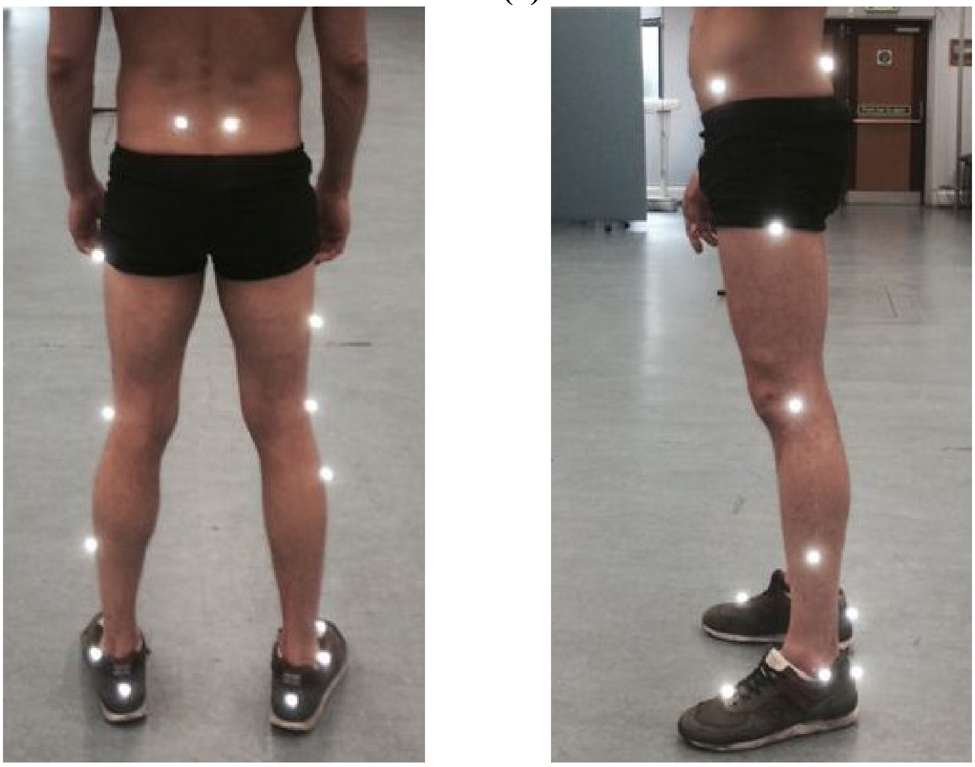


Figure 36: Marker placement according to the lower limb PiG model.
(a) Posterior view and (b) Side view.

The participants were asked to perform a series of provisional walks before any data was captured, to allow the participant to familiarise with the surrounding environment along with the attached markers. During these practice walks, the participants were requested to walk in a normal manner (similar to taking a stroll in the park) at a self-selected walking speed, while looking ahead towards a mark shown on the far side of the laboratory in front of the participant. Practice walks were concluded once the participants felt comfortable with the environment around them. The actual walking trials followed the same procedure taken during the practice walks, but this time data was captured by the motion analysis system. Five successful walking trials were necessary in order to conclude the session. This concluded the over ground walking session in the Conventional Human Movement Analysis Laboratory.

Afterwards, the participants were asked to relocate in the Motek Medical CAREN Extended System to perform the treadmill walks. The participants were required to wear a safety harness that was connected to and overhead weight-bearing structure. This ensured that the participants would not fall from the MOOG platform while the treadmill was moving. Afterwards, the participants were calibrated to the system by taking the same calibration T-pose mentioned earlier and an approximately 200 frame-long capture.

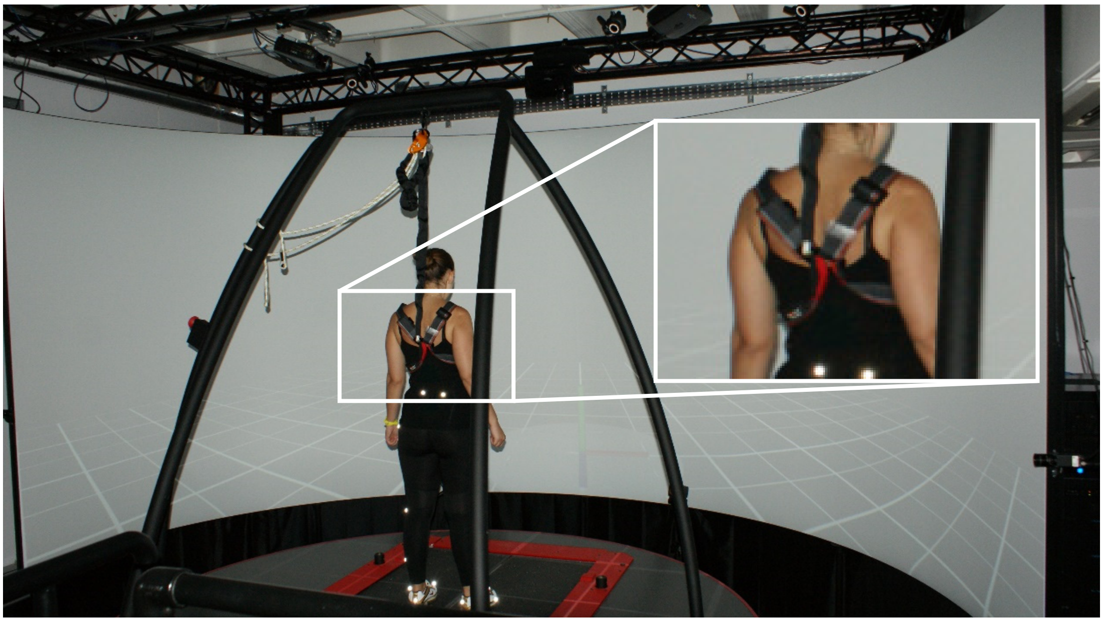


Figure 37: A participant in the CAREN extended system laboratory.
The inset picture focuses on the harness used during treadmill walking.

As mentioned and explained earlier in section 2.1.5, a decision was taken to choose the fixed-speed approach over the self-paced one. In order to select to individualise the treadmill velocity for each participant the following procedure was executed. The participants were asked to stand still on the treadmill, and then a countdown was given, after which the treadmill was accelerated slowly (at 0.2 m/s^2) until the participant sensed that the treadmill velocity approximated his/her CWS, at which the participant signalled the researcher to stop accelerating the treadmill. The treadmill velocity was noted down. Then, the treadmill velocity was increased further to a fast walking speed, and then decelerated slowly (at 0.2 m/s^2) until the participant sensed that the treadmill velocity again approximated his/her CWS. This treadmill velocity was noted down as well, and the mean velocity of the two values was calculated and used as the treadmill velocity throughout the rest of the treadmill walking environments.

The participants were then informed that the velocity chosen for them would be applied throughout the rest of the treadmill walking environments. The first treadmill-walking environment consisted of a black image in front of the participant. The participants were told to walk on the treadmill in a normal manner until they would be signalled that the treadmill velocity would be decelerated and stopped. While the participants were walking, their motion was captured for approximately 1500 frames (equivalent to 15 seconds). The motion capture was executed in this manner to avoid any lurking variable in the data due to the participants' awareness that they are being recorded, thus resulting in gait deviations. Following the data capture, the participant was allowed time to rest to avoid gait deviations due to fatigue. During this time, the researcher asked the participant to comment on any sensations that were felt in this treadmill-walking environment, in comparison with the over ground walking environment used earlier.

The next environment consisted of a grass path (Figure 38) which moved at the same speed of the treadmill, thus simulating an environment that responded to the walking speed of the participant. The motion capture data was obtained in the same way as it was done in the previous treadmill environment. The participant was allowed to rest, and was asked to comment on any sensations experienced when walking in this environment, in comparison to the other two previous environments. Finally, the participant was prepared for the last environment that consisted of the same grass pathway, but this time the lower limb PiG markers were shown over-and-above the grass pathway, and their location moved in real-time mirroring the participants' lower limb motion. The motion capture data was obtained using the same approach taken for the two previous environments. Following the motion capture, the participants were asked to comment on the last treadmill-walking environment in comparison to the previous three environments. This concluded the testing protocol for the comparison of human gait biomechanics.

Figure 38: The simulated virtually realistic grass pathway that was shown infront of the participants during the treadmill walk with optic flow.

3.3.2 ACCURACY ANALYSIS

The accuracy of the optoelectronic motion analysis systems were determined statically and dynamically. The accuracy was determined by measuring the distance between all the reflective markers on the Vicon calibration wand (Figure 21) using the motion analysis systems, and comparing the measured data with the actual inter-marker distances.

The actual inter-marker distances were measured on a milling machine in the Machining Laboratory found in the Department Of Biomedical Engineering in the University of Strathclyde. The actual 3D locations of the markers were obtained by fastening a 2mm drill bit on the jig of the milling machine, and placing the wand securely along the long axis of the milling machine. A small piece of moist tissue paper was attached to one of the markers, and then the drill bit was slowly brought close to one of the edges of the marker until the drill bit touched the moist tissue paper. The milling machine's origin was set at this location, and therefore all subsequent coordinates were considered to be relative to this origin. Consideration was given to touch the marker at a tangent to ensure that the widest part of it is being measured. The location of the centre of the drill bit was shown on the milling machine's display and noted down. This was repeated four times on each of the wand markers (Figure 39), every time noting down the coordinates displayed. Finally, the coordinates of the top-end of the markers, was obtained using the same approach. This approach generated all the necessary data to measure the three coordinates (X, Y and Z) of the centroid of the markers, which were considered the actual coordinates of the wand markers (in relation to the aforementioned origin) in this analysis. The real inter-marker distances were later calculated, in order to compare these values with the ones obtained during the static and dynamic trials.

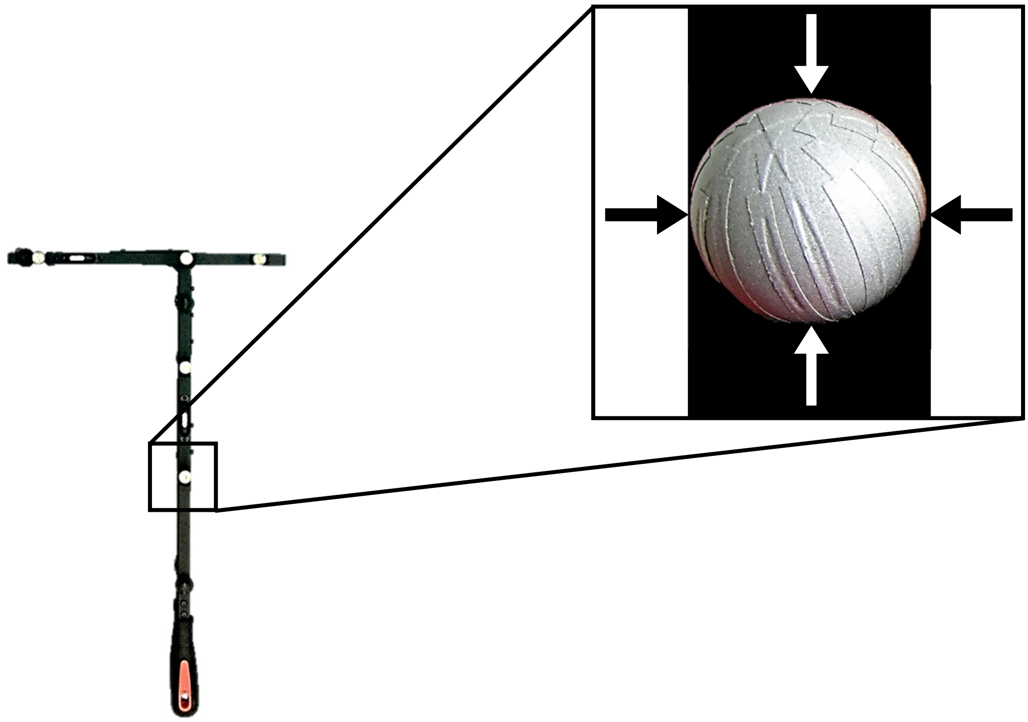


Figure 39: A diagram showing the plan view of a magnified reflective marker (inset) from the calibration wand. The arrows show the location where the drill bit made contact with the marker.

For static testing in the Conventional Human Movement Analysis Laboratory, the calibration wand was placed at 7 positions along the length of the walkway (Figure 40). For each of the positions, the wand was placed at ankle, knee and hip height, using the setups shown in Figure 40. At each of the 21 locations, the coordinate data for the markers was measured by taking a motion capture of approximately 100 frames.

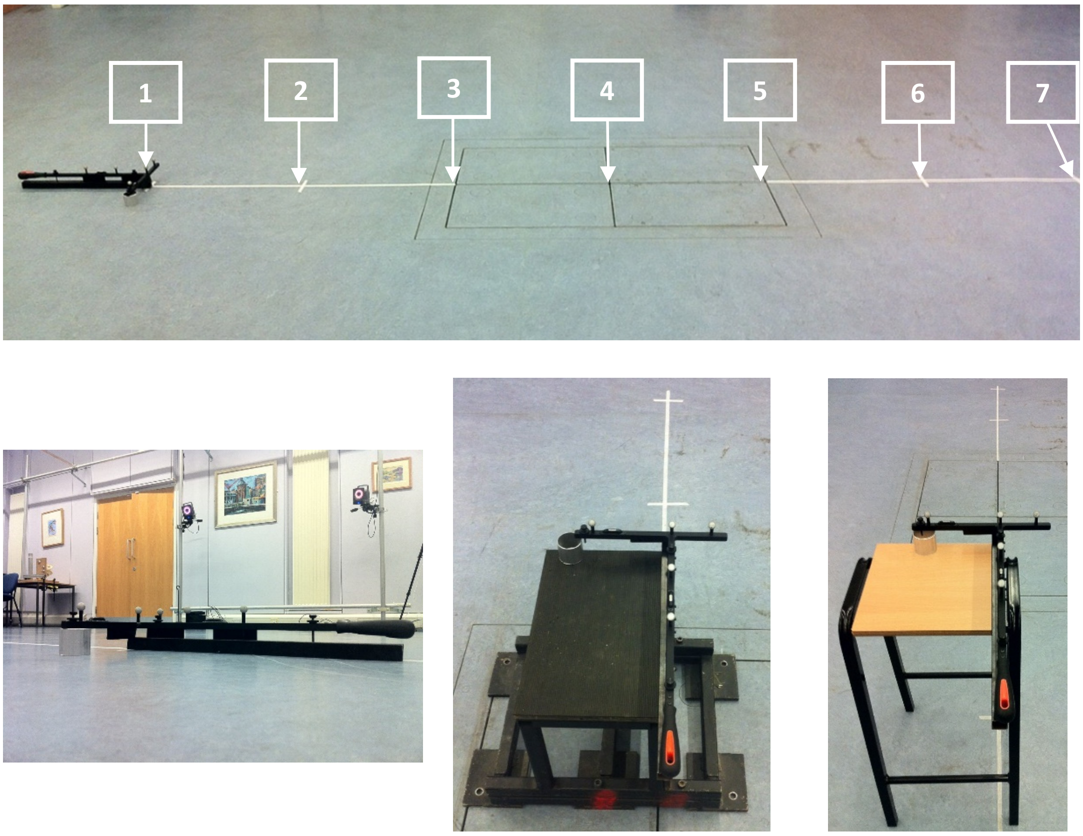


Figure 40: (a) shows the seven positions where the calibration wand was placed along the walkway and the setup used to place the wand at (b) ankle height, (c) knee height, and (d) hip height.

For static testing in the CAREN extended system Laboratory the same approach was taken. The calibration wand was placed at 3 positions along the length of the treadmill (Figure 41). For each of the positions, the wand was placed at ankle, knee and hip height, using the same setups shown in Figure 40. At each of the nine locations, the coordinate data for the markers was measured by taking a motion capture of approximately 100 frames.

The dynamic testing in the Conventional Human Movement Analysis Laboratory and the CAREN extended system Laboratory was performed by waving the calibration wand around the capture volume used during gait motion capture. The wand was waved at three distinct velocities, namely slow, medium and fast. The researcher paid attention to the velocities applied, in order to make sure that they were distinct from each other. Finally, the data obtained from the static and dynamic testing in both of the laboratories was saved for subsequent data processing.

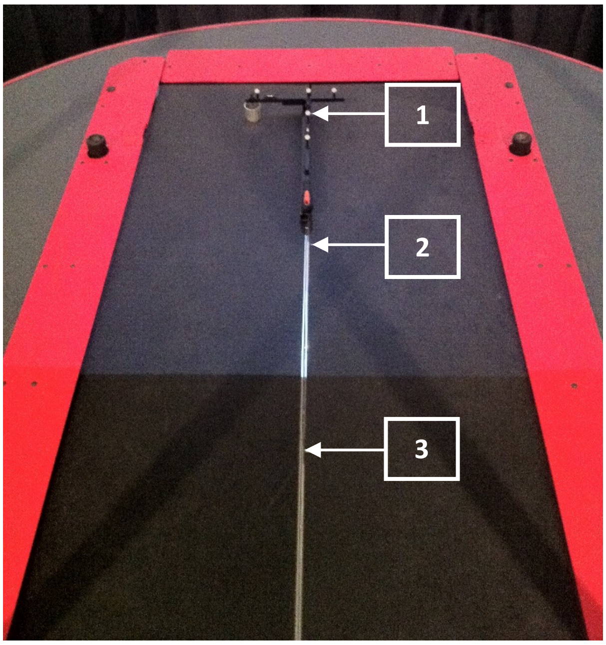
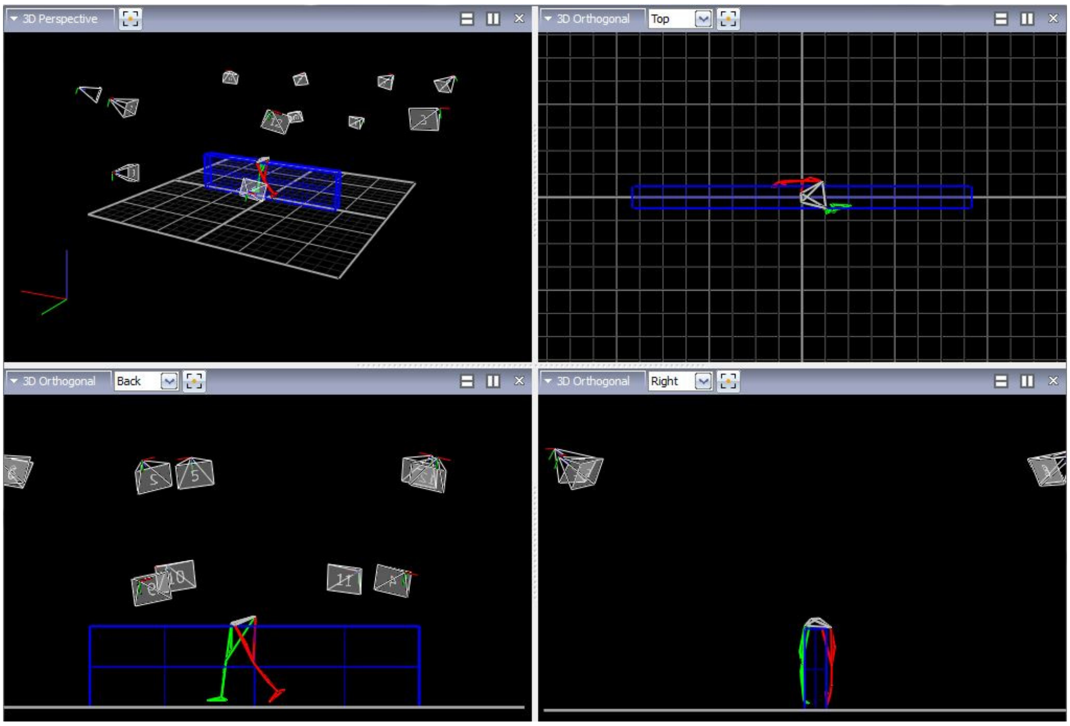


Figure 41: A diagram showing the three positions where the calibration wand was placed along the walkway.

In this study, the accuracy analysis focused on the capture volumes used during gait testing. This is clearly shown in Figure 42, which illustrates the relationship between the volume that was measured for accuracy during the static testing and the volume utilised by the participants during a motion capture. The dynamic testing was not related to the volume covered (although it was still performed within the capture volume), but the idea behind it was to analyse the accuracy of a reflective marker moving at different velocities.

(a)



(b)

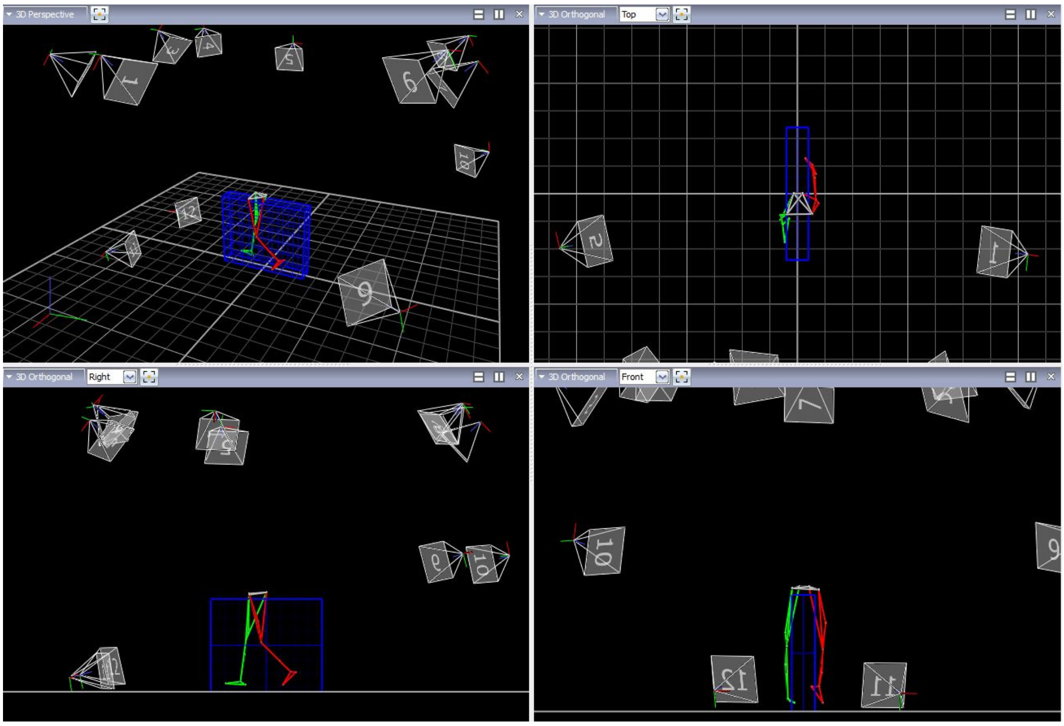


Figure 42: An illustration of the relationship between the volume that was measured for accuracy, shown in blue, during the static testing and the volume utilised by the participants during a motion capture (shown by the lower limb PiG model). (a) The Conventional Human Movement Analysis Laboratory and (b) CAREN extended system Laboratory. (Murphy, 2014)

3.4 DATA PROCESSING AND ANALYSIS

3.4.1 COMPARISON OF HUMAN GAIT BIOMECHANICS

The Vicon data obtained during the trials was initially cropped using Vicon Nexus software (version 1.8.6) to include the region of interest. The data was subsequently checked for any gaps in the marker trajectories, which happen when a marker is only seen by one camera, thus the 3D coordinates of the marker could not be defined by the system. The missing marker trajectories were then filled using mathematical algorithms installed within the software. Finally, the system worked out the foot strikes and the foot offs automatically, and the data was exported as an ASCII file, which was later opened on MS Excel (Office 2013) for further data processing.

In MS Excel, the data was then categorised and used to measure the TSPs and locate the required joint kinematics. For the TSPs, the walking speed was worked out for the Conventional Human Movement Analysis Laboratory only, since the walking speed was constant in the CAREN Extended System Laboratory. The cadence and the stride length were calculated using the measured location of the heel markers at successive foot strikes. The maximum joint kinematics of the hip, knee and ankle were also extracted from the data using functions embedded within Excel to locate the maximum values of each gait parameter.

The data selected from Excel was exported to the IBM SPSS Statistics software (version 22), for statistical analysis. The data was processed one gait parameter at a time in the following manner. Firstly, the data was checked for normality using a Kolmogorov-Smirnov test of normality. This was performed in order to determine if a parametric or non-parametric test would be used to compare the gait parameter between the four different walking conditions. If the data were normally distributed a parametric test would have been performed using a repeated measures design test with a Bonferroni correction, for pair wise comparisons. Since no data obtained in this study was normally distributed, thus all the data was analysed using a non-parametric approach. First, a Friedman’s ANOVA test was performed to check for any statistically significant change between the walking conditions. A Wilcoxon Signed Ranks test with a Bonferroni correction was utilised to follow up on the Friedman’s ANOVA. A Bonferroni correction (Equation 4) increases the fidelity in the statistical analysis performed by applying a correction over-and-above the significance limit of the Wilcoxon test, as shown below:

Equation 4: The Bonferroni correction.

$$Bonferroni\ Correction = \frac{\propto}{number\ of\ comparisons}$$

The Wilcoxon Signed Rank test was utilised to report any statistical difference between the six pairwise comparisons performed.

3.4.2 ACCURACY ANALYSIS

The coordinate data measured using the milling machine was inputted on Excel, and the actual inter-marker distance were calculated. This consisted of working out the 3D coordinates of the centroid of each marker, relative to the aforementioned origin. Pythagoras Theorem for 3-dimensional coordinates was then applied to measure the real distance between the markers.

The Vicon data obtained during the accuracy testing was initially cropped using Vicon Nexus software. All the captures were cropped down to 100 frames. Any gaps found in the marker trajectories of the dynamic trials were filled, and subsequently the data was exported as an ASCII file. The ASCII data was imported in excel for further data processing. The inter-marker distances were calculated using the 3D coordinates measured by the motion analysis systems, along with the above-mentioned Pythagoras Theorem for 3D coordinates. For each individual static and dynamic capture, the mean and standard deviation of the data was calculated for each inter-marker distance (10 data points). The *error* in the measured values was calculated using Equation 5, while the *accuracy* was calculated using Equation 6.

Equation 5: Equation used to calculate Measurement Error

$$Error = \left(\frac{measured\ distance - real\ distance}{real\ distance}\right) \times 100$$

Equation 6: Equation used to measure accuracy.

$$Accuracy = 100 - Measurement\ Error$$

Finally, the mean accuracy of both laboratories was calculated by taking the mean of the accuracies obtained for each static trial.

4 RESULTS

4.1 COMPARISON OF HUMAN GAIT BIOMECHANICS

4.1.1 DESCRIPTIVE STATISTICS

Descriptive statistics for the data of each gait parameter measured are reported below. The full post-processed data set, from which the following results were calculated, could be found in Appendix B. The results are tabulated showing the mean, standard deviation, minimum and maximum values of the data set, for each walking condition. Moreover, the data was visualised on a bar graph, showing the mean and standard deviation values for the data set. It should be noted that the following acronyms are used in the results.

WC1	Over ground walking
WC2	Treadmill walking
WC3	Treadmill walking with the grass pathway
WC4	Treadmill walking with the grass pathway and the interactive lower limb PiG markers.

Walking Speed (m/s)

	N	Mean	Std. Deviation	Minimum	Maximum
WC1	35	1.4440	.13848	1.15	1.72
WC2	35	1.0686	.10282	.94	1.22
WC3	35	1.0686	.10282	.94	1.22
WC4	35	1.0686	.10282	.94	1.22

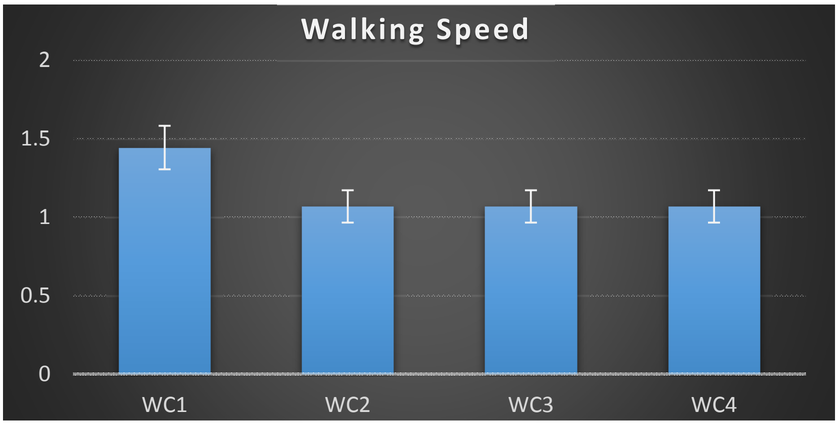


Table 15: Descriptive statistics for walking speed during the 4 different walking conditions.

Cadence (Steps/min)

	N	Mean	Std. Deviation	Minimum	Maximum
WC1	35	115.9909	5.99164	107.91	129.73
WC2	35	108.9114	2.57816	104.27	112.97
WC3	35	109.0129	2.34939	105.94	112.97
WC4	35	107.5743	2.76662	102.33	110.55

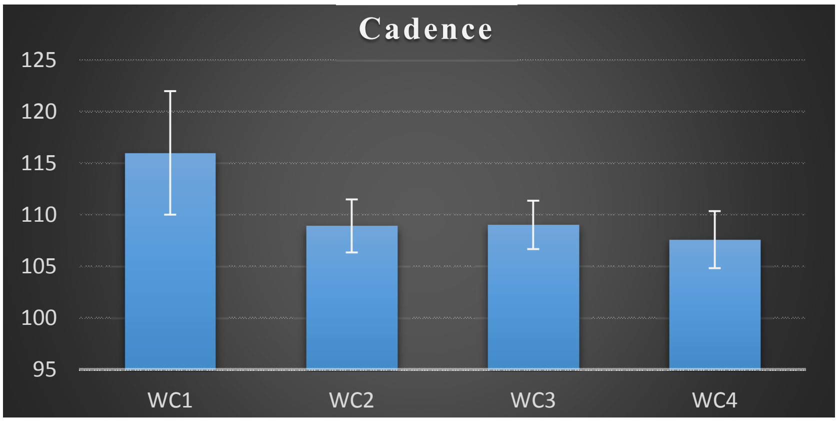


Table 16: Descriptive statistics for cadence during the 4 different walking conditions.

Stride Length (m)

	N	Mean	Std. Deviation	Minimum	Maximum
WC1	70	1.4933	.13927	1.22	1.74
WC2	70	1.2394	.08985	1.11	1.42
WC3	70	1.2467	.10121	1.07	1.46
WC4	70	1.2547	.10304	1.05	1.49

Table 17: Descriptive statistics for stride length during the 4 different walking conditions.

Maximum Hip Flexion Angles (^{0})

	N	Mean	Std. Deviation	Minimum	Maximum
WC1	140	36.2562	4.62599	25.64	43.12
WC2	140	33.9324	6.40450	21.97	44.16
WC3	140	33.3834	6.18354	20.90	43.51
WC4	140	33.6834	6.36351	20.92	44.95

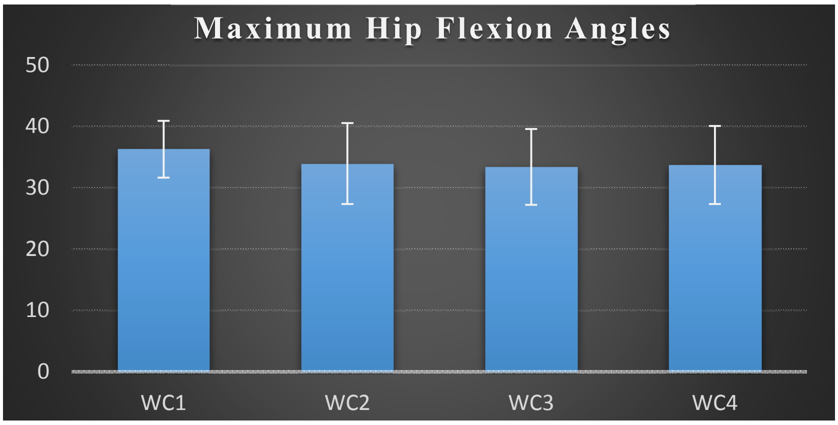


Table 18: Descriptive statistics for the maximum hip flexion angles during the 4 different walking conditions.

Maximum Knee Flexion Angles (^{0})

	N	Mean	Std. Deviation	Minimum	Maximum
WC1	140	63.4209	5.88067	50.73	77.81
WC2	140	63.7897	4.81412	54.64	73.24
WC3	140	64.2724	4.65065	52.53	72.61
WC4	140	64.2400	4.75417	52.26	72.80

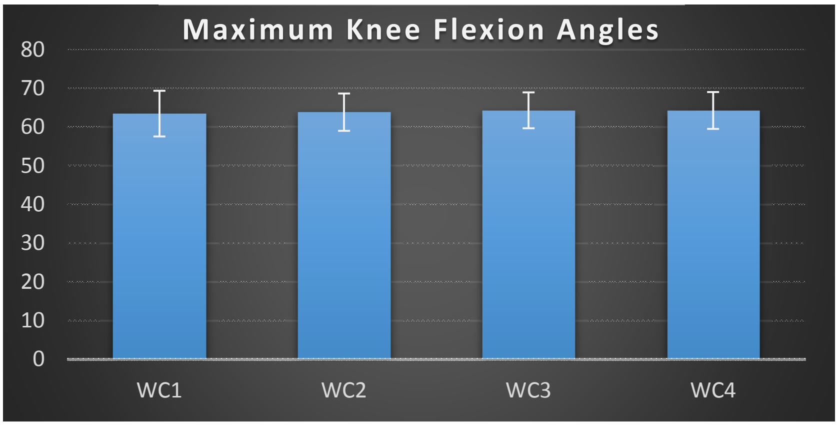


Table 19: Descriptive statistics for maximum knee flexion angles during the 4 different walking conditions.

Maximum Ankle Dorsiflexion Angles (^{o})

	N	Mean	Std. Deviation	Minimum	Maximum
WC1	140	21.1836	7.25461	10.11	33.65
WC2	140	14.9529	2.52563	8.91	19.39
WC3	140	15.6349	2.47727	8.73	20.84
WC4	140	15.6590	2.16065	9.70	20.89

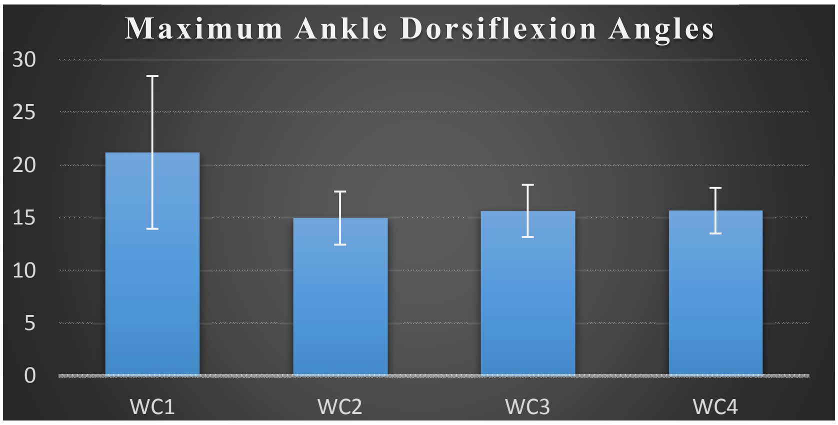
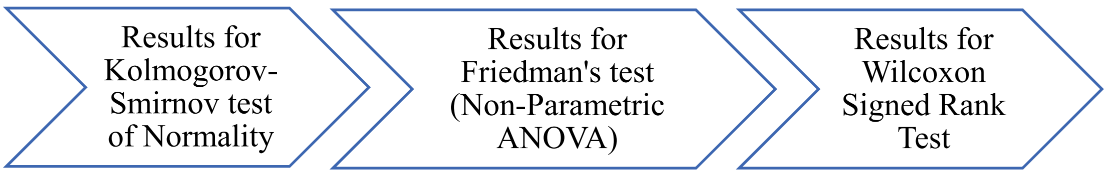


Table 20: Descriptive statistics for maximum ankle dorsiflexion angles during the 4 different walking conditions.

4.1.2 ANALYTICAL STATISTICS

The statistical analysis performed on SPSS gave a sequence of results for each gait parameter investigated. The results will follow the following order for each gait parameter:



- *Walking Speed:*

The Kolmogorov-Smirnov test showed that the walking speeds during WC1, $D(35) = 0.106$, $p > 0.05$ were not significantly different from a normal distribution, but those during WC2, WC3, WC4, $D(7) = 0.006$, $p < 0.05$, were significantly different. Since the data was predominantly not normal, a non-parametric approach was performed.

A Friedman's test (non-parametric ANOVA) reported that the walking speeds were significantly different amid the 4 conditions, $\chi2\ (3) = 0.0$, $p < 0.05$.

Table 21: Results obtained for walking speed using the Wilcoxon Signed Ranks test. The values followed by a * represent statistically significant differences ($p < 0.0083$).

	WC1	WC2	WC3	WC4
WC1		0.000*	0.000*	0.000*
WC2			1.000	1.000
WC3				1.000
WC4				

A Wilcoxon Signed Ranks Test was used to follow up on the Friedman's ANOVA. A Bonferroni correction was applied over-and-above the significance limit of the Wilcoxon Test, and therefore all effects are reported at a 0.0083 level of significance. The walking speeds during WC1 were significantly different from all other WCs. Furthermore, WC2, WC3 and WC4, were not statistically different from any of the other WCs (except for WC1). This occurred since the same walking speed was used for WC2, WC3 and WC4, while the walking speeds for WC1 were self-selected by the participants (CWS).

- *Cadence:*

The Kolmogorov-Smirnov test of normality showed that the cadences during WC1, WC2, WC4, $D(35) = 0.0$, $p < 0.05$ and WC3, $D(35) = 0.012$, $p < 0.05$ were significantly different from a normal distribution. Since the data was not normally distributed, a non-parametric approach was performed.

The Friedman's ANOVA reported that the cadences were significantly different between the 4 WCs, $\chi2\ (3) = 0.0$, $p < 0.05$.

Table 22: Results obtained for cadence using the Wilcoxon Signed Ranks test. The values followed by a * represent statistically significant differences ($p < 0.0083$).

	WC1	WC2	WC3	WC4
WC1		0.000*	0.000*	0.000*
WC2			0.622	0.002*
WC3				0.027*
WC4				

A Wilcoxon Signed Ranks Test with a Bonferroni correction was used to follow up on the Friedman's ANOVA. This statistical analysis showed that the cadences during WC1 were significantly different from all other WCs ($p < 0.0083$). Furthermore, the cadences between WC2 and WC4 were also statistically different ($p < 0.0083$). The cadences did not display a significant change between WC3 and WC2, and WC4 ($p > 0.0083$) respectively.

- *Stride Length:*

The Kolmogorov-Smirnov test of normality showed that the stride lengths for WC1 and WC4, $D(70) = 0.0$, $p < 0.05$, WC3, $D(70) = 0.001$, $p < 0.05$ and WC2 $D(70) = 0.01$, $p < 0.05$, were all significantly different from a normal distribution. Thus, a non-parametric approach was performed.

The Friedman's ANOVA reported that the stride lengths were significantly different between all the 4 conditions, $\chi 2\ (3) = 0.0$, $p < 0.05$.

Table 23: Results obtained for stride length using the Wilcoxon Signed Ranks test. The values followed by a * represent statistically significant differences ($p < 0.0083$).

	WC1	WC2	WC3	WC4
WC1		0.000*	0.000*	0.000*
WC2			0.429	0.003*
WC3				0.242*
WC4				

A Wilcoxon Signed Ranks Test with a Bonferroni correction was used to follow up on the Friedman's ANOVA. This statistical analysis showed that the stride lengths during WC1 were significantly different from all other WCs ($p < 0.0083$). Furthermore, the stride lengths between WC2 and WC4 were also statistically different ($p < 0.0083$). The stride lengths did not show a significant change between WC3 and WC2, and WC4 ($p > 0.0083$) respectively.

- *Maximum Hip Flexion Angles:*

The Kolmogorov-Smirnov test of normality showed that the maximum hip flexion angles for WC1 and WC3, $D(140) = 0.0$, WC2, $D(140) = 0.032$, and WC4, $D(140) = 0.012$, $p < 0.05$ were all significantly different from a normal distribution. Since the data was not normally distributed, a non-parametric approach was performed.

The Friedman's ANOVA reported that the maximum hip flexion angles were significantly different amid all the 4 conditions, $\chi2\ (3) = 0.0$, $p < 0.05$.

Table 24: Results obtained for maximum hip flexion angles using the Wilcoxon Signed Ranks test. The values followed by a * represent statistically significant differences ($p < 0.0083$).

	WC1	WC2	WC3	WC4
WC1		0.001*	0.000*	0.001*
WC2			0.000*	0.110
WC3				0.170
WC4				

A Wilcoxon Signed Ranks Test with a Bonferroni correction was used to follow up on the Friedman's ANOVA. This statistical analysis showed that the maximum hip flexion angles during WC1 were significantly different from all other WCs ($p < 0.0083$). Furthermore, the hip flexion angles in the pairwise comparison between WC3 and WC2 were also statistically different ($p < 0.0083$). Finally, the hip flexion angles did not show a significant change in the comparison between WC4 and WC2, and WC3 ($p > 0.0083$) respectively.

- *Maximum Knee Flexion Angles:*

The Kolmogorov-Smirnov test of normality indicated that the maximum knee flexion angles measured for WC2, $D(140) = 0.001$, WC3, $D(140) = 0.0$ and WC4, $D(140) = 0.015$, $p < 0.05$ were not normally distributed, but the angles during WC1, $D(140) = 0.2$, $p > 0.05$ followed a Gaussian (normal) distribution. Since the data was predominantly not normal, a non-parametric approach was performed.

The Friedman's ANOVA reported that the maximum knee flexion angles were significantly different between the 4 conditions, $\chi2\ (3) = 0.008$, $p < 0.05$.

Table 25: Results obtained for maximum knee flexion angles using the Wilcoxon Signed Ranks test. The values followed by a * represent statistically significant differences ($p < 0.0083$).

	WC1	WC2	WC3	WC4
WC1		0.326	0.055	0.110
WC2			0.001*	0.002*
WC3				0.913
WC4				

A Wilcoxon Signed Ranks Test using a Bonferroni correction reported that the maximum knee flexion angles during WC1 were not significantly different from all other WCs ($p > 0.0083$). The knee flexion angles in the pairwise comparison between WC4 and WC3 also showed that no statistical difference exists between them ($p > 0.0083$). Finally, the knee flexion angles in the comparisons between WC2 and WC3, and WC4 ($p < 0.0083$) respectively, showed a statistically significant change.

- *Maximum Ankle Dorsiflexion angle:*

The Kolmogorov-Smirnov test of normality showed that the maximum ankle dorsiflexion angles collected during WC1 D(140) = 0.0, WC3 D(140) = 0.003 and WC4 D(140) = 0.033, $p < 0.05$ were significantly different from a normal distribution, while the ones collected during WC2 D(140) = 0.062, $p > 0.05$, were not. Since the data was predominantly not normal, a non-parametric approach was performed.

The Friedman's ANOVA reported that the maximum ankle dorsiflexion angles were significantly different amid all the 4 conditions, $\chi2$ (3) = 0.0, $p < 0.05$.

Table 26: Results obtained for maximum ankle dorsiflexion angles using the Wilcoxon Signed Ranks test. The values followed by a * represent statistically significant differences ($p < 0.0083$).

	WC1	WC2	WC3	WC4
WC1		0.000*	0.000*	0.000*
WC2			0.000*	0.001*
WC3				0.703
WC4				

A Wilcoxon Signed Ranks Test, with a Bonferroni correction reported that the maximum ankle dorsiflexion angles during WC1 were significantly different from all other WCs ($p < 0.0083$). Furthermore, the dorsiflexion angles in the pairwise comparisons between WC2 and WC3, and WC4 were also statistically different ($p < 0.0083$). Finally, the ankle dorsiflexion angles did not show a significant change between WC4 and WC3 ($p > 0.0083$).

4.2 ACCURACY ANALYSIS

The full post-processed data set (21000 measured data points), from which the following results were calculated, could be found in Appendix C. The inter-marker distance that were measured using the data collected from the two motion analysis systems was compared to the real inter-marker distances measured using the milling machine. Ten inter-marker distances were measured from the calibration wand, having a range of 72mm to 288mm. The real distances measured are shown in Table 27

Table 27: The actual calibration wand inter-marker distances

Inter-Marker Distance		
A	B	160.30
A	C	240.18
A	D	199.51
A	E	288.16
B	C	79.89
B	D	119.77
B	E	239.99
C	D	143.78
C	E	252.51
D	E	120.22

4.2.1 CONVENTIONAL HUMAN MOTION ANALYSIS LABORATORY

Before the testing for the accuracy analysis was initiated, the cameras were calibrated. The errors recorded for each optoelectronic camera by the motion analysis system, are tabulated in Table 28 for documentation purposes.

Table 28: Camera errors recorded by the system following calibration.

Camera Number	Measured Error (mm)
1	0.277
2	0.196
3	0.288
4	0.302
5	0.203
6	0.332
7	0.171
8	0.340
9	0.373
10	0.371
11	0.232
12	0.234

The results for the static accuracy testing are summarised in Table 29.

Table 29: Accuracy testing results

Location	Average Accuracy	Average Error	St. Dev.	Minimum Accuracy	Maximum Accuracy
Ankle Height	99.87	0.13	0.01	99.79	99.90
Knee Height	99.88	0.12	0.01	99.80	99.90
Hip Height	99.88	0.12	0.01	99.81	99.90
Position 1	99.80	0.20	0.01	99.38	99.95
Position 2	99.88	0.12	0.00	99.67	99.99
Position 3	99.89	0.11	0.00	99.74	100
Position 4	99.90	0.10	0.00	99.80	100
Position 5	99.90	0.10	0.00	99.80	100
Position 6	99.89	0.11	0.00	99.75	100
Position 7	99.90	0.10	0.02	99.61	100
Capture Volume	**99.88**	**0.12**	**0.01**	**99.79**	**99.90**

With respect to the dynamic accuracy testing, the wand speeds used for the slow, medium and fast trials are summarised in Table 30.

Table 30: Descriptive Statistics for the speeds used during dynamic accuracy testing.

Wand Speed	Mean	St. Dev.	Minimum	Maximum
Slow	1.21	0.31	0.55	2.24
Medium	1.69	0.34	0.59	2.74
Fast	1.80	0.44	0.33	2.89

The intra-laboratory velocities used during the dynamic accuracy testing were analysed using IBM SPSS Statistical software, to check if the velocities used were statistically different from each other. The statistical analysis confirmed that the velocities used within this laboratory were statistically different. The results for the dynamic accuracy testing are summarised in Table 31.

Table 31: Dynamic accuracy testing results

Wand Speed	Accuracy	Error	St. Dev.
Slow	99.90	0.1	0.43
Medium	99.90	0.1	0.32
Fast	99.90	0.1	0.38

Conclusively, the average error of the motion capture system within the Conventional Human Movement Analysis Laboratory is **0.12 ± 0.01 mm** or else **99.88%** accurate.

4.2.2 MOTEK MEDICAL CAREN EXTENDED SYSTEM LABORATORY

Before the testing for the accuracy analysis was initiated, the cameras were calibrated. The errors recorded for each optoelectronic camera by the motion analysis system, are tabulated in Table 32 for documentation purposes.

Table 32: Camera errors recorded by the system following calibration.

Camera Number	Measured Error (mm)
1	0.069
2	0.094
3	0.076
4	0.084
5	0.092
6	0.092
7	0.077
8	0.073
9	0.102
10	0.100
11	0.079
12	0.900

The results for the static accuracy testing are summarised in Table 33.

Table 33: Accuracy testing results

Location	Average Accuracy	Average Error	St. Dev.	Minimum Accuracy	Maximum Accuracy
Ankle Height	99.61	0.39	0.03	99.50	99.73
Knee Height	99.70	0.30	0.04	99.62	99.77
Hip Height	99.75	0.25	0.03	99.68	99.79
Position 1	99.74	0.26	0.01	99.12	100
Position 2	99.72	0.28	0.02	99.44	100
Position 3	99.60	0.40	0.03	99.22	99.93
Capture Volume	**99.68**	**0.32**	**0.03**	**99.50**	**99.79**

With respect to the dynamic accuracy testing, the wand speeds used for the slow, medium and fast trials are summarised in Table 34.

Table 34: Descriptive Statistics for the speeds used during dynamic accuracy testing.

Wand Speed	Mean	St. Dev.	Minimum	Maximum
Slow	0.39	0.08	0.19	0.59
Medium	1.17	0.32	0.08	1.90
Fast	1.69	0.38	0.82	2.79

The intra-laboratory velocities used during the dynamic accuracy testing were analysed using IBM SPSS Statistical software, to check if the velocities used were statistically different from each other. The statistical analysis confirmed that the velocities used within this laboratory were statistically different. The results for the dynamic accuracy testing are summarised in Table 35.

Table 35: Dynamic accuracy testing results

Wand Speed	Accuracy	Error	St. Dev.
Slow	99.69	0.31	0.39
Medium	99.69	0.31	0.34
Fast	99.70	0.30	0.43

Conclusively, the average error of the motion capture system within the Motek Medical CAREN Extended system Laboratory is **0.32 ± 0.03 mm** or else **99.68%** accurate.

5 DISCUSSION

The primary aim of this thesis was to compare and evaluate the variation in the gait parameters following walking trials in four different optic flow environments. The secondary aim investigated the accuracy of the motion analysis systems used in the primary objective to provide an insight into the inter-system and intra-system accuracy. In this section an in-depth analysis of the results obtained is performed which relates the measured data with those reported in literature and theory. Based on the limitations of this study any alternative explanations for the reported results will be explained.

This study had a number of limitations that showed a direct and repercussive effect on the results obtained. The fixed treadmill speed used in this study was a major limitation, which is justified since it was the most reliable way of approximating the participants' walking speeds, given that self-selected treadmill speeds create sensations of cautiousness and instability on the participants using it. Due to time limitations on the time-window of the laboratories availability during this study, the participants might have not been given a proper acclimatisation period, although the participants were required to have a basic knowledge of using the treadmill to minimise any lurking variables. Although, the optic flow used in WC3 during this study was presumed to give adequate visual feedback to the participants (Figure 38), it was noticed that this imposed a limitation since the participants were disoriented when walking with this optic flow pattern. Finally, the number of participants used in this study was limited due to the period available for testing during this project

5.1 COMPARISON OF HUMAN GAIT BIOMECHANICS

The primary objective of this study was to compare and evaluate the variation in the gait biomechanics of the four walking conditions. The statistical results obtained during the testing phase of this project (Table 36), reported that the WC1 gait biomechanics showed a significant difference from the other three treadmill environments, except for the maximum knee flexion angles, which showed similarity. The TSPs obtained for WC3, showed similarity with those obtained for WC2 and WC4, but variations were noted in the kinematics of this same relationship. Finally, the entire gait biomechanics evaluated in this study showed similarities between WC3 and WC4. These results show patterns of similarities in the variation of the gait parameters for the different walking conditions, although there still are certain gait parameters that do not follow these relationships. Each investigated parameter will now be discussed individually in relation to the cited work and theory, and any alternative explanations that have been perceived.

Table 36: Summarised Wilcoxon Signed Ranks Test results. The Red boxes indicate significantly different results, the Green boxes indicated statistically similar results, while the yellow box indicates a significant value which was considered to be statistically similar due to the Bonferroni correction.

		Wilcoxon Signed Ranks Test Results P-Value					
		WC1-WC2	WC1-WC3	WC1-WC4	WC2-WC3	WC2-WC4	WC3-WC4
Gait Parameters	Walking Speed	0.000	0.000	0.000	1.000	1.000	1.000
	Cadence	0.000	0.000	0.000	0.622	0.002	0.027
	Stride Length	0.000	0.000	0.000	0.429	0.003	0.242
	Hip	0.001	0.000	0.001	0.000	0.110	0.170
	Knee	0.326	0.055	0.110	0.001	0.002	0.913
	Ankle	0.000	0.000	0.000	0.000	0.001	0.703

Initial analysis of the walking speed data revealed that the walking speeds during over ground walking fell within the mean ranges stated by Murray *et al.* (1964) and Murray *et al.* (1970) (Section 2.1.2.1). The CWSs measured during over ground walking were faster and statistically different from the walking speed used during treadmill walking (Table 15), which agreed with the study by Riley *et al.* (Table 3). The walking speeds during the three treadmill walking conditions were identical since the same treadmill velocity was chosen. Apart from the fact that this approach eliminates the prospect of comparing walking speeds amid the three treadmill conditions, the researcher came to the deduction that this approach might have further implications on the other parameters, because of the way it was conducted. This is because some of the participants commented that although the treadmill speed was based on their own judgement of a comfortable walking speed, they still felt that during treadmill walking with optic flow (WC3 and WC4) their walking speed was faster than the 'imposed' treadmill speed. With reference to Bartlett *et* al.'s and Sheik-Nainar *et al.*'s study, they reported that the walking speed between WC2 and both WC3 and WC4 varies (Table 6), which they related to the effect of optic flow on gait. Thus, the author suggests that in future studies the walking speed protocol should be performed for each of the treadmill environments, which would allow for a better understanding of the effect of the optic flow on this gait parameter. This would have been far more time consuming, but still it would have increased the fidelity in the other gait parameters' results, especially cadence and stride length since they are directly related to walking speed by Equation 3.

Preliminary analysis of the cadence results obtained from this study show agreement with the normal over ground values stated by Perry (1992). Moreover, the data shows agreement with that of Riley *et al*. and Sheik-Nainar *et al.*, that is, the cadence during the treadmill walking conditions is slower than that during over ground walking (Table 16). The cadence values measured, show that while there is a clear difference between the over ground and treadmill walking environments, the variation amongst the three treadmill walking conditions show similar results. The fixed treadmill speed could have affected the cadence such that the participants lowered their cadence in order to match their walking speed to that of the treadmill, leading to relatively proximate cadences amongst the different environments. With retrospective to Sheik-Nainar *et al.*'s study, the researchers reported a significant change between treadmill walking with and without optic flow (Table 6), such that the cadence for treadmill walking with optic flow better approximated that of over ground walking rather than treadmill walking. For the three environments during treadmill walking, a statistically significant difference was found between WC2 and WC4 (Table 36), while the other two pairwise comparisons showed no statistical difference (i.e. similarity). Primarily, despite the effort to ensure the integrity of the statistical modelling in this project, and that the assumptions of the tests chosen were adhered to; it is acknowledged that the statistical analytical methodology used has room for improvement in future projects; such advanced statistical modelling was out with the scope of this thesis. Secondly, it was noted that the participants commented that the WC4 was the most immersive environment out of the two provided, such that they felt more comfortable walking in this environment. The participants' confidence might have been reflected as an increase in walking speed during WC4, but since the treadmill speed was kept constant this was not possible, and thus this might have resulted in the participant having to alter the cadence (and consequently the stride length), thus masking the effect of the optic flow environment. If the participant senses treadmill speed to be slower than the CWS, then the participant will subconsciously alter the cadence and stride length to match the walking speed of the treadmill. With reference to Equation 3 if the walking speed is decreased, then the cadence is decreased and the stride length must be increased. The increase in stride length is evident in the mean stride lengths obtained for WC2 through WC4 (Table 17), while the decrease in cadence is also evident during WC4 (Table 16). Therefore, the participants' immersion during WC4 could have resulted in the small significance values obtained for the pair wise relationships for cadence between WC4 - WC3 and WC4 – WC2. Additionally, the strong similarity found between the cadences of WC2 and WC3, disagrees with the statement, which says that optic flow alters gait parameters making them closer to over ground walking conditions. This could be related to the fixed treadmill speed aforementioned, but nonetheless the participants' reaction to WC3 might have led to such similarities. It was expected that the cadence for WC3 would approximate that of WC4 (as shown in the statistical pair wise comparison) not WC2, but a majority of the participants commented that the environment during WC2 and WC3 caused them to feel unstable and loose balance. Warren *et al.* and Regnaux *et al.*, reported these sensations during lack of optic flow, in contrast with the sensations that our participants felt, which occurred during optic flow. The participants felt that these sensations occurred during optic flow due to the bright screen (in relation to the dark laboratory) along with the mismatch between their walking speed and the treadmill's fixed speed, which made them look towards the floor in order to be precautious about their foot placement. Therefore, due to the similarity in the sensations that were felt during WC2 and WC3 (imbalance and instability); it can be assumed that the cadences were similarly matched by the participants' locomotion control because of the similar sensations that they experienced during both walking conditions.

The Stride lengths obtained during over ground walking show agreement with the normal values reported by Perry (1992). The resulting over ground and treadmill stride lengths were consistent with those reported in literature by Murray *et al.*, Riley *et al.*, and Lee *et al.*, showing a distinct decrease in stride length from the former modality to latter (Table 17). Additionally, the stride lengths during the different treadmill modalities showed conformity with the results reported by Sheik-Nainar *et* *al.* (Table 6). Although the results obtained agreed with the majority of the reviewed literature studies, the statistical analysis (Table 36) still showed similarity for the stride length pair wise comparisons of WC3 - WC2, and WC4 – WC3, while, as expected, the significant difference in the pairwise comparison amongst the stride lengths measured for WC2 and WC4 was reported. The Wilcoxon Signed Ranks test result obtained for stride lengths echoes the one obtained for cadence, and thus enhances agreement with the deduction, that the similarity between WC2 and WC3 can be attributed to the fixed treadmill speed and the negative effect caused by the optic flow during WC3. The fixed treadmill speed seems to have affected both cadence and stride length similarly since they are related to each other. Finally, the similarity between WC3 and WC4 was expected, since the two walking conditions use analogous environments, but the greater immersion for WC4 shows that the quality of the optic flow has a positive effect on the gait parameters.

Initial analysis of the maximum hip flexion angles results revealed that the angles measured during over ground walking approximated normal values stated by Perry (1992). The angles between the over ground and treadmill modalities displayed a distinct decrease in the angles between the former and the latter, agreeing with the studies performed by Riley *et al.* (Table 4) and Lee *et al.* (Table 5). The decrease in the maximum hip flexion angle between over ground and treadmill walking is attributed to the fact that when an individual is walking on a treadmill, the hip's flexion angle is decreased in order to bring the foot down to contact quicker. This is occurs since the contra lateral foot is being dragged behind the centre of mass of the body faster than over ground walking. The statistical analysis for the maximum hip flexion angles (Table 36) reported that the angles between WC2 and WC3 were statistically different, while the angles between the other two pair wise comparisons for the treadmill modalities were statistically similar to each other. The statistical difference between WC2 and WC3 was expected, but this difference does not follow the assumption that optic flow improves the gait cycle as expected, but rather confirms that the optic flow used in WC3 during this study deteriorated the gait cycle since the mean values (Table 18) for WC3 are smaller than WC2. The smaller flexion values obtained could be due to inferior misplacement of the PSIS marker or posterior misplacement of the knee markers; although this artefact should have been eliminated because a repeated-measures design approach was undertaken. With reference to Regnaux *et al.*, it was remarked that when optic flow patterns are applied, the brain would attempt to preserve the kinematic patterns of the over ground ambulation while allowing the TSPs to vary. This statement relates to this study since the TSPs were noted to have been altered by the optic flow, but contrariwise the hip kinematics were not preserved, although it could be argued that the differences of the means obtained are very small and could fall within the range of repeatability of the gait parameters' variability. This decrease in the hip's maximum flexion angle is also potentially associated with the adverse effect that the optic flow had on the participants during WC3. It can be also associated with the fact that a few of the participants were observed to be looking at their feet for a considerable amount of time during this walking condition, which could have affected the hip kinematics. The statistical similarity reported between the pair wise comparison for WC2 – WC4, can be boiled down to a statistical false positive, due to the small statistical significance reported, along with the fact that this parameter is the only one that does not follow the pattern of the other gait parameters during this pair wise comparison. Furthermore, the fixed treadmill speed could also be potentially affecting the maximum hip flexion angles. Finally, the comparison between WC3 and WC4 was expected since the conditions were proved similar, along with a slight increase in the mean values for this parameter.

Preliminary analysis of the maximum knee flexion angles measured during this study show that the values relate to the over ground walking values stated by Perry (1992). Furthermore, the data show agreement with the studies performed by Lee *et al.* and Sheik-Nainar *et al.*, who got similar results between over ground and treadmill walking conditions. This finding is supported by the statistical analysis performed in this study (Table 36), which shows that the maximum knee flexion angles during over ground walking are not statistically different from all other conditions. This signifies that the maximum knee flexion angles are not affected by the surrounding environment's effect on the participants. Additionally the high significance value obtained for the statistical analysis of the angles measured during WC3 –WC4, further agrees with the statement above, suggesting that the maximum knee flexion angles are preserved when walking, regardless what kind of optic flow the individual receives. Finally, statistical difference was reported for the angles measured during WC2 – WC3 and WC2 – WC4, which contradicts the above statement, but with reference to the descriptive statistics for these three walking conditions (Table 19), the differences are small such that the variation measured might fall within the range of repeatability of the gait parameters'. This could also be attributed to fixed treadmill speed, which might have influenced this kinematic parameter adversely in WC3 and WC4.

Initial review of the maximum ankle dorsiflexion angles showed an increase from the normal values stated by Perry (1992). It was noted that the data measured for this gait parameter ranged between 8.73^0 and 33.65^0, which is considerably different from the 10^0 mean value stated by Perry. These values are in agreement with those obtained by Alton et al. Nonetheless, this difference is attributed to marker misplacement since a superiorly placed ASIS marker or a posteriorly placed thigh marker or an anteriorly placed ankle marker or a disagreement between the imaginary axis (due to the heel and toe markers) and real axis of the foot, could all lead to an increase in the ankle's dorsiflexion angle. The angles measured for all the walking conditions showed agreement with the statement that treadmill walking with optic flow improves the gait parameters over treadmill walking with no optic flow, thus bringing them closer to over ground parameters. The statistical analysis (Table 36) reported that the only pair wise comparison that was not statistically different was WC3 – WC4, which shows that the ankle kinematics researched in this study did experience a difference amid the walking environments. This similarity between the aforementioned pair wise comparison signifies that the maximum ankle dorsiflexion angles between these two walking conditions, are not affected by the imbalance and instability experiences by the participants, but by the optic flow since they remained similar to each other when the optic flow did not change.

In conclusion, the results obtained for gait parameters show agreement with the work of previous researchers, who state that there exists a difference between the gait cycle parameters during over ground walking and treadmill walking. Moreover, treadmill walking with optic flow alters the participants' walking speed, in relation to treadmill walking (WC2), which was reflected in the participants' comments, which stated that the treadmill velocity during WC3 and WC4 was noticed to be slower than their walking speed. The effect of the fixed walking speed reflected in the cadence and stride length values alike, since it was speculated that the cadence and stride length alter inversely in order to match the treadmill walking speed. Meanwhile, both cadence and stride length during WC2 were mirrored on WC3 since the participants experienced similar sensations of losing balance and instability, which might have masked the effect of optic flow between these two modalities, thus affecting the participants' gait in the same way for both modalities. The effect of the fixed treadmill speed and the optic flow are believed to have caused the participants to experience imbalanced and unstable sensations, which could have produced different neurological pathways that preserve balance and stability. To counteract this imbalance and instability; the brain might have altered the gait in a manner that did not only affect the TSPs but also the maximum hip flexion angles, and possibility the maximum knee flexion angles. The maximum knee angles measured showed that this gait parameter is not affected by the change in environment, and thus it was speculated that the neurological pathways preserve the knee kinematics during these different conditions in order for the gait cycle to advance. Finally, the maximum ankle dorsiflexion angles show a deviation from the mean values for this parameter, which was attributed to marker misplacement. The maximum angles for the ankle joint were noticed to be significantly different amongst all conditions except for the two conditions using optic flow, which reflects that the variation in the angles is due to the change in optic flow.

5.2 ACCURACY ANALYSIS

The secondary objective of this study challenged the accuracy of the state-of-the-art motion capture systems. This accuracy analysis is based on the assumption that the measured inter-marker distances using the milling machine were considered the actual (golden-standard) inter-marker distances, although unavoidable human and machine error exists within the measured values. While the necessary precautions to get the most accurate inter-marker distances were taken, the measured data could not be compared to manufacturer's values, since this information is not made available to the customers. The measurements captured in both laboratories showed very high accuracies for both the static and dynamic accuracy testing procedures, thus confirming that the protocol followed during this accuracy analysis was carefully planned out.

The Conventional Human Movement Analysis Laboratory reported a mean capture volume accuracy of 99.88% (having a maximum error of 0.12 ± 0.01 mm). The static accuracy followed an increasing trend with height, which can be observed in the average accuracies (Table 29) reported for the ankle, knee and hip height (measured for all the seven positions). This trend was expected since the cameras in the motion laboratory are aimed towards the centre of the capture volume, which would be located towards the hip height range. Additionally, the accuracy tends to increase from position 1 through position 3 and remains the same from position 3 through position 7, along with decreasing variability in the data form position 1 to position 7. This data was as expected, that is, to reach a maximum accuracy towards position 4 (followed by a relative decrease again until position 7), since this lied in the origin of the laboratory where the highest accuracy is expected from a laboratory. Furthermore, the values obtained for the error ± S.D. showed improvement upon the ones reported by Ehara *et al.* (2002) and Kertis (2012). This is attributed to the progress in the technology of these motion capture systems, which always strive to improve the accuracy, to increase the fidelity in the results measured by the system. For the dynamic accuracy testing performed in this laboratory, the accuracy calculated for the three different wand speeds were identical (Table 31), while the range of the accuracies calculated showed an increasing trend with speed, despite the fact that the variability of the data decreased with speed. The difference between the dynamic and static accuracy was not expected to improve during the dynamic accuracy (although the difference is just 0.02%), but the decreasing variability with speed was predicted since the faster the markers move the harder it is for the system to get a clear frame of the marker. The accuracies measured in this laboratory showed improvement over the papers discussed in the literature review.

The Motek Medical CAREN Extended Laboratory reported a mean capture volume accuracy of 99.68% (having a maximum error of 0.32 ± 0.03 mm). The static accuracy also followed an increasing trend with height (Table 33) which can be observed in the average accuracies obtained for the ankle, knee and hip height (measured for all the seven positions). This pattern was expected due to the cameras being aimed and calibrated to have the highest accuracies in the centre of the capture volume, which normally lies at hip height over the origin. Furthermore, the accuracy was highest at position 1 and decreased through position 3 (measured for all the three heights), along with increasing variability in the same direction. The data reported was slightly different from the expected data, which was anticipated to have a maximum accuracy at position 2 (over the origin of the laboratory), while the other two positions were expected to have slightly lower accuracies. With reference to the studies by Ehara *et al.*, Myers *et al.*, Kidder *et al.* and Kertis, the accuracies calculated in this static study fell just below the accuracy measured by Myers *et al.* and improved over those reported by the other researchers. Although Myers *et al.* had better accuracy, the variation of the data reported was nearly twice that obtained in this study. For the dynamic accuracy testing performed in this laboratory, the accuracies reported for the slow and medium wand speeds were identical (Table 35), while the fast wand speed had the highest mean accuracy. The accuracy of the dynamic testing was not expected to show improvement over the static accuracy results. Moreover, the variability of the data does not follow a pattern with the increasing velocity, which may have occurred because of the wand passing close to the border of the capture volume during the waving procedure undertaken. This may have resulted in some of the cameras to calculate an imprecise marker location, since, as seen in the static accuracy results, the most accurate areas are towards position 1 at hip height. This could have been improved in the protocol stage of this testing if the SAMSA protocol setup (Figure 35) was used to measure the dynamic accuracy, as this would have replicated the markers' trajectories for each individual speed. The mean accuracy calculated for the dynamic testing was lower than the minimum accuracy that Myers *et al.* obtained in his study, but larger than the ones evaluated by Kertis and Kidder *et al.*. This higher accuracy is attributed to the fact that Myers *et al.* utilised a 15-camera system, which could have improved the system's ability to accurately capture the moving markers. Additionally, the variation of the accuracy data measured in the dynamic trial showed the lowest variation when compared to the three papers discussed earlier.

The motion capture system installed in the Conventional Human Movement Analysis Laboratory showed an increased accuracy and decreased variability in comparison to the system installed in the Motek Medical CAREN Extended Laboratory. This finding was expected since the system installed in the Conventional laboratory has a much higher pixel-depth resolution than the former system, along with the added Full Marker Grey-Scale feature. The conventional laboratory was found to be the most accurate at hip height, from position 3 to position 7, while the CAREN laboratory was found to be the most accurate at hip height in position 1. In conclusion, both systems performed remarkably, since they both accurately measured the markers within error ranges, which will be masked during human motion analysis due to the considerable amount of variables, which exist that vary the markers location during analysis, such as skin motion artefacts and marker dropouts.

6 CONCLUSIONS

Both motion capture systems whose accuracy was analysed showed excellent levels of accuracy. The comparison of the gait analysis in different optic flow environments showed that the effect of fixed speed had a distinct effect on the cadence and stride length values, while the effect of the optic flow echoed similar results for the TSPs during WC2 to WC3. The maximum knee flexion angles showed that they are not affected by the change in environment, while the ankle dorsiflexion angles showed statistically significant changes that varied with the level of optic flow immersion. Thus, the use of treadmills with virtually realistic environments as a rehabilitation tool should be further investigated by the research community in order to gain a better understanding to the variables that affect human gait. The recent growth in the research of this field is already noticed due to the rising demand of Computer Assisted Rehabilitation Environments (CAREN) such as the installation of the Motek Medical CAREN Extended Laboratory used in this study.

7 SUGGESTIONS FOR FUTURE WORK

If further work were to be performed on this study, the author would like to suggest the following improvements on the work already achieved:

- More time should be allowed for the participants to acclimatise themselves to each treadmill-walking environment, in order to eliminate any confounding variables, which may be related to cautious treadmill walking.

- A method of calculating the participants walking speed in each individual environments should be investigated to implement it in the protocol undertaken in this study. This will avoid having the issue encountered in this study, which resulted in the repercussive effect on other gait variables, thus having less effective results.

- The method of self-pace treadmill speed should be further studied and researched for the effect this mode has on the human gait cycle, so that such a versatile walking mode could be used effectively in different studies while knowing its limitations.

- In this study it was shown that using the appropriate optic flow during treadmill walking clearly has an effect on gait cycle variables. More research should be performed on the effect that optic flow, due to different levels of immersion, has on different gait parameters in order to have a broader knowledge of the effects that different virtually realistic environments show.

- The static and dynamic accuracy testing should utilise the SAMSA protocol, which would allow for a repetitive and standardised way of measuring and comparing the accuracy of motion analysis laboratories. The protocol also includes error thresholds, which represent the limits under which the errors are not within acceptable limits, and furthermore, this allows for a systematic comparison of the laboratories investigated.

- Joint kinetics should be added to the list of gait parameters that would be analysed in such a study. Researchers such as Lee *et al.* commented that kinetic data gave better understanding of the participants' over all gait cycle.

- A Presence Questionnaire (Witmer and Singer, 1994) should be given to the participants to comment and complete. Such a presence Questionnaire investigates underlying factors and research variables, which affect studies that require participants to immerse themselves in a Virtually Realistic Environment. This PQ will ultimately give the researcher a quantitative representation of the subjective perceptions of the degree of immersion in the virtual environment being utilised.

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9 APPENDICES

9.1 APPENDIX A

Ethical Sheets given to participants prior to initiating testing.

Participant Information Sheet

Name of department: Biomedical Engineering

Title of the study: An Evaluation of the Accuracy and Precision of the Motek Medical Computer Assisted Rehabilitation Environment (CAREN) for the assessment of Human Gait Biomechanics

Introduction

Chief Investigator: Dr Andrew Murphy
Status: Research Fellow
Department: Biomedical Engineering
Telephone: 01415482855
E-mail: andrew.j.murphy@strath.ac.uk

Research Student: Mr Andre Attard
Status: Postgraduate student
Department/Institution: Biomedical Engineering
Name of supervisor: Andrew J Murphy
Telephone: +447835266072
E-mail: andre.attard.2013@uni.strath.ac.uk

What is the purpose of this investigation?

The purpose of this project is to compare and evaluate human walking biomechanics of healthy participants under different circumstances. The study will focus on three different walking environments, specifically over ground normal walking, treadmill walking without any virtual reality and treadmill walking with the visual aid.

Do you have to take part?

The participant will be expected to take part in this investigation voluntarily and it is up to the participant's decision to refuse to participate before or during the investigation itself, along with the removal of any data records of the participant from the study without giving any reason whatsoever. This will not in any way affect the participant's relationship with the University of Strathclyde or any of its members.

What will you do in the project?

The participant will be asked to attend a session (location given below) for not longer than 4 hours (30 minutes of which will be solely dedicated to walking), of which the participant will be given prior notice. The participant will need to wear appropriate clothing so that accurate motion of the body while walking can be recorded (appropriate clothing will be provided by the department, if necessary). Male participants will be required to wear tight cycling type shorts with no top on. Female participants will be required to wear tight cycling type shorts and a tight fitting crop type top. All participants will be required to bring with them sports type shoes. The participant will affixed with reflective markers using medical grade non-allergic tape and asked to walk along a short path for a short period of time after which he will be allowed to rest if necessary. Then, the participant is asked to walk for another short period of time on a treadmill, again allowed to rest prior to the final part of the experiment which is a final short walk on the same treadmill used before, but this time using virtual reality aid for the participant. There could be times when the participant will be video recorded and photographed, only if the participant agrees to this beforehand. The experiment offers no incentives for application nor reimbursements to potential participants. The laboratory session will take place in the following location:

The University of Strathclyde,
Department of Biomedical Engineering,
Biomechanics laboratory 2,
Level 1,
Wolfson Centre,
106 Rottenrow,
G4 0NW.

Why have you been invited to take part?
The chosen participants will be developing young adults over the age of 18, falling under the following criteria:

Inclusion criteria

- Able bodied
- 5’2” to 6’2’ in height.
- Normal lower limb function
- Knowledge of using a treadmill
- 20/20 vision (with or without visual aid)

Exclusion criteria

- Musculoskeletal, neurological or sensory deficit
- Pregnancy

What are the potential risks to you in taking part?
This study offers no potential risks to the participant.

What happens to the information in the project?
The participants will be given time to decide whether they wish to be considered for participation in the study. Furthermore, the participants will be asked to consider whether they wish to provide consent for the following:

- Consent to being photographed and video recorded as part of the project.
- Consent for unidentifiable photographs and video recordings to be used in publications or teaching materials.

Any identifiable information
The consent form will be kept confidential, stored for 5 years in a locked cabinet in the Department of Biomedical Engineering. These will be available for those named in this application and will be destroyed on more than 5 years after completion of the study. If consent is given by the participants, video will be taken. Participants will be identifiable from this but these will be stored on password protected non-networked hard drives with secure access only by the named researchers. Additionally, all the information will be saved as a backup in a password protected folder on password protected University of Strathclyde computers and external hard drives. If consent is given all videos will be kept indefinitely.

Any pseudo-anonymous data (anonymised raw data and given a code name, with the key for code names being stored in a separate location from the raw data)
An ID code will link the collected data to the participant. The code list will be stored in a locked cabinet in the Department of Biomedical Engineering. The coded list will only be available for those named in the application and will be destroyed 5 years after completion of the study. Thereby the pseudo-anonymous data will become anonymous. All experimental data will be stored pseudo-anonymously, coded with an ID-number. All videos will be coded with the same ID-number. All experimental data will be kept indefinitely, but will become fully anonymous 5 year after completion of the study; when the master file associating participants' names with their ID number is destroyed. Data will be securely stored and its access and destruction will be in accordance with the University of Strathclyde Data Protection Policy. All computing systems holding electronic data and all hard data will be stored within lock & key, and/or, magnetic swipe card security access enabled offices and laboratories within the Department of Biomedical Engineering of the University of Strathclyde. Videos and all other experimental data will be stored on password protected hard drives with secure access only by the named researchers

The University of Strathclyde is registered with the Information Commissioner's Office who implements the Data Protection Act 1998. All personal data on participants will be processed in accordance with the provisions of the Data Protection Act 1998.

What happens next?

Once the participant agrees with the information given above and would like to participate in this research study, all that will have to be done is to read and sign the consent form. This should be then handed to any of the investigators/researchers mentioned in the following section.

In the case that the participant does not wish to be involved in the project, then the investigators of this study would like to take the opportunity to thank the participant for taking interest in this research project.

If the participant would like to receive feedback about the progress of the study post-testing he is encourage to contact any of the investigators on the contact details given below. If any of the results from this study will be published, the participants will be advised beforehand.

Researcher contact details:
Research Student: Mr Andre Attard
Status: Postgraduate student
Department/Institution: Biomedical Engineering
Name of supervisor: Andrew J Murphy
Telephone: +447835266072
E-mail: andre.attard.2013@uni.strath.ac.uk

Chief Investigator details:
Full Name: Dr Andrew Murphy
Status: Research Fellow
Department: Biomedical Engineering
Telephone: 01415482855
E-mail: andrew.j.murphy@strath.ac.uk

Thank you for reading this information – please ask any questions if you are unsure about what is written here. This investigation was granted ethical approval by the University of Strathclyde Ethics Committee. If you have any further questions/concerns, during or after the investigation, or wish to contact an independent person to whom any questions may be directed or further information may be sought from, please contact:

Linda Gilmour
Secretary to the Departmental Ethics Committee
National Centre for Prosthetics and Orthotics
Department of Biomedical Engineering
Curran Building, 131 St James Road
Glasgow G4 0LS
Tel: 0141 548 3298 E-mail: linda.gilmour@strath.ac.uk

Consent Form for Participants

Name of department: Biomedical Engineering
Title of the study: An Evaluation of the Accuracy and Precision of the Motek Medical Computer Assisted Rehabilitation Environment (CAREN) for the assessment of Human Gait Biomechanics

- I confirm that I have read and understood the information sheet for the above project and the researcher has answered any queries to my satisfaction.
- I understand that my participation is voluntary and that I am free to withdraw from the project at any time, without having to give a reason and without any consequences.
- I understand that I can withdraw my data from the study at any time without giving reason.
- I understand that any information recorded in the investigation will remain confidential and no information that identifies me will be made publicly available.
- I understand that whether I participate in the project or not will in no way affect my standing within the University of Strathclyde.
- I confirm that I meet the inclusion/exclusion criteria.
- I consent to being a participant in the project and for the collection, documentation and usage of data gathered during the experiment.
- I understand that incentives/reimbursements will not be offered for participation.

Optional:
- I consent to the use of unidentifiable audio and video data recorded as part of the project for educational purposes
- I consent to the use of unidentifiable audio and video data recorded as part of the project in future publications [delete which is not being used] Yes/ No

Full Name of Participant:	
Signature of Participant:	Date:

9.2 APPENDIX B

Comparison of Human Gait Biomechanics Post Processed Data Sheet

Conventional Human Movement Analysis Data

		Walking Speed (m/s)	Cadence (Steps/min)	Stride Length (m)				Max Hip Flexion Angle (^{o})					
				1	2	3	4	1	2	3	4	5	6
Participant 3	1	1.15	107.91	1.27	1.28	1.27	1.26	35.55	35.34	35.78	35.94	34.28	34.53
	2	1.36	114.29	1.44	1.42	1.41		40.22	40.49	39.09	38.17	38.89	
	3	1.35	114.07	1.49	1.42	1.37	1.37	41.67	39.86	41.22	42.17	36.93	39.00
	4	1.31	111.63	1.46	1.41	1.36		40.35	41.66	40.60	41.35	38.97	
	5	1.28	109.89	1.42	1.39	1.39	1.35	42.91	38.78	39.39	40.61	39.77	38.52
Participant 4	1	1.71	119.40	1.66	1.70	1.78		39.68	40.06	40.22	40.89	41.47	
	2	1.71	122.45	1.69	1.73			40.98	39.40	39.19	40.05		
	3	1.71	122.45	1.60	1.64	1.75		39.95	37.99	39.14	39.47	40.46	
	4	1.72	117.65	1.74	1.73	1.76		40.89	40.49	39.83	39.60	40.46	
	5	1.69	120.00	1.64	1.66	1.74		40.53	38.64	39.17	40.30	40.96	
Participant 5	1	1.37	112.78	1.47	1.45	1.46	1.49	35.43	36.14	36.90	36.88	38.28	39.77
	2	1.45	110.60	1.60	1.56	1.56		38.77	38.83	41.47	38.37	38.91	
	3	1.42	113.74	1.46	1.50	1.55		38.48	39.30	39.64	39.02	40.21	
	4	1.44	114.29	1.47	1.50	1.55		37.72	39.94	39.87	39.25	39.86	
	5	1.42	115.38	1.48	1.50	1.54		38.77	38.64	40.04	39.83	40.24	
Participant 6	1	1.33	113.64	1.43	1.40	1.40	1.40	39.76	39.53	38.45	38.28	39.53	39.41
	2	1.33	114.50	1.43	1.42	1.39	1.40	36.81	34.78	38.31	39.01	38.99	38.68
	3	1.33	113.21	1.43	1.40	1.40		37.95	39.12	38.06	43.12	39.91	
	4	1.31	111.11	1.44	1.39	1.43	1.41	38.11	38.41	40.37	38.48	38.16	40.45
	5	1.36	114.94	1.42	1.42	1.42	1.42	37.15	39.10	36.91	38.56	40.13	38.99
Participant 7	1	1.33	126.13	1.23	1.29	1.30	1.27	26.54	28.41	27.99	26.37	27.16	26.00
	2	1.36	128.34	1.23	1.30	1.33		30.28	30.14	27.36	28.49	28.91	
	3	1.38	127.66	1.25	1.29	1.34		26.71	28.03	29.67	26.57	26.93	27.64
	4	1.37	129.73	1.24	1.30	1.29		29.81	30.55	27.74	27.41	27.52	
	5	1.34	127.66	1.22	1.28	1.30		28.43	29.16	29.21	26.81	27.96	28.35
Participant 8	1	1.50	111.63	1.58	1.62	1.64		33.30	34.92	26.43	29.56	29.47	
	2	1.44	109.09	1.57	1.58	1.61		33.05	34.65	25.64	28.50	28.22	
	3	1.48	111.11	1.60	1.59	1.60		34.41	33.54	27.12	32.81	29.68	
	4	1.47	109.59	1.60	1.59	1.63		34.67	33.76	27.35	29.44	29.60	
	5	1.46	110.09	1.58	1.62	1.62		34.81	34.71	27.01	31.47	29.56	
Participant 9	1	1.55	113.92	1.63	1.63	1.63		35.30	36.56	39.43	40.29	1.07	
	2	1.52	114.83	1.58	1.57	1.59		35.59	37.85	38.61	39.15	40.15	
	3	1.53	114.65	1.60	1.60			36.29	37.65	36.97	38.84	37.58	
	4	1.54	115.94	1.62	1.58	1.56		35.10	36.78	39.35	39.38	38.34	
	5	1.52	115.38	1.61	1.59	1.56		36.39	36.04	39.23	37.06	37.15	

		Max Knee Flexion Angle (°)						Max Dorsi Flexion Angle (°)					
		1	2	3	4	5	6	1	2	3	4	5	6
Participant 3	1	59.22	59.97	63.99	61.77	62.10		15.07	15.18	16.17	13.81	15.78	15.62
	2	62.56	60.24	63.57	63.70			15.23	14.53	14.90	16.49	14.57	
	3	60.22	60.96	61.30	65.04	64.08		13.82	13.67	15.72	13.43	13.74	15.64
	4	61.55	62.09	64.96	65.46			12.10	13.57	10.98	13.03	13.66	
	5	59.97	61.46	63.14	64.22	63.54		13.12	12.70	14.30	12.73	12.88	12.84
Participant 4	1	77.81	74.67	73.36	74.73			20.86	18.78	19.18	18.92	17.70	
	2	71.45	68.88	71.89				15.09	13.60	13.66	18.31	18.35	
	3	74.70	71.81	71.39	71.10			14.13	12.90	13.67	15.58	15.03	16.40
	4	75.65	72.20	71.54	74.71			15.30	12.87	15.35	16.05		
	5	73.90	73.43	72.19	71.07			21.01	17.92	15.70	17.76	17.35	
Participant 5	1	55.31	56.03	55.45	62.45	62.34		29.77	27.35	27.33	28.46	27.67	27.28
	2	55.37	52.18	59.48	60.27			24.63	27.32	28.09	23.94	25.79	
	3	56.35	55.66	62.14	61.98			24.20	27.35	26.49	27.08	26.50	
	4	55.11	55.25	64.68	63.76			26.68	29.45	28.44	28.42	25.41	
	5	54.89	54.60	63.76	62.44			25.15	27.64	26.56	27.63	25.59	
Participant 6	1	69.29	67.63	66.29	63.77	62.22		32.82	33.09	31.12	30.62	29.22	28.67
	2	65.35	66.12	66.61	60.45	61.11		33.56	29.48	30.67	27.93	28.55	29.64
	3	66.95	66.47	63.08	63.70			32.71	30.30	26.82	29.63	27.42	
	4	66.26	68.19	66.63	54.24	63.31		33.65	32.82	31.09	28.85	29.37	28.87
	5	65.77	67.97	66.68	61.65	60.82		33.28	31.39	30.68	30.50	29.98	29.33
Participant 7	1	59.54	60.16	61.04	54.33	52.24	55.08	17.70	16.48	17.61	18.31	18.79	17.34
	2	57.90	59.63	50.73	53.24			18.24	14.89	16.00	18.07	15.39	
	3	57.22	56.63	52.67	52.08	53.20		16.73	17.43	15.86	17.36	17.21	17.16
	4	58.56	57.56	55.45	51.33			17.78	16.32	16.30	17.64	15.94	
	5	57.50	57.98	52.60	53.95	54.37		15.94	15.36	15.91	16.92	17.42	16.95
Participant 8	1	69.65	69.06	61.57	60.31			14.34	14.11	13.24	12.09	10.50	
	2	69.50	68.14	60.65	59.69			15.35	16.05	14.39	12.74	11.27	
	3	69.87	69.16	63.85	61.05			11.82	14.44	13.46	10.11	12.06	
	4	70.02	68.55	60.84	60.11			14.01	15.26	15.86	12.14	9.06	
	5	70.71	69.34	61.03	61.17			17.53	13.56	13.73	12.02	9.07	
Participant 9	1	61.72	61.97	62.37	65.40	63.04		30.32	25.69	26.93	26.15	30.62	
	2	65.71	67.19	65.70	64.15			31.30	29.98	30.50	24.99	26.54	
	3	67.27	65.10	63.45	63.45			30.88	30.95	30.12	26.94	26.21	
	4	65.16	66.36	64.51	63.33			31.07	31.43	30.86	26.15	27.16	
	5	67.15	65.77	61.46	61.12			30.62	30.50	29.45	27.60	27.82	

Motek Medical CAREN Extended System Laboratory Data

		Walking Speed (m/s)	Cadence (Steps/min)	Stride Length (m)									
				1	2	3	4	5	6	7	8	9	10
Participant 3	TW	1.09	109.09	1.28	1.30	1.27	1.27	1.26	1.29	1.32	1.27	1.28	1.35
	TWOF	1.09	110.20	1.25	1.24	1.31	1.26	1.25	1.37	1.23	1.28	1.31	1.28
	TWOF + M	1.09	106.30	1.31	1.26	1.30	1.30	1.31	1.31	1.37	1.34	1.26	1.34
Participant 4	TW	1.22	112.97	1.36	1.39	1.37	1.38	1.39	1.34	1.37	1.42	1.40	1.37
	TWOF	1.22	110.88	1.46	1.37	1.37	1.34	1.35	1.45	1.40	1.38	1.39	1.36
	TWOF + M	1.22	109.98	1.39	1.49	1.43	1.37	1.41	1.41	1.47	1.41	1.39	1.46
Participant 5	TW	1.01	106.72	1.20	1.19	1.19	1.21	1.17	1.12	1.14	1.17	1.15	1.11
	TWOF	1.01	108.43	1.19	1.18	1.17	1.16	1.18	1.13	1.15	1.14	1.13	1.16
	TWOF + M	1.01	108.87	1.18	1.15	1.19	1.19	1.19	1.13	1.14	1.18	1.13	1.14
Participant 6	TW	0.94	104.27	1.18	1.12	1.16	1.14	1.12	1.20	1.15	1.17	1.13	1.16
	TWOF	0.94	105.94	1.17	1.14	1.15	1.16	1.15	1.17	1.19	1.17	1.18	1.15
	TWOF + M	0.94	102.33	1.14	1.14	1.14	1.16	1.15	1.16	1.13	1.16	1.18	1.18
Participant 7	TW	0.94	109.98	1.11	1.12	1.14	1.12	1.15	1.11	1.13	1.13	1.13	1.15
	TWOF	0.94	108.22	1.20	1.07	1.13	1.14	1.12	1.16	1.14	1.15	1.12	1.07
	TWOF + M	0.94	105.68	1.17	1.15	1.19	1.12	1.16	1.05	1.14	1.15	1.11	1.19
Participant 8	TW	1.17	109.82	1.31	1.31	1.22	1.29	1.25	1.32	1.25	1.24	1.27	1.28
	TWOF	1.17	106.45	1.37	1.38	1.32	1.39	1.33	1.42	1.38	1.36	1.33	1.35
	TWOF + M	1.17	110.55	1.32	1.30	1.29	1.27	1.30	1.28	1.30	1.30	1.26	1.26
Participant 9	TW	1.11	109.53	1.26	1.34	1.24	1.31	1.23	1.31	1.28	1.29	1.31	1.30
	TWOF	1.11	112.97	1.24	1.32	1.22	1.20	1.22	1.30	1.31	1.18	1.21	1.27
	TWOF + M	1.11	109.31	1.32	1.26	1.27	1.28	1.26	1.35	1.40	1.30	1.26	1.33

		Max Hip Flexion Angle (^{o})											
		1	2	3	4	5	6	7	8	9	10	11	12
Participant 3	TW	43.31	43.11	42.61	42.21	41.51	43.81	41.30	43.42	42.27	40.70	42.52	41.57
	TWOF	41.83	43.51	39.79	43.22	42.57	41.75	39.66	40.94	38.50	40.26	41.08	40.51
	TWOF + M	42.27	42.98	43.07	42.79	42.33	43.25	39.75	41.63	42.56	41.88	42.40	44.36
Participant 4	TW	31.96	36.00	33.79	33.88	34.22	35.36	28.13	28.84	30.48	30.33	31.34	30.44
	TWOF	32.50	34.48	33.80	33.59	32.43	33.70	30.26	29.84	29.89	28.47	29.90	33.05
	TWOF + M	34.08	33.36	32.17	32.23	32.36	33.09	31.17	30.97	30.88	30.10	28.85	28.36
Participant 5	TW	39.37	38.90	39.21	38.94	38.03	39.62	44.16	43.54	42.85	43.65	42.87	42.59
	TWOF	37.38	39.19	38.05	37.17	38.92	37.42	43.09	42.02	42.89	42.20	41.45	43.57
	TWOF + M	39.04	38.68	40.44	39.23	37.17	39.95	42.48	44.95	43.55	43.80	43.26	43.14
Participant 6	TW	32.96	32.82	32.11	32.33	33.47	34.55	32.42	33.80	33.04	34.48	34.73	33.99
	TWOF	32.73	35.42	33.91	32.93	34.90	32.37	33.37	35.63	33.90	33.98	34.83	34.34
	TWOF + M	32.49	32.64	31.97	31.67	33.90	34.16	32.39	34.29	32.78	32.50	35.81	35.23
Participant 7	TW	26.26	24.42	22.88	22.84	21.97	24.52	22.55	23.34	23.28	22.30	23.34	23.22
	TWOF	22.05	24.53	22.87	22.15	20.90	21.79	22.36	23.60	23.58	23.13	22.89	22.68
	TWOF + M	23.03	21.38	24.76	22.03	25.37	27.41	24.11	22.20	22.62	20.92	22.86	25.55
Participant 8	TW	32.76	34.26	29.80	31.63	30.67	31.99	23.11	27.58	26.94	25.11	27.68	25.78
	TWOF	32.70	32.53	31.65	31.26	32.28	32.08	22.04	26.01	26.56	26.68	27.66	28.58
	TWOF + M	30.15	30.86	30.24	30.90	32.37	30.43	22.79	28.92	27.73	26.17	26.13	24.37
Participant 9	TW	35.95	35.63	35.05	36.63	36.20	34.42	35.98	36.00	36.40	37.31	37.64	36.74
	TWOF	35.34	33.44	33.59	32.71	35.34	36.57	33.13	36.59	35.64	35.28	34.68	35.79
	TWOF + M	35.09	37.33	36.41	33.57	33.28	35.39	32.76	35.61	37.12	35.89	35.67	34.53

		Max Knee Flexion Angle (°)										
		1	2	3	4	5	6	7	8	9	10	11
Participant 3	TW	60.88	61.30	60.56	60.68	57.25	60.69	64.59	64.11	62.96	64.32	63.10
	TWOF	59.73	59.77	60.59	63.10	60.03	61.68	64.07	65.65	63.39	61.90	63.99
	TWOF + M	60.31	60.11	59.73	61.63	60.86	60.84	63.07	62.57	62.45	64.08	61.05
Participant 4	TW	70.03	72.35	69.04	71.04	70.29	73.24	71.33	69.00	68.01	69.31	69.79
	TWOF	69.97	70.51	70.08	70.09	69.15	68.85	67.40	70.13	69.88	69.64	71.68
	TWOF + M	72.80	71.03	71.20	70.87	71.65	70.77	71.10	69.10	69.88	71.24	70.30
Participant 5	TW	58.13	55.88	58.75	58.46	58.84	58.80	66.71	64.77	66.26	66.54	66.83
	TWOF	59.93	61.56	57.93	60.28	61.20	59.03	67.36	68.17	68.56	66.35	69.35
	TWOF + M	58.37	59.81	61.71	58.22	57.12	60.00	68.87	66.64	68.47	70.02	67.35
Participant 6	TW	64.94	64.05	64.12	64.26	64.09	63.81	58.42	60.69	61.06	60.04	58.93
	TWOF	65.11	66.54	65.60	66.18	66.55	65.11	59.97	59.56	59.50	60.74	60.25
	TWOF + M	65.49	65.54	66.03	65.95	64.61	66.86	60.26	61.39	61.79	62.34	62.78
Participant 7	TW	59.52	58.87	59.47	57.55	58.36	60.09	54.64	57.38	55.81	56.19	55.30
	TWOF	58.40	60.09	58.92	59.48	59.12	57.84	52.53	56.06	56.72	56.84	51.91
	TWOF + M	60.88	60.36	60.74	59.52	60.54	60.41	56.58	52.26	55.26	53.57	55.11
Participant 8	TW	70.70	72.39	70.86	70.00	68.83	69.04	61.82	60.00	58.44	61.91	59.70
	TWOF	71.69	70.58	70.86	70.45	72.61	69.48	60.86	62.11	62.23	62.00	63.20
	TWOF + M	66.79	69.61	69.89	71.09	71.43	70.21	63.50	62.69	62.24	61.71	59.18
Participant 9	TW	66.76	66.77	67.63	67.55	65.84	65.59	67.05	63.54	66.63	66.45	66.95
	TWOF	68.23	65.68	68.15	66.67	67.17	67.46	67.18	65.95	67.15	65.72	65.70
	TWOF + M	65.89	66.02	65.76	65.92	66.63	65.36	63.63	62.77	66.44	64.32	64.23

		Max Dorsi Flexion Angle (°)											
		1	2	3	4	5	6	7	8	9	10	11	12
Participant 3	TW	15.16	15.42	15.93	16.34	18.89	16.00	16.44	16.57	16.22	15.71	15.12	17.54
	TWOF	17.09	17.26	17.30	18.41	18.87	17.61	15.78	15.05	16.00	17.17	16.30	15.23
	TWOF + M	16.90	16.41	16.74	14.98	15.38	17.06	15.12	17.25	14.65	14.74	16.76	16.54
Participant 4	TW	14.84	16.60	16.12	17.31	19.32	15.39	16.54	16.99	16.70	16.33	18.29	
	TWOF	16.83	13.84	18.01	16.71	16.90	18.43	12.86	14.21	17.30	18.81	16.53	14.17
	TWOF + M	15.55	16.99	15.71	17.27	17.71	18.43	17.64	15.37	15.26	16.90	18.28	18.69
Participant 5	TW	10.81	11.41	9.17	12.25	12.35	12.65	15.11	13.87	14.08	14.82	14.43	13.68
	TWOF	12.62	12.74	12.90	13.81	15.73	13.53	15.83	16.25	15.70	16.16	15.77	13.43
	TWOF + M	16.50	16.53	13.95	13.38	13.92	13.01	17.22	18.15	16.02	16.79	16.19	15.21
Participant 6	TW	14.18	12.07	12.65	15.07	14.34	14.33	11.32	10.12	12.25	11.95	12.76	10.02
	TWOF	15.79	13.92	13.38	13.60	14.71	14.99	14.27	11.63	11.11	11.89		
	TWOF + M	18.20	15.60	15.18	14.99	15.87	13.53	9.70	13.61	15.70	14.63	14.56	7.64
Participant 7	TW	15.70	17.77	17.11	17.66	18.20	19.35	17.98	17.59	17.70	17.75	17.99	14.23
	TWOF	15.04	15.11	16.02	16.82	19.19	19.37	14.31	17.62	17.24	17.92	17.16	13.16
	TWOF + M	17.70	19.27	16.77	15.67	18.48	18.25	15.80	15.74	17.77	18.60	17.95	14.53
Participant 8	TW	17.25	13.79	13.99	19.39	18.36	12.03	11.26	10.29	8.91	12.76	7.83	11.37
	TWOF	17.43	16.74	19.03	17.96	20.84	17.84	11.94	12.10	8.73	8.97	12.21	11.31
	TWOF + M	20.89	18.42	15.85	14.91	15.19	17.35	12.68	11.81	9.85	11.64	10.01	10.06
Participant 9	TW	14.73	14.79	15.85	17.28	14.14	15.11	13.76	12.96	16.44	13.18	15.73	12.69
	TWOF	17.77	17.06	13.31	15.41	19.96	16.94	14.83	14.22	14.09	15.63	15.70	15.36
	TWOF + M	14.00	15.08	13.38	17.25	16.70	14.16	13.84	16.73	13.42	10.39	14.01	12.96

9.3 APPENDIX C

Accuracy Analysis Post Processed Data Sheet

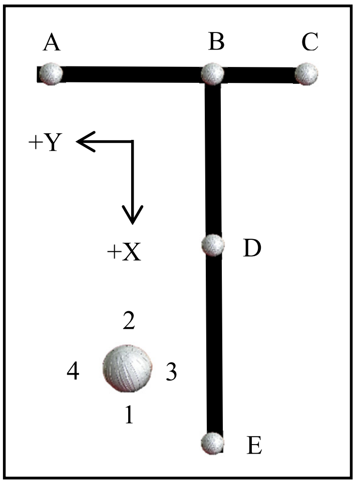
Coordinates for the maker edges measured using the milling machine.

	X	Y	Z	Diameter X	Diameter Y	Average Diameter
A1	0	0	0	8.275	8.215	8.245
A2	-16.55	0				
A3	-8.51	-8.16				
A4	-8.51	8.27				
B1	-0.52	-160.03	1.5	8.19	8.18	8.185
B2	-16.9	-160.03				
B3	-8.51	-168.39				
B4	-8.51	-152.03				
C1	-0.23	-240.03	1.32	8.025	8.115	8.07
C2	-16.28	-240.03				
C3	-8.51	-248.16				
C4	-8.51	-231.93				
D1	118.75	-160.05	0.9	7.865	8.205	8.035
D2	103.02	-160.07				
D3	110.52	-167.82				
D4	110.52	-151.41				
E1	239.34	-159.62	1	8.12	8.22	8.17
E2	223.1	-159.62				
E3	231.09	-168.21				
E4	231.48	-151.77				

Coordinate data for marker centroids.

	X	Y	Z
A	-8.31	0.09	0.00
B	-8.72	-160.21	1.50
C	-8.21	-240.09	1.32
D	111.06	-159.79	0.90
E	231.27	-160.04	1.00

Marker Names.



Conventional Human Movement Analysis Laboratory

				AB				AC			
				Real Mean	Measured Mean	St. Dev.	Accuracy	Real Mean	Measured Mean	St. Dev.	Accuracy
Convention Laboratory	Static	Ankle	1	160.30	160.54	0.01	**99.85**	240.18	240.01	0.02	**99.93**
			2	160.30	160.49	0.01	**99.88**	240.18	240.10	0.01	**99.97**
			3	160.30	160.50	0.01	**99.87**	240.18	240.18	0.01	**100.00**
			4	160.30	160.46	0.01	**99.90**	240.18	240.20	0.01	**99.99**
			5	160.30	160.47	0.01	**99.89**	240.18	240.20	0.01	**99.99**
			6	160.30	160.44	0.01	**99.91**	240.18	240.19	0.01	**100.00**
			7	160.30	160.47	0.02	**99.89**	240.18	240.16	0.02	**99.99**
		Knee	1	160.30	160.52	0.01	**99.86**	240.18	239.98	0.02	**99.92**
			2	160.30	160.47	0.01	**99.89**	240.18	240.11	0.01	**99.97**
			3	160.30	160.45	0.01	**99.90**	240.18	240.15	0.01	**99.99**
			4	160.30	160.44	0.01	**99.91**	240.18	240.16	0.01	**99.99**
			5	160.30	160.43	0.01	**99.92**	240.18	240.18	0.01	**100.00**
			6	160.30	160.39	0.01	**99.94**	240.18	240.11	0.01	**99.97**
			7	160.30	160.39	0.02	**99.94**	240.18	240.06	0.02	**99.95**
		Hip	1	160.30	160.49	0.01	**99.88**	240.18	239.88	0.01	**99.87**
			2	160.30	160.48	0.01	**99.89**	240.18	240.12	0.01	**99.98**
			3	160.30	160.49	0.01	**99.88**	240.18	240.19	0.01	**100.00**
			4	160.30	160.47	0.01	**99.89**	240.18	240.20	0.01	**99.99**
			5	160.30	160.44	0.01	**99.91**	240.18	240.18	0.01	**100.00**
			6	160.30	160.38	0.01	**99.95**	240.18	240.07	0.01	**99.95**
			7	160.30	160.33	0.03	**99.98**	240.18	239.90	0.05	**99.88**
	Dynamic	Slow		160.30	160.35	0.28	**99.97**	240.18	239.97	0.34	**99.91**
		Medium		160.30	160.38	0.24	**99.95**	240.18	240.02	0.24	**99.94**
		Fast		160.30	160.40	0.24	**99.94**	240.18	240.08	0.24	**99.96**

				AD				AE			
				Real Mean	Measured Mean	St. Dev.	Accuracy	Real Mean	Measured Mean	St. Dev.	Accuracy
Convention Laboratory	Static	Ankle	1	199.51	199.04	0.01	**99.76**	288.16	287.60	0.05	**99.80**
			2	199.51	199.09	0.01	**99.79**	288.16	287.68	0.01	**99.83**
			3	199.51	199.18	0.01	**99.83**	288.16	287.74	0.01	**99.85**
			4	199.51	199.19	0.01	**99.84**	288.16	287.76	0.01	**99.86**
			5	199.51	199.23	0.01	**99.86**	288.16	287.79	0.01	**99.87**
			6	199.51	199.22	0.01	**99.85**	288.16	287.73	0.01	**99.85**
			7	199.51	199.25	0.04	**99.87**	288.16	287.81	0.06	**99.88**
		Knee	1	199.51	199.01	0.03	**99.75**	288.16	287.52	0.03	**99.78**
			2	199.51	199.11	0.01	**99.80**	288.16	287.73	0.01	**99.85**
			3	199.51	199.17	0.01	**99.83**	288.16	287.75	0.01	**99.86**
			4	199.51	199.20	0.01	**99.84**	288.16	287.79	0.01	**99.87**
			5	199.51	199.22	0.01	**99.85**	288.16	287.80	0.01	**99.87**
			6	199.51	199.27	0.01	**99.88**	288.16	287.77	0.01	**99.86**
			7	199.51	199.38	0.04	**99.93**	288.16	287.91	0.06	**99.91**
		Hip	1	199.51	199.26	0.01	**99.87**	288.16	287.61	0.01	**99.81**
			2	199.51	199.18	0.01	**99.83**	288.16	287.80	0.01	**99.87**
			3	199.51	199.25	0.01	**99.87**	288.16	287.82	0.01	**99.88**
			4	199.51	199.24	0.01	**99.86**	288.16	287.80	0.01	**99.87**
			5	199.51	199.23	0.01	**99.86**	288.16	287.79	0.01	**99.87**
			6	199.51	199.25	0.01	**99.87**	288.16	287.74	0.01	**99.85**
			7	199.51	199.34	0.06	**99.91**	288.16	287.95	0.06	**99.93**
	Dynamic	Slow		199.51	199.18	0.42	**99.83**	288.16	287.89	0.61	**99.90**
		Medium		199.51	199.21	0.42	**99.85**	288.16	287.83	0.51	**99.88**
		Fast		199.51	199.25	0.49	**99.87**	288.16	287.83	0.56	**99.89**

				BC				BD			
				Real Mean	Measured Mean	St. Dev.	Accuracy	Real Mean	Measured Mean	St. Dev.	Accuracy
Convention Laboratory	Static	Ankle	1	79.89	79.47	0.02	**99.48**	119.77	119.47	0.01	**99.75**
			2	79.89	79.62	0.01	**99.67**	119.77	119.55	0.01	**99.82**
			3	79.89	79.68	0.01	**99.74**	119.77	119.58	0.01	**99.84**
			4	79.89	79.74	0.01	**99.81**	119.77	119.59	0.00	**99.84**
			5	79.89	79.73	0.01	**99.80**	119.77	119.63	0.01	**99.88**
			6	79.89	79.75	0.01	**99.83**	119.77	119.76	0.01	**99.99**
			7	79.89	79.69	0.02	**99.76**	119.77	119.73	0.05	**99.97**
		Knee	1	79.89	79.46	0.02	**99.46**	119.77	119.50	0.02	**99.78**
			2	79.89	79.65	0.01	**99.70**	119.77	119.55	0.00	**99.81**
			3	79.89	79.70	0.01	**99.77**	119.77	119.59	0.01	**99.85**
			4	79.89	79.72	0.01	**99.80**	119.77	119.63	0.01	**99.88**
			5	79.89	79.76	0.01	**99.84**	119.77	119.68	0.01	**99.92**
			6	79.89	79.72	0.01	**99.79**	119.77	119.78	0.01	**100.00**
			7	79.89	79.66	0.02	**99.72**	119.77	119.90	0.02	**99.89**
		Hip	1	79.89	79.39	0.01	**99.38**	119.77	119.67	0.01	**99.91**
			2	79.89	79.65	0.01	**99.71**	119.77	119.62	0.01	**99.87**
			3	79.89	79.70	0.01	**99.77**	119.77	119.64	0.01	**99.89**
			4	79.89	79.73	0.01	**99.80**	119.77	119.65	0.00	**99.90**
			5	79.89	79.74	0.01	**99.82**	119.77	119.70	0.00	**99.94**
			6	79.89	79.69	0.01	**99.75**	119.77	119.77	0.01	**100.00**
			7	79.89	79.57	0.04	**99.61**	119.77	119.83	0.01	**99.95**
	Dynamic	Slow		79.89	79.62	0.25	**99.66**	119.77	119.76	0.44	**99.99**
		Medium		79.89	79.65	0.20	**99.70**	119.77	119.71	0.28	**99.95**
		Fast		79.89	79.69	0.23	**99.75**	119.77	119.75	0.36	**99.98**

				BE				CD			
				Real Mean	Measured Mean	St. Dev.	Accuracy	Real Mean	Measured Mean	St. Dev.	Accuracy
Convention Laboratory	Static	Ankle	1	239.99	239.49	0.04	**99.79**	143.78	143.63	0.02	**99.90**
			2	239.99	239.69	0.01	**99.88**	143.78	143.69	0.01	**99.94**
			3	239.99	239.67	0.01	**99.87**	143.78	143.75	0.01	**99.98**
			4	239.99	239.68	0.01	**99.87**	143.78	143.78	0.01	**100.00**
			5	239.99	239.70	0.00	**99.88**	143.78	143.86	0.01	**99.95**
			6	239.99	239.70	0.01	**99.88**	143.78	143.99	0.01	**99.85**
			7	239.99	239.78	0.06	**99.91**	143.78	143.99	0.03	**99.85**
		Knee	1	239.99	239.51	0.02	**99.80**	143.78	143.71	0.02	**99.95**
			2	239.99	239.72	0.01	**99.89**	143.78	143.73	0.01	**99.97**
			3	239.99	239.72	0.01	**99.89**	143.78	143.76	0.01	**99.99**
			4	239.99	239.74	0.01	**99.90**	143.78	143.82	0.01	**99.98**
			5	239.99	239.71	0.01	**99.88**	143.78	143.92	0.01	**99.90**
			6	239.99	239.74	0.01	**99.90**	143.78	143.97	0.01	**99.87**
			7	239.99	239.89	0.02	**99.96**	143.78	144.07	0.03	**99.80**
		Hip	1	239.99	239.60	0.01	**99.84**	143.78	143.69	0.01	**99.94**
			2	239.99	239.78	0.01	**99.91**	143.78	143.77	0.01	**99.99**
			3	239.99	239.74	0.01	**99.90**	143.78	143.79	0.01	**99.99**
			4	239.99	239.71	0.00	**99.89**	143.78	143.87	0.01	**99.94**
			5	239.99	239.71	0.01	**99.89**	143.78	143.97	0.01	**99.87**
			6	239.99	239.71	0.01	**99.88**	143.78	144.00	0.01	**99.85**
			7	239.99	239.88	0.01	**99.96**	143.78	143.91	0.03	**99.91**
	Dynamic	Slow		239.99	239.83	0.52	**99.93**	143.78	143.84	0.51	**99.96**
		Medium		239.99	239.79	0.34	**99.92**	143.78	143.78	0.35	**100.00**
		Fast		239.99	239.76	0.40	**99.91**	143.78	143.82	0.42	**99.97**

				CE				DE				Mean	
				Real Mean	Measured Mean	St. Dev.	Accuracy	Real Mean	Measured Mean	St. Dev.	Accuracy	**Accuracy**	**St. Dev.**
Convention Laboratory	Static	Ankle	1	252.51	252.06	0.03	**99.82**	120.22	120.02	0.04	**99.84**	**99.79**	**0.02**
			2	252.51	252.25	0.01	**99.90**	120.22	120.14	0.01	**99.94**	**99.86**	**0.01**
			3	252.51	252.27	0.01	**99.91**	120.22	120.09	0.01	**99.90**	**99.88**	**0.01**
			4	252.51	252.31	0.01	**99.92**	120.22	120.10	0.01	**99.90**	**99.89**	**0.01**
			5	252.51	252.37	0.01	**99.94**	120.22	120.07	0.01	**99.88**	**99.89**	**0.01**
			6	252.51	252.37	0.01	**99.95**	120.22	119.94	0.01	**99.78**	**99.89**	**0.01**
			7	252.51	252.50	0.05	**100.00**	120.22	120.05	0.07	**99.86**	**99.90**	**0.04**
		Knee	1	252.51	252.16	0.02	**99.86**	120.22	120.01	0.03	**99.83**	**99.80**	**0.02**
			2	252.51	252.34	0.01	**99.93**	120.22	120.17	0.01	**99.97**	**99.88**	**0.01**
			3	252.51	252.34	0.01	**99.93**	120.22	120.13	0.01	**99.93**	**99.89**	**0.01**
			4	252.51	252.38	0.01	**99.95**	120.22	120.12	0.01	**99.92**	**99.90**	**0.01**
			5	252.51	252.38	0.01	**99.95**	120.22	120.03	0.01	**99.85**	**99.90**	**0.01**
			6	252.51	252.42	0.01	**99.97**	120.22	119.97	0.01	**99.79**	**99.90**	**0.01**
			7	252.51	252.58	0.02	**99.97**	120.22	119.99	0.02	**99.81**	**99.89**	**0.03**
		Hip	1	252.51	252.20	0.01	**99.88**	120.22	119.93	0.01	**99.76**	**99.81**	**0.01**
			2	252.51	252.38	0.01	**99.95**	120.22	120.16	0.01	**99.95**	**99.90**	**0.01**
			3	252.51	252.35	0.01	**99.94**	120.22	120.11	0.01	**99.91**	**99.90**	**0.01**
			4	252.51	252.38	0.01	**99.95**	120.22	120.07	0.01	**99.88**	**99.90**	**0.01**
			5	252.51	252.42	0.01	**99.97**	120.22	120.02	0.01	**99.83**	**99.90**	**0.01**
			6	252.51	252.44	0.01	**99.97**	120.22	119.94	0.01	**99.77**	**99.88**	**0.01**
			7	252.51	252.49	0.02	**99.99**	120.22	120.05	0.01	**99.86**	**99.90**	**0.03**
	Dynamic	Slow		252.51	252.32	0.62	**99.93**	120.22	120.07	0.30	**99.88**	**99.90**	**0.43**
		Medium		252.51	252.30	0.43	**99.92**	120.22	120.08	0.22	**99.88**	**99.90**	**0.32**
		Fast		252.51	252.27	0.52	**99.91**	120.22	120.02	0.30	**99.83**	**99.90**	**0.38**

Motek Medical CAREN Extended System Laboratory Data

				AB				AC			
				Real Mean	Measured Mean	St. Dev.	Accuracy	Real Mean	Measured Mean	St. Dev.	Accuracy
Convention Laboratory	Static	Ankle	1	160.30	160.99	0.01	**99.57**	240.18	240.88	0.01	**99.71**
			2	160.30	160.96	0.01	**99.59**	240.18	240.79	0.01	**99.75**
			3	160.30	161.05	0.03	**99.53**	240.18	241.00	0.01	**99.66**
		Knee	1	160.30	160.84	0.01	**99.66**	240.18	240.86	0.01	**99.72**
			2	160.30	161.14	0.02	**99.47**	240.18	241.04	0.01	**99.64**
			3	160.30	161.21	0.07	**99.43**	240.18	241.20	0.03	**99.58**
		Hip	1	160.30	161.02	0.01	**99.55**	240.18	240.96	0.01	**99.67**
			2	160.30	161.06	0.01	**99.52**	240.18	240.95	0.02	**99.68**
			3	160.30	161.28	0.03	**99.39**	240.18	241.11	0.03	**99.61**
	Dynamic	Slow		160.30	161.04	0.32	**99.53**	240.18	241.04	0.26	**99.64**
		Medium		160.30	160.99	0.27	**99.57**	240.18	241.00	0.29	**99.66**
		Fast		160.30	160.95	0.29	**99.59**	240.18	240.94	0.34	**99.68**

				AD				AE			
				Real Mean	Measured Mean	St. Dev.	Accuracy	Real Mean	Measured Mean	St. Dev.	Accuracy
Convention Laboratory	Static	Ankle	1	199.51	199.90	0.01	**99.81**	288.16	289.18	0.02	**99.64**
			2	199.51	200.16	0.02	**99.67**	288.16	289.41	0.03	**99.57**
			3	199.51	200.25	0.05	**99.63**	288.16	289.90	0.06	**99.40**
		Knee	1	199.51	199.56	0.03	**99.98**	288.16	288.93	0.02	**99.73**
			2	199.51	200.28	0.02	**99.62**	288.16	288.94	0.04	**99.73**
			3	199.51	200.26	0.02	**99.63**	288.16	289.26	0.11	**99.62**
		Hip	1	199.51	200.10	0.03	**99.71**	288.16	288.80	0.01	**99.78**
			2	199.51	200.12	0.04	**99.70**	288.16	288.88	0.02	**99.75**
			3	199.51	200.01	0.03	**99.75**	288.16	288.97	0.03	**99.72**
	Dynamic	Slow		199.51	200.04	0.60	**99.74**	288.16	288.93	0.66	**99.73**
		Medium		199.51	199.97	0.39	**99.77**	288.16	288.85	0.39	**99.76**
		Fast		199.51	199.97	0.48	**99.77**	288.16	288.91	0.52	**99.74**

				BC				BD			
				Real Mean	Measured Mean	St. Dev.	Accuracy	Real Mean	Measured Mean	St. Dev.	Accuracy
Convention Laboratory	Static	Ankle	1	79.89	79.89	0.01	**100.00**	119.77	119.68	0.02	**99.92**
			2	79.89	79.83	0.01	**99.93**	119.77	120.44	0.02	**99.44**
			3	79.89	79.96	0.03	**99.91**	119.77	120.52	0.07	**99.37**
		Knee	1	79.89	80.02	0.01	**99.83**	119.77	119.32	0.03	**99.62**
			2	79.89	79.90	0.02	**99.98**	119.77	120.04	0.03	**99.77**
			3	79.89	79.99	0.06	**99.88**	119.77	120.34	0.04	**99.53**
		Hip	1	79.89	79.94	0.01	**99.93**	119.77	119.54	0.03	**99.80**
			2	79.89	79.89	0.02	**100.00**	119.77	120.15	0.04	**99.68**
			3	79.89	79.83	0.01	**99.93**	119.77	120.16	0.02	**99.68**
	Dynamic	Slow		79.89	80.00	0.31	**99.85**	119.77	120.19	0.35	**99.65**
		Medium		79.89	80.01	0.30	**99.84**	119.77	120.17	0.36	**99.67**
		Fast		79.89	79.99	0.31	**99.87**	119.77	120.17	0.55	**99.67**

				BE				CD			
				Real Mean	Measured Mean	St. Dev.	Accuracy	Real Mean	Measured Mean	St. Dev.	Accuracy
Convention Laboratory	Static	Ankle	1	239.99	240.64	0.02	**99.73**	143.78	144.01	0.01	**99.84**
			2	239.99	241.27	0.03	**99.46**	143.78	144.36	0.02	**99.60**
			3	239.99	241.67	0.08	**99.30**	143.78	144.48	0.05	**99.52**
		Knee	1	239.99	240.58	0.02	**99.75**	143.78	143.39	0.02	**99.73**
			2	239.99	240.32	0.05	**99.86**	143.78	144.05	0.02	**99.81**
			3	239.99	240.86	0.12	**99.63**	143.78	144.48	0.02	**99.51**
		Hip	1	239.99	239.99	0.02	**100.00**	143.78	143.41	0.02	**99.74**
			2	239.99	240.32	0.05	**99.86**	143.78	144.17	0.05	**99.73**
			3	239.99	240.67	0.03	**99.71**	143.78	144.41	0.02	**99.57**
	Dynamic	Slow		239.99	240.72	0.37	**99.69**	143.78	144.35	0.44	**99.61**
		Medium		239.99	240.76	0.35	**99.68**	143.78	144.37	0.41	**99.59**
		Fast		239.99	240.76	0.55	**99.68**	143.78	144.33	0.50	**99.62**

				CE				DE				Mean	
				Real Mean	Measured Mean	St. Dev.	Accuracy	Real Mean	Measured Mean	St. Dev.	Accuracy	Accuracy	St. Dev.
Convention Laboratory	Static	Ankle	1	252.51	253.33	0.02	**99.67**	120.22	120.97	0.02	**99.38**	**99.73**	**0.02**
			2	252.51	253.66	0.04	**99.54**	120.22	120.83	0.04	**99.49**	**99.60**	**0.02**
			3	252.51	253.96	0.07	**99.42**	120.22	121.15	0.09	**99.22**	**99.50**	**0.05**
		Knee	1	252.51	252.92	0.02	**99.83**	120.22	121.27	0.03	**99.12**	**99.70**	**0.02**
			2	252.51	252.79	0.05	**99.89**	120.22	120.28	0.05	**99.95**	**99.77**	**0.03**
			3	252.51	253.39	0.10	**99.65**	120.22	120.53	0.12	**99.74**	**99.62**	**0.07**
		Hip	1	252.51	252.27	0.02	**99.91**	120.22	120.46	0.03	**99.80**	**99.79**	**0.02**
			2	252.51	252.70	0.07	**99.92**	120.22	120.18	0.06	**99.97**	**99.78**	**0.04**
			3	252.51	253.28	0.03	**99.69**	120.22	120.52	0.03	**99.75**	**99.68**	**0.03**
	Dynamic	Slow		252.51	253.30	0.42	**99.68**	120.22	120.53	0.19	**99.74**	**99.69**	**0.39**
		Medium		252.51	253.41	0.45	**99.64**	120.22	120.59	0.21	**99.69**	**99.69**	**0.34**
		Fast		252.51	253.36	0.55	**99.66**	120.22	120.59	0.21	**99.69**	**99.70**	**0.43**