THE INFLUENCE OF BODY MASS ON GAIT PARAMETERS

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DECLARATION

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SUMMARY

This research assessed the relationship between increased body mass and gait. Previous literature has noted that a reduced velocity is adopted by overweight participants. Velocity has been shown to have a confounding influence on gait. It is difficult to determine which changes reported are due to an increase in body mass, and not the result of a reduction in velocity. The author's therefore also assessed the influence of auditory cueing on gait, as a means of constraining velocity. This enabled the direct assessment of the relationship between body mass and gait.

For both adults and children the method of auditory cueing to constrain cadence (and in turn velocity) did not alter gait parameters beyond that anticipated for natural variation in gait. When auditory cueing was used to dictate a cadence, an altered presentation in gait was noted for children but not adults. This suggests that the imposition of a cadence altered the gait presentation of children but not of adults. It may be recommended that in the research setting, an auditory cue be used to dictate velocity in healthy adults control groups. This does not apply to healthy children.

When investigating the influence of body mass on gait parameters in adults few significant differences were noted for an increase in BMI. No significant changes were seen for three dimensional ankle joint moments, or pelvis and ankle joint range. At the hip an increase in flexion and decreased abductor moment was noted at initial contact for adults who were overweight. In the frontal plane, at terminal stance and during swing phase an increase in hip abduction, knee flexion, varus and internal tibial rotation were noted for those with excess body mass. Three dimensional absolute maximum ground reaction forces were significantly greater for adults who were overweight. Once corrected by body mass, maximum vertical ground reaction force was significantly reduced for adults with excess body mass. Finally a significant decrease in maximum total hip absorption was noted for adults who were overweight, indicating a reduction in eccentric muscle activity for this group.

The analysis of the influence of body mass on gait parameters in children was not independent of velocity. Children who were overweight presented with a more flexed attitude with a significant increase in hip and knee flexion noted. An increase in hip abduction and ankle supination was also seen for the overweight group. In the transverse plane increased external rotation at the hip and ankle, and internal rotation at the knee was recorded for children who were overweight. Associated changes in joint moments were also noted. Absolute maximum vertical ground reaction forces were significantly higher for those with excess body mass. Once scaled for body mass, three dimensional maximum ground reaction force values were significantly smaller for overweight children. Maximum total ankle power generation was increased, whereas total hip power absorption was decreased for children who were overweight.

This research has demonstrated that for adults who were overweight, biomechanics independent of velocity during gait were altered. For children, alterations were also seen, however these were not independent of velocity.

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LIST OF FIGURES

LIST OF TABLES

LIST OF APPENDICES

LIST OF PUBLICAITIONS FROM THE WORK IN THIS THESIS

Published Papers

Sheehan K, Gormley J. Gait and increased body weight (potential implications for musculoskeletal disease). *Physical Therapy Reviews*2012; 17(2):91-8.

Published abstracts

Sheehan K, Roche E F, Gormley J. Ground reaction forces in overweight children 2010. Proceedings of the 20th Annual European Childhood Obesity Group, Brussels, November 2010.

Sheehan K, Gormley J. The influence of auditory cueing on sagittal kinematics in healthy adults 2011. Proceedings of the 20th Annual European Society of Movement Analysis for Adults and Children, Vienna, 2011.

Sheehan K, Roche E F, Gormley J. The influence of weight on internal joint moments in children 2011. Proceedings of the 4th School of Medicine Postgraduate Research Day Trinity College Dublin, Dublin, 2011.

LIST OF ABBREVIATIONS

WHO	World Health Organisation
CDC	The United States Centre for Disease Control and Prevention
BMI	Body Mass Index
IOTF	International Obesity Task Force
GRF	Ground reaction force
ASIS	Anterior superior iliac spine
LED	Light emitting diode
CODA	Cartesian optoelectronic dynamic anthropometer
RMS	Root mean square
SS	Self selected
AMNCH	The Adelaide and Meath Hospital, Dublin Incorporating the National Children's Hospital
AMTI	Advanced mechanical technology Inc
AD	Analogue to digital
PSIS	Posterior superior iliac spine
SS2	Self selected trial two
AC	Auditory cue

TABLE OF CONTENTS

CHAPTER 1: INTRODUCTION

1.1 INTRODUCTION TO THE THESIS

The prevalence of overweight and obesity in Ireland is increasing (SLAN 2007; NANS 2011). To combat this trend weight loss is advocated (Hooper et al 2007). Weight loss is achieved with dietary modifications and increased physical activity (Donnelly et al 2009). Walking is often recommended as a method of increasing physical activity levels in populations that are overweight. However, walking has been shown to induce pain in obese individuals (Mattsson et al 1997; Hulens et al 2003). It is necessary to investigate any alterations in the gait cycle which may predispose individuals who are overweight to pain. Should alterations be present, further investigation into appropriate exercise prescription for populations that are overweight may be warranted.

The relationship between adiposity and altered gait has received relatively little attention. With advanced motion capture systems the assessment of gait in individuals who are overweight is more readily examined. Changes in gait parameters have been noted as a result of velocity (Schwartz et al 2008). It is therefore difficult to differentiate the effects of increased body mass from the effects of velocity on gait. It is necessary to account for velocity in either the experimental design stage, or the data analysis stage, to enable a more direct assessment of the influence of body mass on gait.

The present chapter acts as an introduction to this thesis. Overweight and obesity are defined and their prevalence and health costs are discussed. The terminology used in gait analysis is provided. A description of normal gait and the influence of velocity on gait are discussed. The chapter concludes with a history of gait analysis and an overview of modern gait analysis. A review of the existing literature assessing the influence of body mass on gait parameters is provided in Chapter two. Chapter three outlines the methodology adopted in this research. Research design, recruitment, inclusion and exclusion criteria, measures and protocols are discussed. Chapters four and five provide the results of research assessing the effect of an external auditory cue on gait parameters in healthy participants. Results of the assessment of gait in overweight adults and children are discussed in Chapters six and seven. Chapter eight provides a discussion of the research as a whole, critical analysis of this thesis, recommendations as a result of this piece of work, and final conclusions.

1.2 OVERWEIGHT AND OBESITY

Overweight and obesity are defined as abnormal or excessive fat accumulation that may impair health (WHO 2011). The United States Centre for Disease Control and Prevention (CDC) describe overweight and obesity as labels for ranges of weight that are greater than what is generally considered healthy for a given height (CDC 2010).

The aetiology of overweight and obesity involves an energy imbalance between energy intake (consumption of food) and energy expended (physical activity levels). Genetics, environment, disease and medication have also been reported as contributing factors in the development of overweight and obesity (CDC 2011).

1.2.1 Defining overweight and obesity in adults

Body Mass Index (BMI) is the commonly used index for classifying underweight, overweight and obesity in adults. BMI is calculated as weight divided by height squared (kg/m^2). For example, an adult who weighs 80kg and whose height is 1.85m will have a BMI of 23.4 ($80/1.85^2$). P-value BMI the CDC, International Obesity Task Force (IOTF) and World Health Organisation (WHO) all define overweight in Caucasian adults as $\geq$ 25 kg/m^2 and obesity in Caucasian adults is defined as $\geq$ 30 kg/m^2.

1.2.2 Defining overweight and obesity in children

To date there is no gold standard for determining overweight and obesity in children and adolescents. BMI has been increasingly acknowledged as an acceptable indirect measure of overweight and obesity in young people (Lobstein 2004). Gender and age have been shown to have a significant effect on body composition in children (Rolland-Cachera et al 1982). The BMI cut-off points for adults do not take gender and age into account and therefore cannot be used in children.

Growth reference charts have been published by the CDC, IOTF and, WHO (CDC 2010b; Cole et al 2000; WHO 2011b). These charts permit a given BMI to be evaluated against age and gender specific reference values. The cut-off points are describes in terms of percentiles or standard deviation scores (Z-score). A percentile is the value of a variable below which a certain percent of observations fall. BMI Z-scores represent the number of standard deviations that an individual's BMI varies from the mean. A Z-score of 0 is equivalent to the median or 50th percentile; a z-score of +2.00 is approximately equivalent to the 98th percentile and a Z-score of +2.85 is > 99th percentile (Lobstein 2004). A BMI Z-score of >+1.00 from the reference mean is indicative of overweight, and a BMI Z-score of >+2.00 from the reference mean is indicative of obesity (WHO 2007). These values correspond to the 85th and 98th percentile respectively.

According to the WHO website, the WHO cut-offs overestimate overweight and obesity when compared to both IOTF and the CDC cut-off points. The IOTF and CDC demonstrate similar cut-off points for overweight and obesity. The CDC and WHO reference cut-off points are based on United States reference populations whereas the IOTF cut-offs are based on internationally pooled data. As a result, the IOTF cut-off points were selected for this research (Figure 1.1).

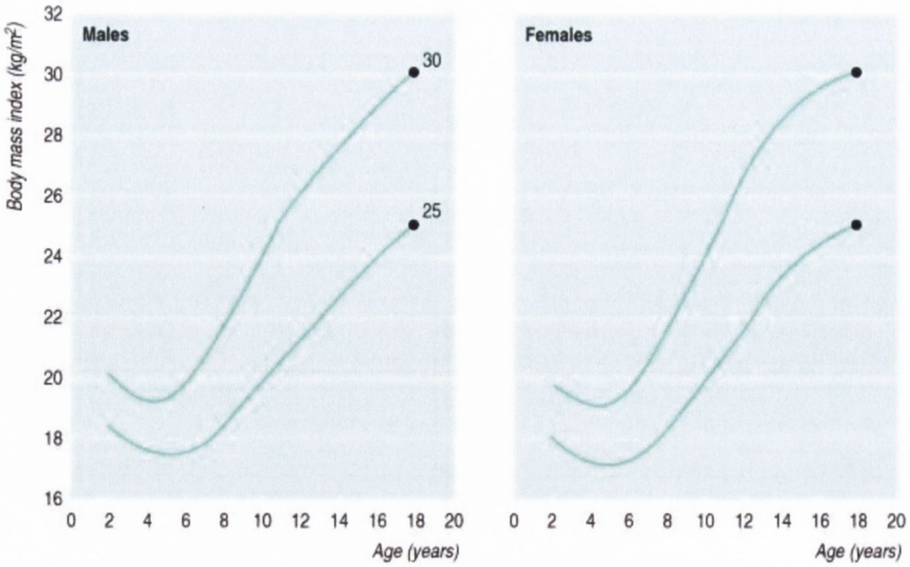


Figure 1.1: Cut off points for body mass index by sex for overweight and obesity, passing through body mass index 25 and 30 kg/m^2 at age 18. ***Cole et al (2000)***

1.2.3 Cost, prevalence and trends in overweight and obesity

In Ireland annually approximately 2,000 premature deaths were attributed to obesity, at an estimated annual cost of €0.4 billion (National Taskforce on Obesity 2005). Indirect costs (absenteeism, presenteeism etc) accounted for €0.37 billion of this total cost (National Taskforce on Obesity 2005). Coping with stigma and discrimination, depression and a reduced quality of life are all intangible costs that have been shown to be associated with obesity (National Taskforce on Obesity 2005).

Prevalence is defined as the number of individuals with a certain disease at a specified time divided by the number of individuals in the population at that time (Sahai & Khurshid 1996). In 2005 approximately 1.6 billion adults were overweight and 400 million were obese worldwide (WHO 2007). In a recent survey in which the prevalence of adult excess body mass in Ireland was assessed, 37% of respondents were found to be overweight and 24% obese (NANS 2011). When compared to females, a higher percentage of males were found to be overweight and obese (NANS 2011). This survey suggests a steady increase in overweight and obesity from earlier surveys (SLAN 1998; SLAN 2002; SLAN 2007).

In 2007 overweight population proportions in Europe ranged from 13% to 32% for primary school-age children (WHO 2007). For older children European overweight population proportions ranged from 14%-34% (WHO 2007). To date the largest epidemiological study investigating the prevalence of childhood overweight and obesity in Ireland was conducted in 2007 by Whelton et al. Of the 19,617 participants aged four-16 years, 23% of boys and 28% of girls were found to be either overweight or obese (Whelton et al 2007). Overweight was found to be the most common in 13 year old girls and obesity in seven year old girls (Whelton et al 2007).

1.2.4 Health consequences of overweight and obesity

The health implications of increased body mass can be divided into different groups that range from non-fatal complaints which can impact on the quality of daily life, to problems which can increase the risk of premature death. In Ireland 58% of type II diabetes mellitus, 21% of heart disease, and 8-42% of various cancers were due to excess body mass (National Taskforce on Obesity 2005). Overweight has been shown to be associated with an increased risk of coronary heart disease, congestive heart failure, type II diabetes mellitus, insulin resistance, cancer, hypertension, dyslipidemia, stroke, liver and gall bladder disease, gall stones, sleep apnoea, respiratory disease, gynaecological problems, psychological disorders, and musculoskeletal disease (NIH 1998; Thompson et al 1999; Reilly et al 2003; Lobstein et al 2004; Wearing et al 2006).

1.2.5 Barriers to exercise in overweight and obese groups

To tackle the health consequences of overweight and obesity, physical activity and dietary modifications are recommended (Donnelly et al 2009). With reference to physical activity, several barriers have been reported for overweight and obese individuals. A recent qualitative study detailed the self-reported barriers to exercise of 80 obese adults, these included: laziness, fear of failure, and little tenacity, lack of pleasure, previous negative experiences of physical activity, injuries, and difficulty of movement (Piana et al 2012). Pain during walking has been reported for overweight adults (Mattsson et al 1997; Hulens et al 2003). For children malalignment of the lower limbs has been shown to be associated with pain during walking (Taylor et al 2006; Wearing et al 2006). Pain and 'difficulty of movement' may be the result of altered biomechanics during gait. Nantel et al (2006) noted that obese children were mechanically less efficient than healthy weight children during gait. In addition, it has been reported that overweight children have difficulty when adapting to different walking velocities (Hills et al 1991a). Difficulty changing velocity may represent a considerable barrier to sports where quick changes in velocity are required (Nantel et al 2011). Altered biomechanics may represent a barrier to physical activity and therefore weight loss in overweight groups.

1.3 TERMINOLOGY IN GAIT

1.3.1 The gait cycle

The gait cycle describes the time interval between two successive occurrences of one of the repetitive events of walking. The cycle may be divided into two phases; stance and swing phase. Stance phase is determined by limb contact with the floor. It may be divided into loading response, mid stance, terminal stance and pre swing. Stance phase may also be subdivided into single limb support and double limb support. Two periods of each are present in the gait cycle. During swing phase the limb is not in contact with the floor. This phase is divided into initial swing, mid swing and terminal swing (Figure 1.2).

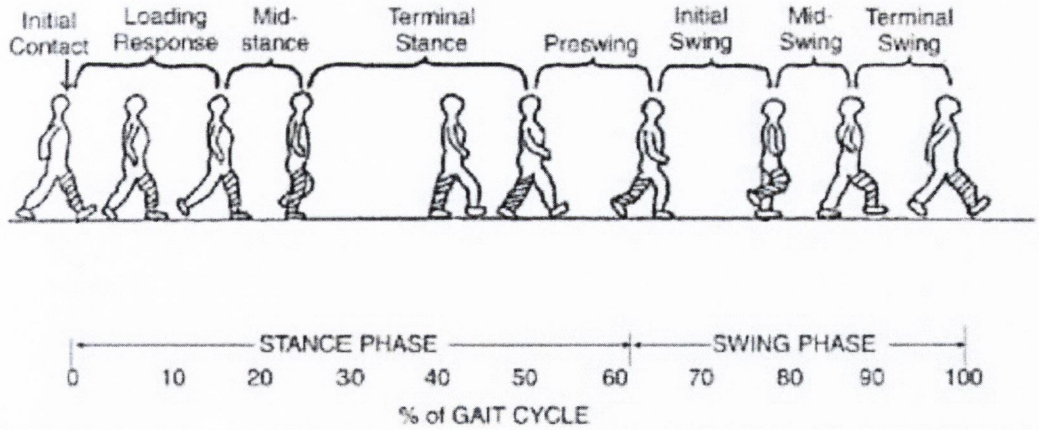


Figure 1.2: Phases of the gait cycle. ***Cuccurullo (2004)***

The following terms are used to identify the events of the gait cycle (Figure 1.3) (Perry 1992):

<u>Initial contact (0%):</u> The initiation of the loading response, the first period of stance phase.

<u>Opposite toe off (10%):</u> End of the first double support phase and the start of mid-stance.

<u>Heel rise (30%):</u> The transition from mid-stance to terminal stance.

<u>Opposite initial contact (50%):</u> Marking the end of the first phase of single support and the initiation of pre-swing, the second phase of double support.

<u>Toe off (60%):</u> Divides pre-swing and initial swing, and is the point at which stance phase ends and the swing phase begins.

<u>Feet adjacent (73%):</u> Separates initial swing from mid-swing. The stance phase leg is passed by the swing phase leg.

Tibia vertical (87%): The division between mid-swing and terminal swing is marked by the tibia of the swing phase leg becoming vertical.

Next initial contact (100%): The completion of one stride and initiation of the ipsilateral next stride.

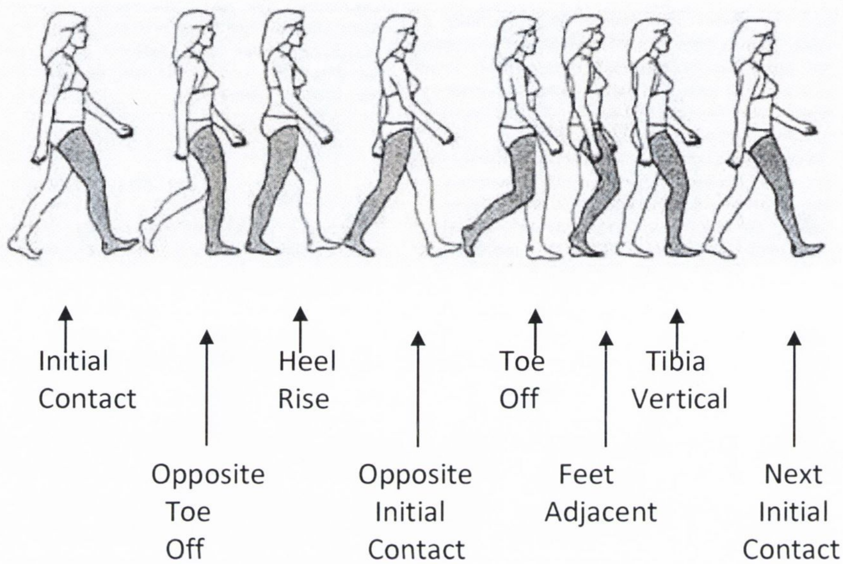


Figure 1.3: Events of the gait cycle.

1.3.2 Temporal and spatial parameters

Temporal and spatial parameters are those gait parameters which are concerned with time and distance. The following are terms used in discussion of temporal and spatial gait parameters:

Stride length: The distance between two successive placements of the same foot (Figure 1.4).

Step length: The distance between successive heel contacts of the two different feet (Figure 1.4).

Base of support/Step width: The lateral distance between the heel centers of two consecutive foot contacts (Figure 1.4).

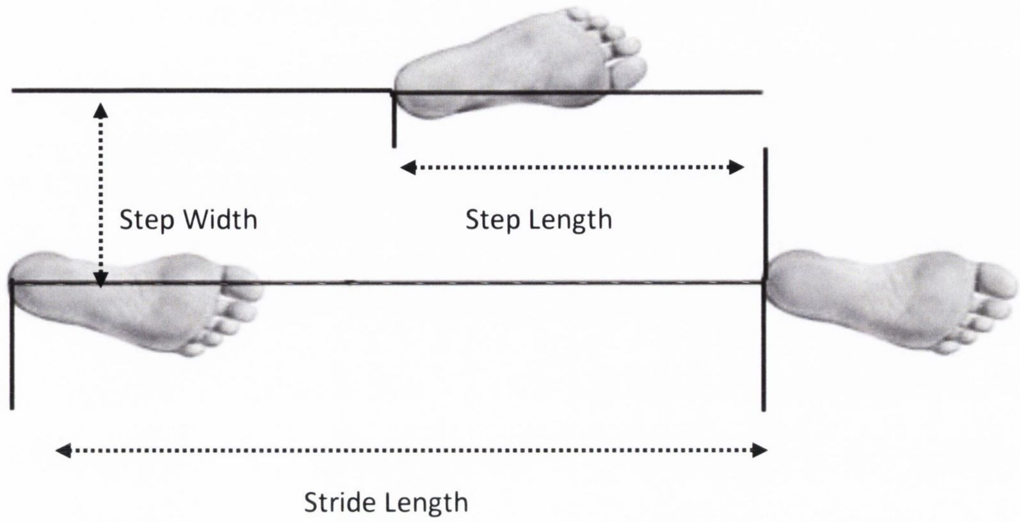


Figure 1.4: Spatial gait parameters.

Velocity: The distance covered by the whole body in a given time.

Cadence/step rate: The number of steps taken per minute. There are two steps in a single gait cycle therefore cadence is a measure of half cycles.

1.3.3 The planes of motion

In biomechanics, motion occurs in three dimension/planes. The sagittal plane, the coronal/frontal plane and the transverse/horizontal plane (Figure 1.5).

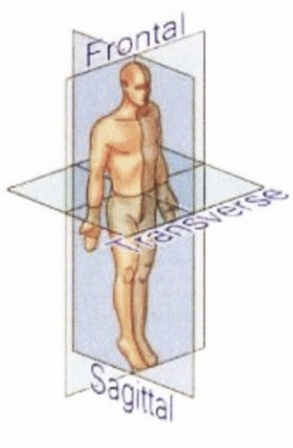


Figure 1.5: The three planes of motion.

The sagittal plane divides the body/a body segment into left and right, perpendicular to the medial-lateral axis.

The coronal/frontal plane divides the body/a body segment into front and back, perpendicular to the anterior-posterior axis.

The transverse/horizontal plane divides the body/a body segment into top and bottom, perpendicular to the inferior-superior axis.

1.3.4 Kinematic parameters

Kinematics refers to the branch of mechanics concerned with motion without reference to force or mass. Joint motion is defined as the movement of one body segment relative to another.

Movements which occur in the sagittal plane include anterior-posterior pelvic tilt, hip flexion-extension, knee flexion-extension and ankle dorsiflexion-plantarflexion. In the frontal plane up-down pelvic obliquity, hip abduction-adduction, knee varus-valgus and ankle pronation-supination occur. Minimal movement is reported to occur at the knee in the frontal plane due to the articular geometry and ligaments (Neumann 2001). Transverse plane movements include forward-backward rotation of the pelvis and internal-external rotation of the thigh and shank.

1.3.5 Kinetic parameters

Kinetics refers to the branch of mechanics concerned with the forces that cause motions of bodies. Kinetics can be subdivided into ground reaction forces (GRF), joint moments and joint powers.

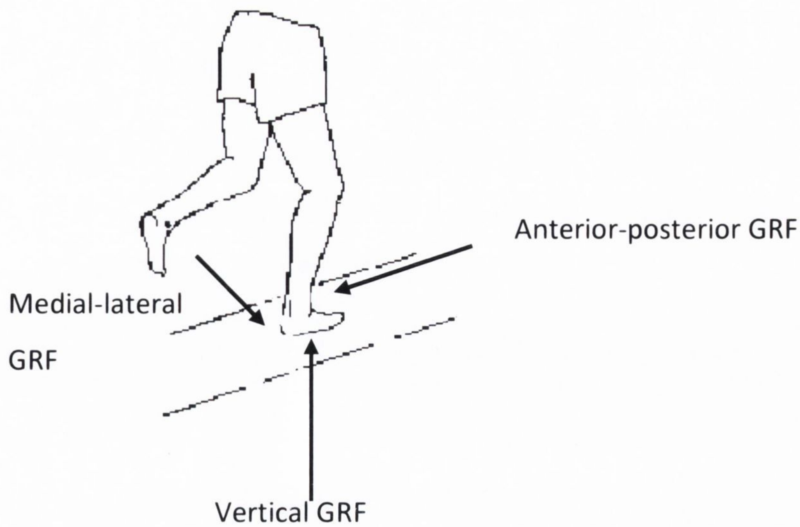


Figure 1.6: Components of ground reaction force.

GRF detail the forces felt by the foot from the ground. Based on Newton's Third Law of Motion, these forces are equal to the force placed on the ground by the foot. The description of GRF follows a Cartesian coordinate system. The forces are presented in terms of their medial-lateral, anterior-posterior and vertical components (Figure 1.6).

The medial-lateral component produces a shear force.I Its magnitude is dependant on the position of the centre of mass and the location of the foot. The anterior-posterior component also produces a shear force; it is applied parallel to the supporting surface and it is greater in magnitude than the medial-lateral component. The vertical component is directed perpendicularly to the supporting surface, and is the largest of the three components.

A moment is the tendency of a force to rotate an object about an axis, fulcrum or pivot (Serway & Jewett 2003). In relation to the body, it is the three dimensional tendency of GRF to rotate joints about their axis or centre of rotation. Joint moments are described in terms of their magnitude, where they act on the body and their direction. The magnitude of a joint moment is dependant on the magnitude of the GRF, and the distance of that force from the axis or centre of rotation (moment = force x perpendicular distance from the fulcrum). Joint moments can be described as internal or external. External joint moments describe the forces acting on the body. Internal moments describe the forces at work within the body to counteract the external forces. Again based on Newton's Third Law of Motion, these moments are equal in magnitude and opposite in direction. An anterior-posterior GRF passing posterior to the knee joint centre will result in an external flexion moment (Figure 1.7). As it is predominantly the quadriceps, the extensors, that counteract this external moment, an internal extensor moment occurs simultaneously (Figure 1.7).

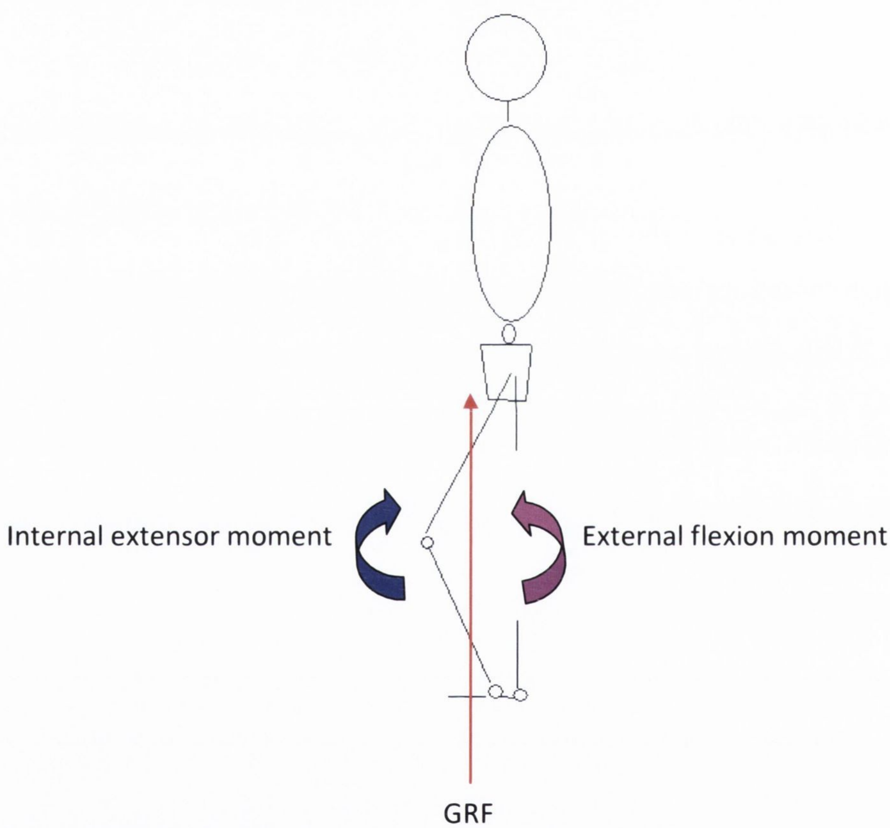


Figure 1.7: Sagittal plane knee joint moments.

Joint power describes the rate of change of energy occurring at a joint (Gordon et al 1980). Power is a scalar quantity described in terms of magnitude and location. It is the rate of change of a joint angle, i.e.

Power = joint moment x angular velocity.

Muscle and other soft tissue structures e.g. ligaments can generate and absorb energy. By calculating power it is possible to identify the type of muscle activity occurring at a joint. Concentric activity, where muscles are actively shortening, generates energy whereas eccentric activity, where muscles are actively lengthening, absorbs energy.

1.4 NORMAL GAIT

Prior to the assessment of pathological gait, it is important to develop an understanding of the standard to which it will be compared i.e. normal gait. Whilst the description of normal gait is readily available, it should be noted that rarely two individuals will walk identically and that a 'range' of normal exists.

1.4.1 Normal gait-kinematics

As previously stated kinematics refers to the branch of mechanics concerned with motion without reference to force or mass. Here the three dimensional motion of the pelvis, hip, knee and ankle during the events of normal gait will be discussed. This description is based on several sources which discussed in detail the presentation of normal kinematic parameters (Murray et al 1964, 1966, 1970; Statham & Murray 1971; Murray 1979; Perry 1992; Messenger 1994; Craik & Oatis 1995; Rose & Gamble 1994; Neumann 2002; Whittle 2002; Nigg & Herzog 2006).

At initial contact the pelvis begins to posterior tilt and counter-clockwise rotation in the frontal plane is initiated. Maximum hip flexion of approximately 30° is present at initial contact, the hip is in an external rotated position at this point. The knee changes from a near fully extended attitude into a position of 5° of flexion, and 3° of relative external rotation. An average of 1.2° of knee abduction is present at the time of initial contact for subjects walking at $1.2ms^{-1}$; this alignment remains unchanged throughout the rest of stance phase (Lafortune et al 1992). The ankle is plantarflexed close to its neutral position (0-5°) at the time of initial contact. The subtalar joint is inverted (2-3°) and the forefoot slightly supinated.

At opposite toe off the pelvis begins to rotate anteriorly and drops on the swinging side. Clockwise rotation of the pelvis is initiated as the opposite anterior superior iliac spine (ASIS) progressively moves forward with the advancing swing leg. After initial contact the hip extends to a position of around 25° of flexion at the time of opposite toe off. The knee continues to flex, reaching the peak of stance phase flexion early in mid-stance after which it begins to extend again. As soon as the foot is flat on the ground, the direction of ankle motion changes from plantarflexion to dorsiflexion. Eversion of the calcaneus continues. Both forefoot pronation and internal tibial rotation reach a peak around opposite toe off, and begin to reverse.

At heel rise the pelvis alters the direction of movement in the sagittal plane as it begins to tilt posteriorly again. The opposite iliac crest is now elevated as the opposite ASIS continues to progress forwards. Hip and knee extension continue until heel rise at which point the knee starts to flex once more. Peak ankle dorsiflexion is reached after heel rise. The tibia becomes increasingly externally rotated and the forefoot becomes increasingly supinated. A relatively neutral position of the calcaneus is reached by heel off, and thereafter an inverted position is adopted.

The pelvis continues to tilt posteriorly, the opposite iliac crest continues to be elevated, and the opposite ASIS continues to progress forwards at opposite initial contact. The hip reaches its most extended position (10-20°) and motion reverses in the direction of flexion. The knee continues to flex and the ankle begins to plantarflex. The forefoot reaches its maximal supination, with coupled external tibial rotation.

The pelvis continues to tilt posteriorly until just after toe off. At toe off the iliac crests are level. Clockwise rotation of the pelvis continues with the opposite ASIS progressing forwards. The hip is in neutral alignment by the time of toe off. A maximal internally rotated position of the hip is reached and external rotation is initiated. The knee is flexed to approximately 35°, and has reached approximately 5° of relative internal rotation at toe off. Peak ankle plantarflexion (15-20°) occurs just after toe off. This enables the ankle to adopt a neutral or dorsiflexed attitude for foot clearance during swing phase. Between heel off and toe off, calcaneal inversion continues until it reaches around 6° of inversion.

At feet adjacent the pelvis tilts anteriorly and the ASIS of the swing leg progressively moves forwards and up. The hip assumes a flexed attitude and return to its neutral position in the frontal plane. During initial swing the knee abducts an additional 5°. Peak knee flexion (60-70°) occurs during mid swing, before the feet are adjacent, by which time the knee has started to extend again. The knee joint externally rotates during swing in preparation for the next heel contact. The ankle moves from a plantarflexed position around toe off, to a neutral or dorsiflexed position at terminal swing. The calcaneus returns to a slightly inverted position. Forefoot supination reduces following toe off, but the forefoot remains slightly supinated until the following initial contact.

When the tibia of the swing leg is vertical to the ground the pelvis begins to posteriorly tilt. Maximum hip flexion of around 30° is reached prior to initial contact. This is achieved by the time tibia vertical occurs and the hip then begins to extend in preparation for initial contact. From mid swing to initial contact a small amount of internal rotation of the hip occurs. The knee moves from peak swing phase flexion at feet adjacent, to just short of full extension prior to the next initial contact. It returns to its slightly abducted position and externally rotates prior to the next initial contact. The ankle lies in a position between a few degrees of plantarflexion and a few degrees of dorsiflexion before initial contact.

1.4.2 Normal gait-kinetics

As stated previously kinetics refers to the branch of mechanics concerned with the forces that cause motions of bodies. It can be divided into GRF, joint moments and joint powers. Hip, knee and ankle kinetics during the events of normal gait will be discussed. This description is based on several sources which discussed in detail the presentation of normal kinetic parameters (Perry 1992; Craik & Oatis 1995; Neumann 2002; Whittle 2002; Nigg & Herzog 2006).

From initial contact to mid-stance the centre of mass is accelerating forwards, the anterior-posterior GRF acts as a braking force and is applied posteriorly. As the stance phase ends and pre-swing begins the acceleration of the centre of mass declines. The GRF changes direction to an anterior and propulsive position. Peak anterior-posterior GRF is approximately 20% of body weight. From initial contact and during the initial 10% of stance phase, laterally directed shear occurs. As the centre of mass takes a medial position after initial stance the subsequent GRF is directed medially. The magnitude of medial-lateral GRF is <5% body weight in normal gait. Vertical GRF is slightly greater than body weight at the start and end of the stance phase as the body's centre of mass moves downwards. During mid-stance it is slightly less than body weight as the body's centre of mass moves upwards.

Anterior-posterior GRF are associated with extensor moments at the hip and knee, and plantarflexor moments at the ankle. From initial contact to 5% of the gait cycle the GRF passes anterior to the hip joint creating a flexion moment. A hip extensor moment is activated to prevent the hip giving way into flexion. This moment accelerates the body forwards and upwards minimising the flexion moment in spite of an increasing vertical load. In mid stance the hip moment reverses from a flexion to an extension moment, with the GRF vector passing posterior to the hip joint. Maximal hip joint extension moment is present at 50% of the gait cycle. The hip is maintained in a stable position through passive action of the anterior hip capsule and ligaments. During pre-swing the extension moment decreases due to unloading of the stance limb.

From initial contact to 5% of the gait cycle the anterior-posterior GRF passes anterior to the knee resulting in an extension moment, this moment stabilises the knee. The GRF then passes posterior to the knee joint and a flexion moment occurs. To prevent buckling of the knee, an extensor moment is generated. By midstance the GRF passes through the knee joint axis and no moments act about the knee. As body mass passes over the supporting foot, the GRF passes anterior to the knee causing an extension moment. At terminal stance the vector begins to travel posteriorly. This extension moment stabilises the knee against the posterior capsule and cruciate ligaments, no internal moments are present at this time. At the end of single support the vector has reached the knee joint axis and once again no moments act about the knee. In pre-swing the vector moves posterior to the knee joint creating a flexion moment.

At initial contact the anterior-posterior GRF is behind the ankle creating a plantarflexion moment. The plantarflexion motion is counteracted by a dorsiflexor moment. The GRF then travels anterior, at 5% of the gait cycle a dorsiflexion moment is created as the vector moves ahead of the ankle joint centre. This moment is controlled by an opposing plantarflexor moment. The dorsiflexion moment continue to increase until terminal stance. With loading of the contra lateral side the magnitude of the vector decreases.

Medial-lateral GRF are associated with hip and knee abductor/adduction moments and pronator/supination moments at the ankle. At the hip a laterally directed vector creates an abduction moment at initial contact. The vector then passes medially towards the hip joint centre resulting in an adduction moment. An abductor moment is required to stabilise the pelvis throughout stance. As per the sagittal plane the loading of the contra lateral side decreases the magnitude of the external moments. The GRF vector passes medially to the knee and ankle in the frontal plane resulting in adduction moments. Stabilisation is achieved by the lateral ligaments of the knee and ankle.

As described earlier, by calculating power it is possible to identify the type of muscle activity occurring at a joint. Due to the concentric action of the hip extensors, followed by the concentric action of the hip flexors, two periods of acceleration occur in the sagittal plane at the hip. One decelerating period occurs at the hip and relates to the stretch of the ligament structure and capsule of the hip, and eccentric action of the hip flexors. At the knee very little power is generated. Instead power is transferred through the knee to the hip or ankle. The first phase of power generation is due to the concentric action of the hamstrings. They provide stability to the knee at initial contact and contribute to hip extension. The second power generation phase occurs during pre-swing with concentric activity of the gastrocnemius. The ligaments of the knee are responsible for power absorption during stance. At the ankle power absorption occurs with eccentric action of the plantarflexors controlling the forward progression of the tibia. Power generation is associated with the concentric work of the plantarflexors accelerating the stance limb into swing.

At the hip in the frontal plane, power absorption during the first half of stance occurs with eccentric action of the abductors. During the second half of stance, power generation occurs with concentric action of the hip abductors. No power is generated at the knee and ankle during normal walking.

1.5 THE INFLUENCE OF VELOCITY ON GAIT

Normal walking velocity for healthy adults ranges from 1.05-1.43 ms^{-1} (Oberg 1993). Pathological gait is often characterised by a reduction in velocity. It is necessary to identify velocity dependant parameters, and account for the influence of velocity on these parameters in the analysis of gait. Inman et al (1981) reported 'every feature of walking changes with speed changes'. This section of the introduction will summarise the literature assessing the validity of this statement.

1.5.1 The influence of velocity on temporal and spatial parameters

There is a consensus as to the relationship between increasing velocity, temporal and spatial parameters (Table 1.1). An increase in velocity was characterised by a reduction in stance, swing and double support duration, cycle and stride time (Murray et al 1966; Andriacchi et al 1977; Murray et al 1984; Kirtley et al 1985; Holden et al 1997; Van der Linden et al 2001; Stansfield et al 2001a; Diop et al 2005; Schwartz et al 2008). In addition, an increase in velocity was associated with an increased single support duration, cadence, stride and step length (Murray et al 1966; Grieve & Gear 1966; Andriacchi et al 1977; Murray et al 1984; Kirtley et al 1985; Holden et al 1997; Van der Linden et al 2001; Stansfield et al 2001a; Schwartz et al 2008). Of the two studies which investigated the relationship between step width and increasing velocity conflicting results were reported (Murray et al 1966; Van der Linden et al 2001).

1.5.2 The influence of velocity on kinematic parameters

A change in velocity has been shown to have significant effects on several kinematic parameters in the sagittal, frontal and transverse plane. The peak values vary with velocity in a variety of linear and non linear ways. The degree to which velocity influences each angle is dependent on the angle itself, and on the timing of the gait cycle.

An increase in velocity was characterised by an increase in anterior pelvic tilt throughout the gait cycle (Murray et al 1966; Kerrigan et al 1998; Stansfield et al 2001a; Lelas 2003; Schwartz et al 2008). Murray et al (1984) reported no change in the position of the pelvis in the sagittal plane for three walking velocities. Pelvic obliquity also increased in magnitude for an increase in velocity (Van der Linden et al 2001; Schwartz et al 2008). A change in the timing of pelvic obliquity has also been reported but this observation has been disputed (Van der Linden et al 2001; Schwartz et al 2008).

	Stance time	Swing time	Double support time	Single support time	Cycle time	Stride length	Step length	Cadence	Stride width
Schwartz et al (2008)	↓		↓				↑	↑	
Diop et al (2005)	↓		↓		↓				
Stansfield et al (2001a)			↓				↑	↑	
Van der Linden et al (2001)			↓	↑	↓	↑	↑	↑	↔
Holden et al (1997)	↓					↑		↑	
Kirtley et al (1985)	↓		↓			↑		↑	
Murray et al (1984)	↓				↓	↑	↑	↑	
Andriacchi et al (1977)	↓	↓					↑	↑	
Grieve & Gear (1966)								↑	
Murray et al (1966)	↓	↓	↓		↓	↑		↑	↑
151 ± 20 cm/s Vs. 218 ± 25 cm/s (Murray et al 1966)	0.65±0.07s Vs. 0.49±0.05	0.41±0.04s Vs. 0.38±0.03s	0.12±0.03s Vs. 0.06±0.03s		1.06±0.09s Vs. 0.87±0.06s	156±13cm Vs. 186±16cm		113steps/min Vs. 138steps/min	7.7±3.5cm Vs. 9.1±4.1

Table 1.1: The influence of increasing velocity on temporal and spatial parameters.

↑ = increases with increasing velocity ↓ = reduces with increasing velocity ↔ = remains unchanged with increasing velocity

In the sagittal plane total hip range of motion has been shown to increase for an increase in walking speed (Murray et al 1966; Murray et al 1984; Frigo et al 1986; Kerrigan et al 1998; Stansfield et al 2001a; Van der Linden et al 2001; Lelas et al 2003; Schwartz et al 2008).This has been shown to be the result of an increase in hip flexion and an increase in hip extension (Murray et al 1966; Frigo et al 1986; Kerrigan et al 1998; Stansfield et al 2001a; Lelas et al 2003; Schwartz et al 2008). These results are disputed by Winter (1983a) who found no change in hip range of motion for an increase in velocity. Van der Linden et al (2001) assessed the relationship between gait parameters at five slow walking velocities and noted an increase in the range of hip motion with an increase in velocity. This was not the case for the two fastest velocities (Van der Linden et al 2001). Schwartz et al (2008) reported a linear relationship for hip flexion at heel strike and maximum hip extension with velocity; however, maximum hip extension exhibited a plateau at high velocities. These results suggested that both linear and non linear relationships exist between velocity and sagittal hip range of motion (Van der Linden et al 2001; Schwartz et al 2008). In the transverse plane a significant difference in total range of hip rotation was reported for a slow velocity when compared to four faster velocities (Van der Linden et al 2001). As for the sagittal plane this suggests a non linear relationship between velocity and transverse plane hip range of motion.

An increase in walking velocity was characterised by an increase in knee sagittal range of motion (Murray et al 1966; Winter 1983a; Kirtley et al 1985; Frigo et al 1986; Murray et al 1984; Holden et al 1997; Stansfield et al 2001a; Van der Linden 2001; Lelas et al 2003; Hanlon & Anderson 2006). This was as a result of increased knee flexion and extension with increasing velocity (Murray et al 1966; Winter 1983a; Murray et al 1984; Kirtley et al 1985; Frigo et al 1986; Holden et al 1997; Van der Linden 2001; Stansfield et al 2001a; Lelas et al 2003; Hanlon & Anderson 2006). Murray et al (1984) reported a significant reduction in knee extension at faster velocities. Kirtley et al (1985) found that knee flexion fell disproportionately at very slow walking speeds, which suggested the relationship between velocity and knee flexion was not linear. For normal gait in the frontal plane a small abduction angle in the stance phase was not significantly affected by a change in velocity (Kirtley et al 1985).

An increase in velocity was characterised by increased dorsiflexion in swing phase (Murray et al 1966; Stansfield et al 2001a). Dorsiflexion during stance phase has been shown to increase and decrease at higher velocities (Murray et al 1984; Frigo et al 1986; Stansfield et al 2001a). The influence of velocity on range of dorsiflexion in stance was dependant on timing of the gait cycle and gender (Frigo et al 1986; Stansfield et al 2001a). Dorsiflexion has been shown to reach a plateau at low velocities, illustrating a non linear relationship (Schwartz et al 2008). Plantarflexion during stance phase was increased at the instant of contralateral heel-strike, and decreased at heel strike with an increase in velocity (Murray et al 1966; Winter 1983a; Murray et al 1984). Kerrigan et al (1998) found no significant difference in plantarflexion between self selected and fast velocities. Stansfield et al (2001a) noted a 5% change in the transition from dorsiflexion to plantarflexion at faster velocities. This observation was disputed by Winter (1983a) who reported a constant pattern of sagittal plane movement for all velocities.

1.5.3 The influence of velocity on kinetic parameters

As for temporal, spatial and kinematic parameters, velocity has been shown to have a significant effect on the kinetic parameters of gait. This is not surprising as joint moments and joint powers take joint angles into account. Kinetic parameters also vary in linear and non linear ways with velocity.

Increased walking velocity resulted in increased peak braking and propulsive GRF (Andriacchi et al 1977; Stansfield et al 2001b; Van der Linden et al 2001; Diop et al 2005; Stansfield et al 2006; Schwartz et al 2008). Andriacchi et al (1977) reported that braking and propulsive GRF varied linearly with increased velocity. Similarly Schwartz et al (2008) reported a linear relationship for peak braking force, but peak propulsive GRF exhibited a plateau at higher velocities. Stansfield et al (2001b) noted that both braking and propulsive force exhibited a plateau at higher velocities. Increased velocity was also associated with an increase in laterally directed GRF (Andriacchi et al 1977; Stansfield et al 2001b; Van der Linden et al 2001). The first peak in vertical GRF increased, and the second peak remained the same for an increase in velocity (Andriacchi et al 1977; Stansfield et al 2001b; Van der Linden et al 2001; Diop et al 2005; Stansfield et al 2006; Schwartz et al 2008). Andriacchi et al (1977) reported that the second peak in vertical GRF increased with increased velocity. Stansfield et al (2001b) noted that the first peak of vertical GRF exhibited a plateau at the higher velocities.

With an increase in velocity there was an increase in hip flexion moment during loading response and the swing phase (Stansfield et al 2001b; Van der Linden et al 2001; Lelas et al 2003; Schwartz et al 2008). Hip flexion moment then decreased in terminal stance; this suggested the presence of a polynomial relationship between velocity and hip flexion moment (Stansfield et al 2001b; Van der Linden et al 2001; Lelas et al 2003; Schwartz et al 2008).

Peak knee flexion moment exhibited a linear relationship with increasing velocity at loading response and pre-swing (Kirtley et al 1985; Holden et al 1997; Stansfield et al 2001b; Lelas et al 2003; Schwartz et al 2008). Peak knee flexion moment has been reported to increase, and decrease in terminal stance for an increase in velocity (Holden et al 1997; Stansfield et al 2001b; Lelas et al 2003). Peak knee flexion moment decreased in terminal swing with an increase in velocity (Stansfield et al 2001b; Van der Linden et al 2001; Schwartz et al 2008). Schwartz et al (2008) noted a plateau for maximum knee flexion moment in late swing at lower walking velocities. Peak knee extension moment demonstrated no relationship with velocity (Kirtley et al 1985; Lelas et al 2003).

Winter (1983a) reported minimal variation in ankle moments with changes in walking velocity. This finding was supported by more recent research (Stansfield et al 2001b; Lelas et al 2003). However, Van der Linden et al (2001) reported a significant increase in peak ankle dorsiflexion moment with an increase in velocity.

Winter (1983b) noted a consistent pattern in joint power with changes in velocity. An increase in velocity was associated with an increase in peak power values (Winter 1983b). Peak sagittal plane hip powers were higher in magnitude for increased velocity (Stansfield et al 2001b; Schwartz et al 2008). Peak hip power generation in early stance has been shown to exceed linear growth for increasing velocity (Schwartz et al 2008). At the knee energy generation and absorption increased with increased velocity (Winter 1983b; Holden et al 1997; Stansfield et al 2001b; Lelas et al 2003). No relationship between ankle power and velocity has been reported (Winter 1983b; Lelas et al 2003). This observation is disputed, increased ankle power at loading response, increased peak ankle power, and decreased ankle power at terminal stance have been reported (Kerrigan et al 1998; Stansfield et al 2001b).

1.6 GAIT ANALYSIS

1.6.1 History of gait analysis

The earliest record of gait analysis can be found in Aristotle's (384-322BC) treatise 'About the Movements of Animals' (Aristotle 1968 *cited by* Baker 2007). Here he described gait and anticipated Newton's third law of motion noting '... for as the pusher pushes, so the pusher is pushed'. After this early attempt to develop an understanding of human biomechanics, very little further analysis occurred until artists revived the study during the renaissance (1450-1527). Leonardo da Vinci described the parallelogram of forces, dividing forces into simple and compound forces, and reported on the projection of the body's centre of gravity in relation to its base of support (Keele 1983).

The 17th century revolution in scientific thinking resulted in a new understanding of mechanics and mathematics. Borelli attempted to integrate physiology and physical science in his 1680 publication 'De Motu Animalum' (Borelli 1679 *cited by* Paul 1998). His descriptions included that of jumping, gait, running, flying and swimming. He also calculated the body's centre of gravity, and outlined its relationship with the body's base of support to help maintain balance during locomotion. During this period Isaac Newton's 'Laws of Motion' were first outlined in 'Philosophioe Naturalis Principia Mathematica' (Newton 1687 *cited by* Paul 2008). While these laws were not applied to the human body here, they provided the fundamental tool for future kinetic and kinematic analysis of movement.

The late 18th century saw a renewed interest in the study of human locomotion. This interest was accelerated by the invention of photography enabling a more detailed, quantifiable study of gait. Brothers Eduard and Wilheim Weber published 'Die Mechanik der Menschlichen Gewerkzeuge' (On the Mechanics of Human Gait Tools) in 1836 (Weber & Weber 1836 *cited by* Paul 1998). Using limited instrumentation such as a telescope with a scale and a stopwatch, the brothers reported over 150 hypotheses on gait.

Etienne Jules Marey moved the analysis of movement from an observational science to one based on quantification. In 1873 he published a study of human limb movements based on experiments he carried out using photographic exposures (Nigg & Herzog 2006). These were taken using a single plate of a subject dressed in black, with brightly illuminated strips placed on the limbs (Nigg & Herzog 2006). In addition, Marey built the first force platform- the dynamometric table (Nigg & Herzog 2006). These inventions enabled the first correlations between ground reaction forces and movement.

Edweard Muybridge's career in the analysis of locomotion began in horses. In order to win a bet, Muybridge was commissioned by Mr Leland Stanford to photograph his horse trotting, to prove there was a period when all four hooves were off the ground. Muybridge was successful and began to document the motion of both animals and humans. From 1884-85 Muybridge produced 20,000 images, the human ones of which were published in 'The Human Figure in Motion' (Nigg & Herzog 2006). Marey quickly saw the potential in Muybridge's use of photography and soon after developed the chronophotograph in 1889.

Wilhelm Braune and Otto Fischer are credited as being the first researchers to perform a mathematical tri-dimensional analysis of gait (Braun & Fischer 1890-1904 *cited by* Paul 2008). The centre of gravity and moments of inertia of the body and all its parts were obtained experimentally with the use of frozen cadavers. A subject with the same measurements was then chosen for the study. Dressed in black with thin light tubes affixed to the body segments the subject's gait was recorded by four cameras. Points were measured from images taken on each side of the subject, and a full three dimensional reconstruction of the true position of that point calculated. Once the point coordinates had been found, the joint centers could be calculated. Knowing the masses and accelerations of the body segments, forces were then able to be estimated for all stages of the gait cycle.

After the First World War, Jules Amar developed the first three component force plate (Baker 2007). The plate was purely mechanical, the force applied to the platform causing the movement of a pointer. From 1938-1939 Elftman developed a similar plate, the pointers of which were photographed by a high speed camera (Nigg & Herzog 2006). The next development in force plate design was that of a 6 component strain gauge force plate created by Cunningham and Brown in the late 1940's (Baker 2007). The first commercially available force plates were piezo-electric in 1969, followed by the strain gauge plates in the early 1970's. There has been little further development in the basic principles of force plate design since this time.

A research group led by Verne Inman and Howard Eberhart was formed at The University of California, Berkeley in 1945 to begin research investigating normal gait. With the use of methods including cinetography, interrupted light photography, bone pins and force plates they developed the science of biomechanics further (Cavanagh 1990). In particular, the analysis of movement in three dimensions allowed for the earliest calculations of the three components of joint moments and forces.

By 1960 research began to focus on the variability of walking, the development of gait from childhood, and the changes in gait in the elderly. The 1970's and 1980's saw vast improvements in measurement methods. The instrumentation to document gait evolved from simple video cameras, to infrared systems with real time coordinate data of limb segments. Reliable force platforms were also now readily available.

1.6.2 Modern gait analysis

Today, three dimensional gait analyses are routinely performed in specialized gait laboratories. With faster processing the analysis is no longer solely being used for research but also as an effective objective clinical tool. Various three dimensional camera systems are commercially available for use in gait laboratories. These systems range from video based recording, electromagnetic-based measurement systems, reflective marker systems, and infra-red light emitting diode (LED) based technology.

Ehara and colleagues completed the earliest study assessing the performance of several of these systems (Ehara et al 1995). All systems were evaluated in identical situations and the relative distance between two markers attached to a rigid bar was assessed (Ehara et al 1995). Of the eight systems which were assessed, video tape recording demonstrated the highest errors and processing times, whereas the active marker system demonstrated the lowest values for error and processing time (Table 1.2) (Ehara et al 1995).

Following on from this study the clinical performance of seven optical based systems, including the CODA (Charnwood Dynamics Ltd., Leicestershire, UK) active marker system, and one electromagnetic-based system were assessed (Richards 1999). Four field tests were conducted to assess the validity of the systems (Richards 1999). Firstly, the ability of the systems to measure the distance between two markers set a known distance apart, and rotating constantly in a volume was investigated (Richards 1999). Four of the systems, including the CODA (Charnwood Dynamics Ltd., Leicestershire, UK), were able to measure the average distance between two markers to within 1 mm of the actual value (Richards 1999). Root mean square (RMS) error describes the difference between an observed measure and a prediction of that measure. The average RMS error was 0.197mm (Richards 1999). Secondly, the ability of the systems to measure motion associated with a static marker was assessed (Richards 1999). The average distance between the two markers was measured to within 1 mm of the actual value for six of the seven optical based systems, the 'Elite' system overestimated the distance by 0.159 mm (Richards 1999). The RMS error was on average 0.214 cm (Richards 1999). Thirdly, the ability of the systems to measure angles in motion was examined (Richards 1999). The angle was measured by all systems on average to within 1.5° of the actual value (Richards 1999). The RMS error was on average 3.034°, the maximum error incurred by the systems ranged from 4.632° - 19.256° (Richards 1999). Finally, the ability of the systems to measure motion of a marker that was moving near to another marker was assessed (Richards 1999). All passive marker systems confused markers placed within 2 mm of each other (Richards 1999). For the CODA system, markers were always correctly identified regardless of how close they were placed to each other. The CODA Motion Dual cx1 (CODA; Charnwood Dynamics Ltd., Leicestershire, UK) LED active marker system was used in conjunction with two AMTI BP400600 Force Platforms (Advanced Mechanical Technology Inc., Massachusetts, USA) for this thesis.

System	Marker	Max no. of markers	Input device	Sampling frequency (Hz)	Marker ID process	Error (mm)	SD (mm)	Mean abs. error (mm)	Max Error (mm)	Processing time (s)
Quick Mag	Coloured	8	Colour CCD* camera	60	Real time	2.6	2.8	3.2	+9.5/-4.6	9
Video locus	Coloured	16	Colour CCD camera	60	Auto	1.0	1.7	1.4	+2.1/-5.2	24
Peak	Reflective/ absorption	35	Video Tape Recorder	60-2000	Auto/manual	5.3	4.2	5.3	+0.6/-14.2	1260
Ariel	Reflective/ absorption	999	Video Tape Recorder	60-400	Auto/manual	5.0	6.0	6.3	+10.8/-26.3	2520
Vicon	Reflective	30	CCD camera	50, 60, 120, 200	Auto	2.3	1.2	2.3	+4.9/-1.6	35
Elite	Reflective	100	CCD camera	50, 100	Auto	3.2	0.9	3.2	+0.4/-5.6	15
Kinemetrix	Reflective	99	CCD camera	50, 100, 200	Auto	3.0	3.8	3.3	+5.0/-12.1	60
Optotrack	LED	256	CCD line sensor	400/ number of markers	Real time	1.0	0.8	0.9	+0.01/-1.2	5

Table 1.2: Performance of 3D camera systems.

*CCD = charge coupled device

Ehara et al (1995)

CHAPTER 2 LITERATURE REVIEW

2.1 INTRODUCTION

The aim of this section is to review the current literature examining the influence of body mass on gait parameters specifically temporal-spatial, kinematic and kinetic parameters of gait.

2.2 SEARCH STRATEGY

Studies included in this review were retrieved from three online databases (PubMed, EmBase, CINAHL) by a single investigator using the following key terms: Obesity, adiposity, body mass, gait, locomotion, ambulation, walking, biomechanics, kinematics, kinetics, temporal, spatial, joint torque, and joint moment. Reference lists of the articles retrieved from the online databases were subsequently hand searched. 21 studies relevant to the present review purpose were collected.

Studies selected for inclusion in this review were studies which assessed the influence of increased body mass in musculoskeletal healthy individuals, on temporal, spatial, kinematic or kinetic parameters at the pelvis, hip, knee or ankle. Those written in English and published from January 1991 to January 2012 were included. 21 studies relevant to the present review purpose were collected.

2.3 STUDY CHARACTERISTICS

The baseline characteristics of participants, velocity and equipment used during testing by each study are presented in Tables 2.1 and 2.2. Ten studies have previously investigated the influence of increased body mass on gait parameters in adults (Table 2.1). 11 studies assessed the relationship between body mass and gait in children (Table 2.2).

Several different techniques were used to measure gait parameters ranging from plantar printing tests to active marker systems. A plantar printing test was adopted by DeSouza et al (2005) in their calculation of temporal spatial characteristics. While this is a cost effective technique it may be prone to human measurement error. The GAITRite system measured temporal and spatial parameters with a sensor embedded walkway connected to the serial port of a computer (Morrisson et al 2008). Several studies used video recording to assess temporal, spatial and kinematic characteristics (DeVita & Hortobagyi 2003; Browning & Kram 2007; Hills & Parker 1991a; 1991b; McGraw et al 2000). However, with video recording the accuracy may be affected by the timing losses involved in recording and replaying a videotape. Techniques used to assess gait parameters which yield the highest degree of accuracy include cine film, passive and active marker systems (Hills & Parker 1991c; Spyropolous et al 1991; Gushue et al 2005; Nantel et al 2006; Vismara et al 2007; Lai et al 2008; Lee et al 2009; McMillan et al 2009; Segal et al 2009; Schultz et al 2009; Schultz et al 2010; Ko et al 2010; McMillan et al 2010; Cimolin et al 2011). In addition when used in conjunction with force plates, kinetic parameters may be calculated.

Author	n	n BMI > 25	Age (years) overweight group	BMI overweight group	Velocity	System
Spyropoulos et al (1991)	21	12	38.9 ± 6.4	*	SS	Cinefilm
DeVita & Hortobagyi (2003)	39	21	39.5 ± 8.8	42.3 ± 7.7	$1.5ms^{-1}$	Video recording, AMTI
De Souza et al (2005)	34	34	47.2 ± 12.9	40.1 ± 6.0	SS	Plantar printing test
Browning & Kram (2007)	20	10	28.9 ± 9.1	35.6 ± 7.1	$0.5ms^{-1}$ $0.75ms^{-1}$ $1.0ms^{-1}$ $1.25ms^{-1}$ $1.5ms^{-1}$ $1.75ms^{-1}$	Video recording, footswitches force measuring treadmill.
Vismara et al (2007)	34	14	29.4 ± 7.9	39.3 ± 3.3	SS	VICON, Kistler
Lai et al (2008)	28	14	35.4 ± 8.8	33.1 ± 4.3	SS	VICON, AMTI
Segal et al (2009)	59	40	49.2 ± 7.3	35.8 ± 5.2	SS	Optotrak, Kistler
Lee et al (2009)	23	9	23.7 ± 4.7	27.7 ± 3.6	SS	Qualisys Motion analysis system, treadmill
Ko et al (2010)	164	108	68.0 ± 1.6	*	SS, fastest possible	Vicon, AMTI
Cimolin et al (2011)	20	10	33.6 ± 5.2	39.3 ± 2.4	SS	Vicon, Kistler

Table 2.1:Previous literature investigating the influence of weight on gait parameters in adults.

SS = Self selected velocity

* = information not provided by the author

Author	n	n BMI > 25	Age (years) overweight group	BMI overweight group	Velocity	System
Hills & Parker (1991a)	20	10	10.5 ± *	26.1 ± 1.6	SS, SS+30%, SS-10%	Video recording
Hills & Parker (1991b)	14	10	*	26.0 ± 1.6	SS, SS+30%, SS-10%	Video recording
Hills & Parker (1991c)	16	12	*	*	SS, SS+30%, SS-10%	Cinefilm
McGraw et al (2000)	20	10	9.1 ± 1.4	30.3 ± 7.9	SS, SS+30%, SS-10%	Peak Motus Video Analysis System
Gushue et al (2005)	23	10	11.9 ± 1.2	29.9 ± 5.4	SS and fast	Optotrak 3020, Kistler
Nantel et al (2006)	20	10	9.7 ± 2.0	26.7 ± 7.1	SS	VICON, AMTI
Morrisson et al (2008)	44	22	9.5 ± 0.6	SDS >1.04	SS	GAITRite
McMillan et al (2009)	14	7	11.9 ± 0.7	2.63 ± 0.24 SDS	SS	Qualisys Motion Analysis System, AMTI
Shultz et al (2009)	20	10	10.4 ± 1.78	30.47 ± 5.54	SS, SS+30%	VICON, Kistler
Shultz et al (2010)	28	14	10.43 ± 1.51	29.74 ± 4.91	SS, SS+30%	VICON, Kistler
McMillan et al (2010)	36	18	15.0 ± 1.5	2.54 ± 0.34 SDS	SS	Hawk System AMTI

Table 2.2: Previous literature investigating the influence of weight on gait parameters in children.

SS = Self selected velocity

* = information not provided by the author

2.4 TEMPORAL AND SPATIAL PARAMETERS

As described in Chapter 1 page 18, velocity has been shown to have a confounding influence on all aspects of gait. It is therefore necessary to review the literature on overweight gait with reference to velocity. From the 21 studies retrieved only 6 noted no between group differences in velocity (Figure 1). The remaining studies either failed to account for velocity in their research, or imposed a constraint on both their experimental and control groups.

The majority of studies examining velocity have concluded that the gait of overweight and obese individuals is characterised by a reduction in self selected velocity (Hills & Parker 1991a; Hills & Parker 1991b; Spyropoulos et al 1991; DeSouza et al 2005; Vismara et al 2007; Lai et al 2008; Ko et al 2010, Cimolin et al 2011). Recently cadence and step length have been described as having a constant linear relationship across a wide range of velocities (Schwartz et al 2008). It is therefore not surprising that both cadence and stride length were reported as reduced in adults and children who are overweight (Hills & Parker 1991a; Hills & Parker 1991b; Hills & Parker 1991c; Spyropoulos et al 1991; DeSouza et al 2005; Vismara et al 2007; Lai et al 2008). It has been hypothesised that overweight and obese individuals are subject to postural instability (Corbeil 2001). This was supported by the reported association between increased adiposity, and increased stance phase and double limb support duration (Spyropoulos et al 1991; McGraw et al 2000; DeVita & Hortobagyi 2003; Browning & Kram 2007; Vismara et al 2007; Lai et al 2008; Morrison et al 2008; Ko et al 2010, Cimolin et al 2011). However, a reduction in velocity was also characterised by an increase in stance phase and double limb support duration (Diop et al 2005; Schwartz et al 2008). It is therefore difficult to differentiate velocity effects from body mass effects on gait parameters. Individuals who are overweight may be subject to postural instability or they may simply be walking at a slower velocity.

Four studies assessed the influence of body mass on temporal and spatial parameters using predetermined velocities in their protocol (DeVita & Hortobagyi

2003; Browning & Kram 2007; Schultz et al 2009; Schultz et al 2010). This enabled an analysis of temporal and spatial parameters independent of velocity. Schultz et al (2009, 2010) used a metronome to control velocity in their study, however little is known about the influence of a metronome on gait. Cadence and stride length were not significantly different between groups, highlighting the association of these parameters with velocity (DeVita & Hortobagyi 2003). Stance phase and double support duration remained increased, suggesting that these differences may be true adaptations in overweight and obese gait (DeVita & Hortobagyi 2003; Browning & Kram 2007).

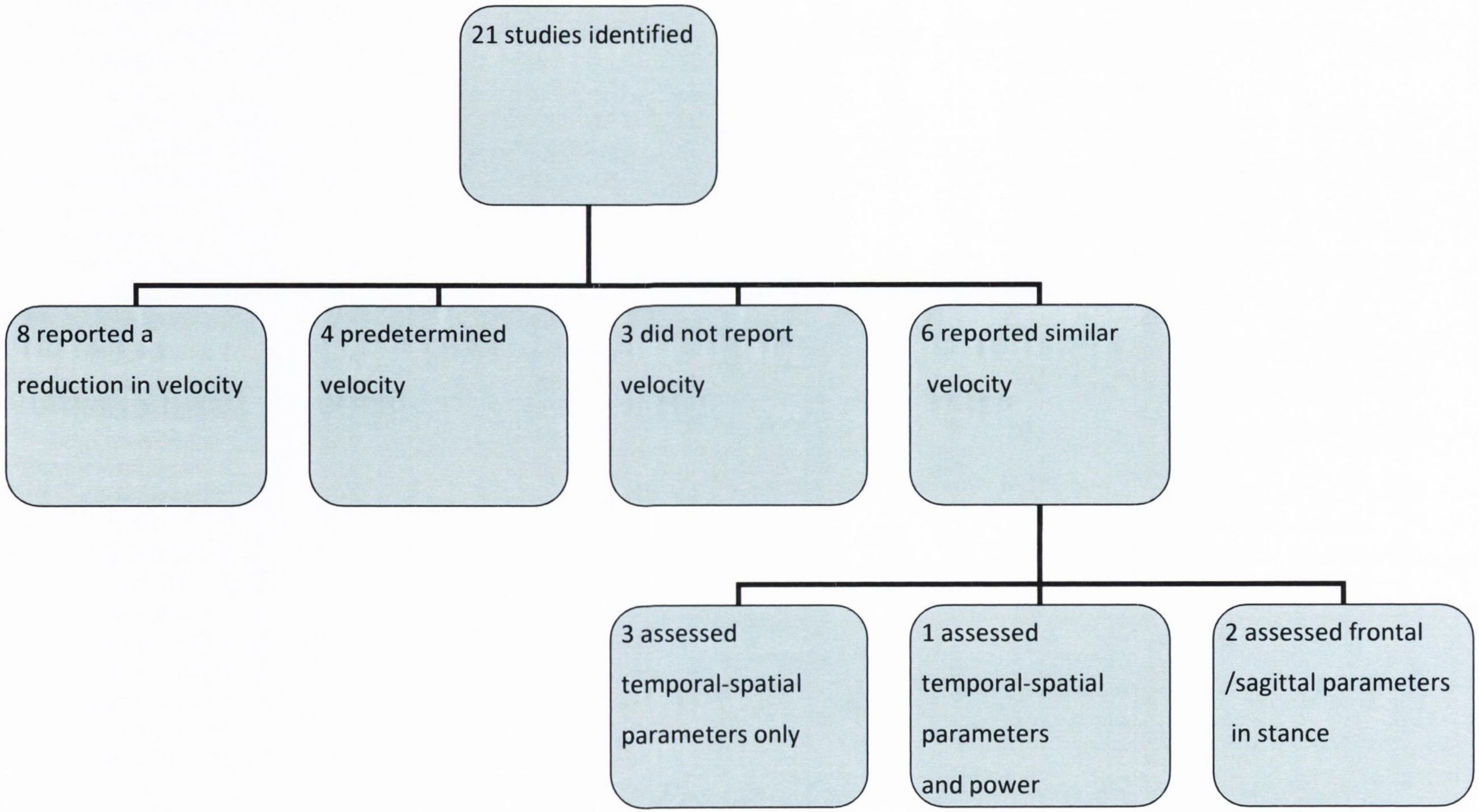


Figure 2.1: Previous literature assessing the influence of weight on body mass with reference to velocity.

Four studies reported no significant difference in velocity between their overweight and healthy weight groups (Hills & Parker 1991c; Nantel et al 2006; McMillan et al 2009; Ko et al 2010). An associated similarity in cadence and stride length was noted between the overweight and healthy weight groups (Hills & Parker 1991c; Bertram & Ruina 2001; Nantel et al 2006). In contrast to the studies by DeVita & Hortobagyi (2003) and Browning & Kram (2007) where velocity was predetermined, no significant difference in stance phase duration and double support duration were reported (DeVita & Hortobagyi 2003; Nantel et al 2006; Browning & Kram 2007; Ko et al 2010). Nantel et al (2006) suggested high activity levels of their obese participants as a possible reason for the similarities between their groups. In research conducted by Ko et al (2010) three groups were assessed; an overweight, obese and healthy weight group. No significant difference in velocity was established between the overweight and healthy weight groups (Ko et al 2010). The obese group however walked significantly slower than the healthy weight control group, but not the overweight group (Ko et al 2010). This suggests the extent to which temporal-spatial parameters vary is dependant on the extent of adiposity.

Step width has been identified as increased in overweight/obese adults (Spyropolous et al 1991; DeSouza et al 2005; Browning & Kram 2007; Ko et al 2010). Step width has been reported to be independent of velocity (Van der Linden et al 2002). Increased step width has been identified as a method of improving stability in neurological populations (Stolze et al 2002). This would support the hypothesis of instability in overweight and obese gait. Step width is influenced by thigh girth and overweight individuals have been shown to have an increased thigh girth (Segal et al 2009).

Velocity has been shown to have a significant relationship with both temporal and spatial parameters (Diop et al 2005; Schwartz et al 2008). Despite a significant difference in velocity between overweight and healthy weight groups, the influence of velocity on reported parameters has been disregarded by several studies (Hills & Parker 1991a; Hills & Parker 1991b; Spyropolous et al 1991; DeSouza et al 2005;

Vismara et al 2007; Lai et al 2008; Ko et al 2010, Cimolin et al 2011). Temporal and spatial results may suggest the presence of dynamic instability in overweight and obese populations. Further investigation assessing the relationship of aforementioned parameters with reference to velocity is warranted.

2.5 KINEMATIC PARAMETERS

2.5.1 The sagittal plane

The sagittal plane kinematic characteristics of individuals with excess body mass demonstrated few differences when compared to their non overweight peers (Hills & Parker 1991a; Spyropolous et al 1991; DeVita & Hortobagyi 2003; Gushue et al 2005; Browning & Kram 2007; Vismara et al 2007; Lai et al 2008; Lee et al 2009; Shultz et al 2009; Ko et al 2010; McMillan et al 2010). There is general agreement that hip, knee and ankle range were within normal limits for individuals who are overweight or obese (Spyropoulos et al 1991; Vismara et al 2007; Browning & Kram 2007; Lai et al 2008; Lee et al 2009; Shultz et al 2009; Ko et al 2010, Cimolin et al 2011). Cimolin et al (2011) noted an increased ankle range for their obese group. In normal gait, sagittal plane hip and knee range decreased with a decrease in velocity (Hills & Parker 1991c; Lelas et al 2003). Several of the studies reported a reduced velocity of walking in the overweight group but no significant difference in hip and knee range (Spyropolous et al 1991; Vismara et al 2007; Lai et al 2008; Ko et al 2010, Cimolin et al 2011). It may be possible that the decreased velocity masked potential significant differences in hip and knee range. The relationship between ankle range and velocity was not well defined. This makes it difficult to consider differences in sagittal plane ankle range in relation to velocity (Schwartz et al 2008; Stansfield et al 2001).

One must note that the total range of movement discussed above provides a general overview of an individual's gait. Examining total range of movement can however mask alterations in the components of range. For example a sagittal plane hip range of 40° will not be deemed significantly different from normal. The components of this hip range may be 40° of hip flexion coupled with 0° of hip extension which would be considered different from normal (Figure 2.2). Despite this, few studies discussed the components of total range (Hills & Parker 1991a; Spyropoulos et al 1991; DeVita & Hortobagyi 2003; Gushue et al 2005; McMillan et al 2010).

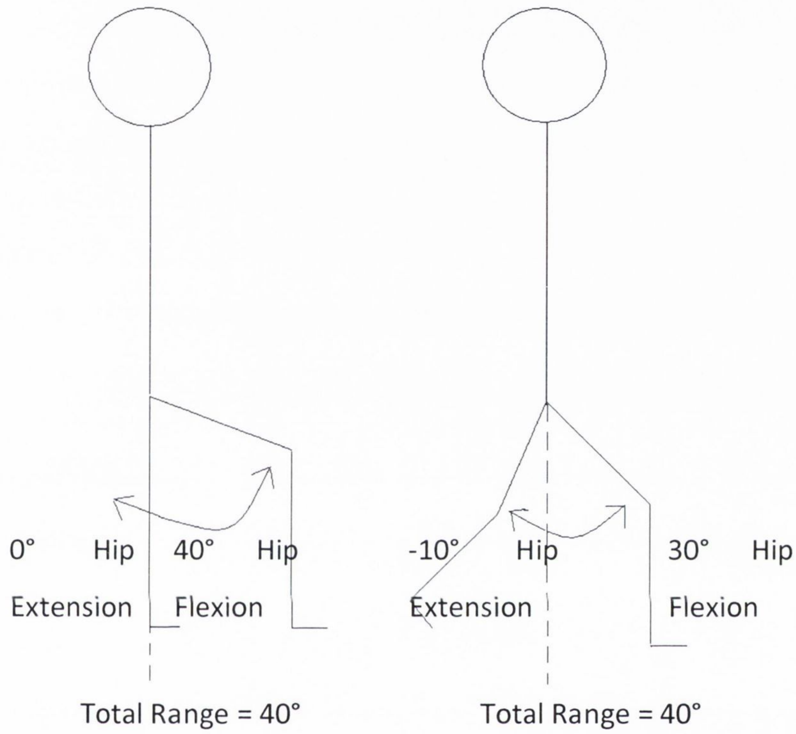


Figure 2.2: Normal sagittal plane hip range equates to 40° and is present both for person A and person B. However, person A exhibits no hip extension which is not a characteristic of normal gait hip kinematics.

At the hip and knee, flexion has been reported as decreased and increased for individuals with excess body mass (Hills & parker 1991a; Spyropoulos et al 1991; DeVita & Hortobagyi 2003; Gushue et al 2005; McMillan et al 2010). The angle at the hip in the sagittal plane is identified as the angle created by the intersection of the pelvic plane and the long axis of the thigh. Pelvic kinematics were not described by these authors so it is therefore difficult to determine the true motion occurring at the hip (Hills & parker 1991a; Spyropoulos et al 1991; DeVita & Hortobagyi 2003; Gushue et al 2005; McMillan et al 2010). Recently Cimolin et al (2011) reported no significant difference between their groups for knee flexion angle at initial contact and mid stance. Where hip and knee flexion were reported as increased, the group which were overweight ambulated at a significantly slower velocity (Hills & parker 1991a). However a reduction in velocity was associated with a reduction in hip and knee flexion (Van der Linden et al 2002). A more flexed posture lowers the body's centre of gravity improving stability. The findings of Hills & Parker (1991a) therefore support the hypothesis of instability in populations who are overweight. In the study by Spyropoulos et al (1991) the obese group ambulated with significantly less hip and knee flexion, but walked significantly slower, and therefore the reduced flexion may simply be as a result of the slower self selected velocity. In contrast, for several studies which reported a reduction in hip and knee flexion, velocity was predetermined and not significantly different between groups (DeVita & Hortobagyi 2003; Gushue et al 2005; McMillan et al 2010). DeVita & Hortobagyi (2003) suggested individuals who are obese reorganise their neuromuscular function to minimise the load across their knee joint. These findings were supported more recently by Gushue et al (2005) who noted a significant reduction in the knee flexion angle of their overweight group, but no significant difference in sagittal knee joint forces between groups. McMillan et al (2010) suggested the more erect position adopted by their obese participants was to compensate for reduced hip and knee extensor strength, and instability within the knee joint structure itself.

At the ankle, an increase in dorsiflexion with an associated decrease in plantarflexion has been reported by one study (Spyropoulos et al 1991). They suggested obese participants alter their ankle kinematics to enable body weight to cross over the ankle early to improve stability (Spyropoulos et al 1991). These findings are disputed by DeVita & Hortobagyi (2003) and Cimolin et al (2011) who found an increase in plantarflexion, and plantarflexion at toe off for individuals who were overweight.

2.5.2 The frontal plane

As for the sagittal plane several studies reported no significant changes in frontal plane hip, knee and ankle range for overweight individuals (Shultz et al 2009; McMillan et al 2010; Ko et al 2010). A decrease in frontal plane hip range was associated with a decrease in velocity; whilst velocity did not influence frontal plane knee range (Schwartz et al 2008; Kirtley et al 1985). For two studies which assessed the influence of obesity on gait, a significant increase in frontal plane hip range was noted (Ko et al 2010; Cimolin et al 2011). In the study by Ko et al (2010) no changes in frontal plane hip range were identified for their overweight group, suggesting that the influence of body mass on gait may be dependant on the extent of adiposity. For obese groups participants ambulated at a significantly slower velocity, suggesting the increase in hip range may be even greater than was reported (Ko et al 2010; Cimolin et al 2011). McMillan et al (2010) reported an increase in frontal plane knee range in their overweight group supporting the association between knee joint instability and individuals who are overweight (Kirtley et al 1985). However, movement occurring at the knee in the frontal plane is minimal. For example, Schultz et al (2009) reported a mean value of 0.62° ± 3.95° for overweight frontal plane swing phase knee adduction. Richards (1999) assessed the ability of several optical based motion analysis systems to measure angles in motion. Angles measured by such systems have on average an accuracy of 1.5° of the actual value (Richards 1999). It is therefore difficult to determine if movements of such a small nature, as frontal plane knee motion, are outside the capabilities of such measurement systems.

Very few studies have assessed the components of frontal plane joint range (Spyropoulos et al 1991; Lai et al 2008; McMillan et al 2009; McMillan et al 2010). Spyropoulos et al (1991) found an increase in hip abduction for their overweight group, whereas an increase in hip adduction has been reported more recently (Lai et al 2008; McMillan et al 2009). These studies failed to describe pelvic kinematics for their populations (Spyropoulos et al 1991; Lai et al 2008; McMillan et al 2009). Altered pelvic obliquity results in altered frontal plane hip kinematics making it difficult to establish a true representation of frontal hip kinematics. Increased knee adduction and increased knee abduction have been reported (Lai et al 2008; McMillan et al 2009; McMillan et al 2010). The components of frontal plane ankle range for individuals that are overweight have not been reported.

2.5.3 The transverse plane

As for the frontal plane research describing the influence of velocity on transverse plane kinematics is minimal (Van der Linden et al 2002; Schwartz et al; 2008). A non linear decrease in transverse plane hip range was associated with a decrease in velocity (Van der Linden 2002). Only two studies have investigated the influence of body mass on transverse plane kinematics (Lai et al 2008; Shultz et al 2009). Both studies reported no significant changes in transverse plane hip, knee and ankle range (Lai et al 2008; Shultz et al 2009). As the overweight groups walked significantly slower, a decrease in transverse plane hip range was anticipated (Lai et al 2008; Schwartz et al 2008; Schultz et al 2009). Therefore transverse plane hip range may actually be increased for the overweight groups (Lai et al 2008; Schultz et al 2009). When assessing the components of total range, Lai et al (2008) reported a significant increase in transverse plane ankle eversion for their obese participants. An ankle strategy has been described, whereby the ankle joint is required to rotate towards the supporting surface, in order to improve postural control (Pintsaar et al 1996). Thus the increased ankle eversion may suggest the presence of reduced stability in overweight groups (Lai et al 2008).

2.6 KINETIC PARAMETERS

2.6.1 Ground reaction forces

Two studies reported on GRF during gait for individuals who are overweight (Lai et al 2008; Browning & Kram 2007). A relationship between increased body mass and GRF was anticipated based on Newton's Third Law of Motion 'for every action there is an equal and opposite reaction'. It is not surprising that peak absolute vertical, anterior-posterior and medial-lateral GRF have been reported as significantly increased in individuals who are overweight (Browning & Kram 2007). These greater joint forces incurred by individuals who are overweight should be taken into account when prescribing exercise clinically.

Another method for analysing GRF and increased adiposity is by normalising the force for body mass (Nkg^{-1}). Normalised peak vertical and anterior-posterior GRF have been reported as similar and reduced for individuals who are overweight (Lai et al 2008; Browning & Kram 2007). A decrease in velocity was associated with a decrease in peak vertical and anterior-posterior GRF (Schwartz et al 2008). This may explain the findings of reduced peak vertical and anterior-posterior GRF in the study by Lai et al (2008). Browning & Kram (2007) reported on normalised medial-lateral peak GRF and found it to be unaltered in populations which are overweight. From the limited research directly assessing the relationship between increased adiposity and GRF, there appears to be no significant association.

2.6.2 Joint moments

As for kinematic parameters, joint moments are also influenced by body mass with absolute moments shown to be significantly increased in overweight groups (DeVita & Hortobagyi 2003; Browning & Kram 2007; Shultz et al 2009). Moments normalised by body mass, and by body mass and height are more frequently reported (DeVita & Hortobagyi 2003; Gushue et al 2005; Browning & Kram 2007; Vismara et al 2007; Lai et al 2008; Segal et al 2009; Shultz et al 2009; McMillan et al 2009; McMillan et al 2010; Ko et al 2010). As discussed previously, there appears to be no alteration in hip, knee and ankle joint range in populations which are overweight. It is therefore reasonable that sagittal, frontal and transverse, hip and knee total joint moments were also reported as unaltered in overweight groups (Shultz et al 2009). Sagittal, frontal and transverse peak hip, knee and ankle moments have also been reported as unaffected by increased body mass (Lai et al 2008; Ko et al 2010).

The components of total joint moments also show differences for individuals who are overweight. In the sagittal plane, the hip flexor moment has been reported as increased at late stance, coinciding with a reduction in hip flexion found by the same study (McMillan et al 2010). Peak hip extensor moment has been reported as similar between overweight and healthy weight groups, whilst hip extensor moment at initial contact was found to be reduced for overweight groups (Browning & Kram 2007; McMillan et al 2010). A more erect posture draws the GRF vector closer to the joint centre. It therefore minimises the level of muscle strength required during gait. Obese individuals may adopt this gait to compensate for reduced muscle strength at the hip. Peak knee extensor moment appears unaffected by increased body mass (DeVita & Hortobagyi 2003; Browning & Kram 2007). Knee flexor moment has been reported as similar and reduced between overweight and healthy weight individuals (Gushue et al 2005; McMillan et al 2010). A reduction in knee flexor moment supports the hypothesis of a more erect posture, adopted to compensate for reduced anti-gravity muscle strength (McMillan et al 2010). Peak ankle dorsiflexor moment for overweight individuals has been reported as unaltered and reduced (Gushue et al 2005; Vismara et al 2007; Lai et al 2008; Shultz et al 2009). Similarly, peak plantarflexor moment has also been found to be reduced in overweight groups (Browning & Kram 2007; Shultz et al 2009; McMillan et al 2010). Peak plantarflexor moment occurs in late stance to enable forward propulsion of the body. A lower peak plantarflexor moment results in a reduced push-off force and therefore a reduced step length and velocity. This alteration may be adopted in an attempt to improve dynamic stability in this population.

Frontal and transverse plane moments have received less attention (Gushue et al 2005; Lai et al 2008; Segal 2009; Shultz et al 2009; McMillan et al 2009; Ko et al 2010; McMillan et al 2010). McMillan et al (2009) noted a reduced hip abductor moment during early stance for their overweight group and suggested hip abductor weakness as a possible explanation for this finding. Two authors reported no difference in knee abductor moment between groups, whereas one study found an increase in knee abductor moment for their overweight group (Gushue et al 2005; Segal et al 2009; McMillan et al 2009). Schultz et al (2009) reported no change in frontal or transverse plane ankle moments between groups. Ko et al (2010) reported a significant decrease in frontal plane peak ankle moments. In addition, an increase in ankle evertor moment has been reported for overweight individuals (Lai et al 2008).

2.6.3 Joint power

Total sagittal hip, knee and ankle power has been reported as unaffected by increased body mass (Nantel et al 2006; Ko et al 2010). Vismara et al (2007) found no change in sagittal peak ankle power for their overweight group. This was recently disputed by Cimolin et al (2011) who found a reduction in peak ankle power in their obese group.

Once again when assessing the components of power-namely generation and absorption, differences in the sagittal plane for those who are overweight were present (Nantel et al 2006; Shultz et al 2010). At the hip, sagittal plane absorptive power was reported as increased in overweight groups (Nantel et al 2006). The authors suggested the earlier shift from hip extensor generation to hip flexor absorption was adopted by their obese participants in an effort to minimise the energy cost of hip extension (Nantel et al 2006). The difference reported may also be due to weakness of the hip extensors (Nantel et al 2006). After this increased energy absorption phase it was noted that hip energy generation was in fact reduced for obese subjects (Nantel et al 2006). This suggests that obese individuals were mechanically less efficient at transferring energy during gait. Hip generative power has however also been found to be increased and therefore requires further research (Shultz et al 2010, Cimolin et al 2011). Schultz et al (2010) also reported a significant increase in knee absorptive power indicating a greater need for control of knee flexion for their overweight group. Cimolin et al (2011) reported a significant increase in ankle absorptive power for their obese group.

Frontal plane powers have received little attention. Ko et al (2010) noted no change in total ankle power, hip generative power or knee absorptive power in overweight individuals. This study also reported an increase in total knee generative and total hip absorptive power (Ko et al 2010). These increases indicate greater frontal plane activity at the knee and hip. As for frontal plane powers, transverse plane powers were addressed by only one author who reported no significant difference for an increase in body mass (Shultz et al 2010).

2.7 CONCLUSION

Of the 21 studies included in this review only six reported similar between group self-selected velocities. These six studies provided a description of temporal spatial parameters, joint power, sagittal and frontal plane biomechanics in children. For the remaining literature it is difficult to differentiate changes in gait due to body mass from changes in gait due to velocity. In addition, the reduction in velocity seen by eight studies may in fact mask considerable alterations in the gait cycle.

Overall for those who are overweight a more tentative gait with a reduced velocity, stride length and cadence, and increased support duration noted. Few alterations in total joint range and joint moments were seen in the sagittal, frontal and transverse planes at the hip knee and ankle. These results suggest that perhaps the presentation of gait for adults and children who are overweight is not in fact altered. However, there are several limitations to this conclusion:

1. Velocity may be masking considerable alterations.
2. Velocity constraints were adopted (e.g. auditory cueing, treadmill), with little information as to how the constraints may affect gait.
3. Conclusion based on total joint range, the components of joint range have been discounted by several studies.
4. Pelvic kinematics were not assessed. This made it difficult to determine the true differences in range and moments acting at the hip.

2.8 KEY FINDINGS AND IMPLICATIONS FOR FUTURE RESEARCH

This review suggests that there are few alterations in the presentation of gait in overweight groups. However it is unclear as to whether these results are due to a difference in self selected velocity, methodological differences between studies or are true results for overweight individuals (Figure 2.3). Absolute ground reaction forces were noted to be increased for overweight adults. This places greater stresses on the skeleton. Further detailed analysis is required to assess the influence of body mass on gait, while accounting for the aforementioned limitations of previous work (Figure 2.3).

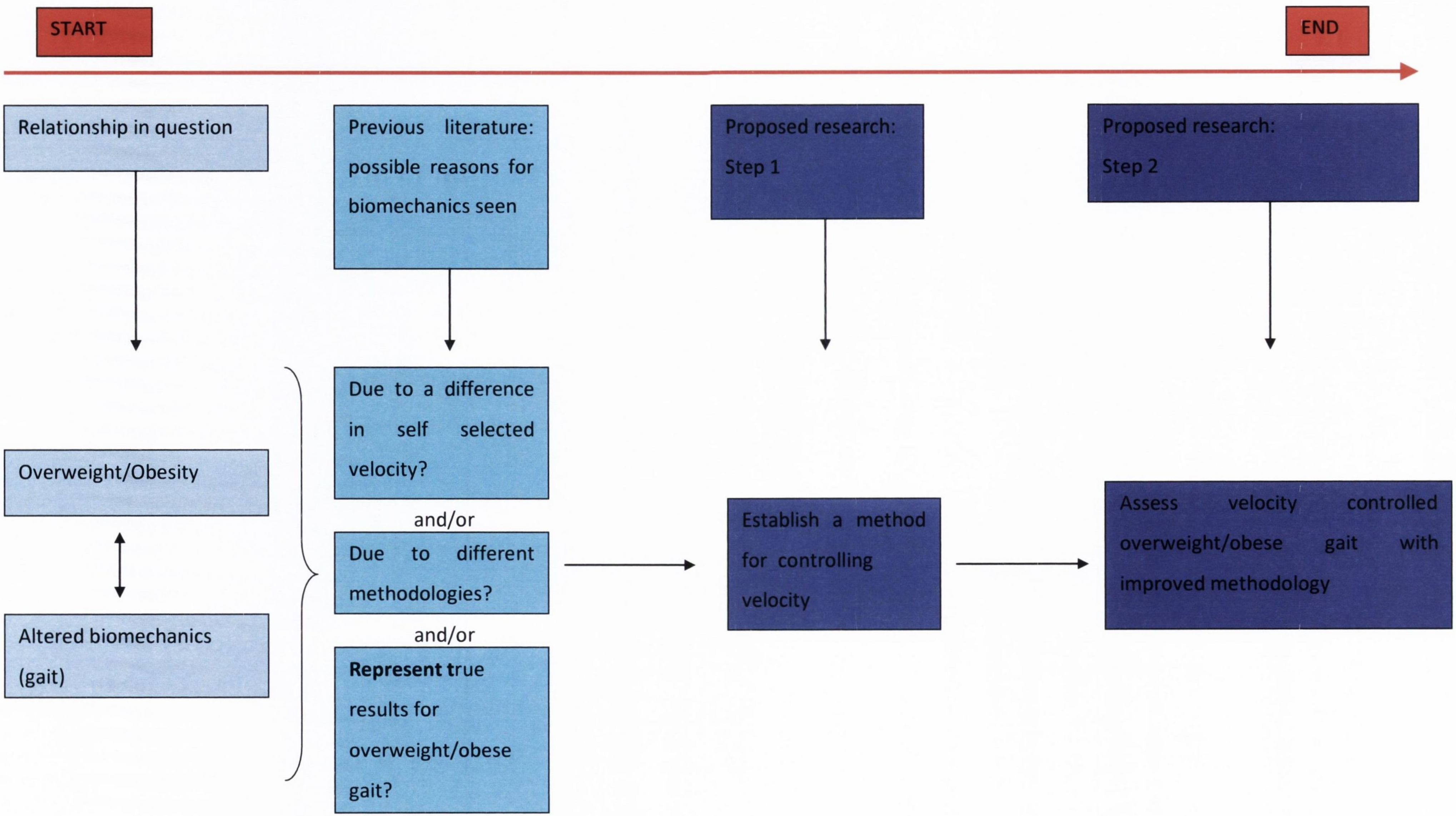


Figure 2.3: Key findings of the literature review and future directions of this work.

2.9 AIM AND OBJECTIVES OF THIS THESIS

Aim:

To investigate the influence of body mass on gait parameters.

Objectives:

To determine the influence of auditory cueing on gait parameters* in healthy adults.

To determine the influence of auditory cueing on gait parameters* in healthy children.

To determine the influence of body mass on gait parameters† in adults.

To determine the influence of body mass on gait parameters† in adults, while accounting for the confounding influence of velocity.

To determine the influence of body mass on gait parameters† in children.

To determine the influence of body mass on gait parameters† in children, while accounting for the confounding influence of velocity.

* Gait parameters refers to: temporal-spatial parameters; maximum and minimum three dimensional kinematics at the hip, knee and ankle.

† Gait parameters refer to: temporal-spatial parameters; three dimensional kinematics at the pelvis, hip, knee and ankle; three dimensional joint moments at the hip, knee and ankle; maximum absolute and normalised anterior, medial and vertical ground reaction forces; and, maximum and minimum total joint power at the hip, knee and ankle.

CHAPTER 3 METHODOLOGY

3.1 INTRODUCTION

This aim of this research was to evaluate the influence of body mass on gait parameters. In an effort to control for the effects of velocity on gait, the influence of an auditory cue on gait parameters was also assessed. Information on the research was provided to the participant, and where applicable to their guardian, prior to written consent being obtained (Appendices 1-7, pages 229-247). All of the studies were granted ethical approval by the Research Ethics Committee of St. James's Hospital/ The Adelaide and Meath Hospital, Dublin Incorporating the National Children's Hospital (AMNCH) (Appendices 9-11, pages 249-251).

3.2 STUDY TITLES

<u>Study 1:</u> The influence of auditory cueing on gait parameters in adults.

<u>Study 2:</u> The influence of auditory cueing on gait parameters in children.

<u>Study 3 (a):</u> The influence of body mass on gait parameters in adults without controlling for the influence of velocity.

<u>Study 3 (b):</u> The influence of body mass on gait parameters in adults while controlling for the influence of velocity.

<u>Study 4 (a):</u> The influence of body mass on gait parameters in children without controlling for the influence of velocity.

Study 4 (b): The influence of body mass on gait parameters in children while controlling for the influence of velocity.

3.3 RESEARCH DESIGN

A cross over study design was chosen to address the influence of using an external auditory cue on gait parameters in participants of a healthy weight (Study 1 and 2). This type of study design required a group of participants (children and adults of a healthy weight) to complete a minimum of two experimental protocols (walking with and without an external auditory cue). Each participant acted in effect as his/her own control.

A cross sectional design was selected as the most appropriate for investigating the effects of body mass on gait parameters (Studies 3(a), 3(b), 4(a) and 4(b)). This epidemiological study design required a sample of the subject population (children and adults who are overweight) of interest to be matched with a sample of controls (children and adults of average weight). Control subjects were matched by gender, age, height and cadence as these demographics have been shown to influence gait parameters (Sutherland 1997; Whittle 2002; Schwartz et al 2008). Control subjects were matched to within 3cm for height. For children, control participants were matched to within two years of age. Adult participants were matched to within five years of age. Gait parameters were then collected in both groups and comparisons made to establish if a difference exists between the groups.

3.4 INCLUSION AND EXCLUSION CRITERIA

Table 3.1 outlines inclusion criteria for this research. For studies with participants from a paediatric population, a lower age limit of seven years was selected. Development of normal gait is said to be accomplished by this age (Studies 2, 4(a) and 4(b)) (Sutherland et al 1980). In Ireland the Children Act (2001) defined a child as any person under the age of 18. An upper age limit of 17 was therefore selected for paediatric research (Studies 2, 4(a) and 4(b)), and a lower limit of 18 was selected for adult research (Studies 1, 3(a) and 3(b)).

Inclusion criteria	
Paediatric research	Adult research
Aged 7-17	Aged > 18
BMI IOTF classification as overweight/obese (Experimental group); BMI IOTF classification as healthy weight (control group)	BMI > 25 (overweight group); BMI < 25 (control group)
Informed consent	

Table 3.1: Inclusion criteria.

Table 3.2 outlines the exclusion criteria for this research. Excessive leg length discrepancy, musculoskeletal injury and disease may lead to altered biomechanics. Subjects who met these criteria were excluded from participation (Studies 1-4(b)). Pregnant females were excluded due to the altered gait associated with changes in body weight, body shape and the endocrine system exhibited during pregnancy (Studies 1-4(b)) (Huang et al 2002).

Exclusion criteria
Leg length discrepancy >5cm
Lower limb musculoskeletal injury in the last 6 months which required medical intervention
A history of musculoskeletal disease
Pregnancy
Allergy to adhesive tape or Cramer Tuf-Skin Tape Adherent Spray

Table 3.2: Exclusion criteria.

3.5 PARTICIPANTS AND RECRUITMENT

3.5.1 Recruitment of children who are overweight (Studies 4(a) and 4(b))

Overweight children attending the Weight Management and Endocrine Clinics at AMNCH were recruited for this research. This took place over a period of nine months from October 2009 to June 2010. Subjects with a BMI z-score > +1 SD and aged 7-17 years were identified from the clinic attendance list and nursing staff.

The author approached potential participants and their guardians in the waiting area of the clinic, where a verbal overview of the study and information leaflets were provided (Appendix 2, page 233). The opportunity to ask questions was given, and an appointment for testing at the Trinity Centre for Health Sciences, St. James's Hospital, Dublin, scheduled for at least one week later.

3.5.2 Recruitment of children of a healthy weight (Studies 2, 4(a) and 4(b))

Children of a healthy weight who were gender, age and height matched were recruited from schools local to AMNCH and St. James's Hospital. This took place over a period of 18 months from March 2010 to September 2011. An e-mail was sent to the principal of each school outlining the research (Appendix 3, page 240). The principal was asked to contact the lead investigator to discuss any questions in relation to the study, and the possible recruitment process.

When a principal decided that their school would take part in the research, s/he was required to send letters and information leaflets to the student's guardians (Appendix 4, page 241). This information outlined the study and provided contact details for the lead investigator. On receiving contact from the child's guardian, the lead investigator answered all questions, and arranged an appointment for testing to take place at least seven days later.

3.5.3 Recruitment of adults who are overweight (Studies 3(a) and 3(b))

Adults attending the hypertension clinic at St. James's Hospital were recruited. This took place over a period of four months from January 2011 to April 2011. Those with a BMI > 25 were identified from the clinic attendance list and nursing staff.

The author approached potential participants in the waiting area of the clinic, where a verbal overview of the study and information leaflets were provided (Appendix 5, page 242). The opportunity to ask questions was given, and an appointment for testing scheduled for at least one week later.

3.5.4 Recruitment of Adults of a healthy weight (Studies 1, 3(a) and 3(b))

Gender, age and height matched adults of a healthy weight were recruited from the Trinity Centre for Health Sciences. This took place over a period of 13 months from September 2010 to October 2011. Posters were placed around the Trinity Centre for Health Sciences (Appendix 7, page 247). An e-mail was sent to the staff and students of the Trinity Centre for Health Sciences (Appendix 6, page 246). An information leaflet was sent as an attachment to the e-mail and the lead investigators contact details were provided (Appendices 1 and 5, pages 229 and 242). When a student or member of staff decided that they would like to take part in the research, s/he was required to contact the lead investigator. On receiving contact from the potential participant, the lead investigator answered all questions and arranged an appointment for testing to take place at least seven days later.

3.5.5 Participants and recruitment continued (Studies 1-4(b))

A one week period was selected between initial contact and testing to enable both the participants, and where applicable their guardian, to consider their participation in the study. Participants and guardians were encouraged to contact the author by phone or e-mail should they have any questions in the interim. Participants over the age of 18 were required to provide written consent at their test appointment

(Appendices 1 and 5, pages 229 and 242). Children were required to provide written assent, and guardians were required to provide written consent for their child's participation (Appendix 2, page 233). All subjects identified during the research period, who agreed to participate, who met the inclusion criteria and did not meet any of the exclusion criteria were included in this research.

To enable informed consent details of the study were relayed to the participant/guardian. These included the reasons for the research being undertaken, the measurement process, any benefits or risks for the participant, confidentiality, and that their involvement was voluntary and they may have ended their participation in the study at any time. Participants over the age of 18 and guardians were provided with an information leaflet reiterating these points (Appendices 1, 2, 5, pages 229, 233 and 242). In addition, a child friendly information leaflet was provided for each child so they better understood their involvement in the research (Appendix 2, page 233).

3.6 MEASURES

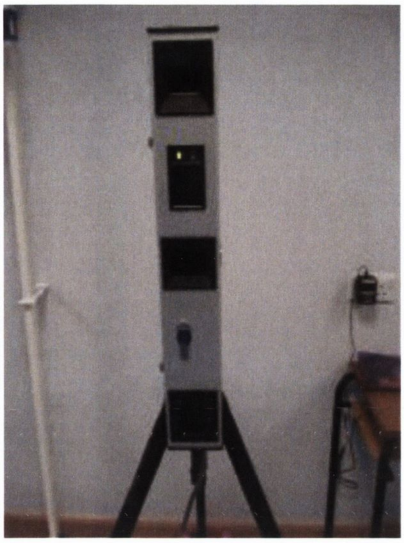
Data from the gait analyses was collected and stored on an external hard-drive. Additional data collected at each participant assessment was initially recorded in paper format on specifically designed data collection forms (Appendix 8, page 248). These were then stored in a locked filing cabinet together with the external hard-drive. All data was later transferred to Microsoft Excel Analyze It for statistical analysis. Table 3.3 below details the measurements used in this research.

Equipment	Measure
CODA Motion Dual cx1 System with Active Hub (Charnwood Dynamics Ltd., Leicestershire, UK)	Kinematic and kinetic gait parameters
AMTI BP400600 Force Platforms (Advanced Mechanical Technology Inc., Massachusetts, USA)	Ground reaction forces and Joint Moments and Powers
Leicester Height Measure(Seca Ltd., Birmingham, UK)	Height

Table 3.3: Equipment and measurements.

3.6.1 CODA Dual CX1 (Charnwood Dynamics Ltd., Leicestershire, UK)

The CODA Dual CX1 (Charnwood Dynamics Ltd., Leicestershire, UK) system enables real-time three dimensional motion capture of LED fitted to predefined locations and the analysis of this motion. The capture system is comprised of two sensors (Figure 3.2). Each sensor unit contains three windows. The two end windows capture horizontal motion while the centre window captures vertical motion. Behind these windows lies a series of black lines placed in front of a linear photodiode sensing array. The photodiode detects light from the LED markers, gathering the information from all the LED markers at a single source, and converts it into current or voltage. The LED markers (Figure 3.1), powered by rechargeable drive boxes (Figure 3.3), are pulsed one at a time in an order that is synchronised with the sensors. This enables the main sensor unit to identify each LED. When a LED marker flashes in front of the windows a shadow of the black lines is cast onto the linear photodiode sensing array. The shadow moves as the marker continues to move, therefore influencing the signal in each pixel of the array. A cross correlation in real time is conducted to identify the point at which the array pattern and a template of the pattern of the black lines match, and therefore the location of the LED. Data is then stored on an active hub to enable analysis to take place.



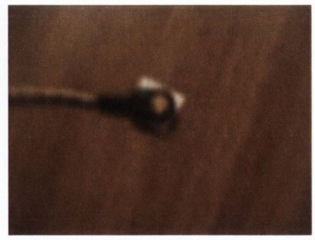


Figure 3.1: CODA motion LED marker.

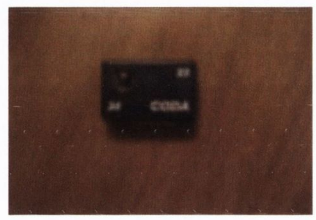


Figure 3.2: CODA motion sensor.

Figure 3.3: CODA motion drive box.

The reliability of optical based biomechanical measurement systems such as the CODA Dual CX1 (Charnwood Dynamics Ltd., Leicestershire, UK) can be affected by three potential sources of variability – system, rater and subject (Monaghan et al 2007). Maintenance checks as recommended by Charnwood Dynamics Ltd., Leicestershire, UK were performed twice annually to minimize system variation. In addition an alignment protocol was conducted on each test day to ensure the laboratory co-ordinate frame was the same for all tests (Chapter 3, page 49). The intra-rater and inter-rater reliability of gait measurements has been identified as a source of variability in gait analysis (Maynard et al 2003). To eliminate inter-rater variability, the assessment process was completed by one investigator for all subjects. A strict protocol for the clinical assessment, identifying joint centre locations and marker placement was devised in an attempt to minimize intra-rater reliability (Chapter 3, page 49).

3.6.2 AMTI BP400600 force platforms

The AMTI BP400600 Force Platform (Advanced Mechanical Technology Inc., Massachusetts, USA) uses strain gages mounted on four strain elements (Figure 3.4). Strain is a measure of the change in shape of an object due to an applied force. When a subject steps onto the force platform, the strain gauges are compressed or stretched causing a change in their shape, and therefore their resistance. The resistance of the strain gages describes the gauges' opposition to the passage of an electrical current. With a change in resistance, bridge excitation at the output terminals occurs. Bridge excitation refers to the conversion of the change in resistance into a voltage that can then be fed into the transducer. The six component transducer measures the three dimensional force and moment components which act on the strain gages. The force plate is then connected to an analogue to digital (AD) converter. This converts the analogue output voltage to a number of bits which can be read by the activehub for analysis.

In the calculation of joint moments and powers, kinematic and kinetic data are combined. In addition, the laboratory coordinate system is required to ensure accurate alignment of the force plates to the motion capture system. Therefore, the techniques adopted to minimize system, rater and subject variability for the CODA dual CX1 (Charnwood Dynamics Ltd., Leicestershire, UK) are also applicable to the AMTI BP400600 force platforms (Advanced Mechanical Technology Inc., Massachusetts, USA) (Chapter 3, page 57). Amplifiers were reset to zero before every test session.

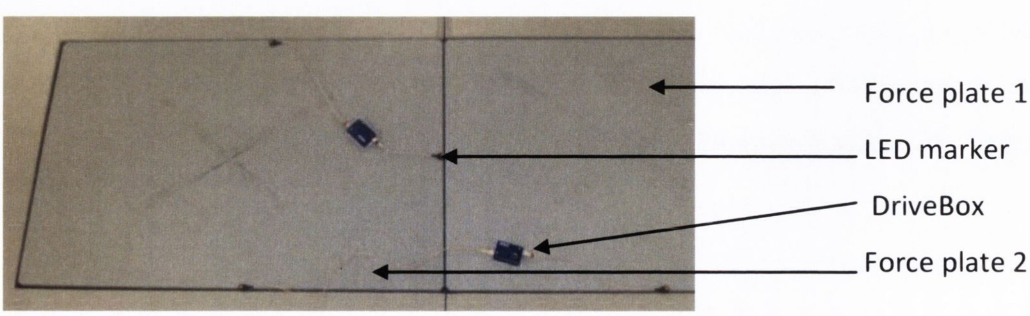


Figure 3.4: Marker placement on force platforms for alignment protocol.

3.6.3 TempoPerfect online digital metronome

TempoPerfect (NCH Software, Canberra, Australia) is an online digital metronome available to download free of charge. Once downloaded the TempoPerfect (NCH Software, Canberra, Australia) simply requires the investigator to input the required tempo/cadence at which a loud beat will be played. Ten participants were used to assess the validity of the metronome. The metronome was set to a cadence of 100 steps per minute and an audible beat was sounded. Each participant walked in turn along the ten meter walkway and allowed one heel strike on each beat. The number of steps, the number of beats and the time were recorded for each participant. Each participant's cadence was calculated manually using the equation:

$$\frac{\text{Number of Steps}}{\text{Time taken to cover 10m}} \quad \text{x } 60 \text{ (seconds)} \quad = \quad \text{cadence}$$

A one sample t-test was used to see if the mean calculated cadence differed significantly from 100. There was no significant difference ($p= 0.2$). The metronome was therefore deemed to be valid.

3.7 GAIT ANALYSIS

3.7.1 The gait laboratory

The gait laboratory at the Trinity Centre for Health Sciences is comprised of two CODAmotion sensors (Charnwood Dynamics Ltd., Leicestershire, UK), and two AMTI force plates (Advanced Mechanical Technology Inc., Massachusetts, USA) embedded in the centre of a 10m x 1m fiber glass raised walkway (See figure 3.5). Prior to a participant attending the laboratory all equipment in the gait laboratory was turned on, and the covers were removed from the cameras. Several uplighters with halogen bulbs were purchased to eliminate the effect of the laboratory florescent lighting on the surface mounted markers and their drive boxes.

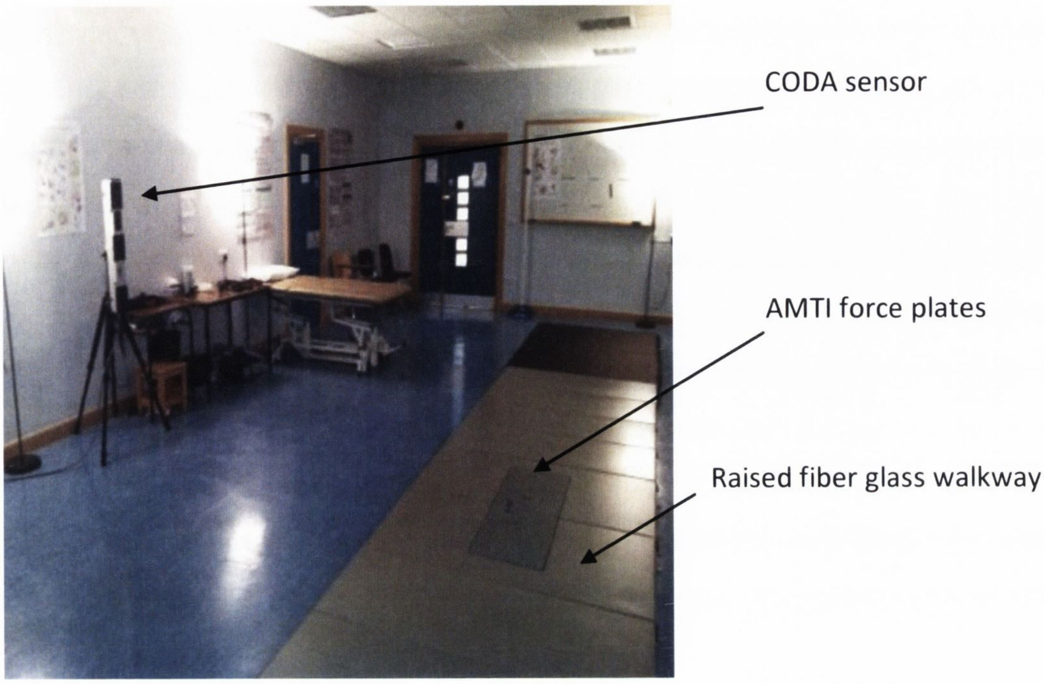


Figure 3.5: Gait laboratory at the Trinity Centre for Health Science.

3.7.2 Calibration

Several calibration procedures were conducted to minimize systematic error. On each test day 'CODAmotion Analysis' was opened on the active hub and the 'Bilateral + Force' configuration file was selected. This file enables the acquisition of bilateral gait and ground reaction forces. The 'BilatGaitAcqCheck.stp' set-up was then loaded.

After a minimum of fifteen minutes force platforms were reset to zero, first manually on the mini amps and then via the active hub. The author ensured no objects were on the force plates while the reset was being conducted.

The CODA sensors (Charnwood Dynamics Ltd., Leicestershire, UK) were aligned relative to the force platforms. Four markers were placed on the force platforms to create a three dimensional axis for the system (Figure 3.4). The origin was selected as the midpoint between the two force plates both vertically and horizontally. It was marked on one force plate with permanent marker to enable the same origin to be used for all acquisitions. The X-axis was defined as the direction of the ten meter walkway, two markers were placed along the ridge of the edges of the two force plates at least one force plate length apart. The Y-axis was perpendicular to the X-axis, with a marker being placed parallel to a marker demarcating the X-axis. The Z-axis was therefore the vertical perpendicular projection up away from the floor. The alignment procedure was completed via the 'Align CODA (s)' tool on the active hub.

3.7.3 Clinical assessment

Participants were required to wear tight fitting shorts or swimming togs, and to remove their shoes and socks. Measurements were collected with the participant positioned supine on a plinth.

Pelvic width, pelvic depth, bilateral knee and ankle joint width were measured using Vernier Calipers (InSize Co. Ltd., Bucharest, Romania) (Figure 3.6). Pelvic width was measured as the distance between the most prominent points of the subject's ASIS. Pelvic depth was measured as the unilateral distance between the right ASIS and the right posterior iliac spine (PSIS). The distance between the medial and lateral femoral condyles was recorded as the knee joint width, and the distance between the medial and lateral malleoli the ankle joint width.

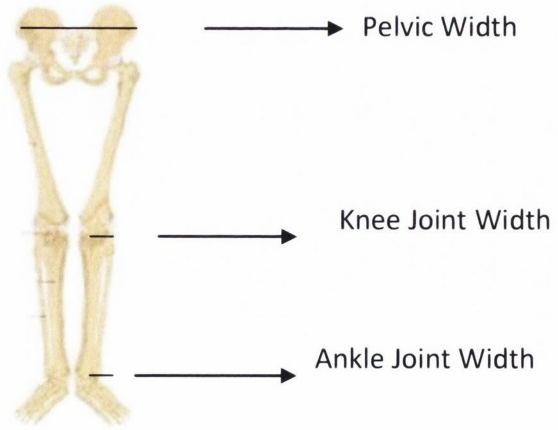


Figure 3.6: Clinical measurements.

Bilateral thigh girth, thigh length, shank length, foot length and leg length were measured using a non-stretch measuring tape (Expert Leisure Supplies, Dublin, Ireland) (Figure 3.7). From the superior border of the patella vertically up the thigh five inches a horizontal mark was made. Thigh girth was measured as the circumference of the thigh at this point. Thigh length was measured as the distance from the most palpable lateral point of the greater trochanter to the lateral knee joint space. Shank length was measured as the distance from the lateral knee joint space to the lateral malleolus. The distance from the posterior border of the calcaneus to the most distal point of the third phalange of the foot was used to estimate foot length. True leg length, the distance from the ASIS to the medial malleolus, was used when measuring leg length.

Lower limb active and passive range of motion was assessed visually. Any movement range which was deemed to be reduced/increased was then measured using a goniometer and protocols described by Clarkson (2000). Femoral neck anteversion was assessed using the Craig test (Ruwe et al 1992). The Craig test required a subject to lie prone with their knee flexed to 90°. The posterior aspect of the greater trochanter was then palpated. Using the foot as a lever the hip was internally and externally rotated until the posterior border of the greater trochanter was parallel to the plinth. The angle created by the longitudinal axis of the shank and the vertical detailed the presence of femoral anteversion/retroversion. An angle of greater than 15° indicated the presence of femoral anteversion whereas an angle less than 8° indicated femoral retroversion. The Craig test can be viewed at http://www.youtube.com/watch?v=OREPrs8aMDw. Participants were also screened for genu valgum with the Q angle test (Insall et al 1976). The Q angle was determined as the angle created by the intersection of two lines. The first a line from the ASIS to the centre of the patella, and second a line from the centre of the patella to the centre of the tibial tubercle. An angle of greater than 15° was deemed indicative of genu valgum.

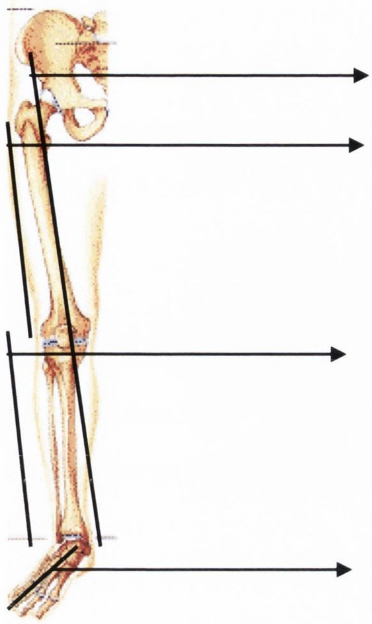


Figure 3.7: Clinical measurements.

To facilitate bilateral gait LED application the ASIS, PSIS, medial and lateral femoral condyles, knee joint centre, ankle joint centre and fifth metatarsal head were all located bilaterally and marked with an eyeliner pencil with the subject in standing. The knee joint centre was identified by palpating the lateral knee joint space and marking a horizontal line at the superior aspect of the space. The fibular head was then located and using a non-stretch measuring tape (Expert Leisure Supplies, Dublin, Ireland) a point 1.5cm anterior to the fibular head was identified. A vertical line was then drawn directed superiorly from that point. The intersection of the two lines was identified as the location of the knee joint centre (Protocol from the Central Remedial Clinic, Dublin; Walsh et al 2000).

3.7.4 Bilateral gait LED application and derivation of 3D segments/joints from marker placements

In order to capture bilateral gait, twenty-two LED markers fixed to predetermined locations on the lower limbs and pelvis were required (Figures 3.8 and 3.9).

Participants were fitted with a pelvic frame; sacral, femoral and tibial wands, and surface mounted LED's for bilateral gait analysis. Markers were attached to the pelvic frame, marker wands and skin using double sided sticky tape. 'Cramer Tuf-Skin Tape Adherent Spray' was applied to the lateral aspect of the foot and heel to aid fixation of the markers.

The pelvic frame was positioned so that the posterior border lay parallel to the PSIS, this point acts as the pelvis origin. The side bars were then affixed and directed anterior in line with the ASIS bilaterally. The sacral wand was attached to the midpoint of the posterior border of the pelvic frame. A total of six markers were affixed to the pelvic frame. Markers were attached to the superior anterior aspect of the side bar bilaterally and to the superior posterior aspects of the side bar bilaterally. To ensure the line adjoining the anterior and posterior markers was parallel to the axis of the side bar, the markers we fixed to the edges of the frame.

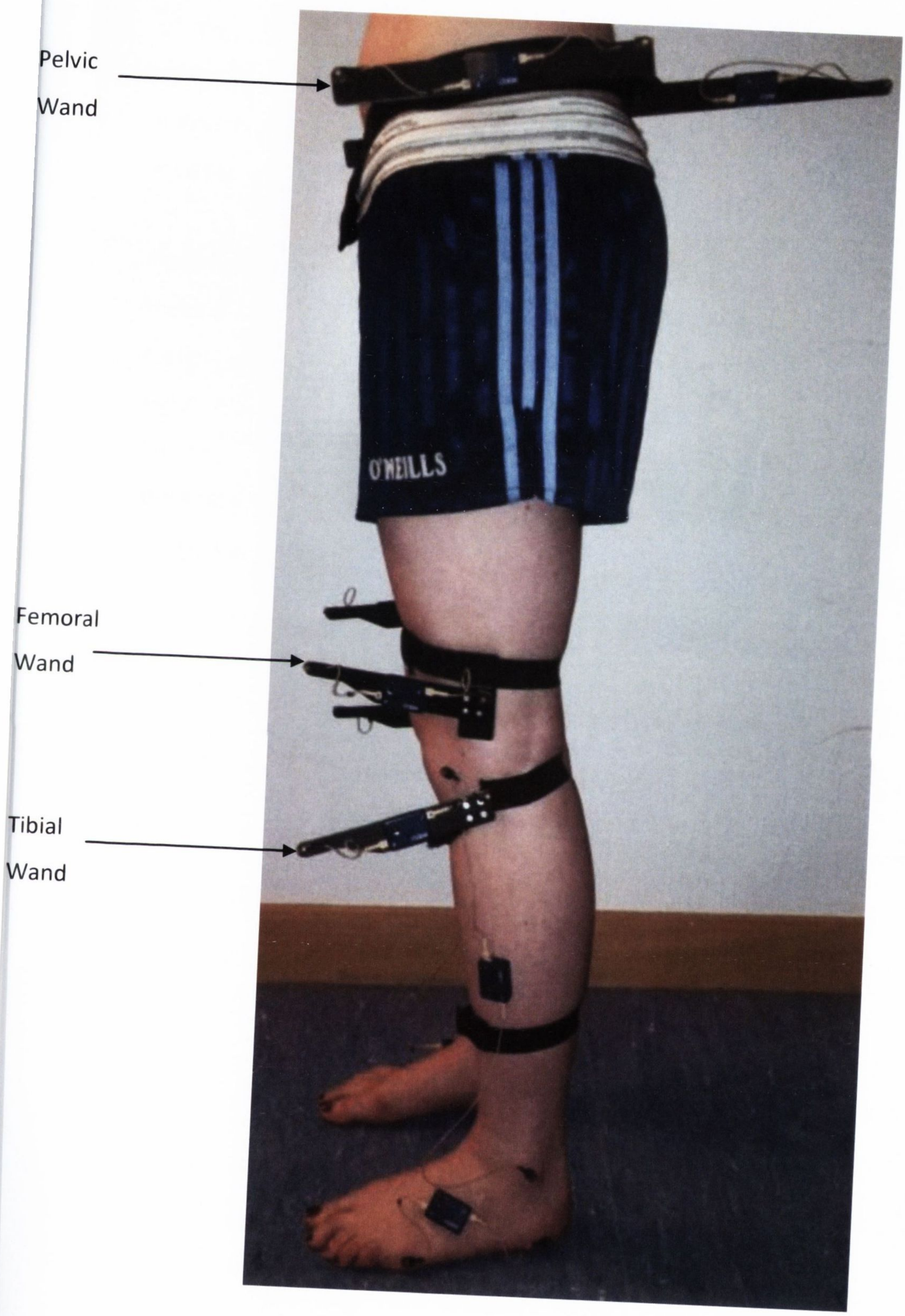


Figure 3.8: Bilateral gait LED application (side view).

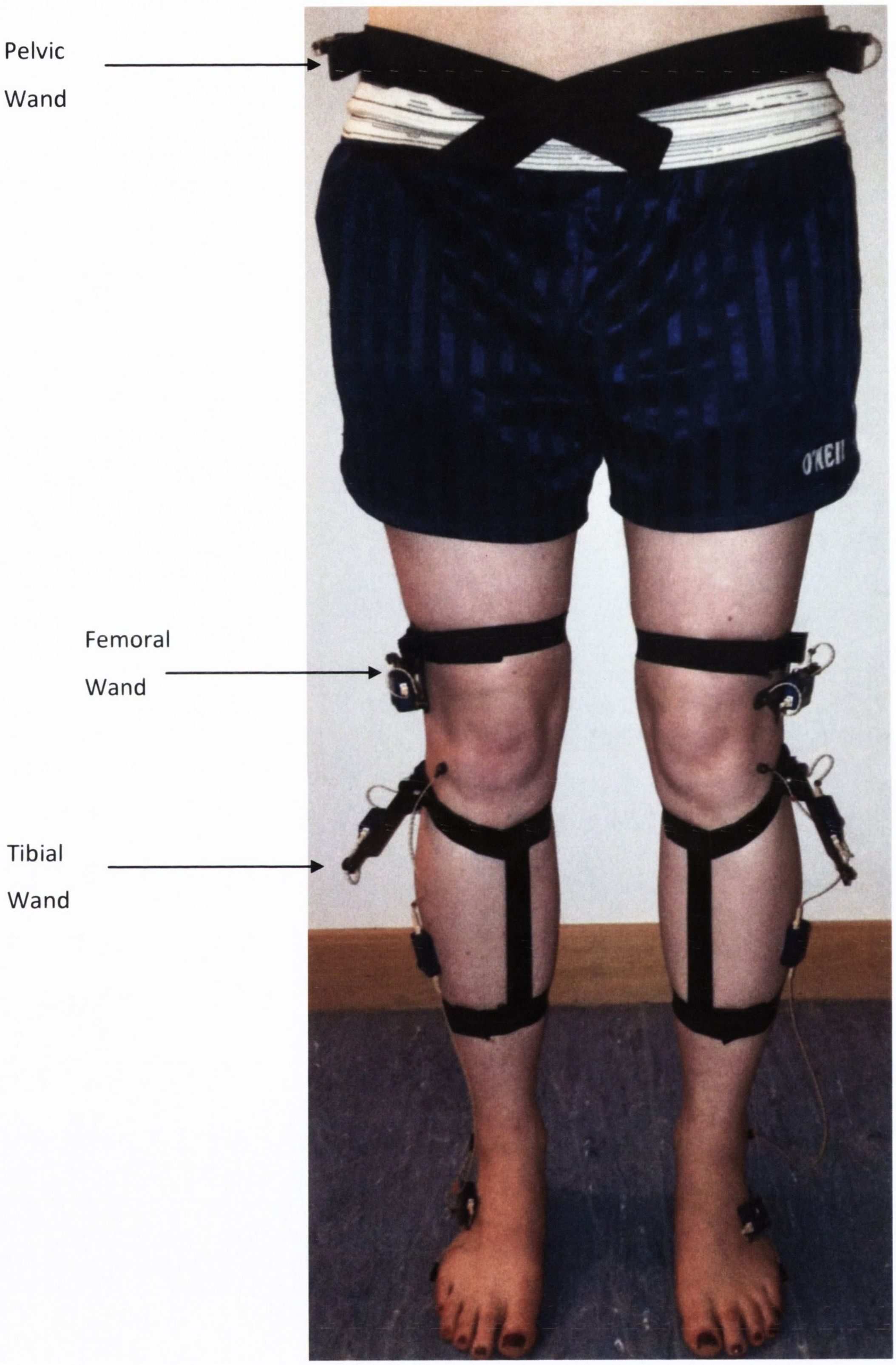


Figure 3.9: Bilateral gait LED application (front view).

The line joining the left ASIS to the right ASIS markers defined the medial-lateral axis of the pelvis. The anterior-posterior axis was then defined as perpendicular to the midpoint of the medial-lateral axis in the plane which contained both ASIS and PSIS markers. The inferior-superior axis was determined as a vector perpendicular to the intersection of the anterior-posterior and medial-lateral axes.

The three axes enabled the creation of the three component vectors. Using clinical measurements of pelvic width and pelvic depth, the precise location of the origin (sacral reference point) was determined. Using the clinical measures the left and right ASIS LED markers were offset (from marker location to clinical measurement) by the CODA software anteriorly by pelvic depth in the plane of the intersection of the three axes, and medio-laterally by half the pelvic width. Two markers were fixed to the posterior aspect of sacral wand, as per the side bars, the markers were fixed to the edges of the frame.

Several predictive methods for locating the hip joint centre have been reported (Seidel et al 1995; Bell et al 1989; Kirkwood et al 1999). These methods identify the hip joint centre as lying close to the groin, and therefore marker placement is difficult. To overcome this, the CODA software (Charnwood Dynamics Ltd., Leicestershire, UK) creates the three dimensional location of the hip joint centre virtually using the published technique by Bell et al (1989). The technique involves the addition of offsets (proportions of an individual's pelvic width) to the mid-point between right and left ASIS reference points.

Femoral and tibial wands were attached to the participants using Velcro strapping. Femoral wands were attached to the lateral aspect of the thigh. The horizontal components of the femoral wands were fitted with two markers, and were aligned perpendicular to the transcondylar axes of the knee joints in sitting. The transcondylar axis was identified as the line joining the medial and lateral femoral condyles which were marked previously. A ruler was held in line with these marks, and the horizontal component of the femoral wand was aligned perpendicular to the ruler (Figure 3.10). A marker was placed over the previously identified knee joint centre (Chapter 3, page 63) bilaterally. The CODA software (Charnwood Dynamics Ltd., Leicestershire, UK) labeled the knee marker as the lateral knee reference point. It then created a virtual medial knee reference point, one knee widths distance from the lateral knee reference point, in a direction perpendicular to the virtual hip and the two thigh wand markers. The knee centre reference point was then determined as the mid point of the medial-lateral reference axis. The thigh inferior-superior axis was determined as the line joining the virtual hip and knee joint centers. The anterior-posterior axis was determined by the horizontal component of the femoral wands. The medial-lateral axis was identified as a vector perpendicular to the intersection of the inferior-superior and anterior-posterior axes.

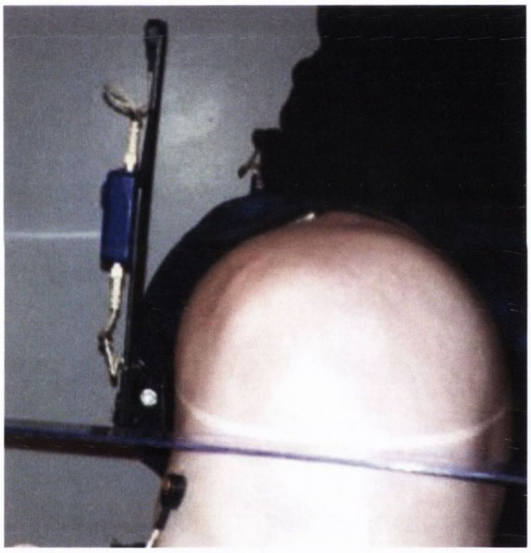


Figure 3.10: Femoral wand alignment procedure.

The edge of the vertical component of the tibial wand was aligned with the anterior crest of the tibia, the proximal aspect of the vertical component forming a 'V' around the tibial tuberosity. The horizontal component was fitted with two markers, and aligned perpendicular to the axis of the ankle joint with the use of the ankle alignment jig (Figure 3.11). Markers were also fixed to the most prominent bony point of the lateral malleolus, fifth metatarsal head, and posterior inferior lateral aspect of the heel. As for the knee joint, the ankle marker was labeled as the lateral ankle reference point. The medial ankle reference point was determined as one ankle joint width from the lateral ankle reference point in a direction perpendicular to the two tibial wand markers. The ankle centre reference point was then determined as the midpoint of these two points. The inferior-superior axis of the shank was determined as the line joining the knee joint center to the ankle joint center. The tibial horizontal component determined the anterior-posterior axis of the shank. The ankle axis was taken as perpendicular to the shank anterior-posterior axis.

The foot segment was defined by the ankle centre, heel and toe markers offset by half an ankle width. The anterior-posterior axis of the foot was taken to be parallel to the line joining heel and toe markers, offset by half an ankle width.

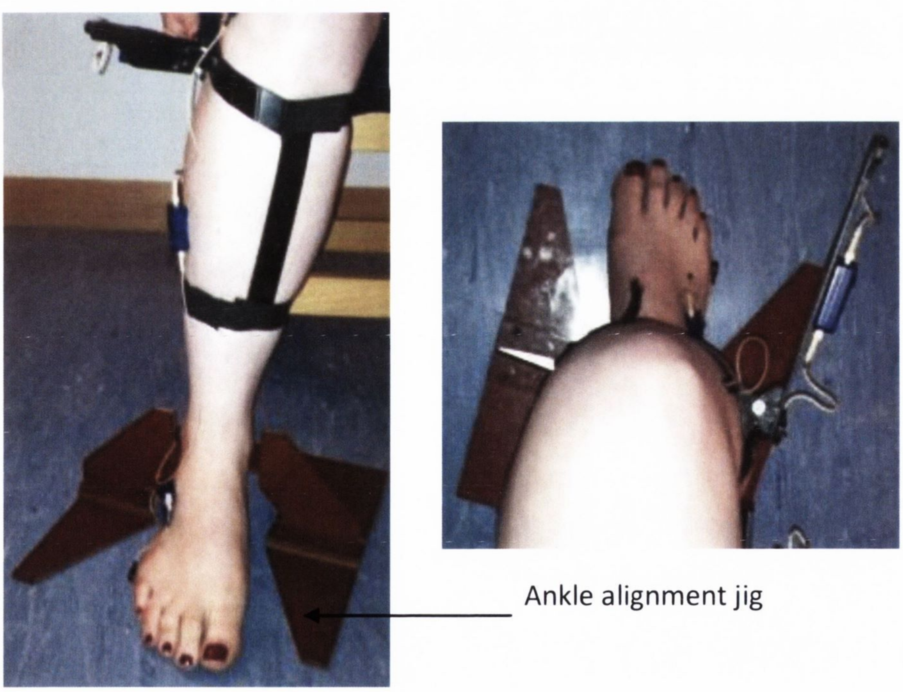


Figure 3.11: Tibial wand alignment procedure.

3.7.5 Data acquisition

Participants were instructed to walk up and down the length of the 10m walkway continuously until data acquisition was completed. Participants were not shown or asked to walk on the AMTI BP400600 Force Platforms (Advanced Mechanical Technology Inc., Massachusetts, USA). This was to avoid subjects trying to 'land' on the force plates. The viewing angle of the CODAmotion CX1 (Charnwood Dynamics Ltd., Leicestershire, UK) cameras is 70°, this angle does not encompass the 10m walkway at the gait laboratory. The longer walkway enables participants to overcome initial acceleration and obtain a steady state velocity for data acquisition. An auto-start option was selected so that data acquisition began as soon as the surface markers came into view of the cameras. As a familiarisation phase, participants were asked to complete ten lengths. During this phase data was collected to enable checks on marker placement, marker visibility and drive box function to occur. Upon completion of the familiarisation phase, data was then collected and stored for each participant. Files were named and stored using predetermined codes.

3.7.6 Identifying the gait cycles

After data acquisition the data collected at the clinical assessment was entered into the active hub using the subject/patient identification tool. This data is required for the calculation of internal joint centers, and for the inverse dynamic calculation of joint moments and powers. Subjects were allocated an ID and their gender, date of birth, age, weight, height, pelvic width, pelvic depth, knee joint widths and ankle joint widths were entered manually. Segment length, mass-ratio, centre of mass position, and radii of gyration were then defined by the CODA software (Charnwood Dynamics Ltd., Leicestershire, UK) from standard reference data.

In order to generate a report of the data, two interlinking gait cycles need to be identified i.e. left initial contact, right initial contact, left toe off, left initial contact, right toe off, right initial contact. The 'cursors' menu and 'move to force-on/force-off' and then 'drop left/right cycle marks' were selected to enable one initial contact and toe off of the left/right foot to be marked. This can then be repeated for the other foot. To identify the following heel strike bilaterally to complete the strides, the stick figure illustration was used (Figure 3.12). The illustration was zoomed in on and slowed down, when the point of foot contact had been determined, the 'cursors' menu and 'drop right/left cycle mark' was selected.

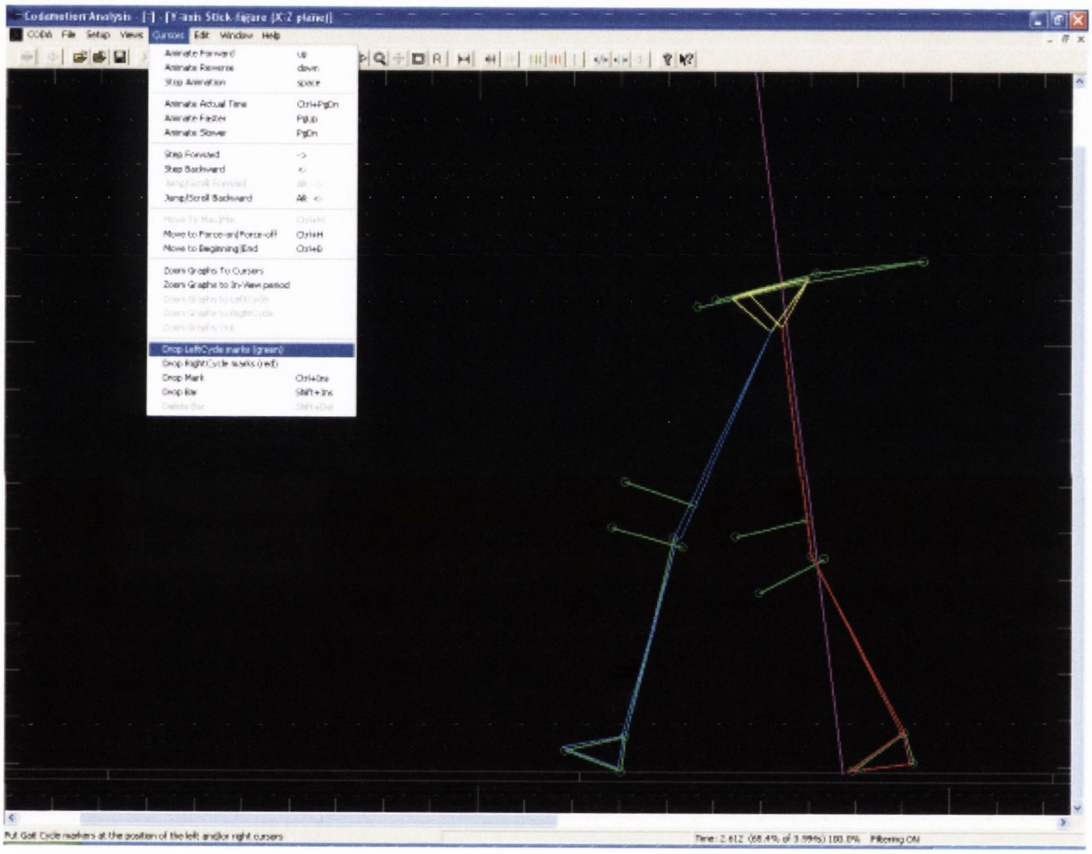


Figure 3.12: Stick figure illustration, with 'Drop Left Cycle marks (green)' function.

In the studies where force data was not collected, the heel strikes were marked as described above. In order to determine toe off the 'L.heel/Toe:L.AnkleZ*Time' graph was used. A toe 'marker velocity' plot was added to the graph. On this graph, the cursor was placed at the peak in the 'marker velocity' plot between the two heel strikes. The point was then marked using the 'cursors' menu and 'drop left/right cycle mark'.

Each data file was the saved as both .mdf and .mdr files. Saving the files as .mdr enabled a report to be generated in the CODAmotion Report Generator (Charnwood Dynamics Ltd., Leicestershire, UK).

3.7.7 Calculating joint angles

A rigid body, free to move in space is said to have six degrees of freedom, three of which may be associated with translational movement, the other three with rotations. In clinical movement analysis such a body is represented by a minimum of three markers, whose measured positions provide sufficient data to allow definition of an embedded co-ordinate frame. For gait analysis it is important to determine the orientation of segment angles and their associated embedded co-ordinate frame. To achieve this Euler angles are used to define the orientation of the distal segment relative to the proximal segment. This is achieved by comparing the attitudes of corresponding axes of the segment-embedded co-ordinate frames.

A distal limb segment in the neutral position, and proximal limb segment in the neutral position, should have embedded coordinate frames which would overlap should they be placed on top of each other. Here the Euler angles equal zero for all three planes of motion. If the distal limb segment is then moved out of the neutral position, a new set of Euler angles are generated and will quantify the rotations, relative to the proximal axes, which have occurred in the movement. Where movement occurs in three planes there are six possible axis rotation sequences, in gait analysis the ZXY Euler sequence is adopted. This sequence orders internal-external rotation first, medial-lateral flexion second, and flexion-extension third. A compound rotation matrix is then automatically calculated by CODAmotion segmental analysis (Charnwood Dynamics Ltd., Leicestershire, UK), enabling the quantification of three dimensional segment rotations.

Hip joint angles were calculated from the distal thigh embedded coordinate frame and proximal pelvic embedded coordinate frame. Knee joint angles were calculated from the distal shank embedded coordinate frame and proximal thigh embedded coordinate frame. Ankle joint angles were calculated from the distal foot embedded coordinate frame and the proximal shank embedded coordinate frame. Pelvic rotations were calculated from the distal pelvic embedded coordinate frame and the laboratory frame (laboratory calibration). Foot rotations were calculated from the foot embedded coordinate frame and the laboratory frame.

3.7.8 Calculating joint kinetics

In the calculation of joint kinetics the lower limbs are considered as a set of individual free bodies/segments which are connected by joints. Each segment is deemed to have a uniform distribution of mass around its inferior-superior axis connecting two joint centres. The location of the centre of mass on this axis is at a fixed proportion of the segment length from the proximal end. The centre of mass then acts as the origin for a segment embedded coordinate frame, whose axes coincide with the principal axes of the segment. Each segment embedded coordinate frame is dependant on three related marker positions which are used to generate three new orthogonal vectors, whose direction represents the position of the limb segment in relation to the laboratory frame.

Each joint is subject to several linear forces:

Weight due to gravity (acting at the segment mass centre)
Total ground reaction force (acting at the centre of pressure)
Joint reaction (+ anthropometric data = joint moments)
These linear forces are calculated using each free body's segment embedded coordinate frame by the CODA motion software (Charnwood Dynamics Ltd., Leicestershire, UK).

3.8 PREDETERMINED PROTOCOLS

3.8.1 Study 1: The influence of auditory cueing on gait parameters in healthy adults.

Order of Measures

Clinical Assessment
Gait Analysis: Self selected velocity
Gait Analysis: TempoPerfect imposed cadence

Height and body mass were measured before gait analysis. Participants were required to walk at their self selected velocity. Gait cycles were then marked and using the CODAmotion Report Generator (Charnwood Dynamics Ltd., Leicestershire, UK) each individual participant's self selected cadence was then identified. Participants were then required to walk at a cadence imposed by the TempoPerfect (NCH Software, Canberra, Australia) digital metronome. The cadence was set at the individual's self selected cadence.

3.8.2 Study 2: The influence of auditory cueing on gait parameters in healthy children.

Order of Measures

Clinical Assessment
Gait Analysis: Self selected velocity
Gait Analysis: TempoPerfect imposed cadence

Height and body mass were measured before gait analysis. Participants were required to walk at their self selected velocity. Gait cycles were then marked and using the CODAmotion Report Generator (Charnwood Dynamics Ltd., Leicestershire, UK) each individual participant's self selected cadence was then identified. Participants were then required to walk at a cadence imposed by the TempoPerfect

(NCH Software, Canberra, Australia) digital metronome. The cadence was set at the individual's self selected cadence.

3.8.3 Studies 3(a) and 3(b): The influence of body mass on gait parameters in adults, with and without controlling for the influence of velocity.

Order of Measures

Height and body mass
Gait Analysis (with and without cue for control group)

Height and body mass were recorded before gait analysis was conducted. Adults who were overweight, and adults of a healthy weight, were required to walk at their self selected velocity. In addition, adults of a healthy weight were required to walk to a beat set by the TempoPerfect (NCH Software, Canberra, Australia) digital metronome. The beat was set to their age, gender and height matched participant who was overweight's self selected cadence.

3.8.4 Studies 4(a) and 4(b): The influence of body mass on gait parameters in children, with and without controlling for the influence of velocity.

Order of Measures

Height and body mass
Gait Analysis (with and without cue for control group)

Height and body mass were recorded before gait analysis was conducted. Children who were overweight, and children of a healthy weight, were required to walk at their self selected velocity. In addition, children of a healthy weight were required to walk to a beat set by the TempoPerfect (NCH Software, Canberra, Australia) digital metronome. The beat was set to their age, gender and height matched participant who was overweight's self selected cadence.

CHAPTER 4 THE INFLUENCE OF AUDITORY CUEING ON GAIT PARAMETERS IN ADULTS

4.1 INTRODUCTION

As discussed previously velocity has been shown to influence gait (Chapter 1, page 18). From the literature review, a reduced velocity is adopted by those who are overweight (Chapter 2, page 30). It therefore appeared necessary to account for velocity in the experimental design. This may be achieved using several techniques such as treadmills, verbal instruction, visual and auditory cueing (Kerrigan et al 1998; Kerrigan et al 2000; Browning & Kram 2007; Kubota et al 2007; Shultz et al 2009; Shultz et al 2010). In order to assess the self selected presentation of gait for those who were overweight, no walking constraints were to be imposed on the experimental group. The use of a treadmill was therefore excluded. The validity and reliability of verbal instruction is difficult to assess as the response to instruction is individual, and the desired velocity may not be achieved. Auditory cueing was selected as the technique for constraining velocity in the healthy control group. However, the influence of auditory cueing on healthy gait is unknown. This study therefore aimed to assess the influence of auditory cueing on gait parameters in healthy adults. This study received ethical approval from the St. James's Hospital/ AMNCH Ethics Committee.

If an auditory cue was found not to influence gait parameters in healthy adults, it could be recommended that, for scientific research in the area of gait, healthy control groups be matched by cadence with the use of an auditory cue. Auditory cueing enables the comparison of gait parameters of healthy and pathological groups at the self selected velocity of the pathological group i.e. a comparison independent of velocity. A direct assessment of the influence of auditory cueing on gait parameters in healthy adults has not previously been investigated.

4.2 OBJECTIVE

The objective of this chapter was to determine the influence of auditory cueing on gait parameters* in healthy adults.

* Gait parameters refers to: temporal-spatial parameters; maximum and minimum three dimensional kinematics at the hip, knee and ankle.

4.3 METHODOLOGY

Staff and student volunteers of the Trinity Centre for Health Sciences, were screened for inclusion/exclusion criteria (Chapter 3, page 51), and recruited for participation (Chapter 3, page 53). The laboratory session was conducted as described in Chapter 3, pages 59-70. The order of measurements is outlined by figure 4.1. At the test session participants were instructed to walk at their own self selected velocity. The participants self selected cadence was then retrieved from their generated gait report (Chapter 3, page 72). Participants were then instructed to walk to a loud auditory cue set to their self-selected cadence as imposed by the TempoPerfect (NCH Software, Canberra, Australia) digital metronome (Chapter 3, page 59). LED markers were not removed between test conditions.

Clinical Assessment	Order of Measures
Gait Analysis: Self selected velocity	
Gait Analysis: TempoPerfect imposed cadence	

Figure 4.1: Order of measures for study 1.

4.4 ANALYSIS

Gait parameters at a participant's self-selected velocity were compared to gait parameters during gait with auditory cueing. In addition, the intra-individual variation was assessed to determine the influence of an auditory cue on gait parameters with respect to natural variation in gait.

Events of the gait cycle were identified as described in Chapter 3, page 71. Left and right stride data graphs containing three-dimensional kinematics for the hip, knee and ankle for each participant were exported as text and transferred to Microsoft Excel 2003. Maximum and minimum joint range was extracted for the hip, knee and ankle in three-dimensions. In addition a report generated by the CODA software (Charnwood Dynamics Ltd., Leicestershire, UK) provided the parameters velocity, stride length and cadence (Chapter 3, page 72).

Data was analysed using Analyse-it for Microsoft Excel (Version 2.20). A 1 way repeated measures analysis of variance (ANOVA) was used to test for a difference in central location (mean) between the three paired samples (self selected velocity trial one, self selected velocity trial two, and auditory cueing trial). An alpha level < 0.05 was considered statistically significant. Where a significant p-value was noted from the repeated measures ANOVA output, post hoc pairwise t-tests with a Holm-Bonferroni were conducted (Holm 1979).

The assumptions of normal distribution and equal variance were assessed with histogram plots, Shapiro-Wilk W test and the f test prior to the conduction of the repeated measures ANOVA. Where data was not normally distributed a Friedman test (non-parametric test for a difference in central location (median) between two or more paired samples) was used. Where a significant p-value was noted from the Friedman test output, post hoc Wilcoxon tests with a Holm-Bonferroni were conducted (Holm 1979).

Finally limit of agreement plots as per Bland and Altman (1986) were generated. These enabled the investigation of the level of agreement between the two trials at the participants self selected velocity, and for the two test conditions. The widths of the limits of agreement between test conditions were interpreted in relation to the width of the limits of agreement for natural variation.

4.5 RESULTS

Fifteen male and fifteen female participants (Age 25.8 ± 8.6 years; Height 173.0 ± 9.4 cm; Body mass 72.9 ± 11.7 kg; BMI 24.3 ± 2.6 kg.m^{-1}) were recruited for the study. Mean values, standard deviations, results of significance testing, and limits of agreement as per Bland and Altman (1986) are presented in Tables (4.1-4.5).

No significant differences for any of the temporal spatial parameters studied were noted for between trials (Table 4.1). A significant p-value was noted for hip extension, knee valgus and external femoral rotation (Table 4.3). Post hoc tests were conducted for these parameters, results are presented in Table 4.4. A significant difference between test conditions was noted for hip extension and external femoral rotation (Table 4.4).

Parameter	SS velocity mean ± SD	SS2 velocity mean ± SD	AC velocity mean ± SD	ANOVA p-value
Velocity (ms^{-1})	1.2 ± 0.2	1.2 ± 0.2	1.2 ± 0.2	p = 0.9
Stride Length (m)	1.4 ± 0.1	1.3 ± 0.1	1.3 ± 0.1	p = 0.7
Cadence (steps/min)	108.5 ±9.8	109.2 ± 9.7	108.2 ± 11.2	p = 0.7

Table 4.1: Statistical analysis of temporal-spatial parameters between trials.
p-value with significance at alpha $\leq$ 0.05. SD = standard deviation.

SS= Self selected velocity trial 1

SS2 = Self selected velocity trial 2

AC = Auditory cue trial

	SS v AC	SS v SS2
Parameter	Limits of agreement	Limits of agreement
Velocity (ms^{-1})	-0.2 – 0.2	-0.2 – 0.2
Stride Length (m)	-0.2 – 0.2	-0.1 – 0.1
Cadence (steps/min)	-14.2 – 13.8	-8.0 – 9.4

Table 4.2: Limits of agreement for temporal-spatial parameters between trials.
SS= Self selected velocity trial 1

SS2 = Self selected velocity trial 2
AC = Auditory cue trial

Plane of movement	Peak value	SS velocity mean ± SD (degrees)	SS2 velocity mean ± SD (degrees)	AC velocity mean ± SD (degrees)	p-value
Sagittal Plane	Hip Flexion	39.1 ± 5.5	38.4 ± 5.5	38.0 ± 5.6	p = 0.2
	Hip Extension	1.2 ± 6.7	1.7 ± 6.7	2.6 ± 6.8	p = 0.0*
	Knee Flexion	69.4 ± 3.7	69.6 ± 3.9	68.9 ± 5.2	*p* = 0.7
	Knee Extension	-4.7 ± 4.7	-5.2 ± 4.3	-4.9 ± 4.2	p = 0.5
	Ankle Dorsiflexion	16.3 ± 3.4	16.4 ± 3.6	16.4 ± 4.1	p = 0.9
	Ankle Plantarflexion	19.2 ± 7.8	19.5 ± 7.1	17.7 ± 5.6	p = 0.2
Frontal Plane	Hip Adduction	5.7 ± 3.1	5.9 ± 2.8	5.2 ± 2.7	p = 0.1
	Hip Abduction	6.9 ± 1.7	6.6 ± 1.8	6.6 ± 2.1	p = 0.6
	Knee Varus	7.7 ± 3.5	7.1 ± 3.6	7.2 ± 3.7	p = 0.5
	Knee Valgus	1.5 ± 3.6	2.4 ± 4.4	2.0 ± 4.2	*p* = 0.05*
	Ankle Supination	15.1 ± 6.3	15.0 ± 7.8	15.4 ± 8.8	p = 0.9
	Ankle Pronation	6.8 ± 5.3	7.6 ± 5.7	8.7 ± 4.7	p = 0.1
Transverse Plane	Internal Femoral Rotation	14.0 ± 5.1	14.4 ± 5.9	13.3 ± 6.0	p = 0.2
	External Femoral Rotation	1.1 ± 5.2	1.1 ± 5.2	2.4 ± 5.6	p = 0.0*
	Internal Tibial Rotation	13.5 ± 7.7	13.4 ± 7.2	14.4 ± 8.8	*p* = 0.4
	External Tibial Rotation	33.3 ± 7.8	32.3 ± 7.0	32.6 ± 8.3	*p* = 0.8
	Foot Internal Alignment	10.6 ± 7.1	9.7 ± 5.8	11.6 ± 8.4	*p* = 0.3
	Foot External Alignment	6.8 ± 6.9	7.7 ± 6.8	6.8 ± 7.8	*p* = 0.4

Table 4.3: Statistical analysis of kinematic parameters between trials. SD = standard deviation. p = ANOVA p-value with significance set at alpha ≤ 0.05. *p* = Friedman test p-value with significance set at alpha ≤ 0.05. * = significance found.

SS= Self selected velocity trial 1
SS2 = Self selected velocity trial 2
AC = Auditory cue trial

Parameter + Trial	p-value	Holm Bonferroni
Hip Extension		
SS v SS2	p = 0.2	not applicable
SS v AC	p = 0.01*	alpha $\leq$0.02
SS2 v AC	p = 0.07	alpha $\leq$0.03
Knee Valgus		
SS v SS2	*p* = 0.2	not applicable
SS v AC	*p* = 0.1	not applicable
SS2 v AC	*p* = 0.6	not applicable
Hip External Rotation		
SS v SS2	p = 0.9	alpha $\leq$0.05
SS v AC	p = 0.02*	alpha $\leq$0.02
SS2 v AC	p = 0.03*	alpha $\leq$0.03

Table 4.4: Post hoc statistical analysis with Holm-Bonferroni correction where applicable. * = significance found.

p = Pairwise t-test p-value.
p = Wilcoxon test p-value.

		SS v AC	SS v SS2
Plane of movement	Peak value	Limits of agreement	Limits of agreement
Sagittal Plane	Hip Flexion	-7.1 – 4.8	-7.5 – 6.0
	Hip Extension	-7.4 – 4.5	-4.6 – 3.6
	Knee Flexion	-8.9 – 7.8	-3.1 – 3.6
	Knee Extension	-5.1 – 5.5	-3.3 – 4.2
	Ankle Dorsiflexion	-4.1 – 4.5	-3.2 – 3.5
	Ankle Plantarflexion	-8.8 – 11.7	-12.5 – 11.9
Frontal Plane	Hip Adduction	-4.2 – 3.1	-3.4 – 3.8
	Hip Abduction	-2.3 – 3.0	-2.0 – 2.5
	Knee Varus	-3.9 – 3.0	-6.4 – 5.2
	Knee Valgus	-4.3 – 3.3	-9.0 – 7.1
	Ankle Supination	-10.5 – 11.1	-8.9 – 7.0
	Ankle Pronation	-10.4 – 6.6	-11.5 – 9.7
Transverse Plane	Internal Femoral Rotation	-7.9 – 6.4	-6.9 – 7.8
	External Femoral Rotation	-6.9 – 4.3	-6.5 – 6.5
	Internal Tibial Rotation	-9.8 – 8.2	-6.9 – 7.1
	Exteral Tibial Rotation	-6.8 – 8.2	-6.7 – 8.7
	Foot Internal Alignment	-8.3 – 10.2	-10.8 – 10.6
	Foot External Alignment	-9.0 – 8.8	-7.0 – 5.1

Table 4.5: Limits of agreement for kinematic parameters between trials.
SS= Self selected velocity trial 1
SS2 = Self selected velocity trial 2
AC = Auditory cue trial

4.6 DISCUSSION

The aim of the present study was to assess the effect of auditory cueing on gait parameters in healthy adults. Participants were assessed under two conditions, walking first at their self selected velocity, followed by walking to an external auditory cue set to their self selected cadence. An analysis of gait parameters during the two conditions was completed and compared to intra-subject variability. This method enabled an investigation into the effect of auditory cueing on gait parameters with respect to natural variation in gait. No study has previously directly assessed the influence of auditory cueing on gait parameters in healthy adults. If a similar level of agreement were to be noted between test conditions as that noted for intra-individual variation then the use of an auditory cue to dictate cadence in control groups will be recommended for the research setting.

In the analysis of temporal-spatial characteristics no significant differences were noted between the trials (Table 4.1). For between test conditions velocity and stride length exhibited similar width in their limits of agreement as for intra-subject variation (Table 4.2). An auditory cue had no effect on velocity and stride length beyond that explained by natural variation in gait. An increase in the widths of the limits of agreement for cadence was noted with the imposition of an auditory cue (Table 4.2). This suggests that the imposition of a cue altered the step rate of healthy individuals, despite being paced at their self selected cadence. While a difference was noted, the difference was not consistent as evident by the lack of significant difference between the two test conditions (Tables 4.1).

Bertram & Ruina (2002) noted an altered cadence-velocity relationship when comparing walking with an auditory cue, and walking on a treadmill. Similarly, the cadence-velocity relationship of this study appears to be influenced by the use of an auditory cue. In the present study the value for cadence and not velocity, were altered with the use of an auditory cue i.e. the level of disagreement was not linear between cadence and velocity (Table 4.2). This was not anticipated based on the reported constant linear association between velocity, cadence and step length (Schwartz et al 2008). Furthermore cadence was set to the self selected velocity of each individual. Despite this with the implementation of an auditory cue, participants were not able to maintain their self-selected cadence i.e. participants were not capable of maintaining the rhythm of the set cadence. As the set cadence was based on participants' self selected cadence, it is reasonable to suggest that for a cadence slower/faster than self-selected, healthy individuals may struggle further to keep to the prescribed stepping rate.

For the kinematic analysis, a significant p-value was noted for hip extension, knee valgus and external femoral rotation. This implies at least two of the three trials for these parameters have different means (Table 4.3). On completion of post hoc pairwise testing no significant difference was noted between pairs for knee valgus (Table 4.4). Mean hip extension and external femoral rotation increased significantly with the addition of an auditory cue (Table 4.4).

In the sagittal plane the limits of agreement were identified as marginally wider for peak hip extension, knee extension, and ankle dorsiflexion with the addition of an auditory cue as compared to natural variation (Table 4.5). Interestingly the level of agreement was found to be less for peak hip flexion and ankle plantarflexion for natural variation (Table 4.5). Knee flexion demonstrated an increase in the limits of agreement for between test conditions (Table 4.5). There was no significant bias demonstrated and so the increased variability may be deemed to be random (Table 4.5). Richards (1999) reported on the accuracy of optical based systems in measuring angles in motion, and found that the systems assessed demonstrated an accuracy of within 1.5° of the actual value. With this in mind, when the limits of agreement for between test conditions are examined in relation to the limits of agreement for natural variation the results are comparable (Table 4.5). Natural variation in gait explains the discrepancy in agreement for the sagittal plane between test conditions.

In the frontal plane similar results were found in the limits of agreement between test conditions and intra-subject variability for peak hip abduction and peak hip adduction (Table 4.5). The imposition of an auditory cue does not have an effect on peak hip values in the frontal plane. For knee varus peak, valgus peak and supination peak the limits of agreement were in fact wider for natural variation in gait (Table 4.5). As for hip flexion and ankle plantarflexion, the imposition of a cue led to a reduction in the variability of these parameters (Table 4.5). Ankle pronation was found to exhibit a reduction in the level of agreement between test conditions with the imposition of an auditory cue (Table 4.5). The results in the frontal plane suggest that the imposition of an auditory cue in healthy adults does not result in an increase in variation beyond that of natural variation.

For the transverse plane the level of agreement was greater for between test conditions at the hip, for external tibial rotation, and internal foot alignment compared to intra-subject variation (Table 4.5). Internal tibial rotation and external foot alignment demonstrated an increased level of disagreement with the addition of an auditory cue (Table 4.5).

For the kinematic variables assessed in the current study, hip extension, knee flexion, ankle supination, internal tibial rotation and external foot alignment appeared to be influenced marginally by the imposition of an auditory cue (Table 4.5). Whilst these results are evident, it should be noted that several parameters exhibited a reduction in the level of variability with the imposition of an auditory cue. These included peak hip flexion, ankle plantarflexion, knee varus, knee valgus, supination, hip internal rotation, hip external rotation, external tibial rotation and internal foot alignment (Tables 4.4-4.5). Auditory cueing can be recommended for healthy adults in the research setting to enable comparisons at a similar cadence to pathological group which they are matched to. In the clinical setting it may not be feasible to match every patient to an age, gender, height and cadence healthy control subject. As a result, the findings of the current chapter may only be applicable to the research setting. In addition auditory cueing does not address the problem of asymmetrical gait presentations as seen in several pathological groups.

4.7 CONCLUSION

This study aimed to assess the influence of auditory cueing on temporal, spatial and kinematic parameters in healthy adults. From the results the imposition of an auditory cue has no effect on velocity or stride length. A difference was noted in the velocity-cadence relationship during gait with an auditory cue. Despite this, few differences were noted in the limits of agreement for three-dimensional kinematic parameters. In fact the imposition of an auditory cue resulted in a decrease in agreement for several parameters. Auditory cueing during gait does not result in altered gait in healthy adults. In the research setting it is often difficult to differentiate the effects of pathology from the effects of velocity on gait parameters. With the use of auditory cueing in research trials, the cadence of healthy matched subjects should be set at their pathological counterparts' cadence. This enables an analysis of the influence of pathology on gait independent of the confounding influence of velocity.

CHAPTER 5 THE INFLUENCE OF AUDITORY CUEING ON GAIT PARAMETERS IN CHILDREN

5.1 INTRODUCTION

This study aimed to assess the influence of auditory cueing on gait parameters in healthy children. This study received ethical approval from the St. James's Hospital/ Adelaide and Meath Hospital incorporating the National Children's Hospital Research Ethics Committee.

5.2 OBJECTIVE

The objective of this chapter was to determine the influence of auditory cueing on gait parameters* in healthy children.

* Gait parameters refers to: temporal-spatial parameters; maximum and minimum three dimensional kinematics at the hip, knee and ankle.

5.3 METHODOLOGY

Children who matched the inclusion/exclusion criteria (Chapter 3, page 51) were recruited from local schools as described in Chapter 3, page 53. The laboratory session was conducted as described in Chapter 3, pages 59-70. The order of measurements is outlined by figure 5.1. At the test session participants were instructed to walk at their own self selected velocity. The participants self selected cadence was then retrieved from their generated gait report (Chapter 3, page 72).

Participants were then instructed to walk to a loud auditory cue set to their self-selected cadence as imposed by the TempoPerfect (NCH Software, Canberra, Australia) digital metronome (Chapter 3, page 59). LED markers were not removed between test conditions.

Order of Measures

Clinical Assessment
Gait Analysis: Self selected velocity
Gait Analysis: TempoPerfect imposed cadence

Figure 5.1: Order of measures for study 2.

5.4 ANALYSIS

The analysis was completed as described in Chapter 4, page 78.

5.5 RESULTS

Ten children were recruited for participation (2 male: 8 female; Age 15.0 ± 1.6 years; Height 166.4 ± 8.9 cm; Body mass 59.1 ± 13.1 kg; BMI 21.1 ± 2.6 kgm^{2-1}). Mean values, standard deviations, results of significance testing, and limits of agreement as per Bland and Altman are presented in Tables 5.1-5.4.

No significant differences for any of the temporal spatial parameters studied were noted for between trials (Table 5.1). A significant p-value was noted for internal tibial rotation (Table 4.3). Post hoc pairwise tests were conducted for internal tibial rotation. For intra individual variation $p = 0.3$, for self selected velocity trial one compared to auditory cueing $p = 0.2$, and for self selected velocity trial one compared to auditory cueing $p = 0.04$. Holm-Bonferroni significance was calculated to be $p \leq 0.02$. Therefore, no significant differences between trials were noted from the post hoc tests.

Parameter	SS velocity mean ± SD	SS2 velocity mean ± SD	AC velocity mean ± SD	ANOVA p-value
Velocity (ms^{-1})	1.2 ± 0.1	1.3 ± 0.2	1.3 ± 0.1	p = 0.5
Stride Length (m)	1.3 ± 0.1	1.3 ± 0.1	1.3 ± 0.1	p = 0.6
Cadence (steps/min)	113.0 ± 4.4	112.4 ± 7.4	114.8 ± 5.6	p = 0.1

Table 5.1: Statistical analysis of temporal-spatial parameters between trials. p-value with significance at alpha $\leq$ 0.05. SD = standard deviation.

SS= Self selected velocity trial 1
SS2 = Self selected velocity trial 2
AC = Auditory cue trial

	SS v AC	SS v SS2
Parameter	Limits of agreement	Limits of agreement
Velocity (ms^{-1})	-0.2 – 0.3	-0.1 – 0.2
Stride Length (m)	-0.2 – 0.3	-0.1 – 0.1
Cadence (steps/min)	-2.6 – 6.3	-8.4 – 7.3

Table 5.2: Limits of agreement for temporal-spatial parameters between trials.

Plane of movement	Peak value	SS velocity mean ± SD (degrees)	SS2 velocity mean ± SD (degrees)	AC velocity mean ± SD (degrees)	ANOVA p-value
Sagittal Plane	Hip Flexion	35.1 ± 5.0	35.6 ± 4.7	35.8 ± 4.4	p = 0.8
	Hip Extension	6.1 ± 4.7	6.1 ± 4.7	5.5 ± 5.7	p = 0.8
	Knee Flexion	71.1 ± 4.0	70.4 ± 4.0	70.4 ± 3.0	p = 0.6
	Knee Extension	-9.2 ± 3.2	-8.8 ± 2.8	-9.0 ± 3.4	p = 0.8
	Ankle Dorsiflexion	16.2 ± 2.4	16.3 ± 2.9	16.1 ± 2.4	p = 0.9
	Ankle Plantarflexion	19.7 ± 5.2	18.8 ± 6.2	18.4 ± 4.5	p = 0.8
Frontal Plane	Hip Adduction	4.5 ± 2.4	4.4 ± 2.1	4.7 ± 2.1	p = 0.3
	Hip Abduction	8.1 ± 2.3	8.7 ± 1.8	8.7 ± 2.6	p = 0.7
	Knee Varus	4.5 ± 3.0	4.9 ± 3.0	4.2 ± 3.6	p = 0.2
	Knee Valgus	9.6 ± 5.1	12.2 ± 6.0	10.0 ± 5.8	p = 0.5
	Ankle Supination	15.4 ± 5.2	13.3 ± 4.5	15.6 ± 5.8	p = 0.3
	Ankle Pronation	4.7 ± 2.5	4.7 ± 2.8	5.2 ± 3.0	p = 0.2
Transverse Plane	Internal Femoral Rotation	11.7 ± 4.6	12.1 ± 4.1	11.8 ± 5.1	p = 0.6
	External Femoral Rotation	3.0 ± 4.5	3.0 ± 4.4	4.1 ± 4.4	p = 0.1
	Internal Tibial Rotation	8.8 ± 5.0	8.5 ± 4.5	9.9 ± 5.0	p = 0.05*
	External Tibial Rotation	25.9 ± 4.6	26.8 ± 4.7	26.5 ± 4.8	p = 0.6
	Foot Internal Alignment	9.1 ± 4.6	9.8 ± 6.1	9.2 ± 4.6	p = 0.8
	Foot External Alignment	11.7 ± 5.8	11.7 ± 4.2	10.3 ± 4.7	p = 0.5

Table 5.3: Statistical analysis of kinematic parameters between trials. SD = standard deviation. p-value with significance set at alpha $\leq$ 0.05. * = significance found.

SS= Self selected velocity trial 1
SS2 = Self selected velocity trial 2
AC = Auditory cue trial

		SS v AC	SS v SS2
Plane of movement	Peak value	Limits of agreement	Limits of agreement
Sagittal Plane	Hip Flexion	-6.1 – 7.7	-3.3 – 4.2
	Hip Extension	-6.5 -7.7	-3.6 – 3.5
	Knee Flexion	-6.4 – 5.0	-4.7 – 3.4
	Knee Extension	-4.1 – 3.7	-4.0 – 3.2
	Ankle Dorsiflexion	-4.5 – 4.2	-4.0 – 4.2
	Ankle Plantarflexion	-8.8 – 11.4	-6.8 – 8.5
Frontal Plane	Hip Adduction	-2.1 – 1.6	-1.7 – 1.8
	Hip Abduction	-1.7 – 3.1	-2.1 – 3.4
	Knee Varus	-3.0 – 2.4	-1.3 – 2.0
	Knee Valgus	-3.4 – 2.6	-1.9 – 1.9
	Ankle Supination	-9.0 – 9.5	-9.8 – 5.6
	Ankle Pronation	-8.9 – 8.1	-8.4 – 3.2
Transverse Plane	Internal Femoral Rotation	-2.6 – 2.8	-2.0 – 2.9
	External Femoral Rotation	-4.4 – 2.2	-3.6 – 3.5
	Internal Tibial Rotation	-5.3 – 3.1	-1.8 – 2.5
	External Tibial Rotation	-7.6 – 6.5	-3.7 – 1.9
	Foot Internal Alignment	-5.8 – 5.9	-6.5 – 7.9
	Foot External Alignment	-9.3 – 12.1	-5.3 – 5.4

Table 5.4: Limits of agreement for kinematic parameters between trials

5.6 DISCUSSION

From chapter 4, it was concluded that auditory cueing should be used to enable velocity matching for gait research in adults. The present chapter aimed to determine the effect of auditory cueing on gait parameters in healthy children. As in chapter 4, an analysis of gait parameters during the two conditions was completed and compared to intra-subject variability.

No significant difference between the mean values for each trial was noted for the temporal-spatial parameters assessed (Table 5.1). Limits of agreement were wider for velocity and stride length between test conditions compared to intra-individual variation (Table 5.2). Limits of agreement for cadence were in fact narrower with the addition of an auditory cue as compared to natural variation (Tables 5.1-5.2).

For the kinematic analysis, no significant differences in parameters were noted for intra-individual variation, or with the addition of an auditory cue for all parameters studied (Table 5.3).

In the sagittal plane for knee extension and ankle dorsiflexion the width of the limits of agreement for intra-individual variation was similar to that seen for the addition of an auditory cue (Table 5.4). The remaining parameters were marginally wider with the addition of an auditory cue as compared to natural variation (Table 5.4). As for the results of chapter 4, when the accuracy of optical based systems is taken into account, the additional error induced by auditory cueing may not be deemed to be of consequence (Richards 1999).

As for adults similar results were found for peak hip abduction and peak hip adduction for frontal plane limits of agreement, between test conditions and intra-individual variability (Table 5.4). The imposition of an auditory cue does not have an effect on peak hip values in the frontal plane. At the knee and ankle, limits of agreement were wider with the addition of an auditory cue (Table 5.4). The largest difference was noted at the ankle, the limits of agreement were wider for gait with an auditory cue as compared to natural variation (Table 5.4). As for the sagittal plane, the additional variability induced by auditory cueing is not in excess of that anticipated from natural variation.

In the transverse plane at the hip, the addition of an auditory cue appeared to have no effect beyond that seen for natural variation (Table 5.4). Larger differences were seen in the width of the limits of agreement at the knee and ankle between test conditions (Table 5.4). At the ankle, the addition of an auditory cue in fact reduced the degree of variation that was seen for intra-individual variation (Table 5.4).

5.7 CONCLUSION

This study aimed to assess the influence of auditory cueing on temporal, spatial and kinematic parameters in healthy children. From the results it appears that the cue successfully controlled velocity as no differences were seen for velocity between the conditions. In addition, for all three planes no significant differences were noted for kinematic parameters. Limits of agreement were seen to be marginally wider with the addition of an auditory cue for some but not all variables. Auditory cue should be used to dictate velocity in healthy children in the research setting.

CHAPTER 6 THE INFLUENCE OF BODY MASS ON GAIT PARAMETERS IN ADULTS

6.1 INTRODUCTION

From evidence assessing the influence of increased adiposity on gait, studies have reported an array of significant differences in gait parameters between healthy and overweight groups (Chapter 2, page 30). These significant differences were however, not clearly defined.

There was poor consistency as to the kinematic and kinetic parameters selected for analysis. This made it hard to construct a comprehensive description of the influence of body mass on gait (Chapter 2, page 30). Few studies reported on the components of joint range/joint moments/joint power (Hills & Parker 1991a, Gushue et al 2005, Nantel et al 2006, Schultz et al 2009, Schultz et al 2010, McMillan et al 2009, McMillan et al 2010). In addition no study reported on pelvic kinematics which, as described in Chapter 3 page 73, are used in the calculation of hip joint kinematics. These are then used in the calculation of joint kinematics more distally. It would appear necessary therefore to include an analysis of the pelvic presentation of overweight groups.

Several studies used cinefilm (Hills & Parker 1991c; Spyropoulos et al 1991), passive (Nantel et al 2006; Vismara et al 2007; Lai et al 2008; Lee et al 2009; McMillan et al 2009; Schultz et al 2009; Schultz et al 2010; Ko et al 2010; Cimolin et al 2011) and active (Gushue et al 2005; Segal et al 2009; McMillan et al 2010) marker systems which would be considered to be the most accurate tools for motion analysis. Several studies used less accurate techniques however, such as plantar printing tests, video analysis, and equipment restricted to temporal-spatial parameters (Hills & Parker 1991a, Hills & Parker 1991b, McGraw et al 2000, DeVita & Hortobagyi 2003, DeSouza et al 2005, Browning & Kram 2007, Morrisson et al 2008). Methodological differences between studies furthered the difficulty in drawing definitive conclusions.

A reduction in gait velocity is a feature associated with increased adiposity. As described previously in Chapter 1 page 18, a change in velocity is associated with changes in several aspects of gait. It was difficult therefore to differentiate velocity effects from adiposity effects on gait parameters in the previous literature. This thesis proposed to provide a description of the relationship between increased adiposity and gait while accounting for the compounding influence of velocity on gait. In Chapter 4 page 76 and Chapter 5 page 88, the reliability of using an auditory cue to dictate velocity in healthy individuals was assessed. For adults and children, the null hypothesis of 'no difference in gait parameters between self selected velocity and self selected velocity dictated by an auditory cue' was accepted. An auditory cue has no impact on temporal-spatial and kinematic parameters, beyond that seen for natural variation in adults and children. From these results when assessing adults and children, it is recommended to control velocity in experimental design.

From the literature review altered gait parameters seen in individuals who were overweight were said to be due to methodological differences between studies, velocity, true differences, or a combination of all three (Chapter 2, page 30). While it is possible to control velocity during gait analysis in adults and children, an investigation into the influence of increased adiposity on gait as compared to healthy individuals, both at their self selected velocity, is still warranted. Individuals who are overweight may choose to ambulate at a reduced velocity, and this reduction may lead to, or enhance an altered presentation of gait.

6.2 OBJECTIVES

The objectives of this chapter were:

To determine the influence of body mass on gait parameters† in adults.

To determine the influence of body mass on gait parameters† in adults, while accounting for the confounding influence of velocity.

† Gait parameters refer to: temporal-spatial parameters; three dimensional kinematics at the pelvis, hip, knee and ankle; three dimensional joint moments at the hip, knee and ankle; maximum absolute and normalised anterior, medial and vertical ground reaction forces; and, maximum and minimum total joint power at the hip, knee and ankle.

6.3 METHODOLOGY

A cross sectional study design as described in Chapter 3, page 51 was selected. Overweight adults who matched the inclusion/exclusion criteria (Chapter 3, page 51) were recruited from the hypertension clinic at St. James's Hospital as described in Chapter 3, page 53. The control group was recruited from the Trinity Centre for Health Sciences as described in Chapter 3, page 53. Participants were required to attend the gait laboratory on one occasion. Measurements were completed in the order outlined in figure 6.1. Each gait laboratory session was completed as described in Chapter 3, pages 59-70. Control healthy weight participants were matched by age, gender, and height. Both groups were asked to walk at their natural, self selected pace (Study 3 (a)). In addition, subjects in the control group were asked to walk in time to a beat, set to the cadence of the overweight adult whom they were matched to (Study 3(b)). The beat was sounded using the TempoPerfect (NCH Software, Canberra, Australia) digital metronome (Chapter 3, page 59).

Order of Measures
Height and body mass
Gait Analysis (with and without cue for control group)

Figure 6.1: Order of measures for study 3.

6.4 ANALYSIS

6.4.1 Data preparation

Once data acquisition was completed the gait cycles were identified and a report generated as described in section 3.7.6. This report provided the temporal-spatial parameters selected for analysis namely velocity, stride length, cadence, percentage stance, single support time, and double support time.

A 'data' graph was created for the 'BilatGaitAcqCheck.stp' setup in CODAmotion Analysis (Charnwood Dynamics Ltd., Leicestershire, UK). This graph included three dimensional, right and left segment rotation plots for the pelvis, hip knee and ankle; three dimensional, right and left ground reaction force plots; three dimensional, right and left joint moments for the hip, knee and ankle; and the total joint power for the left and right hip, knee and ankle. Right and left cursors were moved to the pre marked initial contact and next initial contact for the left leg. The graph was zoomed into this area which was then exported as text. The text file was copied and pasted into an Excel spreadsheet. This was then repeated for the right leg.

Three-dimensional kinematic and kinetic parameters at the seven events of the gait cycle (Figure 1.6) were included in the analysis. Initial contact, opposite toe-off, opposite initial contact, toe-off, and next initial contact were identified as described in section 3.7.6. Heel rise was identified using the stick figure in CODAmotion analysis (Charnwood Dynamics Ltd., Leicestershire, UK), and as occurring at the dorsiflexion peak of the associated limb. Feet adjacent was identified using the stick figure. Tibia vertical was identified using the stick figure and the associated hip flexion peak. The time at which each event occurred was documented for the left and right cycles respectively. Using the whole time period to represent 100%, the time at which each event took place was used to determine at what percentage of the gait cycle each event occurred.

For a left side Excel file all right sided parameters were deleted. For a right sided Excel file all left sided parameters were deleted. The maximum and minimum values for power and ground reaction force parameters in the file were calculated at this point. Each time point identified previously was then located in the Excel files, highlighted, and all other data on the spreadsheet deleted. Left and right files for each subject were then averaged.

For each group (overweight group; healthy weight group; healthy weight group with auditory cue) three master files were created entitled temporal-spatial characteristics, events of the gait cycle, and kinetics maximum and minimum. The files contained the averaged left and right data for each participant.

6.4.2 Data analysis

Mean and standard deviations for each group were calculated. Mean differences, or bias between the groups, were also calculated to determine the magnitude of any significant bias. Reported biases represent the value for the healthy weight group minus the value for the overweight group. Statistical analysis was then conducted.

A statistical significance test asks if the mean difference of two sets of measures is different from zero, and if this difference is due to more than chance variation. An independent t test is often used in data analysis. This test is dependent on two assumptions, that the data sets are normally distributed and exhibit equal variance. Prior to the conduction of this test these assumptions were assessed with histogram plots, Shapiro-Wilk W test and the f test. Analyse-it for Microsoft Excel (Version 2.0) was used for the analysis.

For normally distributed data with equal variance an independent t-test with equal variance was calculated. For variables which exhibited a normal distribution but not equal variance, an independent t-test without normal variance was calculated. For non-normally distributed data with/without equal variance a Mann Whitney U (non parametric) significance test was calculated. For significance testing alpha < 0.05 was considered statistically significant. Graphical presentations were also generated to depict kinematics and joint moments throughout the motion cycle.

6.5 RESULTS STUDY 3(a)

Results presented in the following section represent the analysis of gait parameters at the self selected velocity of participants who were overweight compared to gait parameters at the self selected velocity of participants of a healthy weight (Study 3(a)).

6.5.1 Baseline characteristics

25 adults who were overweight (male: female, 15:10) and 25 adults of a healthy weight (male: female, 15:10) completed the study. 25 pairs were matched by age and gender. 24 pairs were matched by height. Baseline demographics are outlined in Table 6.1. Of the 25 participants in the experimental group, 13 were classified as obese while 12 were classified as overweight (Chapter 1, page 3). A significant difference in body mass, BMI and thigh girth was noted between the groups ($p < 0.002$). No significant difference for age and height were noted. Craig test and Q angle results were not analysed. The measures were difficult to complete in the overweight group, and it was felt that the accuracy of measurements was not acceptable.

Parameter	Overweight group (n = 25) (mean ± SD)	Healthy weight group (n = 25) (mean ± SD)
Age (years)	39.2 ± 16.5	39.8 ± 16.8
Height (cm)	172.0 ± 7.8	172.0 ± 7.6
Body mass (kg)	91.5 ± 10.9	69.4 ± 9.4
BMI (kgm^{2-1})	31.0 ± 4.1	23.4 ± 2.3
Thigh Girth (cm)	53.8 ± 4.0	46.0 ± 3.3

Table 6.1: Baseline characteristics. mean ± SD.

Temporal-spatial and kinematic data was collected for 50 adults. Kinetic data was collected for 46 adults. Kinetic data was not collected for two overweight adults due to an equipment fault and therefore two pairs were excluded from the kinetic analysis.

6.5.2 Temporal-spatial parameters

Temporal-spatial characteristics selected for analysis included velocity, stride length, cadence, percentage stance, single support time, and double support time. Mean, standard deviation, bias, and p-value are presented in Table 6.2. A significant difference in percentage stance phase and double support duration were noted between the two groups (Table 6.2). A significant relationship between increasing BMI and velocity was noted (r = 0.35) (p for trend = 0.0*).

Parameter	Overweight group (mean ± SD)	Healthy weight group (mean ± SD)	Bias	p-value
Velocity (ms^{-1})	1.1 ± 0.2	1.2 ± 0.2	0.1	p = 0.1
Stride Length (m)	1.2 ± 0.1	1.3 ± 0.1	0.0	p = 0.3
Cadence ($stepsmin^{-1}$)	106.2 ± 11.0	110.2 ± 10.6	4.0	p = 0.2
Stance phase (%)	63.8 ± 1.3	62.0 ± 1.4	-1.8	p = 0.0*
Single Support (s)	0.4 ± 0.0	0.4 ± 0.0	0.0	p = 0.6
Double Support (s)	0.2 ± 0.0	0.1 ± 0.0	0.0	*p* = 0.0*

Table 6.2: Results for temporal-spatial parameters. p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05.
*** = significant difference between groups.**

6.5.3 Timing of events

No significant difference in the timing of initial contact and tibia vertical was noted ($p>0.05$). A significant difference in the timing of opposite toe off, heel rise, opposite initial contact, toe off and feet adjacent were noted between the two groups ($p<0.05$). Each event occurred later for adults who were overweight.

6.5.4 Kinematics

Mean, standard deviation, bias, and p-value were calculated for the pelvis, hip, knee and ankle in the sagittal, frontal and transverse planes for each of the seven events. Results are presented in Tables 6.3- 6.14 and Figures 6.2-6.4.

6.5.4.1 Sagittal plane

Adults who were overweight exhibited significantly greater hip flexion at initial contact (Table 6.4) (Figure 6.2). A significant difference in knee flexion was noted at tibia vertical (Table 6.5). No other significant differences were noted between groups (Tables 6.3-6.6).

Event	Overweight group mean ± SD (degrees)	Healthy weight group mean ± SD (degrees)	Bias	p-value
Initial contact	11.8 ± 4.4	9.6 ± 4.8	-1.3	p = 0.1
Opposite toe off	10.2 ± 4.3	9.0 ± 4.8	-1.2	p = 0.4
Heel rise	11.8 ± 4.5	10.1 ± 4.7	-1.0	p = 0.2
Opposite initial contact	11.9 ± 4.4	9.9 ± 4.8	-1.9	p = 0.1
Toe off	10.3 ± 4.4	9.3 ± 4.7	-1.7	p = 0.5
Feet adjacent	11.3 ± 4.0	10.0 ± 4.6	-1.2	p = 0.3
Tibia vertical	11.7 ± 4.4	10.4 ± 4.6	-2.2	p = 0.3

Table 6.3: Results for pelvis anterior-posterior tilt (+/-).p = t-test p-value with significance at alpha $\leq$ 0.05.

Event	Overweight group mean ± SD (degrees)	Healthy weight group mean ± SD (degrees)	Bias	p-value
Initial contact	37.7 ± 6.7	34.1 ± 6.1	-3.5	*p* = 0.0*
Opposite toe off	32.1 ± 6.8	30.0 ± 6.8	-2.1	p = 0.3
Heel rise	3.4 ± 7.9	1.0 ± 6.1	-2.4	p = 0.2
Opposite initial contact	2.1 ± 8.2	-1.1 ± 5.9	-3.1	p = 0.1
Toe off	8.6 ± 8.6	6.6 ± 6.2	-2.1	p = 0.3
Feet adjacent	34.6 ± 6.9	33.3 ± 5.7	-1.3	p = 0.5
Tibia vertical	37.5 ± 6.9	36.1 ± 5.4	-1.4	*p* = 0.3

Table 6.4: Results for hip flexion-extension (+/-).p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

Event	Overweight group mean ± SD (degrees)	Healthy weight group mean ± SD (degrees)	Bias	p-value
Initial contact	10.7 ± 7.2	10.5 ± 4.2	-0.3	p = 0.9
Opposite toe off	24.1 ± 6.7	23.2 ± 5.7	-1.0	p = 0.6
Heel rise	14.7 ± 7.3	14.8 ± 4.2	0.1	*p* = 0.7
Opposite initial contact	17.2 ± 6.7	18.6 ± 3.8	1.4	*p* = 0.2
Toe off	49.6 ± 6.4	49.9 ± 4.5	0.3	p = 0.8
Feet adjacent	53.0 ± 5.8	55.0 ± 4.3	2.0	*p* = 0.2
Tibia vertical	26.2 ± 4.6	24.6 ± 2.8	-1.6	*p* = 0.0*

Table 6.5: Results for knee flexion-extension (+/-).p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

Event	Overweight group mean ± SD (degrees)	Healthy weight group mean ± SD (degrees)	Bias	p-value
Initial contact	1.8 ± 4.5	1.7 ± 3.0	0.0	p = 0.9
Opposite toe off	2.7 ± 3.7	2.5 ± 3.0	0.7	p = 0.8
Heel rise	15.5 ± 3.6	14.9 ± 2.8	-1.1	p = 0.5
Opposite initial contact	14.2 ± 4.4	12.7 ± 3.8	-1.5	p = 0.2
Toe off	-12.2 ± 8.6	-13.3 ± 4.6	-0.6	p = 0.6
Feet adjacent	4.2 ± 4.6	4.9 ± 3.3	-0.2	*p* = 0.9
Tibia vertical	4.6 ± 2.9	4.6 ± 2.9	-0.2	p = 0.9

Table 6.6: Results for ankle dorsiflexion-plantarflexion (+/-).p = t-test p-value with significance at alpha ≤ 0.05. *p* = Mann Whitney U p-value with significance at alpha ≤ 0.05.
*** = significant difference between groups.**

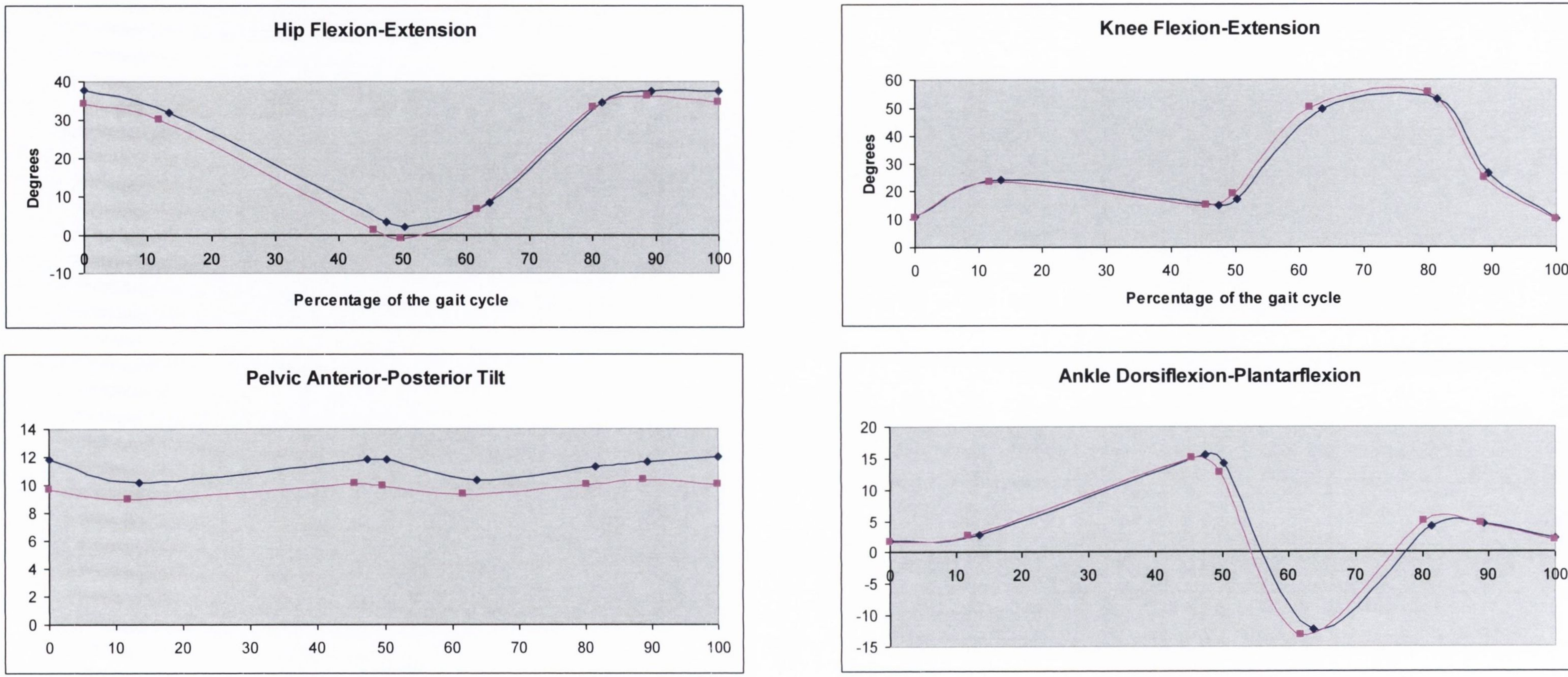


Figure 6.2: Sagittal plane kinematics.

X axis = percentage of one gait cycle.
Y axis = degrees.
The eight events of the gait cycle are demarcated on the lines.

6.5.4.2 Frontal plane

As for the sagittal plane significant changes were seen at the hip and knee (Tables 6.8-6.9) (Figure 6.3). A significant increased in hip abduction was noted for adults who were overweight at toe off and feet adjacent (Table 6.8). A significantly increased knee varus was noted at feet adjacent for adults who were overweight (Table 6.9). No additional significant differences were seen (Tables 6.7-6.10) (Figure 6.3).

Event	**Overweight group mean ± SD (degrees)**	**Healthy weight group mean ± SD (degrees)**	**Bias**	**p-value**
Initial contact	-0.7 ± 0.9	-0.4 ± 1.0	0.4	p = 0.2
Opposite toe off	2.4 ± 1.1	2.5 ± 1.8	0.2	p = 0.7
Heel rise	0.8 ± 1.0	0.6 ± 1.0	-0.2	p = 0.6
Opposite initial contact	0.7 ± 1.0	0.5 ± 1.1	-0.2	*p* = 0.2
Toe off	-2.3 ± 1.2	-2.3 ± 1.6	0.0	p = 0.9
Feet adjacent	-0.6 ± 1.2	-0.2 ± 0.9	0.4	*p* = 0.3
Tibia vertical	-0.4 ± 0.9	0.0 ± 0.7	0.4	p = 0.1

Table 6.7: Results for pelvis up-down obliquity (+/-).p = t-test p-value with significance at alpha ≤ 0.05. *p* = Mann Whitney U p-value with significance at alpha ≤ 0.05.
*** = significant difference between groups.**

Event	Overweight group mean ± SD (degrees)	Healthy weight group mean ± SD (degrees)	Bias	p-value
Initial contact	-2.7 ± 2.5	-1.7 ± 2.0	1.1	p = 0.1
Opposite toe off	2.8 ± 2.9	3.4 ± 3.1	0.7	p = 0.5
Heel rise	3.4 ± 2.3	3.6 ± 2.0	0.2	p = 0.7
Opposite initial contact	3.0 ± 2.5	3.0 ± 1.7	0.1	*p* = 0.9
Toe off	-4.9 ± 2.5	-3.4 ± 1.7	1.5	p = 0.0*
Feet adjacent	-3.6 ± 3.1	-1.1 ± 2.3	2.5	p = 0.0*
Tibia vertical	-2.4 ± 3.2	-0.9 ± 2.3	1.5	*p* = 0.1

Table 6.8: Results for hip adduction-abduction (+/-).p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05.
*** = significant difference between groups.**

Event	Overweight group mean ± SD (degrees)	Healthy weight group mean ± SD (degrees)	Bias	p-value
Initial contact	1.6 ± 3.1	1.4 ± 2.3	-0.3	*p* = 0.6
Opposite toe off	2.6 ± 3.3	2.3 ± 3.1	-0.3	*p* = 0.5
Heel rise	1.1 ± 3.3	1.2 ± 2.9	0.0	p = 0.9
Opposite initial contact	1.3 ± 3.4	1.2 ± 3.1	-0.1	*p* = 0.9
Toe off	2.0 ± 3.5	0.9 ± 3.6	-1.0	p = 0.3
Feet adjacent	6.5 ± 3.8	3.4 ± 5.1	-3.1	p = 0.0*
Tibia vertical	3.9 ± 2.8	3.4 ± 3.3	-0.6	p = 0.5

Table 6.9: Results for knee varus-valgus (+/-).p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05.
*** = significant difference between groups.**

Event	Overweight group mean ± SD (degrees)	Healthy weight group mean ± SD (degrees)	Bias	p-value
Initial contact	6.0 ± 7.1	4.9 ± 4.7	-1.1	p = 0.5
Opposite toe off	0.4 ± 8.5	1.4 ± 6.3	1.0	p = 0.7
Heel rise	0.3 ± 8.8	1.3 ± 7.5	1.1	p = 0.7
Opposite initial contact	-1.2 ± 7.2	0.2 ± 6.1	1.4	p = 0.5
Toe off	4.7 ± 5.1	5.0 ± 3.9	0.3	p = 0.8
Feet adjacent	5.3 ± 5.1	4.1 ± 4.6	-1.2	p = 0.4
Tibia vertical	2.6 ± 9.0	1.1 ± 6.1	-1.5	*p* = 0.7

Table 6.10: Results for ankle supination-pronation (+/-).p = t-test p-value with significance at alpha ≤ 0.05. *p* = Mann Whitney U p-value with significance at alpha ≤ 0.05.
*** = significant difference between groups.**

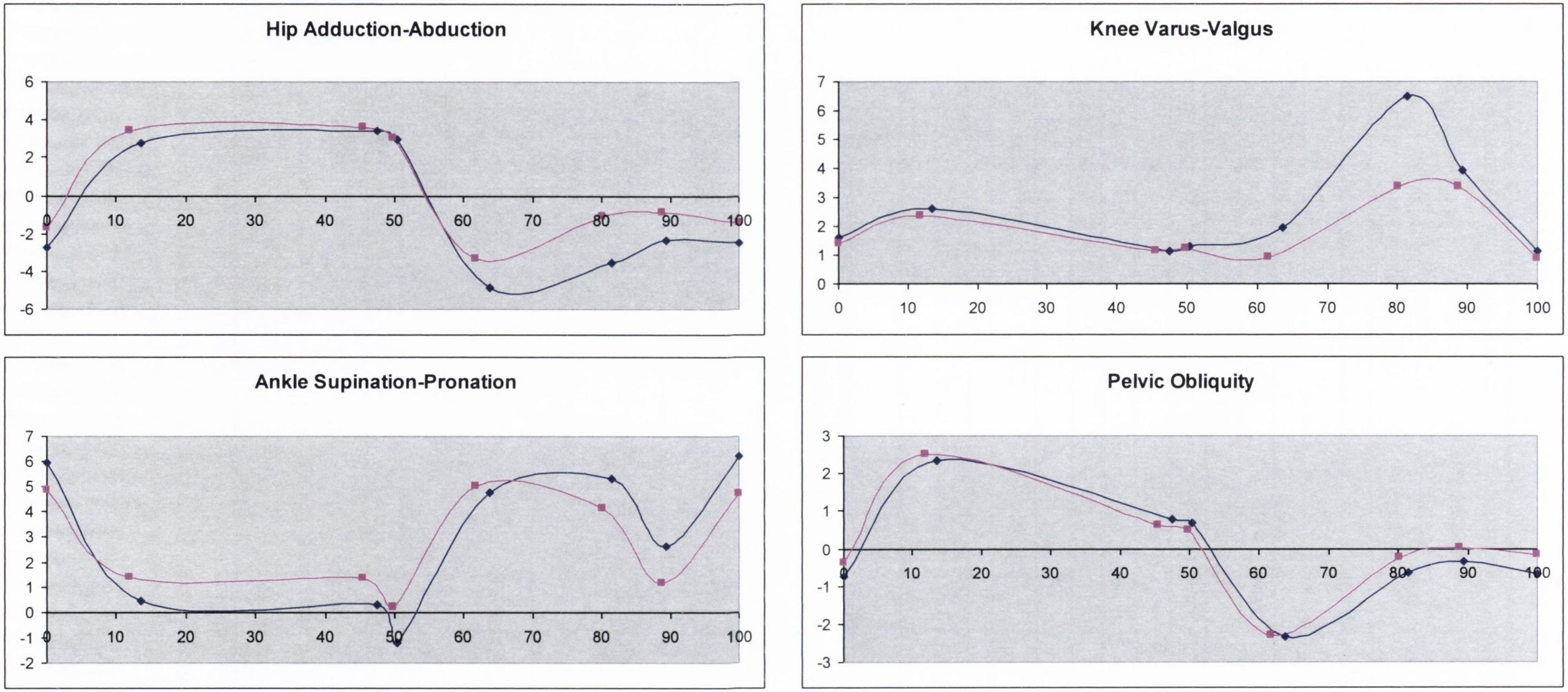


Figure 6.3: Frontal plane kinematics.

X axis = percentage of one gait cycle.

Y axis = degrees.

The eight events are demarcated on the lines.

Overweight Group

Healthy weight Group

6.5.4.3 Transverse plane

As for the sagittal and frontal plane no significant difference in pelvic kinematics in the transverse plane were noted between the two groups (Table 6.11) (Figure 6.4). No significant differences between the two groups were seen at the hip, knee or ankle in the transverse plane (Tables 6.12-6.14) (Figure 6.4).

Event	Overweight group mean ± SD (degrees)	Healthy weight group mean ± SD (degrees)	Bias	p-value
Initial contact	2.5 ± 2.4	2.4 ± 2.8	-0.1	p = 0.9
Opposite toe off	2.3 ± 2.0	2.3 ± 2.4	0.0	p = 1.0
Heel rise	-1.4 ± 2.0	-1.0 ± 3.4	0.5	p = 0.6
Opposite initial contact	-2.2 ± 2.1	-2.1 ± 3.3	0.1	p = 0.9
Toe off	-2.2 ± 1.7	-2.3 ± 2.9	-0.1	*p* = 0.9
Feet adjacent	-2.8 ± 1.2	-3.1 ± 2.0	-0.4	p = 0.5
Tibia vertical	-1.2 ± 1.5	-1.6 ± 2.1	-0.3	p = 0.5

Table 6.11: Results for pelvis forward/backward rotation (+/-).p = t-test p-value with significance at alpha ≤ 0.05. *p* = Mann Whitney U p-value with significance at alpha ≤ 0.05.
*** = significant difference between groups.**

Event	Overweight group mean ± SD (degrees)	Healthy weight group mean ± SD (degrees)	Bias	p-value
Initial contact	5.2 ± 7.8	4.3 ± 6.2	-0.9	p = 0.7
Opposite toe off	5.1 ± 8.8	6.0 ± 7.8	0.9	p = 0.7
Heel rise	2.3 ± 8.6	3.6 ± 7.1	1.3	p = 0.6
Opposite initial contact	2.1 ± 8.1	3.2 ± 7.0	1.1	p = 0.6
Toe off	-2.2 ± 6.1	-0.4 ± 5.8	1.8	p = 0.3
Feet adjacent	2.3 ± 6.0	1.8 ± 6.1	-0.5	p = 0.8
Tibia vertical	3.0 ± 6.8	3.8 ± 5.8	0.8	p = 0.7

Table 6.12: Results for hip internal-external rotation (+/-). p = t-test p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

Event	Overweight group mean ± SD (degrees)	Healthy weight group mean ± SD (degrees)	Bias	p-value
Initial contact	-28.8 ± 8.2	-27.1 ± 7.0	1.7	p = 0.4
Opposite toe off	-21.8 ± 9.0	-20.6 ± 7.1	1.3	p = 0.6
Heel rise	-18.8 ± .8.3	-19.2 ± 7.3	-0.4	p = 0.9
Opposite initial contact	-18.3 ± 8.2	-18.1 ± 7.2	0.2	p = 0.9
Toe off	-18.7 ± 6.6	-19.7 ± 6.1	-1.0	p = 0.6
Feet adjacent	-22.1 ± 5.4	-20.6 ± 7.1	1.5	p = 0.4
Tibia vertical	-27.1 ± 7.4	-26.5 ± 8.2	0.7	p = 0.8

Table 6.13: Results for knee internal-external rotation (+/-).p = t-test p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

Event	Overweight group mean ± SD (degrees)	Healthy weight group mean ± SD (degrees)	Bias	p-value
Initial contact	1.9 ± 6.1	3.6 ± 3.9	1.7	p = 0.2
Opposite toe off	-4.2 ± 6.5	-4.2 ± 4.5	0.0	p = 0.9
Heel rise	-4.4 ± 5.8	-3.2 ± 4.7	1.2	p = 0.4
Opposite initial contact	-2.2 ± 6.1	-0.4 ± 5.2	1.8	p = 0.3
Toe off	7.1 ± 6.5	10.4 ± 5.5	3.3	p = 0.1
Feet adjacent	1.3 ± 6.6	0.5 ± 4.7	-0.8	p = 0.6
Tibia vertical	1.9 ± 6.1	3.1 ± 4.7	1.2	p = 0.5

Table 6.14: Results for ankle internal-external alignment (+/-).p = t-test p-value with significance at alpha ≤ 0.05. * = significant difference between groups.

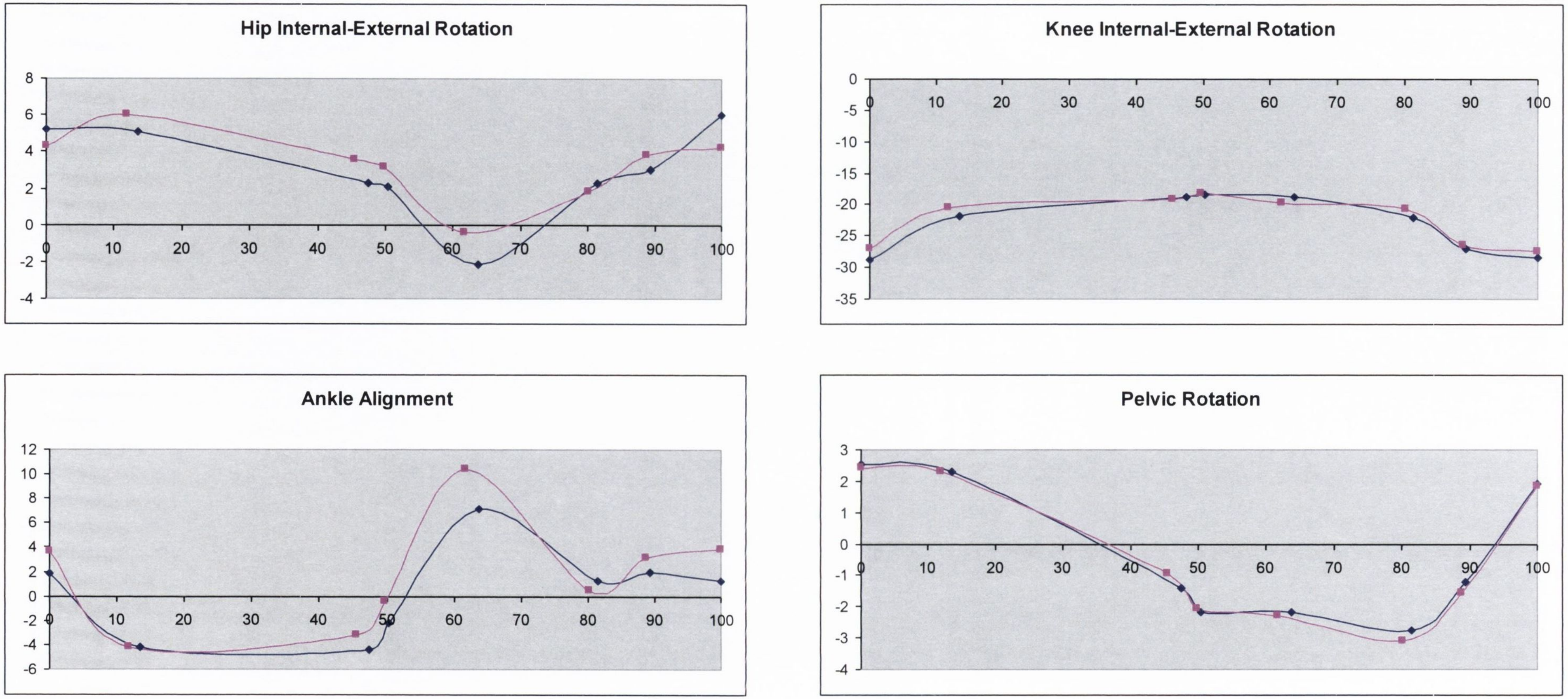


Figure 6.4: Transverse plane kinematics.

X axis = percentage of one gait cycle.

Y axis = degrees.

The eight events are demarcated on the lines.

Overweight Group

Healthy weight Group

6.5.5 Kinetics

Maximum and minimum absolute and normalised by body mass vertical, propulsive and lateral ground reaction forces were assessed. Three dimensional hip, knee and ankle moments normalised by body mass at the seven events of the gait cycle were also selected for analysis. Finally maximum and minimum total hip, knee and ankle power normalised by body mass were assessed. Mean, standard deviation, between group bias, and p-value were calculated and are presented in Tables 6.15-6.27. Values are also presented in Figures 6.5-6.11.

6.5.5.1 Ground reaction forces

Maximum vertical (Z) ground reaction force corrected for body mass was significantly reduced for adults who were overweight (Table 6.25) (Figure 6.5). Maximum propulsive (X) and medial (Y) ground reaction forces corrected for body mass were not significantly different between the two groups (Table 6.15) (Figure 6.5). Maximum absolute vertical (Z), propulsive (X) and medial (Y) ground reaction force was significantly greater for adults who were overweight (Table 6.16) (Figure 6.6).

Parameter (N/kg)	Overweight group (mean ± SD)	Healthy weight group (mean ± SD)	Bias	p-value
Force X	1.7 ± 0.4	1.7 ± 0.3	0.1	p = 0.7
Force Y	0.6 ± 0.2	0.6 ± 0.1	0.0	p = 0.7
Force Z	10.8 ± 0.6	11.2 ± 0.6	0.4	p = 0.0*

Table 6.15: Maximum ground reaction forces. N=Newton. p = t-test p-value with significance at alpha ≤ 0.05. * = significant difference between groups.

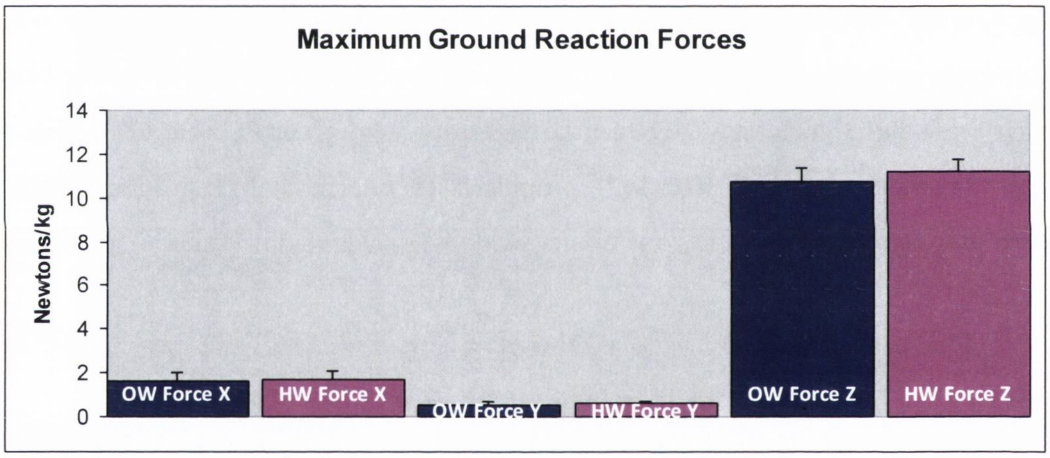


Figure 6.5: Maximum ground reaction forces.

OW = Overweight

HW = Healthy weight

Parameter (N/kg)	Overweight group (mean ± SD)	Healthy weight group (mean ± SD)	Bias	p-value
Force X	151.5 ± 35.5	120.9 ± 25.3	-30.7	*p* = 0.0*
Force Y	53.4 ± 15.7	41.3 ± 9.4	-12.0	p = 0.0*
Force Z	989.5± 114.2	783.1 ± 109.6	-206.4	p = 0.0*

Table 6.16: Maximum absolute ground reaction forces. N=Newton. p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

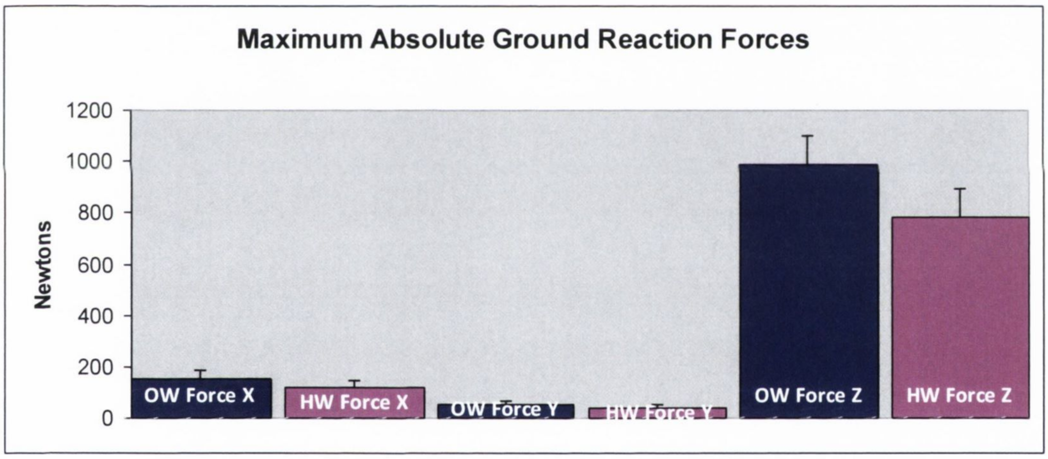


Figure 6.6: Maximum absolute ground reaction forces.

OW = Overweight

HW = Healthy weight

6.5.5.2 Sagittal plane joint moments

There was no significant difference in sagittal plane joint moments at the hip, knee and ankle between groups (Tables 6.17-6.19) (Figure 6.7).

Event	Overweight group mean ± SD ($Nm^{-1}kg^{-1}$)	Healthy weight group mean ± SD ($Nm^{-1}kg^{-1}$)	Bias	p-value
Initial contact	$-3.0x10^{-2}$ ± 0.2	$4.0x10^{-3}$ ± 0.2	0.0	p = 0.5
Opposite toe off	0.7 ± 0.3	0.6 ± 0.3	-0.1	p = 0.5
Heel rise	-0.3 ± 0.4	-0.3 ± 0.2	-0.1	*p* = 0.1
Opposite initial contact	-0.4 ± 0.4	-0.4 ± 0.1	-0.1	*p* = 0.1
Toe off	-0.2 ± 0.1	-0.2 ± 0.2	0.0	p = 0.8
Feet adjacent	-0.1 ± 0.2	-0.1 ± 0.2	0.0	*p* = 0.6
Tibia vertical	$-2.5x10^{-3}$ ± 0.1	$1.3x10^{-2}$± 0.2	0.0	*p* = 0.5

Table 6.17: Results for hip extensor-flexor moments (+/-). Nm = Newton-meter. p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

Event	Overweight group mean ± SD ($Nm^{-1}kg^{-1}$)	Healthy weight group mean ± SD ($Nm^{-1}kg^{-1}$)	Bias	p-value
Initial contact	-0.1 ± 0.1	-0.1 ± 0.1	0.0	p = 0.4
Opposite toe off	0.5 ± 0.2	0.5 ± 0.2	0.1	p = 0.3
Heel rise	0.1 ± 0.2	0.1 ± 0.1	0.0	p = 0.7
Opposite initial contact	0.2 ± 0.1	0.2 ± 0.2	0.0	p = 0.5
Toe off	5.8×10^{-3}± 0.0	-8.1×10^{-3}± 0.0	0.0	p = 0.2
Feet adjacent	-1.7×10^{-2} ± 0.1	-1.8×10^{-2} ± 0.1	0.0	*p* = 0.8
Tibia vertical	-0.1 ± 0.0	-0.1 ± 0.1	0.0	p = 0.6

Table 6.18: Results for knee extensor-flexor moments (+/-).p = t-test p-value with significance at alpha ≤ 0.05. *p* = Mann Whitney U p-value with significance at alpha ≤ 0.05.
*** = significant difference between groups.**

Event	Overweight group mean ± SD ($Nm^{-1}kg^{-1}$)	Healthy weight group mean ± SD ($Nm^{-1}kg^{-1}$)	Bias	p-value
Initial contact	-1.5×10^{-2} ± 0.0	-1.8×10^{-2} ± 0.0	0.0	*p* = 0.5
Opposite toe off	0.3 ± 0.1	0.2 ± 0.2	0.0	*p* = 0.2
Heel rise	1.2 ± 0.1	1.2 ± 0.1	0.1	*p* = 0.2
Opposite initial contact	1.2 ± 0.1	1.2 ± 0.1	0.0	*p* = 0.7
Toe off	-4.6×10^{-3} ± 0.0	-6.0×10^{-3} ± 0.0	0.0	p = 0.4
Feet adjacent	-8.1×10^{-3} ± 0.0	-7.7×10^{-3} ± 0.0	0.0	p =0.5
Tibia vertical	-1.1×10^{-2} ± 0.0	-1.1×10^{-2} ± 0.0	0.0	p = 0.8

Table 6.19: Results for ankle plantarflexor-dorsiflexor moments (+/-).p = t-test p-value with significance at alpha ≤ 0.05. *p* = Mann Whitney U p-value with significance at alpha ≤ 0.05. * = significant difference between groups.

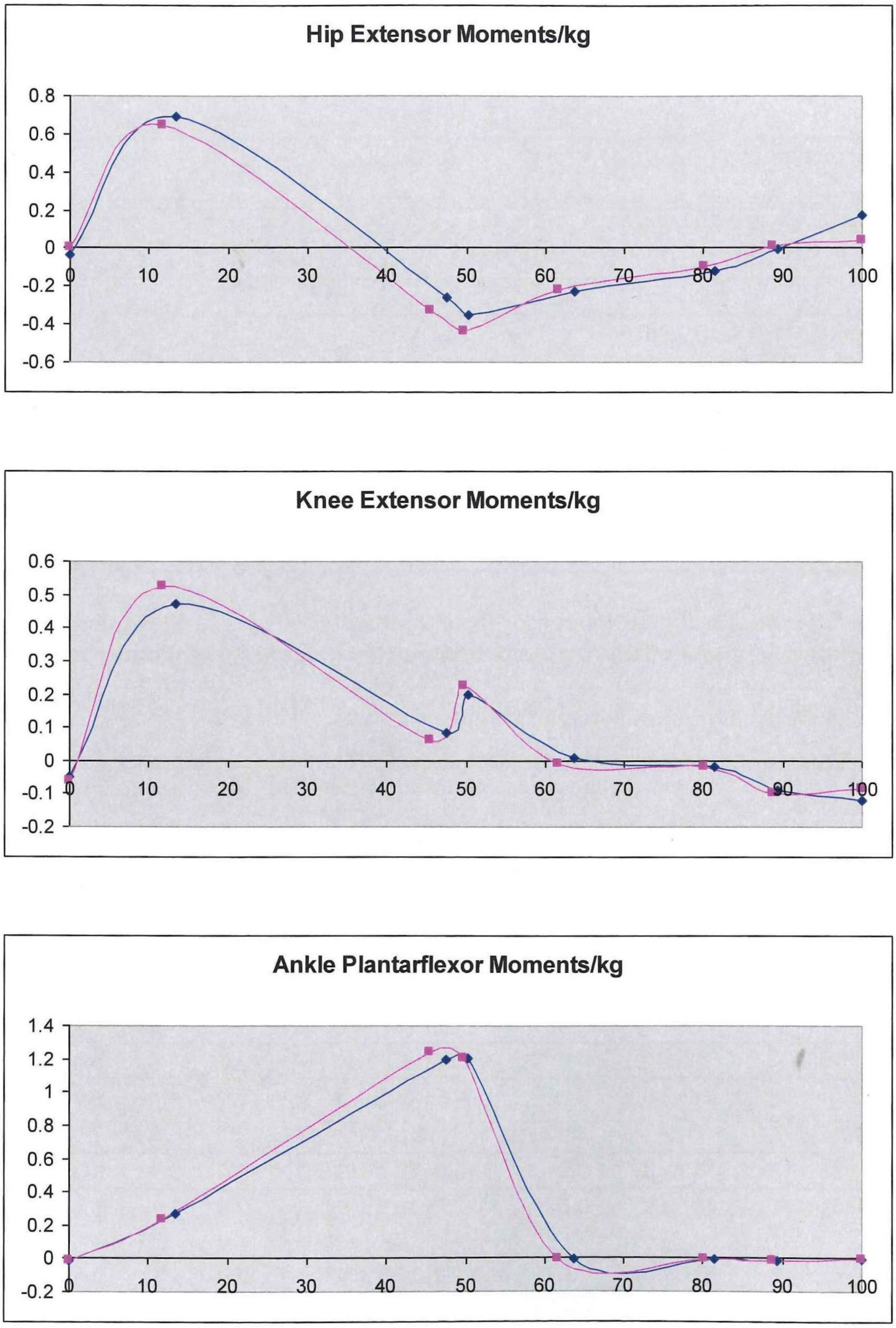


Figure 6.7: Sagittal plane joint moments.
X axis = percentage of one gait cycle.
Y axis = Newton-meters/kg.
The eight events of the gait cycle
are demarcated on the lines.

Overweight Group

Healthy weight Group

6.5.5.3 Frontal plane joint moments

Frontal plane hip joint moments were found to be reduced at initial contact for adults who were overweight (Table 6.20) (Figure 6.8). At the knee and ankle no significant difference in joint moments were noted (Tables 6.21-6.22) (Figure 6.8).

Event	Overweight group mean ± SD ($Nm^{-1}kg^{-1}$)	Healthy weight group mean ± SD ($Nm^{-1}kg^{-1}$)	Bias	p-value
Initial contact	$3.1x10^{-2} \pm 0.1$	$-2.0x10^{-2} \pm 0.1$	-0.1	p = 0.0*
Opposite toe off	0.5 ± 0.1	0.5 ± 0.1	0.0	p = 0.5
Heel rise	0.7 ± 0.1	0.7 ± 0.1	0.0	p = 0.8
Opposite initial contact	0.6 ± 0.1	0.6 ± 0.1	0.0	p = 0.7
Toe off	-0.1 ± 0.1	-0.1 ± 0.1	0.0	p = 0.4
Feet adjacent	$2.4x10^{-2} \pm 0.1$	$2.9x10^{-2} \pm 0.1$	0.0	p = 0.8
Tibia vertical	$5.3x10^{-3} \pm 0.1$	$-1.5x10^{-2} \pm 0.1$	0.0	p = 0.4

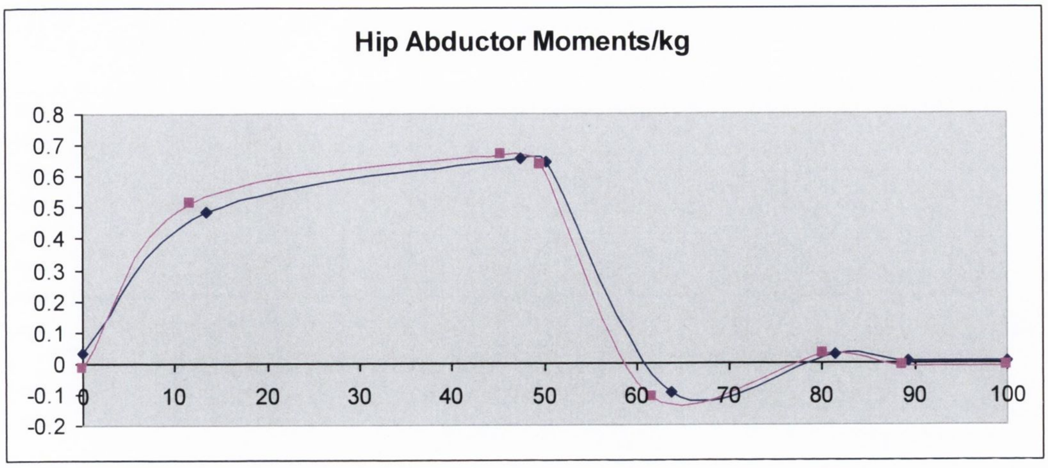
Table 6.20: Results for hip abductor-adductor moments (+/-).p = t-test p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

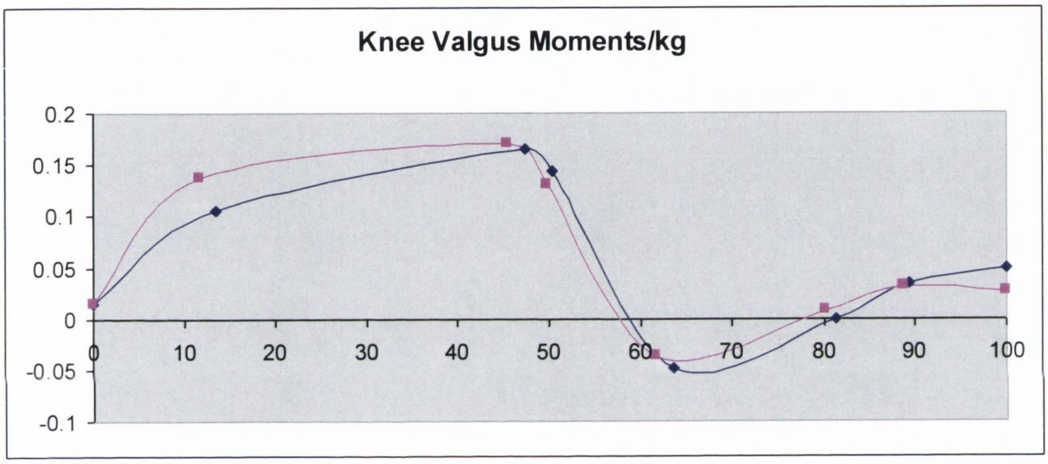
Event	Overweight group mean ± SD ($Nm^{-1}kg^{-1}$)	Healthy weight group mean ± SD ($Nm^{-1}kg^{-1}$)	Bias	p-value
Initial contact	$1.4x10^{-2}$ ± 0.0	$1.5x10^{-2}$ ± 0.0	0.0	p = 0.9
Opposite toe off	0.1 ± 0.2	0.1 ± 0.1	0.0	p = 0.4
Heel rise	0.2 ± 0.2	0.2 ± 0.1	0.0	p = 0.9
Opposite initial contact	0.1 ± 0.2	0.1 ± 0.1	0.0	p = 0.8
Toe off	$-4.8x10^{-2}$ ± 0.0	$-3.8x10^{-2}$ ± 0.0	0.0	p = 0.3
Feet adjacent	$-2.4x10^{-4}$ ± 0.0	$7.6x10^{-3}$ ± 0.1	0.0	*p* = 0.2
Tibia vertical	$3.4x10^{-2}$ ± 0.0	$3.1x10^{-2}$ ± 0.0	0.0	p = 0.7

Table 6.21: Results for Knee valgus-varus moments (+/-).p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$0.05.
*** = significant difference between groups.**

Event	Overweight group mean ± SD ($Nm^{-1}kg^{-1}$)	Healthy weight group mean ± SD ($Nm^{-1}kg^{-1}$)	Bias	p-value
Initial contact	$7.0x10^{-4}$ ± 0.0	$1.6x10^{-3}$ ± 0.0	0.0	*p* = 0.5
Opposite toe off	$4.4x10^{-2}$ ± 0.1	0.1 ± 0.1	0.0	p = 0.1
Heel rise	0.2 ± 0.1	0.2 ± 0.1	0.0	p = 0.7
Opposite initial contact	0.2 ± 0.1	0.2 ± 0.1	0.0	p = 0.8
Toe off	$-1.0x10^{-3}$ ± 0.0	$-4.6x10^{-4}$ ± 0.0	0.0	*p* = 0.6
Feet adjacent	$-2.9x10^{-3}$ ± 0.0	$-1.6x10^{-3}$ ± 0.0	0.0	*p* = 0.2
Tibia vertical	$5.9x10^{-4}$ ± 0.0	$-1.1x10^{-3}$ ± 0.0	0.0	*p* = 0.3

Table 6.22: Results for Ankle pronator-supinator moments (+/-).p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$0.05. * = significant difference between groups.





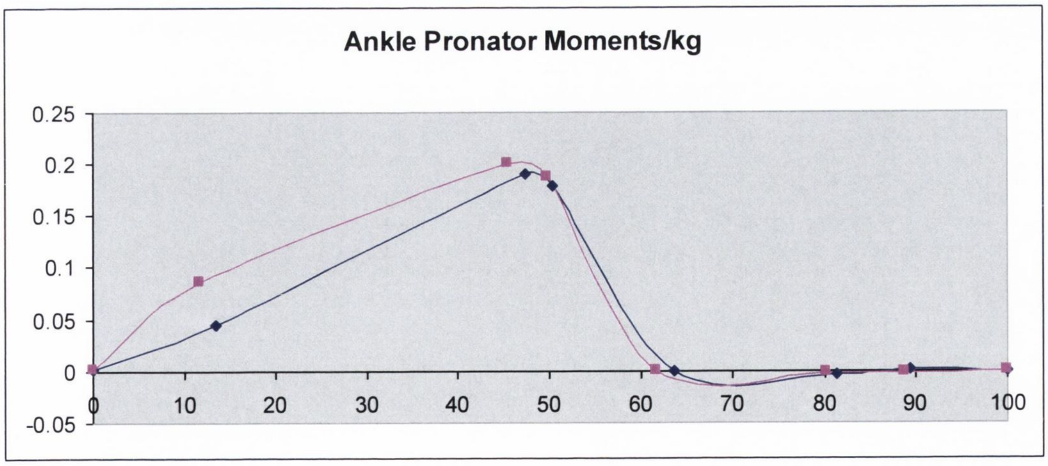


Figure 6.8: Frontal plane joint moments.
X axis = percentage of one gait cycle.
Y axis = Newton-meters/kg.
The eight events of the gait cycle
are demarcated on the lines.

Overweight Group

Healthy weight Group

6.5.5.4 Transverse plane joint moments

No significant differences in transverse plane hip or ankle joint moments were noted between the two groups (Tables 6.23 and 6.25) (Figure 6.9). A significant difference in knee rotator moments was noted at toe off (Tables 6.24) (Figure 6.9).

Event	Overweight group mean ± SD ($Nm^{-1}kg^{-1}$)	Healthy weight group mean ± SD ($Nm^{-1}kg^{-1}$)	Bias	p-value
Initial contact	$-5.0x10^{-4} \pm 0.0$	$-7.0x10^{-4} \pm 0.0$	0.0	p = 0.9
Opposite toe off	0.1 ± 0.0	0.1 ± 0.0	0.0	*p* = 0.9
Heel rise	$-1.1x10^{-2} \pm 0.0$	$-5.2x10^{-3} \pm 0.0$	0.0	p = 0.6
Opposite initial contact	$3.9x10^{-3} \pm 0.0$	$1.3x10^{-2} \pm 0.0$	0.0	*p* = 0.8
Toe off	$-3.2x10^{-2} \pm 0.0$	$-3.0x10^{-2} \pm 0.0$	0.0	p = 0.7
Feet adjacent	$-2.8x10^{-3} \pm 0.0$	$1.6x10^{-3} \pm 0.0$	0.0	p = 0.5
Tibia vertical	$-4.1x10^{-4} \pm 0.0$	$-2.7x10^{-3} \pm 0.0$	0.0	p = 0.5

Table 6.23: Results for Hip external-internal rotator moments (+/-).p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

Event	Overweight group mean ± SD ($Nm^{-1}kg^{-1}$)	Healthy weight group mean ± SD ($Nm^{-1}kg^{-1}$)	Bias	p-value
Initial contact	$1.1x10^{-3}$ ± 0.0	$-1.1x10^{-4}$ ± 0.0	0.0	p = 0.1
Opposite toe off	$9.2x10^{-3}$ ± 0.0	$1.8x10^{-2}$ ± 0.0	0.0	*p* = 0.1
Heel rise	0.1 ± 0.1	0.1 ± 0.0	0.0	p = 0.9
Opposite initial contact	0.1 ± 0.1	0.1 ± 0.0	0.0	p = 0.9
Toe off	$-1.4x10^{-3}$ ± 0.0	$2.4x10^{-4}$± 0.0	0.0	*p* = 0.0*
Feet adjacent	$-5.4x10^{-4}$± 0.0	$-2.8x10^{-4}$± 0.0	0.0	*p* = 0.5
Tibia vertical	$1.5x10^{-3}$± 0.0	$1.5x10^{-3}$± 0.0	0.0	p = 0.7

Table 6.24: Results for Knee external-internal rotator moments (+/-).p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

Event	Overweight group mean ± SD ($Nm^{-1}kg^{-1}$)	Healthy weight group mean ± SD ($Nm^{-1}kg^{-1}$)	Bias	p-value
Initial contact	$4.6x10^{-4}$± 0.0	$9.6x10^{-4}$± 0.0	0.0	p = 0.6
Opposite toe off	$-9.3x10^{-3}$± 0.0	$-2.0x10^{-2}$± 0.0	0.0	*p* = 0.3
Heel rise	$-1.4x10^{-2}$± 0.2	$-3.9x10^{-2}$± 0.2	0.0	p = 0.6
Opposite initial contact	$1.1x10^{-2}$± 0.1	$-1.4x10^{-2}$± 0.1	0.0	p = 0.5
Toe off	$2.2x10^{-3}$± 0.0	$6.1x10^{-4}$± 0.0	0.0	p = 0.1
Feet adjacent	$1.1x10^{-3}$± 0.0	$7.8x10^{-4}$± 0.0	0.0	*p* = 0.4
Tibia vertical	$-1.3x10^{-3}$± 0.0	$-1.4x10^{-3}$± 0.0	0.0	p = 0.9

Table 6.25: Results for Ankle external-internal alignment moments (+/-).p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

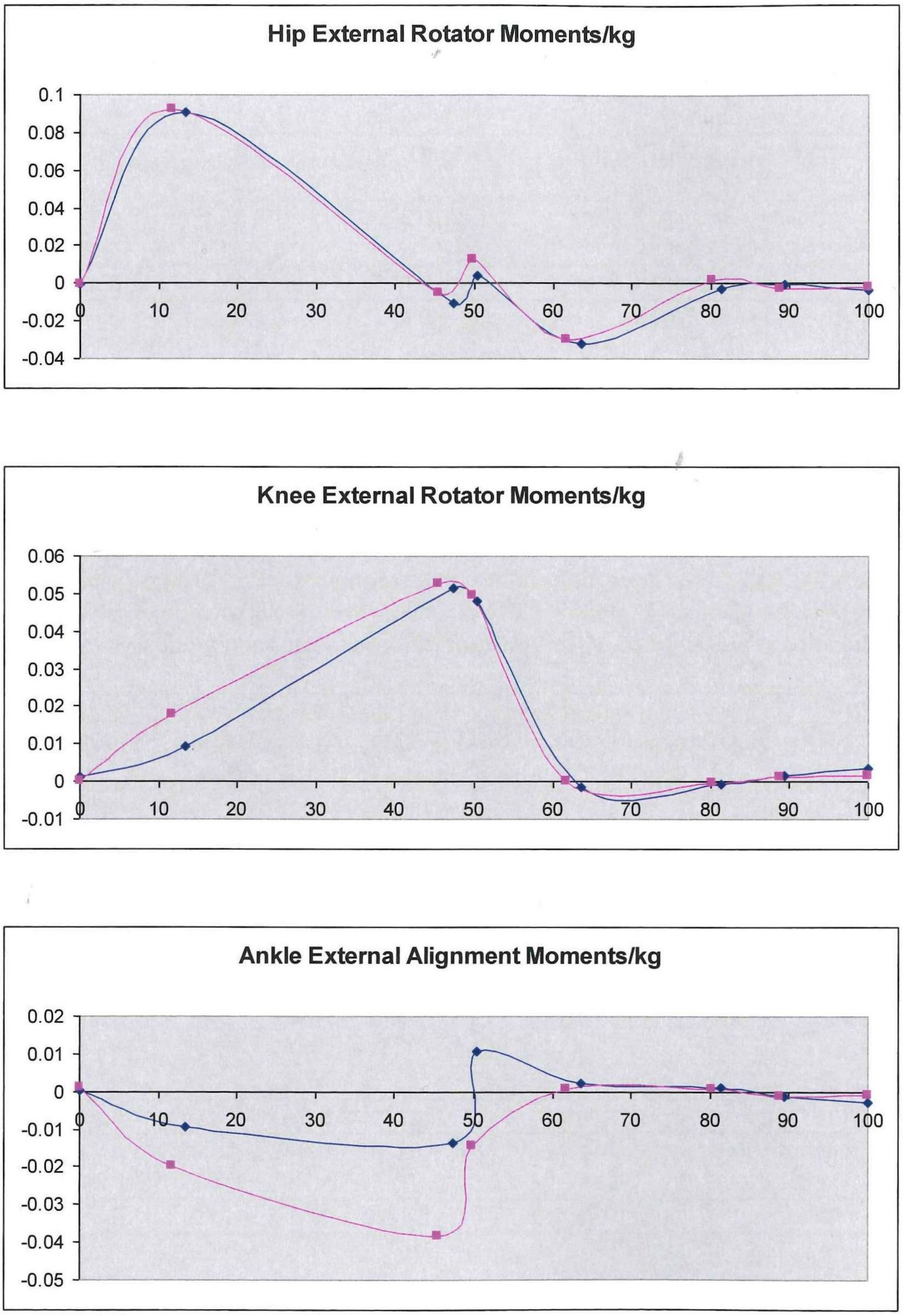


Figure 6.9: Transverse plane joint moments.
X axis = percentage of one gait cycle.
Y axis = Newton-meters/kg.
The eight events of the gait cycle are demarcated on the lines.

Overweight Group

Healthy weight Group

6.5.5.5 Joint power

There was no significant difference in maximum total joint power generation (concentric activity) at the hip, knee and ankle (Table 6.26) (Figure 6.10).

Parameter (W/kg)	Overweight group (mean ± SD)	Healthy weight group (mean ± SD)	Bias	p-value
Hip	1.4 ± 0.3	2.0 ± 1.3	0.6	*p* = 0.1
Knee	1.0 ± 0.7	1.1 ± 0.7	0.0	*p* = 0.7
Ankle	2.7 ± 1.0	3.2 ± 1.2	0.5	p = 0.1

Table 6.26: Maximum total joint power generation. p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05.

*** = significant difference between groups.**

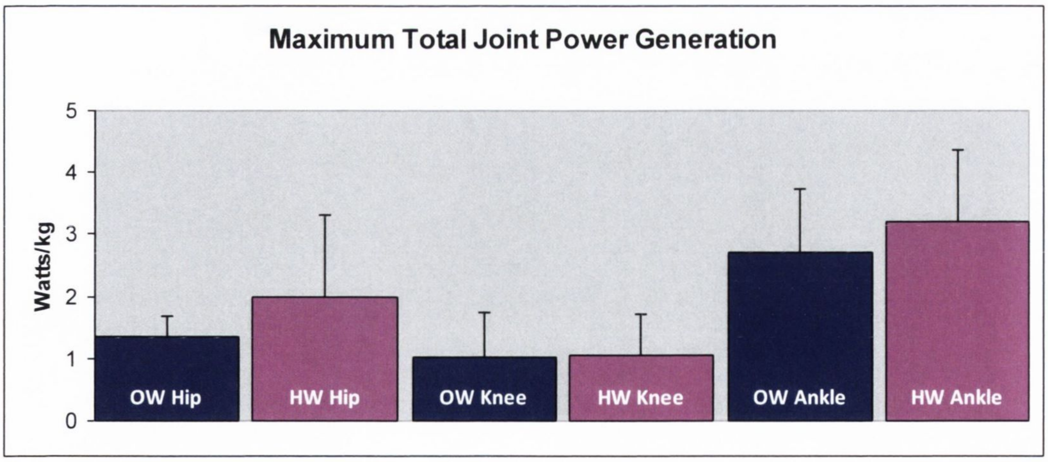


Figure 6.10: Maximum total joint power generation.

OW = Overweight

HW = Healthy weight

A significant decrease in maximum total hip power absorption (eccentric activity) was noted for adults who were overweight (Table 6.27) (Figure 6.11). No significant difference was noted at the knee and ankle (Table 6.27) (Figure 6.11).

Parameter (W/kg)	Overweight group (mean ± SD)	Healthy weight group (mean ± SD)	Bias	p-value
Hip	0.5 ± 0.3	0.9 ± 0.6	0.4	*p* = 0.0*
Knee	1.6 ± 0.6	1.8 ± 0.7	0.2	*p* = 0.1
Ankle	1.1 ± 0.4	1.2 ± 0.5	0.2	*p* = 0.1

Table 6.27: Maximum total joint power absorption. p = t-test p-value with significance at alpha ≤ 0.05. *p* = Mann Whitney U p-value with significance at alpha ≤ 0.05.

*** = significant difference between groups.**

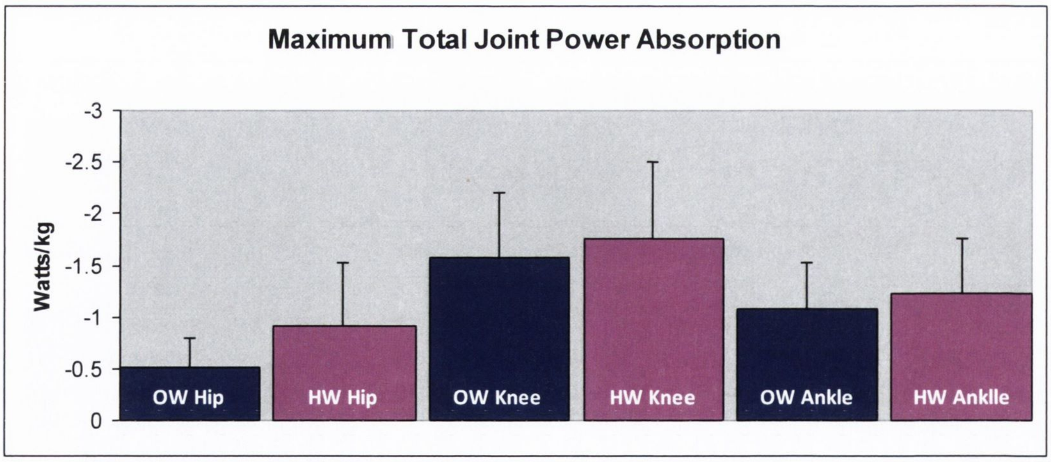


Figure 6.11: Maximum total joint power absorption.

OW = Overweight

HW = Healthy weight

6.6 DISCUSSION

The aim of the current chapter was to assess the influence of body mass on three dimensional gait parameters in adults. 50 subjects were recruited and completed testing. Significant differences in body mass, BMI, and body composition (as seen by the significant difference in thigh girth) were recorded (Harris et al 2000) (Table 6.1).

6.6.1 Temporal-spatial characteristics

The first objective of this chapter was to determine the effects of increased body mass in adults on temporal-spatial parameters. All participants ambulated at their self selected velocity. A significant decreasing trend in velocity was noted for an increase in BMI (p for trend = 0.0*). Despite this, no significant differences in velocity, cadence, stride length, or single support duration were noted between groups (Table 6.2). As discussed previously velocity alters several parameters during the gait cycle in a variety of linear and non-linear ways (Chapter 1, page 18). As no difference in velocity was noted between the two groups, the results presented and discussed here are independent of velocity. This finding is in contrast to earlier research where the self selected velocity of adults with increased body mass was found to be reduced (Spyropoulos et al 1991; Vismara et al 2007; Lai et al 2008; Ko et al 2010, Cimolin et al 2011). For the study by Ko et al (2010), a significant reduction in self selected velocity was noted for their obese and not for their overweight group. The authors proposed that the extent to which increased weight influences velocity is dependant on the extent of the increase weight (Ko et al 2010). This hypothesis is supported by the findings of the current study where the difference in weight between healthy an overweight groups was of a comparable difference for that seen by Ko et al (2010) (Table 6.2).

As no significant difference in velocity was noted between the two groups walking at their self selected velocity, the analysis of data for study 3(b) whereby participants of a healthy weight were cued during gait was not included in this chapter. The data set for study 3 (b) where overweight adults walking at their self selected velocity were compared to healthy adults walking in time to a cue is presented in Appendix 12, page 268. Few differences were noted between the two datasets.

A significant increase in stance phase, and double support duration was found for adults who were overweight (Table 6.2). In contrast to the findings of Ko et al (2010), the results of the present study are in keeping with the findings of earlier research where velocity was constrained with the use of a treadmill (DeVita & Hortobagyi 2003; Browning & Kram 2007). The results note for those with excess body mass, swing phase duration was minimised. This may have been an attempt to protect the joints of the lower limbs by reducing the time spent loading one and not both limbs. In addition by increasing the time spent in double support and stance phase, stability is improved suggesting the presence of instability during gait for adults who are overweight (England & Granata 2007).

6.6.2 Kinematics and joint moments

Additional objectives of this chapter included to determine the effects of increased adiposity in adults on three-dimensional kinematic parameters, and joint moments at seven events of the gait cycle. As described earlier, each event of the gait cycle was identified manually for each participant and expressed in terms of a percentage to enable between group comparisons (Chapter 6, page 98). The timing of opposite toe off, heel rise, opposite initial contact, toe off and feet adjacent were significantly different between the two groups. All events which were significantly different in timing occurred later for the overweight group. This result coincides with the increased stance and reduced swing phase duration noted for the group of adults who were overweight (Andriacchi et al 1977).

6.6.2.1 The sagittal plane

Very few differences were noted in the sagittal plane between the two groups (Tables 6.3-6.6, 6.17-6.19) (Figure 6.2). At the pelvis no significant differences were noted for an increase in body mass (Table 6.3). It is therefore possible to state that any alterations seen at the hip are due to femoral of pelvic kinematics, and not from pelvic on femoral kinematics. In addition, no significant differences in ankle kinematics were noted between the two groups (Table 6.6). The results at the ankle are in keeping with the findings of earlier research (Browning & Kram 2007; Ko et al 2010). However, DeVita & Hortobagyi (2003) dispute these findings as an increase in ankle plantarflexion was noted. Finally, there was no significant difference between groups for joint moments in the sagittal plane at the hip, knee or ankle (Tables 6.17-6.19). Supporting these findings, no changes in peak hip or knee extensor moments were noted by earlier research (Browning & Kram 2007; Ko et al 2010). Once again, DeVita & Hortobagyi (2003) disputed the findings, as they noted a reduction in normalised knee extensor joint moments.

A significant increase in hip flexion was noted at initial contact for adults who were overweight (Table 6.4). No other significant differences were noted. The increase in hip flexion seen at initial contact may be an effort to minimise joint loads across the hip as the swing limb comes into contact with the ground. Alternatively, it may be seen to be an attempt to regain stability, lowering the centre of gravity as the body travels from single to double support (Watkins 2010). Few changes were noted at the knee in the sagittal plane for adults who were overweight. A significant increase in knee flexion at tibia vertical was found for adults who were overweight (Table 6.5). No other differences were noted. Piazza & Delph (1996) reported on the influence of muscles on knee flexion during swing phase. They noted an increase in knee flexion was associated with inactivity of the rectus femoris (Piazza & Delph 1996). This suggests the presence of muscle weakness in adults who are overweight. However, as muscle activity was not directly assessed here, and no difference was noted at toe off or feet adjacent it is difficult to draw any definitive conclusions.

Previous work which has noted no change in velocity and assessed the influence of body mass on sagittal plane hip and knee kinematics have found no difference between groups at midstance, or for total joint range (Browning & Kram 2007; Ko et al 2010). As for ankle kinematics and knee joint moment parameters, DeVita & Hortobagyi (2003) disputed these conclusions as they noted a reduction in hip and knee flexion for adults who were overweight.

6.6.2.2 The frontal plane

Similar to the sagittal plane, few differences were noted in the frontal plane for those who were overweight, as compared to their healthy weight counterparts (Tables 6.7-6.10, 6.20-6.22) (Figure 6.3). No significant differences were noted at the pelvis, once again alterations in hip kinematics may be deemed to be due to femoral on pelvic and not pelvic on femoral movement (Table 6.7).

At toe off and feet adjacent adults who were overweight demonstrated a significant increase in hip abduction (Table 6.8). Kerrigan et al (2000) defined circumduction as an increase in thigh angle during mid-swing. The left and right lower limbs pass each other when one limb is in swing phase. The circumduction seen at toe off and feet adjacent may be an effort adopted to enable the limbs to pass each other without friction, due to the increased thigh girth seen for adults who were overweight. Alternatively the gluteus medius for adults who are overweight may be required to compensate for reduced muscle strength of the hip flexors, including the rectus femoris, to enable clearance of the foot during swing phase. However, as muscle strength/activity was not assessed it is difficult to assess this hypothesis directly. No differences in hip abductor moments were noted during swing phase between the two groups (Table 6.20). However, a decrease in hip abductor moments was seen in the frontal plane at initial contact for adults who were overweight (Table 6.20).

Optical based systems on average measure angles in motion to within 1.5° of the actual value (Richards 1999). Joint range of motion at the knee and ankle in the frontal plane is small during gait, 1.5° may be the difference between a reported varus/valgus alignment, and pronation/supination. The results for the knee and ankle discussed here should therefore be viewed with a degree of caution.

A significant increase in varus alignment was noted at feet adjacent for adults who were overweight (Table 6.9). This alignment places greater stress on the medial compartment of the tibio-femoral joint. However, feet adjacent occurs during swing phase where the limb is in a non-weight bearing state. In addition, no altered frontal knee joint moments were found throughout the gait cycle (Table 6.21).

At the ankle an increase in supination was seen at feet adjacent (Table 6.10). A degree of supination is anticipated during the swing phase protecting the foot as a rigid structure is created (Johanson et al 2008; Jenkyn et al 2009). As for the knee, there was no difference in frontal plane ankle joint moments between the two groups (Table 6.22).

6.6.2.3 The transverse plane

As for the frontal plane, range of motion in the transverse plane is small and conclusions may therefore be subject to error due to the accuracy of the CODA Motion (Richards 1999).

There were no significant differences in transverse plane kinematics at the pelvis, hip, knee or ankle between the two groups (Tables 6.11-6.14) (Figure 6.4). Only one study has previously investigated transverse plane kinematics for adults who are overweight (Lai et al 2008). Similar to the results presented here, no differences in kinematic parameters were noted between groups (Lai et al 2008). Despite there being no significant differences between the two groups for kinematic parameters in the transverse plane, a significant internal rotator moment was seen at the tibia at toe off for adults who were overweight (Table 6.24). This prevented the tibia from becoming externally rotated excessively at toe off.

6.6.3 Ground reaction forces

There was no significant difference in maximum anterior-posterior and medial lateral three-dimensional maximum GRF between the two groups, once corrected for body mass (Table 6.15) (Figure 6.5). Similar findings were found in earlier research where velocity was constrained with the use of a treadmill (Browning & Kram 2007). In contrast to the study by Browning & Kram (2007) a significant reduction in maximum vertical GRF corrected for body mass was noted for adults who were overweight (Table 6.15) (Figure 6.5). As for the previous literature, in the current study maximum absolute ground reaction forces were significantly greater for adults who were overweight for all three planes (Browning & Kram 2007) (Table 6.16) (Figure 6.5).

6.6.4 Joint power

There was no significant difference in power generation/ concentric muscle activity between the two groups (Table 6.26) (Figure 6.10). These results support previous work where total sagittal and frontal hip, knee and ankle power has been reported as similar between groups (Vismara et al 2007; Ko et al 2010). However, the results are disputed by Cimolin et al (2011) who noted a reduction in peak ankle power and increased hip generative power for their obese group (Cimolin et al 2011). In addition previous literature has reported an increase in total frontal knee generative power (Ko et al 2010).

A significant decrease in hip power absorption was noted for adults who were overweight (Table 6.27) (Figure 6.10). This is in contrast to earlier work where an increase in frontal hip absorptive power was reported (Ko et al 2010). For the current study, this suggests that eccentric activity of the hip extensors, adductors and internal rotators was reduced for adults who were overweight. During eccentric activity greater demands are placed on the soft tissue structures concerned (Lindstedt et al 2002). Therefore the reduction in power absorption at the hip may be due to muscle weakness. The result is a reduction in shock attenuation by the muscles of the hip for the adults who were overweight (Lindstedt et al 2001). No significant difference in power absorption was seen at the knee or ankle between the two groups (Table 6.27) (Figure 6.10). Similarly, Ko et al (2010) reported no change in knee absorptive power for overweight individuals. Cimolin et al (2011) reported a significant increase in ankle absorptive power for their obese group.

6.7 CONCLUSION

In comparison to the previous literature assessing the influence of body mass on gait parameters in adults fewer between group differences were noted. A difference in velocity may account for several between group differences reported by the earlier research. For the current chapter no difference in velocity was found between the two groups. The analysis was therefore independent of velocity, and represented the overweight groups' presentation of gait at their self selected velocity. Despite this, several differences were identified and discussed by the current chapter, with the frontal plane exhibiting the majority of changes for adults who were overweight.

CHAPTER 7 THE INFLUENCE OF BODY MASS ON GAIT PARAMETERS IN CHILDREN

7.1 INTRODUCTION

Similar to adults, children who are overweight may choose to ambulate at a reduced velocity, and this reduction may lead to, or enhance an altered presentation of gait. It is therefore necessary to assess gait parameters in children who are overweight with, and without constrained velocity.

7.2 OBJECTIVES

The objectives of this chapter were:

To determine the influence of body mass on gait parameters† in children.

To determine the influence of body mass on gait parameters† in children, while accounting for the confounding influence of velocity.

† Gait parameters refer to: temporal-spatial parameters; three dimensional kinematics at the pelvis, hip, knee and ankle; three dimensional joint moments at the hip, knee and ankle; maximum absolute and normalised anterior, medial and vertical ground reaction forces; and, maximum and minimum total joint power at the hip, knee and ankle.

7.3 METHODOLOGY

A cross sectional study design as described in Chapter 3, page 51 was selected. Children who matched the inclusion/exclusion criteria (Chapter 3, page 51) were recruited from the Weight Management and Endocrine Clinics at AMNCH, and local schools as described in Chapter 3, page 53. Participants were required to attend the gait laboratory on one occasion. Measurements were completed in the order outlined in figure 7.1. Each gait laboratory session was completed as described in Chapter 3, pages 59-70. All participants were requested to ambulate at their self selected velocity. Those in the control group were also required to ambulate in time to a beat, set to the cadence of the overweight child whom they were matched to. The beat was sounded using the TempoPerfect (NCH Software, Canberra, Australia) digital metronome (Chapter 3, page 59).

	Order of Measures
Height and body mass	↓
Gait Analysis (with and without cue for control group)	

Figure 7.1: Order of measures for study 4.

7.4 ANALYSIS

Data preparation and analysis was completed as described in Chapter 6, page 98.

7.5 RESULTS STUDY 4 (a)

Results presented in the following section represent the analysis of gait parameters at the self selected velocity of participants who are overweight, as compared to gait parameters at the self selected velocity of participants of a healthy weight.

7.5.1 Baseline characteristics

25 children who were overweight (male: female, 8:17) and 25 age and gender matched children of a healthy weight (male: female, 8:17) completed the study. 23 pairs were matched by height to within 2 cm. Two pairs were not matched by height. Baseline demographics are outlined in Table 7.1. A significant difference in body mass, BMI and thigh girth was noted between the groups ($p < 0.00$). 19 children in the experimental group were classified as obese according to IOTF cut-offs (Cole et al 2000). Six children in the experimental group were classified as overweight according to IOTF cut-offs (Cole et al 2000). Children were of a similar height to the adult groups, no significant difference for age and height were noted. As for adults the Craig test and Q angle results were not analysed due to uncertainty with regard to the accuracy of the measurements.

Parameter	Overweight group (n = 25) (mean ± SD)	Healthy weight group (n = 25) (mean ± SD)
Age (years)	12.4 ± 3.0	12.7 ± 2.8
Height (cm)	157.5 ± 13.6	156.6 ± 13.9
Body mass (kg)	78.9 ± 27.3	49.7 ± 13.9
BMI (kgm^{2-1})	30.9 ± 5.9	19.8 ± 2.6
Thigh girth (cm)	55.9 ± 6.7	42.7 ± 4.5

Table 7.1: Baseline Characteristics. Mean ± standard deviation.

Temporal-spatial and kinematic data was collected in 50 children. Kinetic data was collected in 48 children. A clean foot contact (whereby only one foot contacted the force platform) with the force plate was not collected for one overweight child, and the pair was excluded from the kinetic analysis as a result. Cued data was collected for 24 participants of a healthy weight.

7.5.2 Temporal-spatial parameters

Temporal-spatial characteristics selected for analysis included velocity, stride length, cadence, percentage stance, single support time, and double support time. Mean, standard deviation, bias, and p-value are presented in Table 7.2. Significant between group differences for velocity, stride length, percentage stance, single support duration, and double support duration were noted (Table 7.2).

Event	Overweight group (mean ± SD)	Healthy weight group (mean ± SD)	Bias	p-value
Velocity (ms^{-1})	1.2 ± 0.2	1.3 ± 0.2	0.2	p = 0.0*
Stride Length (m)	1.1 ± 0.2	1.3 ± 0.2	0.1	p = 0.0*
Cadence (stepsmin^{-1})	123.9 ± 16.1	124.3 ± 15.8	0.5	p = 0.6
Stance phase (%)	63.4 ± 2.1	60.3 ± 2.3	3.1	p = 0.0*
Single Support (s)	0.4 ± 0.1	0.4 ± 0.0	0.0	p = 0.0*
Double Support (s)	0.1 ± 0.0	0.1 ± 0.0	0.0	p = 0.0*

Table 7.2: Results for temporal-spatial parameters. p = t-test p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

7.5.3 Timing of events

No significant difference in the timing of opposite toe off, heel rise, opposite initial contact, and toe off were noted between the groups p = 0.5 p = 0.9, p = 0.2, and p = 0.5 respectively. However significant differences in the timing of feet adjacent and tibia vertical were documented p = 0.0 and p = 0.0 respectively.

7.5.4 Kinematics

Mean, standard deviation, bias, and p-value were calculated for the pelvis, hip, knee and ankle in the sagittal, frontal and transverse planes for each of the seven events. Results are presented in Tables 7.3- 7.14 and Figures 7.2-7.4.

7.5.4.1 Sagittal plane

There was no significant difference in anterior/posterior pelvic tilt between the two groups for all seven time points (Table 7.3) (Figure 7.2).

Event	Overweight group mean ± SD (degrees)	Healthy weight group mean ± SD (degrees)	Bias	p-value
Initial contact	9.2 ± 7.3	9.5 ± 6.1	0.3	p = 0.9
Opposite toe off	6.0 ± 7.7	8.5 ± 5.7	2.5	p = 0.2
Heel rise	9.1 ± 7.1	10.2 ± 6.1	1.1	p = 0.6
Opposite initial contact	9.5 ± 7.3	10.0 ± 6.0	0.6	p = 0.8
Toe off	5.9 ± 7.6	9.0 ± 5.7	3.1	p = 0.1
Feet adjacent	7.7 ± 8.2	10.3 ± 6.3	2.6	p = 0.2
Tibia vertical	8.3 ± 7.9	10.4 ± 6.3	2.1	p = 0.3

Table 7.3: Results for pelvis anterior-posterior tilt (+/-).p = t-test p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

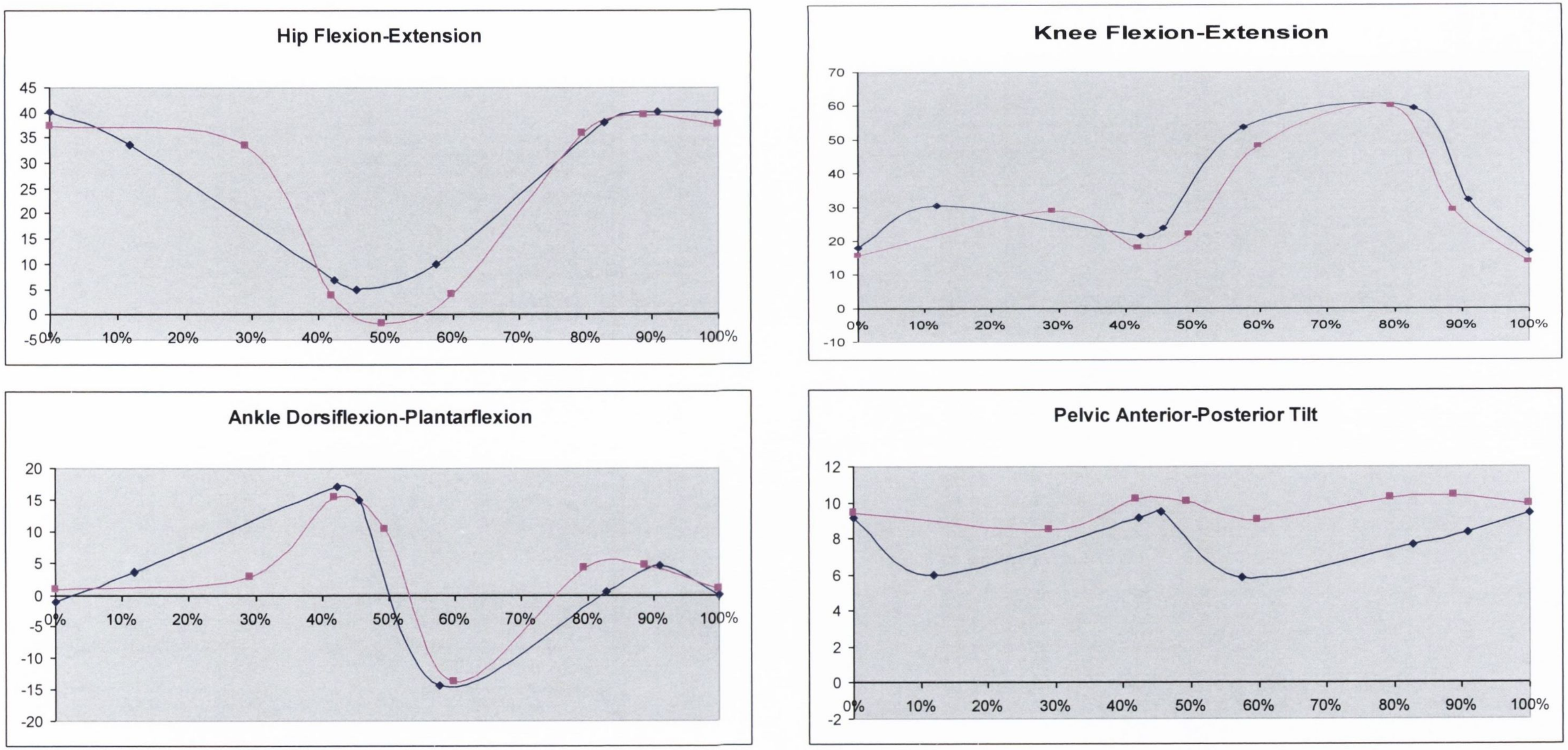


Figure 7.2: Sagittal plane kinematics.

X axis = percentage of one gait cycle. Y axis = degrees.

The eight events are demarcated on the lines.

Overweight Group

Healthy weight Group

At the hip a significant difference between the groups was noted at opposite initial contact and toe off for hip flexion (Table 7.4) (Figure 7.2). A significant increase in knee flexion was noted for children who were overweight at heel rise, toe off, and tibia vertical (Table 7.5) (Figure 7.2). No additional significant differences were noted between the two groups at the hip or knee (Tables 7.5-7.6).

Event	**Overweight group mean ± SD (degrees)**	**Healthy weight group mean ± SD (degrees)**	**Bias**	**p-value**
Initial contact	40.2 ± 7.2	37.3 ± 7.0	2.9	p = 0.2
Opposite toe off	33.5 ± 7.6	33.5 ± 7.7	0.1	p = 0.97
Heel rise	6.9 ± 8.7	3.6 ± 7.2	3.3	p = 0.2
Opposite initial contact	4.9 ± 9.0	-2.0 ± 5.2	6.9	*p* = 0.0*
Toe off	10.1 ± 8.6	3.8 ± 4.4	6.3	*p* = 0.0*
Feet adjacent	38.1 ± 7.2	35.8 ± 6.1	2.3	p = 0.2
Tibia vertical	40.2 ± 7.2	39.4 ± 6.9	0.8	p = 0.2

Table 7.4: Results for hip flexion-extension (+/-).p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

Event	Overweight group mean ± SD (degrees)	Healthy weight group mean ± SD (degrees)	Bias	p-value
Initial contact	18.0 ± 7.3	15.6 ± 4.7	2.4	*p* = 0.1
Opposite toe off	30.3 ± 6.9	28.9 ± 6.3	1.3	p = 0.5
Heel rise	21.6 ± 6.4	17.8 ± 4.5	3.8	p = 0.0*
Opposite initial contact	23.8 ± 5.6	22.1 ± 3.5	1.7	p = 0.2
Toe off	53.7 ± 6.3	48.0 ± 5.1	5.7	p = 0.0*
Feet adjacent	59.2 ± 6.4	59.8 ± 7.0	0.6	p = 0.7
Tibia vertical	32.3 ± 5.2	29.0 ± 4.3	3.3	*p* = 0.0

Table 7.5: Results for knee flexion-extension (+/-).p = t-test p-value with significance at alpha $\leq$ 0.05. ***p*** **= Mann Whitney U p-value with significance at alpha $\leq$ 0.05.**

*** = significant difference between groups.**

A significant increase in dorsiflexion was recorded at opposite initial contact in the overweight group (Table 7.6) (Figure 7.1). A significant increase in plantarflexion was recorded at feet adjacent in the overweight group (Table 7.6) (Figure 7.1). No significant difference was noted at the remaining five time points (Table 7.6) (Figure 7.2).

Event	Overweight group mean ± SD (degrees)	Healthy weight group mean ± SD (degrees)	Bias	p-value
Initial contact	-1.0 ± 9.1	0.9 ± 5.9	1.9	*p* = 0.7
Opposite toe off	2.9 ± 2.9	3.0 ± 7.7	0.2	p = 0.9
Heel rise	17.1 ± 3.0	15.5 ± 3.8	1.6	p = 0.1
Opposite initial contact	14.9 ± 4.8	10.3 ± 6.5	4.6	p = 0.0*
Toe off	-14.4 ± 5.5	-13.8 ± 5.9	0.6	p = 0.7
Feet adjacent	0.5 ± 5.6	4.1 ± 4.3	3.6	*p* = 0.0*
Tibia vertical	4.6 ± 4.8	4.5 ± 2.6	0.2	p = 0.9

Table 7.6: Results for ankle dorsiflexion-plantarflexion (+/-). p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

7.5.4.2 Frontal plane

No significant differences were noted at the pelvis between the two groups for six of the seven events (Table 7.7) (Figure 7.3). A significant difference was noted at tibia vertical (Table 7.7).

Event	Overweight group mean ± SD (degrees)	Healthy weight group mean ± SD (degrees)	Bias	p-value
Initial contact	-1.4 ± 1.2	-0.9 ± 1.4	0.5	p = 0.2
Opposite toe off	1.6 ± 1.2	2.2 ± 1.4	0.6	p = 0.1
Heel rise	1.5 ± 1.2	1.5 ± 1.1	0.0	p = .99
Opposite initial contact	1.4 ± 1.2	0.9 ± 1.1	0.4	p = 0.2
Toe off	-1.5 ± 1.2	-1.9 ± 1.4	0.4	*p* = 0.2
Feet adjacent	-1.1 ± 1.2	-1.5 ± 1.4	0.3	p = 0.4
Tibia vertical	-1-1 ± 1.0	-0.9 ± 1.3	0.3	p = 0.0*

Table 7.7: Results for pelvis up-down obliquity (+/-). p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05.

*** = significant difference between groups.**

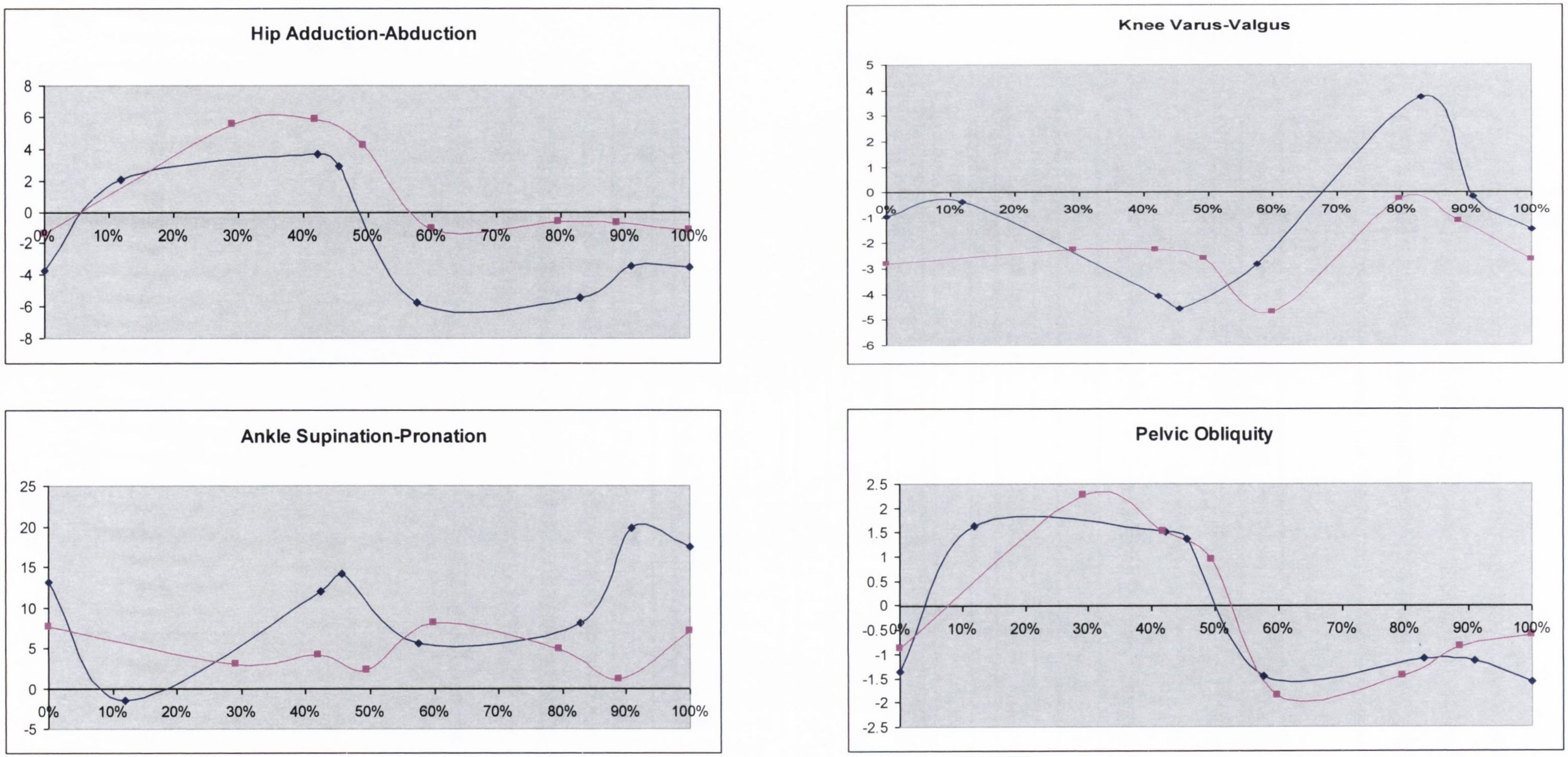


Figure 7.3: Frontal plane kinematics.

X axis = percentage of one gait cycle.

Y axis = degrees.

The eight events are demarcated on the lines.

Overweight Group

Healthy weight Group

No significant difference in hip abduction/adduction between the groups was noted at opposite initial contact (Table 7.4) (Figure 7.3). A significant difference was noted for the remaining six events with overweight children demonstrating a consistently increased level of hip abduction (Table 7.4).

Event	Overweight group mean ± SD (degrees)	Healthy weight group mean ± SD (degrees)	Bias	p-value
Initial contact	-3.8 ± 2.3	-1.4 ± 2.3	2.3	p = 0.0*
Opposite toe off	2.0 ± 2.1	5.5 ± 2.6	3.5	*p* = 0.0*
Heel rise	3.7 ± 2.3	5.9 ± 2.4	2.2	p = 0.0*
Opposite initial contact	2.9 ± 2.4	4.2 ± 2.5	1.3	p = 0.1
Toe off	-5.8 ± 2.8	-1.1 ± 2.7	4.7	p = 0.0*
Feet adjacent	-5.4 ± 2.4	-0.7 ± 2.4	4.8	p = 0.0*
Tibia vertical	-3.4 ± 2.2	-0.8 ± 2.2	2.7	p = 0.0*

Table 7.8: Results for hip adduction-abduction (+/-). p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05.

*** = significant difference between groups.**

No significant differences in frontal plane knee motion for six of the seven events were noted (Table 7.9) (Figure 7.3). A significant increase in knee varus was recorded at feet adjacent (Table 7.9) (Figure 7.3).

Event	**Overweight group mean ± SD (degrees)**	**Healthy weight group mean ± SD (degrees)**	**Bias**	**p-value**
Initial contact	-1.0 ± 2.9	-2.8 ± 3.8	1.9	p = 0.1
Opposite toe off	-0.4 ± 3.9	-2.3 ± 4.3	1.9	p = 0.1
Heel rise	-4.1 ± 4.0	-2.3 ± 3.8	1.8	p = 0.1
Opposite initial contact	-4.6 ± 4.0	-2.6 ± 3.9	2.0	*p* = 0.1
Toe off	-2.8 ± 5.1	-4.7 ± 4.7	1.9	p = 0.2
Feet adjacent	3.7 ± 5.7	-0.3 ± 5.5	4.0	p = 0.0*
Tibia vertical	-0.2 ± 4.2	-1.2 ± 3.7	1.0	p = 0.4

Table 7.9: Results for knee varus-valgus (+/-). p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

A significant increase in supination at initial contact, heel rise, opposite initial contact and tibia vertical was noted in the overweight group (Table 7.10) (Figure 7.3). No significant difference was noted at the remaining 2 time points (Table 7.10) (Figure 7.3).

Event	**Overweight group mean ± SD (degrees)**	**Healthy weight group mean ± SD (degrees)**	**Bias**	**p-value**
Initial contact	13.2 ± 7.9	7.6 ± 5.9	5.6	p = 0.0*
Opposite toe off	-1.6 ± 6.6	3.0 ± 7.7	4.6	p = 0.0*
Heel rise	12.0 ± 9.2	4.1 ± 7.6	7.9	*p* = 0.0*
Opposite initial contact	14.2 ± 8.3	2.3 ± 7.7	11.9	p = 0.0*
Toe off	5.6 ± 4.0	8.0 ± 5.2	2.4	p = 0.1
Feet adjacent	8.0 ± 6.4	4.9 ± 4.7	3.2	p = 0.1
Tibia vertical	19.8 ± 7.3	1.0 ± 7.2	18.6	p = 0.0*

Table 7.10: Results for ankle supination-pronation (+/-). p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05.

*** = significant difference between groups.**

7.5.4.3 Transverse plane

A significant difference in forward/backward pelvic rotation between the two groups was noted at tibia vertical (Table 7.11). No additional significant differences were noted (Table 7.11) (Figure 7.4).

Event	**Overweight group mean ± SD (degrees)**	**Healthy weight group mean ± SD (degrees)**	**Bias**	**p-value**
Initial contact	3.3 ± 3.1	4.6 ± 3.8	1.4	p = 0.2
Opposite toe off	2.9 ± 2.3	2.0 ± 3.3	-1.0	p = 0.1
Heel rise	-2.1 ± 2.8	-2.2 ± 2.7	-0.1	p = 0.9
Opposite initial contact	-3.2 ± 3.3	-4.4 ± 3.4	-1.2	p = 0.2
Toe off	-3.6 ± 2.5	-2.8 ± 3.2	0.8	*p* = 0.1
Feet adjacent	-3.1 ± 2.3	-2.4 ± 2.3	0.7	*p* = 0.5
Tibia vertical	-0.8 ± 2.6	0.5 ± 2.1	1.3	p = 0.0*

Table 7.11: Results for pelvis forward/backward rotation (+/-). p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05.

*** = significant difference between groups.**

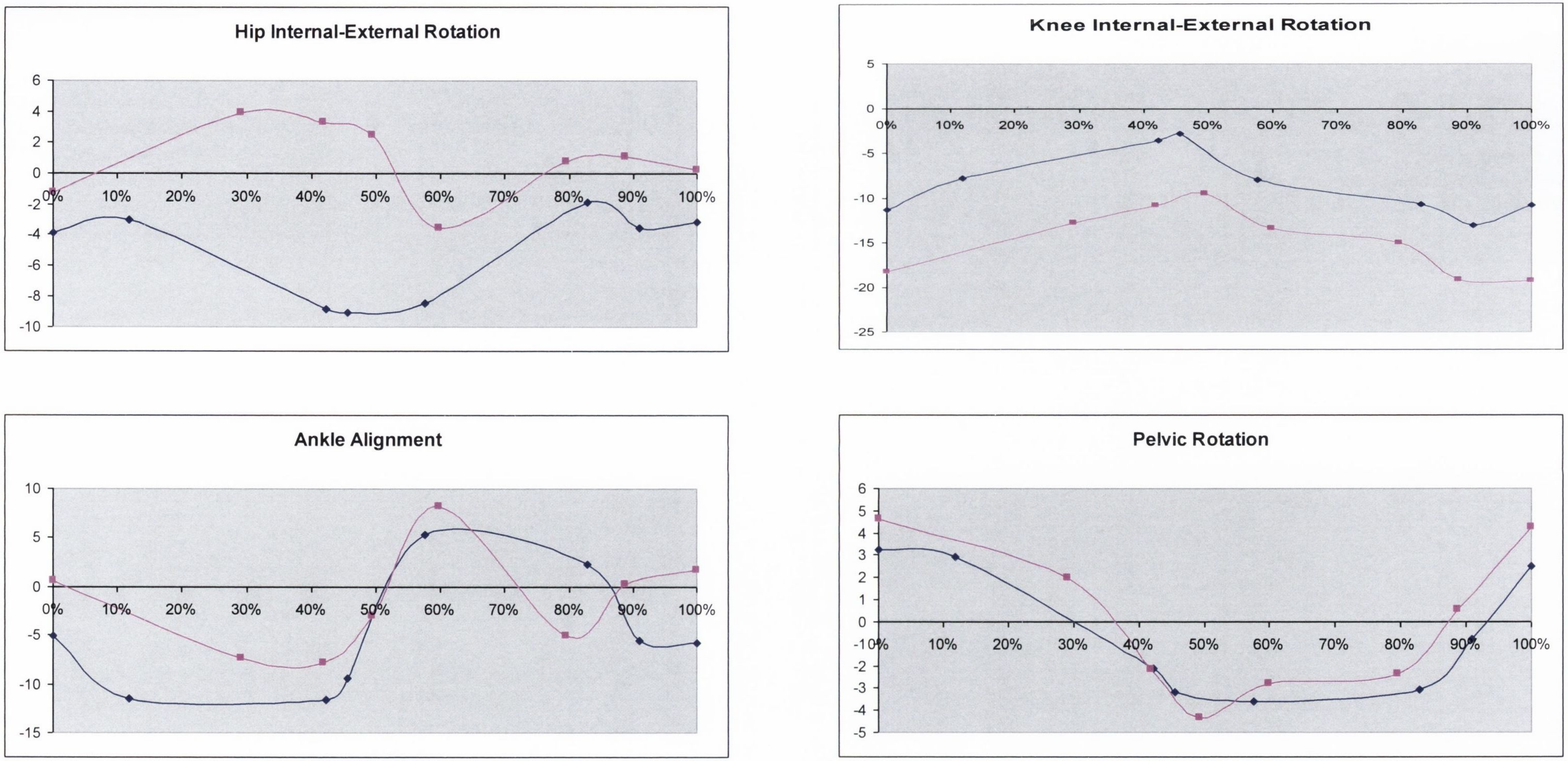


Figure 7.4: Transverse plane kinematics.

X axis = percentage of one gait cycle.

Y axis = degrees.

The eight events are demarcated on the lines.

Overweight Group

Healthy weight Group

A significant increase in hip external rotation was noted for children who were overweight at opposite toe off, heel rise, opposite initial contact, toe off, and tibia vertical(Table 7.12) (Figure 7.4). No significant differences in hip rotation at initial contact and feet adjacent were recorded (Table 7.12) (Figure 7.4).

Event	**Overweight group mean ± SD (degrees)**	**Healthy weight group mean ± SD (degrees)**	**Bias**	**p-value**
Initial contact	-3.9 ± 6.9	-1.3 ± 6.2	2.6	p = 0.2
Opposite toe off	-3.0 ± 8.2	3.9 ± 5.7	6.9	p = 0.0*
Heel rise	-8.8 ± 9.8	3.3 ± 6.2	12.1	p = 0.0*
Opposite initial contact	-9.1 ± 9.4	2.4 ± 6.1	11.5	p = 0.0*
Toe off	-8.5 ± 6.7	-3.6 ± 4.9	12.4	p = 0.0*
Feet adjacent	-1.9 ± 6.1	0.7 ± 5.4	2.6	p = 0.1
Tibia vertical	-3.6 ± 6.4	1.1 ± 5.3	4.6	p = 0.0*

Table 7.12: Results for hip internal-external rotation (+/-). p = t-test p-value with significance at alpha ≤ 0.05. * = significant difference between groups.

A significant increase in knee internal rotation was recorded for six of the seven events (Table 7.8) (Figure 7.4). No significant difference was noted at opposite toe off (Table 7.8) (Figure 7.4).

Event	**Overweight group mean ± SD (degrees)**	**Healthy weight group mean ± SD (degrees)**	**Bias**	**p-value**
Initial contact	-11.3 ± 9.7	-18.3 ± 10.9	7.0	p = 0.0*
Opposite toe off	-7.8 ± 10.9	-12.8 ± 8.0	5.0	p = 0.1
Heel rise	-3.5 ± 10.3	-10.8 ± 7.2	7.2	*p* = 0.0*
Opposite initial contact	-2.8 ± 10.1	-9.5 ± 6.9	6.6	*p* = 0.0*
Toe off	-7.9 ± 8.8	-13.3 ± 6.3	5.5	p = 0.0*
Feet adjacent	-10.6 ± 9.5	-15.0 ± 7.7	4.5	p = 0.1
Tibia vertical	-13.0 ± 10.8	-19.1 ± 7.5	6.1	p = 0.0*

Table 7.13: Results for knee internal-external rotation (+/-).p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$0.05.

*** = significant difference between groups.**

A significant external rotated alignment of the ankle was adopted by the overweight group at initial contact, opposite initial contact and tibia vertical (Table 7.14) (Figure 7.4). There was no difference between groups at opposite toe off, heel rise and toe off (Table 7.14) (Figure 7.4). A significant internal rotated alignment of the ankle was adopted by the overweight group at feet adjacent (Table 7.14) (Figure 7.4).

Event	**Overweight group mean ± SD (degrees)**	**Healthy weight group mean ± SD (degrees)**	**Bias**	**p-value**
Initial contact	-5.1 ± 7.2	0.6 ± 6.2	5.7	p = 0.0*
Opposite toe off	-11.5 ± 9.0	-7.4 ± 7.5	4.1	*p* = 0.1
Heel rise	-11.6 ± 7.9	-7.9 ± 7.0	3.7	p = 0.1
Opposite initial contact	-9.5 ± 7.3	-3.1 ± 6.8	6.4	p = 0.0*
Toe off	5.3 ± 7.5	8.2 ± 8.2	2.8	p = 0.2
Feet adjacent	2.2 ± 8.4	-5.2 ± 6.7	7.4	p = 0.0*
Tibia vertical	-5.6 ± 8.4	0.3 ± 7.2	5.8	*p* = 0.0*

Table 7.14: Results for ankle internal-external alignment (+/-).p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05.

*** = significant difference between groups.**

7.5.5 Kinetics

Maximum and minimum absolute and normalised by body mass vertical, propulsive and lateral ground reaction forces were assessed. Three dimensional hip, knee and ankle moments normalised by body mass at the seven events of the gait cycle were also selected for analysis. Finally maximum and minimum total hip, knee and ankle power normalised by body mass were assessed. Mean, standard deviation, between group bias, and p-value were calculated and are presented in Tables 7.15-7.27. Values are also presented in Figures 7.5-7.11.

7.5.5.1 Ground reaction forces

Maximum vertical (Z), propulsive (X) and medial (Y) ground reaction forces corrected for body mass were significantly reduced in children who were overweight (Table 7.15) (Figure 7.5). Maximum absolute vertical (Z) ground reaction force was significantly greater in children who were overweight (Table 7.16) (Figure 7.6). There was no significant difference in absolute propulsive (X) or absolute medial ground reaction force (Y) between the groups (Table 7.16) (Figure 7.6).

Parameter (N/kg)	Overweight group (mean ± SD)	Healthy weight group (mean ± SD)	Bias	p-value
Force X	1.4 ± 0.8	2.2 ± 0.7	0.8	p = 0.0*
Force Y	0.5 ± 0.2	0.7 ± 0.3	0.2	*p* = 0.0*
Force Z	11.0 ± 2.1	12.4 ± 2.3	1.4	p = 0.0*

Table 7.15: Maximum ground reaction forces. p = t-test p-value with significance at alpha ≤ 0.05. *p* = Mann Whitney U p-value with significance at alpha ≤ 0.05. * = significant difference between groups.

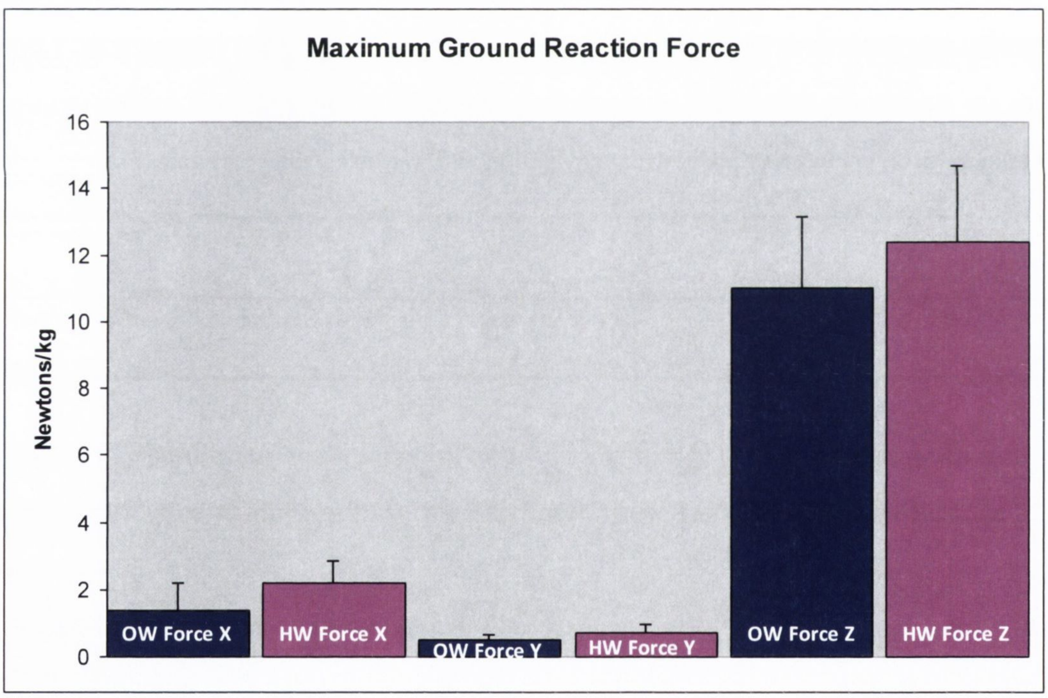


Figure 7.5: Maximum ground reaction forces.

OW = Overweight

HW = Healthy weight

Parameter (N)	Overweight group (mean ± SD)	Healthy weight group (mean ± SD)	Bias	p-value
Force X	103.7 ± 53.6	103.7 ± 29.5	0.1	$p = 0.99$
Force Y	41.3 ± 23.1	33.6 ± 8.8	-7.7	$p = 0.3$
Force Z	851.1 ± 258.7	605.5 ± 168.7	-245.6	$p = 0.0$*

Table 7.16: Maximum absolute ground reaction forces. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

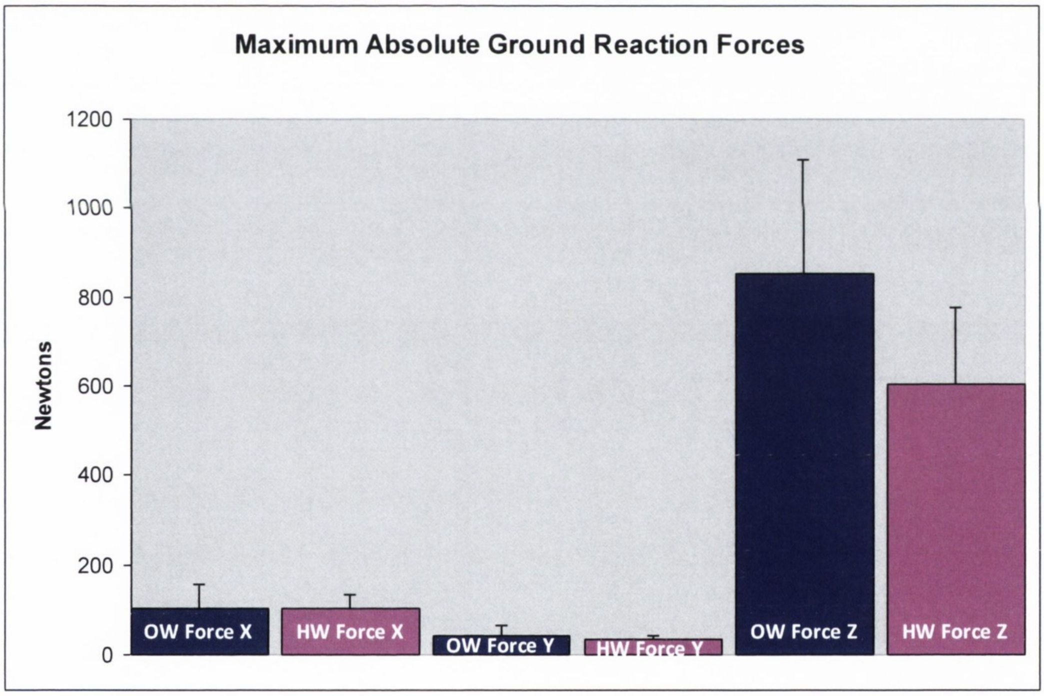


Figure 7.6: Maximum absolute ground reaction forces.

OW = Overweight

HW = Healthy weight

7.5.5.2 Sagittal plane joint moments

A significant reduction in hip flexor moment was noted at heel rise and opposite initial contact for the overweight group (Table 7.17) (Figure 7.7). No significant changes in hip moments were noted at the remaining five time points (Table 7.17) (Figure 7.7).

Event	**Overweight group mean ± SD ($Nm^{-1}kg^{-1}$)**	**Healthy weight group mean ± SD ($Nm^{-1}kg^{-1}$)**	**Bias**	**p-value**
Initial contact	0.1 ± 0.2	0.2 ± 0.3	0.1	*p* = 0.4
Opposite toe off	0.8 ± 0.2	0.7 ± 0.3	0.1	p = 0.1
Heel rise	-0.1 ± 0.2	-0.2 ± 0.2	0.1	p = 0.0*
Opposite initial contact	-0.2 ± 0.2	-0.6 ± 0.2	0.3	p = 0.0*
Toe off	-0.3 ± 0.1	-0.4 ± 0.2	0.1	*p* = 0.2
Feet adjacent	-3.2×10^{-2} ± 0.2	-0.1 ± 0.2	0.1	*p* = 0.5
Tibia vertical	0.1 ± 0.4	0.2 ± 0.2	0.1	p = 0.2

Table 7.17: Results for hip extensor-flexor moments (+/-). p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

A reduced knee extensor moment was noted in the overweight group at opposite toe off (Table 7.18) (Figure 7.7). In contrast, a significantly increased knee extensor moment was noted at heel rise for the children who were overweight (Table 7.18) (Figure 7.7).

Event	**Overweight group mean ± SD ($Nm^{-1}kg^{-1}$)**	**Healthy weight group mean ± SD ($Nm^{-1}kg^{-1}$)**	**Bias**	**p-value**
Initial contact	-0.1 ± 0.1	-0.2 ± 0.1	0.0	*p* = 0.8
Opposite toe off	0.5 ± 0.2	0.6 ± 0.2	0.2	p = 0.0*
Heel rise	0.2 ± 0.3	$4.3x10^{-2}$ ± 0.1	0.2	p = 0.0*
Opposite initial contact	0.3 ± 0.2	0.3 ± 0.1	0.0	p = 0.6
Toe off	$1.2x10^{-2}$ ± 0.1	$7.3x10^{-4}$± 0.1	0.0	p = 0.5
Feet adjacent	-0.1 ± 0.1	$-1.8x10^{-2}$ ± 0.1	0.1	*p* = 0.1
Tibia vertical	-0.2 ± 0.2	-0.2 ± 0.1	0.0	p = 0.6

Table 7.18: Results for knee extensor-flexor moments (+/-).p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05.

*** = significant difference between groups.**

At the ankle a reduction in plantarflexor moment was found for children who were overweight at heel rise and opposite initial contact (Table 7.19) (Figure 7.7). An increased dorsiflexor moment was noted at feet adjacent, at tibia vertical a decrease in dorsiflexor moment was found (Table 7.19).

Event	Overweight group mean ± SD ($Nm^{-1}kg^{-1}$)	Healthy weight group mean ± SD ($Nm^{-1}kg^{-1}$)	Bias	p-value
Initial contact	$-3.1 \times 10^{-3} \pm 0.0$	$-1.2 \times 10^{-2} \pm 0.0$	0.0	*p* = 0.2
Opposite toe off	0.4 ± 0.3	0.2 ± 0.2	0.1	*p* = 0.1
Heel rise	1.0 ± 0.2	1.1 ± 0.2	0.1	p = 0.0*
Opposite initial contact	1.0 ± 0.1	1.1 ± 0.2	0.1	p = 0.0*
Toe off	$-1.2 \times 10^{-2} \pm 0.0$	$-9.7 \times 10^{-3} \pm 0.0$	0.0	*p* = 0.0*
Feet adjacent	$-7.3 \times 10^{-3} \pm 0.0$	$-8.4 \times 10^{-3} \pm 0.0$	0.0	*p* = 0.0*
Tibia vertical	$-1.6 \times 10^{-2} \pm 0.0$	$-1.4 \times 10^{-2} \pm 0.0$	0.0	p = 0.4

Table 7.19: Results for ankle plantarflexor-dorsiflexor moments (+/-). p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

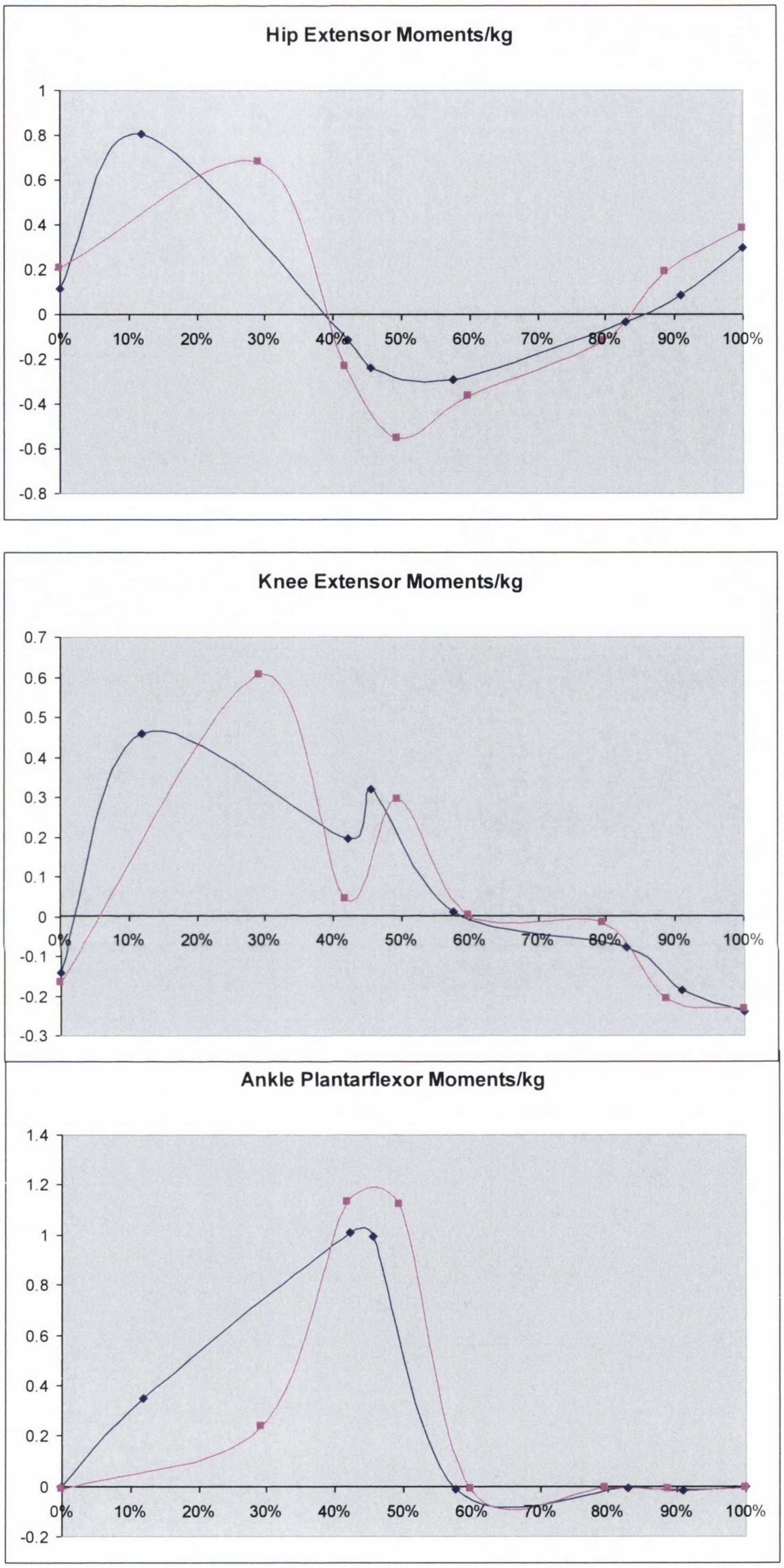


Figure 7.7: Sagittal plane joint moments.
X axis = percentage of one gait cycle.
Y axis = Newton-meters/kg.
The eight events of the gait cycle are demarcated on the lines.

Overweight Group

Healthy weight Group

7.5.5.3 Frontal plane joint moments

A significant increase in hip abductor moment was noted for children who were overweight at initial contact, toe off, feet adjacent, and tibia vertical (Table 7.20) (Figure 7.8).

Event	**Overweight group mean ± SD ($Nm^{-1}kg^{-1}$)**	**Healthy weight group mean ± SD ($Nm^{-1}kg^{-1}$)**	**Bias**	**p-value**
Initial contact	$3.8 \times 10^{-2} \pm 0.1$	-0.1 ± 0.1	0.1	p = 0.0*
Opposite toe off	0.5 ± 0.2	0.5 ± 0.1	0.0	p = 0.3
Heel rise	0.6 ± 0.2	0.5 ± 0.2	0.0	p = 0.5
Opposite initial contact	0.5 ± 0.2	0.5 ± 0.2	0.0	p = 0.9
Toe off	-0.1 ± 0.1	-0.2 ± 0.1	0.1	*p* = 0.0*
Feet adjacent	0.1 ± 0.1	0.0 ± 0.1	0.1	p = 0.0*
Tibia vertical	0.1 ± 0.1	-0.1 ± 0.1	0.2	p = 0.0*

Table 7.20: Results for hip abductor-adductor moments (+/-). p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

At the knee, a significant increase in knee valgus moment was noted for all time points excluding initial contact and tibia vertical (Table 7.21) (Figure 7.8).

Event	**Overweight group mean ± SD ($Nm^{-1}kg^{-1}$)**	**Healthy weight group mean ± SD ($Nm^{-1}kg^{-1}$)**	**Bias**	**p-value**
Initial contact	$3.8x10^{-2}$± 0.0	$2.2x10^{-2}$± 0.0	0.0	p = 0.1
Opposite toe off	0.2 ± 0.1	± 0.2	0.2	*p* = 0.0*
Heel rise	0.2 ± 0.1	0.1 ± 0.2	0.1	p = 0.0*
Opposite initial contact	0.2 ± 0.1	0.1 ± 0.2	0.1	p = 0.0*
Toe off	$-2.4x10^{-2}$± 0.0	$-4.5x10^{-2}$± 0.0	0.0	p = 0.0*
Feet adjacent	0.1 ± 0.1	$1.4x10^{-2}$± 0.0	0.1	p = 0.0*
Tibia vertical	0.1 ± 0.1	± 0.1	0.0	*p* = 0.1

Table 7.21: Results for knee valgus-varus moments (+/-). p = t-test p-value with significance at alpha ≤ 0.05. *p* = Mann Whitney U p-value with significance at alpha ≤ 0.05.

*** = significant difference between groups.**

In the frontal plane, a significant increase in ankle pronator moment was noted at heel rise, opposite initial contact, and feet adjacent (Table 7.22) (Figure 7.8).

Event	**Overweight group mean ± SD ($Nm^{-1}kg^{-1}$)**	**Healthy weight group mean ± SD ($Nm^{-1}kg^{-1}$)**	**Bias**	**p-value**
Initial contact	$1.7x10^{-3}$± 0.0	$8.8x10^{-4}$± 0.0	0.0	p = 0.7
Opposite toe off	0.1 ± 0.1	$4.3x10^{-2}$± 0.1	0.0	*p* = 0.8
Heel rise	0.4 ± 0.1	0.2 ± 0.1	0.1	p = 0.0*
Opposite initial contact	0.4 ± 0.1	0.2 ± 0.1	0.2	p = 0.0*
Toe off	$-3.3x10^{-3}$± 0.0	$-3.6x10^{-3}$± 0.0	0.0	p = 0.8
Feet adjacent	$6.8x10^{-3}$± 0.0	$-2.8x10^{-3}$± 0.0	0.0	p = 0.0*
Tibia vertical	$-2.5x10^{-3}$± 0.0	$4.6x10^{-5}$± 0.0	0.0	p = 0.4

Table 7.22: Results for ankle pronator-supinator moments (+/-). p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

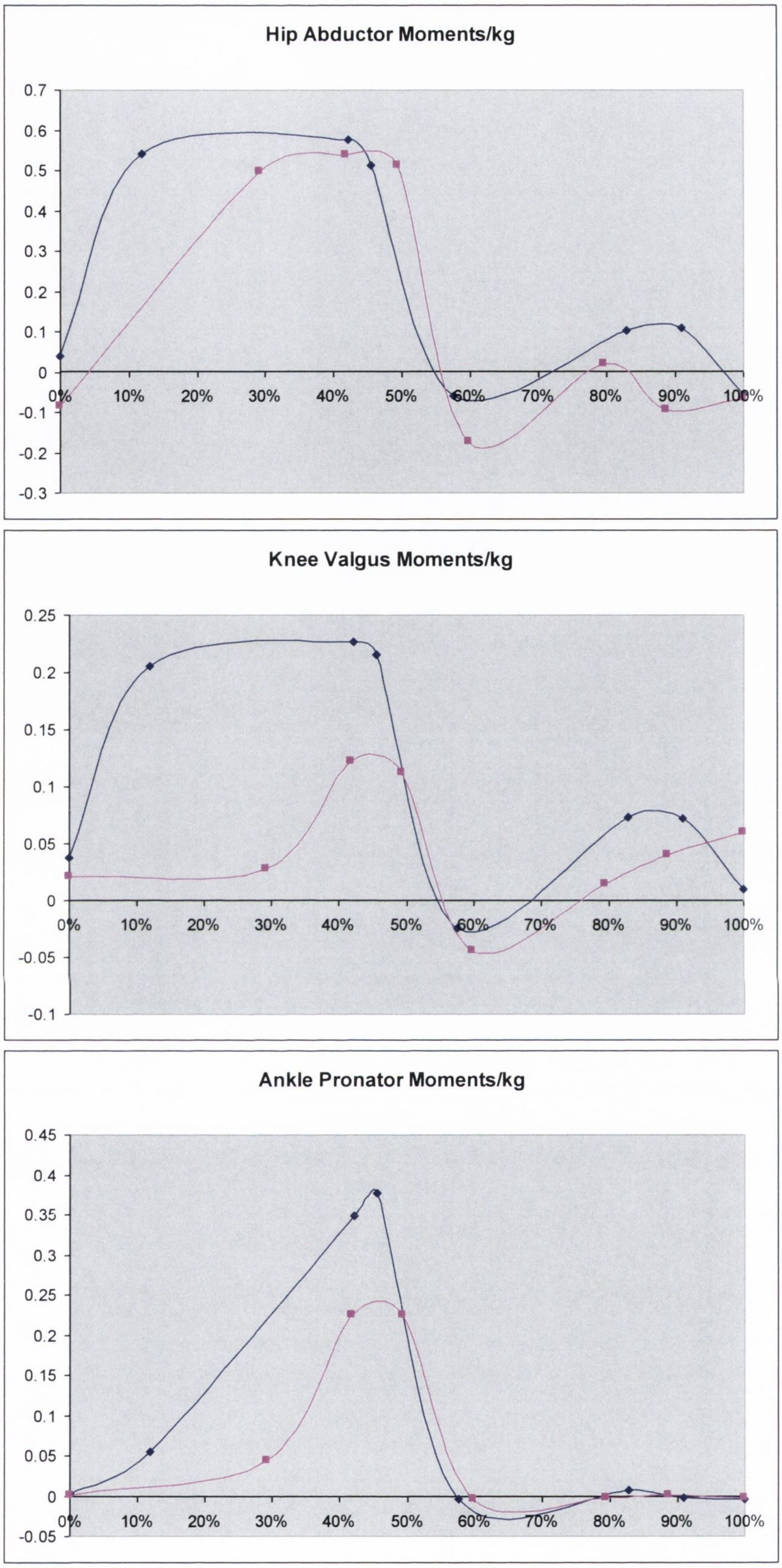


Figure 7.8: Frontal plane joint moments.

Overweight Group

X axis = percentage of one gait cycle.

Healthy weight Group

Y axis = Newton-meters/kg.

The eight events of the gait cycle are demarcated on the lines.

7.5.5.4 Transverse plane joint moments

At the hip in the transverse plane, a significant increase in hip external rotator moment was noted at five time points for children who were overweight (Table 7.23) (Figure 7.9). No significant differences were noted at opposite initial contact and toe off (Table 7.23) (Figure 7.9).

Event	Overweight group mean ± SD ($Nm^{-1}kg^{-1}$)	Healthy weight group mean ± SD ($Nm^{-1}kg^{-1}$)	Bias	p-value
Initial contact	$-9.9 \times 10^{-4} \pm 0.0$	$-1.7 \times 10^{-2} \pm 0.0$	0.0	*p* = 0.0*
Opposite toe off	0.1 ± 0.0	0.1 ± 0.0	0.0	p = 0.0*
Heel rise	$3.3 \times 10^{-2} \pm 0.0$	$1.6 \times 10^{-3} \pm 0.0$	0.0	p = 0.0*
Opposite initial contact	$4.1 \times 10^{-2} \pm 0.0$	$3.0 \times 10^{-2} \pm 0.0$	0.0	p = 0.2
Toe off	$-1.9 \times 10^{-2} \pm 0.0$	$-3.1 \times 10^{-2} \pm 0.0$	0.0	p = 0.1
Feet adjacent	0.1 ± 0.1	$7.4 \times 10^{-3} \pm 0.0$	0.0	p = 0.0*
Tibia vertical	$1.8 \times 10^{-2} \pm 0.0$	$-2.3 \times 10^{-2} \pm 0.0$	0.0	p = 0.0*

Table 7.23: Results for hip external-internal rotator moments (+/-).p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

At the knee, an increased knee external rotator moment was noted from opposite toe off to feet adjacent for children who were overweight (Table 7.24) (Figure 7.9). No significant differences in knee rotator moment were noted at initial contact and tibia vertical (Table 7.24) (Figure 7.9).

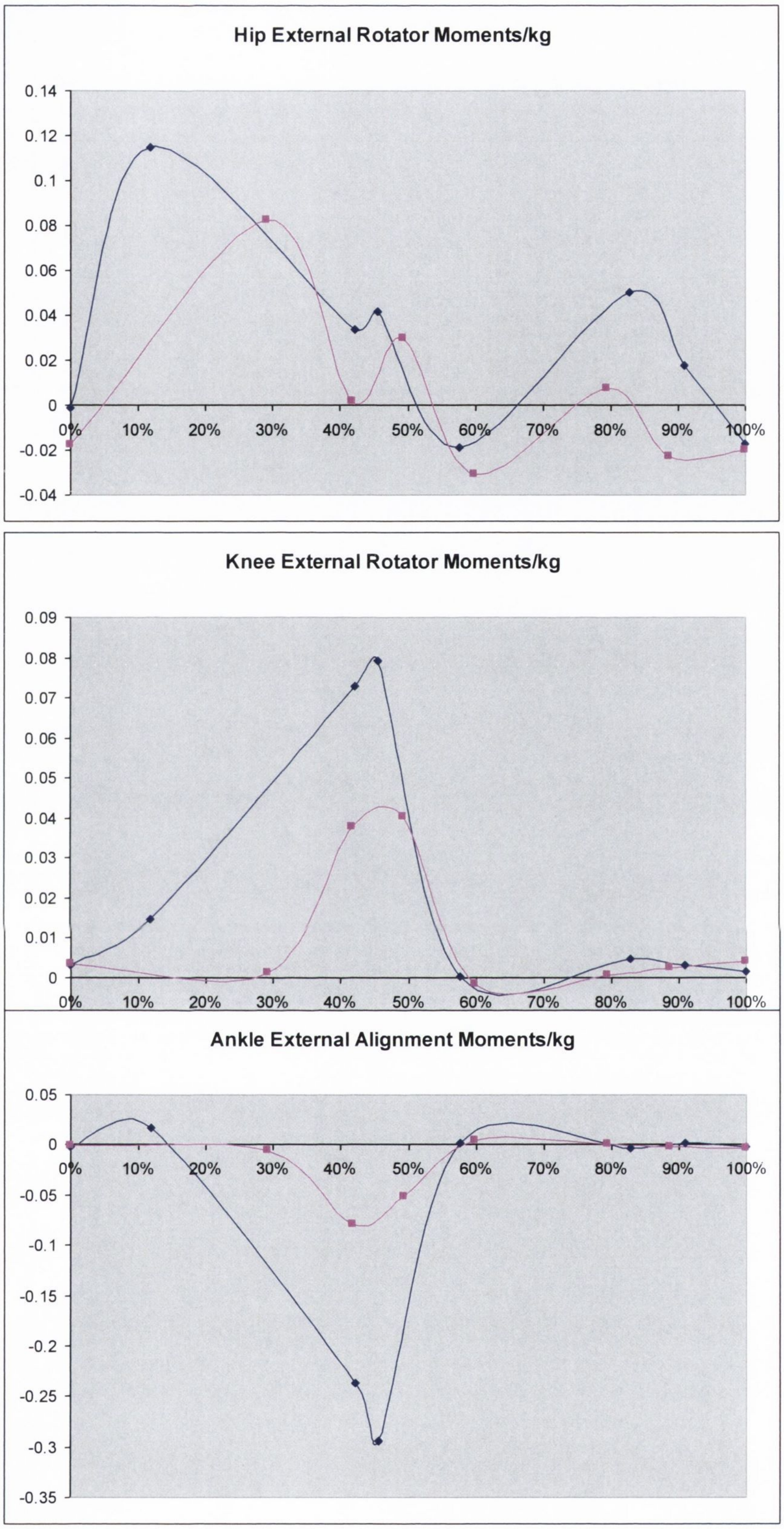
Event	**Overweight group mean ± SD ($Nm^{-1}kg^{-1}$)**	**Healthy weight group mean ± SD ($Nm^{-1}kg^{-1}$)**	**Bias**	**p-value**
Initial contact	$2.9 \times 10^{-3} \pm 0.0$	$3.2 \times 10^{-3} \pm 0.0$	0.0	*p* = 0.9
Opposite toe off	$1.4 \times 10^{-2} \pm 0.0$	$1.2 \times 10^{-3} \pm 0.0$	0.0	p = 0.0*
Heel rise	0.1 ± 0.0	$3.8 \times 10^{-2} \pm 0.1$	0.0	p = 0.0*
Opposite initial contact	0.1 ± 0.0	$4.0 \times 10^{-2} \pm 0.1$	0.0	p = 0.0*
Toe off	$2.3 \times 10^{-4} \pm 0.0$	$-1.8 \times 10^{-3} \pm 0.0$	0.0	p = 0.0*
Feet adjacent	$4.6 \times 10^{-3} \pm 0.0$	$4.3 \times 10^{-4} \pm 0.0$	0.0	p = 0.0*
Tibia vertical	$2.9 \times 10^{-3} \pm 0.0$	$2.4 \times 10^{-3} \pm 0.0$	0.0	p = 0.7

Table 7.24: Results for knee external-internal rotator moments (+/-). p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

For ankle joint moments in the transverse plane an increased internal joint alignment moment was noted at heel rise, opposite initial contact, toe off and feet adjacent for children who were overweight (Table 7.25) (Figure 7.9). An increase in external alignment joint moment was found at tibia vertical (Table 7.25) (Figure 7.9).

Event	**Overweight group mean ± SD ($Nm^{-1}kg^{-1}$)**	**Healthy weight group mean ± SD ($Nm^{-1}kg^{-1}$)**	**Bias**	**p-value**
Initial contact	$-2.3x10^{-3} \pm 0.0$	$-1.3x10^{-3} \pm 0.0$	0.0	p = 0.4
Opposite toe off	$1.7x10^{-2} \pm 0.1$	$-6.4x10^{-3} \pm 0.0$	0.0	*p* = 0.3
Heel rise	-0.2 ± 0.2	-0.1 ± 0.1	0.2	p = 0.0*
Opposite initial contact	-0.3 ± 0.2	-0.1 ± 0.1	0.2	p = 0.0*
Toe off	$2.0x10^{-3} \pm 0.0$	$4.4x10^{-3} \pm 0.0$	0.0	p = 0.0*
Feet adjacent	$-3.9x10^{-3} \pm 0.0$	$5.2x10^{-4} \pm 0.0$	0.0	p = 0.0*
Tibia vertical	$2.3x10^{-3} \pm 0.0$	$-2.1x10^{-3} \pm 0.0$	0.0	*p* = 0.0*

Table 7.25: Results for ankle external-internal alignment moments (+/-). p = t-test p-value with significance at alpha $\leq$ 0.05. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.



Overweight Group

Healthy weight Group

Figure 7.9: Transverse plane joint moments.
X axis = percentage of one gait cycle.
Y axis = Newton-meters/kg.
The eight events of the gait cycle are demarcated on the lines.

7.5.5.5 Joint Power

There were no significant differences for maximum total hip and knee joint power generation (concentric activity) (Table 7.26) (Figure 7.10). A significant increase in maximum total ankle power generation was noted for children who were overweight (Table 7.26) (Figure 7.10).

Parameter (W/kg)	Overweight group (mean ± SD)	Healthy weight group (mean ± SD)	Bias	p-value
Hip	1.8 ± 0.6	2.5 ± 1.6	0.7	$p = 0.1$
Knee	2.2 ± 2.4	1.6 ± 1.5	0.6	$p = 0.6$
Ankle	5.9 ± 2.3	4.1 ± 1.7	1.8	$p = 0.0$*

Table 7.26: Maximum total joint power generation. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

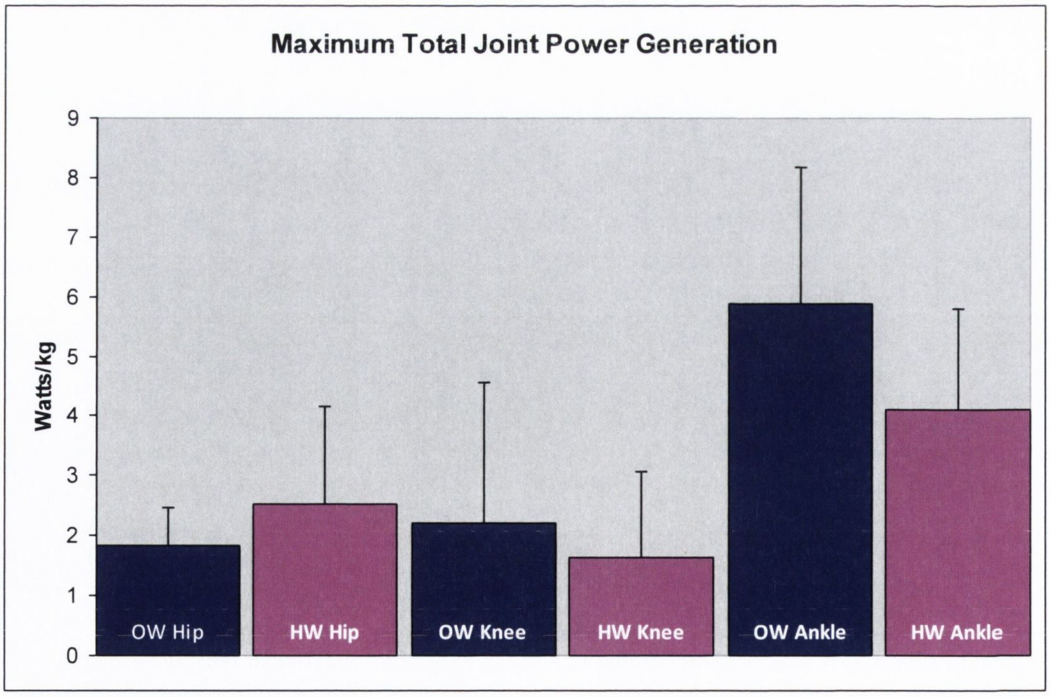


Figure 7.10: Maximum total joint power generation.

OW = Overweight HW = Healthy weight

Maximum total hip joint power absorption (eccentric activity) was reduced for children who were overweight (Table 7.26) (Figure 7.11). No significant differences were noted for maximum knee and ankle joint power absorption (Table 7.26) (Figure 7.11).

Parameter (W/kg)	Overweight group (mean ± SD)	Healthy weight group (mean ± SD)	Bias	p-value
Hip	0.6 ± 0.4	1.2 ± 1.0	0.6	$p = 0.0$*
Knee	2.8 ± 1.6	2.6 ± 1.1	0.2	$p = 0.9$
Ankle	1.8 ± 0.6	1.7 ± 1.1	0.1	$p = 0.2$

Table 7.27: Maximum total joint power absorption. *p* = Mann Whitney U p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

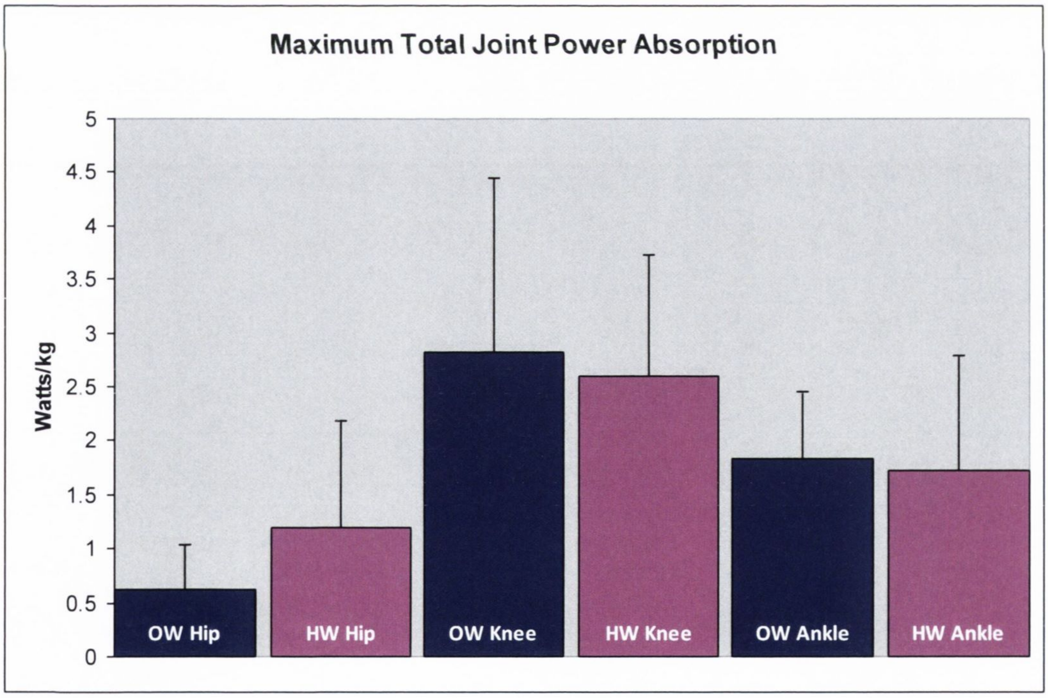


Figure 7.11: Maximum total joint power absorption.

OW = Overweight HW = Healthy weight

7.6 RESULTS STUDY 4(b)

The results presented here represent gait parameters for children who are overweight, as compared to gait parameters for children of a healthy weight walking in time to an auditory cue. The auditory cue represented the self selected cadence of their matched overweight participant. Results for temporal-spatial parameters are presented in Table 7.28. A significant difference in velocity, stride length, double support, and single support duration were noted (Table 7.28). No significant differences for cadence or stance phase duration were noted between the two groups (Table 7.28).

Parameter	**Overweight group (mean ± SD)**	**Healthy weight group (mean ± SD)**	**Bias**	**p-value**
Velocity (ms^{-1})	1.2 ± 0.2	1.3 ± 0.2	0.2	p = 0.0*
Stride Length (m)	1.1 ± 0.2	1.3 ± 0.1	0.1	*p* = 0.0*
Cadence (steps/min)	124.7 ± 15.9	124.8 ± 43.6	0.1	*p* = 0.8
Stance phase (%)	63.4 ± 2.1	62.1 ± 12.3	-1.3	*p* = 0.6
Single Support (s)	0.4 ± 0.1	0.4 ± 0.1	0.0	p = 0.0*
Double Support (s)	0.1 ± 0.0	0.1 ± 0.0	0.0	*p* = 0.0*

Table 7.28: Results for temporal-spatial parameters. p = t-test p-value with significance at alpha ≤ 0.05. *p* = Mann Whitney U p-value with significance at alpha ≤ 0.05. * = significant difference between groups.

7.7 DISCUSSION

The aim of this chapter was to provide a comprehensive overview of the presentation of gait in children who are overweight. 50 children were recruited and assessed, twenty-five of which were classified as overweight/obese according to IOTF cut off points (Chapter 1, page 3).

In order to eliminate the confounding influence of velocity, children of a healthy weight were required to walk in time to a beat. This beat was set to their overweight matches' self selected cadence. However, a significant difference in velocity remained (Table 7.28). Children who were of a healthy weight were not capable of maintaining the prescribed imposed cadence. Therefore the full analysis of cued data for study 4(b) was not included in this thesis.

The results presented in section 7.5 and discussed here are from the analysis of gait parameters with both groups ambulating at their self selected velocity.

7.7.1 Temporal-spatial characteristics

Children who were overweight were found to reduce their walking velocity with short strides taken at a similar cadence to their non-overweight counterparts (Table 7.2). Hills & Parker (1991a, 1991b) also noted a reduced walking velocity in their overweight groups. In contrast to the current chapter, they found this reduction to be the result of shorter stride length and a reduced cadence (Hills & Parker 1991a). Due to the reported constant linear association between velocity, stride length and cadence the finding of the current chapter, whereby cadence was not different between the groups, was unexpected (Schwartz et al 2008). During gait, the centre of gravity is moved alternately from left to right with weight transference from the back foot to the forward foot (Watkins 2010). As weight is transferred towards the toes, momentum carries the body forward initiating swing of the trailing leg

(Watkins 2010). Momentum is the quantity of motion of a moving body, measured as a product of its mass and velocity (Oxford Dictionaries 2011). The higher frequency of steps taken by the children who were overweight may be the result of an inability to control momentum and therefore dynamic stability during gait.

A reduction in velocity is associated with an increase in support time (Schwartz et al 2008). An increase in support time was adopted by children who were overweight, with an increase in single and double support duration noted (Table 7.2). These results are supported by earlier research by McGraw et al (2000).

For the overweight children here, a reduced velocity, shortened stride length and increase in support time may have been adopted in an attempt to improve stability, as a result of reduced muscle strength, in an effort to minimise loads across weight bearing joints, or to minimise the energy cost of walking (Andriacchi et al 1977; Elsworth et al 2005; Goldberg & Neptune 2007; England & Granata 2007).

7.7.2 Kinematics and joint moments

Feet adjacent and tibia vertical occurred later in the children who were overweight. This was expected as a reduction in the duration of swing phase is associated with a reduction in velocity (Andriacchi et al 1977).

7.7.2.1 The sagittal plane

Several differences in joint kinematics and moments were noted in the sagittal plane. At initial contact there were no significant differences between the two groups for all joints assessed (Tables 7.3-7.6; 7.17-7.19).

No significant differences in pelvic kinematics were noted (Table 7.3). This is the first study to assess pelvic kinematics in individuals who are overweight. The position of the pelvis contributes to the components of joint range at the hip (Chapter 3, page

73). It is possible therefore for us to state that differences seen at the hip are from femoral on pelvic kinematics, and not from pelvic on femoral kinematics.

At the hip in the sagittal plane significant changes were seen at heel rise, opposite initial contact and toe off (Table 7.4, 7.17). An increase in hip flexion at opposite initial contact and toe off were found, with an associated reduced hip flexor moment at heel rise and opposite initial contact. In the study by Hills & Parker (1991a), an increase in peak hip flexion was found. While this chapter did not directly assess peak hip flexion, from figure 7.2 peak hip flexion occurs at tibia vertical where no significant difference between the two groups were found (Table 7.4). In contrast to the results of this chapter, two studies reported a decrease in hip flexion for their groups of children who were overweight (Gushue et al 2005; McMillan et al 2010). In addition a significant decrease in hip extensor moment at initial contact, and an increase in hip flexor moment at late stance was noted (McMillan et al 2010).

Knee flexion was found to increase at heel rise, toe off and tibia vertical in children who were overweight (Table 7.5). Hills & Parker (1991a) noted an increase in peak knee flexion which occurs close to feet adjacent, and where no significant between group differences were found for the current sample (Figure 7.2). As for the hip, two studies reported a reduction in flexion for their overweight groups (Gushue et al 2005, McMillan et al 2010). In contrast to the research by Gushue et al (2005), alterations in knee joint moments were found in the present chapter. Knee extensor moment was noted as reduced at opposite toe off, and increased at heel rise (Table 7.18). In dispute of an earlier study by McMillan et al (2010) knee flexor moment was found to be increased at feet adjacent (Table 7.18).

With a more flexed attitude the body's centre of gravity is lowered and stability improves (Watkins 2010). The results from the hip and knee in the sagittal plane support the hypothesis of instability in children who are overweight. The stance phase events which exhibited between group differences occurred during the latter half of the stance phase, and prior to the swing phase where stability is compromised the most (Table 7.3-7.5) (Watkins 2010). This may suggest that children who are overweight adopt a more flexed attitude during terminal stance in preparation for the challenge of stability during swing phase.

With increased flexion the GRF vector moves further from the joint centre, the loads placed on weight bearing joints are minimised, and greater demands are placed upon the muscles (Chapter 1, page 12). The results at the hip and knee may be deemed an effort to minimise loads across weight bearing joints (Table 7.4-7.5). Increased muscle strength and energy expenditure is required to maintain a flexed presentation during gait (Hicks et al 2008). This suggests children who are overweight do not alter their gait in order to compensate for muscle weakness or to minimise the energy cost of walking.

At the ankle a reduced ankle plantarflexor moment was recorded at heel rise with no associated change in ankle kinematic parameters (Table 7.6, 7.19). This suggests that for children who were overweight heel rise occurred as a result of the significant increase in knee flexion as opposed to the plantarflexor moment generated by the children of a healthy weight (Table 7.5, 7.19). Ankle dorsiflexion was noted to increase at opposite initial contact with a resultant reduction in ankle plantarflexor moment in the overweight group (Table 7.6, 7.19). Reduced plantarflexor moment at terminal stance results in a reduced push-off force, and therefore a reduced step length and velocity (Yang & Pai 2010). This would help to improve stability during gait and further supports the relationship between dynamic instability and increased adiposity. In support of these findings, McMillan et al (2010) noted a reduction in peak ankle plantarflexor moment for their overweight group. Finally, at feet adjacent an increase in ankle plantarflexion was recorded (Table 7.6). The increased ankle plantarflexion may explain the increased knee flexion at tibia vertical, shortening the limb to enable clearance of the toes during swing phase.

7.7.2.2 The frontal plane

No significant differences in pelvic kinematics were noted between the two groups for six of the seven events (Table 7.7). An increase in pelvic obliquity was noted at tibia vertical for children who were overweight (Table 7.7). As pelvic differences were few, alterations at the hip can be presumed to be the result of altered femoral on pelvic kinematics, and not pelvic on femoral kinematics (Chapter 3, page 73). An increase in hip abduction was noted for six of the seven events (Table 7.8). An increase in hip abduction is associated with an increased base of support and a lowering of the centre of gravity (Watkins 2010). If this difference is an adaptation made by children who are overweight it may suggest the presence of instability during the gait cycle. However, this increase in hip abduction may be explained by the significant difference in thigh girth between the two groups (Table 7.1).

With an increase in hip abduction the GRF vector passes medial to the hip joint centre, and an external hip adduction moment is created with a resultant internal hip abductor moment. An increase in hip abductor moment was noted at the start and end of the stance phase, and for the swing phase (Table 7.8). From opposite toe off to opposite initial contact no significant difference in hip abductor moment was noted between the groups (Table 7.20). This is the period where the contralateral limb is abducted in swing phase. Weight is transferred onto the stance limb and the GRF vector is displaced laterally, closer to the hip joint centre, resulting in a reduced hip abductor moment. These findings are in contrast to earlier work by McMillan et al (2009) who found a significant increase in hip adduction and an associated reduction in hip abduction moments.

As discussed in Chapter 6, page 95, joint range of motion in the frontal plane at the knee and ankle is small, and may therefore be subject to misinterpretation as a result of the degree of accuracy of the CODA Motion (Charnwood Dynamics Ltd., Leicestershire, UK) (Richards 1999). This should be kept in mind as the results for frontal knee and ankle movement of the current study are discussed.

A significant increase in knee varus was noted at feet adjacent for the children who were overweight (Table 7.9). With varus alignment greater loads are placed upon the medial tibio-femoral compartment. As a result of this increase, children who were overweight exhibited significantly increased knee valgus moment during feet adjacent (Table 7.21). Whilst no additional kinematic differences were noted between the two groups, an increased knee valgus moment was found from opposite toe off to feet adjacent for children who were overweight (Table 7.21). This internal valgus moment prevented the knee joint from being positioned in knee varus for six of the seven events of the gait cycle, minimising the effects of increased medial joint loading. This finding is supported by the study by McMillan et al (2009) who noted a significant increase in knee valgus moment for their overweight group. However later work disputes the results, with a reduced knee abduction moment reported for children who were overweight (McMillan et al 2010).

Increased supination was noted at the ankle at initial contact, heel rise, opposite initial contact, and tibia vertical (Table 7.10). An associated increased pronator moment was noted at heel rise, opposite initial contact and feet adjacent (Table 7.22). An increase in supination results in 'locking' of the joints distal to the subtalar joint and provides a degree of protection to these joints with a strong and stable base (Johanson et al 2008). However in excessive supination the foot acts as a rigid lever at time points where it should be more flexible to enable adequate GRF attenuation (Johanson et al 2008).

7.7.2.3 The transverse plane

Only one study has previously addressed transverse plane biomechanics in children who are overweight (Schultz et al 2009). They reported on total joint range and joint moments and found no significant difference between their groups at the hip, knee and ankle (Schultz et al 2009). In contrast, several differences were found between the groups of the current chapter (Tables 7.11-7.14, 7.23-7.25).

No significant differences in pelvic kinematics were noted for six of the seven events (Table 7.11). A decrease in forward rotation of the pelvis was noted at tibia vertical (Table 7.11). Once again, hip motion can be deemed to be as a result of femoral on pelvic motion and not pelvic on femoral motion. At the hip, children who were overweight exhibited a significant increase in external rotation at each time point, excluding initial contact and feet adjacent (Table 7.12). An increase in external rotator moment was also noted for each time point excluding opposite initial contact and toe off (Table 7.23). An increase in external rotation widens the base of support, improving stability (Watkins 2010). With both joint motion and joint moments encouraging external rotation, it may be reasonable to suggest that children who are overweight are actively rotating their hips externally to improve stability (Table 7.12 and 7.23).

An increase in internal rotation at the knee/tibia was noted for children who were overweight at six of the seven events of the gait cycle (Table 7.13). An associated increase in knee/tibial external rotator moment was also recorded for five of the seven events (Table 7.24).

Increased external alignment/out-toeing at the ankle was noted at initial contact and opposite initial contact in children who were overweight (Table 7.14). Increased internal rotator moments at the foot were noted from heel rise to toe off (Table 7.25). At initial contact and opposite initial contact, the body is transferring from a position of single limb support to double limb support (Chapter 1, page 8). Immediately prior to these time points balance is challenged as inertia is overcome and the body falls forwards (Watkins 2010). An increase in the external rotated position of the foot enables a greater base of support for the body to 'fall onto', and greater medial-lateral stability at the transition stages (Watkins 2010). During the swing phase the ankle has adopted an internally rotated position by feet adjacent, before returning to its externally rotated position in preparation for initial contact at tibia vertical (Table 7.14). The internally rotated position may be the result of increased internal rotator moments at toe off, which are then counteracted by increased external rotator moments by the time of tibia vertical (Table 7.25).

7.7.3 Ground reaction forces

This chapter also proposed to determine the effects of increased adiposity in children on maximum absolute ground reaction forces and maximum ground reaction forces corrected for body mass. Absolute maximum vertical GRF was found to be significantly greater in children who were overweight (Table 7.16; Figure 7.6). This would be anticipated due to the significantly greater body mass of the children who were overweight (Chapter 1, page 10).

Absolute maximum propulsive and medial ground reaction forces were not significantly different between the two groups (Table 7.16; Figure 7.6). Children who were overweight demonstrated reduced velocity, reduced stride length, and reduced plantarflexor moment in terminal stance (Table 7.2, 7.19). These significant changes all result in a reduction of the propulsive component of GRF, and therefore the load on weight bearing joints (Yang & Pai 2010). As described in Chapter 1 page 11, the medial component of the GRF is dependent on the position of the foot relative to the body's centre of mass. An increase in hip abduction and an externally rotated position of the foot, as seen in the children who were overweight, results in an increased base of support and an increased medial GRF (Simpson & Jiang 1999). A reduction in velocity is also associated with an increased medially directed GRF (Van der Linden et al 2001). With a mean difference of 7.7N representing 21% of the overall mean value of 37.5N, it is surprising that no significant differences in medial GRF were noted between the two groups.

All maximum GRF corrected for body weight were found to be reduced for children who were overweight (Table 7.15; Figure 7.5). This suggests that the temporal-spatial and kinematic changes seen for children who are overweight may have been the result of efforts to minimise the maximum loads incurred by their joints.

7.7.4 Joint power

The final objective of this chapter was to determine the effects of increased adiposity in children on maximum total hip, knee and ankle joint power generation and absorption (Tables 7.26-7.27; Figure 7.10-7.11).

A significant decrease in power absorption at the hip was noted for children who were overweight (Table 7.27). This suggests that eccentric activity of muscles at the hip was reduced for children who were overweight. Eccentric activity enables shock absorbing during gait, where energy absorbed is dissipated as heat or, as an energy source for use as kinetic energy during a concentric contraction (Lindstedt et al 2001). With a reduction in maximum eccentric activity at the hip, reduced energy is available to encourage forward progression, and protect the hip against increase loads. This may be a result of weakness of these muscle groups as eccentric activity requires greater muscle strength (Lindstedt et al 2002).

In contrast to the current chapter, Nantel et al (2006) found a significant increase in sagittal plane hip absorption. While no difference in power generation i.e. concentric activity at the hip was noted for the current chapter, a significant increase and a significant decrease in power generation have been previously reported at the hip (Nantel et al 2006, Schultz et al 2010). In contrast to work by Schultz et al (2010) who found an increase in power absorption at the knee, no significant differences in power generation or absorption were noted for children who were overweight in the current study (Tables 7.26-7.27).

A significant increase in total ankle power generation was found (Table 7.26). This suggests that concentric muscle work at the ankle is increased for children who are overweight. An individual's metabolic rate decreases when they are aiding forces (eccentric activity) and increases when they are resisting forces (concentric activity)

(Lindstedt et al 2001). This implies that children who are overweight are required to exert more energy to maintain forward progression during gait.

7.7 CONCLUSION

From the results of this chapter, children who are overweight adopt several changes to their gait pattern as compared to their non-overweight counterparts. These changes may be deemed as efforts to improve stability, and reduce loads across the weight bearing joints. However, several of the results discussed here may be explained by the significant reduction in velocity for children who are overweight.

CHAPTER 8 DISCUSSION AND FURTHER RESEARCH

8.1 INTRODUCTION

The main aim of this piece of work was to assess the relationship between body mass and gait parameters. More specifically temporal, spatial, three-dimensional kinematic, three-dimensional joint moments, maximum GRF and maximum and minimum total joint powers were examined. The question of the ability of auditory cueing to constrain velocity during gait was also posed.

From the literature review, several studies have previously addressed the question of the association between increased body mass and gait parameters (Chapter 2, page 30). However, the description of gait was poorly defined with several studies limiting their findings to temporal, spatial, total joint range of motion/moments/powers. The components of joint range/moments/powers were addressed by few studies. Reporting on total range can mask considerable discrepancies in the components of total range. In addition, there is sparse evidence outlining the presentation of gait in both frontal and transverse planes. Finally, pelvic kinematics have not previously been reported for this cohort. The present thesis aimed to investigate the influence of body mass on gait parameters in children and adults.

Age, gender, height and velocity have all been shown to influence gait parameters (Sutherland 1997; Whittle 2002; Schwartz et al 2008). For several studies age, gender and height were accounted for in the methodological design, or corrected for post-hoc using statistical methods. Ideally during gait analysis the patient in question should be free to walk in the way which is natural to them i.e. with no constraints, at their self selected velocity and over level ground. For those studies which assessed individuals who were overweight at their self selected velocity, the majority noted a significant difference between groups for velocity (Chapter 2, page 36). Despite this velocity was not controlled a priori, corrected for post hoc, or accounted for in the discussion of results. It was therefore difficult to determine whether the reported results were due to increased body mass as opposed to the difference in velocity between groups. The current thesis aimed to assess the influence of auditory cueing on gait parameters in healthy adults and children.

The forthcoming discussion is divided into two sections. Firstly, the influence of auditory cueing on gait parameters in health adults and children is examined. Secondly, the influence of body mass on gait parameters in adults and children is addressed.

8.2 THE USE OF AUDITORY CUEING IN GAIT ANALYSIS

Research in the area of velocity and gait has been explored in Chapter 1. In summary, with the exception of step width, for an increase/decrease in velocity there is a significant change in the temporal-spatial characteristics of gait (Table 1.1, page 20). An alteration in velocity has significant effects on sagittal, frontal and transverse joint range (Chapter 1, page 19). The peak values vary with velocity in a variety of linear and non linear ways. The degree to which velocity influences each angle is dependent on the angle itself and on the timing of the gait cycle. With the presence of a relationship between joint range and joint moments/powers, velocity has too been shown to alter kinetic parameters of the gait cycle (Chapter 1, page 22).

Previous research has used treadmills, verbal instruction and auditory cueing to dictate velocity, thus enabling between group comparisons independent of velocity (Kerrigan et al 1998; Kerrigan et al 2000; Browning & Kram 2007; Kubota et al 2007; Shultz et al 2009; Shultz et al 2010). Whilst treadmill training prescribes a set velocity, it does not give a true representation of a patients' self selected gait presentation. In addition several aspects of gait have been shown to be different for treadmill versus over-ground walking (Riley et al 2007; Terrier & Deriaz 2011). Verbal instruction and auditory cueing used in a control group enable analysis at a pathological groups' self selected velocity. Verbal instruction, while less disruptive than a treadmill, may not always result in a comparable velocity. The repeatability of gait with auditory cueing and the influence of auditory cueing on gait have received little attention. Bertram & Ruina (2001) assessed the influence of constraining velocity with a treadmill at several velocities, followed by cadence with auditory cueing at several related frequencies. A significant difference in the slope of speed-frequency curves for constrained velocity versus constrained cadence was established for ten of the twelve conditions (Bertram & Ruina 2001). This was not anticipated based on the reported constant linear association between velocity, cadence and step length (Schwartz et al 2008). The authors suggested the constraint resulted in unnatural gait and *'therefore the results generated from these situations are not indicative of natural walking'* (Bertram & Ruina 2001). Treadmill walking has previously been shown to result in altered gait (Terrier & Deriaz 2011). The use of a treadmill in the study by Bertram & Ruina (2001) may have influenced their results.

Post-hoc methods aimed at predicting gait parameters at any given velocity from a set of reference values have been assessed (Lelas et al 2003; Hanlon & Anderson 2006; Stansfield et al 2006). Kinetic parameters were found to be more predictable than kinematic parameters (Lelas et al 2003). For all parameters predictions from the same speed range as that on which the reference data was based, were found to be more accurate than predictions based on speeds outside of the reference range (Hanlon & Anderson 2006; Stansfield et al 2006). It was therefore recommended that predictions be based upon reference data of slow-walking healthy adults (Hanlon & Anderson 2006). This conclusion is hardly surprising as healthy adults asked to walk slowly, at a velocity encompassed by the range in velocity adopted by their pathological counterparts, will not be walking significantly faster and therefore prediction should not be required.

In statistical analyses, a priori statistical significance testing is more powerful than post hoc significance testing (Kuehne 1993). To maximise the power of results, a priori correction for velocity was selected for this piece of work. As discussed previously, a linear relationship between velocity, cadence and stride length have been reported (Schwartz et al 2008). Therefore velocity may be constrained by controlling cadence or stride length. Controlling cadence with the use of an auditory cue was selected to be examined further. Applying visual aids to dictate step and stride length would result in stride length, velocity and cadence constraints. However step width, and associated kinematic and kinetic parameters in the frontal plane would also be controlled. In addition the use of various cadences as dictated by an auditory cue was deemed to be a more feasible experimental protocol, opposed to various visual aid placements dependant on step and stride length.

In order to adopt the use of an auditory cue, the influence of cueing on gait was assessed first (Chapter 4, page 76 and Chapter 5, page 88). Healthy participants were required to ambulate at their self selected velocity, and also in time to an auditory cue set at their self selected cadence. Results of a self selected gait trial as compared to a cued gait trial were analysed with respect to natural variation in gait. The results of these studies enabled an assessment of the method (auditory cue) by which the imposition of velocity was made as opposed to the actual imposed velocity. In order to determine whether an auditory cue increases variation in the presentation of gait, a comparison to the natural variation in gait is required.

For adults (Study 1) the limits of agreement for cadence were wider with the imposition of an auditory cue (-14.2 – 13.8) than for natural variation (-8.0 – 9.4). This was not the case for velocity or stride length where similar widths were noted between test conditions as compared to natural variation. This suggests that the imposition of an auditory cue resulted in a difference in the velocity-cadence relationship. Despite this, no significant difference between test conditions were noted for temporal, spatial and, with the exception of hip extension and external femoral rotation, three-dimensional kinematic parameters studied. When compared to natural variation few discrepancies in the limits of agreements for three-dimensional kinematic parameters were recorded, the addition of an auditory cue actually reducing variation as compared to natural variation for several parameters. It was concluded that auditory cueing may be used in healthy adults, as the majority of discrepancies noted may be explained by natural variation in gait.

For Studies 3 (a) and 3(b), healthy adults were required to ambulate at their self selected velocity, and to an auditory cue set to the cadence of their overweight match. It is known from Study 1 that the method of auditory cueing to impose a cadence does not alter the presentation of gait in healthy adults. In Study 3 (a), no significant difference was noted between groups ambulating at their self selected velocity ($p = 0.1$). In addition no significant difference was noted between adults who were overweight ambulating at their self selected velocity as compared to adults of a healthy weight ambulating to an auditory cue (Study 3(b)) ($p = 0.2$). Therefore, the data sets act in effect as a source of information to assess the influence of imposing a cadence on healthy adults. The analyses of both data sets yielded similar findings for temporal, spatial, kinematic and kinetic parameters. It is granted that as no significant difference in velocity was found between groups ambulating at their self selected velocity, the imposed cadence may not have been too dissimilar to the healthy participants self selected cadence. However, the results give an indication of the influence of imposing a cadence on healthy adults. Therefore, not only the method of auditory cueing, but the imposition of a cadence different from a self selected cadence does not appear to alter gait in healthy adults.

Similarly for children (Study 2) no significant differences were seen between test conditions for the all temporal, spatial and three-dimensional kinematic parameters assessed. Compared to natural variation, limits of agreement were similar or marginally wider with the addition of an auditory cue for all parameters. The largest discrepancy in limits of agreement was found for foot external alignment with limits of agreement of -9.3 – 12.1 for the addition of an auditory cue as compared to -5.3 – 5.4 for natural variation. As for adults it was found that auditory cueing does not result in excessive additional variation in children beyond that seen for natural variation.

Similar to adults, in Studies 4 (a) and 4 (b) healthy children were required to ambulate at their self selected cadence, and to an auditory cue set to the cadence of their overweight match. A significant difference in velocity was noted between overweight and healthy weight groups walking at their self selected velocity ($p<0.05$). From Study 2, auditory cueing did not alter the presentation of gait in healthy children ambulating at their self selected cadence. However, in Study 4 (b) the significant difference in velocity persisted when children of a healthy weight walked in time to a cadence dictated by the auditory cue ($p<0.05$). Whilst children managed to maintain the prescribed cadence, velocity and stride length remained significantly different between the two groups. Therefore, the imposition of a cadence different to self selected cadence in healthy children resulted in an altered presentation of gait.

Free walking requires limited attention, and so there is a residual attention capacity for an additional/dual task (Laessoe et al 2008). Attention is required when walking to an auditory cue, and so it may be described as dual task walking. Where one task requires greater attention, more of the attention resources must be allocated to this task (Huang & Mercer 2001). In situations where dual-task activity exceeds the resource capacity dual task interference occurs (Abernethy 1988). With interference comes a reduction in the quality of the tasks (Abernethy 1988). For Study 4 (b) children maintained the prescribed cadence with the auditory cue at the expense of their gait presentation. It has previously been noted that children's response to dual/triple task activities is different to adults (Schachar & Logan 1990). Adults ensure the primary task is conducted correctly (gait), whereas children rush when dual/triple task requirements are presented to them, at the expense of the quality of the primary task (Schachar & Logan 1990). Children were also shown to rush more for an increase in the number of tasks (Schachar & Logan 1990). For the current work greater demands were placed upon the residual attention capacity when ambulating to a prescribed cadence, as opposed to a self selected with an auditory cue.

The results of this thesis indicate for gait analysis research, a control group of healthy adults should be matched to the pathological group of interests self selected velocity with the use of external auditory cueing. This however is not the case for children, who demonstrated an altered gait in response to an imposed cadence. For adults, while the use of a cue allows for velocity to be constrained, it does not address the issue of asymmetry during pathological gait. In addition, due to time constraints it may not be feasible for use clinically. Auditory cueing is recommended in the research setting to enable a clearer picture of the presentation of gait independent of velocity for pathological groups.

8.3 THE INFLUENCE OF BODY MASS ON GAIT PARAMETERS

8.3.1 Adults

For Studies 3 (a) and 3 (b) (Chapter 6, page 100) gait for adults who were overweight was compared to gait for age, gender and height matched adults of a healthy weight. Analyses of gait with both groups ambulating at their self selected velocity, and with adults of a healthy weight walking in time to a cue set to the cadence of their overweight match were completed. No significant differences in velocity were noted for both analyses. The resulting analyses were similar and further discussion relates to data from both groups ambulating at their self selected velocity.

The absence of a significant difference between groups for velocity was surprising. There is a general consensus from the previous literature that a reduction in velocity is present for adults who are overweight (Chapter 2, page 30). In the study by Ko et al (2010) no significant difference in velocity was noted between their overweight and healthy weight groups, whereas a significant difference in velocity was noted between their obese and healthy weight groups. They noted a reduction in preferred gait speed with increasing BMI (p for trend <0.001) (Ko et al 2010). Similarly a reduction in velocity was noted for an increase in BMI for adults of the current thesis (p for trend = 0.0*). This suggests the possibility of a 'threshold effect' for a reduction in velocity with increased BMI.

From the previous literature, three studies have compared overweight groups to healthy weight groups ambulating at a similar velocity. DeVita & Hortobagyi (2003) compared groups ambulating at their self selected velocity and at a fast velocity of $1.5ms^{-1}$. A significant difference in velocity was noted for the self selected velocity whilst no significant difference was noted at the fast velocity (DeVita & Hortobagyi 2003). Browning & Kram (2007) compared groups walking on a treadmill at six predetermined velocities. From these studies, while independent of velocity, neither study assessed the self selected presentation of gait for adults who were overweight (DeVita & Hortobagyi 2003; Browning & Kram 2007). In order to assess the possible reasons for or implications of altered gait for adults who are overweight, it is important to assess and evaluate their natural presentation. Ko et al (2010) is the only study previously published that found no significant difference in self selected velocity between adults who were overweight as compared to their healthy weight peers. The results of the current study will therefore be compared to the findings of the study by Ko et al (2010).

From the current thesis stance phase duration was extended for adults who were overweight, with an increase in stance phase and double support duration noted ($p<0.05$). This was in contrast to the study by Ko et al (2010) who noted no alteration in stance phase duration for their overweight group as compared to their healthy weight group. For the study by Ko et al (2010), sagittal and frontal total hip, knee and ankle range of moment, peak hip, knee and ankle moments, and total hip, knee and ankle generative and absorptive power were assessed (Ko et al 2010). No significant differences were noted for all parameters studies between the two groups (Ko et al 2010). Similarly for the current piece of research, no changes were noted for frontal, sagittal and transverse pelvic kinematics, ankle kinematics, ankle joint moments, or for peak total power generation at the hip, knee or ankle. Finally, no alterations to peak total power absorption at the knee and ankle were noted.

As discussed previously, total range of motion can mask considerable alterations in the components of joint range. This is confirmed by the significant between group differences noted in the components of joint range by this thesis. At initial contact a significant increase in hip flexion was noted, and a significant decrease in hip abductor moment ($p<0.05$). In the frontal plane an increase in hip abduction was noted at toe off and feet adjacent for adults who were overweight ($p<0.05$). Significant differences were seen at the knee for all three planes of motion. In the sagittal plane an increased knee flexion was noted at tibial vertical, for the frontal plane an increase in varus was seen at feet adjacent, and for the transverse an increased internal rotator moment was seen at toe off ($p<0.05$). For adults who were overweight absolute GRF were significantly higher for all three dimensions. Vertical GRF corrected for body mass was significantly reduced for adults who were overweight ($p<0.05$). A significant reduction in maximum total hip power absorption was also recorded ($p<0.05$).

For the study by Ko et al (2010) the aim was to characterise the gait pattern of overweight older adults. As such the mean age of their participants was higher than that of the current study (67 years as compared to 39 years) (Ko et al 2010). In addition, those with musculoskeletal complaints that did not result in '...*severe pain*' were not excluded (Ko et al 2010). Knee osteoarthritis has been shown to lead to altered biomechanics (Heiden et al 2009). For the study by Ko et al (2010) participants with knee osteoarthritis were included in both the healthy and overweight groups. Whilst for their study comparisons were made between similar groups, meeting their study objectives, a description of gait for those with increased weight as compared to healthy controls was not provided (Ko et al 2010). The current thesis contributes new information to the area of gait analysis in adults who are overweight, by providing the first comprehensive overview of the influence of weight on gait in adults independent of velocity, and musculoskeletal disease.

Interestingly, for the current research changes in kinematics and joint moments occurred at the end of stance phase (toe off), during swing phase (feet adjacent and tibia vertical) and at the start of the stance phase (initial contact). No changes were seen from opposite toe off to toe off. Stance phase duration was longer for adults who were overweight despite ambulating at a similar velocity to their non-overweight peers. The changes seen during the swing phase may be the result of body fat distribution, and efforts to minimise the swing phase duration by increasing the acceleration of the swinging limb. With a significant increase in thigh girth, overweight adults may be required to increase their hip abduction, to allow the swing limb to clear the stance phase limb as the limbs pass each other. The increase in hip flexion allows the swinging limb to return to neutral in the frontal plane before coming into contact with the ground, minimising stresses in the frontal plane. This is supported by the reduction in hip abductor moment seen at this point. The increase in knee flexion seen at tibia vertical shortens the swing limb, reducing the strength requirements of the hip abductors.

Alternatively these changes may be efforts to minimize joint loading, improve dynamic stability, or as a result of muscle imbalance (Piazza & Delph 1996; England & Granata 2007). Balance, stability and muscle strength were not assessed directly by the current study. It is therefore difficult to discuss their role in the altered parameters found. An indication as to the presence of anti gravity muscle weakness in adults who are overweight may be drawn from the results of this study. An increase in knee flexion during swing phase of gait has been shown to be associated with inactivity of the rectus femoris (Piazza & Delph 1996). In addition, an increase in hip flexion at initial contact may be as a result of hip extensor weakness, as peak hip extension occurs at terminal stance for normal gait (Whittle 2002). The hypothesis of reduced anti gravity muscle strength is also supported by the decrease in maximum total hip power absorption seen for adults who were overweight. This may indicate reduced muscle mass in this cohort. As discussed previously, eccentric activity requires greater strength from the muscle groups involved (Lindstedt et al 2002). In addition, a significant reduction in vertical GRF corrected for body mass seen for adults who were overweight would support the hypothesis of minimising joint loads by altering gait.

Absolute GRF was found to be increased for adults who were overweight for all three planes. Based on Newtonian equations this result was to be expected. Nonetheless, the increased forces passing across the articular surfaces of the weight bearing joints should not be dismissed due to their clinical relevance. The articular surface size of the femoral head has been shown not to be altered by an increase in body mass (Ruff et al 1991). In addition, joint cartilage at the knee has been shown not to adapt in response to increased mechanical stimuli (Eckstein et al 2002). Despite the increase in absolute three dimensional GRF, with the exception of an increase in hip flexion at initial contact, no changes were seen during the stance phase to accommodate for these increased absolute loads. Therefore, for adults who are overweight the articular surfaces of the hip/cartilage at the knee are subjected to greater absolute forces, and as a result may be prone to pathology.

The changes in gait for adults who were overweight were seen at the start and end of the stance phase and during the swing phase. The magnitude of difference was largest for hip frontal plane kinematics where bias represented 36% and 106% of the overall mean for toe off and feet adjacent respectively. In the frontal plane hip abductor moments corrected for body mass were not seen to be significantly increased for adults who were overweight. However, a significant increase in hip abduction was noted during the swing phase. In the frontal plane, for the increase in knee varus at feet adjacent bias represented 31% of the overall mean. For the knee in the sagittal plane, at tibia vertical the increase in knee flexion represented 6% of the overall mean value.

From the results of Study 3 (a) assessing the influence of body mass on gait parameters in adults, adults with increased body mass exhibited few differences as compared to their healthy weight peers. Changes in the frontal plane demonstrated the greatest between group biases and could be due to body fat distribution, with a significant difference in thigh girth noted. Alternatively, the changes seen may be the result of muscle weakness, instability or to reduce joint loads.

8.3.2 Children

Similar to adults who were overweight, gait of children with increased body mass was compared to gait of children of a healthy weight (Chapter 7, page 131). Participants were matched by age, gender and height. In addition auditory cueing was used to match children by cadence, and in turn velocity. However, significant between group differences were noted for velocity for both studies. While auditory cueing had a minimal effect on the gait characteristics of healthy children, the imposition of a cadence different to a self-selected cadence resulted in an altered presentation of gait. Therefore the data from healthy children ambulating in time to a cue was excluded, and the results discussed are those from both overweight and healthy weight children ambulating at their self-selected velocity.

For studies assessing the influence of body mass on gait parameters in children two previously noted a significant reduction in self selected velocity for children who were overweight (Hills & Parker 1991a, 1991b). Interestingly several studies noted no change in self selected velocity for an increase in body mass for children (Hills & Parker 1991c, Nantel et al 2006; McMillan et al 2009). More surprisingly, several studies failed to record velocity, despite early researches suggestion that a difference in velocity presents for children who are overweight (Hills & parker 1991a, 1991b; McGraw et al 2000; Gushue et al 2005; Morrisson et al 2008; McMillan et al 2010). In the research by Schultz et al (2009; 2010), velocity was controlled in both groups with the use of a metronome. It has been noted that the imposition of a velocity on children results in an altered presentation of gait (Chapter 5, page 88).

As in the present study, the previous research by Hills & Parker (1991a, 1991b) noted a reduction in velocity for their overweight group. For the present study, stride length was reduced, and an increase in percentage stance, single support and double support duration were noted for children with increased body mass ($P<0.05$). Hills & Parker (1991b) noted an increase in stride length despite the reduced velocity of their overweight group. This may be explained by the height differences between groups, as the overweight group was on average 12.8cm taller (Hills & parker 1991b).

From the results presented in Chapter 7, page 131, it appears that the presentation of gait for children who are overweight varies considerably from that of healthy children. However, it is know that velocity alters gait in a variety of linear and non-linear ways (Schwartz et al 2008). As the imposition of a cadence which differs from a self selected cadence altered the presentation of gait in children, it was not possible to assess their gait independent of velocity. This makes it difficult to determine which parameters are as a result of increased BMI and which are due to the reduction in velocity seen. Schwartz et al (2008) supplied supplementary material with their paper discussing the influence of velocity on gait parameters in children. Using this data set the author has attempted to determine which results from Study 4 (a) may be as a result of a reduction in velocity (Schwartz et al 2008).

For the study by Schwartz et al (2008) in order to correct for inter-individual differences (height and weight) gait parameters were expressed in terms of dimensionless speed. Dimensionless speed was calculated as described by Hof (1996) as:

$$V = \frac{\text{velocity}}{\sqrt{g l_0}}$$

where g = gravity (9.8 ms^{-1}) and l_0 = leg length.

Dimensionless speed was calculated for the two groups of the current study. The overweight group's dimensionless speed was 0.403 whereas the healthy weight group's dimensionless speed was 0.480 for the present study. Both of these speeds were encompassed by the 'free' speed 0.429 ± 0.026 (mean ± 2 standard deviations) from the study by Schwartz et al (2008). The dimensionless speed of the healthy weight group for the current study was closer to the 'fast' dimensionless speed (0.56 ± 0.062) of the study by Schwartz et al (2008), than the dimensionless speed of the overweight group for the current study was to the 'slow' dimensionless speed (0.29 ± 0.054) of the study by Schwartz et al (2008). Therefore, the change in gait parameters between 'free' and 'fast' dimensionless speed of the study by Schwartz et al (2008) were compared to the findings for children for the current study.

The calculated percentage time points for the seven events of the gait cycle for each group of the present study were identified on the Schwartz et al (2008) dataset. Percentages identified as the events of the cycle for the healthy weight group at the 'fast' velocity were compared to the percentages indentified as the events of the cycle for the overweight group at the 'free' velocity. Maximum and minimum GRF and joint powers were also calculated for 'free' and 'fast' velocities. Values were deemed to be increased, decreased or similar (where a difference of $\leq$ 0.1 was noted) (Tables 8.1, 8.3, 8.5). These findings were then compared to the significant findings for children who were overweight from the current study (Tables 8.2, 8.4, 8.5).

Range	Initial contact	Opposite toe off	Heel rise	Opposite initial contact	Toe off	Feet adjacent	Tibia vertical
Pelvic tilt	↓	↓	↓	↓	↓	↓	↓
Pelvic obliquity	↓	↓	↓	↓	↓	↓	↓
Pelvic rotation	↓	↑	↓	↓	↓	↓	↓
Hip flexion	↓	↑	↓	↓	↑	↔	↓
Hip abduction	↓	↑	↑	↑	↓	↓	↓
Hip external rotation	↓	↓	↓	↓	↓	↑	↑
Knee flexion	↓	↑	↑	↓	↓	↓	↓
Knee varus	↓	↑	↓	↓	↓	↑	↓
Tibial internal rotation	↓	↓	↓	↓	↓	↑	↓
Ankle dorsiflexion	↓	↓	↑	↑	↓	↑	↓
Ankle external alignment	↑	↓	↓	↑	↑	↑	↓
Ankle supination	Information not supplied by Schwartz et al (2008)						

Table 8.1: Trend in kinematics anticipated for a reduction in velocity in children (Schwartz et al 2008).

↑ = increases for a reduction in velocity

↓ = reduces for a reduction in velocity

↔ = remains unchanged for a reduction in velocity

Range	Initial contact	Opposite toe off	Heel rise	Opposite initial contact	Toe off	Feet adjacent	Tibia vertical
Pelvic tilt	↔	↔	↔	↔	↔	↔	↔
Pelvic obliquity	↔	↔	↔	↔	↔	↔	↓
Pelvic rotation	↔	↔	↔	↔	↔	↔	↓
Hip flexion	↔	↔	↔	↑	↑	↔	↔
Hip abduction	↑	↑	↑	↔	↑	↑	↑
Hip external rotation	↔	↑	↑	↑	↑	↔	↑
Knee flexion	↔	↔	↑	↔	↑	↔	↑
Knee varus	↔	↔	↔	↔	↔	↑	↔
Tibial internal rotation	↑	↔	↑	↑	↑	↑	↑
Ankle dorsiflexion	↔	↔	↔	↑	↔	↓	↔
Ankle external alignment	↑	↔	↔	↑	↔	↓	↑
Ankle supination	↑	↔	↑	↑	↔	↔	↑

Table 8.2: Significant kinematic changes seen for children who were overweight.
↑ = increases for an increase in body mass
↓ = reduces for an increase in body mass
↔ = remains unchanged for an increase in body mass

Moment	Initial contact	Opposite toe off	Heel rise	Opposite initial contact	Toe off	Feet adjacent	Tibia vertical
Hip extensor	↓	↑	↓	↓	↓	↓	↓
Hip abductor	↓	↑	↑	↑	↓	↑	↓
Hip external rotator	Information not supplied by the Schwartz et al (2008) paper						
Knee extensor	↓	↓	↓	↑	↓	↓	↓
Knee valgus	↔	↑	↑	↔	↓	↔	↓
Tibial external rotator	Information not supplied by the Schwartz et al (2008) paper						
Ankle plantarflexor	↔	↑	↓	↑	↑	↔	↔
Ankle internal alignment	Information not supplied by the Schwartz et al (2008) paper						
Ankle pronator	Information not supplied by the Schwartz et al (2008) paper						

Table 8.3: Trend in joint moments anticipated for a reduction in velocity in children (Schwartz et al 2008).

↑ = increases for a reduction in velocity

↓ = reduces for a reduction in velocity

↔ = remains unchanged for a reduction in velocity

Moment	Initial contact	Opposite toe off	Heel rise	Opposite initial contact	Toe off	Feet adjacent	Tibia vertical
Hip extensor	↔	↔	↑	↑	↔	↔	↔
Hip abductor	↑	↔	↔	↔	↑	↑	↑
Hip external rotator	↑	↑	↑	↔	↔	↑	↑
Knee extensor	↔	↓	↑	↔	↔	↓	↔
Knee valgus	↔	↑	↑	↑	↑	↑	↔
Tibial external rotator	↔	↑	↑	↑	↑	↑	↔
Ankle plantarflexor	↔	↔	↓	↓	↔	↔	↔
Ankle internal alignment	↔	↔	↑	↑	↑	↑	↓
Ankle pronator	↔	↔	↑	↑	↔	↑	↔

Table 8.4: Significant differences in joint moments seen for children who were overweight.

↑ = increases for an increase in body mass

↓ = reduces for an increase in body mass

↔ = remains unchanged for an increase in body mass

	Column 2	Column 3
Parameter	Anticipated trend for a decrease in velocity	Significant changes seen for children who were overweight
Max GRF X	↓	↓
Max GRF Y	Not provided by Schwartz et al (2008)	↓
Max GRF Z	↓	↓
Max Hip Generative	↓	↔
Max Hip Absorptive	↓	↓
Max Knee Generative	↓	↔
Max Knee Absorptive	↓	↔
Max Ankle Generative	↓	↑
Max Ankle Absorptive	↑	↔

Table 8.5: Trend in GRF and joint power anticipated for a reduction in velocity, and significant changes in GRF and joint power seen for children who were overweight (Schwartz et al 2008).

Column 2: ↑ = increases for a reduction in velocity
↓ = reduces for a reduction in velocity

Column 3: ↑ = increases for an increase in body mass
↓ = reduces for an increase in body mass
↔ = remains unchanged for an increase in body mass

Tables 8.1-8.5 suggest that several of the between group differences noted for children may be explained by the difference in velocity seen. For kinematic parameters the significant differences noted at the pelvis and for hip flexion at toe off, hip abduction at opposite toe off and heel rise, and hip external rotation at tibia vertical may be due to a reduction in velocity (Tables 8.1-8.2). In addition, knee flexion at heel rise, varus and internal rotation at feet adjacent, ankle dorsiflexion at opposite initial contact, and external alignment at initial contact and opposite initial contact may also be as a result of a reduction in velocity (Tables 8.1-8.2). For kinetic parameters the increased hip abductor moment seen at feet adjacent, and knee valgus moment at opposite toe off and heel rise, and reduced knee extensor moment at opposite toe off and feet adjacent, and ankle plantarflexor moment at heel rise may be explained by the reduction in velocity (Tables 8.3-8.4). Similarly the reduction noted in peak maximum vertical and propulsive GRF, and reduction in maximum hip absorptive power may be explained by the reduction in self selected velocity (Table 8.5).

The remaining significant differences were in fact opposite to the trends anticipated for a reduction in velocity (Tables 8.1-8.5). This suggests that the difference may be of an even greater magnitude than reported in the present study. It should also be noted that while no significant differences were noted between groups for several of the parameters studied, for a reduction in velocity altered range is anticipated (Tables 8.1-8.5). For example, with a reduction in velocity a decrease in anterior-posterior pelvic tilt is anticipated, however for the current thesis no significant differences were noted between groups. This suggests that for children who are overweight there may in fact be an increase in pelvic tilt.

Possible reasons for the altered parameters, which from the previous analysis are not explained by velocity, will be discussed here. Schwartz et al (2008) did not provide data on the influence of velocity on frontal ankle kinematics and joint moments, or transverse plane joint moments. It was therefore difficult to examine the relationship between velocity and the significant differences found. The results should be viewed with a degree of caution as alterations in the transverse plane have previously been reported for a change in velocity (Van der Linden et al 2001).

As for adults the majority of alterations in the presentation of gait for children who were overweight occurred during the latter half of stance phase, and for the swing phase. A total of 75% of significant differences in kinematics and 70% of significant changes in joint moments were seen from opposite initial contact to tibia vertical. As for adults these changes may be due to instability, muscle weakness or efforts to minimize joint loading (Piazza & Delph 1996; England & Granata 2007). Once again balance, stability and muscle strength were not assessed directly.

Alterations in sagittal plane kinematics and joint moments appear present in children who are overweight. The gait pattern adopted by children who are overweight enables a lowering of the centre of gravity towards the floor improving stability at terminal stance. In addition these motions draw the GRF vector further from the joint centre minimising loads incurred on the joint, with particular attention to the swing phase where one limb is required to take the total load (Watkins 2010). As joint moments are a product of kinematics it is not surprising that changes have been reported. Significant increases in joint moments acting at the hip and knee were found. This suggests that soft tissue at the hip and knee is required to cope with the increased loads incurred.

Alterations in the frontal plane are evident in children who are overweight. With excessive hip abduction, possibly due to an increase in thigh girth, the GRF vector is shifted medially resulting in greater stress on the medial compartment at the knee and ankle. To account for the increased load medially, children who were overweight exhibited an increased knee valgus moment and increased ankle pronator moment. While the increased moments at the knee appear to have controlled the degree of varus occurring at the joint, at the ankle the pronator moments did not prevent excessive supination. This increase in supination creates a rigid structure of the foot, one which is poor at attenuating joint loads (Johanson et al 2008). Therefore, as for the sagittal plane, greater stresses are placed upon the knee, and more proximally the hip.

As for the sagittal and frontal planes, alterations in the presentation of gait in the transverse plane presented. With increased external rotation at the hip and foot, the tibia offers counterbalance with internal rotation. The overall externally rotated position of the limbs results in a widened base of support and improved stability (Watkins 2010). The changes adopted in this plane support the hypothesis of reduced dynamic stability in children who are overweight.

The increase in total ankle power generation noted for children who were overweight indicated higher concentric activity at the ankle. It was hypothesised previously that greater energy expenditure was required for children who were overweight to maintain forward progression. As no changes were noted in sagittal ankle joint moments, it may be suggested that the concentric activity is as a result of frontal and transverse plane muscle activity. Frontal muscle activity at the ankle is predominantly associated with stability and this finding supports the hypothesis of dynamic instability for overweight populations (Brunt et al 1992).

8.4 CRITICAL ANALYSIS OF THIS WORK

Monaghan et al (2007) provide an account of the sources of variation in gait analysis experimental procedures, notably subject, rater and system.

Areas of subject variation were identified as natural variation in gait, natural variation in velocity, variation in velocity between trials, footwear, a short runway, and real differences due to pathological change (Monaghan et al 2007). In the present studies these possible sources of variation were addressed using adults and children who matched the inclusion/exclusion criteria, and who ambulated barefoot over the 10m walkway. Participants were also instructed to walk continuously up and down the walkway in an effort to encourage a true self-selected velocity. In addition, the first ten lengths were considered a familiarisation phase and no data were collected during this period. In order to address natural variation in gait, for studies 1 and 2 intra-individual variation was assessed and considered in the analysis of cued data results. For studies 3 (b) and 4 (b) participants were matched by velocity using external auditory cueing. While auditory cueing did not appear to alter gait parameters in adults, for children the imposition of a cadence different from a self selected cadence resulted in an altered presentation of gait. It was not possible to constrain velocity in healthy children. The analysis therefore, did not control for the difference in velocity between the two groups. It should be noted that the author has highlighted several differences which may be attributable to the reduced velocity, using previous research which investigated the influence of velocity on gait in children (Schwartz et al 2008).

Possible sources of rater variation include marker placement, identification of anatomical landmarks, wand alignment, anthropometric measurement, data processing, choice of statistical analysis, and different rater measurements (Monaghan et al 2007). To minimise the influence of rater variation one rater completed all test sessions and data processing. In addition, for studies 1 and 2 the markers were not removed between the two test conditions. For studies 3 (a), 3 (b), 4 (a) and 4 (b) the experimental group was tested prior to the control group. The experimental group was tested over a period of nine months for children, and four months for adults. In addition, the control group was tested over a period of 18 months for children, and 13 months for adults. The author acknowledges that with equipment familiarisation over time, rater variation may have been reduced. However, as participants were matched by age, gender height and cadence equipment familiarisation was unavoidable.

System variation includes calibration, precision of computer algorithms, uncertainty in construction of an embedded coordinate system, downstream errors associated with Euler angle calculations, number of cameras, alignment procedure, relative skin/marker movement error, and force plate drift/noise (Monaghan et al 2007). The CODA Motion sensors (Charnwood Dynamics Ltd., Leicestershire, UK) were aligned on each test day and force plates reset. It is anticipated for an increase in body mass skin/marker movement error may increase. However, from a recent study, it was noted that soft tissue artefact was found to have the greatest effect on transverse plane motion at the thigh (Gao & Zheng 2008). For adults few changes were noted in the transverse plane. For children, the differences noted were consistently in the same direction i.e. externally rotated at the hip, internally rotated at the tibia and external foot alignment. It therefore is the authors belief that soft tissue artefact had minimal impact on the results reported by this thesis. The accuracy of the CODA Motion (Charnwood Dynamics Ltd., Leicestershire, UK) has been discussed previously in Chapter 1, page 34 and referred to throughout this thesis. As frontal and transverse plane kinematics and kinetics were of particular interest, it is regrettable that the results are limited by the systems error of approximately 1.5° (Richards 1999).

When running statistical tests with alpha set to ≤ 0.05 there is a one in twenty chance of obtaining Type 1 error (a false statistically significance result). When running multiple significance tests this probability of Type 1 error increases. For the current thesis 165 tests were run for each of the studies 3 (a), 3 (b), and 4 (a). As such, the likelihood of obtaining a Type 1 error increased. Often a Bonferroni correction to the alpha level is adopted to account for this error (Tukey 1977). This correction involves dividing the alpha level by the number of tests conducted i.e. 0.05/165 – 0.00303' (Tukey 1977). While this reduces the possibility of Type 1 error, as a result Type 2 error is increased (Perneger 1998). In other words, the probability of omitting a statistically significant result is increased. As Bonferroni corrects for the entire study's error rate it is relevant when the overall null hypothesis is the main interest i.e. is overweight gait different to healthy weight gait (Morgan 2007)? For the current thesis it was of interest to detect changes in the individual tests, where the Bonferroni correction is not as relevant (Morgan 2007). It was therefore not appropriate to adopt a Bonferroni correction. The author acknowledges that this increased the probability of Type 1 error in the results presented.

For the current research there was no single dominant variable of interest. As such, it was difficult to calculate statistical power. A sample of 50 for each of the body mass studies was selected based on previous literature in the area (Chapter 2, pages 32-33). The recruitment of subjects represented a significant barrier to sample size, particularly for children where the recruitment period lasted a total of 27 months.

Due to a limitation in the number of available markers, trunk and foot parameters were not assessed. Few significant differences were noted at the pelvis for studies 3 (a), 3 (b) and 4 (a). It may be hypothesised that alterations seen distally were not influenced by altered trunk mechanics. Similarly, few changes at the ankle in studies 3 (a), 3 (b) and 4 (a) (that were not explained by the reduction in velocity seen for children) were noted. This would suggest that alterations seen more proximally were not influenced by altered foot mechanics.

The overweight and obese groups of this research were described in terms of BMI. Body composition (lean and fat mass) may have provided useful information. For example, the relationship between altered power generation and absorption may be associated with a reduction in muscle mass. At the Trinity Centre for Health Sciences body composition analysis is completed with a Tanita Body Composition Analyser MC-180MA. The Tanita Body Composition Analyser MC-180MA has not been validated in children. In addition, for bioelectrical impedance analysis participants are required to fast from food. It was felt that this would represent a significant barrier to recruitment. As overweight participants were recruited from weight management and hypertension clinics, it was felt that an increase in BMI would represent an increase in fat mass as compared to muscle mass. However, as body composition was not assessed directly this assumption not be confirmed.

This study was confined to the assessment of the influence of body mass on gait parameters in otherwise healthy adults and children. In addition, the influence of auditory cueing on gait parameters in healthy adults and children was addressed. Although these were the objectives of this thesis it must be noted that the findings of this work provide information relating purely to these objectives. The information may not be applied to individuals with additional musculoskeletal complaints.

8.5 FUTURE RESEARCH

It is often assumed that abnormal joint biomechanics result in musculoskeletal pathology. However, the literature in this area has not established cause and effect, but is more often based on observations and opinion (Wearing et al 2006b). The relationship between increased body mass and tibiofemoral osteoarthritis has been well assessed (Felson et al 1988; Spector et al 1994; Chang et al 2005; Reijman et al 2007; Marks 2007; Grotle et al 2008). In addition, altered biomechanics have been implicated in the onset and progression of tibio-femoral osteoarthritis (Elahi et al 2000; Sharma et al 2001; Cerejo et al 2002; Felson et al 2004; Astephen et al 2005; Cahue et al 2004; Felson et al 2005). Further cross sectional investigation of the presentation of gait for overweight adults with and without tibiofemoral osteoarthritis may provide more information of the relationship between body mass, altered biomechanics and tibiofemoral osteoarthritis.

For the current thesis, changes in the presentation of gait were noted for both adults and children who were overweight. However, it is not known whether obesity pre-disposes an individual to altered biomechanics, or altered biomechanics contributes towards the development of obesity. In addition the long term implications are unknown. About 10% of skeletal injuries in children are injuries to the epiphyseal plate, articular cartilage and apophysis (Maffulli 1990). Both acute and repetitive loading can injure the plates potentially resulting in the premature closure of the epiphyseal junction and termination of bone growth (Hall 2012). With a higher incidence of slipped capital femoral epiphysis seen for children who are overweight, this cohort may be at risk of altered bone growth.

Motivation to participate in physical activity has been shown to be reduced for those who are overweight (Deforche et al 2006). In addition psychological input and behavioural modification have been shown to improve weight loss for adults who are obese (Hainer et al 2008). In a study assessing six minute walk test performance, walking ability of obese women was hampered by several factors including perceived discomfort and musculoskeletal pain (Hulens et al 2003). From the results of this thesis there appears to be alterations in the gait cycle for individuals who are overweight. These alterations may represent a barrier to exercise in this cohort. Non-weight bearing exercise has been recommended for overweight '*.....patients with severe arthritis and problems with mobility*' (Hainer et al 2008). In an effort to reduce pain as a result of exercise, it may be suggested that non-weight bearing exercise may be recommended for all patients who are overweight. A clinically based randomised controlled trial assessing the influence of a non-weight bearing exercise programme as compared to a weight bearing exercise programme on barriers to exercise (including pain), weight loss, and exercise adherence is recommended. This would enable clinicians to assess the feasibility and effectiveness of implementing a non-weight bearing exercise protocol for those with increased body mass.

This thesis provided an objective and concise overview of the presentation of gait for adults and children with increased body mass. It did not aim to provide objective information as to why alterations may be present. Possible reasons for the changes seen have been discussed, including body fat distribution, altered dynamic stability, reduced muscular strength and redistribution of joint loads. However, these hypotheses were not objectively assessed. Future research should investigate these issues so that exercise prescription may be tailored further to improve gait.

8.6 CONCLUSIONS

This study aimed to investigate the influence of body mass on gait parameters in children and adults by providing a concise overview of temporal, spatial, kinematic and kinetic parameters. As velocity has been seen to influence several components of the gait cycle, the author also assessed the possibility of constraining velocity with the use of an auditory cue.

From this thesis auditory cueing did not alter the gait presentation for adults of a healthy weight. In addition, the imposition of a cadence did not appear to influence gait. For the scientific setting, further research assessing gait parameters in adults should be matched by cadence to a healthy control group with the use of an auditory cue. This was not the case for children, where the imposition of a reduced cadence with auditory cueing resulted in an altered presentation of gait. It is therefore not recommended that auditory cueing be used to match children by velocity in research trials assessing paediatric gait.

Alterations in the presentation of gait were noted for both children and adults who were overweight with adults exhibiting fewer changes than children. However, the adult analysis was independent of velocity whereas the analysis for children was not. From the previous literature, it appears that several of the changes noted for children may be explained by a reduction in velocity. Alterations were seen predominantly in the frontal plane for adults, and frontal and transverse planes for children. For both adults and children who were overweight, the hip and knee joints exhibited the majority of significant differences for an increase in body mass.

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APPENDICES

Appendix 1: Information and consent form, study 1

To Investigate the Effect of Auditory Cueing on Gait Parameters in Healthy and Overweight Adults.

Introduction:

Gait analysis is frequently used to assess altered walking patterns associated with disease. In order to investigate the changes in walking characteristics, healthy subjects are often required for comparison. The speed at which a person chooses to walk can greatly affect their walking pattern. Therefore the differences in walking characteristics between a disease group and a healthy group may be due to differences in self selected walking speed.

To remove the influence of speed on walking characteristics, a loud beat may be used to control the speed at which healthy subjects walk. However, the influence of p-value a loud beat on walking characteristics has not been previously assessed.

Therefore the purpose of this study is to assess the affects of p-value a beat on walking characteristics. As subjects recruited for control groups will vary in body mass index, this study aims to look at the effect of p-value a beat on healthy adult and overweight walking.

Procedures:

Should you wish to take part in the study, you will be required to attend the Gait Laboratory at the Trinity Centre for Health Sciences St. James's Hospital for two, two hour visits. You will be required to wear shorts during the assessment. On arrival measurements of your legs will be made e.g. leg length, knee width, as required by the gait analysis system. You will then be fitted with surface mounted markers with the use of double-sided sellotape and Velcro straps. After the equipment is attached to you, you will be asked to walk along a 10m walkway several times. There will then be a short interval, the equipment will remain attached to you during this interval.

You may bring a book/games console etc for this time period. After the short break you will be asked to walk along the walkway again. You will be asked to step in the rhythm of a beat which will be played through a computer. One week later you will be asked to return to repeat the study again.

Inclusion criteria:

Aged over 18.

Body mass index of 20-25 (healthy weight group).

Body mass index of 25-30 (overweight group).

Informed consent obtained before measurements are taken.

Exclusion criteria:

A history of musculoskeletal disease.

Lower limb musculoskeletal injury in the last 6 months which required medical intervention.

Leg length discrepancy of greater than 2 inches.

Benefits:

An analysis of your walking pattern will not be given after the test, however it is hoped the results of this study will help future medical practice to treat walking problems. Should abnormalities in your walking pattern be noted when your walking results are analysed, you will be notified to enable remedial action to be taken.

Risks:

There are no perceived risks.

Confidentiality:

Your identity will remain confidential. Your name will not be published and will not be disclosed to anyone outside the hospital.

Voluntary Participation:

You have volunteered to participate in this study. You may quit at any time. If you decide not to participate, or if you quit, you will not be penalised and will not give up any benefits which you had before entering the study.

Stopping the study:

You understand that the research team may stop your participation in the study at any time without your consent.

Permission:

This project has Research Ethics Committee approval from Trinity College Dublin and AMNCH/SJH Ethics Committee.

Further information:

You can get more information or answers to your questions about the study, your child's participation in the study and their rights, from Katie Sheehan at 01896 3613/0863400010 or via e-mail to sheehakj@tcd.ie. If the study team learns of important new information that might affect your desire to remain in the study, you will be informed at once.

SJH / AMNCH RESEARCH ETHICS COMMITTEE.

CONSENT FORM

Title of research study: To Investigate the Effect of Auditory Cueing on Gait Parameters in Healthy and Overweight Adults.

This study and this consent form have been explained to me. The researcher has answered all my questions to my satisfaction. I believe I understand what will happen if I agree to be part of this study.

I have read, or had read to me, this consent form. I have had the opportunity to ask questions and all my questions have been answered to my satisfaction. I freely and voluntarily agree to be part of this research study, though without prejudice to my legal and ethical rights. I have received a copy of this agreement.

Name of sponsor:

PARTICIPANT'S NAME:________________________________

PARTICIPANT'S SIGNATURE: __________________________

Date:____________

Statement of investigator's responsibility: I have explained the nature, purpose, procedures, benefits, risks of, or alternatives to, this research study. I have offered to answer any questions and fully answered such questions. I believe that the participant understands my explanation and has freely given informed consent.

Researcher's signature:________________________________

Date:_____________

Appendix 2: Information and consent form, studies 2, 4(a) and 4(b)

The effects of weight on gait parameters in children

Introduction:

Recently it has been noted that children who are overweight sometimes complain of muscle and joint pain.

This study aims to see if weight has an effect on the way in which the legs move during walking in overweight children. To explore whether there is a different pattern of walking compared to that in average weight children. If so, this may be a reason for the joint and muscle pain that children troubled with weight may have.

Very few previous studies have examined the effects of weight on walking characteristics.

People who are fitter and more active often have a better walking pattern than people who are less fit and active. This study also aims to see if fitness and activity levels in children have an effect on the way in which they walk.

Finally, this study will see if the results found are the same for boys and girls.

Procedures:

Should you wish to participate with us in this study to help answer these important questions, you and your child would be required to attend The Trinity Centre for Health Sciences, St. James's Hospital once. At this visit, three measurements would be made- physical activity levels, physical fitness level and, walking characteristics.

DAY OF TESTING

Your child should wear lightweight clothing and good footwear for the visit so they are comfortable while exercising.

Two testers will be present at the session.

You are welcome to stay for the testing which will last around 1 hour.

Your child will first complete a short questionnaire on joint pain. Height, weight and body composition will then be measured.

Your child's fitness will be measured by a walking test. This test will be carried out in a quiet hallway. Your child will be asked to walk up and down the corridor as many times as they can in 6 minutes.

Your child will then undergo a second walking measurement test.

For the walking measurements to be made your child will be asked to bring tight fitting shorts and vest, or, a swimsuit to wear.

A clinical assessment of your child will be conducted. During this assessment measurements of your child's legs will be taken, for example- knee width, leg length, and ankle movement.

Light weight markers will be attached to your child's legs with double sided tape and Velcro strapping. These markers will allow a computer to measure exactly the way in which your child's legs move when they are walking.

Your child will then be asked to walk across a flat walkway several times, first at their chosen pace and then to a fast pace and slow pace (control group only).

After the walking tests, your child will be given a small portable device similar to a pedometer called an RT3 accelerometer. You will be provided with a stamped addressed envelope.

This device will measure the amount of time your child spends inactive and, in light, moderate and vigorous activity.

S/he will be required to wear the lightweight portable device attached to the top of his/her trousers with a clip at the back of the device during the day for the next 7 days. It may be removed while they are sleeping at night.

After the 7 days you will be required to send the device back to the Trinity Centre for Health Sciences in the stamped addressed envelope provided.

Benefits:

You and your child will be given information about your child's activity and fitness levels.

An analysis of your child's walking pattern will not be given after the test, however it is hoped the results of this study will help future medical practice to treat walking problems in overweight children. Should abnormalities in your child's walking pattern be noted when his/her walking results are analysed, you will be notified to enable remedial action to be taken.

Risks:
There are no perceived risks for your child.

Exclusion from participation:
Your child cannot participate in this study if they have any of the following-
Any history of cardiac or respiratory disease.
Poorly controlled asthma.
Currently taking prescribed medication.
Under the age of 7 or older than 17 years.
Pregnancy.

Confidentiality:
Your child's identity will remain confidential. Your child's name will not be published and will not be disclosed to anyone outside the study group. Data which will identify your child will be kept after the study is completed, however it will not be used in future unrelated studies without your specific permission being obtained.

Voluntary participation:
If you decide you would like your child to participate in this study, please contact Katie Sheehan at 018963613/0863400010, or via e-mail to sheehakj@tcd.ie. Alternatively you may fill out the enclosed expression of interest form, place it in the envelope provided and send it by post, the lead investigator will then contact you to arrange an appointment.
If you decide you would like your child to take part in the study, you may withdraw your child at any time.

Stopping the study:

You understand the investigators may withdraw your child's participation in the study at any time without your consent.

Permission:

This project has Research Ethics Committee approval from Trinity College Dublin and AMNCH/SJH Ethics Committee.

Further information:

You can get more information or answers to your questions about the study, your child's participation in the study and their rights, from Katie Sheehan at 01896 3613/0863400010 or via e-mail to sheehakj@tcd.ie. If the study team learns of important new information that might affect your desire to remain in the study, you will be informed at once.

Walking study- Information for your child

What does the study want to see?

My name is Katie Sheehan and I am a physiotherapist. I would like to see the way different young people walk.

I would also like to see how fit different young people are and how much walking they do in a week.

What would you do?

If you think you would like to be in my study, you will visit me at St. James's Hospital. When you get there I will tell you everything that we will do. You can then ask me questions before we start. If you still want to be in the study, I will ask you to write your name on a page which says you would like to be in the study.

On this day I will ask you some questions about your legs and then I will check how tall you are and how much you weigh. You will then have to do a walking test. The walking test will be done in a quiet hallway; you will have to walk up and down the hallway as many times as you can in six minutes.

We will then do the second walking test. For the second walking test you will need to wear swimming togs or a vest and shorts. Before this test, I will measure some parts of your legs. I will then stick some tiny pads onto your legs p-value sticky tape and straps. You will then do the second walking test; you will need to walk along an indoor path a few times. *You will then be asked to walk in time to two different loud beats (control group only).*

The next thing I would like to see is how much walking you do in a week. After your walking test I will give you a small measuring box the same size as an i-pod, to take home. You will need to clip this box to your trousers during the day time every day for a week. The box will count the number of steps you take during the week.

After 1 week you will need to send this box to me in the post.

What will you find out from all these tests?

After all the tests you will find out how fit you are and how much walking you do in a week.

Is this study safe for you?

This study is very safe and will not make you feel sick.

Will people know I am doing the tests?

The only people who will know you are doing the study are your parents and the people doing the study.

What do I do if I would like to do the study?

If you would like to do the study, you should tell your parents and they can tell me.

What do I do if I change my mind?

If you change your mind and do not want to be in the study, you can tell your parents and they can tell me.

What do you do if you have any questions?

If you have any questions, you can tell your parents and they can telephone me to ask me. You can also ask me when you see me at the hospital *(overweight group).*

SJH / AMNCH RESEARCH ETHICS COMMITTEE CONSENT FORM

Title of research study: To Investigate Gait Parameters in Normal Weight and Overweight Children

.

This study and this consent form have been explained to me. The researcher has answered all my questions to my satisfaction. I believe I understand what will happen if I agree to be part of this study.

I have read, or had read to me, this consent form. I have had the opportunity to ask questions and all my questions have been answered to my satisfaction. I freely and voluntarily agree to be part of this research study, though without prejudice to my legal and ethical rights. I have received a copy of this agreement.

Name of sponsor:

PARTICIPANT'S NAME:_______________________________

PARTICIPANT'S SIGNATURE: _________________________

Date:____________

NAME OF CONSENTER, PARENT or GUARDIAN:___________________________

SIGNATURE: ____

Date:_______________

Statement of investigator's responsibility: I have explained the nature, purpose, procedures, benefits, risks of, or alternatives to, this research study. I have offered to answer any questions and fully answered such questions. I believe that the participant understands my explanation and has freely given informed consent.

Researcher's signature:_______________________________

Date:_____________

Appendix 3: Letter sent to school principals

Katie Sheehan MISCP,

Lead Investigator,

Physiotherapy Research Room,

Trinity Centre for Health Sciences,

St. James's Hospital,

Dublin 8.

Dear Principal,

My name is Katie Sheehan, and I am physiotherapist from Trinity College Dublin. I am currently undertaking a research project for my PhD, and am investigating the effects of body weight on the walking characteristics in young people. It is hoped that this piece of research will help future treatment of walking difficulties in overweight young people.

To complete my research, I will need a number of young people to volunteer their time. I am hoping that your school will be interested in being a part of this useful research. Should you wish your school to participate, I will supply you with information leaflets to distribute to the guardians of the pupils. This leaflet will contain my contact details should the guardians have any questions or should they wish their daughter/son to participate.

With this letter I have included a copy of the information leaflet which contains more detail on what is involved. Please do not hesitate to contact me should you require any further information or answers to your questions about the study, student participation in the study and their rights. You may contact me by telephone at 8963613, 0863400010 or via e-mail at sheehakj@tcd.ie.

Look forward to hearing from you,

Yours truly,

____

Katie Sheehan MISCP

Appendix 4: Letter to parents

Katie Sheehan MISCP

Physiotherapy Department

Trinity Centre for Health Sciences

St. James's Hospital

Dublin 8

Dear Parent/Guardian,

My name is Katie Sheehan and I am physiotherapist from Trinity College, Dublin. I am currently carrying out a research study investigating the effects of body weight on walking characteristics in young people.

It is hoped that this piece of research will help future treatment of walking difficulties in overweight young people.

To complete my research, I will need a number of young people to volunteer their time. I am hoping that your daughter/son will be interested in being a part of this useful research. Please read the enclosed information leaflet, which contains more detail on what is involved.

I would be happy to answer any of your questions. To contact me or to arrange participation please call me on *01 8963613/0863400010,* e-mail me at *sheehakj@tcd.ie* or, complete the enclosed expression of interest form and I will contact you.

Many Thanks,

____

Katie Sheehan

Appendix 5: Information and consent form, studies 3(a) and 3(b)

Title of study: To Investigate Gait Parameters in Normal Weight and Overweight Adults.

Introduction:

It has been noted that adults who are overweight often complain of muscle and joint pain. This study aims to see if weight has an effect on the way in which the legs move during walking in adults who are overweight. To explore whether there is a different pattern of walking compared to that in a healthy adult. If so, this may be a reason for the joint and muscle pain that adults troubled with weight may have. Very few previous studies have examined the effects of weight on walking characteristics.

Procedures:

Should you wish to take part in the study, you will be required to attend the Gait Laboratory at the Trinity Centre for Health Sciences St. James's Hospital for a one hour visit. You will be required to wear shorts during the assessment. On arrival measurements of your legs will be made e.g. leg length, knee width, as required by the gait analysis system. You will then be fitted with surface mounted markers with the use of double-sided sellotape and Velcro straps. After the equipment is attached to you, you will be asked to walk along a 10m walkway several times.

After walking at your natural pace, you will be required to walk in time to a beat played by a loud metronome *(healthy weight group information leaflet only).*

Inclusion criteria:

Aged over 18.

Body mass index of greater than 25 (overweight group information leaflet).

Body mass index of less than 25 (healthy weight group information leaflet).

Informed consent obtained before measurements are taken.

Exclusion criteria:

A history of musculoskeletal disease.

Lower limb musculoskeletal injury in the last 6 months which required medical intervention.

Leg length discrepancy of greater than 2 inches.

Benefits: An analysis of your walking pattern will not be given after the test, however it is hoped the results of this study will help future medical practice to treat walking problems. Should abnormalities in your walking pattern be noted when your walking results are analysed, you will be notified to enable remedial action to be taken.

Risks:

There are no perceived risks.

Confidentiality:

Your identity will remain confidential. Your name will not be published and will not be disclosed to anyone outside the hospital.

Voluntary Participation:

You have volunteered to participate in this study. You may quit at any time. If you decide not to participate, or if you quit, you will not be penalised and will not give up any benefits which you had before entering the study.

Stopping the study:

You understand that the research team may stop your participation in the study at any time without your consent.

Permission:

This project has Research Ethics Committee approval from Trinity College Dublin and AMNCH/SJH Ethics Committee.

Further information:

You can get more information or answers to your questions about the study, your participation in the study and their rights, from Katie Sheehan at 01896 3613/0863400010 or via e-mail to sheehakj@tcd.ie. If the study team learns of important new information that might affect your desire to remain in the study, you will be informed at once.

SJH / AMNCH RESEARCH ETHICS COMMITTEE.

CONSENT FORM

Title of research study: To Investigate Gait Parameters in Normal Weight and Overweight Adults.

This study and this consent form have been explained to me. The researcher has answered all my questions to my satisfaction. I believe I understand what will happen if I agree to be part of this study.

I have read, or had read to me, this consent form. I have had the opportunity to ask questions and all my questions have been answered to my satisfaction. I freely and voluntarily agree to be part of this research study, though without prejudice to my legal and ethical rights. I have received a copy of this agreement.

Name of sponsor:

PARTICIPANT'S NAME:______________________________

PARTICIPANT'S SIGNATURE: __________________________

Date:____________

Statement of investigator's responsibility: I have explained the nature, purpose, procedures, benefits, risks of, or alternatives to, this research study. I have offered to answer any questions and fully answered such questions. I believe that the participant understands my explanation and has freely given informed consent.

Researcher's signature:______________________________

Date:_____________

Appendix 6: E-mail to staff and students

Subject: Gait Analysis Study

Dear staff and students,

It has been noted that adults who are overweight often complain of muscle and joint pain. This study aims to see if weight has an effect on the way in which the legs move during walking in adults who are troubled with weight. If so, this may be a reason for the joint and muscle pain that adults troubled with weight may have.

Volunteers of a healthy weight are needed for this study. Volunteers will be matched to participants troubled with weight who have already been tested.

Should you wish to take part in the study, you will be required to attend the Gait Laboratory at the Trinity Centre for Health Sciences St. James's Hospital for one 40 minute visits. You will be required to wear shorts during the assessment. On arrival measurements of your legs will be made e.g. leg length, knee width, as required by the gait analysis system. You will then be fitted with surface mounted markers with the use of double-sided sellotape and Velcro straps. After the equipment is attached to you, you will be asked to walk along a 10m walkway barefoot several times. You will be required to walk at your natural speed and to a slow audible beat played through a laptop.

Please find further information in the attached information leaflet. You can get more information or answers to your questions about the study, your participation in the study and their rights, from Katie Sheehan at 01896 3613/0863400010 or via e-mail to sheehakj@tcd.ie.

Look forward to hearing from you,

____

Katie Sheehan

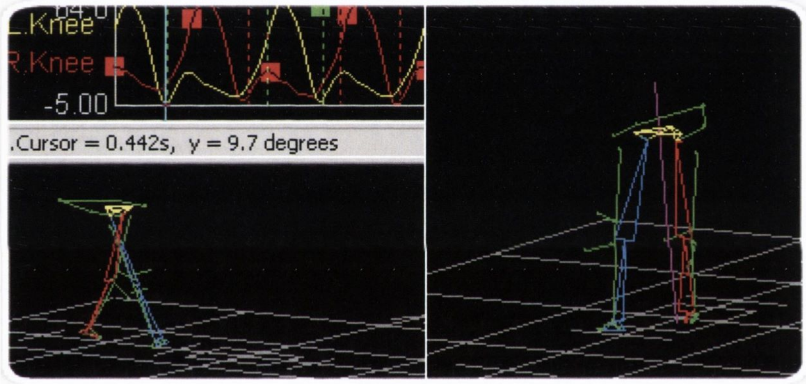
Appendix 7: Recruitment Poster

*****Volunteers Needed *****

To Investigate Gait Parameters in Normal Weight and Overweight Adults.

Department of Physiotherapy

Introduction: It has been noted that adults who are overweight often complain of muscle and joint pain. This study aims to see if weight has an effect on the way in which the legs move during walking in adults who are troubled with weight. If so, this may be a reason for the joint and muscle pain that adults troubled with weight may have.Volunteers of a healthy weight are needed for this study. Volunteers will be matched to participants troubled with weight who have already been tested.



Procedures: Should you wish to take part in the study, you will be required to attend the Gait Laboratory at the Trinity Centre for Health Sciences St. James's Hospital for one 40 minute visits. You will be required to wear shorts during the assessment. On arrival measurements of your legs will be made e.g. leg length, knee width, as required by the gait analysis system. You will then be fitted with surface mounted markers with the use of double-sided sellotape and Velcro straps. After the equipment is attached to you, you will be asked to walk along a 10m walkway barefoot several times. You will be required to walk at your natural speed and to a slow audible beat played through a laptop.

Further information: You can get more information or answers to your questions about the study, your participation in the study and their rights, from Katie Sheehan at 01896 3613/0863400010 or via e-mail to sheehakj@tcd.ie.

Appendix 8: Clinical assessment form

Clinical Assessment

Subject Code:___________________

Gender:________________________

Date:__________________________

Researcher:_____________________

ANTHROPOMETRIC DATA (all in mm except height and weight):

Height		
Weight		
Pelvic depth		
Pelvic width		
Knee joint width	L	R
Ankle joint width	L	R
Thigh girth	L	R
Thigh length	L	R
Shank length	L	R
Foot length	L	R
Leg length	L	R

PASSIVE RANGE OF MOTION DATA (in degrees):

Joint	Movement	Passive ROM
Ankle	Dorsiflexion	
	Plantarflexion	
Knee	Extension	
	Flexion	
Hip	Extension	
	Flexion	
	Abduction	
	Internal rotation	
	External rotation	

SKELETAL OBSERVATIONS:

Genu Valgum	L	R
Genu Varum	L	R
Femoral Torsion	L	R

Appendix 9: Ethical approval Study 1

THIS NOTEPAPER MUST NOT BE USED FOR PRESCRIPTIONS OR INVOICING PURPOSES

SJH/AMNCH Research Ethics Committee Secretariat
Dan Lynch Ph: 4142860 email: Dan.Lynch@amnch.ie
Ursula Ryan Ph: 4142342 email: Ursula.Ryan@amnch.ie
Secretariat Fax 4142371

SJH/AMNCH
Research Ethics Committee
THE ADELAIDE & MEATH HOSPITAL, DUBLIN
INCORPORATING
THE NATIONAL CHILDREN'S HOSPITAL

TALLAGHT, DUBLIN 24, IRELAND
TELEPHONE +353 1 4142000

Ms. Katie Jane Sheehan
Physiotherapy Postgraduate Student
Trinity Centre for Health Sciences
St. James's Hospital
James Street
Dublin 8

July 21st 2010

Please quote this reference in any follow up to this letter: 2010/06/05 Chairman's Action

Re: To Investigate the Effect of Auditory Cueing on Gait Parameters in Healthy and Overweight Adults.

Dear Katie,

Thank you for your recent submission of the above proposal to the SJH/AMNCH Research Ethics Committee.

The Chairman, having reviewed the proposal, has given ethical approval on behalf of the Committee.
The following documents were reviewed:

1. Administrative Application.
2. Confidential Research Protocol, 2006 Edition.
3. Patient Information Sheet & Consent Form.
4. Clinical Assessment Form.

Yours sincerely,

Ms. Ursula Ryan
Secretary
SJH/AMNCH Research Ethics Committee.

Appendix 10: Ethical approval Studies 2, 4(a) and 4(b)

THIS NOTEPAPER MUST NOT BE USED FOR PRESCRIPTIONS OR INVOICING PURPOSES

THE ADELAIDE & MEATH HOSPITAL, DUBLIN

Research Ethics Committee

THE NATIONAL CHILDREN'S HOSPITAL

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Secretariat Fax 4142371

Ms. Katie Sheehan
Postgraduate Physiotherapy Research Room
Trinity Centre for Health Sciences
St. James's Hospital
James Street
Dublin 8

May 21st 2009

Please quote this reference in any follow up to this letter: 2009/05/02 Chairman's Action

Re: The Effects of Weight Gait Parameters in Children.

Dear Katie,

Thank you for your recent submission of the above proposal to the SJH/AMNCH Research Ethics Committee. The Chairman has given ethical approval to this proposal on behalf of the Committee.

Yours sincerely,

Daniel R. Lynch,
Secretary,
SJH/AMNCH Research Ethics Committee.

Appendix 11: Ethical approval Studies 3(a) and 3(b)

THIS NOTEPAPER MUST NOT BE USED FOR PRESCRIPTIONS OR INVOICING PURPOSES

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Ms. Katie Sheehan
Physiotherapy Department
Trinity Centre for Health Sciences
St. James's Hospital
Dublin 8

January 27th. 2011

REC reference: 2011/01/08

(Please quote REC reference and ~~EudraCT number on all correspondence~~)

RE: To Investigate Gait parameters in Healthy and Overweight Adults.

Dear Ms. Sheehan,

The SJH / AMNCH Research Ethics Committee, having reviewed the above application, decided to give ethical approval to this proposed study, subject to the following:

- Title should be changed to '…Normal Weight and Overweight Adults."
- Same changes to be made to information leaflet.

Yours sincerely,

p Colm O'Neill

Dr. Ray McDermott,
Chairman, SJH/AMNCH Research Ethics Committee

Appendix 12: Data set of overweight adults at SS and healthy weight adults at AC

Temporal-spatial parameters

Parameter	Overweight Group (mean ± SD)	Healthy Weight Group (mean ± SD)	Bias	p-value
Velocity (ms^{-1})	1.1 ± 0.2	1.2 ± 0.2	0.1	p = 0.2
Stride Length (m)	1.2 ± 0.1	1.3 ± 0.1	0.1	p = 0.1
Cadence (steps/min)	106.2 ± 11.0	105.4 ± 10.6	-0.8	p = 0.9
Percentage stance phase	63.8 ± 1.3	61.5 ± 1.5	-2.3	p = 0.0*
Single Support (s)	0.4 ± 0.0	0.4 ± 0.0	0.0	p = 0.0*
Double Support (s)	0.2 ± 0.0	0.1 ± 0.0	0.0	p = 0.0*

Results for temporal-spatial parameters. SD = standard deviation. p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

Timing of events

No significant difference in the timing of initial contact and tibia vertical were noted between the two groups (p>0.05). The timing of opposite toe off, heel rise, opposite initial contact, toe off, and feet adjacent were found to be significantly different between the two groups (p<0.00).

Kinematics Sagittal Plane

Event	Overweight Group (mean ± SD)	Healthy Weight Group (mean ± SD)	Bias	p-value
Initial contact	11.8 ± 4.4	9.6 ± 5.7	-2.2	p = 0.1
Opposite toe off	10.2 ± 4.3	8.9 ± 5.8	-1.3	p = 0.4
Heel rise	11.8 ± 4.5	10.2 ± 5.7	-1.7	p = 0.3
Opposite initial contact	11.9 ± 4.4	9.9 ± 5.9	-2.0	p = 0.2
Toe off	10.3 ± 4.4	9.1 ± 5.9	-1.3	p = 0.4
Feet adjacent	11.3 ± 4.0	9.6 ± 5.4	-1.7	p = 0.2
Tibia vertical	11.7 ± 4.4	10.1 ± 5.4	-1.6	p = 0.3

Results for pelvis anterior-posterior tilt (+/-). SD = standard deviation. p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

Event	Overweight Group (mean ± SD)	Healthy Weight Group (mean ± SD)	Bias	p-value
Initial contact	37.7 ± 6.7	33.9 ± 6.3	-3.7	p = 0.0*
Opposite toe off	32.1 ± 6.8	29.8 ± 7.4	-2.3	p = 0.3
Heel rise	3.4 ± 7.9	1.5 ± 7.1	-1.9	p = 0.4
Opposite initial contact	2.1 ± 8.2	-1.7 ± 6.9	-3.8	p = 0.1
Toe off	8.6 ± 8.6	5.3 ± 7.1	-3.3	p = 0.1
Feet adjacent	34.6 ± 6.9	32.5 ± 6.0	-2.1	p = 0.3
Tibia vertical	37.5 ± 6.9	35.9 ± 6.0	-1.6	p = 0.3

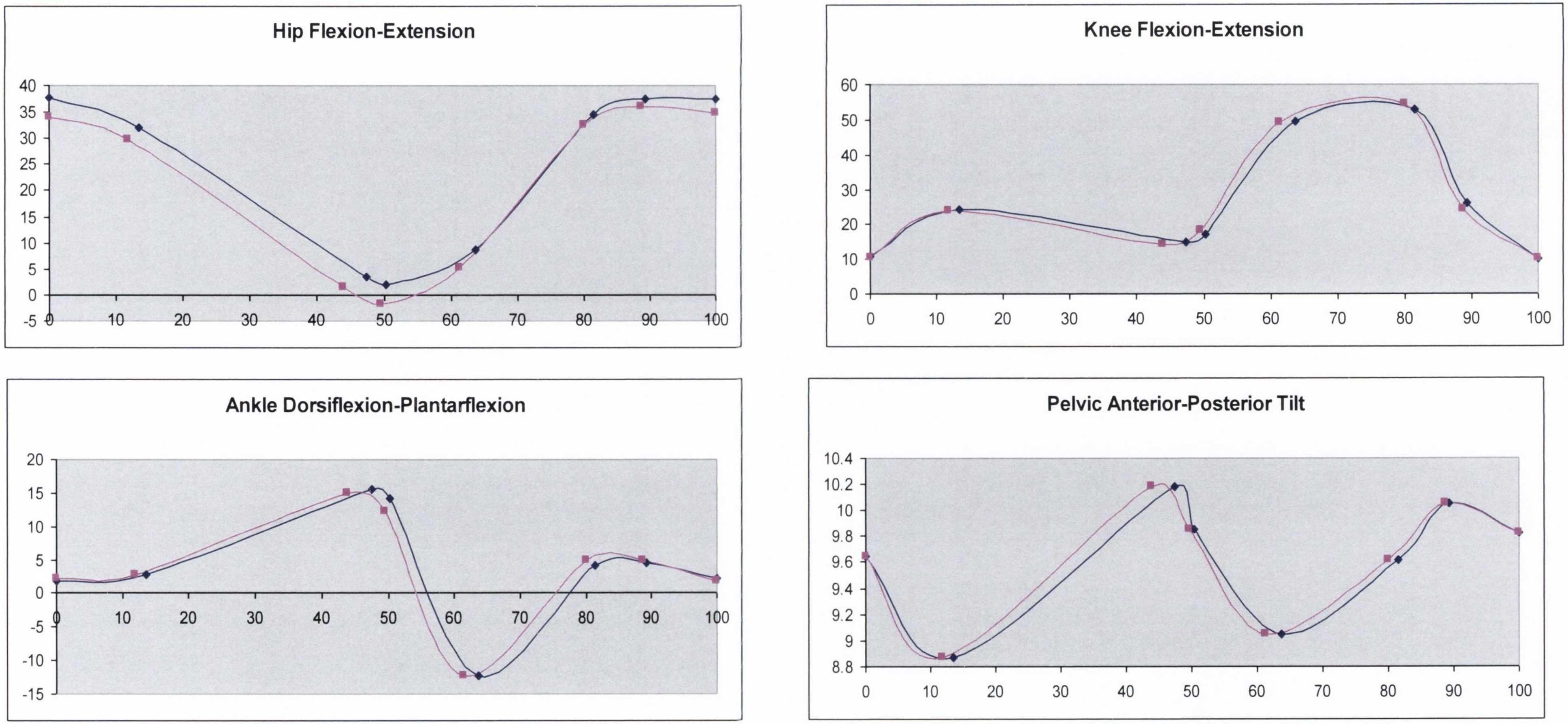
Results for hip flexion-extension (+/-). SD = standard deviation. p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

Event	Overweight Group (mean ± SD)	Healthy Weight Group (mean ± SD)	Bias	p-value
Initial contact	10.7 ± 7.2	10.3 ± 3.5	-0.4	p = 0.8
Opposite toe off	24.1 ± 6.7	23.7 ± 5.9	-0.4	p = 0.8
Heel rise	14.7 ± 7.3	14.1 ± 4.5	-0.7	p = 0.7
Opposite initial contact	17.2 ± 6.7	18.4 ± 4.3	1.2	p = 0.5
Toe off	49.6 ± 6.4	49.1 ± 5.0	-0.5	p = 0.8
Feet adjacent	53.0 ± 5.8	54.4 ± 5.1	1.4	p = 0.5
Tibia vertical	26.2 ± 4.6	24.4 ± 3.1	-1.8	p = 0.0*

Results for knee flexion-extension (+/-). SD = standard deviation. p-value with significance at alpha $\leq$ 0.05. = significant difference between groups.

Event	Overweight Group (mean ± SD)	Healthy Weight Group (mean ± SD)	Bias	p-value
Initial contact	1.8 ± 4.5	2.2 ± 3.1	0.3	p = 0.8
Opposite toe off	2.7 ± 3.7	2.8 ± 2.4	0.1	p = 0.9
Heel rise	15.5 ± 3.6	15.0 ± 2.8	-0.5	p = 0.6
Opposite initial contact	14.2 ± 4.4	12.3 ± 4.2	-1.9	p = 0.1
Toe off	-12.2 ± 8.6	-12.4 ± 5.3	-0.1	p = 0.9
Feet adjacent	4.2 ± 4.6	4.9 ± 2.8	0.7	p = 0.9
Tibia vertical	4.6 ± 2.9	4.8 ± 3.1	0.2	p = 0.7

Results for Ankle dorsiflexion-plantarflexion (+/-). SD = standard deviation. p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.



Sagittal plane kinematics.
X axis = percentage of one gait cycle.
Yaxis=degrees.

Overweight Group

Healthy weight Group

Frontal Plane

Event	Overweight Group (mean ± SD)	Healthy Weight Group (mean ± SD)	Bias	p-value
Initial contact	-0.7 ± 0.9	-0.8 ± 1.3	0.0	p = 0.9
Opposite toe off	2.4 ± 1.1	2.5 ± 1.8	0.1	p = 0.6
Heel rise	0.8 ± 1.0	0.5 ± 0.8	-0.3	p = 0.3
Opposite initial contact	0.7 ± 1.0	0.4 ± 0.9	-0.3	p = 0.1
Toe off	-2.3 ± 1.2	-2.5 ± 1.5	-0.2	p = 0.7
Feet adjacent	-0.6 ± 1.2	-0.5 ± 0.9	0.1	p = 0.9
Tibia vertical	-0.4 ± 0.9	-0.2 ± 0.8	0.1	p = 0.7

Results for Pelvis up-down obliquity (+/-). SD = standard deviation. p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

Event	Overweight Group (mean ± SD)	Healthy Weight Group (mean ± SD)	Bias	p-value
Initial contact	-2.7 ± 2.5	-1.6 ± 2.9	1.1	p = 0.2
Opposite toe off	2.8 ± 2.9	3.8 ± 3.0	1.0	p = 0.2
Heel rise	3.4 ± 2.3	3.5 ± 1.8	0.1	p = 0.8
Opposite initial contact	3.0 ± 2.5	2.7 ± 1.8	-0.2	p = 0.7
Toe off	-4.9 ± -3.5	-3.5 ± 2.1	1.4	p = 0.0*
Feet adjacent	-3.6 ± 3.1	-1.4 ± 2.0	2.2	p = 0.0*
Tibia vertical	-2.4 ± 3.2	-1.3 ± 1.8	1.0	p = 0.2

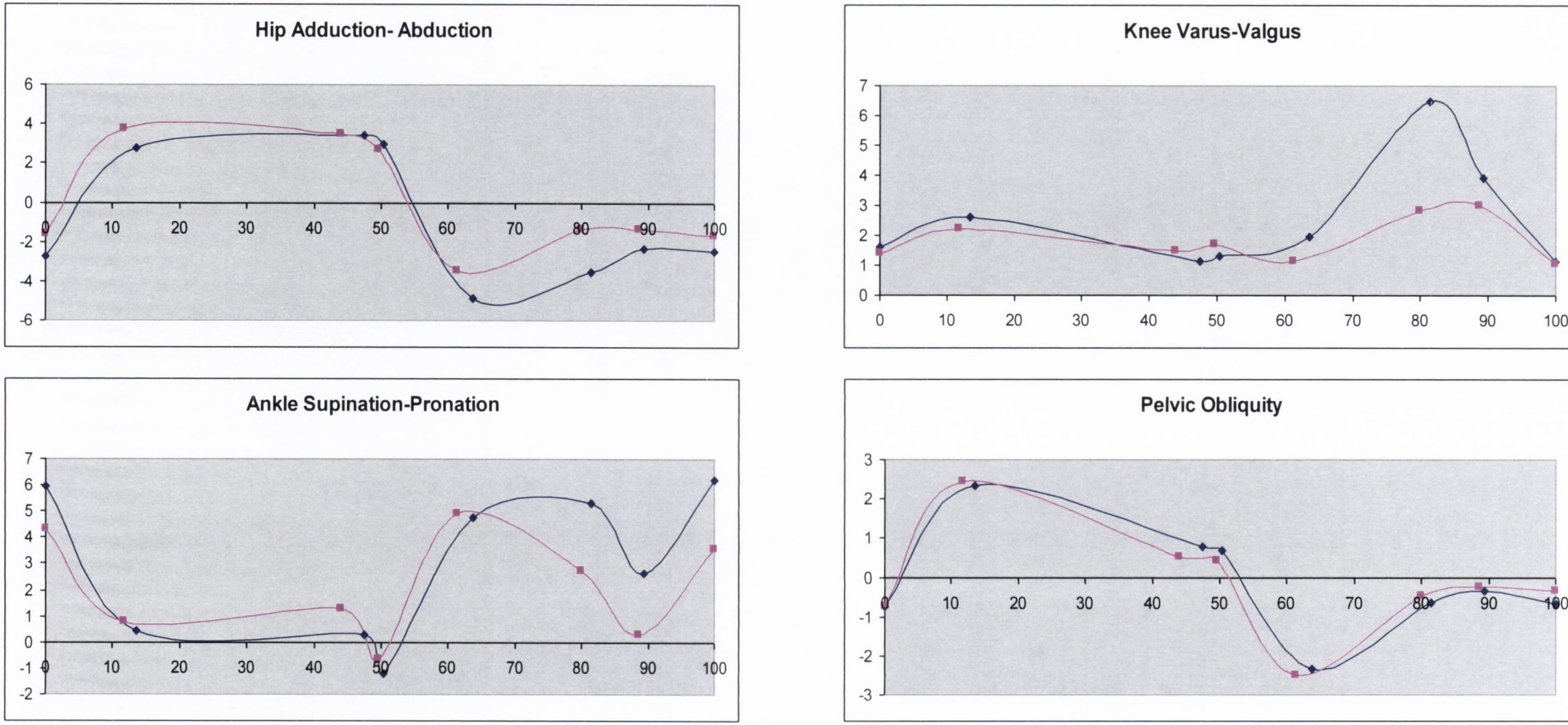
Results for hip adduction-abduction (+/-). SD = standard deviation. p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

Event	Overweight Group (mean ± SD)	Healthy Weight Group (mean ± SD)	Bias	p-value
Initial contact	1.6 ± 3.1	1.4 ± 2.7	-0.3	p = 0.8
Opposite toe off	2.6 ± 3.3	2.2 ± 3.3	-0.4	p = 0.6
Heel rise	1.1 ± 3.3	1.5 ± 3.1	0.4	p = 0.7
Opposite initial contact	1.3 ± 3.4	1.7 ± 3.2	0.4	p = 0.7
Toe off	2.0 ± 3.5	1.1 ± 4.2	-0.8	p = 0.5
Feet adjacent	6.5 ± 3.8	2.8 ± 4.8	-3.6	p = 0.0*
Tibia vertical	3.9 ± 2.8	3.0 ± 3.0	-0.9	p = 0.3

Results for knee varus-valgus (+/-). SD = standard deviation. p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

Event	Overweight Group (mean ± SD)	Healthy Weight Group (mean ± SD)	Bias	p-value
Initial contact	6.0 ± 7.1	4.3 ± 5.7	-1.6	p = 0.4
Opposite toe off	0.4 ± 8.5	0.8 ± 7.0	0.4	p = 0.9
Heel rise	0.3 ± 8.8	1.3 ± 6.5	0.9	p = 0.7
Opposite initial contact	-1.2 ± 7.2	-0.7 ± 6.3	0.5	p = 0.8
Toe off	4.7 ± 5.1	4.9 ± 3.6	0.2	p = 0.9
Feet adjacent	5.3 ± 5.1	2.7 ± 4.1	-2.6	p = 0.0*
Tibia vertical	2.6 ± 9.0	0.3 ± 7.0	-2.3	p = 0.6

Results for Ankle supination-pronation (+/-). SD = standard deviation. p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.



Frontal plane kinematics.
X axis = percentage of one gait cycle.
Y axis = degrees.

Overweight Group

Healthy weight Group

Event	Overweight Group (mean ± SD)	Healthy Weight Group (mean ± SD)	Bias	p-value
Initial contact	2.5 ± 2.4	2.2 ± 4.3	-0.4	p = 0.8
Opposite toe off	2.3 ± 2.0	2.2 ± 2.1	-0.3	p = 0.8
Heel rise	-1.4 ± 2.0	-0.3 ± 2.6	1.1	p = 0.1
Opposite initial contact	-2.2 ± 2.1	-1.9 ± 2.8	0.3	p = 0.7
Toe off	-2.2 ± 1.7	-1.9 ± 1.9	0.2	p = 0.7
Feet adjacent	-2.8 ± 1.2	-3.0 ± 1.7	-0.2	p = 0.6
Tibia vertical	-1.2 ± 1.5	-1.5 ± 1.7	-0.3	p = 0.6

Results for Pelvis forward/backward rotation (+/-). SD = standard deviation. p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

Event	Overweight Group (mean ± SD)	Healthy Weight Group (mean ± SD)	Bias	p-value
Initial contact	5.2 ± 7.8	4.9 ± 7.3	-0.4	p = 0.9
Opposite toe off	5.1 ± 8.8	6.5 ± 7.7	1.4	p = 0.5
Heel rise	2.3 ± 8.6	4.0 ± 6.8	1.7	p = 0.4
Opposite initial contact	2.1 ± 8.1	3.9 ± 6.9	1.8	p = 0.4
Toe off	-2.2 ± 6.1	0.2 ± 6.8	2.4	p = 0.2
Feet adjacent	2.3 ± 6.0	2.0 ± 6.5	-0.3	p = 0.9
Tibia vertical	3.0 ± 6.8	3.7 ± 6.2	0.7	p = 0.7

Results for hip internal-external rotation (+/-). SD = standard deviation. p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

Event	Overweight Group (mean ± SD)	Healthy Weight Group (mean ± SD)	Bias	p-value
Initial contact	-28.8 ± 8.2	-27.2 ± 7.4	1.6	p = 0.5
Opposite toe off	-21.8 ± 9.0	-20.1 ± 7.3	1.8	p = 0.5
Heel rise	-18.8 ± 8.3	-19.2 ± 7.0	-0.4	p = 0.9
Opposite initial contact	-18.3 ± 8.2	-18.2 ± 6.9	0.1	p = 0.9
Toe off	-18.7 ± 6.6	-18.9 ± 5.9	-0.2	p = 0.9
Feet adjacent	-22.1 ± 5.4	-20.8 ± 8.4	1.3	p = 0.5
Tibia vertical	-27.1 ± 7.4	-25.6 ± 7.9	1.5	p = 0.5

Results for knee internal-external rotation (+/-). SD = standard deviation. p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

Event	Overweight Group (mean ± SD)	Healthy Weight Group (mean ± SD)	Bias	p-value
Initial contact	1.9 ± 6.1	3.8 ± 4.7	1.9	p = 0.2
Opposite toe off	-4.2 ± 6.5	-4.1 ± 4.3	0.1	p = 0.9
Heel rise	-4.4 ± 5.8	-3.7 ± 4.7	0.7	p = 0.6
Opposite initial contact	-2.2 ± 6.1	0.1 ± 5.1	2.3	p = 0.2
Toe off	7.1 ± 6.5	9.1 ± 5.9	2.0	p = 0.3
Feet adjacent	1.3 ± 6.6	0.4 ± 4.4	-0.9	p = 0.6
Tibia vertical	1.9 ± 6.1	3.4 ± 4.3	1.5	p = 0.3

Results for Ankle internal-external alignment (+/-). SD = standard deviation. p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

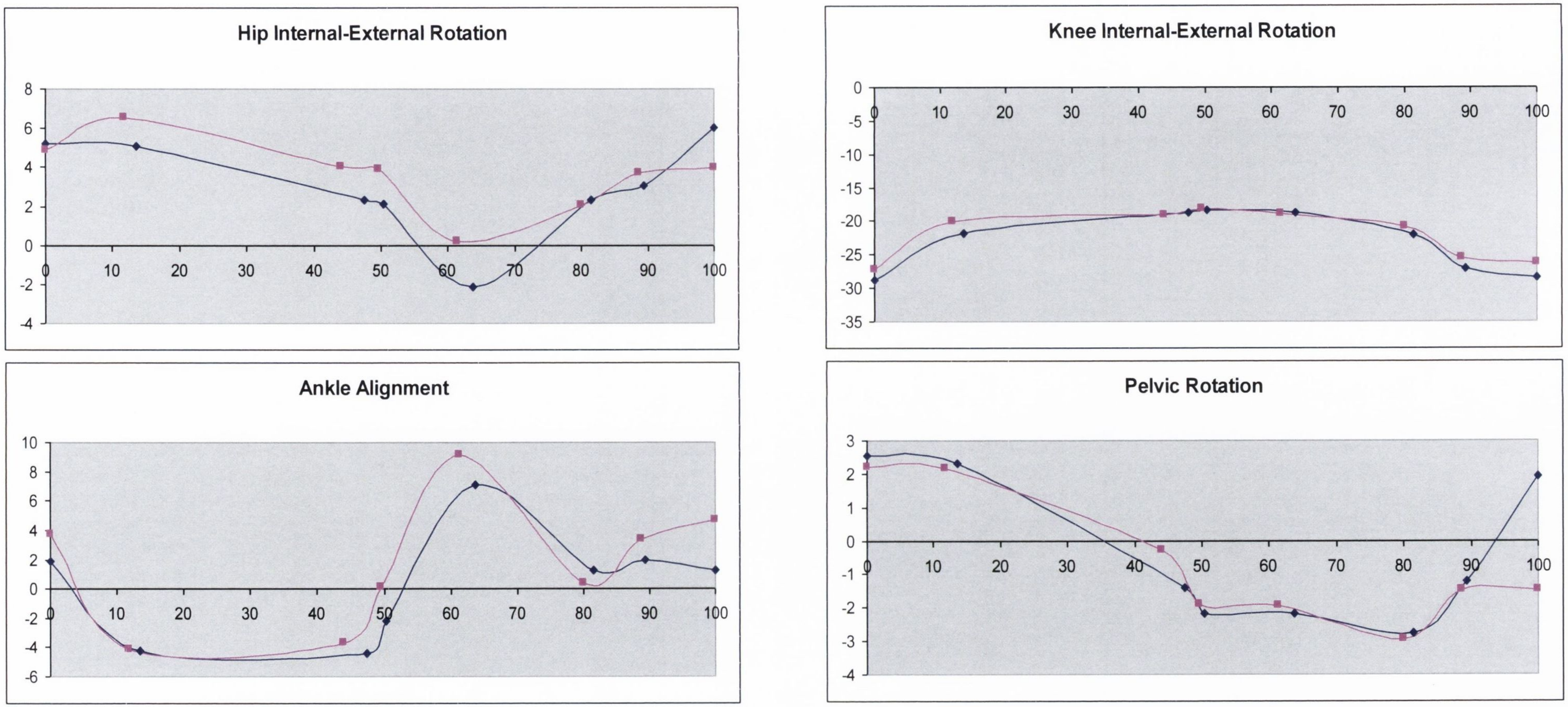
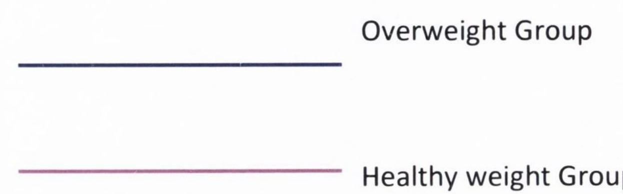


Figure 6.3: Transverse plane kinematics.

X axis = percentage of one gait cycle.

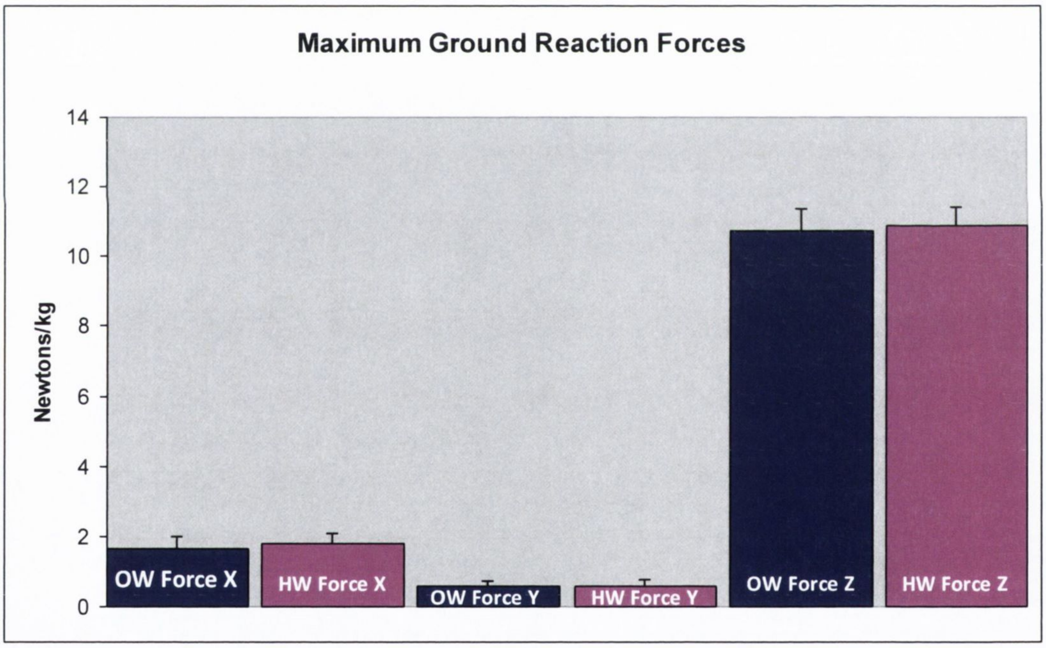
Y axis = degrees.



Kinetics Ground Reaction Forces

Parameter (N/kg)	Overweight Group (mean ± SD)	Healthy Weight Group (mean ± SD)	Bias	p-value
Force X	1.7 ± 0.4	1.8 ± 0.3	0.2	p = 0.1
Force Y	0.6 ± 0.2	0.6 ± 0.2	0.0	p = 0.8
Force Z	10.8 ± 0.6	10.9 ± 0.6	0.1	p = 0.4

Maximum ground reaction forces. SD = standard deviation. p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.



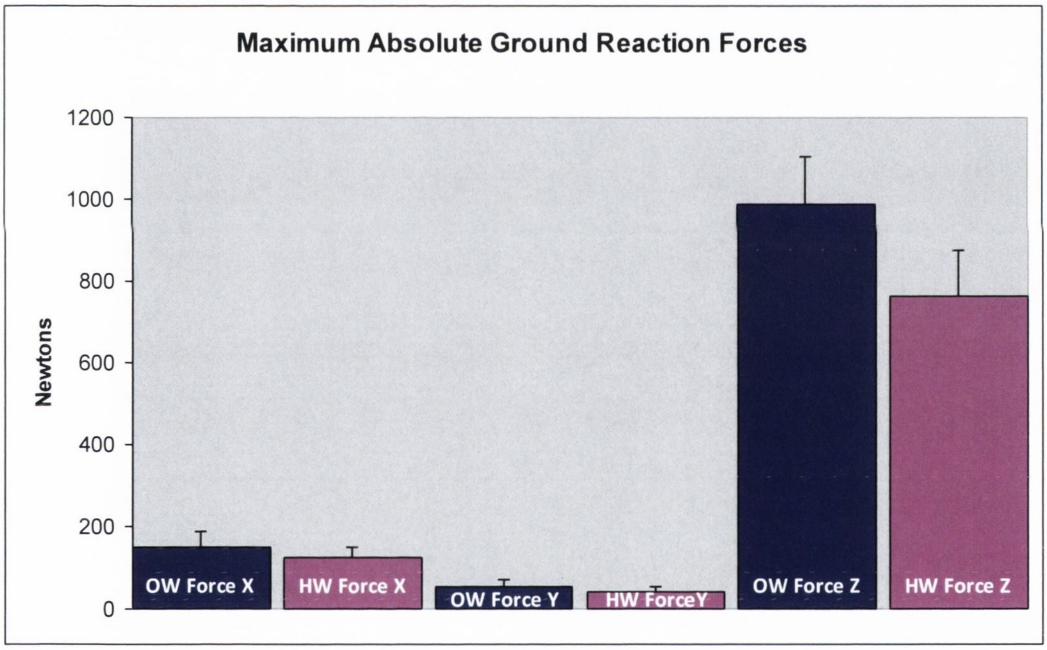
Maximum ground reaction forces.

OW = Overweight

HW = Healthy weight

Parameter (N/kg)	Overweight Group (mean ± SD)	Healthy Weight Group (mean ± SD)	Bias	p-value
Force X	151.5 ± 35.5	125.3 ± 24.2	-26.2	p = 0.0*
Force Y	53.4 ± 15.7	41.2 ± 12.4	-12.1	p = 0.0*
Force Z	989.5 ± 114.2	762.7 ± 112.5	-226.8	p = 0.0*

Maximum absolute ground reaction forces. SD = standard deviation. p-value with significance at alpha ≤ 0.05. * = significant difference between groups.



Maximum absolute ground reaction forces.

OW = Overweight

HW = Healthy weight

Sagittal plane joint moments

Event	Overweight Group (mean ± SD)	Healthy Weight Group (mean ± SD)	Bias	p-value
Initial contact	0.0 ± 0.2	-0.1 ± 0.2	0.0	p = 0.7
Opposite toe off	0.7 ± 0.3	0.6 ± 0.2	-0.1	p = 0.2
Heel rise	-0.3 ± 0.4	-0.3 ± 0.2	0.0	p = 0.6
Opposite initial contact	-0.4 ± 0.4	-0.4 ± 0.1	-0.1	p = 0.2
Toe off	-0.2 ± 0.1	-0.3 ± 0.2	0.0	p = 0.9
Feet adjacent	-0.1 ± 0.2	-0.1 ± 0.1	0.1	p = 0.1
Tibia vertical	0.0 ± 0.1	0.1 ± 0.1	0.1	p = 0.1

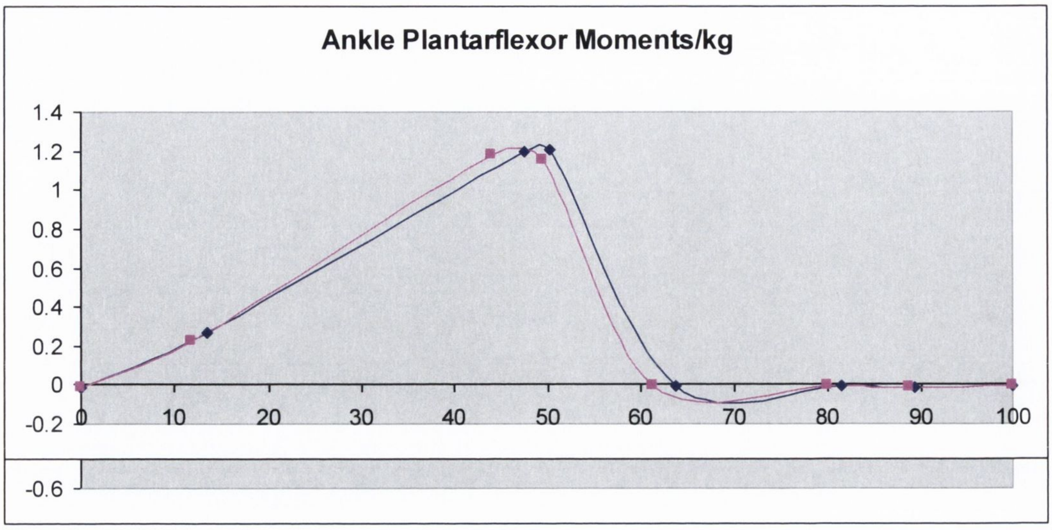
Results for Hip extensor-flexor moments (+/-). SD = standard deviation. p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

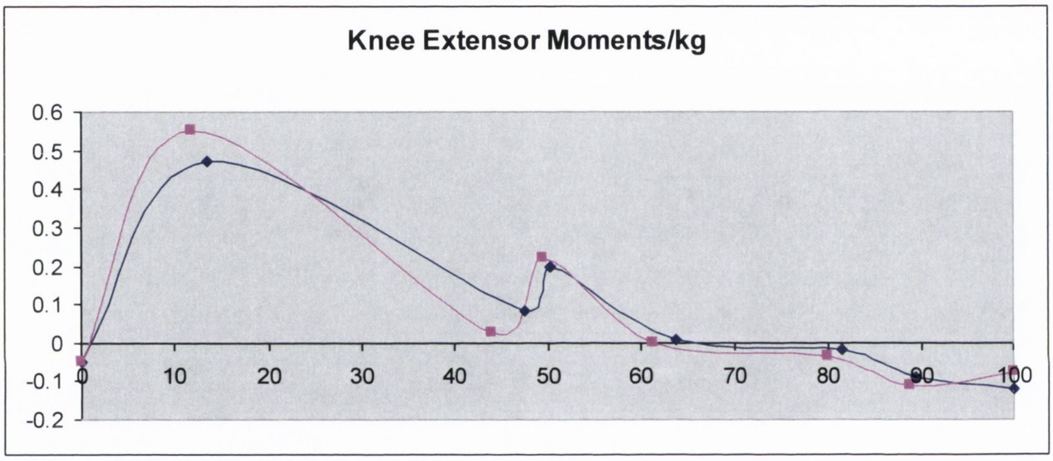
Event	Overweight Group (mean ± SD)	Healthy Weight Group (mean ± SD)	Bias	p-value
Initial contact	-0.1 ± 0.1	-0.1 ± 0.1	0.0	p = 0.9
Opposite toe off	0.5 ± 0.2	0.6 ± 0.2	0.1	p = 0.2
Heel rise	0.1 ± 0.2	0.0 ± 0.2	-0.1	p = 0.2
Opposite initial contact	0.2 ± 0.1	0.2 ± 0.2	0.0	p = 0.6
Toe off	0.0 ± 0.0	0.0 ± 0.0	0.0	p = 0.2
Feet adjacent	0.0 ± 0.1	0.0 ± 0.0	0.0	p = 0.2
Tibia vertical	-0.1 ± 0.0	-0.1 ± 0.0	0.0	p = 0.1

Results for Knee extensor-flexor moments (+/-). SD = standard deviation. p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

Event	Overweight Group (mean ± SD)	Healthy Weight Group (mean ± SD)	Bias	p-value
Initial contact	0.0 ± 0.0	0.0 ± 0.0	0.0	p = 0.4
Opposite toe off	0.3 ± 0.1	0.2 ± 0.1	-0.1	p = 0.2
Heel rise	1.2 ± 0.1	1.2 ± 0.2	0.0	p = 0.9
Opposite initial contact	1.2 ± 0.1	1.2 ± 0.2	-0.1	p = 0.5
Toe off	0.0 ± 0.0	0.0 ± 0.0	0.0	p = 0.6
Feet adjacent	0.0 ± 0.0	0.0 ± 0.0	0.0	p = 0.9
Tibia vertical	0.0 ± 0.0	0.0 ± 0.0	0.0	p = 0.3

Results for Ankle plantarflexor-dorsiflexor moments (+/-). SD = standard deviation. p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.





Sagittal plane joint moments.

X axis = percentage of one gait cycle.

Y axis = Newton-meters/kg.

The eight events of the gait cycle

are demarcated on the lines.

Overweight Group

Healthy weight Group

Frontal plane joint moments

Event	Overweight Group (mean ± SD)	Healthy Weight Group (mean ± SD)	Bias	p-value
Initial contact	0.0 ± 0.1	0.0 ± 0.1	-0.1	p = 0.0*
Opposite toe off	0.5 ± 0.1	0.5 ± 0.1	0.0	p = 0.4
Heel rise	0.7 ± 0.1	0.6 ± 0.1	0.0	p = 0.6
Opposite initial contact	0.7 ± 0.1	0.6 ± 0.1	-0.1	p = 0.2
Toe off	-0.1 ± 0.1	-0.1 ± 0.1	0.0	p = 0.7
Feet adjacent	0.0 ± 0.1	0.1 ± 0.1	0.0	p = 0.2
Tibia vertical	0.0 ± 0.1	0.0 ± 0.1	0.0	p = 0.2

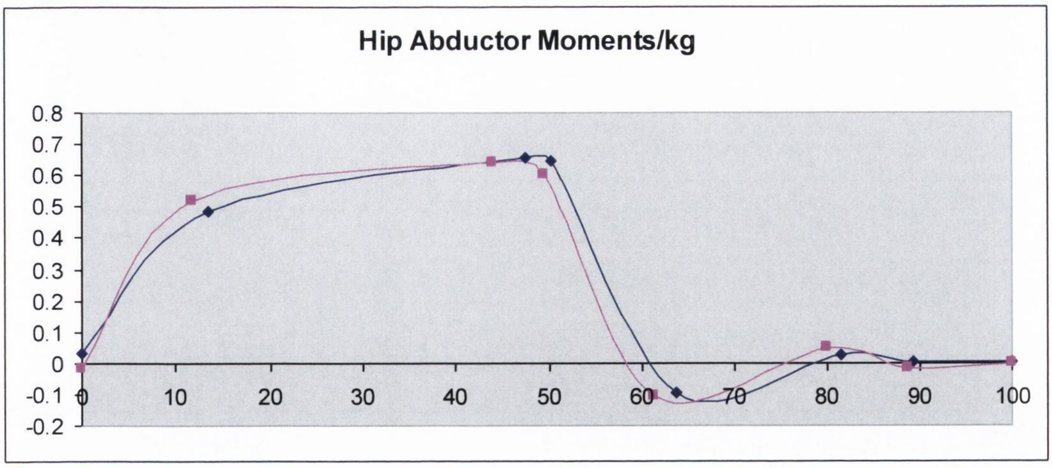
Results for Hip abductor-adductor moments (+/-). SD = standard deviation. p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

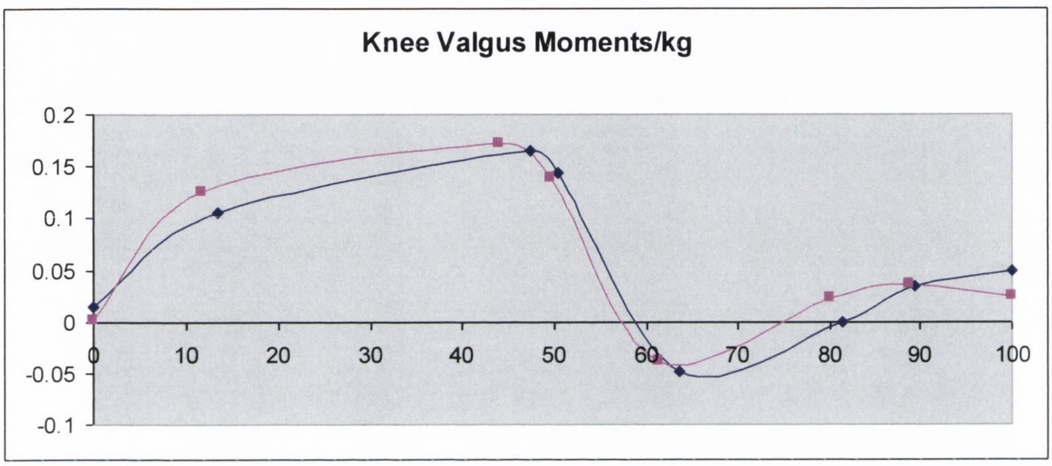
Event	Overweight Group (mean ± SD)	Healthy Weight Group (mean ± SD)	Bias	p-value
Initial contact	0.0 ± 0.0	0.0 ± 0.0	0.0	p = 0.1
Opposite toe off	0.1 ± 0.2	0.1 ± 0.1	0.0	p = 0.6
Heel rise	0.2 ± 0.2	0.2 ± 0.1	0.0	p = 0.9
Opposite initial contact	0.1 ± 0.2	0.1 ± 0.1	0.0	p = 0.9
Toe off	-0.1 ± 0.0	0.0 ± 0.0	0.0	p = 0.2
Feet adjacent	0.0 ± 0.0	0.0 ± 0.0	0.0	p = 0.0*
Tibia vertical	0.0 ± 0.0	0.0 ± 0.0	0.0	p = 0.8

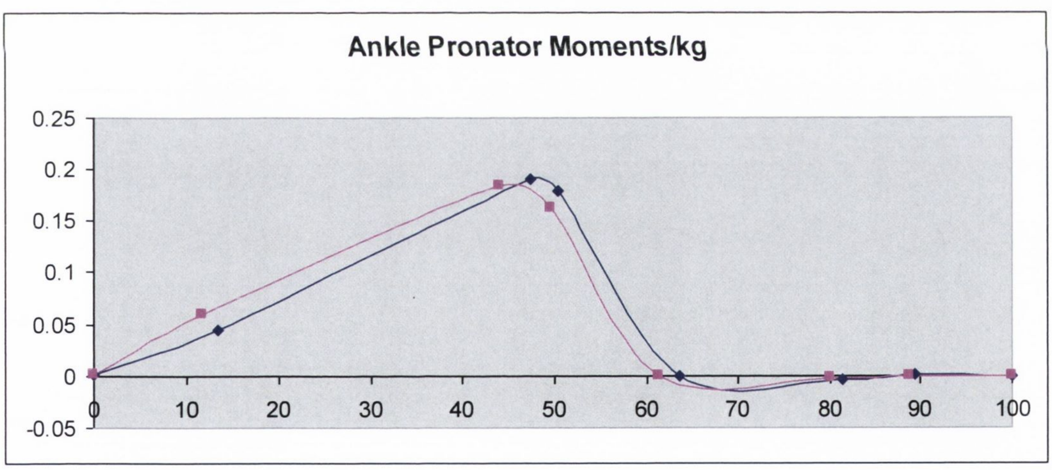
Results for Knee valgus-varus moments (+/-). SD = standard deviation. p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

Event	Overweight Group (mean ± SD)	Healthy Weight Group (mean ± SD)	Bias	p-value
Initial contact	0.0 ± 0.0	0.0 ± 0.0	0.0	p = 0.9
Opposite toe off	0.0 ± 0.1	0.1 ± 0.1	0.0	p = 0.5
Heel rise	0.2 ± 0.1	0.2 ± 0.1	0.0	p = 0.8
Opposite initial contact	0.2 ± 0.1	0.2 ± 0.1	0.0	p = 0.6
Toe off	0.0 ± 0.0	0.0 ± 0.0	0.0	p = 0.5
Feet adjacent	0.0 ± 0.0	0.0 ± 0.0	0.0	p = 0.2
Tibia vertical	0.0 ± 0.0	0.0 ± 0.0	0.0	p = 0.3

Results for Ankle pronator-supinator moments (+/-). SD = standard deviation. p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.







Frontal plane joint moments.

X axis = percentage of one gait cycle.

Y axis = Newton-meters/kg.

The eight events of the gait cycle are demarcated on the lines.

Overweight Group

Healthy weight Group

Transverse plane joint moments

Event	Overweight Group (mean ± SD)	Healthy Weight Group (mean ± SD)	Bias	p-value
Initial contact	0.0 ± 0.0	0.0 ± 0.0	0.0	p = 0.4
Opposite toe off	0.1 ± 0.0	0.1 ± 0.0	0.0	p = 0.9
Heel rise	0.0 ± 0.0	0.0 ± 0.0	0.0	p = 0.8
Opposite initial contact	0.0 ± 0.0	0.0 ± 0.0	0.0	p = 0.4
Toe off	0.0 ± 0.0	0.0 ± 0.0	0.0	p = 0.6
Feet adjacent	0.0 ± 0.0	0.0 ± 0.0	0.0	p = 0.1
Tibia vertical	0.0 ± 0.0	0.0 ± 0.0	0.0	p = 0.5

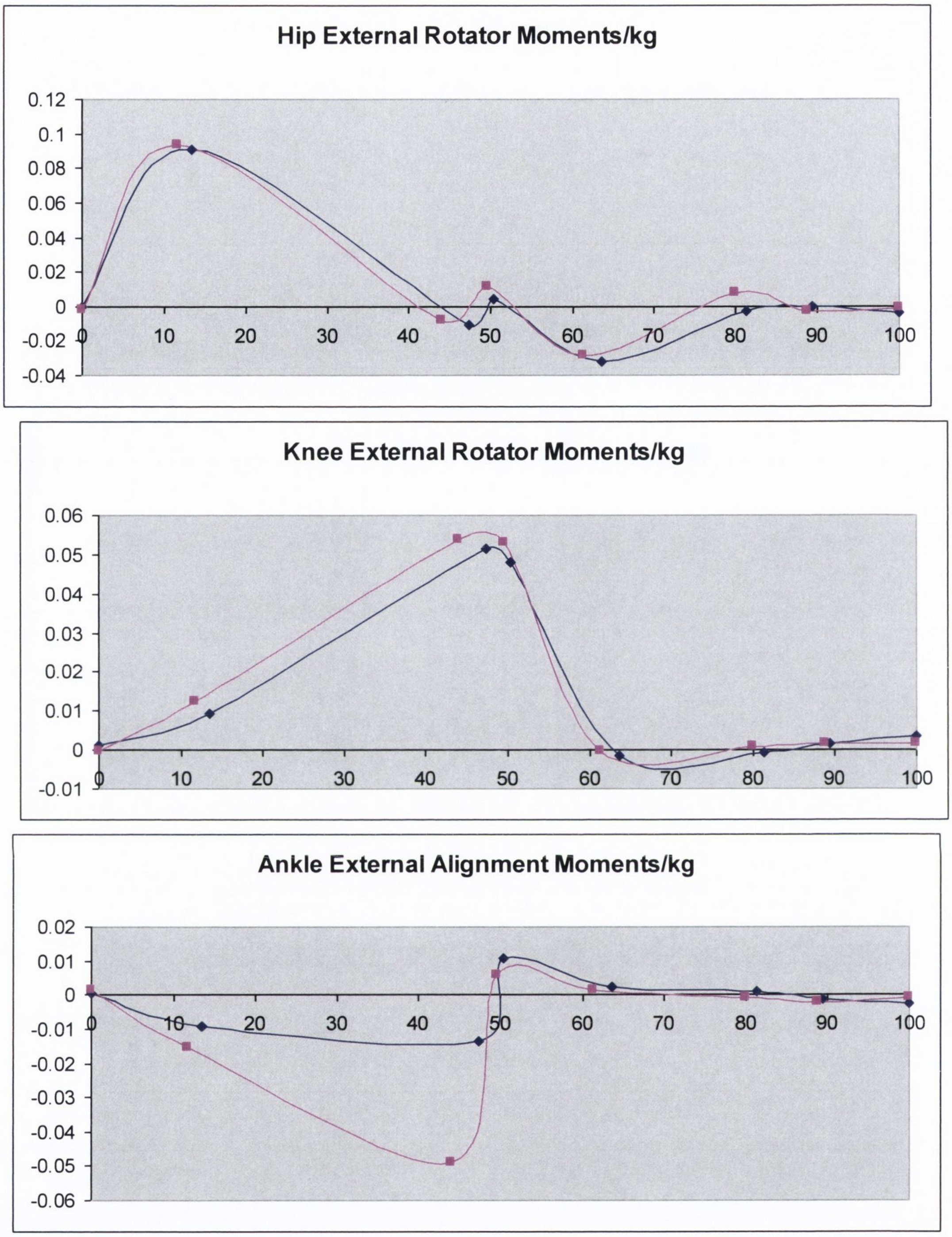
Results for Hip external-internal rotator moments (+/-). SD = standard deviation. p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

Event	Overweight Group (mean ± SD)	Healthy Weight Group (mean ± SD)	Bias	p-value
Initial contact	0.0 ± 0.0	0.0 ± 0.0	0.0	p = 0.1
Opposite toe off	0.0 ± 0.0	0.0 ± 0.0	0.0	p = 0.6
Heel rise	0.1 ± 0.1	0.1 ± 0.0	0.0	p = 0.9
Opposite initial contact	0.1 ± 0.1	0.1 ± 0.0	0.0	p = 0.7
Toe off	0.0 ± 0.0	0.0 ± 0.0	0.0	p = 0.1
Feet adjacent	0.0 ± 0.0	0.0 ± 0.0	0.0	p = 0.0*
Tibia vertical	0.0 ± 0.0	0.0 ± 0.0	0.0	p = 0.7

Results for Knee external-internal rotator moments (+/-). SD = standard deviation. p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.

Event	Overweight Group (mean ± SD)	Healthy Weight Group (mean ± SD)	Bias	p-value
Initial contact	0.0 ± 0.0	0.0 ± 0.0	0.0	p = 0.2
Opposite toe off	0.0 ± 0.0	0.0 ± 0.0	0.0	p = 0.4
Heel rise	0.0 ± 0.2	-0.1 ± 0.2	0.0	p = 0.5
Opposite initial contact	0.0 ± 0.1	0.0 ± 0.1	0.0	p = 0.9
Toe off	0.0 ± 0.0	0.0 ± 0.0	0.0	p = 0.4
Feet adjacent	0.0 ± 0.0	0.0 ± 0.0	0.0	p = 0.0*
Tibia vertical	0.0 ± 0.0	0.0 ± 0.0	0.0	p = 0.3

Results for Ankle external-internal alignment moments (+/-). SD = standard deviation. p-value with significance at alpha $\leq$ 0.05. * = significant difference between groups.



Transverse plane joint moments.

X axis = percentage of one gait cycle.

Y axis = Newton-meters/kg.

The eight events of the gait cycle are demarcated on the lines.

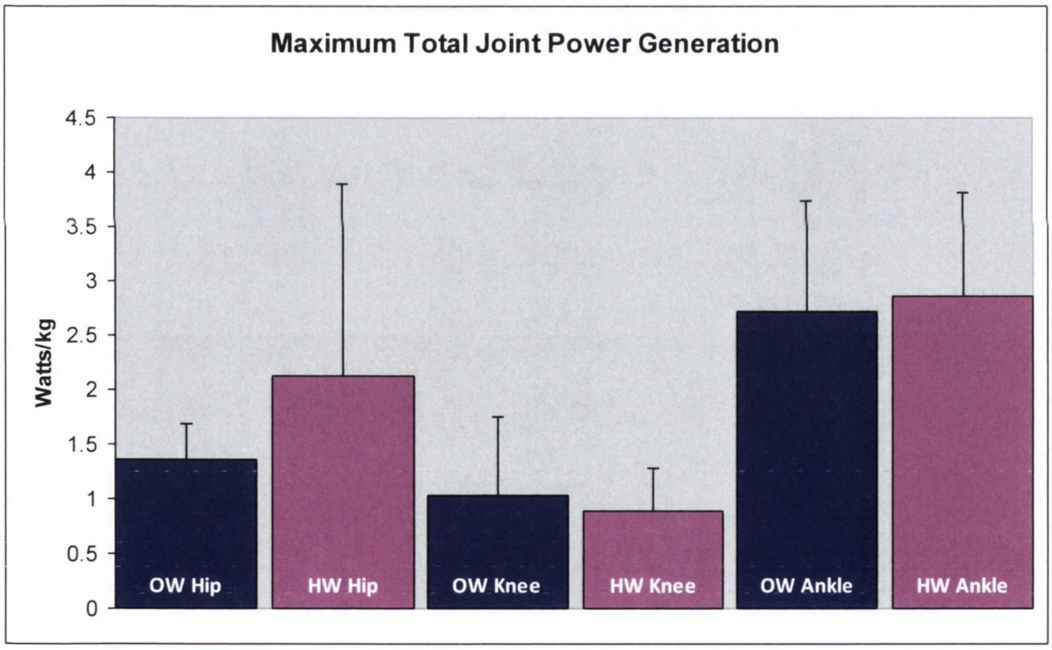
Overweight Group

Healthy weight Group

Joint Power

Parameter (W/kg)	Overweight Group (mean ± SD)	Healthy Weight Group (mean ± SD)	Bias	p-value
Hip	1.4 ± 0.3	2.1 ± 1.8	0.8	p = 0.2
Knee	1.0 ± 0.7	0.9 ± 0.4	-0.2	p = 0.8
Ankle	2.7 ± 1.0	2.9 ± 1.0	0.1	p = 0.6

Maximum total joint power generation. SD = standard deviation. p-value with significance at alpha ≤ 0.05. * = significant difference between groups.



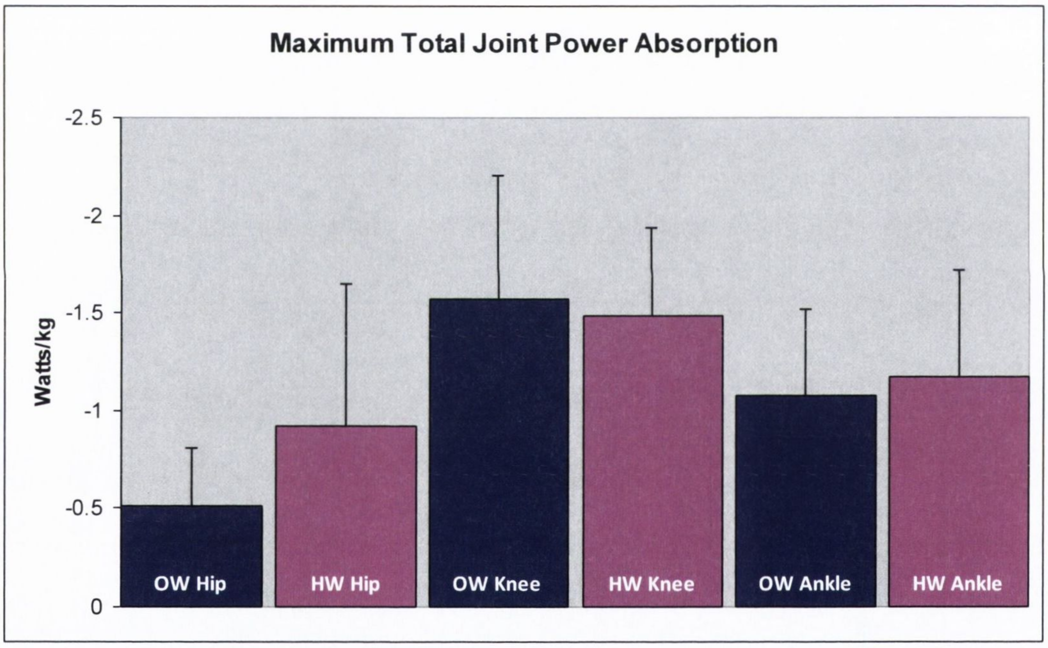
Maximum total joint power generation.

OW = Overweight

HW = Healthy weight

Parameter (W/kg)	**Overweight Group (mean ± SD)**	**Healthy Weight Group (mean ± SD)**	**Bias**	**p-value**
Hip	0.5 ± 0.3	0.9 ± 0.7	-0.4	p = 0.0*
Knee	1.6 ± 0.6	1.5 ± 0.5	0.1	p = 0.9
Ankle	1.1 ± 0.4	1.2 ± 0.5	-0.1	p = 0.4

Maximum total joint power absorption. SD = standard deviation. p-value with significance at alpha ≤ 0.05. * = significant difference between groups.



Maximum total joint power absorption.

OW = Overweight

HW = Healthy weight