Gender differences in landing biomechanics associated with ACL injury.

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Summary.

The incidence of non-contact anterior cruciate ligament (ACL) injury is reported to be 6 to 8 times greater in females than males competing in the same activities. Injury to the ACL occurs as a result of insufficient stability of the tibiofemoral joint, which fails to prevent abnormal movement of the femur on the tibia. The stability of the tibiofemoral joint is maintained by passive (non-contractile) and dynamic (contractile) mechanisms. The relative significance of the various passive and dynamic mechanisms in maintaining the stability of the tibiofemoral joint is not clear. The purpose of the review of literature was to present a risk factor model for ACL injury based on a review of passive and dynamic stability mechanisms. Current evidence suggests that the greater incidence of ACL injury in females is largely due to gender differences in dynamic stability rather than passive stability. The purpose of the present project was to examine gender differences in biomechanics during landing. The project was conducted in three stages. The aim of study one was to investigate the effects of gender on frontal and sagittal plane kinematics and kinetics in university volleyball players when performing opposed block jump landings, using 3-D motion analysis. The results suggest that the gender differences in lower limb alignments in normal upright standing do not totally account for the gender differences in landing kinematics on landing and may indicate less dynamic stability of the knee in females compared to males in the passive phase of landing. The aim of study two was to examine the effect of opposition on the kinematics and kinetics of landing from a volleyball block in male and female university volleyball players. Differences in sagittal plane knee kinematics and kinetics during opposed and unopposed trials suggest that the effect of opposition may significantly alter subjects' neuromuscular responses during landing. However, differences in frontal plane kinematics and kinetics between males and females appear to be consistent in unopposed and opposed conditions. The aim of study three was to investigate lower limb coordination and stiffness in male and female university volleyball players performing block jump landings. The results suggest reduced symmetry between left and right legs and reduced stability of coordination between the left and right knees in females compared to males. Also, males exhibited significantly greater absolute and normalised leg and knee stiffness than females. Overall, the results suggest reduced dynamic stability of the knee in females compared to males which may contribute to the greater incidence of ACL injury in females compared to males. Future research should investigate possible methods of improving the dynamic stability of the knee in females during landing.

Declaration.

I hereby declare that this thesis has been composed by myself, that the work is the result of my own investigations except where assistance has been otherwise acknowledged, that the work has not been previously submitted in candidature for any other degree, that all sources of information have been specifically acknowledged by means of reference, and that consent is provided for the thesis to be made available for photocopying and for inter-library loan. Some of the findings presented in this thesis have been published in peer-reviewed journals, as follows:

Hughes, G. & Watkins, J. (2006). A risk factor model for anterior cruciate ligament injury. *Sports Medicine.* 36 (5): 411-428.

Hughes, G., Watkins, J. & Owen, N. (2007). The effects of gender on lower limb kinematics during volleyball block jump landings. *Journal of Sports Sciences,* 25 (3): 241.

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Hughes, G. & Watkins, J. (In Press). Gender differences in lower limb coordination during volleyball block jump landings. *Journal of Sports Sciences.*

Hughes, G. & Watkins, J. (In Press). Gender differences in knee and leg stiffness during volleyball block jump landings. *Journal of Sports Sciences.*

Signature.. .... Date....2/8/07................

Gerwyn T G Hughes

Contents

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List of Abbreviations

Abbreviation	Meaning
ACL	Anterior cruciate ligament
AL	Start of active loading
ANKLE	Ankle joint centre
AP	Active loading phase
ASPT	Anterior shear force applied to the proximal end of the tibia
C7	7th cervical vertebrae
CLAV	Jugular notch where clavicles meet sternum
CG	Centre of gravity
CP	Centre of pressure
CRP	Continuous relative phase
DRP	Discrete relative phase
EMG	Electromyography
EXT	External moment
F_R	Resultant ground reaction force
Fx	Anterioposterior ground reaction force
Fy	Mediolateral ground reaction force
Fz	Vertical ground reaction force
GRAV	Gravitational moment
GRF	Ground reaction force
HIP	Hip joint centre
IC	Initial ground contact during landing
INW	Intercondylar notch width
KNEE	Knee joint centre
LANK	Lateral malleolus of the left ankle

List of Abbreviations (cont.)

Abbreviation	Meaning
LANKLE	Left ankle joint centre
LASI	Left anterior superior iliac spine
LBHD	Left back head
LELB	Left lateral epicondyle estimating left elbow joint axis
LFHD	Left front head
LFIN	Head of the second metacarpal of the left hand
LFRA	Left forearm, in line with LELB and LWRA
LHEE	Posterior aspect of the Achilles tendon of the left leg at the same height as the second metatarsal marker
LHIP	Left hip joint centre
LKNE	Lateral epicondyle of the left femur
LKNEE	Left knee joint centre
LPSI	Left posterior superior iliac spine
LSHO	Left acromio-clavicular joint
LTHI	Lower lateral surface of the left thigh along the line between the hip and knee joint markers
LTIB	Lower lateral surface of the left tibia along the line between knee and ankle joint markers
LTOE	Superior proximal end of the second metatarsal of the left foot
LUPA	Left upper arm, in line with LELB and LSHO
LWRA	Left medial wrist
LWRB	Left lateral wrist
MAX	Either maximum flexion or valgus/varus angle during landing

List of Abbreviations (cont.)

Abbreviation	Meaning
MDM	Motion dependent moment
MIN	Minimum inter-joint distance
ML	Start of muscle latency
MSE	Mean squared error
MUS	Muscle moment
MVC	Maximum voluntary contraction
NET	Net joint moment
NWI	Ratio of INW to width of femoral epicondyles
PCL	Posterior cruciate ligament
PP	Passive loading phase
PSD	Power spectral density
PTTSA	Patella tendon-tibia shaft angle
RANK	Lateral malleolus of the right ankle
RANKLE	Right ankle joint centre
RASI	Right anterior superior iliac spine
RBAK	Middle of right scapula
RBHD	Right back head
RELB	Right lateral epicondyle estimating right elbow joint axis
RFHD	Right front head
RFIN	Head of the second metacarpal of the right hand
RFRA	Right forearm, in line with RELB and RWRA RSHO Right acromio-clavicular joint
RHEE	Posterior aspect of the Achilles tendon of the right leg at the same height as the second metatarsal marker

List of Abbreviations (cont.)

Abbreviation	Meaning
RHIP	Right hip joint centre
RKNE	Lateral epicondyle of the right femur
RKNEE	Right knee joint centre
ROM	Range of motion
RPSI	Right posterior superior iliac spine
RT	Muscle reaction time
RTHI	Lower lateral surface of the right thigh along the line between the hip and knee joint markers
RTIB	Lower lateral surface of the right tibia along the line between knee and ankle joint markers
RTOE	Superior proximal end of the second metatarsal of the right foot
RUPA	Right upper arm, in line with RELB and RSHO
RWRA	Right medial wrist
RWRB	Right lateral wrist
STRN	Xiphoid process of the sternum
T10	10^{th} thoracic vertebrae
TPT	Time to peak torque
V_o	Zero velocity of whole body centre of gravity
vDRP	Variability of discrete relative phase
VL	Vicon – laboratory environment
VLS	Vicon – laboratory environment – human subject
W	Weight

List of Figures

List of Figures (cont.)

List of Figures (cont.)

List of Figures (cont.)

List of Figures (cont.)

Figure	Title	Page

List of Figures (cont.)

List of Figures (cont.)

List of Figures (cont.)

List of Figures (cont.)

List of Figures (cont.)

List of Figures (cont.)

List of Figures (cont.)

List of Figures (cont.)

List of Figures (cont.)

List of Tables

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Chapter 1

Introduction and Review of Literature

A risk factor model for anterior cruciate ligament injury.

1.1 ACL injuries.

Anterior Cruciate ligament (ACL) injuries are common and approximately 70% of these injuries occurr in sport (Faegin, 1988; Johnson, 1988; Smith et al., 1988). ACL reconstruction was the 6th most common orthopaedic procedure performed in the US in 1999 and 2000 with approximately 100,000 being performed annually at a cost of over $2 billion for surgery alone (Gottlob et al., 1999; Griffin et al., 2000; Huston et al., 2000).

Up to two thirds of patients who have complete ACL tears develop knee instability and, subsequently, damage to the menisci and articular surfaces which significantly affects knee function and leads to a decrease in level of activity (Satku et al., 1986; Kannus and Jarvinen, 1987; Smith et al., 1988; Irvine and Glasgow, 1992). Noyes et al. (1983) found that in a group of individuals with rupture of the ACL, 31% of patients reported overall difficulty in walking alone, 44% had difficulties with activities of daily living including walking and 77% had difficulties with playing sport as a direct result of their ACL injury.

Between 70% and 90% of ACL injuries have been reported to occur in non-contact situations (McNair et al., 1993; Mykelbust et al., 1997; Griffin et al., 2000). A non-contact situation is where there is no direct contact with the knee when the injury occurs (Yu et al., 2002). Most ACL injuries appear to occur in situations involving one or more of the following manoeuvres: foot strike with knee close to full extension (Boden et al., 2000; Olsen et al., 2004), landing (Hume and Steele, 1997; Otago and Neal, 1997), deceleration (Miller et al., 1995) and rapid change of direction (Bartold, 1997). For example, in a study by Olsen et al. (2004) video tapes of game situations in which ACL injury occurred in team handball were analysed to try to identify the mechanisms for ACL injury. Three handball experts were used to identify possible risk factors such as body position, type of movement, with or without the ball, phase of play, balance, attention, speed and contact. Three physicians were used to identify factors relating to the knee position such as estimated flexion, rotation, varus-valgus angle, foot position and degree of weight bearing on the leg. The most common action being performed at the time of injury was a plant and cut movement (12 of the 20 cases analysed) accompanied by forceful valgus and external-internal rotation with the knee close to full extension. Of these twelve injuries, four occurred during two-foot stance, eight were in single foot stance and all occurred during the push off. The next most common injury mechanism was a one-legged jump shot landing, where athletes were jumping and landing on the same leg. This accounted for four cases and all occurred with a forceful valgus, external rotation and knee close to full extension on landing. Therefore it was concluded that movements that put the ACL at risk involved a single foot stance, forceful valgus knee movement with the knee close to full extension accompanied with external or internal rotation of the tibia. Not surprisingly, the incidence of ACL injury is relatively high in sports such as basketball, netball, handball, soccer and volleyball that are characterised by a high frequency of landing, decelerating and rapid changes of direction (Arendt and Dick, 1995; Griffin et al., 2000; Olsen et al., 2004).

1.2 Gender differences in ACL injuries.

The incidence of non-contact ACL injury in females has been reported to be 6 to 8 times greater than in males competing in the same sports (Chandy and Grana, 1985; Gray et al., 1985; Ferretti et al., 1992; Paulos, 1992; Malone et al., 1993; Lidenfeld et al., 1994; Irelan, 1994; Arendt and Dick, 1995; Gwinn et al., 2000). Arendt and Dick (1995) reported the incidence of ACL injury in collegiate basketball and soccer for males and females over the period 1989-93. Data were collected for 461 male and 278 female soccer teams and 531 male and 576 female basketball teams. ACL injury was reported in terms of athlete-exposure, where athlete-exposure took into account games and practise sessions. For soccer, female ACL injury incidence of 0.39 per 1000 athlete-exposures compared to 0.13 per 1000 athlete-exposures for males were reported. For basketball, the incidence of ACL injury was 0.29 per 1000 athlete exposures for females and 0.07 per 1000 athlete exposures for males. Malone et al. (1993) documented the injuries of 402 male and 385 female basketball players from 29 institutions in three Division 1 collegiate basketball conferences over a five year period. 62 females and 9 males sustained ACL injury, which corresponded to an incidence of 16.1% in females and 2.2% in males.

1.3 Anatomy of the knee complex.

To understand the possible mechanisms which contribute to ACL injury and therefore possible causes of the gender difference in ACL injury, there is a need to have a brief understanding of the anatomy of the knee complex, how stability of the joint in maintained and the role of the ACL within the tibiofemoral joint.

The knee joint complex consists of two joints: the patellofemoral joint and the tibiofemoral joint. The tibiofemoral joint is the articulation between the proximal tibia and the distal femur. The patellofemoral joint is the articulation between the patella and the anterior femoral plateaus.

1.3.1 The tibiofemoral joint.

The proximal articular surface of the tibiofemoral joint is formed by large medial and lateral plateaus of the distal end of the femur (Figure 1.1 a and b). The two plateaus are separated by the intercondylar notch. The articular surfaces of the proximal tibia that correspond to the femoral articular surfaces are two shallow concave medial and lateral plateaus.

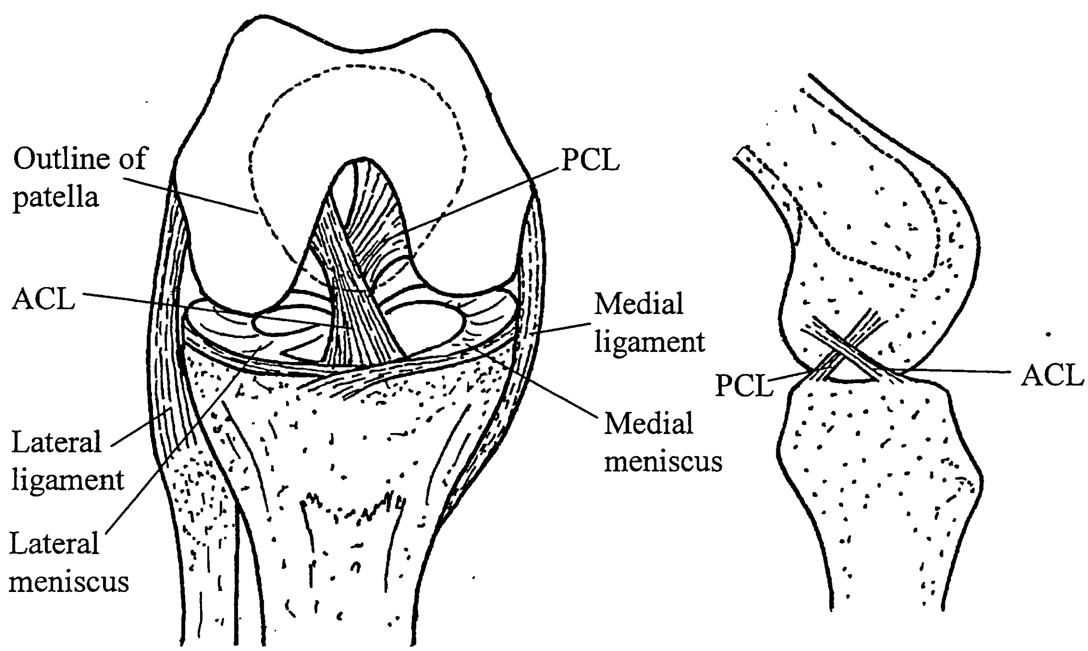


Figure 1.1 a) Anterior aspect of the right knee in flexion, b) Normal orientation of the cruciate ligaments in the partially flexed knee (adapted from Watkins 1999).

The tibiofemoral joint is stabilised by the interaction of dynamic and passive stabilisers. Dynamic stability is provided by the muscles that cross the tibiofemoral joint. These are primarily the quadriceps, hamstrings and triceps surae. Passive stability is provided by the non-contractile structures of the knee. These structures are the joint capsule, lateral and medial menisci and four extracapsular ligaments: lateral ligament, medial ligament, ACL and the posterior cruciate ligament (PCL) (Figure 1.1 a and b).

The ACL and PCL are located in the intercondylar notch and cross each other with the ACL passing laterally to the PCL. The distal end of the ACL is attached to the posterior aspect of the anterior intercondylar area of the tibial table. The proximal end is attached to the posterior medial aspect of the lateral femoral plateau. The distal end of the PCL is attached to the posterior aspect of the intercondylar area of the tibial table, and the proximal end is attached to the anterior inferior lateral aspect of the medial femoral plateau (Figure 1.1 a and b).

1.3.2 The function of the cruciate ligaments.

The function of the cruciate ligaments is to ensure normal movement between the articular surfaces of the femoral and tibial plateaus. Figure 1.2 shows a diagram of the forces acting on the proximal tibia due to the hamstrings, quadriceps, ACL and PCL when the knee is close to full extension.

From Figure 1.2, it can be seen that when the knee is close to full extension, i.e the position in which most non-contact ACL injuries occur, the ACL and the hamstrings work together to help prevent anterior dislocation of the tibia relative to the femur by resisting forward movement of the tibial plateaus (Figure 1.3a and b). Similarly, the PCL works with the quadriceps to help prevent posterior dislocation of the tibia relative to the femur by restricting backward movement of the tibial plateaus (Figure 1.3a and c). When the knee is at deeper flexion angles, the line of

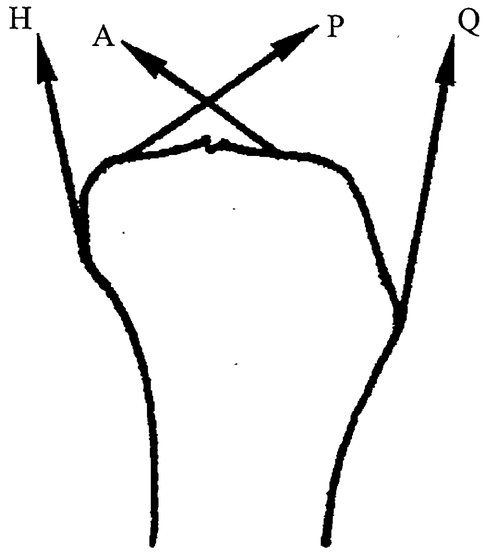


Figure 1.2 Diagram of forces acting on the proximal tibia in the sagittal plane. Forces due to the Hamstrings (H), Quadriceps (Q), Anterior cruciate ligament (A) and Posterior cruciate ligament (P).

action of the quadriceps changes so that it acts with the hamstrings and the ACL in resisting forward movement of the tibial plateaus. This may be why ACL injury is uncommon when the knee is more flexed.

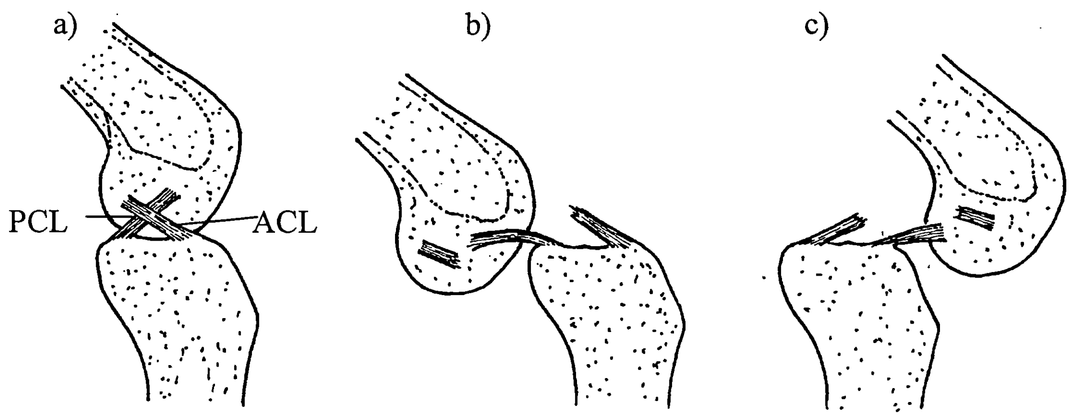


Figure 1.3 a) Normal orientation of the cruciate ligaments in the partially flexed knee; b) ACL tear due to posterior dislocation of the femur on the tibia; c) PCL tear due to anterior dislocation of the femur on the tibia (adapted from Watkins 1999).

In addition to helping to prevent posterior dislocation of the femur on the tibia (ACL) and anterior dislocation of the femur on the tibia (PCL), the cruciate ligaments also help prevent hyperextension of the tibiofemoral joint, medial and lateral displacement of the tibia relative to the femur and internal rotation of the tibia relative to the femur. At present, there is little empirical information on the extent to which landing, decelerating and cutting movements cause abnormal movement of the tibiofemoral joint and, therefore, abnormal strain on the ACL and other knee ligaments.

1.4. Risk factor models.

The risk factors associated with ACL injury in general and the gender difference in ACL injury incidence in particular have been grouped in various ways (Lysens et al., 1984; Lorenzton, 1988; Nigg, 1988; Huston et al., 2000; Griffin et al., 2000; Yu et al., 2002). One of the most frequently used categorisations of injury risk factors is intrinsic-extrinsic. Intrinsic factors are personal, physical and psychological characteristics that distinguish individuals from each other and extrinsic factors concern environmental conditions and the manner in which activities are administered (Watkins, 1999). With regard to ACL injuries, Griffin et al. (2000) categorised risk factors into three intrinsic groups (anatomical, hormonal, biomechanical) and one extrinsic group (environmental). While these categories may be of some help in identifying the possible cause of an injury, in most cases the cause of the injury is likely to be the result of a complex interaction of intrinsic and extrinsic factors (Lysens et al., 1984). The main limitation of current models of the aetiology of injury in general and ACL injury in particular is the failure to show how the various risk factors interact. A reasonable first step in trying to understand the interaction between factors would be to describe the relationship between the various factors. ACL injury is caused by excessive load on the ACL which is due to abnormal movement of the tibiofemoral joint. The latter is due to the failure of the passive and/or dynamic support mechanisms to adequately stabilise the joint; see Figure 1.4. The purpose of this chapter is to present a risk factor model for ACL injury based upon a review of the components of the passive and dynamic support mechanisms.

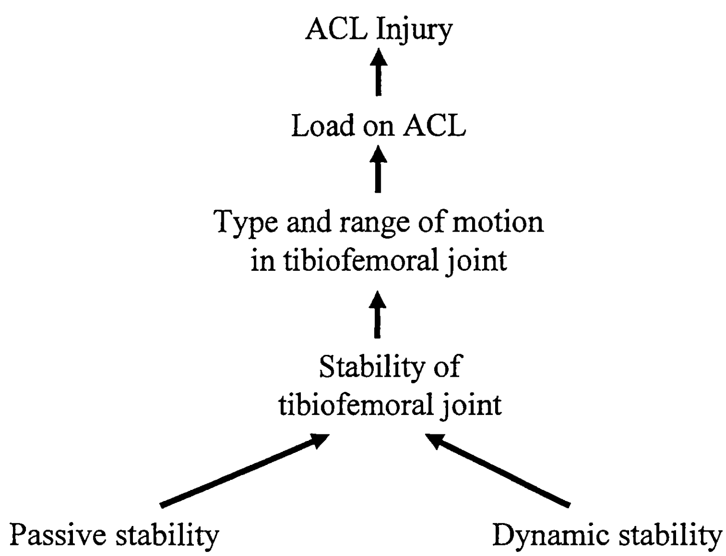


Figure 1.4. Model of ACL injury based on passive and dynamic stability of the tibiofemoral joint.

1.5 Passive stability risk factors.

As shown in Figure 1.5, the passive stability of the tibiofemoral joint depends upon the geometry of the articular surfaces and ligament laxity.

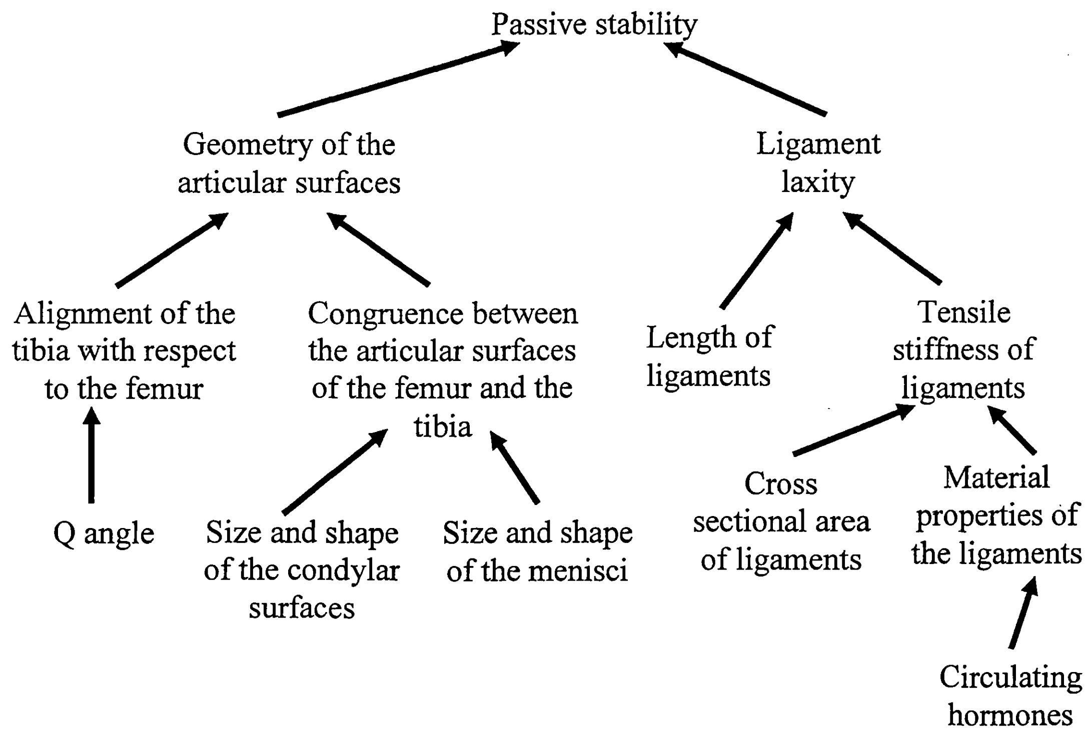


Figure 1.5. Factors affecting passive stability of the tibiofemoral joint.

1.5.1 Geometry of the articular surfaces

The geometry of the articular surfaces of the tibiofemoral joint depends upon the alignment of the femur with respect to the tibia and the congruence between the articular surfaces of the femur and tibia; see Figure 1.5.

1.5.1.1 Alignment of the femur with respect to the tibia

There is considerable evidence that the Q angle, i.e. the acute angle between the line connecting anterior superior iliac spine to the middle of the patella and the line connecting the tibial tuberosity to the centre of the patella (Hungerford and Barry, 1979) is, on average, larger in females than males (Horton and Hall, 1989; Hsu et al., 1990; Woodland and Francis, 1992; Guerra et al., 1994; Herrington and Nester, 2004); see Figure 1.6.

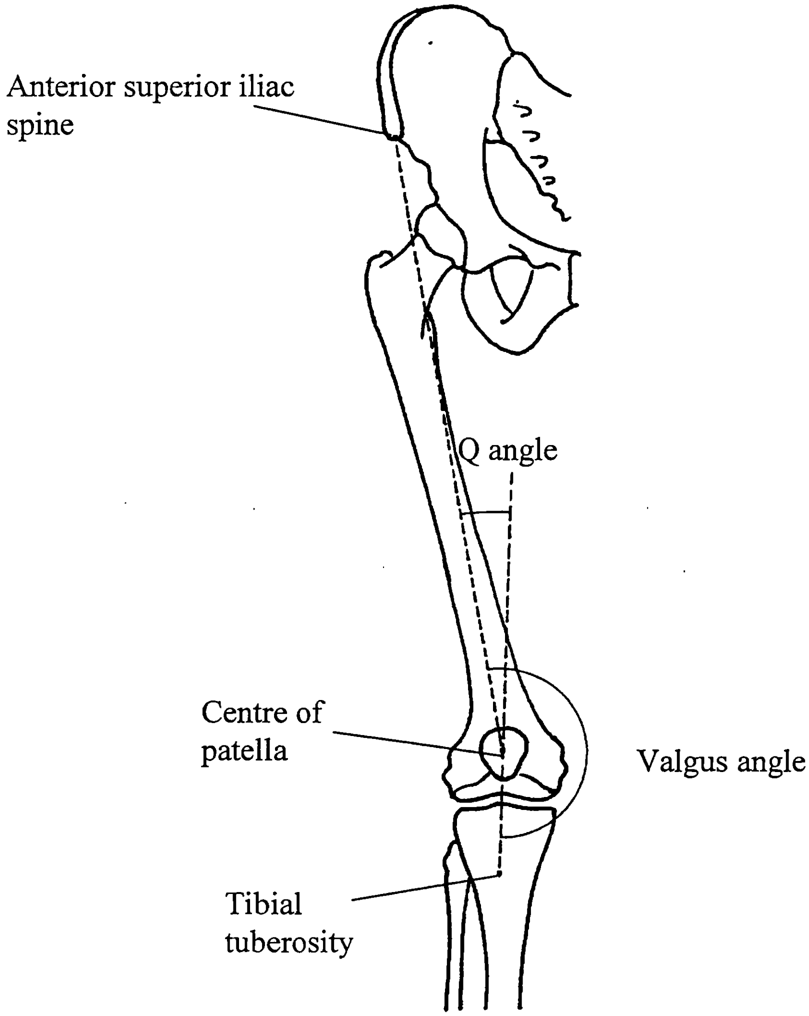


Figure 1.6 Q angle and valgus angle of the right leg (adapted from Huston et al. 2000).

For example, Herrington and Nester (2004) measured the Q angles of 51 male and 58 female physically active subjects with no history of lower limb injury. Q angle was measured with subjects standing and quadriceps relaxed. The results showed mean Q angle to be significantly greater in females (mean Q angle 13.9°) than in males (mean Q angle 11.5°). The larger the Q angle, the larger knee valgus angle;

see Figure 6. Since increased valgus angle during dynamic movement has been associated with an increased likelihood of ACL injury (Boden et al., 2000; Olsen et al., 2004), some studies have investigated the relationship between Q angle and ACL injury incidence. For example, Shambaugh et al. (1991) investigated the relationship between lower extremity alignment and knee injury in 45 recreational athletes. The results showed that athletes who had sustained knee injuries had significantly larger Q angles than the players who had not been injured (mean Q angle in knee injured 14°, non-injured 10°). However, other studies have demonstrated no apparent relationship between Q-angle and ACL injury (Gray et al., 1985) indicating little evidence that the increased Q angle in females increases the risk of ACL injury. Also, the apparent valgus angle of the knee observed when subjects are performing cutting and landing manoeuvres is likely to be different than the static Q-angle measured anatomically.

1.5.1.2 Congruence between the articular surfaces of the femur and tibia

Congruence between the articular surfaces of the femur and tibia depends upon the size and shape of the plateau surfaces of the femur and tibia and the size and shape of the menisci. The femoral plateaus (convex in sagittal plane) and tibial plateaus (shallow concave in the sagittal plane) (Figure 1.3) are not very congruent, but in the healthy knee, the congruence between the femoral and tibial plateaus is normally quite high due to the menisci; see Figure 1.1a. Damage to the menisci, especially complete radial tears, has been shown to decrease congruence and decrease passive stability of the joint (Williams et al., 1995). However, there would appear to be no empirical evidence of damaged menisci increasing the incidence of ACL injury. The congruence between the femoral and tibial plateaus is also affected by the width of the intercondylar notch (INW); the wider the notch, the lower the congruence; see Figure 1.7.

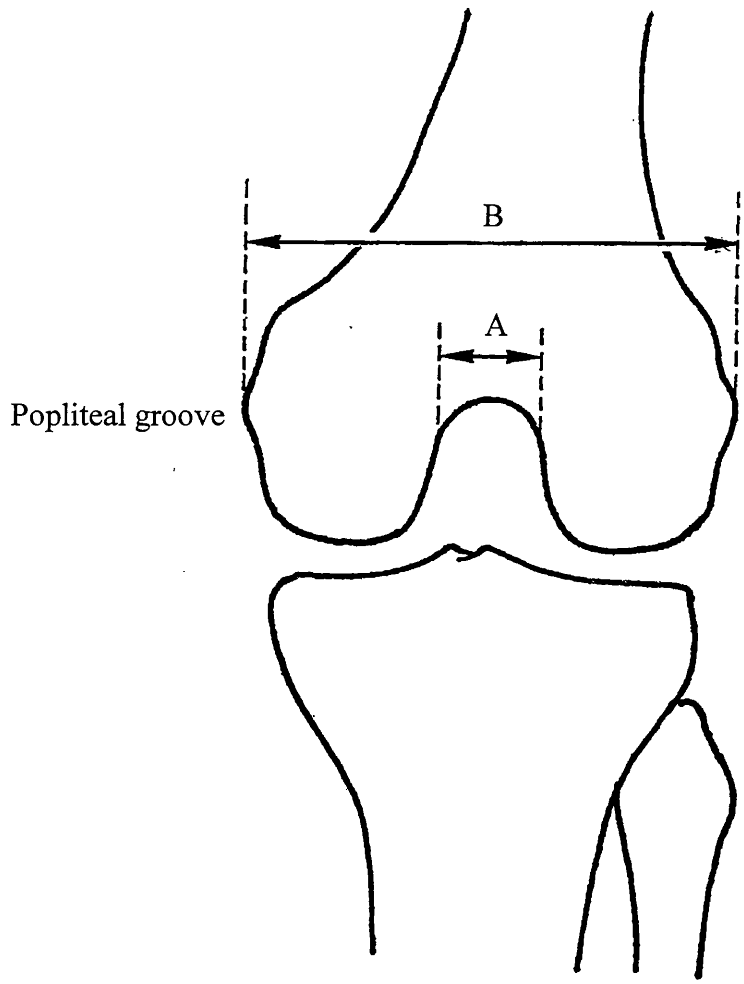


Figure 1.7 Intercondylar notch width (A) and notch width index (A/B) (adapted from Huston et al. 2000).

However, the narrower the notch, the smaller the space available for the cruciate ligaments. Some studies have reported that females have a smaller INW (Shelbourne et al., 1986; Charlton et al., 2002; Uhorchak et al., 2003) and NWI (ratio of INW to width of femoral epicondyles) (Souryal et al., 1988) than males. Also, a number of studies have reported greater incidence of unilateral and bilateral ACL injury in subjects with smaller INW (Shelbourne et al., 1986; Lund-Hanssen et al., 1994; Ireland et al., 2001; Uhorchak et al., 2003) and NWI (Souryal et al., 1988; Ireland et al., 2001). For example, Uhorchak et al. (2003) carried out a four-year study of 711 male and 113 female army cadets. INW was measured at the start of the four-year period via radiographic assessment using digital callipers. During the four-year period 24 non-contact ACL injuries occurred, 16 to males and 8 to females. The results showed INW to be significantly smaller in females (mean INW 15.6 mm) compared to males (mean INW 17.7 mm) and INW to be significantly smaller in non-contact ACL injured subjects (mean INW 13.8 mm) than in non-injured subjects (mean INW 17.5 mm). Also, Shelbourne et al. (1986) investigated the relationship of the INW with bilateral ACL injury incidence and the rupture rate of the reconstructed ACL. This was to identify whether subjects who had a narrow INW were more likely to rupture either the reconstructed ACL or the ACL of the other knee than those subjects with a wide INW. 714 subjects who underwent patella tendon graft ACL reconstruction were investigated and divided into narrow (<15 mm) and wide (>16 mm) INW groups. The findings showed females to have narrower INW than males of the same height and the incidence of bilateral rupture after initial rupture of the other ACL was found to be higher for the narrow INW group than for the wide INW group. However, the findings showed no difference between males and females for contralateral ACL injury incidence and no difference in rupture rate of the reconstructed ACL between the narrow and wide groups. Also, other studies have not found an association between ACL injury and INW or NWI (Schichendantz and Weiker, 1993; LaPrade and Burnett, 1994; Teitz et al., 1997) and there is no clear evidence that differences in INW or NWI influence the incidence of ACL injury.

1.5.2 Ligament laxity.

Joint laxity refers to the degree of instability in a joint, i.e. the range of movement in directions that are considered abnormal for that joint when no muscles are active (Watkins, 1999). Several studies have reported that females exhibit greater joint laxity than males (Hutchinson and Ireland, 1995; Huston and Wojtys, 1996) and knee joint laxity has been proposed as potentially contributing to the greater incidence of ACL injuries in females (Gray et al., 1985; Hutchinson and Ireland, 1995; Uhorchak et al., 2003). It is not clear what causes differences in joint laxity in healthy knees, but it is reasonable to assume that it is due to differences in the length and tensile stiffness (resistance to stretching) of ligaments; see Figure 1.5. The longer the ligaments and the lower the tensile stiffness, the greater the laxity. The tensile stiffness of a ligament will depend upon the cross sectional area and the effect of circulating hormones on the material properties of the ligament. Some studies have reported that the cross sectional area of the ACL is relatively larger in males than in females (Muneta et al., 1985; Charlton et al., 2002). For example, Charlton et al. (2002) used magnetic resonance imaging to measure ACL volume in 98 knees of males and females and found that ACL volume was significantly relatively smaller in females than males. However, there is no evidence that cross sectional area of the ACL influences the incidence of ACL injury. The effect of hormones on the material properties of the ACL has been identified as a potential cause of the greater incidence of ACL injuries in females. Due to the identification of oestrogen and progesterone receptors in the human ACL (Liu et al., 1996), some studies have investigated the effects of the variations in the levels of female sex hormones on the material properties of ligaments (Liu et al., 1997; Slauterbeck et al., 1999; Yu et al., 1999). For example, Slauterbeck et al. (1999) reported that the administration of oestrogen significantly decreased the tensile strength of the ACL in rabbits. These findings suggest that the levels of female sex hormones may affect the strength of the ACL which may provide some explanation for the greater incidence of ACL injury in females. Consequently, several studies have investigated the time of occurrence of non-contact ACL injury in females in relation to the phase of the menstrual cycle (Wojtys et al., 1998; Wojtys et al., 2002; Mykelbust et al., 1997; Slauterbeck et al., 2002). However, the findings are inconsistent, see Figure 1.8. Some studies reported significantly higher incidence of ACL injury between days 10-14 (Wojtys et al., 1998; Wojtys et al., 2002), whereas others reported significantly higher incidence during days 1-2 of the menstrual cycle (Slauterbeck et al., 2002). Also, days of significantly lower incidence of ACL injury have been reported between days 1-9 (Wojtys et al., 1998), days 8-14 (Mykelbust et al., 1997) and days 15-28 of the menstrual cycle (Wojtys et al., 2002).

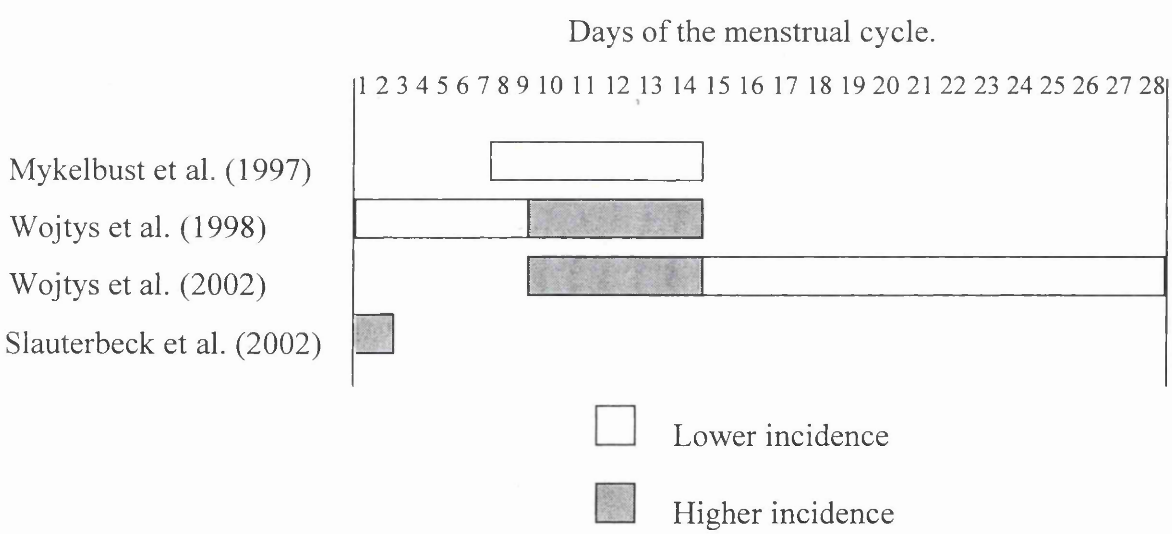


Figure 1.8. Reported days of significantly higher and lower incidence of ACL injury.

The inconsistency of these findings may be attributable to the relatively small samples used. However, on current evidence, the influence of changes in hormone concentrations on the incidence of ACL injuries in females is not clear.

1.6 Dynamic stability risk factors.

During dynamic movements such as landing and cutting (side-stepping), dynamic stability in the form of muscle activity is necessary to provide adequate joint stability. Figure 1.9 shows the factors that affect muscle function and, therefore, the dynamic stability of the tibiofemoral joint: the patellar tendon-tibia shaft angle, muscle activity pattern (in terms of net joint torque), muscle reaction time, time to peak torque and muscle stiffness.

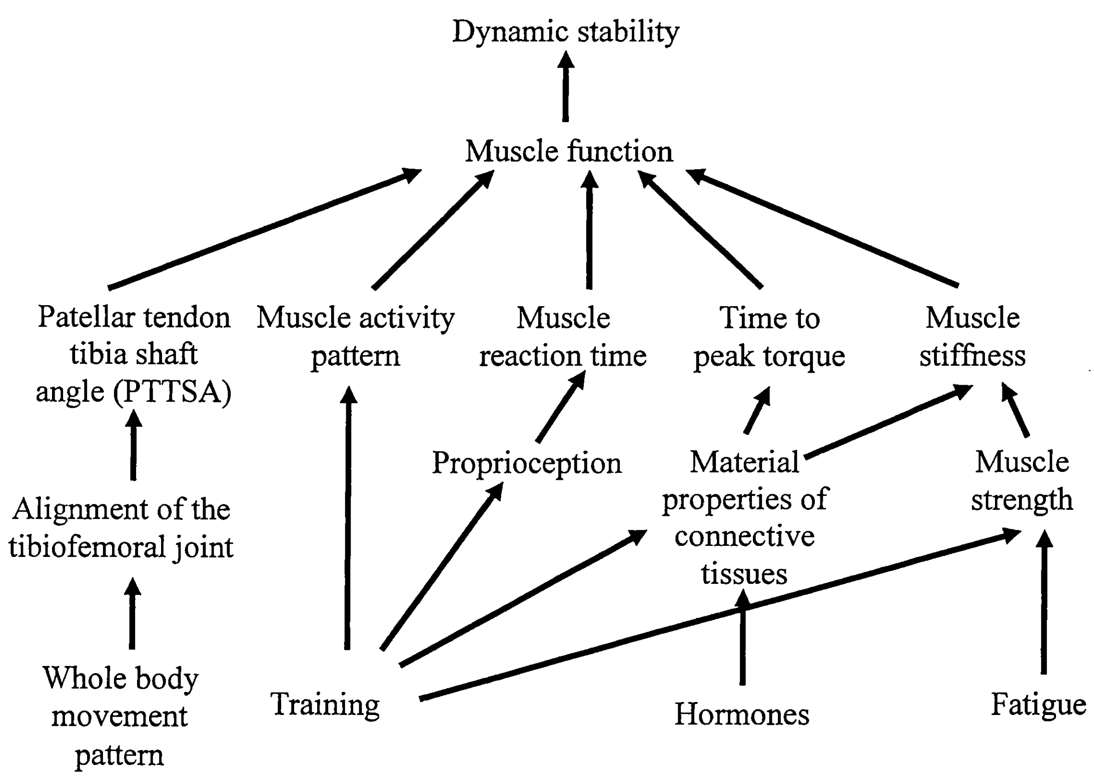


Figure 1.9 Factors that affect dynamic stability of the tibiofemoral joint.

1.6.1. Patellar tendon – tibia shaft angle.

The patella tendon-tibia shaft angle (PTTSA) is the angle (in the sagittal plane) between the long axis of the tibia and the line of action of the patellar ligament (Figure 1.10). When the knee is close to full extension, contraction of the quadriceps, which acts through the patellar tendon, causes an anterior shear force applied to the proximal end of the tibia (ASPT). The greater the PTTSA, the greater the ASPT and the greater the potential strain on the ACL.

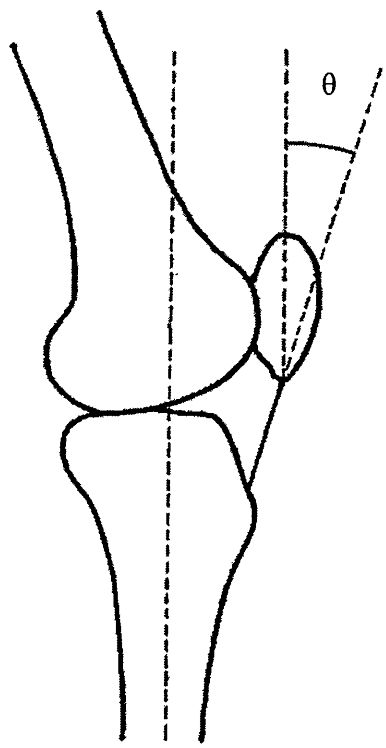


Figure 1.10 Patella tendon-tibia shaft angle (θ) (adapted from Nunley et al. 2003).

Nunley et al. (2003) investigated the effect of knee angle on the PTTSA. Regression analysis showed that the PTTSA decreased linearly in both males and females from 0° of knee flexion to 90° of knee flexion (Table 1.1), i.e. the PTTSA and, therefore, the ASPT was largest when the knee was close to full extension. A number of studies including Boden et al. (2000) and Olsen et al. (2004) have reported that non-contact ACL injury appears to occur more frequently when the knee is close to full extension than when flexed. Nunley et al. (2003) also showed that the PTTSA was, on average, 3.6° greater in females than in males over the 0°-90° range of knee flexion, i.e. for a given angle of knee flexion, the ASPT is likely to be 13.0% greater in females than in males over the 0°-90° range of knee flexion. Therefore, the PTTSA may contribute a greater risk of ACL injury in females than in males.

Table 1.1 Mean and range of patellar tendon-tibia shaft angle (PTTSA) in male and female recreational athletes over the 0°-90° range of knee flexion (from Nunley et al. 2003).

	Mean*	Knee Flexion	Range of PTTSA
Males	22.0°	0°	12.2° to 27.8°
		90°	-11.3° to -0.1°
Females	25.7°	0°	13.3° to 34.8°
		90°	2.1° to 5.4°

*Mean patellar tendon-tibia shaft angle over the 0°-90° range of knee flexion.

1.6.2 Whole body movement pattern

The patellar tendon-tibia shaft angle depends upon the knee angle which, in turn, depends upon the whole body movement pattern. Kinematic and kinetic analyses of drop-jumps, stop-jumps and cutting movements have been undertaken in order to try to identify movement characteristics that may account for gender differences in the incidence of ACL injuries. A summary of these studies can be seen in Table 1.2. In general, the studies indicate that in landing and cutting manoeuvres, females tend to land with less knee flexion (Malinzak et al., 2001; Chappell et al., 2002;

Table 1.2 Summary of recent studies that investigated the kinematics and kinetics of landing and cutting manoeuvres in males and females.

	Subjects	Level	Task	Results
Salci et al. (2004)	8F-8M	University volleyball	Vertical landing	Females displayed reduced knee and hip flexion angles at ground contact, greater peak knee extension moment and greater normalised ground reaction force
Decker et al. (2003)	9F-12M	Recreational volleyball and basketball	Vertical landing	Females had reduced knee flexion at ground contact, greater range of motion and greater peak angular velocities of ankle, hip and knee in sagittal plane. Females showed greater energy absorption and peak powers in the knee extensors and ankle plantar flexors
Chappell et al. (2002)	10F-10M	Recreational	Forward, backward and vertical jump landing	Females landed with reduced knee flexion and had greater proximal tibia anterior shear force, greater knee extension moments and greater knee valgus moments
Ford et al. (2003)	47F-34M	High school basketball	Vertical drop Jump	Increased knee valgus motion and maximum angle in females
Malinzak et al. (2001)	9F-11M	Recreational	Cutting	Females displayed reduced flexion, greater knee valgus angle and greater quad activation
James et al. (2004)	19M-19F	High school and collegiate basketball	Cutting	Females displayed greater ground reaction force at maximum knee flexion, reduced knee flexion at ground contact and greater range of knee flexion during stance phase

M = males, F = females

Salci et al., 2004; James et al., 2004; Decker et al., 2003), exhibit greater knee valgus (Malinzak et al., 2001; Ford et al., 2003), display greater peak knee extension moments (Chappell et al., 2002; Salci et al., 2004) and produce greater normalised ground reaction force (Salci et al., 2004; James et al., 2004) when compared to males. As described earlier, (section 1.1) these types of movement are likely to increase the load on the ACL. These findings tend to indicate differences in landing and cutting movement patterns between males and females. However, lack of appropriate standardisation in task demands may have invalidated meaningful comparison between females and males. For example, dropping down from a raised platform set at the same height for both males and females (Salci et al., 2004; Decker et al., 2003; Ford et al., 2003) may result in significantly different task demands.

1.6.3 Muscle activity pattern

When providing dynamic stability of the knee, the activity of the knee flexors and extensors should, ideally, result in a zero shear load on the proximal tibia and, therefore, minimal strain on the knee ligaments. However, if the shear load exerted by the quadriceps is greater than the shear load exerted by the hamstrings, a resultant anteriorly directed shear force may be exerted on the proximal end of the tibia, which will increase ACL strain (Figure 1.11). This is known as quadriceps dominance and is defined as an imbalance between the recruitment patterns of the knee flexors and extensors (Ford et al., 2003). A number of studies have found that females exhibit greater quadriceps dominance than males in activities associated with ACL injury (Huston and Wojtys, 1996; Malinzak et al., 2001; Zeller et al., 2003).

Malinzak et al. (2001) investigated 3D knee joint motion and EMG activities of the hamstrings and quadriceps in 9 female and 11 male recreational athletes while running, cross-cutting and side-cutting. Surface electromyographic (EMG) signals were recorded during maximum voluntary contraction (MVC) tests for vastus lateralis and vastus medialis oblique and for the biceps femoris and medial hamstrings so that EMG data could be expressed as a percentage of the EMG recorded for MVC. EMG data indicated that females had greater quadriceps activation (17-40% MVC greater) yet reduced hamstring activation (20% MVC less) than males. Huston and Wojtys (1996) investigated knee activation patterns

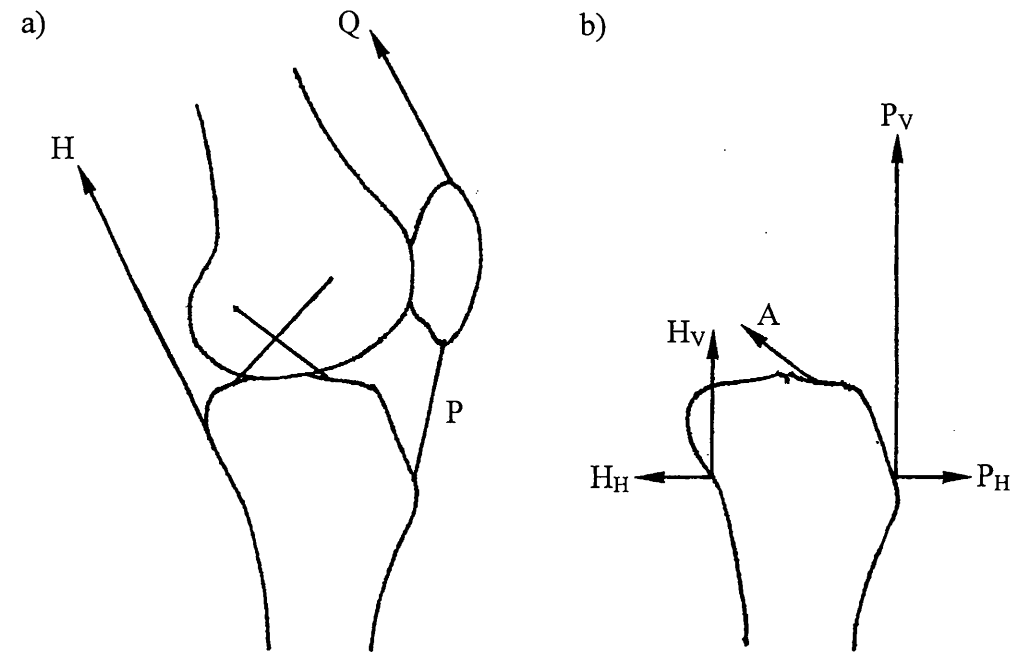


Figure 1.11 Forces exerted by the quadriceps and hamstrings on the tibia. a) relationship of the hamstrings (H), quadriceps (Q) and patellar ligament (P) to the anterior cruciate ligament (A). b) components of H and P in relation to A: if P_H is greater than H_H, the ACL will be put under strain.

in response to anterior tibial translation in elite basketball and volleyball players and non athletes. Female athletes tended to respond to anterior tibial displacement of the knee by firstly contracting the quadriceps, whereas male athletes and male and female non-athletes responded by firstly contracting the hamstrings. Zeller et al. (2003) investigated the kinematics and EMG activity of the quadriceps and hamstrings of 9 male and 9 female collegiate athletes while performing a one-legged squat using the dominant limb. Analysis of all muscles tested showed females to display greater total muscle activation than males. Analysis of each muscle independently showed females to produce significantly greater total and maximal activation of the rectus femoris compared to males.

The findings of these studies (Huston and Wojtys, 1996; Malinzak et al., 2001; Zeller et al., 2003) suggest that, compared to males, females tend to exhibit a quadriceps dominant mode of producing dynamic joint stability of the knee which may increase the risk of ACL injury. However, there is little empirical information and further research is necessary to clarify this possible gender difference.

1.6.4 Muscle reaction time and time to peak torque

Muscle reaction time (RT) and time to peak torque (TPT) clearly affect the speed with which dynamic stability can be achieved and, as such, may affect the likelihood of joint injury (Beard et al., 1994). Cowling and Steele (2001) investigated the effect of gender on lower limb muscle synchrony during landing in 7 male and 11 female subjects. The task involved subjects accelerating forwards three steps to receive a netball chest pass from 3 m directly in front and then land on their dominant limb on a force platform in single limb stance. EMG data was recorded, from the rectus femoris, vastus medialis, vastus lateralis, semimembranosus, biceps femoris and the medial head of the gastrocnemius muscles. The main findings were that males exhibited a delayed semimembranosus onset relative to ground contact (Males 113 ± 46 ms, females 173 ± 54 ms) and delayed peak activity relative to peak tibiofemoral shear force (Males 54 ± 27 ms, females 77 ± 15 ms) compared to females. It was suggested that the delayed onset of semimembranosus activity in males allows peak semimembranosus activity to better coincide with high anterior force exerted on the proximal tibia by the quadriceps thereby acting as an ACL synergist via increased joint compression and posterior tibial drawer, reducing the chance of ACL injury.

Wojtys et al. (1996a) investigated the effect of three types of training on RT and TPT in 16 male and 16 female healthy subjects. Subjects were placed into one of four groups; isokinetic, isotonic, agility and control. The three training groups exercised for 30 minutes, three times a week for 6 weeks, whereas the control group participated only in activities they were involved in prior to the study. RT was assessed as the muscular response to an anteriorly directed 133.42 N (30-pound) force applied to the posterior aspect of the tibia, with the subject sitting with the knee flexed at 30°. Surface EMG was recorded for the gastrocnemius, lateral hamstring, medial hamstring, lateral quadriceps and medial quadriceps muscles. The readings from these muscles were then used to calculate three components of RT; spinal reflex (between 26 to 130 ms after tibial displacement), intermediate response (between 110 to 216 ms after tibial displacement), and voluntary muscle activity (between 156 to 431 ms after tibial displacement). The findings showed that the agility exercises resulted in a significant reduction in all three components of RT. Isokinetic training resulted in a significant decrease in the voluntary muscle activity component of RT. No significant improvements in any of the components of reaction time were seen as a result of isotonic training. All groups showed a reduction in TPT in knee extension isokinetic strength test, with the agility group showing the biggest reduction (39 ms). For knee flexion, the agility group again showed the greatest reduction (38 ms) after training. For ankle plantar flexion during isokinetic strength testing, TPT reduced by 15 ms for the isokinetic group, with the other two training groups showing an increase. The findings of this study show that agility and isokinetic training may reduce anterior tibial translation and therefore reduce the risk of ACL injury.

1.6.5 Muscle stiffness and muscle strength

As the quadriceps and hamstring muscles contract they act in a way to increase the joint contact forces and limit shear movement within the tibiofemoral joint. The ability of the muscles to resist movement within the tibiofemoral joint (maintain a particular joint angle) refers to muscle stiffness. The greater the ability of the muscles of the knee to prevent tibiofemoral shear movement, the less likely the passive structures of the knee, such as the ACL, will be put under strain. Therefore, muscle stiffness may be an important factor in preventing ACL injury. Due to this, a number of studies have investigated the muscular stiffness of males and females to determine its link to the gender difference in ACL injury incidence (Granata et al., 2002a; Granata et al., 2002b; Wojtys et al., 2003).

Granata et al. (2002a) investigated the muscle stiffness of the quadriceps and hamstrings in healthy subjects (12 male, 11 female). Subjects were required to support their lower leg at 45° in two positions. The first required the hamstring to support the weight of the leg and the other required the quadriceps to support the weight of the leg. Weights of 0 kg, 6 kg and a weight corresponding to 20% of maximum voluntary exertion were added to a fixed ankle-foot orthosis. A sudden transient perturbation was then applied to the lower leg and the resulting knee flexion-extension motion recorded by an accelerometer attached to the heel of the orthosis. Subjects were asked not to attempt to control the natural motion of the tibia after the perturbation and to maintain constant muscle activity, which was measured by EMG electrodes on the quadriceps and hamstrings. The flexion-extension oscillations of the knee were then used to calculate muscle stiffness. After accounting for differences in applied moment resulting from leg mass, females demonstrated reduced muscle stiffness for the quadriceps and the hamstrings for all torque levels compared to males, with results showing females' stiffness to be between 55.8 to 73.9% of males. Granata et al. (2002b) continued the work of the previous study to evaluate whether females also demonstrated reduced leg stiffness in functional tasks, such as two-legged hopping, compared to males. Fifteen male and fifteen female healthy subjects took part in the study. Subjects were required to perform repeated two-legged hopping on a force platform. Subjects were required to hop at three different hopping frequencies of 2.5 Hz, 3.0 Hz and preferred frequency. Leg stiffness was found to be significantly greater in males than females at all hopping frequencies with females exhibiting 73-81% of the leg stiffness of males. Wojtys et al. (2003) investigated the hypothesis that females are less able to volitionally increase the apparent torsional stiffness of the knee by maximally activating the knee muscles. Torsional stiffness is the ability of the muscles of the knee to prevent rotation of the knee joint. There were two groups of subjects. The first group consisted of twelve male and twelve female National Collegiate Athletic Association Division 1 athletes competing in basketball, volleyball and soccer. The second group of subjects was fourteen male and fourteen female collegiate endurance athletes competing in cycling, crew and running. Testing was performed using a weighted pendulum which applied an 80 N force, directed medially, to the lateral aspect of the right forefoot. The resulting internal rotation was measured optically. Trials were carried out with and without maximal activation of the quadriceps and hamstring. The maximal rotations of the leg were greater in females than males for trials with (27% greater) and without (16% greater) activation of the knee muscles. Also, females displayed significantly smaller (18%) volitional increase in torsional stiffness of the knee than males. This finding was particularly evident in those athletes in group 1, where there was a 42% difference between genders.

The findings of the studies examining the muscle stiffness of the knee suggest that females exhibit less muscular protection of the knee than males when the knee is externally loaded. This suggests that gender differences in the muscular stiffness of the knee may account, at least in part, for the greater incidence of ACL injury in females. However, further investigation into the apparent variation in the stiffness of the knee between males and females of varying athletic levels is needed to validate this.

One factor that may influence muscle stiffness is muscle strength. A number of studies have reported significantly lower muscular strength of the hamstrings and the quadriceps in females compared to males, even when normalised to body weight (Huston and Wojtys, 1996; Kanehisa et al., 1996; Hakkinen et al., 1997; Salci et al., 2004). This suggests that the lower levels of muscle stiffness exhibited by females compared to males may be due, at least in part, to lower levels of muscle strength (absolute and relative). Lower levels of strength may increase the risk of ACL injury.

1.6.6 Effects of fatigue on dynamic stability

Fatigue has been suggested as a risk factor for non-contact ACL injury (Chappell et al., 2001; Fagenbaum and Darling, 2003) due to its affect on lower extremity muscle activity. However, the findings are conflicting. Chappell et al. (2001) investigated the effects of fatigue on knee joint kinetics and kinematics of 10 male and 10 female recreational athletes whilst performing drop-jump tasks. The subjects performed three drop-jump tasks before and after undertaking repeated 30 m sprints and vertical jumps to fatigue the subjects' lower limbs. The findings showed that both males and females significantly increased the peak proximal tibia anterior shear force and decreased knee flexion on landing when fatigued compared to non-fatigued. Also, after fatigue, males displayed decreased knee varus moment, whereas females showed increased knee valgus moment. Fagenbaum and Darling (2003) investigated the effect of fatigue on knee kinematics and muscle activation during jump landings for 6 male and 8 female varsity basketball players. Three landing tasks were performed, one where subjects jumped from both feet as high as possible and landed on the dominant leg, one where subjects jumped down from a 25.4 cm high platform and landed on the dominant leg, and finally where subjects jumped down from a 50.8 cm high platform and landed on the dominant leg. Each of these tasks were performed before and after subjects were fatigued by using an isokinetic dynamometer. Subjects undertook knee flexion-extension exercises on the dynamometer until they were fatigued to two levels. These were when subjects could no longer produce 50% and then 25% of maximum extensor torque. EMG data for the hamstrings and quadriceps was collected along with knee flexion angle which was measured using a potentiometer incorporated into two freely pivoting orthoplast lever arms placed on the lateral aspect of the dominant knee. The findings showed that females exhibited greater knee flexion angle at ground contact than males. Also, females displayed greater knee flexion accelerations as a result of ground contact than males, contradictory to a number of other studies (Malinzak et al., 2001; Chappell et al., 2002; Salci et al., 2004; James et al., 2004; Decker et al., 2003). These findings were shown for all jumps and all fatigue levels. Quadriceps and hamstring activities were similar in both males and females. Wojtys et al.

(1996b) investigated the effect of hamstring and quadriceps muscle fatigue on anterior tibial translation and muscle reaction time in six male and four female healthy subjects. Subjects undertook knee examination, arthrometer measurements of tibial translation, subjective functional assessment and an anterior tibial translation stress test before and after fatiguing exercise. The muscle recruitment patterns of the hamstrings, quadriceps and gastrocnemius demonstrated no change between fatigued and non-fatigued tests of anterior tibial translation. However, the results showed a 32.5% increase in anterior tibial translation for the fatigued test compared to the non-fatigued test. Muscle responses originating in the spinal cord and cortical level exhibited significant slowing after fatigue of the hamstrings and quadriceps. These findings suggest that muscle fatigue alters the neuromuscular response to anterior tibial translation which, in turn may reduce the dynamic stability of the knee. In general, the results of studies on the effect of fatigue on lower extremity kinematics and kinetics suggest that fatigue may increase the risk of ACL injury. However, there is currently no evidence to suggest that fatigue has a greater effect on the incidence of ACL injury in females compared to males

1.7 Composite model.

Figure 1.12 shows a composite model of the passive and dynamic risk factors that affect the stability of the tibiofemoral joint. It is reasonable to assume that the apparent greater incidence of ACL injury in females compared to males is due to the gender differences with regard to some or all of the passive and dynamic stability risk factors. The only evidence (uni-variate correlation based on small samples) in support of gender differences with regard to some risk factors, such as Q angle, joint laxity, INW, NWI, ACL cross sectional area and changes in

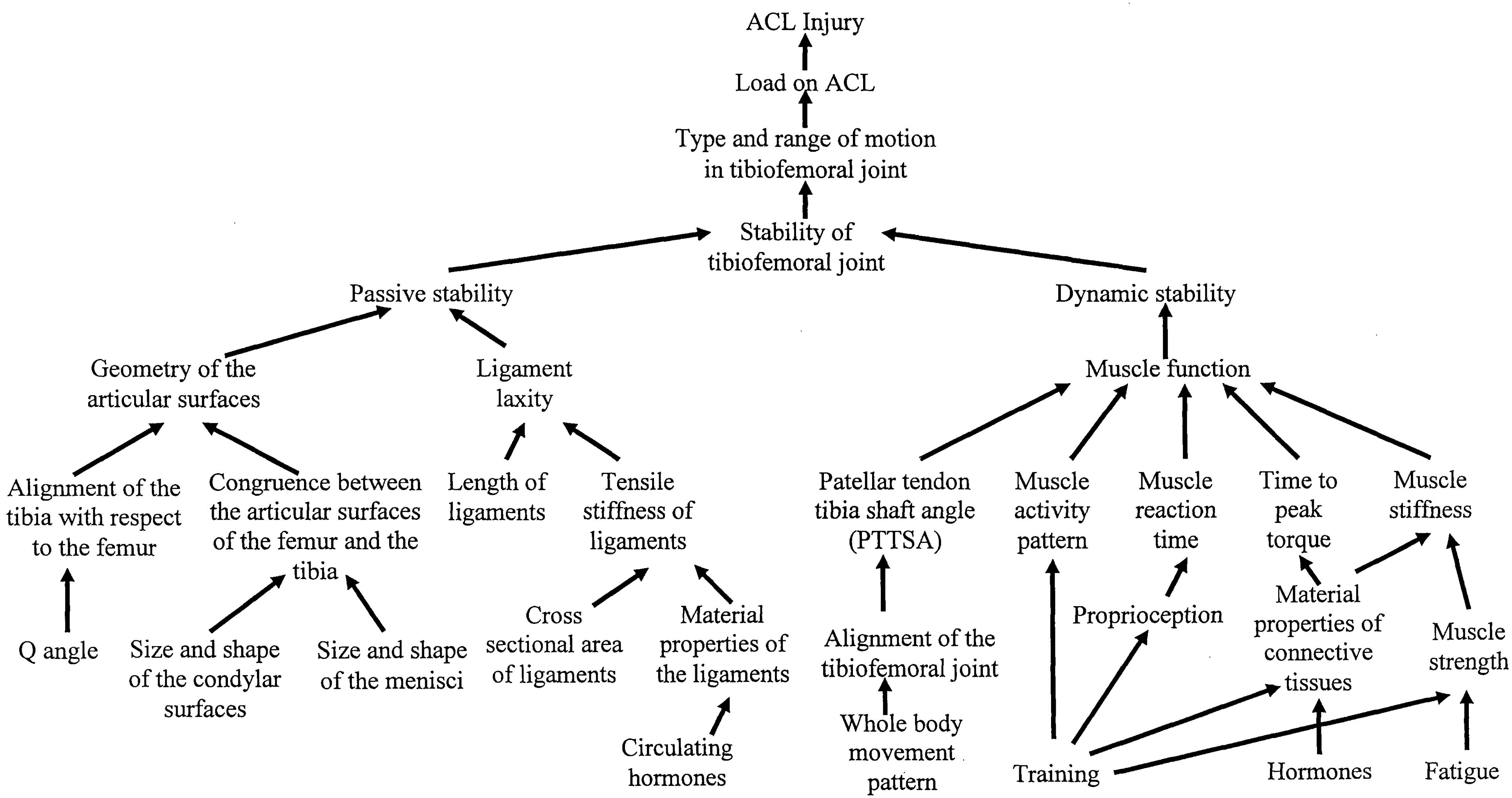


Figure 1.12 Proposed risk factor model of ACL injury based on passive and dynamic stability of the tibiofemoral joint.

concentration of circulating hormones, is fairly weak. However, the evidence in support of gender differences is much stronger with regard to some of the dynamic stability factors, especially PTTSA, muscle activity pattern, time to peak torque and muscle stiffness.

1.8 Rationale for present study.

ACL injury is a common injury which frequently occurs in sport. ACL injuries occur most frequently in non-contact situations with the athlete in single foot stance, forceful valgus knee collapse with knee close to full extension accompanied with external or internal rotation of the tibia. The situation appears to occur in two particular types of movement; landing from a jump and rapid change of direction initiated on one leg, such as a side-cutting and cross-cutting manoeuvre. ACL injury occurs as a result of a lack of stability provided by the dynamic and passive stability mechanisms of the knee. Current evidence suggests that the greater incidence of ACL injury in females is due to gender differences in the dynamic stabilising structures rather than the passive stability structures. Consequently, future research should investigate these factors and, in particular, the interaction between the factors that affect dynamic stability of the knee.

The research indicates that gender differences in landing/cutting kinematics and kinetics appear to account, at least in part, for the gender differences in the incidence of non-contact ACL injury. However, previous studies investigating gender differences in knee kinematics only report absolute angular displacement data with no reference to possible gender differences in lower limb alignment during normal upright standing. Also, previous studies report gender differences in knee kinematics and kinetics during unopposed landing/cutting manoeuvres, but there would appear to be no reported data on the effect of opposition on landing kinematics and kinetics. Finally, since the cause of ACL injury is likely to be the result of a complex interaction of risk factors (Lysens et al., 1984), composite variables, including measures of coordination and stiffness, may provide more insight into the causes of ACL injury in general and the greater incidence of ACL injury in females in particular. The lower leg stiffness in females reported by Granata et al. (2002b) will be reflected in lower stiffness in one or more of the hip, knee and ankle joints. Therefore, the higher incidence of ACL injury in females may be associated with lower knee joint stiffness. However, no previous investigation has been made into the gender differences in lower limb coordination or absolute and normalised leg and knee stiffness during landing/cutting manoeuvres. The purpose of the current project is to investigate landing biomechanics in males and females in order to identify possible risk factors associated with the gender difference in the incidence of non-contact ACL injury. The experimental phase of the project shall consist of three studies:

1. A kinematic (absolute and relative to lower limb alignment during upright standing) and kinetic (relative to body weight and/or height) analysis of male and female volleyball players (sport in which ACL injury is common) during landing from opposed volleyball block jumps.
2. A kinematic and kinetic analysis of male and female volleyball players during landing from unopposed and opposed volleyball block jumps.
3. Examination of lower limb coordination and stiffness in male and female volleyball players when landing from opposed volleyball block jumps.

Chapter 2

Reliability

Reliability of the Vicon 512 system and the lower body plug-in gait marker set.

2.1 Introduction.

Within any motion analysis system there is some degree of systematic error (noise) due to the instrumentation and software used which will affect the reliability of the measurements. The purpose of this study was to examine the reliability of the Vicon 512 system by comparing the noise (reflected in the variation of the position of markers and calculated joint centres) inherent in two systems: Vicon – laboratory environment (VL) and Vicon – laboratory environment – human subject (VLS). As in all actions of the human body, static balance is maintained by constant monitoring of body position by various proprioceptors which results in constant adjustment towards a mean posture (Roberts, 1995). Due to the inevitable minor movements of the body resulting from maintenance of static balance, it was hypothesised that the noise in the VLS system would be greater than in the VL system. The difference in the average noise levels between the two systems would indicate the noise variation associated with normal static balance in the human body.

2.1.1 Objectives.

The objectives of this study were:

1. To determine the level of noise in the 2 systems as reflected in the variation in:

- The 3D co-ordinates of markers,
- The distance between markers,
- The angles formed by 3 markers,
- The centre of pressure (CP),

2. To compare the noise levels of the systems with regard to the four measurements (coordinates, distance, angle, CP) in order to establish baseline variation between the VL and VLS systems.

2.2 Method.

2.2.1 Measurement system.

A Vicon 512 system (Vicon, Oxford, England) sampling at 120 Hz was used to determine 3D coordinates of retro-reflective markers (25 mm diameter) and the distance and angles derived from the coordinate data. Figure 2.1 shows the global coordinate system of the testing laboratory. An Advanced Mechanical Technology Incorporated (AMTI) force plate sampling at 600 Hz was used to determine CP. The coordinate system and dimensions of the force plate are shown in Figure 2.2.

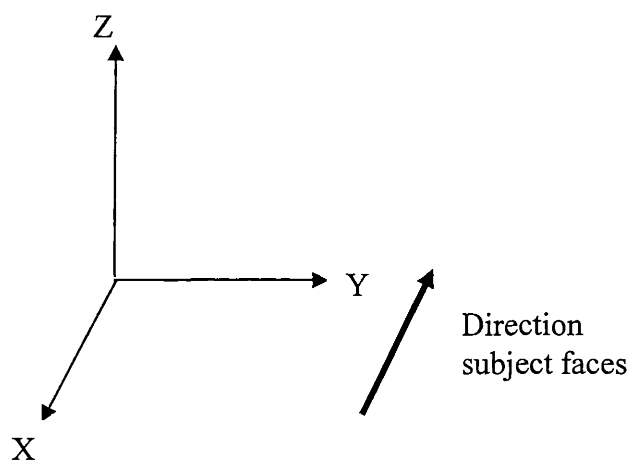


Figure 2.1 Global coordinate system of the testing laboratory.

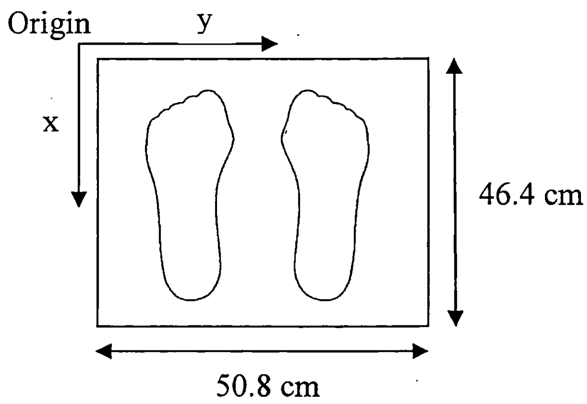


Figure 2.2 Coordinate system and dimensions of the force plate used to measure CP.

2.2.2 Inanimate static structure.

Markers were placed on an inanimate static structure which was then placed on the centre of the force plate. The dimensions of the inanimate static structure are shown in Figure 2.3. Three trials lasting 5 seconds were recorded. The inanimate static structure was not moved between trials. Means, standard deviations and coefficients of variation of the x, y, and z coordinates of markers, distances between markers, acute angle between the line made by markers 1 and 2 and the extended line made by markers 2 and 3 and the location of the CP were then calculated for each trial over a one second period (4-5 s).

2.2.3 Human subjects.

Two physically active male subjects were used in the study. Subject 1 was aged 26, mass 83 kg and height 187 cm. Subject 2 was aged 22, mass 75 kg and height 174 cm.

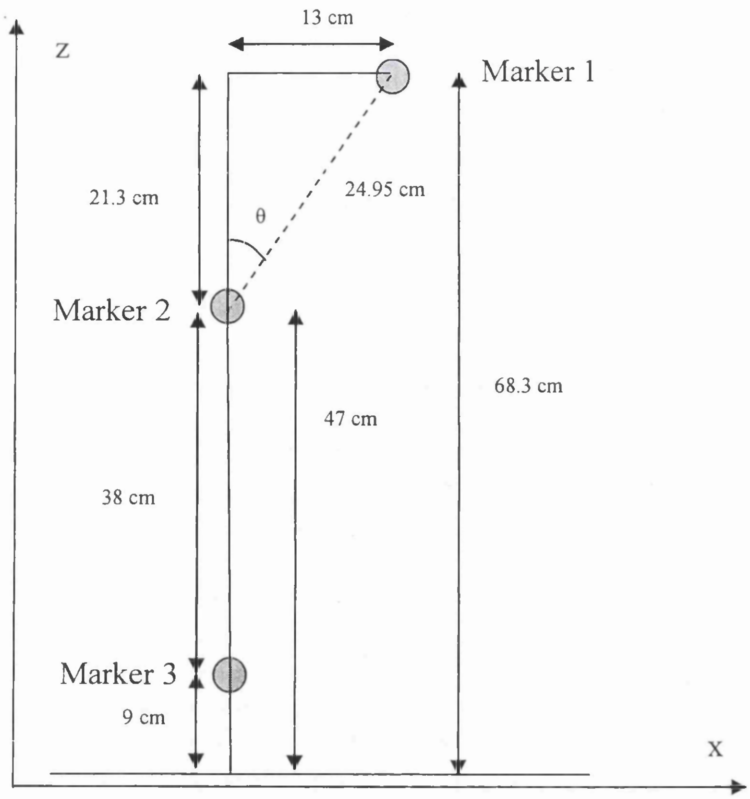


Figure 2.3 Dimensions of the inanimate static structure and angle measured (θ) (Measurements taken from the centre of markers).

2.2.3.1 Marker placement.

Markers were placed directly on the skin in accordance with the Vicon system's lower body plug-in gait marker set; see Figure 2.4.

- Anterior superior iliac spines (LASI and RASI),
- Posterior superior iliac spines (LPSI and RPSI),
- Lower lateral surface of each thigh along the line between the hip and knee joint markers (LTHI and RTHI),
- Lateral epicondyle of each femur (LKNE and RKNE),
- Lower lateral surface of each tibia along the line between knee and ankle joint markers (LTIB and RTIB),

- Lateral malleolus of each ankle (LANK and RANK),
- Superior proximal end of the second metatarsal of each foot (LTOE and RTOE),
- Posterior aspect of the Achilles tendon of each leg at the same height as the second metatarsal marker (LHEE and RHEE).

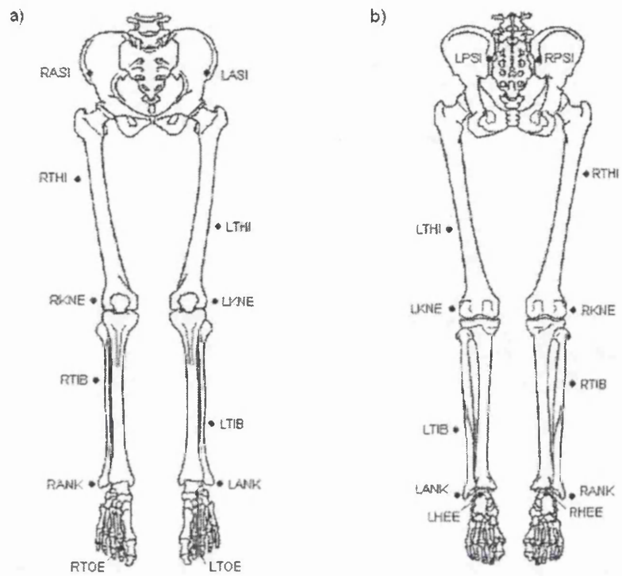


Figure 2.4 Locations of markers in the Vicon lower body plug-in gait marker set, a) anterior aspect; b) posterior aspect.

2.2.3.2 Calculation of joint centres.

From the location of the markers placed on the body, combined with required anthropometric measurements of each subject entered into the system, the Vicon system calculated the 3D coordinates of hip, knee and ankle joint centres. The subject measurements required were height, mass, leg length, knee width and ankle width (Table 2.1). The Vicon system determines the position of;

1) Hip joint centre (HIP) from the LASI, RASI, LPSI and RPSI markers and leg length measurement,

2) Knee joint centre (KNEE) from HIP, knee marker, thigh marker and knee width measurement,

3) Ankle joint centre (ANKLE) from KNEE, ankle marker, tibia marker and ankle width measurement.

Table 2.1 Anthropometric measurements.

Measurement	Method
Height (cm)	Stadiometer
Mass (kg)	Weighing scales
Leg length (cm)	Steel tape (anterior superior iliac spine to medial malleolus)
Knee width (cm)	Bone callipers
Ankle width (cm)	Bone callipers

The Vicon system uses the Newington-Gage model to define the positions of the HIP in the pelvis segment in which pelvis size and leg length are used as scaling factors (Davis et al, 1991).

1) The origin of the pelvis segment is taken as the midpoint between the LASI and RASI markers.

2) The Y axis is coincident with the line joining the RASI and LASI markers with the LASI direction as positive.

3) The X axis is coincident with the line joining the origin and the mid point of the line joining the LPSI and RPSI with forward as positive.

4) The Z axis is perpendicular to the x and y axes with upwards as positive (Figure 2.5).

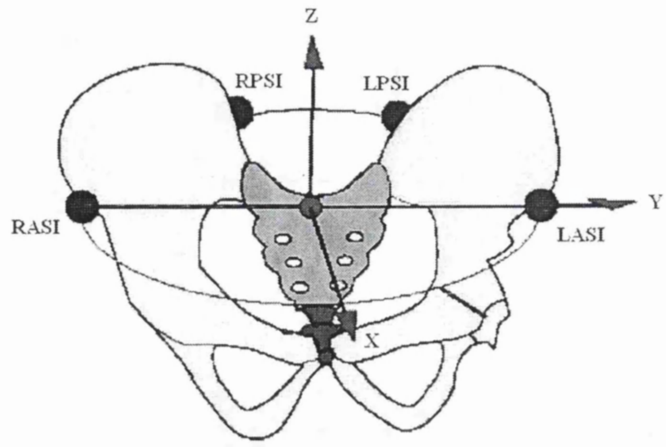


Figure 2.5 Local coordinate system of the pelvis segment.

When calculating the HIP;

1) The inter-ASIS distance is calculated as the mean distance between the LASI and RASI markers for each frame in the trial for which there is a valid position for each marker.

2) The LASI to left greater trochanter and RASI to right greater trochanter distances are calculated from the left and right leg lengths using the following formula (Davis et al, 1991), where all distances are measured in mm;

$$AsisTrocDist = 0.1288 \times LegLength - 48.56$$

3) The value C is calculated from the mean leg length (Davis et al, 1991);

$$C = MeanLegLength \times 0.115 - 15.3$$

4) The x, y and z offsets (distances from the origin of the pelvis segment to the hip joint centres in the x, y and z directions) for the two hip joint centres (LHIP and RHIP) from the origin of the pelvis segment are calculated as follows (Davis et al, 1991);

$$x = C \times \cos(\theta) \times \sin(\beta) - (AsisTrocDist + r) \times \cos(\beta)$$

$$y = -[C \times \sin(\theta) - d]$$

$$z = -C \times \cos(\theta) \times \cos(\beta) - (AsisTrocDist + r) \times \sin(\beta)$$

where d is half the inter-ASIS distance, r is the marker radius, θ is taken as 0.5 radians, and β as 0.314 radians. For the right joint centre, the Y offset is negative with respect to the origin.

The following worked example indicates the offsets for the hip joint centres for a subject of leg length 960 mm and inter-ASIS distance 250 mm, illustrated in Figure 2.6.

$$AsisTrocDist = 0.1288 \times 960 - 48.56 = 75.01mm$$

$$C = 960 \times 0.115 - 15.3 = 95.1$$

$$x = 95.1 \times \cos 0.5 \times \sin 0.314 - (75.01 + 12.5) \times \cos 0.314 = -57.5mm$$

$$y = -[95.1 \times \sin 0.5 - (0.5 \times 250)] = 79.4mm$$

$$z = -95.1 \times \cos 0.5 \times \cos 0.314 - (75.01 + 12.5) \times \sin 0.314 = -106.4mm$$

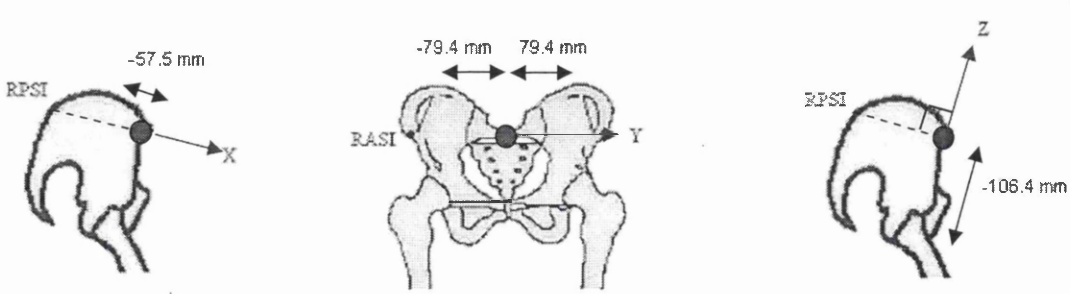


Figure 2.6 Illustration of the calculated hip joint centre offsets from the pelvis segment origin in the x, y and z directions.

Knee and ankle offsets are used to calculate the knee and ankle joint centres. These offsets are calculated by adding half the measured joint width and marker diameter to give the distance from the centre point of the marker to the joint centre. Figure 2.7 shows an illustration of how the left knee joint centre is determined and Figure 2.8 shows an illustration of how the left ankle joint centre is determined. The KNEE is determined from the position of the HIP, the LTHI, and the LKNE, together with the knee offset. The KNEE is located at a distance of the knee offset from the LKNE marker in a direction perpendicular to the line from the HIP to LKNE (Figure 2.7). The ANKLE is determined using the KNEE, LTIB, LANK and the ankle offset. Thus the ANKLE is at a distance of the ankle offset from the LANK marker in a direction perpendicular to the line from the KNEE to LANK (Figure 2.8).

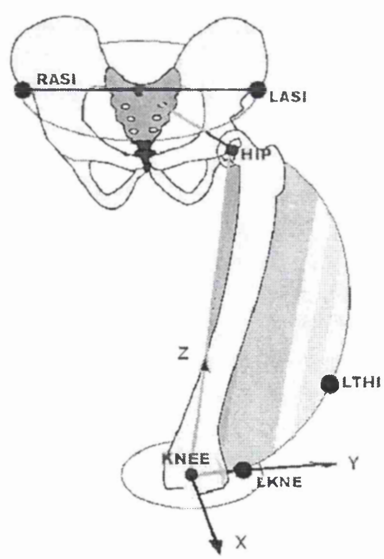


Figure 2.7 Calculation of knee joint centre.

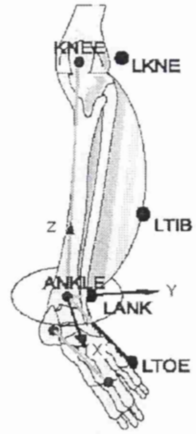


Figure 2.8 Calculation of ankle joint centre.

2.2.3.3 Task.

Subjects were instructed to stand still on a force plate with their feet placed either side of a block of wood 7.5 cm wide to standardise the distance between the feet; see Figure 2.9. Figure 2.9 shows the dimensions of the subjects' feet when standing in the required position which indicates the base of support for each subject. Arms were held horizontally (shoulders abducted 90^{o}) in order to prevent the markers on the pelvis being obscured. A skeleton figure of a subject in the testing position is shown in Figure 2.10. Each trial lasted 5 seconds with only one second (4-5 s) of each trial being used for analysis. Three trials were taken of the subjects in the required position without moving any markers with approximately one minute between trials. Subjects were allowed to move away from the force plate between trials. Markers were then removed and replaced an hour later and the task was repeated.

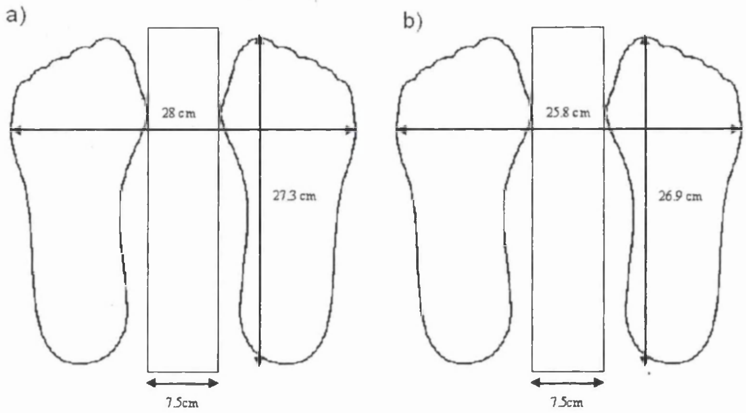


Figure 2.9 Dimensions of the feet of a) Subject 1 and b) Subject 2, when standing in the required position.

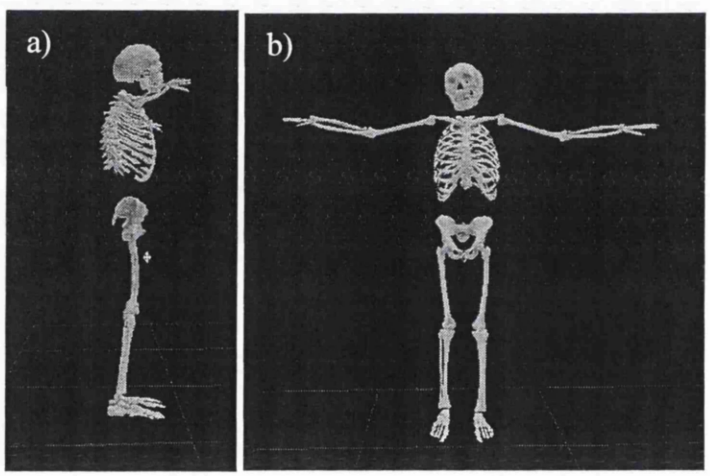


Figure 2.10 Skeleton figure of subject in the testing position. a) lateral aspect; b) anterior aspect.

Since the subjects were required to adopt an upright standing posture in each trial, the markers were placed on the subjects while they were standing upright in order to minimise the likelihood of marker movement as a result of skin and muscle movement when changing body position between trials. Subjects' CP was also recorded using the force plate. Means, standard deviations and coefficients of variation of the x, y, z coordinates of the hip, knee and ankle joint centres, distances between hip, knee and ankle joint centres, angles of the knee (flexion/extension and valgus/varus), and the location of the CP with respect to each subject's base of support were then calculated for each trial.

2.2.3.4 Angular definitions.

The Plug-in gait system measures angles with respect to normal planes of motion of the hip, knee and ankle joints. The measurement of knee flexion/extension is based on the thigh axis (line connecting the hip joint and knee joint centres) and the shank axis (line connecting the knee and ankle joint centres) projected onto the plane of knee flexion/extension (as determined by the plug-in gait marker system). The flexion/extension angle is the angle between the distal extension of the thigh axis and the shank axis. A positive angle corresponds to knee flexion relative to the fully extended position. Figure 2.11 illustrates the angle measured.

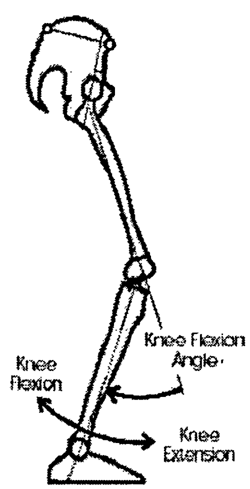


Figure 2.11 Knee flexion/extension: see text for definition.

The measurement of knee valgus/varus is based on the thigh axis and the shank axis projected onto the plane of knee valgus/varus (defined as perpendicular to the knee flexion/extension axis). The valgus/varus angle is the angle between the distal extension of the thigh axis and the shank axis. A positive angle indicates varus and a negative angle indicates valgus. Figure 2.12 illustrates the angle measured.

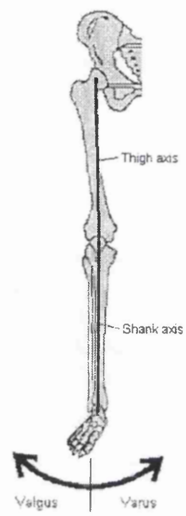


Figure 2.12 Knee valgus/varus: see text for definition.

Since the calculation of angles in the Plug-in gait model requires 3D coordinate data for a particular set of markers (4 on the pelvis, 5 on the leg), it was not possible to calculate angles in the inanimate static structure (3 points) corresponding to knee flexion/extension and knee valgus/varus in the human subjects. However, Vicon allows angles in the 3D workspace to be determined manually by digitising (on the 3D monitor display) the relevant points. For identified points 1, 2 and 3, the angle in 3D space is defined as the acute angle between the extended line connecting points 1 and 2 and the line connecting points 2 and 3 in the plane common to all 3 points. See Figure 2.13.

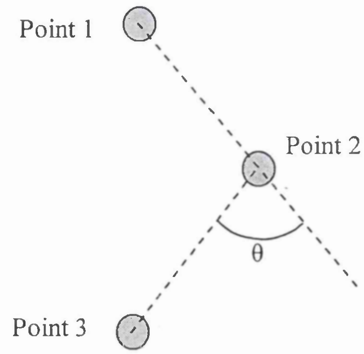


Figure 2.13 Angle measured in the 3D workspace; see text for definition.

The manual system of determining angles was used to compare the variation in the human subject knee angle (where points 1, 2 and 3 were the HIP, KNEE and ANKLE respectively) and the angle between markers 1, 2 and 3 in the inanimate static structure (Table 2.2).

Table 2.2 Angles calculated by the Vicon system.

		Angle projected onto a Plane	
	3D	**Flexion / extension**	**Valgus / varus**
Inanimate static structure	A	-	-
Human subject	B	C	D

2.3 Results.

2.3.1 3D Co-ordinates of markers.

Table 2.3 shows the range of coefficients of variation of the x, y and z coordinates of the three markers in the inanimate static structure. The raw data for individual trials are shown in Appendix D (Table D1).

Table 2.3. Range of coefficients of variation for x, y and z co-ordinates of three markers placed on the inanimate static structures across three trials.

	Range of CV (%)		
Coordinates	**Marker 1**	**Marker 2**	**Marker 3**
X	0.043 - 0.053	0.054 - 0.064	0.055 - 0.063
Y	0.028 - 0.031	0.024 - 0.027	0.019 - 0.024
Z	0.014 - 0.018	0.034 - 0.039	0.107 - 0.127

Tables 2.4 and 2.5 show the range of coefficients of variation of x, y and z coordinates of calculated hip, knee and ankle joint centres for Subject 1 and Subject 2 respectively. For Subject 1, the raw data for individual trials can be seen in Appendix E (Tables E1 and E2). For Subject 2, the raw data for individual trials can be seen in Appendix F (Tables F1 and F2).

Table 2.4 Range of coefficients of variation for x, y and z co-ordinates of the hip, knee and ankle joint centres of Subject 1 across six trials.

	Range of CV (%)					
Coordinates	**Left HIP**	**Left KNEE**	**Left ANKLE**	**Right HIP**	**Right KNEE**	**Right ANKLE**
X	0.41 - 1.73	0.49 - 1.63	0.66 - 1.38	0.37 - 1.45	0.85 - 1.22	2.64 - 7.50
Y	0.20 - 0.72	0.14 - 0.38	0.08 - 0.29	0.35 - 1.26	0.19 - 1.05	0.20 - 1.12
Z	0.01 - 0.04	0.02 - 0.04	0.17 - 2.29	0.01 - 0.06	0.03 - 0.08	0.61 - 1.64

Table 2.5 Range of coefficients of variation for x, y and z co-ordinates of the hip, knee and ankle joint centres of Subject 2 across six trials.

	Range of CV (%)					
Coordinates	**Left HIP**	**Left KNEE**	**Left ANKLE**	**Right HIP**	**Right KNEE**	**Right ANKLE**
X	0.03 - 0.44	0.2 - 0.36	0.15 - 0.41	0.19 – 0.44	0.15 - 0.26	0.1 - 0.1
Y	0.17 - 0.51	0.2 - 0.56	0.09 - 0.32	0.13 – 0.38	0.11 - 0.29	0.06 - 0.09
Z	0.01 - 0.04	0.03 - 0.07	0.13 - 0.36	0.01 – 0.03	0.02 - 0.08	0.16 - 0.41

Figures 2.14 to 2.16 illustrate the difference in the coefficients of variation in the x, y and z coordinates between static markers on the inanimate static structure and the subject joint centres. The data for the static markers relates to three trials for each marker. The data for both Subject 1 and Subject 2 relates to 6 trials each (2 sessions of 3 trials) for each joint centre (HIP, KNEE and ANKLE) and is a combination of

both left and right joint centres. For these figures, the ends of the line for each marker indicates the maximum and minimum coefficient of variation measured for that marker over the trials recorded.

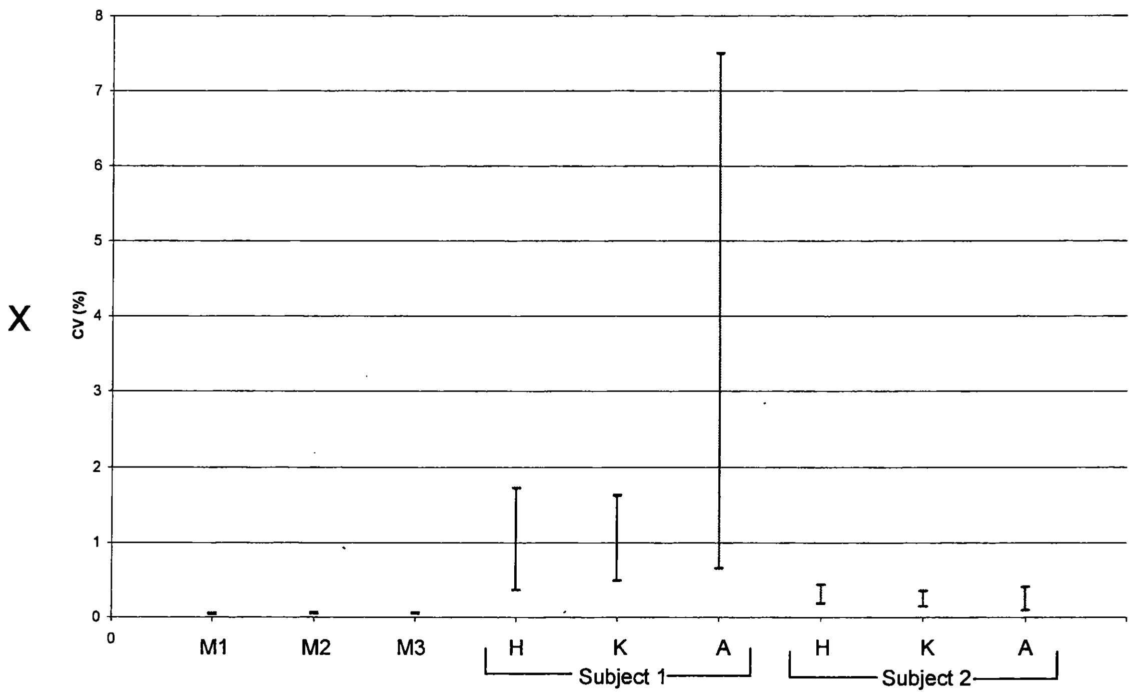


Figure 2.14 Comparison of the coefficients of variation of static markers (M1, M2, M3) on the inanimate static structure (3 trials) and subject joint centres (H, Hip; K, Knee; A; Ankle) (6 trials for each subject, 2 sessions of 3 trials) with respect to the x coordinate (anterioposterior).

The results show greater variation in the location of calculated joint centres than in the location of markers placed on the inanimate static structure. In the x coordinates, the maximum coefficient of variation for the static markers is 0.06%, compared to 7.5% and 0.44% for Subjects 1 and 2 respectively. This relates to a maximum standard deviation of 0.21 mm for the static markers compared to 6.31 mm for Subject 1 and 1.26 mm for Subject 2. For the y coordinates, the maximum coefficient of variation for the static markers is 0.03%, compared to 1.26% and 0.56%

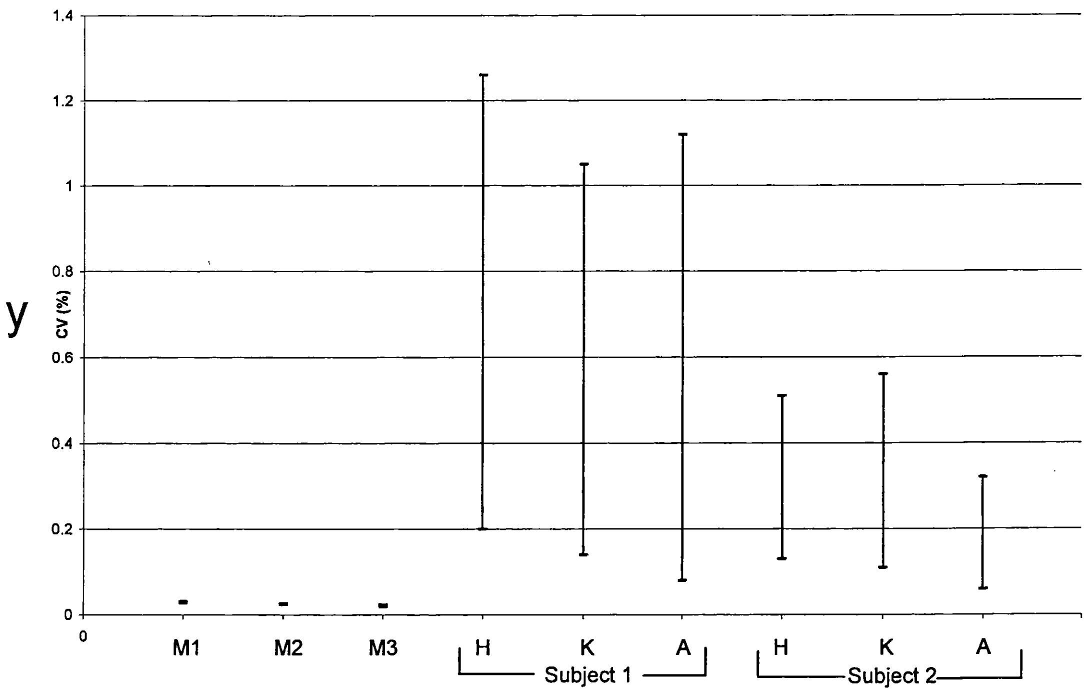


Figure 2.15 Comparison of the coefficients of variation of static markers (M1, M2, M3) on the inanimate static structure (3 trials) and subject joint centres (H, Hip; K, Knee; A, Ankle) (6 trials for each subject, 2 sessions of 3 trials) with respect to the y coordinate (mediolateral).

for Subjects 1 and 2 respectively. This relates to a maximum standard deviation of 0.23 mm for the static markers compared to 2.56 mm for Subject 1 and 1.13 mm for Subject 2. In the z coordinates, the maximum coefficient variation for the static markers is 0.13%, compared to 2.29% and 0.41% for Subjects 1 and 2 respectively. This relates to a maximum standard deviation of 0.18 mm for the static markers compared to 1.48 mm for Subject 1 and 0.38 mm for Subject 2. The results also show greater variation in the x, y and z coordinates of the joint centres of Subject 1 than the joint centres of Subject 2. This indicates that Subject 1 was unable to stand as still as Subject 2 during trials.

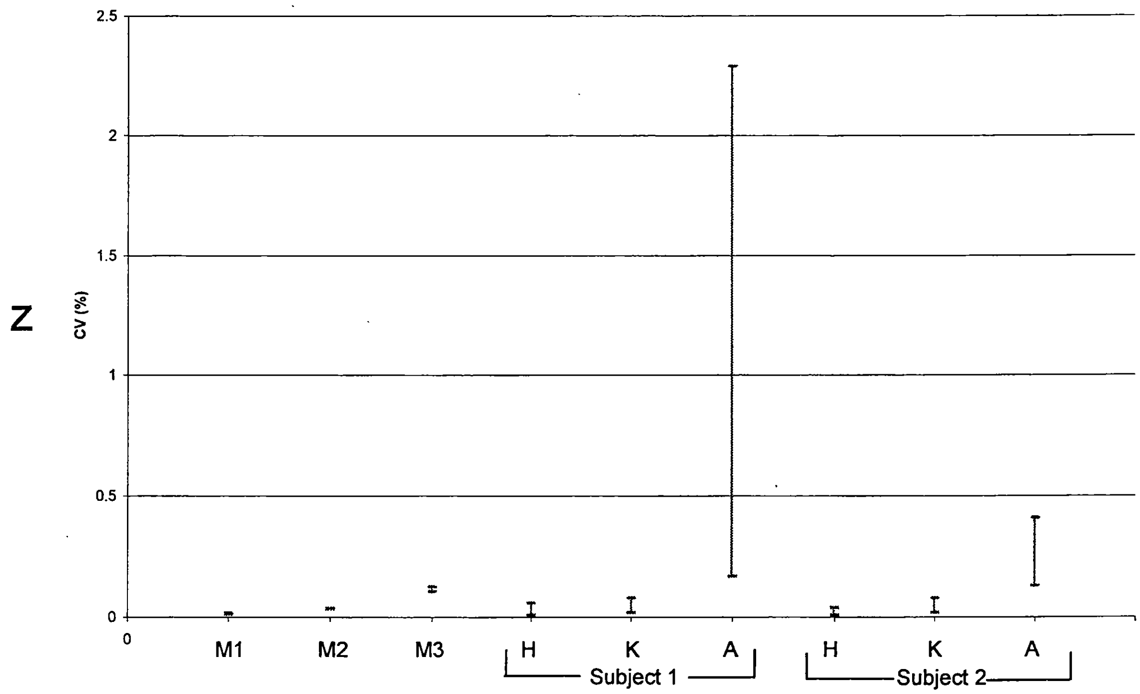


Figure 2.16 Comparison of the coefficients of variation of static markers (M1, M2, M3) on the inanimate static structure (3 trials) and subject joint centres (H, Hip; K, Knee; A, Ankle) (6 trials for each subject, 2 sessions of 3 trials) with respect to the z coordinate (vertical).

2.3.2 Distance between markers.

The range of coefficients of variation of the distances between the static markers is shown in Table 2.6. The raw data can be seen in Appendix D (Table D2).

Table 2.6 Range of coefficients of variation for the distances between markers on the static structure (3 trials).

Distance between markers	Range of CV (%)
1 and 2	0.08 – 0.09
2 and 3	0.05 – 0.06

The range of coefficients of variation of the distances between the joint centres of Subjects 1 and 2 are shown in Tables 2.7 and 2.8 respectively. The distance between the HIP and KNEE relates to the length of the thigh and the distance between the KNEE and the ANKLE relates to the length of the shank. The raw data can be seen in Appendix E (Table E3) for Subject 1 and Appendix F (Table F3) for Subject 2.

Table 2.7 The length of thigh and shank of Subject 1 (six trials).

Segment	Range of CV (%)
Right thigh length	0.05 - 0.12
Right shank length	0.09 - 0.23
Left thigh length	0.03 - 0.08
Left shank length	0.03 - 0.21

Table 2.8 The length of thigh and shank of Subject 2 (six trials).

Segment	Range of CV (%)
Right thigh length	0.04 - 0.08
Right shank length	0.04 - 0.08
Left thigh length	0.04 - 0.10
Left shank length	0.06 - 0.09

Figure 2.17 illustrates the difference in the coefficients of variation of the distances between static markers and the length of the thigh and shank for Subjects 1 and 2. The data for the static markers relates to three trials for each marker. The data for both Subject 1 and Subject 2 relates to 6 trials each (2 sessions of 3 trials) and the results for segment length are presented separately for the left and right sides of the body. For these figures, the ends of the line for each marker indicate the maximum

and minimum coefficient of variation measured for that distance over the trials recorded.

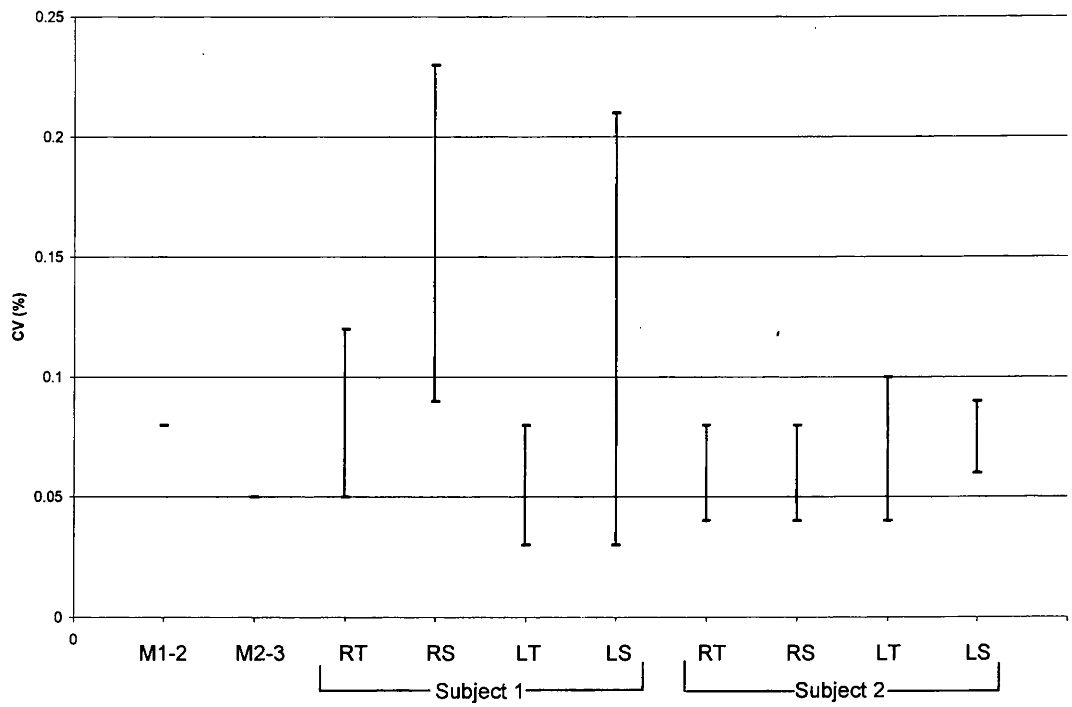


Figure 2.17 Comparison of the coefficients of variation of distances between static markers (M1-2, M2-3) (3 trials) and subjects' thigh and shank lengths (RT, Right thigh; RS, Right shank; LT, Left thigh; LS, Left shank) (6 trials for each subject, 2 sessions of 3 trials).

The results show the maximum coefficient of variation for the distance between the markers on the inanimate static structure to be 0.09%, relating to a maximum standard deviation of 0.24 mm. The results for Subject 1 indicate a maximum coefficient of variation of 0.23% corresponding to a maximum standard deviation of 1.06 mm and for Subject 2 a maximum coefficient of variation of 0.1% corresponding to a maximum standard deviation of 0.38 mm. These results indicate greater absolute and relative variation in the lengths of human body segments compared to the lengths between markers on a static structure.

2.3.3 Angle between markers.

The range of coefficients of variation of the angle made by the markers in the 3D workspace and the angle made by the joint centres in the 3D workspace are shown in Table 2.9. The raw data can be seen for the static structure markers in Appendix D (Table D3), for Subject 1 in Appendix E (Table E4) and for Subject 2 in Appendix F (Table F4).

Table 2.9 Range of coefficients of variation for the angle made between markers (3 trials) and the angle made between joint centres (6 trials).

Angle	**Range of CV (%)**
Static markers	0.16 - 0.21
Subject 1- Right knee	1.66 - 7.76
Subject 1- Left knee	1.27 - 2.48
Subject 2- Right knee	0.64 - 1.89
Subject 2- Left knee	1.15 - 1.87

Figures 2.18 illustrates the difference in the coefficients of variation of the angle between the three static markers and the angle made at the knee (angle between HIP, KNEE and ANKLE) for Subjects 1 and 2 in the 3D workspace. The data for the static markers relates to three trials for each marker. The data for both Subject 1 and Subject 2 relates to 6 trials each (2 sessions of 3 trials) and the results for the left and right knee angles are presented separately. For these figures, the ends of the line for each marker indicate the maximum and minimum coefficient of variation measured for that angle over the trials recorded.

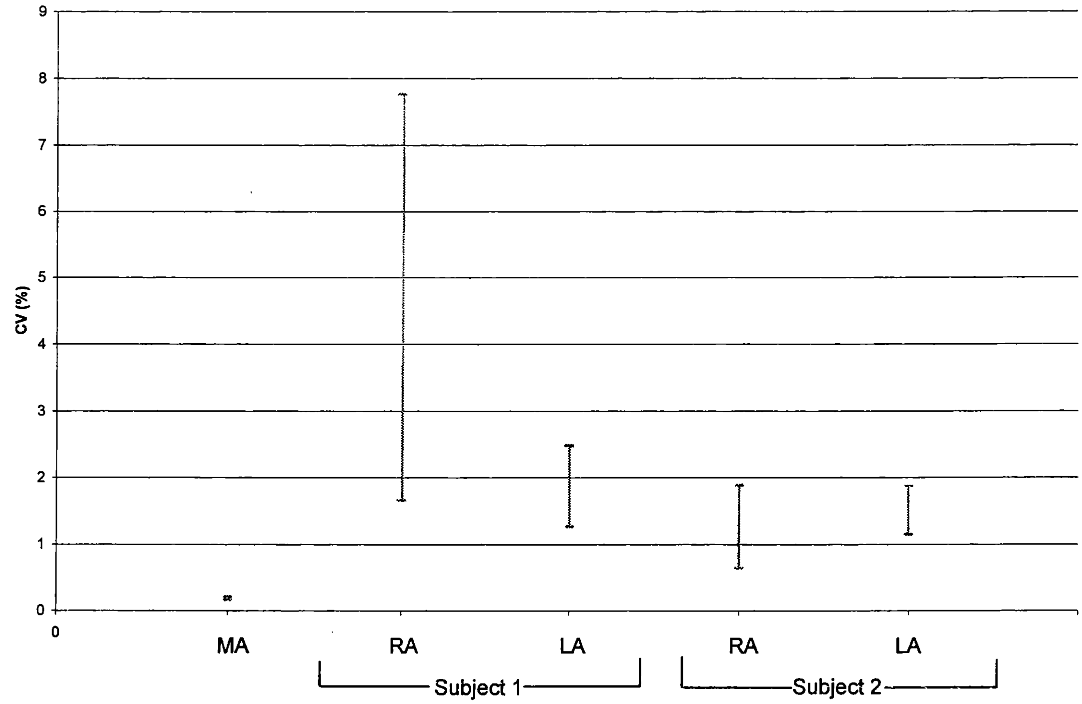


Figure 2.18 Comparison of the coefficients of variation of the angle in the 3D workspace of static markers (MA) (3 trials) and the knee angles of the human subjects (RA, Right knee angle; LA, Left knee angle) (6 trials for each subject, 2 sessions of 3 trials).

The results show that the greater variation in the coordinates of joint centres compared to markers placed on the inanimate static structure (Figures 2.14 to 2.16) is also reflected in greater variation in the knee angles of the human subjects measured compared to the angle between the markers on the static structure. The results show a greater maximum coefficient of variation for the knee angle of the human subjects (Subject 1- 7.76% and Subject 2- 1.89%) compared to the angle between the three markers placed on the inanimate static structure (0.21%). This corresponds to a greater absolute maximum variation of 0.59^{o} and 0.2^{o} for Subjects

1 and 2 respectively, compared to 0.06° for the angle between the markers of the static structure.

2.3.4 Centre of pressure.

Figure 2.19 shows the excursion of the CP of Subject 2 in trial 3 of session 1. Data from this trial is also used in Figures 2.20, 2.21 and 2.22. The raw CP data for this trial can be seen in Appendix F (Table F5). This trial is typical of all trials for both subjects. The maximum excursion in the x (anteroposterior) and y (mediolateral) directions was 8 mm and 13 mm respectively.

Figure 2.19 shows the movement of the CP relative to the subject's base of support. Figure 2.20 shows the movement of the subject's CP relative to the movement of the CP of the static structure.

Figure 2.20 shows greater excursion in the CP of Subject 2 than for the inanimate static structure. For example, the range of the movement of the CP of the static structure shown in Figure 2.20 is approximately 4 mm in the x direction and 3 mm in the y direction compared to approximately 8 mm in the x direction and 13 mm in the y direction for Subject 2. The standard deviation of CP movement for Subject 2 in Figure 2.20 is 1.98 mm in the x direction and 2.96 mm in the y direction. This is typical of all other trials for Subject 1 and 2 (see Tables E5 and F5 in Appendix E and F).

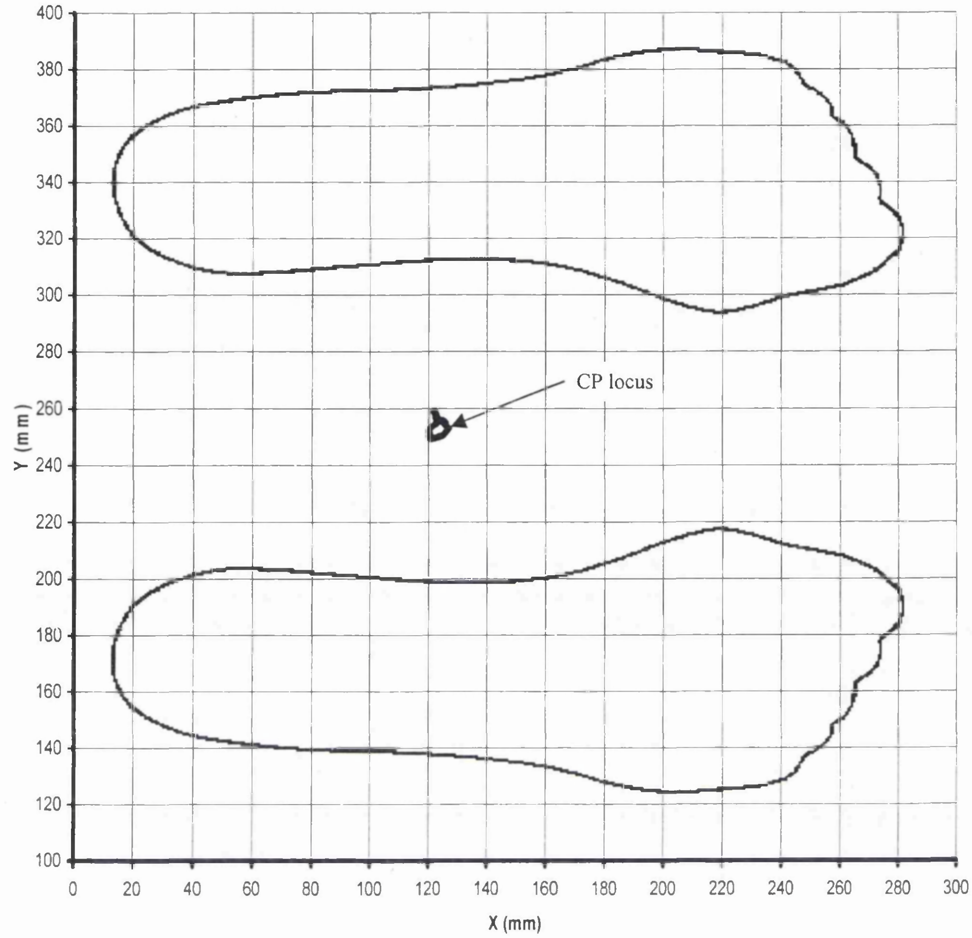


Figure 2.19 CP excursion in the x-y plane of Subject 2 standing still with the subject's feet dimensions superimposed on top.

Figure 2.21 displays the x and y coordinates of the CP of Subject 2 in trial 3 of session 1 standing on the force plate against time over three seconds.

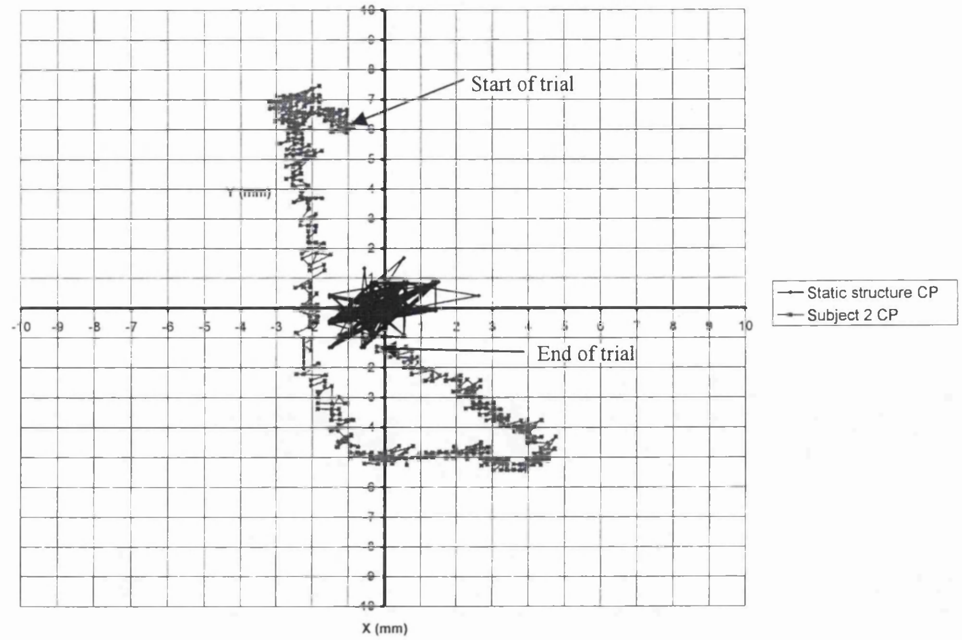


Figure 2.20 Typical CP excursions exhibited by Subject 2 and the inanimate static structure.

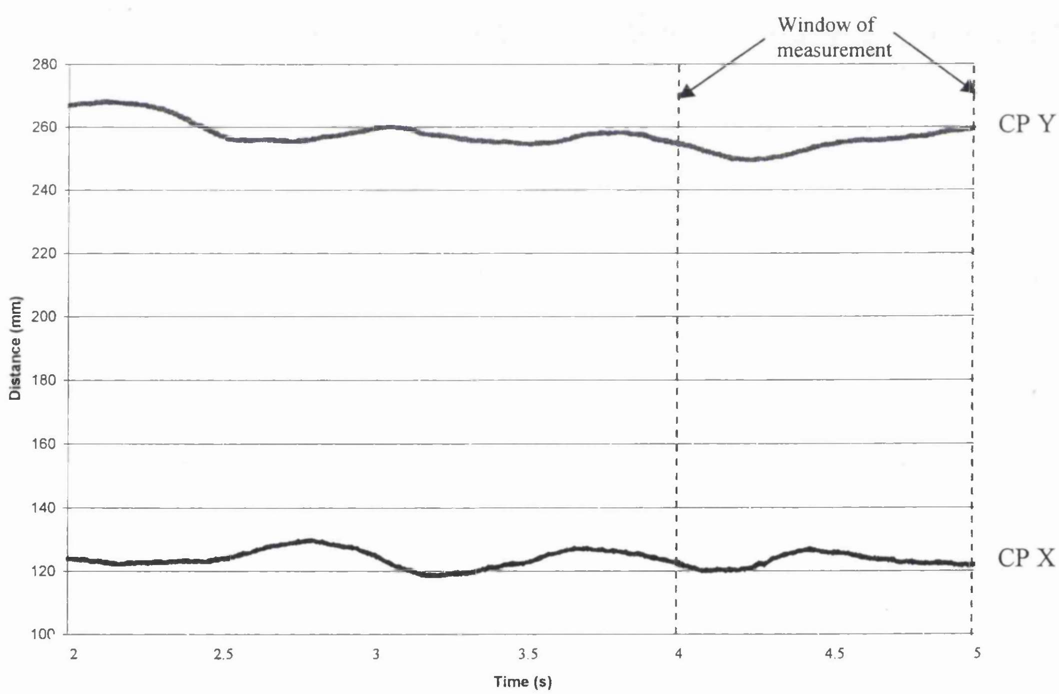


Figure 2.21 Typical CP x and y coordinate-time graphs of Subject 2 standing still on the force plate.

From Figure 2.21, it can be seen that the CP of Subject 2 oscillates approximately 1 cm in both the x and y directions, with approximately two of the x-y cycles taking place in approximately 1.8 s, i.e 1.1 Hz. This movement of the centre of pressure indicates movement of the subject's centre of mass which is, presumably, reflected in the greater variation in the position of the joint centres (and derived measures) than in the position of markers in the inanimate static structure.

2.3.5 Knee flexion/extension and valgus/varus.

Figure 2.22 shows a typical knee flexion/extension and valgus/varus angle-time graphs from the Vicon system using the plug-in gait model. The raw plug-in gait model output data of knee flexion/extension and valgus/varus shown in Figure 2.22 can be seen in Appendix F (Tables F6 and F7).

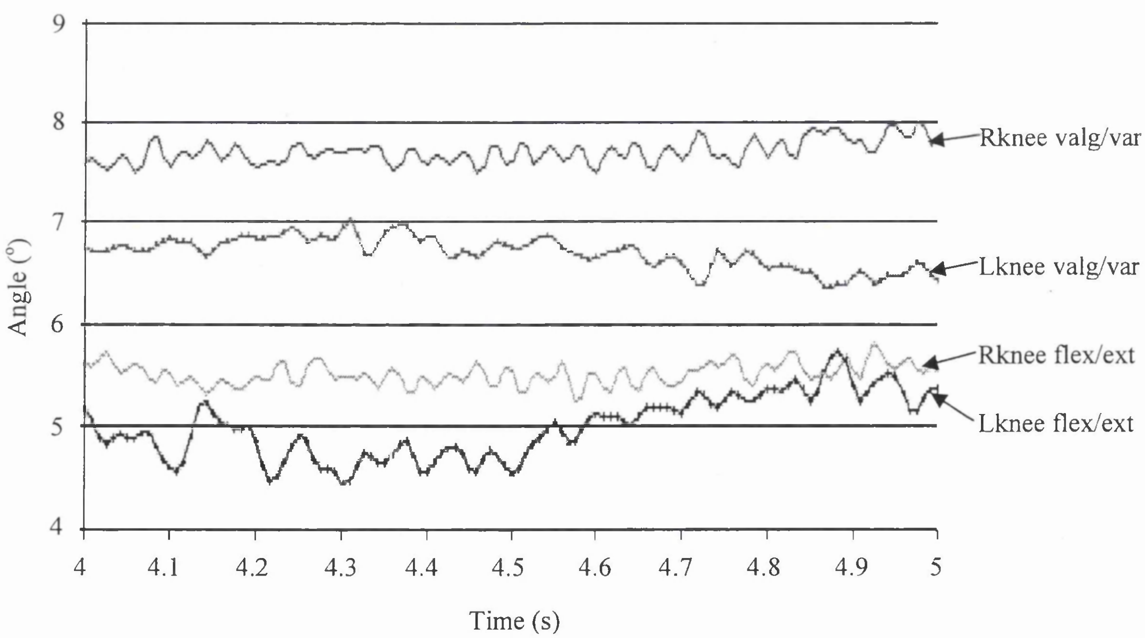


Figure 2.22 Typical right and left knee flexion/extension and valgus/varus angle-time graphs of Subject 2 (Trial 3 of session 1).

2.4 Discussion.

The variation in the measurements was, as expected, less for trials of the inanimate static structure than for human subject trials. This is due to the static structure being internally perfectly static whereas the human subjects were not internally perfectly static, i.e. some movement of markers placed on human subjects when standing still due to muscular activity necessary to maintain their static standing position is inevitable. This muscular activity not only causes movement of markers placed on the subjects' skin but also causes small movement of the whole body. This movement was also evident in the greater movement of the CP for the human subjects compared to the static structure.

The results for the VL system (inanimate static structure) indicate the baseline noise within the system. The results for the static structure indicate that this baseline noise was ± 0.21 mm ± 0.23 mm and ± 0.18 mm for marker x, y and z coordinates respectively, ± 0.24 mm for the distance between two markers and ± 0.06° for the angle between three markers in the 3D workspace. These results were taken from the maximum standard deviation observed for the corresponding variables during the experimental trials.

The results for the VLS system (human subject) indicate the baseline noise within the system. The results indicate that the noise within the VLS system was ± 6.31 mm, ± 2.56 mm and ± 1.48 mm for joint centre x, y and z coordinates respectively, ± 1.06 mm for the length of body segments and ± 0.59° for the angle between three joint centres in the 3D workspace. The greater variation in the x coordinate measurements reflects the greater freedom of movement in this direction due to human anatomical structure. These results were taken from the maximum standard deviation observed for the corresponding variables during the experimental trials. This indicates that any variation in joint centre measurements below this limit may be termed noise. Any variation in joint centres measurements beyond this limit is likely to indicate joint movement.

When collecting subsequent data in experimental trials with human subjects it will be necessary to take a static trial in order to provide a reference for the interpretation of movement in dynamic trials. Also, provided that markers are not removed and replaced between static and dynamic trials, the error due to differences in marker placement will be eliminated.

2.5 Summary and conclusions.

The results indicate that the Vicon 512 motion analysis system reliably determines the positions of joint centres with a very small variation. This small variation results in variation of corresponding distance and angular measurements taken. The small amount of variation in these variables may be termed noise and subsequent measurements taken with the Vicon system using the lower body plug-in gait marker placement will be accurate to the static variation indicated.

Chapter 3

Experimental Study 1

Gender differences in knee kinematics and kinetics during landing from volleyball block jumps.

3.1 Introduction.

3.1.1 Landing mechanics.

The purpose of landing from a jump is to reduce the linear momentum of the human body to zero. The linear momentum ($m.v$) of the body on landing is equal to the product of the mass m of the body and the linear velocity v of the body at initial ground contact. During landing (the period from initial contact with the floor, Figure 3.1a, to the instant when the downward velocity of the body is zero, Figure 3.1b) there are two forces acting on the body, the body weight W and the ground reaction force R (assumed to be vertical for illustration).

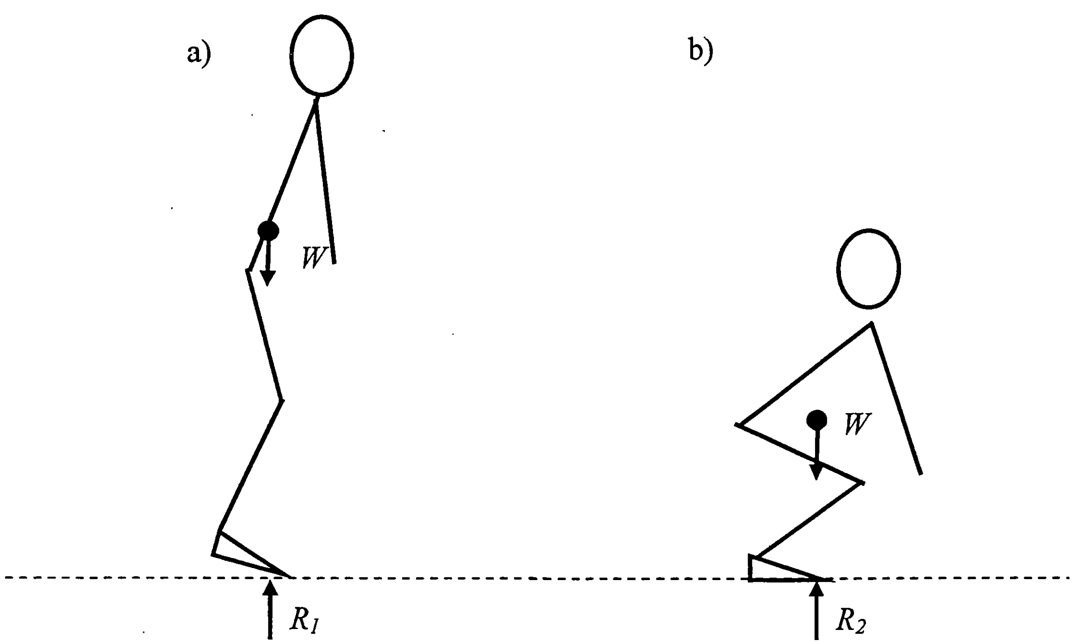


Figure 3.1 Vertical forces acting on the body during landing. a) instant of ground contact; b) instant when the downward velocity of the body CG is zero.

The resultant force F (positive upward) acting on the whole body CG is given by,

$$F = R - W \qquad (3.1)$$

Consequently,

$$F = R - W = m.a \quad \text{.......} \quad (3.2)$$

where m = mass of the body and a = acceleration of the CG.

The acceleration of the CG is given by,

$$a = \frac{v - u}{t} \quad \text{.......} \quad (3.3)$$

where u = downward velocity of the CG at t_1 (Figure 1a), $v = 0$ at t_2 (body at rest, Figure 3.1b) and $t = t_2 - t_1$.

From equations (3.2) and (3.3),

$$F = \frac{m\,(v - u)}{t} \quad \text{.......} \quad (3.4)$$

Since $v = 0$,

$$F = -\frac{m.u}{t} \quad \text{.......} \quad (3.5)$$

where F = average resultant force acting on the body during landing.

As $F = R - W$ and W is constant, it follows that F is directly proportional to R, i.e the greater the deceleration, the greater the ground reaction force. The greater the ground reaction force, the higher the risk of injury to the lower limbs. It is clear from equation (3.5) that F is inversely proportional to t i.e., the greater the length of time in which the momentum of the person is reduced to zero, the smaller the average force required and, therefore, the less chance of injury. Conversely, the higher the rate of change of momentum of the body, the greater the average force and the higher the risk of injury.

Therefore, when landing, it is advisable to maximise the period of controlled deceleration in order to minimise the average force and, therefore, the peak force acting on the body. During landing the ankle, knee and hip joints will move from a position of extension to flexion. The time in which this occurs will reflect the magnitude of the average and peak forces acting on the body and therefore the likelihood of injury to the lower limbs. When the lower limb joints move from a position of extension to flexion, the extensor muscles of the ankle, knee and hip joints dissipate the linear momentum of the body as strain energy in muscle – tendon units by eccentric contraction. The strain energy is subsequently dissipated as heat. The average and peak ground reaction force will therefore be determined by the lower limb joint muscle moments and the coordination between the muscles (Figure 3.2). In reducing the body's linear momentum to zero, the coordination of the body will be reflected in the angular kinematics of the joints and the linear kinematics of the movement of the whole body centre of gravity (CG). Figure 3.2 shows the relationships between the mechanical factors which influence landing.

After initial contact with the ground during landing it takes a finite time for muscles to fully respond to changes in external loading. This time lag is referred to as muscle latency. Muscle latency varies between 30 ms and 75 ms (Nigg et al., 1984; Watt and Jones, 1971). Consequently, muscles cannot fully respond to changes in external load that occur in less than the latency period of muscles. In these circumstances the body is forced to respond passively to the external load. This type of loading is referred to as passive loading. After the passive loading phase, the magnitude and direction of the ground reaction force is completely controlled by conscious muscular activity, referred to as the active loading phase.

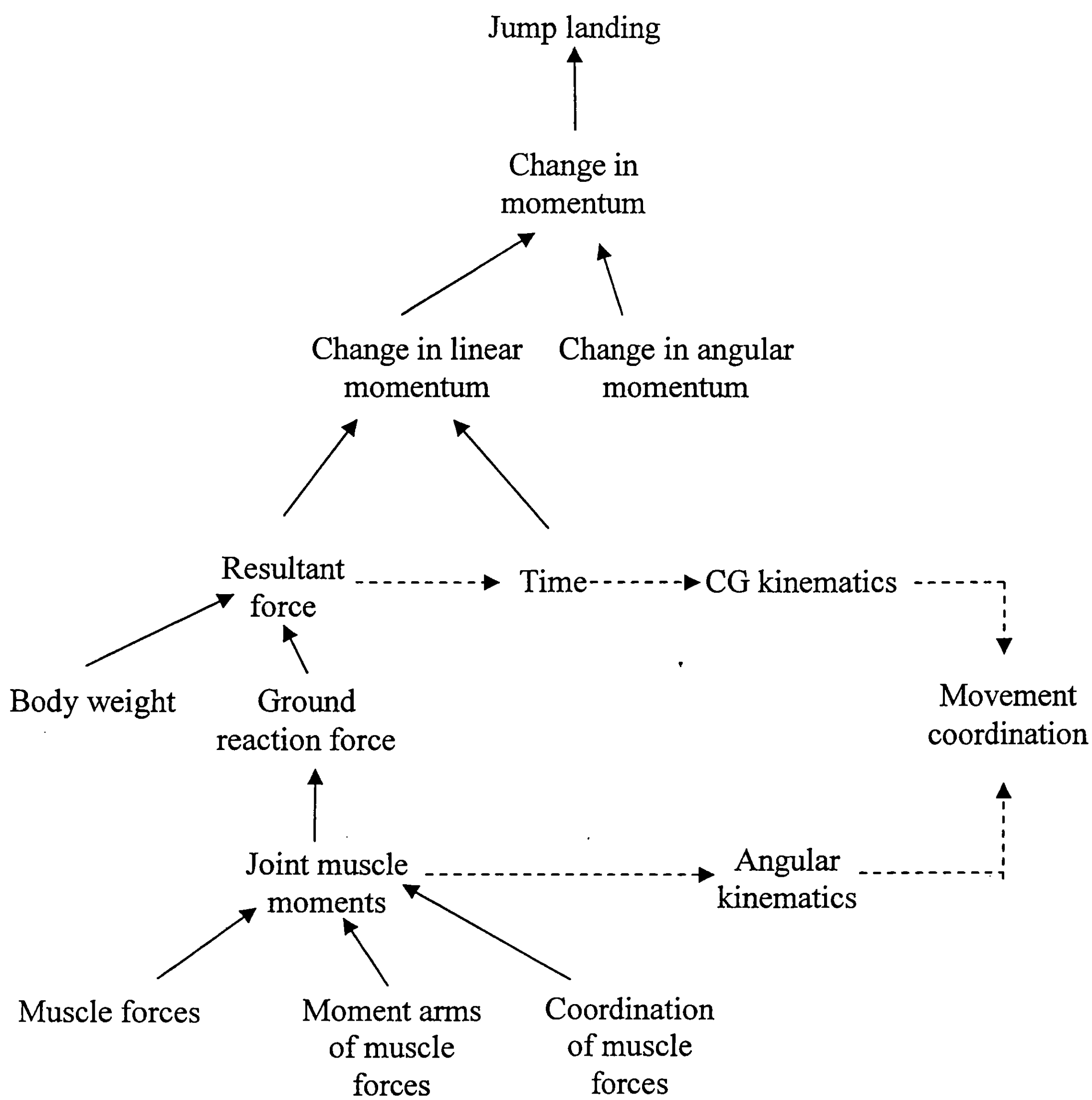


Figure 3.2 Model of landing mechanics.

By definition, the body is unable to control passive loading and therefore, the body is most vulnerable to injury from high passive loads. It is, perhaps, not surprising that ACL injury appears to occur most often just after initial ground contact (Olsen et al., 2004; Boden et al., 2000), i.e. during passive loading.

3.1.2 Landing/cutting kinematics.

Whilst the muscle moments about the joints of the lower limbs largely determine the movement patterns of the lower limbs, the resulting angular kinematics of the joints may provide some indication of strain on the passive support structures, especially the ligaments; the greater the range of abnormal joint movement, the greater the strain on associated ligaments (Watkins, 1999). A number of studies which have investigated the sagittal plane kinematics of landing and/or cutting manoeuvres report that females tend to land with the knees more extended than males (Yu et al., 2006; James et al., 2004; Decker et al., 2003; Malinzak et al., 2001) and exhibit a greater range of knee flexion than males (Decker et al., 2003). Some studies have reported a smaller maximum knee flexion angle in females than males, i.e. a more extended knee at the end of the landing movement (Yu et al., 2006; Salci et al., 2004; Malinzak et al., 2001). For a given load on the patellar ligament, the more extended the knee, the greater the strain on the ACL is likely to be due to the effect of knee flexion on the patella tendon-tibia shaft angle (PTTSA) (angle between the long axis of the tibia and the line of action of the patellar ligament in the sagittal plane) (Nunley et al., 2003; Li et al., 1999) (Figure 3.3). When the knee is close to full extension, contraction of the quadriceps, which acts through the patellar tendon, causes an anterior shear force applied to the proximal end of the tibia (ASPT). The greater the PTTSA, the greater the ASPT and the greater the potential strain on the ACL.

Nunley et al. (2003) investigated the effect of knee angle on the PTTSA. Regression analysis showed that, in both males and females, the PTTSA decreased linearly as knee flexion decreased from 0° of knee flexion to 90° of knee flexion.

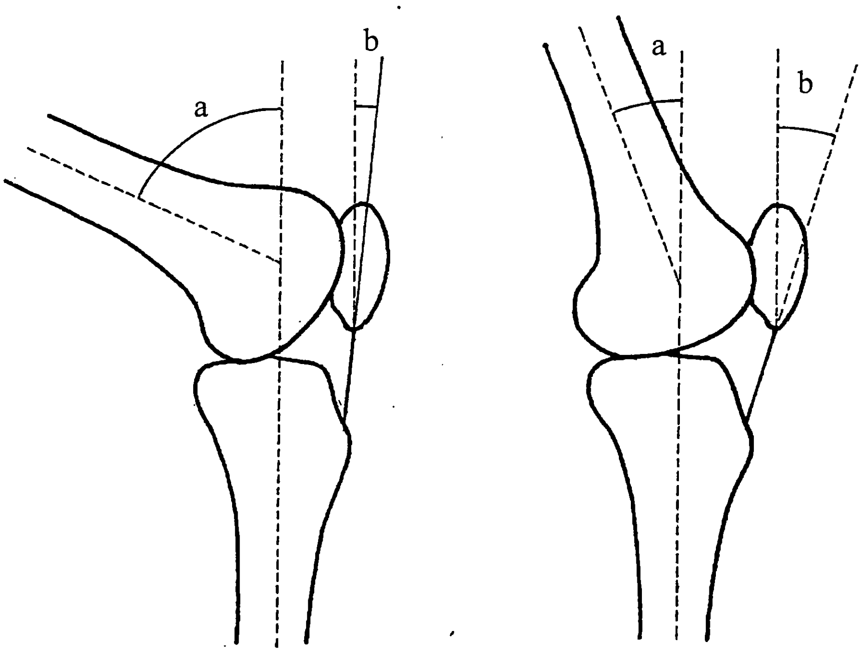


Figure 3.3 Effects of knee flexion angle (a) on patella tendon tibia shaft angle (b) (adapted from Nunley et al. 2003).

Therefore, the PTTSA and consequently the ASPT was largest when the knee was close to full extension. A number of studies have reported that non-contact ACL injury appears to occur more frequently when the knee is close to full extension than when flexed (Boden et al., 2000; Olsen et al., 2004). Consequently, if females do tend to make ground contact with knees more extended than males, this may increase the risk of ACL injury relative to males.

A number of studies which have investigated the frontal plane kinematics of landing/cutting report that females tend to exhibit greater maximum knee valgus angle and greater range of motion (from initial contact to maximum) when landing than males (Kernozek et al., 2005; Ford et al., 2003; Malinzak et al., 2001). Boden et al. (2000) and Olsen et al. (2004) have reported that non-contact ACL injury appears to occur more frequently when the knee exhibits a valgus movement (relative to normal upright standing position). Consequently, the reported greater maximum knee valgus angle in females when landing may increase the risk of ACL injury relative to males. A summary of the reported differences between males and females in lower limb sagittal and frontal plane kinematics in landing/cutting manoeuvres is shown in Table 3.1.

Table 3.1 Sagittal and frontal plane kinematics in landing/cutting movements in males and females.

Study.	Task	Sagittal plane kinematics.	Frontal plane kinematics.
Salci et al. (2004)	40 cm and 60 cm vertical drop landing.	F displayed smaller maximum knee flexion angles than M (M: 79.6 ± 17.9°; F: 59.3 ± 9.5°)	
Decker et al. (2003)	60 cm vertical drop landing.	F had smaller knee flexion at ground contact (M: 30.0 ± 7.7°; F: 22.8 ± 8.0°) and greater range of motion (M: 63.4 ± 9.3°; F: 75.8 ± 9.1°) than M.	
Ford et al. (2003)	31 cm vertical drop-jump landing.		Increased knee valgus motion (M: 5.3 ± 0.5 cm; F: 7.3 ± 0.5 cm) and maximum angle (M: 14.25 ± 1.95°; F: 20.05 ± 2.5°) in F compared to M.
Malinzak et al. (2001)	Running, side-cutting and cross-cutting.	F displayed smaller knee flexion throughout stance phase than M (mean of 8° less throughout stance phase. No absolute mean data provided).	F exhibited greater knee valgus angle throughout stance phase than M (mean of 11° more throughout stance phase. No absolute mean data provided).
James et al. (2004)	Cutting.	F exhibited smaller knee flexion at ground contact than M (M: 46.0 ± 8.05°; F: 40.2 ± 8.04°).	
Kernozek et al. (2005)	60 cm vertical drop landing.		F exhibited greater peak (M: -0.66 ± 6.90°; F: 24.85 ± 8.45°) and range of motion (M: 7.08 ± 6.61°; F: 26.50 ± 9.00°) of knee valgus angle than M.
Yu et al. (2006)	Stop-jump landing.	F exhibited smaller knee flexion at ground contact (M: 31.92 ± 10.30°; F: 23.95 ± 8.31°) and smaller maximum knee flexion (M: 77.36 ± 10.59°; F: 68.54 ± 9.28°) than M.	

F = females, M = males.

Lack of standardisation in the demands of the tasks that subjects are required to perform will influence the movement patterns exhibited and reduce the likelihood of meaningful comparisons. For example, dropping down from a raised platform set at the same height for both males and females (Salci et al., 2004; Decker et al., 2003; Ford et al., 2003) may result in significantly different task demands. Also, dropping down from a raised platform usually requires an initial jump to clear the platform. This could lead to differences between subjects with regard to the impulse of the initial jump which would affect linear momentum of the body on landing.

With regard to movement of the knee during landing and cutting manoeuvres, many studies only report absolute angular displacement – time data (Kernozek et al., 2005; Ford et al., 2003; Malinzak et al., 2001) with no reference to the subjects' natural lower leg alignments. There is considerable evidence that the Q angle, i.e. the acute angle between the line connecting anterior superior iliac spine to the middle of the patella and the line connecting the tibial tuberosity to the centre of the patella (Hungerford and Barry, 1979) is, on average, larger in females than males (Horton and Hall, 1989; Hsu et al., 1990; Woodland and Francis, 1992; Guerra et al., 1994; Herrington and Nester, 2004). For example, Herrington and Nester (2004) measured the Q angles of 51 male and 58 female physically active subjects with no history of lower limb injury. Q angle was measured with subjects standing upright with quadriceps relaxed. The mean Q angle was significantly greater in females (mean Q angle $13.9^{\circ} \pm 5.35^{\circ}$) than in males (mean Q angle 11.5°

± 5.05°) ($p = 0.02$). The larger Q angle in females may contribute to some extent to the larger maximum knee valgus angle reported in some studies for females on landing (Kernozek et al., 2005; Ford et al., 2003; Malinzak et al., 2001), but there would appear to be no reported data concerning change in lower leg alignment on landing relative to normal lower leg alignment in females or males.

3.1.3 Landing/cutting kinetics.

During landing the ankle, knee and hip joints will move from a position of extension to flexion as the downward linear momentum of the body is reduced to zero. These joint movements are determined by the net joint moments acting about the joints. It takes a certain amount of time (latency period of the muscles) for the muscles to fully respond to the ground reaction force. During this period of passive loading, the body is vulnerable to injury from high ground reaction force (GRF) and/or high external moments about the joints arising from the ground reaction force. During active loading however, the magnitude and direction of the GRF is controlled by conscious muscular activity, i.e., the muscles determine the magnitude and direction of the GRF in order to try to prevent excessive GRF moments about the lower limb joints and therefore reduce the risk of injury.

Figure 3.4 shows a sagittal plane free body diagram and an acceleration diagram of the shank and foot segment during landing.

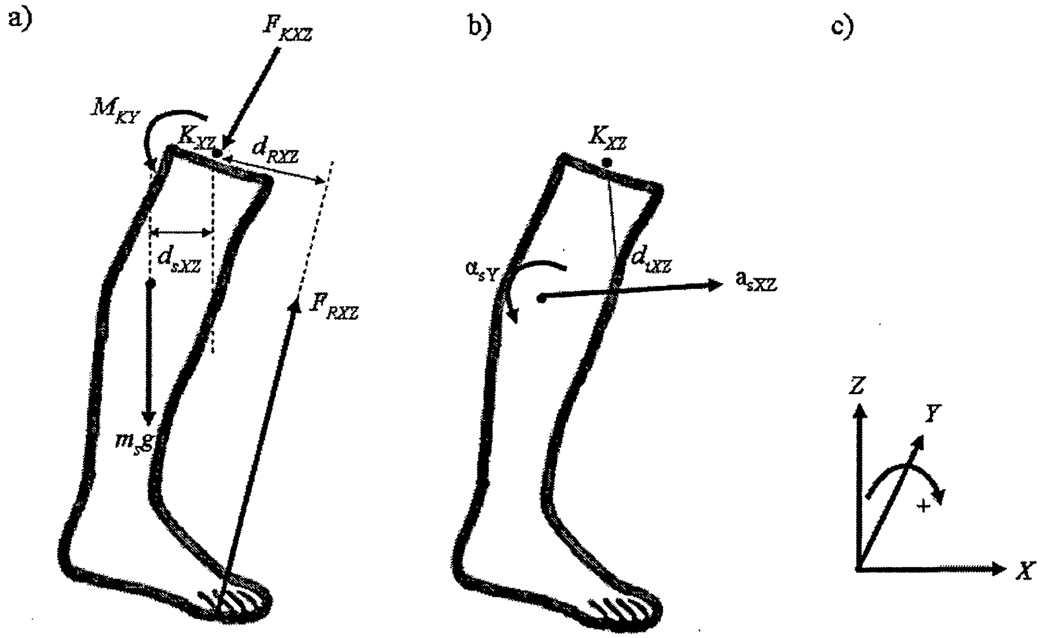


Figure 3.4. Sagittal plane free body diagram (a), acceleration diagram (b) and corresponding axis system (c) of the shank and foot segment during landing. F_{KXZ} = the component of the knee joint reaction force in the sagittal plane (*XZ*), M_{KY} = the resultant moment of the force distribution (muscle and articular) about the *Y* axis through the knee joint centre, K_{XZ} = location of the knee joint centre in the sagittal plane (*XZ*), F_{RXZ} = ground reaction force vector in the sagittal plane (*XZ*), d_{RXZ} = moment arm of the ground reaction force about the knee joint centre in the sagittal plane (*XZ*), $m_s.g$ = weight of the shank and foot segment acting at the CG, d_{sXZ} = moment arm of the weight of the shank and foot segment in the sagittal plane (*XZ*), α_{sY} = angular acceleration of the shank and foot segment about the *Y* axis through the CG, a_{sXZ} = linear acceleration of the shank and foot segment CG in the sagittal plane (*XZ*), d_{iXZ} = moment arm of the linear acceleration of the shank and foot segment CG in the sagittal plane (*XZ*).

In accordance with Newton's second law of motion, the sum of the moments of the applied forces and torques acting on the shank and foot segment is equal to the rate of change of angular momentum of the segment, i.e. taking moments about the Y axis through K_{XZ},

$$-M_{KY} - m_s.g.d_{sXZ} - F_{RXZ}.d_{RXZ} = -m_s.a_{sXZ}.d_{iXZ} - I_s.\alpha_{sY} \quad (3.6)$$

$$M_{KY} = m_s.a_{sXZ}.d_{iXZ} + I_s.\alpha_{sY} - m_s.g.d_{sXZ} - F_{RXZ}.d_{RXZ} \quad (3.7)$$

where M_{KY} = muscle moment (*MUS*): the resultant moment of the force distribution about the Y axis through the knee joint centre, i.e. the moment exerted by the muscle and articular forces about the knee joint. $m_s.a_{sXZ}.d_{iXZ}$ = motion dependent moment (*MDM*): the moment acting on the segment due to the motion of adjacent segments in the sagittal plane, i.e. the thigh. $m_s.g.d_{sXZ}$ = gravitational moment (*GRAV*): the moment acting on the segment due to its weight. $F_{RXZ}.d_{RXZ}$ = external moment (*EXT*): the moment acting on the segment due to the GRF. $I_s.\alpha_{sY}$ = net joint moment (*NET*): sum of *MUS*, *MDM*, *GRAV* and *EXT*.

A frontal plane free body diagram and an acceleration diagram of the shank and foot segment during landing are shown in Figure 3.5.

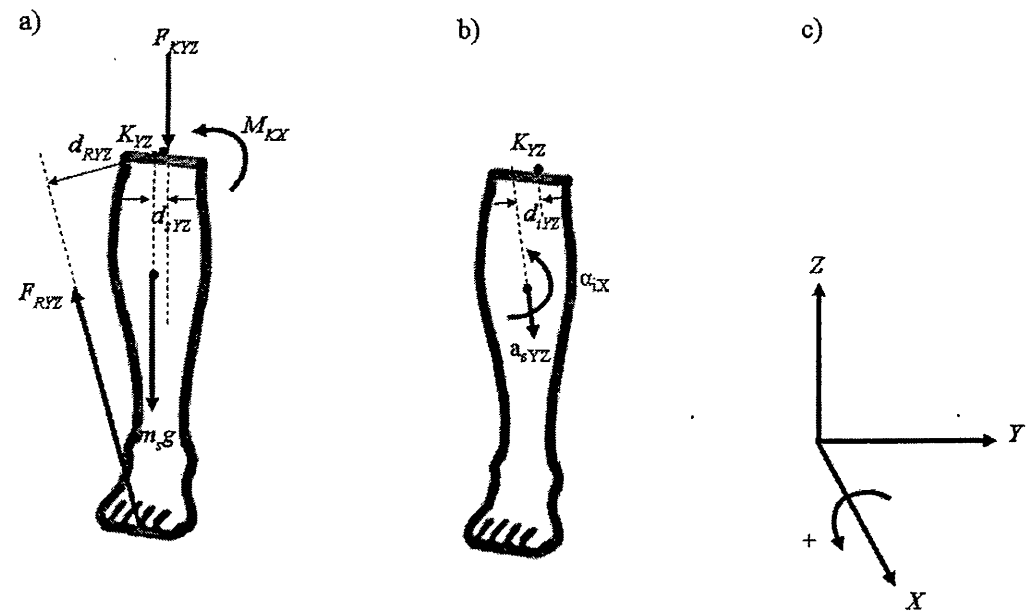


Figure 3.5 Frontal plane free body diagram (a), acceleration diagram (b) and corresponding axis system (c) of the shank and foot segment during landing. F_{KYZ} = the component of the knee joint reaction force in the frontal plane (YZ), M_{KX} = the resultant moment of the force distribution (muscular and articular) about the X axis through the knee joint centre, K_{YZ} = location of the knee joint centre in the frontal plane (YZ), F_{RYZ} = ground reaction force vector in the frontal plane (YZ), d_{RYZ} = moment arm of the ground reaction force about the knee joint centre in the frontal plane (YZ), $m_s.g$ = weight of the shank and foot segment acting at the CG, d_{sYZ} = moment arm of the weight of the shank and foot segment in the frontal plane (YZ), α_{sX} = angular acceleration of the shank and foot segment about the X axis through the CG, a_{sYZ} = linear acceleration of the shank and foot segment CG in the frontal plane (YZ), d_{iYZ} = moment arm of the linear acceleration of the shank and foot segment CG in the frontal plane (YZ).

In accordance with Newton's second law of motion, the sum of the moments of the applied forces and torques acting on the shank and foot segment is equal to the rate of change of angular momentum of the segment, i.e. taking moments about the X axis through K_{YZ},

$$M_{KX} + m_s.g.d_{sYZ} - F_{RYZ}.d_{RYZ} = m_s.a_{sYZ}.d_{iYZ} + I_s.\alpha_{sX} \quad (3.8)$$

$$M_{KX} = m_s.a_{sYZ}.d_{iYZ} + I_s.\alpha_{sX} - m_s.g.d_{sYZ} + F_{RYZ}.d_{RYZ} \quad (3.9)$$

where M_{KX} = muscle moment (*MUS*): the resultant moment of the force distribution about the X axis through the knee joint centre, i.e. the moment exerted by the muscle and articular forces about the knee joint. $m_s.a_{sYZ}.d_{iYZ}$ = motion dependent moment (*MDM*): the moment acting on the segment due to the motion of adjacent segments in the frontal plane, i.e. the thigh. $m_s.g.d_{sYZ}$ = gravitational moment (*GRAV*): the moment acting on the segment due its weight. $F_{RYZ}.d_{RYZ}$ = external moment (*EXT*): the moment acting on the segment due to the GRF. $I_s.\alpha_{sX}$ = net joint moment (*NET*): sum of *MUS*, *MDM*, *GRAV* and *EXT*.

Equations (3.7) and (3.9) show that *MUS* is the sum of four possible components, i.e.

$$MUS = MDM + NET + GRAV + EXT \quad (3.10)$$

The signs of the four components (positive or negative) will depend on the axis system being used.

Based on equation (3.10), three methods of determining *MUS* are reported in the literature. These are the quasi-static model (Hewett et al., 1996; Harrison et al., 1986; Alexander and Vernon, 1975; Smith, 1975), the ground reaction force and segment weight model (Watkins and Nicol, 1986) and the inverse dynamics model (Yu et al., 2006; Kernozek et al., 2005; Salci et al., 2004; Decker et al., 2003;

Chappell et al., 2002). For the quasi-static model, it is assumed that the segment mass and the linear and angular accelerations of the segment CG are zero. Therefore, the *MDM*, *NET* and *GRAV* are assumed to be zero. Consequently, from equation (3.10), *MUS* is considered to be equal and opposite to *EXT* (Figure 3.6a). The ground reaction force and segment weight model takes into account *GRAV* and *EXT* when calculating *MUS* (Figure 3.6b). This approach assumes that *MDM* and *NET* are zero and, consequently, *MUS* is equated to the sum of *GRAV* and *EXT*; see equation (3.10). The inverse dynamics approach, expressed in equation (3.10), takes into account all four possible components of *MUS* (Figure 3.6c).

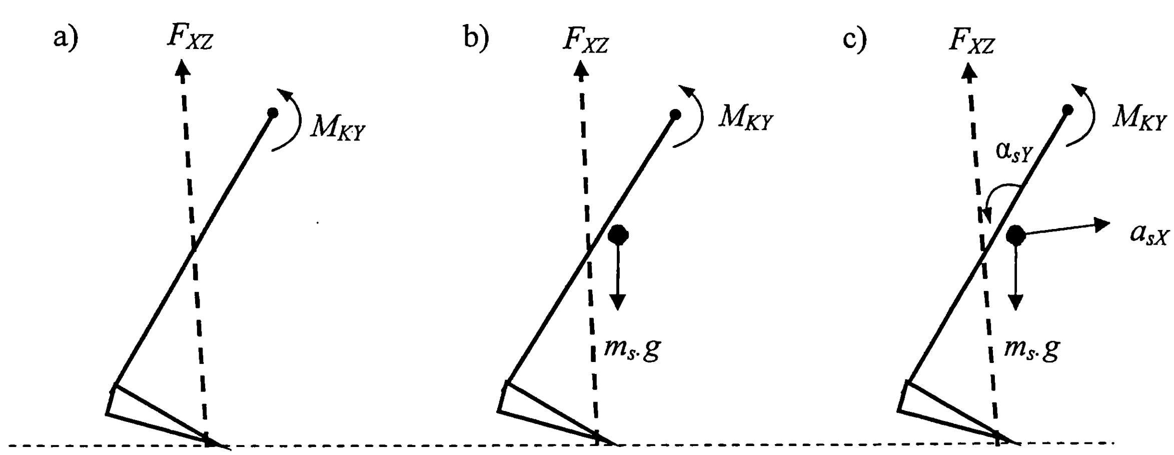


Figure 3.6 Sagittal plane diagram of a) Quasi-static model (F_{XZ} = ground reaction force vector in the sagittal plane, M_{KY} = muscle moment about the Y axis through the knee joint centre), b) Ground reaction force and segment weight model ($m_s.g$ = weight of the shank and foot segment acting at the segment CG), c) Inverse dynamics model (a_{sXZ} = linear acceleration of the shank and foot segment CG in the sagittal plane, α_{sY} = angular acceleration of the shank and foot segment about the Y axis through the CG).

The inverse dynamics approach to calculating the muscle moments acting about a joint is the most accurate method as it takes into consideration all of the possible component moments. However, when the segment mass is small and the linear and angular accelerations of the segment CG are small, such that the *GRAV*, *NET* and *MDM* are small relative to *EXT*, the more closely *EXT* will approximate *MUS* (Winter, 1990). When this is the case, the quasi-static model for calculating the muscle moment is justifiable (Hewett et al., 1996; Harrison et al., 1986; Alexander and Vernon, 1975; Smith, 1975). Alexander and Vernon (1975) found that in two 68 kg male subjects landing from a 0.81 m vertical drop the effect of the *GRAV*, *MDM* and *NET* acting on the shank and foot segment was small in relation to *EXT* when calculating *MUS* about the knee joint centre using the quasi-static model. For example, the peak *MUS* about the knee was estimated at 120 N.m using the quasi-static model. This was decreased by 9 N.m when *GRAV*, *MDM* and *NET* were included. The quasi-static model is more accurate the closer, in terms of the links in the skeletal chain, the joint under consideration is to the point of application of the GRF, i.e. in terms of the lower limb, determination of *MUS* by the quasi-static model is likely to be most accurate for the ankle, then the knee, then the hip (Alexander and Vernon, 1975).

In the sagittal plane, a knee extension *MUS* (negative M_{KY} with respect to Figure 3.4) is the resultant extensor moment exerted by the muscles about the mediolateral (*Y*) axis through the knee joint centre in response to an *EXT* that tends to flex the knee. A knee flexion *MUS* (positive M_{KY} with respect to Figure 3.4) is the resultant flexor moment exerted by the muscles about the mediolateral (Y) axis through the knee joint in response to an *EXT* that tends to extend the knee.

In the frontal plane, a knee varus *MUS* (positive M_{KX} with respect to Figure 3.5) is the resultant varus moment exerted by the muscles about the anterioposterior (X) axis through the knee joint centre in response to an *EXT* that tends to abduct (move into a valgus position) the knee. A knee valgus *MUS* (negative M_{KX} with respect to Figure 3.5) is the resultant valgus moment exerted by the muscles about the anterioposterior (X) axis through the knee joint in response to an *EXT* that tends to adduct (move into a varus position) the knee.

Studies examining *MUS* about the knee (calculated via inverse dynamics) during landing indicate that females tend to exhibit greater normalised peak knee extension *MUS* (negative M_{KY} with respect to Figure 3.4) (Chappell et al., 2002; Salci et al., 2004; Yu et al., 2006) and greater normalised peak ground reaction force (Yu et al., 2006; Kernozek et al., 2005; Salci et al., 2004) than males. There is very little empirical data available on knee *MUS* in the frontal plane during landing. Chappell et al. (2002) found females to display greater normalised knee valgus *MUS* (negative M_{KX} with respect to Figure 3.5) than males, whereas Kernozek et al. (2005) found females to display lower normalised knee varus *MUS* (positive M_{KX} with respect to Figure 3.5) than males in landing manoeuvres. Table 3.2 shows the results of a number of studies that have reported group mean data for ground reaction force and *MUS* about the knee in landing manoeuvres.

Table 3.2. Group mean data for ground reaction force and muscle moments about the knee in landing manoeuvres in males and females.

Study.	Task	Sagittal plane knee moment.	Frontal plane knee moment.	Ground reaction forces.
Salci et al. (2004)	40 cm and 60 cm vertical drop landing.	F displayed significantly greater peak knee extension *MUS* than M at 40 cm drop landing (M; 0.1±3.2 N.m/kgBM: F; 3.0±2.2 N.m/kgBM).		F exhibited significantly greater normalised peak vertical ground reaction force than M in both 40 and 60 cm drop landing (mean- M: 3.8±0.7 BW: F; 5.4±0.9 BW) .
Decker et al. (2003)	60 cm vertical drop landing.	No significant difference between M and F peak knee extensor *MUS* (M; 17.69±4.57 %BW.ht: F; 15.31±3.3 %BW.ht).		No significant difference between M and F peak normalised vertical ground reaction force (M; 3.67±0.92 BW: F; 3.39±0.89 BW).
Chappell et al. (2002)	Forward, backward and vertical stop-jump landing.	F exhibited a significantly greater knee extension *MUS* than M in all tasks (mean estimated from graphs (+ flex, – ext) M; +0.05±0.2 BW.ht: F; -0.03±0.05 BW.ht).	F displayed a significantly greater knee valgus *MUS* than M in all tasks (mean estimated from graphs (+ var, – val) M; +0.02±0.05 BW.ht: F; -0.02±0.06 BW.ht).	
Kernozek et al. (2005)	60 cm vertical drop landing.	No significant difference between M and F peak knee extension *MUS* (M; 1.75±0.37 N.m/kgBM: F; 1.70±0.27 N.m/kgBM).	F displayed significantly lower peak knee varus *MUS* than M (M; 1.61±0.72 N.m/kgBM: F; 0.93±0.69 N.m/kgBM).	F exhibited significantly greater normalised peak vertical ground reaction force than M (M: 3.51±0.63 BW: F; 4.71±0.71 BW).
Yu et al. (2006)	Stop-jump landing.	F displayed significantly greater peak knee extension *MUS* than M (M; 0.15±0.04 BW.ht; F; 0.18±0.05 BW.ht).		F exerted significantly greater normalised peak vertical ground reaction force than M (M; 2.16±0.60 BW: F; 2.67±0.95 BW).

F = females, M = males.

The greater *EXT* about any particular axis through the knee the greater the resultant *MUS* about the knee joint is likely to be and therefore, the greater the risk of overloading the muscles about the knee joint. Since knee joint stability (i.e., prevention of abnormal joint movement) is maintained by dynamic (contractile) and passive (non-contractile) support structures, the greater the load on the muscles, i.e. dynamic support structures, the greater the extent to which stability of the knee joint is likely to be maintained by the passive support structures, in particular the ACL, PCL, lateral and medial ligaments. If the load on the passive support structures exceeds their strength, injury is likely to occur. Consequently, the reported increased incidence of ACL injury in females during landing movements may be due, in part, to greater peak normalised knee extension *MUS* and greater normalised ground reaction force. Further investigation is needed concerning the influence of *MUS* in the frontal plane during landing/cutting on the gender difference in the incidence of non-contact ACL injury.

ACL injury is likely to occur due to abnormal movement of the tibiofemoral joint. In the sagittal plane, an imbalance of quadriceps muscle force over hamstring muscle force resulting in anterior shear force acting on the proximal end of the tibia is likely to cause an abnormal movement of the tibiofemoral joint (anterior displacement of the tibia relative to the femur) which will increase ACL strain. Since the hamstrings and the ACL work together to prevent anterior displacement of the tibia relative to the femur, the greater the knee flexion *MUS* for a given knee angle, the greater the load on the hamstring muscles and therefore the greater the risk of ACL strain.

The *MUS* about a particular axis through a joint is the sum of the moments of the various muscles which, in turn, depends on both the muscle forces and the moment arms of the muscles. Figure 3.7 shows the forces acting on the proximal end of the tibia due to the ACL, PCL, quadriceps and hamstrings and their moment arms in the sagittal plane when the knee is close to full extension, i.e., when non-contact ACL injury is most common.

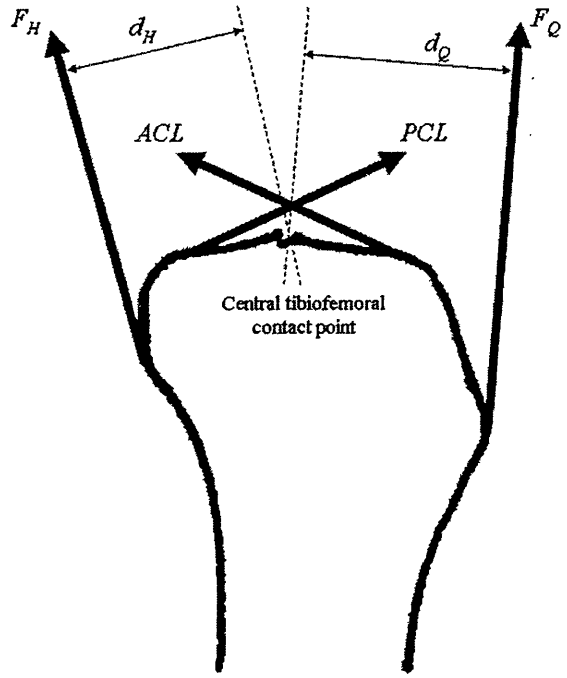


Figure 3.7 The forces acting on the proximal end of the tibia due to the quadriceps and hamstrings and their moment arms in the sagittal plane. F_Q = force exerted by the quadriceps, F_H = force exerted by the hamstrings, d_Q = moment arm of the quadriceps (patella tendon), d_H = moment arm of the hamstrings, *ACL* = force exerted by the ACL and *PCL* = force exerted by the PCL.

Kellis and Baltzopoulos (1999) calculated the moment arms of the patella tendon and the hamstrings for ten male subjects in the sagittal plane during submaximal knee flexion-extension movement at very slow (non constant) angular velocity using videofluoroscopy. Moment arms were taken as the perpendicular distance between the muscle tendon and the central contact point of the tibiofemoral joint. Between 0-10° of knee flexion, the mean moment arm of the patella tendon was found to be 36.9 ± 3.2 mm and the mean moment arm of the hamstrings was found to be 23.9 ± 2.6 mm. Other studies report values ranging from 30 mm to 40 mm for the moment arm of the patella tendon (Herzog and Read, 1993; Grood et al., 1984;

Smidt, 1973) and ranges from 20 mm to 41.3 mm for the moment arm of the hamstrings (Wretenburg et al., 1996; Herzog and Read, 1993; Smidt, 1973). These data suggests that the mechanical advantage of the quadriceps may be greater than that of the hamstrings. Since the hamstrings work with the ACL to prevent anterior dislocation of the proximal tibia relative to the distal femur, this reduced mechanical advantage of the hamstrings relative to the quadriceps may increase the risk of overloading the hamstring muscles, which in turn may cause an anterior shear force on the proximal end of the tibia which may strain the ACL.

Figure 3.8 shows the forces acting on the proximal end of the tibia due to the ACL, PCL, semimembranosus, semitendinosus, gracilis and biceps femoris and their moment arms in the frontal plane when the knee is close to full extension, i.e., when non-contact ACL injury is most common.

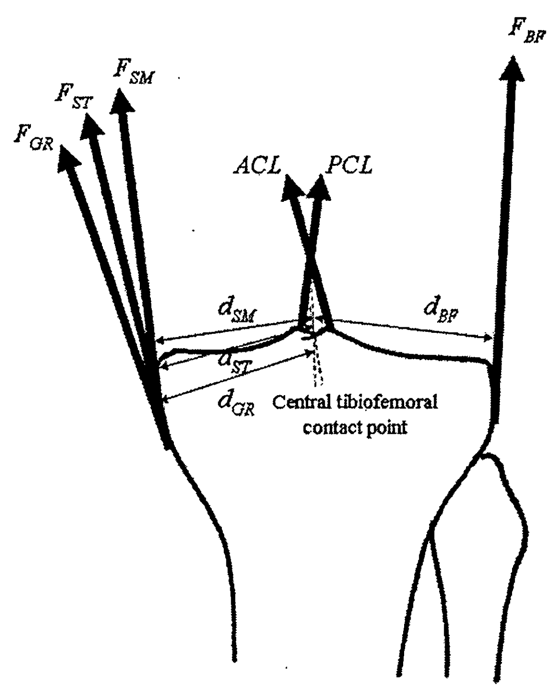


Figure 3.8 Anterior aspect of the proximal end of the left tibia and the forces acting on the proximal end of the tibia due to the semimembranosus,

semitendinosus, gracilis and biceps femoris and their moment arms in the frontal plane. F_{SM} = force exerted by the semimembranosus, F_{ST} = force exerted by the semitendinosus, F_{GR} = force exerted by the gracilis, F_{BF} = force exerted by the biceps femoris, d_{SM} = moment arm of the semimembranosus, d_{ST} = moment arm of the semitendinosus, d_{GR} = moment arm of the gracilis, d_{BF} = moment arm of the biceps femoris, ACL = force exerted by the ACL and PCL = force exerted by the PCL.

Wretenburg et al. (1996) calculated the moment arms of the semimembranosus, semitendinosus, gracilis and biceps femoris in the frontal plane for ten male and seven female subjects using MRI measurements. Moment arms were taken as the perpendicular distance between the muscle tendon and the central contact point of the tibiofemoral joint and were measured with no muscle contraction. Mean data for the moment arms of the semimembranosus, semitendinosus, gracilis and biceps femoris in the frontal plane when the knee is fully extended (0° of flexion) are presented in Table 3.3.

The absolute moment arms of the semimembranosus, semitendinosus, gracilis and biceps femoris in the frontal plane were significantly greater in males than females. Even when normalised to height, the moment arms of all muscles were still greater in males than females. These data suggests that the mechanical advantage of the semimembranosus, semitendinosus, gracilis and biceps femoris in the frontal plane may be greater in males than females. Since these muscles work with the passive support structures of the knee to prevent abnormal movement of the knee joint in the frontal plane, this reduced mechanical advantage in females compared to males

Table 3.3 Mean moment arms of the semimembranosus, semitendinosus, gracilis and biceps femoris in the frontal plane when the knee is fully extended (Wretenburg et al., 1996).

Muscle	Mean moment arm of males		Mean moment arm of females	
	(mm)	(% of height)	(mm)	(% of height)
Biceps femoris	48.8 ± 4.5	2.70 ± 0.25	42.3 ± 3.2	2.56 ± 0.19
Semitendinosus	33.5 ± 7.2	1.85 ± 0.40	24.9 ± 7.2	1.51 ± 0.44
Semimembranosus	29.7 ± 5.0	1.64 ± 0.28	21.6 ± 4.7	1.31 ± 0.28
Gracilis	37.5 ± 5.6	2.07 ± 0.31	32.2 ± 7.7	1.95 ± 0.47

may increase the risk of overloading the semimembranosus, semitendinosus, gracilis and biceps femoris, which in turn may increase the likelihood of an abnormal movement of the knee joint in the frontal plane which may strain the passive support structures of the knee.

The moment arms of the muscles acting about the knee would appear to be fairly similar in the sagittal and frontal planes. However, the structure of the knee joint only allows one main degree of freedom, i.e. angular motion about a mediolateral axis (knee flexion/extension). The normal ranges of motion in the other five degrees of freedom (3 linear planes and 2 angular) are very small. Consequently, the quadriceps and hamstrings facilitate knee flexion and extension, but tend to stabilise the knee with respect to the other 5 degrees of freedom. Therefore, due to the structure of the knee, an *EXT* acting about the knee in the frontal plane is more likely to induce abnormal movement of the knee joint than similar *EXT* in the sagittal plane, which in turn is more likely to overload the stabilising structures (passive and dynamic) of the knee.

3.1.4 Aim.

The aim of the study was to investigate the effects of gender on knee kinematics and kinetics in university volleyball players performing block jump landings in opposed conditions.

The objectives were to determine, during landing from a volleyball block jump in opposed conditions:

i) The angular displacement of the knee joint in the sagittal plane (absolute and relative to normal, static, upright standing, lower limb alignment),

ii) The angular displacement of the knee joint in the frontal plane (absolute and relative to normal static upright standing lower limb alignment),

iii) The linear displacement of the inter-hip, inter-knee and inter-ankle joint centre distances (absolute and relative to normal static upright standing lower limb alignment),

iv) The muscle moment-time histories about the tibiofemoral joint in the sagittal and frontal planes, corresponding to

v) The associated GRF force-time and GRF moment arm-time histories.

3.2 Method.

3.2.1 Subjects.

Six female (Mean age 21.2 ± 1.3 years, mass 57.6 ± 7.5 kg and height 164.8 ± 7.5 cm) and six male (Mean age 21.6 ± 3.3 years, mass 70.1 ± 3.1 kg and height 175.7 ± 8.6 cm) university volleyball players participated in the study. All subjects were right leg dominant and had no previous history of hip, knee or ankle injury. Ethical approval was granted for the study by the University Ethics Committee (Ethical approval form shown in Appendix A, B and C) and written consent forms were signed by all subjects prior to data collection.

3.2.2 Measurement system.

Two adjacent AMTI force platforms embedded into the laboratory floor sampling at 600 Hz were used to measure the ground reaction forces (GRF) on the right and left legs and the centre of pressure (CP) on the right leg during landing (the system was unable to measure the centre of pressure on the left leg). A time synchronised 12 camera Vicon 512 system (Vicon, Oxford, England) sampling at 120 Hz was used to determine 3D coordinates of 39 retro-reflective markers (25 mm diameter).

The laboratory was set up with a rope fixed horizontally to act as a volleyball net at a height of 2.43 m for male subjects and 2.24 m for female subjects (height of a standard volleyball net). The net was placed 5 cm in front of and parallel to the adjacent force platforms (Figure 3.9). In addition to the net, a volleyball was suspended from the ceiling so that it was positioned 5 cm above the height of the net (2.48 m for males and 2.29 m for females) and with the centre of the ball 10 cm in front of the line of the net (the other side of the net to where the subject (blocker) was standing). The ball was positioned vertically above the line separating the two force platforms.

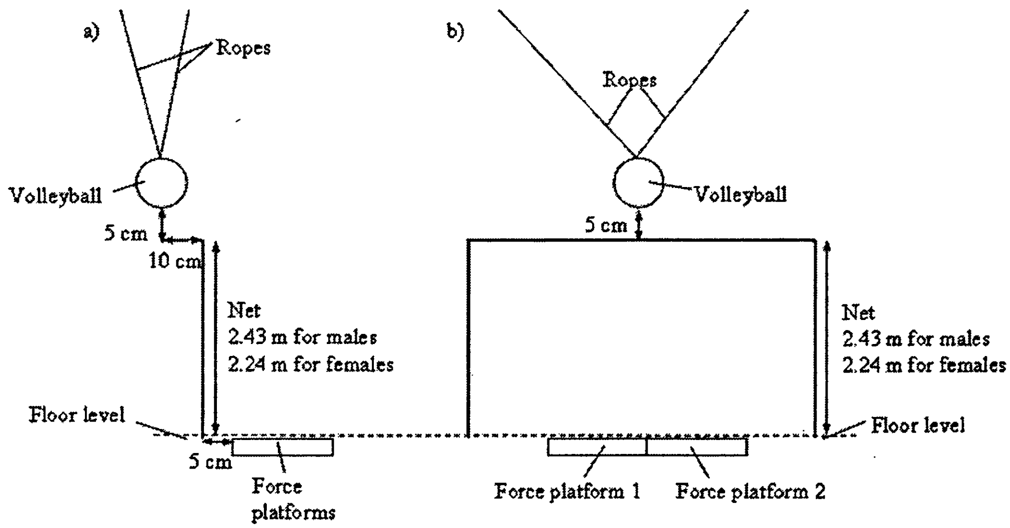


Figure 3.9 Laboratory set up; a) left lateral aspect, b) frontal aspect.

3.2.3 Marker placement.

Markers were placed directly on each subject (on skin or on clothing covering the skin) in accordance with the Vicon system's full body plug-in gait marker set; see Figure 3.10. All subjects wore tight fitting clothing in order to minimise movement of markers relative to the anatomical locations they were intended to designate. The anatomical locations were as follows:

- 4 head markers; right front head (RFHD), left front head (LFHD), right back head (RBHD), left back head (LBHD),
- 7th cervical vertebrae (C7),
- 10th thoracic vertebrae (T10),
- Jugular notch where clavicles meet sternum (CLAV),

- Xiphoid process of the sternum (STRN),
- Middle of right scapula (RBAK),
- Each acromio-clavicular joint (LSHO and RSHO),
- Lateral epicondyle estimating each elbow joint axis (LELB and RELB),
- Upper arms, in line with ELB and SHO (LUPA and RUPA),
- Medial wrists (LWRA and RWRA),
- Lateral wrists (LWRB, RWRB),
- Forearms, in line with ELB and WRA (LFRA and RFRA),
- Head of the second metacarpal of each hand (LFIN and RFIN),
- Anterior superior iliac spines (LASI and RASI),
- Posterior superior iliac spines (LPSI and RPSI),
- Lower lateral surface of each thigh along the line between the hip and knee joint markers (LTHI and RTHI),
- Lateral epicondyle of each femur (LKNE and RKNE),
- Lower lateral surface of each tibia along the line between knee and ankle joint markers (LTIB and RTIB),
- Lateral malleolus of each ankle (LANK and RANK),
- Superior proximal end of the second metatarsal of each foot (LTOE and RTOE),
- Posterior aspect of the Achilles tendon of each leg at the same height as the second metatarsal marker (LHEE and RHEE).

From the location of the markers placed on the body, combined with required anthropometric measurements of each subject entered into the system, the Vicon system calculated the 3D coordinates of hip, knee and ankle joint centres. The

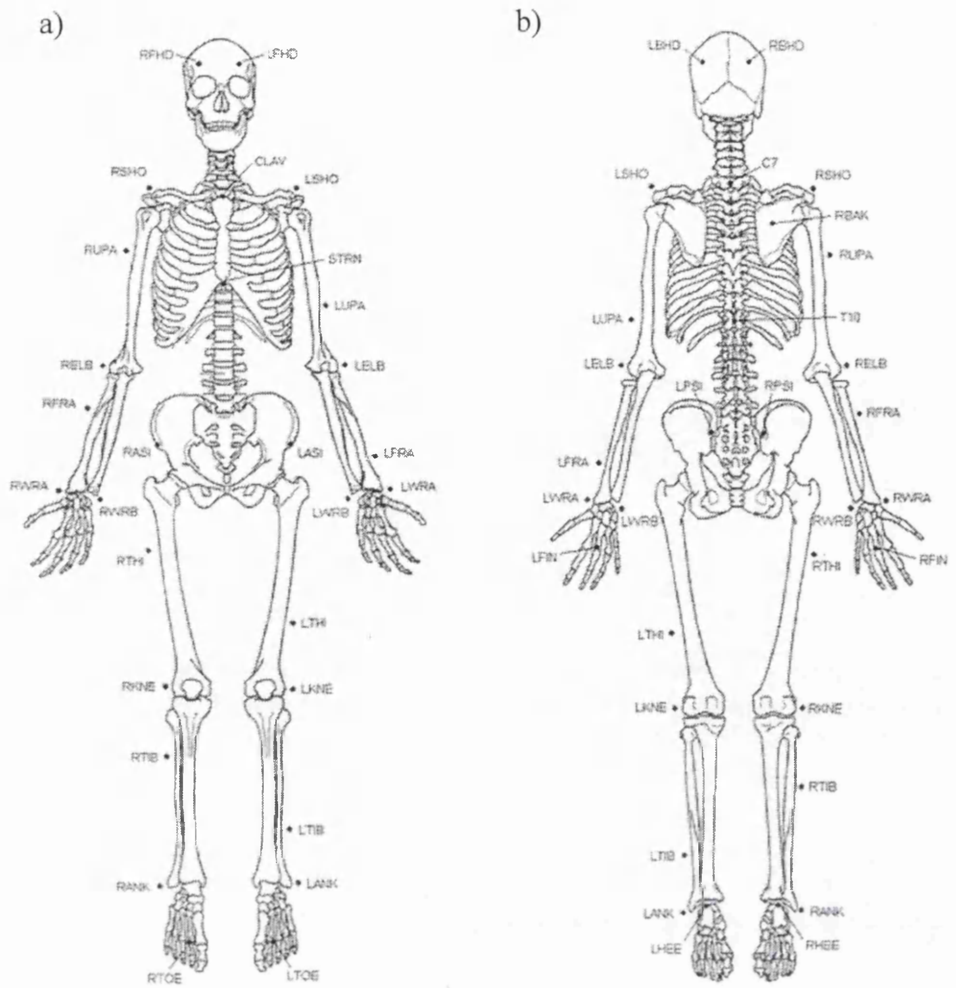


Figure 3.10 Locations of markers in the Vicon full body plug-in gait marker set: a) anterior aspect; b) posterior aspect.

subject anthropometric measurements required were height, mass, leg length, knee width, ankle width, elbow width, wrist width and hand thickness (Table 3.4). The Vicon system determines the position of;

1) Hip joint centre (HIP) from the LASI, RASI, LPSI and RPSI markers and leg length measurement,

2) Knee joint centre (KNEE) from HIP, knee marker, thigh marker and knee width measurement,

3) Ankle joint centre (ANKLE) from KNEE, ankle marker, tibia marker and ankle width measurement.

Table 3.4 Anthropometric measurements.

Measurement	Method
Height (cm)	Stadiometer
Mass (kg)	Weighing scales
Leg length (cm)	Steel tape (anterior superior iliac spine to medial malleolus)
Knee width (cm)	Bone callipers
Ankle width (cm)	Bone callipers
Elbow width (cm)	Bone callipers
Wrist width (cm)	Bone callipers
Hand thickness (cm)	Bone callipers

3.2.4 Angular definitions.

In the Plug-in gait system, the measurement of knee flexion/extension is based on the thigh axis (line connecting the hip joint and knee joint centres) and the shank axis (line connecting the knee and ankle joint centres) projected onto the plane of knee flexion/extension (as determined by the plug-in gait marker system). The flexion/extension angle is the angle between the distal extension of the thigh axis and the shank axis. A positive angle corresponds to knee flexion relative to the fully extended position. Figure 3.11 illustrates the angle measured.

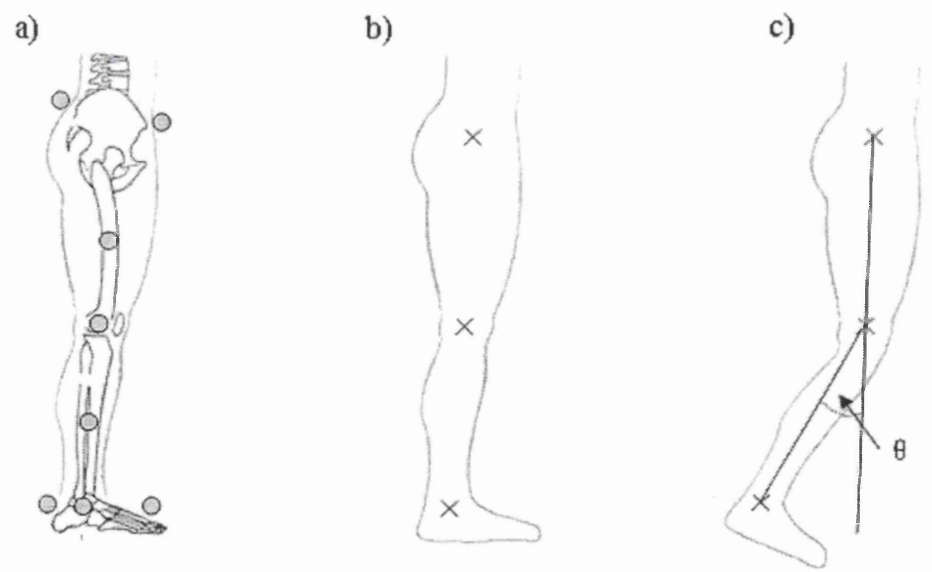


Figure 3.11 Knee flexion/extension: see text for definition. a) Markers placed on skin over bone landmarks. b) Derived estimated joint centres. c) Knee flexion/extension angle θ.

The measurement of knee valgus/varus is based on the thigh axis and the shank axis projected onto the plane of knee valgus/varus (defined as perpendicular to the knee flexion/extension axis). The valgus/varus angle is the angle between the distal extension of the thigh axis and the shank axis. A positive angle indicates varus and a negative angle indicates valgus. Figure 3.12 illustrates the angle measured.

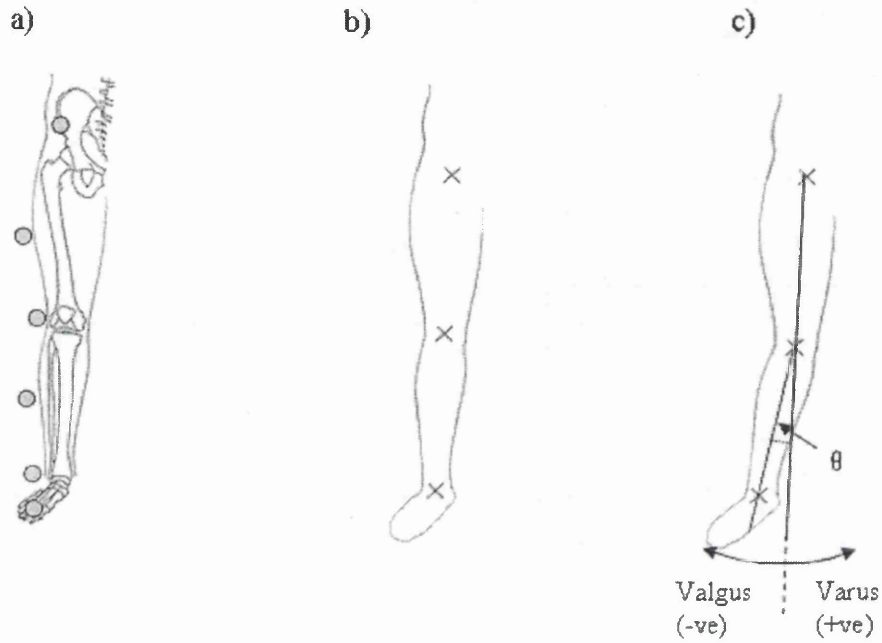


Figure 3.12 Knee valgus/varus: see text for definition. a) Markers placed on skin over bone landmarks. b) Derived estimated joint centres. c) Knee valgus/varus angle θ.

3.2.5 Moment definitions.

The quasi-static model (Figure 3.6a) was used to estimate *MUS* about the knee joint centre of the right leg in the sagittal and frontal planes during landing. The GRF moment was calculated using the cross product $\boldsymbol{r} \times \boldsymbol{F}$ where $\boldsymbol{r}$ = position vector of the point of application of F (centre of pressure) with respect to the knee joint centre and $\boldsymbol{F}$ = ground reaction force vector. In the sagittal plane, a GRF moment that tends to extend the knee is called a GRF knee extensor moment which, using the quasi-static approach, is considered to be equal and opposite to a corresponding flexor moment exerted by the muscles (M_{Kf}) (Figure 3.13a). Similarly, a GRF moment that tends to flex the knee is called a GRF flexor moment and results in a corresponding extensor moment exerted by the muscles (M_{Ke}) (Figure 3.13b). With respect to a right side view of the subject, a negative GRF moment about the Y (mediolateral) axis through the knee joint centre corresponds to a GRF extensor moment and a positive GRF moment about the Y axis of the knee corresponds to a GRF flexor moment (Figure 3.13).

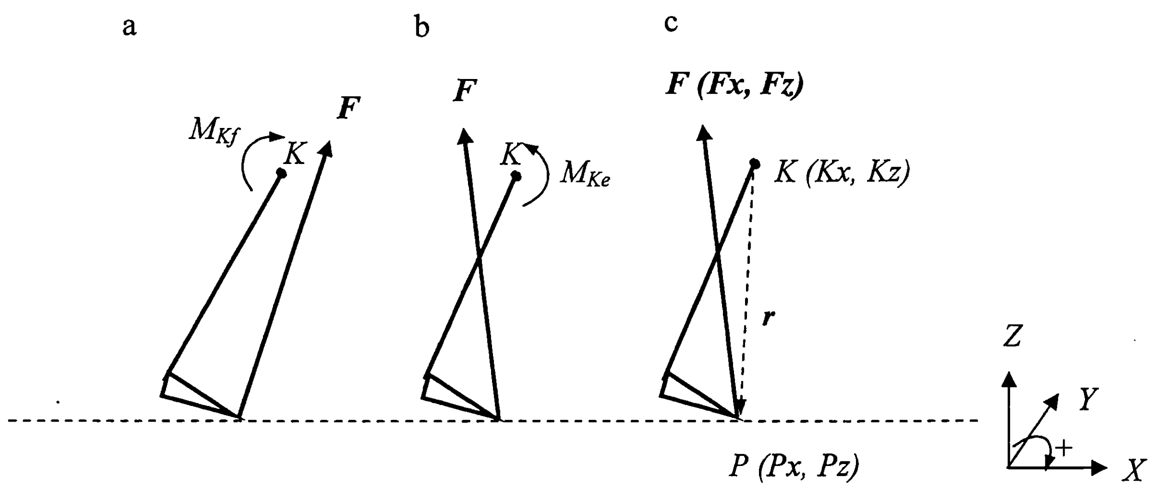


Figure 3.13 Sagittal plane of the right lower leg, a) A GRF extensor moment about the knee (+ ve), b) a GRF flexor moment about the knee (– ve), c) position vector of the centre of pressure (P) with respect to the knee joint centre (K). $\boldsymbol{F}$ = ground reaction force vector, $\boldsymbol{r}$ = position vector of P with respect to K. $\boldsymbol{M_K} = \boldsymbol{r} \times \boldsymbol{F}$.

The *GRF* moment about K in the sagittal plane = $\boldsymbol{r} \times \boldsymbol{F}$

where $\boldsymbol{r}$ = position vector of the centre of pressure with respect to K and $\boldsymbol{F}$ = ground reaction force vector in the sagittal plane (resultant of Fz and Fx), i.e.,

$\boldsymbol{r} \times \boldsymbol{F} = (rx.Fz - rz.Fx)$

where $rx = Px - Kx$

and $rz = Pz - Kz$

The GRF moment arm about the knee in the sagittal plane was calculated by dividing the GRF moment in the sagittal plane by the magnitude of the GRF in the sagittal plane.

In the frontal plane, a GRF moment that tends to adduct the knee (move into a varus position) is called a GRF knee varus moment which, using the quasi-static approach, is considered to be equal and opposite to a corresponding valgus moment exerted by the muscles (M_{valg}) (Figure 3.14a). Similarly, a GRF moment that tends to abduct the knee (move into a valgus position) is called a GRF knee valgus moment and results in a corresponding varus moment exerted by the muscles (M_{var}) (Figure 3.14b). With respect to the right leg of the subject (viewed from the front), a positive GRF moment about the X (anterioposterior) axis through the knee joint centre corresponds to a GRF varus moment and a negative GRF moment about the X axis of the knee corresponds to a GRF valgus moment (Figure 3.14).

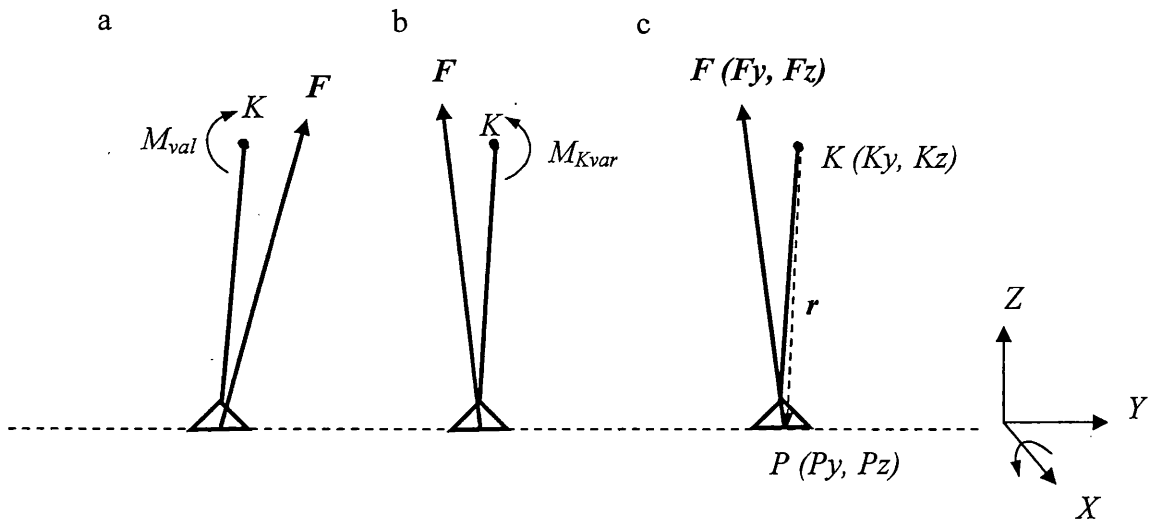


Figure 3.14 Frontal plane of the right lower leg, a) a GRF varus moment about the right knee (+ ve), b) a GRF valgus moment about the right knee (– ve), c) position vector of the centre of pressure (*P*) with respect to the knee joint centre (*K*). $\boldsymbol{F}$ = ground reaction force, $\boldsymbol{r}$ = position vector of P with respect to K. $\boldsymbol{M_K} = \boldsymbol{r} \times \boldsymbol{F}$.

Moment of *GRF* about *K* in the frontal plane = $\boldsymbol{r} \times \boldsymbol{F}$

where $\boldsymbol{r}$ = position vector of the centre of pressure with respect to *K* and $\boldsymbol{F}$ = ground reaction force vector in the frontal plane (resultant of *Fz* and *Fy*), i.e.,

$\boldsymbol{r} \times \boldsymbol{F} = (ry.Fz - rz.Fy)$

where $ry = Py - Ky$

and $rz = Pz - Kz$

The GRF moment arm about the knee in the frontal plane was calculated by dividing the GRF moment in the frontal plane by the magnitude of the GRF in the frontal plane.

3.2.6 Static reference position.

Prior to dynamic trials, a static trial was recorded for each subject while standing in the normal upright position. The purpose of the static trial was to provide reference data (angles and distances between joint centres during normal upright standing) for knee flexion/extension, knee valgus/varus, inter-hip distance (distance between the hip joint centres), inter-knee distance (distance between the knee joint centres) and inter-ankle distance (distance between the ankle joint centres) in order to facilitate analysis of knee motion in dynamic trials relative to normal upright standing. Subjects were instructed to stand still with their feet placed apart at a standardised distance of 10% of their leg length (i.e. for a leg length of 1.1 m, the feet were placed 11 cm apart). Arms were held horizontally (shoulders abducted 90°) in order to prevent the markers on the pelvis being obscured. A skeleton figure of a subject in the testing position is shown in Figure 3.15. Each trial lasted 5 seconds with only one second (4-5 s) of each trial being used for analysis.

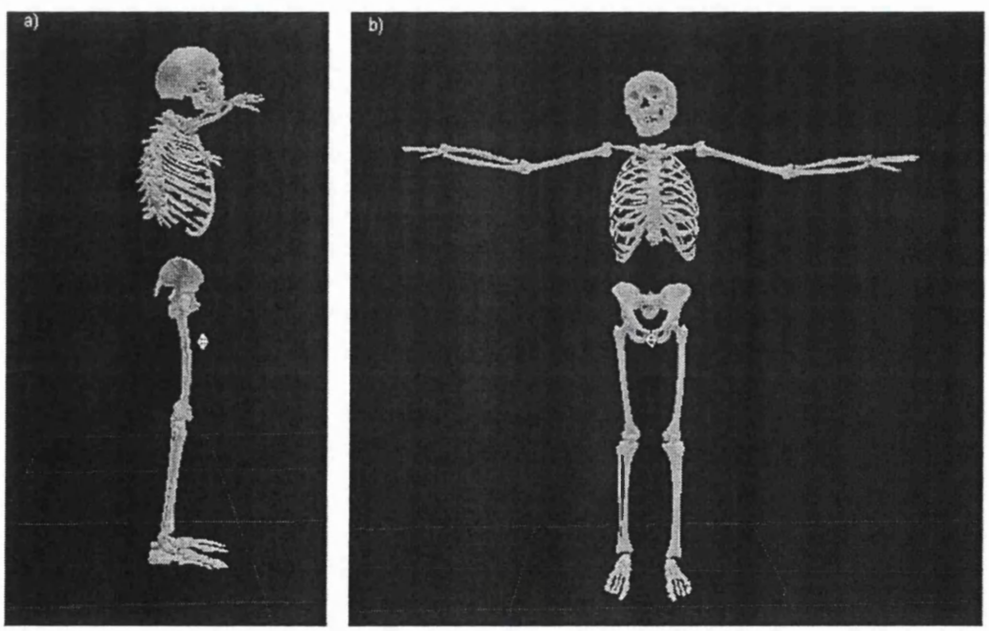


Figure 3.15 Skeleton figure of subject in the static reference position: a) right lateral aspect; b) anterior aspect.

Whereas the inter-hip distance was essentially fixed, changes in knee flexion/extension, knee valgus/varus, inter-knee and inter-ankle distances were expected during dynamic trials. In order to compare these changes between subjects, the relative and absolute knee flexion/extension, knee valgus/varus, inter-knee and inter-ankle distances were measured. The relative inter-joint distances were calculated as a percentage of the corresponding static reference value using the following formula:

$$d_{rel} = \frac{d_{dyn}}{d_{sta}} \times 100 \qquad (3.11)$$

where d_{rel} = relative inter-joint distance,

d_{dyn} = inter-joint distance measured during dynamic trials,

d_{sta} = inter-joint distances measured during the static reference trial.

Since the knee flexion/extension and knee valgus/varus angles measured during the static reference trial were small compared to values measured during dynamic trials, relative values calculated as a percentage of the corresponding static reference value were very large and difficult to interpret. Consequently, for knee flexion/extension and valgus/varus, relative values were calculated by subtracting the static reference value from the value measured during dynamic trials, as shown in equation 3.12.

$$\theta_{rel} = \theta_{dyn} - \theta_{sta} \qquad (3.12)$$

where θ_{rel} = relative angle,

θ_{dyn} = angle measured during dynamic trials,

θ_{sta} = angle measured during the static reference trial.

2.7 Landing Task.

Prior to data collection all subjects performed a 10-min warm up consisting of lower limb stretching and running/jogging on a treadmill at self determined speeds. When this was completed, subjects practised the jumping and landing task until comfortable with the procedure.

The jumping and landing task was made as realistic as possible by having subjects attempt to block an actual volleyball spike shot performed by an experienced volleyball player. A volleyball spike shot involves hitting a volleyball at a sharp angle into the opponent's court by jumping near the net and hitting the ball down hard from above. At the start of each trial, the subject stood with each foot on a separate force plate. The subject then timed his/her blocking action in order to try to block the ball as it was spiked. The ball was spiked from the same suspended position in order to eliminate variation in the position and velocity of the ball. On landing, each foot landed on a separate force plate. Data were recorded for three successful trials for each subject. Trials where each foot did not land entirely on a separate force plate were discarded.

3.2.8 Processing of the 3D coordinate data.

Since the raw data output by the Vicon system were noisy, data were smoothed by filtering prior to analysis. The 3D coordinate data were filtered using a Woltring Filter. To alter the filter settings a mean squared error (MSE) tolerance value was entered into the Vicon system. The MSE method allows the noise level to be input, and a spline function is fitted to the data points in accordance with the specified level of tolerance. Consistent application of this processing method ensured the same level of smoothing for all marker trajectories. Based on a primary consideration of minimising high frequency artefacts whilst maintaining the detail of the signal at all lower frequencies, it was determined that it would be most appropriate to use a MSE value of 50 as a suitable setting for filtering the data. This was determined by analysing the effects of a number of different filter settings for sample data of a number of different jumps and from a number of different subjects. In determining a suitable MSE value, the data were analysed using a Welch periodogram with nFFT (number of points used in the fast Fourier transform) = 512, a Hanning window with window length = 128 and overlap = 64 (50%) to provide power spectral density (PSD) plots. PSD plots quantify the magnitude of power in a narrow frequency band (in this case the bandwidth is 1/120 Hz). From the PSD plots, the estimated lowest frequency with appreciable signal attenuation, frequencies with 50% of signal attenuation and almost complete signal attenuation could be determined for the MSE value of 50. The filter setting determined to be most appropriate for these data (i.e. MSE = 50) corresponded to a low-pass filter of cut-off frequency 10 Hz and stop-band frequency of 30 Hz.

3.2.9 Data analysis.

The angular displacement of the knee in the sagittal (flexion/extension) and frontal (valgus/varus) planes was determined between initial ground contact (IC) and either (depending upon which occurred later in each trial) maximum flexion or maximum valgus angle (MAX). The angular displacement – time data were then normalised with respect to average trial time for both legs combined and for separate dominant and non-dominant legs. Angular displacement mean data (IC, MAX and range of motion (ROM)) for combined dominant and non-dominant legs were based on 36 trials for males and 36 trials for females (6 subjects × 3 trials × 2 legs). Angular displacement mean data for each leg (dominant and non-dominant) were based on 18 trials for males and 18 trials for females (6 subjects × 3 trials × 1 leg).

Absolute and relative inter-hip, inter-knee and inter-ankle distances were also determined and normalised with respect to average trial time. Inter-joint mean data (IC, minimum (MIN) and range of motion (ROM)) were based on 18 trials for males and 18 trials for females (6 subjects × 3 trials).

The GRF, moment arm of the GRF about the knee and the moment of the GRF about the knee of the right leg in the sagittal (flexion/extension) and frontal (valgus/varus) planes were determined between IC and MAX in each trial. All data were then normalised with respect to average trial time. GRF was normalised to body weight (in newtons), moment arm normalised to height (in metres) and knee moments were normalised to body weight (in newtons) and height (in metres). GRF, moment arm of the GRF about the knee and moment of the GRF about the knee mean data (IC and MAX) were based on 18 trials for males (6 subjects × 3 trials × 1 leg) and 18 trials for females (6 subjects × 3 trials × 1 leg).

Independent-samples t-tests were carried out on the angular displacement, inter-joint distances, GRF, GRF moment arm and GRF moment about the knee data in the sagittal and frontal planes to examine gender differences and differences between dominant and non-dominant legs. Prior to statistical analysis, data were assessed for normality using the Lilliefors test. All data were found to display normal distribution. Furthermore, Bartlett's test for equality of variances determined whether the variation of the data for the two groups (males and females) were the same. If the significance value for Bartlett's test was greater than 0.05, equal variances were assumed and the appropriate significance value was identified to indicate significant differences between groups. However, if the significance value was 0.05 or less for Bartlett's test, equal variances were not assumed and the results for the assumption of non-equal variance between groups were referred to. All parameters displayed equal variances between groups other than maximum GRF in the sagittal and frontal planes, sagittal plane GRF angle at AL and moment arm and moment in the frontal plane at ML.

The results of Chapter 2 were used to select an appropriate level of precision for reporting these data. This was calculated using the following equation:

$$\text{Level of precision} = \frac{\max \mathrm{CV}}{100} \times \sigma$$

Where maxCV = maximum coefficient of variation for that variable measured in Chapter 2, σ = value measured in the present experimental study.

For example, for angles, maxCV = 7.7%, mean knee flexion angle for males = 4.68°, therefore:

$$\text{Level of precision} = \frac{7.7}{100} \times 4.68 = 0.36$$

Angles are therefore precise to the nearest 0.4°, and are reported to 1 decimal place (DP).

This procedure was used to evaluate the number of DP separately for each type of parameter (angle, distance, force, normalised moment arm and normalised moment). For variables which are normalised to height and/or weight (ground reaction force, moment arm and moment), the level of precision is divided by mean height and/or weight. It was concluded that angles are reported to 1 DP, distance (in mm) reported to 0 DP, normalised force reported to 3 DP, normalised moment arm reported to 3 DP and normalised moment reported to 4 DP.

3.3 Results.

All Figures show variables plotted against normalised time and against absolute mean trial time between IC and MAX. Absolute mean contact time was 0.190 ± 0.040 s for males and 0.194 ± 0.057 s for females. As there was no significant difference between contact time for males and females, mean contact time of 0.192 s was used. Static reference data are reported in Table 3.5. There was no significant difference between males and females knee flexion/extension, knee valgus/varus, inter-hip distance, inter-knee distance or inter-ankle distance in the static reference position.

Table 3.5 Group mean results for knee flexion/extension, knee valgus/varus, inter-hip distance, inter-knee distance and inter-ankle distance in the static reference position for males and females (Mean ± standard deviation) *.

		Male	Female
Knee flexion (+ve) / extension (–ve) (°)	Left	4.3 ± 5.8	2.9 ± 3.3
	Right	5.1 ± 2.6	2.6 ± 4.4
	Mean of left and right	4.7 ± 4.2	2.8 ± 3.7
Knee varus (+ve) / valgus (–ve) (°)	Left	-0.2 ± 3.6	-1.4 ± 3.5
	Right	-2.7 ± 3.6	-2.7 ± 2.5
	Mean of left and right	-1.4 ± 3.7	-2.1 ± 2.9
Inter-joint distances (mm)	Inter-hip distance	160 ± 15	154 ± 14
	Inter-knee distance	147 ± 17	163 ± 9
	Inter-ankle distance	158 ± 25	169 ± 16

* No significant differences between males and females in the static reference position.

3.3.1 Sagittal plane kinematics.

3.3.1.1 Absolute changes in knee flexion.

In the sagittal plane, females exhibited significantly less knee flexion at IC (males: 19.4 ± 6.4°; females: 15.1 ± 6.2°) and greater MAX knee flexion (males: 62.1 ± 11.6°; females: 68.2 ± 12.2°) which resulted in a significantly greater ROM of knee flexion (males: 42.7 ± 13.9°; females: 53.1 ± 13.1°) than males (Table 3.6 and Figure 3.16). Males and females showed no significant difference in sagittal plane kinematics between dominant and non-dominant legs during landing (Table 3.7 and Figures 3.17 and 3.18). The magnitude of the standard deviation of the knee flexion data (combined and for each leg) at 1% normalised time intervals were very similar between IC and MAX. This is illustrated in Figure 3.16.

Table 3.6 Group mean results for absolute and relative knee flexion/extension and valgus/varus (– varus; + valgus) angles at IC, MAX and ROM (Mean ± standard deviation).

		Males		Females	
		Absolute (°)	Relative (°)	Absolute (°)	Relative (°)
Flexion	IC	19.4 ± 6.4[1]	14.7 ± 6.4	15.1 ± 6.2[1]	12.4 ± 6.2
	MAX	62.1 ± 11.6[2]	57.4 ± 11.6[3]	68.2 ± 12.2[2]	65.5 ± 12.2[3]
	ROM	42.7 ± 13.9[4]	N/A	53.1 ± 13.1[4]	N/A
Valgus/varus	IC	-2.8 ± 5.9	-1.4 ± 5.9	-1.6 ± 2.8	0.5 ± 2.8
	MAX_{VAL}	-2.9 ± 7.9[5]	-1.5 ± 7.9[6]	-10.4 ± 7.7[5]	-8.3 ± 7.7[6]
	MAX_{VAR}	0.6 ± 9.1	2.0 ± 9.1	N/A	N/A
	ROM	3.5 ± 9.6[7]	N/A	8.8 ± 7.8[7]	N/A

[1-7]: Significant difference between males and females ($p < 0.05$).

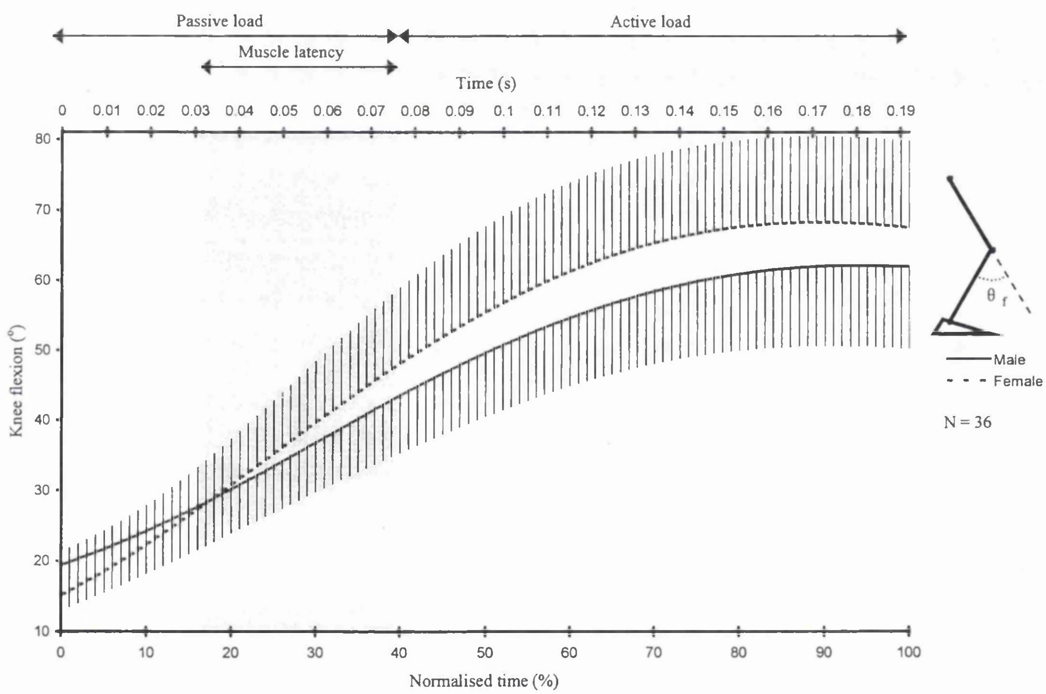


Figure 3.16 Knee flexion (θ_f) between IC and MAX for males and females. The standard deviation at 1% normalised time intervals in indicated by the vertical lines.

Table 3.7 Group mean results for absolute knee flexion/extension and valgus/varus (– varus; + valgus) at IC, MAX and for ROM for male and female subjects' dominant and non-dominant legs (Mean ± standard deviation).

		Males		Females	
		Non-dominant	Dominant	Non-dominant	Dominant
Flexion (°)	IC	17.1 ± 6.4	21.7 ± 5.7	16.7 ± 6.1	13.5 ± 6.0
	MAX	61.2 ± 12.3	63.0 ± 11.2	68.3 ± 14.7	68.2 ± 9.5
	ROM	44.1 ± 15.1	41.3 ± 13.0	51.6 ± 13.9	54.7 ± 12.5
Valgus/varus (°)	IC	-4.0 ± 5.6	-1.6 ± 6.1	-1.1 ± 2.7	-2.1 ± 3.0
	MAX	2.5 ± 8.9	-1.4 ± 9.2	-13.9 ± 8.7[1]	-6.8 ± 4.5[1]
	ROM	6.5 ± 12.0	0.2 ± 5.2	12.8 ± 7.6[2]	4.7 ± 5.8[2]

[1+2] significant difference between dominant and non-dominant legs in females ($p < 0.05$).

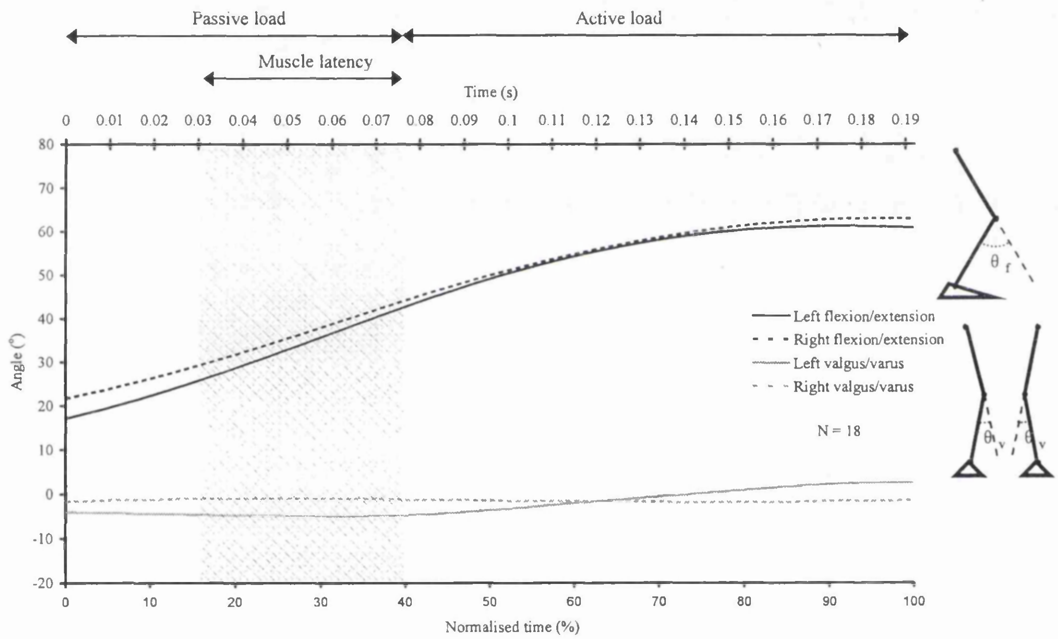


Figure 3.17 Dominant and non-dominant leg knee flexion/extension (θ_f) and valgus (–ve) / varus (+ve) (θ_v) between IC and MAX for males.

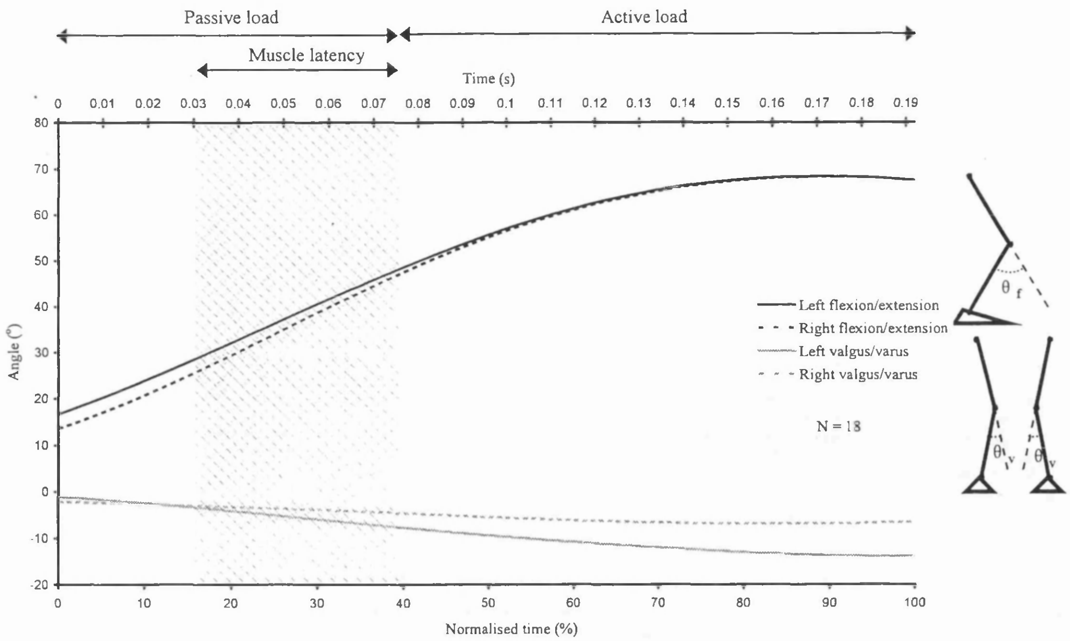


Figure 3.18 Dominant and non-dominant leg knee flexion/extension (θ_f) and valgus (–ve) / varus (+ve) (θ_v) between IC and MAX for females.

The average angular velocity of knee flexion during the two main phases of the landing, passive loading and muscle latency phase (PP) and active loading phase (AP) were determined for males and females. Females displayed significantly greater average knee flexion angular velocity than males during PP (males: 312.8°/s; females: 428.4°/s), but average knee flexion angular velocity during AP was not significantly different between males and females (males: 162.8°/s; females: 172.7°/s) (Table 3.8). The slopes of the curves in Figure 3.16 also indicate that peak knee flexion angular velocity (peak gradient of curve) occurred at approximately 25% of normalised time for males and females but peak knee flexion angular velocity was significantly greater in females than males (males: 335.2°/s; females: 439.6°/s) (gradient of knee flexion – time curves in muscle latency period in Figure 3.16 and Table 3.8).

Table 3.8 Angular velocity of knee flexion and valgus/varus during the two main phases of the landing, i.e. passive load and muscle latency phase and active phase.

		Passive load and muscle latency phase						
		t_{IC} (s)	t_{PP} (s)	Δt (s)	θ_{IC} (°)	θ_{PP} (°)	$\Delta\theta$ (°)	ω (°/s)
Flexion	Males	0	0.075	0.075	19.4	42.9	23.5	312.8[1]
	Females	0	0.075	0.075	15.1	47.2	32.1	428.4[1]
Valgus/varus	Males	0	0.075	0.075	-2.8	-2.9	0.1	1.7[2]
	Females	0	0.075	0.075	-1.6	-6.1	4.5	59.7[2]
		Active phase						
		$t_{PP'}$ (s)	t_{MAX} (s)	Δt (s)	θ_{PP} (°)	θ_{MAX} (°)	$\Delta\theta$ (°)	ω (°/s)
Flexion	Males	0.075	0.192	0.117	42.9	61.9	19.1	162.8
	Females	0.075	0.192	0.117	47.2	67.5	20.2	172.7
Valgus/varus	Males	0.075	0.192	0.117	-2.9	0.6	3.5	29.7
	Females	0.075	0.192	0.117	-6.1	-10.2	4.2	35.8

[1+2]: significant difference between males and females ($p < 0.05$).

t_{IC} = time at IC; t_{PP} = duration of PP; t_{MAX} = time at MAX; θ_{IC} = angle at IC; θ_{PP} = angle at end of PP; θ_{MAX} = angle at MAX; ω = average angular velocity.

3.3.1.2 Relative changes in knee flexion.

Relative to the static reference position in the sagittal plane, there was no significant difference between males and females in knee flexion at IC (males: 14.7 ± 6.4°; females: 12.4 ± 6.2°). Females, however, displayed significantly greater MAX knee flexion than males (males: 57.4 ± 11.6°; females: 65.5 ± 12.2°) (Table 3.6).

3.3.2 Frontal plane kinematics.

3.3.2.1 Absolute changes in knee valgus/varus.

In the frontal plane, females tended to contact the ground in a slight valgus position (–ve values) (-1.6 ± 2.8°) which progressively increased between IC and MAX position (-10.4 ± 7.7°). In contrast, males tended to contact the ground in a slight valgus position (-2.8 ± 5.9°) and moved into a slight varus position (+ve values)

(0.6 ± 9.1°) at MAX (Table 3.6 and Figure 3.19). The amount of valgus at IC was not significantly different between males (-2.8 ± 5.9 °) and females (-1.6 ± 2.8°). However, the ROM (males: 3.5 ± 9.6°; females: 8.8 ± 7.8°) and the MAX valgus angle (males: -2.9 ± 7.9°; females: -10.4 ± 7.7°) were significantly greater in females compared to males (Table 3.6 and Figure 3.19). Males showed no significant difference in frontal plane kinematics between dominant and non-dominant legs during landing (Table 3.7 and Figure 3.17). However, females' non-dominant leg displayed significantly greater maximum knee valgus angle (non-dominant: -13.9 ± 8.7°; dominant: -6.8 ± 4.5°) and range of motion (non-dominant: 12.8 ± 7.6°; dominant: 4.7 ± 5.8°) compared to the dominant leg (Table 3.7 and Figures 3.18).

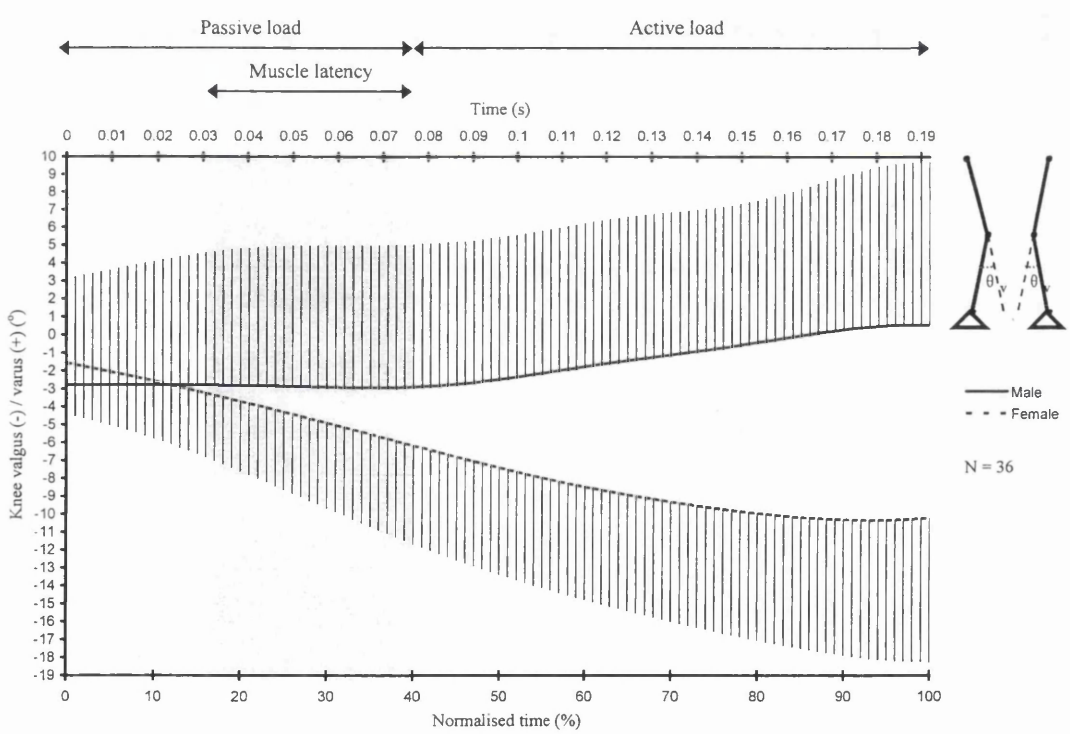


Figure 3.19 Knee valgus/varus (θ_v) between IC and MAX for males and females. The standard deviation at 1% normalised time intervals in indicated by the vertical lines.

The average angular velocity of knee valgus/varus during PP and AP were determined for males and females (Table 3.8). Females displayed significantly greater average knee valgus angular velocity than males during PP (males: 1.7°/s, associated with increasing valgus; females: 59.7°/s, associated with increasing valgus). During AP, the average knee varus angular velocity exhibited by males was similar to the average knee valgus angular velocity exhibited by females (males: 29.7°/s, associated with decreasing valgus: -2.9° to 0.6°; females: 35.8°/s, associated with increasing valgus: -6.1° to -10.2°).

3.3.2.2 Relative changes in knee valgus/varus.

In the frontal plane, the relative amount of valgus at IC was not significantly different between males and females (males: -1.4 ± 5.9°; females: 0.5 ± 2.8°). Females, however, displayed significantly greater MAX knee valgus angle than males (males: -1.5 ± 7.9°; females: -8.3 ± 7.7°) (Table 3.6).

3.3.3 Inter-hip, inter knee and inter ankle displacements.

3.3.3.1 Absolute changes in inter-joint displacements.

Due to the hip joint centres being fixed in the pelvis, it was not surprising that the inter-hip distance remained constant throughout the landing (Table 3.9 and Figure 3.20). There was no significant difference in absolute inter-hip distance between males and females in the static reference position (males: 160 ± 15 mm; females: 154 ± 14 mm), at IC (males: 160 ± 15 mm; females: 154 ± 14 mm) or at MIN (males: 160 ± 15 mm; females: 154 ± 14 mm) (Table 3.9). There was no significant

Table 3.9 Group mean results for absolute and relative (to normal upright standing alignment) inter-hip, inter-knee and inter-ankle distances at IC, minimum distance (MIN) and for range of motion during landing (ROM) (Mean ± standard deviation).

		Males		Females	
		Absolute (mm)	Relative (%)	Absolute (mm)	Relative (%)
Inter-hip distance	IC	160 ± 15	100	154 ± 14	100
	MIN	160 ± 15	100	154 ± 14	100
Inter-knee distance	IC	244 ± 33	165.4 ± 26.6[1]	228 ± 29	140.1 ± 14.6[1]
	MIN	234 ± 39[2]	158.8 ± 28.7[3]	200 ± 35[2]	123.0 ± 20.6[3]
	ROM	10 ± 16[4]	6.6 ± 12.1[5]	28 ± 18[4]	17.1 ± 10.9[5]
Inter-ankle distance	IC	311 ± 58	196.5 ± 73.3	289 ± 46	170.4 ± 19.5
	MIN	269 ± 59	170.2 ± 68.9	267 ± 46	156.4 ± 20.7
	ROM	42 ± 27[6]	26.3 ± 17.4[7]	24 ± 16[6]	14.0 ± 9.1[7]

[1-7]: Significant difference between males and females ($p < 0.05$).

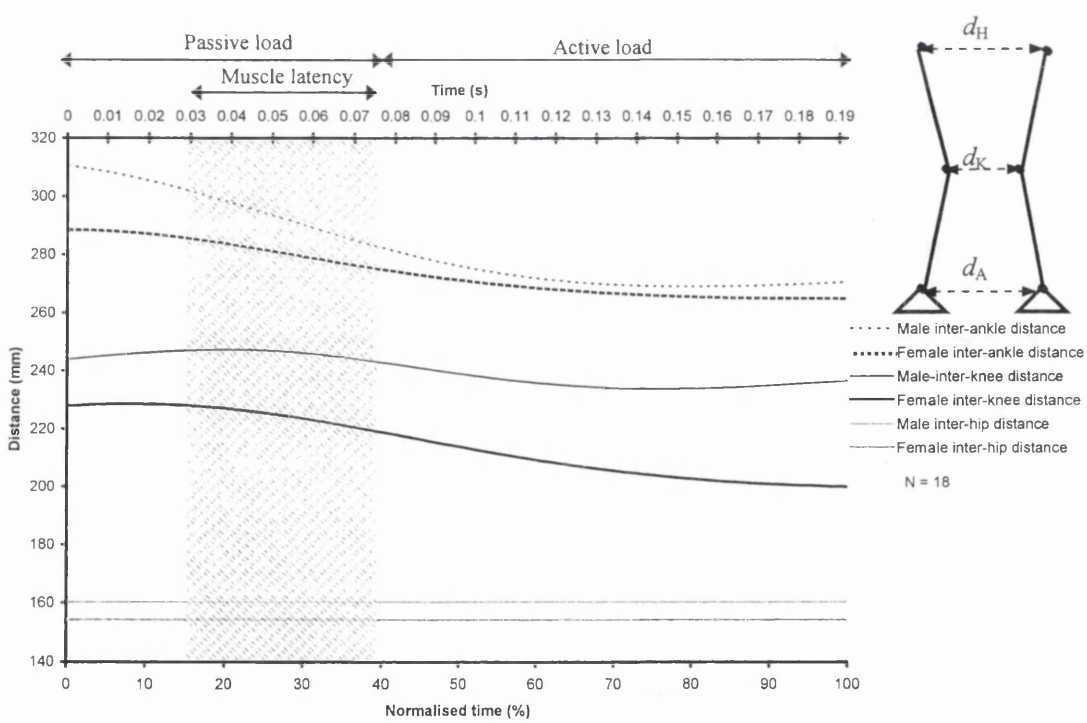


Figure 3.20 Absolute inter-hip (d_H), inter-knee (d_K) and inter-ankle (d_A) joint centre distances between IC and MAX for males and females.

difference in absolute inter-knee distance at IC between males and females (males: 244 ± 33 mm; females: 228 ± 29 mm). The absolute minimum inter-knee distance was significantly longer for males than females (males: 234 ± 39 mm; females: 200 ± 35 mm) and the change in the inter-knee distance between IC and MAX was significantly smaller for males than females (males: 10 ± 16 mm; females: 28 ± 18 mm). There was no significant difference between males and females in absolute inter-ankle distance at IC (males: 311 ± 58 mm; females: 289 ± 46 mm) or minimum distance (males: 269 ± 59 mm; females: 265 ± 46 mm). However, the change in inter-ankle distance between IC and MAX was significantly greater in males than females (males: 42 ± 27 mm; females: 23.71 ± 16.47 mm) (Table 3.9 and Figure 3.20). The mean positions of hip, knee and ankle joint centres of both legs at IC and MIN for males and females are shown in Figure 3.21.

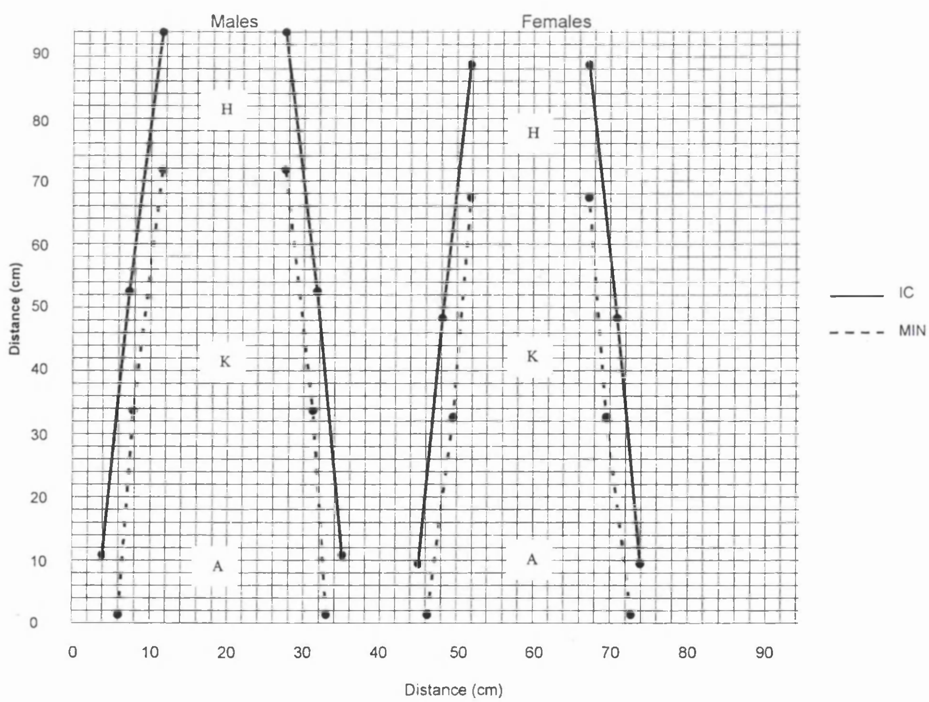


Figure 3.21 Mean positions of hip (H), knee (K) and ankle (A) joint centres of both legs at IC and MIN for males and females.

3.3.3.2 Relative changes in inter-joint displacements.

The changes in relative inter-joint distances were similar to the changes in absolute inter-joint distances. Whereas there was no significant difference in absolute inter-knee distance at IC between males and females, the relative inter-knee distance at IC was significantly longer in males than females (males: 165.4 ± 26.6%; females: 140.1 ± 14.6%) (Table 3.9 and Figure 3.22). The relative minimum inter-knee distance was significantly longer in males than females (males: 158.8 ± 28.7%; females: 123.0 ± 20.6%) and the change in the relative inter-knee distance between IC and MAX was significantly smaller in males than females (males: 6.6 ± 12.1%; females: 17.1 ± 10.9%). There was no significant difference between males and females in relative inter-ankle distance at IC (males: 196.5 ± 73.3%; females: 170.4 ± 19.5%) or relative minimum distance (males: 170.2 ± 68.9%; females: 156.4 ± 20.7%). However, the change in relative inter-ankle distance was significantly greater in males than females (males: 26.3 ± 17.4%; females: 14.0 ± 9.1%) (Table 3.9 and Figure 3.22).

Inter-knee distance relative to inter-hip distance was greater in males than females at IC (males: 152.3 ± 20.6%; females: 147.7 ± 19.1%) and MIN (males: 146.2 ± 24.6%; females: 129.7 ± 22.4%) (Table 3.10). However, inter ankle distance relative to inter-hip distance was greater in males than females at IC (males: 193.9 ± 36.5%; females: 187.0 ± 30.0%) but less in males than females at MIN (males: 168.3 ± 36.6%; females: 171.8 ± 29.7%).

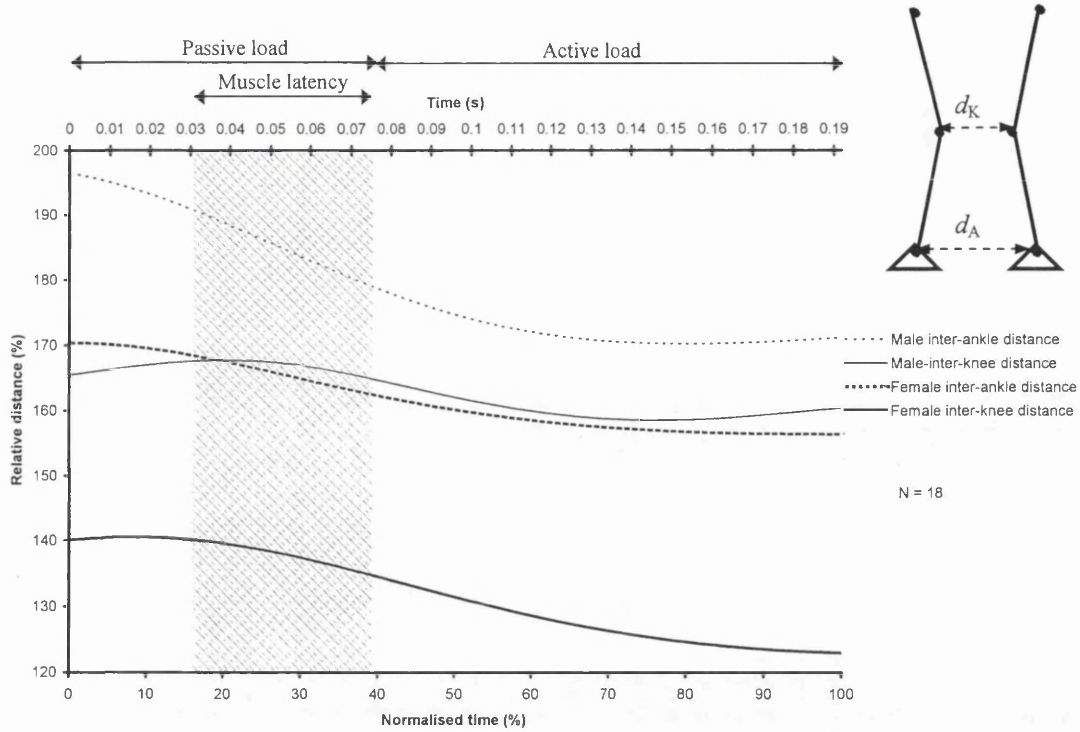


Figure 3.22 Relative inter-knee (d_K) and inter-ankle (d_A) joint centre distances between IC and MAX for males and females.

Table 3.10 Inter-knee and inter-ankle distances at IC and MIN relative to inter-hip distance (mean ± standard deviation).

	Males	Females
A: Inter-hip distance at IC (mm)	160 ± 15	154 ± 14
B: MIN inter-hip distance (mm)	160 ± 15	154 ± 14
C: Inter-knee distance at IC (mm)	244 ± 33	228 ± 29
D: Inter-ankle distance at IC (mm)	311 ± 58	289 ± 46
E: MIN inter-knee distance (mm)	234 ± 39	200 ± 35
F: MIN inter-ankle distance (mm)	269 ± 58	265 ± 46
C/A x 100 (%)	152.3 ± 20.6	147.7 ± 19.1
D/A x 100 (%)	193.9 ± 36.5	187.0 ± 30.0
E/B x 100 (%)	146.2 ± 24.6	129.7 ± 22.4
F/B x 100 (%)	168.3 ± 36.6	171.8 ± 29.7

3.3.4 Sagittal plane kinetics.

Group mean curves for normalised GRF (divided by body weight (BW)), angle of the GRF, knee angle, normalised GRF moment arm (divided by height (ht)) and normalised GRF moment (divided by body weight and height (BW.ht)) (+ve = extensor GRF moment, – ve = flexor GRF moment) throughout the landing period are shown for males and females in the sagittal plane in Figure 3.23.

With regard to normalised GRF (Figure 3.23a), the overall shapes of the curves were similar for males and females, i.e. increase during PP followed by decrease during AP. For most of the landing period, the normalised GRF was greater for males than females. The main difference between males and females occurred during PP where females exhibited a smaller initial peak which also occurred earlier in the landing phase than in males. Females exhibited significantly greater normalised GRF at IC than males (males: 0.049 ± 0.085 BW; females: 0.133 ± 0.120 BW). There was no significant difference between males and females' normalised GRF at the start of the muscle latency period (ML), the start of the active loading period (AL), at MAX or maximum normalised GRF.

With regard to the angle of the GRF ($>90^{\circ}$ = posterior to the vertical, $<90^{\circ}$ = anterior to the vertical) during landing (Figure 3.23b), the shape of the graph was similar in males and females. However, the decrease in the angle of the GRF during PP occurred earlier in females than in males. Males and females showed no significant difference in sagittal plane GRF angle at IC, ML, AL, MAX or maximum and minimum values (Figure 3.23b).

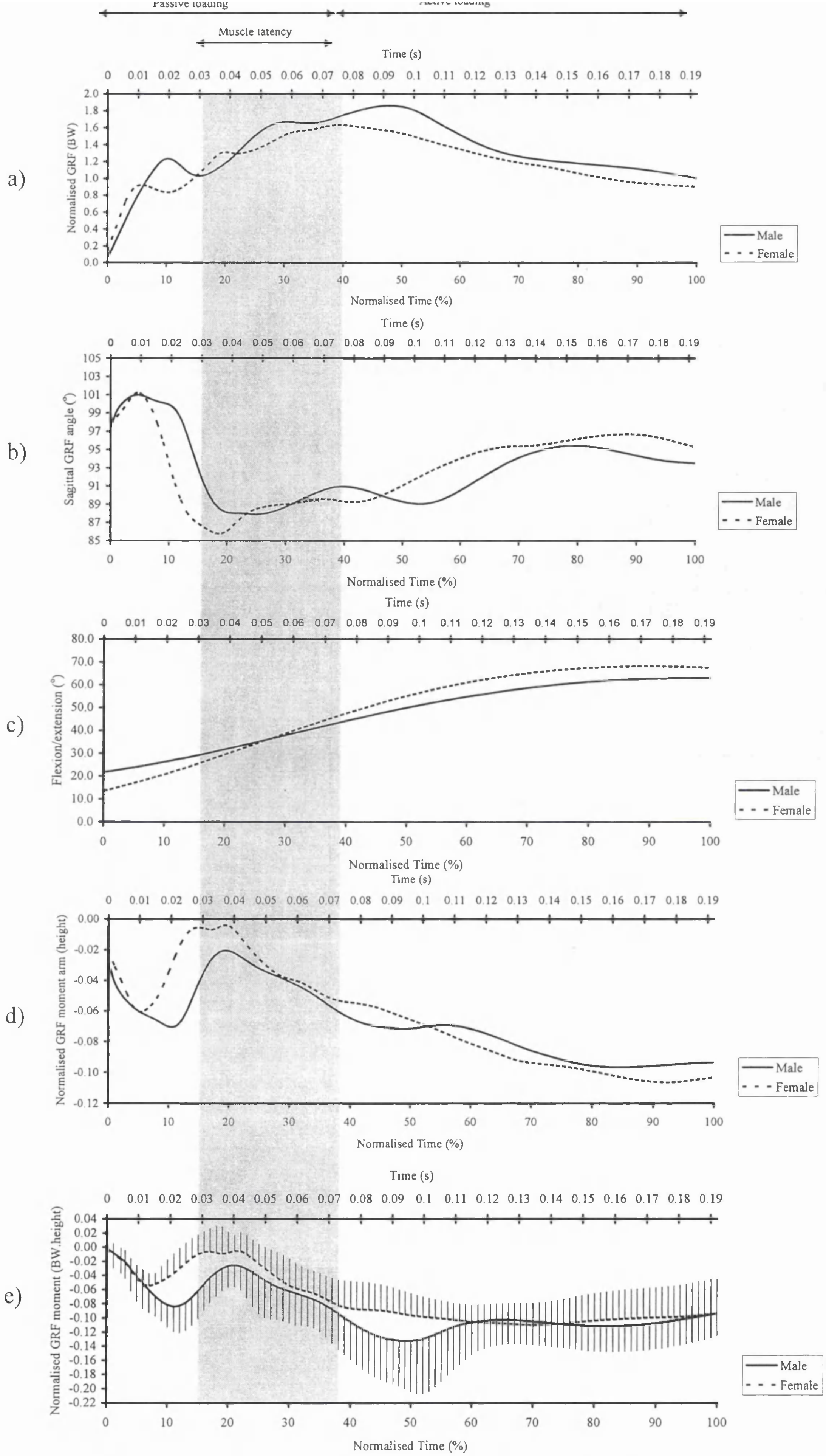


Figure 3.23 Sagittal plane normalised GRF, GRF angle, knee angle, normalised GRF moment arm and normalised GRF moment between IC and MAX for males and females.

Females and males exhibited a progressive increase in knee flexion during the landing phase (Figure 3.23c). Females exhibited significantly less knee flexion at IC (males: $21.7 \pm 5.7^{\circ}$; females: $13.5 \pm 6.0^{\circ}$) and greater MAX knee flexion (males: $63.0 \pm 11.2^{\circ}$; females: $68.2 \pm 9.5^{\circ}$) (Table 3.11 and Figure 3.23c). There was no significant difference in knee flexion angle between males and females at ML or AL.

The pattern of the normalised GRF moment arm (Figure 3.23d) largely mirrored the pattern of the normalised GRF moment (Figure 3.23e) in both males and females. During PP, females exhibited a smaller peak in normalised GRF moment arm and normalised GRF moment than males, which occurred earlier during the landing phase in females than in males. During AP, the normalised GRF moment arm and normalised GRF moment were very similar in males and females. Females displayed a significantly smaller normalised GRF flexor moment arm at the ML than males (males: -0.041 ± 0.204 ht; females: -0.006 ± 0.198 ht). The significantly smaller normalised GRF moment arm at ML in females resulted in a significantly smaller normalised GRF flexor moment in females than males at ML (males: -0.0433 ± 0.0353 BW.ht; females: -0.0065 ± 0.0325 BW.ht). There was no significant difference in the normalised GRF moment arm or the normalised GRF moment between males and females at IC, AL, at MAX or the maximum and minimum values. The magnitude of the standard deviation of the normalised GRF moment data at 1% normalised time intervals were very similar between IC and MAX in males and females (Figure 3.23e). Mean stick figures of the angle of the knee, the magnitude and angle of the GRF and the GRF moment arm in the sagittal plane for males and females at IC, ML, AL and MAX are shown in Figures 3.24 and 3.25.

Table 3.11 Group mean results for sagittal plane normalised GRF, GRF angle, knee angle, normalised GRF moment arm and GRF normalised moment about the knee (+ve = extensor GRF moment, – ve = flexor GRF moment) at IC, ML, AL, MAX maximum and minimum (Mean ± standard deviation).

Sagittal plane		IC	ML (0.03 s)	AL (0.075 s)	MAX	Maximum	Minimum
Normalised GRF (BW)	Male	0.049 ± 0.085[1]	1.052 ± 0.170	1.772 ± 0.485	0.972 ± 0.415	1.861 ± 0.595	NA
	Female	0.133 ± 0.120[1]	1.160 ± 0.287	1.625 ± 0.415	0.894 ± 0.378	1.631 ± 0.427	NA
GRF angle (°)	Male	97.4 ± 5.1	93.0 ± 6.3	90.9 ± 4.2	93.5 ± 3.0	101.0 ± 3.6	87.9 ± 2.2
	Female	97.7 ± 3.3	86.9 ± 2.9	89.4 ± 3.4	95.3 ± 4.3	101.2 ± 5.1	85.8 ± 5.0
Flexion / extension (°)	Male	21.7 ± 5.7[2]	28.8 ± 5.3	43.6 ± 7.8	63.0 ± 11.2[3]	NA	NA
	Female	13.5 ± 6.0[2]	24.9 ± 5.0	46.7 ± 9.1	68.2 ± 9.5[3]	NA	NA
Normalised GRF moment arm (ht)	Male	-0.028 ± 0.106	-0.041 ± 0.204[4]	-0.062 ± 0.111	-0.093 ± 0.073	-0.097 ± 0.072	-0.021 ± 0.115
	Female	-0.021 ± 0.070	-0.006 ± 0.198[4]	-0.055 ± 0.094	-0.104 ± 0.129	-0.106 ± 0.124	-0.004 ± 0.075
Normalised GRF moment (BW.ht)	Male	-0.0014 ± 0.0166	-0.0433 ± 0.0353[5]	-0.1110 ± 0.0541	-0.0908 ± 0.0303	-0.1325 ± 0.0681	-0.0097 ± 0.0166
	Female	-0.0028 ± 0.0182	-0.0065 ± 0.0325[5]	-0.0876 ± 0.038	-0.0923 ± 0.048	-0.1100 ± 0.0309	-0.0055 ± 0.0227

[1-5] Significant difference between males and females.

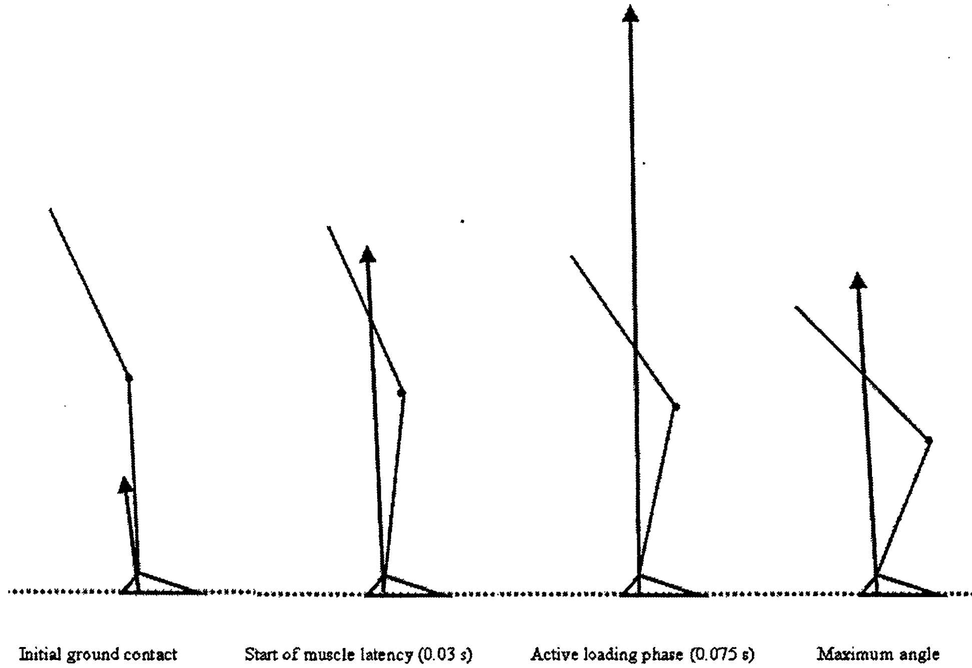


Figure 3.24 Mean stick figures of males' dominant lower leg and GRF vector in the sagittal plane at initial ground contact (IC), start of muscle latency (ML), start of active loading (AL) and maximum angle of the knee (MAX).

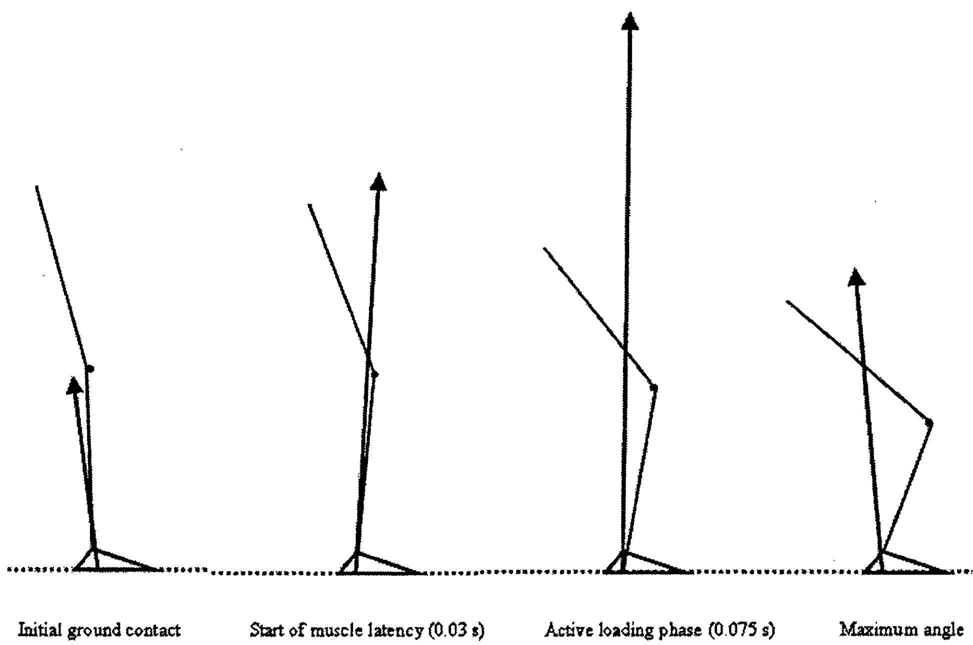


Figure 3.25 Mean stick figures of females' dominant lower leg and GRF vector in the sagittal plane at initial ground contact (IC), start of muscle latency (ML), start of active loading (AL) and maximum angle of the knee (MAX).

Injury to the passive support structures of the knee is likely to occur as the result of a large instantaneous GRF moment about the knee. Since the mean data are likely to hide these peak moments, the peak normalised GRF flexion moment (–ve) and the peak normalised GRF extension moment (+ve) observed at any instant throughout all trials for all male subjects and throughout all trials for all female subjects was identified (Table 3.12 and Figure 3.26 and 3.27). The peak normalised GRF extension moment occurred early during the trial for males and females, i.e. during passive loading. The peak normalised GRF flexion moment occurred later in the trial for both males and females, i.e. during active loading. The peak normalised GRF extension moment was greater in females than males, but the peak normalised GRF flexor moment was greater in males than females. The normalised GRF moment arm was greater for females than males in both the peak normalised GRF extensor moment and the peak normalised GRF flexor moment. Also, the knee is less flexed for females at peak normalised GRF extensor moment than for the males. Stick Figures showing the angle of the knee, relative magnitude of the GRF, angle of the GRF and the normalised GRF moment arm at the peak normalised GRF extension moment and peak normalised GRF flexion moment about the knee in the sagittal plane found in all male subjects across all trials and all female subjects across all trials (corresponding to data sets in Table 3.12) are shown in Figures 3.26 and 3.27.

Table 3.12 Peak extension and flexion normalised GRF moments and corresponding data for all males across all trials and for all females across all trials.

	Males		Females	
	Extension	**Flexion**	**Extension**	**Flexion**
Subject mass (kg)	70.0	74.0	52.8	56.5
Subject height (m)	1.75	1.91	1.69	1.69
Normalised trial time (%)	19.2	50.0	14.3	88.9
Knee angle (flex/ext) (°)	30.2	46.4	14.3	62.8
Fx (BW)	0.239	-0.056	0.241	-0.248
Fz (BW)	1.621	2.856	2.012	1.530
F_R (BW)	1.639	2.857	2.026	1.550
Moment arm (ht)	0.016	-0.101	0.034	-0.118
Moment (BW.ht)	0.0266	-0.2896	0.0706	-0.1840

3.3.5 Frontal plane kinetics.

Group mean curves for normalised GRF, angle of the GRF, knee angle, normalised GRF moment arm and normalised GRF moment (+ve = varus GRF moment, –ve = valgus GRF moment) throughout the landing period are shown for males and females in the frontal plane in Figure 3.28.

Since Fy (mediolateral force) and Fx (anterioposterior force) were small relative to Fz (vertical force) during landing, the resultant normalised GRF in the frontal plane (Figure 3.28a) was very similar to the resultant normalised GRF in the sagittal plane. Therefore as with the resultant normalised GRF in the sagittal plane, the resultant normalised GRF in the frontal plane was similar in shape in males and females, was greater for males than females during most of the landing phase and the main difference between males and females occurred during PP where females exhibit a smaller initial peak which occurred earlier in the landing phase than in males. Females exhibited significantly greater normalised GRF at IC than males

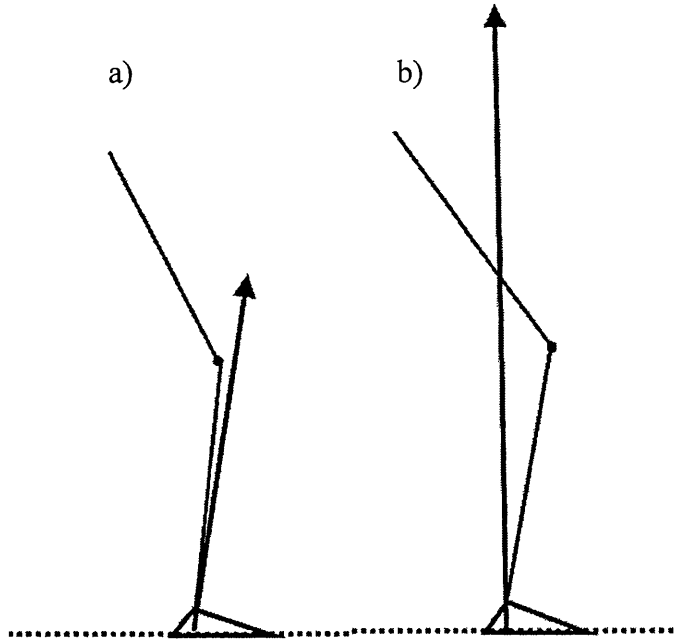


Figure 3.26 Stick figures of a) peak extension GRF moment and b) peak flexion GRF moment about the knee for males.

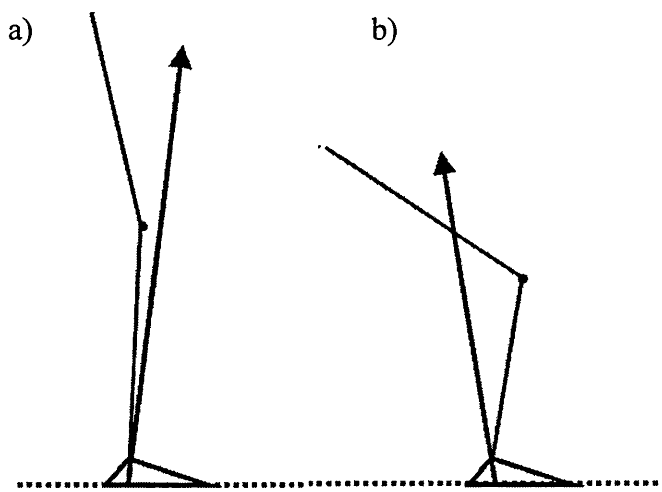


Figure 3.27 Stick figures of a) peak extension GRF moment and b) peak flexion GRF moment for females.

(males: 0.049 ± 0.085 BW; females: 0.132 ± 0.119 BW). There was no significant difference between males and females' normalised GRF at ML, AL, MAX or maximum GRF.

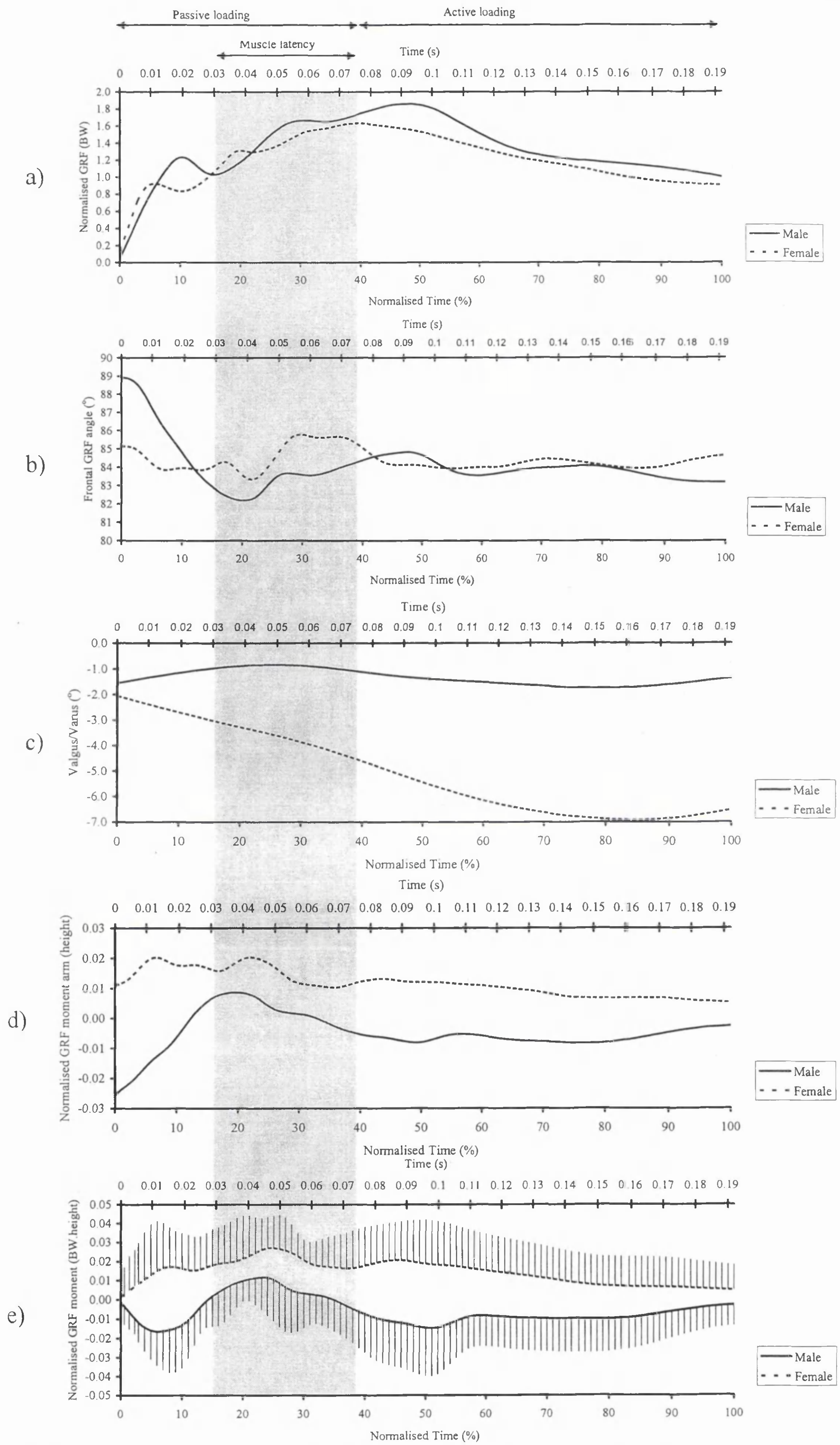


Figure 3.28 Frontal plane normalised GRF, GRF angle, knee angle, normalised GRF moment arm and normalised GRF moment between IC and MAX for males and females.

The GRF angle in the frontal plane (>90° = lateral to the vertical, <90° = medial to the vertical) (Figure 3.28b) was significantly greater at IC in males than females (males: 88.9 ± 5.3°; females: 85.1 ± 5.3°). After IC, the angle of the GRF rapidly decreased in males until it reached a similar value to that of the females close to ML. During the muscle latency phase, the angle of the GRF was then greater in females than males until AL, where the angle of the GRF for males and females remained very similar until MAX. As the maximum GRF angle occurred at or very close to IC in males and females, the maximum GRF angle in the frontal plane was also significantly greater in males than females (males: 88.9 ± 5.3°; females: 85.8 ± 4.5°). Males and females showed no significant difference in frontal plane GRF angle at ML, AL, MAX or minimum GRF angle.

In the frontal plane, females tended to contact the ground in a slight valgus position (–ve values) (-2.1 ± 3.0°) which progressively increased between IC and MAX position (-6.5 ± 4.7°). In contrast, males tended to contact the ground in a slight valgus position (-1.6 ± 6.1°) and maintained a slight valgus position (maximum: -0.8 ± 7.5°; minimum: -1.8 ± 8.8°) throughout the landing phase (Figure 3.28c). The amount of valgus at IC, ML and AL was not significantly different between males and females. However, the maximum valgus angle (males: -1.8 ± 8.8°; females: -6.9 ± 4.4°) was significantly greater in females compared to males (Table 3.13 and Figure 3.28c).

Table 3.13 Group mean results for frontal plane normalised GRF, GRF angle, knee angle, normalised GRF moment arm and normalised GRF moment about the knee (+ve = varus GRF moment, –ve = valgus GRF moment) at IC, ML, AL, MAX maximum and minimum (Mean ± standard deviation).

Frontal plane		IC	ML (0.03 s)	AL (0.075 s)	MAX	Maximum	Minimum
Normalised GRF (BW)	Male	0.049 ± 0.085[1]	1.054 ± 0.173	1.778 ± 0.486	0.977 ± 0.418	1.864 ± 0.595	NA
	Female	0.132 ± 0.119[1]	1.150 ± 0.302	1.601 ± 0.412	0.890 ± 0.378	1.604 ± 0.421	NA
GRF angle (°)	Male	88.9 ± 5.3[2]	83.0 ± 4.3	84.3 ± 2.6	83.2 ± 2.4	88.9 ± 5.3[3]	82.2 ± 1.6
	Female	85.1 ± 5.3[2]	84.6 ± 3.3	85.3 ± 2.8	84.6 ± 5.5	85.7 ± 4.5[3]	83.3 ± 2.7
Valgus / varus (°)	Male	-1.6 ± 6.1	-0.1 ± 7.0	-1.1 ± 7.8	-1.4 ± 9.2[4]	-0.8 ± 7.5	-1.8 ± 8.8
	Female	-2.1 ± 3.0	-3.0 ± 3.2	-4.5 ± 4.4	-6.8 ± 4.5[4]	-2.1 ± 3.0	-6.9 ± 4.4
Normalised GRF moment arm (ht)	Male	-0.025 ± 0.061[5]	0.006 ± 0.101	-0.005 ± 0.044[6]	-0.003 ± 0.026	0.009 ± 0.049[7]	-0.025 ± 0.061
	Female	0.010 ± 0.073[5]	0.017 ± 0.069	0.0012 ± 0.048[6]	0.005 ± 0.034	0.020 ± 0.057[7]	0.005 ± 0.034
Normalised GRF moment (BW.ht)	Male	-0.0012 ± 0.0098[8]	0.0058 ± 0.0173	-0.0085 ± 0.0212[9]	-0.0025 ± 0.0106	0.0116 ± 0.0170[10]	-0.0164 ± 0.0176
	Female	0.0013 ± 0.0184[8]	0.0192 ± 0.0199	0.0187 ± 0.0200[9]	0.0047 ± 0.0127	0.0208 ± 0.0199[10]	0.0047 ± 0.0127

[1-10] Significant difference between males and females.

As in the sagittal plane, the pattern of the normalised GRF moment arm (Figure 3.28d) largely mirrored the pattern of the normalised GRF moment (Figure 3.28e) in both males and females in the frontal plane. The normalised GRF moment arm and the normalised GRF moment remained in varus throughout the landing phase in females. For females, Figures 3.28d and 3.28e show a slight increase in varus normalised GRF moment arm and varus normalised GRF moment during PP and a slight decrease during AP. However, for males, the normalised GRF moment arm and normalised GRF moment in the frontal plane were valgus at IC, which increased slightly then decreased until it changed to a varus normalised GRF moment arm and moment close to ML. The normalised GRF moment arm and normalised GRF moment in the frontal plane then changed back to valgus at approximately 30% normalised time and remained in valgus until MAX. The normalised GRF moment and the normalised moment arm of the GRF in the frontal plane at IC were significantly different between males and females. For males at IC, a valgus (–ve) normalised moment arm of the GRF (-0.025 ± 0.061 ht) and therefore a valgus normalised GRF moment about the knee (-0.0012 ± 0.0098 BW.ht) were observed. For females, at IC, a small varus normalised moment arm of the GRF (0.010 ± 0.073 ht) and therefore a varus normalised GRF moment about the knee (0.0013 ± 0.0184 BW.ht) were observed. At AL, the valgus normalised GRF moment arm (-0.005 ± 0.044 ht) and the valgus normalised GRF moment in males (-0.0085 ± 0.0212 BW.ht) were significantly different from the varus normalised GRF moment arm (0.0012 ± 0.048 ht) and the varus normalised GRF moment (0.0187 ± 0.0200 BW.ht) in females. The maximum normalised varus GRF moment arm (males: 0.009 ± 0.049 ht; females: 0.020 ± 0.057 ht) and the maximum normalised varus GRF moment about the knee (males: 0.0116 ± 0.0170 BW.ht; females: 0.0208 ± 0.0199 BW.ht) were significantly greater in females than males. There was no significant difference in the normalised GRF moment arm or normalised GRF moment about the knee in the frontal plane at ML, MAX or maximum normalised valgus GRF moment arm or moment between males and females. The magnitude of the standard deviation of the GRF valgus/varus moment data at 1% normalised time intervals were very similar between IC and MAX. This is illustrated in Figure 3.28e. Mean stick figures of the angle of the knee, the magnitude and angle of the GRF and the normalised GRF moment arm in the frontal plane for males and females are shown in Figures 3.29 and 3.30.

Injury to the passive support structures of the knee is likely to occur as the result of a large instantaneous GRF moment about the knee. As stated in previously, a large moment about the knee in the frontal plane is more likely to strain the passive support structures of the knee than a similar moment about the knee in the sagittal plane. Since the mean data are likely to hide these peak moments, the peak normalised GRF varus moment (+ve) and the peak normalised GRF valgus moment (–ve) observed at any instant throughout all trials for all male subjects and throughout all trials for all female subjects was identified (Table 3.14 and Figures 3.31 and 3.32). The peak normalised GRF valgus and varus moments occurred earlier during the trial for the female subjects than the male subjects. The peak normalised GRF varus moment was greater in the females than males, but the peak normalised GRF valgus moment is greater in males than females. The normalised GRF moment arm was greater for females than males in both the peak normalised GRF varus moment and the peak normalised GRF valgus moment. Stick Figures showing the angle of the knee, relative magnitude of the GRF, angle of the GRF and the normalised GRF moment arm at the peak normalised GRF valgus moment

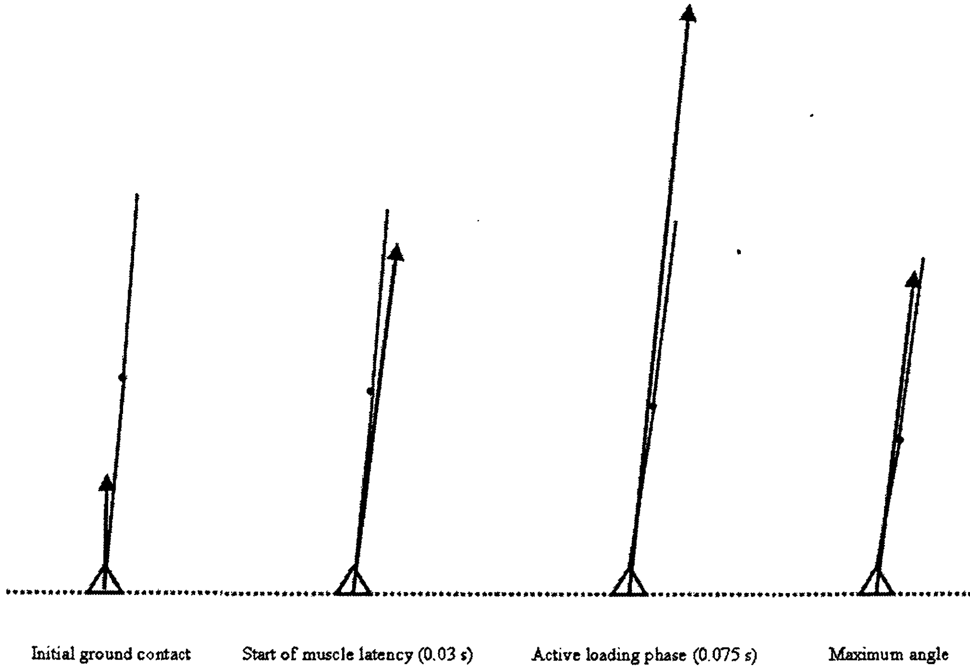


Figure 3.29 Mean stick figures of males' dominant lower leg and GRF vector in the frontal plane at initial ground contact (IC), start of muscle latency (ML), start of active loading (AL) and maximum angle of the knee (MAX).

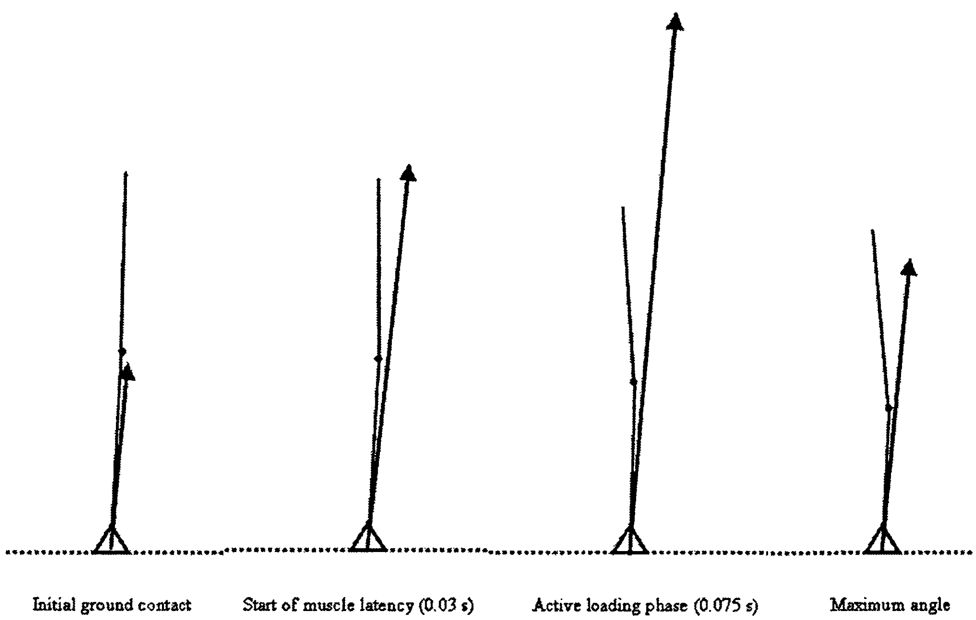


Figure 3.30 Mean stick figures of females' dominant lower leg and GRF vector in the frontal plane at initial ground contact (IC), start of muscle latency (ML), start of active loading (AL) and maximum angle of the knee (MAX).

and peak normalised GRF varus moment about the knee in the frontal plane for males and females (corresponding to data sets in Table 3.14) are shown in Figure 3.31 and 3.32.

Table 3.14 Peak valgus and varus normalised GRF moments and corresponding data for all males across all trials and for all females across all trials.

	Males		Females	
	Varus	**Valgus**	**Varus**	**Valgus**
Subject mass (kg)	65.0	71.1	52.8	52.8
Subject height (m)	1.70	1.75	1.69	1.69
Normalised trial time (%)	23.8	50.0	4.7	15.8
Knee angle (valg/var) (^{o})	3.9	2.4	-4.1	-10.7
Fy (BW)	0.428	0.032	0.233	-0.033
Fz (BW)	2.344	2.763	2.011	1.183
F_R (BW)	2.383	2.763	2.025	1.184
Moment arm (ht)	0.030	-0.026	0.048	-0.031
Moment (BW.ht)	0.0720	-0.0716	0.0979	-0.0378

3.4 Discussion.

3.4.1 Sagittal plane kinematics.

Table 3.6 and Figure 3.16 show that females tended to land with less absolute knee flexion than males, a finding strongly supported by previous literature (Yu et al., 2006; Salci et al., 2004; James et al., 2004; Decker et al., 2003; Chappell et al., 2002; Malinzak et al., 2001). The more extended the knees on ground contact, the greater the risk of ACL strain (Nunley et al., 2003; Li et al., 1999). Maximum absolute knee flexion angle and range of motion of knee flexion was found to be significantly greater in females than males, contrary to a number of other studies (Yu et al., 2006; Kernozek et al., 2005; Malinzak et al., 2001). For example, mean

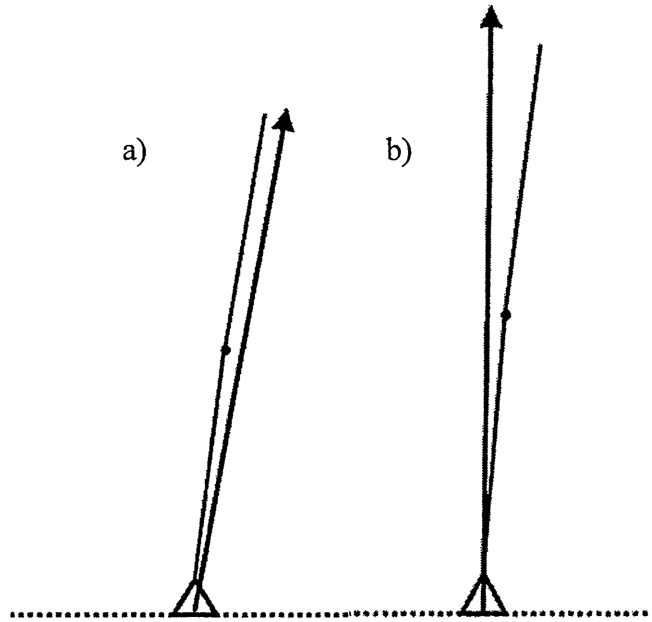


Figure 3.31 Stick figures of a) peak varus GRF moment and b) peak valgus GRF moment about the knee for males.

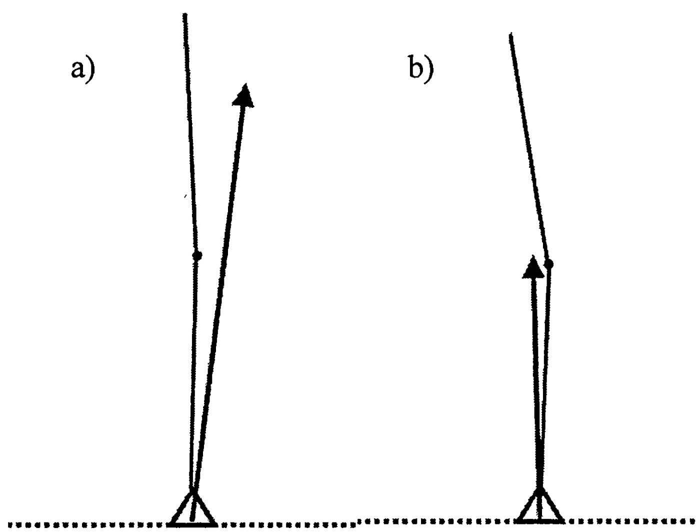


Figure 3.32 Stick figures of a) peak varus GRF moment and b) peak valgus GRF moment about the knee for females.

maximum knee flexion of $88.9 \pm 11.4^{\circ}$ for males and $78.3 \pm 13.4^{\circ}$ for females were reported by Kernozek et al. (2005) compared to $62.1 \pm 11.6^{\circ}$ for males and $68.2 \pm 12.2^{\circ}$ for females in this study. These differences could be due to different task demands in the two studies. The present study involved an opposed jumping and landing task, whereas the Kernozek et al. (2005) study involved an unopposed drop landing task.

Females displayed significantly greater average knee flexion angular velocity during PP than males, but there was no significant difference in average knee flexion angular velocity between males and females during AP. During PP, the lower limb muscles do not have complete control over the landing manoeuvre and therefore, the significantly greater average knee flexion angular velocity during PP in the females may indicate less dynamic stability of the knee than males during PP. The lower the level of dynamic stability, the greater the dependence on passive support structures, especially ligaments, for the maintenance of joint stability. Ligament strain is more likely as joint angular velocity increases due to the time required by the neuromuscular system to control the movement. Consequently, the significantly greater knee flexion angular velocity during PP in females may increase the likelihood of ACL strain compared to males.

Relative to the static reference position, there was no significant difference between males and females knee flexion at IC (Table 3.6). This suggests that the reduced absolute knee flexion at IC in females compared to males may, to some extent, be accounted for by their natural lower limb alignment, i.e. knees more extended in females than males during normal upright standing (Table 3.5). Females, however, showed significantly greater MAX relative knee flexion than males (Table 3.6).

When comparing the motion of the dominant and non-dominant legs in the sagittal plane (Figures 3.17 and 3.18 and Table 3.7), no significant differences were observed at IC, MAX or for ROM for males or females. This indicates a highly symmetrical landing pattern in the sagittal plane which, it is reasonable to assume, would facilitate greater dynamic balance during landing compared to a less symmetrical pattern.

3.4.2 Frontal plane kinematics.

Table 3.6 and Figure 3.19 show that females exhibited significantly greater absolute and relative maximum knee valgus angle and significantly greater range of motion of knee valgus angle than males. A number of studies have reported greater absolute maximum knee valgus angle and greater absolute knee valgus range of motion in females compared to males (Kernozek et al., 2005; Ford et al., 2003; Malinzak et al., 2001). However, no other studies have reported comparable relative data. The greater relative maximum knee valgus angle displayed by females compared to males in this study suggests that the reported greater absolute maximum knee valgus angle in females compared to males during landing tasks (Kernozek et al., 2005; Ford et al., 2003; Malinzak et al., 2001) is unlikely to have been accounted for by differences in static lower limb alignments.

The absolute values reported in this study are different to previous results. For example, Ford et al. (2004) reported maximum knee valgus (–ve) / varus (+ve) angle values of $-14.3 \pm 2.0^{\circ}$ for males and $-20.1 \pm 2.5^{\circ}$ for females, compared to $0.6 \pm 9.1^{\circ}$ for males and $-10.4 \pm 7.7^{\circ}$ for females in this study. There are a number of possible reasons for these differences which include subjects' age and playing standard, task demands and the method of measuring the knee valgus angle. In Ford et al. (2004) the subjects used were high school athletes whereas university athletes were used in this study. The effect of opposition in the present study may have resulted in differing levels of conscious control over the landing manoeuvre than in the Ford et al. (2004) study which involved an unopposed drop landing task. Finally, the valgus angle measured in Ford et al. (2004) was based on skin-mounted markers placed over the greater trochanter, lateral epicondyle of the knee and the lateral malleolus of the ankle, whereas the valgus angle in this study was determined from the estimated hip, knee and ankle joint centres in the Vicon plug-in gait model.

When comparing dominant and non-dominant legs in males (Figure 3.17 and Table 3.7), no significant difference was observed in valgus/varus angles at IC, MAX or ROM. Females, however, showed significantly greater MAX knee valgus angle and ROM in the non-dominant limb compared to the dominant limb (Table 3.7). Taken together, these results may indicate a higher level of dynamic stability in males compared to females. Ford et al. (2004) found a significant difference in maximum knee valgus angle in females between dominant and non-dominant limbs during a drop landing task. However, they reported that the greater range of knee valgus was in the dominant limb (dominant: $26.7 \pm 2.2^{\circ}$; non-dominant: $12.5 \pm 2.8^{\circ}$). As with maximum knee valgus, these differences may be due to differences in subjects and methodology.

As with knee flexion, females displayed significantly greater average knee valgus angular velocity than males during PP (Table 3.10), but there was no significant difference in average knee angular velocity in the frontal plane between males (varus) and females (valgus) during AP. The combination of significantly greater knee flexion angular velocity in females during PP, significantly greater knee valgus angular velocity in females during PP, significantly greater maximum knee valgus angle during landing in females and significantly greater knee valgus ROM during landing in females may reflect lower dynamic stability and, in turn, increased risk of knee ligament strain. Increased knee valgus angle (relative to static reference position) is a major risk factor for ACL injury (Olsen et al., 2004; Boden et al., 2000). It appears that females are vulnerable to excessive knee valgus motion during the PP which, in turn, is likely to increase the risk of ACL strain. The greater vulnerability to excessive valgus during PP may be due, at least in part, to lower torsional stiffness of the knee joint in females compared to males. Torsional stiffness refers to the ability to prevent joint motion (maintain a particular joint angle) in response to external loading. A number of studies have indicated that females have lower torsional stiffness in the knee joint than males (Wojtys et al., 2003; Granata et al., 2002a). Lower torsional stiffness in the females in the present study may be reflected in Figure 3.19 which shows a progressive increase in knee valgus in females between IC and MAX. In contrast, the males demonstrate a fairly neutral position with regard to knee valgus/varus between IC and MAX.

3.4.3 Inter-hip, inter knee and inter ankle displacements.

The results of the absolute inter-knee distances indicate that females' knees move significantly closer together and move through a greater range of motion during landing than males (Table 3.9), which is also reported by Ford et al. (2004). In the

Ford et al. (2004) study, inter-knee distance was measured from markers placed on the lateral epicondyles of each femur, whereas in this study inter-knee distance was measured from estimated knee joint centres. Each estimated knee joint centre incorporates an offset equivalent to the sum of half the knee width and the marker radius. The knee joint centre is located as the offset from the LKNE (Figure 3.10) marker in a direction perpendicular to the line from the hip joint centre to LKNE. To compare data from this study with that of Ford et al. (2004) the average knee offsets of 122 mm for males and 117 mm for females were applied to the Ford et al. (2004) data. The amended Ford et al. (2004) data for minimum inter-knee distance (males: 224 ± 6 mm; females 204 ± 6 mm) is similar to the results of the present study (males: 234 ± 39 mm; females: 200 ± 35 mm). However, the amended Ford et al. (2004) data for inter-knee ROM (males: 53 ± 5 mm; females: 73 ± 5 mm) indicate greater ROM compared to the present results (males: 10 ± 16 mm; females: 28 ± 18 mm).

Table 3.10 and Figure 3.21 show that males and females tended to have a fairly wide base of support at IC (inter-ankle distance relative to inter-hip distance), irrespective of knee motion. However, whereas there was no significant difference in the frontal plane kinematics of the dominant and non-dominant legs in males, there was a significant difference in females, suggesting a lower level of dynamic stability in females compared to males.

To the author's knowledge, no data have been reported for absolute or relative inter-ankle distances during landing. Therefore no comparisons can be made between the results of this study and previous studies. The absolute inter-ankle results indicate that the males moved their ankles together after IC through a greater ROM than females. From Table 3.9 and Figures 3.20 and 3.21 it can be seen that males' ankles are wider apart at IC and move together more quickly than in females for the first 40% of normalised contact time. Thereafter, the absolute inter-ankle distance is similar in males and females. This is likely to be because the heels are in contact with the ground during this period. The movement patterns indicate that after the toes make contact with the ground, females' ankles move vertically downward to the ground until the heels make contact, whereas for males, the ankles are brought in towards each other as the heels move down to the ground. When looking at the simultaneous linear motion of the knees and ankles on landing (Figure 3.20), a continuous inward movement of the ankles is shown by males and females, however, this inward movement of the ankles is greater in males than females. At the same time, the movement of the knees in males show an out – in – out action resulting in minimum net movement. In contrast, the females' knees show continuous inward movement. These combined linear movement patterns of the knees and ankles are likely to reduce knee valgus in males and increase knee valgus in females which is particularly evident during PP (Figure 3.19).

The results of the relative inter-joint distances were similar to the results of the absolute inter-joint distances. The only difference between the absolute and relative findings was that females' knees were significantly closer together than males at IC relative to the static reference position (Table 3.9 and Figure 3.22). Consequently, the variation between males and females static lower limb alignments does not totally account for the significantly shorter MIN inter-knee distance in females compared to males on landing.

Relative to inter-hip distance, inter-knee distance was greater in males than females at both IC and MIN (Table 3.10). Inter-ankle distance relative to inter-hip distance was greater in males than females at IC but smaller in males than females at MIN (Table 3.10). Taken together, these results seem to indicate that in comparison to females, males adopt a wider base of support at IC and then reduce the width of this base of support, by moving the ankles together, in order to maintain a fairly neutral knee valgus/varus angle.

3.4.4 Ground reaction force.

Peak GRF in both the frontal and sagittal planes was not significantly different between females and males, even when normalised to body weight. This is different to a number of other studies which found females to exert greater normalised GRF than males when landing (Salci et al., 2004; Yu et al., 2006; Kernozek et al., 2005). This may be due to other studies having males and females dropping down from the same fixed height, whereas this study had subjects jumping up to block a ball at a height of 2.43 m for males and 2.24 m for females. It is unlikely females jump as high as males when playing those sports where non-contact ACL injury is particularly common, particularly volleyball as the net is 0.19 m higher for males than females. The normalised GRF was significantly greater in females than males at IC which may be attributed to females landing with less knee flexion than males.

The pattern of the normalised GRF moment about the knee in both frontal and sagittal planes followed a very similar pattern to the pattern of the normalised GRF moment arm about the knee in the frontal and sagittal planes respectively. The normalised GRF was very similar between males and females but the resulting GRF moments about the knee, particularly in the frontal plane, were different. This suggests that the main differences in the normalised GRF moment between males and females occur as the result of different GRF moment arms rather than the magnitude of the normalised GRF.

3.4.5 Sagittal plane kinetics.

The normalised GRF moment arm about the knee joint centre at ML was significantly smaller in females than males. This resulted in a significantly smaller normalised GRF moment about the knee in the sagittal plane in females compared to males at ML. This suggests that through training, females may have developed a strategy of landing which minimises the GRF moment about the knee in the sagittal plane in an attempt to reduce the likely strain on the dynamic and passive support structures of the knee. Subjects unfamiliar with performing volleyball block jump landings may have displayed a different pattern to the normalised GRF moment about the knee which may alter the likelihood of ACL strain compared to trained subjects. However, this was not investigated in the present study.

The pattern of the normalised GRF moment arm and normalised GRF moment about the knee in the sagittal plane were similar in males and females. This suggests that the greater MAX flexion in females compared to males may be due to reduced quadriceps muscle strength in females compared to males rather than any difference in the normalised GRF moment produced about the knee in the sagittal plane.

The male and female group mean peak normalised GRF knee flexor moment in this study was very similar to that reported by Hewett et al. (1996). For example, values for the mean peak normalised GRF flexor moment reported by Hewett et al. (1996) were 0.104 BW.ht for trained females and 0.158 BW.ht for untrained males compared to 0.110 BW.ht for trained females and 0.1325 BW.ht for trained males in the present study.

The peak instantaneous normalised GRF flexor moment in the sagittal plane reported by Alexander and Vernon (1975) for two male subjects during landing from a height of 81 cm was -0.14 BW.ht. This involved a normalised GRF moment arm of -0.099 ht and a normalised resultant GRF of 1.42 BW. The peak instantaneous normalised GRF moment about the knee in the sagittal plane observed in this study for a male subject was -0.29 BW.ht which involved a GRF moment arm of -0.10 ht and a normalised resultant GRF of 2.86 BW. The results indicate that the peak normalised GRF moment arm is very similar between both studies, but the normalised resultant GRF in the present study was approximately double that measured by Alexander and Vernon (1975) which resulted in a normalised GRF moment about the knee in this study of approximately double that measured by Alexander and Vernon (1975). Other studies report mean peak normalised vertical GRF of between 6.2 and 2.16 BW for females and 6.1 and 2.16 BW for males (combined left and right leg GRF) during landing (Yu et al., 2006; Kernozek et al., 2005; Salci et al., 2004; Decker et al., 2003; Hewett et al., 1996). There are a number of possible reasons for the differences between the results of the present study and those of Alexander and Vernon (1975) which include subjects' age and playing standard, the number of subjects, task demands and the method of measuring the GRF. In Alexander and Vernon (1975) there were only two male subjects aged 37 and 43 years, whereas 6 female and 6 male university athletes of mean age 21.2 ± 1.3 years for females and 21.6 ± 3.29 years for males were used in this study. The effect of opposition in the present study may have resulted in different levels of conscious control over the landing manoeuvre than in the Alexander and Vernon (1975) study which involved an unopposed drop landing task. Another limitation of the Alexander and Vernon (1975) study was that half the total GRF acting on both feet was used to calculate the moment about one knee rather than separate GRF measurements for left and right legs. This assumes an equal load on each foot. In the present study the GRF on the right foot was measured separately.

The mean peak normalised GRF moments in the sagittal plane in this study were -0.1325 ± 0.0681 BW.ht for males and -0.1100 ± 0.0309 BW.ht for females. The mean peak normalised GRF moment arms in the sagittal plane were -0.097 ± 0.072 ht for males and -0.106 ± 0.124 ht for females. The mean peak normalised resultant GRF was 1.861 ± 0.595 BW for males and 1.631 ± 0.427 BW for females. These mean peak values are similar to the instantaneous peak values measured by Alexander and Vernon (1975).

The peak normalised GRF flexion moment (–ve) and the peak normalised GRF extension moment (+ve) observed at any instant throughout all trials for all male subjects and throughout all trials for all female subjects (Table 3.12 and Figure 3.26 and 3.27) showed that the peak normalised GRF extension moment occurred early during the trial for males and females, i.e. during passive loading. Whereas, the peak normalised GRF flexion moment occurred later in the trial for both males and females, i.e. during active loading. Since the muscles are in control of the landing during active loading but not during passive loading, it is more likely that the passive support structures of the knee will be placed under strain during passive loading. Therefore, the peak normalised GRF extension moments are more likely to strain the passive support structures of the knee than the peak normalised GRF flexor moments. The peak normalised GRF extension moment was greater in females than males, but the peak normalised GRF flexor moment was greater in males than females. A GRF extensor moment results in a knee flexor *MUS* produced largely by hamstring muscle activity. Since the hamstrings work with the ACL to prevent anterior displacement of the tibia relative to the femur, the greater the load on the hamstring muscles, the greater the extent to which stability of the knee joint (anterior displacement of the tibia relative to the femur) is likely to be maintained by the ACL. Therefore, the greater peak normalised GRF extensor moment about the knee in females compared to males is likely to increase ACL strain in females relative to males. The knee is less flexed for females at peak normalised GRF extensor moment than for the males. ACL strain is likely to be increase with reduced knee flexion therefore peak extensor normalised GRF moment is more likely to cause ACL strain in females than males.

3.4.6 Frontal plane kinetics.

There was very little difference in the magnitude of the normalised GRF and the GRF angle between males and females during landing. However, the resulting normalised GRF moment was different in males and females. This, at least in part, may due to the females landing with their feet closer together than males.

In males, the normalised moment arm of the GRF about the knee was fairly small (Figure 3.29). This resulted in a small moment about the knee in the frontal plane for males. In females however, the normalised moment arm of the GRF about the knee was greater than in males (Figure 3.30). This produced a larger GRF moment about the knee in the frontal plane for females which is likely to put the passive support structures of the knee under strain.

Hewett et al. (1996) reported values of 0.021 BW.ht for mean peak normalised varus GRF moment for trained females. These values are similar to those reported in the present study of 0.0208 BW.ht for females. Hewett et al. (1996) reported values of -0.017 BW.ht for mean peak normalised valgus GRF moment for trained females. However, in this study, throughout the landing phase used for analysis (between IC and MAX) the mean GRF moment acting about the knee remained a varus moment for females. In untrained males, Hewett et al. (1996) reported values of 0.037 BW.ht for mean peak normalised varus GRF moment and -0.049 BW.ht for mean peak normalised valgus GRF moment. These values appear slightly higher than those measured in the present study for trained males, which are a mean peak normalised varus GRF moment of 0.0116 BW.ht and a mean peak normalised valgus GRF moment of -0.0164 BW.ht. The present study differs from Hewett et al. (1996) because the left leg (non-dominant) was analysed by Hewett et al. (1996) but the right leg (dominant) was analysed in this study.

The peak normalised GRF varus moment (–ve) and the peak normalised GRF valgus moment (+ve) observed at any instant throughout all trials for all male subjects and throughout all trials for all female subjects was identified (Table 3.14 and Figures 3.31 and 3.32) showed that the peak normalised GRF valgus and varus moment occurred earlier during the trial for the female subject than the male subject. The earlier in the landing phase the less likely muscular control is likely to have been gained over the landing. Since the muscles are less likely to be in control of the landing for the females peak normalised GRF valgus and varus moments than males, it is more likely the passive support structures of the knee will be placed under strain for the female subject than the male. The peak normalised GRF varus moment was greater in the females than males, but the peak normalised GRF valgus moment is greater in males than females. Using the quasi-static model, a varus GRF moment is considered to result in an equivalent valgus *MUS* and a valgus GRF moment is considered to result in an equivalent varus *MUS*. Therefore females may appear to exert a greater valgus *MUS* than males during landing, whereas males exert a greater varus *MUS* than females during landing. This suggests that the valgus *MUS* exhibited by females during landing causes their knees to move into a greater valgus position, whereas males exert a varus *MUS* during landing which prevents an increase in the valgus angle of the knee which in turn, reduces the risk of strain to the passive support structures of the knee in males compared to females. The normalised GRF moment arm was greater for females than males in both the peak normalised GRF varus moment and the peak normalised GRF valgus moment.

3.4.7 Training implications.

The results of the present study provide a number of training suggestions to reduce the likelihood of ACL strain and therefore non-contact ACL injury in females:

- Increase hamstring and quadriceps strength. This increased strength may reduce MAX knee flexion angle and ROM of knee flexion in females compared to males by increasing the knee extension muscle moment exerted and therefore providing greater resistance to knee flexion.
- Land with feet further apart and bring the ankles in together just after IC. This appears to be a way males reduce the valgus angle of the knee during landing and therefore may reduce valgus movement of the knee in females.
- Increase hip abductor strength. This increased strength may help prevent the knees moving into a valgus position during landing due to the muscles providing more resistance to valgus movement during landing.

These training suggestions may be beneficial for coaches and athletes competing not just in volleyball, but by a number of sports characterised by a high frequency of landing manoeuvres, which may help reduce the likelihood of ACL injury.

3.5 Summary and conclusions.

In the sagittal plane, the main findings were:

- For both males and females, the landing period was characterised by a progressive increase in knee flexion.
- Absolute knee flexion at ground contact was significantly smaller in females than males, but there was no significant difference in relative knee flexion on ground contact.

- Maximum knee flexion (absolute and relative) and range of motion of knee flexion were significantly greater in females than males.

- There was no significant difference in knee flexion (on ground contact, at maximum knee flexion or range of motion) between dominant and non-dominant legs in males or females.

- Angular velocity of the knees in the sagittal plane was significantly greater in females than males in the passive phase of landing, but not significantly different in the active phase.

- The normalised GRF moment arm and moment in the sagittal plane were similar in pattern between males and females but females displayed significantly smaller flexion normalised GRF moment arm and moment at ML than males.

These results suggest that the greater maximum knee flexion (absolute and relative), greater range of motion of knee flexion and greater angular velocity during the passive phase of landing in females compared to males is more likely to be due to reduced dynamic stability of the knee in females compared to males than due to greater normalised GRF moments acting about the knee joint centre in the sagittal plane.

In the frontal plane, the main findings were:

- Females exhibited a progressive increase in knee valgus during landing. In contrast there was very little change in knee valgus/varus angle during landing in males.

- There was no significant difference in knee valgus/varus at ground contact between males and females.

- Females displayed significantly greater maximum valgus angle and range of motion knee valgus angle than males in absolute and relative terms.

- There was no significant difference in knee valgus/varus at ground contact, maximum knee valgus/varus angle or range of knee valgus/varus between dominant and non-dominant legs in males, but there was a significant difference in maximum valgus and range of motion between the dominant and non-dominant legs in females.

- Angular velocity of the knees in the frontal plane was significantly greater in females than males in the passive phase of landing, but not significantly different in the active phase.

- The patterns of the normalised GRF moment arm and moment in the frontal plane were different between males and females. Females' normalised GRF moment arm and moment remained in varus throughout landing, whereas for males, the normalised GRF moment arm and moment changed between valgus and varus during landing.

- Females' varus normalised GRF moment arms and moment were significantly different from males' valgus normalised GRF moment arm and moment at IC and AL.

- The maximum varus normalised GRF moment arm and moment was significantly greater in females than males.

These results suggest the difference in normal alignment of the knees in the frontal plane does not account for the difference in maximum valgus angle on landing between males and females and may indicate reduced dynamic stability of the knee in females compared to males in resisting abnormal movement of the knees.

For the ground reaction force, the main findings were:

- The overall patterns of the magnitude and the angle of the normalised GRF were similar between males and females in both the sagittal and frontal planes during landing.
- Females displayed significantly greater GRF at IC (frontal and sagittal planes) and significantly smaller GRF angle at IC in the frontal plane than males.

Overall, the result of the study may indicate less dynamic stability of the knee in females compared to males, particularly in the passive phase of landing. The lower the dynamic stability, the greater the dependence on the passive support structures, especially the ligaments, for the maintenance of joint stability. As ACL injuries occur most frequently in the passive phase of landing manoeuvres, the present results suggest that lack of dynamic stability of the knee in the passive phase could be a contributing factor in the reported greater incidence of ACL injury in females compared to males. Training programmes for females should incorporate exercises and practices to improve the dynamic stability of the knee in the passive phase of ground contact.

Chapter 4

Experimental Study 2

Knee kinematics and kinetics during landing from unopposed and opposed volleyball block jumps.

4.1 Introduction.

The results of Chapter 3 suggest that females have less dynamic stability of the knee compared to males in the passive phase when landing from an opposed volleyball block jump. The lower the dynamic stability, the greater the dependence on the passive support structures, especially the ligaments, for the maintenance of joint stability. As ACL injuries occur most frequently in the passive phase of landing manoeuvres, these results suggest that lack of dynamic stability of the knee in the passive phase could be a contributing factor in the reported greater incidence of ACL injury in females compared to males when landing from opposed volleyball block jumps.

4.1.1 Aim.

It is reasonable to assume that the attentional demand of jumping and landing in an opposed context will be less than that in an unopposed context (Chen et al., 1996; Lajoie et al., 1993) which, in turn, is likely to affect the neuromuscular response when landing. A number of studies have examined gender differences in kinematics and kinetics during landing/cutting manoeuvres in unopposed and opposed contexts, but direct comparison of the results is not possible due to differences in task demands. The purpose of the present study was to examine the effect of opposition on the kinematics and kinetics of landing from a volleyball block in male and female university volleyball players.

The objectives of the study were to examine, during landing from a volleyball block jump, the effect of unopposed and opposed conditions on:

i) The angular displacement of the knee joint in the sagittal plane,

ii) The angular displacement of the knee joint in the frontal plane,

iii) The linear displacement of the inter-hip, inter-knee and inter-ankle joint centre distances,

iv) The muscle moment-time histories about the tibiofemoral joint in the sagittal and frontal planes, corresponding to the associated GRF force-time and GRF moment arm-time histories.

4.2. Method.

The methods used in this study were similar to those in Chapter 3. The positions of the suspended volleyball, net and force plates are shown in Figure 3.9. The subjects were six female (Mean age 21.2 ± 1.3 years, mass 57.6 ± 7.5 kg and height 164.8 ± 7.5 cm) and six male (Mean age 21.6 ± 3.3 years, mass 70.1 ± 3.1 kg and height 175.7 ± 8.6 cm) university volleyball players. All subjects were right leg dominant and had no previous history of hip, knee or ankle injury. Ethical approval was granted for the study by the University Ethics Committee and written consent forms were signed by all subjects prior to data collection. Subjects were required to perform two landing tasks: unopposed volleyball block jump and landing and opposed volleyball block jump and landing.

1) Unopposed: At the start of each trial, the subject stood with each foot on a separate force plate. The subject was then instructed to jump up and pretend to block the suspended volleyball.

2) Opposed: At the start of each trial, the subject stood with each foot on a separate force plate. The subject then timed his/her blocking action in order to try to block the ball as it was spiked. The ball was spiked from the same suspended position in order to eliminate variation in the position and velocity of the ball. On landing,

each foot landed on a separate force plate. Data were recorded for three successful trials for each subject. Trials where each foot did not land entirely on a separate force plate were discarded.

Independent-samples t-tests were used to assess gender differences and paired samples t-tests were used to asses differences between landing tasks for the angular displacement, inter-joint distances, normalised GRF, normalised GRF moment arm and normalised GRF moment about the knee data in the sagittal and frontal planes. Normality of the data and between-group homogeneity of variance were assessed prior to statistical analysis, as described in Chapter 3, section 3.2.9.

4.3 Results.

For the purpose of analysis, data for the three trials of each subject was averaged. The mean data were then used to calculate the overall group mean data. All Figures show variables plotted against normalised time and against absolute mean trial time between IC and MAX. For the unopposed trials, absolute mean contact time was 0.203 ± 0.068 s for males and 0.213 ± 0.061 s for females. For the opposed trials, the same data are presented as in Chapter 3. Absolute mean contact time was 0.190 ± 0.040 s for males and 0.194 ± 0.057 s for females. As there was no significant difference between contact time for males and females during unopposed and opposed trials, a mean contact time of 0.200 s was used. Since the results of chapter 3 indicated that the differences in normal alignment of the knees in upright standing did not account for the observed differences in knee kinematics on landing from opposed block jumps between males and females, only absolute changes in knee kinematics are reported in this chapter.

4.3.1 Sagittal plane kinematics.

In the sagittal plane, there was no significant difference in knee flexion at IC between males and females. However, there was a significant difference in knee flexion at IC between unopposed and opposed trials (males unopposed: 20.3 ± 4.7°; males opposed: 19.4 ± 6.4°; females unopposed: 19.5 ± 6.9°; females opposed: 15.1 ± 6.2°). Females displayed significantly greater MAX knee flexion than males and MAX knee flexion was significantly greater during unopposed trials than opposed trials (males unopposed: 67.2 ± 12.9°; males opposed: 62.1 ± 11.6°; females unopposed: 78.0 ± 8.1°; females opposed: 68.2 ± 12.2°). This resulted in a significantly greater ROM of knee flexion in females than males and significantly greater ROM of knee flexion during unopposed trials than opposed trials (males unopposed: 46.9 ± 14.9°; males opposed: 42.7 ± 13.9°; females unopposed: 58.6 ± 7.4°; females opposed: 53.1 ± 13.1°) (Table 4.1 and Figure 4.1). Males and females showed no significant difference in sagittal plane kinematics between dominant and non-dominant legs during landing in both unopposed and opposed conditions (Table 4.2 and Figures 4.2 and 4.3).

The gradient of the curves in Figure 4.1 indicate differences between males and females and between unopposed and opposed trials in average angular velocity of knee flexion during the two main phases of the landing, passive loading and muscle latency phase (PP) and active loading phase (AP). Females displayed significantly greater average knee flexion angular velocity than males during PP and average knee flexion angular velocity was significantly greater during unopposed trials than opposed trials during PP (males unopposed: 377.6°/s; males opposed: 312.8°/s;

Table 4.1. Group mean results (combined right and left legs) for knee flexion/extension and valgus/varus (– varus; + valgus) angles at IC, MAX and ROM for males and females during unopposed and opposed trials (Mean ± standard deviation).

		Males		Females	
		Unopposed (°)	Opposed (°)	Unopposed (°)	Opposed (°)
Flexion	IC*	20.3 ± 4.7	19.4 ± 6.4	19.5 ± 6.9	15.1 ± 6.2
	MAX*†	67.2 ± 12.9	62.1 ± 11.6	78.0 ± 8.1	68.2 ± 12.2
	ROM*†	46.9 ± 14.9	42.7 ± 13.9	58.6 ± 7.4	53.1 ± 13.1
Valgus/varus	IC	-2.2 ± 5.3	-2.8 ± 5.9	-2.1 ± 3.4	-1.6 ± 2.8
	MAX_{VAL}†	-2.2 ± 5.3	-2.9 ± 7.9	-13.9 ± 11.3	-10.4 ± 7.7
	MAX_{VAR}	1.0 ± 9.6	0.6 ± 9.1	N/A	N/A
	ROM†	3.2 ± 8.0	3.5 ± 9.6	11.8 ± 10.3	8.8 ± 7.8

* : Significant difference between unopposed and opposed trials ($p < 0.05$).

† : Significant difference between males and females ($p < 0.05$).

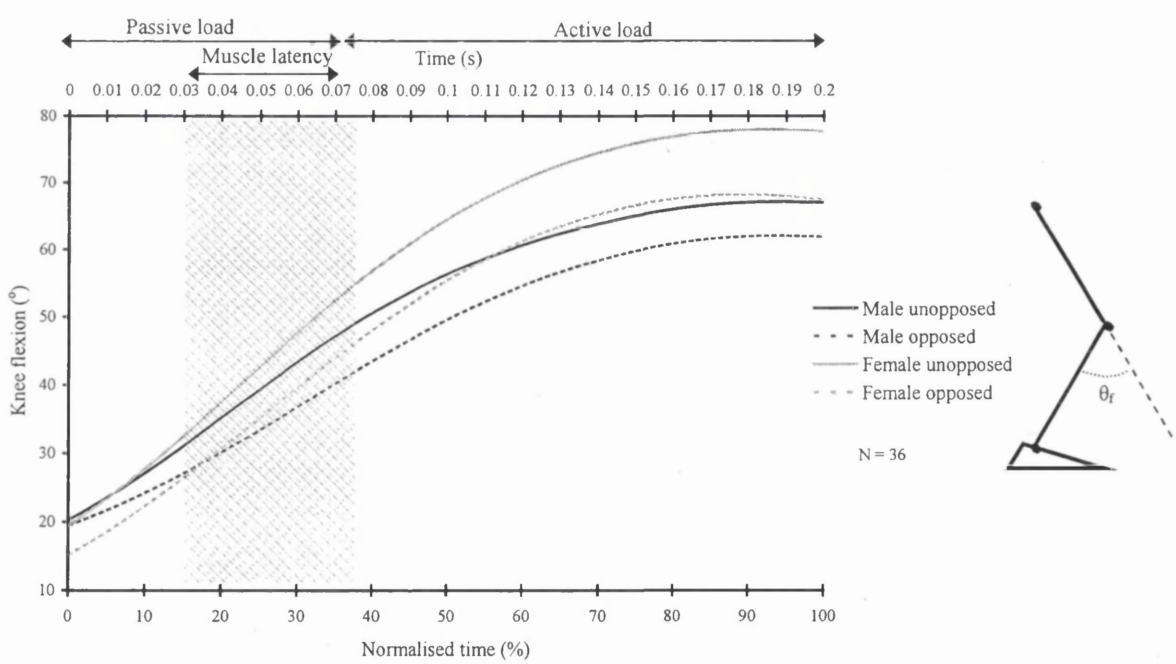


Figure 4.1. Knee flexion (θ_f) between IC and MAX for males and females during unopposed and opposed trials.

Table 4.2. Group mean results for knee flexion/extension and valgus/varus (– varus; + valgus) at IC, MAX and for ROM for male and female subjects' dominant and non-dominant legs during unopposed and opposed trials (Mean ± standard deviation).

		Males				Females			
		Unopposed		Opposed		Unopposed		Opposed	
		Non-dominant	Dominant	Non-dominant	Dominant	Non-dominant	Dominant	Non-dominant	Dominant
Flexion (°)	IC	18.1 ± 4.3	22.4 ± 4.2	17.1 ± 6.4	21.7 ± 5.7	22.3 ± 6.9	16.7 ± 5.9	16.7 ± 6.1	13.5 ± 6.0
	MAX	67.2 ± 12.4	67.3 ± 13.5	61.2 ± 12.3	63.0 ± 11.2	80.2 ± 9.3	76.2 ± 6.9	68.3 ± 14.7	68.2 ± 9.5
	ROM	49.0 ± 14.3	44.9 ± 15.8	44.1 ± 15.1	41.3 ± 13.0	57.9 ± 8.3	59.5 ± 6.6	51.6 ± 13.9	54.7 ± 12.5
Valgus/varus (°)	IC	-3.1 ± 5.1	-1.3 ± 5.5	-4.0 ± 5.6	-1.6 ± 6.1	-2.6 ± 3.3	-1.7 ± 3.5	-1.1 ± 2.7	-2.1 ± 3.0
	MAX	-3.2 ± 7.3	-0.0 ± 10.1	2.5 ± 8.9	-1.4 ± 9.2	-19.4 ± 8.9[1]	-8.5 ± 11.0[1]	-13.9 ± 8.7[2]	-6.8 ± 4.5[2]
	ROM	5.9 ± 9.6	1.3 ± 5.1	6.5 ± 12.0	0.2 ± 5.2	16.8 ± 7.2[3]	6.9 ± 10.7[3]	12.8 ± 7.6[4]	4.7 ± 5.8[4]

[1-4]: Significant difference between dominant and non-dominant legs in females ($p < 0.05$).

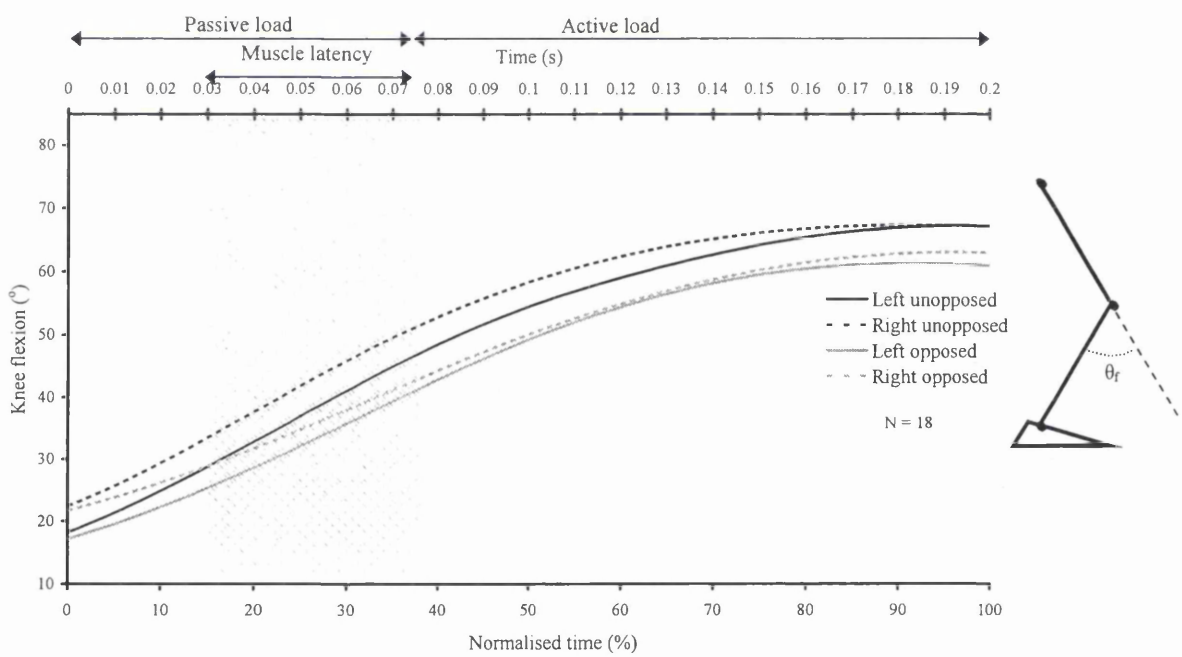


Figure 4.2. Dominant and non-dominant leg knee flexion/extension (θ_f) between IC and MAX for males during unopposed and opposed trials.

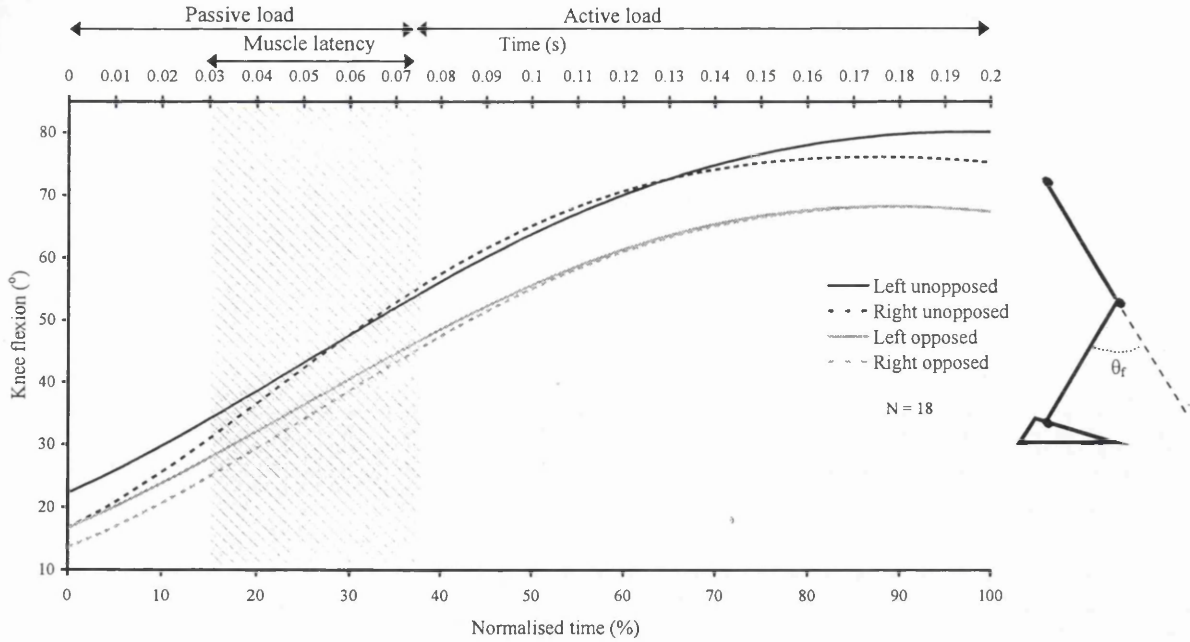


Figure 4.3. Dominant and non-dominant leg knee flexion/extension (θ_f) between IC and MAX for females during unopposed and opposed trials.

females unopposed: 463.9°/s; females opposed: 428.4°/s). Average knee flexion angular velocity during AP was not significantly different between males and females or between unopposed and opposed trials (males unopposed: 144.3°/s; males opposed: 162.8°/s; Females unopposed: 183.3°/s; females opposed: 172.7°/s) (Table 4.3). The slopes of the curves in Figure 4.1 also indicate that peak knee flexion angular velocity occurred at approximately 25% of normalised time for males and females during unopposed and opposed trials (males unopposed: 20%; males opposed; 31%: females unopposed: 22%; females opposed: 25%), but peak knee flexion angular velocity was significantly greater in females than males and greater during unopposed trials than opposed trials (males unopposed: 410.7°/s; males opposed: 335.2°/s; females unopposed: 505.8°/s; females opposed: 439.6°/s) (gradient of knee flexion – time curves in muscle latency period in Figure 4.1 and Table 4.3).

Table 4.3. Angular velocity of knee flexion and valgus/varus during the two main phases of the landing, i.e. passive load and muscle latency phase and active phase.

Passive load and muscle latency phase									
			t_{IC} (s)	t_{PP} (s)	Δt (s)	θ_{IC} (°)	θ_{PP} (°)	$\Delta\theta$ (°)	ω (°/s)
Flexion *†	Males	Unopposed	0	0.075	0.075	20.3	48.6	28.3	377.6
		Opposed	0	0.075	0.075	19.4	42.9	23.5	312.8
	Females	Unopposed	0	0.075	0.075	19.5	54.3	34.8	463.9
		Opposed	0	0.075	0.075	15.1	47.2	32.1	428.4
Valgus/varus †	Males	Unopposed	0	0.075	0.075	-2.2	-1.5	0.7	9.6
		Opposed	0	0.075	0.075	-2.8	-2.9	0.1	1.7
	Females	Unopposed	0	0.075	0.075	-2.1	-7.8	5.6	75.1
		Opposed	0	0.075	0.075	-1.6	-6.1	4.5	59.7
Active phase									
			t_{PP} (s)	t_{MAX} (s)	Δt (s)	θ_{PP} (°)	θ_{MAX} (°)	$\Delta\theta$ (°)	ω (°/s)
Flexion	Males	Unopposed	0.075	0.203	0.128	48.6	67.0	18.5	144.3
		Opposed	0.075	0.192	0.117	42.9	61.9	19.1	162.8
	Females	Unopposed	0.075	0.203	0.128	54.3	77.7	23.5	183.3
		Opposed	0.075	0.192	0.117	47.2	67.5	20.2	172.7
Valgus/varus	Males	Unopposed	0.075	0.203	0.128	-1.5	-0.0	1.5	11.3
		Opposed	0.075	0.192	0.117	-2.9	0.6	3.5	29.7
	Females	Unopposed	0.075	0.203	0.128	-7.8	-13.8	6.0	47.0
		Opposed	0.075	0.192	0.117	-6.1	-10.2	4.2	35.8

* : Significant difference between unopposed and opposed trials ($p < 0.05$).

† : Significant difference between males and females ($p < 0.05$).

t_{IC} = time at IC; t_{PP} = duration of PP; t_{MAX} = time at MAX; θ_{IC} = angle at IC; θ_{PP} = angle at end of PP; θ_{MAX} = angle at MAX; ω = average angular velocity.

4.3.2 Frontal plane kinematics.

In the frontal plane, during both unopposed and opposed trials, females tended to contact the ground in a slight valgus position (–ve values) which progressively increased between IC and MAX position. In contrast, during both unopposed and opposed trials, males tended to contact the ground in a slight valgus position and moved into a slight varus position (+ve values) at MAX (Table 4.1 and Figure 4.4). The amount of valgus at IC was not significantly different between males and females or between unopposed and opposed trials (males unopposed: -2.2 ± 5.3°;

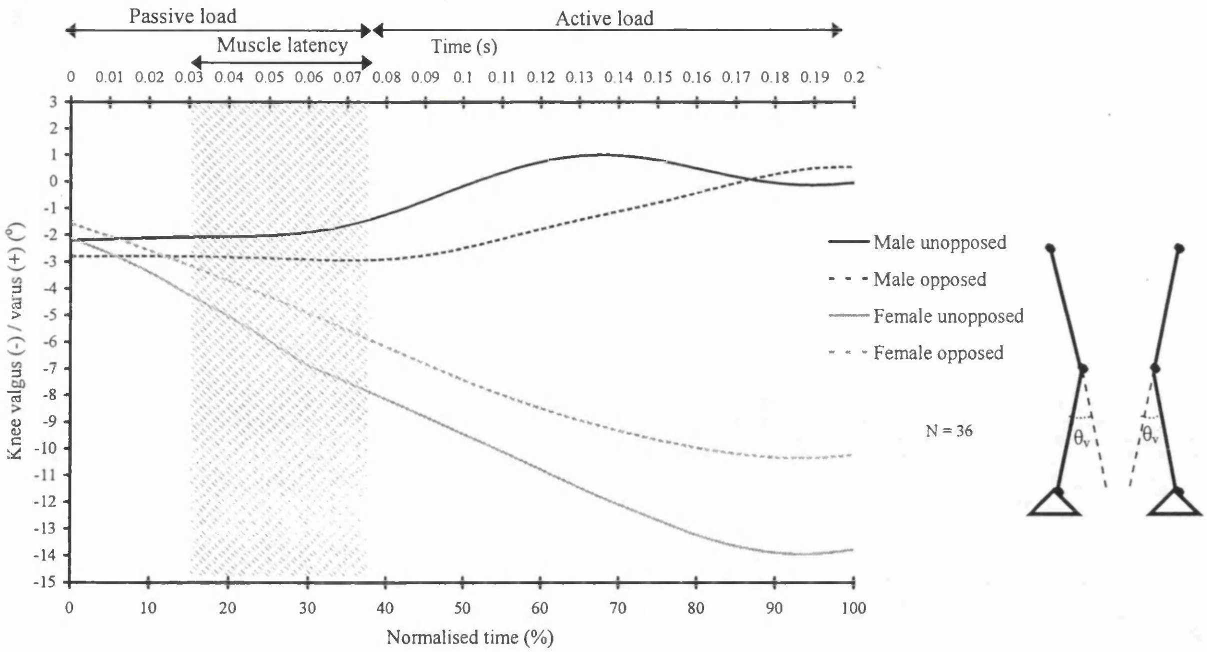


Figure 4.4. Knee valgus/varus (θ_v) between IC and MAX for males and females during unopposed and opposed trials.

males opposed: -2.8 ± 5.9°; females unopposed; -2.1 ± 3.4°; females opposed: -1.6 ± 2.8°). However, the ROM (males unopposed: 3.2 ± 8.0°; males opposed: 3.5 ± 9.6°; females unopposed: 11.8 ± 10.3°; females opposed: 8.8 ± 7.8°) and the MAX valgus angle (males unopposed: -2.2 ± 5.3°; males opposed: -2.9 ± 7.9°; females unopposed: -13.9 ± 11.3°; females opposed: -10.4 ± 7.7°) were significantly greater in females compared to males. There was no significant difference in MAX knee valgus angle and ROM of knee valgus between unopposed and opposed trials (Table 4.1 and Figure 4.4). Males showed no significant difference in frontal plane kinematics between dominant and non-dominant legs during both unopposed and opposed trials (Table 4.2 and Figure 4.5). However, in females, during both unopposed and opposed trials, the non-dominant leg displayed significantly greater maximum knee valgus angle (non-dominant unopposed: -19.4 ± 8.9°; non-dominant opposed: -13.9 ± 8.7°; dominant unopposed: -8.5 ± 11.0°;

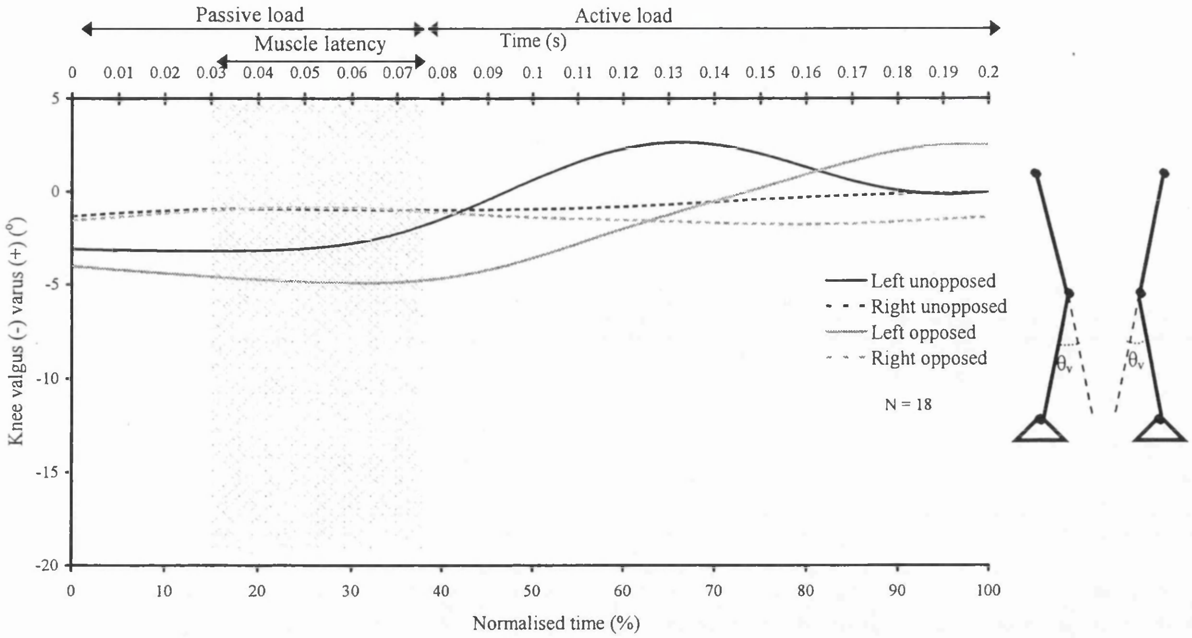


Figure 4.5. Dominant and non-dominant leg knee valgus (–ve) / varus (+ve) (θ_v) between IC and MAX for males during unopposed and opposed trials.

dominant opposed: -6.8 ± 4.5°) and range of motion (non-dominant unopposed: 16.8 ± 7.2°; non-dominant opposed: 12.8 ± 7.6°; dominant unopposed: 6.9 ± 10.7°; dominant opposed: 4.7 ± 5.8°) compared to the dominant leg (Table 4.2 and Figures 4.6).

The gradient of the curves in Figure 4.4 indicate differences between males and females but not between unopposed and opposed trials in average angular velocity of knee valgus/varus during PP and AP (Table 4.3). Females displayed significantly greater average knee valgus angular velocity than males during PP but there was no significant difference between unopposed than opposed trials during PP (males unopposed: 9.6°/s, associated with decreasing valgus; males opposed:

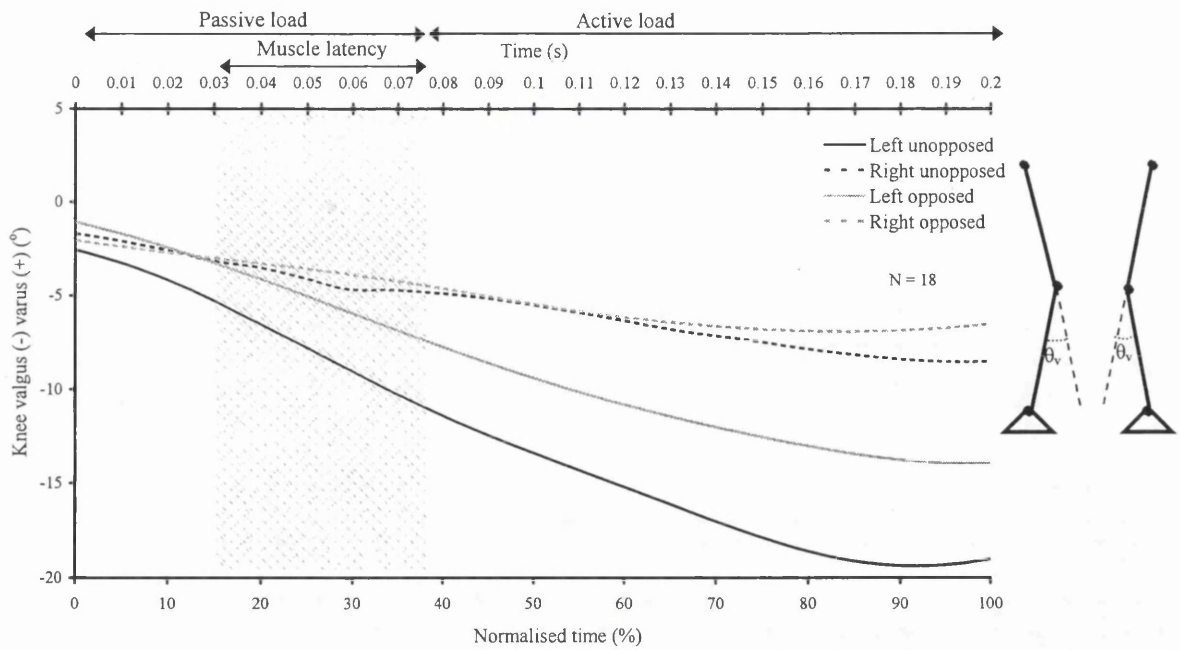


Figure 4.6. Dominant and non-dominant leg knee valgus (–ve) and varus (+ve) (θ_v) between IC and MAX for females during unopposed and opposed trials.

1.7°/s, associated with increasing valgus; females unopposed: 75.1°/s, associated with increasing valgus; females opposed: 59.7°/s, associated with increasing valgus). During AP, the average knee varus angular velocity exhibited by males was similar to the average knee valgus angular velocity exhibited by females and the average knee valgus/varus angular velocity was similar between unopposed and opposed trials (males unopposed: 11.3°/s, associated with decreasing valgus; males opposed; 29.7°/s, associated with decreasing valgus; females unopposed: 47.0°/s, associated with increasing valgus; females opposed: 35.8°/s, associated with increasing valgus).

4.3.3 Inter-hip, inter knee and inter ankle displacements.

Due to the hip joint centres being fixed in the pelvis, it was not surprising that the inter-hip distance remained constant throughout landing (Table 4.4). There was no significant difference in inter-hip distance between males and females or between unopposed and opposed trials at IC or at MIN. There was no significant difference in inter-knee distance at IC (males unopposed: 252 ± 25 mm; males opposed: 244 ± 33 mm; females unopposed: 247 ± 35 mm; females opposed: 228 ± 29 mm) between males and females or between unopposed and opposed trials. However, MIN inter-knee distance was significantly smaller in females than males (males unopposed: 235 ± 24 mm; males opposed: 234 ± 39 mm; females unopposed: 213 ± 44 mm; females opposed: 200 ± 35 mm) and the ROM of inter-knee distance was significantly greater in females than males (males unopposed: 17 ± 19 mm; males opposed: 10 ± 16 mm; females unopposed: 34 ± 24 mm; females opposed: 28 ± 18 mm) (Table 4.4 and Figure 4.7). There was no significant difference in MIN inter-knee distance or ROM of inter-knee distance between unopposed and opposed trials. There was no significant difference between males and females or between unopposed and opposed trials for inter-ankle distance at IC (males unopposed: 318 ± 40 mm; males opposed: 311 ± 58 mm; females unopposed: 324 ± 70 mm; females opposed: 289 ± 46 mm) or MIN inter-ankle distance (males unopposed: 275 ± 33 mm; males opposed: 269 ± 59 mm; females unopposed: 304 ± 71 mm; females opposed: 265 ± 46 mm). The ROM of inter-ankle distance was significantly greater in males than females but not significantly different between unopposed and opposed trials (males unopposed: 43 ± 22 mm; males opposed: 42 ± 27 mm; females unopposed: 19 ± 12 mm; females opposed: 24 ± 16 mm) (Table 4.4 and Figure 4.8).

Table 4.4. Group mean results for inter-hip, inter-knee and inter-ankle distances at initial ground contact (IC), minimum distance (MIN) and for range of motion during landing (ROM) during unopposed and opposed trials (Mean ± standard deviation).

		Males		Females	
		Unopposed (mm)	Opposed (mm)	Unopposed (mm)	Opposed (mm)
Inter-hip distance	IC	160 ± 15	160 ± 15	154 ± 14	154 ± 14
	MIN	160 ± 15	160 ± 15	154 ± 14	154 ± 14
Inter-knee distance	IC	252 ± 25	244 ± 33	247 ± 35	228 ± 29
	MIN†	235 ± 24	234 ± 39	213 ± 44	200 ± 35
	ROM†	17 ± 19	10 ± 16	34 ± 24	28 ± 18
Inter-ankle distance	IC	318 ± 40	311 ± 58	324 ± 70	289 ± 46
	MIN	275 ± 33	269 ± 59	304 ± 71	265 ± 46
	ROM†	43 ± 22	42 ± 27	19 ± 12	24 ± 16

* : No significant difference between unopposed and opposed trials ($p < 0.05$).

† : Significant difference between males and females ($p < 0.05$).

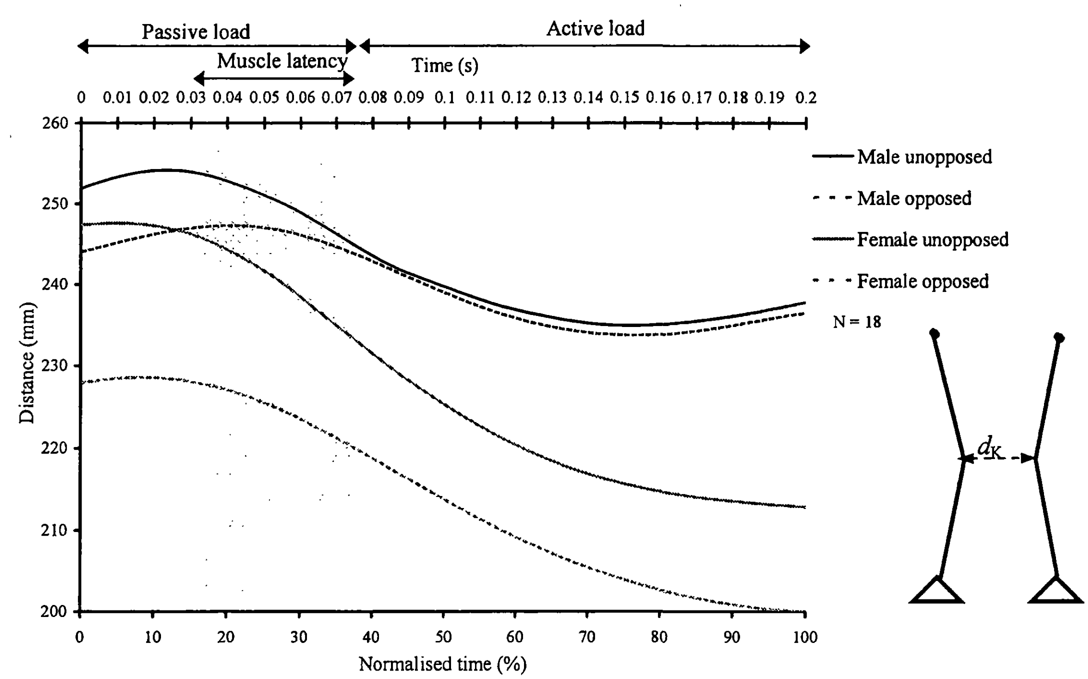


Figure 4.7. Absolute inter-knee joint centre distance (d_K) between IC and MAX for males and females during unopposed and opposed trials.

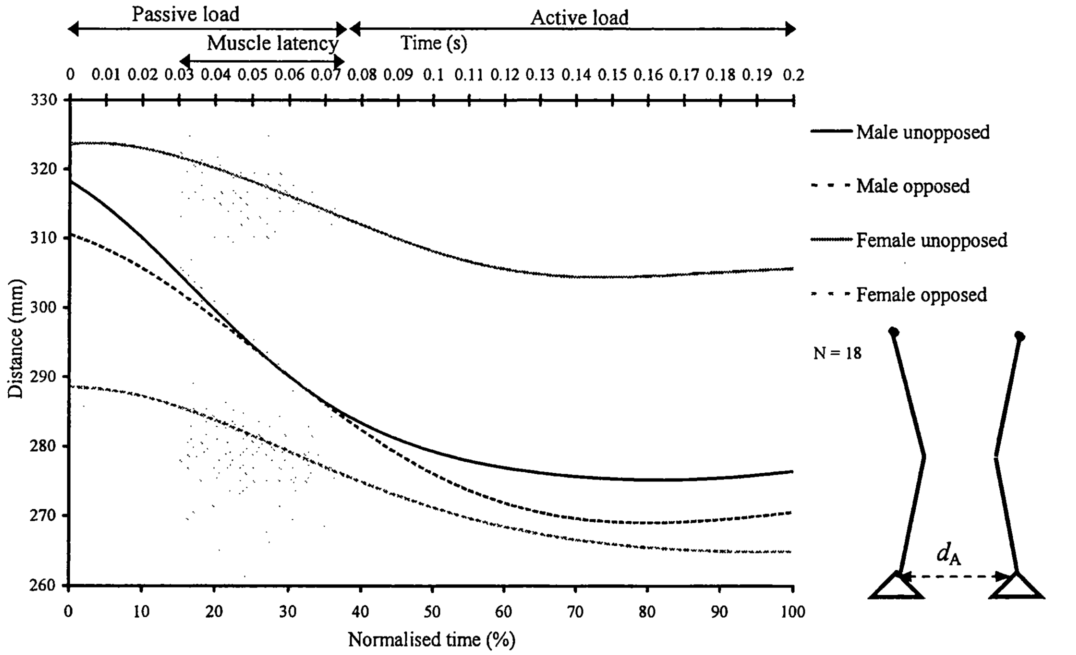


Figure 4.8. Absolute inter-ankle joint centre distance (d_A) between IC and MAX for males and females during unopposed and opposed trials.

Inter-knee distance relative to inter-hip distance was greater in males than females at IC during opposed trials but not during unopposed trials (males unopposed: 157.4 ± 18.6%; males opposed: 152.3 ± 20.6%; females unopposed: 160.3 ± 20.6%; females opposed: 147.7 ± 19.1%). MIN inter-knee distance relative to inter-hip distance was greater in males than females during both unopposed and opposed trials (males unopposed: 146.9 ± 22.9%; males opposed: 146.2 ± 24.6%; females unopposed: 138.1 ± 23.0%; females opposed: 129.7 ± 22.4%) (Table 4.5). As with inter-knee distance, inter-ankle distance relative to inter-hip distance was greater in males than females at IC during opposed trials but not during unopposed trials (males unopposed: 198.8 ± 27.7%; males opposed: 193.9 ± 36.5%; females unopposed: 209.7 ± 37.9%; females opposed: 187.0 ± 30.0%). Inter-ankle distance relative to inter-hip distance was less in males than females at MIN during both unopposed and opposed trials (males unopposed: 172.1 ± 26.7%; males opposed: 168.3 ± 36.6%; females unopposed: 197.6 ± 41.8%; females: 171.8 ± 29.7%).

Table 4.5. Inter-knee and inter-ankle distances at IC and MIN relative to inter-hip distance for males and females during unopposed and opposed trials (mean ± standard deviation).

	Males		Females	
	Unopposed	Opposed	Unopposed	Opposed
A: Inter-hip distance at IC (mm)	160 ± 15	160 ± 15	154 ± 14	154 ± 14
B: MIN inter-hip distance (mm)	160 ± 15	160 ± 15	154 ± 14	154 ± 14
C: Inter-knee distance at IC (mm)	252 ± 24	244 ± 33	247 ± 35	228 ± 29
D: Inter-ankle distance at IC (mm)	318 ± 40	311 ± 58	324 ± 71	289 ± 46
E: MIN inter-knee distance (mm)	235 ± 24	234 ± 39	213 ± 44	200 ± 35
F: MIN inter-ankle distance (mm)	275 ± 33	269 ± 59	304 ± 71	265 ± 46
C/A x 100 (%)	157.4 ± 18.6	152.3 ± 20.6	160.3 ± 20.6	147.7 ± 19.1
D/A x 100 (%)	198.8 ± 27.7	193.9 ± 36.5	209.7 ± 37.9	187.0 ± 30.0
E/B x 100 (%)	146.9 ± 22.9	146.2 ± 24.6	138.1 ± 23.0	129.7 ± 22.4
F/B x 100 (%)	172.1 ± 26.7	168.3 ± 36.6	197.6 ± 41.8	171.8 ± 29.7

4.3.4 Sagittal plane kinetics.

Group mean curves for sagittal plane normalised GRF, angle of the GRF, knee angle, normalised GRF moment arm and normalised GRF moment (+ve = extensor GRF moment, – ve = flexor GRF moment) throughout the landing period are shown for males and females during both unopposed and opposed conditions in Figure 4.9.

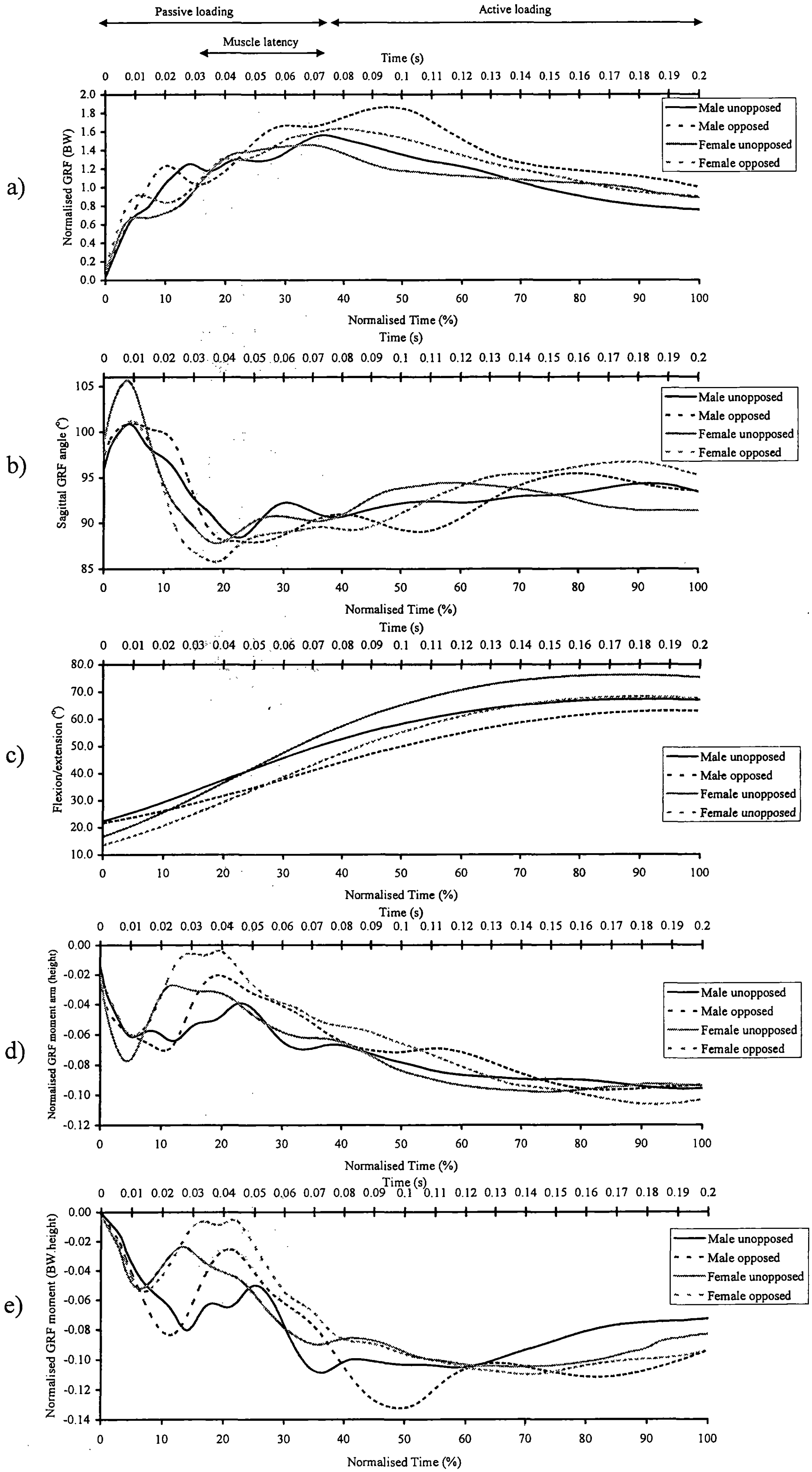


Figure 4.9. Sagittal plane normalised GRF, GRF angle, knee angle, normalised GRF moment arm and normalised GRF moment between IC and MAX for males and females during unopposed and opposed trials.

With regard to normalised GRF (Figure 4.9a), the overall shapes of the curves were similar for males and females and for unopposed and opposed trials, i.e. increase during PP followed by decrease during AP. For most of the landing period, the normalised GRF was greater for males than females and greater for opposed trials than unopposed trials. The initial peak in normalised GRF during PP occurred earlier during opposed trials than unopposed trials and the maximum normalised GRF during landing occurred later in opposed trials than unopposed trials. Females exhibited significantly greater normalised GRF at IC than males (males unopposed: 0.027 ± 0.045 BW; males opposed: 0.049 ± 0.085 BW; females unopposed: 0.090 ± 0.097 BW; females opposed: 0.133 ± 0.120 BW) but there was no significant difference between unopposed and opposed trials. Normalised GRF at the start (AL) of the active loading phase (AP) (males unopposed: 1.519 ± 0.594 BW; males opposed: 1.772 ± 0.485 BW; females unopposed: 1.365 ± 0.477 BW; females opposed: 1.625 ± 0.415 BW) and maximum normalised GRF (males unopposed: 1.561 ± 0.663 BW; males opposed: 1.861 ± 0.595 BW; females unopposed: 1.457 ± 0.477 BW; females opposed: 1.631 ± 0.427 BW) were significantly greater during opposed trials than unopposed trials but not significantly different between males and females. There was no significant difference in normalised GRF between males and females or between unopposed and opposed trials at the start of the muscle latency period (ML) or at MAX.

With regard to the angle of the GRF (>90° = posterior to the vertical, <90° = anterior to the vertical) in the sagittal plane during landing (Figure 4.9b), the shape of the graph was similar in males and females and for unopposed and opposed conditions. The angle of the GRF in the sagittal plane at IC (males unopposed: 96.0 ± 8.7°; males opposed: 97.4 ± 5.1°; females unopposed: 99.0 ± 2.7°; females opposed: $97.7 \pm 3.3^{\circ}$) and maximum angle of the GRF in the sagittal plane (males unopposed: $100.9 \pm 6.8^{\circ}$; males opposed: $101.0 \pm 3.6^{\circ}$; females unopposed: $105.7 \pm 4.7^{\circ}$; females opposed: $101.2 \pm 5.1^{\circ}$) were significantly greater in females than males but not significantly different between unopposed and opposed trials. The angle of the GRF in the sagittal plane was significantly greater in males than females at ML (males unopposed: $92.7 \pm 8.2^{\circ}$; males opposed: $93.0 \pm 6.3^{\circ}$; females unopposed: $89.6 \pm 4.3^{\circ}$; females opposed: $86.9 \pm 2.9^{\circ}$) but not significantly different between unopposed and opposed trials. The minimum angle of the GRF in the sagittal plane was significantly smaller in opposed trials than unopposed trials (males unopposed: $88.4 \pm 5.8^{\circ}$; males opposed: $87.9 \pm 2.2^{\circ}$; females unopposed: $87.8 \pm 3.2^{\circ}$; females opposed: $85.8 \pm 5.0^{\circ}$) but there was no significant difference between males and females. There was no significant difference in the angle of the GRF in the sagittal plane at AL or at MAX between males and females or between unopposed and opposed trials.

Females and males during both unopposed and opposed trials exhibited a progressive increase in knee flexion during the landing phase (Figure 4.9c). Dominant leg knee flexion was significantly greater in males than females and significantly greater in unopposed trials than opposed at IC (males unopposed: $22.4 \pm 4.2^{\circ}$; males opposed: $21.7 \pm 5.7^{\circ}$; females unopposed: $16.7 \pm 5.9^{\circ}$; females opposed: $13.5 \pm 6.0^{\circ}$). Dominant leg MAX knee flexion was significantly greater in females than males and significantly greater in unopposed trials than opposed trials (males unopposed: $67.0 \pm 14.1^{\circ}$; males opposed: $63.0 \pm 11.2^{\circ}$; females unopposed: $75.3 \pm 6.4^{\circ}$; females opposed: $68.2 \pm 9.5^{\circ}$) (Table 4.6 and Figure 4.9c). There was no significant difference in knee flexion angle between males and females or between unopposed and opposed trials at ML or AL.

The pattern of the normalised GRF moment arm (Figure 4.9d) largely mirrored the pattern of the normalised GRF moment (Figure 4.9e) in both males and females and in unopposed and opposed conditions. During PP, females exhibited a smaller peak in normalised GRF moment than males, which occurred earlier during the landing phase in females than in males. During AP, the normalised GRF moment arm and normalised GRF moment were similar in males and females during both unopposed and opposed trials. Females displayed a significantly smaller normalised GRF flexor moment arm at the ML than males (males unopposed: -0.055 ± 0.047 ht; males opposed: -0.041 ± 0.204 ht; -0.030 ± 0.026 ht; females opposed: -0.006 ± 0.198 ht) but there was no significant difference between unopposed and opposed trials. The significantly smaller normalised GRF moment arm at ML in females resulted in a significantly smaller normalised GRF flexor moment in females than males at ML and normalised GRF flexor moment was significantly smaller in opposed trials than unopposed at ML (males unopposed: -0.0647 ± 0.0516 BW.ht; males opposed: -0.0433 ± 0.0353 BW.ht; females unopposed: -0.0347 ± 0.0567 BW.ht; females opposed: -0.0065 ± 0.0325 BW.ht). There was no significant difference in the normalised GRF moment arm or the normalised GRF moment between males and females or between unopposed and opposed trials at IC, AL, at MAX or the maximum and minimum values. Mean stick figures of the angle of the knee, the magnitude and angle of the GRF and the GRF moment arm in the sagittal plane for males and females during unopposed and opposed trials at IC, ML, AL and MAX are shown in Figures 4.10 and 4.11.

Table 4.6. Group mean results for sagittal plane normalised GRF, GRF angle, knee angle, normalised GRF moment arm and GRF normalised moment about the knee (+ve = extensor GRF moment, – ve = flexor GRF moment) at IC, ML, AL, MAX maximum and minimum (Mean ± standard deviation).

Sagittal plane			IC	ML (0.03 s)	AL (0.075 s)	MAX	Maximum	Minimum
Normalised GRF (BW)	Male	Unopposed	0.027 ± 0.045†	1.178 ± 0.471	1.519 ± 0.594*	0.752 ± 0.194	1.561 ± 0.663*	NA
		Opposed	0.049 ± 0.085†	1.052 ± 0.170	1.772 ± 0.485*	0.972 ± 0.415	1.861 ± 0.595*	NA
	Female	Unopposed	0.090 ± 0.097†	1.140 ± 0.435	1.365 ± 0.477*	0.873 ± 0.210	1.457 ± 0.477*	NA
		Opposed	0.133 ± 0.120†	1.160 ± 0.287	1.625 ± 0.415*	0.894 ± 0.378	1.631 ± 0.427*	NA
GRF angle (°)	Male	Unopposed	96.0 ± 8.7†	92.7 ± 8.2†	90.7 ± 7.3	93.4 ± 4.6	100.9 ± 6.8†	88.4 ± 5.8*
		Opposed	97.4 ± 5.1†	93.0 ± 6.3†	90.9 ± 4.2	93.5 ± 3.0	101.0 ± 3.6†	87.9 ± 2.2*
	Female	Unopposed	99.0 ± 2.7†	89.6 ± 4.3†	90.4 ± 4.2	91.4 ± 2.6	105.7 ± 4.7†	87.8 ± 3.2*
		Opposed	97.7 ± 3.3†	86.9 ± 2.9†	89.4 ± 3.4	95.3 ± 4.3	101.2 ± 5.1†	85.8 ± 5.0*
Flexion / extension (°)	Male	Unopposed	22.4 ± 4.2*†	33.4 ± 5.4	51.5 ± 12.1	67.0 ± 14.1*†	NA	NA
		Opposed	21.7 ± 5.7*†	28.8 ± 5.3	43.6 ± 7.8	63.0 ± 11.2*†	NA	NA
	Female	Unopposed	16.7 ± 5.9*†	30.9 ± 6.7	55.7 ± 11.1	75.3 ± 6.4*†	NA	NA
		Opposed	13.5 ± 6.0*†	24.9 ± 5.0	46.7 ± 9.1	68.2 ± 9.5*†	NA	NA
Normalised GRF moment arm (ht)	Male	Unopposed	-0.014 ± 0.043	-0.055 ± 0.047†	-0.067 ± 0.029	-0.096 ± 0.025	-0.096 ± 0.048	-0.014 ± 0.043
		Opposed	-0.028 ± 0.106	-0.041 ± 0.204†	-0.062 ± 0.111	-0.093 ± 0.073	-0.097 ± 0.072	-0.021 ± 0.115
	Female	Unopposed	-0.020 ± 0.018	-0.030 ± 0.026†	-0.063 ± 0.027	-0.094 ± 0.025	-0.098 ± 0.046	-0.020 ± 0.018
		Opposed	-0.021 ± 0.070	-0.006 ± 0.198†	-0.055 ± 0.094	-0.104 ± 0.129	-0.106 ± 0.124	-0.004 ± 0.075
Normalised GRF moment (BW.ht)	Male	Unopposed	-0.0004 ± 0.0078	-0.0647 ± 0.0516*†	-0.1014 ± 0.0283	-0.0720 ± 0.0291	-0.1053 ± 0.0433	-0.0044 ± 0.0078
		Opposed	-0.0014 ± 0.0166	-0.0433 ± 0.0353*†	-0.1110 ± 0.0541	-0.0908 ± 0.0303	-0.1325 ± 0.0681	-0.0097 ± 0.0166
	Female	Unopposed	-0.0018 ± 0.0107	-0.0347 ± 0.0567*†	-0.0860 ± 0.0357	-0.0822 ± 0.0291	-0.1048 ± 0.0292	-0.0100 ± 0.0107
		Opposed	-0.0028 ± 0.0182	-0.0065 ± 0.0325*†	-0.0876 ± 0.038	-0.0923 ± 0.048	-0.1100 ± 0.0309	-0.0055 ± 0.0227

*: Significant difference between unopposed and opposed trials.

†: Significant difference between males and females.

a)

b)

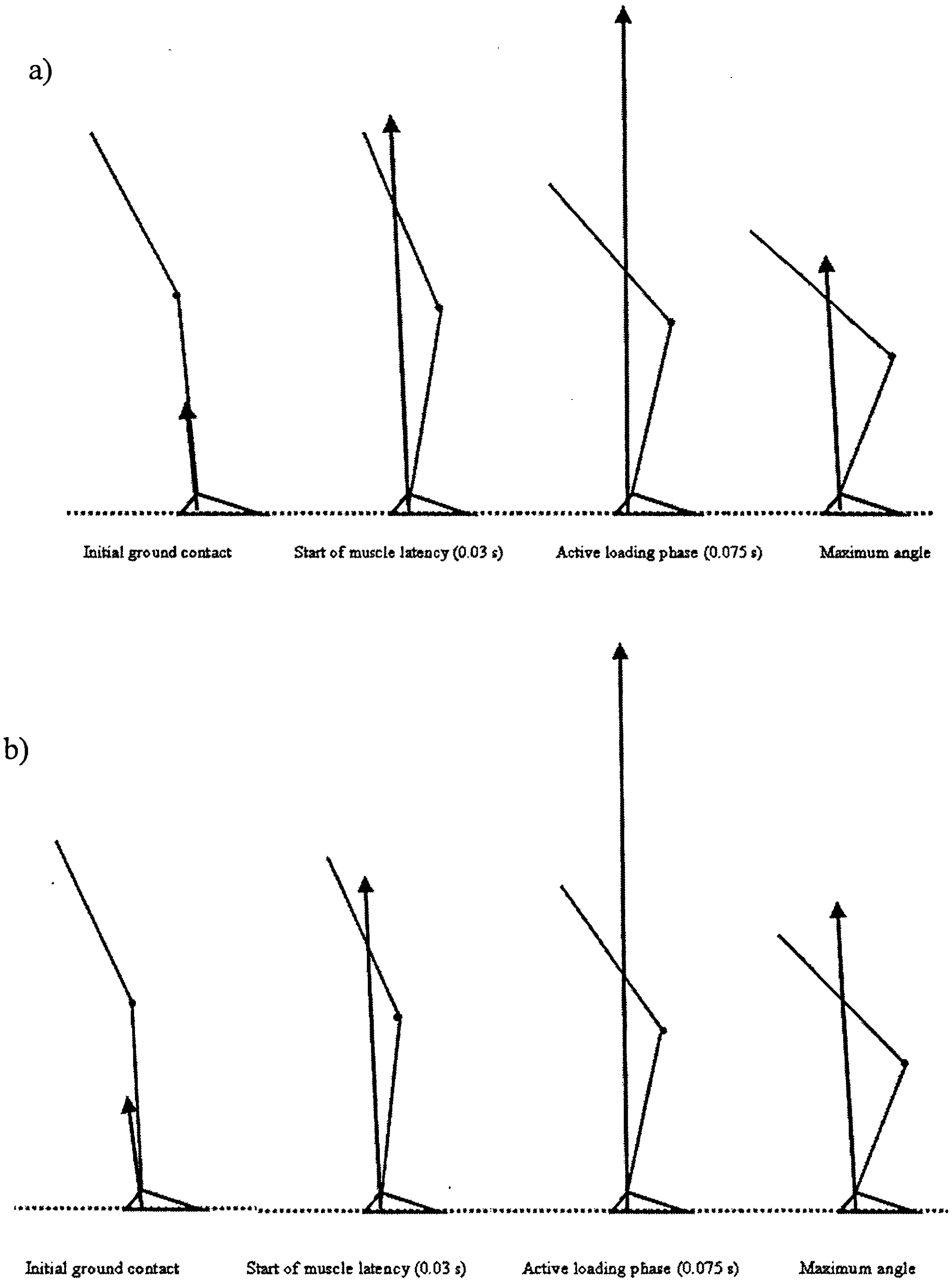
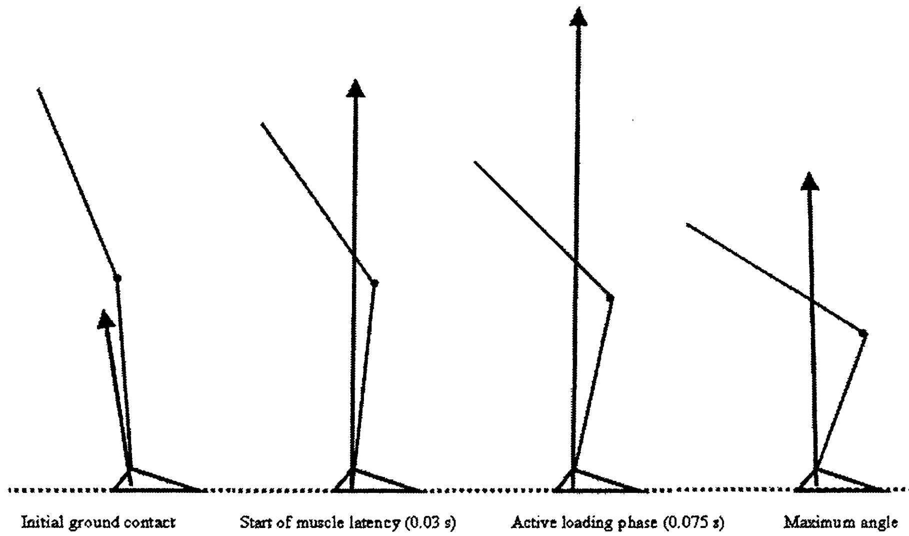


Figure 4.10. Mean stick figures of males' dominant lower leg and GRF vector in the sagittal plane at initial ground contact (IC), start of muscle latency (ML), start of active loading (AL) and maximum angle of the knee (MAX) during a) unopposed trials and b) opposed trials.

a)



b)

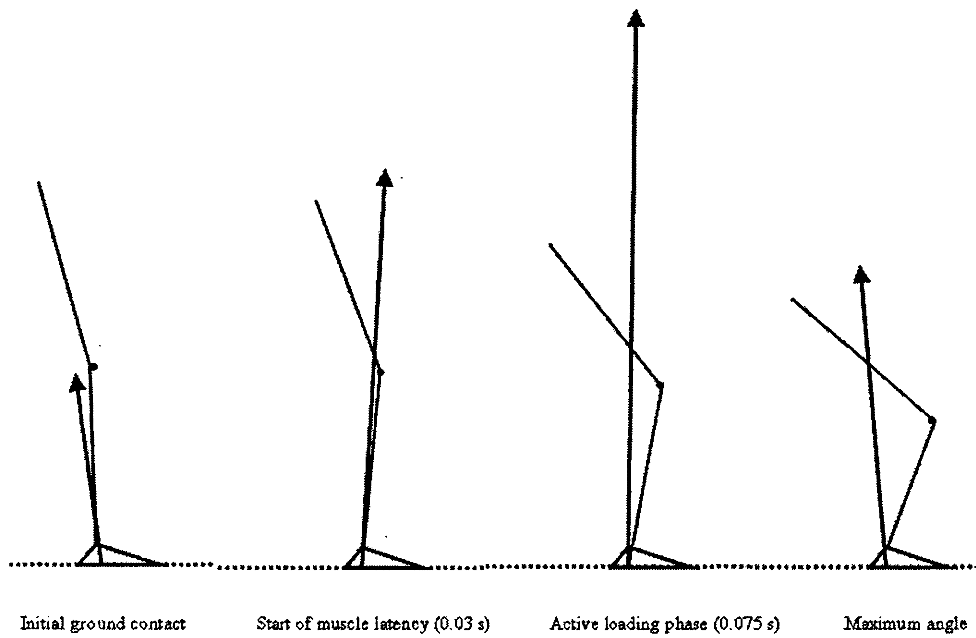


Figure 4.11. Mean stick figures of females' dominant lower leg and GRF vector in the sagittal plane at initial ground contact (IC), start of muscle latency (ML), start of active loading (AL) and maximum angle of the knee (MAX) during a) unopposed trials and b) opposed trials.

The peak normalised GRF flexion moment (–ve) and the peak normalised GRF extension moment (+ve) observed at any instant throughout all unopposed and all opposed trials for males and throughout all unopposed and all opposed trials for females were identified (Table 4.7 and Figure 4.12 and 4.13) (i.e. individual data points for the peak values identified over a number of trials). The peak normalised GRF extension moment occurred during passive loading for males and females during both unopposed and opposed trials. The peak normalised GRF flexion moment occurred during active loading for both males and females during both unopposed and opposed conditions. The peak normalised GRF extension moment was greater during opposed trials than during unopposed trials for males and females, but the peak normalised GRF flexor moment was greater during unopposed trials than opposed trials for males and females. The peak normalised GRF extension moment was greater in males than females during unopposed trials, but was greater in females than males during opposed trials. The peak normalised GRF flexor moment was greater in males than females for both unopposed and opposed conditions. Also, the knee was less flexed for females at peak normalised GRF extensor moment than for the males during both unopposed and opposed trials. Stick Figures showing the angle of the knee, relative magnitude of the GRF, angle of the GRF and the normalised GRF moment arm at the peak normalised GRF extension moment and peak normalised GRF flexion moment about the knee in the sagittal plane found in all male subjects across all unopposed and all opposed trials and all female subjects across all unopposed and all opposed trials (corresponding to data sets in Table 4.7) are shown in Figures 4.12 and 4.13.

Table 4.7. Peak extension and flexion normalised GRF moments and corresponding data for all males across all unopposed and all opposed trials and for all females across all unopposed and all opposed trials (peak individual data points observed for each group).

	Males				Females			
	Unopposed		Opposed		Unopposed		Opposed	
	Extension	Flexion	Extension	Flexion	Extension	Flexion	Extension	Flexion
Subject mass (kg)	65.5	65.5	70.0	74.0	56.5	55.2	52.8	56.5
Subject height (m)	1.70	1.70	1.75	1.91	1.70	1.80	1.69	1.69
Normalised trial time (%)	11.1	22.2	19.2	50.0	21.0	33.3	14.3	88.9
Knee angle (flexion) (°)	24.9	50.5	30.2	46.4	22.5	42.7	14.3	62.8
Fx (BW)	0.085	0.187	0.239	-0.056	0.011	0.130	0.241	-0.248
Fz (BW)	1.298	3.345	1.621	2.856	0.902	2.890	2.012	1.530
F_R (BW)	1.306	3.35	1.639	2.857	0.910	2.895	2.026	1.550
Moment arm (ht)	0.0172	-0.0913	0.0160	-0.1013	0.0186	-0.0856	0.0343	-0.1181
Moment (BW.ht)	0.0225	-0.3059	0.02659	-0.2896	0.0170	-0.2479	0.0706	-0.1840

FR = Resultant GRF in the sagittal plane.

4.3.5 Frontal plane kinetics.

Group mean curves for normalised GRF, angle of the GRF, knee angle, normalised GRF moment arm and normalised GRF moment (+ve = varus GRF moment, –ve = valgus GRF moment) throughout the landing period are shown for males and females in unopposed and opposed conditions in the frontal plane in Figure 4.14.

Since Fy (mediolateral force) and Fx (anterioposterior force) were small relative to Fz (vertical force) during landing, the resultant normalised GRF in the frontal plane (Figure 4.14a) was very similar to the resultant normalised GRF in the sagittal plane. Therefore as with the resultant normalised GRF in the sagittal plane, the resultant normalised GRF in the frontal plane was similar in shape in males and females during unopposed and opposed trials and was greater for males than females and greater for opposed trials compared to unopposed trials during most of

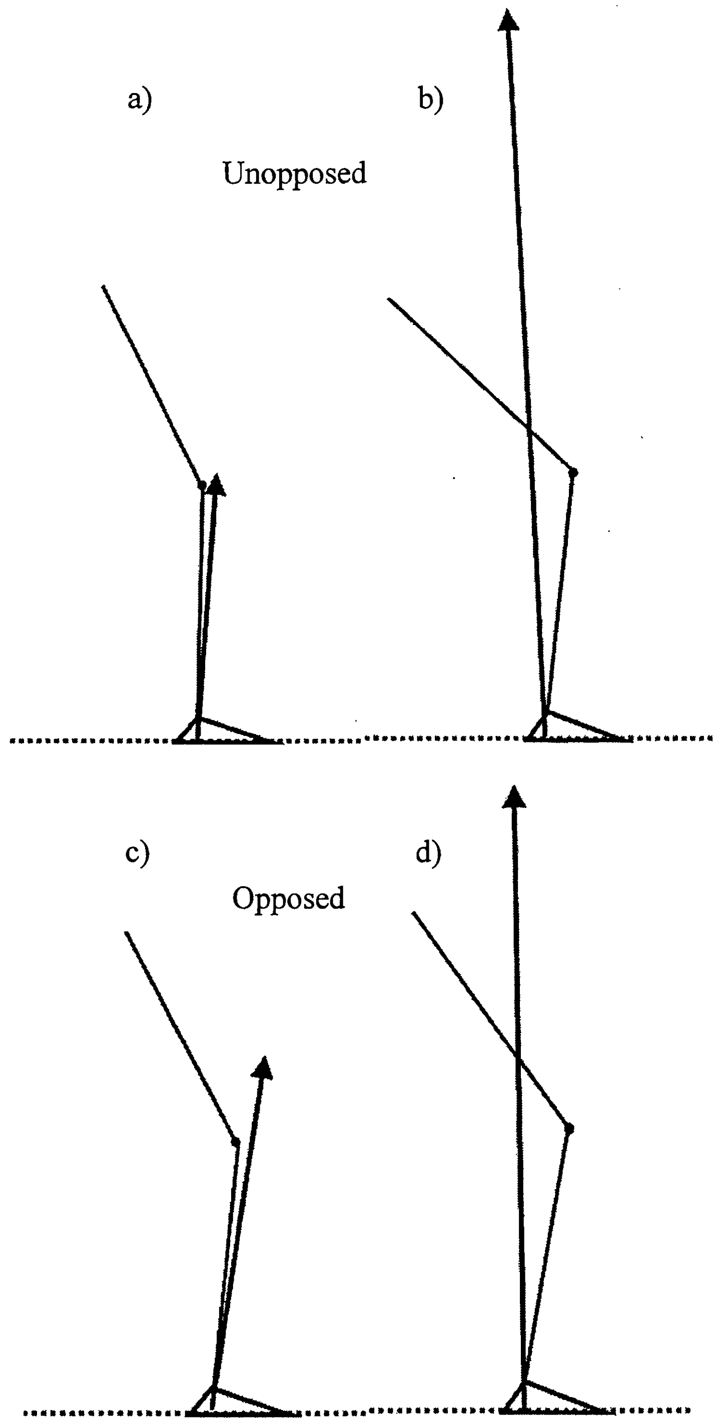


Figure 4.12. Stick figures for males of a) peak extension GRF moment about the knee during unopposed trials, b) peak flexion GRF moment about the knee during unopposed trials, c) peak extension GRF moment about the knee during opposed trials and d) peak flexion GRF moment during opposed trials about the knee.

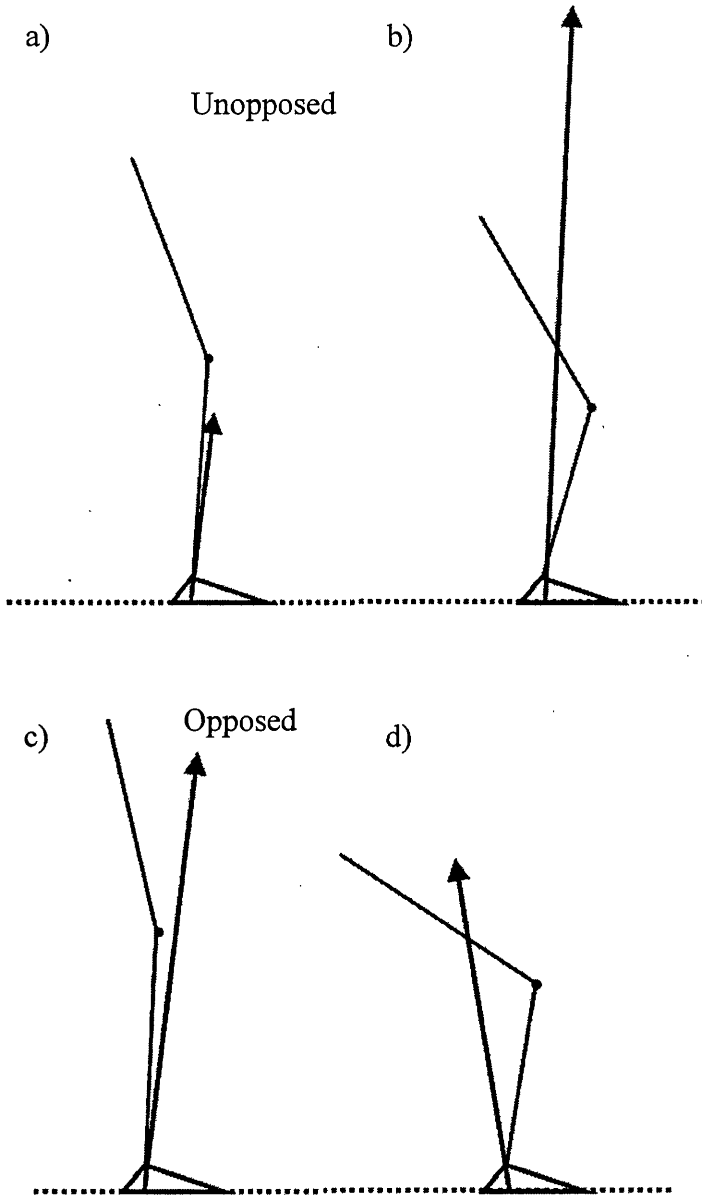


Figure 4.13. Stick figures for females of a) peak extension GRF moment about the knee during unopposed trials, b) peak flexion GRF moment about the knee during unopposed trials, c) peak extension GRF moment about the knee during opposed trials and d) peak flexion GRF moment during opposed trials about the knee.

the landing phase. Females exhibited significantly greater normalised GRF at IC than males (males unopposed: 0.027 ± 0.027 BW; males opposed: 0.049 ± 0.085 BW; females unopposed: 0.090 ± 0.096 BW; females opposed: 0.132 ±

a)
b)
c)
d)
e)

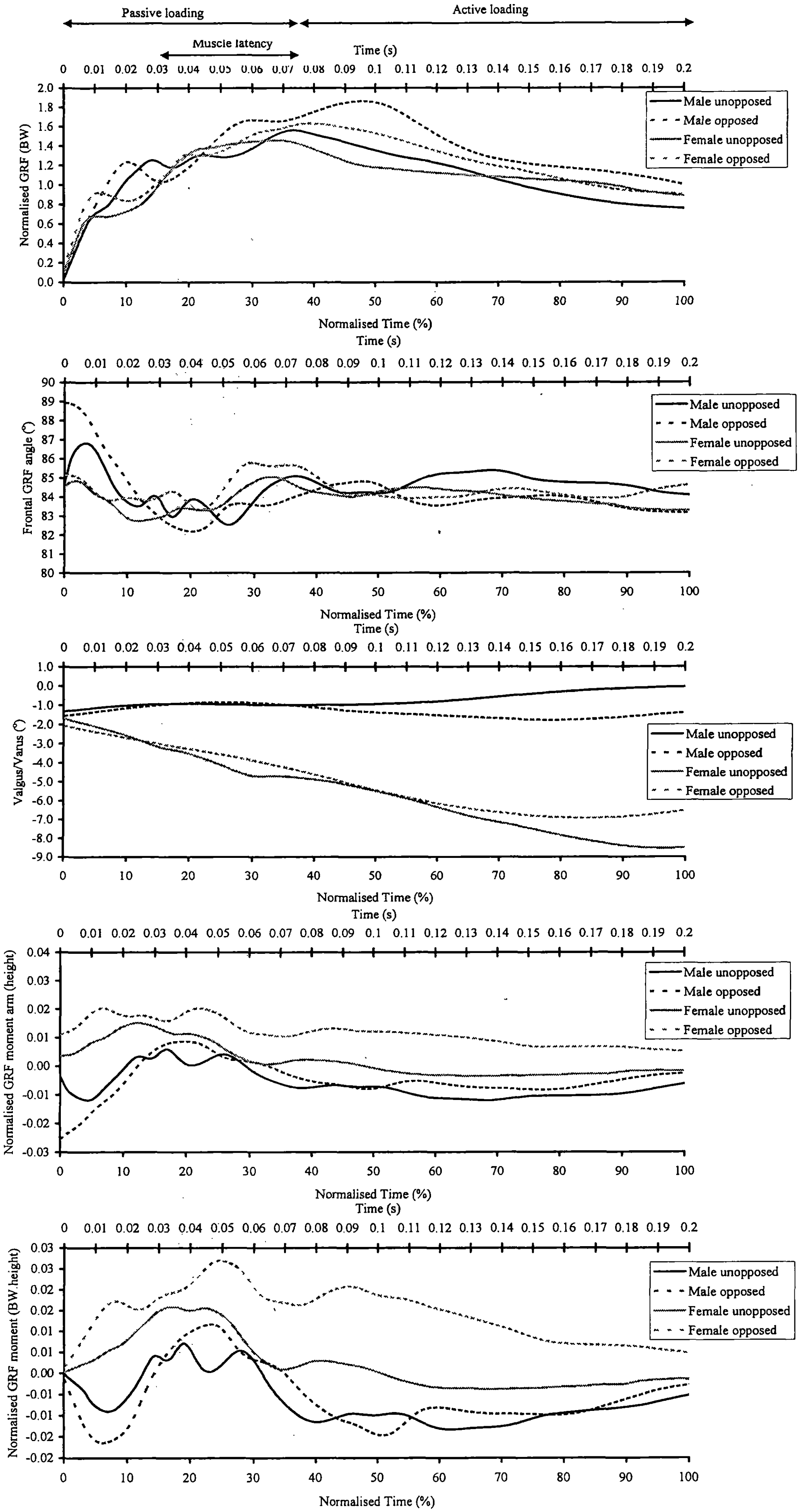


Figure 4.14. Frontal plane normalised GRF, GRF angle, knee angle, normalised GRF moment arm and normalised GRF moment between IC and MAX for males and females during unopposed and opposed trials.

0.119 BW) but there was no significant difference between unopposed and opposed conditions. Normalised GRF in the frontal plane was significantly greater at AL (males unopposed: 1.514 ± 0.594 BW; males opposed: 1.778 ± 0.486 BW; females unopposed: 1.368 ± 0.472 BW; females opposed: 1.601 ± 0.412 BW) and maximum (males unopposed: 1.559 ± 0.666 BW; males opposed: 1.864 ± 0.595 BW; females unopposed: 1.458 ± 0.479 BW; females opposed: 1.604 ± 0.421 BW) for opposed trials compared to unopposed trials but there was no significant difference between males and females. There was no significant difference between males and females or between unopposed and opposed trials for normalised GRF at ML or MAX.

The pattern of the GRF angle in the frontal plane (>90° = lateral to the vertical, <90° = medial to the vertical) was similar for males and females during unopposed and opposed trials, i.e., rapid decrease between IC and ML followed by a slight increase between ML and AL followed by fairly constant value between AL and MAX. The GRF angle in the frontal plane (Figure 4.14b) was significantly greater at IC in opposed trials than in unopposed trials (males unopposed: 84.7 ± 7.9°; males opposed: 88.9 ± 5.3°; females unopposed: 84.6 ± 6.1°; females: 85.1 ± 5.3°) but there was no significant difference between males than females. The maximum GRF angle in the frontal plane was significantly greater in males than females (males unopposed: 86.8 ± 3.8°; males opposed: 88.9 ± 5.3°; females unopposed: 84.9 ± 4.3°; females opposed: 85.8 ± 4.5°) but there was no significant difference between unopposed and opposed trials. The was no significant difference between males and females or between unopposed and opposed conditions for the GRF angle in the frontal plane at ML, AL, MAX or minimum GRF angle.

In the frontal plane, during both unopposed and opposed trials, females tended to contact the ground in a slight valgus position (–ve values) which progressively increased between IC and MAX position. In contrast, during both unopposed and opposed trials, males tended to contact the ground in a slight valgus position and maintained a slight valgus position throughout the landing phase (Figure 4.14c). The amount of valgus at IC, ML and AL were not significantly different between males and females or between unopposed and opposed trials. However, the knee valgus angle at MAX (males unopposed: $-0.0 \pm 10.1^{\circ}$; males opposed: $-1.4 \pm 9.2^{\circ}$; females unopposed: $-8.5 \pm 10.9^{\circ}$; females opposed: $-6.8 \pm 4.5^{\circ}$) and the maximum knee valgus angle (males unopposed: $-1.3 \pm 5.5^{\circ}$; males opposed: $-1.8 \pm 8.8^{\circ}$; females unopposed: $-8.5 \pm 11.0^{\circ}$; females opposed: $-6.9 \pm 4.4^{\circ}$) were significantly greater in females compared to males but there was no significant difference between unopposed and opposed trials (Table 4.8 and Figure 4.14c).

As in the sagittal plane, the pattern of the normalised GRF moment arm (Figure 4.14d) largely mirrored the pattern of the normalised GRF moment (Figure 4.14e) in both males and females during unopposed and opposed conditions in the frontal plane. The normalised GRF moment arm and the normalised GRF moment remained in varus throughout the landing phase in opposed trials for females and was varus between IC and approximately 50% normalised time for females during unopposed trials. For females, Figures 4.14d and 4.14e show a slight increase in varus normalised GRF moment arm and varus normalised GRF moment during PP and a slight decrease during AP for both unopposed and opposed trials. However, for males during both unopposed and opposed trials, the normalised GRF moment arm and normalised GRF moment in the frontal plane were valgus at IC, which

Table 4.8. Group mean results for frontal plane normalised GRF, GRF angle, knee angle, normalised GRF moment arm and normalised GRF moment about the knee (+ve = varus GRF moment, –ve = valgus GRF moment) at IC, ML, AL, MAX maximum and minimum (Mean ± standard deviation).

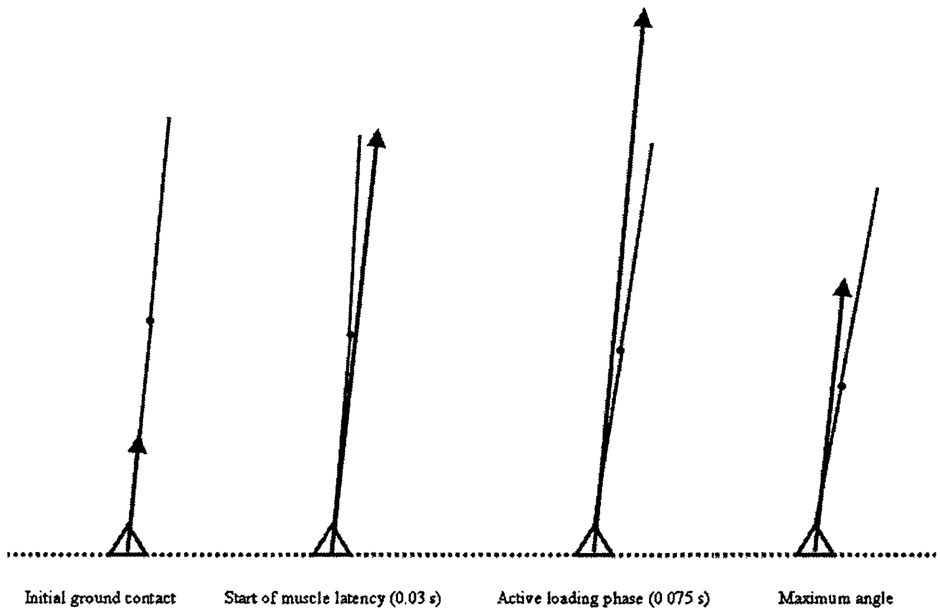
Frontal plane			IC	ML (0.03 s)	AL (0.075 s)	MAX	Maximum	Minimum
Normalised GRF (BW)	Male	Unopposed	0.027 ± 0.027†	1.177 ± 0.471	1.514 ± 0.594*	0.754 ± 0.196	1.559 ± 0.666*	NA
		Opposed	0.049 ± 0.085†	1.054 ± 0.173	1.778 ± 0.486*	0.977 ± 0.418	1.864 ± 0.595*	NA
	Female	Unopposed	0.090 ± 0.096†	1.149 ± 0.440	1.368 ± 0.472*	0.879 ± 0.212	1.458 ± 0.479*	NA
		Opposed	0.132 ± 0.119†	1.150 ± 0.302	1.601 ± 0.412*	0.890 ± 0.378	1.604 ± 0.421*	NA
GRF angle (°)	Male	Unopposed	84.7 ± 7.9*	83.9 ± 4.8	85.0 ± 2.1	84.1 ± 3.1	86.8 ± 3.8†	82.5 ± 4.9
		Opposed	88.9 ± 5.3*	83.0 ± 4.3	84.3 ± 2.6	83.2 ± 2.4	88.9 ± 5.3†	82.2 ± 1.6
	Female	Unopposed	84.6 ± 6.1*	82.9 ± 3.4	84.4 ± 2.1	83.3 ± 2.1	84.9 ± 4.3†	82.7 ± 3.5
		Opposed	85.1 ± 5.3*	84.6 ± 3.3	85.3 ± 2.8	84.6 ± 5.5	85.8 ± 4.5†	83.3 ± 2.7
Valgus / varus (°)	Male	Unopposed	-1.3 ± 5.5	-0.9 ± 6.4	-1.0 ± 7.5	-0.0 ± 10.1†	-0.0 ± 10.1	-1.3 ± 5.5†
		Opposed	-1.6 ± 6.1	-0.1 ± 7.0	-1.1 ± 7.8	-1.4 ± 9.2†	-0.8 ± 7.5	-1.8 ± 8.8†
	Female	Unopposed	-1.7 ± 3.5	-3.2 ± 4.2	-4.8 ± 6.3	-8.5 ± 10.9†	-1.7 ± 3.5	-8.5 ± 11.0†
		Opposed	-2.1 ± 3.0	-3.0 ± 3.2	-4.5 ± 4.4	-6.8 ± 4.5†	-2.1 ± 3.0	-6.9 ± 4.4†
Normalised GRF moment arm (ht)	Male	Unopposed	-0.004 ± 0.020	0.003 ± 0.022†	-0.008 ± 0.017*†	-0.006 ± 0.018	0.006 ± 0.021†	-0.012 ± 0.018
		Opposed	-0.025 ± 0.061	0.006 ± 0.101†	-0.005 ± 0.044*†	-0.003 ± 0.026	0.009 ± 0.049†	-0.025 ± 0.061
	Female	Unopposed	0.003 ±0.012	0.014 ± 0.011†	0.002 ± 0.009*†	-0.002 ± 0.010	0.015 ± 0.015†	-0.004 ± 0.021
		Opposed	0.010 ± 0.073	0.017 ± 0.069†	0.001 ± 0.048*†	0.005 ± 0.034	0.020 ± 0.057†	0.005 ± 0.034
Normalised GRF moment (BW.ht)	Male	Unopposed	-0.0001 ± 0.097	0.0037 ± 0.0316†	-0.0115 ± 0.0202*†	-0.0047 ± 0.0145	0.0072 ± 0.0263†	-0.0133 ± 0.0189
		Opposed	-0.0012 ± 0.0098	0.0058 ± 0.0173†	-0.0085 ± 0.0212*†	-0.0025 ± 0.0106	0.0116 ± 0.0170†	-0.0164 ± 0.0176
	Female	Unopposed	0.0003 ± 0.0152	0.0159 ± 0.0194†	0.0029 ± 0.0122*†	-0.0015 ± 0.0091	0.0159 ± 0.0194†	-0.0038 ± 0.0118
		Opposed	0.0013 ± 0.0184	0.0192 ± 0.0199†	0.0187 ± 0.0200*†	0.0047 ± 0.0127	0.0208 ± 0.0199†	0.0047 ± 0.0127

*: Significant difference between unopposed and opposed trials.
†: Significant difference between males and females.

increased slightly then decreased until it changed to varus close to ML. The normalised GRF moment arm and normalised GRF moment in the frontal plane then changed back to valgus at approximately 30% normalised time and remained in valgus until MAX. The varus normalised GRF moment arm (males unopposed: 0.003 ± 0.022 ht; males opposed: 0.006 ± 0.101 ht; females unopposed: 0.014 ± 0.011 ht; females opposed: 0.017 ± 0.069 ht) and moment (males unopposed: 0.0037 ± 0.0316 BW.ht; males opposed: 0.0058 ± 0.0173 BW.ht; females unopposed: 0.0159 ± 0.0194 BW.ht; females opposed: 0.0192 ± 0.0199 BW.ht) at ML were significantly greater in females than males but there was no significant difference between unopposed and opposed trials. At AL, the valgus normalised GRF moment arm (males unopposed: -0.008 ± 0.017 ht; males opposed: -0.005 ± 0.044 ht) and the valgus normalised GRF moment in males (males unopposed: -0.0115 ± 0.0202 BW.ht: males opposed: -0.0085 ± 0.0212 BW.ht) were significantly different from the varus normalised GRF moment arm (females unopposed: 0.002 ± 0.009 ht; females opposed: 0.001 ± 0.048 ht) and the varus normalised GRF moment (females unopposed: 0.0029 ± 0.0122 BW.ht; females opposed: 0.0187 ± 0.0200 BW.ht) in females and frontal plane normalised GRF moment arm and moment were significantly different between unopposed and opposed trials at AL. The maximum normalised varus GRF moment arm (males unopposed: 0.006 ± 0.021 ht; males opposed: 0.009 ± 0.049 ht; females unopposed: 0.015 ± 0.015 ht; females opposed: 0.020 ± 0.057 ht) and the maximum normalised varus GRF moment about the knee (males unopposed: 0.0072 ± 0.0263 BW.ht; males opposed: 0.0116 ± 0.0170 BW.ht; females unopposed: 0.0159 ± 0.0194 BW.ht; females opposed: 0.0208 ± 0.0199 BW.ht) were significantly greater in females than males but there was no significant difference between unopposed and opposed trials. There was no significant difference in the normalised GRF moment arm or normalised GRF moment about the knee in the frontal plane at IC, MAX or at minimum between males and females or between unopposed and opposed conditions. Mean stick figures of the angle of the knee, the magnitude and angle of the GRF and the normalised GRF moment arm in the frontal plane for males and females during unopposed and opposed trials are shown in Figures 4.15 and 4.16.

The peak normalised GRF varus moment (+ve) and the peak normalised GRF valgus moment (–ve) observed at any instant throughout all unopposed and all opposed trials for males and throughout all unopposed and all opposed trials for females were identified (Table 4.9 and Figures 4.17 and 4.18) (i.e. individual data points for the peak values identified over a number of trials). The peak normalised GRF valgus and varus moments occurred earlier during the trial for the female subjects than the male subjects during both unopposed and opposed trials. The peak normalised GRF valgus and varus moments occurred earlier during unopposed trials than opposed trials for males but earlier during opposed trials than unopposed trials for females. The peak normalised GRF varus moment was similar in males and females during unopposed trials but the peak normalised GRF varus moment was greater in females than males during opposed trials. The peak normalised GRF valgus moment was greater in males than females during both unopposed and opposed trials and the peak normalised GRF valgus moment was greater during opposed trials than unopposed trials for both males and females. For both the peak normalised GRF varus moment and the peak normalised GRF valgus moment, the normalised GRF moment arm was greater during unopposed trials than opposed trials for males but greater during opposed trials than unopposed trials for females. Also, the normalised GRF moment arm at peak normalised GRF varus moment and the peak normalised GRF valgus moment were greater in males than females

a)



b)

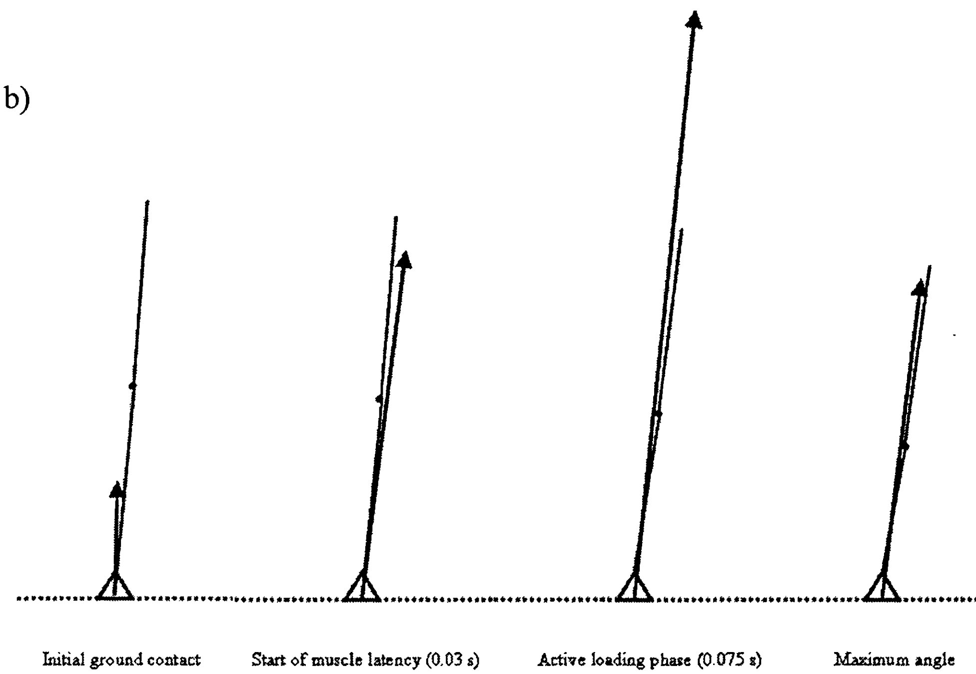


Figure 4.15. Mean stick figures of males' dominant lower leg and GRF vector in the frontal plane at initial ground contact (IC), start of muscle latency (ML), start of active loading (AL) and maximum angle of the knee (MAX) during a) unopposed trials and b) opposed trials.

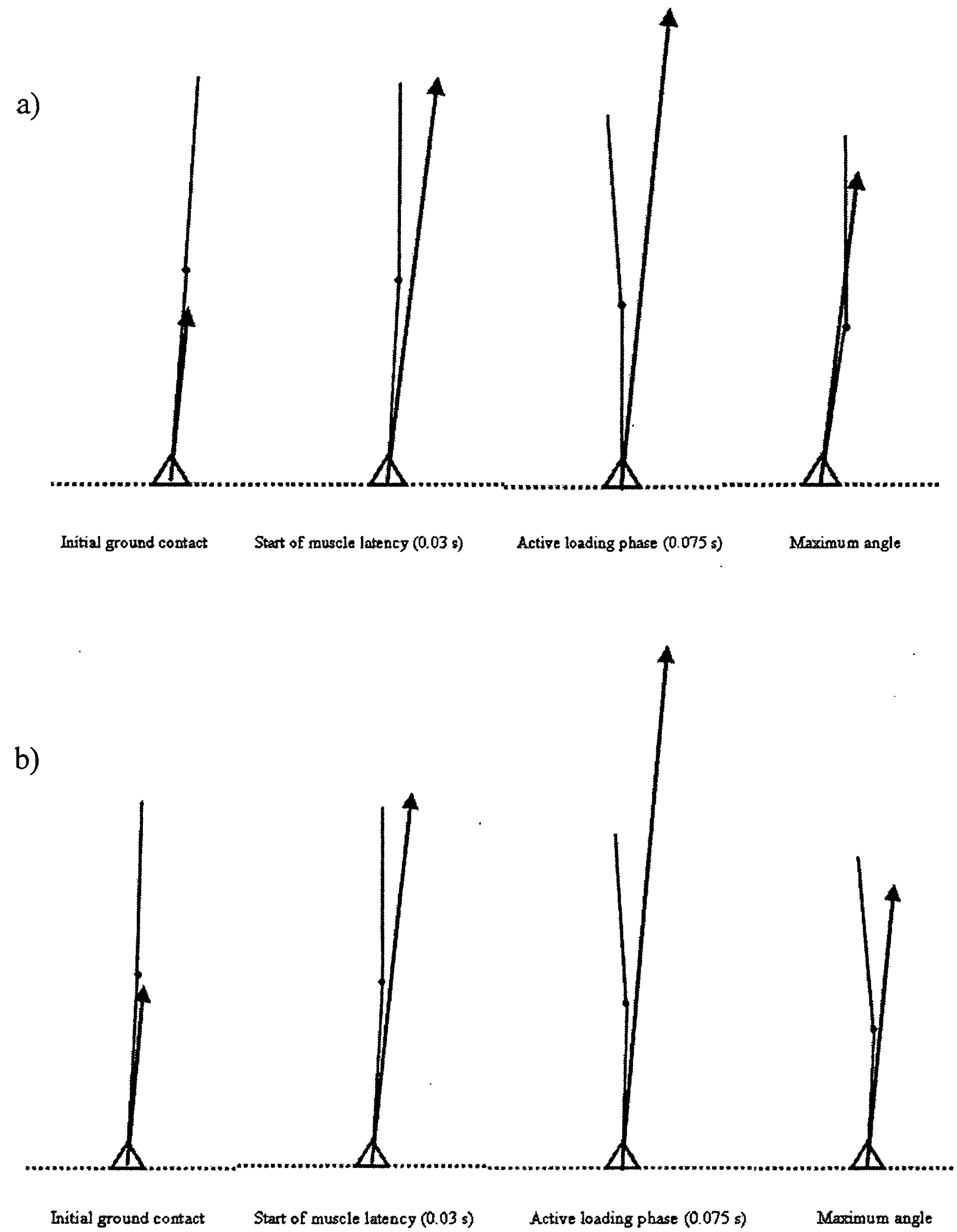


Figure 4.16. Mean stick figures of females' dominant lower leg and GRF vector in the frontal plane at initial ground contact (IC), start of muscle latency (ML), start of active loading (AL) and maximum angle of the knee (MAX) during a) unopposed trials and b) opposed trials.

during unopposed trials but greater in females than males during opposed trials. Stick Figures showing the angle of the knee, relative magnitude of the GRF, angle of the GRF and the normalised GRF moment arm at the peak normalised GRF valgus moment and peak normalised GRF varus moment about the knee in the frontal plane for males during unopposed and opposed trials and females during unopposed and opposed trials (corresponding to data sets in Table 4.9) are shown in Figure 4.17 and 4.18.

Table 4.9. Peak valgus and varus normalised GRF moments and corresponding data for all males across all unopposed and all opposed trials and for all females across all unopposed and opposed trials (peak individual data points observed for each group).

	Males				Females			
	Unopposed		Opposed		Unopposed		Opposed	
	Varus	Valgus	Varus	Valgus	Varus	Valgus	Varus	Valgus
Subject mass (kg)	65.5	70.0	65.0	71.1	52.8	70.7	52.8	52.8
Subject height (m)	1.70	1.74	1.7	1.75	1.59	1.63	1.69	1.69
Normalised trial time (%)	22.2	31.4	23.8	50.0	21.6	16.7	4.7	15.8
Knee angle (valgus/varus) (°)	4.7	4.5	3.9	2.4	-6.4	-0.8	-4.1	-10.7
Fy (BW)	0.464	-0.274	0.428	0.032	0.319	0.186	0.233	-0.033
Fz (BW)	2.217	2.209	2.344	2.763	2.435	1.858	2.012	1.183
F_R (BW)	2.225	2.262	2.383	2.763	2.455	1.894	2.025	1.184
Moment arm (ht)	0.0336	-0.0309	0.0300	-0.0258	0.0303	-0.0131	0.0479	-0.0314
Moment (BW.ht)	0.0748	-0.0698	0.0720	-0.0716	0.0744	-0.0248	0.0979	-0.0378

FR = Resultant GRF in the frontal plane.

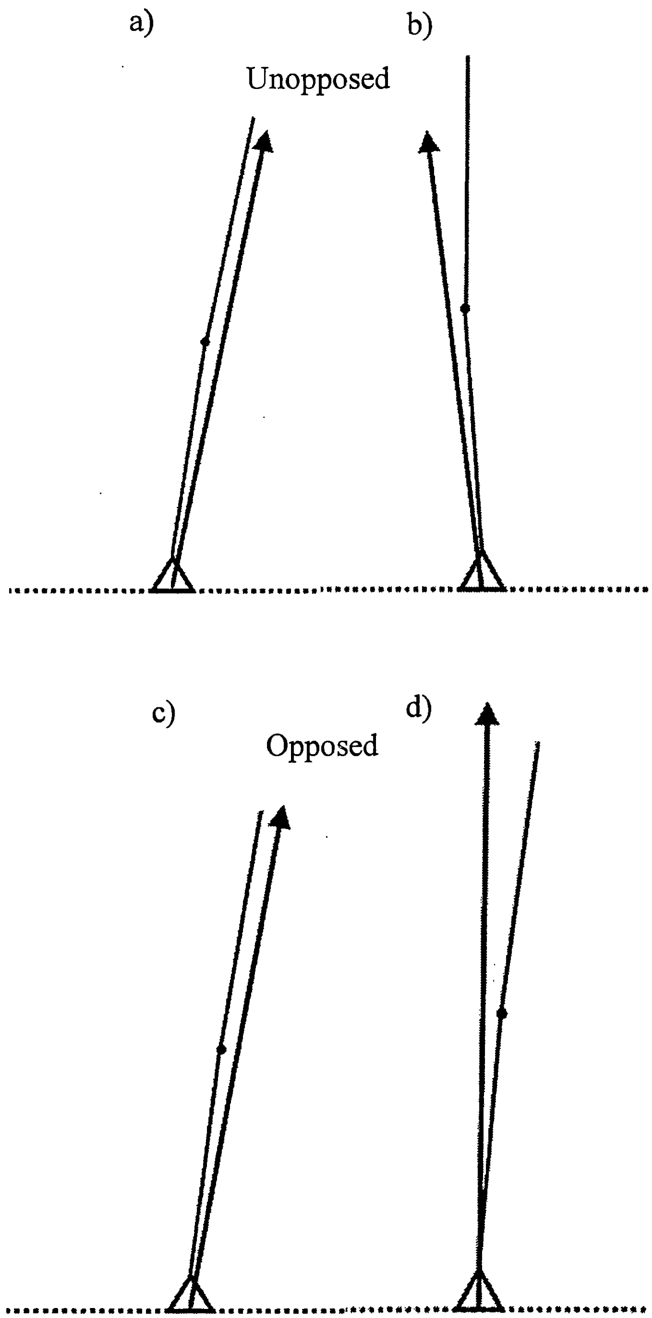


Figure 4.17. Stick figures for males of a) peak varus GRF moment about the knee during unopposed trials, b) peak valgus GRF moment about the knee during unopposed trials, c) peak varus GRF moment about the knee during opposed trials and d) peak valgus GRF moment during opposed trials about the knee.

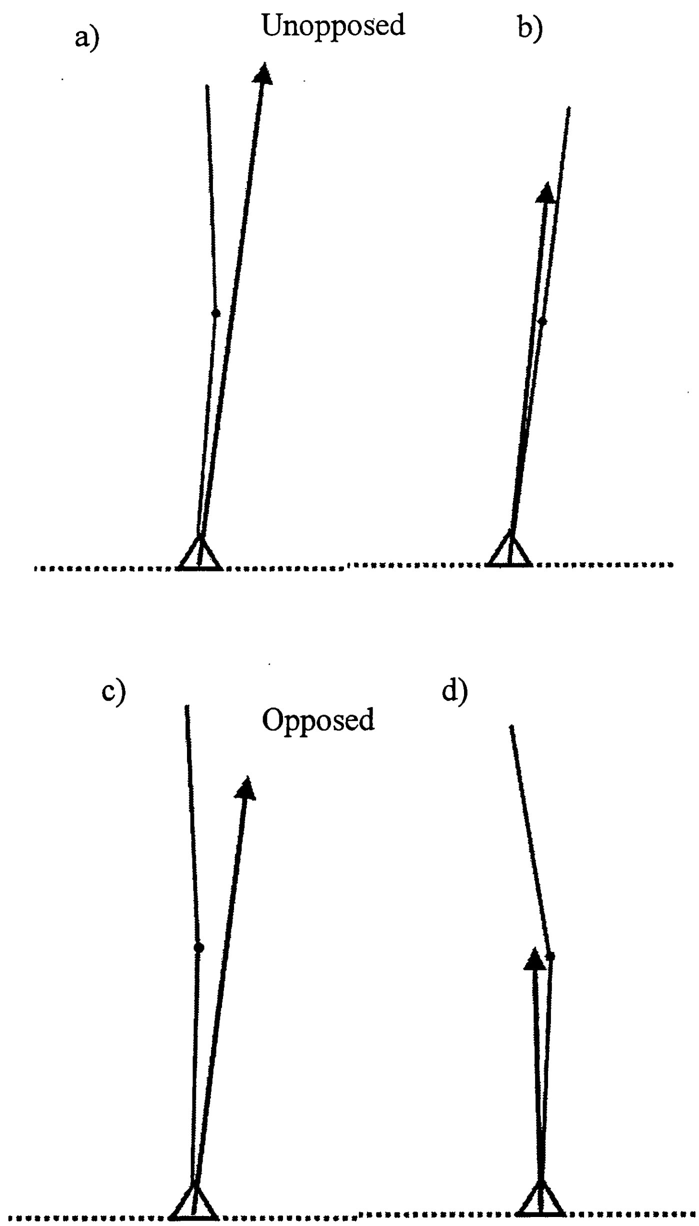


Figure 4.18. Stick figures for females of a) peak varus GRF moment about the knee during unopposed trials, b) peak valgus GRF moment about the knee during unopposed trials, c) peak varus GRF moment about the knee during opposed trials and d) peak valgus GRF moment during opposed trials about the knee.

4.4. Discussion.

4.4.1. Sagittal plane kinematics.

The results indicate differences in sagittal plane kinematics between males and females and between unopposed and opposed trials. Knee flexion at IC was significantly greater during unopposed trials than opposed trials, but there was no significant difference between males and females. Increased knee flexion is likely to reduce ACL strain (Nunley et al., 2003; Li et al., 1999), therefore during unopposed trials subjects may increase knee flexion at IC compared to opposed trials to reduce the likelihood of ACL strain. MAX knee flexion, ROM of knee flexion and angular velocity of knee flexion during PP were significantly greater in females than males and significantly greater during unopposed trials than opposed trials. This may be due to subjects consciously increasing their knee flexion during unopposed trials in an attempt to reduce the impact of the ground reaction forces during landing and therefore reduce the risk of injury. However, during opposed trials, due to the greater attentional demand, subjects were, perhaps, less able to consciously increase the amount of knee flexion during landing. These results indicate that sagittal plane kinematics changed significantly with the introduction of opposition to the landing task and highlight the need for ecologically valid task demands in studies designed to examine differences in the incidence of injuries between males and females in specific sports.

The results of the present study indicate that values of maximum knee flexion measured during unopposed trials were nearer to values reported by previous studies where subjects performed unopposed landing than those measured during opposed conditions. For example, mean maximum knee flexion of $88.9 \pm 11.4^{\circ}$ for males and $78.3 \pm 13.4^{\circ}$ for females were reported by Kernozek et al. (2005) compared to $67.2 \pm 12.9^{\circ}$ for males and $78.0 \pm 8.1^{\circ}$ for females during unopposed trials and $62.1 \pm 11.6^{\circ}$ for males and $68.2 \pm 12.2^{\circ}$ for females during opposed trials

There was no significant difference in knee flexion at IC, MAX or ROM of knee flexion between dominant and non-dominant legs for males or for females during unopposed and opposed trials. This indicates a highly symmetrical landing pattern in the sagittal plane for males and females during both unopposed and opposed conditions which indicates a high level of dynamic balance during landing compared to a less symmetrical pattern.

4.4.2. Frontal plane kinematics.

The results indicate differences in frontal plane kinematics between males and females but not between unopposed and opposed trials. There was no significant difference in knee valgus at IC between males and females or between unopposed and opposed trials. MAX knee valgus and ROM of knee valgus was significantly greater for females than males, but there were no significant differences between MAX knee valgus or ROM of knee valgus during unopposed and opposed trials. Females also displayed significantly greater average knee valgus angular velocity than males during PP but there was no significant difference between unopposed and opposed trials. These results indicate that differences in frontal plane kinematics between males and females during landing were consistent between unopposed and opposed conditions and may indicate decreased dynamic stability of the knee joint in females compared to males.

The values of maximum knee valgus angle reported in this study are different to previous results but as with the sagittal plane kinematics, the results of the present study indicate values of maximum knee valgus angle measured during unopposed trials were nearer to values reported by previous studies where subjects performed unopposed landing than those measured during opposed conditions. For example, Ford et al. (2004) reported maximum knee valgus (–ve) / varus (+ve) angle values of $-14.3 \pm 2.0^{\circ}$ for males and $-20.1 \pm 2.5^{\circ}$ for females, compared to $-2.2 \pm 5.3^{\circ}$ for males and $-13.9 \pm 11.3^{\circ}$ for females during unopposed trials and $-2.9 \pm 7.9^{\circ}$ for males and $-10.4 \pm 7.7^{\circ}$ for females during opposed trials in this study. There are a number of possible reasons for these differences which include subjects' age and playing standard and the method of measuring the knee valgus angle. In Ford et al. (2004) the subjects used were high school athletes whereas university athletes were used in this study. The valgus angle measured in Ford et al. (2004) was determined from markers placed on the skin over the greater trochanter, lateral epicondyle of the knee and the lateral malleolus of the ankle, whereas in this study, the valgus angle was based on estimated hip, knee and ankle joint centres using the Vicon plug-in gait model.

Males showed no significant difference in frontal plane kinematics between dominant and non-dominant legs during both unopposed and opposed trials, but in females, during both unopposed and opposed trials, the non-dominant leg displayed significantly greater maximum knee valgus angle and range of motion compared to the dominant leg. These results may indicate a higher level of dynamic stability in males compared to females during both unopposed and opposed trials and indicate that differences in frontal plane kinematics between males and females during landing were consistent between unopposed and opposed conditions.

4.4.3. Inter-hip, inter knee and inter ankle displacements.

MIN inter-knee distance was significantly smaller for females than males and ROM of inter-knee distance was significantly greater in females than males. ROM of inter-ankle distance was significantly greater in males than females. There was no significant difference in inter-knee distance at IC and there was no significant difference in inter-ankle distance at IC or MIN between males and females. There was no significant difference in inter-knee or inter-ankle distance at IC, MIN or ROM between unopposed and opposed trials. As with frontal plane kinematics, these results indicate that gender differences in inter-knee and inter-ankle distances did not change between unopposed and opposed conditions. The pattern of the inter-knee (Figure 4.7) and inter-ankle distances (Figure 4.8) indicate a large similarity between unopposed and opposed trials in males, but a difference between unopposed and opposed trials in females. For both males and females, inter-knee distance was smaller during opposed trials than during unopposed trials throughout the landing manoeuvre but females displayed a greater change between unopposed and opposed trials than males. Also, for both males and females, inter-ankle distance was smaller during opposed trials than during unopposed trials throughout the landing manoeuvre but again females exhibited a greater change between unopposed and opposed trials than males. During unopposed trials, inter-ankle distance was greater in females than males throughout landing but during opposed trials, inter-ankle distance was smaller in females than males throughout landing. This may indicate that during unopposed trials, females consciously increased inter-ankle distance during landing to increase the base of support and therefore increase dynamic stability relative to males and relative to opposed conditions. However, during opposed trials, due to the greater attentional demand, females were, perhaps, less able to consciously increase the inter-ankle distance during landing and therefore exhibited a decreased base of support and decreased dynamic stability relative to males and relative to unopposed conditions.

4.4.4. Ground reaction force.

Maximum normalised GRF and normalised GRF at AL was significantly greater during opposed trials than unopposed trials but there was no significant difference between males and females. This may be due to the greater MAX knee flexion and ROM of knee flexion during unopposed trials compared to opposed trials. Subjects may consciously increase knee flexion during unopposed trials compared to opposed trials to reduce the impact forces acting on the legs during landing and therefore reduce the likelihood of injury during landing. The normalised GRF was significantly greater in females than males at IC which may indicate a greater risk of injury in females compared to males at or close to IC. ACL injury is reported to occur close to IC (Olsen et al., 2004; Boden et al., 2000). If females exert greater normalised GRF at IC they are at greater risk of a large GRF moment about the knee which suggests females may be more likely to overload the passive support structures of the knee at IC compared to males. During the trials analysed in this study, subjects performed largely balanced landing patterns which resulted in relatively small moment arms about the knee at IC and therefore no injuries occurred. However, during an unbalanced landing, the greater normalised GRF at IC in females compared to males combined with a large moment arm about the knee may result in an overloading of the passive support structures of the knee in females.

4.4.5. Sagittal plane kinetics.

The angle of the GRF in the sagittal plane at IC and maximum angle of the GRF in the sagittal plane were significantly greater in females than males but not significantly different between unopposed and opposed trials. The angle of the GRF in the sagittal plane was significantly greater in males than females at ML but not significantly different between unopposed and opposed trials. The change in the angle of the GRF in the sagittal plane between IC and ML was 3.33° for males during unopposed trials, 4.45° for males during opposed trials, 9.40° for females during unopposed trials and 10.82° for females during opposed trials. These results suggest far greater control of the angle of the GRF in the sagittal plane in males compared to females during both unopposed and opposed conditions. Since ACL injury is most likely during passive loading, this greater change in the angle of the GRF in the sagittal plane between IC and ML in females compared to males might contribute to the greater incidence of ACL injury in females compared to males. The minimum angle of the GRF in the sagittal plane was significantly smaller in opposed trials than unopposed trials but there was no significant difference between males and females. Since subjects were performing a vertical landing, the angle of the GRF should be close to 90° throughout landing. During unopposed landings, the minimum angle of the GRF in the sagittal plane was nearer to 90° than during opposed trials suggesting a more controlled landing during unopposed trials than opposed trials.

Whilst the sagittal plane kinematics of the knee indicate significant differences in MAX knee flexion and ROM of knee flexion between males and females and between unopposed and opposed conditions, only significant differences at ML were observed in the sagittal plane normalised GRF moment arm and moment about the knee between males and females and between unopposed and opposed conditions. These significant differences in sagittal plane normalised GRF moment arm and moment about the knee occurred at a critical stage during landing, i.e., during passive loading. Females displayed a significantly smaller normalised GRF flexor moment arm at the ML than males but there was no significant difference between unopposed and opposed trials. The significantly smaller normalised GRF moment arm at ML in females resulted in a significantly smaller normalised GRF flexor moment in females than males at ML and normalised GRF flexor moment was significantly smaller in opposed trials than unopposed at ML. ACL injury is most likely during passive loading as the muscles are in control of the landing during active loading but not during passive loading. Through practice, females may have learned to reduce the GRF moment about the knee in the sagittal plane at ML to reduce the likely strain on the passive support structures of the knee, particularly during the opposed trials. Since the subjects have greater conscious control during unopposed trials than opposed, females are more able to control a larger GRF moment in the sagittal plane in unopposed conditions even at ML. Subjects unfamiliar with performing volleyball block jump landings may have displayed a different pattern to the normalised GRF moment about the knee which may alter the likelihood of ACL strain compared to trained subjects. However, this was not investigated in the present study.

The peak normalised GRF flexion moment (–ve) and the peak normalised GRF extension moment (+ve) observed at any instant throughout all unopposed and all opposed trials for males and throughout all unopposed and all opposed trials for females indicate that the peak normalised GRF extension moment occurred during passive loading for males and females during both unopposed and opposed trials whereas the peak normalised GRF flexion moment occurred during active loading for both males and females during both unopposed and opposed conditions. ACL injury is more likely during passive loading therefore the peak normalised GRF extension moment is more likely to cause injury than the peak normalised GRF flexor moment. Also, a GRF extensor moment is likely to be associated with a knee flexor *MUS* produced largely by hamstring muscle activity. Since the hamstrings work with the ACL to prevent anterior displacement of the tibia relative to the femur, the greater the load on the hamstring muscles, the greater the extent to which stability of the knee joint (anterior displacement of the tibia relative to the femur) is likely to be maintained by the ACL. Therefore, the greater the peak normalised GRF extensor moment about the knee the greater the likely ACL strain. The peak normalised GRF extension moment was greater during opposed trials than during unopposed trials for males and females which indicates ACL strain is most likely during opposed conditions. The peak normalised GRF extension moment was greater in males than females during unopposed trials, but was greater in females than males during opposed trials. This may indicate that due to the greater attentional demand during opposed trials, females are less able to control the normalised GRF extensor moment than males and are therefore at greater risk of injury. Also, the knee was less flexed for females at peak normalised GRF extensor moment than for the males during both unopposed and opposed trials. ACL strain is likely to be increase with reduced knee flexion therefore peak extensor normalised GRF moment is more likely to cause ACL strain in females than males during both unopposed and opposed trials.

4.4.6. Frontal plane kinetics.

The GRF angle in the frontal plane was significantly greater (closer to vertical) at IC in opposed trials than in unopposed trials but there was no significant difference between males than females. This may be due to subjects landing with feet further apart during unopposed trials and that therefore they were pushing out with the feet at IC more during unopposed trials than opposed. The maximum GRF angle in the frontal plane was significantly greater (closer to vertical) in males than females but there was no significant difference between unopposed and opposed trials. Due to subjects performing a vertical landing manoeuvre, it is expected that the angle of the GRF should be close to vertical throughout landing. Since the maximum angle of the GRF in the frontal plane is closer to vertical in males than females, it suggests that males reduce the body's downward linear momentum more efficiently than females and there is less lateral movement in the frontal plane in males compared to females during both unopposed and opposed conditions.

The normalised GRF moment arm and the normalised GRF moment remained in varus throughout the landing phase in opposed trials for females and was varus between IC and approximately 50% normalised time for females during unopposed trials. However, for males during both unopposed and opposed trials, the normalised GRF moment arm and normalised GRF moment in the frontal plane were fairly small throughout landing and changed between valgus and varus. The varus normalised GRF moment arm and moment at ML were significantly greater in females than males but there was no significant difference between unopposed and opposed trials. At AL, the valgus normalised GRF moment arm and the valgus normalised GRF moment in males were significantly different from the varus normalised GRF moment arm and the varus normalised GRF moment in females and frontal plane normalised GRF moment arm and moment were significantly different between unopposed and opposed trials at AL. The frontal plane kinematics show clear differences between males and females which are consistent between unopposed and opposed conditions. Females display significantly greater MAX knee valgus angle during landing compared to males. This greater knee valgus in females is likely to increase the risk of ligament strain in females compared to males. However, the corresponding frontal plane normalised GRF moment indicates a compensatory GRF moment pattern which acts as to prevent further valgus movement of the knee joint, i.e., a varus GRF moment. This suggests that females manage their knee valgus angle by exerting a varus GRF moment. Assuming the quasi-static approach, the results indicate a valgus *MUS* throughout landing which suggests that it is the muscles which cause the increased valgus angle in females compared to males rather than the external moment due to the GRF. However, during this study the subjects examined were uninjured subjects performing landing manoeuvres in which injury did not occur. The results appear to suggest that the kinematics pattern exhibited by females puts them at greater risk of ACL injury than males but in most cases they avoid injury by exerting a varus GRF moment about the knee which prevents injury. However, in some situations, possibly during an unbalanced landing manoeuvre, females either fail to or are unable to exert a sufficient varus GRF moment about the knee and an injury occurs.

The maximum normalised varus GRF moment arm and the maximum normalised varus GRF moment about the knee were significantly greater in females than males but there was no significant difference between unopposed and opposed trials.

Assuming the quasi-static model, this indicates that females displayed a significantly greater valgus normalised *MUS* during landing compared to males, which is reported by a number of previous studies (Kernozek et al., 2005; Chappell et al., 2002). As with the frontal plane kinematics, the normalised GRF moment about the knee in the frontal plane shows significant differences between males and females which are largely consistent between unopposed and opposed conditions.

The peak normalised GRF varus moment (+ve) and the peak normalised GRF valgus moment (–ve) observed at any instant throughout all unopposed and all opposed trials for males and throughout all unopposed and all opposed trials for females indicated that the peak normalised GRF valgus and varus moments occurred earlier during the trial for the female subjects than the male subjects during both unopposed and opposed trials. Since muscular control is less likely to have been gained early during landing, the peak normalised valgus GRF moment and peak normalised varus GRF moment are more likely to cause strain to the passive support structures of the knee in females compared to males during both unopposed and opposed trials. The peak normalised GRF varus moment was similar in males and females during unopposed trials but the peak normalised GRF varus moment was greater in females than males during opposed trials. This may be due to females being less able to control the peak varus normalised GRF moment during opposed trials compared to males and compared to unopposed conditions. The peak normalised GRF valgus moment was greater in males than females during both unopposed and opposed trials and the peak normalised GRF valgus moment was greater during opposed trials than unopposed trials for both males and females. This suggests that males and females were less able to control the normalised valgus GRF moment during opposed trials compared to unopposed trials.

4.5. Summary and conclusions.

Differences in knee kinematics and kinetics during opposed and unopposed trials suggest that coaches should implement training programs that involve ecologically valid landing manoeuvres. Future research into landing kinematics and kinetics should include opposition during the landing task as the effect of opposition may significantly alter subjects' neuromuscular responses during landing, particularly in the sagittal plane. Differences in frontal plane kinematics and kinetics between males and females however, appear to be consistent in unopposed and opposed conditions. Therefore the results of this study, in which comparison between unopposed and opposed trials was examined, may validate the results of many other studies (Kernozek et al., 2005; Ford et al., 2003; Malinzak et al., 2001) which have investigated gender differences in frontal plane kinematics and kinetics of landing during unopposed conditions.

Chapter 5

Experimental Study 3

Lower limb coordination and stiffness during landing from volleyball block jumps.

5.1 Introduction.

Whilst the previous two previous studies have examined changes in knee kinematics and kinetics with respect to time, no data have been provided regarding the movement of the hip and the ankle. Since knee joint kinematics are influenced by hip and ankle joint kinematics, it would be appropriate to examine the effect of hip and ankle joint movement on the movement of the knee. Whilst the angle–time histories provide a measure of movement patterns of individual joints, the relative movement of one joint with respect to another, i.e. the coordination, may provide insight into the relationship between the lower limb joints.

The results of Chapters 3 and 4 indicate that females exhibit reduced dynamic stability of the knee joint compared to males. Since a number of the muscles of the lower limb cross more than one joint (quadriceps: hip and knee, hamstrings: hip and knee, gastrocnemius: knee and ankle), an indication of the dynamic stability of the leg as a whole may provide more information regarding possible gender differences in dynamic stability. One factor which may provide a more complete analysis of the dynamic stability of the leg is stiffness.

5.1.1 Coordination.

Coordination refers to the relative timing of motion between body segments (Jensen et al., 1994). In any particular whole body movement, each body segment can be thought of as an independent oscillator that oscillates through a particular range of motion during the movement. The coordination between any two oscillators is the degree of coupling (relative timing or relative phasing) between these two oscillators. The methods of quantifying coordination can be classified into two approaches (Hamill et al., 2000). These are referred to as 'discrete' and 'continuous' methods.

5.1.1.1 Discrete methods.

Discrete methods of analysing joint coordination are used to analyse the relative timing between key events in a movement cycle, such as the maximum angle of two different joints. The advantages of these methods are that they are fairly simple, no manipulations to these data are needed, such as normalisation, and discrete methods are more suitable for discrete movements (e.g. throwing, landing) than continuous methods. However, discrete methods only evaluate coordination at one point in a movement. There are two types of discrete analysis of coordination: time series approach and return maps.

5.1.1.1.1 Time series.

Discrete relative phase (DRP) angle can be determined for a discrete event during a movement using the time series of two joints or body segments to provide a temporal analysis of coordination. For example, the DRP between two joints may be assessed by firstly determining an initial start point (t_s), such as initial ground contact. The timing of the maximum angles of the two joints being analysed is then identified (t_1 and t_2). Finally, a finish time is determined (t_f), such as the end of the stance phase or the point at which the body CG velocity is reduced to zero. The DRP angle is then calculated using the following formula (Hamill et al., 2000):

$$\mathrm{DRP} = \frac{t_1 - t_2}{t_f - t_s} \times 360^{\circ} \qquad (5.1)$$

The DRP angle can range between 0° and 360°, where 0° indicates the timing of the two key events are perfectly in phase and an angle between 1° and 360° indicates the degree to which the timing of the two key events are out of phase. The average DRP angle between the two events over a number of trials is reflected in the mean DRP angle. The standard deviation of the DRP angle over a number of trials indicates the stability (variability) of the coupling; the lower the standard deviation, the greater the stability of the coupling.

5.1.1.1.2 Return maps.

Return maps are plots of one oscillator against another with respect to the cycle of one of the oscillators. This allows the identification of the frequency ratio between the oscillators. The method is useful for examining the coordination between oscillators with different frequencies, such as the frequency of heel strike and breathing in running, where more than one breath is taken per stride.

5.1.1.2 Continuous methods.

In contrast to discrete methods, continuous methods may provide a spatial and temporal analysis of coordination over a period of time, typically a cycle. The two main continuous methods are 'relative motion' and 'continuous relative phase'.

5.1.1.2.1 Relative motion.

Relative motion refers to the motion of one joint or segment relative to the motion of another joint or segment. The simplest qualitative illustration of relative motion is an angle – angle plot, where the angle of one joint is plotted against the angle of another (Schmidt and Lee, 1999; Anderson and Sidaway, 1994).

Quantification of relative motion is based on vector coding techniques, originally devised by Freeman (1961). The orientation of the vector between two adjacent points on an angle – angle plot relative to the right horizontal is calculated, called the coupling angle (γ) (Hamill et al., 2000) (Figure 5.1).

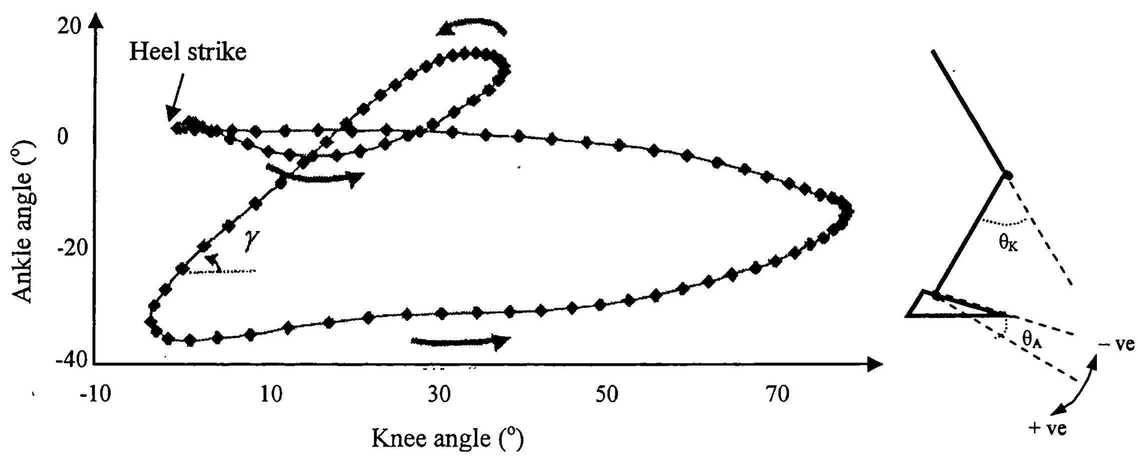


Figure 5.1. Angle – angle diagram of knee (θ_K) – ankle (θ_A) joint coupling for one leg during one complete stride cycle of running and illustration of coupling angle γ (Hamill et al., 2000).

The coupling angle will range between 0° and 360°. Angles of 90°, 180° and 270° indicate movement of one joint only and angles of 45°, 135°, 225° and 315°

indicate equal movement of both joints on the angle – angle plot. The advantage of relative motion analysis is that no normalisation procedures are needed and it provides an indication of coordination over a complete cycle of movement. However, whilst this method provides information on the spatial coordination between two joints or segments, no indication of temporal coordination is provided.

5.1.1.2.2 Continuous relative phase.

Continuous relative phase (CRP) is calculated from the relative phase angle of a pair of joints throughout a movement cycle (Haddad et al., 2006; Kurz and Stergiou, 2002; Hamill et al., 1999; Van Emmerick and Wagenaar, 1996). The phase angle (Φ) is derived from a normalised phase – plane plot (joint angular velocity plotted as a function of joint angular displacement) of a joint for each cycle. The angle and angular velocity of a joint are normalised between –1 and +1, where –1 is the minimum value and +1 is the maximum value in either each cycle or for all cycles using the following equations (Hamill et al., 1999).

Horizontal axis (angle): $$\theta_i = \frac{2 \times [\theta_i - \min(\theta_1)]}{\max(\theta_i) - \min(\theta_i)} \qquad (5.2)$$

where θ = segment or joint angle, i = data point within cycle.

Vertical axis (angular velocity): $$\omega_i = \frac{\omega_i}{\max\{\max(\omega_i), \max(-\omega_i)\}} \qquad (5.3)$$

where ω = segment or joint angular velocity, i = data point within cycle.

The phase angle (Φ) is given by the angle between a line from the origin of the graph to the current data point (θ, ω) and the right horizontal (Figure 5.2).

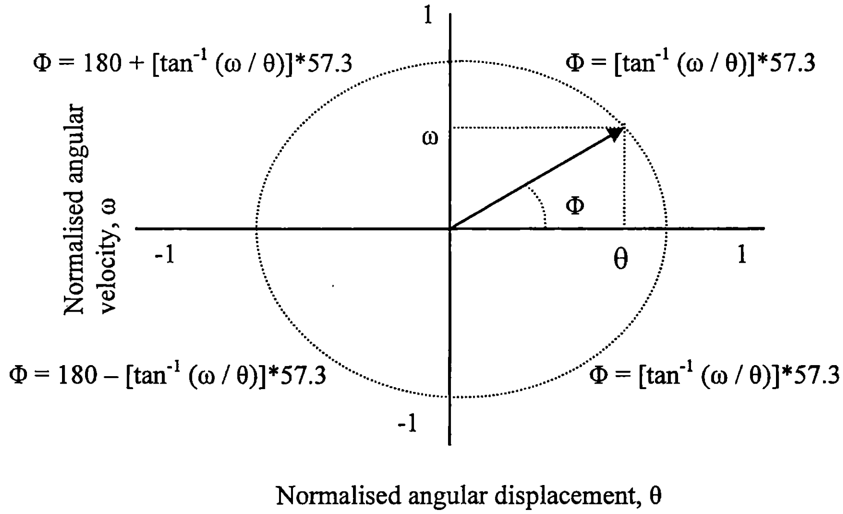


Figure 5.2. Calculation of the relative phase angle Φ (Hamill et al., 1999).

The CRP angle is then calculated as the difference between the phase angles of two joints at corresponding time intervals throughout a stride cycle. The mean CRP angle over a number of cycles provides an indication of the type of relationship between joints (in-phase, CRP = 0^{o}; anti-phase, CRP = 180^{o}). The standard deviation of the CRP angle over a number of trials provides an indication of the variability in coordination between cycles. Ensemble CRP and CRP variability history can be obtained and plotted against normalised time to examine the changes in coordination throughout a cycle (Figure 5.3). The advantage of CRP is that it evaluates both the spatial and temporal coordination between two joints or segments during an entire cycle. However, CRP requires complex calculations and normalisation procedures and is only appropriate for cyclic movements such as walking, running and hopping and when angle – time histories of joint motions are sinusoidal (Diedrich and Warren, 1995).

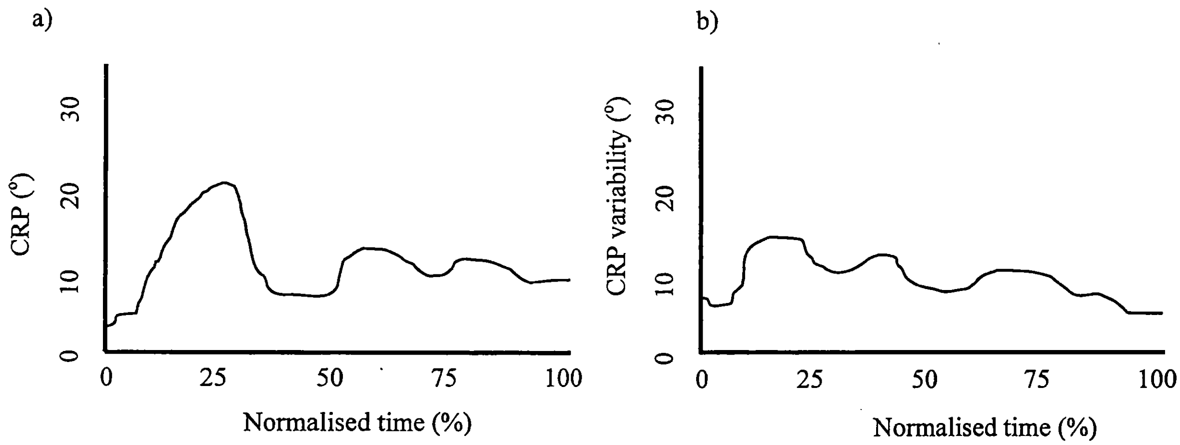


Figure 5.3. Example ensemble CRP (a) and CRP variability (b) – time graphs over a complete stride cycle of running (Hamill et al., 2000).

Table 5.1 shows a summary of studies that have examined coordination using a variety of methods. The methods used are largely determined by the type of movement being analysed. Since a landing manoeuvre is a discrete action, CRP analysis is not a suitable method of analysing coordination. In this case, angle – angle plots are likely to provide the most appropriate illustration (qualitative) and time series – based DRP is likely to provide the most appropriate measure (quantitative) of coordination between the hip, knee and ankle joints.

5.1.2 Stiffness.

Whilst analysis of the relative phase between pairs of joints provides an indication of the coordination between the joints, it is the coordination between all of the joints in the kinetic chain that determine the effectiveness and efficiency of the movement as a whole. In a landing manoeuvre, it is the coordination between the

Table 5.1. Summary of previous studies examining coordination during continuous and discrete tasks.

Study	Subjects	Task	Method	Results
Hamill et al. (1999)	Subjects with PFP and healthy subjects (no indication of numbers)	Running	CRP	PFP sufferers exhibited lower variability in CRP of the lower limb joint couplings than healthy subjects.
Van Emmerick and Wagenaar (1996)	7 M healthy	Walking	CRP and DRP	CRP and DRP between thorax an pelvis significantly increased as walking speed increased.
Heiderscheit et al. (2002)	8 F with PFP and 8 F healthy subjects.	Running	Coupling angle	Reduced joint coordination variability for thigh rotation-leg rotation coupling of the PFP group compared to healthy group.
Van Uden et al. (2003)	5 M and 2 F ACL reconstructed and 2 F and 11 M healthy subjects.	One-legged hopping	CRP	Mean CRP and variability of CRP between knee rotation-ankle rotation was significantly lower for the ACL reconstructed limb than for the healthy subjects.
Whiting and Zernicke (1982)	4 M runners	Slow walk, fast walk and running	Vector coding	Cross-correlation values of movement patterns changed during different locomotion speeds with most similar values obtained between similar locomotion speeds (i.e. slow walk – fast walk, fast walk – run).
Anderson and Sidaway (1994)	5 M and 1 F novices and 3 M soccer players	Kicking	Angle – angle diagrams	Coordination pattern of hip flexion-knee flexion of novices altered after practise to become more similar to the coordination pattern of experts.
Tempardo et al. (1997)	14 M novice and 6 M expert volleyball	Volleyball serve	Cross-correlation	Significant difference in the cross-correlation of the shoulder-wrist anterior-posterior coordinates between experts and novices but not for the elbow-shoulder and wrist-shoulder coordinate couplings.

M = males, F = females. PFP = patellofemoral pain.

hip, knee and ankle joints that largely determine the dynamic stability of the leg. The dynamic stability of the leg is reflected in the stiffness of the leg, i.e. the resistance of the leg to compression (flexion of hip, knee and ankle joints) during landing. During activities such as hopping, running and landing the actions of the body's musculoskeletal components, in particular, the muscle – tendon units, behave like springs, storing elastic energy when stretched and returning it when they recoil (Alexander, 1988). The stiffness of a spring is given by dividing the change in the force exerted on the spring by the resulting change in length of the spring (Houk, 1974). Leg stiffness refers to the stiffness of the leg when modelled as a single linear spring and joint stiffness refers to the torsional stiffness of an individual joint when modelled as a spring.

5.1.2.1 Leg stiffness.

During landing, the musculoskeletal system behaves much like a single linear spring (leg spring). The body can therefore be modelled as a simple spring – mass system, consisting of a single linear spring and a point mass equal to body weight, W (Figure 5.4) (Farley and Morgenroth, 1999; Farley and Gonzalez, 1996; Blickhan, 1989).

Previous studies have calculated leg stiffness during hopping (Granata et al., 2002b; Farley and Morgenroth, 1999) and running (Arampatzis et al., 1999; Farley and Gonzalez, 1996). In these activities maximum vertical ground reaction force (GRF) has been shown to occur at the same time as maximum vertical displacement of the whole body CG, i.e. at the same instant when downward velocity is reduced to zero. Leg stiffness in these studies was calculated as the ratio of the change in vertical GRF to the vertical displacement (change in vertical height) of the body's CG during the period from foot-strike to the instant when downward velocity was reduced to zero (Farley and Morgenroth, 1999):

$$k_{leg} = \frac{\Delta F}{\Delta L} \quad \text{.......} \quad (5.4)$$

where k_{leg} = leg stiffness, ΔF = change in vertical GRF and ΔL = vertical displacement of the CG.

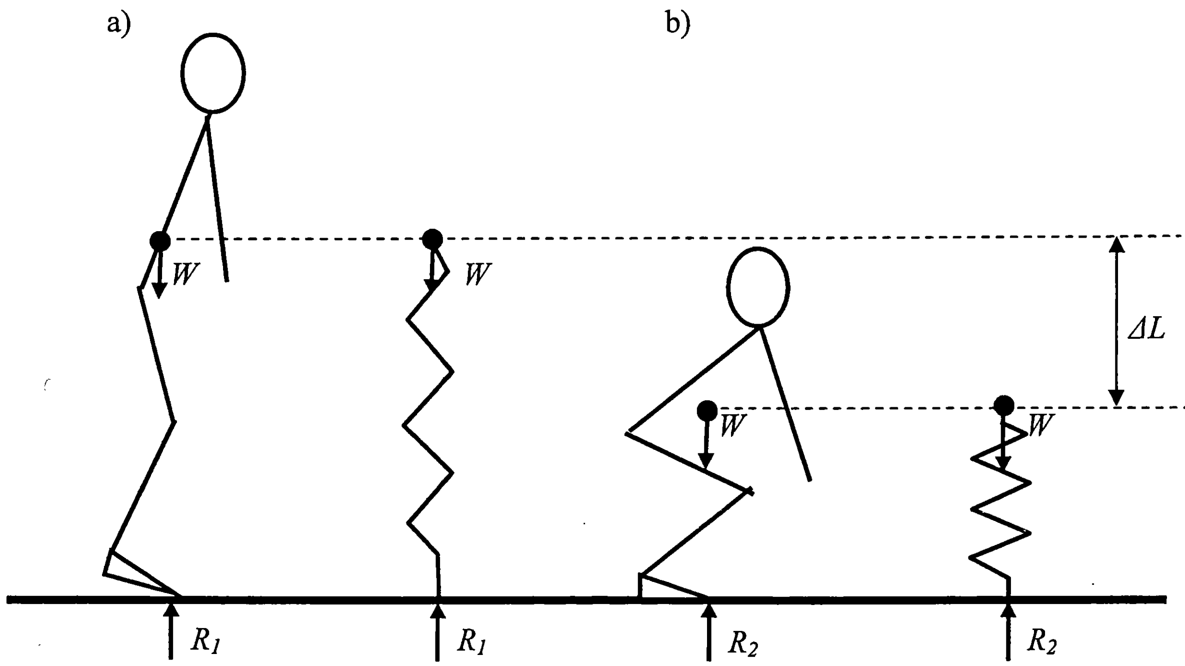


Figure 5.4. Spring – mass model for vertical landing at a) instant of ground contact and b) instant when the downward velocity of the body CG is zero (V_o), where R = vertical ground reaction force and ΔL = vertical displacement of the CG.

5.1.2.2 Joint stiffness.

Leg stiffness is dependent on the combined stiffness of the lower limb joints (Farley and Ferris, 1998; Greene and McMahon, 1979). Joint stiffness is given by the ratio of the change in joint moment (equivalent to the muscle moment in the quasi static method of determining joint moment) to the joint angular displacement (change in joint angle) during the same period of time (Farley and Morgenroth, 1999):

$$k_{joint} = \frac{\Delta M}{\Delta \theta} \quad \text{.......} \quad (5.5)$$

where k_{joint} = joint stiffness, ΔM = change in joint muscle moment and $\Delta\theta$ = joint angular displacement.

Table 5.2 provides a summary of a number of studies that have examined leg and joint stiffness in a number of different activities.

Whilst studies have examined gender differences in leg stiffness during two-legged hopping (Granata et al., 2002b), to the author's knowledge no study has investigated gender differences in leg stiffness during tasks in which non-contact ACL injury is common, such as landing. Furthermore, previous studies (Granata et al., 2002b; Farley and Morgenroth, 1999) only report absolute leg stiffness without normalising for body weight and height.

Joint stiffness can vary depending on many factors, including muscle activation (Neilsen et al., 1994; Weiss et al., 1988), joint angle (Gottlieb and Agarwal, 1978), range of motion and angular velocity (Kearney and Hunter, 1982). Since the angle, range of motion and angular velocity of the lower limb joints will also influence the vertical displacement of the whole body CG and therefore influence the leg stiffness (i.e. for a given change in vertical GRF, smaller angular displacement of the joints during landing will result in less vertical displacement of the body's CG and a higher value of leg stiffness), the reduced leg stiffness in females compared to males reported by Granata et al. (2002b) may be due, at least in part, to reduced stiffness of one or more of the hip, knee and ankle joints in females compared to males. These differences in leg stiffness and possibly joint stiffness may indicate less dynamic stability of the legs in females compared to males, which may in

Table 5.2. Summary of previous studies examining leg stiffness and joint stiffness.

			Leg stiffness		Joint stiffness	
Study	Subjects	Task	Method	Results	Method	Results
Farley and Morgenroth (1999)	2 M and 3 F	Maximum height and preferred height two-legged hopping	$k_{leg} = \frac{\Delta F}{\Delta L}$	Leg stiffness was significantly greater for maximum height hopping (29.3 ± 2.9 kN/m) compared to preferred height hopping (14.5 ± 0.7 kN/m).	$k_{joint} = \frac{\Delta M_{joint}}{\Delta \theta_{joint}}$	Ankle joint stiffness and knee joint stiffness significantly greater when hopping for maximum height compared to preferred height (ankle- 7.0 ± 0.4 to 13.4 ± 1.8 N.m/°; knee- 6.4 ± 1.4 to 11.0 ± 1.5 N.m/°).
Arampatzis et al. (1999)	13 (no indication of gender)	Running at velocities between 2.5 and 6.5 m/s.	$k_{leg} = \frac{\Delta F}{\Delta L}$	Leg stiffness significantly increased from 25.3 ± 4.2 to 35.2 ± 4.3 kN/m as running speed increased.	$k_{joint} = \frac{2W_{joint}}{\Delta \theta_{joint^2}}$	Ankle joint stiffness remained constant whereas knee joint stiffness significantly increased from 7 ± 2 to 17 ± 3 N.m/° as running speed increased.
Granata et al. (2002b)	15 M and 15 F	Two-legged hopping at 2.5Hz, 3Hz and preferred frequency	1) Regression slope of *F* against *L* graph. 2) Natural frequency of equivalent mass-spring system.	Both methods produced near identical results. Leg stiffness was significantly greater in M (33.9 ± 8.7 kN/m) than F (26.3 ± 6.5 kN/m) at all hopping frequencies.		
Farley and Gonzalez (1996)	4 M	Running at -26% to +36% preferred stride frequency.	$k_{leg} = \frac{\Delta F}{\Delta L}$	Leg stiffness significantly increased from 7.0 ± 0.2 to 16.3 ± 0.4 kN/m as stride frequency increased.		
Kuitanen et al. (2002)	10 M	Sprint running at 70 to 100% maximum speed.	$k_{leg} = \frac{\Delta F}{\Delta L}$	Leg stiffness significantly increased from 103 ± 25 to 171 ± 40 kN/m as running speed increased.	$k_{joint} = \frac{\Delta M_{joint}}{\Delta \theta_{joint}}$	Ankle joint stiffness remained constant whereas knee joint stiffness significantly increased from 17 ± 6 to 24 ± 13 N.m/° as running speed increased.

M = males, F = females

some way account for the greater incidence of ACL injury in females compared to males. However, to the author's knowledge, no study has examined gender differences in knee joint stiffness when performing tasks such as hopping, running or landing.

5.1.3 Aim.

The aim of the study was to investigate lower limb coordination and stiffness in male and female university volleyball players performing block jump landings.

The objectives were to determine, during landing from a volleyball block jump:

i) Angle – angle plots of the hip – knee, knee – ankle and hip – ankle joint couplings in the sagittal plane,

ii) The DRP angle of the hip – knee, knee – ankle and hip – ankle joint couplings in the sagittal plane,

iii) The absolute and normalised leg stiffness between IC and maximum vertical GRF as calculated from the vertical GRF and vertical CG displacement data,

iv) The absolute and normalised knee joint stiffness between IC and maximum knee joint moment as calculated from the knee joint moment and knee flexion angular displacement data.

5.2 Method.

Five female (mean age 21.8 ± 0.5 years, mass 58.9 ± 8.0 kg and height 1.66 ± 0.08 m) and five male (mean age 21.0 ± 3.5 years, mass 69.9 ± 3.5 kg and height 1.76 ±

0.10 m) university volleyball players with no previous history of hip, knee or ankle injury participated in the study. Ethical approval was granted for the study by the University Ethics Committee (Ethical approval form shown in Appendix A, B and C) and written consent forms were signed by all subjects prior to data collection. Subjects were required to perform opposed volleyball block jump and landings, where subjects were instructed to jump up and block a volleyball as it was spiked against them from a suspended position of 10 cm above a net of height 2.43 m for male subjects and 2.24 m for female subjects. Each subject performed three trials. A 12 camera Vicon 512 system (Vicon, Oxford, England) sampling at 120 Hz was used to determine 3D coordinates of 39 retro-reflective markers. From the markers and anthropometric data (height, weight, leg length, knee width, ankle width, elbow width, wrist width and hand thickness) of each subject, the Vicon system calculated the 3D coordinates of the location of the whole body CG and the hip, knee and ankle angles in the sagittal plane. Ground reaction force (GRF) was measured using two AMTI force plates sampling at 600 Hz. Detailed explanation of the methods used in this study are provided in the method section of Chapter 3. In addition to these methods, hip and ankle angles in the sagittal plane were determined as described in section 5.2.1, the whole body CG was located as described in section 5.2.2 and leg and knee stiffness was calculated as described in section 5.2.3.

5.2.1 Angular definitions.

In the Plug-in gait system, the measurement of hip flexion/extension is based on the pelvic transverse axis (line connecting ASIS and PSIS markers) and the thigh axis (line connecting the hip joint and knee joint centres) projected onto the plane of hip flexion/extension (as determined by the plug-in gait marker system). The hip flexion/extension angle is the angle between the thigh axis and the line perpendicular to the pelvic transverse axis which passes through the hip joint centre. A positive angle corresponds to hip flexion (knee anterior to the trunk) relative to the fully extended position (Figure 5.5).

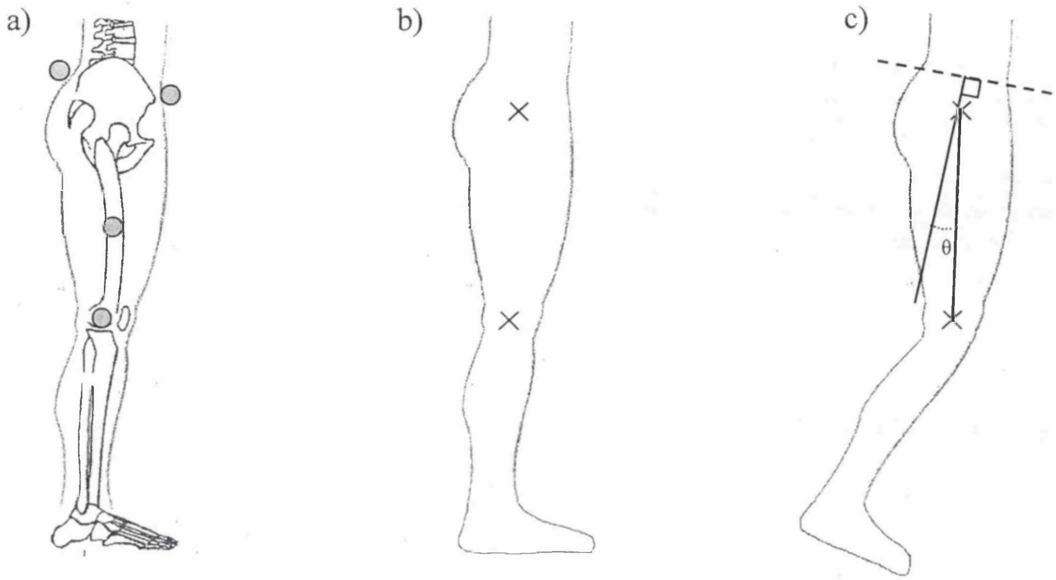


Figure 5.5. Hip flexion/extension: see text for definition. a) Markers placed on skin over bone landmarks. b) Derived estimated joint centres. c) Hip flexion/extension angle θ.

The measurement of knee flexion/extension is based on the thigh axis (line connecting the hip joint and knee joint centres) and the shank axis (line connecting the knee and ankle joint centres) projected onto the plane of knee flexion/extension (as determined by the plug-in gait marker system). The knee flexion/extension angle is the angle between the distal extension of the thigh axis and the shank axis.

A positive angle corresponds to knee flexion relative to the fully extended position (Figure 5.6).

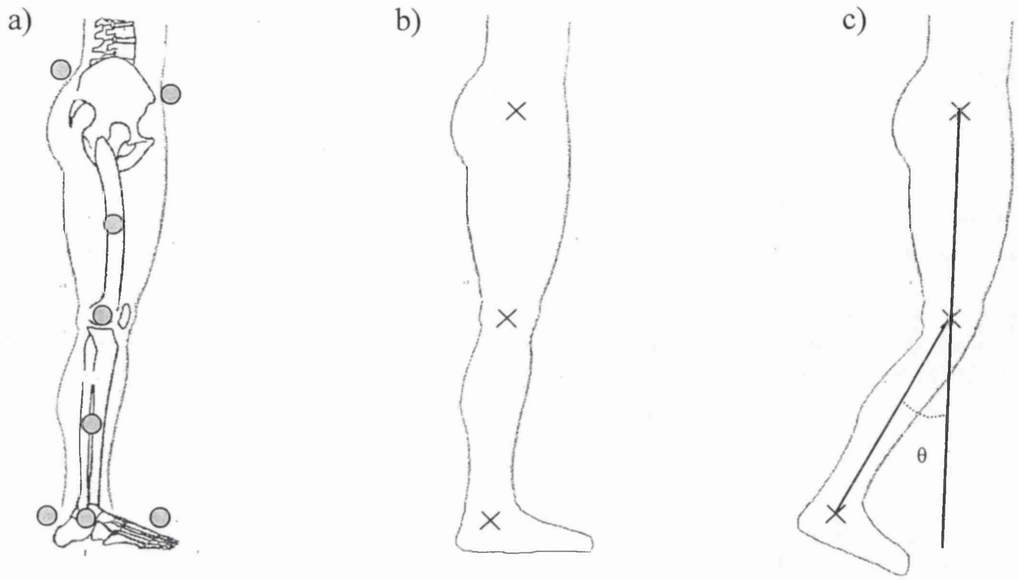


Figure 5.6. Knee flexion/extension: see text for definition. a) Markers placed on skin over bone landmarks. b) Derived estimated joint centres. c) Knee flexion/extension angle θ.

The measurement of ankle plantar/dorsiflexion is based on the foot axis (line connecting the ankle joint centre and the toe marker) and the shank axis (line connecting the knee and ankle joint centres) projected onto the plane of ankle dorsi/plantar flexion (as determined by the plug-in gait marker system). The ankle dorsi/plantar flexion angle is the angle between foot axis and a line perpendicular to the shank axis. A positive angle indicates dorsiflexion and a negative angle indicates plantarflexion (Figure 5.7).

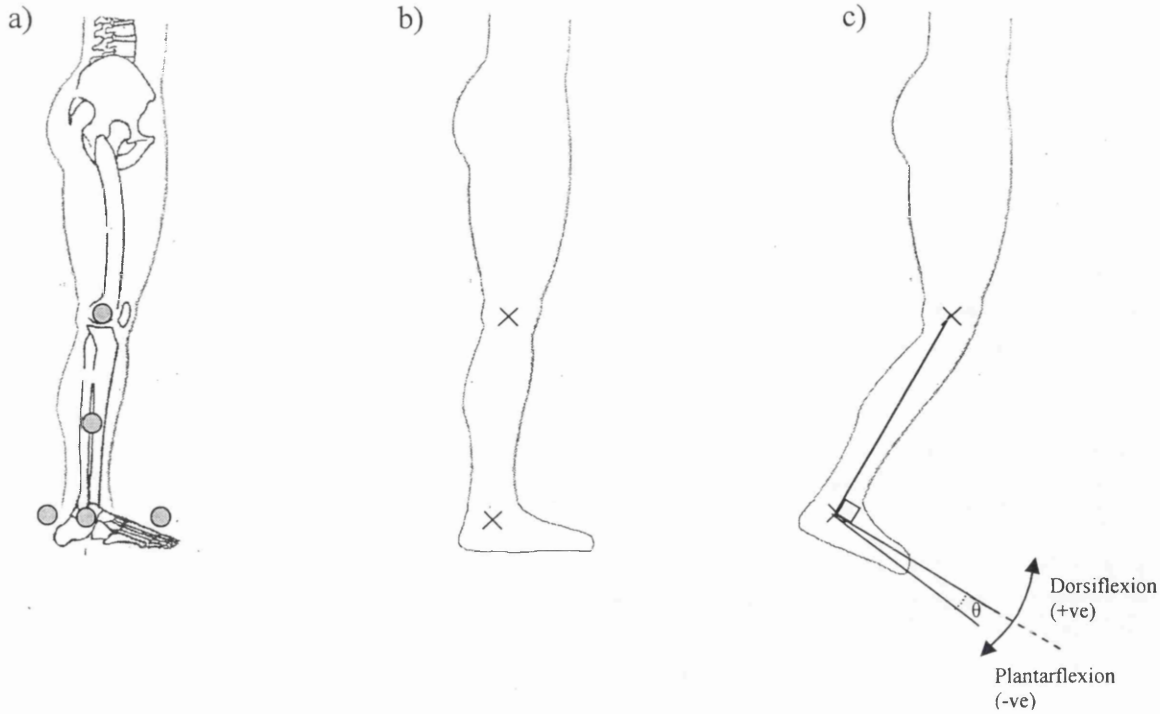


Figure 5.7. Ankle plantar/dorsiflexion: see text for definition. a) Markers placed on skin over bone landmarks. b) Derived estimated joint centres. c) Ankle plantar/dorsiflexion angle θ.

From the angle – time histories of the hip, knee and ankle, DRP analysis was carried out to quantify coordination and angle – angle plots where used to illustrate coordination between hip – knee, knee – ankle and hip – ankle joint couplings. DRP angle was calculated for hip – knee, knee – ankle and hip – ankle joint couplings for right legs, left legs and combined legs for males and females. Also, the DRP angle was calculated between corresponding joints in the right and left leg, i.e., left knee coupled with right knee. The DRP angle was calculated using the following equation:

$$\text{DRP angle } \Phi = \frac{t_2 - t_1}{t_f - t_s} \times 360^{o} \qquad (5.6)$$

where t_2 = time of maximum angle of proximal/left joint, t_1 = time of maximum angle of distal/right joint, t_f = time of zero velocity of the whole body CG and t_s = time of initial ground contact (IC).

For hip – knee, knee – ankle and hip – ankle joint couplings, a positive DRP angle indicates that the proximal joint takes longer to reach its maximum angle than the distal joint and a negative DRP angle indicates that the distal joint takes longer to reach its maximum angle than the proximal joint. For left – right joint couplings, a positive DRP angle indicates that the joint in the left leg takes longer to reach its maximum angle than the corresponding joint in the right leg (e.g. left knee reaches its maximum angle after the right knee) and a negative DRP angle indicates that the joint in the right leg takes longer to reach its maximum angle than the corresponding joint in the left leg (e.g. right knee reaches its maximum angle after the left knee).

5.2.2 Whole body centre of gravity location.

The location of the whole body CG was determined by the Vicon system based on a 15 segment model defined by the plug-in gait marker set (right and left hand, right and left forearm, right and left upper arm, right and left foot, right and left shank, right and left thigh, head and neck, thorax, pelvis). The Vicon system incorporates anthropometric data of Dempster (1955) as reported by Winter (1990).

5.2.3 Leg and knee stiffness calculations

Leg stiffness was calculated as the ratio of the change in vertical GRF to the vertical displacement of the whole body CG between IC and the maximum vertical GRF:

$$k_{leg} = \frac{\Delta F}{\Delta L} \quad \text{.......................................} \quad (5.7)$$

where k_{leg} = leg stiffness, ΔF = change in vertical GRF and ΔL = vertical displacement of the CG.

Knee joint stiffness was calculated as the ratio of the change in knee joint moment to the knee flexion angular displacement between IC and the maximum knee joint moment:

$$k_{joint} = \frac{\Delta M}{\Delta \theta} \quad \text{.......................................} \quad (5.8)$$

where k_{joint} = knee joint stiffness, ΔM = change in knee joint moment and $\Delta\theta$ = knee flexion angular displacement.

5.3 Results.

5.3.1 Angle – angle plots.

The angular displacement – time data were standardised with respect to average trial time (between IC and maximum knee flexion). Angular displacement mean data were based on 30 trials for males and 30 trials for females (5 subjects × 3 trials × 2 legs). Independent-samples t-tests were carried out on the hip, knee and ankle angular displacement data to examine gender differences. Normality of the data and between-group homogeneity of variance were assessed prior to statistical analysis, as described in Chapter 3, section 3.2.9. All parameters were found to display normal distribution. All parameters displayed equal variances between groups other than the maximum angle of the hip.

There was no significant difference in hip flexion at IC between males and females (males: 13.9 ± 5.8°; females: 13.7 ± 5.9°) (Table 5.3). Maximum hip flexion was significantly greater in females than males (males: 29.2 ± 7.9°; females: 39.1 ± 11.9°). Consequently, the ROM of hip flexion was significantly greater in females than males (males: 15.2 ± 9.3°; females: 25.3 ± 12.0°).

Table 5.3. Group mean results for hip flexion, knee flexion and ankle plantar/dorsiflexion (– plantarflexion; + dorsiflexion) angles at IC, maximum angle and ROM (Mean ± standard deviation).

		Hip flexion (°)	**Knee flexion (°)**	**Ankle plantar/dorsiflexion (°)**
	IC	13.9 ± 5.8	19.6 ± 6.4[1]	-17.1 ± 10.3[2]
Males	Maximum	29.2 ± 7.9[3]	62.6 ± 12.1	31.4 ± 6.5
	ROM	15.3 ± 9.3[4]	43.0 ± 14.4[5]	48.5 ± 12.3[6]
	IC	13.7 ± 5.9	14.8 ± 6.3[1]	-25.1 ± 9.3[2]
Females	Maximum	39.1 ± 11.9[3]	67.9 ± 12.5	31.5 ± 5.9
	ROM	25.3 ± 12.0[4]	53.1 ± 13.8[5]	56.5 ± 11.1[6]

[1-6]: Significant difference between males and females ($p<0.05$).

Knee flexion at IC was significantly greater in males than females (males: 19.6 ± 6.4°; females: 14.8 ± 6.3°) (Table 5.3). There was no significant difference in the maximum knee flexion angle between males and females (males: 62.6 ± 12.1°;

females: 67.9 ± 12.5°). The ROM of knee flexion was significantly greater in females than males (males: 43.0 ± 14.4°; females: 53.1 ± 13.8°).

At IC, ankle plantarflexion was significantly greater in females compared to males (males: -17.1 ± 10.3°; females: -25.1 ± 9.3°) (Table 5.3). There was no significant difference in the maximum angle of ankle dorsiflexion between males and females (males: 31.4 ± 6.5°; females: 31.5 ± 5.9°). However, the ROM of ankle plantar/dorsiflexion was significantly greater in females than males (males: 48.5 ± 12.3°; females: 56.5 ± 11.1°).

Figure 5.8(a) shows the relationship between the angular displacement of the hip and the angular displacement of the knee in the sagittal plane. The relationship between hip flexion and knee flexion was fairly linear in males and females with males and females exhibiting greater knee flexion relative to hip flexion. Males and females exhibited a similar rate of hip flexion relative to knee flexion (males: 0.4 ± 0.1; females: 0.5 ± 0.1) (Table 5.4).

Figure 5.8(b) shows the relationship between the angular displacement of the knee and the angular displacement of the ankle in the sagittal plane. The relationship between knee flexion and ankle plantar/dorsiflexion was fairly linear in males and females with both exhibiting greater ankle plantar/dorsiflexion relative to knee flexion. Males and females exhibited a similar rate of knee flexion relative to ankle plantar/dorsiflexion (males: 1.4 ± 0.3; females: 1.5 ± 0.5) (Table 5.4).

Figure 5.8(c) shows the relationship between the angular displacement of the hip and the angular displacement of the ankle in the sagittal plane. The relationship between hip flexion and ankle plantar/dorsiflexion was fairly linear in males and females with males and females exhibiting greater ankle plantar/dorsiflexion relative to hip flexion. Males exhibited a greater rate of hip flexion relative to ankle plantar/dorsiflexion than females (males: 4.1 ± 0.8; females: 3.7 ± 1.0) (Table 5.4).

Table 5.4. Peak gradient of angle – angle coupling graphs.

	Hip – knee	**Knee – ankle**	**Hip – ankle**
Males	0.4 ± 0.1	1.4 ± 0.3	4.1 ± 0.8
Females	0.5 ± 0.1	1.5 ± 0.5	3.7 ± 1.0

5.3.2 Discrete relative phase.

The DRP results are presented in two sections; comparison of DRP of intra-limb joint couplings (hip – knee, knee – ankle and hip – ankle joint couplings) between the left and right legs and comparison of DRP of inter-limb joint couplings (hip, knee and ankle) between the left and right legs. The lower the DRP angle between two joints in the same leg or between corresponding joints in both legs the tighter the coupling between the two joints (the closer the two events are in time in each cycle) and the lower the variability of the discrete relative phase (vDRP) (indicated by standard deviation of DRP) the more stable the coupling between the two joints.

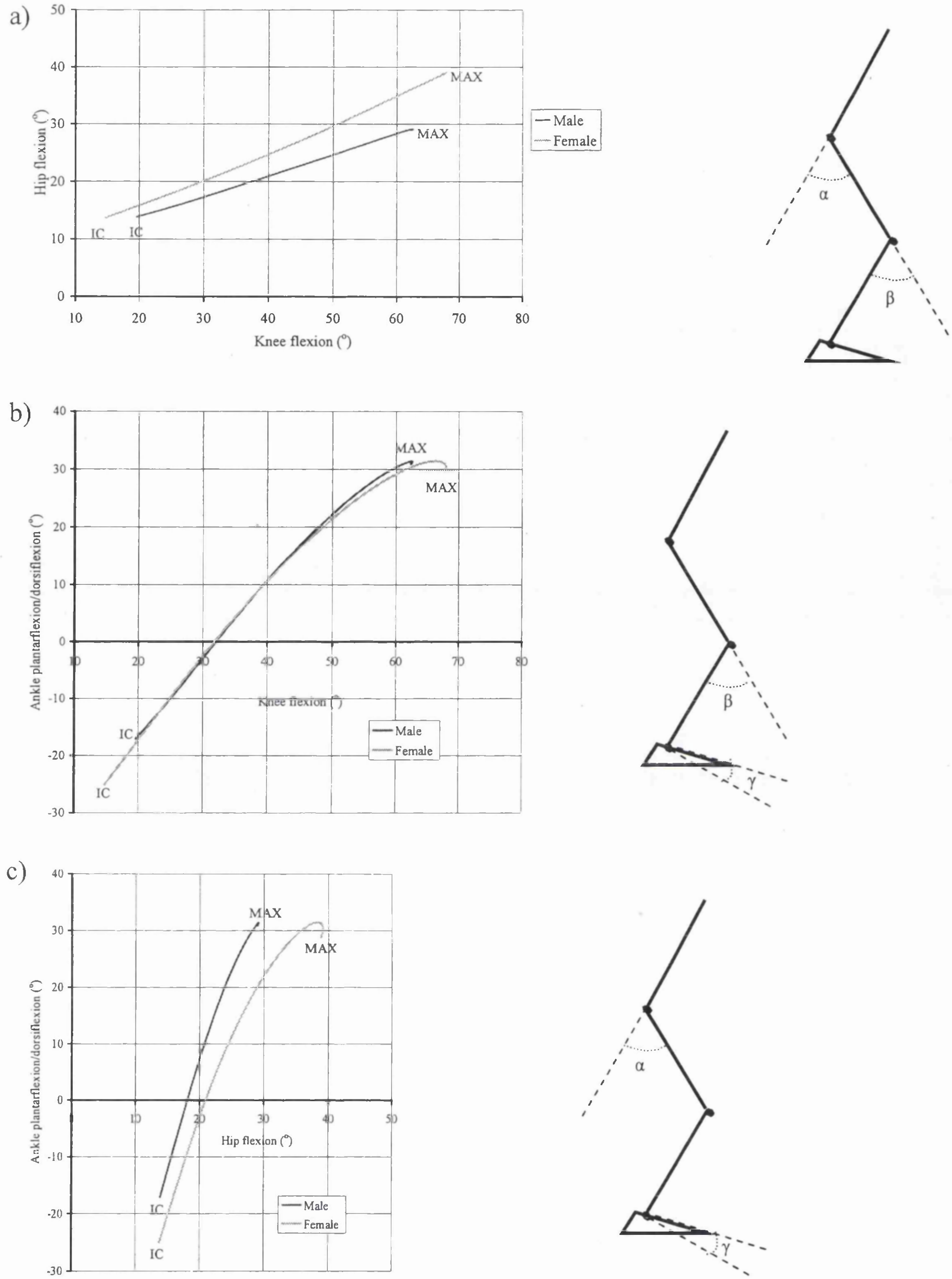


Figure 5.8. Angle – angle diagrams of a) hip flexion (α) – knee flexion (β) joint coupling, b) knee flexion (β) – ankle plantar/dorsiflexion (γ) joint coupling and c) hip flexion (α) – ankle plantar/dorsiflexion (γ) between IC and maximum knee flexion for males and females.

5.3.2.1 Intra-limb joint couplings.

The mean DRP angle and vDRP angle for hip – knee, knee – ankle and hip – ankle joint couplings in each leg and in combined legs are shown in Table 5.5 and Figures 5.9 and 5.10. The DRP angle for the hip – knee joint coupling was negative for the left leg, the right leg and combined legs in both males and females, indicating that the knee reached its maximum angle after the hip. The DRP angle was positive for the knee – ankle and hip – ankle joint couplings for the left leg, the right leg and combined legs in both males and females indicating that the ankle reached its maximum angle before both the hip and knee. These values indicate that on average the ankle reached its maximum angle first, then the hip and finally the knee for both left and right legs in both males and females.

Table 5.5. The DRP and vDRP angles for the hip – knee, knee – ankle and hip – ankle joint couplings.*

		Hip – Knee (°)	**Knee – Ankle (°)**	**Hip – Ankle (°)**
		DRP ± vDRP	**DRP ± vDRP**	**DRP ± vDRP**
Males	Combined	-9.73 ± 50.57	25.44 ± 52.49	15.71 ± 56.19
	Left	-8.74 ± 60.31	33.18 ± 55.55	24.44 ± 44.30
	Right	-10.72 ± 40.71	17.69 ± 49.92	6.97 ± 66.45
	Right – left difference	1.98	15.49	17.47
Females	Combined	-13.11 ± 57.10	45.70 ± 62.36	32.58 ± 67.82
	Left	-0.78 ± 40.08	57.53 ± 33.76	58.31 ± 60.76
	Right	-25.44 ± 68.81	33.86 ± 81.33	6.85 ± 66.46
	Right – left difference	24.66	23.67	51.46

* No significant differences between males and females.

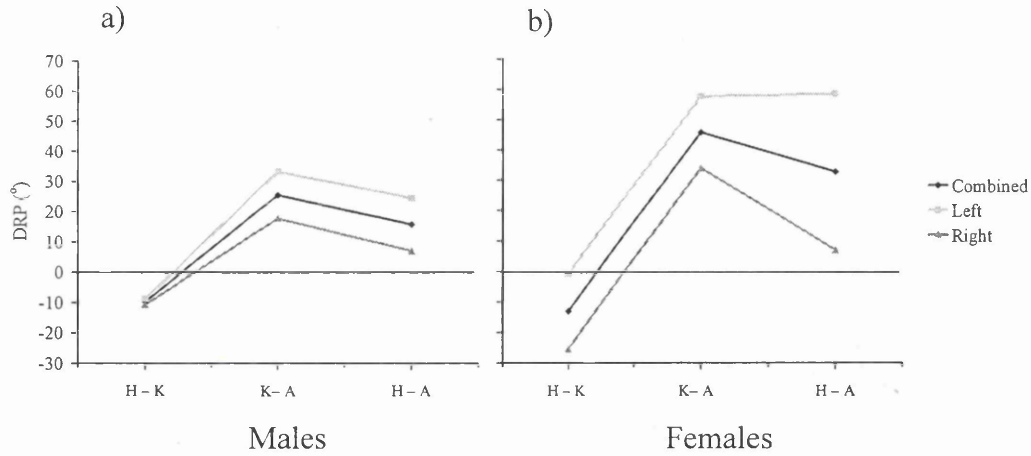


Figure 5.9. DRP angles for the hip – knee, knee – ankle and hip – ankle joint couplings for a) males and b) females.

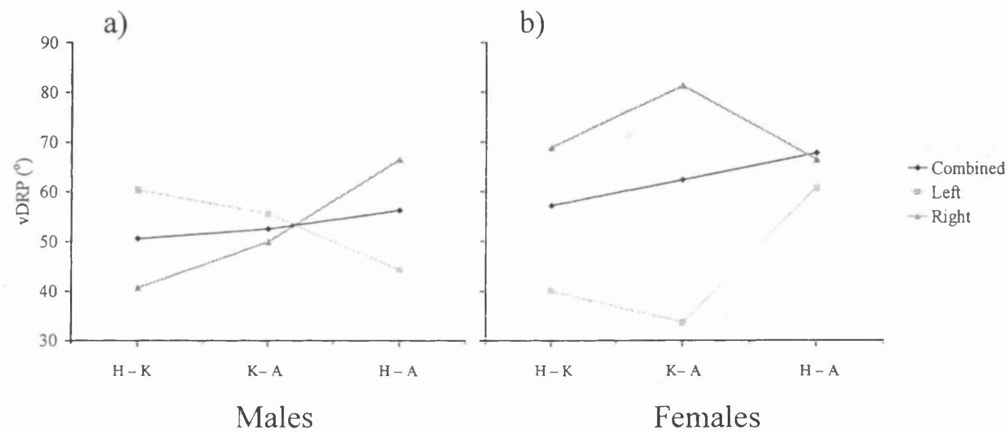


Figure 5.10. vDRP angles for the hip – knee, knee – ankle and hip – ankle joint couplings for a) males and b) females.

Independent-samples t-tests were carried out on the DRP angles for hip – knee, knee – ankle and hip – ankle joint couplings in the sagittal plane to examine gender differences. Normality of the data and between-group homogeneity of variance were assessed prior to statistical analysis, as described in Chapter 3, section 3.2.9. All parameters were found to display normal distribution and equal variances between groups. Whilst there was no significant differences in the DRP angles between males and females, the DRP angle and the vDRP angle were greater for females than males for combined legs hip – knee (males- DRP: -9.73, vDRP: 50.57; females- DRP: -13.11, vDRP: 57.10), knee – ankle (males- DRP: 25.44, vDRP: 52.49; females- DRP: 45.70, vDRP: 62.36) and hip – ankle (males- DRP: 15.71, vDRP: 56.19; females- DRP: 32.58, vDRP: 67.82) joint couplings (Table 5.5 and Figures 5.9 and 5.10).

For left legs, the DRP angle and the vDRP angle were greater for males than females for the hip – knee joint coupling (males- DRP: -8.74, vDRP: 60.31; females- DRP: 0.78, vDRP: 40.08). For the knee – ankle joint coupling, the DRP angle was greater for females than males but the vDRP angle was greater for males than females (males- DRP: 33.18, vDRP: 55.55; females- DRP: 57.53, vDRP: 33.76). For the hip – ankle joint coupling (males- DRP: 24.44, vDRP: 44.30; females- DRP: 58.31, vDRP: 60.76) the DRP angle and the vDRP angle were greater for females than males (Table 5.5 and Figures 5.9 and 5.10).

For right legs, the DRP angle and the vDRP angle for the hip – knee (males- DRP: -10.72, vDRP: 40.71; females- DRP: -25.44, vDRP: 68.81) and knee – ankle (males- DRP: 17.69, vDRP: 49.92; females- DRP: 33.86, vDRP: 81.33) joint couplings were greater for females than males. Similar values of the DRP angle and the vDRP angle were displayed by males and females for the hip – ankle joint coupling (males- DRP: 6.97, vDRP: 66.45; females- DRP: 6.85, vDRP: 66.46) (Table 5.5 and Figures 5.9 and 5.10).

For combined legs, the DRP angle was largest for the knee – ankle joint coupling and smallest for the hip – knee joint coupling in both males and females indicating that the tightest coupling existed between the hip and knee joints and the least tight coupling existed between the knee and ankle joints in both males and females. For combined legs, the vDRP angle was largest for the hip – ankle joint coupling and smallest for the hip – knee joint coupling in both males and females indicating that the least stable joint coupling was the hip – ankle and the most stable joint coupling was the hip – knee.

The difference in the DRP angles of corresponding joint couplings (hip – knee, knee – ankle and hip – ankle) between the right and left legs shows that the DRP angle for hip – knee (males: 1.98; females: 24.66), knee – ankle (males: 15.49; females: 23.67) and hip – ankle (males: 17.47; females: 51.46) joint couplings were more similar between left and right legs in males than females (Table 5.5 and Figure 5.9), which indicates greater symmetry between the movements of the legs in males than females.

5.3.2.2 Inter-limb joint couplings.

Table 5.6 and Figure 5.11 show the DRP angles and the vDRP angles for the left – right hip, knee and ankle joint couplings for males and females. The DRP angle for the left – right hip joint coupling in females was positive indicating that the right hip reached its maximum angle before the left hip. For the left – right joint couplings of the hips in males and the knees and ankles in both males and females, the DRP angle was negative indicating that the left joint reached its maximum angle before the right joint (Table 5.6 and Figure 5.11).

Table 5.6. DRP and vDRP angles of left – right joint couplings for the hip, knee and ankle joints.

	Hip (°) **DRP ± vDRP**	**Knee (°)** **DRP ± vDRP**	**Ankle (°)** **DRP ± vDRP**
Males	-2.07 ± 41.48	-3.75 ± 20.01	-19.60 ± 65.93
Females	19.67 ± 33.58	-9.17 ± 52.39	-31.62 ± 42.04

* No significant differences between males and females.

Independent samples t-tests were carried out on the DRP angles for left – right joint couplings (e.g. left knee coupled with right knee) to examine gender differences. Normality of the data and between-group homogeneity of variance were assessed prior to statistical analysis, as described in Chapter 3, section 3.2.9. All parameters were found to display normal distribution and equal variances between groups. Whilst there was no significant differences between males and females, the DRP angles for the left – right joint couplings were greater for females than males for the hip (males: -2.07; females: 19.67), knee (males: -3.75; females: -9.17) and ankle (males: -19.60; females: -31.62). The vDRP angle was greater for males than females for the hip (males: 41.48; females: 33.58) and the ankle (males: 65.93; females: 42.04) but the vDRP angle for the knee was greater for females than males (males: 20.01; females: 52.39). The DRP (symmetry) and the vDRP (stability) results indicate greater symmetry between hip, knee and ankle joints and greater stability between the left and right knees in males compared to females.

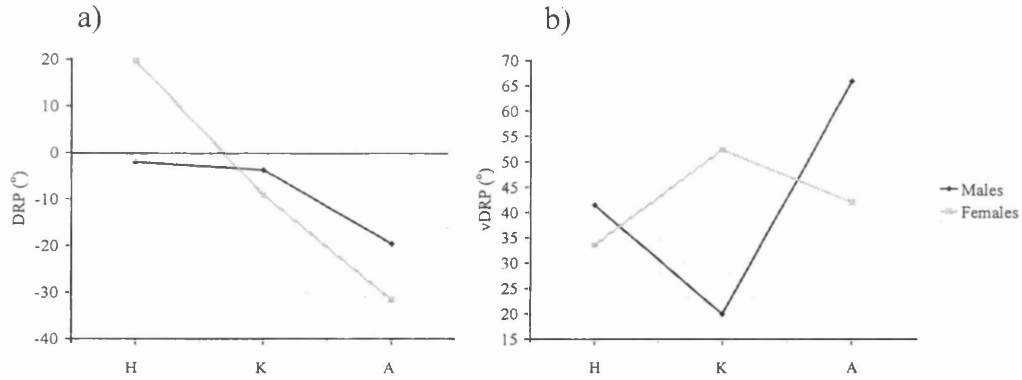


Figure 5.11. a) DRP angle and b) vDRP angle of left – right joint couplings for the hip (H), knee (K) and ankle (A) joints for males and females.

5.3.3 Leg stiffness.

Vertical GRF, vertical CG displacement and leg stiffness mean data were based on 15 trials for males and 15 trials for females (5 subjects × 3 trials). Absolute and normalised (with respect to weight and height) leg stiffness data were calculated for males and females. Independent-samples t-tests were carried out on the absolute and normalised change in vertical GRF, absolute and normalised vertical displacement of the CG and absolute and normalised leg stiffness to examine gender differences. Normality of the data and between-group homogeneity of variance were assessed prior to statistical analysis, as described in Chapter 3, section 3.2.9. All parameters were found to display normal distribution. All parameters displayed equal variances between groups other than the absolute and normalised change in vertical GRF.

Both the absolute and normalised change in vertical GRF were significantly greater in males than females (males: absolute: 2501.2 ± 692.0 N, normalised: 3.64 ± 1.01 BW; females: absolute: 1659.3 ± 411.3 N, normalised: 2.87 ± 0.60 BW) (Table 5.7). There was no significant difference in the absolute or normalised vertical displacement of the CG between males and females (males: absolute: 0.167 ± 0.044 m, normalised: 0.095 ± 0.025 ht; females: absolute: 0.161 ± 0.048 m, normalised: 0.097 ± 0.029 ht). However, both absolute and normalised leg stiffness were significantly greater in males than females (males: absolute: 15.02 ± 8.82 kN/m, normalised: 38.55 ± 20.91 BW/ht; females: absolute: 10.29 ± 3.56 kN/m, normalised: 29.61 ± 7.94 BW/ht). The change in normalised vertical GRF and normalised vertical CG displacement at the corresponding points in time for males and females is shown in Figure 5.12.

Figure 5.12 shows normalised vertical GRF and normalised CG height between IC and maximum knee flexion. Absolute mean time between IC and maximum knee flexion was 0.189 ± 0.042 s for males and 0.191 ± 0.033 s for females. As there was no significant difference in absolute mean time between IC and maximum knee flexion between males and females, the time histories of relevant variables were plotted against a combined mean contact time of 0.190 s.

Table 5.7. Group mean results for change in vertical GRF, vertical displacement of the CG, leg stiffness, change in knee joint moment, angular displacement of the knee and knee stiffness during landing (mean ± standard deviation).

		Males	Females
Change in vertical GRF	Absolute (N)	2501.2 ± 692.0[1]	1659.3 ± 411.3[1]
	Normalised (BW)	3.64 ± 1.01[2]	2.87 ± 0.60[2]
Vertical CG displacement	Absolute (m)	0.167 ± 0.044	0.161 ± 0.048
	Normalised (ht)	0.095 ± 0.025	0.097 ± 0.029
Leg stiffness	Absolute (kN/m)	15.02 ± 8.82[3]	10.29 ± 3.56[3]
	Normalised (BW/ht)	38.55 ± 20.91[4]	29.61 ± 7.94[4]
Change in knee joint moment	Absolute (N.m)	121.6 ± 76.6	93.2 ± 31.8
	Normalised (BW.ht)	0.101 ± 0.063	0.097 ± 0.033
Knee flexion displacement	Absolute (°)	28.8 ± 9.9[5]	51.9 ± 14.2[5]
Knee stiffness	Absolute (N.m/°)	4.23 ± 1.08[6]	1.79 ± 0.90[6]
	Normalised (BW.ht/°)	0.0035 ± 0.0009[7]	0.0019 ± 0.0009[7]

[1-7]: Significant difference between males and females ($p<0.05$).

5.3.4 Knee stiffness.

Since the force-measurement system was unable to determine the centre of pressure on the left foot, knee joint moment data were only collected for the right leg. Therefore mean data were based on 15 trials for males and 15 trials for females (5 subjects × 3 trials × 1 leg). Absolute and normalised (with respect to weight and height) knee joint stiffness data were calculated for males and females. Independent-samples t-tests were carried out on the absolute and normalised change in knee joint moment, knee flexion angular displacement and the absolute and normalised knee stiffness to examine gender differences. Normality of the data and between-group homogeneity of variance were assessed prior to statistical analysis, as described in Chapter 3, section 3.2.9. All parameters were found to display normal distribution and equal variances between groups.

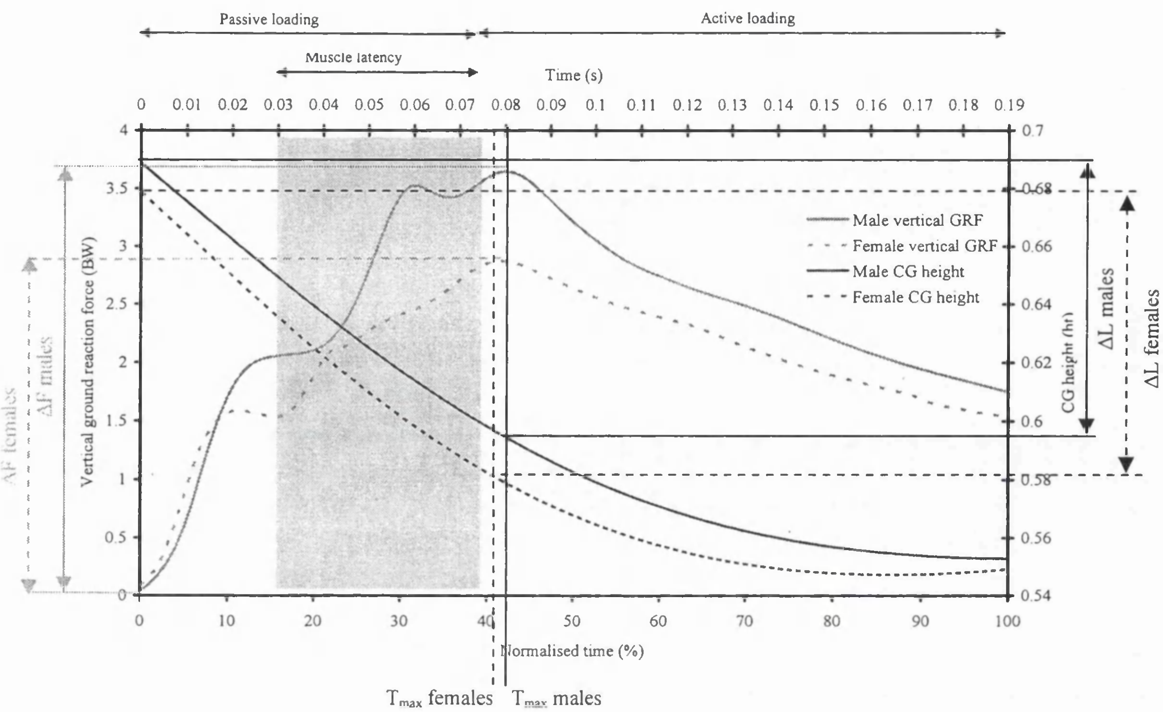


Figure 5.12. Normalised vertical GRF and CG height between IC and maximum knee flexion for males and females (ΔF = change in normalised vertical GRF, ΔL = normalised vertical displacement of the CG and T_{max} = time of maximum normalised vertical GRF).

There was no significant difference in the absolute or normalised change in knee joint moment between males and females (males: absolute: 121.6 ± 76.6 N.m, normalised: 0.101 ± 0.063 BW.ht; females: absolute: 93.2 ± 31.8 N.m, normalised: 0.097 ± 0.033 BW.ht) (Table 5.7). Knee flexion angular displacement was significantly greater in females than males (males: 28.8 ± 9.9^{o}; females: 51.9 ± 14.2^{o}). Consequently, both absolute and normalised knee joint stiffness were significantly greater in males than females (males: absolute: 4.23 ± 1.08 N.m/°, normalised: 0.0035 ± 0.0009 BW.ht/°; females: absolute: 1.79 ± 0.90 N.m/°, normalised: 0.0019 ± 0.0009 BW.ht/°). The change in normalised knee joint moment and knee flexion angular displacement at the corresponding points in time for males and females is shown in Figure 5.13.

Figure 5.13 shows normalised knee joint moment and knee flexion angle between IC and maximum knee flexion. As stated previously, there was no significant difference in absolute mean time between IC and maximum knee flexion between males and females, therefore the time histories of relevant variables were plotted against a combined mean contact time of 0.190 s.

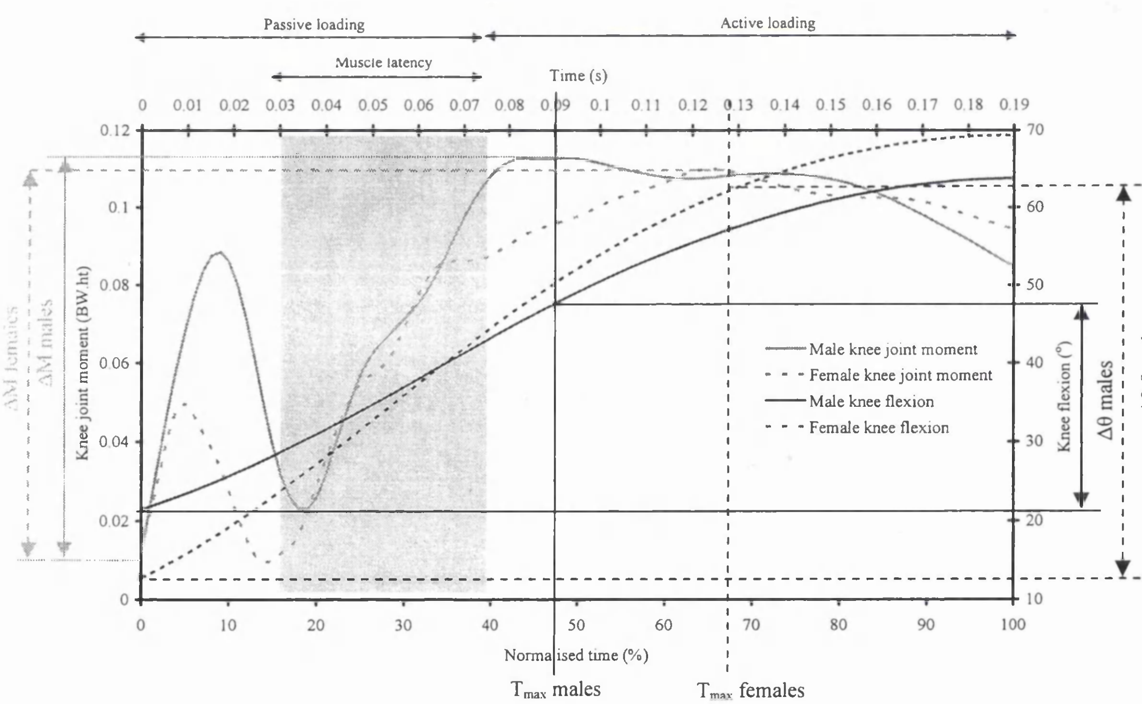


Figure 5.13. Normalised knee joint moment and knee flexion between IC and maximum knee flexion for males and females (ΔM = change in normalised knee joint moment, Δθ = knee flexion angular displacement and T_{max} = time of maximum normalised knee joint moment).

5.4 Discussion.

5.4.1 Coordination.

The angle – angle diagram of the hip – knee joint coupling (Figure 5.8a) shows that the knee moved through a greater angle relative to the hip during landing in both males and females. Peak gradient of the hip – knee curve was greater in females than males and range of motion of the hip and knee were greater in females than males (Tables 5.3 and 5.4).

The angle – angle diagram of the knee – ankle joint coupling (Figure 5.8b) shows that the ankle range of motion was greater relative to the knee during landing in both males and females but range of motion of knee flexion and ankle plantar/dorsiflexion was greater in females than males (Table 5.3). The peak gradients of the ankle – knee curves were similar for males and females (Table 5.4).

The angle – angle diagram of the hip – ankle joint coupling (Figure 5.8c) shows that the ankle range of motion was greater relative to the hip during landing in both males and females but range of motion of hip flexion and ankle plantar/dorsiflexion greater in females than males (Table 5.3). The peak gradients of the ankle – hip curves were similar for males and females (Table 5.4). Range of motion was significantly greater in all joints in females compared to males (Table 5.3). This increased range of motion in the hip, knee and ankle exhibited by females is consistent with the significantly lower leg stiffness and knee joint stiffness exhibited by females (Table 5.7) and may indicate less dynamic stability of the hip, knee and ankle joints in females compared to males.

During a two-footed landing, it is reasonable to assume that the DRP angle of two events in one leg (intra-limb joint coupling) should be very similar to the corresponding DRP angle in the other leg (i.e. the smaller the difference in the DRP angle between the two legs the greater the symmetry between the two legs). Similarly, the DRP angle of the corresponding event in left and right legs (inter-limb joint coupling) should be very similar (i.e. the smaller the DRP angle for right – left hip, knee and ankle joints the greater the symmetry between the two legs). Comparison of the DRP angles of corresponding joint couplings (hip – knee, knee – ankle and hip – ankle) between the right and left legs shows that the DRP angles for all three joint couplings were more similar between left and right legs in males than in females (Table 5.5 and Figure 5.9). Comparison of the DRP angle between corresponding joints (hip, knee and ankle) in the right and left legs show a greater DRP angle in females for left – right hip, knee and ankle joint couplings compared to males (Table 5.6 and Figure 5.11). These results indicate less symmetry and, therefore, less coordination between the left and right legs during landing in females compared to males. The reduced symmetry in females compared to males may indicate greater asymmetry in loading on the passive support structures of the legs during landing in females compared to males.

The stability of the coordination (indicated by the vDRP angles) between the hip – knee, knee – ankle and hip – ankle joint couplings indicate that males displayed a similar level of stability between left and right legs during landing (Figure 5.10a), whereas females displayed a large difference in the stability of hip – knee and knee – ankle joint couplings between right and left legs (Figure 5.10b). For combined legs, the stability of the coordination between hip – knee, knee – ankle and hip – ankle joint couplings were greater for males than females. The stability of the coordination between left – right joint couplings was less for males than females for the hip and ankle joints but less for females than males for the knee joints (Table 5.6 and Figure 5.11). The reduced stability in the coordination between the hip – knee, knee – ankle and hip – ankle joint couplings along with the reduced stability in the coordination between the left and right knees in females compared to males may indicate less dynamic stability of the legs, in particular the knees, in females compared to males which may be a contributory factor in the gender difference in the incidence of non-contact ACL injury.

5.4.2 Leg stiffness.

Since the males weighed more than the females, it was not surprising that the change in absolute vertical GRF was significantly greater for males than females. However, even when normalised to body weight, the change in vertical GRF was still significantly greater in males than females. There was no significant difference between males' and females' absolute or normalised vertical displacement of the CG. However, the resulting absolute and normalised leg stiffness were significantly greater in males than females (Table 5.7). This suggests reduced dynamic stability of the leg in females compared to males during landing which may be a contributory factor in the increased incidence of non-contact ACL injury in females compared to males.

Granata et al. (2002b) reported mean values of leg stiffness during hopping of 33.9 ± 4.2 kN/m for males and 26.3 ± 6.5 kN/m for females. These values are greater than the values of leg stiffness observed for males (15.02 ± 8.82 kN/m) and females (10.29 ± 3.56 kN/m) in the present study. This is likely to be due to differences in the task; hopping involves storage and utilisation of strain energy and, therefore, maintenance of leg stiffness, whereas landing involves dissipation of strain energy and, therefore, a fairly rapid reduction in leg stiffness following the initial impact phase.

5.4.3 Knee stiffness.

There was no significant difference in the absolute or normalised change in knee joint moment between males and females during landing. However, the knee flexion angular displacement was significantly greater in females than males. Consequently, the absolute and normalised knee joint stiffness was significantly greater in males than females during landing (Table 5.7). The reduced absolute and normalised knee stiffness in females compared to males may contribute to the reduced absolute and normalised leg stiffness in females compared to males. The reduced knee joint stiffness in females compared to males may also indicate reduced dynamic stability of the knee during landing which may contribute, at least in part, to the greater incidence of non-contact ACL injury in females compared to males.

Farley and Morgenroth (1999) reported values of between 6.4 ± 1.4 N.m/° and 11.0 ± 1.5 N.m/° for knee stiffness during hopping and Arampatzis et al. (1999) reported an increase in knee stiffness from 7 ± 2 N.m/° to 17 ± 3 N.m/° as running velocity increased from 2.5 to 6.5 m/s. These values of knee stiffness are greater than the values of knee stiffness in the present study (males: 4.23 ± 1.08 N.m/°; females:

1.79 ± 0.90 N.m/°). This again is likely to be due to differences in the task subjects were required to perform.

5.5 Summary and conclusions.

Range of motion of hip, knee and ankle joints were greater for females than males during landing. Comparison of the DRP angles of the hip – knee, knee – ankle and hip – ankle joint couplings between left and right legs and comparison of the DRP angle for left – right hip, knee and ankle joint couplings indicate less symmetry between the left and right legs during landing in females compared to males which may indicate greater asymmetry in loading on the passive support structures of the joints of the legs in females compared to males during landing which, in turn, may influence stain on the ACL. Furthermore, the reduced stability in the coordination between hip – knee, knee – ankle and hip – ankle joint couplings and the reduced stability in the coordination between the left and right knee joints in females compared to males during landing may indicate reduced dynamic stability of the legs during landing in females compared to males.

Males exhibited significantly greater absolute and normalised leg stiffness and significantly greater absolute and normalised knee joint stiffness during landing compared to females. In conjunction with the coordination data, this may indicate reduced dynamic stability of the legs in females compared to males. The reduced dynamic stability of the legs in females compared to males during landing may contribute to the greater incidence of ACL injury in females compared to males. Future research should investigate the effects of coordination and strength training on leg stiffness and knee joint stiffness in females during landing.

Chapter 6

General summary and recommendations for future research.

6. General summary and future research.

The incidence of non-contact anterior cruciate ligament (ACL) injury is reported to be 6 to 8 times greater in females than males competing in the same activities. Injury to the ACL occurs as a result of insufficient stability of the tibiofemoral joint, which fails to prevent abnormal movement of the femur on the tibia. The stability of the tibiofemoral joint is maintained by passive (non-contractile) and dynamic (contractile) mechanisms. The passive mechanisms include the shape of the articular surfaces, the menisci, the ligaments and the joint capsule. The dynamic mechanisms consist of the muscle-tendon units that cross the joint, in particular, the quadriceps and hamstrings. The relative significance of the various passive and dynamic mechanisms in maintaining the stability of the tibiofemoral joint is not clear. A number of intrinsic and extrinsic risk factors have been proposed to account for the gender difference in the incidence of ACL injuries. However, most of the proposed risk factors have arisen from uni-variate correlation studies based on relatively small samples. The purpose of the review of literature was to present a risk factor model for ACL injury based on a review of passive and dynamic stability mechanisms. From the review of literature it was concluded that ACL injuries occur most frequently in non-contact situations with the athlete in single foot stance, forceful valgus knee collapse with knee close to full extension accompanied with external or internal rotation of the tibia. The situation appears to occur in two particular types of movement; landing from a jump and rapid change of direction initiated on one leg, such as a side-cutting and cross-cutting manoeuvre. Current evidence suggests that the greater incidence of ACL injury in females is largely due to gender differences in dynamic stability rather than passive stability of the knee joint. In addition, gender differences in landing/cutting kinematics and kinetics appear to account, at least in part, for the gender differences in the incidence of non-contact ACL injury. However, previous studies investigating gender differences in knee kinematics only report absolute angular displacement rather than angular displacement relative to lower limb alignment during normal upright standing. Also, previous studies report gender differences in knee kinematics and kinetics during unopposed landing/cutting manoeuvres, but there would appear to be no reports on the effect of opposition on landing kinematics and kinetics. Finally, since the cause of ACL injury is likely to be the result of a complex interaction of risk factors (Lysens et al., 1984), composite variables, including measures of coordination and stiffness, may provide more insight into the causes of ACL injury in general and the greater incidence of ACL injury in females in particular. There would appear to be no reports of lower limb coordination or leg and knee stiffness during landing/cutting manoeuvres in males and females. The purpose of the current study was to investigate landing biomechanics in males and females in order to try to identify possible risk factors associated with the greater incidence of non-contact ACL injury in females compared to males. The experimental phase of the project consisted of three studies:

1. A kinematic (absolute and relative to lower limb alignment during upright standing) and kinetic (relative to body weight and/or height) analysis of male and female volleyball players during landing from opposed volleyball block jumps.

2. A kinematic and kinetic analysis of male and female volleyball players during landing from unopposed and opposed volleyball block jumps.

3. Examination of lower limb coordination and stiffness in male and female volleyball players when landing from opposed volleyball block jumps.

Prior to undertaking the experimental studies, an investigation into the reliability of the motion analysis system used to conduct the experimental studies was undertaken. The objectives of the reliability investigation were to determine and compare the level of noise in the Vicon – laboratory environment and Vicon – laboratory environment – human subject systems as reflected in the variation in the 3D co-ordinates of markers, the distance between markers, the angles formed by 3 markers, and the centre of pressure. Three markers were placed on an inanimate static structure and were compared to the estimated joint centres derived from markers placed on two physically active male subjects, in accordance with the Vicon system's lower body plug-in gait marker set, while standing still on a force plate with their feet placed either side of a block of wood 7.5 cm wide. The variation in the measurements was, as expected, less for trials of the inanimate static structure than for human subject trials. The results of the Vicon – laboratory environment – human subject system indicate that the Vicon 512 motion analysis system reliably determines the positions of joint centres with a very small variation (standard deviations of $\pm$ 6.31 mm, $\pm$ 2.56 mm and $\pm$ 1.48 mm for joint centre x, y and z coordinates respectively). These small variations in joint coordinates correspond to variation in static distance measures of $\pm$ 1.06 mm (linear) and $\pm$ 0.59^{o} (angular). The variation in the static distance measures is the noise in the Vicon – laboratory environment – human subject system.

The purpose of study one was to investigate the effects of gender on frontal and sagittal plane kinematics (absolute and relative to lower limb alignment during upright standing) and kinetics (normalised) in university volleyball players when performing opposed block jump landings. Six female and six male university volleyball players performed a static reference trial and three dynamic trials where subjects were instructed to jump up and block a volleyball suspended 10 cm above a net of height 2.43 m for male subjects and 2.23 m for female subjects as it was spiked against them by an opposing player. Sagittal plane and frontal plane kinematics (knee flexion/extension, knee valgus/varus, inter hip distance, inter-knee distance, inter-ankle distance) and kinetics (normalised ground reaction force (GRF), GRF angle, normalised GRF moment arm and normalised GRF moment) were determined during landing.

Absolute knee flexion at ground contact was significantly smaller in females than males, but there was no significant difference in relative knee flexion on ground contact. Maximum knee flexion (absolute and relative) and range of motion of knee flexion were significantly greater in females than males. There was no significant difference in knee flexion (on ground contact, at maximum knee flexion or range of motion) between dominant and non-dominant legs in males or females. The time histories of the normalised GRF moment arm and moment in the sagittal plane were similar in males and females. Females displayed significantly smaller normalised GRF moment arm and moment (flexion) at the start of muscle latency (ML) than males. Since the normalised GRF moment in the sagittal plane was similar in males and females,. the greater maximum knee flexion (absolute and relative) and range of motion of knee flexion in females compared to males is more likely to be due to reduced dynamic stability of the knee in resisting knee flexion in females compared to males rather than due to greater normalised GRF moments acting about the knee joint centre in the sagittal plane during landing.

In the frontal plane, females exhibited a significantly greater maximum valgus angle and significantly greater range of motion of knee valgus/varus compared to males in both absolute and relative terms. Consequently, the difference in normal alignment of the knees in the frontal plane in normal upright standing does not account for the difference in maximum valgus angle on landing between males and females. There was no significant difference in knee valgus/varus angle at ground contact, maximum knee valgus/varus angle or range of knee valgus/varus between dominant and non-dominant legs in males, but there was a significant difference in maximum valgus angle and range of motion between the dominant and non-dominant legs in females. The time histories of the normalised GRF moment arm and moment in the frontal plane were different between males and females. Females' varus normalised GRF moment arms and moment were significantly different from males' valgus normalised GRF moment arm and moment at initial ground contact and at the start of active loading (AL). The maximum varus normalised GRF moment arm and moment was significantly greater in females than males.

Angular velocity of the knees in both frontal and sagittal planes were significantly greater in females than males in the passive phase of landing, but not significantly different in the active phase. These results may indicate less dynamic stability of the knee in females compared to males in the passive phase. The lower the dynamic stability, the greater the dependence on the passive support structures, especially the ligaments, for the maintenance of joint stability. As ACL injuries occur most frequently in the passive phase of landing manoeuvres, the present results suggest that lack of dynamic stability of the knee in the passive phase could be a contributing factor in the reported greater incidence of ACL injury in females compared to males.

The aim of study two was to examine the effect of opposition on the kinematics and kinetics of landing from a volleyball block in male and female university volleyball players. Six female and six male university volleyball players performed two landing tasks designed to mimic a volleyball block jump and landing. 1) Unopposed: Subjects were instructed to jump up and pretend to block a volleyball suspended 10 cm above a net of height 2.43 m for male subjects and 2.23 m for female subjects. 2) Opposed: Subjects jumped to block the suspended volleyball as it was spiked against them by an opposing player. Sagittal plane and frontal plane kinematics (knee flexion/extension, knee valgus/varus, inter hip distance, inter-knee distance, inter-ankle distance) and kinetics (normalised GRF, GRF angle, normalised GRF moment arm and normalised GRF moment) were determined during landing.

The results of the study showed that there was no significant difference in knee flexion at ground contact between males and females or between unopposed and opposed conditions. Maximum knee flexion, range of motion of knee flexion and angular velocity of knee flexion during the passive phase of landing were significantly greater in females than males and significantly greater during unopposed trials than opposed trials. Maximum knee valgus angle and range of motion of knee valgus were significantly greater in females than males, but there were no significant differences in maximum knee valgus and range of motion of knee valgus between unopposed and opposed trials. Maximum normalised GRF and normalised GRF at AL were significantly greater during opposed trials than unopposed trials but there was no significant difference between males and females. Normalised knee GRF flexor moment was significantly smaller in females than males at ML and significantly smaller in opposed trials than unopposed. The knee varus normalised GRF moment at ML was significantly greater in females than males but there was no significant difference between unopposed and opposed trials. At AL, the knee valgus normalised GRF moment in males was significantly different from the varus normalised GRF moment in females and the frontal plane normalised GRF moment was significantly different between unopposed and opposed trials. The changes that occurred in knee kinematics and kinetics as a result of opposition during landing from a volleyball block jump suggest that in training programmes, coaches should incorporate exercises that simulate opposed landing manoeuvres as the effect of opposition may significantly alter subjects' neuromuscular responses during landing, particularly in the sagittal plane. Frontal plane kinematics and kinetics for males and for females appear to be similar in unopposed and opposed conditions.

The aim of study three was to investigate lower limb coordination and stiffness in male and female university volleyball players performing block jump landings. Five male and five female subjects performed opposed volleyball block jump tasks. Coordination was assessed using angle – angle plots of the hip – knee, knee – ankle and hip – ankle joint couplings and discrete relative phase (DRP) of the hip – knee, knee – ankle and hip – ankle joint couplings along with right – left joint couplings (i.e. left knee coupled with right knee). Leg stiffness was calculated as the ratio of the change in vertical GRF to the change in vertical displacement of the centre of gravity between ground contact and maximum vertical GRF. Knee stiffness was calculated as the ratio of the change in knee joint moment to the change in knee flexion angular displacement between ground contact and maximum knee joint moment.

Comparison of the DRP angles of the hip – knee, knee – ankle and hip – ankle joint couplings between left and right legs and comparison of the DRP angle for left – right hip, knee and ankle joint couplings indicated reduced symmetry between the left and right legs during landing in females compared to males which, it is reasonable to assume, may indicate greater likelihood of ligament strain in females compared to males during landing. Furthermore, females exhibited reduced stability in the coordination between hip – knee, knee – ankle and hip – ankle joint couplings and reduced stability in the coordination between the left and right knee joints than males during landing. These differences in the stability of coordination may indicate reduced dynamic stability of the legs during landing in females compared to males.

Males exhibited significantly greater absolute and normalised leg stiffness and significantly greater absolute and normalised knee joint stiffness during landing compared to females. In conjunction with the coordination data, this may indicate reduced dynamic stability of the leg in females compared to males. The reduced dynamic stability of the leg in females compared to males during landing may contribute to the greater incidence of ACL injury in females compared to males.

It is recommended that future research should investigate methods of improving the dynamic stability of the knee in females during landing, particularly during the passive phase of ground contact, which may reduce the likelihood of ACL strain and therefore non-contact ACL injury in females. Training methods to increase the dynamic stability of the knee in females should include; (i) increasing hamstring and quadriceps strength in order to maximise knee joint stiffness which, in turn, may reduce maximum knee flexion angle and range of motion of knee flexion during landing, and (ii) increasing hip abductor strength and positioning feet further apart on landing, which may reduce valgus movement of the knee during landing. It is also recommended that future research should investigate the effects of coordination and strength training on leg stiffness and knee joint stiffness in females during landing. Furthermore, future research into landing biomechanics should include opposition during the landing task as the effect of opposition may significantly alter subjects' neuromuscular responses during landing, particularly in the sagittal plane.

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Appendix A: Application for ethical approval.

UNIVERSITY OF WALES SWANSEA
DEPARTMENT OF SPORTS SCIENCE
DEPARTMENTAL ETHICS ADVISORY COMMITTEE

APPLICATION FOR ETHICAL COMMITTEE APPROVAL OF A RESEARCH PROJECT

In accordance with Departmental Safety Policy, all research undertaken in the department must be approved by the Departmental Ethics Advisory Committee **prior to** data collection. **Applications for approval should be typewritten on this form using the template available in the Public Folders.** The researcher(s) should complete the form in consultation with the project supervisor. Where appropriate, the application must include the following appendices:
(A) subject information sheet;
(B) subject consent form;
(C) subject health questionnaire.

After completing sections 1-12 of the form, eight copies of the form should be handed into the Department Administrator who will submit the application for consideration by the Departmental Ethics Advisory Committee. The applicant(s) will be informed of the decision of the Committee in due course.

1. DRAFT TITLE OF PROJECT
The relationship between leg strength and landing kinematics and kinetics in court games players.

2. NAMES AND STATUS OF RESEARCH TEAM
Gerwyn Hughes- Postgraduate student.
Prof James Watkins- Supervisor.

3. RATIONALE
Between 70% and 90% of anterior cruciate ligament (ACL) injuries have been reported to occur in non-contact situations (no direct contact with the knee) (McNair et al., 1993, Mykelbust et al., 1997, Griffin et al., 2000). Most non-contact ACL injuries appear to occur in situations involving one or more of the following manoeuvres: foot strike with knee close to full extension (Boden et al., 2000, Olsen et al., 2004), rapid changes in direction (Bartold, 1997), landing (Hume and Steele, 1997, Otago and Neal, 1997) and deceleration (Miller et al., 1995). The incidence of ACL injury is therefore relatively high in sports such as basketball, netball and volleyball that are characterised by a high frequency of landing, decelerating and rapid changes of direction.

The main function of the ACL is to help prevent anterior displacement of the proximal end of the tibia relative to the distal end of the femur (ADTRF). Therefore, movements which tend to cause ADTRF will load the ACL (Ma et al., 2000, Lervat et al., 2000). Consequently, excessive ADTRF will damage the ACL. The tibiofemoral joint is stabilised by passive and dynamic stabilisers. Passive stability is provided by the non-contractile structures of the knee, in particular, the joint capsule, menisci and ligaments. Dynamic stability is provided by the muscles that cross the tibiofemoral joint, in particular, the quadriceps, hamstrings and the triceps surae. Insufficient dynamic stability will increase the risk of injury to the passive support structures.

The incidence of non-contact ACL injury in females has been reported to be 6 to 8 times greater than in males competing in the same sports (Chandy and Grana, 1985, Ferretti et al., 1992, Paulos, 1992, Malone et al., 1993, Lidenfeld et al., 1994, Arendt and Dick, 1995, Gwinn et al., 2000). Various risk factors have been reported to be associated with the apparent increased incidence of ACL injuries in females. However, the only risk factors for which there appears to be strong empirical evidence are those factors related to dynamic stability, in particular the patella tendon tibia shaft angle, muscle activity pattern, muscle strength and muscle stiffness.

Patella tendon-tibia shaft angle (PTTSA):
Nunley et al., (2003) investigated the effects of knee angle on PTTSA in males and females. They found that the PTTSA ranged from 12.2° to 27.8° for males and from 13.3° to 34.8° for females at 0° knee flexion. Regression analysis showed that the PTTSA in females was, on average, 3.6° greater

than males throughout the 0° to 90° knee flexion range and that PTTSA decreased linearly between 0° and 90° knee flexion. Consequently, the smaller the angle of knee flexion, the greater the PTTSA. A number of studies including Boden et al., (2000) and Olsen et al., (2004) have reported non-contact ACL injuries to be associated with actions in which the knee joint is close to full extension. Furthermore, for a given angle of knee flexion and a given force exerted by the patella ligament, the anterior shear component of the patella ligament force will tend to be greater in females than males. Therefore, the PTTSA may contribute a greater risk of ACL injury in females than in males.

Muscle activity pattern:

When providing dynamic stability of the knee, the activity of the knee flexors and extensors should, ideally, result in a zero shear load on the proximal tibia and, therefore, minimal strain on the knee ligaments. However, if the shear load exerted by the quadriceps is greater than the shear load exerted by the hamstrings, a resultant anteriorly directed shear force may be exerted on the proximal end of the tibia, which will increase ACL strain. This is known as quadriceps dominance (Hewett et al., 2001). A number of studies have found that females exhibit greater quadriceps dominance than males in activities associated with ACL injury (Huston and Wojtys, 1996, Malinzak et al., 2001, Zeller et al., 2003).

Muscle stiffness:

As the quadriceps and hamstring muscles contract they act in a way to increase the joint contact forces and limit shear movement within the tibiofemoral joint. The ability of the muscles to resist movement (maintain a particular joint angle) within the tibiofemoral joint refers to muscle stiffness. The greater the ability of the muscles of the knee to prevent tibiofemoral shear movement, the less likely the passive structures of the knee, such as the ACL, will be put under strain. Therefore, muscle stiffness may be an important factor in preventing ACL injury. A number of studies have reported that females exhibit decreased muscle stiffness of the knee compared to males (Granata et al., 2002a, Granata et al., 2002b, Wojtys et al., 2003).

Muscle Strength:

One factor that may influence muscle stiffness is muscle strength. A number of studies have reported that females possess significantly less muscular strength of the hamstrings and quadriceps than males, even when normalised to body weight (Kanehisa et al., 1996, Hakkinen et al., 1997, Salci et al., 2004). This suggests that the lower levels of muscle stiffness exhibited by females compared to males may be due, at least in part, to lower levels of muscle strength (absolute and relative). Lower levels of strength may increase the risk of ACL injury.

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5. AIMS and OBJECTIVES

The purpose of the study is to investigate the relationship between leg strength and the kinematics and kinetics of landing.

The objectives of the study are to 1) assess the isokinetic strength of the hip, knee and ankle muscles, 2) assess the kinematics and kinetics of landing technique and 3) determine the relationship between leg strength and landing technique.

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6. METHODOLOGY

6.1 Study Design

10 female and 10 male recreational/elite athletes (18-30 years of age) without any previous history of hip/knee or ankle injury will participate in the study. Subjects will also be screened for resting blood pressure and those found to have abnormal blood pressure will be excluded from taking part. The subjects will participate in sports involving a lot of jump landings such as volleyball, netball, soccer and basketball. The subjects will be required for testing on two separate days within a week of each other. One testing session will take approximately 1 hour and will involve subjects undertaking maximal exertion isometric strength testing. The other testing session will take up to half an hour and will involve subjects performing maximal vertical jump and landing maneuvers.

6.2 Experimental Procedures

Strength Test.

Isokinetic leg strength will be assessed using a dynamometer. Tests will be carried out for hip flexion/extension and adduction/abduction, knee flexion/extension, and ankle plantarflexion/dorsiflexion and inversion/eversion. Both eccentric and concentric tests will be undertaken for each movement and tests will be carried out on both legs. Each test will include 3 sets of 5 repetitions with 5 minutes rest between sets.

For hip flexion/extension, range of motion will be 0° to 75° flexion and angular velocity will 30°/sec. For hip adduction/abduction, range of motion will be abduction 45° and adduction 25° and angular velocity will be 30°/sec. For knee flexion/extension, range of motion will be 0° to 90° flexion and angular velocity will be 60 and 120°/sec. For ankle dorsiflexion/plantarflexion, range of motion will be plantarflexion 55° to dorsiflexion 30° and angular velocity will be 90°/sec. For ankle inversion/eversion, range of motion will be Eversion 30° to inversion 45° and angular velocity will be 90°/sec.

Landing Test.

Subjects will be required to perform maximal vertical jump and stop-landing manoeuvres. The task will involve subjects running forward three steps then jumping to touch a mark 3.2m high (height of basketball ring) and landing with each foot on a separate force plate. Each subject will perform 3 trials.

Equipment.

The ground reaction force on each foot will be recorded at a frequency of 600Hz. Three-dimensional co-ordinates of 39 retro-reflective markers will be measured using a 12 camera Vicon 512 system sampling at 120Hz. Markers will be placed on the skin or tight fitting clothing over the following anatomical landmarks on both the right and left sides of the body;

- Anterior superior illiac spine (ASIS),
- Posterior superior illiac spine (PSIS),
- Lower lateral surface of the thigh along the line between the hip and knee joint centres (THI),
- Lateral epicondyle of the knee (KNE),
- Lower lateral surface of tibia along the line between knee and ankle joint centres (TIB),
- Lateral malleolus of the ankle (ANK),
- Proximal end of the second metatarsal (TOE),
- Achilles tendon at the same height as the second metatarsal marker (HEE),

The 3-D spatial coordinates of the joint centres of the hip, knee and ankle joints of both legs will be determined from the 3-D spatial coordinates of the markers.

6.3 Data Analysis Techniques

Kinematics: The angle-time histories of the hip (flexion-extension, abduction-adduction, internal rotation-external rotation), knee (flexion-extension, valgus-varus, internal rotation-external rotation) and ankle (plantarflexion-dorsiflexion, inversion-eversion) joints of both legs will be determined for the continuous period comprising the fall prior to landing and the landing period (from initial contact to the instant when the body comes to rest).

Kinetics: The moment-time histories of the hip(flexion-extension, abduction-adduction, internal rotation-external rotation), knee (flexion-extension, valgus-varus, internal rotation-external rotation) and ankle (plantarflexion-dorsiflexion, inversion-eversion) joints of both legs will be determined during the full landing period.

The isokinetic strength of the hip flexors, extensors, adductors and abductors, the knee extensors and flexors and the ankle plantar flexors and dorsi flexors will be assessed.

All measurements will be normalised to body weight. Results of kinematics and kinetic variables will be correlated to hip flexion/extension and adduction/abduction, knee flexion/extension, and ankle plantarflexion/dorsiflexion and inversion/eversion strength to determine how leg strength parameters relate to landing kinetics and kinematics. Also, a multivariate model will be used to analyse the data to determine relationships between variables. Results will be compared between genders and between dominant and non-dominant legs.

6.4 Storage and Disposal of Data and Samples

The data will be stored in personal computer files but the subjects will remain anonymous. The principal applicant will have sole access to and full responsibility for the storage and disposal of data.

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7. LOCATION OF THE PREMISES WHERE THE RESEARCH WILL BE CONDUCTED.

Motion Analysis Laboratory and Exercise Physiology Laboratory, University of Wales Swansea.

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8. SUBJECT RISKS AND DISCOMFORTS

The subjects may experience some discomfort while undertaking strength testing because they will be required to perform maximal exertions. There is also the possibility of raised blood pressure as a result of isokinetic strength testing. Individuals who have a history of serious hip, knee or ankle injury and abnormal blood pressure will be excluded from taking part in the study.

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9. INFORMATION SHEET AND INFORMED CONSENT

Have you included a Subject Information Sheet for the participants of the study? YES

Have you included a Subject Consent Form for the participants of the study? YES

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10. COMPUTERS

Are computers to be used to store data? YES

If so, is the data registered under the Data Protection Act? YES

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11. STUDENT DECLARATION

Please read the following declarations carefully and provide details below of any ways in which your project deviates from them. Having done this, each student listed in section 2 is required to sign where indicated.

1. I have ensured that there will be no active deception of participants.
2. I have ensured that no data will be personally identifiable.
3. I have ensured that no participant should suffer any undue physical or psychological discomfort
4. I certify that there will be no administration of potentially harmful drugs, medicines or foodstuffs.
5. I will obtain written permission from an appropriate authority before recruiting members of any outside institution as participants.
6. I certify that the participants will not experience any potentially unpleasant stimulation or deprivation.
7. I certify that any ethical considerations raised by this proposal have been discussed in detail with my supervisor.
8. I certify that the above statements are true with the following exception(s):

Student signature: [signature] Date: 3/3/05

~~Student signature:~~ ~~Date:~~

---

12. SUPERVISOR'S DECLARATION

In the supervisor's opinion, this project (delete those that do not apply):

- Does not raise any significant issues.
- ~~Raises some ethical issues, but I consider that appropriate steps and precautions have been taken and I have approved the proposal.~~
- ~~Raises ethical issues that need to be considered by the Departmental Ethics Committee.~~

* ~~Raises ethical issues such that it should not be allowed to proceed in its current form.~~

Supervisor's signature: James Watkins Date: 3/3/05

13. ETHICS COMMITTEE DECISION (COMMITTEE USE ONLY)

ETHICAL APPROVAL: GRANTED REJECTED (delete as appropriate)

The ethical issues raised by this project have been considered by members of the Departmental Ethical Approval Committee who made the following comments:

....
....
....
....
....
....
....

Please ensure that you take account of these comments and prepare a revised submission that should be shown to your supervisor/ resubmitted to the Department Ethical Approval Committee (delete as appropriate).

Signed: M. J. Lewis. Date: 3.3.05

(Chair, Departmental Ethics Advisory Committee)

Appendix B: Subject information sheet.

DEPARTMENT OF SPORTS SCIENCE

SUBJECT INFORMATION SHEET

Date :

Contact Details:

Name- Gerwyn Hughes

Address- 28 Brynmill Terrace, Brynmill, Swansea, SA2 0BA.

Telephone number- 07737472670

E-mail- 189895@swan.ac.uk

1. Study title

The relationship between leg strength and landing kinematics and kinetics in court games players.

2. Invitation paragraph

I am inviting you to take part in my study investigating the relationship between leg strength and landing technique. There is little research investigating the relationship between leg strength and landing technique and we aim to provide more understanding between these two factors, which may provide some information for injury prevention. Your involvement would be most important in carrying out this study.

3. What is the purpose of this study?

The purpose of my study is to examine how muscular strength in a variety of lower limb movements relates to landing technique.

4. Why have I been chosen?

The reason you have been chosen to take part in my study is because you have played at a sufficient level in your sport and therefore have the required level of experience in performing landing manoeuvres. Your taking part is completely voluntary and you are free to withdraw from the study at any time you wish, without giving a reason. It is my job as the researcher to keep your well being during the study, and if you have any problems or concerns feel free to contact me.

5. What will happen to me if I take part?

You will be required to perform up to 10 landing manoeuvres in a laboratory while being filmed with markers placed on your body. This will take up approximately half an hour of your time. On a different day, you will be required to undertake isokinetic leg strength testing in a variety of lower limb movements. This procedure should take up approximately one hour of your time.

6. What are the possible disadvantages of taking part?

There are no disadvantages of taking part in the study. You may experience some discomfort during leg strength testing because you will be required to perform maximal exertion. However, there are no safety concerns when undertaking this testing.

7. What are the possible benefits of taking part?

The possible benefits that you will receive are improvements in technique due to being highlighted of some key biomechanical principles that can be applied to your landing technique. You will be provided with information regarding your leg strength which may be of use to your strength training program. Also, your involvement will be contributing to the development of knowledge and understanding of injury prevention.

8. Will my taking part in the study be kept confidential?

Your privacy will be respected at all times and you will remain anonymous throughout the study. The results of the study will be used in my thesis and may be used in a published academic research paper.

Appendix C: Subject consent form.

DEPARTMENT OF SPORTS SCIENCE
SUBJECT CONSENT FORM

Contact Details:

Name- Gerwyn Hughes

Address-[redacted]

Telephone Number-[redacted]

E-mail-[redacted]

Subject Age:

Project Title:
The relationship between leg strength and landing kinematics and kinetics in court games players.

Please initial box

1. I confirm that I have read and understood the information sheet dated ....../....../...... (version number ....) for the above study and have had the opportunity to ask questions. ☐

2. I understand that my participation is voluntary and that I am free to withdraw at any time, without giving any reason, without my medical care or legal rights being affected. ☐

3. I understand that sections of any of data obtained may be looked at by responsible individuals from the University of Wales Swansea or from regulatory authorities where it is relevant to my taking part in research. I give permission for these individuals to have access to these records. ☐

4. I agree to take part in the above study.

Name of Subject	Date	Signature
Name of Person taking consent	Date	Signature
Researcher	Date	Signature

Appendix D: Chapter 2 inanimate static structure raw data.

Table D1. Means (X), standard deviations (SD) and coefficients of variation (CV) of the x, y and z coordinates of markers on the inanimate static structure.

	Trial 1			Trial 2			Trial 3		
Marker 1	x (mm)	y (mm)	z (mm)	x (mm)	y (mm)	z (mm)	x (mm)	y (mm)	z (mm)
X	389.24	744.95	683.41	389.22	744.90	683.38	389.23	744.86	683.36
SD	0.17	0.21	0.12	0.21	0.23	0.10	0.18	0.23	0.11
CV (%)	0.04	0.03	0.02	0.05	0.03	0.01	0.05	0.03	0.02
Marker 2	x (mm)	y (mm)	z (mm)	x (mm)	y (mm)	z (mm)	x (mm)	y (mm)	z (mm)
X	274.16	800.12	467.34	274.17	800.10	467.35	274.11	800.08	467.31
SD	0.17	0.22	0.18	0.15	0.22	0.16	0.15	0.19	0.18
CV (%)	0.06	0.03	0.04	0.05	0.03	0.03	0.05	0.02	0.04
Marker 3	x (mm)	y (mm)	z (mm)	x (mm)	y (mm)	z (mm)	x (mm)	y (mm)	z (mm)
X	274.84	802.36	86.04	274.85	802.41	86.05	274.91	802.42	86.02
SD	0.15	0.16	0.11	0.17	0.18	0.09	0.17	0.19	0.10
CV (%)	0.06	0.02	0.13	0.06	0.02	0.11	0.06	0.02	0.12

Table D2. Means, standard deviations and coefficients of variation of the distances between markers of the inanimate static structure.

	Trial 1	Trial 2	Trial 3
Marker 1-2 (mm)			
X	250.90	250.90	251.00
SD	0.24	0.19	0.21
CV (%)	0.09	0.08	0.08
Marker 2-3 (mm)			
X	381.30	381.30	381.30
SD	0.21	0.19	0.22
CV (%)	0.06	0.05	0.06

Table D3. Means, standard deviations and coefficients of variation of the angle made between markers of the inanimate static structure.

	Trial 1	Trial 2	Trial 3
Angle (°)			
X	30.50	30.50	30.50
SD	0.06	0.05	0.05
CV (%)	0.21	0.16	0.16

Table D4. Means, standard deviations and coefficients of variance of the COP of the inanimate static structure on the force plate.

Trial 1	COP:X	COP:Y
X (mm)	377.61	306.63
SD (mm)	0.68	0.47
CV (%)	0.18	0.15
Trial 2	COP:X	COP:Y
X (mm)	377.49	306.66
SD (mm)	0.73	0.51
CV (%)	0.20	0.17
Trial 3	COP:X	COP:Y
X (mm)	377.53	306.59
SD (mm)	0.75	0.48
CV (%)	0.20	0.16

Appendix E: Chapter 2 subject 1 raw data.

Table E1. Means, standard deviations and coefficients of variance of x, y and z coordinates of calculated joint centres of Subject 1 in session 1.

Session 1

Trail 1	LKNEE:X	LKNEE:Y	LKNEE:Z	LHIP:X	LHIP:Y	LHIP:Z	LANKLE:X	LANKLE:Y	LANKLE:Z
X (mm)	102.81	350.07	538.07	186.16	323.97	944.09	82.38	327.29	59.88
SD (mm)	1.53	0.92	0.21	2.08	1.08	0.20	0.55	0.50	0.99
CV (%)	1.49	0.26	0.04	1.12	0.33	0.02	0.66	0.15	1.66
Trial 2	LKNEE:X	LKNEE:Y	LKNEE:Z	LHIP:X	LHIP:Y	LHIP:Z	LANKLE:X	LANKLE:Y	LANKLE:Z
X (mm)	107.13	359.86	535.55	160.44	345.64	946.65	69.11	331.66	60.04
SD (mm)	0.73	0.63	0.12	2.20	0.70	0.10	0.56	0.26	0.10
CV (%)	0.68	0.18	0.02	1.37	0.20	0.01	0.82	0.08	0.17
Trial 3	LKNEE:X	LKNEE:Y	LKNEE:Z	LHIP:X	LHIP:Y	LHIP:Z	LANKLE:X	LANKLE:Y	LANKLE:Z
X (mm)	96.21	347.93	537.12	164.71	325.15	949.74	61.68	328.80	64.38
SD (mm)	1.06	0.54	0.23	0.68	1.05	0.38	0.82	0.89	1.48
CV (%)	1.10	0.16	0.04	0.41	0.32	0.04	1.33	0.27	2.29
Trail 1	RKNEE:X	RKNEE:Y	RKNEE:Z	RHIP:X	RHIP:Y	RHIP:Z	RANKLE:X	RANKLE:Y	RANKLE:Z
X (mm)	134.23	158.10	517.44	196.29	180.55	943.69	94.08	188.70	67.90
SD (mm)	1.28	0.59	0.29	1.89	1.06	0.14	6.02	2.12	1.12
CV (%)	0.96	0.37	0.06	0.96	0.59	0.01	6.40	1.12	1.64
Trial 2	RKNEE:X	RKNEE:Y	RKNEE:Z	RHIP:X	RHIP:Y	RHIP:Z	RANKLE:X	RANKLE:Y	RANKLE:Z
X (mm)	138.16	184.37	519.48	174.21	202.56	949.65	84.16	201.79	66.03
SD (mm)	1.18	0.34	0.14	2.31	0.71	0.25	6.31	1.60	1.08
CV (%)	0.85	0.19	0.03	1.33	0.35	0.03	7.50	0.80	1.63
Trial 3	RKNEE:X	RKNEE:Y	RKNEE:Z	RHIP:X	RHIP:Y	RHIP:Z	RANKLE:X	RANKLE:Y	RANKLE:Z
X (mm)	121.29	160.82	516.85	177.42	181.95	948.33	79.14	194.87	66.06
SD (mm)	1.39	1.05	0.22	0.78	1.04	0.59	2.78	0.63	0.41
CV (%)	1.15	0.65	0.04	0.44	0.57	0.06	3.52	0.32	0.61

Table E2. Means, standard deviations and coefficients of variance of x, y and z coordinates of calculated joint centres of Subject 1 in session 2.

Session 2

Trial 1	LKNEE:X	LKNEE:Y	LKNEE:Z	LHIP:X	LHIP:Y	LHIP:Z	LANKLE:X	LANKLE:Y	LANKLE:Z
X (mm)	74.38	359.43	509.93	150.01	340.87	961.27	65.94	334.11	58.49
SD (mm)	0.36	0.51	0.11	0.85	0.91	0.31	0.91	0.98	1.03
CV (%)	0.49	0.14	0.02	0.57	0.27	0.03	1.38	0.29	1.77
Trial 2	LKNEE:X	LKNEE:Y	LKNEE:Z	LHIP:X	LHIP:Y	LHIP:Z	LANKLE:X	LANKLE:Y	LANKLE:Z
X (mm)	69.04	367.18	509.23	149.03	351.16	958.59	53.11	335.03	59.30
SD (mm)	0.67	1.24	0.16	2.58	1.18	0.28	0.72	0.84	0.81
CV (%)	0.97	0.34	0.03	1.73	0.34	0.03	1.35	0.25	1.37
Trial 3	LKNEE:X	LKNEE:Y	LKNEE:Z	LHIP:X	LHIP:Y	LHIP:Z	LANKLE:X	LANKLE:Y	LANKLE:Z
X (mm)	60.17	365.43	508.95	134.64	345.98	960.07	57.11	336.94	60.95
SD (mm)	0.98	1.40	0.17	1.11	2.49	0.37	0.53	0.52	0.78
CV (%)	1.63	0.38	0.03	0.83	0.72	0.04	0.94	0.16	1.27
Trial 1	RKNEE:X	RKNEE:Y	RKNEE:Z	RHIP:X	RHIP:Y	RHIP:Z	RANKLE:X	RANKLE:Y	RANKLE:Z
X (mm)	103.22	171.80	515.11	163.78	198.02	958.31	60.07	201.71	70.00
SD (mm)	1.26	0.59	0.24	0.66	0.90	0.23	1.58	0.40	0.48
CV (%)	1.22	0.35	0.05	0.40	0.45	0.02	2.64	0.20	0.68
Trial 2	RKNEE:X	RKNEE:Y	RKNEE:Z	RHIP:X	RHIP:Y	RHIP:Z	RANKLE:X	RANKLE:Y	RANKLE:Z
X (mm)	91.23	177.55	516.64	160.16	208.05	957.99	50.15	203.06	70.66
SD (mm)	0.96	0.56	0.25	2.32	1.17	0.16	1.44	0.56	0.60
CV (%)	1.05	0.32	0.05	1.45	0.56	0.02	2.87	0.28	0.85
Trial 3	RKNEE:X	RKNEE:Y	RKNEE:Z	RHIP:X	RHIP:Y	RHIP:Z	RANKLE:X	RANKLE:Y	RANKLE:Z
X (mm)	91.79	174.35	517.32	147.73	203.05	958.08	57.60	202.36	71.11
SD (mm)	0.99	1.83	0.43	0.55	2.56	0.27	1.93	0.49	0.57
CV (%)	1.07	1.05	0.08	0.37	1.26	0.03	3.35	0.24	0.80

Table E3. Means, standard deviations and coefficients of variance of the distances between calculated joint centres of Subject 1.

Session 1				**Session 2**		
	Trial 1	**Trial 2**	**Trial 3**	**Trial 1**	**Trial 2**	**Trial 3**
RHIP-RKNEE						
X (mm)	431.30	432.10	436.40	448.10	447.70	445.20
SD (mm)	0.38	0.30	0.29	0.23	0.47	0.52
CV (%)	0.09	0.07	0.07	0.05	0.11	0.12
RKNEE-RANKLE						
X (mm)	452.40	457.00	453.40	448.20	448.60	448.40
SD (mm)	1.06	0.52	0.40	0.40	0.70	0.81
CV (%)	0.23	0.12	0.09	0.09	0.16	0.18
LHIP-LKNEE						
X (mm)	415.30	414.80	419.20	458.00	456.70	457.60
SD (mm)	0.23	0.29	0.27	0.23	0.15	0.35
CV (%)	0.06	0.07	0.06	0.05	0.03	0.08
LKNEE-LANKLE						
X (mm)	479.20	477.90	473.50	452.20	451.40	448.90
SD (mm)	1.01	0.16	0.26	0.99	0.82	0.73
CV (%)	0.21	0.03	0.06	0.22	0.18	0.16

Table E4. Means, standard deviations and coefficients of variation of the angle of the knee in the 3D workspace between calculated joint centres of Subject 1.

Session 1				**Session 2**		
	Trial 1	**Trial 2**	**Trial 3**	**Trial 1**	**Trial 2**	**Trial 3**
Right Angle						
X (°)	7.60	5.10	7.40	7.50	8.00	7.80
SD (°)	0.59	0.20	0.14	0.12	0.18	0.27
CV (%)	7.76	3.97	1.85	1.66	2.19	3.40
Left Angle						
X (°)	11.10	6.00	7.80	10.10	10.10	10.80
SD (°)	0.16	0.13	0.13	0.13	0.25	0.22
CV (%)	1.48	2.22	1.70	1.27	2.48	2.00

Table E5. Means, standard deviations and coefficients of variance of the COP of Subject 1 on the force plate.

	Session 1		**Session 2**	
Trial 1	COP:X	COP:Y	COP:X	COP:Y
X (mm)	230.56	269.13	238.16	245.16
SD (mm)	3.04	1.29	1.38	3.66
CV (%)	1.32	0.48	0.58	1.49
Trial 2	COP:X	COP:Y	COP:X	COP:Y
X (mm)	260.97	257.35	230.08	242.97
SD (mm)	3.03	1.59	1.53	2.14
CV (%)	1.16	0.62	0.66	0.88
Trial 3	COP:X	COP:Y	COP:X	COP:Y
X (mm)	255.86	259.71	251.73	246.36
SD (mm)	2.14	1.88	0.54	1.26
CV (%)	0.84	0.73	0.21	0.51

Table E6. Means, standard deviations and coefficients of variance of the angles of the knee of Subject 1 in session 1.

	Session 1			
Trail 1	Left knee flex/ext	Left knee valg/var	Right knee flex/ext	Right knee valg/var
X (°)	-8.87	6.74	-2.34	7.19
SD (°)	0.19	0.15	0.80	0.39
CV (%)	-2.19	2.29	-34.06	5.39
Trial 2	Left knee flex/ext	Left knee valg/var	Right knee flex/ext	Right knee valg/var
X (°)	-2.41	5.54	2.38	4.42
SD (°)	0.20	0.13	0.84	0.30
CV (%)	-8.48	2.42	35.07	6.71
Trial 3	Left knee flex/ext	Left knee valg/var	Right knee flex/ext	Right knee valg/var
X (°)	-4.77	5.85	-1.18	7.28
SD (°)	0.29	0.15	0.50	0.11
CV (%)	-6.03	2.49	-42.47	1.46

Table E7. Means, standard deviations and coefficients of variance of the angle of the knee of Subject 1 in session 2.

	Session 2			
Trial 1	Left knee flex/ext	Left knee valg/var	Right knee flex/ext	Right knee valg/var
X (°)	-6.37	7.84	-1.69	7.33
SD (°)	0.18	0.14	0.37	0.10
CV (%)	-2.79	1.76	-21.66	1.30
Trial 2	Left knee flex/ext	Left knee valg/var	Right knee flex/ext	Right knee valg/var
X (°)	-6.18	8.01	-2.96	7.46
SD (°)	0.21	0.20	0.30	0.13
CV (%)	-3.35	2.46	-10.12	1.69
Trial 3	Left knee flex/ext	Left knee valg/var	Right knee flex/ext	Right knee valg/var
X (°)	-7.05	8.25	-2.25	7.49
SD (°)	0.32	0.11	0.42	0.17
CV (%)	-4.60	1.35	-18.83	2.32

Appendix F: Chapter 2 subject 2 raw data.

Table F1. Means, standard deviations and coefficients of variation of x, y and z coordinates of calculated joint centres of Subject 2 in session 1.

Session 1

Trail 1	LHIP:X	LHIP:Y	LHIP:Z	LKNEE:X	LKNEE:Y	LKNEE:Z	LANKLE:X	LANKLE:Y	LANKLE:Z
X (mm)	198.66	230.39	927.39	238.55	191.36	499.71	305.57	206.51	80.69
SD (mm)	0.88	0.60	0.15	0.47	0.38	0.20	0.53	0.20	0.19
CV (%)	0.44	0.26	0.02	0.20	0.20	0.04	0.17	0.10	0.24
Trial 2	LHIP:X	LHIP:Y	LHIP:Z	LKNEE:X	LKNEE:Y	LKNEE:Z	LANKLE:X	LANKLE:Y	LANKLE:Z
X (mm)	235.31	219.47	929.43	254.21	182.09	500.53	312.30	192.68	80.41
SD (mm)	1.02	1.13	0.33	0.51	1.03	0.19	0.82	0.28	0.25
CV (%)	0.43	0.51	0.04	0.20	0.56	0.04	0.26	0.15	0.31
Trial 3	LHIP:X	LHIP:Y	LHIP:Z	LKNEE:X	LKNEE:Y	LKNEE:Z	LANKLE:X	LANKLE:Y	LANKLE:Z
X (mm)	230.10	216.99	929.28	252.35	179.37	501.33	307.70	194.99	80.18
SD (mm)	0.73	0.57	0.11	0.53	0.44	0.24	0.47	0.17	0.11
CV (%)	0.32	0.26	0.01	0.21	0.24	0.05	0.15	0.09	0.13
Trial 1	RHIP:X	RHIP:Y	RHIP:Z	RKNEE:X	RKNEE:Y	RKNEE:Z	RANKLE:X	RANKLE:Y	RANKLE:Z
X (mm)	197.18	304.87	926.37	230.48	349.80	497.46	280.39	328.66	76.47
SD (mm)	0.87	0.60	0.08	0.44	0.46	0.14	0.27	0.18	0.12
CV (%)	0.44	0.20	0.01	0.19	0.13	0.03	0.10	0.06	0.16
Trial 2	RHIP:X	RHIP:Y	RHIP:Z	RKNEE:X	RKNEE:Y	RKNEE:Z	RANKLE:X	RANKLE:Y	RANKLE:Z
X (mm)	236.69	293.95	927.77	256.21	338.60	497.05	299.56	316.72	76.43
SD (mm)	0.89	1.12	0.18	0.51	0.81	0.38	0.30	0.24	0.12
CV (%)	0.37	0.38	0.02	0.20	0.24	0.08	0.10	0.08	0.16
Trial 3	RHIP:X	RHIP:Y	RHIP:Z	RKNEE:X	RKNEE:Y	RKNEE:Z	RANKLE:X	RANKLE:Y	RANKLE:Z
X (mm)	232.61	291.43	927.20	252.78	340.65	496.89	298.95	321.19	75.87
SD (mm)	0.73	0.57	0.08	0.39	0.39	0.12	0.31	0.25	0.13
CV (%)	0.31	0.20	0.01	0.15	0.11	0.02	0.10	0.08	0.18

Table F2. Means, standard deviations and coefficients of variance of x, y and z coordinates of calculated joint centres of Subject 2 in session 2.

Session 2

Trial 1	LKNEE:X	LKNEE:Y	LKNEE:Z	LHIP:X	LHIP:Y	LHIP:Z	LANKLE:X	LANKLE:Y	LANKLE:Z
X (mm)	241.95	170.18	496.97	209.39	211.97	928.41	304.15	183.81	81.65
SD (mm)	0.74	0.60	0.35	0.43	0.73	0.14	0.64	0.29	0.21
CV (%)	0.31	0.36	0.07	0.20	0.34	0.02	0.21	0.16	0.26
Trial 2	LKNEE:X	LKNEE:Y	LKNEE:Z	LHIP:X	LHIP:Y	LHIP:Z	LANKLE:X	LANKLE:Y	LANKLE:Z
X (mm)	238.31	166.37	496.24	201.55	207.75	926.38	306.00	178.11	80.37
SD (mm)	0.85	0.69	0.21	0.57	0.35	0.17	1.26	0.56	0.29
CV (%)	0.36	0.42	0.04	0.28	0.17	0.02	0.41	0.32	0.36
Trial 3	LKNEE:X	LKNEE:Y	LKNEE:Z	LHIP:X	LHIP:Y	LHIP:Z	LANKLE:X	LANKLE:Y	LANKLE:Z
X (mm)	247.21	167.25	497.13	225.64	205.87	928.51	302.73	182.49	81.70
SD (mm)	0.66	0.37	0.17	0.58	0.35	0.09	0.84	0.21	0.19
CV (%)	0.27	0.22	0.03	0.26	0.17	0.01	0.28	0.12	0.24
Trial 1	RKNEE:X	RKNEE:Y	RKNEE:Z	RHIP:X	RHIP:Y	RHIP: Z	RANKLE:X	RANKLE:Y	RANKLE: Z
X (mm)	239.01	329.73	492.30	209.43	271.73	927.06	283.37	310.25	76.62
SD (mm)	0.60	0.97	0.14	0.39	0.72	0.11	0.29	0.29	0.32
CV (%)	0.25	0.29	0.03	0.19	0.27	0.01	0.10	0.09	0.41
Trial 2	RKNEE:X	RKNEE:Y	RKNEE:Z	RHIP:X	RHIP:Y	RHIP: Z	RANKLE:X	RANKLE:Y	RANKLE: Z
X (mm)	241.35	324.55	491.42	202.29	267.51	924.97	291.28	303.39	76.30
SD (mm)	0.54	0.41	0.10	0.60	0.35	0.28	0.28	0.20	0.20
CV (%)	0.22	0.13	0.02	0.30	0.13	0.03	0.10	0.07	0.27
Trial 3	RKNEE:X	RKNEE:Y	RKNEE:Z	RHIP:X	RHIP:Y	RHIP: Z	RANKLE:X	RANKLE:Y	RANKLE: Z
X (mm)	250.13	326.06	491.96	228.46	265.55	926.49	295.20	309.14	76.16
SD (mm)	0.64	0.37	0.10	0.62	0.35	0.09	0.30	0.23	0.15
CV (%)	0.26	0.11	0.02	0.27	0.13	0.01	0.10	0.07	0.19

Table F3. Means, standard deviations and coefficients of variation of the distances between calculated joint centres of Subject 2.

Session 1				Session 2		
	Trial 1	**Trial 2**	**Trial 3**	**Trial 1**	**Trial 2**	**Trial 3**
RHIP-RKNEE						
X (mm)	433.10	433.40	433.60	439.60	439.20	439.30
SD (mm)	0.25	0.23	0.24	0.19	0.36	0.19
CV (%)	0.06	0.05	0.06	0.04	0.08	0.04
RKNEE-RANKLE						
X (mm)	424.50	423.90	424.10	418.50	418.60	418.80
SD (mm)	0.17	0.19	0.19	0.34	0.24	0.22
CV (%)	0.04	0.04	0.05	0.08	0.06	0.05
LHIP-LKNEE						
X (mm)	431.20	431.10	430.20	434.90	433.70	433.70
SD (mm)	0.24	0.18	0.19	0.44	0.17	0.29
CV (%)	0.06	0.04	0.04	0.10	0.04	0.07
LKNEE-LANKLE						
X (mm)	425.10	424.30	425.00	420.20	421.50	419.40
SD (mm)	0.28	0.27	0.25	0.38	0.37	0.25
CV (%)	0.07	0.06	0.06	0.09	0.09	0.06

Table F4. Means, standard deviations and coefficients of variation of the angle of the knee in the 3D workspace between calculated joint centres of Subject 2.

	Session 1			Session 2		
	Trial 1	Trial 2	Trial 3	Trial 1	Trial 2	Trial 3
Right Angle						
X (°)	9.20	9.40	9.90	10.40	10.50	10.60
SD (°)	0.08	0.06	0.07	0.20	0.10	0.09
CV (%)	0.84	0.64	0.73	1.89	0.93	0.82
Left Angle						
X (°)	8.00	8.30	8.50	8.50	8.30	8.40
SD (°)	0.09	0.11	0.12	0.12	0.16	0.10
CV (%)	1.15	1.32	1.41	1.40	1.87	1.19

Table F5. Means, standard deviations and coefficients of variance of the COP of Subject 2 on the force plate.

	Session 1		Session 2	
Trial 1	COP:X	COP:Y	COP:X	COP:Y
X (mm)	145.77	254.75	100.72	266.38
SD (mm)	5.68	2.46	0.87	2.59
CV (%)	3.90	0.96	0.86	0.97
Trial 2	COP:X	COP:Y	COP:X	COP:Y
X (mm)	123.09	277.92	97.63	278.87
SD (mm)	3.04	2.33	3.69	1.77
CV (%)	2.47	0.84	3.78	0.63
Trial 3	COP:X	COP:Y	COP:X	COP:Y
X (mm)	123.05	254.21	88.22	273.72
SD (mm)	1.98	2.96	5.67	3.95
CV (%)	1.61	1.16	6.43	1.44

Table F6. Means, standard deviations and coefficients of variance of the angles of the knee of Subject 2 in session 1.

	Session 1			
Trail 1	Left knee flex/ext	Left knee valg/var	Right knee flex/ext	Right knee valg/var
X (°)	3.66	7.29	4.16	8.12
SD (°)	0.17	0.11	0.11	0.11
CV (%)	4.67	1.47	2.66	1.40
Trial 2	Left knee flex/ext	Left knee valg/var	Right knee flex/ext	Right knee valg/var
X (°)	4.99	6.69	5.50	7.71
SD (°)	0.31	0.15	0.11	0.12
CV (%)	6.18	2.30	1.98	1.50
Trial 3	Left knee flex/ext	Left knee valg/var	Right knee flex/ext	Right knee valg/var
X (°)	4.34	7.23	5.88	7.88
SD (°)	0.10	0.09	0.14	0.10
CV (%)	2.32	1.23	2.45	1.25

Table F7. Means, standard deviations and coefficients of variance of the angles of the knee of Subject 2 in session 2.

	Session 2			
Trial 1	Left knee flex/ext	Left knee valg/var	Right knee flex/ext	Right knee valg/var
X (°)	3.92	7.53	5.03	9.21
SD (°)	0.21	0.10	0.19	0.18
CV (%)	5.44	1.28	3.80	1.92
Trial 2	Left knee flex/ext	Left knee valg/var	Right knee flex/ext	Right knee valg/var
X (°)	3.95	7.31	4.96	9.27
SD (°)	0.29	0.19	0.14	0.14
CV (%)	7.42	2.64	2.73	1.50
Trial 3	Left knee flex/ext	Left knee valg/var	Right knee flex/ext	Right knee valg/var
X (°)	4.20	7.53	6.53	8.57
SD (°)	0.18	0.12	0.17	0.12
CV (%)	4.31	1.60	2.63	1.36