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THE EFFECT OF TRUNK ORIENTATION UPON KNEE JOINT LOADING DURING UNANTICIPATED SIDESTEP CUTTING MANOEUVRES.

by

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Abstract

The purpose of this investigation was to examine the relationship between sagittal and frontal plane trunk orientation and the most commonly accused attributors to anterior cruciate ligament injury. Twenty-one male (mean age 31 years; height 1.8 metres; weight 75 kilograms) with at least 10 years soccer experience, participated in this repeated measures study of experimental design. Forty-four reflective markers were attached to specific landmarks to identify the lower limb and trunk segments. Three-Dimensional and force plate data were recorded for 5 successful trials. Participants were instructed to achieve an approach speed of 3m/s, facilitated by timing gates, complete a right-foot contact on the force plate and execute a sidestep cutting manoeuvre, responding to the movements of the defender. The independent variable tested was trunk orientation (sagittal and frontal plane). Peak values of 7 selected dependent variables were subjected to a two-tailed Spearman's rho correlation co-efficient test. A bonferroni-corrected p value of 0.007 was used to denote statistical significance. No significant correlations were found to exist. Certain patterns emerged from the results. Subjects who exhibited knee valgus moments also exhibited a posteriorly directed anterior-posterior ground reaction force, whereas some subjects exhibited an anteriorly directed anterior-posterior ground reaction force and exhibited knee varus loading. The results of this study suggest natural trunk movements during sidestep cutting manoeuvres has no relationship with the selected dependent variables in the confines of this experiment.

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1.0 Introduction

The knee is the major weight bearing joint and is located between two of the body's longest bones (Bartlett & Bussey, 2012 pp55). The knee comprises of a complex system of ligaments that bind the tibia and femur together, the joint capsule and tendons (Powers, 2010). This structure makes the knee highly susceptible to soft tissue injury through different planes of motion.

The main function of the ACL is the restraint of anterior shear force (Fleming et al., 2001). Anterior shear force acts at the proximal end of the tibia and is as a result of posterior ground reaction forces (force generated by the foot in the backward direction) following ground contact, and quadriceps muscle contraction (Yu & Garrett, 2007; Li et al., 1999). This causes anterior tibial translation (Fleming et al., 2001), which means the tibia moves anteriorly with respect to the femur, forcing the distal end of the femur and the proximal end of the tibia apart, which strains the ACL. The most widely believed cause of ACL injury is knee valgus loading (Quatman & Hewett, 2009; Cochrane et al., 2007; Boden et al., 2000). Valgus refers to the outward angulation of the distal segment. This can occur due to a pure abduction of the tibia with respect to the femur or as a result of concomitant internal and external rotations of the femur and tibia respectively (Quatman & Hewett, 2009). The opposite of which is varus loading (Cochrane et al., 2009). Internal rotation involves the knee rotating towards the body's midline, the opposite of which is external rotation (Cochrane et al., 2009). The anatomy of the knee joint and supporting structures makes it susceptible to different types of loading. Certain activities and/or movements can exacerbate loading mechanisms.

Soccer is the most popular sport in the world, with over 240 million participants (Wong & Hong, 2005). The incidence of soccer related injuries in adult male players is estimated at 10-35 per 1000 playing hours, with the most severe and debilitating injuries occurring in the lower extremities (Agel et al., 2007). Of the lower extremity joints, the knee sustains the greatest proportion of injuries (Powers, 2010). The anterior cruciate ligament (ACL) is the most important, yet the most frequently injured (Fleming et al., 2001).

McLean et al. (2005) reported that there are approximately 100,000 ACL injuries per year in the United States, with a treatment bill estimated at $1.7 billion. Similarly Cochrane et al. (2007), on a study of Australian Rules football reported an estimated cost of knee injuries to be A$100 million. Bartlett and Bussey (2012 pp55) stated that 32.6% of all reported sporting injuries involved the knee, and the most common activity leading to injury was soccer (35%). Ishida et al. (2012) noted that not only is there a treatment cost, but also the cost of time away from the sport for professional athletes and the burden that places on the team, as well as time away from work for non-professional athletes.

Quatman and Hewett (2009) and Lohmander et al. (2007) reported that Osteoarthritis develops in up to 50% of all ACL injuries, 10 to 20 years post injury. Caraffa et al (1996) stated that ACL injuries produced the longest disability time, are the most costly and lead to the highest percentage of permanent disability than any other injury. Similarly Bartlett and Bussey (2012 pp56) wrote that the ACL suffers the most frequent disruption, and once ruptured up to 80% of all patients lose rotational stability at the knee.

The most common cause of ACL injury is as a result of a non-contact episode. Non-contact episodes account for 70% of all reported ACL injuries (Agel et al., 2007; Boden et al., 2000; Miyaska et al. as cited by Quatman & Hewett, 2009) and occur from high-risk sporting manoeuvres, such as, deceleration, landing and pivoting or sidestep cutting manoeuvres that do not involve contact with another player (Yu & Garrett, 2007; Boden et al., 2000). These types of high risk sporting manoeuvre are synonymous with dynamic, invasion type games such as soccer (Brophy et al., 2010). Non-contact ACL injuries occur due to the person themselves generating excessive loads about the knee (Yu & Garrett, 2007). The most common cause of ACL injury is from a sidestep cutting manoeuvre (Koga et al., 2010), where a player rapidly changes direction to avoid contact with another player, or to follow the path of a moving ball (Besier et al., 2001a).

The amount of deceleration required to perform a sidestep cutting manoeuvre is dependent upon the approach speed and the angle at which cutting is required (Vanrenterghem et al., 2012), and as such, higher approach speeds and greater cutting angles increase the amount of deceleration required to perform the manoeuvre and ultimately increase the chance of ACL injury (Vanrentergham et al., 2012).

There have been many studies investigating knee loading and ACL injury, yet no conclusive evidence exists as to why, or indeed what is happening to cause ACL injury. In fact, there is actually conflicting evidence surrounding this area, meaning any ACL injury prevention programs employed by athletes and coaches may in fact be ineffective. Many factors are believed to be contributors to ACL injury, as during movements the knee can move through all 3 planes of motion, as well as being susceptible to external forces. For example, Boden et al. (2000) identify knee valgus collapse as the major cause of ACL injury, whereas, Berns et al. (as cited in Fleming et al., 2001) found that knee valgus moments had no effect on ACL strain. Valgus collapse is where the tibia rotates externally against the internal rotation of the femur (Quatman & Hewett, 2009), creating a twist movement, loading the ACL perpendicular to the ligament fibres. However, mechanically, the primary function of the ACL is restraint of anterior shear forces (Markolf et al. as cited in Fleming et al., 2001), and it is in fact the primary function of the medial collateral ligament (MCL) to restrain knee valgus loading and internal rotation during weight acceptance (Besier et al., 2001a). Furthermore, Mazzocca et al. (as cited in Yu & Garrett, 2007) demonstrated minimal ACL strain under knee valgus with MCL intact, and went on to say that ACL rupture due to knee valgus collapse cannot occur without a complete rupture of the MCL first. Concomitant ACL and MCL rupture is in fact rare, Fayed et al. (as cited in Yu & Garrett, 2007) showed that only 5 out of 84 ACL injuries examined were accompanied by MCL rupture, and Miyaska et al. (as cited in Quatman & Hewett, 2009) identified a less than 30% concomitant incidence rate.

The primary function of the ACL is anterior shear force restraint (Fleming et al., 2001). Anterior shear force acts perpendicular to the ACL at the proximal end of the tibia, causing an anterior translation of the tibia with respect to the femur (Kulas et al., 2010), and is a product of landing forces and muscular contraction (Blackburn & Padua, 2009). Anterior shear force is affected by; small knee flexion angles, strong quadriceps muscle contraction and great posterior ground reaction forces (GRF) (Yu & Garrett, 2007).

Quadriceps muscle force is believed to be a major contributor to anterior shear force as the muscles act through the patella tendon at the proximal end of the tibia (Yu & Garrett, 2007). As such, much of the research has centred on the relationship between quadriceps muscle force and ACL rupture (DeMorat et al., 2004; Beynnon et al., 1995; Durselen et al., 1995). However, McLean et al. (2004b) stated that the quadriceps muscle alone cannot generate a large enough force to rupture the ACL. Therefore, this suggests that there must be another contributing factor. That is not to say that the quadriceps muscle force is not related, as Yu and Garrett (2007) reported that small knee flexion angles, posterior GRF, as well as a strong quadriceps muscle force can all increase ACL loading. The latter two are inextricably linked as greater posterior GRF induces a greater knee flexion moment, which is countered by a greater quadriceps muscle force (Blackburn & Padua, 2009).

Most studies identify posterior GRF as a factor of ACL injury as solely inducing a quadriceps contraction to counter the subsequent flexion moment occurring at the knee on touchdown (Blackburn & Padua, 2009; Yu & Garrett, 2007). These studies fail to acknowledge the equal and opposite reactions transmitted up the shank creating a resultant force acting across the ACL proceeding ground contact. This force has the potential to create an anterior force at the proximal end of the tibia, as well as an equal and opposite force at the distal end of the femur (Appendix 1; Figure 1). It is believed (Kulas et al., 2010) that posterior shear force reduces ACL strain. This is because posterior shear force is viewed only as a result of hamstrings muscle contraction acting at the proximal end of the tibia, as this is where the hamstrings attach. This therefore resists anterior tibial translation (Kulas et al., 2010). However, shear force at the distal end of the femur, as a result of posterior GRF at touchdown, and not as a result of hamstrings muscle contraction, and therefore without the counterbalancing effect of the hamstrings against the quadriceps, added to anterior shear force produced by the quadriceps during the weight acceptance phase to counter the flexion moment generated as a result of the forward and downward movement of the body on touchdown, could create a greater net force acting across the ACL, straining the ACL further, producing potentially injurious amounts of loading. Highlighting the important influence of posterior GRF. However, many early studies excluded these interactions as the studies were performed on cadaveric knees.

Much of the initial research pertaining to this injury has utilized cadaveric knees (DeMorat et al., 2004; Markolf et al. as cited in Fleming et al., 2001; Li et al., 1999; and Durselen et al., 1995), non-weight bearing situations (Beynnon et al., 1995), simulated loading (Fleming et al., 2001), or in cohort studies (Cochrane et al., 2007; Hewett et al., 2005) utilizing available video footage and post-injury questionnaires.

This could explain why so many different theories exist, or why much of the evidence is conflicting. While these studies can provide important structural and mechanical properties of the ACL, the relevance of these types of study must be questioned, in particular cadaver and non-weight bearing studies, as no study has ever acknowledged or reported an ACL injury in a non-weight bearing knee. In fact, it is thought that weight bearing itself may have a protective response by reducing anterior-posterior laxity and stiffening the tibiofemoral joint, which stabilizes the knee by forcing the tibiofemoral surfaces together (Fleming et al., 2001).

With this in mind, the most recent research (Blackburn & Padua, 2009; Sigward & Powers, 2006; McLean et al., 2004a; McLean et al., 2004b; Pollard et al., 2004; Besier et al., 2001a) has utilized live trials in which a subject performs under specific conditions, measuring GRF and joint angle changes. However, whilst many studies have used live weight bearing trials to assess the ACL in different situations, many studies do not replicate the game like scenarios in which many of these injuries occur. Sigward and Powers (2006), McLean et al. (2005), McLean et al. (2004a) and Besier et al. (2001a) all employed a pre-planned trial system, where the subject knew which direction to sidestep cut from the force plate before the trial began. Besier et al. (2001b) highlighted the difference between pre-planned trials and trials that were unanticipated, where the subject was signalled on approach to tell them which direction to leave the force plate. The study found significant changes resulting in the authors to hypothesise that postural changes were implemented on approach for the pre-planned trials as the subjects prepared for the direction of the manoeuvre. Unanticipated experiments are facilitated by the addition of either a crossover option (Jamison et al. 2012; Besier et al. 2001b) or a straight run through (Pollard et al. 2004). The subject would be signalled on approach by a lighting system, pointing the subject in either direction (Jamison et al., 2012; Pollard et al., 2004).

Mclean et al. (2004a), highlighted the need to replicate game like situations to obtain truer results, and as the sidestep cutting manoeuvre is primarily an evasive manoeuvre, a defender was incorporated. Their study found significant increases in peak medial GRF, hip flexion, hip abduction, knee flexion and knee valgus when compared to trials with no defender. The defender applies a physical stimulus to the trial, an opponent for the subject to evade. This again highlights the importance of replicating the game like scenarios in which these types of injuries occur, as the real loading patterns that occur in these movements are significantly different from simple laboratory based representations.

Another explanation for the conflicting evidence surrounding ACL injury maybe due to the almost exclusive focus on the changes in the knee joint biomechanics during sporting manoeuvres. McLean et al. (2004a) recognised that excessive loading may be as a result of abnormal neuromuscular control elsewhere in the lower extremity. As the knee is apart of a multi-segmented kinetic chain (Kulas et al., 2010) it should not be analysed as a single unit. Much of the research fails to recognise the effect the upper body can have on loading characteristics of the lower extremities. The trunk segment accounts for 36% of the body’s total mass (Vanrenterghem et al., 2010), which is the greatest proportion of mass concentrated in one segment. Therefore, the trunk can influence the location of the body’s centre of mass (CM), and the orientation can determine where the resultant GRF vector will act, influencing the moment arm (Powers, 2010) and therefore affecting how the knee will load. For example, a lateral lean away from the body's midline can change the vertical GRF vector, causing it to pass either side of the knee joint centre, resulting in a rotational moment (Powers, 2010).

Blackburn and Padua (2009) examined the effects of sagittal plane trunk position on landing forces in landing tasks. The study found that landing with the trunk flexed reduced landing forces and quadriceps activity in comparison to a trunk-extended landing. A prospective study by Hewett et al. (2005) demonstrated that subjects who sustained an ACL injury exhibited 20% greater landing forces than those that did not. Kulas et al. (2010) also demonstrated interactions between trunk position and changes in knee loading, specifically changes in knee anterior shear force, quadriceps and hamstrings muscle forces, during landing tasks. These studies start to highlight an interdependent relationship between the upper and lower extremities. Jamison et al. (2012) examined the relationship between knee moments and lateral trunk position during sidestep cutting manoeuvres. The study found a significant positive association between lateral lean away from the direction of the cut and knee abduction moments. This will increase strain on the ACL as lateral causes the GRF vector to pass lateral to the knee joint centre. This will increase the moment arm increasing the knee abduction moment (Jamison et al., 2012; Powers, 2010).

Therefore, the purpose of this study is to investigate the relationship between trunk orientation and knee joint loading during unanticipated sidestep cutting manoeuvres typically experienced in soccer. To make the movement more realistic a defensive opponent will be used to facilitate the sidestep manoeuvre. The first hypothesis is that there will be a strong correlation between trunk lean in the sagittal plane and posterior GRF, as forward trunk rotation has been shown to increase posterior GRF at touchdown. The second hypothesis is that there will be a strong correlation between frontal plane trunk angulation and valgus/varus knee loading, as lateral leaning, shifts the vertical GRF vector, which can change both type and severity of loading.

2.0 Review of Literature

2.1 Knee Loading and ACL injuries

Many theories exist as to the causes of ACL injuries during sidestep cutting manoeuvres. However, most of the common explanations are not without doubts and conflict. For example, the most widely believed cause of ACL injury is knee valgus loading (Quatman & Hewett, 2009; Cochrane et al. 2007; Boden et al., 2000). Markolf et al. (as cited in Yu & Garrett, 2007) demonstrated increased strain when a 10Nm knee valgus moment was applied to the ACL, but only when applied with 100N anterior shear force. The same study demonstrated increased ACL strain when 10Nm knee varus and internal rotation moments were applied with a 100N anterior shear force. However, structurally the ACL does not offer restraint to this type of movement (Fleming et al., 2001). Berns et al. (as cited in Yu & Garrett, 2007) suggested that varus/valgus loading did not strain the ACL. However, this study only applied pure loads to the ACL, combinations of loads were not used. Also, both studies were done on cadaveric knees and therefore were non-weight bearing. Fleming et al. (2001) demonstrated increased strain on the ACL when a varus/valgus moment was applied to the ACL under weight bearing conditions when compared to non-weight bearing knees ($p<0.01$). However, ACL strain remained constant, even when varus/valgus moments were increased (4% increase in ACL strain across all loads when transitioned from non-weight bearing to weight bearing), leading the authors to suggest that the ACL is not a restraint to varus/valgus loading ($p = 0.32$). However, as with the study by Berns et al. (as cited in Yu & Garrett, 2007), Fleming et al.

(2001) measured ACL strain by measuring the change in length of the anteromedial bundle. Hollis et al. (as cited in Quatman & Hewett, 2009) demonstrated a fourfold greater change in length of the posterolateral bundle compared with the anteromedial bundle when a varus/valgus load was applied. Zantop et al. (2007), by applying a 10Nm valgus load with a 134N anterior tibial load, demonstrated that the posterolateral bundle offers greater resistance to anterior tibial translation at 0° to 30° knee flexion, the range experienced in sidestep cutting manoeuvres (Cochrane et al., 2007), as well as rotation movements near full extension. This suggests that the true nature of ACL loading could have been under quantified in both studies, as both studies examined ACL strain between 0° and 30° knee flexion.

Both Berns et al. (as cited in Yu & Garrett, 2007) and Markolf et al. (as cited in Fleming et al., 2001) went on to demonstrate greater ACL strain when a valgus load was applied to the knee along with anterior shear force. However, discrepancies arise as Fleming et al. (2001) identified internal rotation moments as adding strain to the ACL, whereas, Markolf et al. (as cited in Fleming et al., 2001) demonstrated increased ACL strain when an internal rotation moment was added to the ACL but only when applied with anterior shear force. Markolf et al. (as cited in Fleming et al., 2001) also demonstrated a decrease in ACL strain when an internal rotation moment was replaced by an external rotation moment, even to the point that ACL strain was less with a combination of external rotation moment and anterior shear force than with anterior shear force alone. This is contrary to the belief of knee valgus collapse being the major cause of ACL injury as external rotation is an element of knee valgus collapse (Markolf et al. as cited in Fleming et al., 2001).

There is clear evidence to indicate the multi-planar nature of knee loading, which suggests that characterising knee loading, and therefore knee injury as a single-plane incident could be a gross oversimplification. Whilst evidence indicates that sagittal plane loading (anterior shear force) is the main protagonist to ACL injury, it also demonstrates that as anterior shear force is applied to the ACL, it becomes susceptible to strain through other planes of motion, strain that would otherwise have no loading effect on the ACL. The same can be said about weight bearing knees.

Past research suggested that weight bearing can provide stability to the knee joint. Barger et al. (as cited in Fleming et al., 2001) and Hseish and Walker (as cited in Fleming et al., 2001) believed that compressive loads, as a result of weight bearing, actually provide a protective function for the knee by forcing the tibiofemoral surfaces together. That being said, no ACL injuries occur in non-weight bearing situations. Even though the ACL of a non-weight bearing knee is strained by anterior shear force and internal torques (Fleming et al., 2001). In fact, past research, as outlined above, seems to suggest the contrary, that the ACL is actually at greater risk of injury during weight bearing, as the ACL becomes susceptible to different types of loading, loading that does not strain the ACL in non-weight bearing situations. This is increasingly evident as many studies suggest that with the knee near full extension, and therefore experiencing higher compressive forces, susceptibility of the ACL to injury is greater (Blackburn & Padua, 2009; Fleming et al., 2001; Beynnon et al. 1995; Durselen et al. 1995). However, Barger et al. (as cited in Fleming et al., 2001) and Hseish and Walker (as cited in Fleming et al., 2001) only take in to account the affects about the knee, failing to account for the impact of the proximal body segments, and the influence this can have on the protective function of the surrounding musculature.

In a study of pre-planned sidestep cutting manoeuvres with female soccer players, Sigward and Powers (2007) demonstrated a correlation between knee valgus and increased hip abduction angle (P= 0.002; effect size 0.79) and increased hip internal rotation angle (p= 0.008; effect size 0.71). However, in a similar study McLean et al. (2004a) demonstrated that whilst females displayed larger peak knee valgus, they in fact exhibit smaller peak hip flexion, hip abduction, hip internal rotation, knee flexion angles and knee internal rotation angles than males ($p < 0.003$). In the study by Sigward and Powers (2007) the participants (aged 14 to 18) where split into groups, normal frontal plane moment and excessive knee valgus moment groups. Excessive knee valgus was defined as being greater than 0.59 Nm/Kg-Bwt, however this was determined from data from a previous study on male subjects.

Similarly conflicting evidence exists as to whether the quadriceps can themselves generate a large enough force to rupture the ACL, with Markolf et al. (as cited in Quatman & Hewett, 2009) suggesting they can, whereas, McLean et al. (2004a) suggest that they cannot. No study, to the authors knowledge, provides critical loads that can rupture the ACL with many just quoting ‘excessive’ loads rupture the ACL (Powers, 2010; Quatman & Hewett, 2009; Yu & Garrett, 2007). DeMorat et al. (2004) loaded the ACL with a

4500N quadriceps muscle force. The study identified 3 complete and 3 partial tears out of 11 cadaveric knees tested. However, the loading was applied in a non-weight bearing scenario, and as such the results are limited.

Quadriceps muscle force is influenced by posterior GRF (Powers, 2010). Posterior GRF induces a flexion moment at the knee, which is countered by the quadriceps creating an extension moment (Yu & Garrett, 2007). The greater the posterior GRF, the greater the flexion moment and, the greater the countering extension moment needs to be (Yu & Garrett, 2007). Increased quadriceps muscle force increases anterior shear force as the quadriceps act through the patella tendon that attaches at the proximal end of the tibia, causing anterior tibial translation which strains the ACL (Kulas et al., 2010). Li et al. (1999) found that concomitant quadriceps and hamstring contraction reduced anterior tibial translation and therefore anterior shear force, at all knee flexion angles except 0-15° knee flexion. The authors believe this to be due to the fact that the hamstrings lose mechanical advantage near full extension. Other studies (DeMorat et al. 2004; Markolf et al. as cited in Fleming et al., 2001) have reported decreases in ACL strain with increased knee flexion (greater than 30°), as well as significantly lower ACL strain with simultaneous quadriceps and hamstrings contraction (Beynnon et al., 1995). However, Sigward and Powers (2006) and Besier et al. (2001a) both demonstrate knee flexion angles of approximately 30° during the weight acceptance phase when performing sidestep cutting manoeuvres, and Cochrane et al. (2007) reported knee flexion angles of less than 30° in 91.7% of ACL injuries analysed in the study. This suggests a limited involvement of the hamstrings during sidestep cutting manoeuvres, leaving the knee susceptible to anterior tibial translation.

In a 2001 study simulating the effect of weight-bearing and external loading on ACL strain, Fleming et al. discovered that a application of a pure posterior shear force significantly strained the ACL in weight-bearing conditions across all posterior loads, whereas anterior shear forces only significantly strained the ACL with loads less than 40N. This is contrary to other available research (Kulas et al., 2010). However, the Kulas et al. (2010) study only attributes posterior shear force as a product of hamstring muscle contraction, and it is therefore seen as an ACL strain-reliever. Li et al., (1999) and Beynnon et al. (1995) both report decreases ACL strain with increased hamstrings muscle contraction, as this counters anterior tibial translation. As posterior shear force is a product of hamstring contraction, and is opposite in direction to anterior shear force it is believed to reduce ACL strain (Kulas et al., 2010). This is due to the fact that the hamstrings also act at the proximal end of the tibia, much like the quadriceps and can therefore, oppose quadriceps muscle force. However, the posterior GRF at touchdown will create a joint reaction force equal in magnitude and opposite in direction at proximal and distal ends of the segments from the sole of the foot. This will create a resultant force in the anterior direction at the proximal end of the tibia, and an equal and opposite force at the distal end of the femur. As such, this will increase anterior tibia translation by forcing the proximal end of the tibia and the distal end of the femur further apart, thus applying a posterior shear force to the ACL which will be unaided by a hamstrings muscle contraction, as hamstrings influence is limited at the small knee flexion angles exhibited in sidestep cutting manoeuvres (Cochrane et al., 2007; Sigward & Powers, 2006; Li et al., 1999). Therefore, this has the potential, when added to the anterior shear force created by the quadriceps muscle contraction to counter knee flexion moment experienced upon touchdown, to increase the shear force across the ACL. However, to the author's knowledge only the study by Fleming et al. (2001) applies a posterior shear force to the ACL and as a result is the only study that considers the effects of a pure posterior shear force and its effects on ACL loading.

Whilst many researchers agree that involvement of the hamstrings, and therefore quadriceps-hamstrings co-contraction is a good thing as it is a protective influence (Li et al., 1999; Beynnon et al., 1995; Durselen et al., 1995), Sigward and Powers (2006b) stated that this is predominantly a protective response exhibited by novice participants. Their study demonstrated a negative relationship between number of years experience and quadriceps-hamstrings co-contraction ($r = -0.32$; $p = 0.04$). Whilst not a strong correlation, the study demonstrated decreased hamstrings involvement in experienced subjects. The study also identified significantly lower peak knee joint moments ($p = 0.03$) and net knee joint moment impulse ($p = 0.03$) for novice athletes when compared to their experienced counterparts. However, the researchers failed to account for multiple comparisons when setting the p value and as such, are susceptible to type 1 errors.

Much like Fleming et al. (2001) much of previous research has exclusively focussed on the knee joint. This could provide possible explanations for the misinterpretations of past experiments. Sigward and Powers (2006); McLean et al. (2004a); Besier et al. (2001a); Besier et al. (2001b) are all examples of studies that focus exclusively on changes in knee loading as a result of sidestep cutting manoeuvres, and attribute kinematic changes to the demands of the task, without first considering other mechanical changes the body must implement to execute the task. The knee is apart of a multi-segment kinetic chain including the lower extremities, the pelvis and trunk segments (Kulas et al., 2010), and therefore highlights the potential of other segments to impact the knee. Powers (2010) highlights the interdependency between the hip and the knee joint, by demonstrating how kinematic changes about the hip have direct consequences on the knee joint, as the two joints have a common segment, the femur. Powers (2010) states that during the weight acceptance phase in running the hip adducts, flexes and internally rotates. Excessive amounts cause a medial shift in the knee joint centre with respect to the foot. As the foot is planted on the ground and supporting the weight of the body it remains fixed, this causes the tibia to abduct and the foot to pronate, resulting in dynamic knee valgus. This highlights the interdependent nature of the lower extremities. Powers (2010) also highlighted how a posterior trunk lean in gait increases the knee flexion moment, which increases the demand on the quadriceps. This could increase anterior shear forces, placing greater strain on the ACL. Whilst the study highlights how excessive amounts of loading can bring about unwanted changes and lead to injury, no values are quoted for what is deemed excessive loading. And whilst Powers (2010) highlights the relationship between the hip and knee joints, and demonstrates how weaknesses in one can impact the other, the study fails to fully extrapolate that notion into the upper extremities, mainly the largest of the body's segment, the trunk.

2.2 Impact of the Trunk

Many studies in this area omit the movements of the trunk during sidestep cutting manoeuvres. The trunk consists of 36% of the total body mass (Vanrenterghem et al., 2012), and is part of a complex, multi-segmented kinetic chain involving the lower extremities, pelvis and trunk segments (Kulas et al., 2010). This suggests that the biomechanical interaction of segments can contribute to the overall risk of ACL injury (McLean et al., 2004a). The trunk can impact the centre of pressure on contact with the ground and thus, effect where the resultant GRF vector will act with respect to the knee (Powers, 2010). This in turn will dictate the direction and magnitude of the moment (Powers, 2010) and therefore the type and amount of loading the knee experiences. For example, lateral lean, away from the body's midline over the stance foot at touchdown, will shift the centre of mass outside the base of support. This creates a valgus moment (Appendix 1: Figure 2) about the knee, as the force vector passes lateral to the knee joint centre (Powers, 2010). External moments are resisted by the muscles, ligaments and the knee joint capsule (Powers, 2010). Therefore, any changes of this nature have the ability to influence knee joint loading and can cause injury.

2.2.1 Trunk Orientation

Kulas et al. (2010) demonstrated how trunk orientation could directly influence knee loading. In a study examining interactions between trunk-load, trunk-position and the consequent adaptive changes in the lower extremities in jump-landing tasks, Kulas et al. (2010) found that changes in alignment of proximal body segments during landing impacted knee joint loading. For example, the correlation between trunk-flexion angle and hamstrings impulse was 0.8 ($R^2 = 0.64$, $p < 0.001$). This means that as trunk flexion increased, hamstrings impulse also increased. Hamstrings have been shown to decrease anterior shear force generated by the quadriceps (Li et al., 1999) and therefore reducing risk of ACL injury. Results from the study also identified that employment of landing strategies had a direct effect on knee loading (Appendix 1: Figure 3), reporting a marked difference in anterior shear force between trunk-extended, and trunk-flexed groups, during the weight acceptance phase. However, the Kulas et al. (2010) study was based on landing tasks and not sidestep cutting manoeuvres, and therefore requires the complete deceleration of vertically falling mass and as such the loading mechanics vary. For example, typical knee flexion angles experienced in sidestep cutting manoeuvres are up to 30° (Cochrane et al., 2007), whereas Kulas et al. (2010) report a range of motion about the knee of 53° ± 13° (trunk-extended) and 54° ± 11° (trunk-flexed group) following impact, allowing a greater range of motion about the knee to decelerate the falling mass.

Small knee flexion angles are widely believed to increase ACL loading (Fleming et al. 2001; Beynnon et al. 1995; Durselen et al. 1995), and trunk angle has been shown to directly affect knee flexion angles (Blackburn & Padua, 2009). Beynnon et al. (1995) demonstrated a reduction in ACL strain with increased knee flexion angles, stating a 36% decrease in ACL strain at 15° knee flexion but an 85% decrease at 30°. In 2009 Blackburn and Padua demonstrated a mean increase of 22° in knee flexion angles on landing with a flexed trunk. The authors believe this flexed landing posture improves the lower extremity's ability to attenuate the landing forces, and as quadriceps force requirement is reduced, decreasing the risk of ACL injury. The results show a decrease in posterior GRF and quadriceps EMG amplitude ($p < 0.001$) with a flexed trunk on landing. This is concurrent with the findings of Li et al. (1999) who demonstrated increased involvement of the hamstrings with greater knee flexion angles. However, Kulas et al. (2010) demonstrated similar knee flexion angles across trunk-flexed and trunk-extended groups, which led the authors to suggest that knee flexion angle had no bearing on ACL loading in that particular study and changes in loading were as a result of the added trunk load. However, changes in trunk angle cause changes to the force vector moment arm, due to movements of the CM (Appendix 1: Figure 4). This is an oversight by Kulas et al. (2010). The results of their study also show that whilst average hamstrings muscle force increased in the trunk-flexed group (no load), average anterior shear force decreased in the trunk-extended group (no load). This is contrary to the finding by Li et al. (1999). The increased average hamstrings muscle force could be as a result of increases in hamstrings activity as the hamstrings contract eccentrically to facilitate flexion of the trunk. This was not addressed in the study.

Quatman and Hewett (2009) stated that approximately 85% of the knee joints restraint to anterior tibial translation, which is caused by anterior shear force, is provided by the ACL at 20-30° knee flexion. This potentially highlights an oversight in the 2010 study by Kulas et al. as the changes in trunk orientation with the added load may limit lower extremity kinematics responses, muscle activity and the consequent changes in anterior shear force. These results could indicate a potential ‘danger zone’ in knee flexion with changes in trunk orientation.

Kinematic changes can also cause changes to the external forces which the body experiences during ground contact.

2.2.2 Trunk kinetics

Extended trunk postures on landing are believed to increase GRFs (Yu et al., 2006). Increased GRFs increase external knee flexion moments (McNitt-Gray, 1993), which initiates an increase in quadriceps muscle force to resist the movement (Blackburn & Padua, 2009). Increases in quadriceps muscle force cause increases in anterior shear force (Kulas et al. 2010) through the patella tendon, which causes anterior tibial translation, which increases the risk of ACL injury. Hewett et al. (2005), in a prospective study, found that individuals who sustained ACL injury had 20% greater landing forces than their counterparts. However, this study was specific to females. Conversely, the reverse is believed to be true. Lower GRFs are associated with lower quadriceps muscle force requirements, therefore, potentially reducing the risk of ACL injury (Blackburn & Padua, 2009). They identified that active trunk flexion during jump-landing tasks decreases vertical GRF ($p=0.001$) as well as decreasing quadriceps EMG amplitude ($p=0.001$), which has the potential to decrease the risk of ACL injury. The authors believe that a flexed landing posture improves the body’s ability to attenuate forces.

Most of the studies on trunk kinematics and the kinetic affects at the knee are specifically done on jump-landing (Kulas et al., 2010; Powers, 2010; Blackburn & Padua, 2009) or stop-landing task (Yu et al., 2006). Few studies, to the author's knowledge, exist examining the effects of trunk orientation and sidestep cutting manoeuvres. One such study by Jamison et al. (2012) examined trunk orientation in the frontal plane during run-to-cut manoeuvres. This study collected force and 3D kinematic data from 45° unanticipated sidestep cutting manoeuvres at a self-selected pace, to determine whether trunk control was correlated with knee loading. The authors hypothesised that increased trunk movement, away from the direction of the sidestep cutting manoeuvre, would have a significant positive association with knee abduction and internal rotation moments. The results of the study identify a positive association with knee abduction moment ($p=0.002$), suggesting that lateral lean away from the direction of the sidestep cutting manoeuvre increases knee abduction moments, which could potentially increase loading on the ACL and cause injury. The study also identified that the same lean decreases internal rotation moments ($p=0.021$), which led the authors to suggest that this would decrease ACL load and the risk of ACL injury. Internal rotation moments have been shown to increase ACL strain under weight bearing conditions (Fleming et al., 2001) and when applied with anterior shear force (Markolf et al. as cited in Fleming et al., 2001). A lateral lean away from the body's midline causes the vertical GRF vector to pass lateral to the knee joint centre (Jamison et al., 2012; Powers, 2010), which will create a valgus moment about the knee joint (Powers, 2010). Limitations of the study by Jamison et al. (2012) include: failure to account for multiple comparisons when setting the p-value and as a consequence are susceptible to type 1 errors; recruitment of both male and female subjects, yet made no allowance for gender, and no gender comparisons were made; level of participants was not defined; and trunk motion analysis was limited to one plane of motion, failing to consider anterior and posterior movements of the trunk and the consequent effects on knee loading. These may have influenced the results as research demonstrates differences in ACL loading between males and females (Brophy et al., 2010; Powers, 2010; Quatman & Hewett, 2009; Sigward & Powers, 2006; McLean et al., 2004a; Pollard et al., 2004). Sigward and Powers (2006) demonstrate significant differences in key sagittal and frontal plane moments, with males exhibiting greater peak knee flexor moments than females (p=0.025), and larger net joint impulse at the knee (p=0.01) during early deceleration. Whereas females demonstrate greater knee valgus with greater peak adductor moment (0.005) during weight acceptance. Also, the subjects used in the study where described as experienced collegiate athletes, although one subject could not identify leg dominance, which must question level of experience. Pollard et al. (2004) highlighted the need for experienced subjects to perform these sport specific movements, as familiarity with the task is essential to provide accurate results. Consequently, with recruiting athletes with at least 10 years soccer experience Pollard et al. (2004) demonstrated no gender differences in peak knee adduction moment during weight acceptance. The Pollard et al. (2004) study only identified gender differences in frontal plane hip motion. However, whilst mechanically, hip adduction can contribute to knee abduction (Pollard et al., 2004) no gender differences occurred. Again this study failed to incorporate the trunk segment into the analysis, and also omitted the main contributor to ACL, anterior shear force.

A prospective study by Zazulak et al. (2007) found that trunk displacement was greater in athletes who experienced ACL injuries when compared with those that did not (p=0.05). The study identified lateral trunk displacement as the strongest predictor of ACL injury (p=0.009). However, the study suggested the predictors were stronger predictors of ACL injury in females (91% accuracy) than males. Whilst highlighting the relationship between trunk movement and ACL injury, the study failed to identify the root cause. Also, the perturbation test employed in the study, which involved the subject isometrically pulling against a cable until the cable was released, measured proceeding trunk deviations. Whilst it is believed to be a good representation of a dynamic task (Jamison et al., 2012) it does mean that the results could be due to the different fitness levels of the different participants. No base level of fitness was identified, and the subjects were a selection of collegiate athletes, with no sport specified.

2.3 Game Like Scenarios

In order to fully understand the mechanics of the knee, and therefore the mechanisms that lead to ACL injury, it is important to analyse the movements that are attributed to causing the injuries (McLean et al., 2004a), and how the movements effect the body’s segments, limbs, joints and supporting structures, in the context of which they are most commonly injured. Therefore, any experiment designed to test these injuries should replicate the game situation as close as possible.

2.3.1 Pre-planned versus Unanticipated

A sidestep cutting manoeuvre involves a sudden deceleration and a rapid change in direction away from the support leg (Cochrane et al., 2007). It is the movement away from the support leg that characterises the movement as a sidestep cut, the opposite of which is a crossover manoeuvre (Cochrane et al., 2007). The difference in movements creates different loading patterns and as such, in some early studies (McLean et al., 2005; McLean et al., 2004a; Besier et al., 2001a), the direction of the movement was known to the subject in advance. This knowledge led to the subjects planning the movement in advance. Besier et al. (2001b) highlighted the presence of specific postural changes (changes in position of CM, altered muscle activation and changes in reflex muscle activation) when movements are anticipated, as the subject prepares themselves for the movement. McLean et al. (2004a) whilst stressing the need for unanticipated trials and the need for experiments to be more game like, implemented pre-planned trials in the experiment. Besier et al. (2001b) performed a similar investigation to their 2001a study, but also examined the differences between pre-planned and unanticipated sidestep cutting manoeuvres by comparing joint loading. Besier et al. (2001b) found that varus/valgus and internal/external rotation moments were up to twice the magnitude during the unanticipated trials when compared to the pre-planned trials, demonstrating a 129% increase in internal rotation moment and a 12.3 times increase in external valgus moment at weight acceptance during the unanticipated trial, when executing a 30° sidestep cutting manoeuvre. Unfortunately, the study didn't incorporate other key factors of ACL injury such as anterior shear force, posterior GRF, quadriceps muscle force or knee flexion angle for further comparison. However, the results are indicative enough to suggest that all experiments wishing to further understand the mechanics of ACL injury, specifically during sidestep cutting manoeuvres, must implement unanticipated trials.

2.3.2 Defensive opponent

With the increasing need to make experiments as much like game situations as possible McLean et al. (2004a) introduced a defensive opponent. A plastic skeleton was used to replicate a defender, and placed 20cm behind the force platform. The study was designed to compare knee joint loading with and without the defensive opponent whilst performing pre-planned sidestep cutting manoeuvres. McLean et al. (2004a) noted an increase in deceleration and concomitant increase in knee joint rotations during the weight acceptance phase. This is more indicative of game situations, as a sidestep cutting manoeuvre is most commonly employed to avoid a collision with another player (McLean et al., 2004a). The study found increases in peak medial GRF, hip flexion, hip abduction, knee flexion and knee valgus ($p=0.003$, alpha level adjusted for multiple comparisons). Whlst peak medial GRF, hip flexion, hip abduction and knee valgus will all put the knee in a sub-optimal position for loading and increase the risk of ACL injury, increases In knee flexion have been shown to reduce the risk of injury (Li et al., 1999). However, the study only recruited 'active' subjects for the experiment, not task specific subjects/athletes from a sport that utilises this specific movement, which calls into question the accuracy of the results. Pollard et al. (2004) and Besier et al. (2001a) highlight the importance of using subjects who are familiar with the type of movement involved in order to generate accurate data.

In 2010 Boros and Plumlee incorporated a live defensive opponent. The defensive opponent also added an unanticipated nature to the trial as the defender would move slightly left or right when the subject was 2m away from the force platform and the subject would cut or crossover in the opposite direction. However, this study was investigating ankle biomechanics with reference to ankle bracing, therefore no knee loading results were obtained.

Most of the other studies in this area, whilst being based around dynamic, invasion type games such as soccer (Brophy et al., 2010; Sigward & Powers, 2006) Australian rules football (Cochrane et al., 2007) basketball (Xie et al., 2012) fail to incorporate a defensive opponent nor acknowledge the importance of replicating the game situation as best as possible to provide more accurate results to aid in the understanding of this type of injury.

Whilst there is conflicting evidence as to the causes of ACL injury from a sidestep cutting manoeuvre, many studies fail to replicate the injurious movement within laboratory trials, which may lead to researchers concluding different or incomplete results. The influence of trunk orientation on knee loading during sidestep cutting manoeuvres is scarce, and the existing evidence has specific failings in the experimental parameters and therefore fails to provide a full and clear picture as to the involvement of the trunk, and how differences in trunk orientations during sidestep cutting manoeuvres may result in different types, as well as different magnitudes of knee loading.

3.0 Method

3.1 Participants

Twenty one men (mean age 31 years; height 1.8 metres; weight 75 kilograms) with at least 10 years soccer experience (Pollard et al., 2004) participated in this study. The sample size was calculated by the G Power sample size calculation software (Erdfelder et al., 1996) to achieve an effect size of 0.8 (Yu et al., 2006) with a significance level of 0.05 and a 0.8 power level (Yu et al., 2006; McLean et al., 2004a). All subjects where healthy, with no previous ACL injury, no lower extremity surgery, neurological disorder, chronic lower extremity injury or acute lower extremity injury within the past 12 months. All subjects were members of a soccer team, and where recruited by the author following training sessions. These guidelines were set in accordance with the previous research by Blackburn and Padua (2009) and Besier et al. (2001a).

Ethical approval was granted o the 16th August 2013, by Dr. Stephen Fallows, Chair of the Faculty Research Ethics Committee.

Informed consent was obtained, by the author, prior to any testing procedures, the subject had read the participant information sheet (Appendix ??) and had any questioned answered by the author.

3.2 Design

The design of this study was a true repeated measures experimental design. All subjects where required to familiarise themselves with the task prior to the commencement of testing. This was to ensure that the subject achieved the required approach speed, and contacted the force platform in the centre of the plate, with the right foot.

The independent variable was trunk orientation. The dependant variables that where tested were; posterior GRF (measured using a Kistler force platform), knee varus/valgus moment, knee internal/external rotation moment, knee flexion/extension angle, knee flexion/extension moment, hip adduction/abduction moment and hip internal/external rotation moment (were all calculated through the Visual 3D system).

3.3 Procedures

3.3.1 Pilot Study

Following a pilot study, changes to the original set-up had to be made. From the research an approach speed of 4 m/s (±0.4) was chosen as this provides an adequate degree of difficulty, whilst minimising the risk of injury (Vanrenterghem et al., 2012). However, due to the spatial restrictions in the laboratory, this approach speed was not achievable. The approach speed was changed to 3 m/s (±0.3). This approach speed is still inline with previous research (McLean et al., 2004a; Besier et al. 2001b).

3.3.2 Lower Limb Trunk Body Model

The Lower Limb Trunk (LLT) model outlined by Vanrenterghem et al. (2010) is based on a 6 degrees of freedom model and incorporates Functional Hip and Knee axes, obtained from specific recordings to define their location. This model is employed as it provides CM data whilst being focussed on lower limb motion, providing accurate knee joint kinematics. The model utilises 44 markers which include: Anterior Sacral Iliac Crest; Posterior Sacral Iliac Crest; Iliac Crest; Greater Trochanter; upper leg cluster; Medial and Lateral

Epicondyles; lower leg cluster; Medial and Lateral Maleolus; and 1st and 5th metatarsal heads for both left and right sides, the model also incorporates markers on the C7 and T8 vertebrae, Sternum, Xiphoid Process and left and right Acromion.

3.3.3 Testing

All testing took place in the biomechanics laboratory at the University of Chester. Subjects were required to wear tight fitting shorts, running trainers and perform the task shirtless. The recording space was calibrated using the calibration frame and wand prior to each recording session.

Figure 5 (Appendix 1) is a representation of the experimental set-up. An approach speed of 3 m/s (±0.3) (Besier et al., 2001b), facilitated by timing gates, and a cutting angle of 45° (Vanrenterghem et al., 2012; Boros & Plumlee, 2010; Sigward & Powers, 2006) were selected for this experiment. Data collection incorporated synchronised Qualisys Track Manager (QTM) 3D system (Qualisys AB, Gothenburg, Sweden) comprising 6 cameras and sampling at 240Hz, and a Kistler force platform sampling at 960Hz.

Following the subjects usual pre-match warm up, the subjects familiarised themselves with the task, and calculated the correct run-up, to achieve the desired approach speed and minimise ‘targeting’ of the force platform (Vanrenterghem et al., 2012). Forty-four reflective markers were fitted according to the LLT body model outlined by Vanrenterghem et al. (2010) and a static trial was taken. Following the static trial, 10 markers were removed, as per Vanrenterghem et al. (2010), for the dynamic trails. Functional joint trials were then administered for the left and right hip, and the left and right knee.

This method creates a functional joint, as the computer algorithm searches for a stationary point, as one segment moves relative to the other (Schwartz & Rozumalski, as cited in Vanrenterghem et al., 2010). The subject was required to stand within the 3D camera system and perform the movements whilst maintaining a rate of one cycle per second. To create a functional knee axis the subject was required to flex and extend the knee joint through approximately 20° for 5 seconds (Vanrenterghem et al., 2010). For the hip the subject was required to move the hip through approximately 30° range of motion in all three planes of motion, for 10 seconds (Vanrenterghem et al., 2010).

The defensive opponent was included to make the trials more realistic (McLean et al., 2004a), and served to make the trial unanticipated (Boros & Plumlee, 2010). The defender was situated behind the force platform, in relation to the oncoming subject (McLean et al., 2004a).

The subjected was instructed to run towards the force plate and react to movements of the defender in an evasive manner, moving the opposite way to the defender, once contact was made with the defender. The defender was instructed to make a move left or right as the subject approached the penultimate stride (Boros & Plumlee, 2010). The defender was instructed to randomise movements. A taped line indicting a 45° sidestep cut was on the floor as a guide for the subject when sidestep cutting (Sigward & Powers, 2006) but not for the crossover option. The subject had to complete 5 successful (deemed by approach speed and cutting angle) sidestep cutting manoeuvre trials. If at any time the trial was unsuccessful the subject was informed, so that adjustments could be made.

3.4 Data Processing and Reduction

All recorded data were labelled and tracked in QTM (Qualisys AB, Gothenburg, Sweden) in accordance with the LLT model outlined by Vanrenterghem et al. (2010). Kinematic and force data were exported from the QTM software (Qualisys AB, Gothenburg, Sweden) and imported into Visual 3D 3-Dimensional biomechanical modelling and calculation software program (version 5; C-Motion Inc, Rockville, MD). A model consisting of the lower extremities, pelvis and trunk segments was constructed from the marker data, in accordance with the LLT model outlined by Vanrenterghem et al. (2010). All raw data (kinematic and kinetic) were subjected to a fourth order Butterworth low-pass filter, with a cut-off frequency of 20Hz (Vanrenterghem et al., 2010). All data were normalised to 100% of the stance phase. The stance phase was defined as the period from initial right foot contact to toe-off, as determined by the vertical GRF from the force plate recordings.

3.5 Statistical Analysis

The purpose of this study was to determine the relationship between trunk orientation and posterior GRF, knee varus/valgus moment, knee internal/external rotation moment, knee flexion/extension angle, knee flexion/extension moment, hip adduction/abduction moment and hip internal/external rotation moment. Having not met the criteria for parametric tests in the Shapiro-Wilk test of normality (Appendix 2; Table 1), data were subjected to a two-tailed Spearman's rho correlation co-efficient.

SPSS (version 21; SPSS Inc, Chicago, IL) was used for data analysis. The Bonferroni correction was applied to the p value to reduce the risk of type 1 errors due to the study having multiple comparisons. As such the p value was calculated at 0.007.

4.0 Results

All peak (mean of all trials) and standard deviation data for all tested variables, for all subjects, are presented in Table 2 (Appendix 2).

The results of the analysis demonstrate no significant correlation between sagittal plane trunk angulation and posterior GRF (r_s = 0.069; n = 21; p = 0.767). Analysis also demonstrated no significant correlation between sagittal plane trunk angulation and knee flexion angle (r_s = -0.218; n = 21; p = 0.342) and knee flexion moment (r_s = 0.019; n=21; p = 0.933), leading to the rejection of the first hypothesis. In order to test the second hypothesis four variables were tested and correlated with frontal plane trunk angulation. Peak knee internal/external rotation moment (r_s = 0.449; n = 21; p = 0.041), peak hip abduction/adduction moment (r_s = -0.410; n = 21; p = 0.065), peak hip internal/external rotation moment (r_s = - 0.560; n = 21; p = 0.008) and peak knee valgus/varus moment (r_s = -0.434; n = 21; p = 0.049), whilst all demonstrating a weak correlation, were all insignificant correlations. Therefore, the second hypothesis was also rejected.

Whilst being insignificant, peak knee flexion angle is an interesting result (Figure 6). Mean peak knee flexion angle of all subjects from all trials is 55° (± 6.2). This is a substantial change from previously quoted values for knee flexion whilst performing sidestep cutting manoeuvres.

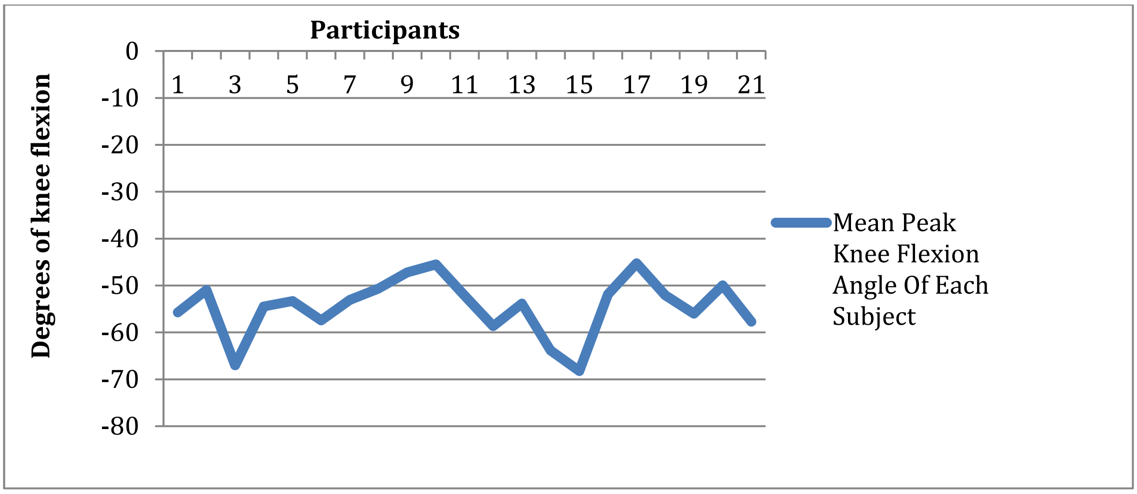

Figure 6. Mean peak knee flexion of all 21 subjects.

Similarly, only 5 out of all 21 subjects demonstrated a knee valgus moment (Figure 7) during the weight acceptance phase. Knee valgus loading is deemed to be one of the main causes of ACL injury (Cochrane et al., 2007). All subjects displaying knee valgus moments following touchdown also exhibited knee internal rotation moments (Figure 8) and hip adduction moments (Figure 9) through the weight acceptance phase. The converse is true of the remaining subjects. The remaining subjects demonstrated knee varus, knee external rotation and hip abduction moments.

The results also show that not all subjects exhibit a knee extension moment following touchdown as would be expected (Figure 10). Six subjects exhibit a flexion moment following touchdown indicating an active knee flexion moment. Furthermore, all subjects demonstrating a knee flexion moment exhibit an anteriorly directed force during the same period (Figure 11), rather than the posteriorly directed force as expected during the weight acceptance phase. This demonstrates two distinct loading patterns in the anterior/posterior GRF trace (Figure 11). Whilst six subjects exhibit a large propulsive force following touchdown, the other 15 subjects have large braking phases following touchdown, which is what is expected.

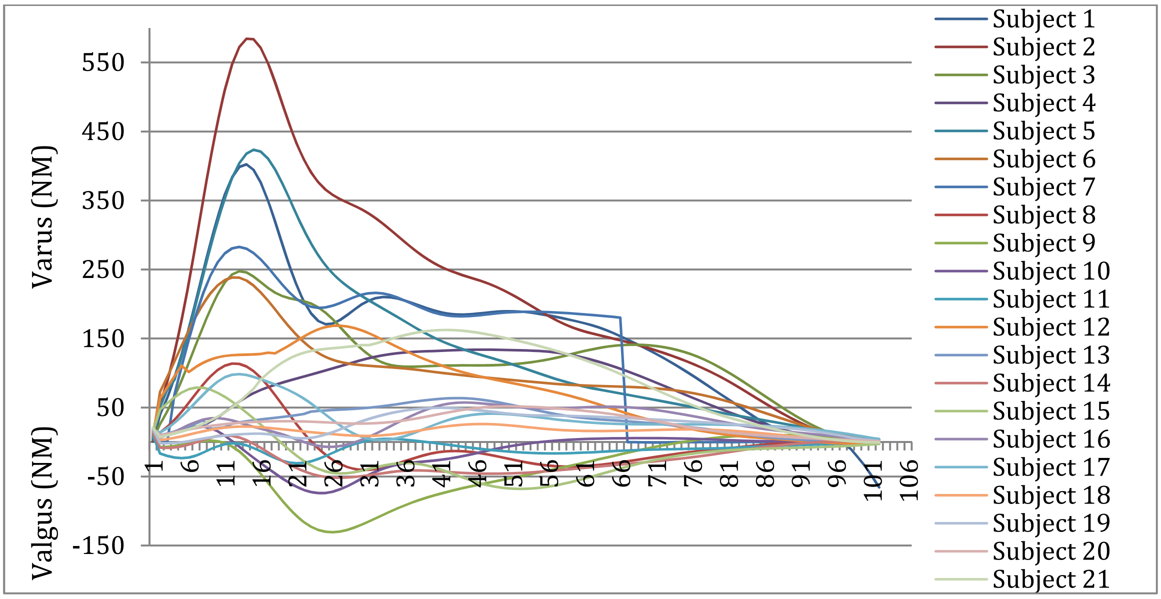

Figure 7. Mean knee valgus(-)/varus(+) moments of all 21 subjects from touchdown to takeoff. Only 5 subjects (9, 10. 11, 14 and 15) demonstrate knee valgus loading whilst performing an unanticipated sidestep cutting manoeuvre. All data normalised to 100% of the stance phase. Weight acceptance phase is up to point 25.

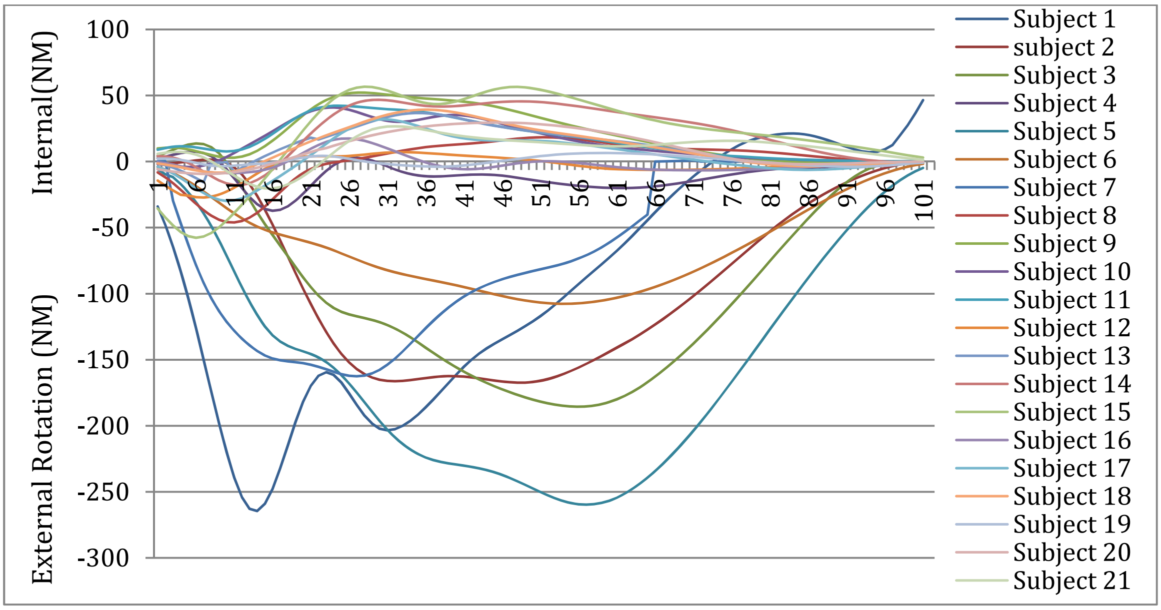

Figure 8. Mean Knee internal/external rotation moments for touchdown to takepff, (-) denotes external rotation and (+) denotes internal rotation moments. Peak knee external rotation moments much greater than the peak knee internal rotation moments. Data normalised to 100% of the stance phase. Weight acceptance phase is up to point 25.

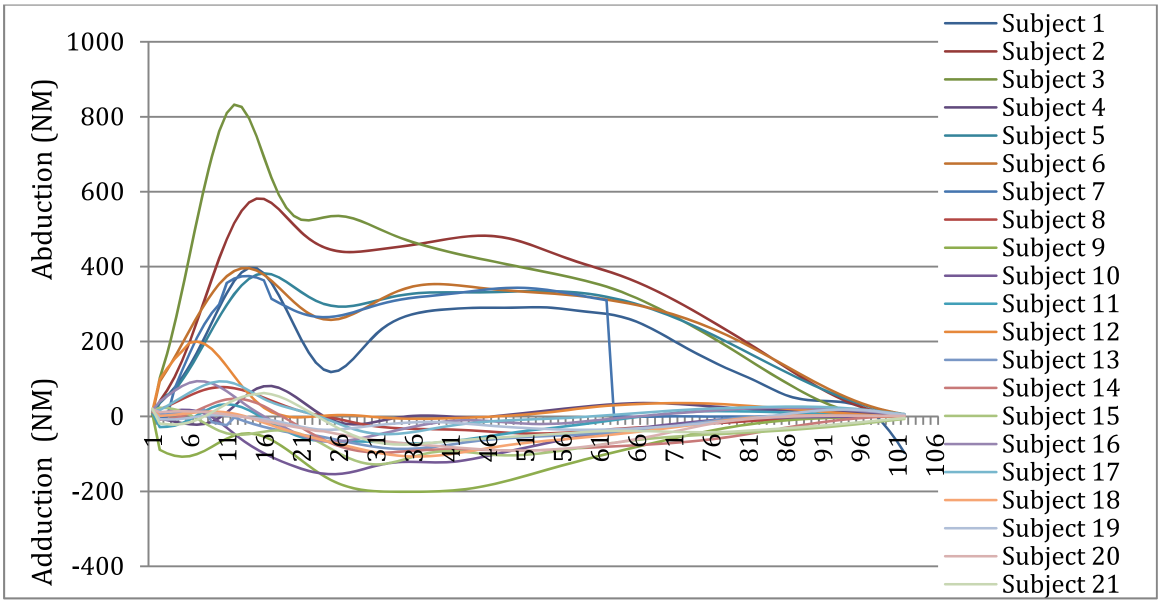

Figure 9. Mean peak hip abduction (-)/adduction (+) rotation moments from touchdown to takeoff. Most subjects exhibit an adduction moment, however a minority to exhibit a abduction moment. Data normalised to 100% of the stance phase. Weight acceptance phase is to point number 25.

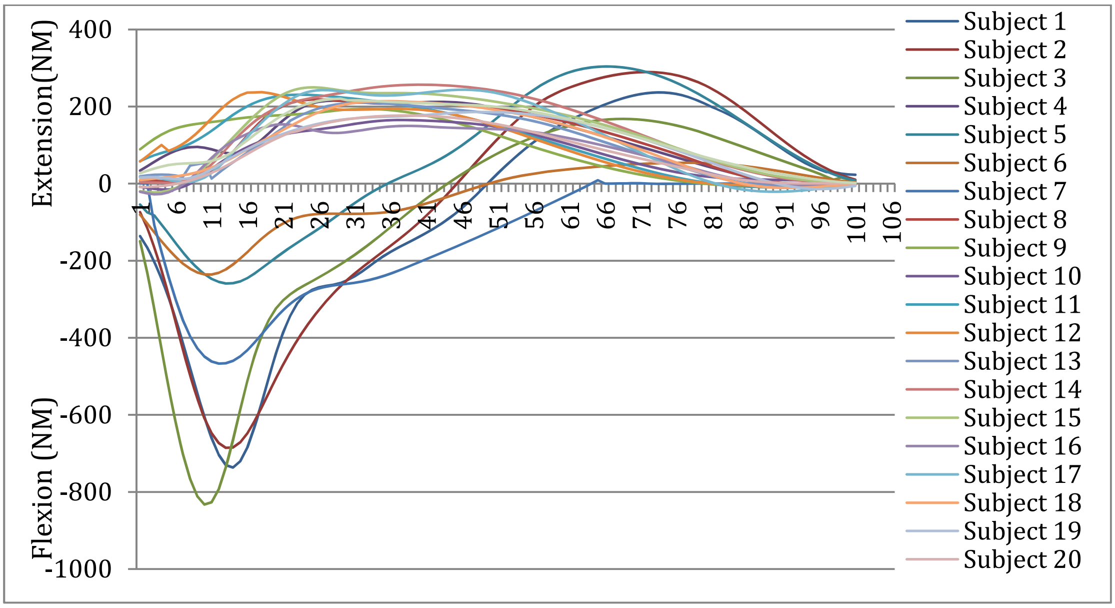

Figure 10. Mean knee flexion/extension moment. All subjects were expected to exhibit knee extension moments to counter the external flexion moment crested by the posterior GRF at touchdown. All data normalised to 100% stance phase. Weight acceptance phase ends at point 25.

Although differing in degree, all subjects exhibit an initial right trunk lean (away from the direction of the sidestep cutting manoeuvre) upon touchdown (Figure 12). Whilst rotating left, towards the direction of the sidestep cutting manoeuvre, all but 4 subjects (9, 13, 18 & 19) keep a rightward lean. The remaining 4 transition to a left lean before take-off. All subjects exhibit different degrees of anterior-posterior trunk lean (-1.886 ± 18.314) throughout the entire stance phase (Figure 13). Subject 12, who exhibits the second highest peak force of 1.42 Bwt (Appendix 2; Table 3), also demonstrates one of the highest peak anterior trunk leans during the weight acceptance phase (20.2°), as well as demonstrating the greatest sagittal plane trunk movement during the weight acceptance phase, a 19.5° change (-0.72° at touchdown to 20.2° at peak sagittal plane trunk angle). However, this is seemingly an isolated event.

Hip internal/external rotation moment is the closest to a significant comparison in this study. Frontal plane trunk angle and hip internal/external rotation moment yielded a comparison of $r_s = -0.560$; $n = 21$; $p = 0.008$. This is only just insignificant due to the p value being set at 0.007 to account for multiple comparisons. This would also represent the strongest correlation of the study between peak hip internal/external rotation moment and frontal plane trunk angle. All the subjects exhibiting an internal rotation moment following touchdown all produced an anteriorly directed force during the same phase (Figure 14). This together with the knee flexion moment could be an adaptive response, as none of these subjects exhibit a knee valgus moment.

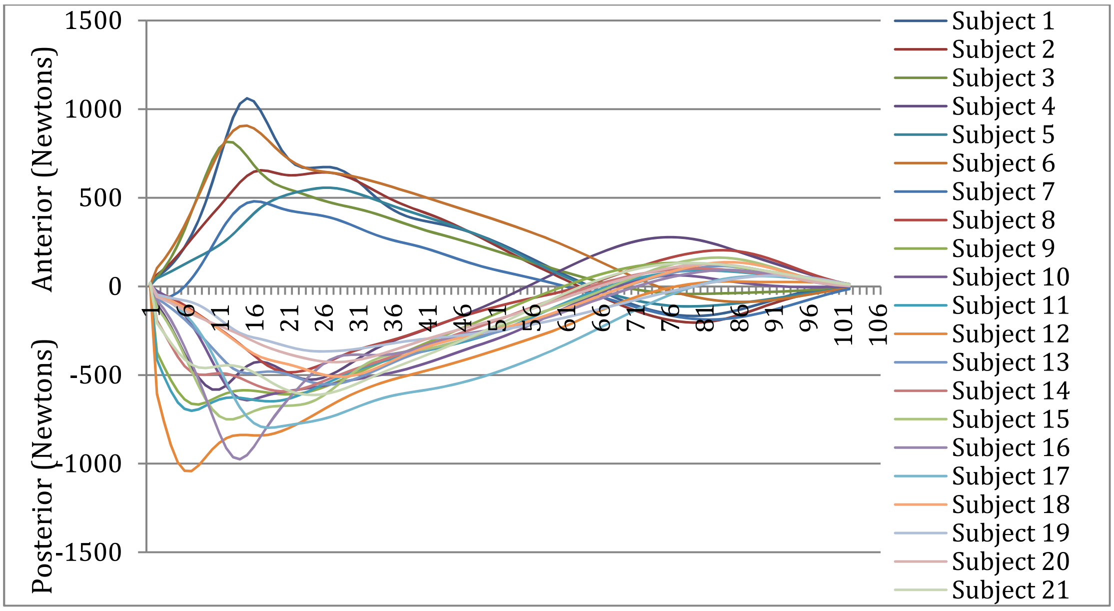

Figure 11. Mean Anterior/Posterior GRF. 6 subjects exhibit an anterior directed force, rather than the posteriorly directed force normally seen. All data normalised to 100% of the stance phase. Point 25 denotes the end of the weight acceptance phase.

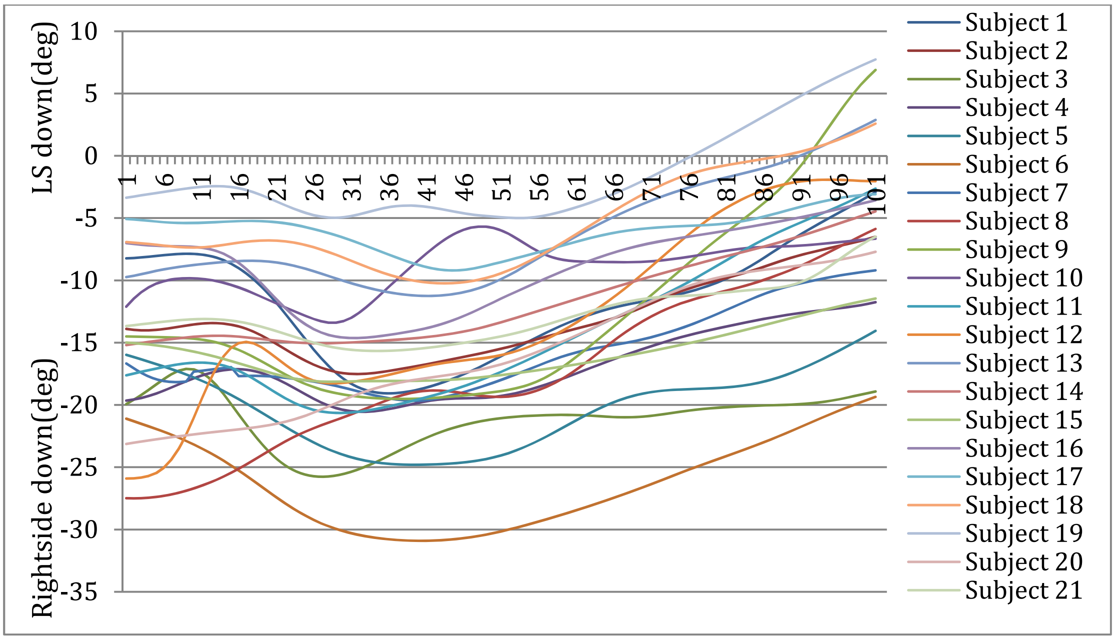

Figure 12 Mean frontal plane trunk angulation. All subjects are leaning away from the direction of the cut at touchdown, rotating towards the direction of the cut through the stance phase. All data normalised to 100% of the stance phase. Weight acceptance phase ends at point 25.

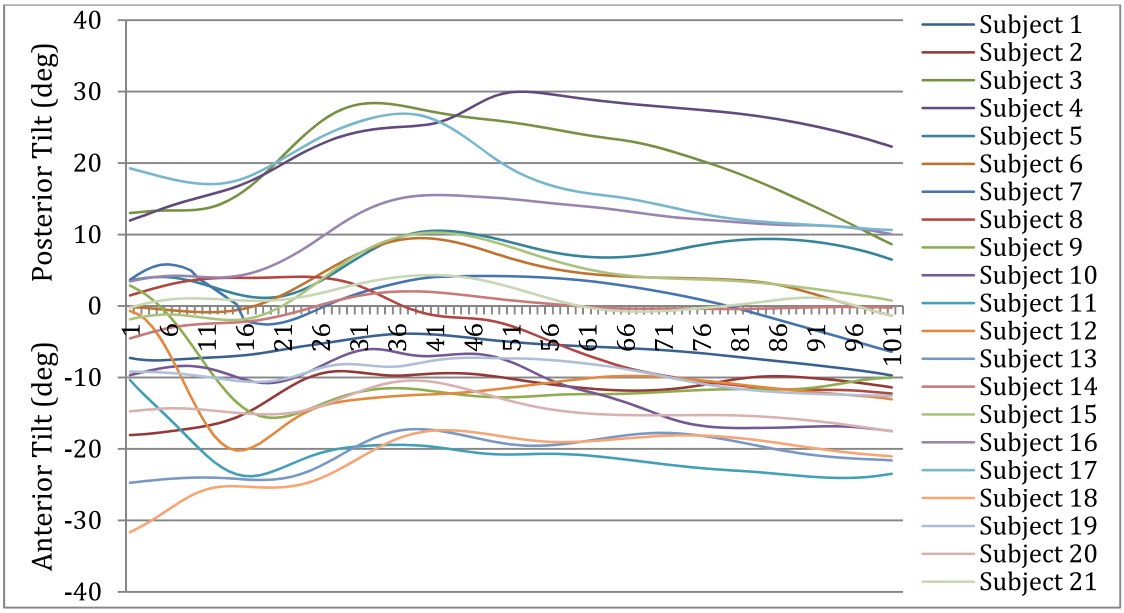

Figure 13. Mean Sagital plane trunk angulation from touchdown to take off. All data normailsied to 100% of the stance phase. Weight acceptance phase ends at point 25.

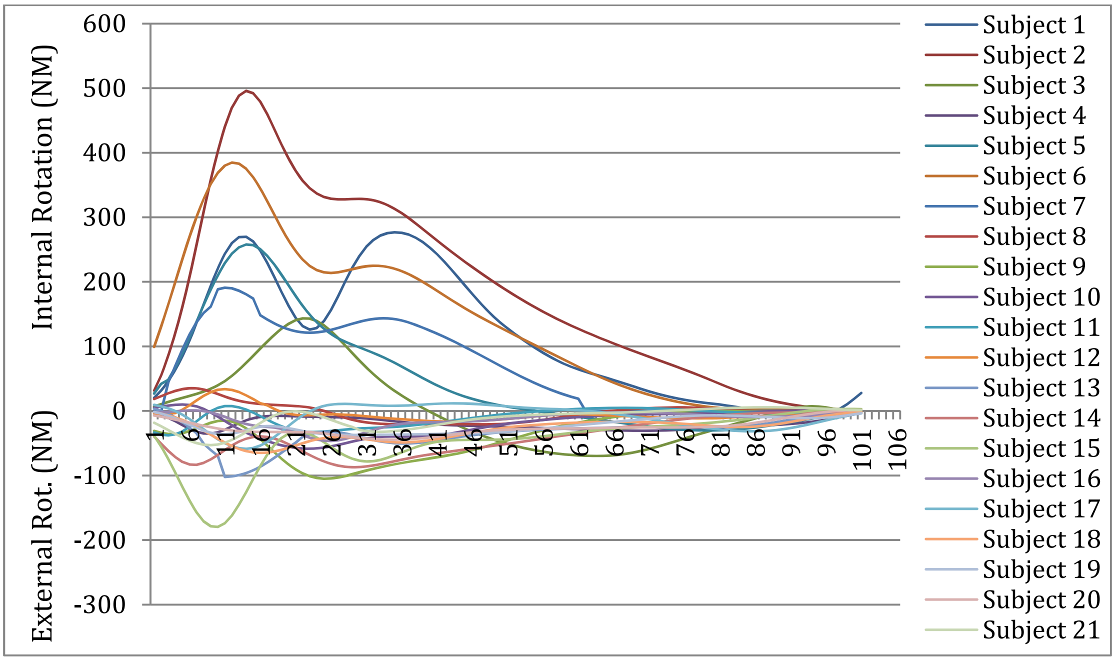

Figure 14. Mean hip internal/external rotation moments. All data normalised to 100% stance phase. Weight acceptance phase ends at point 25.

5.0 Discussion

This study has investigated the role of the trunk in terms of its movements (sagittal and frontal plane) during the stance phase of a sidestep cutting manoeuvre and its relationship to lower extremity kinematics and consequent forces that have the potential increase strain on the ACL. The experiment was designed to replicate the scenario in which most ACL injuries commonly occur, in order to investigate whether trunk angle changes influence knee joint loading to determine whether any specific movement could be ascertained as having a direct impact on ACL injury.

From the results, no relationship is shown to exist between sagittal plane trunk angulation and posterior GRF, and as such no relationship would be expected between trunk angulation and knee flexion/extension angle or knee flexion/extension moment, as is the case. Previous research (Kugler & Janshen, 2010) has suggested that a forward orientated GRF is correlated with a forward orientate body position ($r = 0.93$; $p < 0.001$). However the results (Figure 11 & Figure 13) show no relationship as subjects demonstrating a large initially posteriorly directed GRF have differing degrees of trunk angle. Previous research has also shown a very strong relationship between posterior GRF and knee flexion/extension angle and knee flexion/extension moment, as it is posterior GRF that causes knee flexion upon touchdown and this in turn generates an extension moment about the knee to oppose the external knee flexion moment (Yu & Garrett, 2007). Therefore, correlations between these variables would be expected. However, after examining the results, not all subjects exhibit an extension moment following touchdown (Figure 10). Six subjects (1, 2, 3, 5, 6 & 7) exhibit knee flexion moments. These subjects also exhibit a large anteriorly directed force following touchdown (Figure 11) rather than the expected braking (posteriorly directed) phase. This could be due to the subjects having active knee flexion upon touchdown to aid in deceleration. The limited space of the laboratory may also have made the subjects believe they must decelerate heavily to avoid hitting the wall. Whilst timing gates where used to ensure a 3 m/s approach speed, they were placed 1m behind the force plate, allowing a small area for deceleration prior to touchdown. Also, speed was not monitored following the cut, as subjects were not asked to exit the cutting manoeuvre at any specific speed. This could have negatively affected the results.

This deceleration could also explain the measured knee flexion angles during the touchdown phase. All studies examining sidestep cutting manoeuvres that measure knee flexion angles quote knee flexion angles of up to 30° during weight acceptance (Cochrane et al., 2007; Sigward & Powers, 2006; Besier et al., 2001a; Li et al., 1999). Whereas, all subjects tested in this study demonstrate greater knee flexion angles (mean 55°; SD ± 6.2). In fact there is a minimum peak knee flexion angle of 40° across all subjects (Appendix 2; Figure 15). This could be due to greater absorption of momentum as the subjects seek to decelerate not only to perform the sidestep cutting manoeuvre but also to avoid a collision with the wall. Whilst the area was small, the space was adequate to perform the manoeuvre. However, the subjects unfamiliarity with the laboratory and experimental procedure, may have caused apprehension on the part of the subject which may have influenced their performance. The knee flexion angle results of this study are similar to the Kulas et al. (2010) study which tested jump-landing tasks, and therefore required the complete deceleration of falling mass, which is aided by a greater range of knee flexion. However, another possible explanation could be due to the experience level of the subjects in the current study. No subjects in the present study exhibited knee flexion angles under 30° (Figure 15; Appendix 2) at the end of the initial weight acceptance phase. The weight acceptance phase was identified by the tough proceeding the initial peak (Besier et al., 2001a) in the vertical GRF (Appendix 2; Figure 16). In fact the smallest knee flexion angle at the end of the weight acceptance phase was 36° by subject 10 (Appendix 2; Figure 15). Small knee flexion angles are believed to increase ACL loading (Fleming et al., 2001; Beynnon et al., 1995; Durselen et al., 1995), by limiting hamstrings involvement (Li et al., 1999). Initial trough in vertical GRF occurs at approximately 25% of the stance, which is in keeping with previous studies as most critical part of the stance phase (Sigward & Powers, 2007; Pollard et al., 2004). However, subject 10, whilst exhibit the smallest knee flexion angle during weight acceptance and therefore is at the greatest risk of ACL, exhibits a small knee valgus moment (0.1Nm/Kg-Bwt), a 0.83 Bwt peak posterior GRF and 0.18Nm/Kg-Bwt knee extension moment, none of which are deemed extreme when compared to other subjects. Neither are these results deemed to be at injurious levels (Sigward & Powers, 2007).

Previous studies have used varying methods of recruitment. Sigward and Powers (2006a) stated that they recruited players with experience (male 12.4 ± 3.0 years). However, the subjects identified dominant leg by which leg they could kick the ball the furthest with, and one subject was unable to identify a dominant leg. The validity of this test must be questioned, as preferred kicking leg is known amongst experienced players. Cochrane et al. (2007), whilst doing a study on professional Australian Rules Footballers, undertook the study in retrospect, and therefore they used available video footage of past games. This may have influenced key parameters such as the viewing angle, which may not have been optimal to accurately measure knee flexion angle. Also, the study designated knee flexion into 2 groups identified as 0-30° and 30-60°, which is not the most accurate identification. Zazulak et al. (2007) sampled collegiate athletes, however no sport was identified, which would again question the experience of the participating athletes for this type of movement. This could identify a possible conflict with previous research, as the experience of the subject could have brought about adaptive changes (greater knee flexion) to avoid injury. Previous research has demonstrated a greater protective influence of the hamstrings at knee flexion angles greater than 30° (DeMorat et al., 2004; Markolf as cited in Fleming et al., 2001; Beynnon et al., 1995), therefore the hamstrings could be providing a protective response. Though, as with the spatial confinements, this is only speculative and this warrants further analysis to identify whether there is a difference between key variables dependant on experience level. Whilst many previous studies have compared results based on gender (Sigward & Powers, 2006a; McLean et al., 2004a; Pollard et al., 2004), age (Sigward & Powers, 2006a) and past injury (Georgoulis et al., 2003), only one study has compared experience level (Sigward & Powers, 2006b). As outlined earlier, this study identified some conflicting evidence, such as significant differences between novice and experienced athletes in peak knee joint moment ($p = 0.03$) and net joint moment impulse ($p = 0.03$), as well as a negative correlation between number of years experience and quadriceps-hamstring co-contraction (r = -0.32; p = 0.04), leading the authors to suggest that novice athletes employ a protective mechanism (co-contraction) and are at less risk of ACL injury as they exhibit lesser knee joint moments. However this study, whilst conducting multiple comparisons for analysis, failed to apply a p-value correction and is therefore subject to type 1 errors. As such this evidence must be treated with caution, and further novice versus experienced athlete research must be conducted to determine the full extent of the relationship.

The knee flexion angle results also tie in with the knee valgus moment results. The two subjects that exhibited the highest peak knee valgus moment (Figure 7) subject 9 (0.15Nm/Kg-Bwt) and subject 10 (0.1Nm/Kg-Bwt) also exhibited some of the smallest knee flexion angles at 25% stance (46° and 36° respectively), as well as exhibiting some of the smallest peak knee flexion angles (47° and 45° respectively). However, none of the subjects exhibit excessive knee valgus moments as defined by Sigward and Powers (2007) to be greater than 0.59Nm/Kg-Bwt, as the highest peak valgus moment in the present study (subject 9) equates to 0.15Nm/Kg-Bwt. All subjects exhibiting a knee valgus moment all have a posteriorly directed force during the weight acceptance phase (Figure 11). Again, whilst not tested in this experiment, manipulation of anterior/posterior force application may have protective applications, and could be implemented in any ACL protective strategies, as none of the subjects exhibiting an anteriorly directed force following touchdown exhibited a knee valgus moment.

Sagittal plane trunk angulation has been shown to have a direct affect on posterior GRF (Yu et al., 2006). Blackburn and Padua (2009) demonstrating decreased peak posterior GRF with a flexed landing. Whilst no significant correlation was found the subject with the highest peak posterior GRF (Appendix 2; Table 3 subject 16; 1.53Bwt) had a posterior trunk lean at touchdown and through the weight acceptance phase, whereas the smallest peak posterior GRF (Appendix 2; Table 3 subject 19; 0.49Bwt) had an anterior trunk lean at touchdown and through weight acceptance phase. That being said, neither subject exhibited the greatest anterior or posterior trunk lean at touchdown (Figure 13) and neither had the greatest range of motion through the weight acceptance phase, meaning that the results could have been coincidence as the lack of correlation may suggest.

The results of the study failed to highlight a correlation between frontal plane trunk angulation and knee internal/external rotation moment, hip abduction/adduction moment, hip internal/external rotation moment or knee valgus moment. However, Jamison et al. (2012) identified a positive association between knee valgus moment and lateral lean. All subjects demonstrate an initial lateral lean away from the direction of the cut (Figure 12) as in the study by Jamison et al. (2012), yet only 6 exhibit a knee valgus moment, with the subject exhibiting the greatest lateral lean in the weight acceptance phase not being one of them. However, as outlined earlier, the study by Jamison et al. (2012) had several limitations, one of which included failure to acknowledge gender differences in knee loading mechanics. This may have led to inaccurate results as males and females have been shown to load different with females exhibiting greater knee valgus (Quatman & Hewett, 2010; McLean et al., 2005). As the Jamison et al. (2012) study is the only study to examine frontal plane trunk angle and sidestep cutting manoeuvres it is difficult to determine accuracy of results. Although, mechanically a lateral lean over the stance leg will shift the centre of pressure laterally due to the shift in the CM, which will cause the vertical GRF vector to act laterally with respect to the knee joint, thus creating a knee valgus moment (Powers, 2010). This is not shown in the results of this study. This may indicate a threshold where lateral lean over the stance leg and away from the direction of the sidestep cutting manoeuvre becomes injurious, as mechanically, the further the lean, the greater lateral shift of the CM and therefore the centre of pressure, which will increase the moment arm of the result vertical GRF, increasing the knee valgus moment.

Whilst the relationship between hip abduction and knee valgus is somewhat controversial in past research as outlined above (Sigward & Powers, 2007; Powers, 2010) all subjects demonstrating a knee valgus moment through the weight acceptance phase also exhibit a hip adduction moment (Figure 9). Sigward and Powers (2007) identified hip abduction moments in the excessive knee valgus group when performing sidestep cutting manoeuvres. However, other subjects also demonstrate a hip abduction moment during weight acceptance whilst exhibiting a knee varus moment, albeit a small varus moment. This may be due to subjects trying to anticipate the manoeuvre and turning their body in preparation. A problem highlighted by Besier et al. (2001b). If the subject had implemented mechanical changes to facilitate the sidestep cutting manoeuvre prior to touchdown, this may have affected the results (Besier et al., 2001b). A method of detecting such events in future research would be to incorporate the foot into the analysis, as toe direction would provide an indicator as to whether any mechanical changes have been implemented prior to touchdown (Ishida et al., 2012). Also, pelvic rotation angle, as the lower extremities make up a multi-segment kinematic chain (Kulas et al., 2010) and dynamic knee valgus is as a consequence of hip, knee and ankle orientations and moments (Ishida et al., 2012; Quatman et al., 2010; Yu & Garrett, 2007), omission of any segment could undermine any analysis. This highlights a shortcoming of the present study.

All subjects exhibiting knee valgus also exhibit an external rotation moment at the knee (Figure 8), both peak and during weight acceptance phase. External rotation is deemed to be part of dynamic knee valgus collapse (Hewett et al., 2005), and as such, is believed to contribute to increased ACL loading and in some cases injury. However, all subjects exhibiting external knee rotation moments, whilst undergoing a knee valgus moment, display relative small moments (Sigward & Powers, 2007). One possible explanation for this is the approach speed. As approach speed increases so do the corresponding knee loading kinetics (Vanrenterghem et al., 2012). An approach speed of 3 m/s may be insufficient to illicit problematic knee loading, although Cochrane et al. (2007) identified ACL injuries with approach speeds of 3 m/s. Markolf et al. (as cited in Yu & Garrett, 2007) demonstrated decreased ACL strain when loaded with a combination of knee valgus and external rotation moments, to the point where strain was lower with the combination of forces than with each singular load. The same study also went on to demonstrate reduced ACL strain when loaded with anterior shear force and an external rotation moment. This could provide one explanation as the relatively small knee valgus moments experienced by subjects in this study, as they all exhibit a knee external rotation moment. This could in effect provide a protective response for the ACL. However, as previously stated, the study by Markolf et al. (as cited in Yu & Garrett, 2007) used cadaveric knees and where, therefore, non-weight bearing. That being said, a causal link may exist between knee valgus and knee external rotation moments.

As the lower extremities make up a multi-segmented kinematic chain, changes in one part will cause changes in another. Therefore, a relationship could exist between knee internal/external rotation and hip internal/external rotation. The same group that exhibited knee external rotation moment exhibited hip internal rotation moment. This is expected as most studies agree that the hip undergoes internal rotation during sidestep cutting manoeuvres (Sigward & Powers, 2007; McLean et al., 2004a) and it is the closest to a significant comparison in this study, as well as presenting the strongest correlation (r_s = - 0.560; n = 21; p = 0.008). This is also congruent with the study by Powers (2010) who highlighted the interdependence of the lower limbs, especially the hip and knee joints as they share a common segment, the femur. Therefore, changes at one end of the segment are expected to cause changes at the opposite end. However, further examination of the result (Figure 8 & Figure 14) demonstrates that while opposite rotation is occurring at the opposite end of the femur, the magnitudes differ greatly. Subject 2 demonstrates the greatest peak hip internal rotation at 495.765 Nm, whereas, peak knee external rotation is not the greatest at -166.24 Nm. Nonetheless, the results do suggest a relationship could exist.

5.1 Limitations

A major limitation of the present study, which may have influenced the results, is the spatial restrictions of the testing laboratory. Whilst the area was adequate to safely execute the sidestep cutting manoeuvre to the desired criteria, unfamiliarity of the test added to the small testing area could have affected the performance of the participants as they may have feared colliding with the wall, and whilst crash mats where used to cover the wall in case of any accidents, this may not have been enough to appease the participants. This may have negatively affected the results as a sidestep cutting manoeuvre is employed to avoid another player, or to rapidly change direction to retrieve the ball, as such players usually accelerate out of the manoeuvre. This also leads to another limitation, not controlling the exit speed following the sidestep cutting manoeuvre. As there was no predefined exit speed, only an angle, participants were not accelerating away from the cut, as they would in actual game like scenarios. This did mean that once the subjects had passed the timing gate, set 1 metre behind the force plate, they could begin to decelerate. This may account for some of the results, specifically the different loading patterns exhibited in the anterior-posterior GRF data.

This study was based around the movements of the trunk during sidestep cutting manoeuvres. This study has been critical of past research into this area that has omitted the trunk segment, yet this study omitted a segment, the foot. As all injurious kinetic forces are created and/or influenced by the subjects interaction with the ground and the transfer of that generated force from the ground to the knee must first be transmitted up the foot. Foot direction during the stance phase has also been shown to have an effect.

5.2 Areas for further study

Whilst all the results of this study proved to be insignificant several areas have been illuminated that require further investigation. As discussed earlier research into experience level of the subject is sparse. It is feasible that experienced athletes develop protection mechanisms.

Another area of interest is manipulation of the anterior-posterior GRFs at touchdown. Normally subjects exhibit a posteriorly directed force at touchdown, creating an extension moment about the knee, however, some subjects in this study exhibited an anteriorly directed force. These subjects exhibited a knee varus moment through weigh acceptance rather than the injurious knee valgus moment. Whilst this may have be coincidental the potential to avoid knee valgus loading means that it warrants further investigation.

In conclusion, whilst this study failed to uncover any significant relationships between trunk orientation and the key contributors to knee joint loading, several interesting points have arisen that could provide useful data moving forward. Trunk orientation should be included in analysis on sidestep cutting manoeuvres as it is the largest of the bodies segments, and whilst not demonstrated in this study, has been shown to affect lower extremity kinematics and kinetics. In order to fully understand sidestep cutting manoeuvres all lower extremity segments together with the trunk should be used to provide a full and comprehensive analysis.

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Appendix 1

Additional Methodology

Figure 1. Representation of how posterior ground reaction force is transmitted up the foot and through the shank to the knee joint following ground contact.

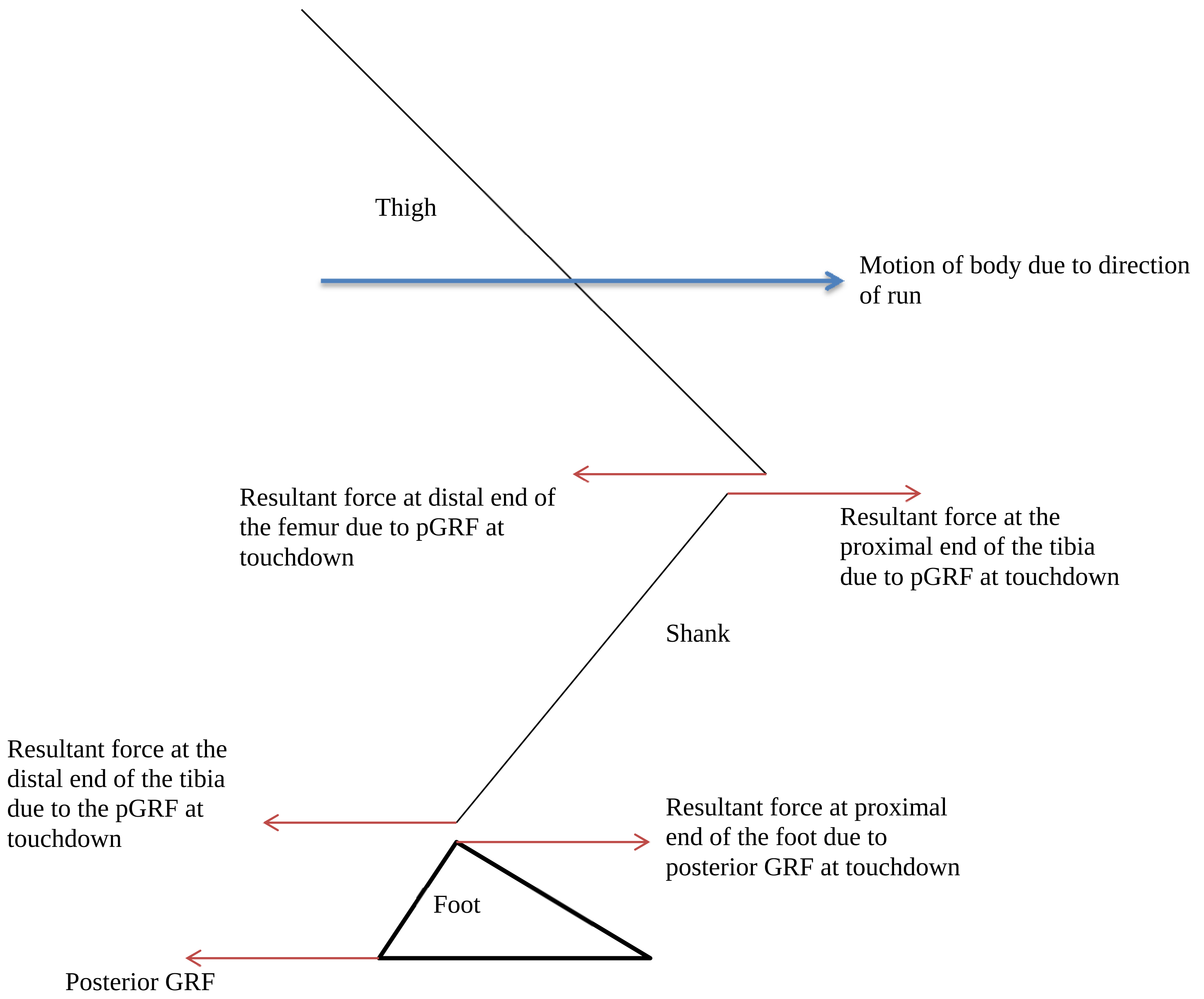


Figure 2. A) Demonstrating upright trunk position on landing, and how GRF vector acts up through the centre of the body to the CM. B) demonstrating how a lateral lean on touchdown shifts the CM over the standing leg which shifts the GRF vector, creating a valgus moment at the knee.

A) B)

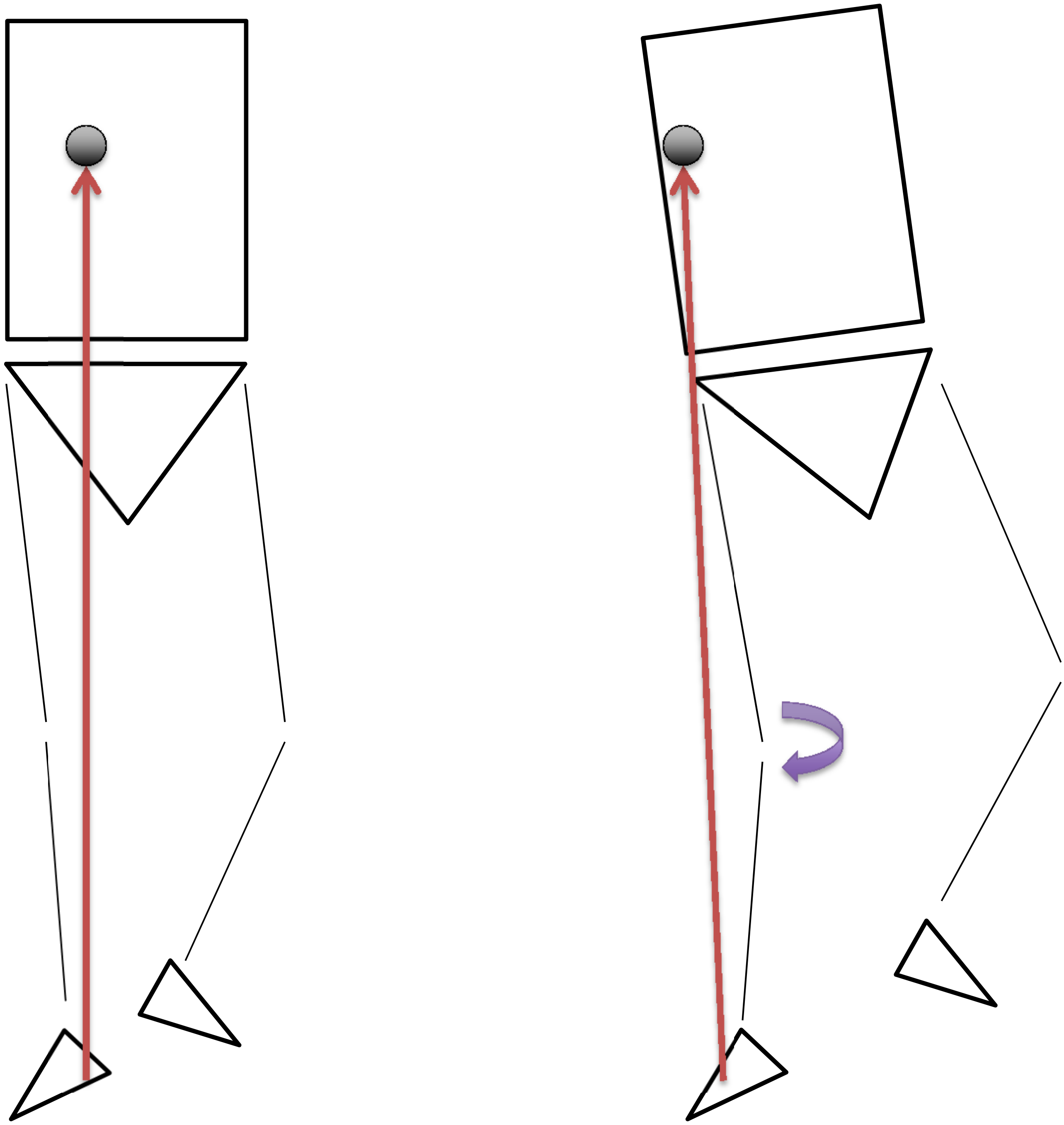


Figure 3. Effect of trunk orientation on Centre of Mass location and impact on moment arm for knee loading

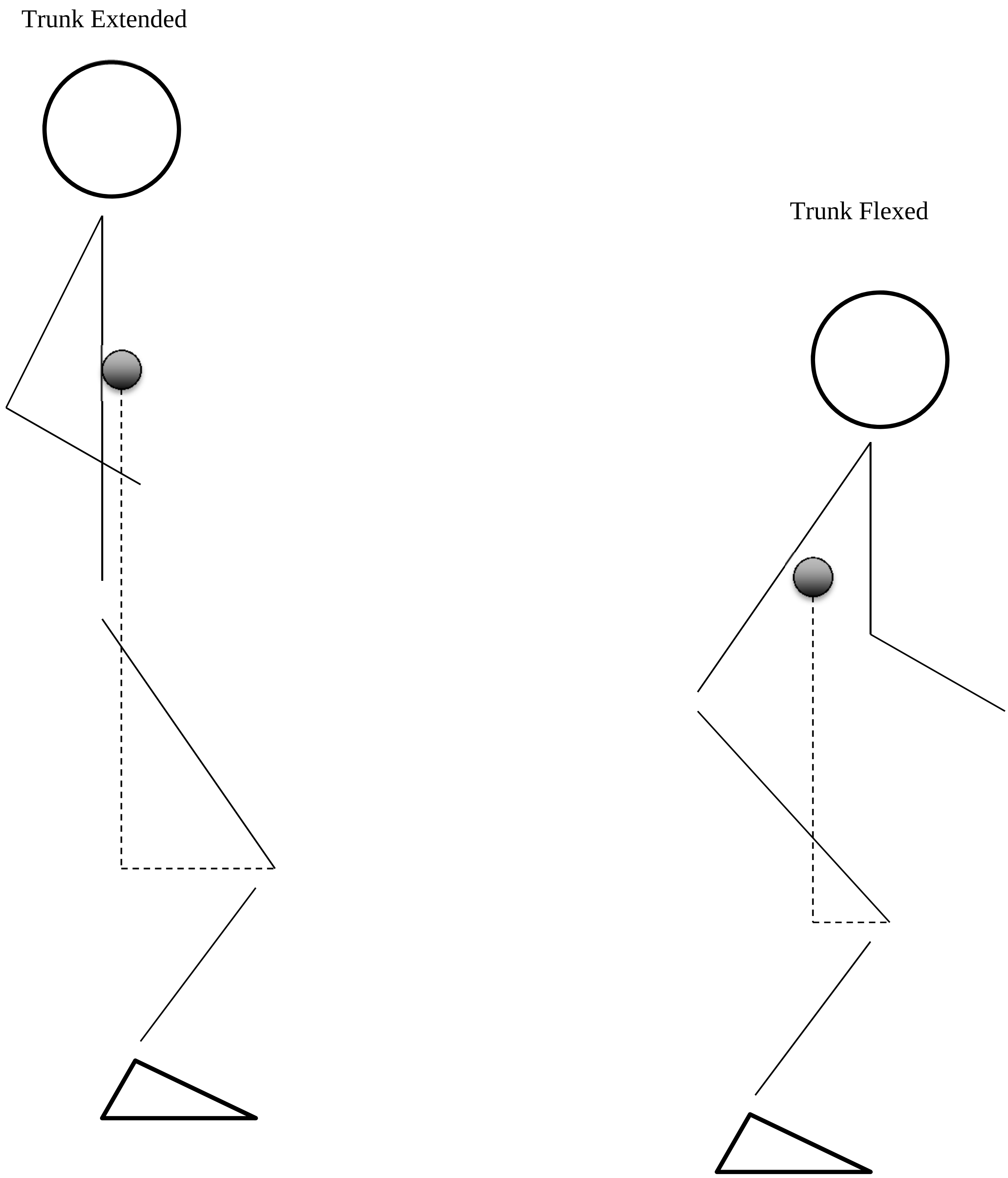


Figure 4. A diagrammatical representation of how the orientation of the trunk at touchdown can affect the vertical GRF and consequently the loading at the knee.

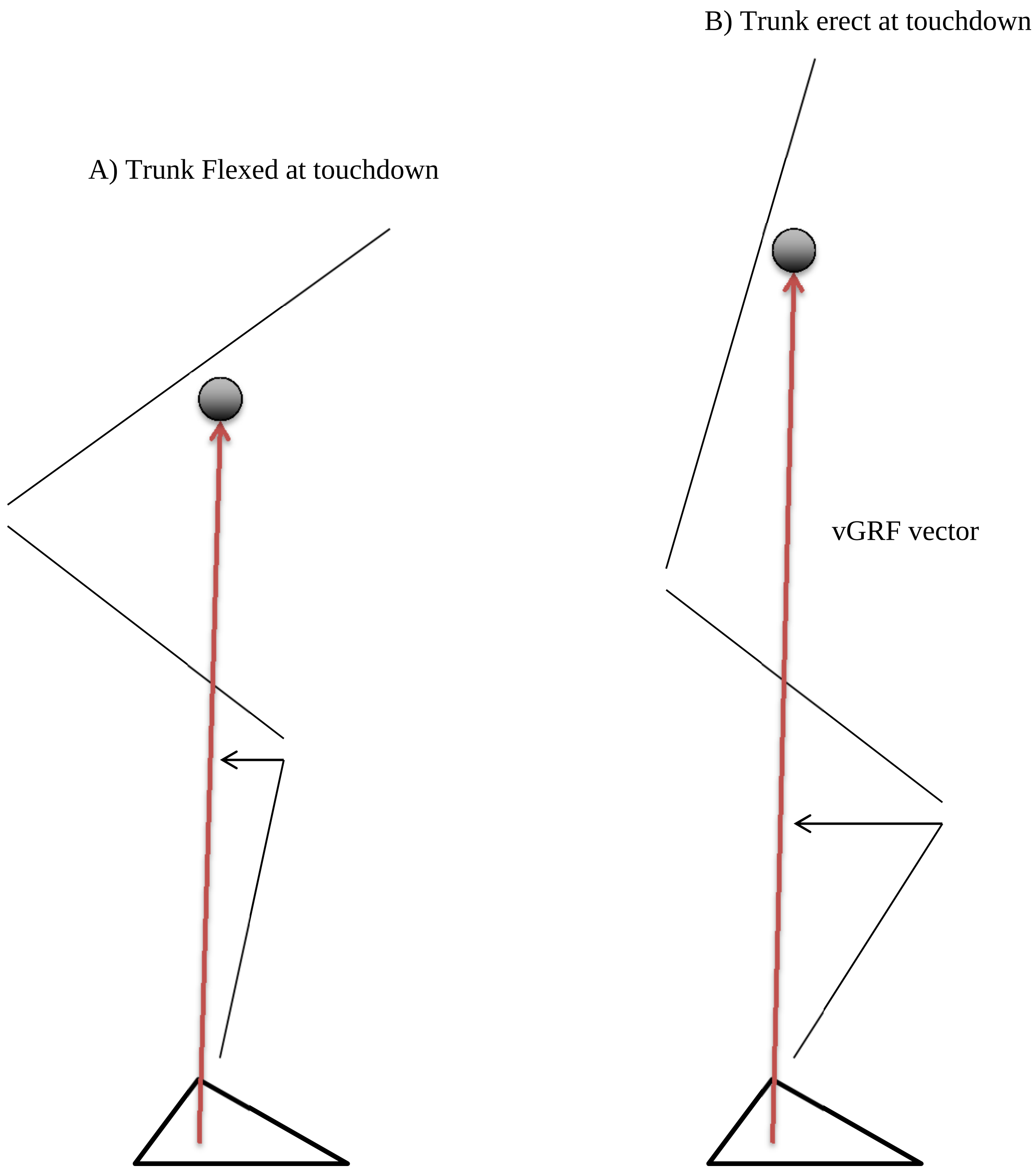
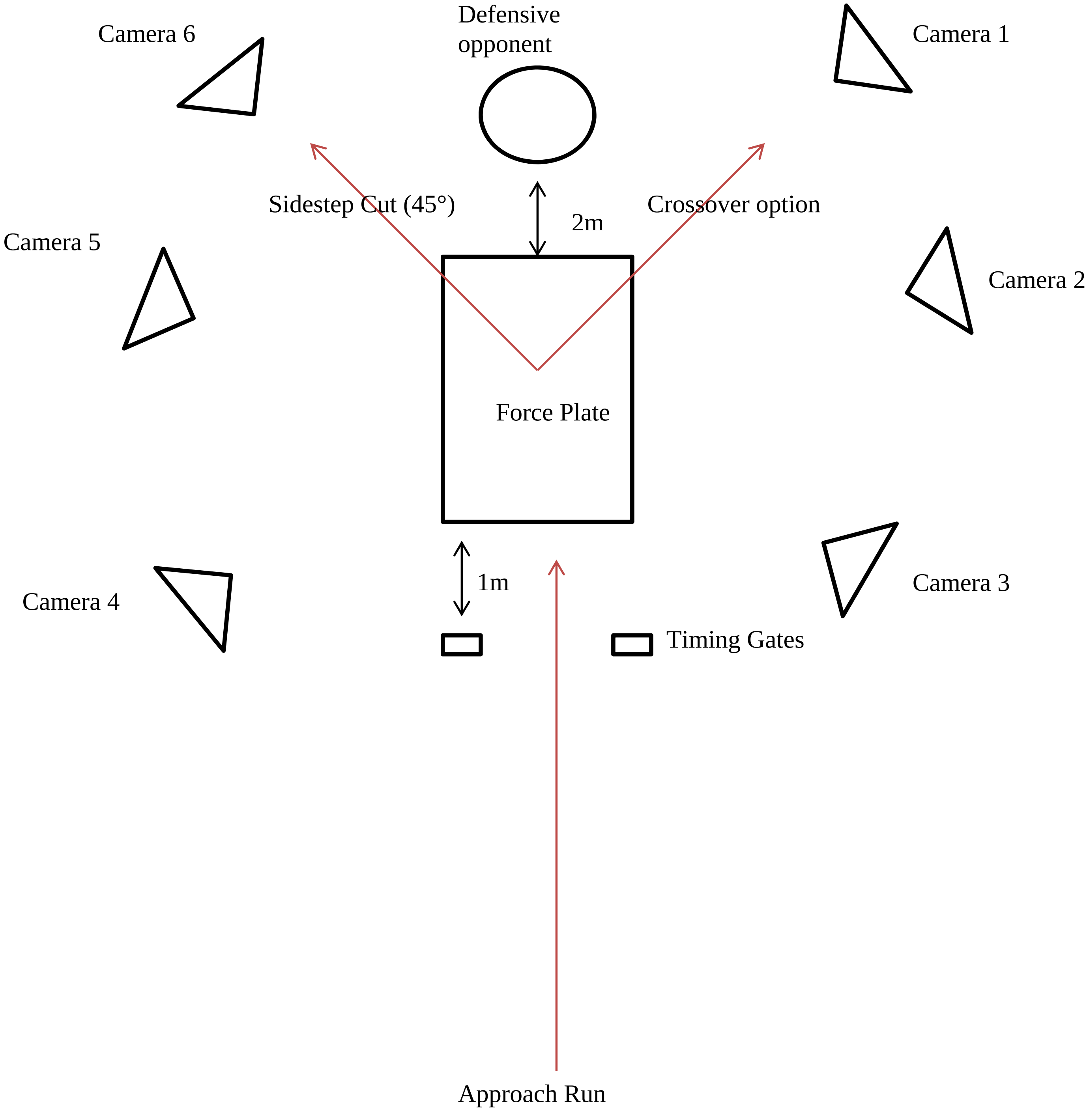


Figure 5. Experimental set-up.



Appendix 2 Additional Results

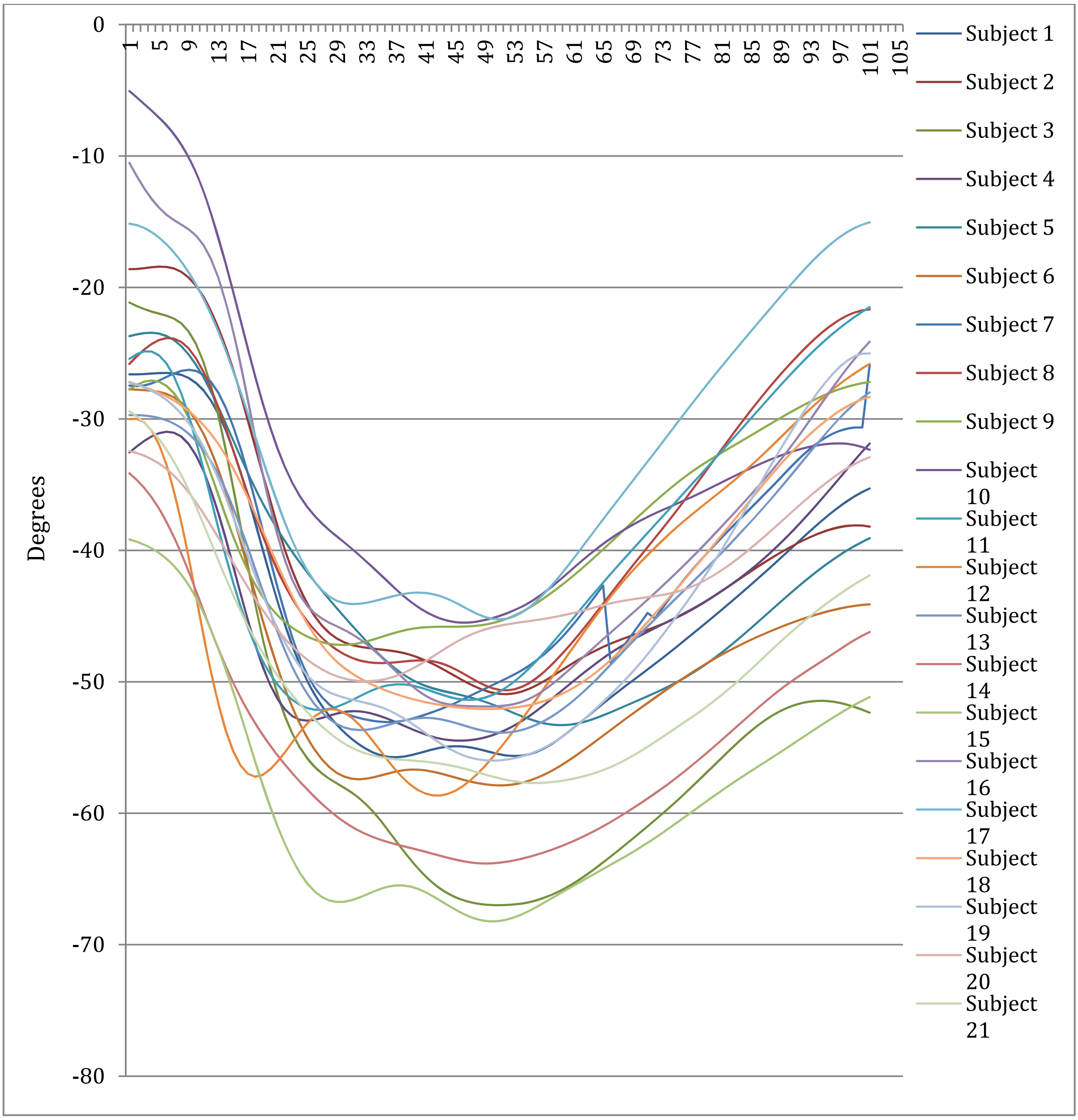


Figure 15. Mean knee flexion angles of all 21 subjects. Data normalised for 100% of the stance phase. (– denotes flexion).

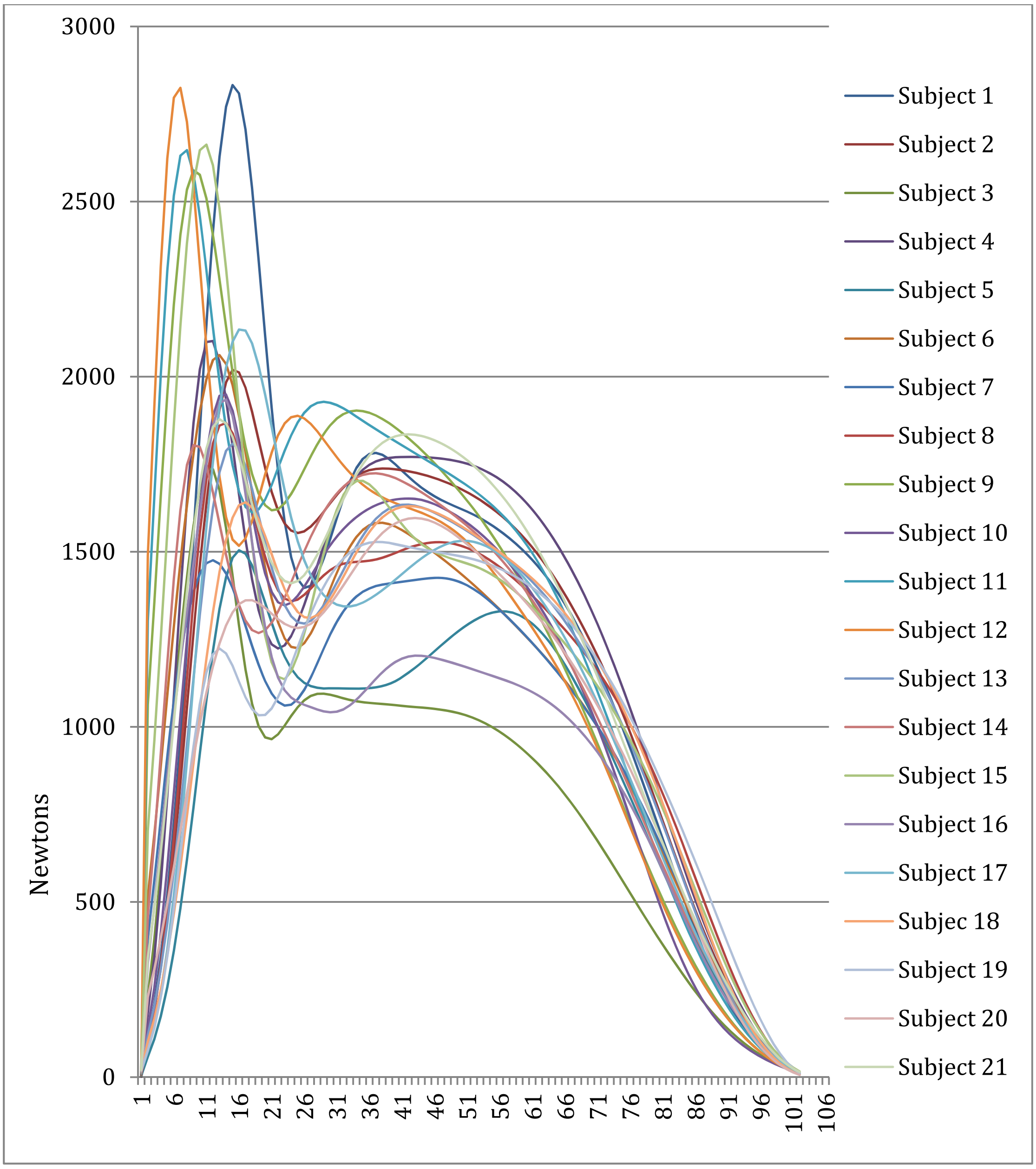


Figure 16. Mean vertical ground reaction force of all 21 subjects. Data normalised for 100% of the stance phase. Trough following the initial peak in force denotes the end of the weight acceptance phase.

Table 1. Tests of Normality Results

Tests of Normality

	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
	Statistic	df	Sig.	Statistic	df	Sig.
pGRF	.286	21	.000	.816	21	.001
kneeflexangle	.122	21	.200*	.934	21	.165
kneeflexmoment	.399	21	.000	.654	21	.000
kneeintrotmoment	.214	21	.013	.824	21	.002
hipabdmom	.200	21	.028	.874	21	.011
hipintrotmoment	.288	21	.000	.835	21	.002
kneevalgusmoment	.145	21	.200*	.940	21	.214
fronttrunkang	.084	21	.200*	.990	21	.997
sagtrunkang	.142	21	.200*	.952	21	.371

*. This is a lower bound of the true significance.

a. Lilliefors Significance Correction

Subject	mean ant(-)/ post(+) GRF	SD	mean hip abd(-) /add(+) moment	SD	hip int(+)/ext(-) rot moment	SD	knee flex(-) /ext(+) angle	SD	knee flex(-) /ext(+) moment	SD
1	1061.456909	145.165192	396.445374	98.964264	276.79834	59.415249	-55.738701	1.798887	-736.580017	175.803848
2	655.435364	103.721489	581.484558	82.342415	495.765961	74.931046	-50.939266	0.542299	-685.7099	98.198433
3	815.353088	124.26535	832.174438	80.750679	143.412125	28.632008	-66.993431	1.929688	-832.700439	14.760177
4	-581.705933	94.734543	81.144974	11.651892	-35.236401	5.091466	-54.476273	0.729135	215.392105	7.665527
5	556.51416	12.221562	382.05545	76.852127	257.813904	51.914436	-53.294449	0.474263	303.89621	61.683189
6	906.833496	30.768961	395.866455	169.586594	384.841095	161.743515	-57.414169	0.713024	-235.842072	99.349556
7	480.282623	78.033783	374.837006	117.312675	190.835022	59.758575	-53.063461	1.347387	-467.055054	156.483765
8	-484.762878	47.505978	78.260391	11.001676	35.115993	5.001804	-50.6469	1.092538	218.404617	2.487728
9	-666.215271	86.781601	-201.126541	4.888764	-104.927681	5.852709	-47.199821	1.143409	194.013214	8.247084
10	-642.000366	12.216881	-154.23938	4.262983	-58.475693	2.548095	-45.499668	1.521931	164.940994	1.393956
11	-648.14502	37.589375	-71.702835	6.653804	-37.789207	6.273871	-52.141182	0.530575	230.722931	3.026731
12	-1041.585327	40.272362	200.146042	25.828049	33.879284	5.758465	-58.647537	1.763791	237.26384	4.540819
13	-552.638123	14.472168	-86.735329	6.198211	-102.415848	10.556996	-53.865307	2.077473	209.821701	5.131775
14	-592.059937	25.951462	-96.177254	6.424389	-87.166718	3.386065	-63.822449	1.77857	256.792969	3.601731
15	-749.897766	82.133278	-127.811066	13.80132	-179.728088	6.27001	-68.219292	1.200458	249.737274	3.190906
16	-975.194885	36.029289	93.963974	2.088974	-46.476242	2.034613	-51.876144	0.419765	154.104599	4.838942
17	-797.245239	50.097969	93.65921	9.740256	-58.654358	7.658214	-45.250183	1.77392	243.61969	9.533722
18	-509.124359	35.216484	-107.092064	13.296681	-64.943848	4.504071	-52.070118	1.351652	214.005356	4.089273
19	-367.010284	25.192884	-35.808395	8.334649	-37.40139	3.416708	-55.996323	0.605993	185.797073	5.694402
20	-426.995605	37.250057	-91.039444	3.00518	-44.597466	1.238663	-49.954453	0.99435	176.88916	6.957396
21	-612.742188	45.320366	-74.80854	4.041558	-53.060345	3.993458	-57.697735	0.789869	220.459412	2.976663

Subject	knee int(+)/ext(-) moment	SD	frontal plane trunk angle; right(-)/ left(+)	SD	sagital plane trunk angle; Ant(-)/ post(+)	SD	mean knee valgus(-)/ vaus(+) moment	SD
1	-264.49765	68.16053	-19.070335	0.884409	-9.713291	1.484696	402.283081	102.981224
2	-167.318985	27.88199	-17.521162	1.49153	-18.067446	0.594021	584.389343	79.446548
3	-185.536453	15.979225	-25.76029	2.298563	28.400993	0.493366	247.495224	34.768543
4	-37.295147	4.250057	-20.54664	0.563819	29.993225	1.055631	133.390442	7.824166
5	-259.678131	53.372429	-24.799829	1.437472	10.532386	1.241746	423.54892	87.130363
6	-107.661942	46.996498	-30.901957	1.147702	9.504826	0.353076	238.385468	102.217133
7	-162.530457	54.527908	-19.662752	0.89794	5.826692	2.950122	282.666107	94.831894
8	-46.105473	2.717303	-27.499464	0.368018	4.095448	2.453813	113.68631	7.373951
9	52.149685	1.020475	-19.513056	1.468141	-15.650112	0.731964	-130.664856	10.296808
10	40.929539	0.669475	-13.395385	1.621846	-10.784249	1.722905	-73.950958	1.865274
11	42.263844	1.496157	-20.642012	1.05499	-23.810411	1.448678	-30.765951	6.006921
12	-27.146046	3.028476	-25.909197	3.352368	-20.225193	1.375449	168.638855	5.400099
13	36.747776	2.826379	-11.243917	1.248255	-24.738094	0.492835	63.587784	8.698881
14	46.773319	4.039127	-15.188267	0.510077	-4.551136	0.296989	-51.822868	4.197695
15	-57.662666	4.282226	-18.127918	0.289312	10.262889	0.99902	78.961792	8.177641
16	17.462967	1.796134	-14.626487	0.484817	15.523832	0.491505	57.364754	3.199686
17	31.886559	2.358905	-9.201042	0.60554	26.911156	1.955023	98.490967	8.649748
18	37.884811	5.772544	-10.233657	0.828773	-31.657694	1.209321	26.042503	14.047166
19	6.535251	3.124551	-4.995685	1.147857	-10.608533	0.425389	49.658356	8.658425
20	29.600887	2.585728	-23.13047	1.015861	-15.175618	1.428679	51.714584	10.224169
21	26.709133	1.027041	-15.667221	1.011834	4.320524	1.96876	162.474365	3.434351

Table 2. All Mean and standard deviation data for all subjects for all trials.

Table 3. Anterior-Posterior Force expressed as Body weights

Subject	Peak A-P GRF	Weight (KG)	N/Kg-BW
1	1061.45	86	1.26
2	655.44	88	0.76
3	815.35	65	1.28
4	-581.71	85	0.70
5	556.51	65	0.87
6	906.83	62	1.49
7	480.28	77	0.64
8	-484.76	76	0.65
9	-666.22	87	0.78
10	-642.00	79	0.83
11	-648.15	73.5	0.9
12	-1041.59	75	1.42
13	-552.64	70	0.8
14	-592.06	78	0.77
15	-749.90	67	1.14
16	-975.19	65	1.53
17	-797.25	76	1.07
18	-508.97	75	0.69
19	-367.01	77	0.49
20	-427.00	75	0.58
21	-612.74	76	0.82

Table 4. Descriptive Statistics.

Descriptives

			Statistic	Std. Error
pGRF	Mean		-246.259407	146.3951742
	95% Confidence Interval for Mean	Lower Bound	-551.634389	
		Upper Bound	59.115576	
	5% Trimmed Mean		-274.492119	
	Median		-552.638123	
	Variance		450062.488	
	Std. Deviation		670.8669671	
	Minimum		-1041.5853	
	Maximum		1061.4569	
	Range		2103.0422	
	Interquartile Range		1175.5785	
	Skewness		.918	.501
	Kurtosis		-.741	.972
kneeflexangle	Mean		-54.514612	1.3462646
	95% Confidence Interval for Mean	Lower Bound	-57.322871	
		Upper Bound	-51.706354	
	5% Trimmed Mean		-54.270515	
	Median		-53.294449	
	Variance		38.061	
	Std. Deviation		6.1693596	
	Minimum		-68.2193	
	Maximum		-45.2502	
	Range		22.9691	
	Interquartile Range		6.7629	
	Skewness		-.774	.501
	Kurtosis		.467	.972
kneeflexmoment	Mean		24.665413	80.8230197
	95% Confidence Interval for Mean	Lower Bound	-143.928452	
		Upper Bound	193.259277	
	5% Trimmed Mean		56.654352	
	Median		209.821701	
	Variance		137179.571	
	Std. Deviation		370.3776058	
	Minimum		-832.7004	
	Maximum		303.8962	
	Range		1136.5966	
	Interquartile Range		274.8621	

	Skewness		-1.566	.501
	Kurtosis		.873	.972
kneeintrotmoment	Mean		-45.070913	22.7222494
	95% Confidence Interval for Mean	Lower Bound	-92.468695	
		Upper Bound	2.326868	
	5% Trimmed Mean		-38.280210	
	Median		6.535251	
	Variance		10842.313	
	Std. Deviation		104.1264280	
	Minimum		-264.4977	
	Maximum		52.1497	
	Range		316.6473	
	Interquartile Range		172.4125	
	Skewness		-1.036	.501
	Kurtosis		-.212	.972
hipabdmom	Mean		117.309382	60.5823266
	95% Confidence Interval for Mean	Lower Bound	-9.063137	
		Upper Bound	243.681901	
	5% Trimmed Mean		95.824702	
	Median		78.260391	
	Variance		77074.584	
	Std. Deviation		277.6230973	
	Minimum		-201.1265	
	Maximum		832.1744	
	Range		1033.3010	
	Interquartile Range		472.0546	
	Skewness		1.089	.501
	Kurtosis		.582	.972
hipintrotmoment	Mean		43.218497	39.0549862
	95% Confidence Interval for Mean	Lower Bound	-38.248777	
		Upper Bound	124.685771	
	5% Trimmed Mean		30.558460	
	Median		-37.789207	
	Variance		32031.131	
	Std. Deviation		178.9724306	
	Minimum		-179.7281	
	Maximum		495.7660	
	Range		675.4940	
	Interquartile Range		228.9227	
	Skewness		1.278	.501
	Kurtosis		.811	.972
kneevalgusmoment	Mean		137.884487	38.4918062
	95% Confidence Interval for Mean	Lower Bound	57.591986	

		Upper Bound	218.176987	
	5% Trimmed Mean		128.273536	
	Median		98.490967	
	Variance		31114.002	
	Std. Deviation		176.3916154	
	Minimum		-130.6649	
	Maximum		584.3893	
	Range		715.0542	
	Interquartile Range		205.0899	
	Skewness		.924	.501
	Kurtosis		.793	.972
fronttrunkang	Mean		-18.458893	1.4358930
	95% Confidence Interval for Mean	Lower Bound	-21.454113	
		Upper Bound	-15.463672	
	5% Trimmed Mean		-18.513443	
	Median		-19.070335	
	Variance		43.298	
	Std. Deviation		6.5800886	
	Minimum		-30.9020	
	Maximum		-4.9957	
	Range		25.9063	
	Interquartile Range		9.9542	
	Skewness		.109	.501
	Kurtosis		-.365	.972
sagtrunkang	Mean		-1.886181	3.9965536
	95% Confidence Interval for Mean	Lower Bound	-10.222846	
		Upper Bound	6.450483	
	5% Trimmed Mean		-2.017380	
	Median		-4.551136	
	Variance		335.421	
	Std. Deviation		18.3145092	
	Minimum		-31.6577	
	Maximum		29.9932	
	Range		61.6509	
	Interquartile Range		27.2564	
	Skewness		.264	.501
	Kurtosis		-.941	.972

Table 5. Correlations (Sagittal plane trunk & posterior GRF)

Correlations

			sagtrunkang	pGRF
Spearman's rho	sagtrunkang	Correlation Coefficient	1.000	.069
		Sig. (2-tailed)	.	.767
		N	21	21
	pGRF	Correlation Coefficient	.069	1.000
		Sig. (2-tailed)	.767	.
		N	21	21

Table 6. Correlations (sagittal plane trunk angle & knee flexion angle)

Correlations

			sagtrunkang	kneeflexangle
Spearman's rho	sagtrunkang	Correlation Coefficient	1.000	-.218
		Sig. (2-tailed)	.	.342
		N	21	21
	kneeflexangle	Correlation Coefficient	-.218	1.000
		Sig. (2-tailed)	.342	.
		N	21	21

Table 7. Correlations (Sagittal plane trunk angle & knee flexion moment)

Correlations

			sagtrunkang	kneeflexmoment
Spearman's rho	sagtrunkang	Correlation Coefficient	1.000	.019
		Sig. (2-tailed)	.	.933
		N	21	21
	kneeflexmoment	Correlation Coefficient	.019	1.000
		Sig. (2-tailed)	.933	.
		N	21	21

Table 8. Correlations (Frontal plane trunk angle & knee flexion/extension moment)

Correlations

			fronttrunkang	kneeintrotmoment
Spearman's rho	fronttrunkang	Correlation Coefficient	1.000	.449*
		Sig. (2-tailed)	.	.041
		N	21	21
	kneeintrotmoment	Correlation Coefficient	.449*	1.000
		Sig. (2-tailed)	.041	.
		N	21	21

*. Correlation is significant at the 0.05 level (2-tailed).

Table 9. Correlations (Frontal plane trunk angle & hip abduction/adduction moment)

Correlations

			fronttrunkang	hipabdmom
Spearman's rho	fronttrunkang	Correlation Coefficient	1.000	-.410
		Sig. (2-tailed)	.	.065
		N	21	21
	hipabdmom	Correlation Coefficient	-.410	1.000
		Sig. (2-tailed)	.065	.
		N	21	21

Table 10. Correlations (frontal plane trunk angle & hip internal/external rotation moment

Correlations

			fronttrunkang	hipintrotmoment
Spearman's rho	fronttrunkang	Correlation Coefficient	1.000	-.560**
		Sig. (2-tailed)	.	.008
		N	21	21
	hipintrotmoment	Correlation Coefficient	-.560**	1.000
		Sig. (2-tailed)	.008	.
		N	21	21

**. Correlation is significant at the 0.01 level (2-tailed).

Table 11. Correlations (Frontal plane trunk angle & knee valgus moment)

Correlations

			fronttrunkang	kneevalgusmoment
Spearman's rho	fronttrunkang	Correlation Coefficient	1.000	-.434*
		Sig. (2-tailed)	.	.049
		N	21	21
	kneevalgusmoment	Correlation Coefficient	-.434*	1.000
		Sig. (2-tailed)	.049	.
		N	21	21

*. Correlation is significant at the 0.05 level (2-tailed).

Appendix 3. Additional Materials

University of Chester

Faculty of Applied Sciences
Research Ethics Committee

frec@chester.ac.uk

16th August 2013

Dear Matthew,

Study title: **The effect of truck orientation upon knee joint loading during unanticipated sidestep cutting manoeuvres.**
FREC reference: **837/13/MS/SES**
Version number: **1**

Thank you for sending your application to the Faculty of Applied Sciences Research Ethics Committee for review.

I am pleased to confirm ethical approval for the above research, provided that you comply with the conditions set out in the attached document, and adhere to the processes described in your application form and supporting documentation.

The final list of documents reviewed and approved by the Committee is as follows:

Document	Version	Date
Application Form	1	July 2013
Appendix 1 – List of References	1	July 2013
Appendix 2 – C.V. for Lead Researcher	1	July 2013
Appendix 3 – Participant Information Sheet	1	July 2013
Appendix 4 – Participant Consent Form	1	July 2013
Appendix 5 – Risk Assessment Form	1	July 2013
Appendix 6 – Sample Size Calculation	1	July 2013
Appendix 7 – Experimental Set-up	1	July 2013
Appendix 8 – Equipment availability and proficiency	1	July 2013

Response to FREC request for further information and clarification		August 2013
Appendix 3 – Participant Information Sheet	2	July 2013
Appendix 9 – C.V. for Defender	1	July 2013

With the Committee's best wishes for the success of this project.

Chair, Faculty Research Ethics Committee

Enclosures: Standard conditions of approval.

Cc. Supervisor/FREC Representative

Physical Sciences Research Laboratories
Preparation for Student Research Project

Student name	Matthew Smith
Study title	The effect of trunk orientation upon knee joint loading during unanticipated sidestep cutting manoeuvres
Equipment & consumables required	3D motion analysis Kistler Force Plate Timing Gates Calibration Frame 3D Body Marker Sets Tape measure
Testing venue	Biomechanics Lab, Chester University

- [ ] The equipment required is currently available and working.
- [ ] The consumables are available, or will be ordered as required.
- [ ] Sufficient laboratory time should be available to complete the testing required.
- [ ] The student is proficient with the equipment he/she wishes to use.
- [ ] A risk assessment has been completed. Copy attached.

Signed ………………………………………………………………
Date……………………
(Technician)

Once completed, this form must be submitted to the Department Ethics Committee along with your Ethical Approval Application.