The Relationship Between Lower Limb Biomechanical Variables During Common Screening Tasks

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Thesis Summary

Abnormal lower-limb mechanics during functional activities have been reported as being associated with several knee injuries. Hence it is important to develop screening tests to identify healthy individuals who may be susceptible to knee injury and then to design individual intervention programmes. There is limited literature exploring the associations between lower-limb biomechanical variables during athletic tasks associated with knee-joint injuries. A better understanding of inter-task performance would offer insights into the consistency of motor patterns employed by healthy individuals during common screening tasks.

This thesis comprises four themed studies. The first study aimed to examine the reliability of using 3D motion analysis to measure the biomechanical variables during single-leg squats (SLS), single-leg landing (SLL), running and sidestep cutting tasks. The findings of first study revealed that within-day measurements are more reliable than those between days across all tasks, while transverse-plane variables are less reliable compared to other planes of movement.

The second study established reference values for lower-limb biomechanical variables during these tasks in a large population sample (90 healthy participants). Furthermore, gender differences in biomechanical variables were also assessed. Significant differences were noticed in knee-flexion, knee-valgus and hip-adduction peak angles across all tasks and both genders.

The third study examined the relationships between lower-limb biomechanical variables during these tasks. A significant relationship has been reported across all tasks between the following variables: peak knee-abduction angle and moment, hip-internal and hip-adduction rotation angles. The findings support the hypothesis that those individuals who exhibit misalignment strategies, specifically in frontal and transverse planes, during SLS & SLL will also show the same movements during running and cutting tasks. However, it must be stressed that the use of squat or landing alone should not be considered as a replacement to find individuals at risk of running or cutting mechanics since several variable showed weak or no correlation.

The final study aimed to examine the effectiveness of an augmented feedback protocol on SLS performance and if changing squat performance would be reflected in a change in performance in SLL, running and side-step cutting tasks. Training resulted in a significant reduction in knee-valgus angle and moment and hip-flexion angles during single-leg squatting. Additionally, these improvements remained a few days later, proposing motor patterns might have improved and these improvements would sustain, thus reducing the risk of injury in the longer time. Furthermore, significant reductions in knee-valgus angle and moment were also noticed in landing after squat feedback training, but no significant improvements were transferred to run and cut tasks.

This thesis has expanded the understanding about using 3D movement-analysis systems and established reference values when performing common screening tasks. Furthermore, feedback was used to improve performance strategies, which could reduce the risk of knee injuries in a quick and easy manner. However, the results of this study do not confirm that the alterations reported in biomechanical variables were solely due to the SLS feedback-training programme.

Chapter 1: Introduction

1.1. Introduction

Of all the lower-limb joints, injury to the knee joint complex sustains the highest percentage of injuries in sport (Hootman, Dick, & Agel, 2007; Powers, 2010; Starkey, 2000). The majority of anterior cruciate ligament (ACL) and patellofemoral pain syndrome (PFPS) injuries happen during non-contact and overuse mechanisms (Agel, Arendt, & Bershadsky, 2005; Olsen, Myklebust, Engebretsen, & Bahr, 2004), which are generally considered as being preventable if the injury mechanisms and factors lead to the injury can be recognised and prevention actions are taken.

Although the prevalence of ACL and PFPS injuries are relatively low comparing to other lower limb injuries, the short-term disability and higher risk of osteoarthritis (OA) associated with ACL and PFPS injuries have made exploration into their mechanisms, prevention and risk factors a focus for research. Notwithstanding this interest, no definite profile of the ACL or PFPS injured individuals has been determined; several things can possibly cause these injuries.

Risk factors for ACL and PFJ injuries can be categorised into three broad categories: anatomic, hormonal and biomechanical factors (Uhorchak et al., 2003; Griffin et al., 2000). The biomechanical variables of lower-extremity can be altered, and therefore an understanding of these factors has great possibility to decrease risk of injury. Dynamic knee valgus is a mixture of motions of the lower limbs, including transverse- and frontal-plane motions at the knee, hip and ankle, which contribute to lower limb alignment during loading maneuvers (Hewett et al., 2005). Furthermore, greater knee valgus angle is linked to PFPS during single-leg squat (SLS) and running tasks (Crossley, Zhang, Schanche, Bryant, & Cowan, 2011; Dierks, Manal, Hamill, &Davis, 2008) and with ACL during single leg landing (SLL) and side-step cutting tasks (Krosshaug et al., 2007; Hewett et al., 2005; Olsen et al., 2004).

Several investigations have documented lower-limb biomechanics during various functional movements which mimic the real situation of PFPS injuries, such as SLS and running (Baldon et al., 2011; Bazett-Jones et al., 2013; Besier, Lloyd, Cochrane, &

Ackland, 2001; David, Stergiou, & Stefanyshyn, 2015; Dwyer, Boudreau, Mattacola, Uhl, & Lattermann, 2010; Ferber, Davis, & Williams, 2003; Graci, Van Dillen, & Salsich, 2012; Horan, Watson, Carty, Sartori, & Weeks, 2014; Nakagawa, Moriya, Maciel, & Serrao, 2012a; Nguyen, Shultz, Schmitz, Luecht, & Perrin, 2011; Noehren, Pohl, Sanchez, Cunningham, & Lattermann, 2012; Weeks, Carty, & Horan, 2012; Willy & Davis, 2011; Yamazaki, Muneta, Ju, & Sekiya, 2010; Zeller, McCrory, Kibler, & Uhl, 2003; Zwerver, Bredeweg, & Hof, 2007), or ACL injuries, such as SLL and side-step cutting tasks (Ali, Rouhi, & Robertson, 2013; Beaulieu, Lamontagne, & Xu, 2008; Garrison, Hart, Palmieri, Kerrigan, & Ingersoll, 2005; Jones, Herrington, Munro, & Graham-Smith, 2014; Jorrakate, Vachalathiti, Vongsirinavarat, & Sasimontonkul, 2011; Kiriyama, Sato, & Takahira, 2009; McLean, Huang, & van den Bogert, 2005a; Nagano, Ida, Akai, & Fukubayashi, 2007; Orishimo, Kremenic, Pappas, Hagins, & Liederbach, 2009; Orishimo, Liederbach, Kremenic, Hagins, & Pappas, 2014; Pappas, Hagins, Sheikhzadeh, Nordin, & Rose, 2007; Pollard, Davis, & Hamill, 2004; Russell, Palmieri, Zinder, & Ingersoll, 2006; Schmitz, Kulas, Perrin, Riemann, & Shultz, 2007; Sigward & Powers, 2006a; Yeow, Lee, & Goh, 2010). The number of subjects participating in each of the aforementioned studies was limited, making the generalization of findings difficult. Consequently, further screening research on large-scale population to identify those who exhibit poor lower-limb biomechanics related with increased risk of injury is needed.

The majority of attempts exploring lower-limb biomechanics and its relation to knee injuries have been conducted using three-dimensional (3D) motion-analysis systems (Souza & Powers, 2009; Hewett et al., 2005; Ford, Myer, & Hewett, 2003). 3D motion-analysis allows researchers to quantify frontal, sagittal, and transverse planes of motion during different screening maneuvers and is considered to be the “gold standard” of motion analysing. For an outcome measurement to be valuable, it must provide stable or reproducible values with small errors in measurement (Rankin & Stokes, 1998). Understanding the reliability and measurement errors related with each of these screening instruments is critical (Batterham & George, 2003). Considering that the project’s main aims are to establish reference values for 3D lower-limb biomechanical variables during a number of screening tasks in a physically active population, and to find the links between those variables, it is important to utilise methods that provide stable and reproducible values with small errors in measurement.

A number of researchers have reported correlated the biomechanical variables between different screening performance tasks within the cohort (Harty, DuPont, Chmielewski, & Mizner, 2011; Jones et al., 2014; Kristianslund & Krosshaug, 2013; Whatman, Hing, & Hume, 2011). However, the outcomes of the aforementioned attempts only apply to female athletes; therefore, applying these findings to other populations should be done with caution. In reviewing the aforementioned research, no reports were found that investigated the correlation between dynamic knee-valgus variables in a large-scale population during distinctly different screening tasks and that linked to both ACL and PFPS injuries in a healthy population. Such data would offer further understandings into the potential poor biomechanics in causal factors associated to both injuries, and thus facilitate more effective screening of individuals at risk of these injuries.

A number of researchers have reported that feedback training can decrease some ACL and PFPS risk factors, such as knee-valgus angle and moment (Ford, DiCesare, Myer, & Hewett, 2015; Mizner, Kawaguchi, & Chmielewski, 2008; Munro & Herrington, 2014), increase flexion of knee joint (Herman et al., 2009; Onate et al., 2005), increase hip-flexion and -abduction angles (Herman et al., 2009) and reduce hip-adduction and -internal rotation angles (Willy, Scholz, & Davis, 2012). There seems to be general agreement on reducing vertical-peak ground-reaction forces after feedback training (Herman et al., 2009; Cronin, Bressel, & Fkinn, 2008; Onate et al., 2005; Prapavessis & McNair, 1999). Few investigations to date have studied whether the effect of augmented feedback of simple tasks such as SLS would translate into an improvement in performance in more complex tasks, such as running and cutting.

1.2. Thesis aims

- **General aim**

  The overall aim of this thesis is to study lower-extremity biomechanics during commonly assessed tasks in a healthy population.

- **Specific aims**

1. Investigate the reliability of using a 3D motion-analysis system to measure lower-limb kinematic and kinetic variables during single-leg squat, single-leg landing, running and cutting tasks.

2. Establish reference values for lower-limb biomechanical variables during a series of lower-limb loading tasks in a physically active population.

3. Investigate the relationship between lower-limb kinematic and kinetic variables across a number of lower-limb loading tasks in a physically active population.

4. Investigate the effect of an augmented feedback protocol on single-leg squat performance and if changing squat performance would be reflected in a change in performance in single-leg landing, running and sidestep cutting tasks.

1.3. Thesis structure

Chapter (1)

- Thesis Summary.
- Introduction.

Chapter (2)

- Literature Review.
- Thesis Aims.

Chapter (3)

Reliablity of using 3D motion analysis to measure biomechanical variables during SLS, SLL, RUN and CUT tasks.

(Study- 1)

Chapter (4)

Establish typical values for 3D biomechancial variables during SLS, SLL, RUN and CUT tasks.

(Study- 2)

Chapter (5)

Examine the relationship between 3D variables collected during SLS, SLL, RUN and CUT tasks.

(Study- 3)

Chapter (6)

Invetigate the effect of a feedback programme for biomechancial variables during SLS, SLL, RUN and CUT tasks.

(Study- 4)

Chapter (7)

- General discussion.
- Suggestions for future work.

Chapter 2: Literature Review

2.1. Incidence of ACL and PFPS injuries

The prevalence of anterior cruciate ligament (ACL) injury is only 0.1–0.4 per 1,000 athlete exposures in different sporting activities such as soccer (Mihata, Beutlet, and Boden, 2006; Agel et al., 2005; Mandelbaum et al., 2005) and basketball (Meeuwisse et al., 2003; Lombardo et al., 2005). Furthermore, it has been reported that there is only one ACL injury per 5,000 healthy individuals in Switzerland (De Loes et al., 2000). The highest risk of ACL injury is found to be among individuals between the ages of 15 and 25 years who are involved in cutting and pivoting sports (Myklebust et al., 1998), whereas the incidence of patellofemoral pain syndrome (PFPS) is larger, at 1.09 injuries per 1,000 athletic exposures (Myer, Ford, Khoury, Succop, & Hewett, 2010). Other researchers have reported that PFPS can affect up to 30% of young students between 13 and 19 years old (Blond and Hansen et al., 1998). The incidence of ACL and PFPS seems to be a lesser problem compare to other injuries, such as hamstring strains and ankle sprains, with occurrence rates of up to 3.19 per 1,000 exposures (Deitch al., 2006). However, the penalties of ACL and PFPS injuries, in terms of sport involvement, higher risk of OA and functional limitations, placing these injuries among the most serious sport injuries.

Several investigations have reported that females are 2 to 6 times more likely to experience ACL or PFPS injuries in comparison to male across a range of sports (Deitch, Starkey, Walters, & Moseley, 2006; Agel et al., 2005; Arendt, Agel, & Dick, 1999; Hewett, Lindenfeld, Riccobene, & Noyes, 1999). In their retrospective study, Lohmander et al. (2004) observed that around 75–80% of young females who had had an ACL injury 12 years earlier suffer from early onset knee osteoarthritis (OA), pain and functional limitations. Another study reported that around 40% of individuals with an ACL tear had signs of knee OA 6 to 11 years after their injury (Myklebust, Holm, Maehlum, Engebretsen, & Bahr, 2003).

2.2. Common mechanisms of ACL and PFPS Injuries

Around 75% of ACL injuries occur during game time and up to 70% of ACL injuries happen in non-contact circumstances (Agel et al., 2005; Olsen et al., 2004). The majority of ACL injuries happen during single leg landing (SLL), deceleration or changing-direction manoeuvres (Olsen et al., 2004; Boden, Dean, Feagin, & Garrett, 2000; Myklebust, Maehlum, Holm, & Bahr, 1998). Furthermore, most non-contact ACL injuries appear to occur close to foot strike, in knee-abduction and minimal-flexion positions (Boden et al., 2000; Olsen et al., 2004). The incidence of ACL injuries is relatively high in sports such as football, basketball, netball, handball and volleyball, which are characterised by frequent landing and decelerating and changes in direction (Griffin et al., 2000).

PFPS injury is caused by patella maltracking during knee flexion and extension actions (Powers, Ward, Fredericson, Guillet, & Shellock, 2003). Maltracking of patella lead to increased patellofemoral joint (PFJ) contact pressure, and this leads to a pathological effect. PFPS patients exhibit more PFJ stress during SLS as a result of a reduction of PFJ contact area (Farrokhi, Colletti, & Powers, 2011).

Figure 2.1. Example of an ACL injury during a sidecut task (Alentorn-Geli et al., 2009)

2.3. Risk factors for ACL and PFPS Injuries

The risk factors for ACL injuries have been explained previously; Uhorchak et al. (2003) and Griffin et al. (2000) simplified these and divided risk factors into three broad categories: anatomic, hormonal and biomechanical. Anatomical risk factors are based on the lower limb alignment and ACL geometry, such as quadriceps angle (Mizuno et al., 2001; Boden et al., 2000), increased knee-joint anterior laxity (Griffin et al., 2006), impingement of the ACL against the intercondylar notch (Fung, Hendrix, Koh, & Zhang, 2007). The hormonal profiles of males and females may contribute to the disproportion in the rates of injury. Several studies have documented that female sex hormones can influence the mechanical properties of the ACL, as well as the flexibility of tendon and muscles around the knee joint (Slauterbeck et al., 2002; Myklebust et al., 1998). Biomechanical risk factors are described in detail in Section 2.4.

The risk factors leading to PFPS injury have concentrated on misalignment of patella as a main injury risk factor. Four main factors influence the patella alignment, i.e. vastus medialis muscle properties, illiotibial band (ITB) tightness, increasing Q-Angle and biomechanical factors. Tang et al. (2001) reported that patients with PFPS showed a significantly reduced activation ratio between the vastus lateralis and vastus medialis obliqus in comparison to asymptomatic persons during an open kinetic-chain task. With regard to ITB, patients with PFPS display a significantly reduced the length of ITB, when measured using a modified obers test (Hudson & Darthuy, 2009). These findings suggest the associations between the length of ITB and the positioning of the patella, and therefore ITB tightness may associate with PFPS development. Furthermore, an increased quadriceps angle is linked to lateral PF contact pressure and patellar dislocation, while reducing the Q angle may not shift the patella medially, but rather increase the medial tibiofemoral contact pressure through increasing the knee valgus alignment, which may develop PFPS (Mizuno et al., 2001; Waryasz & McDermott, 2008).

2.4. Biomechanical risk factors for ACL and PFPS

2.4.1. Frontal- and transverse-planes motion

Alteration in hip and knee frontal- and transverse-plane motion and loading during functional activities are often described as “apparent knee valgus”, “dynamic valgus” or “dynamic misalignment”. Dynamic valgus is a combination of hip internal rotation and adduction, knee valgus and foot pronation, as shown in Figure 2.2. This pattern has been suggested to be a critical factor in both ACL and PFPS injuries (Willson & Davis, 2008a; Hewett et al., 2005; Ireland, 1999).

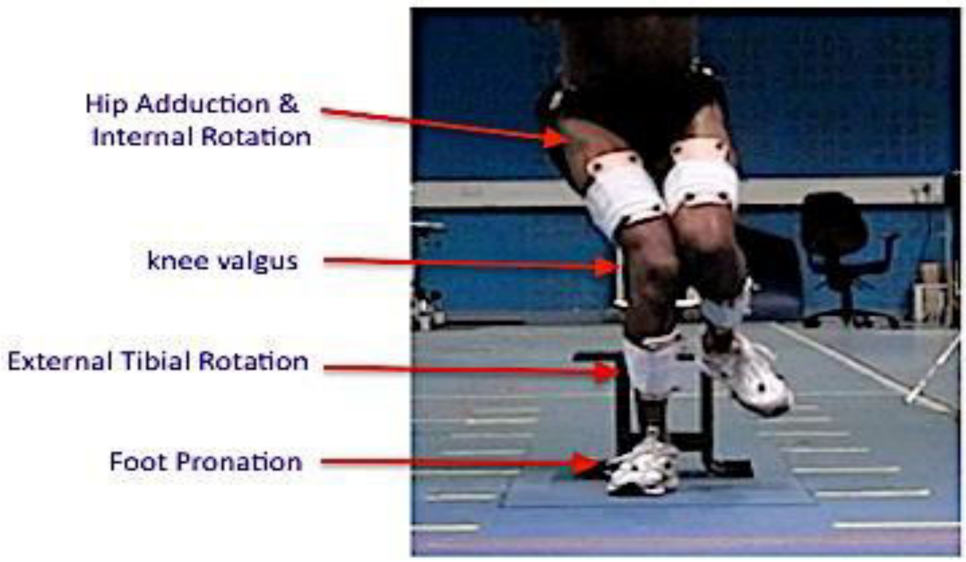


Figure 2.2. Dynamic knee-valgus pattern

Hip internal rotation (HIR) has previously been reported as a contributing factor in dynamic knee valgus position (Powers et al., 2003; Ireland, 1999). Hip-internal rotation leads to knee-external rotation, which in turn causes ACL impingement on the lateral femoral condyle wall, consequently increasing the risk of injury (Fung et al., 2007). Higher hip internal rotation motion can also influence the position of the patella and increase the PFJ forces (Powers, 2010; Lee, Morris, & Csintalan, 2003). Previous studies have reported that females with PFPS performed SLS and running tasks with a higher hip-internal rotation angle compares to control groups (Nakagawa et al., 2012b; Souza & Powers, 2009).

Hewett et al (2005) reported that strong correlation between higher hip-adduction moment and knee-valgus moment in ACL injured individuals. Several authors reported that PFPS patients display higher hip adduction in comparison to controls individuals during different screening tasks (Nakagawa et al., 2012b; Souza & Powers, 2009). Where higher hip adduction motion are noticed, they are between 2.4° - 5.5° greater in PFPS patients. Higher hip adduction may lead to an increase in Q angle, which is a static calculation of the location of the quadriceps muscles forces acting on the patella (Mizuno et al., 2001). A larger Q angle is considered to increase the possibility of sustaining PFPS by pulling the quadriceps laterally on the patella leading to increase the contact pressure of lateral PF (Mizuno et al., 2001).

Knee-valgus angle (KVA) also refers to the knee-abduction angle. Several authors have documented that greater knee-valgus angle during running, landing and sidestep cutting manoeuvres is linked to, and predicts, ACL injuries (McLean, et al., 2005a; Hewett et al., 2005; Boden et al., 2000) and PFPS injuries (Myer, Ford, Khoury, et al., 2010; Boling et al., 2009; Stefanyshyn et al., 2006). Hewett and colleagues (2005) conducted a prospective attempt on 205 female athletes from different sports. At the end of the season, there were nine ACL injuries. Those who had torn their ACL demonstrated significantly higher knee-valgus angles during a double jumping at baseline screening. Females who had torn their ACL exhibited a 5° knee-valgus angle at initial contact and a peak knee valgus of 9°, which was 8.4° higher than healthy females at initial contact and 7.6° greater at peak value of knee-valgus angle.

Higher knee-internal rotation angle (KIR) can lead to more strain on the ACL (Oh, Lipps, Ashton-Miller, & Wojtys, 2012); a previous investigation reported that knee-internal rotation does not cause ACL impingement whereas external rotation does (Fung et al., 2007). Furthermore, external knee rotation can also cause more lateral-patella tracking (Noehren, Barrance, Pohl, & Davis, 2012), increased PFJ forces (Lee et al., 2003) and a reduced PFJ contact area.

Tiberio (1987) reported that to extend the knee coupled with internal rotation of the tibia, the femur must also internally rotate and this leads to a higher hip-adduction angle (Tiberio, 1987). A recent attempt found some correlation between hip-adduction angle and foot eversion during walking (Barton et al., 2012). The investigators summarised that the foot kinematics influence the femoral motion and this could be a risk factor for PFPS. Witvrouw et al. (2000) reported that decreasing the flexibility of the calf muscles causes compensatory pronation of the foot to attain the required dorsiflexion range of motion.

In summary, increased peaks of knee valgus and internal rotation, hip internal rotation and adduction movements together with superficial knee flexion during landing or changing-direction tasks are frequently seen in ACL injuries (Koga et al., 2010; Hewett et al., 2009; Olsen et al., 2004; Boden et al., 2000; Krosshaug et al., 2007; Olsen, Myklebust, Engebretsen, & Bahr, 2004) and place more strain on the ACL (Berns, Hull, & Patterson, 1992; Markolf, Burchfield, Shapiro, Shepard, Finerman, & Slauterbeck, 1995). Hewett et al. (2005), however, reported that only knee valgus angle and moment and vertical GRF during a drop-jump task were significant predictors of ACL injuries. Similar changes in lower-limb posture can increase the load applied on the PFJ, with reduced knee flexion angle, increased hip-internal rotation and increased knee-valgus load having been linked to the PFPS development (Boling et al., 2009; Myer et al., 2010; Stefanyshyn et al., 2006). Combining these actions has been named as the dynamic-knee valgus position (Munro et al., 2012b; Hewett et al., 2005) and females often exhibit postures which contribute to dynamic-knee valgus more than their men counterparts, and this is widely believed to be one of the primary causes for the disproportion in injury rates (Hewett et al., 2005; Hewett, Myer, & Ford, 2004; Ferber, Davis, & Williams, 2003; Ford, Myer, & Hewett, 2003; Zeller, McCrory, Kibler, & Uhl, 2003).

2.4.2. External valgus moment (KVM)

High-knee abduction moment is a common risk factor for ACL & PFPS injuries (Hewett et al., 2005; Stefanyshyn et al., 2006). Prospectively, Hewett et al. (2005) found that female young athletes who sustained an ACL injury had a peak value of knee-abduction moment during landing 2.5 times greater than that of uninjured athletes. Furthermore, this study found that KVM was a stronger predictor of ACL injury than knee-flexion angle. Furthermore, Fukuda and colleagues (2003) reported that an extra 10Nm of isolated valgus load causes more pressure on the ACL in cadaveric knees.

In a prospective study, Stefanyshyn et al. (2006) supported the association between excessive knee-valgus moment and PFPS. In their study, the participants who developed

PFPS after six months of running showed a significantly greater knee-abduction impulse during baseline measurement compared to that of an age-matched group who did not develop PFPS. Paoloni et al. (2010) compared frontal-plane kinetic patterns of the knee between young adults with PFPS and age-matched healthy controls while the participants walked 10 m on a level surface at a self-selected speed. Using three-dimensional (3D) kinetic analysis, the study found that patients with PFPS displayed significantly higher knee-abduction moment than healthy control group during loading of the stance leg.

Figure 2.3. Video-graphic depiction of an athlete with a kinematic pattern that is likely to display high knee-valgus moment (Myer et al., 2010)

2.4.3. Sagittal-plane loading

The loading on the ACL can be altered by a change in the sagittal plane of motion. Previous literature has found that the greatest stain on the ACL often happens near to full extension position (Berns, Hull, & Patterson, 1992; Markolf et al., 1995). Previous studies have reported that females often land with less than 25° of knee flexion, which on average is 5–10° less than their male counterparts (Chappell et al., 2005; Decker, Torry, Wyland, Sterett, & Richard Steadman, 2003; Malinzak, Colby, Kirkendall, Yu, & Garrett, 2001; Malinzak et al., 2001). Moreover, females exhibit smaller flexion angles and less force absorption at the hip, which may lead to increased knee loading (Chappell et al., 2005; Decker et al., 2003). Both in-vivo and in-vitro trials found that the anterior translation and ACL initiated by quadriceps contraction peak at around 15-30° of knee flexion angle (Beynnon et al., 1995). A possible explanation for this may link to the angle between the patellar tendon and the tibial axis (Pandy & Shelburne, 1997).

The reduction in knee flexion in female athletes together with increased the activation of quadriceps and reduced activation of hamstring may all contribute to more strain on the ACL and increased likelihood of injury. The ACL can be protected by the posterior GRF and synergistic muscle contraction (Markolf et al., 1995; McLean et al., 2004). Biomechanical modelling has proved that knee frontal plane loading is more important in ACL injuries (McLean et al., 2004). As stated earlier, knee frontal or transverse motion can significantly increase the strain placed on the ACL (Markolf et al., 1995). This suggests the importance of higher dynamic knee-valgus motion during screening tasks as a possible mechanism for ACL injuries.

2.4.4. Vertical ground-reaction force

Measuring ground-reaction force (GRF) demonstrates the amount of loading on the body that takes place during impact. The weight of the person acts in a downward direction, while the GRF is upwards on impact. The GRF comprises a three-component vector representing forces in the X (anterior-posterior), Y (medial-lateral), Z (vertical) directions (Rowe, Durward, & Baer, 1999).

McNair and Prapavessis (1999) provided normative values for vertical GRF for a landing task in an adolescent population (154 males and 80 females). Their findings suggest that the average vertical GRF should be at 4.6 (±1.8) times body weight for individuals participating in sports involving jumping and landing activities and 4.4 (±1.5) times body weight for subjects in sports not involving jumping activities. In their prospective investigation, Hewett and colleagues (2005) found that female athletes who sustained an ACL tear had a 20% higher ground-reaction force during a jump-landing task compared to their uninjured counterparts. On the other hand, Boiling et al. (2009) observed 1597 participants prospectively during a period of 2.5 years at the United States Naval Academy. They reported that participants who developed PFPS injury had a significantly less vertical ground-reaction force during a jump-landing task at baseline screening than those who did not develop PFPS (2.6 vs 2.9 times BW).

2.5. ACL and PFPS injury prevention programmes

Studies have begun to examine possible routes towards the prevention of ACL and PFPS injuries by modification of the risk factors. Neuromuscular intervention training programmes and movement-technique alterations are at the forefront of research in this area, because these approaches address modifiable risk factors. Neuromuscular training programmes have shown some success in decreasing potential biomechanical risk factors (Myer et al., 2007; Pollard, Sigward, Ota, Langford, & Powers, 2006; Lephart et al., 2005; Irmischer et al., 2004).

Not all interventions programmes have been effective in decreasing ACL and PFPS injury rates. For instance, Pfeiffer et al. (2006) performed a randomized controlled trial and reported no decline in the ACL injury rate in female players who joined a training programme which consisted of 20 minutes of plyometric-based exercise two times per week focusing on lower limb alignment and mechanics during landing from a jump and deceleration with changing direction while running. Myer et al. (2007) found that female athletes who are at high ACL tear risk, as classified by their greater knee-valgus moment during drop vertical jump task (DVJ), were able to significantly reduce their knee-valgus moment after six weeks of training that included plyometric, core strengthening, balance training, speed and resistance training. In spite of the fact that women reduced their knee-valgus moment, they still did not reduce their moments to low-risk values.

In a double-blinded RCT, Crossley and colleagues (2002) divided 71 patient with PFPS into intervention and control groups. The intervention group received a protocol included quadriceps strengthening and retraining and patella taping and mobilisation, whilst the control individuals received a sham intervention. A questionnaire revealed that the intervention group had significant reductions in pain and improvements in function, while no changes were noticed in the control group. However, it is not clear whether the combination of interventions or only one intervention caused the improvements. Supporting this, Herrington (2000) found that only applying patella taping significantly improved function and reduced pain.

2.5.1. Feedback training

Another approach that is being explored in this area is the effect of feedback training on modifying potential risk factors. Feedback can be traditionally classified into two types: sensory and augmented. Sensory feedback is information naturally available from performing a motor task and is received through the performer's sensory systems, e.g. hearing, vision, or touch (Prapavessis & McNair, 1999). Augmented feedback is external information about the motor task that supplements naturally available information. Augmented feedback is usually under the control of a skills instructor (e.g. physiotherapist or coach) who can control it in many different ways to enhance learning or performance, or both.

Augmented feedback is defined as information provided by an external source that can be added to intrinsic feedback to alter activity patterns of the body. Feedback methods include verbal instructions, real-time visuals (Davis, 2005; White et al., 2005 Dingwell et al., 1996; Missier et al., 1989) or auditory information (Cronin et al., 2008; McNair, Prapavessis, & Callender, 2000). Other studies have used verbal instructions and videotape reviews (Onate et al., 2005; Onate et al., 2001).

A number of researchers have reported that feedback training can reduce knee-valgus angle and moment (Barrios, Crossley, & Davis, 2010; Ford et al., 2015; Mizner et al., 2008), increase knee flexion angle (Herman et al., 2009; Onate et al., 2005), increase hip-flexion and -abduction angles (Herman et al., 2009) and reduce hip internal-rotation and adduction angles (Willy, Scholz, et al., 2012). There seems to be general agreement on reducing vertical peak-ground reaction forces after feedback training (Herman et al., 2009; Cronin et al., 2008; Onate et al., 2005; Prapavessis & McNair, 1999). A summary of feedback studies is reported in Table 2-1.

In preliminary work, Ford et al. (2015) compared the effects of two different modes of visual feedback during a squat on drop vertical-jump landing mechanics. Four young females (high-school soccer players) received visual feedback depicting their knee valgus moment values while performing a double-leg squat task. Following training, knee abduction moment reduced by 33% during a drop vertical jump compare to baseline screening, and maximum knee-abduction angle decreased by 31.5%, suggesting a carryover of the effects of feedback between tasks. In separate training, participants also received visual kinematic feedback regarding knee-valgus angle, but that technique only helped the athletes hit the target KAM range 29.3% of the time. Following the training, knee-valgus angle and moment were not significantly different from the baseline.

The augmented feedback model used by Onate et al. (2005) combined visual and verbal feedback. By using this mode of feedback, individuals can compare their performance against an expert. This mode of feedback has proved effective in reducing the knee-valgus moment as well as the vertical GRF (Onate et al., 2005). Several authors have found that verbal instructions alone can increase knee-flexion angle and reduce vertical GRF (Milner et al., 2012; Mizner et al., 2008), although it is unknown whether these changes in knee-valgus angles can be achieved for longer periods.

A protocol combining verbal and visual feedback can have the same effect as verbal feedback alone on clean power performance (Rucci and Tomporowski, 2010). Further, verbal feedback alone made higher alterations in performance than video only, supporting a verbal mode of feedback being a main factor may lead to improvement in performance. However, their visual and verbal feedback programme involved only video of subjects' performance. Onate et al. (2005) found that a self-and-expert model was more effective than viewing participants' performance only. It may be that the most key feature of a video-and-verbal feedback practice, which would improve the performance, is expert performance as well as verbal instructions.

Recently, Munro & Herrington (2014) used the same feedback protocol as Onate (2005), with a landing-error scoring system (LESS), to find out whether this would decrease frontal plane projection angle (FPPA) during a drop-jump task (DJ) a single-leg squat (SLL) task in 28 recreational athletes (eight were used as a control group). A significant reduction in vertical GRF (2.73 vs. 2.55 * BW) and FPPA (4.0° vs. -19.9°) was noticed after feedback training in the experimental groups. There were no changes noticed in the control group. These findings would have been more interesting if they addressed whether the immediate changes in performance seen were sustained for a longer period of time, thus more research is required in this topic.

The effect of feedback over longer periods of time has not been investigated properly (Willy et al., 2012; Onate et al., 2005; Prapavessis et al., 2003). Onate et al. (2005) tested the effectiveness of their protocol seven days after the first day of testing. Willy et al. (2012) noticed that improvement after feedback was retained for up to three months in the absence of feedback. Conversely, Prapavessis et al. (2003) did not notice any effect of feedback (instructions) after three months compared to baseline testing.

To dare, previous literature has tended to focus on the effectiveness of feedback training on specific tasks (Table 2.1, Section 2.7.1). Willy and Davis (2011) found that a motor-learning and hip-strengthening intervention improved strength and SLS performance, but these changes were not transferred to improved running performance. However, it is not clear whether the changes were due to motor learning or increase in muscular strength.aisal1234 In another attempt, Willy et al. (2012) found that mirror and verbal feedback during a treadmill activity results in improvements to running performance (reduction in hip-adduction angle and moment), and these improvements were transferred to SLS and step down. It is not known whether these changes would occur if their feedback training was based on a simple task, such as SLS training instead of treadmill running. Further research about the sustainability and transferability of these improvements is required to prove using the feedback training as a tool for reducing the injury risk.

When assessing SLS mechanics, distal and proximal variables should be taken into account as these can influence the loading on the lower limbs (Herrington and Munro, 2014; Myer et al., 2008). A qualitative analysis of single-leg squats (QASLS) takes these variables into account. Specifically, it involves movement strategies occurring in the feet, knees, pelvis, trunk and arms (Herrington and Munro, 2014). High scores on QASLS, which indicates poor SLS performance, are linked to 3D motion that may increase the injury risk (Herrington and Munro, 2014). Therefore, using QASLS as a source for feedback is likely to improve lower limb biomechanics when performing SLS task.

Table 2.1. Summary of Feedback studies

Study	Population	Tasks	Feedback	Finding
Munro & Herrington (2014)	28 students	SLL & DJ	1. Combination of expert & self	1. Significant reduction in (2.73 vs 2.55 * bw) and FPPA (4.0° vs -19.9°) post feedback. 2. No changes were evident in the control group.
Ford et al. (2015)	4 F-athletes	DVJ	1. Kinetic visual-FB 2. Kinematic visual-FB	1. Kinetic: reduced KVM 33% & KVA 31.5%. 2. Kinematic FB; no sig. difference.
Willy et al. (2012)	10 F-runners (with PFPS)	RUN, SLS & step	Mirror & verbal FB	1- Reduction hip-add. & hip-abd. moment. 2- Improvement in pain & function. 3- Both remain during the 3rd visit (3 m).
Crowell et al. (2011)	10 runners	Treadmill RUN	Real-time video (Pre, post & 1 month)	1- Reduction in tibial acceleration (50%), VGRF (20%) & force rate (20%). 2- Reduction maintained after 1 month.
Barrios et al. (2010)	8 healthy participants with varus	Treadmill WALK	Video FB (Pre, post & 1 month)	1. 19% reduction in KVM & 2° add angle. 2. Increase in hip int. rot. (8°) & hip adduction (3°).
Dempsey et al. (2009)	12 M- athletes	45° CUT	Oral and visual-FB	1. 36% reduction in peak KVM. 2. No change in flexion & int. rot Moment.
Herman et al. (2009)	58 F- athletes	Double-leg landing	1. Strength +FB 2. FB no strength	1- In FB vs GRF: reduced & increased hip flex. & abd., knee flex. & ant. shear force. 2- Hip abd. increase in ST-FB only.
Cronin et al. (2008)	15 F. volley-ballers	Leg-spike jump	Expert	1- 23% reduction in vertical GRF. 2-No differences in ML & AP forces.
Mizner et al. (2008)	37 F- athletes	DVJ	Verbal inst.	1. Increased knee-flexion angle. 2. Reduction vs GRF, KVA & KVM.

Continued

Table 2.1: Continued

Study	Population	Tasks	Feedback	Finding
Walsh et al. (2007)	25 basket-ballers	Drop jumps	Expert	1. No diff. in men after instruction. 2. Sig. reduction in KVA & force in females.
Onate et al. (2005)	51 rec. athletes	Jump, land	3 groups: expert, self & combination.	1. Self & comb. reduced GRF & increase knee FLX. 2. Exp. did not change more than control.
Prapavess et al. (2003)	61 students	Double-leg landing	1_Inst. & aud. FB 2_Control	1. FB reduced GRF during sessions 2-4. 2. No diff. between groups at session 5.
Cowling et al. (2003)	24 F- athletes	SLL	Verbal inst.	1. Increased knee flexion, reduce vs GRF. 2. No change in muscle activity.
Onate et al. (2001)	63 students (42 f.)	Vertical jump	4 groups: augmented, sensory, CON I (2 mins) & CON II (1 wk.)	Aug. reduced vs GRF in both sessions (2 mins & 1 week) compared to SEN, CON I, & CON II.
Prapavess et al. (1999)	91 students (35 f.)	Jump (30 cm)	1 AUG 2 Sensory	1. Sig. reduction in GRF with aug group (4.5 vs 3.5) & with sensory (4.5 vs 4.3).

M = Males; F= Females; athle = athletes; DVJ = Drop Vertical Jump; FB= Feedback; Inst= Instruction; KVM= Knee Valgus Moment; KVA= Knee Valgus Angle; GRF= Ground Reaction Force. VGRF= Vertical Ground Force. AP= Anterior-Posterior Force; ML= Medial-lateral Force.

2.6. Laboratory Assessment of Lower Limb Motion

2.6.1. Screening Tasks

It is noticeable in the previous literature that several movement tasks have been used to assess biomechanical risk factors for PFPS, including single-leg squatting (Whatman et al., 2011; Dwyer et al., 2010; Zwerver et al., 2007; Hass et al., 2005; DiMattia, Livengood, Uhl, Mattacola, & Malone, 2005; Zeller et al., 2003) and running (Queen et al., 2006; Ferber et al., 2003; Malinzak et al., 2001), and biomechanical risk factors for ACL injuries including single-leg landing (Yeow et al., 2010; Pappas, Sheikhzadeh, Hagins, & Nordin, 2007; Hass et al., 2005; McLean et al., 2004; Ford et al., 2003; Lephart, Ferris, Riemann, Myers, & Fu, 2002; Malinzak et al., 2001; Myklebust et al., 1998) and cutting tasks (Vanrenterghem, Venables, Pataky, & Robinson, 2012; McLean, Walker, & van den Bogert, 2005b; McLean et al., 2004; Pollard et al., 2004; Houck, 2003; Houck & Yack, 2003; Malinzak et al., 2001; Colby et al., 2000; Schot, Dart, & Schuh, 1995; Andrews, McLeod, Ward, & Howard, 1977).

2.6.1.1. Single-leg squat (SLS)

The SLS is a very simple test of knee alignment that often used in a clinical setting (Willson, Ireland, & Davis, 2006). During the descend phase of the squat, the body weight helps to pull the individual into a knee-flexed position; therefore, the quadriceps muscles act eccentrically to control knee flexion (Shields et al., 2005; Zeller et al., 2003). The SLS is said to stimulate a common athletic situation, i.e. requiring control of the body over a planted leg, prompting Claiborne et al. (2006a) to describe it as a controlled, yet dynamic, manoeuvre that can be extrapolated to many functional actions, such as single leg landing, running and changing direction tasks.

Patients with PFPS diagnosed demonstrate knee valgus during a squat test, which may be related to imbalance in the soft tissue and biomechanical misalignment of the lower extremities (Willson et al., 2006). The SLS, therefore, is an appropriate functional task to investigate in relation to PFPS as it concerns the injury mechanism, aggravating factors, assessment, diagnosis, treatment and rehabilitation and the evaluation of treatment progression.

Yamazaki et al. (2010) compared the 3D angles of SLS between ACL injured individuals and controls. Sixty-three ACL injured patients (32 male, 31 female) performed single-legged half squats the day prior to ACL reconstruction, and these were compared to 26 healthy control individuals with no knee injuries. When comparing the injured and uninjured legs within participants, the injured leg of both male and female individuals showed more knee adduction than the uninjured leg. Gender differences indicate that more external hip rotation (M = 38.8° ± 12.6°; F = 7.9° ± 49.3°) and knee varus (M = 16.9° ± 15.1°; F = 8.9° ± 8.2°) were present in the female subjects compared to males for both the injured and uninjured legs. The injured leg of the male individuals showed less knee and hip external rotation angles, less knee flexion and more knee varus than those of the uninjured leg of the male subjects.

Gender disparities when performing SLS have been noticed (Table 2.2). When compared to their male counterparts, females perform SLS tasks with less knee flexion (Dwyer et al., 2010), greater knee valgus (Zeller et al., 2003), more peak hip-adduction angle (Zeller et al., 2003) and more external hip rotation (Yamazaki et al., 2010). Women also show a more erect position (less torso flexion) than men (Graci et al., 2012). It has been argued that this posture may expose women to the risk of ACL injury by increasing the demand on the quadriceps to maintain control of the centre of mass (Griffin et al., 2000).

As can been seen in Table 2-1, there is no official standard for an SLS. Zeller et al. (2003) instructed their subjects to stand on their dominant extremity, cross their arms over their chest, squat down as far as possible and return to a single-leg stance position without losing their balance. This was to be done within five seconds. Their protocol does not state whether the five SLSs performed were to be done concurrently, without a rest between them, in order to rule out fatigue effects. Herrington (2013) asked his participants to squat down as far as possible, to at least 45° of knee flexion, but not more than 60°, for 5 seconds. The angle of knee-flexion was measured using a goniometer during practice trials (maximum of three). There was also a counter for each participant over this 5-second period in which the first count initiated the movement, the third indicated the lowest point of the squat and the fifth indicated the end of the movement, before returning to the start position.

Claiborne et al. (2006a) asked their subjects to squat to approximately 60°, but it is unclear how subjects knew when they had reached 60°. Their squats were, however, controlled by the same investigator. They were also non-consecutive, with a two-minute rest after each SLS, to avoid fatigue. Five to seven SLSs were done in order to obtain three acceptable trials. Yamazaki et al. (2010) instructed their participants to cross their arms over their chest and perform a half squat while keeping correct balance, with the duration of the squat being ten seconds or less. Subjects performed two single-leg half squats with both the injured and uninjured legs, while subjects in the control group performed squats with the dominant leg.

DiMattia et al. (2005) were more specific in their method for SLS, ensuring that the arms were in a standard position (straight out in front of the subject at 90°); the contralateral leg was positioned at 45° hip flexion and 90° knee flexion off the ground and each SLS, lasting six seconds, was limited to 60° of knee flexion for the dominant leg. There is, therefore, a range of methodologies for an SLS. Dwyer et al. (2010) instructed their participants to squat down as far as possible and return to a single-leg stance without losing their balance, as they believed this better represented a clinical setting.

Biomechanical studies of SLS have focused on narrow demographic healthy cohorts (Tables 2.2) or included participants with musculoskeletal problems such as PFPS (Herrington, 2014; Powers et al., 2003; Willson & Davis, 2008a, 2008b; Willson et al., 2006) or ACL injuries (Yamazaki et al., 2010). Whilst extensive research has been carried out on SLS biomechanics, no single study has provided reference values for lower-limb biomechanical variables in a large-scale healthy population. Such information could be used to assess previous and upcoming research, especially intervention studies, and also by practitioners who use SLS tasks to evaluate individual performance during training or rehabilitation.

Table 2.2. Summary of literature reporting 3D variables during SLS tasks in Healthy Participants

Study	N	SLS technique	Joint angles (degrees)					Moments (Nm/Kg)	
			Knee flexion	Knee valgus	Knee Int. Rot.	Hip adduction	Hip int. rot.	Knee flexion	Knee valgus
Zwerver et al. (2007)	5 F+M	Squat on dominant leg to max. knee flexion	67	-	-	-	-	0.23	-
Zeller et al. (2003)	9 F	Squat as far as possible then stand in a balanced position	95.4	-7.0	-	17.8	-	-	-
	9 M		89.5	-5.1	-	14.6	-	-	-
DiMattia (2005)	50 M+F	Squat to 60° of KF (depth limited by a block)	-	-4.0	-	8.0	-	-	-
Graci et al. (2012)	9 F	Squat on right leg to max. KF (L leg kept back)	69.7	-1.3	-	17.3	-1.0	-	-
	10 M		76.4	7.0	-	13.5	-0.7	-	-
Horan (2014)	22 M+F	Squat slowly with arms across chest (no depth limit)	90.1	-	-	14.7	-15	-	-
Nguyen et al. (2011)	60 M+F	Squat to 60° of KF (5-sec. count)	-	-0.1	-	11.4	-2.3	-	-
Richards (2008)	10 F+M	Squat slowly to 90° of KF (Self-assessed)	70.9	-	-	-	-	1.18	-
Weeks et al. (2012)	9 F	Squat with arms across chest	71.5	-	-	20.8	-1.2	-	-
	13 M		86.2	-	-	15.5	-5.5	-	-

Table 2.2: continued

Study	N	SLS technique	**Joint angles (degrees)**					**Moments (Nm/Kg)**	
			Knee flexion	Knee valgus	Knee Int. Rot.	Hip adduction	Hip. Int. Rot.	Knee flexion	Knee valgus
Willy (2011)	11 F+M	Squat to 60° of KF (arms held horizontal)	-	-	-	11.4	-6.6	-	-
Yamazaki et al. (2010)	12 F	Half squat in a balanced poistion (arms across chest)	66.2	-	-	-	-	-	
	14 M		77.8	-	-	-	-	-	-
Baldon et al. (2011)	16 F	Squat on dom. leg to 75° of KF (using an adjustable support)	-	-4.7	-	4.16	2.46	-	-
	16 M		-	-0.3	-	0.01	0.45	-	-
Deursen et al. (2014)	16 F+M	NA	74		-	-	-	0.06	-
Nakagawa et al. (2012b)	20 F	Squat ≤60º of KF for 4+ sec. without losing balance	65.2	-7.2	-	14.3	9.7	-	-
	20 M		67.4	-4.2	-	7.2	9.5	-	-
Dwyer et al. (2010)	21 F	Squat on dom. leg as low as possible in a balanced position	60.0	-12.4	-	22.4	-	-	-
	21 M		66.8	-14.1	-	18.3	-	-	-
Silva et al. (2014)	22 F	Squat on dom. leg as low as possible (4-sec. count)	62.1	-6.11	-	22.3	-2.5	-	-
	22 M		65.3	3.82	-	16.6	1.85	-	-

M =males; F = females; Int. Rot. = internal rotation; sign conventions (- knee valgus angle; + knee flexion angle; + hip adduction angle; + hip and knee internal rotation angles; + knee flexion and valgus moments).

2.6.1.2. Single-leg landing

Single-leg landing (SLL) is a common manoeuvre in athletic activity, and also a common mechanism for an ACL injury (McLean et al., 2004; Pollard, Sigward, & Powers, 2010). Misalignment of the lower extremities may occur during landing, which could potentially be due to an inefficiency in neuromuscular control (McLean et al., 2004).

The type of landing technique that an individual exhibits as well as how they absorb the force upon landing may be associated with the potential for experiencing an ACL injury (Cortes et al., 2007; McLean et al., 2004; Yu, Lin, & Garrett, 2006). Females tend to land in a more erect position when landing from a jump (Blackburn & Padua, 2009; Cortes et al., 2007). This position is associated with greater ground-reaction forces (GRFs) and more strain being placed on the ACL (Blackburn & Padua, 2009). High GRFs require a greater amount of eccentric quadriceps activation to counter the force without sustaining an injury (Blackburn & Padua, 2009). Males, on the other hand, demonstrate greater knee flexion and ankle dorsiflexion on ground contact during landing than females, thus reducing GRF (Cortes et al., 2007).

DeVita and Skelly (1992) observed that an erect landing resulted in reduced knee flexion (approximately 77°) while a softer landing resulted in an increased knee-flexion angle (approximately 117°), and so a soft landing resulted in a smaller ground-reaction force. Hewett et al. (2005) observed a relationship between peak GRF during landing and ACL injury. Among adolescent basketball, volleyball and soccer players, those with ACL injuries had a 20% greater peak-ground reaction force when compared to healthy controls (Myer et al., 2005). These studies indicate that landing with a greater vertical GRF increases the risk of sustaining an ACL injury.

Fong et al. (2011) evaluated the relationship between ankle dorsiflexion and landing biomechanics. Thirty-five healthy volunteers (17 male, 18 female) were recruited. The results demonstrated a significant correlation between ankle dorsiflexion and knee-flexion displacement ($r = 0.646$, $P = 0.029$) and between vertical ($r = -0.411$, $P = 0.014$) and posterior ($r = -0.412$, $P = 0.014$) ground-reaction forces. The authors suggest that greater knee displacement and smaller ground-reaction forces during landing were indicative of a landing posture consistent with reduced ACL injury risk by limiting the forces the lower limbs must absorb. The studies of both DeVita & Skelly (1992) and Fong et al. (2011) used double-legged landing tasks in their investigations. Pappas et al. (2007) found that both females and males performing SLL with higher knee valgus, hip adduction and vertical GRF compared with double-leg landings. Hence, single-leg landing tasks are a more common mechanism for ACL injuries (Faude et al., 2005).

Ford et al. (2006) compared dynamic frontal-plane excursion between females and males during single-legged landings. Collegiate basketball and soccer athletes (11 female, 11 male) performed medial and lateral drop landings from a 13.5 cm block. Females demonstrated greater maximum eversion than males during medial landings. The authors noted that higher amounts of eversion or pronation could cause an increased valgus load at the knee, which in turn places a significant amount of stress on the ACL. In this study, the female subjects also exhibited greater knee-abduction angles, knee frontal-plane excursion and hip frontal-plane excursion during both types of landing.

Zhang et al. (2000) found that knee flexion increased as the landing height increased from 46° to 48° and 53° for 30 cm, 50 cm and 70 cm, respectively, and from 52° to 56° and 63° for 32 cm, 62 cm and 103 cm in height, respectively. However, the exact instructions given to the participants for landing are not mentioned. In addition, this knee-flexion increase could be a common strategy to attenuate ground-reaction forces upon impact. During single-leg landing, Yeow et al. (2010) noticed that as the height increased from 30 cm to 60 cm, the ground-reaction force increased significantly from that at the lower height. With this increase in landing height, the knee becomes more flexed as well. This study also found that when the landing height increased from 30 cm to 60 cm, the knee flexion angle increased; and this was for the initial contact and the peak ground-reaction force. However, the population tested was too homogeneous; it was composed only of males and therefore it is possible that sex, due to differences in neuromuscular control strategies, may have had a confounding effect on the results. In addition, since only recreationally active adults were evaluated in this study, the results cannot be generalised to other athletic or symptomatic populations. Cortes et al. (2007)

reported that if individuals have a high level of experience in an activity, such as landing, that plays an important role in the way they land and absorb energy. It is suggested that the sport training and background of an athlete contributes to the neuromuscular and landing strategies that individuals exhibit (Colby et al., 2000; Cortes et al., 2007). In support of this, Cowley et al. (2006) found differences in the ground-reaction forces (GRF) and stance times between female soccer and basketball players when performing a drop landing and a cutting task.

It can be noticed in Table 2-3 that there are inconsistencies in landing techniques; some researchers instruct their participants to land on the dominant leg (Ali et al., 2013; Orishimo et al., 2009; Orishimo et al., 2014; Russell et al., 2006; Schmitz et al., 2007; Yeow et al., 2010), while others focus on the right leg (Garrison et al., 2005; Kiriyama et al., 2009; Nagano et al., 2007). Variations in arm position also exist, some researchers instructed their participants to cross their arms against their chest (Pappas, Hagins, et al., 2007; Pappas, Sheikhzadeh, et al., 2007), while others specify an abducted position (Garrison et al., 2005). And some participants are asked to keep their hands on their iliac crests when landing to reduce any variability from swinging arms (Ali et al., 2013; Nagano et al., 2007; Schmitz et al., 2007).

To the best of our knowledge, only two studies (McNair & Prapavessis, 1999; Herrington & Munro, 2010) have utilized a large population to present reference values during target tasks. McNair & Prapavesis (1999) recruited 234 adolescent participants to obtain normative data on vertical ground-reaction forces only during landing. Surprisingly, females showed a greater force compared to their male counterparts (4.2 vs 4.6 *body weight) and recreational athletes greater force compared to competitive ones (4.5 vs 4.4 *body weight); and individuals participating in sports involving jumping and landing activities produced greater force compared to individuals in sports that did not involve jumping activities (4.6 vs 4.4 * body weight).

Using a two-dimensional system, Herrington and Munro (2010) obtained normative numbers for knee-valgus angle during drop-jump landing for a population of 100 physically active participants. They suggest that average knee valgus ranges between 5° and12° for females and between 1° and 9° for males. Although both studies reported valuable data during landing and stepping tasks, they only investigated vertical GRF data or 2D valgus angle. However, there are no reference values for either kinematic or kinematic data during single-leg landing in a healthy population. Table 2.3 gives a summary of previous studies providing 3D variables collected from SLL tasks with healthy participants.

Table 2.3. Summary of literature reporting 3D variables during SLL tasks with healthy participants

Study	Sample	Height (cm)	Joint angles (degrees)						Moments (Nm/Kg)	
			Knee flexion	Knee valgus	Knee Int. Rot.	Hip Flexion	Hip adduction	Hip Int. Rot.	Knee flexion	Knee valgus
			Values at peak							
Pappas et al. (2007)	32 athletes	40	72.2	0.96	-	-	8.4	-	-	-
Kiriyama et al. (2009)	81 F healthy	20	55.0	-3.00	13.7	-	-		-	-
	88 M healthy		56.0	-2.00	10.1	-	-		-	-
Orishimo et al. (2009)	21 F dancers	30	55.1	-11.5	-	23.3	15.4		3.2	-1.5
	12 M dancers		58.2	-8.4	-	23.2	15.3		3.1	-1.7
Ali et al. (2013)	12 healthy	30	27.9	-	-	21.5	-	-	-	-0.13
		50	30.4	-	-	20.3	-	-	-	-0.11
Orishimo et al. (2014)	10 F dancers	30	57.0	-	-	-	-	-	2.5	-
	10 F athletes		56.0	-	-	-	-	-	2.8	-
	10 M dancers		54.3	-	-	-	-	-	2.8	-
	10 M athletes		54.2	-	-	-	-	-	2.8	-
Garrison et al. (2005)	8 F footballers	60	-	-	-	-	-	-	1.10	0.14
	8 F footballer		-	-	-	-	-	-	1.30	0.26

Table 2.3: Continued

Study	Sample	Height (cm)	Joint Angles (Degrees)						Moments (Nm/Kg)	
			Knee flexion	Knee valgus	Knee Int. Rot.	Hip flexion	Hip adduction	Hip Int. Rot.	Knee flexion	Knee valgus
			Values at initial contact							
Schmitz et al. (2007)	14 F healthy	30	42.5	-	-	21.6	-	-	-	-
	14 M healthy		38.9	-	-	16.7	-	-	-	-
Nagano et al. (2007)	19 healthy	30	31.2	-2.30	12.6	-	-	-	-	-
	18 M uni. athletes		27.8	-1.40	9.4	-	-	-	-	-
Russell et al. (2006)	16 F healthy	60	18.0	-0.65	-	-	-	-	-	-
	16 F healthy		17.0	3.85	-	-	-	-	-	-
			Values at maximal knee flexion							
Yeow et al. (2010)	10 M healthy	60	61.0	-	-	-	-	-	-	-
		30	59.0	-	-	-	-	-	-	-
Russell et al. (2006)	16 F healthy	60	59.0	3.13	-	-	-	-	-	-
	16 F healthy		58.0	15.2	-	-	-	-	-	-
Orishimo et al. (2009)	21 F dancers	30	58.7	-1.70	-	28.7	0.9	-	1.40	-0.4
	12 M dancers		59.2	-3.20	-	20.0	4.8	-	1.60	-0.6

M = males; F = females; Int. Rot. = Internal rotation; sign conventions = (- knee-valgus angle; + knee-flexion angle; + hip-adduction angle; + hip and knee internal-rotation angles; + knee flexion and valgus moment).

2.6.1.3. Running

Most recreational sporting enthusiasts engage in running-based sports which involve repetitive high-magnitude feet impact with the ground (Hreljac, 2004). Patellofemoral pain syndrome (PFPS) is the most prevalent type of knee pain among runners (Taunton et al., 2002; Willy, Manal, Witvrouw, & Davis, 2012). Stefanyshyn et al (2006) in their prospective study have reported that the occurrence of high knee frontal loading while running can predict PFPS. Retrospective cohort investigations do not always support the idea of increased knee valgus in PFPS individuals compared to control groups (Bolgla, Malone, Umberger, & Uhl, 2008; Dierks et al., 2008), although it could be debated that PFPS individuals may avoid dynamic knee-valgus because of pain.

Gender differences have been noticed during running biomechanics. Specifically, females exhibit higher peak-knee valgus (Ferber et al., 2003; Malinzak et al., 2001) as well as hip-internal rotation and adduction (Ferber et al., 2003; Souza & Powers, 2009). In contrast, Willson and Davis (2008a) reported greater hip adduction but not greater hip-internal rotation while running. However, neither study focused on habitual runners who specifically reported suffering from PFP during running (Souza & Powers, 2009; Willson & Davis, 2008a). A recent study that focused on runners with PFP reported that they had less hip adduction and no differences in hip-internal rotation when compared to a healthy control group (Dierks et al., 2008). The inclusion of males and females may have influenced the results of this study, as previous gender differences have been reported in running (Ferber et al., 2003). The various cohorts and tasks reported in the literature to date have limited applicability to female runners, and thus further research is required on that particular population.

It has been suggested that this increased non-sagittal plane motion contributes to PFPS, and it is females who are predominantly affected as 68% of sufferers of PFPS are females (Taunton et al., 2002). Ferber et al. (2003) compared hip-and-knee stance-phase angles and moments in 20 male and 20 female recreational runners. The females showed significantly greater peak-knee valgus, hip-adduction and hip-internal-rotation angles compared to men. Greater hip adduction will result in an increased knee-joint moment as the lever arm between the line of action of the ground-reaction force and the knee-joint centre increases (Stefanyshyn et al., 2006).

Standardizing the running speed is useful when comparing kinematics and kinetics between and within subjects. Using the same speed for all subjects allows for a comparison between subjects that is not affected by the running speed. An increase in running speed has been shown to change the kinematics and kinetics of the lower extremities, thus most researchers agree that the running speed should be standardised (Queen et al., 2006; Pollard et al., 2004; Malinzak et al., 2001; Colby et al., 2000; Kadaba et al., 1989). Among subjects, standardization of the running speed allows for a comparison of conditions, without the kinematics or kinetics changing due to it. Others believe that the running speed should be standardised among subjects (Stergiou et al., 1999b). The acceptable over-ground running speed controlled by photocells in running studies has ranged from 1.5 to 6 $m \cdot s^{-1}$, with an average running speed of approximately 4 $m \cdot s^{-1}$ (Diss, 2001; Ferber et al., 2003; Ferber et al., 2002; Stergiou, N. et al., 1999b; Wank, Frick, & Schmidtbleicher, 1998). In most studies, subjects have been asked to maintain a running speed within 5%–8% of a predetermined speed for an acceptable trial (Stergiou et al., 1999a; Vanrenterghem et al., 2012).

Despite the wealth of literature on running biomechanics, no studies were found which provide reference values for kinematic and kinetic variables in a healthy population. This information could provide valuable insights for screening athletes at high risk of PFPS and ACL injuries.

Table 2.4. Summary of literature reporting 3D variables during a RUN task for healthy participants

Study	**Sample**	**Speed** ($m{\cdot}s^{-1}$)	**Joint angles (degrees)**						**Moments (Nm/Kg)**	
			Knee flexion	Knee valgus	Knee Int. Rot.	Hip flexion	Hip adduction	Hip Int. Rot.	Knee flexion	Knee valgus
			Values at Peak values							
B-Jones et al. (2013)	19 healthy	4.0 ±0.5	45.1	-1.6	2.95	35.8	12.7	6.33	2.41	-0.91
Bischof et al. (2010)	19 F healthy	3.3 ± 5%								
Ferber et al. (2003)	20 F runners	3.7±5 %	46.0	-6.4	0.79	38.8	9.2	11.2	1.14	-0.47
	20 M runners		45.0	-4.5	2.7	33.3	5.6	7.0	1.31	-0.51
Irene et al. (1999)	20 runners	3.3 ± 5%	-	-	-	-	-	-	1.63	-0.65
Noehren et al. (2012)	16 F runners		-	-	6.4	-	17.8	5.2	-	-
Besier et al. (2001)	11 M healthy	3.0 ±0.2	47	-	-	-	-	-	2	1.2
			Without Normalisation							
David et al. (2015)	12 M runners	4.1 ±0.1	-	-	-	-	-	-	37.0	87.4
		2.9 ±0.1	-	-	-	-	-	-	29.6	75.9

M = males; F = females; Int. Rot. = internal rotation; sign conventions= (- knee-valgus angle; + knee flexion angle; + hip-adduction angle; + hip and knee internal-rotation angles; + knee flexion and valgus moment

2.6.1.4. Sidestep cutting manueuvre

The cutting manoeuvre is a specific movement often performed to change direction quickly while running in a sporting activity (Besier, Lloyd, Ackland, et al., 2001; Schot et al., 1995). The two most common cutting techniques in the literature are the sidestep cut and the crossover cut. The sidestep cut is performed by planting with the foot opposite to the intended change in direction, while the crossover cut is performed by planting with the foot on the same side as the intended change in direction (Andrews et al., 1977; Houck, 2003). Both techniques comprise three separate phases: deceleration, plant and cut, and take off (see Fig. 2.4).

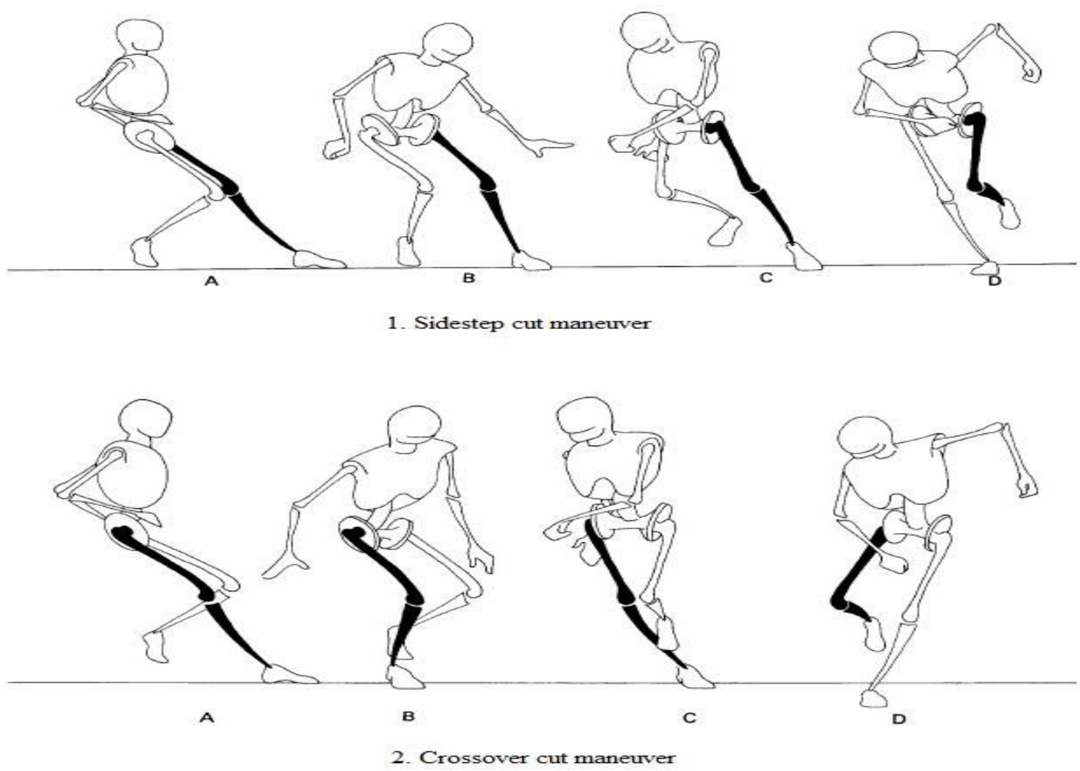


Figure 2.4. Schematic representation of plant-and-cut phase during sidestep (1) and crossover (2) tasks (Adapted from Andrews et al., 1977)

The goal during the deceleration phase is to decrease the momentum by using the greatest amount of force possible in the shortest amount of time in order to begin to move in a new direction. During this phase, the majority of forces occur in the antero-posterior direction. It is possible for the knee to flex as far as 90° during the deceleration phase. Two sources of resistance against excessive anterior translation of the femur onto the tibia are the extensor mechanism and the medial collateral ligament. Normally, a little damage is done to the PCL and medial collateral ligament due to the extensor mechanism being the main decelerating force (Andrews et al., 1977).

A change in direction occurs during the plant-and-cut phase, which also can be defined as the stance phase (Cross, Gibbs, & Bryant, 1989). While most of the required deceleration has already taken place, the pivot foot remains in contact with the ground and the hip rotators turn the torso in the desired direction (Andrews et al., 1977). The free leg swings in the direction of the cut and starts to accelerate the individual in a new direction. During the plant-and-cut phase of sidestep cutting there is a great amount of stress placed on the medial structures of the pivoting leg's knee. This phase adds a rotational component that was not present in the deceleration phase, which in turn increases the risk of injury (Andrews et al., 1977). The taking-off phase begins once the body has realigned itself in the new direction. This phase mimics that of a normal gait pattern except that the individual is leaning forward more than normal to increase their acceleration back to normal. Once again, the majority of movement in this phase is in the antero-posterior direction (Andrews et al., 1977).

Sidestep cutting has been shown to be a mechanism that can cause non-contact ACL injuries (Besier, Lloyd, Ackland, et al., 2001; Besier, Lloyd, Cochrane, et al., 2001; J. R. Houck, Duncan, & Haven, 2005; McLean et al., 2004; Schot et al., 1995). Sidestep cutting manoeuvres generally generate a valgus moment in the knee during the stance phase (Besier, Lloyd, Ackland, et al., 2001; Besier, Lloyd, Cochrane, et al., 2001). Bendjaballah et al. (1997) reported that the load on the ACL can be six times higher at as little as 5°, from neutral, of knee valgus. Even small changes in valgus motion can considerably increase the valgus load on the knee.

The amount of deceleration needed in side cutting is related to the angle and speed at which the manoeuvre is performed. A 90° sidestep cut has a very different momentum profile than a 45° degree sidestep cut or a straight-ahead run (Schot et al., 1995). The majority of studies standardise the cutting angle at or around 45° (Landry, McKean, Hubley-Kozey, Stanish, & Deluzio, 2007; McLean, et al., 2005a; McLean et al., 2004; O'Connor & Bottum, 2009; Pollard et al., 2004; Sigward & Powers, 2006b). This angle is acute enough to require substantial deceleration, but shallow enough for the change in direction to be achieved within the time constraint of a single foot contact. In Premier League football matches, Bloomfield et al. (2007) noticed that changes in direction frequently reached higher angles of between 90° and 180°, which may lead to higher knee-valgus moment.

According to McLean (2005a), in a sidestep cutting task a correlation exists between the internal rotation position of the lower extremities and the degree of hip flexion. Due to this high initial internal rotation of the hips at initial contact, the medial muscle groups can be weakened, leaving the knee susceptible to valgus-load injury. Unlike the lower-extremity joints, the trunk is often ignored in cutting-motion studies. It has been speculated that the torque generated by the lower extremities, pelvis and torso is what actually changes the direction by applying a force to the ground (Schot et al., 1995).

When compared to their male counterparts, females perform cutting tasks with less knee flexion (Malinzak et al., 2001), a smaller peak-knee flexor moment (Sigward & Power, 2006b) and greater knee valgus (McLean et al., 2004; Malinzak et al., 2001). Gender differences in external knee-valgus moments are also exhibited during pre-planned sidestep cutting tasks (Sigward & Powers, 2006a). It is suggested that the increased load on the ACL during cutting tasks is the result of valgus torque applied to the knee (Sigward & Powers, 2006a). Because females typically exhibit greater knee valgus than males, consequently greater valgus torque is applied to the knee, which may result in ACL injury during cutting tasks. In addition, previous comparison studies have found that females are more quadriceps-dominant than males during cutting (Chappell et al., 2002; Malinzak et al., 2001). Despite a rapidly growing body of research on cutting biomechanics, see Table 2.5, reference values for lower-limb joint angles and moments when performing this task are still unclear.

Table 2.5. Summary of literature reporting 3D variables during a sidestep cutting task in healthy participants

Study	Sample	Cut angle	Speed ($m{\cdot}s^{-1}$)	Joint angles (degrees)						Moments (Nm/Kg)	
				Knee blexion	Knee valgus	Knee Int. Rot.	Hip flexion	Hip adduction	Hip Int. Rot.	Knee flexion	Knee valgus
				Values at Peak values							
Bealulie (2008)	15 F healthy	45°	4.0-5.0	57.9	-15.3	19.8	-	-	-	-	-
	15 M healthy			57.3	-5.3	22.9	-	-	-	-	-
McLean et al. (2005)	10 healthy	45°	4.5-5.5	-	-	-	-	-	-	-	0.63
	10 M healthy			-	-	-	-	-	-	-	0.42
Jones et al. (2014)	20 F healthy	90°	4.0-5.0	-	-13	-11	-	0	9	-	1.18
		180	3.6-4.4	-	-14	-6	-	1	5	-	1.13
				Values at peak knee-valgus moment							
Jorrakate (2011)	10 F healthy	45°	3.6±0.2	28.6	2.8	-6.9	42.0	-4.2	11.4	-	0.5
	10 F athletes		4.3±0.3	35.0	5.0	0.5	51.6	-3.4	15.4	-	0.6
	10 M healthy		4.5 ±0.4	35.1	-0.8	1.2	52.7	-9.8	2.5	-	0.8
	10 M athletes		4.7±0.1	34.5	4.0	1.0	52.6	-7.4	10.5	-	0.6
				Values at initial contact							
Bealulie (2008)	15 F healthy	45°	4.0-5.0	17.9	-2.9	-2.7	-	-	-	-	-
	15 M healthy			15.6	1.2	0.17	-	-	-	-	-
Jorrakate (2011)	10 F healthy	45°	3.6± 0.2	27.0	1.3	-3.4	47.1	-4.0	6.4	-	-
	10 F athletes		4.3± 0.3	34.0	3.8	3.3	57.8	-2.5	11.8	-	-
	10 M healthy		4.5 ± 0.4	31.9	0.8	3.8	57.6	-10.7	-3.5	-	-
	10 M athletes		4.7± 0.1	32.6	1.3	3.3	59.3	-7.6	8.3	-	-

Table 2.5. Continued

Study	Sample	Cut angle	Speed (m·s⁻¹)	**Joint angles (degrees)**						**Moments (Nm/Kg)**	
				Knee flexion	Knee valgus	Knee Int. Rot.	Hip flexion	Hip adduction	Hip Int. Rot.	Knee flexion	Knee valgus
				Values at 45° of knee-flexion angle							
Sigward (2006b)	15 F athletes	45°	5.5-7.0	-	-	-	-	-	-	1.4	-0.43
	15 M athletes			-	-	-	-	-	-	2.1	0.01
Pollard (2004)	12 F athletes	45°	4.0-5.0	45	-2.39	6.3	-	-3.43	3.3	-	0.37
	12 M athletes			45	-1.53	6.1	-	-9.07	3.5	-	0.31
				Values in peak-stance phase							
McLean et al. (2004)	8 females	30-40°	4.5-5.5	57.2	-14.2	14.3	43.2	-	8.4	-	-
	8 males			63.1	-12.1	19.2	54.1	-	14.6	-	-
				Values in first 20% of stance phase							
Sigward & Powers (2007)	38 females (soccer)	45°	5.5-7.0				46.2	7.7	4.2		0.2
	23 F high V moment						48.5	12.8	9.3		1.2

M = males; F = females; Int. Rot. = internal rotation; sign conventions= (- knee-valgus angle; + knee-flexion angle; + hip-adduction angle; + hip and knee internal-rotation angles; + knee flexion and valgus moment

2.6.2. Relationship between biomechanical variables during different screening tasks

An understanding of how the risk factors hypothesized behave under different task constraints might provide better insights into possible risky motions. The intrinsic differences in the control mechanisms of various tasks and how those tasks are conducted in laboratory experiments have been of recent concern. A few studies have compared lower-limb biomechanics across tasks within the same population. Jones and Colleagues examined the relationship between single-legged landing, 90° cutting and pivoting (180° turn) in 20 female soccer players. The authors found strong correlations for peak knee-abduction angles across tasks (R = 0.63-0.86), but only moderate correlations between SLL and cutting (R = 0.46), cutting and pivoting (R = 0.56) and SLL and pivoting (R = 0.43) across tasks for peak knee-abduction moments.

Whatman and colleagues (2011) investigated the links between lower-limb kinematics during jogging (2.9 ± 0.4 $m{\cdot}s^{-1}$) and those occurring during five screening tasks (lunge, small knee bend (SKB), single-leg small knee bending, one-metre hop and step down from 20 cm). They reported moderate to very large associations between kinematic variables recorded during the functional tests in relation to jogging (r= 0.53 to 0.93). The highest associations (r≥ 0.70) for more than 3 tasks were for the frontal plane motion in ankle, knee, and hip joints, and hip internal rotations. High correlation was also reported in peak pelvic tilt (r= 0.60 to 0.72), while trunk angles exhibited the poorest associations (r= 0.15 to 0.53). Despite the small sample size and control velocity, this study demonstrates the potential of using SKB tasks when evaluating the alignment of lower-limb.

Whatman et al. (2013) conducted another correlational study between double-legged tasks (drop jump and SKB) and single-legged tasks (single-leg SKB & treadmill jogging) in 23 uninjured young athletes (aged 10–12 years). Correlations for peak knee-valgus angles between drop jumping and SKB were moderate (r= 0.60-0.63), and moderate to large between running and single-leg SKB (r= 0.64-0.84). The highest correlation in their study was found with hip-internal rotation, between SKB and drop jump (r= 0.82-0.87). However, the target participants in studies by Whatman and colleagues (Whatman et al., 2013; Whatman et al., 2011) were young athletes (11±1 and 22±4 years, respectively); whether the same levels of correlation exist in older participants is unclear. Another drawback of their studies is that authors do not clarify whether their participants were males or females.

Earl et al. (2007) compared the movement patterns of 18 men and 19 women athletes during a single-leg step down (SLSD) and drop vertical jump (DVJ). The authors found that the SLSD task resulted in greater hip adduction (16° compared to 1°), greater eversion (12° compared to 8°) and less knee flexion in both females and males. DVJ produced more frontal-plane motion in the knee (3.5° compared to 0.3°). Women had larger peak hip-internal rotation value in the step down than in the drop vertical jump (5° compared to 2°). When averaged, in both tasks, women had greater knee abduction than men (4° compared to 0°). These findings suggest using stepping down to evaluate hip control and bilateral drop vertical jump to assess excessive knee-valgus measures.

Pappas et al. (2007) compared bilateral vs unilateral landings of recreational athletes (16 males and 16 females). They reported that unilateral landing resulted in increased knee valgus (0.96° vs -1.4°), decreased knee flexion at initial contact (15.1° vs 20.8°), decreased peak knee flexion (72.2° vs 93.3°), decreased relative hip adduction (1.13° vs 8.4°) and more vertical ground-reaction force VGRF (3.2 vs 2.7 BW). During both types of landing, females landed with increased knee valgus and normalized VGRF compared to males. In 2008, Willson and Davis conducted a study to compare lower-limb angles in females with and without PFPS when performing single-leg squats, running and repetitive single-leg jumps. They found that a group with PFPS had 3.5° greater hip-adduction angles and 3.5° fewer internal hip rotation during SLS, running, and jumping than a control individuals. The control group showed hip-external rotation excursion during the loading phase of running but internal-rotation excursion during the loading phase of single-leg jumps (Willson and Davis, 2008a). However, their findings should be interpreted with caution since the differences between groups were relatively small and no measurement errors were reported for the study.

Imwalle et al. (2009) compared lower-extremity kinematics during 45° and 90° cutting tasks in 19 female soccer players. They found hip and knee-internal rotation angles ($p = 0.008$) were increased when performing cutting task at 90° compared with cutting at 45°. Hip flexion ($p < 0.001$) was also larger in the 90° cutting. The only significant predictor of knee abduction during both tasks was hip adduction (R = 0.49). The findings suggest that the mechanisms underlying increased knee-abduction measures in athletic women during cutting tasks are primarily frontal-plane motions at the hip.

In another correlational study, McLean et al. (2005b) evaluated 20 athletes during three unilateral tasks involving rapid directional change: a sidestep, a side jump and a 180° cut during a shuttle run. They found significant correlations with peak frontal-plane angle across tasks (r = 0.84-0.89), but no significant relationship between dynamic knee abduction and standing static abduction measurements. It should be noted that all tasks were alike, as they were all unilateral with very physical demands. The peak lower-extremity joint motions of the female cohort were similar across tasks in all three-movement planes, with most joint motions differing by only a few degrees.

In another attempt at linking cutting tasks with running, Besier et al. (2001) examined the external moments around the knee joint of 11 male soccer players during running, sidestepping and crossover cutting tasks. They found that the external sagittal loads were similar across tasks, whereas the external frontal and horizontal moments placed on the joint increased dramatically during cutting tasks compared with forward running. However, the findings of both of the aforementioned studies should be interpreted with caution due to the small sample sizes. Also, the similarities in the nature of the tasks may have led to high inter-task correlations.

Apart from Kristianslund & Krosshaug (2013), all of the aforementioned studies were conducted on relatively small sample sizes. Kristianslund & Krosshaug (2013) conducted a large-scale study (n=120) to examine the association between a cutting task and a drop vertical jump task. They observed weak correlation with knee-valgus moment (p= 0.13), but greater correlation with valgus angles (p= 0.71). A note of caution is due here since these findings were collected only from elite female handball players, making the findings less generalisable to other populations. In reviewing the literature, no studies were found that investigated the inter-task correlation of kinematic and kinetic variables in a large sample of recreational athletes during distinctly different movement tasks related to common knee injuries.

2.6.3. Movement-analysis techniques

Motion of the knee joint occurs in three planes (sagittal, frontal, transverse) with six degrees of freedom (3 rotations and 3 translations allowing 12 directional motions) between the femoral condyles and tibial plateau (Quatman, 2009). The knee joint can rotate in the frontal plane by adduction and abduction, in the sagittal plane by flexion and extension, and in the transverse plane by internal and external rotation. Knee-joint translation occurs in the sagittal plane anteriorly and posteriorly, in the frontal plane medially and laterally, and in the transverse plane via compression and distraction (see Fig. 2.5).

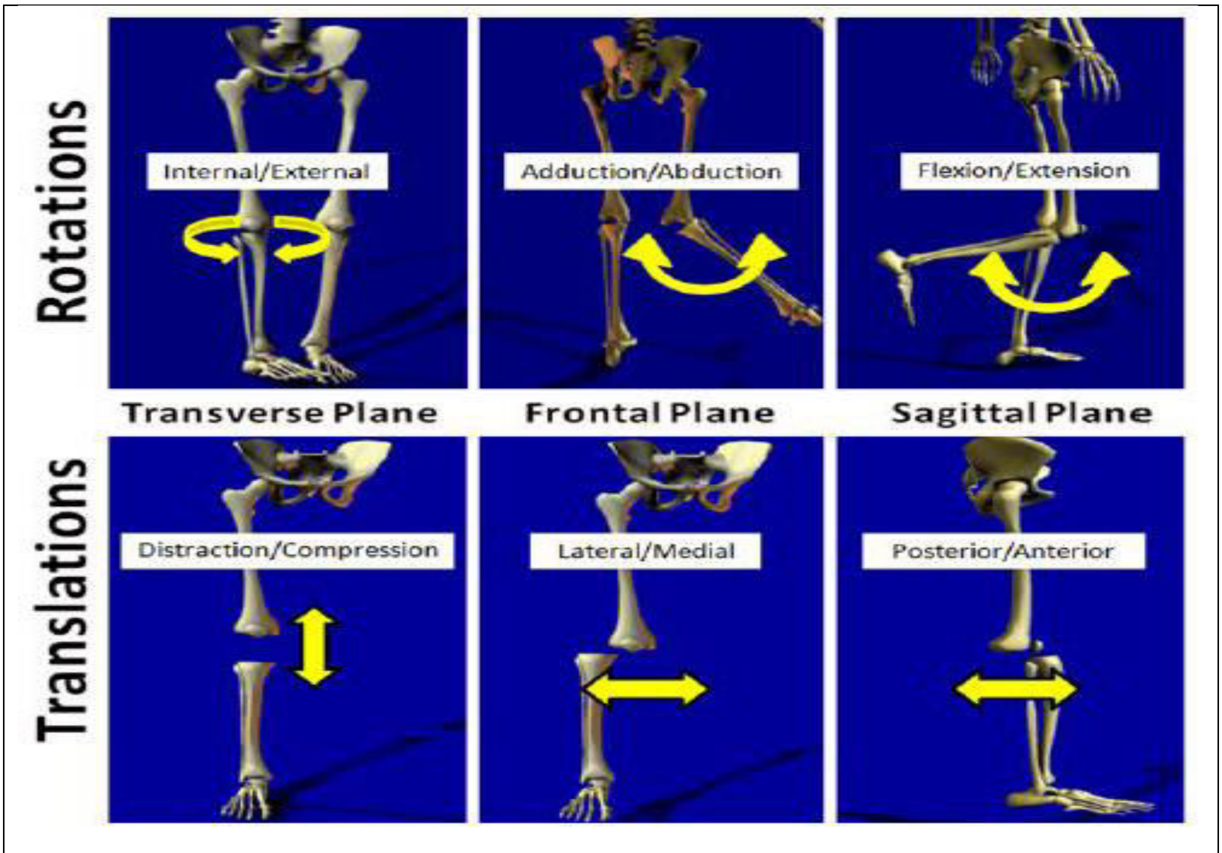


Figure 2.5. Rotations and translations of the knee joint. Adapted from Quatman (2009)

Most studies which measure lower-limb kinetic and kinematic commonly use 3D motion-analysis systems (Cappozzo, Catani, Leardini, Benedetti, & Croce, 1996; Ferber et al., 2003; Ford et al., 2003; Hewett et al., 2005; Jones et al., 2014; McLean, Neal, Myers, & Walters, 1999; Milner, Westlake, & Tate, 2011; Sigward & Powers, 2006a). This allows researchers to quantify all motion planes during dynamic tests and is assumed as the "gold standard" of movement analysis (Meldrum et al., 2013; Munro et al., 2012b). By fixing reflective markers on specific anatomical landmarks, the skeletal system can be recreated and biomechanical features can be recorded and measured during functional tasks.

2.6.4. Reliability of using 3D motion-analysis techniques

The reliability of an outcome measurement reflects how reproducible or repeatable it is under a given set of conditions. For an outcome measurement to be valuable, it must provide stable or reproducible values with small measurement errors (Rankin & Stokes, 1998). Understanding the reliability and measurement errors related with each of these screening tools is essential. There are two types of measurement errors: systemic bias and random error. The former can be used as an indicator of whether a learning effect or fatigue exists. The latter occurs due to unpredictable biological, psychological and mechanical factors, which cannot be avoided, even if the source of the errors is anticipated (Portney & Watkins, 2009). Biological and psychological factors include lack of attention, motivation and fluctuations in the performance of the subject. Mechanical factors include instrumentation or equipment problems. Uncontrolled confounding variables may also contribute to noise in measurements (Batterham & George, 2003).

Despite the widespread use of SLS, SLL, running and cutting tasks in the literature investigating the etiology of PFPS and ACL injuries, as reported in Tables 2.1–2.4, only a few attempts have examined the consistency of biomechanical measures during these tasks (Sankey et al., 2015; Nakagawa et al., 2014; Stephenson et al., 2012; Milner et al., 2011; Noehren et al. 2010; Ford et al., 2007; Queen, Gross, & Liu, 2006; Ferber et al., 2003; Besier et al., 2001b). While within-day reliability is important, interventional research requires that outcome measures are stable from day to day (Bland and Altman, 1986). Only four of these studies investigated the within- and between-days reliability of their measures (Nakagawa et al., 2014; Noehren et al. 2010; Ford et al., 2007; Ferber et al., 2003).

Nakagawa et al. (2014) investigated the within- and between-days reliability of 3D angles during SLS in young individuals (10 males and 10 females, aged 20±1.7 years). They found that the within-days ICCs of hip and knee joint angles were higher than those of between days (ICC average 0.94 vs. 0.91, respectively). This trend has also been found during running (Queen et al., 2006; Ferber et al., 2002) and landing tasks (Milner et al., 2011; Ford et al., 2007). Between-days reliability is mainly affected by the misapplication of markers (Ford et al., 2007; Queen et al., 2006; Ferber et al., 2002). Skin artefacts is another issue that might affect both within- and between-days measurements (Cappozzo et al., 1996).

Additionally, previous studies have noticed differences in the consistency of measurements in sagittal, frontal and transverse motion planes. The sagittal plane has the lowest variability among measurements during running, stop jump and drop vertical landings (Milner et al., 2011; Ford et al., 2007; Queen et al., 2006; Ferber et al., 2002). Marker placement has a great influence on frontal and transverse planes of movement (Kadaba et al., 1989), which may justify the reduction in between-sessions consistency. In their systematic review, McGinley et al. (2009) found greater errors in knee and hip rotations during gait analysis compared to other planes of movement. As dynamic knee valgus is a combination of motions in the frontal and transverse planes, knowing the measurement errors in these planes is important when assessing individuals with a high risk of knee injury using 3D motion-analysis techniques (Munro, 2014; Ferber et al., 2002).

A number of researchers have noticed higher reliability for GRF data compared to kinematic values (Ferber et al., 2002; Kadaba et al, 1989; Winter et al., 1984). They have suggested that GRF data are representative of the sum of all segmental masses and accelerations, and that less variability will be seen compared to individual joint kinetic or kinematic patterns (Winter, 1984). Moreover, no markers are needed to collect GRF data and these are therefore less variable (Ferber et al., 2002).

All of the studies reviewed above suffer from the fact that they only focused on relative reliability using intra-class correlation (ICC). ICC appears to be easy to interpret, but the closer to one the higher the reliability, and so ICC alone cannot provide a full picture of reliability since it does not indicate the amount of disagreement between measurements. It should therefore be used in combination with standard error of measurement (SEM) (Rankin and Stokes, 1998), which is very useful for practitioners wanting to determine individual improvement (Munro et al., 2012b; Domholdt, 2005). The calculation of SEM depends on the standard deviation of measurements, which allows the clinician to be 68% confident that the true value lies within ±1 SEM of an observed value (Portney and Watkins 1993).

Only a few researchers provide SEM values for different screening tasks, such as SLS (Nakagawa et al., 2014), running (Ferber et al., 2002), double-legged drop jumps with a 7-week gap between sessions (Ford et al., 2007) and a 10-week gap (Whatman et al., 2013). Nakagawa et al. (2014) noticed higher SEM values in sagittal-plane motion (2.6 and 1.3 for hip and knee flexion angles, respectively) compared to other planes. In the same vein, Ferber et al. (2006) found that between-days SEM values for hip, knee and ankle sagittal motion during running were higher than for other planes (1.03°, 2.21° and 2.22°, respectively). This may be explained by the larger range of motion in the sagittal plane compared to other planes. Despite their common use in the ACL literature, no single study provides SEM for single-leg-landing and changing-direction tasks.

In addition to calculating SEM, measuring the smallest detectable difference (SDD) has been advised to determine the minimum change needed to be 95% confident that the change is more than a measurement error (Atkinson & Nevill, 1998; Eliasziw et al., 1994). SDD is based on SEM calculation, but it is more conservative (2.7 SEMs) (Ries, Echternach, Nof, & Blodgett, 2009). From their reliability testing, Nakagawa et al. (2013) provide within- and between-days SDD values for lower-limb angles during SLS in young individuals. None of the aforementioned studies provide SDD values for lower-limb angles and moments during SLL, run and cut tasks.

In summary, the reliability of SLS, SLL, RUN and CUT have been investigated before (Sankey et al., 2015; Nakagawa et al., 2013; Whatman et al., 2013; Ford et al., 2007; Queen et al., 2006; Ferber et al., 2002; Besier et al., 2001). However the findings are sparse and focus on specific populations such as young individuals or top athletes (Nakagawa et al., 2013; Whatman et al., 2013; Sankey et al., 2015; Ferber et al., 2002).

Some previous studies have only examined single parts of consistency (i.e. kinematics or kinetic data alone or within- or between-days reliability). In reviewing the literature, no research was found that investigated within- and between-days reliability and associated measurement error (SEM and SDD) of lower-limb biomechanical variables during SLS, SLL, RUN and CUT together for the same cohort.

This information is essential to assess earlier and forthcoming studies, especially interventional ones, and likewise for practitioners who use these screening tasks to evaluate individual performance during training or rehabilitation. Without measurement-error values, changes in performance cannot be evaluated properly as it is not known whether these changes can be attributed to the intervention or to measurement errors, such as marker position or re-application, static alignment or task difficulty (Malfait et al., 2014; Whatman et al., 2011; Ford et al., 2007).

2.7. Literature gaps

- Although several studies have been conducted to address lower-limb biomechanics during various screening tasks which mimic the real situation of PFPS injuries, such as single-leg squats and running (Tables 2-1 and 2-4), or ACL injuries, such as single-leg landing and cutting tasks (Tables 2-3 and 2-5), the numbers of subjects participating in those studies were limited, making the generalisation of findings difficult. Also, there are no reference values for either kinematic or kinematic data for single-leg squats, single-leg landing, running and 90°cutting tasks with the same cohort population.

- Several attempts have been made to examine the correlation of the biomechanical variables of two (Whatman et al., 2013; Whatman et al., 2011; Imwalle et al., 2009; Willson and Davis, 2008; Earl et al., 2007; Pappas et al., 2007) or more (McLean et al., 2005b; Besier et al., 2001) functional tasks (see Section 2.9); so far, large-scale correlational studies have been reported that link kinematic and kinematic data during single-leg squats, single-leg landing, running or 90°cutting tasks.

- Previous studies have shown that feedback training can reduce some ACL and PFPS risk factors, a summary of feedback studies is reported in Table 4. Most of the investigations up to this point have not dealt with individuals displaying poor motion, i.e. excessive angles, moments or forces. Another question that needs to be asked, however, is whether the effect of augmented feedback on a specific task can spread to tasks.

Chapter 3: Reliability of lower-limb biomechanical variables collected during single-leg squat, single-leg landing, running and cutting tasks.

3.1. Aims

The aims of this chapter are to:

a. Assess the within-day and between-day reliability of measuring 3D biomechanical variables during single-leg squat (SLS), single-leg landing (SLL), running (RUN) and sidestep cutting (CUT) tasks.
b. Establish the standard measurement error (SEM) and smallest detectable changes (SDD) collected from these tasks for healthy participations.

3.2. Background

Abnormal lower-limb mechanics during functional activities has been found to be associated with ACL (Hewett et al., 2005) and PFPS (Willson & Davis, 2008a) injuries. The majority of studies investigating lower-limb biomechanics and its relation to knee injury have been done by analysing 3D motion-analysis systems (Ford et al., 2003; Hewett et al., 2005; Souza & Powers, 2009). 3D analysis allows researchers to calculate all three motion planes during dynamic tasks and is assumed to be the "gold standard" of motion analysis (Meldrum et al., 2013; Munro et al., 2012b).

For an outcome measurement to be valuable, it must provide stable or reproducible values with small measurement errors (Rankin & Stokes, 1998). Understanding of the reliability and measurement errors associated with each of these screening tools is important (Batterham & George, 2003). A key factor in 3D motion analysis is the ability to measure kinematic and kinetic variables reliably, both within and between days. Several authors have reported that measuring biomechanical variables within the same session is often record less variability than in different sessions (Ferber et al., 2002; Ford et al., 2007; Milner et al., 2011; Queen et al., 2006). Marker-placement error has the most influence on between-days reliability (Ferber et al., 2002; Queen et al., 2006).

Sagittal-plane variables have the greatest reliability compared to those for frontal and transverse planes during running (Queen et al., 2006; Ferber et al., 2002), drop vertical jump (Malfait et al., 2014; Ford et al., 2007) or single-leg squat (Nakagawa, Moriya, Maciel, & Serrao, 2014). Frontal and transverse motions, especially dynamic-knee valgus, is seen as key to the high-risk motions related to both ACL and PFJ injuries (Myer, Ford, Barber Foss, et al., 2010; Hewett et al., 2005). Therefore measurement errors in these planes may have a great influence on the identification of individuals with high-risk of injuries using 3D movement analysis techniques

Markers positioning accounts for the greatest errors in 3D motion analysis (Malfait et al., 2014; Ford et al., 2007). Uncertainty in identifying markers' locations affects the calculations determining the positions of joint centres, which leads to errors in joint kinematic and kinetic calculations (Baker, 2006). This uncertainty is mainly due to the fact that markers are positioned on bony prominences (rather than flat surfaces), thus introducing variability and increasing measurement errors (Cappozzo et al., 1996). Also, these bony prominences might be covered by layers of muscles and adipose tissue, making it more difficult to palpate (Baker, 2006). One way in which these errors can be reduced is to place markers on rigid plates fixed to the thigh and shank, as this has been demonstrated to result in less movement than those applied directly to the skin (Manal, McClay, Stanhope, Richards, & Galinat, 2000).

Considering that the project's main aims are to establish reference values for lower-limb kinematics and kinetics during a set of athletic tasks in a physically active population, and to find out the links between those variables, it is, therefore, important to conduct the study using appropriate tools that give stable and reproducible values with small measurement errors.

3.3. Aims

The first aim is to examine the within-day and between-days reliability of using a 3D movement-analysis system to measure lower-limb kinematic and kinetic variables during single-leg squat (SLS), single-leg landing (SLL), running (RUN) & cutting (CUT) tasks. The second aim is to establish the standard measurement error (SEM) and smallest detectable changes (SDD) during these tasks for healthy participations.

3.4. Hypotheses

Based on the previous literature review, three hypotheses are formulated:

- H_1= Within-day reliability for kinetic and kinematic variables will be greater than between-days reliability.
- H_2= Vertical GRF data will be more reliable than joint angles and moments across all tasks.
- H_3= Transverse-plane variables will be less reliable compared to sagittal and frontal planes of movement across all tasks.

3.5. Methods

3.5.1. Pilot study

Prior to starting data collection for the reliability study, a pilot study was conducted to test the differences between right and left legs when performing screening tasks. If performance appeared to be symmetrical, which was shown later to be the case, then for time considerations in both in testing and data processing, only one leg was then tested for most of the participants. Ten healthy participants (5 female, 5 male) were asked to complete three acceptable trials for each leg (starting with right leg) during SLS, SLL and RUN tasks. The cutting task could only be performed with the right leg because of limited laboratory space. Therefore this task was not taken into account in piloting. The protocol and procedure for the pilot study were exactly the same in terms of reliability, as explained in the following sections.

3.5.2. Reliability-study methodology

3.5.2.1. Participants

The demographic characteristics of fifteen recreationally active participants are summarised in Table 3-1; these were all of university students and staff who volunteered for the study. None of the studies reviewed above did a sample-size calculation for reliability testing. However, Wimmer and Dominick (2003) suggest that the sample size for reliability studies should be between 10% and 25% of that of the main study. Therefore, a sample of 15 healthy participants was chosen to represent 15% of the target sample for the main study of this thesis.

Participants were healthy without any lower limb injuries or musculoskeletal complaints for at least six months before the testing. Before starting the data collection, all participants read and signed a written informed consent statement approved by the Research, Innovation and Academic Engagement Ethical Approval Panel at the University of Salford.

Participants were tested twice on their first visit (two sessions), with a 1-hour gap between the sessions to investigate within-day consistency. Participants were then tested after seven days (one session) at the same time as the first session, to assess the between-days reliability of using 3D motion analysis to measure biomechanical variables during SLS, SLL, RUN and CUT tasks. Before each session, participants were asked to warm up on a stationary bicycle.

Table 3.1. Participants' demographics

Characteristic	Gender	
	Males (N= 7)	Females (N= 8)
Age (years)	25.0 (±6.4)	26.6 (±3.5)
Height (cm)	171.0 (±6.7)	163.0 (±5.4)
Mass (kg)	69.7 (±10.7)	63.0 (±8.0)

3.5.2.2 Instrumentation

A motion-analysis system consist of ten cameras (Pro-Reflex, Qualisys), with a sample frequency of 240 Hz, and three force platforms (AMTI, USA) fixed into the running track, sampled at 1200 Hz, was used to gather biomechanical data for lower limbs. This system uses infrared (IR) cameras and passive retro-reflective markers. To enable connection to the cameras, Qualisys proprietary software, Qualisys Track Manager (QTM), was used. There are three stages in the collection of coordinate data using the Qualisys Pro-reflex system: calibration, data collection and 3D reconstruction of retro-reflective markers.

The capture volume size is an important issue, since it affects the system resolution and therefore the accuracy with which position data can be collected. The most appropriate camera position is that which minimises the blind space surrounding the chosen capture volume in the camera's field of view (Richards et al., 2008; Pantano, White, Gilchrist, & Leddy, 2005). Since the variables of interest in this study were collected during the stance phase of running, cutting, SLS & SLL tasks, the ten cameras were positioned in an umbrella configuration around the three force platforms to make sure they could accommodate the selected movements (Fig. 3.1). A Brower Timing Gate System (TC-Timing System, USA) was used to monitor running and cutting times.

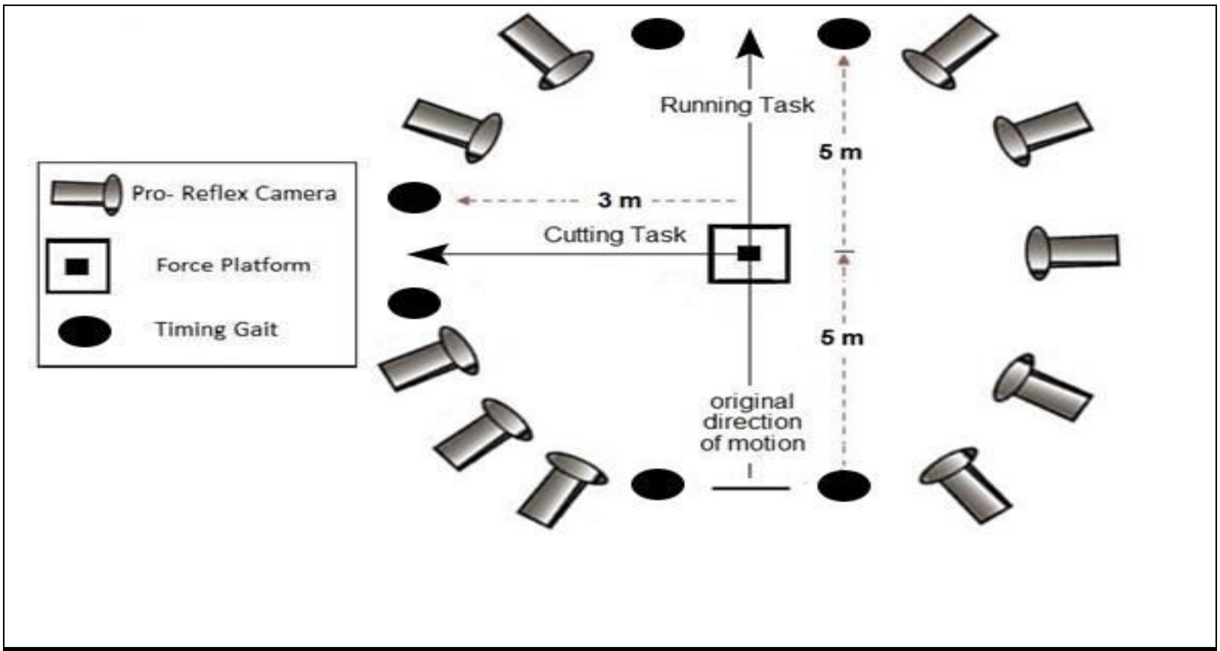


Figure 3.1. Data-collection set-up

3.5.2.3 System calibration

Each IR camera gives a 2D image that needs to be converted into a 3D workplace for the analysis of coordinate data. The purpose of this is to ensure the creation of 3D coordinates of marker position using a direct-linear transformation technique, and to facilitate global references (Richards et al., 2008). Marker position in 3D space can only be located according to the accuracy with which the system is calibrated (Payton & Bartlett, 2008). The lower the residuals, the more accurate the calibration and 3D marker coordinates from measurements.

A rigid L-frame was used in the static calibration of the motion-capture system and its relationship to the laboratory reference frame (Fig. 3.2). A handheld wand with reflective markers (Fig. 3.2) was positioned at each end, at a fixed and known distance of 750.43 mm, and these were used to calibrate the volume that would be used during dynamic trials. A capture time of 45 seconds was used to enable the calibration volume to be successfully calibrated, ensuring that both the lower-floor level and height were covered completely so that at least two cameras could see the wand (Richards et al., 2008).

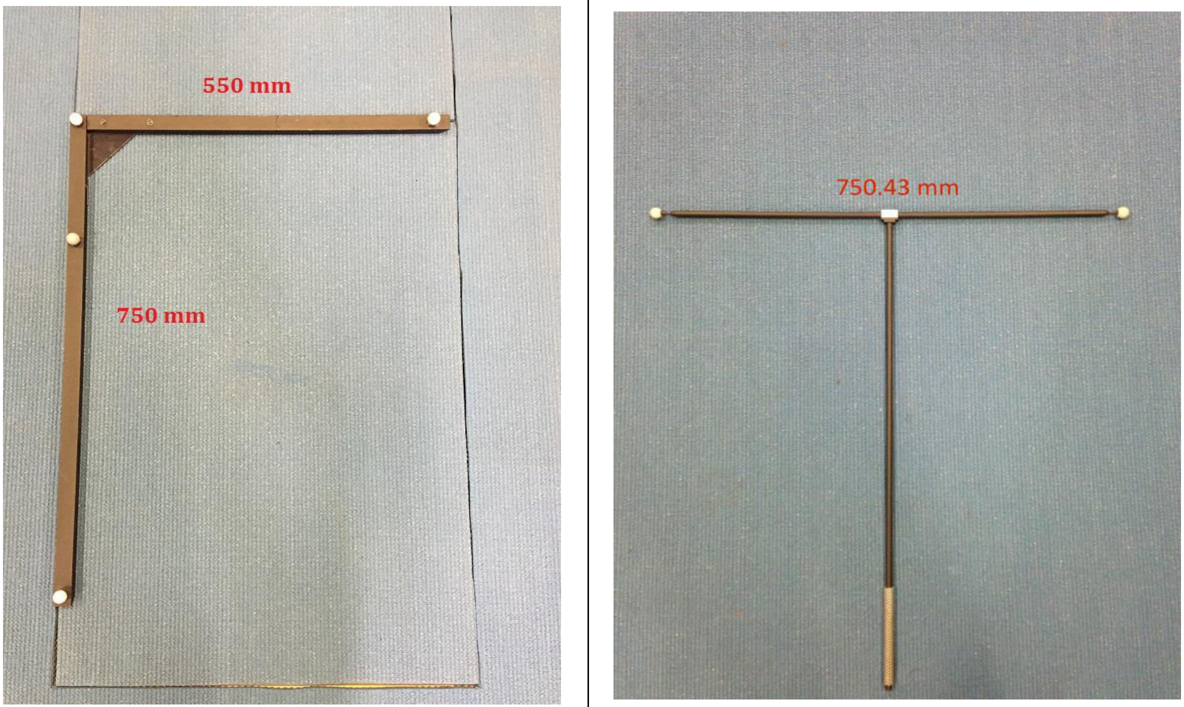


Figure 3.2. Calibration L-frame (left) and handheld wand (right).

3.5.2.4 Marker placement

Prior to each testing session, reflective markers of 14.5 mm diameter were used in all trials of data collection. The markers were attached to the skin using hypoallergenic adhesive tape attached to a flat-based marker (Fig. 3.3). To define the orientation and position of a segment in three-dimensional space, three non-co-linear markers were used (Cappozzo et al., 1996); and during capture time, at least two cameras could see each marker at any instant (Payton & Bartlett, 2008).

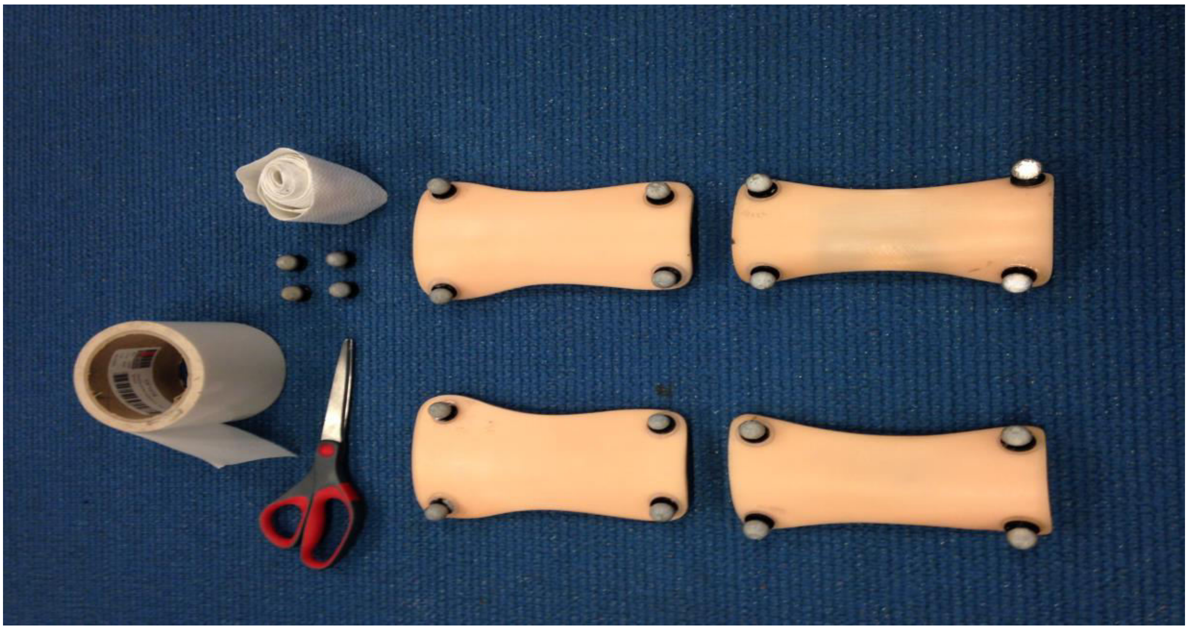


Figure 3.3. Cluster plates, reflective markers and adhesive tape.

A total of twenty anatomical markers were used on each participant in order to describe the anatomical reference frame and centres of joints rotation. Markers were placed on lateral and medial aspects of joints, on anatomical landmarks, at the proximal and distal ends of the segment. Specifically, foot markers were placed on the 1st, 2nd, 5th metatarsal heads and calcaneal tubercle, ankle markers were attached on medial and lateral malleolus, knee markers were attached on lateral and medial femoral condyle, thigh markers were attached on greater trochanter, and finally pelvis markers were attached on right and left anterior superior iliac spine (ASIS), right and left posterior superior iliac spine (PSIS), and right and left iliac crest.

Following a satisfactory capture of all the static markers, the anatomical markers were detached, keeping only 28 as tracking markers (16 markers over 4 cluster plates, 8 markers attached to standard shoes, and 4 markers on ASISs & PSISs). These cluster were securely fastened to the antero-lateral aspect of the thigh and shank of both legs. Manal and colleagues (2000) found that the use of rigid clusters is the optimal configuration, compared to individual skin markers (Manal et al, 2000). Both static and tracking markers are illustrated in Figure 3.4.

Figure 3.4. Static (left) and tracking (right) marker sets

3.5.2.5. Conducting the tests

Before testing, participants wore compression shorts and standard shoes (New Balance, UK) to control the shoe-surface interface. They started with three minutes of low intensity warm-up on a cycle ergometer and were then familiarised with the testing procedure by practising each of the four tasks until they feel comfortable with them; this was typically two and three trials. After familiarisation, the principal researcher attached a total of 40 markers to the participant's lower limb, as explained in Section 3.5.2.4. In order to conduct a static standing trials, each participant was asked to stand in a stationary position on the force plate. It was ensured that the arms of the participant were held clear of the markers so as not to compromise any detection of them. The anatomical markers were then removed and the participant was asked to do the various tasks, starting with SLS, then SLL, RUN and ending with the CUT task.

3.5.2.5.1 Single-leg squat:

The subjects in the current study were taught to stand on their right leg holding their left leg with approximately 45° of knee flexion without allowing the legs to touch each other, then start to squat down as far as they could (but no lower than the position of the thigh being parallel to the ground) and return to a single-leg stance without losing their balance. Consistent with the work of Dwyer et al. (2010) and Zeller et al. (2003), the squat depth was not controlled as this better represented a clinical setting in which normal inter-participant variability will occur. During practice trials, there was a counter for each participant to measure a 5-second period: the first count initiates the movement, the third indicates the lowest point of the squat and the fifth indicates the end (Herrington, 2014). This standardised the test for all participants, thereby reducing the effect of velocity on knee angles and movement patterns.

3.5.2.5.2. Single-leg landing

The participant dropped from a 30-cm step on their right leg, going as far down vertically as possible onto a mark 30 cm from the bench. This height was similar to that used by other researchers (Hargrave, Carcia, Gansneder, & Gansneder, 2003; McNair & Prapavessis, 1999; Yeow et al., 2010). The arms effects were reduced by asking the participants to keep them crossed against their chest (Decker et al., 2003; Pappas, Sheikhzadeh et al., 2007; Pflum, Shelburne, Torry, Decker, & Pandy, 2004).

3.5.2.5.3. Running task

Subjects were required to run at their perceived maximal velocity and to make contact with the force platform with their right foot whist running along a 10 m runway. Their times were measured using timing gates (Fig. 3.1).

3.5.2.5.4. Cutting tasks

As presented in Figure 3.1, subjects were requred to make contact with the force platform using their right foot and immediately turn 90° to the left and run 3 metres in that direction through the second timing gate. Cones were placed at 90° from the original movement direction and were used to guide the participants to cut at an angle of 90°.

To ensure consistent speeds for the running and cutting tasks, a set of Brower timing lights (Draper, UT) was used. These were set at approximately hip height for all participants to ensure that only one body part, such as the lower torso, broke the beam, Yeadon et al. (1999). The time to complete the run and cut tasks was used to monitor each subject's performance on each test occasion. The speed was then calculated by dividing the distance by the time. In order to compare the findings with the literature, participants were asked to redo their trial if the speed fell below 4 m/sec. for running and 3 m/sec. for cutting tasks.

Participants were asked to complete three successful trials for each task, and they were given about one to one and a half minutes between trials to diminish the effect of fatigue (Cortes et al., 2010; Beaulieu et al., 2008). The markers were then removed and replaced for within-day reliability (1st and 2nd sessions) and between-day sessions (1st and 3rd sessions).

3.5.2.6. Data processing

Visual3D motion (Version 4.21, C-Motion Inc. USA) was used to calculate joint kinematic and kinetic data. Motion and force-plate data were filtered using a Butterworth 4th order bi-directional low-pass filter with cut-off frequencies of 12Hz and 25Hz, respectively, with the cut-off frequencies based on a residual analysis (Yu et al., 1999). All lower-extremity segments were modelled as conical frustra, with inertial parameters estimated from anthropometric data (Dempster, Gabel, & Felts, 1959). Joints angles was calculated using an X-Y-Z Euler rotation sequence, where X equals flexion-extension, Y equals abduction-adduction/ varus-valgus and Z equals internal-external rotation. Joint kinetic data were calculated using three-dimensional inverse dynamics, and joint-moment data were normalized to body mass and presented as external moments referenced to the proximal segment. External moments are described in this study, e.g. an external knee-valgus load will lead to abducting the knee (valgus position), and an external knee-flexion load will tend to flex the knee (Malfait et al., 2014).

The calibration anatomical systems technique (CAST) was used to define the 6 degrees of freedom movement of each segment during the dynamic tasks (Cappozzo et al., 1996). A static trial, where the participant stood on the force plates with all markers in view of the cameras, was done with all the anatomical and tracking markers and the Qualisys software prior to extraction for post-processing software. The positions of these anatomical markers offered reference points to identify bone movement through only the tracking markers set during the movement trials.

As can be seen in Figure 3-5, the model used had seven rigid segments attached to the joint. Each segment is considered to have six variables that describe its position (3 variables describe the position of the origin, and 3 variables describe the rotation) in 3D space. Specifically, 3 variables describe the segment translation along three perpendicular axes (vertical, medial-lateral and anterior-posterior) and 3 variables describe the rotation about each axis of the segment (sagittal, frontal and transverse). The subject's body mass (in kilogrammes) and height (in metres) were entered into the software for use in kinetic calculations. Each segment of the pelvis, thigh, shank and foot was modelled to determining the proximal and distal joint/radius. The hip-joint centre is automatically calculated by using ASIS and PSIS markers using the regression equation from Bell , Brand & Pedersen (1989).

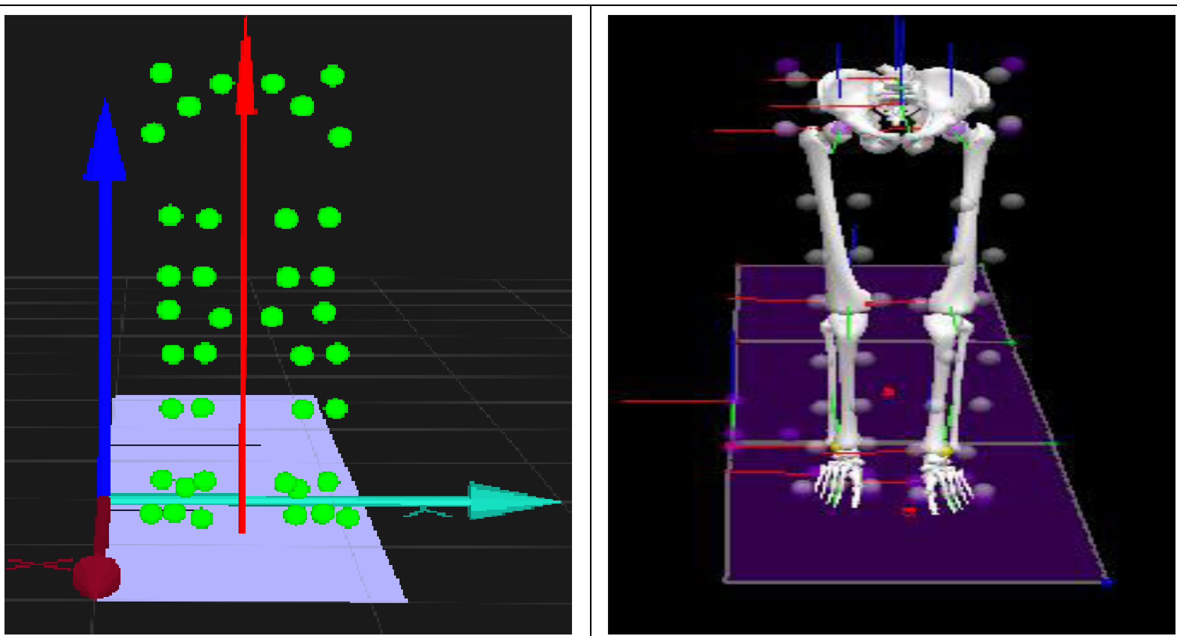


Figure 3.5. QTM™ static models (left), and Visual 3D™ bone model (right)

For the running and cutting tasks, kinematics and kinetic data were normalised to 100% of the right-leg contact phase. This was defined from right-leg initial contact (IC) to toe-off (TO). The initial contact was defined when vertical GRF first exceeded 10 Newtons (N). Toe-off (TO) was defined when VGRF fell under 10 N. During the SLS task, the starting phase began when the right knee exceeded 15° of flexion, and ended when returning to this point while ascending after the task. During the SLL task, the event was defined from IC until 15° ascending of knee flexion of the right leg; this was chosen to make sure that maximum knee flexion was included in the SLL cycle.

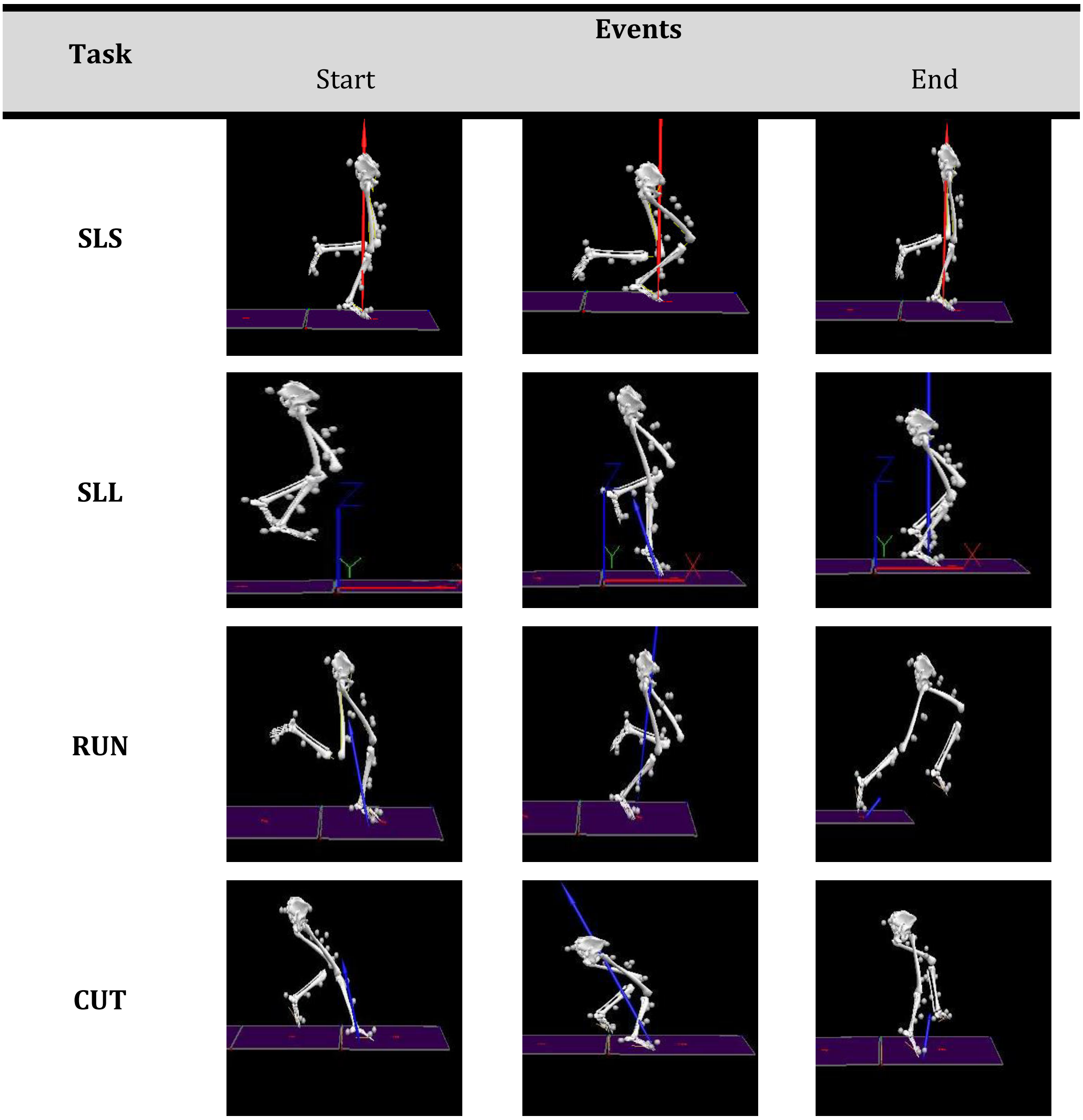


Figure 3.6. Events during SLS, SLL, RUN and CUT tasks.

3.5.2.7. Main outcome measures

On the basis of their frequent use in relation to possible biomechanical risk factors for ACL and PFPS injuries and gender-comparison studies, as discussed in Chapter 2, the following variables were measured for the right leg during each trial:

a) Peaks of hip-flexion, adduction and internal-rotation angles and moments.
b) Peaks of knee-flexion, valgus and internal-rotation angles.
c) Peaks of knee-flexion and valgus moments.
d) Peak ankle dorsiflexion angle and moment.
e) Peak vertical ground-reaction force (VGRF).

3.5.2.8. Statistical Analysis

Statistical analysis was performed using SPSS (v. 21). The means of three trials from the first and second sessions were used for within-day reliability and the mean of the first and third session for between-days reliability. Intra-class correlation coefficients (ICC), model 3.3, were used to assess relative reliability. Since the principal investigator performed all the measurements, these results are not generalisable to other raters, thus the two-way-mixed model was used (Shrout and Fleiss, 1979). The first number indicates the use of the two-way-mixed model of ICC, whereas the second number represents the use of an average measurement (Portney & Watkins, 2009). The levels of ICC were interpreted according to the criteria shown in Table 3.2 (Coppieters, Stappaerts, Janssens, & Jull, 2002).

Table 3.2. ICC values and corresponding levels

ICC Value	Interpretation
Less than 0.40	Poor
0.40 – 0.75	Fair
0.75 – 0.90	Good
More than 0.90	Excellent

Although the ICC appears to be easy to interpret, the closer it is to one the greater is the reliability, it alone cannot provide a full picture of reliability and should be complemented by confidence intervals (CI). Additionally, ICC does not provide any indication of the amount of disagreement between measurements. A low standard error of measurement (SEM) with high ICC indicates good reliability of a measure. Therefore, SEM and smallest detectable difference (SDD) were used in conjunction with ICC and a CI of 95%.

Calculation of SEM was done using the formula: $\mathrm{SD}\sqrt{1-\mathrm{ICC}}$ (Denegard & Ball, 1993). The following formula was used to calculate SDD values: SDD = 1.96 * ($\sqrt{2}$) * SEM (Kropmans et al., 1999). Both SEM & SDD are expressed in the units of the measurement tool used (degrees for joints angles, Newton-metres per kilogramme for moments around joints) (Blankevoort, van Heuvelen, & Scherder, 2013; Bruton, Conway, & Holgate, 2000).

3.6. Results

3.6.1. Pilot-study results

A normality check revealed that all variables in the pilot study were distributed normally. Paired t-tests revealed that there were no significant differences between right and left legs during the three tasks (see Table 3.3). The highest differences between limbs were found in hip adduction during SLL and in knee-valgus moment during RUN tasks (p=0.06 and 0.07, respectively). Knee-valgus angles were very similar between legs across the three tasks. Therefore, the decision was made to test the right leg throughout the study.

Table 3.3. Differences between legs during SLS, SLL and RUN tasks

Variable	SLS			SLL			RUN		
	Right	Left	*p*-value	Right	Left	*p*-value	Right	Left	*p*-value
Joint Angles (°)									
Hip adduction	11.3	10.0	**0.29**	1.02	2.13	**0.06**	7.96	8.14	**0.87**
Hip flexion	78.64	76.1	**0.45**	52.3	51.0	**9.65**	39.7	38.9	**0.65**
Hip Int. Rot.	10.6	10.0	**0.68**	6.81	5.55	**0.79**	6.40	5.21	**0.46**
Knee valgus	4.31	4.09	**0.79**	4.12	4.00	**0.94**	-0.35	-0.9	**0.42**
Knee flexion	88.6	83.4	**0.09**	65.9	67.0	**0.61**	44.5	43.5	**0.51**
Dorsiflexion	38.8	36.7	**0.16**	28.3	28.3	**0.97**	28.7	27.5	**0.20**
Moments (Nm/Kg)									
Hip adduction	-0.94	-1.01	**0.06**	-1.85	-2.03	**0.26**	-1.43	-2.02	**4.19**
Hip flexion	-1.16	-1.14	**0.84**	-1.73	-2.01	**0.46**	-1.58	-1.60	**0.79**
Knee valgus	-0.22	-0.27	**0.32**	0.03	-0.09	**0.41**	0.11	0.06	**0.07**
Knee flexion	1.96	1.82	**0.07**	3.29	3.40	**0.44**	3.36	3.28	**0.73**
Dorsiflexion	-0.86	-0.84	**0.83**	-2.20	-2.19	**0.95**	-2.43	-2.50	**0.26**
Force (* body weight)									
VGRF (*BW)	1.12	1.12	**0.93**	3.67	3.45	**0.09**	2.42	2.44	**0.68**

Sign conventions; all variables are reported in positive values apart from the following variables are presented in negative values (knee valgus angle, hip adduction and flexion moments, and ankle dorsiflexion moment).

Based on the use of three trials of SLS, all variables were normally distributed (Shapiro-WLK ≥ 0.05) apart from hip-adduction moment in the second session (p= 0.031), as shown in Appendix B-1. The within-day ICC values for all variables (ICC = 0.70–0.95; Table 3.3) were generally greater than for between days (ICC= 0.63–0.94). Within-day ICCs were good to excellent, apart from peak-ankle dorsiflexion moment (ICC = 0.70). The poorest between-day ICC value was for hip-adduction moment (ICC = 0.63). SEM values, as shown in Table 3.4, range from 1.18°–4.48° for joint angles and between 0.06 and 0.13 Nm-kg for both sagittal and frontal-plane moments. Hip flexion recorded the highest SEM values for both within- & between-days reliability (4.48° & 5.42°, respectively). Furthermore, the SDD values for the hip flexion were high as well (within day = 12.41°; between days= 15.02°).

Table 3.4. Within- & between-days ICC (95%CI), Mean, SEM & SDD values during the SLS task

Variable	Within day				Between days			
	ICC (95% CI)	Mean	SEM	SDD	ICC (95% CI)	Mean	SEM	SDD
Joint Angles (°)								
Hip adduction	0.93 (0.81-0.98)	16.19	1.91	5.29	0.94 (0.83-.098)	15.9	1.52	4.21
Hip flexion	0.93 (0.81-0.98)	68.37	4.48	12.4	0.88 (0.68-.096)	69.1	5.42	15.0
Hip Int. Rot.	0.78 (0.46-0.92)	6.08	2.86	7.92	0.78 (0.46-.092)	6.58	3.29	9.11
Knee valgus	0.87 (0.66-0.95)	-3.64	1.74	4.82	0.84 (0.59-.094)	-3.32	1.82	5.04
Knee flexion	0.94 (0.83-0.98)	91.78	2.31	6.40	0.84 (0.59-.094)	92.3	3.48	9.64
Knee Int. Rot.	0.78 (0.46-0.92)	3.75	2.05	5.68	0.82 (0.54-.094)	3.15	2.58	7.15
Dorsiflexion	0.95 (0.86-0.98)	42.83	1.18	3.27	0.95 (0.86-.098)	43.1	1.11	3.07
Moments (Nm/Kg)								
Hip adduction	0.94 (0.83-0.98)	-1.08	0.06	0.16	0.63 (0.19-0.86)	-1.09	0.13	0.36
Hip flexion	0.95 (0.86-0.98)	-0.75	0.09	0.24	0.81 (0.51-0.93)	-0.79	0.18	0.49
Knee valgus	0.78 (0.46-0.92)	0.10	0.07	0.19	0.78 (0.46-.092)	0.07	0.08	0.22
Knee Flexion	0.87 (0.66-0.95)	1.95	0.09	0.24	0.94 (0.83-0.98)	1.96	0.06	0.16
Dorsiflexion	0.70 (0.31-0.89)	-1.08	0.13	0.36	0.81 (0.52-0.93)	-1.06	0.11	0.30
Force (*body weight)								
Vertical GRF	0.89 (0.70-0.96)	1.13	0.02	0.05	0.88 (0.68-.096)	1.12	0.02	0.05

Sign conventions; all variables are reported in positive values apart from the following variables are presented in negative values (knee valgus angle, hip adduction and flexion moments, and ankle dorsiflexion moment).

In the SLL task, 6 out of 39 variables during all sessions were non-normally distributed (Appendix B-2). As shown in Table 3.5, within-day ICC values for SLL ranged between (0.57–0.98, while the between-day ICCs ranged between 0.61–0.96) The SEM values ranged between 1.10°–5.20° for angles and between 0.09–0.58 Nm-kg for moments. Hip-internal rotation angle recorded the highest SDD values for both within- and between-days reliability (8.67° & 14.41°, respectively). The within-day ICC value for hip-adduction moment recorded the lowest among all the variables at 0.57.

Table 3.5. Within- & between-days ICC (95%CI), Mean, SEM & SDD values during the SLL task

Variable	**Within day**				**Between days**			
	ICC (95% CI)	Mean	SEM	SDD	ICC (95% CI)	Mean	SEM	SDD
Joint Angles (°)								
Hip adduction	0.92 (0.78-0.97)	8.80	1.75	4.85	0.81 (0.52-.093)	7.90	2.38	6.59
Hip flexion	0.98 (0.94-0.99)	49.2	1.83	5.07	0.90 (0.73-0.97)	49.7	3.77	10.4
Hip Int. Rot.	0.76 (0.42-0.91)	7.14	3.13	8.67	0.60 (0.15-0.85)	6.48	5.20	14.4
Knee valgus	0.92 (0.78-0.97)	-5.84	1.71	4.73	0.61 (0.16-0.85)	-6.15	3.60	9.97
Knee flexion	0.96 (0.89-0.99)	70.2	2.62	7.26	0.95 (0.86-0.98)	70.0	2.92	8.98
Knee Int. Rot.	0.60 (0.15-0.85)	5.21	2.95	8.17	0.66 (0.24-0.87)	4.64	3.97	11.0
Dorsiflexion	0.97 (0.91-0.99)	28.6	1.10	3.04	0.96 (0.89-0.99)	28.4	1.26	3.49
Moments (Nm/Kg)								
Hip adduction	0.57 (0.10-0.83)	-1.84	0.58	1.60	0.74 (0.38-0.90)	-2.03	0.24	0.66
Hip flexion	0.91 (0.75-0.97)	-2.25	0.28	0.77	0.83 (0.57-0.94)	-2.43	0.51	1.41
Knee valgus	0.80 (0.50-0.93)	0.64	0.18	0.49	0.66 (0.24-0.87)	0.59	0.22	0.60
Knee flexion	0.94 (0.83-0.98)	3.35	0.09	0.24	0.90 (0.73-0.97)	3.37	0.12	0.33
Dorsiflexion	0.95 (0.86-0.98)	-2.41	0.24	0.66	0.71 (0.33-0.89)	-2.47	0.75	2.07
Force (*body weight)								
Vertical GRF	0.98 (0.94-0.99)	4.36	0.12	0.33	0.95 (0.86-0.98)	4.42	0.20	0.55

Sign conventions; all variables are reported in positive values apart from the following variables are presented in negative values (knee valgus angle, hip adduction and flexion moments, and ankle dorsiflexion moment).

A normality check for the running task revealed that 6 out of 39 were non-normally distributed (Appendix B-3). As shown in Table 3.6, within-day ICC values for kinematic and kinetic variables collected during the run trials ranged between 0.64–0.94, while the between-day ICCs ranged between 0.51–0.91. SEM values ranged between 1.98°–5.14° for angles and between 0.09–0.58 Nm-kg for moments. The poorest ICC value was for hip-adduction angle in between-day measurement, at 0.51. Hip-flexion angle recorded the highest SEM and SDD values for both within- and between-days reliability (SEM= 5.14° & 4.74°; SDD= 14.24° & 13.13°, respectively). The average speed during running was 4.99± 0.5 $m \cdot s^{-1}$, with ICC values of 0.91 to 0.95.

Table 3.6. Within- & between-days ICC (95%CI), Mean, SEM & SDD values during the run task

Variables	Within-day				Between-days			
	ICC (95%CI)	Mean	SEM	SDD	ICC (95%CI)	Mean	SEM	SDD
Joint Angles (°)								
Hip adduction	0.75 (0.40-0.91)	17.3	1.99	5.51	0.51 (0.02-0.80)	17.1	2.49	6.90
Hip flexion	0.74 (0.38-0.90)	54.7	5.14	14.2	0.65 (0.23-0.87)	55.3	4.74	13.1
Hip Int. Rot.	0.76 (0.42-0.91)	2.54	2.46	6.81	0.72 (0.35-0.90)	3.03	3.08	8.53
Knee valgus	0.94 (0.83-0.98)	-7.04	0.98	2.71	0.61 (0.16-0.85)	-7.23	2.41	6.68
Knee flexion	0.63 (0.19-0.86)	53.5	3.68	10.2	0.67 (0.26-0.88)	53.7	3.23	8.95
Knee Int. Rot.	0.74 (0.38-0.90)	5.25	2.84	7.87	0.58 (0.12-0.84)	3.47	3.62	10.0
Dorsiflexion	0.78 (0.46-0.92)	33.1	1.98	5.48	0.71 (0.33-0.89)	33.0	2.42	6.70
Moments (Nm/Kg)								
Hip adduction	0.64 (0.21-0.86)	-2.38	0.39	1.08	0.69 (0.29-0.88)	-2.36	0.30	0.83
Hip flexion	0.81 (0.52-0.93)	-2.84	0.44	1.21	0.83 (0.57-0.94)	-2.84	0.38	1.05
Knee valgus	0.85 (0.61-0.95)	0.36	0.07	0.19	0.72 (0.35-0.90)	0.35	0.09	0.24
Knee flexion	0.70 (0.31-0.89)	2.63	0.22	0.60	0.58 (0.12-0.84)	2.67	0.25	0.69
Dorsiflexion	0.89 (0.70-0.96)	-3.06	0.15	0.41	0.91 (0.75-0.97)	-3.04	0.14	0.38
Force (*body weight)								
Vertical GRF	0.92 (0.78-0.97)	2.69	0.14	0.38	0.84 (0.59-0.94)	2.66	0.18	0.49

Sign conventions; all variables are reported in positive values apart from the following variables are presented in negative values (knee valgus angle, hip adduction and flexion moments, and ankle dorsiflexion moment).

All cutting variables were normally distributed (Shapiro-WLK ≥ 0.05) apart from hip-flexion angle in the third session (p= 0.009), as shown in Appendix B-4. It can be seen from the data in Table 3.7 that the within-day ICC values for kinematic and kinetic variables collected during the cutting task ranged between 0.63–0.96, while the between-day ICCs ranged between 0.42–0.92. SEM values ranged between 1.73° and 5.15° for angles and between 0.14–0.56 Nm-kg for moments. The poorest ICC value was for knee-internal rotation angle in between-day measurement, at 0.42. Hip-internal rotation angle recorded the highest SEM and SDD values for both within- and between-days reliability (SEM= 3.81° & 5.15°; SDD= 10.56° & 14.27°, respectively). The average speed was 3.8 ± 0.4 m/sec. with ICC values between 0.89 and 0.94.

Table 3.7. Within- & between-days ICC (95%CI), Mean, SEM & SDD values during the CUT task

Variable	**Within day**				**Between days**			
	ICC (95%CI)	Mean	SEM	SDD	ICC (95%CI)	Mean	SEM	SDD
Joint Angles (°)								
Hip adduction	0.65 (0.23-0.87)	-7.15	3.37	9.14	0.60 (0.15-0.85)	-7.84	3.02	8.37
Hip flexion	0.93 (0.81-0.98)	48.41	2.49	6.90	0.75 (0.40-0.91)	49.1	4.98	13.8
Hip Int. Rot.	0.80 (0.50-0.93)	6.84	3.81	10.56	0.51 (0.02-0.80)	6.51	5.15	14.2
Knee valgus	0.93 (0.81-0.98)	-11.8	1.73	4.79	0.79 (0.48-0.92)	-11.6	3.02	8.37
Knee flexion	0.96 (0.89-0.99)	66.26	2.04	5.65	0.83 (0.57-0.94)	65.9	4.16	11.5
Knee Int. Rot.	0.63 (0.19-0.86)	7.31	2.71	7.51	0.42 (-0.1-0.76)	5.48	4.09	11.3
Dorsiflexion	0.88 (0.68-0.96)	30.95	2.24	6.20	0.80 (0.50-0.93)	30.2	3.82	10.5
Moments (Nm/Kg)								
Hip adduction	0.79 (0.48-0.92)	-0.76	0.22	0.60	0.88 (0.68-0.96)	-0.81	0.13	0.36
Hip flexion	0.94 (0.83-0.98)	-2.70	0.27	0.74	0.84 (0.59-0.94)	-2.91	0.56	1.55
Knee valgus	0.93 (0.81-0.98)	1.43	0.18	0.49	0.92 (0.78-0.97)	1.40	0.20	0.55
Knee flexion	0.82 (0.54-0.94)	3.30	0.16	0.44	0.83 (0.57-0.94)	3.25	0.18	0.49
Dorsiflexion	0.88 (0.68-0.96)	-2.46	0.14	0.38	0.87 (0.66-0.95)	-2.46	0.16	0.44
Force (*body weight)								
Vertical GRF	0.95 (0.86-0.98)	3.09	0.18	0.49	0.88 (0.68-0.96)	3.08	0.28	0.77

Sign conventions; all variables are reported in positive values apart from the following variables are presented in negative values (knee valgus angle, hip adduction and flexion moments, and ankle dorsiflexion moment).

3.7. Discussion

The purposes of this chapter were to:

1. Examine the with- and between-days reliability of using a 3D motion-analysis system to measure lower-limb biomechanical variables during SLS, SLL, RUN and CUT tasks.
2. Establish standard measurement error (SEM) and smallest detectable changes (SDD) during these tasks in healthy participants.

In the current study, the majority of between-day ICC values for joint angles, moments and vertical GRF were lower than within-day values across all tasks. Other investigators have found a similar trend during running (Ferber et al., 2002; Queen et al., 2006), drop vertical jumps (Ford et al., 2007), stepping down (Nakagawa et al., 2014), small knee-bending (Whatman et al., 2011; Whatman et al., 2013) and a 45° cutting task (Sankey et al., 2015). Transverse-plane variables (hip and knee-internal rotation angles) are less reliable compared to other planes of movement, which is in line with previous investigations (Ferber et al., 2002; Ford et al., 2007; Malfait et al., 2014; Nakagawa et al., 2014; Queen et al., 2006).

With respect to the 2nd hypothesis of this study, vertical GRF data were highly reliable during all tasks, with ICCs ranging between 0.84 and 0.98, which is in line with previous investigations (Ferber et al., 2002; Kadaba et al., 1989; Winter, 1984). These results may be explained by GRF values being representative of the sum of all segmental masses, accelerations and gravitational forces. Thus, no markers were needed to gather GRF data and so these did not suffer from marker-placement error and can be assumed to be more repeatable (Ferber et al., 2002; Kadaba et al., 1989; Winter, 1984).

Several factors influence both within- and between-days reliability, such as skin-marker movement, referenced static alignment and task difficulty (Ferber et al., 2002; Ford et al., 2007; Manal et al., 2000). Kadaba et al. (1989) attribute the variability of between-days measures to marker reapplication. In the current study, only one investigator attached the markers in all trials. The decreased between-days ICC values indicate that differences in marker replacement influenced the reliability, even when controlling for the tester.

To decrease this variability within the study, the CAST marker-based protocol (Cappozzo, Catani, Croce, & Leardini, 1995) was employed, which has the benefit of offering improved anatomical relevance compared to the modified Helen Hayes marker set (Kadaba et al., 1989) as it attempts to reduce skin-movement artefacts by attaching markers to the centre of segments rather than single markers close to the joints, as in the Helen Hayes model (Collins, Ghoussayni, Ewins, & Kent, 2009).

This study provides SEM and SDD reference values for SLS, SLL, RUN and CUT tasks that may be useful for evaluating intervention outcomes, (Tables 3.4–3.7). SEM is very useful for clinicians wanting to determine individual improvement (Munro et al., 2012b; Domholdt, 2005). The calculation of SEM depends on the standard deviation of measurements that allow the clinician to be 68% confident that the true value lies within ±1 SEM of an observed value (Portney and Watkins 1993). The SDD is based on SEM calculation, but it is more conservative (2.7 SEMs). If a score change is larger than the SDD, this difference is not caused by measurement error or patient variability with a probability of 95% (Ries, Echternach, Nof, & Blodgett, 2009; Wilken, Rodriguez, Brawner, & Darter, 2012).

The SEM values for peak knee-valgus angle during the SLS task were between 1.7° and 1.8° for within-day and between-days measures, respectively. This means that there was a 68% confidence that participants' true measures fell within a range of 3.6° if there was a 1-week gap between repeat measures. This range reduced to 3.4° if the two measurements were taken on the same day. Subsequently, there was a 95% chance that the true value lay within 5.0° if the gap between measures was 7 days and 4.8° when both measures were taken on the same day. Nakagawa et al. (2014) reported lower values than the ones reported in the current study for knee-valgus angle during the same task (SEM=0.5-1.5°; SDD=1.3-3.7°, within-day and between-days, respectively). This might be because their participants were younger than the current study's participants (21±1.1 vs. 26±4.1 years) and the between-days interval was shorter than in the current study (3 vs 7 days), resulting in improved ICC values and consequently lower SEM values.

To the best of the author's knowledge, this is the first study to provide measurement errors for a 90° sidestep cutting task. This task produced high SEM and SDD values for knee-valgus angle (SEM=1.73–3.02°; SDD=4.79º–8.37°). This finding suggests that an improvement of at least 8.3° in knee-valgus angle during cutting would be needed to say that the intervention had a significant effect above the measurement error with 95% confidence if the time interval between the two sessions was one week. None of the cutting-task literature provides measurement errors for knee-valgus angle (Besier et al., 2001; Sankey et al., 2015; Stephenson et al., 2012). The value of valgus angle reported in this study is lower than previous findings for the same cutting angle (Jones et al. 2014). However, the target population (female soccer players) and approach speed (4.0–5.0 m/sec.) might explain these differences.

Across all tasks, the highest SEM and SDD values were found with hip-flexion angles, particularly in between-day sessions (SEM= 3.7º–5.42°; SDD= 10.4º–15.0°), but these represent between 7.5% and 10.1% for SEM and between 20.9% and 28.0% for SDD when comparing their mean values (SLS= 69.1°; SLL=49.7°; RUN=55.4°; CUT=49.1°). This may be explained by the larger range of motion in the sagittal plane compared to other planes. Nakagawa et al. (2014) reported lower SEM and SDD values for hip flexion during an SLS task during both within day and between days (SEM=1.7 and 2.6°; SDD=4.7 and 7.1°). This might be because their participants were younger than the current study's participants (21±1.1 vs 26±4.1 years) and the between-days interval was shorter than in the current study (3 vs 7 days), resulting in improved ICC values and consequently lower SEM values. In the SLL task, the within-day and between-days SEM values for hip-flexion angles reported in the current study are lower than those reported for drop jumps with a 7-week gap (Ford et al., 2007) and a 10-week gap (Whatman et al., 2013). A direct comparison with RUN and CUT tasks in previous work is not possible, as none of the aforementioned running and cutting studies included the hip-flexion angle in their reliability analyses (Sankey et al., 2015; Queen et al., 2006; Ferber et al., 2002).

With regard to the third hypothesis of this study, the transverse-plane angles (hip and knee-internal rotation) demonstrated high levels of variability compared to other planes, specifically for between-days measurements of knee-internal rotation angle during cutting tasks (ICC=0.42; SEM=4.09°; SDD=11.3°). The ICC value was fair, based on the interpretations used in this study (Coppieters et al., 2002), but unfortunately the lower band of the 95% confidence interval crossed the zero level (-0.1–0.76). This finding is rather disappointing and, therefore, knee-internal rotation during cutting will not be carried forward to other chapters in this thesis. The cluster movement might explain the decline in cutting-rotation motion, since the cutting trials were done after completing SLS, SLL and RUN tasks. Another explanation for this decline might be the more dynamic nature of the cutting task compared to the other tasks in this study. Noehren et al. (2010) was the only attempt to improve between-days reliability by using a marker placement device. They found the largest reduction in SEM values was in the transverse plane during running tasks (reducing SEM to 57% and improving ICC by 7%). Future research should focus on this issue and how to improve the reliability of knee-rotation measurements taken during cutting tasks.

The generalisability of current study results is subject to several limitations. For example, these data only apply to our laboratory setting and models, though they are consistent with those previously reported; this, along with participant's ability to apply markers, could affect the results obtained in other workplaces. Moreover, the squat depth was not sufficiently controlled for each participant, though this reflects normal practice. Subjects were instructed to squat down on their right extremity as far as possible and return to a single-legged standing position without losing their balance.

An additional limitation of the current study is that participants wore standard trainers on a mondo running surface, which fails to represent typical shoe-surface interactions in real games, such as studded boots on grass and trainers on AstroTurf. Another limitation is that an uninjured population was assessed, but given that tasks are used as screening session, these should be helpful to researchers conducting out similar investigation. The reliability of these screening tasks in individuals with lower-limb injuries, such as ACL and PFPS, needs more exploration, since these injuries have been associated to excessive hip adduction and internal rotation, and to knee valgus and external rotation, during different functional tasks (Hewett et al., 2004; Willson & Davis, 2008a).

3.8. Conclusion

Based on the results of this study, all the hypotheses are accepted and the following results can be highlighted:

- The majority of between-day ICC values for joint angles, moments and vertical GRF were lower than within-day values across all tasks.
- Vertical GRF were more reliable than moments and angles results across all tasks.
- Transverse-plane variables (hip and knee internal-rotation angles) were less reliable compared to sagittal and frontal planes of movement across all tasks.

Chapter 4: Developing reference values for lower-limb biomechanics variables during single-leg squat (SLS), single-leg landing (SLL), running (RUN) and sidestep cutting (CUT) tasks

4.1. Aims

The aims of this chapter are to:

a. Establish reference values for 3D biomechanical variables during SLS, SLL, RUN and CUT tasks in a physically active population.
b. Differentiate between males and females when performing these tasks.

4.2. Background:

To understand what is an abnormal performance, typical or reference performances must first be defined. The literature lacks clear guidance with respect to reference values across a range of tasks, so the purpose of this study is to present reference data to better understand abnormal or suboptimal performance when it occurs. Several studies have been conducted to examine lower-limb biomechanics during various movements which mimic the real situation of PFPS injuries, such as single-leg squat and running (Tables 2-1 and 2-4), or ACL injuries, such as single-leg landing and cutting tasks (Tables 2-3 and 2-5). But the numbers of subjects participating in all of the aforementioned studies were limited. In addition to this, different biomechanical models were chosen for these studies (i.e. modified Helen Hayes or Calibrated Anatomical System Technique), making generalizing findings to a healthy population difficult. Also, the results from different marker sets cannot be directly compared (Collins et al., 2009).

Another question related to what a reference performance is the impact of gender. Previous literature only shows differences in adolescent females or limited tasks involving both genders (Beaulieu et al., 2008; Sigward and Powers, 2006; Ford et al., 2005; McLean et al. 2005b; Malinzak et al., 2001). It is therefore worth considering across tasks if males and females do actually perform differently.

To the best of our knowledge, only two studies (McNair & Prapavessis, 1999; Herrington & Munro, 2010) have utilized a large population to present reference values for target tasks. In a technical report, McNair & Prapavesis (1999) recruited 234 adolescent participants to obtain normative data on vertical ground-reaction forces only during landing, while Herrington and Munro (2010) used a two-dimensional system to obtain normative numbers for knee-valgus angle during drop jump landing for a population of 100 physically active participants. Although both studies reported valuable data for landing and jumping, they only investigated vertical GRF data or 2D valgus angles. In reviewing the literature, far too little attention has been paid to reference values for both kinematic and kinematic data during common screening tasks in the same population.

The aims of this study are, first, to provide reference values for both kinematic and kinematic data during single-leg squats (SLS), single-leg landing (SLL), running (RUN) and 90° cutting (CUT) tasks in the same cohort population. The second aim is to differentiate between males and females when performing these tasks. Based on the available literature, it has been hypothesized that angles and moments during cutting and running will be greater than those obtained from SLS and SLL tasks. Furthermore, females in all tasks will demonstrate higher knee-valgus, hip-adduction and internal rotations compared to their male counterparts.

4.2.1 Hypotheses

Based on the aforementioned literature review, two hypotheses are formulated:

- H_1= Joint angles and moments during cutting and running will be greater than those obtained from SLS and SLL tasks.
- H_2= Females in all tasks will demonstrate higher knee-valgus, hip-adduction and internal rotations compared to their male counterparts.

4.3. Methods

4.3.1. Participants

A total of 90 healthy participants, whose demographics are listed in Table 4.1, all of whom were from a university population (students and staff) took part of this study. The same inclusion and exclusion criteria were employed as earlier explained in Chapter 3 (section 3.5.2.1). Before testing, each participant read and signed a written informed consent statement, approved by the Research, Innovation and Academic Engagement Ethical Approval Panel at the University of Salford (Appendix A-2).

Table 4.1: Demographic information for all participants

Demographic		Number	Mean	SD	Min	Max
Age (years)	Females	35	26.6	3.8	20	34
	Males	55	27.0	5.3	18	38
	All	90	26.8	4.7	18	38
Height (cm)	Females	35	165.3	5.96	152.4	178.5
	Males	55	173.7	6.87	159.0	188.0
	All	90	170.5	7.68	154.4	188.0
Mass (kg)	Females	35	62.29	6.63	53.0	78.40
	Males	55	74.39	11.2	53.0	104.0
	All	90	69.6	11.3	53.0	104.0
Running speed (m/sec.)	Females	35	4.45	0.40	4.05	5.3
	Males	55	5.15	0.40	4.30	6.0
	All	90	4.86	0.54	4.05	6.0
Cutting speed (m/sec.)	Females	35	3.51	0.25	3.01	4.04
	Males	55	3.76	0.34	3.15	4.50
	All	90	3.66	0.34	3.01	4.50

Centimetre (cm); kilogramme (kg); meters per second (m/sec.); minimum (Min); maximum (Max); standard deviation (SD)

4.3.2. Procedure

Kinematic data were collected using a ten-camera motion analysis system (Pro-Reflex, Qualisys, Sweden), sampled at 240 Hz. Kinetic data were collected using three force platforms embedded into the floor (AMTI, USA), sampled at 1200 Hz. The same instrumentation, calibration, filtration, training shoes, marker list and biomechanical model, was used as earlier described in the reliability study (Chapter three, Sections 3.5.2.2–7). The tasks (SLS, SLL, RUN and CUT) were conducted as previously described in Sections 3.5.2.5.1–4.

4.3.3. Statistical Analysis

Descriptive analysis (mean and standard deviation) covers each dependent variable in the target tasks (SLS, SLL, RUN, CUT). A Shapiro-WILK test was used to check whether data were normally distributed or not (parametric or non-parametric). Gender differences were examined using an independent t-test for parametric variables and a Mann-Whitney U test for non-parametric variables. Limb differences were examined using a paired t-test for parametric variables and a Wilcoxon Rank Test for non-parametric variables. The p-value was set at 0.05. All statistical analyses were performed using SPSS (v. 21, SPSS Inc., USA).

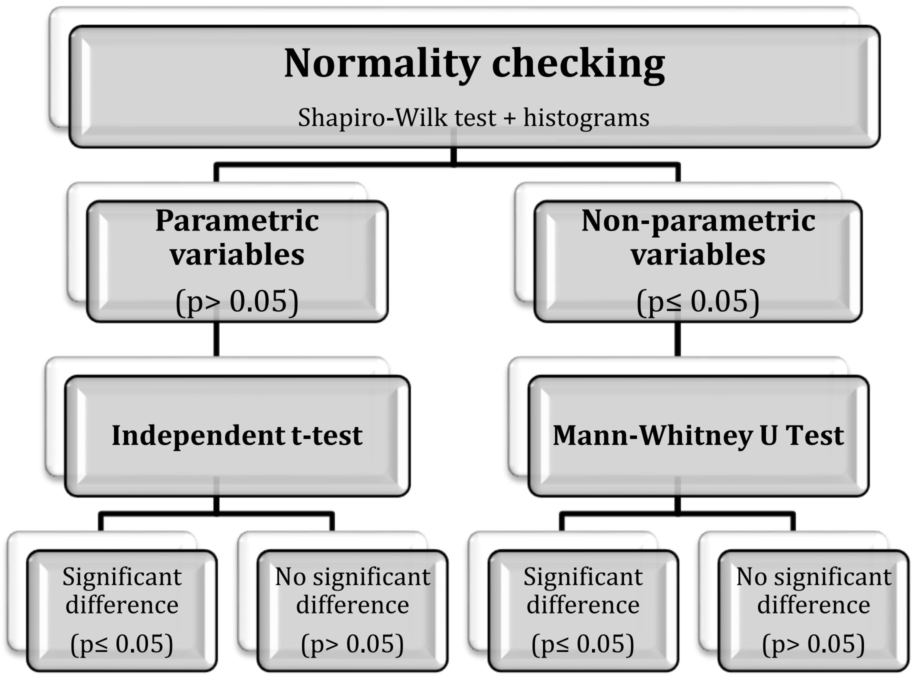


Figure 4.1: Statistical analysis outline for study two

4.4. Results

A total of 90 healthy participants completed three acceptable trials using their right leg in SLS, SLL, RUN and CUT tasks. In terms of the participants' primary recreational activities, 30% of the participants were soccer players, 12% were runners, 10% were cyclists, 7% volley ballers and 2% were rugby players. The rest did different sports, such as badminton, tennis and gymnastics.

4.4.1. SLS variables

Normality testing revealed that all variables were normally distributed apart from knee-internal rotation angle, ankle dorsiflexion angle and hip-adduction moment. See Appendix C-1 for the Shapiro Wilk test findings and histograms for all variables.

Table 4.2 presents a summary of normal values for 3D variables in an SLS task for a healthy population. The females performed the SLS tasks with significantly greater knee-valgus and hip-adduction angles ($p=0.04$; $p<0.001$, respectively), and less knee flexion, but not significantly so (93.1° vs. 91.4°), compared to their male counterparts.

Figures 4.2–4.3 illustrate time-normalised curves for hip and knee frontal-plane motions, hip-transverse motion and knee frontal-plane moments. The females started the squat in a quite neutral knee-frontal position and proceeded to a further knee adduction position as knee flexion angles increased and then ended the squat with a valgus position, whereas the males ended with a neutral position. All participants maintained their hip-internal rotation position throughout the squat cycle. Hip-adduction angle increased throughout squatting, reaching a peak point, for both females and males, at around 70% of the whole squat cycle.

Table 4.2. Reference values for 3D biomechanical variables during the SLS task

Variable	All participants			Females			Males			*P*-value	*ES*	PWR
	Mean	SD	Med	**Mean**	SD	Med	**Mean**	SD	Med			
Joint Angle (°)												
Hip add	**14.01**	6.4	14.6	**17.08**	6.4	16.6	**12.04**	5.6	11.5	<0.001*	0.83	0.96
Hip flexion	**73.7**	13.9	72.8	**72.58**	12.8	71.5	**74.39**	14.7	74.1	0.55	0.39	0.82
Hip Int. Rot.	**6.02**	7.80	7.41	**6.28**	6.5	7.31	**6.12**	8.6	7.51	0.87	0.02	0.20
Knee valgus	**-1.52**	4.1	-1.5	**-2.46**	3.9	-2.52	**-0.82**	4.3	-0.44	0.04*	0.38	0.80
Knee flexion	**92.4**	8.8	91.9	**91.30**	7.3	90.7	**93.13**	9.7	92.7	0.34	0.21	0.54
Knee Int. Rot.	**-0.85**	6.1	-0.81	**-1.61**□	6.5	-1.14	**-0.52**	5.9	-0.20	0.17	0.17	0.12
Dorsiflexion	**42.53**	5.1	43.1	**42.73**	4.9	44.09	**42.4**□	5.3	43.1	0.87	0.06	0.06
Moment (Nm/Kg)												
Hip add	**-1.13**	0.96	-1.03	**-1.08**	0.2	-1.08	**-1.16**□	1.2	-1.01	0.70	0.09	0.07
Hip flexion	**-0.94**	0.44	-0.88	**-0.86**	0.4	-0.76	**-0.99**	0.4	-0.92	0.16	0.33	0.32
Knee valgus	**0.002**	0.17	-0.02	**0.04**	0.1	0.02	**-0.02**	0.1	-0.04	0.11	0.38	0.81
Knee flexion	**1.94**	0.48	1.98	**1.90**	0.2	1.89	**1.97**□	0.6	2.10	0.47	0.16	0.76
Dorsiflexion	**-1.05**	0.24	-1.06	**-1.01**	0.2	-1.03	**-1.07**	0.2	-1.08	0.24	0.30	0.27
Force (* body weight)												
*VGRF (*BW)*	**1.13**	0.04	1.13	**1.12**	0.04	1.12	**1.13**	0.03	1.14	0.24	0.28	0.25

Int. Rot. = Internal Rotation; BW = body weight; SD = Standard Deviation; Med = Median; (*) Significantly different (p≤0.05); (n) = non-parametric variable; ES = Effect size; PWR = Power; Sign conventions; all variables are reported in positive values apart from the following variables are presented in negative values (knee valgus angle, hip adduction and flexion moments, and ankle dorsiflexion moment).

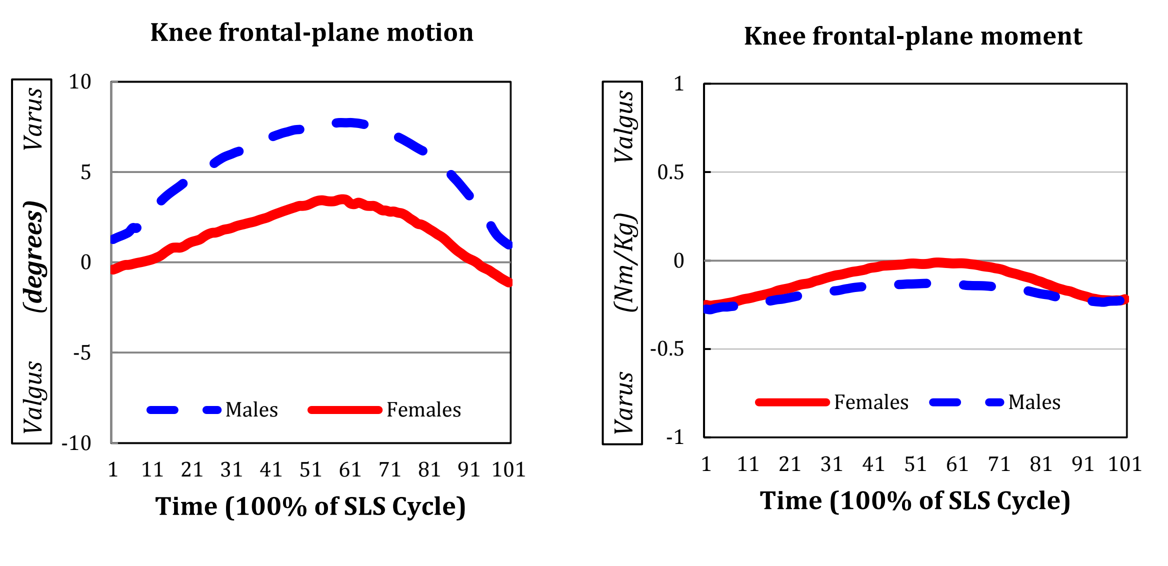


Figure 4.2. Ensemble average plot of knee frontal-plane motion (left) and moment (right) during the SLS task.

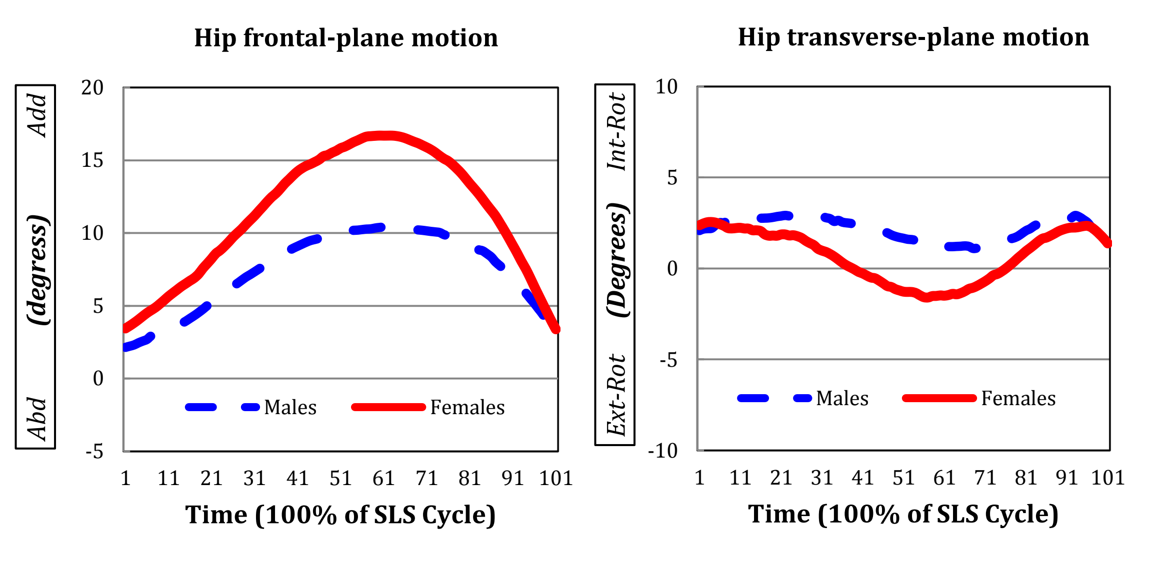


Figure 4.3. Ensemble average plot hip frontal motion (left) and hip transverse motion (left) during the SLS task

4.4.2. SLL variables

The normality checking process found that all joint angles were normally distributed apart from knee-internal rotation angle. Kinetically, knee-flexion moment was the only normally distributed variable among the tested moments. See Appendix C-2 for the Shapiro Wilk test findings and histograms for all variables measured in the SLL trials.

Table 4.3 shows the descriptive measures for peak values for hip-, knee- and ankle-joint angles and moments around these joints during the SLL task. Compared to their male counterparts, females performed the SLL with a significantly greater knee-hip adduction angle and flexion moment ($p<0.01$ and $p<0.01$, respectively), greater-knee valgus angle ($p=0.04$) and less knee flexion moment ($p=0.05$). The males produced significantly higher vertical GRF during SLL ($p=0.006$)

It can be noticed from Figures (4.4–4.5) that both genders reached their peak valgus angle and moment at around 20% of the stance phase. Participants touched the ground with an abducted hip position, which quickly changed to hip adduction, and this was maintained throughout the SLL cycle. The same scenario happened for hip-rotation motion, starting with external rotation, then the males progressed towards internal rotation while the females fluctuated around the neutral line during the entire cycle.

Table 4.3. Reference values for 3D biomechanical variables during the SLL task

Variables	All Participants			Females			Males			*P*-value	*ES*	PWR
	Mean	SD	Med	**Mean**	SD	Med	**Mean**	SD	Med			
Joint Angle (°)												
Hip add.	***7.72***	*6.07*	*8.10*	***10.40***	*5.05*	*10.54*	***6.02***	*6.0*	*5.45*	*<0.001**	*0.78*	*0.95*
Hip flexion	***53.8***	*11.4*	*52.6*	***56.2***	*12.8*	*58.36*	***52.3***	*10.1*	*50.65*	*0.11*	*0.33*	*0.95*
Hip Int. Rot.	***6.19***	*7.5*	*6.5*	***5.25***	*6.11*	*6.05*	***7.05***	*8.3*	*6.81*	*0.18*	*0.10*	*0.95*
Knee valgus	***-4.22***	*4.9*	*-3.9*	***-5.11***	*5.43*	*-5.03*	***-3.49***	*4.68*	*-3.33*	*0.08**	*0.31*	*0.90*
Knee flexion	***71.3***	*9.81*	*69.9*	***72.23***	*11.0*	*69.9*	***70.68***	*9.29*	*70.15*	*0.32*	*0.15*	*0.90*
Knee Int. Rot.	***1.44***	*7.13*	*2.03*	***1.00***	*7.64*	*0.22*	***1.92***	*6.83*	*2.93*	*0.23*	*0.11*	*0.95*
Dorsiflexion	***28.82***	*5.0*	*28.5*	***30.1***	*5.35*	*29.65*	***28.0***	*4.81*	*27.69*	*0.05**	*0.41*	*0.65*
Moment (Nm/Kg)												
Hip add.	***-1.93***	*0.47*	*-1.90*	***-1.81***	*0.49*	*-1.73*	***-1.99***	*0.45*	*-1.95*	*0.07*	*0.38*	*0.95*
Hip flexion	***-2.06***	*0.92*	*-1.75*	***-1.65***	*0.57*	*-1.52*	***-2.30***	*1.01*	*-2.09*	*0.003**	*0.79*	*0.95*
Knee valgus	***0.49***	*0.31*	*0.45*	***0.57***	*0.40*	*0.47*	***0.44***	*0.23*	*0.42*	*0.16*	*0.39*	*0.95*
Knee flexion	***3.39***	*0.48*	*3.35*	***3.25***	*0.42*	*3.23*	***3.49***	*0.49*	*3.47*	*0.05**	*0.52*	*0.68*
Dorsiflexion	***-2.18***	*0.67*	*-2.06*	***-2.02***	*0.48*	*-1.90*	***-2.27***	*0.74*	*-2.17*	*0.15*	*0.40*	*0.95*
Force (* body weight)												
*VGRF (*BW)*	***4.10***	*0.72*	*4.04*	***3.93***	*0.69*	*3.73*	***4.24***	*0.71*	*4.24*	*0.006**	*0.44*	0.61

Int. Rot. = Internal Rotation; BW = body weight; SD = Standard Deviation; Med = Median; (*) Significantly different (p≤0.05); (n) = non-parametric variable; ES = Effect size; PWR = Power; Sign conventions; all variables are reported in positive values apart from the following variables are presented in negative values (knee valgus angle, hip adduction and flexion moments, and ankle dorsiflexion moment).

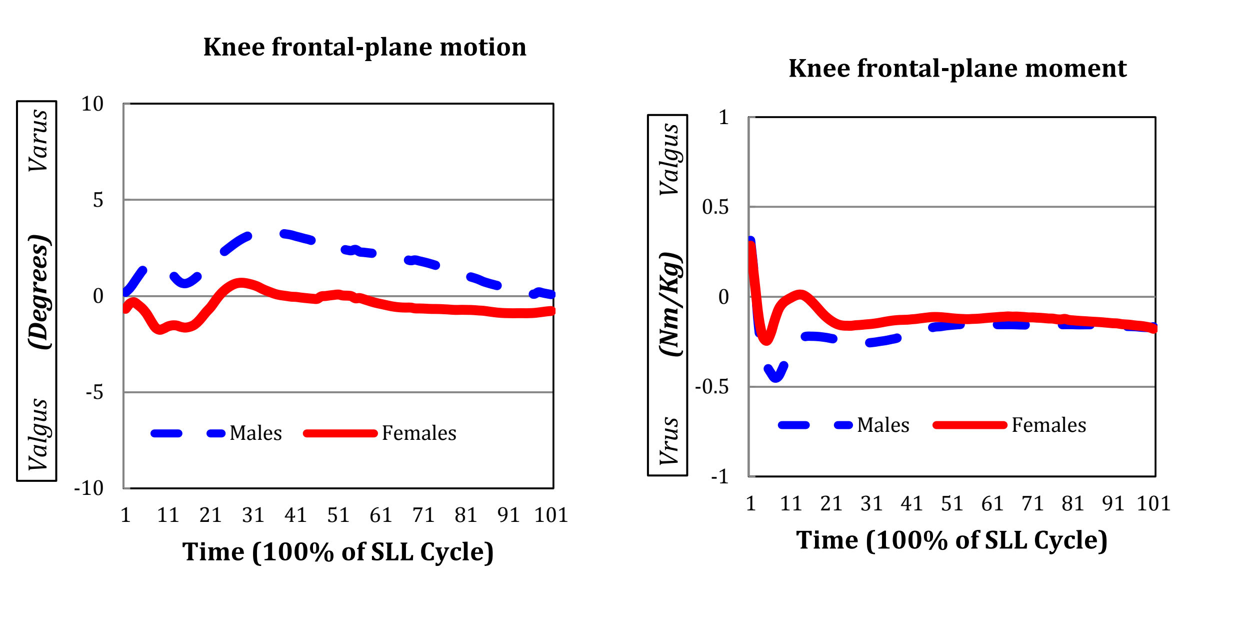


Figure 4.4: Ensemble average plot of knee frontal-plane motion (left) and moment (right) during the SLL task.

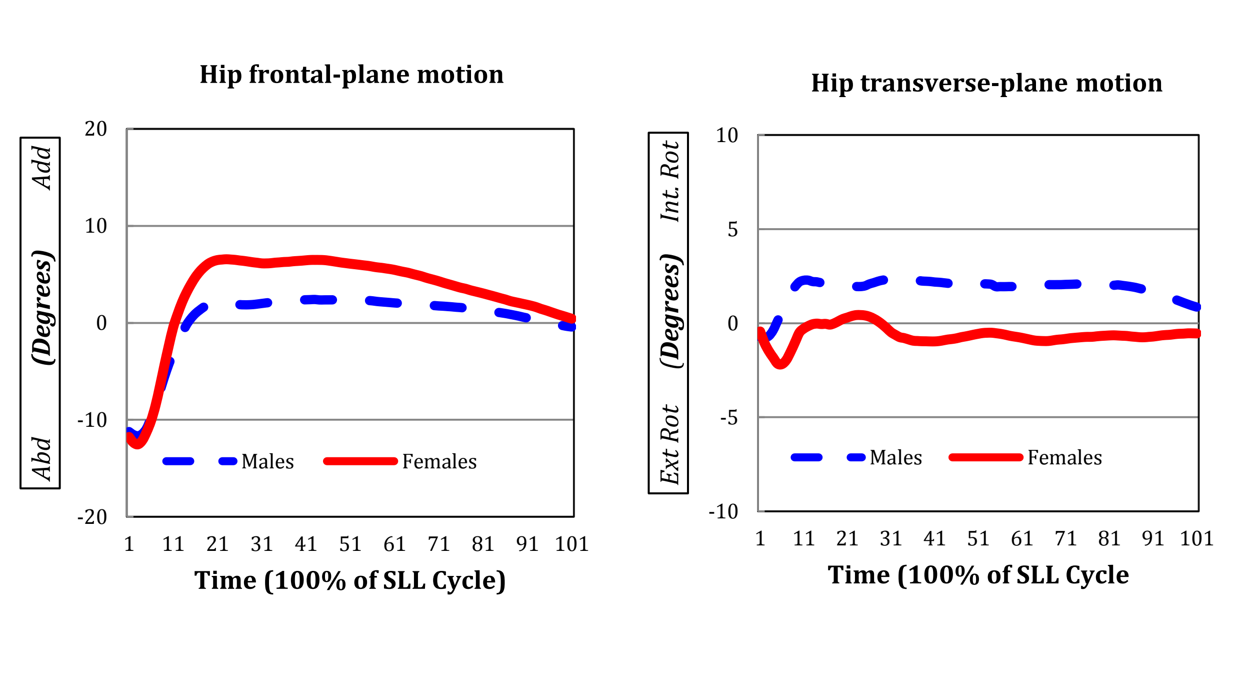


Figure 4.5: Ensemble average plot of hip frontal-plane motion (left) and hip-transverse motion (right) during the SLL task.

4.4.3. RUN variables

A Shapiro-Wilk test ($p \leq 05$) and histograms (Appendix C-3) confirmed the normality of the majority of the variables with the exception of the following ones: knee-valgus and internal-rotation angles, knee-flexion moment, dorsiflexion moment, vertical GRF.

Table 4.4 summarises the average of peak values for lower-limb angles and moment during running trials. Females demonstrated significantly greater peak hip-adduction ($p=<0.001$) and hip-flexion angles and moment ($p=0.01$; $p=<0.001$, respectively), knee-valgus angle ($p=0.01$) and lower valgus moment (0.25 vs. 0.34, NmKg), and dorsiflexion moment ($p=0.02$) compared to men. The participants performed the running trial with an average speed of 4.9 (±0.5) m/sec.

Ensemble average plots of hip and knee frontal-plane motions; hip-transverse motion and knee frontal-plane moments in the stance phase are graphically displayed in Figures 4.6–4.7. A visual inspection of these graphs reveals that the male participants touched the ground in a knee -varus position, and this decreased, leading to the knee-valgus angle being greatest at the toe-off position. The females started in and maintained a valgus position throughout the SLL task without touching the varus.

Both males and females touched the ground during the RUN task, with their hips slightly adducted, and progressed to a more adducted position, which lead to the hip-adduction angle being highest in the mid-stance phase. The pattern of hip rotation is almost identical in both gender groups, fluctuating between a neutral and an external-rotation position, as can be seen in Figure 4.7.

Table 4.4. Reference values for 3D biomechanical variables during the RUN task

Variables	**All participants**			**Females**			**Males**			***P*-value**	***ES***	**PWR**
	Mean	SD	Med	**Mean**	SD	Med	**Mean**	SD	Med			
Joint Angle (°)												
Hip add.	***15.17***	*4.58*	*15.3*	***17.46***	*3.99*	*17.07*	***13.59***	*4.24*	*13.91*	*<0.001**	*0.96*	*0.99*
Hip flexion	***59.78***	*10.5*	*59.6*	***56.28***	*10.5*	*55.47*	***61.92***	*10.0*	*60.92*	*0.01**	*0.55*	*0.96*
Hip Int. Rot.	***2.67***	*6.58*	*3.34*	***4.33***	*5.96*	*4.05*	***1.96***	*7.20*	*3.21*	*0.20*	*0.30*	*0.54*
Knee valgus	***-5.22***	*4.38*	*-4.73*	***-6.65***□	*4.37*	*-5.95*	***-4.31***	*4.17*	*-4.10*	*0.02**	*0.54*	*0.95*
Knee flexion	***56.4***	*5.73*	*56.38*	***55.7***	*5.79*	*54.99*	***56.9***	*5.69*	*56.93*	*0.31*	*0.20*	*0.95*
Knee Int. Rot.	***1.55***	*6.75*	*1.71*	***2.12***□	*7.31*	*1.89*	***1.06***	*6.39*	*1.05*	*0.23*	*0.15*	*0.95*
Dorsiflexion	***34.8***	*4.74*	*35.22*	***34.8***	*5.38*	*35.45*	***34.6***	*4.33*	*34.85*	*0.82*	*0.77*	*0.95*
Moment (Nm/Kg)												
Hip add.	***-2.25***	*0.65*	*-2.18*	***-2.18***	*0.41*	*-2.12*	***-2.29***	*0.77*	*-2.23*	*0.42*	*0.18*	0.95
HipfFlexion	***-3.10***	*0.95*	*-3.08*	***-2.56***	*0.61*	*-2.40*	***-3.45***	*0.97*	*-3.57*	*<0.001**	*0.77*	0.95
Knee valgus	***0.31***	*0.17*	*0.28*	***0.25***□	*0.16*	*0.22*	***0.34***	*0.18*	*0.31*	*0.007**	*0.52*	0.95
Knee flexion	***2.97***	*0.48*	*2.93*	***2.90***	*0.48*	*2.86*	***3.02***	*0.48*	*3.04*	*0.28*	*0.79*	0.95
Dorsiflexion	***-3.03***	*0.41*	*-2.95*	***-2.90***□	*0.43*	*-2.8*	***-3.11***	*0.38*	*-3.07*	*0.01**	*0.51*	0.95
Force (* body weight)												
*VGRF (*BW)*	***2.62***	*0.39*	*2.54*	***2.60***	*0.36*	*2.58*	***2.63***□	*0.41*	*2.51*	*0.99*	*0.20*	0.95

Int. Rot. = Internal Rotation; BW = body weight; SD = Standard Deviation; Med = Median; (*) Significantly different (p≤0.05); (n) = non-parametric variable; ES = Effect size; PWR = Power; Sign conventions; all variables are reported in positive values apart from the following variables are presented in negative values (knee valgus angle, hip adduction and flexion moments, and ankle dorsiflexion moment).

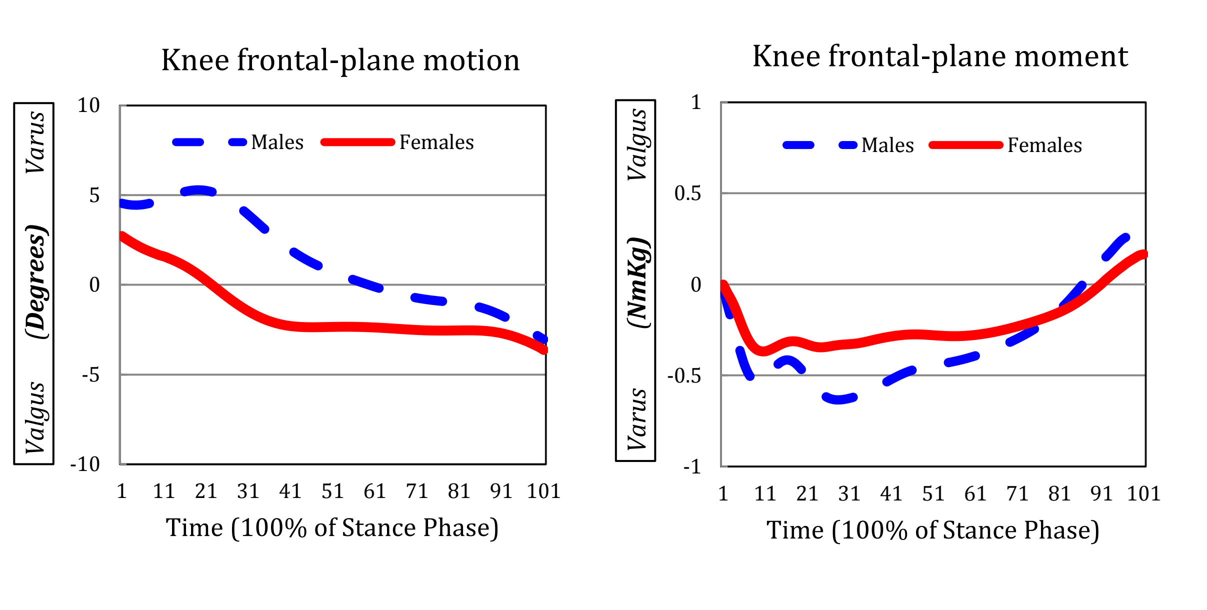


Figure 4.6: Ensemble average plot of knee frontal-plane motion (left) and moment (right) during the RUN task

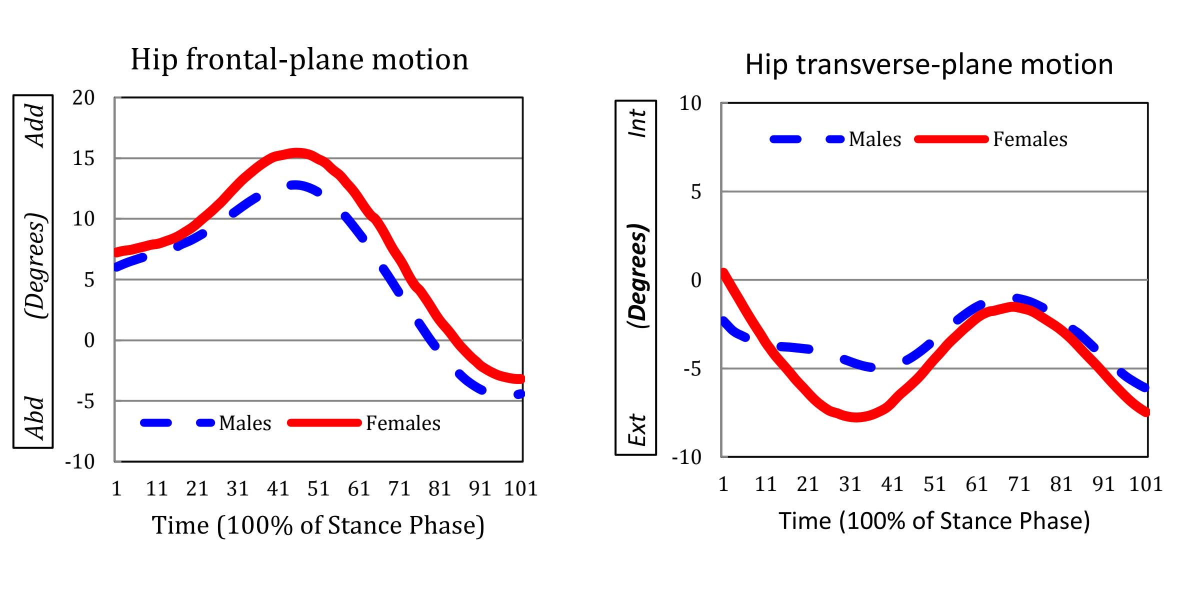


Figure 4.7. Ensemble average plot of hip frontal motion (left) and hip transverse motion (left) during the RUN task

4.4.4. CUT variables

Normality was confirmed for the majority of the variables using a Shapiro-Wilk test ($p \leq 05$) and histograms, see Appendix C-4, with the exception of knee-flexion angle, hip-flexion moment, knee-valgus moment and ankle-dorsiflexion moment.

Table 4.5 summarises the averages of peak values for lower-limb angles and moments during cutting trials. Females landed with a significantly greater knee-hip adduction angle ($p=0.05$), knee-valgus angle and moment ($p=0.05$; $p=<0.001$, respectively) and lower hip and knee flexion moment and angle ($p=0.05$; $p=0.02$, respectively). Participants performed the cutting task at an average speed of 3.8 ± 0.4 $m \cdot s^{-1}$.

As shown in Figures 4.8–4.9, knee-valgus angle and moment peaked at 10% of the stance phase. Males and females sustained an abducted hip position during the cutting task. On the contrary, they touched the ground with hip-internal rotation and then moved into an external-rotation position, leading to the hip-external rotation angle being greatest in the late stance phase.

Table 4.5. Reference values for 3D biomechanical variables during the CUT task

Variable	**All participants**			**Females**			**Males**			***P*-value**	***ES***	**PWR**
	Mean	SD	Med	**Mean**	SD	Med	**Mean**	SD	Med			
Joint Angle (°)												
Hip add.	***-6.75***	*5.3*	*-6.4*	***-5.07***	*6.01*	*-4.87*	***-7.48***	*4.9*	*-7.14*	*0.05**	*0.43*	*0.52*
Hip flexion	***50.8***	*10.4*	*51.59*	***49.4***	*11.0*	*49.51*	***51.7***	*10.0*	*51.75*	*0.30*	*0.41*	*0.95*
Hip Int. Rot.	***7.83***	*9.53*	*7.83*	***8.04***	*9.15*	*8.69*	***7.94***	*9.70*	*7.05*	*0.89*	*0.10*	*0.95*
Knee valgus	***-8.36***	*6.1*	*-7.64*	***-9.97***	*6.32*	*-9.71*	***-7.19***	*5.8*	*-6.41*	*0.02**	*0.45*	*0.75*
Knee flexion	***66.3***	*8.3*	*65.83*	***64.2***	*7.60*	*64.49*	***67.6***□	*8.5*	*66.78*	*0.10*	*0.42*	*0.46*
Knee Int. Rot.	*This variables found to be un-reliable based on reliability study finding (chapter 3)*											
Dorsiflexion	***28.1***	*8.47*	*27.9*	***28.0***	*7.75*	*28.48*	***28.1***	*8.9*	*27.57*	*0.97*	*0.10*	*0.95*
Moment (Nm/Kg)												
Hip add.	***-0.89***	*0.46*	*-0.85*	***-1.00***	*0.52*	*-0.97*	***-0.83***	*0.43*	*-0.82*	*0.11*	*0.35*	0.95
Hip flexion	***-2.96***	*1.3*	*-2.64*	***-2.53***□	*1.17*	*-2.25*	***-3.23***□	*1.3*	*-2.85*	*0.003**	*0.56*	0.95
Knee valgus	***1.21***	*0.67*	*1.03*	***0.88***	*0.37*	*0.79*	***1.38***□	*0.74*	*1.33*	*0.001**	*0.46*	0.95
Knee flexion	***3.47***	*0.63*	*3.44*	***3.28***	*0.50*	*3.23*	***3.59***	*0.68*	*3.55*	*0.02**	*0.68*	0.66
Dorsiflexion	***-2.50***	*0.66*	*-2.40*	***-2.42***□	*0.57*	*-2.28*	***-2.55***	*0.71*	*-2.48*	*0.33*	*0.98*	0.95
Force (* body weight)												
*VGRF (*BW)*	***2.97***	*0.73*	*2.83*	***2.89***	*0.80*	*2.68*	***3.00***	*0.69*	*2.95*	*0.66*	*0.15*	0.95

Int. Rot. = Internal Rotation; BW = body weight; SD = Standard Deviation; Med = Median; (*) Significantly different (p≤0.05); (n) = non-parametric variable; ES = Effect size; PWR = Power; Sign conventions; all variables are reported in positive values apart from the following variables are presented in negative values (knee valgus angle, hip adduction and flexion moments, and ankle dorsiflexion moment).

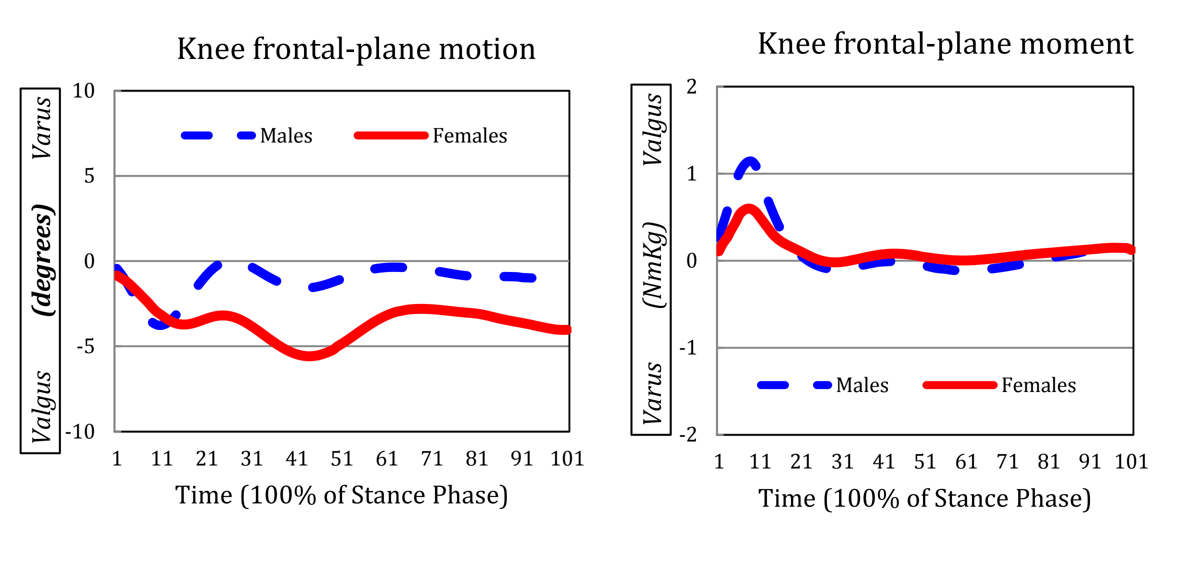


Figure 4.8. Ensemble average plot of knee frontal-plane motion (left) and moment (right) during the CUT task.

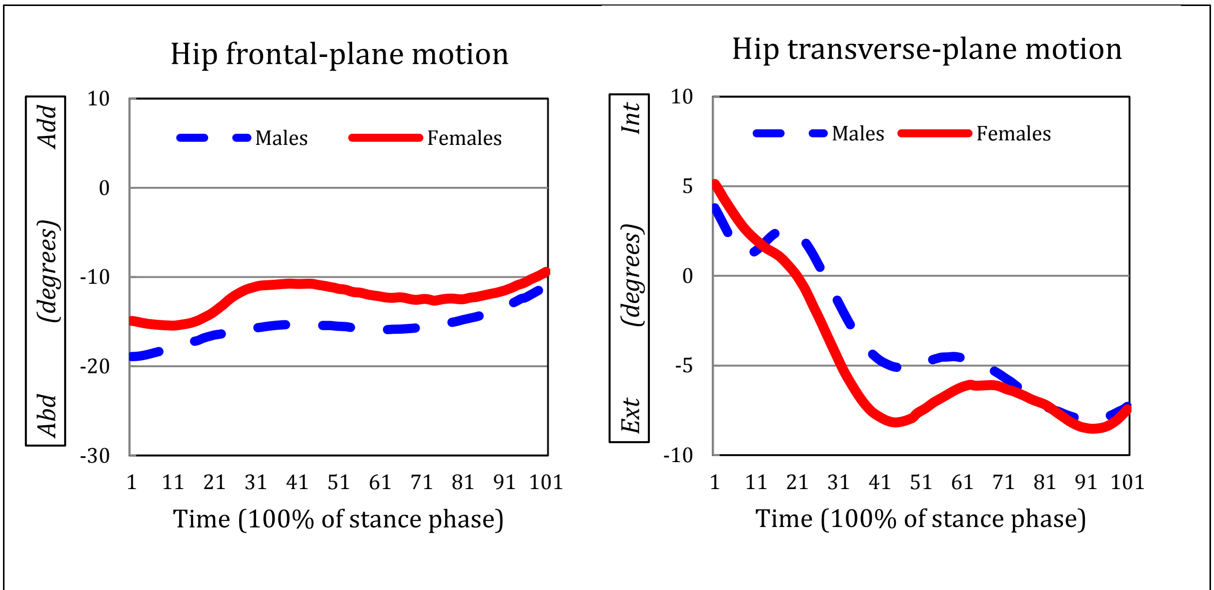


Figure 4.9. Ensemble average plot of hip frontal motion (Lleft) and hip transverse motion (left) during the CUT task.

4.5. Discussion

The goals of this chapter were:

c. To develop reference values for 3D biomechanical variables during SLS, SLL, RUN and CUT tasks in a physically active population.
d. To differentiate between males and females when performing these tasks.

Several studies have been conducted to address lower-limb biomechanics during various functional performance tasks which mimic the real situation of PFPS injuries, such as single-leg squats and running (Chapter 2, Tables 2-1 and 2-4), or ACL injuries, such as single-leg landing and cutting tasks (Chapter 2, Tables 2-3 and 2-5). The existing literature fails to establish reference values for biomechanical variables when performing functional tasks, due to the limited numbers of subjects participating and the different methods employed. However, it is important to establish reference values for both kinematic and kinematic data in a healthy population.

According to the SLS findings in the present study (Table 4.3), the average values for a healthy female range from 1.4° to -6.2° for knee valgus and between 10.6° and 23.4° for hip adduction during their SLS performances. Likewise, males' average performances were found to range from 3.4º to -5.1° and from 6.4° to 17.6° for knee valgus and hip adduction, respectively. What is surprising is that the knee-valgus values for both genders were lower than those reported by previous researchers for these tasks (Baldon et al., 2011; DiMattia et al., 2005; Nakagawa et al., 2012b; Zeller et al., 2003). A possible explanation for this might be differences in marker-list models between studies. However, the results from different marker sets cannot be directly compared (Collins et al., 2009). The CAST marker base used in the current study has the advantage of offering improved anatomical relevance compared to the modified Helen Hayes marker set (Kadaba et al., 1989), as it attempts to reduce skin-movement artefacts by attaching markers to the centres of segments rather than single markers close to the joints, as in the Helen Hayes model.

Gender disparities in SLS variables were observed in the current study. As expected, females showed significantly larger knee-valgus and hip-adduction motion and less knee flexion motion, but not significantly so, compared to males; these differences were greater than the SEM values reported in the reliability study of in thesis (Table 3.4, Chapter 2). Other researchers have found similar differences between the genders for SLS (Dwyer et al., 2010; Yamazaki et al., 2010; Zeller et al., 2003). A possible explanation for this might be the strength difference between the genders; this study did not include a strength assessment but previous investigations have shown that females exhibit lower peak isometric and isokinetic strength measures for hips and knees compared with males (Claiborne, Armstrong, Gandhi, & Pincivero, 2006b; Dwyer et al., 2010; Jacobs, Uhl, Mattacola, Shapiro, & Rayens, 2007; Willson et al., 2006). The current study and that of Dwyer et al. (2010) do not report differences between the genders in transverse-plane motion (hip and knee internal rotations), whereas Zeller et al. (2003) did report differences in hip transverse motion. These differences could be due to the different target populations. Zeller et al. (2003) targeted young athletes while the present study involved older participants more representative of the general active population.

With regard to SLL, an average individual should perform this task with a knee-valgus angle of between 0.6° and -9.0°. Other researchers have reported lower values than this for the same task (Kiriyama et al., 2009; Pappas, Hagins et al., 2007). However, other research has reported larger knee-valgus angles (-11.5° for females and -8.4° for males) for professional ballet dancers on landing (Orishimo et al., 2009). Participants in the current study produced lower vertical GRF (4.10* body weight) than the forces reported in McNair & Prapavesis's (1999) research (4.4 vs 4.5 for recreational and competitive athletes, respectively).

Gender differences in landing trials have been noticed in hip-adduction, knee-valgus and ankle-dorsiflexion angles ($p=0.001$, $p=0.08$ and $p=0.05$, respectively), and these differences are greater than SEM values reported previously (Table 3.5, Chapter 3). Females exhibited higher peak knee-flexion angle and less moment (72.6 vs 70.5 and 3.3 vs 3.5), contrary to a number of other studies' reports for SLL (Ali et al., 2013; Kiriyama et al., 2009; Orishimo et al., 2009) and double-legged landing (Salci, Kentel, Heycan, Akin, & Korkusuz, 2004; Yu et al., 2006) tasks. A possible explanation for this is the neuromuscular control differences between genders (Jacobs et al., 2007), and therefore more knee flexion in females indicates less dynamic stability of the knee compared to their male counterparts.

According to the RUN findings in the current study, knee-valgus angle was found to be in the range of -2.2° to -11.0° for females and -0.1° to 8.4° for males. Likewise, hip-adduction angle was found to be 21.6° and 17.7° for females and males, respectively. These values are similar to those reported for recreational athletes (Ferber et al., 2003; Ferber et al., 2002; Malinzak et al., 2001). It is possible, therefore, that a larger hip-adduction angle contributes to a larger knee-valgus angle by increasing the leverage between the GRF vector and the knee joint (Stefanyshyn et al., 2006). Again, sex differences were noticed in the running variables, males had smaller hip-adduction and knee valgus angles and moment compare to females; again these differences are higher than the SEM values reported in the reliability study in this thesis (Table 3.6, Chapter 3). This trend has been found in running investigations (Ferber et al., 2003; Malinzak et al., 2001).

Not surprisingly, cutting-task performances produced the largest knee-valgus angles and moments compared to all tasks. The reference value for knee-valgus angle was found to be around -8.3° (±6.1). These values are similar to those reported previously for a 90° cutting task (Jones, Herrington, & Graham-Smith, 2015; Jones et al., 2014; Kristianslund & Krosshaug, 2013). Furthermore, McLean et al. (2005) reported similar values for 45° cutting (11±4°) for National Collegiate Athletic Association (NCAA) Division 1 basket ballers. Peak knee-valgus moment during 90° cutting tasks in the current study was similar to those reported previously for the same cutting angle (Jones et al., 2014). However, these values were substantially greater than those reported for a 45° cutting task (Sigward & Powers, 2006a; McLean, et al., 2005b; Pollard et al., 2004). Bloomfield et al. (2007) observed premier-league football matches, they noticed that soccer players frequently change direction by between 90° and 180° in both directions. However, cutting at such angles has a very different momentum profile than a 45° degree sidestep cut (Havens and Sigward, 2014; Schot et al., 1995).

It is important to bear in mind the variations in speed ranges during cutting tasks reported in the literature (Figure 4.10). However, speeds in the current study were very similar to those reported in the relaiblity study in this thesis (Chapter 2). A number of authors have selected higher approach speeds of 5.0–7.0 m/sec. (Pollard et al., 2004; Sigward and Powers, 2007; Pollard et al., 2007). Higher approach speeds might result in more knee-valgus loading or lower task achievement. Despite the differences in cutting angles between studies, Sigward and Powers (2007) report similar knee-valgus moment to that in the current study (1.2±0.4 vs 1.2±0.6, Nmkg), this can be explained by the differences in approach speed between studies.

The good match between desirable and actual speed at touch-down in terms of both direction and angle shows that timing gates are suitable for checking entry speed, but they do not allow for the evaluation of actual task achievement. Pollard et al.'s study (2004) is the only that required approach and exit speeds of 5.5–6.5 and 4.5–5.5 m/sec., respectively. Also, with the limitations of comparing studies with different progression speeds in mind, a standardized speed may be preferable. Based on task achievement, lower speeds are to be preferred; however, the knee-joint loading at such speeds is too low if the purpose is to evaluate ACL injury risk, i.e. the risk of damaging the ACL when performing the task. Whilst wishing to induce sufficient loading, it is important to keep the safety of participants in mind at all times. Therefore, based on the trade-off between task achievement and loading, we propose a progression speed of 4 m/sec. to be most suitable for investigating lower-limb loading associated with a dynamic sidecutting manoeuvre.

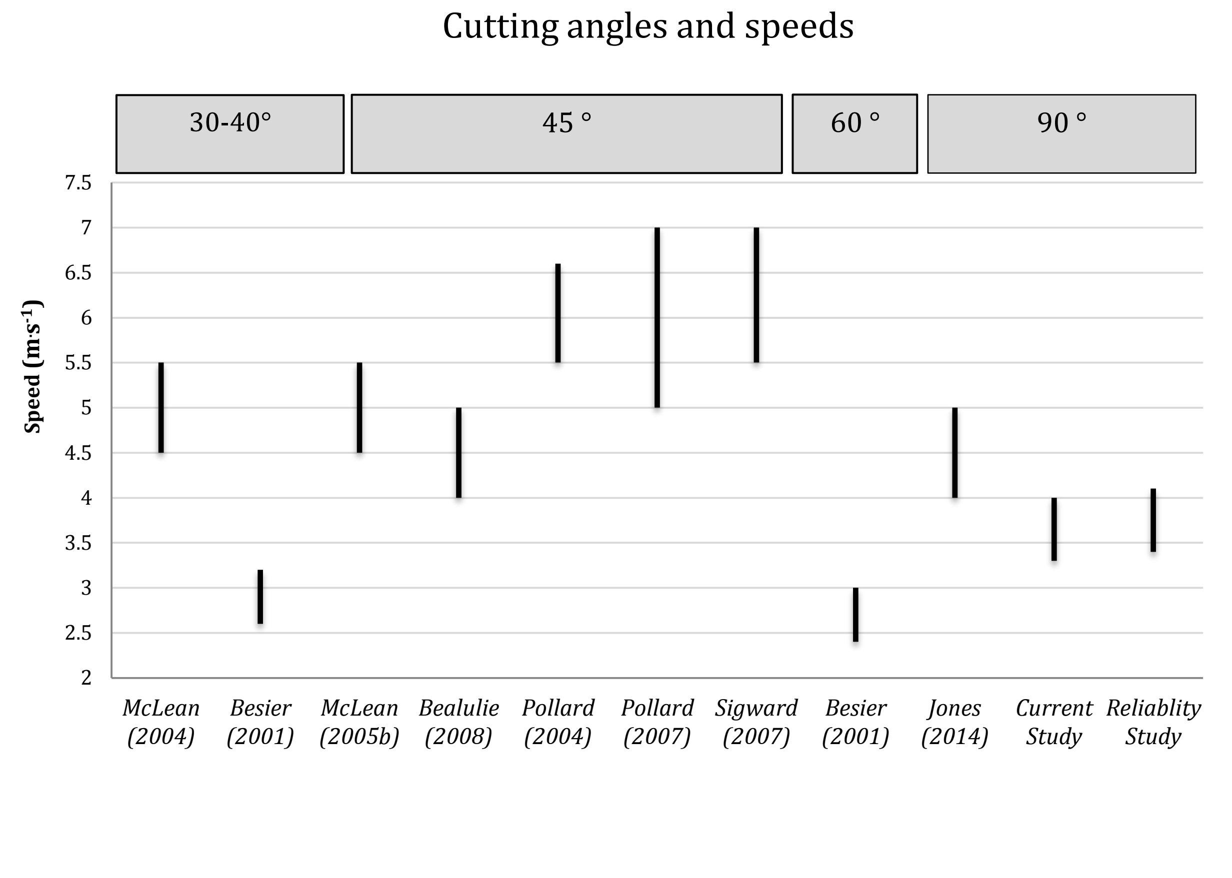


Figure 4.10. *Allowed variation in cutting speeds as reported in previous literature on 3D biomechanical variables during side-step cutting at different angles.*

As with any study, there are some limitations of the present study. The first limitation relates to the study population. The participants represented a healthy population without lower-extremity problems. Therefore, it is not possible to generalize the results obtained to very athletic, inactive or patient populations. Another concern is that the levels and kinds of physical activity of the participants were not taken into account. It is possible that some subjects may have had more experience of squatting, landing, running or cutting tasks. An issue not addressed in this study is whether leg dominancy might affect the results. However, Clark (2001) concludes in his review article that the impact of leg dominancy has yet to be clearly established in the literature. Kicking a ball can be considered to be an example of skill dominance versus stance dominance, with regard to the objective of the task (Clark, 2001). Future studies should address the impact of leg dominancy, and the way it is defined, on lower-limb angles and loading during screening tasks.

Other limitations related to task standardisation, such as squat depth, running and cutting velocities, and cutting trials being in a pre-planned rather than an unanticipated situation, which is known to elevate knee-joint loads (Besier, Lloyd, Ackland, et al., 2001). Lastly, the highly controlled lab environment is another limitation of this study and so the ecological validity of the findings should be considered. With ongoing technological evolutions, investigators should seek to transfer the findings from such standardized methods into more ecologically valid evaluations of loading and injury risk in actual sports environments and training sessions. Finding reference values for joint angles and loading during commonly assessed screening tasks in non-injured individuals may help to find ways of identifying at-risk individuals for non-contact knee injuries associated with misalignment, such ACL and PFPS. However, more research is required to discover the underlying causes of poor mechanics when performing squatting, landing, running or changing direction manoeuvres, this would help in devising more efficient injury-prevention protocols.

4.6. Conclusion

On the basis of the outcomes from the population tested, the study hypotheses are accepted and the following observations can be made:

1. The study has established reference values for lower-limb biomechanical variables for a healthy population when performing single-leg squat, single-leg landing, running and cutting tasks.
2. Measurements of knee frontal-plane motion and moment during simple tasks (single-leg squat and single-leg landing) were lower than those gathered for complex tasks (running and cutting).
3. Across all tasks, females had significantly greater hip-adduction and knee-valgus angles compared to males.
4. No significant differences were noticed for vertical GRF produced during SLS, RUN and CUT tasks.
5. Knee-valgus moments were significantly different between the genders during RUN and CUT tasks.

Chapter 5: Relationship between lower-limb biomechanical variables during common screening tasks.

5.1. Aim

The aim of this chapter is to examine the connection between 3D biomechanical variables SLS, SLL, run and side-step cutting tasks.

5.2. Background

Dynamic knee valgus is a mixture of actions involving frontal- and transverse-plane motion at lower limb joints, which contribute to lower limb malalignment during loading tasks (Munro, Herrington, & Comfort, 2012; Hewett et al., 2005). Moreover, increased dynamic knee valgus is associated with PFPS during running and single-leg squat tasks (Dierks et al., 2008; Willson & Davis, 2008a; Stefanyshyn et al., 2006) and with ACL injury during landing and cutting tasks (Krosshaug et al., 2007; Hewett et al., 2005; Olsen et al., 2004).

Several attempts have been made to correlate the biomechanical variables, within the same population, among functional screening tests, such as single-leg landing (SLL), 90° and 180° cutting tasks (Jones et al., 2014), SLL, stepping and drop jump (Harty et al., 2011), shuttle run, side jump and 45º cutting (McLean et al., 2005), 45° and 90° cutting (Imwalle et al., 2009), drop vertical jump (DVJ) and 35° cutting (Kristianslund & Krosshaug, 2013), stepping down and drop vertical jump (Earl et al., 2007) and bilateral and unilateral landing (Pappas et al., 2007). Whatman et al. (2011) investigated the link between jogging and those variables involved during five simple tasks (single and bilateral small-knee bending, lunge, hop and step down), and more recently, in 2013, the same team published new work on the correlation between double (small-knee bend and drop jump) and single (single-leg small-knee bend and treadmill jogging) movements. The majority of these attempts have focused on females players (Harty et al., 2011; Jones et al., 2014; Kristianslund & Krosshaug, 2013; Whatman et al., 2013; Whatman et al., 2011; Imwalle et al., 2009; McLean et al., 2005b; Besier et al., 2001). Other authors (Earl et al., 2007 and Pappas) have included both genders in their studies.

Whatman and colleagues' studies would have been more useful if the authors had clarified whether their participants were males or females (Whatman et al., 2013; Whatman et al., 2011).

Apart from Kristianslund & Krosshaug (2013), all of the aforementioned studies were conducted on relatively small sample sizes. Kristianslund & Krosshaug (2013) conducted a large-scale study (n=120) to examine the association between cutting and drop vertical jump tasks. They observed weak correlation for knee-valgus moment (p= 0.13), while the correlations were stronger for valgus angles (p= 0.71). A note of caution is due here, since these findings were only collected from elite female handball players, making the findings less generalisable to other populations. Another weakness with the aforementioned literature is that it fails to take the coefficient of determination (R^2) into account. Including R^2 is useful as it gives the proportion of variance of one variable that is predictable from the other one (Jones et al., 2014). The study by Jones et al. (2014) is the only comprehensive correlation analysis, as they included R^2 in it. They found that 40% of variance in knee-valgus angle during cutting is explained by the valgus angle during SLL. This value reduced with knee-valgus moment to 21%. However, these results were only based on data from female soccer players and it is unclear whether their findings are applicable to other populations.

In reviewing the literature, there are no studies that have investigated the inter-task correlation of kinematic and kinetic variables in a large sample of recreational athletes during distinctly different movement tasks related to common knee injuries. A better knowledge of the inter-task performance would offer insights linked to how consistent male and female individuals in motor patterns during specific sport tests. The aim of this study, however, was to investigate the association between lower-limb biomechanical variables during SLS, SLL, RUN, and CUT tasks.

5.2.1 Hypotheses

Based on the abovementioned literature, two hypotheses are formulated:

- H_1= Knee valgus angle during all study tasks will report higher correlation than knee valgus moments.
- H_2= Female participants will exhibit higher correlations compare to male participants.
- H_3= Knee valgus angles will exhibit higher correlation between SLS and SLL tasks compare to its correlation with other tasks.

5.3. Methods

5.3.1. Participants

Ninety recreational athletes, 55 males and 35 females (age 26.8 ± 4.7 years; height 170.5 ± 7.6 cm; and mass 69.6 ± 11.3 kg) took part. Participants were free from lower-limb injuries for last six months and to have no history of lower-limb surgery. A recreational athlete was defined as participating in physical activity for at least one hour, three times per week.

5.3.2. Procedure

A ten-camera motion analysis system (Pro-Reflex, Qualisys), sampled at 240 Hz, and three force platform fixed into the ground (AMTI, USA), sampled at 1,200 Hz, were used to gather biomechanical measures during the stance phase of SLS, SLL), RUN and CUT tasks. The same instrumentation, calibration, filtration, training shoes, marker list and biomechanical model were used as previously outlined in the reliability study in Chapter 3 (Sections 3.4.2–7).

5.3.3. Statistical Analysis

All statistical analyses were performed in SPSS (v. 21 (SPSS Inc.). Normality for each variable was checked with a Shapiro-Wilk test and histograms (see Appendix 3). Pearson's correlation coefficient (r) was used to explore the relationships between 3D variables and SLS, SLL, RUN and CUT tasks for parametric data. Relationships involving nonparametric variables were explored using Spearman's rank correlation (ρ). Furthermore, the coefficient of determination (R^2) was used in parametric data to represent the amount of variability in one screening test, which is explained by a second screening test (Swearingen et al., 2011). Table 5.1 illustrates the interpretation of the strength of correlation coefficients used in this study (Hopkins, Marshall, Batterham, & Hanin, 2009).

Table 5.1: Correlation coefficient scores and levels of association (Hopkins et al., 2009)

Correlation coefficient score	Level of association
(0.1–0.3)	Small
(0.3–0.5)	Moderate
(0.5–0.7)	Large
(0.7–0.9)	Very Large
(0.9–1.0)	Extremely large

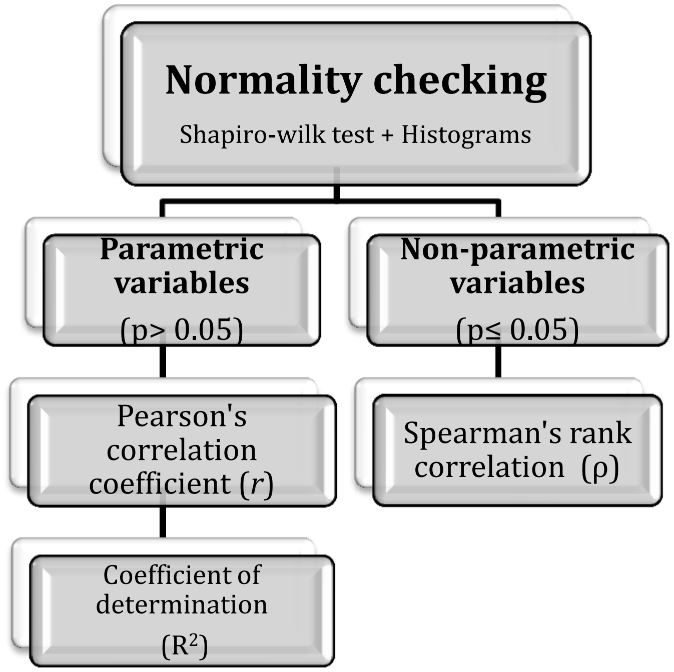


Figure 5-1: Statistical analysis outline for the correlation study

5.3.4. Results

Normality checking results for each variable are listed in Appendix C. Tables 5.2–5.7 illustrate the associations between biomechanical variables during SLS, SLL, Run and Cut tasks. Tables 5.4–5.5 contain correlation findings for female participants and Tables 5.6–5.7 for males. Furthermore, scatter plots for the following variables: knee-valgus angle and knee valgus moment, hip-internal rotation, hip adduction, can be seen in Appendix D.

Pearson and Spearman correlation coefficients revealed that hip-internal rotation angle had the strongest correlation during SLS with SLL and RUN (r =0.73; ρ =0.60, respectively). These correlations improved when applied to each gender separately, as seen in Tables 5.4–5.7. During cutting, hip-internal rotation angle exhibited small to moderate correlations between tasks (≤0.43). No correlations were noticed in hip-internal rotation moment except between SLL and RUN (ρ = 0.61).

Knee-valgus angle during the SLS task showed strong correlations with SLL, RUN and CUT (r =0.62; ρ =0.59; r =0.57, respectively). These relationships strengthened to a very large extent when applied to female individuals (r =0.75; ρ =0.51; r =0.65, respectively). Knee-valgus moment showed only weak correlation with SLS, SLL & RUN (0.15–0.25). No correlation was found between knee-valgus moment during cutting and SLS or SLL (ρ = 0.06-0.1), but there was a moderate correlation with RUN (r = 0.50).

Hip-adduction angle during the SLS task showed moderate correlation with SLL, RUN and CUT (r =0.42; ρ =0.48; r =0.40, respectively). These relationships were weaker for male participants (r =0.25; r=0.39; r =0.39 respectively). Hip-adduction moments recorded small to moderate correlations between tasks (0.21–0.41).

Table 5.2. Relationships between 3D biomechanical variables for SLS with SLL, RUN & CUT tasks

Variables	SLS vs. SLL	SLS vs. RUN	SLS vs. CUT
Joint angle (°)			
Hip flexion	r = 0.37** (R^2=0.14)	r =0.31** (R^2=0.09)	r = 0.01 (R^2=0.00)
Hip adduction	r = 0.42** (R^2=0.18)	r = 0.48** (R^2=0.23)	r = 0.40** (R^2=0.16)
Hip Int. Rot.	r = 0.73** (R^2=0.53)	r = 0.60** (R^2=0.36)	r = 0.36** (R^2=0.13)
Knee flexion	r = 0.29** (R^2=0.08)	r = 0.33** (R^2=0.11)	ρ = 0.20
Knee valgus	r = 0.62** (R^2=0.39)	ρ = 0.59**	r = 0.57** (R^2=0.32)
Knee Int. Rot.	r = 0.76** (R^2=0.58)	r = 0.63** (R^2=0.39)	*Unreliable variable*
Dorsiflexion	ρ = 0.43**	ρ = 0.45**	ρ = 0.00
Moments (Nm/kg)			
Hip adduction	ρ = 0.45**	ρ = 0.40**	ρ = 0.36**
Hip Int. Rot.	ρ = 0.26*	ρ = 0.26*	ρ = 0.13
Knee valgus	ρ= 0.23*	ρ= 0.25*	ρ = 0.16
Force (*body weight)			
Vertical GRF	r = 0.22* (R^2=0.05)	ρ = -0.07	ρ = -0.01

(ρ) Spearman & (r) Pearson correlation coefficients; (R^2) Coefficient of determination; (*) Statistically significant at p≤ .05; (**) statistically significant at p≤.01

Table 5.3 Relationships between 3D biomechanical variables for SLL, RUN & CUT tasks

Variables	**SLL vs. RUN**	**SLL vs. CUT**	**RUN vs. CUT**
Joint angle (°)			
Hip flexion	r = 0.33** (R^2=0.11)	r =0.27** (R^2=0.07)	r= 0.35** (R^2=0.12)
Hip adduction	r = 0.52** (R^2=0.27)	r = 0.22* (R^2=0.05)	r = 0.35** (R^2=0.12)
Hip Int. Rot	r = 0.67** (R^2=0.45)	r = 0.54** (R^2=0.29)	r = 0.53** (R^2=0.28)
Knee flexion	r = 0.18 (R^2=0.03)	ρ = 0.25*	ρ = 0.29**
Knee valgus	ρ = 0.58**	r = 0.64** (R^2=0.41)	ρ = 0.76**
Knee Int. Rot.	r = 0.64** (R^2=0.41)	ρ = 0.63**	*Unreliable variable*
Dorsiflexion	r = 0.29** (R^2=0.08)	r = 0.19 (R^2=0.04)	r = -0.22* (R^2=0.05)
Moment (Nm/kg)			
Hip adduction	ρ = 0.41**	r = 0.30** (R^2=0.10)	ρ = 0.21
Hip Int. Rot.	ρ = 0.61**	ρ = -0.02	ρ = 0.09
Knee valgus	ρ = 0.15	ρ = 0.13	r = 0.50** (R^2=0.16)
Force (*body weight)			
Vertical GRF	ρ = 0.14	ρ = 0.35**	ρ = 0.31**

(ρ) Spearman & (r) Pearson correlation coefficients; (R^2) Coefficient of determination; (*) Statistically significant at $p \leq .05$; (**) statistically significant at $p \leq .01$

Table 5.4. Relationships between 3D biomechanical variables for SLS with SLL, RUN & CUT tasks in female participants

Variables	SLS vs. SLL	SLS vs. RUN	SLS vs. CUT
Joint Angle (°)			
Hip flexion	r = 0.66** (R^2=0.44)	r =0.27 (R^2=0.07)	r = 0.22 (R^2=0.05)
Hip adduction	r = 0.46** (R^2=0.21)	r = 0.39* (R^2=0.15)	r = 0.31 (R^2=0.09)
Hip Int. Rot.	r = 0.80** (R^2=0.64)	r = 0.61** (R^2=0.36)	r = 0.21 (R^2=0.04)
Knee flexion	r = 0.39* (R^2=0.15)	r = 0.10 (R^2=0.01)	r = 0.39* (R^2=0.15)
Knee valgus	r = 0.75** (R^2=0.56)	ρ = 0.51**	r = 0.65** (R^2=0.42)
Knee Int. Rot.	ρ = 0.80**	ρ = 0.62**	*Un-reliable variable*
Dorsiflexion	r = 0.53** (R^2=0.28)	r = 0.42* (R^2=0.18)	r = -0.23 (R^2=0.05)
Moment (Nm/kg)			
Hip adduction	ρ = 0.73**	r = 0.43** (R^2=0.19)	r = 0.34* (R^2=0.12)
Hip Int. Rot.	ρ = 0.55**	r = 0.48** (R^2=0.22)	ρ = 0.53**
Knee valgus	ρ= 0.32	ρ= 0.16	r = 0.25 (R^2=0.06)
Force (*body weight)			
Vertical GRF	ρ = 0.03	r = 0.07 (R^2=0.01)	r = 0.11 (R^2=0.01)

(ρ) Spearman & (r) Pearson correlation coefficients; (R^2) Coefficient of determination; (*) Statistically significant at $p\leq .05$; (**) statistically significant at $p\leq.01$

Table 5.5. Relationships between 3D biomechanical variables for SLL, RUN & CUT tasks in female participants

Variables	SLL vs. RUN	SLL vs. CUT	RUN vs. CUT
Joint angle (°)			
Hip flexion	r = 0.41* (R^2=0.17)	r = 0.40* (R^2=0.16)	r = 0.57** (R^2=0.33)
Hip adduction	r = 0.48** (R^2=0.23)	r = 0.24 (R^2=0.06)	r = 0.50** (R^2=0.25)
Hip Int. Rot.	r = 0.66** (R^2=0.44)	r = 0.34* (R^2=0.12)	r = 0.43** (R^2=0.19)
Knee flexion	r = 0.07 (R^2=0.01)	r = 0.42* (R^2=0.18)	r = 0.40* (R^2=0.16)
Knee valgus	ρ = 0.55**	r = 0.74** (R^2=0.55)	ρ = 0.79**
Knee Int. Rot.	ρ = 0.60**	ρ = 0.77**	*Un-reliable variable*
Dorsiflexion	r = 0.19 (R^2=0.04)	r = -0.18 (R^2=0.03)	r = 0.11 (R^2=0.01)
Moments (Nm/kg)			
Hip adduction	ρ = 0.43**	ρ = 0.39*	r = 0.43** (R^2=0.19)
Hip Int. Rot.	ρ = 0.70**	ρ = 0.18**	ρ = 0.49**
Knee valgus	ρ = 0.32	ρ = 0.49**	ρ = 0.49**
Force (*body weight)			
Vertical GRF	ρ = 0.09	ρ = 0.69**	r = 0.38* (R^2=0.14)

(ρ) Spearman & (r) Pearson correlation coefficients; (R^2) Coefficient of determination;
(*) Statistically significant at p≤ .05; (**) statistically significant at p≤.01

Table 5.6. Relationships between 3D biomechanical variables for SLS with SLL, RUN & CUT tasks in male participants

Variables	SLS vs. SLL	SLS vs. RUN	SLS vs. CUT
Joint angle (°)			
Hip flexion	r = 0.22 (R^2=0.05)	r = 0.33* (R^2=0.11)	r = -0.11 (R^2=0.01)
Hip adduction	r = 0.25 (R^2=0.06)	r = 0.39** (R^2=0.15)	r = 0.39** (R^2=0.15)
Hip Int. Rot.	r = 0.71** (R^2=0.50)	r = 0.61** (R^2=0.37)	r = 0.43** (R^2=0.18)
Knee flexion	r = 0.27 (R^2=0.07)	r = 0.44** (R^2=0.20)	ρ = 0.13
Knee valgus	r = 0.53** (R^2=0.29)	r = 0.63** (R^2=0.40)	r = 0.49** (R^2=0.42)
Knee Int. Rot.	ρ = 0.59**	r = 0.58** (R^2=0.34)	*Unreliable variable*
Dorsiflexion	ρ = 0.37**	ρ = 0.45**	ρ = 0.13
Moment (Nm/kg)			
Hip adduction	ρ = 0.35**	ρ = 0.41**	ρ = 0.36**
Hip Int. Rot.	ρ = 0.06	ρ = 0.13	ρ = 0.12
Knee valgus	r = 0.13 (R^2=0.02)	r = 0.41** (R^2=0.17)	ρ = 0.25
Force (*body weight)			
Vertical GRF	r = 0.19 (R^2=0.04)	ρ = -0.12	r = -0.09 (R^2=0.01)

(ρ) Spearman & (r) Pearson correlation coefficients; (R^2) Coefficient of determination; (*) Statistically significant at $p \leq .05$; (**) statistically significant at $p \leq .01$

Table 5.7. Relationships between 3D biomechanical variables for SLL, RUN & CUT tasks in male participants

Variables	SLL vs. RUN	SLL vs. CUT	RUN vs. CUT
<u>Joint angle (°)</u>			
Hip flexion	r = 0.38** (R^2=0.14)	r =0.21 (R^2=0.04)	r= 0.16 (R^2=0.03)
Hip adduction	r = 0.40** (R^2=0.16)	r = 0.10 (R^2=0.10)	r = 0.16 (R^2=0.03)
Hip Int. Rot.	r = 0.72** (R^2=0.52)	r = 0.63** (R^2=0.39)	r = 0.58** (R^2=0.34)
Knee flexion	r = 0.39** (R^2=0.15)	ρ = 0.21	ρ = 0.15
Knee valgus	r = 0.59** (R^2=0.35)	r = 0.53** (R^2=0.28)	r = 0.72** (R^2=0.52)
Knee Int. Rot.	ρ = 0.62**	ρ = 0.59**	*Un-reliable variable*
Dorsiflexion	r = 0.38** (R^2=0.14)	r = 0.42** (R^2=0.17)	r = 0.30* (R^2=0.00)
<u>Moment (Nm/kg)</u>			
Hip adduction	r = 0.29* (R^2=0.10)	r = 0.34* (R^2=0.11)	r = 0.17 (R^2=0.03)
Hip Int. Rot.	ρ = 0.57**	ρ = -0.18	ρ = -0.13
Knee valgus	r = 0.06** (R^2=0.00)	ρ= -0.02	ρ = 0.39**
<u>Force (*body weight)</u>			
Vertical GRF	ρ = 0.29*	r = 0.15 (R^2=0.02)	ρ = 0.32*

(ρ) Spearman & (r) Pearson correlation coefficients; (R^2) Coefficient of determination; (*) Statistically significant at p≤ .05; (**) statistically significant at p≤.01

5.4. Discussion

The goal of this chapter was to examine the relationship between 3D biomechanical measures during SLS, SLL, RUN and CUT tasks. There are further similarities in several joint angles and a few moments across tasks. The findings clearly demonstrate that there are significant relationships between SLS, SLL, RUN and CUT tasks for a number of variables. Other researchers have reported similar correlations between different screening tasks, such as landing, pivoting and turning (Jones et al., 2014) and step-down, single-leg landing and drop vertical jump (Harty et al., 2011), and between jogging and squatting (Whatman et al., 2011), side-jump, shuttle and 55° side cutting (Mclean et al., 2005b).

Several significant correlations were noticed between SLS variables and those that occur during SLL, RUN and CUT tasks. Specifically, the associations between SLS and SLL in the current study were high for knee valgus, hip and knee internal-rotation motions (0.62, 0.73, and 0.76, respectively) coupled with high R^2 percentages (39%, 53%, and 58%, respectively). A lack of previous literature correlating SLS with SLL makes comparisons virtually impossible. Furthermore, significant correlations were seen between SLS and RUN and knee valgus (p=0.59) and hip-internal rotation (r= 0.60, R^2=36%) and knee-internal rotation angles (r=0.63, R^2=39%). These findings are comparable to those reported by Whatman et al. (2011) for similar tasks. They noticed moderate correlations with knee-valgus angle (r= 0.66) and high correlation with hip-internal rotation angle (r=0.87) during a small-knee bending (SKB) task compared to jogging at low speed (2.9± 0.4 $m{\cdot}s^{-1}$). However, the observed increase in their correlation results compared to the current study can be attributed to the differences in methods between both studies and running speeds (2.9± 0.4 vs 4.9± 0.5 m/sec.). Furthermore, they observed strong correlations between SKB and stepping down for hip-internal rotation and knee-valgus angles (r= 0.84, and 0.76, respectively), and fair correlation for hip adduction (r= 0.65). Perhaps the most serious drawback of their study is that the authors offer no explanation about normality checking for their data. Furthermore, their investigation would have been more convincing if they had included R^2 in their correlation testing.

Across all tasks, knee-valgus moment showed small to moderate correlations (*r* and *p* values ranged between 0.15 and 0.50). This could be due to different technical parameters in each task which will effect knee-abduction moments at the knee, such as foot-progression angle which is the angle of foot orientation during initial contact relative to the original travel direction (Jones et al., 2014; Andrews et al., 1996). Sigward and Powers (2007) observed a significant correlation between foot-progression angle and higher knee-valgus moment (r=0.39, p-value 0.001). In the current study, females demonstrated moderate correlation for knee-valgus moment between SLL, RUN and CUT tasks (0.32-0.49). Previous researchers reported small to moderate correlation in female athletes for knee-valgus moment between sidestep cutting and landing (Jones et al., 2014) and between drop vertical jump and sidestep cutting (Kristianslund & Krosshaug, 2013).

The R^2 values (Tables 5.1–5.6) represent the amount of variability of one variable in one screening test, which is explained by the same variable in another task. The highest R^2 values were found in hip and knee internal-rotation angles between SLS and SLL tasks, suggesting that 53% of the variability in hip-internal rotations and 58% in knee-internal rotation angles during SLS can be explained by knowing the same variables for SLL tasks. These findings are improved when the correlation is measured for female participants only (n=35), so that up to 64% of the variability in this variable for SLS can be explained by SLL tasks. Direct comparison with R^2 values in the literature is virtually impossible since no studies have been conducted to measure the correlation between SLS and SLL tasks.

The generalisability of the current study is subject to certain limitations. For instance, the cutting task in the current study could only be performed with the right leg because of limited laboratory space. Thus, comparisons were only made for the right leg between tasks. Other limitations relate to task standardization. These include the squat depth, running and cutting velocities. A further limitation is the uninjured individuals that we investigated. It is uncertain whether these correlations would be effected the level of activity, therefore these results may not be appropriate to elite athletes, adolescents or older participants. Furthermore, it should be stressed that lowering the reliability level for the transverse-plane variables reported in the reliability study of this thesis (Chapter 3) affects the level of certainty for correlation findings, so while they provide useful information these results need to be interpreted with caution. In terms of directions for future research, further work is warranted to find the underlying causes of the weak correlations for kinetic variables in these tasks.

Notwithstanding these limitations, the current findings propose that those individuals who exhibit misalignment strategies, specifically in the frontal and transverse planes, during SLS and SLL may also show these during running and cutting tasks. However, it should be stressed that the use of squats or landings alone should not be considered sufficient to identify individuals at risk from running or cutting mechanics since several variables showed weak or no correlation. Admittedly, just because several SLS variables correlate with other tasks does not mean that SLS causes this or that these are the only two factors involved in the relationship. Therefore, the results of current study do not imply causation since only experimental studies can establish cause and effect.

5.5. Conclusion

In this healthy population, several joint angles during single-leg squats significantly correlated with those collected from SLL, RUN and CUT tasks. This could potentially reduce the time and the tasks required for screening, as just one easy task could give an idea of which individuals may exhibit poor movement strategies related to a number of other complex tasks. However, it should be stressed that the use of squats or landings alone should not be considered sufficient to identify individuals at risk from running or cutting mechanics since several variable showed weak or no correlation. What this chapter does not confirm is if manipulation of the performance of simple tasks such as SLS and SLL has an impact on the performance of more dynamic tests such as running and changing direction tasks.

Chapter 6: Use of augmented feedback to modify movement patterns during common screening tasks.

6.1. Aims

The aims of this study are:

1. Investigate the effect of verbal and visual feedback protocols on single-leg squat performance (SLS).
2. Investigate if a significant change in SLS performance through a feedback protocol is reflected in a change in the performance of single-leg landing (SLL), running (RUN) and sidestep cutting (CUT) tasks.

6.2. Background

Dynamic knee valgus is a combination of frontal and transverse plane motions at the hip, knee and ankle, which contribute to lower limb malalignment during athletic tasks (Hewett et al., 2005; Munro & Herrington, 2014). Increased dynamic knee valgus is associated with PFPS injury during SLS and running tasks (Crossley et al., 2011; Willson and Davis, 2009) and with ACL injury during landing and cutting tasks (Hewett et al., 2005; Olsen et al., 2004).

As discussed earlier in Chapter Two, previous researchers have shown that feedback training can reduce knee valgus angle and moment (Ford et al., 2015; Mizner et al., 2008; Munro & Herrington, 2014), increase knee flexion motion (Herman et al., 2009; Onate et al., 2005), increase hip-flexion and abduction angles (Herman et al., 2009) and reduce hip-adduction and internal-rotation angles (Willy, Scholz, et al., 2012). There seems to be general agreement on reducing vertical peak ground-reaction forces after feedback training (Cronin et al., 2008; Herman et al., 2009; Onate et al., 2005; Prapavessis & McNair, 1999).

Onate et al. (2005) assessed the effects of different modes of feedback on a jump-landing task. Fifty-one participants were assigned, randomly, into four groups (self, expert model, both self and expert, and control). All feedback modes resulted in higher knee flexion and lower vertical GRF, and these improvements were retained for a week. Self and combination feedback has been shown to result in decreased vertical GRF and knee-valgus moments, increased hip-abduction and flexion angles, and increased knee-flexion angles. These changes were significantly more than the effects of expert and control groups. In line with this, Mizner et al. (2008) reported that verbal feedback alone has been shown to decrease vertical GRF, knee-valgus angle and moment and to increase knee-flexion angles, although it is unclear whether these last for several weeks or even days.

Rucci and Tomporowski (2010) reported that video and verbal feedback had no larger effect than verbal feedback alone during power, clean performance. Moreover, verbal feedback alone made greater effect in performance than video only, which suggests that verbal feedback is a key factor for performance improvement. However, the video and verbal feedback protocol in Rucci and Tomporowski's (2010) was only video of participants' performances, which has previously been shown to be less effective than a combination of self and expert models (Onate et al., 2005). It may be that the most essential feature of the verbal and video feedback protocol, which would result in the greatest improvement in performance, is expert modelling combined with verbal instructions (Munro & Herrington, 2014). Most of the investigations up to this point have not studied whether the effect of augmented feedback for simple tasks such as SLS would translate into an improvement in performance in more complex tasks, such as running and cutting.

Recently, Munro & Herrington (2014) used the same feedback protocol as Onate (2005) with its landing-error scoring system (LESS) to determine whether this would reduce FPPA during drop jump (DJ) and single-leg squat (SLL) tasks in 28 recreationally athletes (eight of them used as a control group). A significant reduction in vertical GRF (2.73 vs. 2.55 * BW) and FPPA (4.0° vs. -19.9°) were noticed post feedback in intervention groups. No changes were evident in the control group. These findings would have been more interesting if they addressed whether the instant improvements in performance were sustained for a longer time, and so more research is required in this extent.

Few of the aforementioned investigations have dealt with individuals displaying poor motion, i.e. excessive angles, moments or high force rates. Another question that needs to be asked is the retention effect of feedback. Only a few researchers have looked at the effect of feedback for longer periods of time (Willy et al., 2012; Onate et al., 2005; Prapavess et al., 2003). Onate et al. (2005) retested their participants a week after first testing. The reduction after their feedback protocol was retained for a week. Willy et al. (2012) found that improvements in running, SLS and stepping mechanics were maintained in the absence of feedback after one month, and three months as well. Conversely, Prapavess et al. (2003) did not notice any effect of feedback (instructions) after three months compared to baseline testing.

To date, the literature has tended to focus on the effects of feedback training on specific tasks (Table 2.5, Section 2.7.1). Willy and Davis (2011) found that hip-strengthening and motor-learning interventions improved strength and SLS performance but these benefits were not transferred to improved running performance. However, it is not clear whether the changes were due to an increase in strength or motor learning. In another attempt, Willy et al. (2012) found that mirror and verbal feedback while using a treadmill resulted in improvements to running performance (reduction in hip-adduction angle and moment), and these improvements were transferred to SLS and step down. It is not known whether these changes would occur if their feedback training were based on a simple task, such as SLS training, instead of treadmill running. Further research into the sustainability and transferability of these changes is required to support using the feedback training as a tool for reducing risk of knee injuries prior to designing prevention programmes.

When assessing SLS mechanics, distal and proximal variables should be taken into account as they can influence loading through the lower limbs (Herrington and Munro, 2014; Myer et al., 2008). A qualitative analysis of single-leg squats (QASLS) takes these variables into account. Specifically, it involves movement strategies occurring in the feet, knee, pelvis, trunk and arms (Herrington and Munro, 2014). High scores on QASLS, which indicates poor SLS performance, are linked to 3D motion that may increase the injury risk (Herrington and Munro, 2014). Therefore, using QASLS as a basis for feedback is likely to improve lower biomechanics when doing an SLS task.

The goal of this study, therefore, is to investigate the effects of visual and verbal feedback protocols on SLS biomechanics and if changing the squat task is reflected in SLL, RUN and CUT tasks.

6.2.1 Hypotheses

Based on the abovementioned literature, two hypotheses are formulated:

- H_1= Using visual and verbal feedback based on QASLS will improve SLS performance.
- H_2= This mode of feedback will result in greater improvement in SLS compared to SLL, RUN and CUT tasks.

6.3. Methods

6.3.1. Participants

Eleven recreationally active female volunteers (age 27.2 ±4.2 years, height 165.5 ±4.4 cm, weight 61.0 ±6.2) were recruited for the study. Before data collection, an a priori power analysis was conducted using SLS data from the reliability study (Chapter 3). Using knee-valgus angle differences between sessions (effect size= 0.91), it was found that 11 participants were required to adequately power this study (power = 0.80; α = 0.05).

A recreationally active individual is defined as someone who has taken part in half an hour of physical activity, three times weekly for the past six months. Participants were excluded if they had a current lower-limb injury, a history of same-side lower-limb surgery or did not meet the activity requirements. Lower-limb injury is defined as any injury that has prevented someone completing their normal exercise routine in the six months prior to testing. All subjects gave written consent to participate (Appendix A-4) and completed a medical screening questionnaire (Appendix A-5). Ethical approval was gained from the University of Salford Research and Ethics Committee prior to initiation of the study (Appendix A-3).

Table 6.1: Demographic measurements for all participants

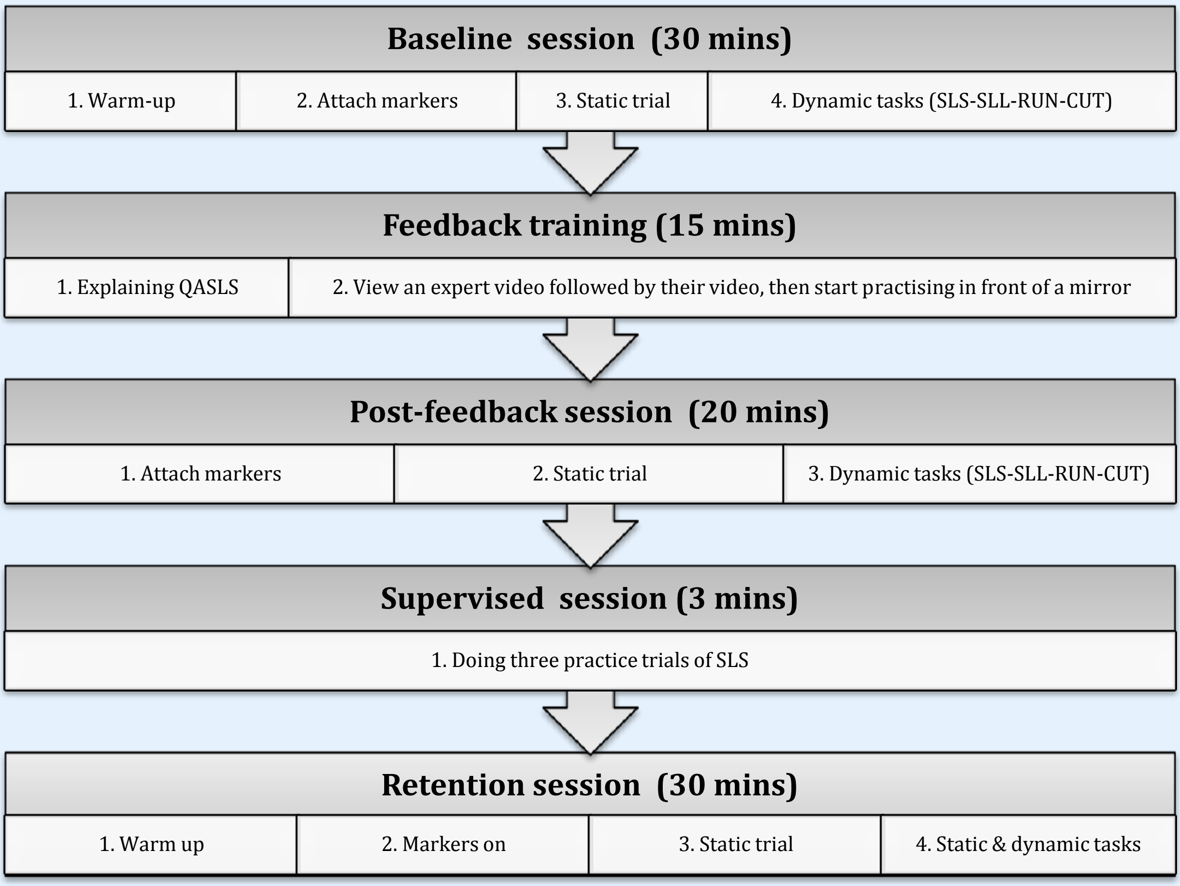
Participants	Age (years)	Height (cm)	Mass (kg)
1	25	166.0	70
2	30	165.0	67.0
3	30	165.0	55.0
4	24	175.5	60.0
5	20	171.0	74.0
6	23	164.0	63.0
7	25	163.0	58.3
8	29	162.0	58.0
9	31	161.0	54.6
10	26	158.0	53.0
11	29	160.0	66.1
Mean	**27.2**	**165.5**	**61.0**
SD	**4.2**	**4.4**	**6.2**

6.3.2. Study Procedure

Each participant attended the human performance laboratory on three occasions (baseline, post feedback and retention), all of which are described below.

6.3.3. Baseline screening tasks

Participants undertook a baseline session, during which they performed three trials of the single-leg squat (SLS), single-leg landing (SLL), running (RUN) and cutting (CUT) tasks, as previously described in Chapter 3 (Sections 3.4–6).



Notes;

- *Feedback training session starts 5 minutes after the baseline session.*
- *A 5-minute gap was allowed before starting the post-feedback session.*
- *The warm-up protocol was 3 for minutes at low intensity on a cycle ergometer.*
- *The retention session was 7 days after the 1st session (same time of day).*
- *Supervised session was 3 days after the baseline session.*

Figure 6.1: Flowchart for the three sessions

6.3.4. Visual and verbal feedback protocol

The feedback protocol was based on the 'model plus self' combination used by Munro et al. (2013) and Onate et al. (2005). Participants were asked to perform the SLS task in front of a mirror to allow them to view and self-correct their own technique based on a qualitative analysis of the single-leg squat (QASLS) tool, which was devised by Herrington and Munro (2014).

The QASLS is a scoring tool that assesses movement strategies occurring in individual body regions (arms, trunk, pelvis, thighs, knees, feet). Optimal behaviour involves minimal deviation or body movement from that prescribed, so the arms do not move, the trunk is slightly flexed, but held still, the pelvis stays in a mid position with minimal tilt, the thighs stay parallel and in an approximately vertically orientation, the patellae point towards the middle of the feet and the feet demonstrate minimal wobble. Therefore, the use of QASLS as a basis for feedback is likely to improve the performance of a single-leg squatting task.

Table 6.2. Qualitative analysis of single-leg loading (QASLS)

No.	Strategy	Yes	No
1	Excessive arm movement to maintain balance		
2	Leaning in any direction		
3	Loss of the horizontal plane		
4	Excessive tilt or rotation		
5	Weight-bearing thigh moves into hip adduction		
6	Non-weight-bearing thigh not held in a neutral stance		
7	Patella pointing towards second toe (noticeable valgus)		
8	Patella pointing past the inside of the foot (significant valgus)		
9	Touches down with non-weight-bearing foot		
10	Stance leg wobbles noticeably		

The subjects first watched 2 trials of the model video, then by their own trials. In each case the sagittal plane video was viewed first. Each trial was viewed two times, first at normal speed and secondly in slower movement, controlled by the main investigator. To help review the technique on display in each trial, participants were asked to complete a checklist (Table 6.2). The checklist focused on performance considerations that would lead to optimal performance. The main investigator described the criteria and revised the video with the participants to make sure they understand the techniques (Appendix E-1). The feedback sessions lasted 15 minutes on average. Following a feedback session, participants relaxed for 5 minutes; then the post feedback session started with marker placement and capturing static trials, followed by screening tasks with the sequence of SLS, SLL, RUN and lastly CUT.

Subjects were asked to return after three days to do a supervised exercise session. During this session, the participants were monitored via a checklist again (Table 6.1) in front of the principal investigator. The follow-up session (retention) was after one week at the same time of day, and the participants were analysed with a 3D screening of the four tasks (SLS, SLL, RUN & CUT). During this session, the same baseline procedure was repeated. No additional feedback was given whilst participants were performing the screening tests.

6.4. Data analysis

All statistical analysis was done with SPSS (v. 21, SPSS Inc.). Means and standard deviations for all study variables were calculated. Normality for each variable was checked using a Shapiro-Wilk test. For parametric variables, paired t-tests were carried out to determine whether changes in the dependent variables occurred from baseline to post feedback/ one-week follow-up sessions for each task. Wilcoxon rank tests were used for comparisons involving nonparametric variables.

The alpha level was set at $p = 0.05$ and corrected p value was set at $p=0.007$ to minimise the likelihood of a type-1 error occurring. This p-value was determined through seven t-tests for each task (knee-valgus angle, knee-valgus moment, knee-internal rotation, knee flexion, hip adduction, hip-internal rotation and hip flexion). For significant comparisons, effect sizes were calculated using the Cohen δ method (Thomas, Nelson, & Silverman, 2005), which defines 0.2, 0.5 and 0.8 as small, medium and large, respectively.

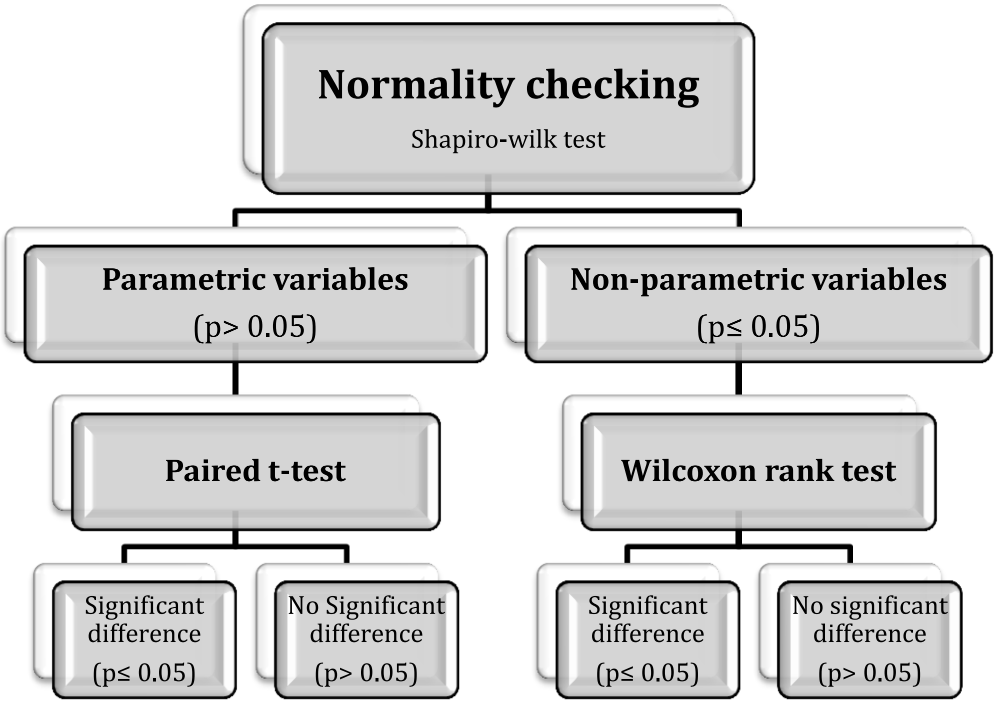


Figure 6.2: Statistical analysis outline for the feedback study

6.5. Results

Normality checking revealed that the majority of variables were normally distributed with the exception of knee-internal rotation across all tasks, knee-valgus moment during SLS and CUT tasks and hip adduction during SLS and SLL. Further details about normality checking can be seen in Appendices E3–6.

It is apparent from Table 6.3 that there were changes in the SLS variables from baseline to post feedback and the follow-up sessions. Immediately after feedback, there were significant differences in knee-valgus angle and moment and hip-flexion angle (effect sizes 0.91, 0.98 and 0.96, respectively). These changes sustained during the retention session were compared to the baseline (effect sizes 1.12, 1.30 and 1.61, respectively). Hip-adduction angle decreased after feedback, and even with follow-up, but these changes were not significant.

It can be seen from the data in Table 6.4 that knee-valgus moment during SLL decreased significantly ($p=0.00$, $ES=0.74$) post feedback. These changes disappeared during the retention session. Feedback decreased knee-valgus angle to 2.5° post feedback and to 1.8° during follow-up screening. A reduction in hip-flexion angle during the follow-up session was noticed but this reduction was above the accepted statistical level. However, no significant change was noticed in running variables post feedback or in the follow-up session, these comparisons can be seen in Table 6.5. The only significant change in cutting was found in hip-flexion angle immediately after feedback training ($p=0.05$), but this disappeared in the retention session ($p=0.45$). Table 6.6 presents Baseline, post-feedback and follow-up results for cutting task.

Table 6.3 Baseline, post-feedback and follow-up results for the SLS task

Task	Baseline	Post FB	MD (1+2)	Sig	ES	PWR	Follow-up	MD (1+3)	Sig	ES	PWR
Joint angle (°)											
Hip flexion	79.7 ±11	67.9 ±12	11.8	0.00	0.96	0.95	61.5 ±11	18.1	0.00	1.61	0.99
Hip ADD	16.2 ±7.5	13.9 ±5.6	2.3	0.06	0.34	0.30	12.0 ±6.0	4.12	0.06	0.78	0.95
Hip Int. Rot.	3.4 ±6.1	3.7 ±1.9	0.29	0.44	0.07	0.10	3.0 ±7.1	0.43	0.42	0.09	0.95
Knee flex	92.5 ±8.4	86.7 ±8.8	5.8	0.01	0.67	0.70	86.1 ±6.3	6.4	0.01	0.86	0.90
Knee valgus	-3.0 ±2.7	-0.6 ±2.6	2.4	0.01	0.91	0.87	0.4 ±3.3	3.4	0.00	1.12	0.95
Knee Int.Rt	-3.3 ±6.6	-3.4 ±6.7	0.04	0.48	0.01	0.10	-4.8 ±4.7	1.42	0.25	0.24	0.23
Moment (NmKg)											
Knee valgus	0.08 ±.14	-0.04 ±.10	0.04	0.00	0.98	0.91	-0.11 ±.15	0.19	0.00	1.30	0.99

Post FB = Post Feedback session; MD^{1+2} = Mean difference between baseline & post feedback sessions; MD^{1+3} = Mean difference between baseline and follow-up sessions; ES = Effect size; Sig = Significance level (p=0.05); ADD= Adduction; Int. Rt. = Internal rotation; PWR = Power.

Table 6.4 Baseline, post-feedback and follow-up results for the SLL task

Task	Baseline	Post FB	MD (1+2)	Sig	ES	PWR	Follow-up	MD (1+3)	Sig	ES	PWR
Joint angle (°)											
Hip flexion	63.1 ±9.3	61.8 ±6.9	1.36	0.24	0.16	0.13	60.3 ±8.4	2.88	0.02	0.32	0.30
Hip ADD	9.0 ±5.3	9.6 ±5.8	0.56	0.26	0.10	0.09	10.3 ±2.9	1.27	0.18	0.29	0.23
Hip Int. Rot.	3.4 ±5.0	4.1 ±6.9	0.74	0.32	0.12	0.10	3.7 ±5.3	0.33	0.44	0.06	0.08
Knee flex	74.6 ±9.1	79.3 ±9.4	4.70	0.10	0.50	0.49	76.8 ±9.7	2.20	0.41	0.23	0.19
Knee valgus	-5.5 ±4.1	-3.0 ±4.4	2.50	0.02	0.60	0.60	-3.7 ±3.9	1.80	0.06	0.44	0.50
Knee Int.Rt	-2.6 ±7.9	-3.1 ±8.5	0.48	0.29	0.05	0.09	-2.9 ±3.7	0.32	0.45	0.05	0.09
Moment (NmKg)											
Knee valgus	0.46 ±0.1	0.36 ±.16	0.10	0.00	0.74	0.77	0.45 ±0.3	0.01	0.49	0.04	0.09

Post FB = Post-feedback session; MD^{1+2} = Mean difference between baseline & post-feedback sessions; MD^{1+3} = Mean difference between baseline and follow-up sessions; ES = Effect size; Sig = Significance level (p=0.05); ADD = Adduction; Int. Rt. = Internal rotation; PWR=Power.

Table 6.5 Baseline, post-feedback and follow-up results for the RUN task

Task	Baseline	Post FB	MD (1+2)	Sig	ES	PWR	Follow-up	MD (1+3)	Sig	ES	PWR
Joint angle (°)											
Hip flexion	61.6 ±11	61.9 ±12	0.30	0.36	0.02	0.1	59.2 ±8.7	2.37	0.14	0.23	0.20
Hip ADD	15.1 ±4.4	14.3 ±4.9	0.98	0.06	0.17	0.15	15.4 ±4.0	0.44	0.37	0.07	0.10
Hip Int. Rot.	4.51 ±5.3	3.4 ±5.7	1.02	0.26	0.18	0.15	2.5 ±4.7	2.01	0.12	0.40	0.35
Knee flex	55.3 ±6.7	54.9 ±7.9	0.40	0.64	0.05	0.09	53.8 ±5.3	1.50	0.06	0.24	0.20
Knee valgus	-6.3 ±4.7	-6.5 ±5.3	0.16	0.38	0.03	0.08	-6.5 ±4.3	0.21	0.42	0.04	0.06
Knee Int.Rt	0.9 ±9.2	0.9 ±8.4	0.01	0.48	0.00	0.5	2.4 ±5.1	1.43	0.31	0.19	0.16
Moment (NmKg)											
Knee valgus	0.22 ±.08	0.23 ±0.1	0.01	0.28	0.08	0.08	0.22 ±0.1	0.00	0.20	0	0.05

Post FB = Post-feedback session; MD^{1+2}= Mean difference between baseline & post feedback sessions; MD^{1+3} = Mean difference between baseline and follow-up sessions; ES = Effect size; Sig = Significance level (p=0.05); ADD= Adduction; Int. Rt. = Internal rotation; PWR = Power.

Table 6.6 Baseline, post-feedback and follow-up results for the CUT task

Task	Baseline	Post FB	MD (1+2)	Sig	ES	PWR	Follow-up	MD (1+3)	Sig	ES	PWR
Joint angle (°)											
Hip flexion	53.5 ±9	51.4 ±9.3	2.12	0.05	0.22	0.20	53.8 ±8.0	0.28	0.41	0.03	0.06
Hip ADD	-6.3 ± 6.8	-5.5 ±7	0.74	0.54	0.10	0.11	-5.0 ±3.5	1.25	0.11	0.23	0.20
Hip Int. Rot.	8.4 ± 7.6	8.4 ±6.9	0.07	0.97	0.01	0.05	7.3 ± 6.1	1.10	0.56	0.15	0.13
Knee flex	62.7 ±7.9	62.2 ±8.5	0.50	0.72	0.06	0.07	62.3 ±9	0.40	0.85	0.04	0.06
Knee valgus	-10.8 ±4.9	-10.1 ±6.2	0.70	0.26	0.12	0.11	-10.8 ±5.4	0.00	0.49	0.00	0.05
Knee Int.Rt	*Knee internal rotation angle during cutting task = unreliable variable (Chapter 3)*										
Moment (NmKg)											
Knee valgus	0.79 ±0.2	0.72 ±0.4	0.07	0.27	0.20	0.35	0.82 ±0.2	0.03	0.36	0.13	0.11

Post FB = Post-feedback session; MD^{1+2} = Mean difference between baseline & post-feedback sessions; MD^{1+3} = Mean difference between baseline and follow-up sessions; ES = Effect size; Sig = Significance level (p=0.05); ADD= Adduction; Int. Rt. = Internal rotation; PWR = Power.

6.6: Discussion

The aims of this study were to:

a). Investigate the effect of the verbal and visual feedback protocol on single-leg squat performance (SLS).

b). Investigate if a significant change in SLS performance through a feedback protocol would be reflected in a change in performance in single-leg landing (SLL), running (RUN) and sidestep cutting (CUT) tasks.

An increase in knee valgus positions during screening tests has been associated to ACL and PFPS injuries (Myer et al., 2010; Hewett et al., 2005), and consequently a decrease in these movements has the possibility to decrease the injuries risk. As discussed earlier in the literature review, in Chapter Two, earlier investigations have reported that feedback training can improve frontal-plane kinematics and kinetics during drop-jump and landing tasks (Ford et al., 2015; Munro and Herrington, 2014; Herman et al., 2009; Mizner et al., 2008; Walsh et al., 2007), increase sagittal-plane motion during double-leg landing and jump landing (Herman et al., 2009; Onate et al., 2005) and reduce hip-adduction and internal-rotation angles during running (Willy, Scholz, et al., 2012). It was not known prior to this research whether the effects of augmented feedback on simple tasks such as SLS would improve SLS biomechanics and whether these changes would translate to improvements in performance in more dynamic tests, such as running and cutting.

The most obvious finding to emerge from the current study is that the use of a combination of expert and self-model video instruction and verbal cues based on the optimal performance of SLS significantly reduces knee-valgus angle and moment during the same task. Specifically, knee-valgus angle reduced from -3.0° to -0.6° from baseline to post feedback. The reduction of 2.4° is greater than the SEM value (1.7°) but less than the SDD values (4.8°) for knee valgus of SLS in the female population as reported in the reliability study in this thesis. The SEM and SDD values for female participants can be seen in Appendices E6 and E7. During the retention session, knee valgus reduced by up to 0.4°, and again this fell within the SDD value, although outside SEM, when this variable was measured after a week’s gap.

Further reductions were noticed in knee-valgus moment after the feedback and during the follow-up sessions (0.12 and 0.19 Nm/Kg, respectively). The follow-up changes are equal to the SDD values. Furthermore, a reduction of 11.8° in hip-flexion angle was noticed post feedback and 18.1° during the follow-up session. Therefore, we are 95% confident that the changes in knee-valgus moment and hip-flexion angle were not caused by measurement errors. Hip-adduction angle decreased after feedback, and even with follow-up, but these changes were not significant. No other significant changes were noticed in the SLS task.

The second aim of the current study was to investigate if a significant change in SLS performance through a feedback protocol would be reflected in changes in performance in single-leg landing, running and sidestep cutting tasks. The findings of the current study reveal that changing SLS tasks is reflected in SLL task performance by reducing knee-valgus angle and moment. Feedback decreased knee-valgus angle to 2.5° post feedback and to 1.8° during follow-up screening, these changes fell outside the SEM range for this task but within the SDD level of error. The changes in knee-valgus moment were significant during the immediate session ($p=0.004$) but this reduction disappeared during the retention session ($p=0.49$). Using a feedback mode, Munro and Herrington (2014) reported greater reduction in frontal-plane motion in female athletes during a single-leg landing task. It should be noted that the authors targeted female athletes with higher angles compared to the normal range for the same task and a normal population (Herrington and Munro, 2010). The value of knee-frontal angle during SLL in their study was substantially greater than in the current study for the same task (8.7° vs 5.5°). It is difficult to compare the current study with Munro and Herrington (2014) due to the differences in methods employed (3D peak-knee valgus angle vs 2D peak-frontal projection angle). However, the reduction in frontal-knee motion in both of these studies fell outside the SEM range for this task but within the SDD level of error.

In contrast, no significant changes were noticed in any of the running or cutting variables post feedback training or in the follow-up session; these comparisons can be seen in Tables 6.4–6.5. A potential reason for the lack of a transfer effect of SLS to other tasks might be the specificity of SLS training for more complex tasks, such cutting and running. There is abundant room for further progress in determining the transferability of feedback between tasks with more training using tasks that are similar in nature, such as landing, running and cutting, coupled with protocols such as the Landing Error Score System (LESS).

In the current study, the visual and verbal feedback model was based on the self and expert combination used by previous investigations (Munro and Herrington, 2014; Onate et al., 2005). Using this model, individuals can compare their own performance against an expert's performance, and this model has been shown previously to reduce knee-valgus moment, increase hip-abduction and flexion angles and increase knee-flexion angle during a landing task (Munro and Herrington, 2014; Herman et al., 2009; Onate et al., 2005). Mizner et al. (2008) found that verbal feedback alone during drop vertical jumping can reduce knee-valgus angle and moment, as well as increase knee-flexion motion, although it unclear whether these can be achieved consistently.

Rucci and Tomporowski (2010) noticed that verbal and visual feedback produced the same effect as verbal feedback alone on power clean performance. Moreover, verbal feedback alone produced greater alterations to performance than video only, supporting a verbal mode of feedback being a key component leading to changes in performance. However, their visual and verbal feedback protocol only involved video of the subject's performance. Onate et al. (2005) found that a self and expert model was more effective than viewing the participant's performance only. It may be that the most important aspect of a verbal and video feedback protocol, which results in the greatest improvement in performance, is expert performance as well as verbal instruction.

In the current study, the reductions knee valgus angle and moment during the SLS and SLL tasks following visual and verbal feedback indicate that participants were able to adjust their lower-extremity frontal -positioning and torque as a result of this simple training. This method of training is useful for enhancing the awareness and visual understanding of important kinematic and kinetic factors that may be related to injury. A particularly salient finding is that the feedback provided during a squat task can be transferred to SLL, but not for more dynamic mechanics such as RUN and CUT tasks. A possible reason for the lack of a transferability effect of SLS to other tasks might be the specificity of SLS training for more complex tasks such cutting and running. It should in be borne in mind that the retention period was not for a full week; it was four days following the supervised session, which was three days after the original test. Although it lasted for three minutes for each participant, it can still be counted as a training session and therefore the retention for this training programme was in reality for four days. The supervision was done to ensure the participants' compliance with the training programme. The findings of the current study must be interpreted with caution since the post hoc analysis for SLL, RUN and CUT tasks was underpowered (Tables 6.3–6.5). On the other hand, most variables during SLS had good power and this is due to the a priori analysis for this study that was conducted based on one of the SLS variables.

The generalisability of the findings of the current study is subject to some limitations. For instance, all subjects were recreationally athletes. It is unclear whether these findings were influenced by age or activity levels, therefore these results may not be applicable to elite athletes, adolescents or older age groups. Since no males were recruited, these findings cannot be generalized across genders. As the previous chapter showed, males behave differently to females across certain tasks. Another concern is that the level and kind of physical activity of participants were not taken into account. It is possible that some individuals may have responded differently to the feedback protocol. Lastly, only single group was examined and was not compared to a control. This should be undertaken in order to confirm that alterations reported in biomechanical variables are not solely due to time effects and are the result of an SLS feedback-training programme.

6.8: Conclusion

On the basis of the study outcomes obtained from the population tested, the following conclusions can be drawn:

1. The use of a combination of expert and self-model video instruction and verbal cues based on optimal performance in SLS significantly reduced knee-valgus angle and moment during the same task. These changes were sustained for a week after the baseline.
2. Changing the SLS tasks is reflected in the SLL task by reducing the knee-valgus angle and moment.
3. Changing the SLS tasks is not reflected in the performance on RUN and CUT tasks.

Chapter 7: Summary, conclusions and suggestions for future work

7.1. Summary

It is important to develop screening tests to identify athletes who may be predisposed to knee injuries and then to design individual intervention programmes. Several studies have been conducted to address lower-limb biomechanics during various screening movements, which mimic the real situation of patella-femoral pain syndrome (PFPS) injuries, such as single-leg squat and running, or anterior cruciate ligament (ACL) injuries, such as single-leg landing and cutting tasks. The available literature fails to define reference values for lower-limb biomechanical variables, as the numbers of subjects participating in all of the reviewed studies were limited, making the generalization of findings difficult. Furthermore, there is limited literature that explores the relationship between lower-limb biomechanical variables during athletic tasks and associated knee-joint injuries. A better understanding of inter-task performance would offer insights into the consistency of motor patterns employed by healthy individuals during screening movement tasks.

The majority of studies investigating lower-limb biomechanics and its relation to knee injuries have undertaken their investigations by analysing three-dimensional (3D) motion-analysis systems. Using 3D motion analysis systems allows researchers to calculate motion in all directions during dynamic tasks and is postulated as the "gold standard" of motion analysis. It is important to ensure that any assessment tool used in research or clinical assessment is valid and reliable if used on the same day or even after a period of several days. Understanding of the reliability and measurement errors associated with each of these screening tasks is essential. Without these values, changes in performance cannot be accurately assessed, as it is unknown whether the differences are due to measurement errors or true changes in performance. By knowing SEM values, researchers can accurately determine whether changes or improvements are more than the measurement error of a test, while SDD scores allow for a determination of whether any observed changes in a specific variable over time are due to a true change in performance. Therefore, reporting measurement errors will help to interpret findings by not overestimating small changes or ignoring meaningful improvements because of high variability. Also, it will confirm that results are from the intervention itself, as demonstrated by its larger effect compared to measurement error.

The modification of high-risk movement strategies is an important element to prevent ACL and PFJ injuries. Earlier investigators have reported that feedback training can decrease some misalignments in lower-limb biomechanical variables. Most of the investigations up to this point have not dealt with individuals displaying poor motion, i.e. excessive angles, moments or high force rates. Another question that needs to be asked, however, is whether the effect of augmented feedback of a specific task will transfer to tasks.

<u>The aims of this thesis were to:</u>

1. Investigate the reliability of using 3D movement-analysis techniques to measure lower-limb kinematic and kinetic variables during single-leg squat, single-leg landing, running and cutting tasks.
2. Establish reference values for lower-limb kinematic and kinetic variables during a series of lower-limb loading tasks in a physically active population.
3. Investigate the relationship between lower-limb kinematic and kinetic variables across a series of lower-limb loading tasks in a physically active population.
4. Investigate the effect of an augmented feedback protocol on single-leg squat performance and if changing squat performance would be reflected in a change in performance in single-leg landing, running and sidestep cutting tasks.

7.2. Conclusion

With regard to the first aim, this was to establish the within-day and between-days reliability of using 3D motion analysis to measure biomechanical variables collected from single-leg squat, single-leg landing, running and cutting tasks. This study found that the majority of between-day ICC values for joint angles, moments and vertical GRF were lower than within-day values across all tasks. Transverse-plane angles (hip and knee-internal rotation) demonstrated high levels of variability when compared to other planes, specifically between-days measurements of knee-internal rotation angle during cutting tasks (ICC=0.42; SEM=4.09°; SDD=11.3°). Although the ICC value was fair, unfortunately, the lower band of the 95% confidence interval crossed the zero level (-0.1–0.76). This finding was rather disappointing and, therefore, knee-internal rotation during cutting was not carried forward to other studies in this thesis. Cluster movement might explain the decline in cutting-rotation motion, since the cutting trials were done after completing the SLS, SLL and RUN tasks. Another possible explanation for this decline may be the more dynamic nature of the cutting task compared to the other tasks in this study. Future research should focus on this issue and how to improve the reliability of knee-rotation data collected during cutting tasks.

Vertical GRF data were highly reliable during all tasks, with ICCs ranging between 0.84 and 0.98. These results may be explained by GRF values being representative of the sum of all segmental masses, accelerations and gravitational forces. Thus, no markers are needed to gather GRF data and so these will not suffer from marker placement error and can therefore be assumed to be more repeatable.

This study provides SEM and SDD reference values for SLS, SLL, RUN and CUT tasks that may be useful for evaluating intervention outcomes. The highest SEM and SDD values were found with hip-flexion angles in between-day sessions across all tasks (SEM = 3.7º–5.42°; SDD = 10.4º–15.0°), but these represent between 7.5 and 10.1 per cent for SEM and between 20.9 and 28.0 per cent for SDD when comparing their means (SLS= 69.1°; SLL=49.7°; RUN=55.4°; CUT=49.1°). This may be rationalised by the greater range of motion in the sagittal plane compared to other planes.

Across all tasks, knee-valgus angle exhibited good to excellent within-day reliability (ICC= 0.87-0.94), while between-days sessions demonstrated good reliability during both SLS and CUT tasks (ICC=0.79–0.84) and fair reliability during SLL and RUN tasks (ICC=0.61–0.61). Therefore, according to the current findings, if the knee-valgus angle was measured during SLS before and after an intervention, with a week's gap, we could be confident that the true score lies within 1.8° of the observed score in both sessions. Furthermore, a change of at least 5.0° would be needed to say that the intervention had a significant effect above measurement error with 95% confidence.

In order to achieve the second aim of this thesis, 90 healthy participants were recruited in order to establish reference values for lower-limb biomechanical variables for a healthy population when performing SLS, SLL, RUN and CUT tasks. This study found that knee frontal-plane motion and moment during simple tasks (single-leg squat and single-leg landing) were lower than those gathered from complex tasks (running and cutting). Across all tasks, females showed significantly more hip-adduction and knee valgus angles compared to males. No significant differences were noticed for the vertical GRF produced SLS, RUN and CUT tasks. Knee-valgus moments were significantly different between the genders during RUN and CUT tasks.

Regarding the third aim of this thesis, a correlational study was conducted to investigate the associations between biomechanical variables during SLS, SLL, RUN and CUT tasks. The findings clearly demonstrate that there were significant relationships in peak knee-abduction, hip-adduction and hip internal-rotation angles across substantially different functional tasks. Knee-valgus moment showed small to moderate correlations across tasks, whereas females sustained moderate correlation in knee-valgus moment between SLL, RUN and CUT tasks (0.32–0.49). This could be because the different technical parameters of each task affect knee-abduction moments at the knee, such as foot-progression angles during cutting. The lack of significant correlations between hip and knee frontal-plane moments collected from the cutting task with those collected from the SLS, SLL and RUN tasks may be due to the nature of performing the cutting task, as individuals often place their foot laterally toward the new direction of movement to generate medial GRF to facilitate the direction change. The thigh involved is placed in an abducted position to begin with, whereas during squatting, landing and even during running, it is directly under the body.

Finally, the last study aimed to investigate the effect of a verbal and visual feedback protocol on the performance of an SLS task and if a significant change in SLS performance through the feedback protocol would be reflected in a change in performance in SLL, RUN and CUT tasks. Training resulted in a significant reduction in knee-valgus angle and moment and hip-flexion angles during single-leg squats. Additionally, these improvements remained several days later, suggesting motor patterns might have changed and improvements would endure, thus future prospective cohort studies are needed to determine if injury risk can be reduced in the long term. Furthermore, significant reductions in knee-valgus angle and moment were also noticed on landing after squat feedback training, but no significant improvements were transferred to run and cut tasks.

7.3. Suggestions for future work

Based on the results of this thesis, several questions are raised for forthcoming research. Primarily, the reliability study revealed that the CAST model should be used to measure kinematic and kinetic variables during SLS, SLL, running and cutting tasks in future investigations. Next, attempts to include different athletic populations, involving a range of different sporting activities and injured individuals, would be helpful in order to discover whether average biomechanical variables differ between sports. This would help to detect those athletes who are considered as representing excessive joint angles or moments, which put athletes at greater risk of knee injuries.

Considering the reference values for screening tasks in the current thesis and those reported in previous literature (Chapter 2, Tables 2.2–2.5), upcoming research is needed on large populations during different screening tasks, or different modes of a specific task. Future studies are required to find out whether simple 2D screening tasks can approximate to 3D estimates of lower-limb angles and loading to allow the wide use of such tasks by clinicians. Finding reference values for joint angles and loading during commonly assessed screening tasks in non-injured individuals may provide some way of identifying at-risk individuals for non-contact knee injuries associated with misalignment, such ACL and PFPS. Also, more research is required to discover the underlying causes of poor mechanics when performing squatting, landing, running and changing-direction manoeuvres, as this would help to devise more efficient injury-prevention protocols.

Further work on feedback training is warranted. Whether a week gap results in knee-valgus, moment and hip-flexion angles as noted in current study being retained over a longer period remains to be seen. Furthermore, future research should move in the direction of using similar feedback protocols for other simple tasks. The error-measurement statistics presented in the reliability study, in Chapter 3, will also allow investigators to determine precisely whether alterations in biomechanical variables are due to intervention or measurement errors.

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Appendix (A)

Appendix (A-1)

University of Salford MANCHESTER

Research, Innovation and Academic Engagement Ethical Approval Panel

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9 August 2012

Dear Faisal,

<u>RE: ETHICS APPLICATION HSCR12/46</u> – Within day and between days reliability of using two types of movement analysis systems during a series of typical athletic tasks

Following your responses to the Panel's queries, based on the information you provided, I am pleased to inform you that application HSCR12/46 has now been approved.

If there are any changes to the project and/ or its methodology, please inform the Panel as soon as possible.

Yours sincerely,

Rachel Shuttleworth

Rachel Shuttleworth
College Support Officer (R&I)

Appendix (A-2)

Research, Innovation and Academic Engagement Ethical Approval Panel

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18 October 2012

Dear Faisal,

<u>RE: ETHICS APPLICATION HSCR12/64</u> – The relationship between kinetic & kinematic variables during a number of typical athletic tasks

Following your responses to the Panel's queries, based on the information you provided, I am pleased to inform you that application HSCR12/64 has now been approved.

If there are any changes to the project and/ or its methodology, please inform the Panel as soon as possible.

Yours sincerely,

Rachel Shuttleworth

Rachel Shuttleworth
College Support Officer (R&I)

Appendix (A-3)

Research, Innovation and Academic Engagement Ethical Approval Panel

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20 March 2015

Dear Faisal,

RE: ETHICS APPLICATION HSCR14/116 – The use of augmented feedback intervention to modify movement patterns during common screening tasks

Based on the information you provided, I am pleased to inform you that application HSCR14/116 has been approved.

If there are any changes to the project and/ or its methodology, please inform the Panel as soon as possible.

Yours sincerely,

Sarah Starkey

Sarah Starkey
Engagement & Innovation Assistant

Appendix (A-4)

You are invited to take part in a study to find out

The Relationship Between Lower Limb Biomechanical Variables During Common Screening Tasks

- *As a participant in this study, you would be asked to undertake tests which included assessment of Single leg Squat, Single Leg Landing, Running and Changing Direction Manoeuvres.*

- *Your participation would involve only one session for approximately an hour at Human Performance Laboratory, Mary Sea-Cole Building.*

- *This study has been reviewed by, and received ethics clearance through, the Office of Research Ethics, University of Salford.*

- *If you decide that you would like to take part in the study and for further information, please contact the researcher:*

Mr Faisal Alenezi, PhD Student

PO43 Brain Blatchford Building, University of Salford, M6 6PU.

E-mail (F.S.Alenezi@edu.salford.ac.uk).

Appendix (A-5)

Informed Consent Form

1. *Faisal Alenezi,* who is a Postgraduate research student at the University of Salford, has requested my participation in a research study. My involvement in the study and its purpose has been fully explained to me.

2. My participation in this research will involve a number of tests, which include Single Leg Squat, Single Leg Landing, Running and changing direction manoeuvres.

3. I understand the requirements of the study and my involvement and the possible benefit of my participation in this research.

4. I have been informed that I will not be compensated for my participation.

5. I understand that the results of this research may be published but that my name or identity will not be revealed at any time. In order to keep my records confidential, *Faisal Alenezi* will store all information as numbered codes in computer files that will only be available to him.

6. I have been informed that any questions I have at any time concerning the research or my participation will be answered by Mr. Faisal Alenezi and I can contact him at :
(F.S.Alenezi@edu.salford.ac.uk).

7. I understand that I may withdraw my consent and participation at any time without objection from the researcher, then all information about me will be destroyed and not to be used in the study.

Name:. …………………………………… **Signed:**. **………………**………… **Date:** ……………………………..

Appendix (A-6)

Health Questionnaire

Tick which type of exercise activity the subject will be participating in:

Maximal exercise ☐. Submaximal exercise ☐. Other ☐ ....

1. **Personal information**

Surname: ....

Forename(s) ....

Date of birth: ....

Age ....

Height (cm): ....

Weight (kg) ....

2. **Additional information**

a. Please state when you last had something to eat / drink ....

b. Tick the box that relates to your present level of activity:

Inactive ☐ moderately active ☐ highly active ☐

c. Give an example of a typical weeks exercise:

....

d. If you smoke, approximately how many cigarettes do you smoke a day ....

3.	Are you currently taking any medication that might affect your ability to participate in the test as outlined?	YES	NO
4.	Do you suffer, or have you ever suffered from, cardiovascular disorders? e.g. Chest pain, heart trouble, cholesterol etc.	YES	NO
5.	Do you suffer, or have you ever suffered from, high/low blood pressure?	YES	NO
6.	Has your doctor said that you have a condition and that you should only do physical activity recommended by a doctor?	YES	NO
7.	Have you had a cold or feverish illness in the last 2 weeks?	YES	NO
8.	Do you ever lose balance because of dizziness, or do you ever lose consciousness?	YES	NO
9.	Do you suffer, or have you ever suffered from, respiratory disorders?	YES	NO

	e.g. Asthma		
10	Are you currently receiving advice from a medical advisor i.e. GP or Physiotherapist **not** to participate in physical activity because of back pain or any musculoskeletal problems?	YES	NO
11	Do you suffer, or have you ever suffered from diabetes?	YES	NO
12	Do you suffer, or have you ever suffered from epilepsy/seizures?	YES	NO
13	Do you know of any reason, not mentioned above, why you should not exercise? e.g. Head injury, pregnant, hangover, eye injury or anything else.	YES	NO
14	Do you have any allergies, especially in relation to reflective markers?	YES	NO

Appendix (B)

Appendix B-1

Tests of Normality for SLS task			
	Shapiro-Wilk		
	Statistic	df	Sig.
SLS_1st session_knee _valgus angle	.978	15	.957
SLS_2nd session_knee_valgus_angle	.926	15	.238
SLS_3rd session_knee_valgus_angle	.922	15	.208
SLS_1st session_knee_flexion_angle	.971	15	.875
SLS_2nd session_knee_flexion_angle	.980	15	.970
SLS_3rd session_knee_flexion_angle	.951	15	.538
SLS_1st session_knee_valgus_moment	.958	15	.661
SLS_2nd session_knee_valgus_moment	.948	15	.496
SLS_3rd session_knee_valgus_moment	.961	15	.715
SLS_1st session_hip_adduction_angle	.940	15	.377
SLS_2nd session_hip_adduction_angle	.981	15	.976
SLS_3rd session_hip_adduction_angle	.939	15	.367
SLS_1st session_hip_flexion_angle	.894	15	.077
SLS_2nd session_hip_flexion_angle	.934	15	.317
SLS_3rd session_hip_flexion_angle	.903	15	.107
SLS_1st session_dorsiflexion_angle	.957	15	.646
SLS_2nd session_dorsiflexion_angle	.987	15	.997
SLS_3rd session_dorsiflexion_angle	.899	15	.090
SLS_1st session_vertical GRF	.949	15	.517
SLS_2nd session_vertcial GRF	.951	15	.547
SLS_3rd session_vertical GRF	.947	15	.478
SLS_1st session_hip_internal_rotation_angle	.979	15	.965
SLS_2nd session_hip_internal_rotation_angle	.940	15	.376
SLS_3rd session_hip_internal_rotation_angle	.974	15	.913
SLS_1st session_knee_flexion_moment	.920	15	.195
SLS_2nd session_knee_flexion_moment	.973	15	.902
SLS_3rd session_knee_flexion_moment	.943	15	.426
SLS_1st session_knee_internal_rotation_angle	.971	15	.873
SLS_2nd session_knee_internal_rotation_angle	.949	15	.508
SLS_3rd session_knee_internal_rotation_angle	.987	15	.997
SLS_1st session_hip_adduction_moment	.910	15	.137
SLS_2nd session_hip_adduction_moment	.867	15	.031
SLS_3rd session_hip_adduction_moment	.933	15	.307
SLS_1st session_dorsiflexion_moment	.988	15	.998
SLS_2nd session_dorsiflexion_moment	.962	15	.729
SLS_3rd session_dorsiflexion_moment	.968	15	.820
SLS_1st session_hip_flexion_moment	.946	15	.466
SLS_2nd session_hip_flexion_moment	.986	15	.995
SLS_3rd session_hip_flexion_moment	.929	15	.264

Appendix B-2

Tests of Normality for SLL task			
	Shapiro-Wilk		
	Statistic	df	Sig.
SLL_1^{st} session_knee _valgus angle	.916	15	.168
SLL_2^{nd} session_knee_valgus_angle	.913	15	.150
SLL_3^{rd} session_knee_valgus_angle	.943	15	.418
SLL_1^{st} session_knee_flexion_angle	.983	15	.984
SLL_2^{nd} session_knee_flexion_angle	.960	15	.701
SLL_3^{rd} session_knee_flexion_angle	.944	15	.429
SLL_1^{st} session_knee_valgus_moment	.698	15	.000
SLL_2^{nd} session_knee_valgus_moment	.916	15	.169
SLL_3^{rd} session_knee_valgus_moment	.941	15	.401
SLL_1^{st} session_hip_adduction_angle	.958	15	.650
SLL_2^{nd} session_hip_adduction_angle	.969	15	.843
SLL_3^{rd} session_hip_adduction_angle	.953	15	.567
SLL_1^{st} session_hip_flexion_angle	.986	15	.995
SLL_2^{nd} session_hip_flexion_angle	.942	15	.403
SLL_3^{rd} session_hip_flexion_angle	.962	15	.728
SLL_1^{st} session_dorsiflexion_angle	.944	15	.431
SLL_2^{nd} session_dorsiflexion_angle	.934	15	.314
SLL_3^{rd} session_dorsiflexion_angle	.917	15	.174
SLL_1^{st} session_vertical GRF	.944	15	.430
SLL_2^{nd} session_vertcial GRF	.977	15	.946
SLL_3^{rd} session_vertical GRF	.935	15	.327
SLL_1^{st} session_hip_internal_rotation_angle	.944	15	.436
SLL_2^{nd} session_hip_internal_rotation_angle	.925	15	.231
SLL_3^{rd} session_hip_internal_rotation_angle	.973	15	.898
SLL_1^{st} session_knee_flexion_moment	.897	15	.087
SLL_2^{nd} session_knee_flexion_moment	.962	15	.724
SLL_3^{rd} session_knee_flexion_moment	.964	15	.763
SLL_1^{st} session_knee_internal_rotation_angle	.954	15	.585
SLL_2^{nd} session_knee_internal_rotation_angle	.958	15	.658
SLL_3^{rd} session_knee_internal_rotation_angle	.971	15	.866
SLL_1^{st} session_hip_adduction_moment	.902	15	.101
SLL_2^{nd} session_hip_adduction_moment	.821	15	.007
SLL_3^{rd} session_hip_adduction_moment	.957	15	.648
SLL_1^{st} session_dorsiflexion_moment	.812	15	.005
SLL_2^{nd} session_dorsiflexion_moment	.587	15	.000
SLL_3^{rd} session_dorsiflexion_moment	.522	15	.000
SLL_1^{st} session_hip_flexion_moment	.920	15	.191
SLL_2^{nd} session_hip_flexion_moment	.916	15	.167
SLL_3^{rd} session_hip_flexion_moment	.807	15	.005

Appendix B-3

Tests of Normality for RUN task			
	Shapiro-Wilk		
	Statistic	df	Sig.
RUN_1st session_knee _valgus angle	.956	15	.628
RUN _2nd session_knee_valgus_angle	.959	15	.682
RUN _3rd session_knee_valgus_angle	.927	15	.244
RUN _1st session_knee_flexion_angle	.973	15	.904
RUN _2nd session_knee_flexion_angle	.933	15	.307
RUN _3rd session_knee_flexion_angle	.950	15	.523
RUN _1st session_knee_valgus_moment	.904	15	.108
RUN _2nd session_knee_valgus_moment	.909	15	.128
RUN _3rd session_knee_valgus_moment	.892	15	.073
RUN _1st session_hip_adduction_angle	.928	15	.258
RUN _2nd session_hip_adduction_angle	.970	15	.862
RUN _3rd session_hip_adduction_angle	.961	15	.712
RUN _1st session_hip_flexion_angle	.929	15	.260
RUN _2nd session_hip_flexion_angle	.917	15	.172
RUN _3rd session_hip_flexion_angle	.933	15	.301
RUN _1st session_dorsiflexion_angle	.946	15	.457
RUN _2nd session_dorsiflexion_angle	.968	15	.829
RUN _3rd session_dorsiflexion_angle	.869	15	.033
RUN _1st session_vertical GRF	.877	15	.043
RUN _2nd session_vertcial GRF	.770	15	.002
RUN _3rd session_vertical GRF	.851	15	.018
RUN _1st session_hip_internal_rotation_angle	.982	15	.983
RUN _2nd session_hip_internal_rotation_angle	.917	15	.172
RUN _3rd session_hip_internal_rotation_angle	.943	15	.422
RUN _1st session_knee_flexion_moment	.955	15	.613
RUN _2nd session_knee_flexion_moment	.962	15	.724
RUN _3rd session_knee_flexion_moment	.982	15	.982
RUN _1st session_knee_internal_rotation_angle	.957	15	.639
RUN _2nd session_knee_internal_rotation_angle	.909	15	.132
RUN _3rd session_knee_internal_rotation_angle	.935	15	.327
RUN _1st session_hip_adduction_moment	.948	15	.500
RUN _2nd session_hip_adduction_moment	.920	15	.195
RUN _3rd session_hip_adduction_moment	.876	15	.042
RUN _1st session_dorsiflexion_moment	.982	15	.983
RUN _2nd session_dorsiflexion_moment	.934	15	.318
RUN _3rd session_dorsiflexion_moment	.967	15	.809
RUN _1st session_hip_flexion_moment	.876	15	.042
RUN _2nd session_hip_flexion_moment	.953	15	.577
RUN _3rd session_hip_flexion_moment	.950	15	.518

Appendix B-4

Tests of Normality for CUT task			
	Shapiro-Wilk		
	Statistic	df	Sig.
CUT_1st session_knee _valgus_angle	.893	15	.075
CUT _2nd session_knee_valgus_angle	.959	15	.666
CUT _3rd session_knee_valgus_angle	.973	15	.898
CUT _1st session_knee_flexion_angle	.902	15	.103
CUT _2nd session_knee_flexion_angle	.920	15	.191
CUT _3rd session_knee_flexion_angle	.935	15	.327
CUT _1st session_knee_valgus_moment	.894	15	.078
CUT _2nd session_knee_valgus_moment	.920	15	.194
CUT _3rd session_knee_valgus_moment	.902	15	.102
CUT _1st session_hip_adduction_angle	.950	15	.519
CUT _2nd session_hip_adduction_angle	.950	15	.524
CUT _3rd session_hip_adduction_angle	.964	15	.762
CUT _1st session_hip_flexion_angle	.964	15	.754
CUT _2nd session_hip_flexion_angle	.954	15	.585
CUT _3rd session_hip_flexion_angle	.831	15	.009
CUT _1st session_dorsiflexion_angle	.951	15	.533
CUT _2nd session_dorsiflexion_angle	.978	15	.952
CUT _3rd session_dorsiflexion_angle	.964	15	.764
CUT _1st session_vertical GRF	.883	15	.052
CUT _2nd session_vertcial GRF	.951	15	.544
CUT _3rd session_vertical GRF	.921	15	.198
CUT _1st session_hip_internal_rotation_angle	.959	15	.673
CUT _2nd session_hip_internal_rotation_angle	.966	15	.802
CUT _3rd session_hip_internal_rotation_angle	.941	15	.401
CUT _1st session_knee_flexion_moment	.970	15	.855
CUT _2nd session_knee_flexion_moment	.958	15	.658
CUT _3rd session_knee_flexion_moment	.958	15	.652
CUT _1st session_knee_internal_rotation_angle	.969	15	.842
CUT _2nd session_knee_internal_rotation_angle	.913	15	.148
CUT _3rd session_knee_internal_rotation_angle	.967	15	.818
CUT _1st session_hip_adduction_moment	.950	15	.517
CUT _2nd session_hip_adduction_moment	.966	15	.793
CUT _3rd session_hip_adduction_moment	.948	15	.490
CUT _1st session_dorsiflexion_moment	.929	15	.267
CUT _2nd session_dorsiflexion_moment	.927	15	.243
CUT _3rd session_dorsiflexion_moment	.958	15	.654
CUT _1st session_hip_flexion_moment	.911	15	.140
CUT _2nd session_hip_flexion_moment	.899	15	.092
CUT _3rd session_hip_flexion_moment	.887	15	.061

Appendix (C)

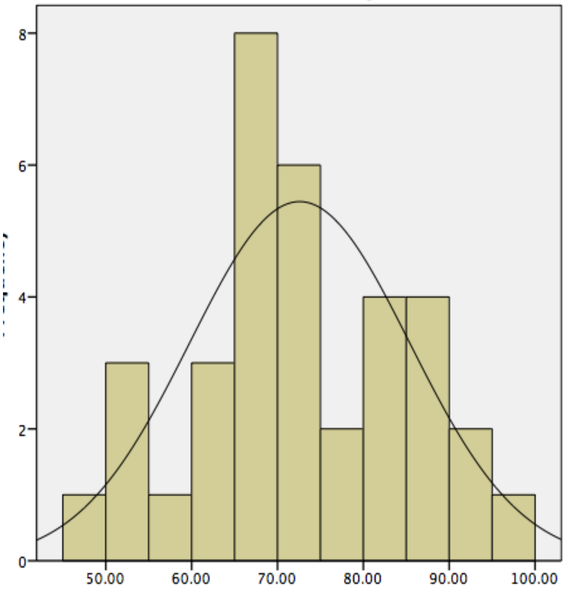
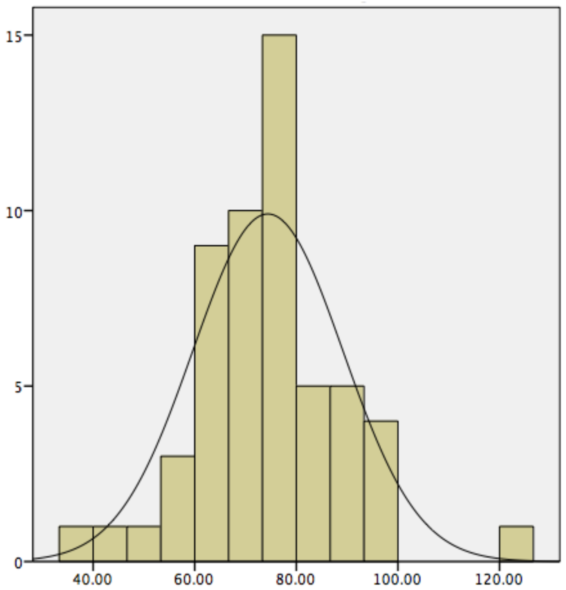
Tests of Normality for SLS variables in Male Participants

	Shapiro-Wilk		
	Statistic	df	Sig.
Hip Flexion Angle	.974	55	.290
Hip Flexion Moment	.985	55	.728
Hip Adduction Angle	.973	55	.261
Hip Adduction Moment	.951	55	.027
Hip Internal Rotation Angle	.974	55	.275
Hip Internal Rotation Moment	.876	55	.000
Knee Flexion angle	.979	55	.433
Knee Flexion Moment	.957	55	.050
Knee Valgus Angle	.968	55	.145
Knee Valgus Moment	.987	55	.793
Knee Internal Rotation Angle	.958	55	.051
Ankle Dorsiflexion Angle	.956	55	.043
Ankle Dorsiflexion Moment	.989	55	.888
Vertical GRF	.982	55	.590

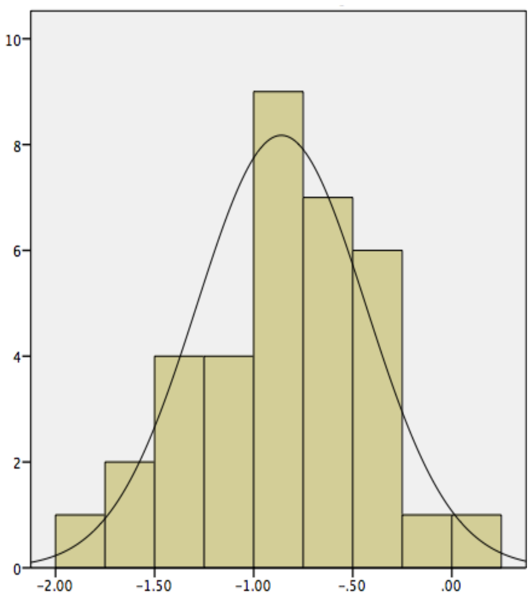
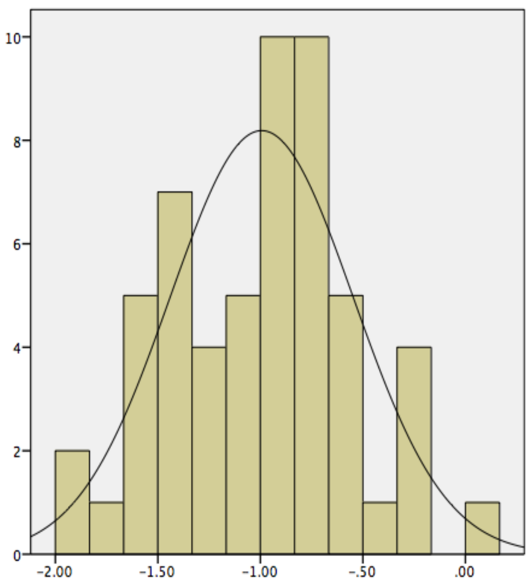
Tests of Normality for SLS variables in Female Participants

	Shapiro-Wilk		
	Statistic	df	Sig.
Hip Flexion Angle	.970	35	.442
Hip Flexion Moment	.983	35	.850
Hip Adduction Angle	.977	35	.654
Hip Adduction Moment	.989	35	.972
Hip Internal Rotation Angle	.990	35	.980
Hip Internal Rotation Moment	.939	35	.054
Knee Flexion angle	.970	35	.433
Knee Flexion Moment	.974	35	.556
Knee Valgus Angle	.980	35	.759
Knee Valgus Moment	.980	35	.752
Knee Internal Rotation Angle	.920	35	.014
Ankle Dorsiflexion Angle	.951	35	.118
Ankle Dorsiflexion Moment	.978	35	.705
Vertical GRF	.966	35	.345

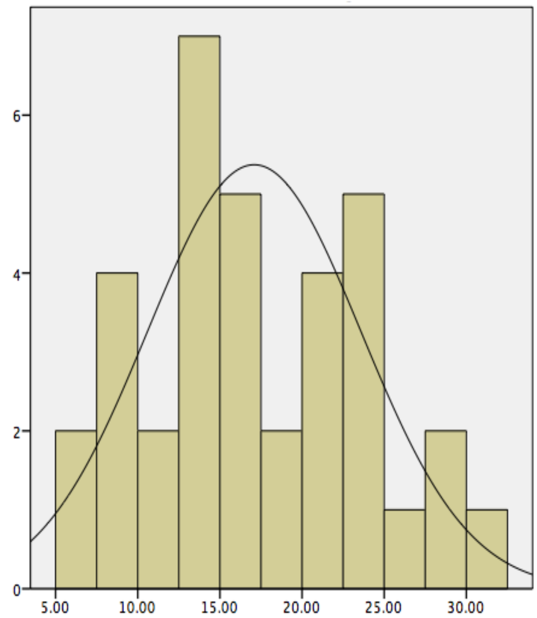
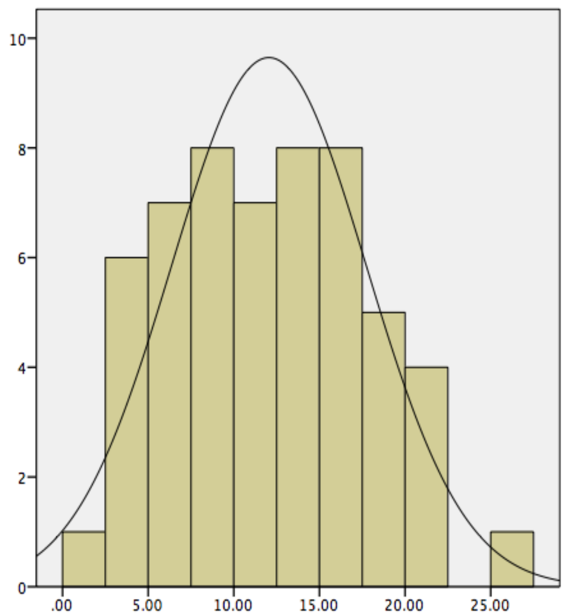
Hip Flexion Angle During SLS Task

Females	Males
	

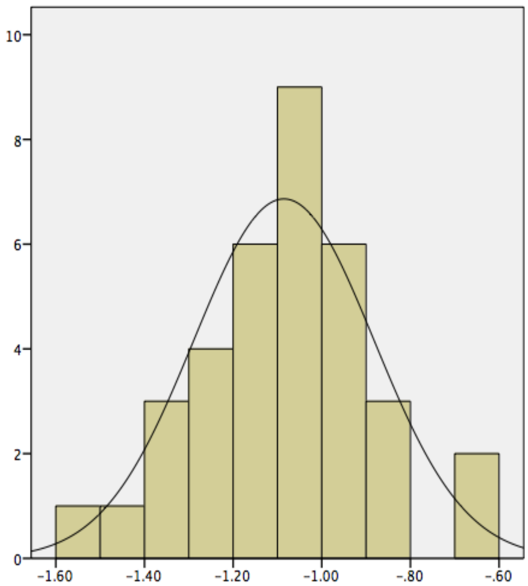
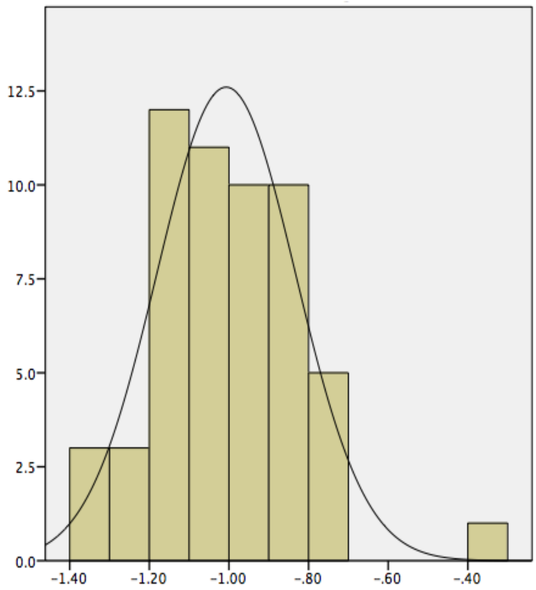
Hip Flexion Moment During SLS Task

Females	Males
	

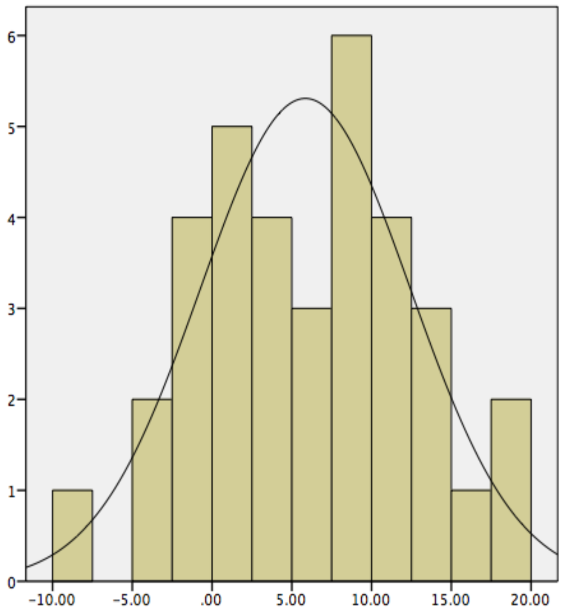
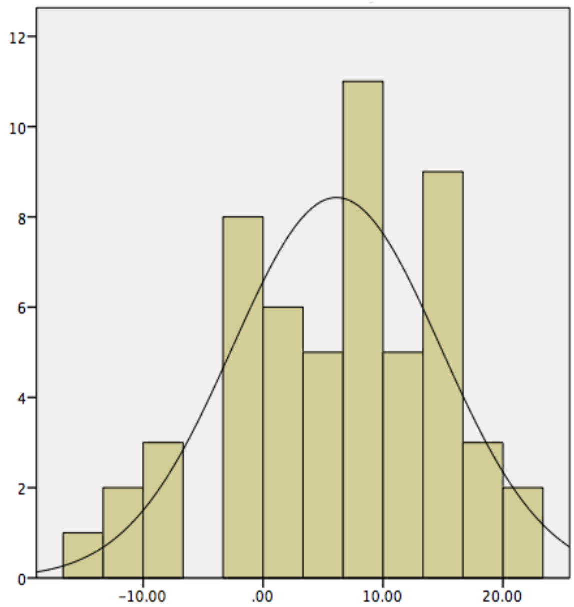
Hip Adduction Angle During SLS Task

Females	Males
	

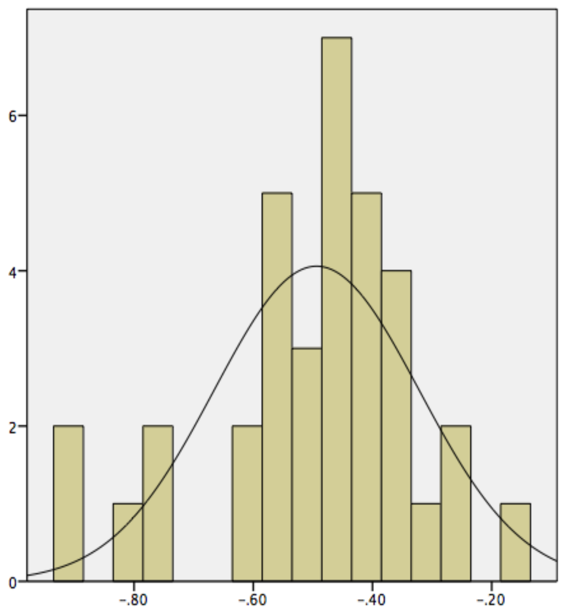
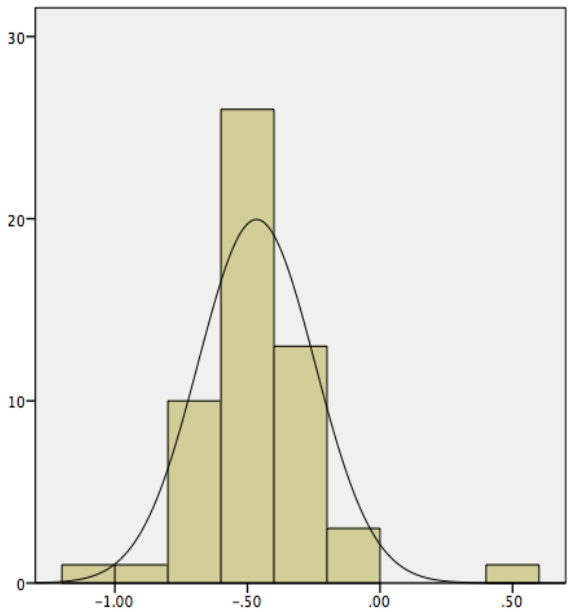
Hip Adduction Moment During SLS Task

Females	Males
	

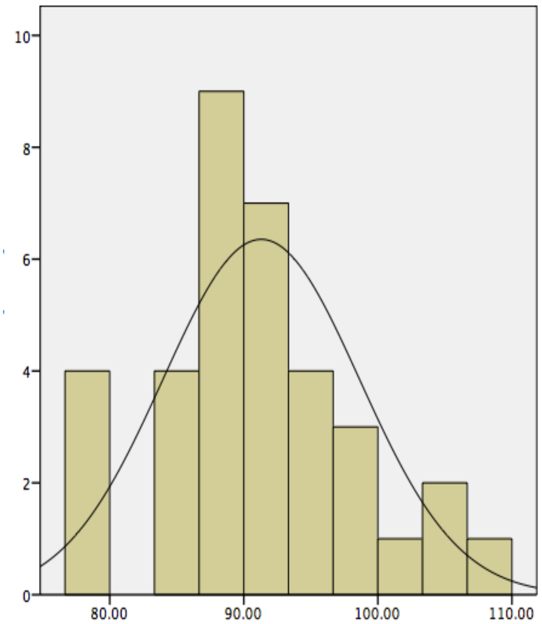
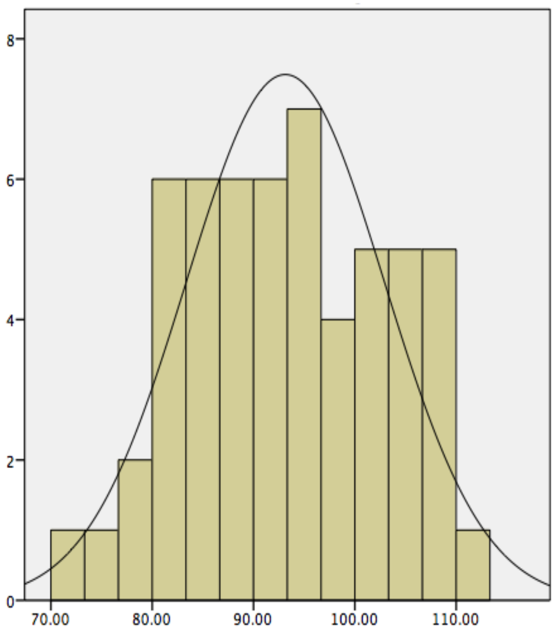
Hip Internal Rotation Angle During SLS Task

Females	Males
	

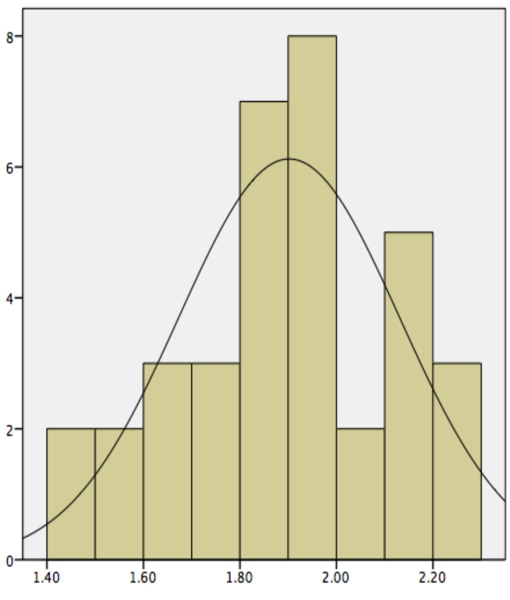
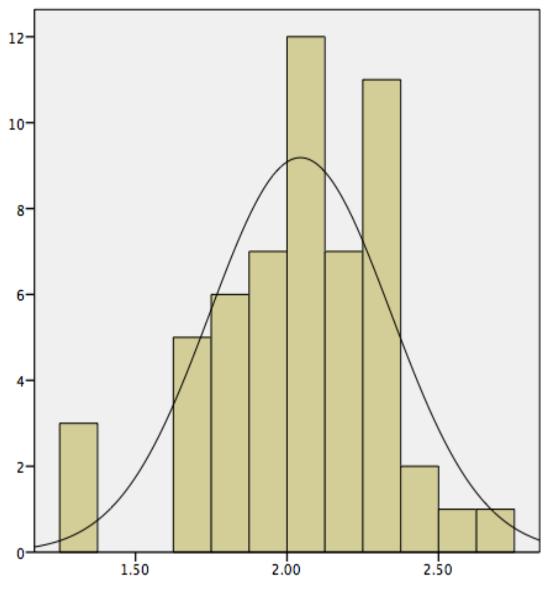
Hip Internal Rotation Moment During SLS Task

Females	Males
	

Knee Flexion Angle During SLS Task

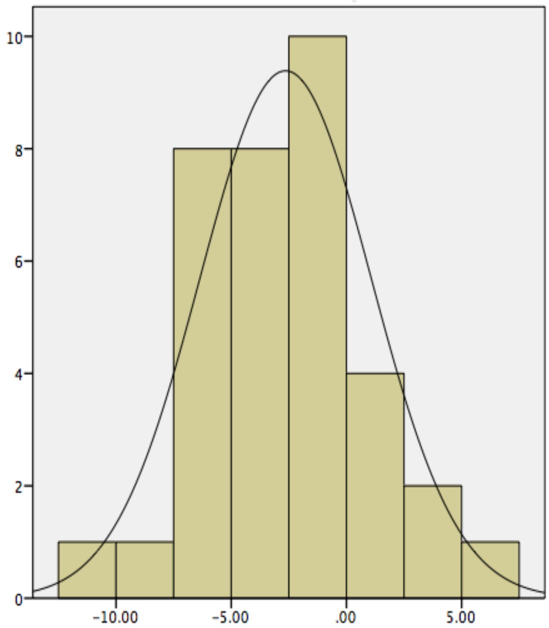
Females	Males
	

Knee Flexion Moment During SLS Task

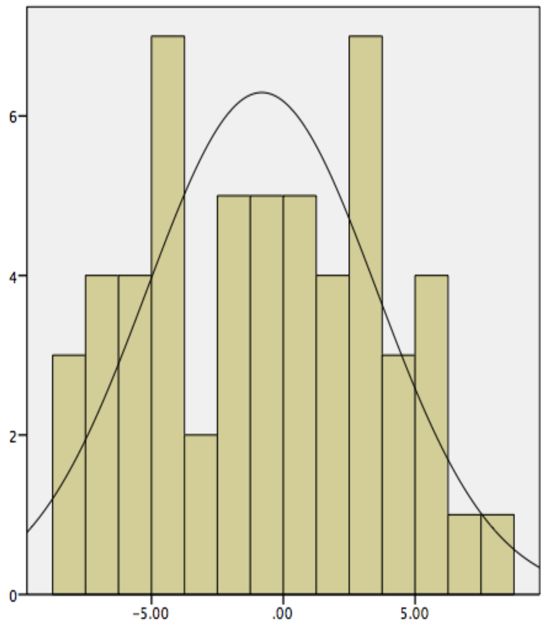
Females	Males
	

Knee Valgus Angle During SLS Task

Females

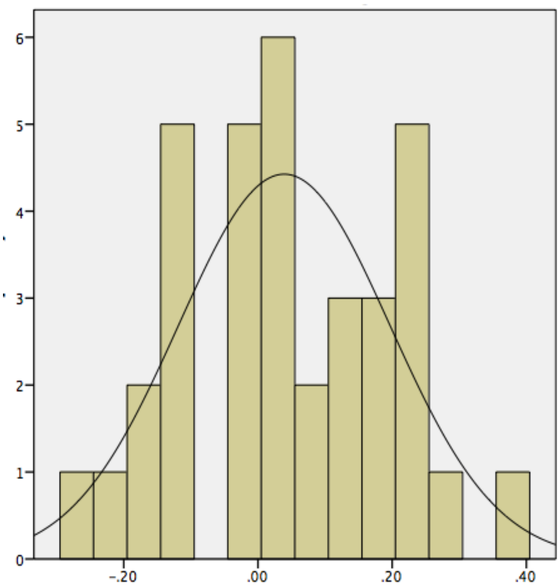


Males

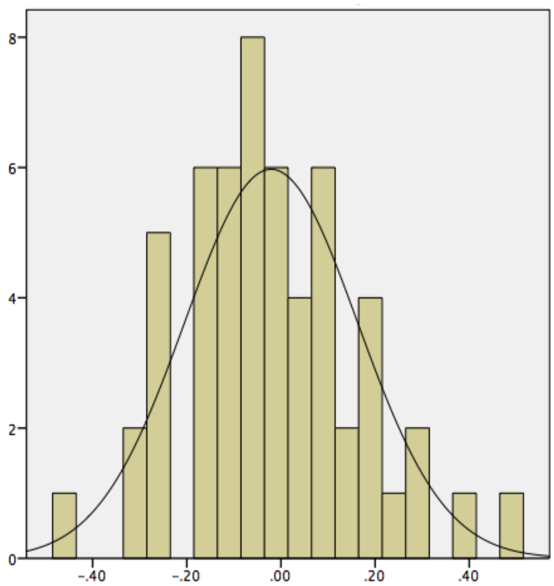


Knee Valgus Moment During SLS Task

Females

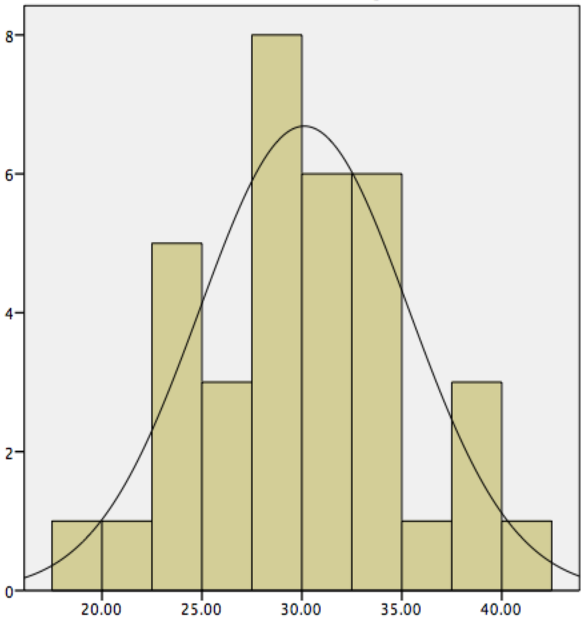


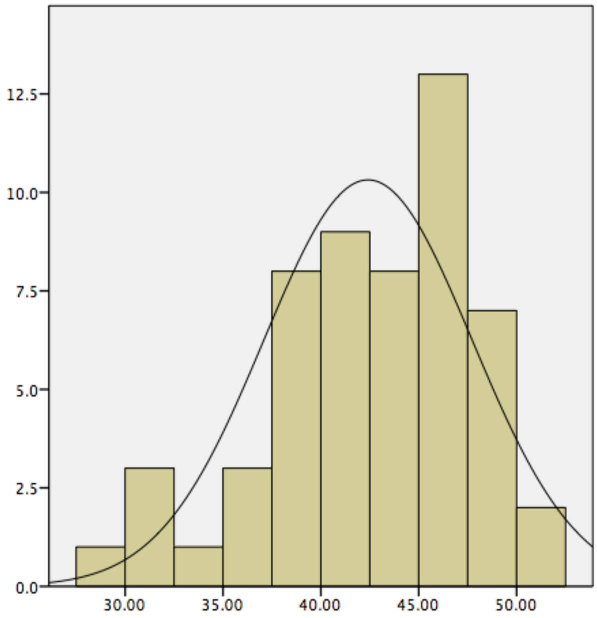
Males



Ankle Dorsiflexion Angle During SLS Task

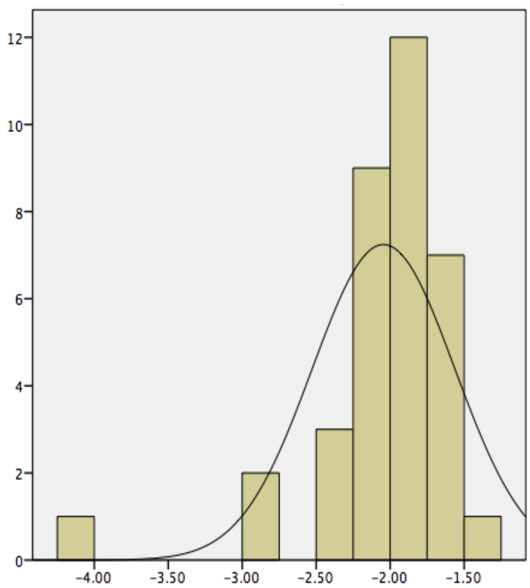
Females	Males

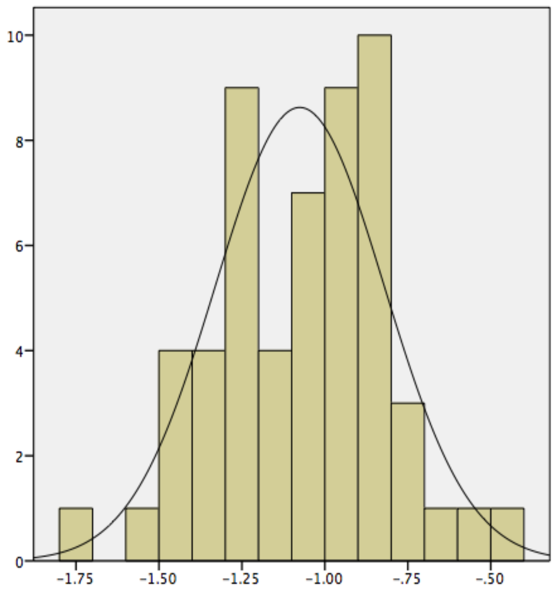




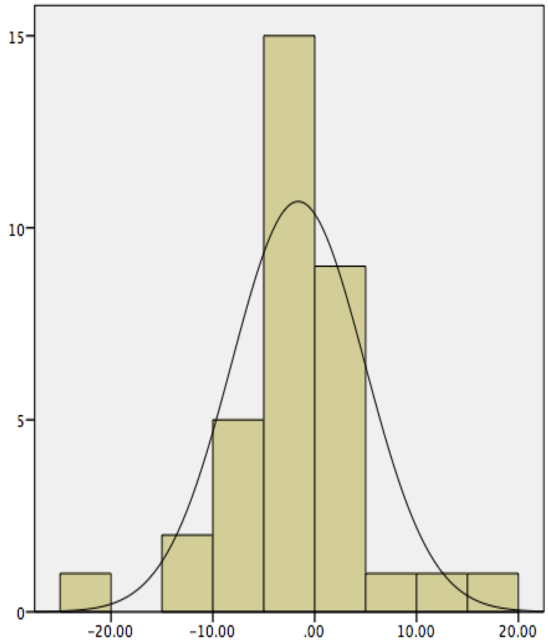
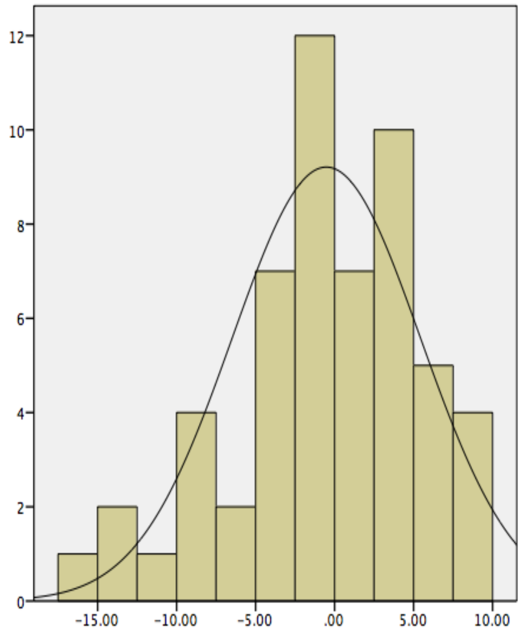
Ankle Dorsiflexion Moment During SLS Task

Females	Males

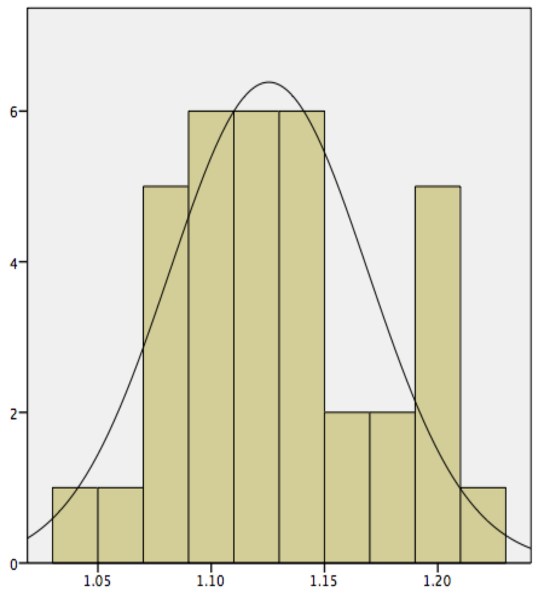
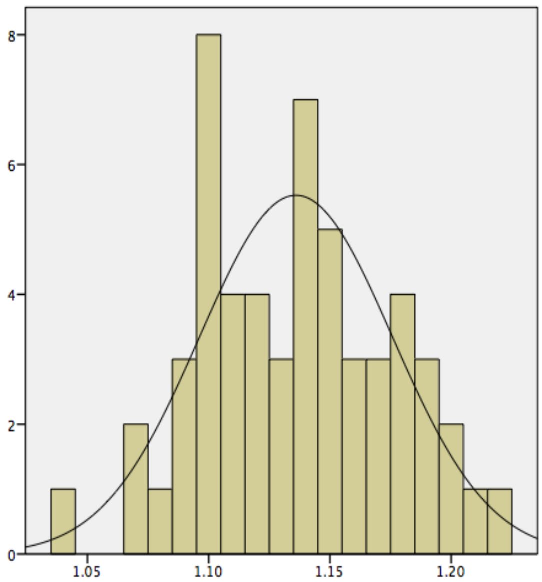




Knee Internal Rotation Angle During SLS Task

Females	Males
	

Vertical Ground Reaction Force During SLS Task

Females	Males
	

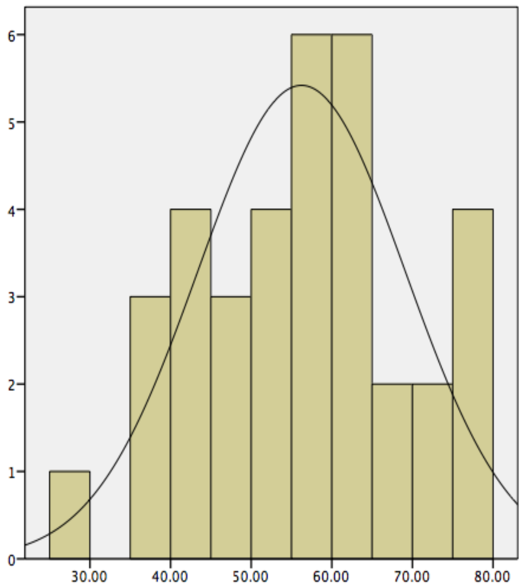
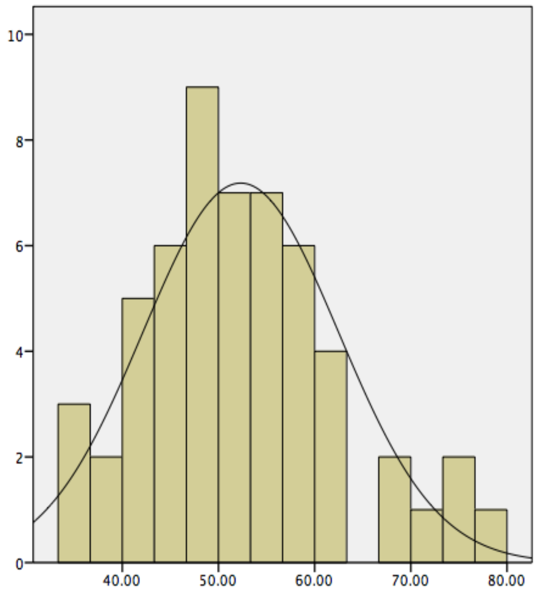
Tests of Normality for SLL variables in Male Participants

	Shapiro-Wilk		
	Statistic	df	Sig.
Hip Flexion Angle	.970	55	.190
Hip Flexion Moment	.909	55	.001
Hip Adduction Angle	.980	55	.468
Hip Adduction Moment	.989	55	.877
Hip Internal Rotation Angle	.987	55	.832
Hip Internal Rotation Moment	.941	55	.010
Knee Flexion angle	.982	55	.570
Knee Flexion Moment	.987	55	.816
Knee Valgus Angle	.984	55	.676
Knee Valgus Moment	.964	55	.100
Knee Internal Rotation Angle	.955	55	.039
Ankle Dorsiflexion Angle	.975	55	.295
Ankle Dorsiflexion Moment	.830	55	.000
Vertical GRF	.973	55	.256

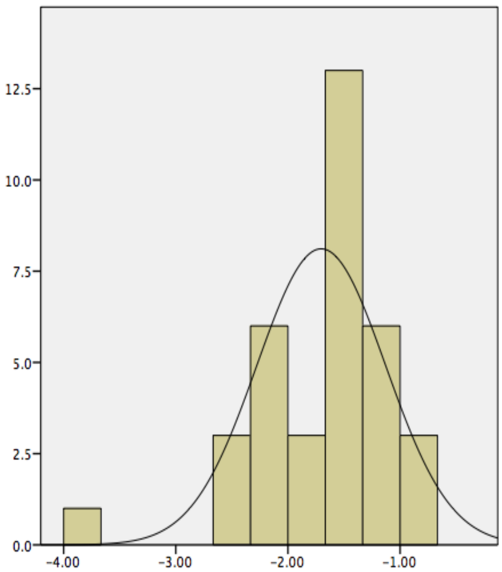
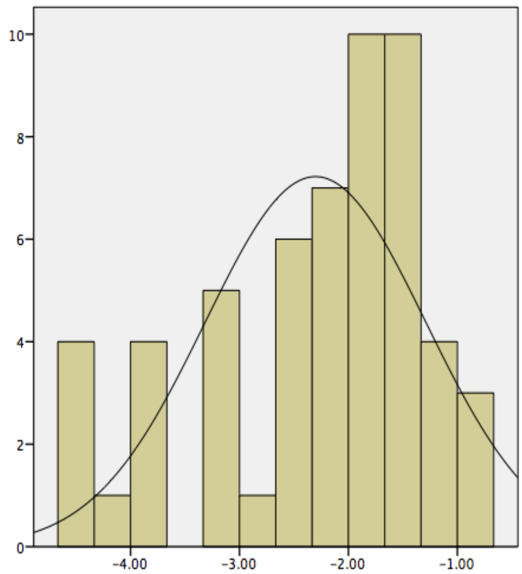
Tests of Normality for SLL variables in Female Participants

	Shapiro-Wilk		
	Statistic	df	Sig.
Hip Flexion Angle	.974	35	.568
Hip Flexion Moment	.889	35	.002
Hip Adduction Angle	.985	35	.909
Hip Adduction Moment	.927	35	.022
Hip Internal Rotation Angle	.979	35	.740
Hip Internal Rotation Moment	.882	35	.001
Knee Flexion angle	.964	35	.305
Knee Flexion Moment	.959	35	.209
Knee Valgus Angle	.990	35	.983
Knee Valgus Moment	.796	35	.000
Knee Internal Rotation Angle	.906	35	.006
Ankle Dorsiflexion Angle	.981	35	.806
Ankle Dorsiflexion Moment	.794	35	.000
Vertical GRF	.860	35	.000

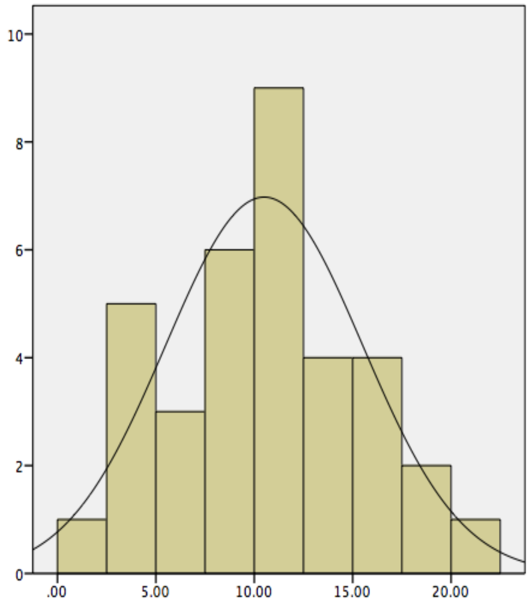
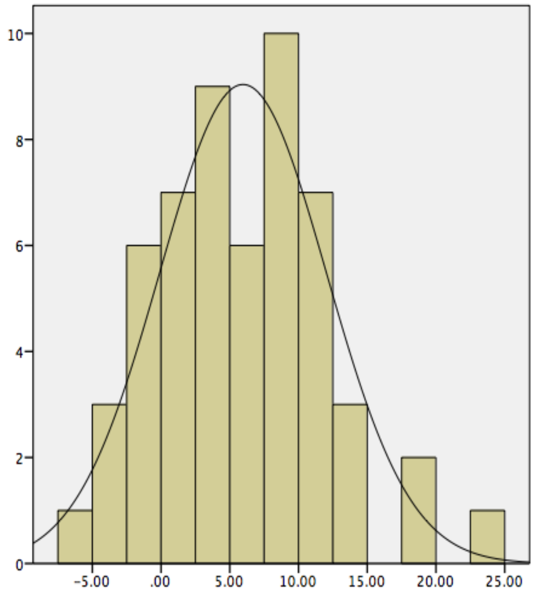
Hip Flexion Angle During SLL Task

Females	Males
	

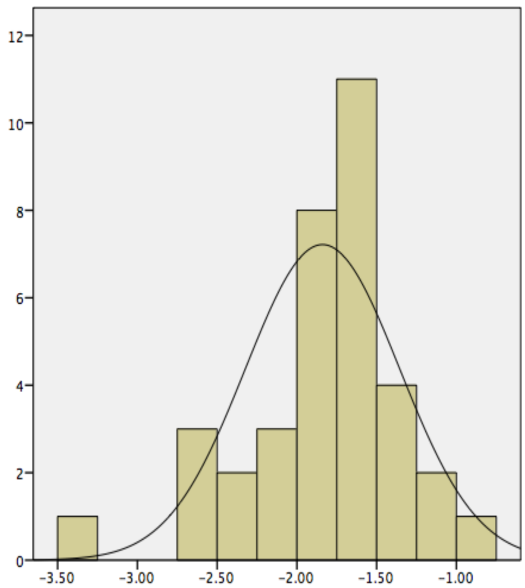
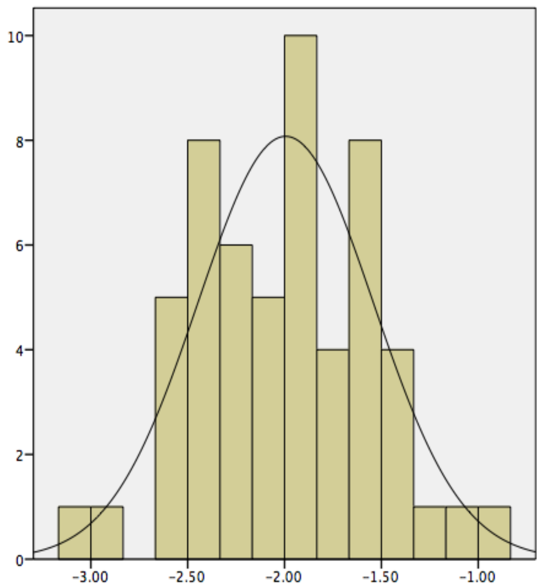
Hip Flexion Moment During SLL Task

Females	Males
	

Hip Adduction Angle During SLL Task

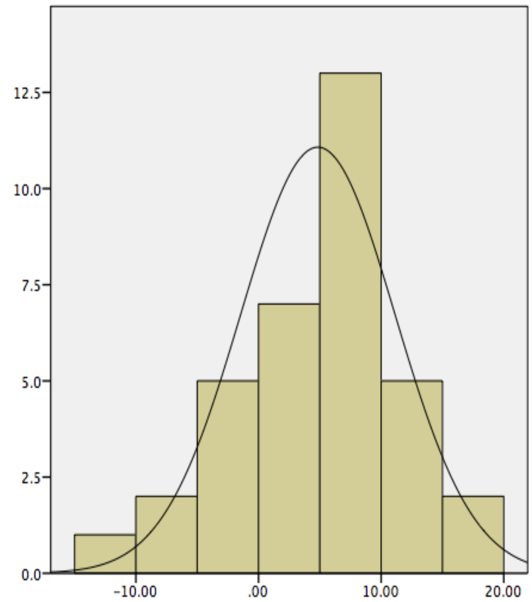
Females	Males
	

Hip Adduction Moment During SLL Task

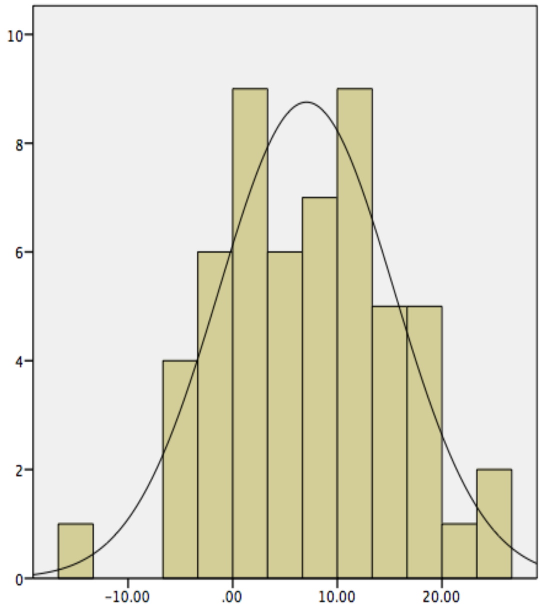
Females	Males
	

Hip Internal Rotation Angle During SLL Task

Females

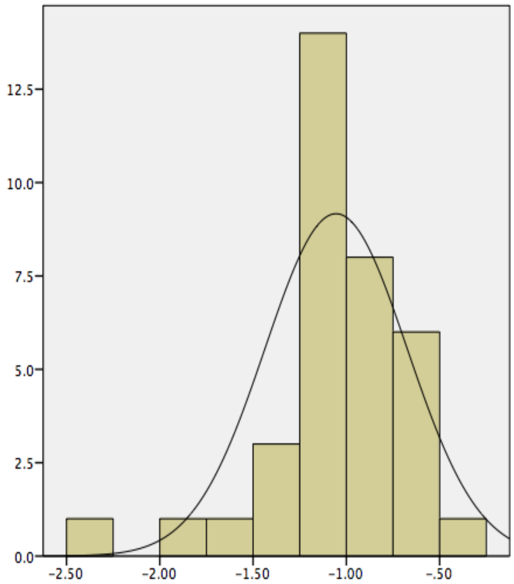


Males

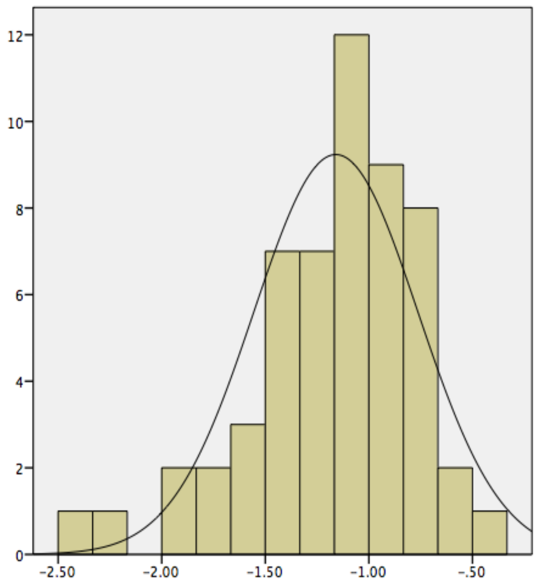


Hip Internal Rotation Moment During SLL Task

Females

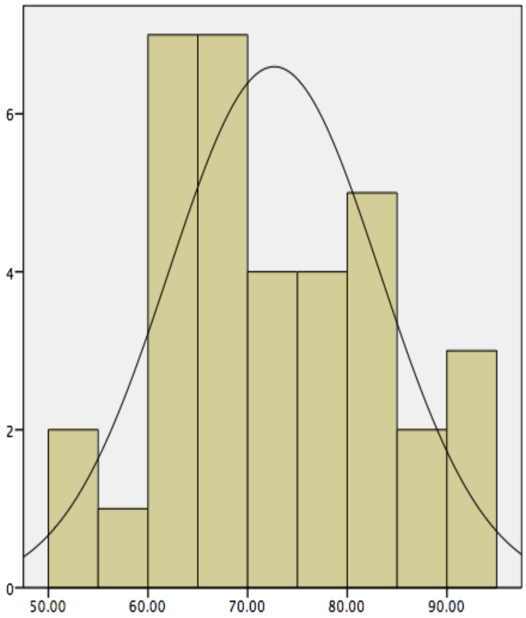


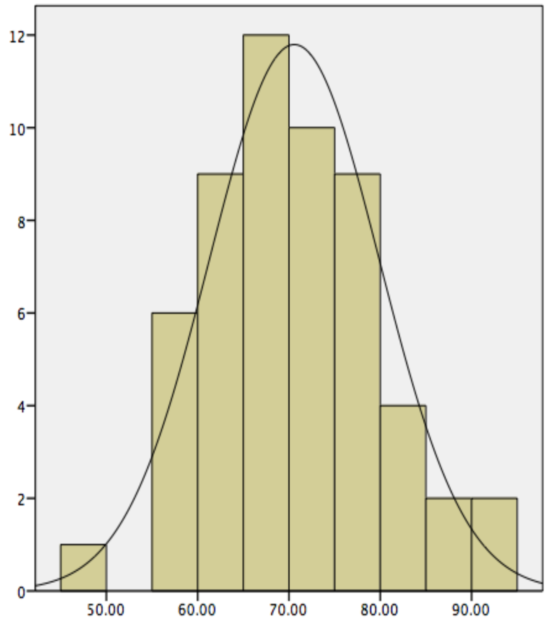
Males



Knee Flexion Angle During SLL Task

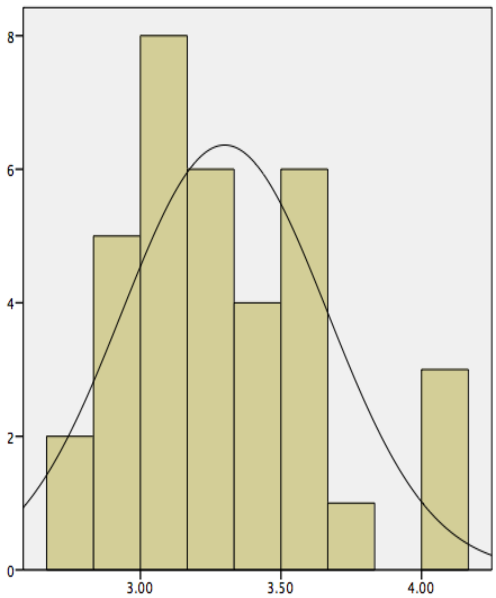
Females	Males

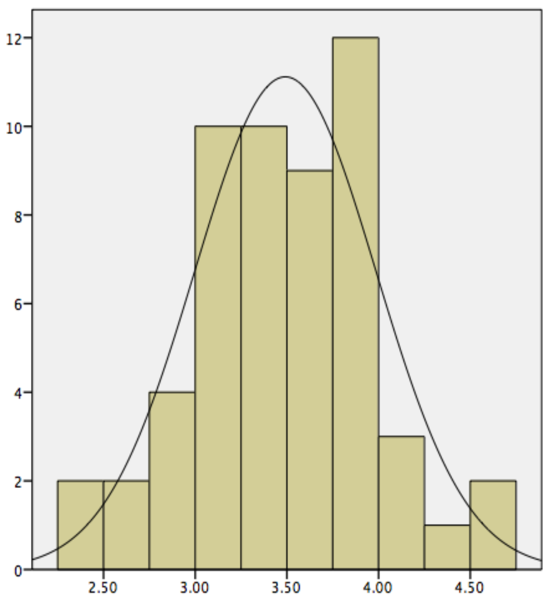




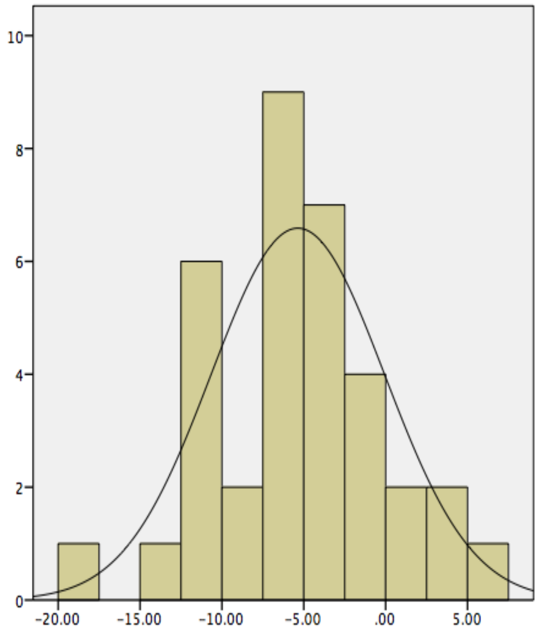
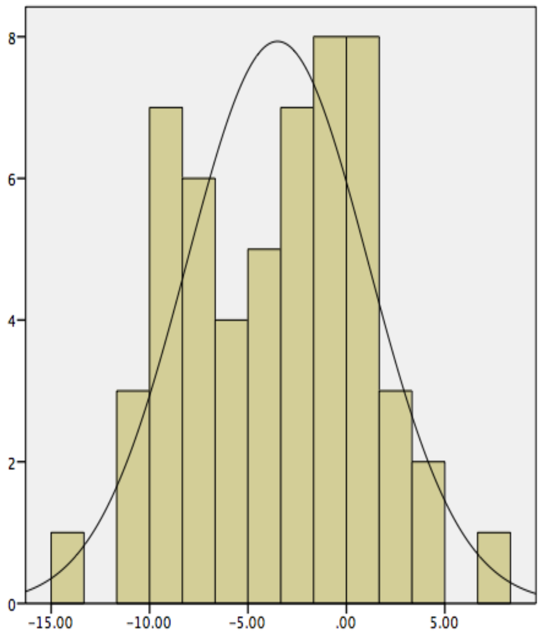
Knee Flexion Moment During SLL Task

Females	Males

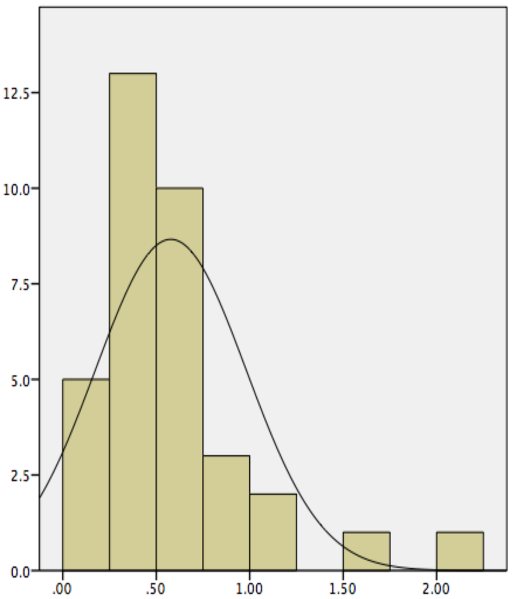
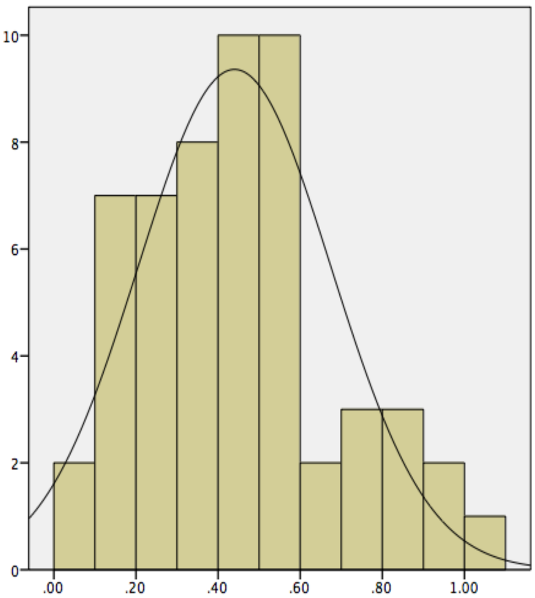




Knee Valgus Angle During SLL Task

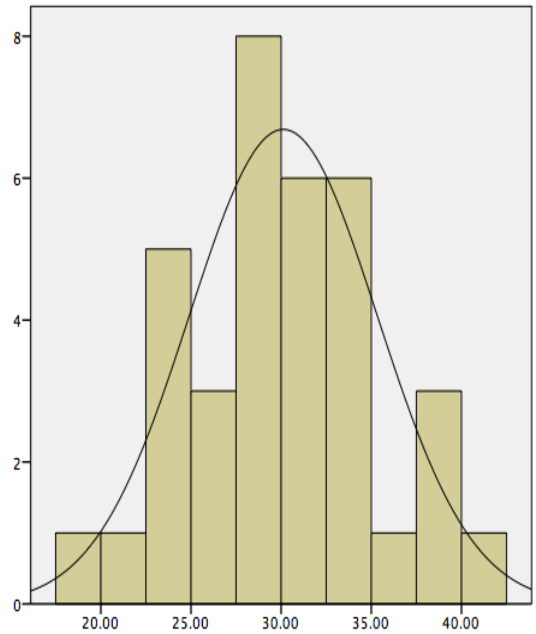
Females	Males
	

Knee Valgus Moment During SLL Task

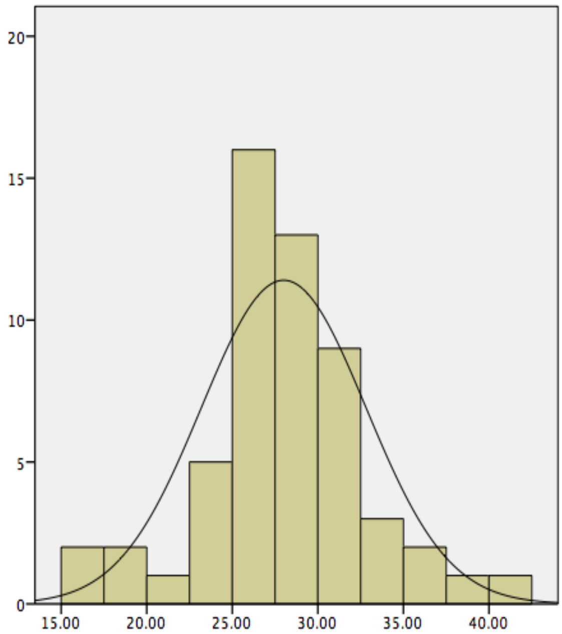
Females	Males
	

Ankle Dorsiflexion Angle During SLL Task

Females

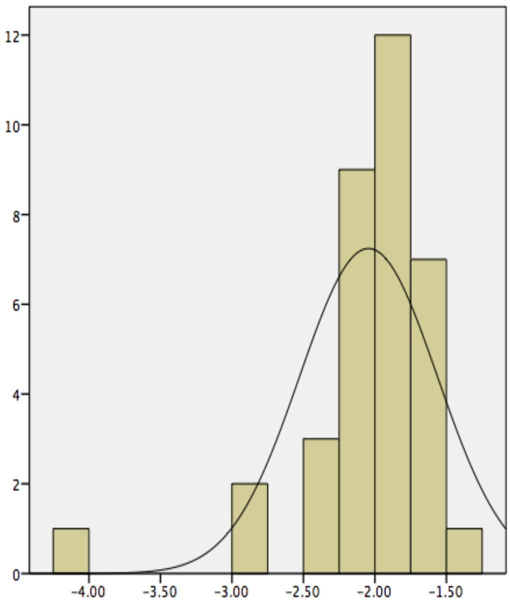


Males

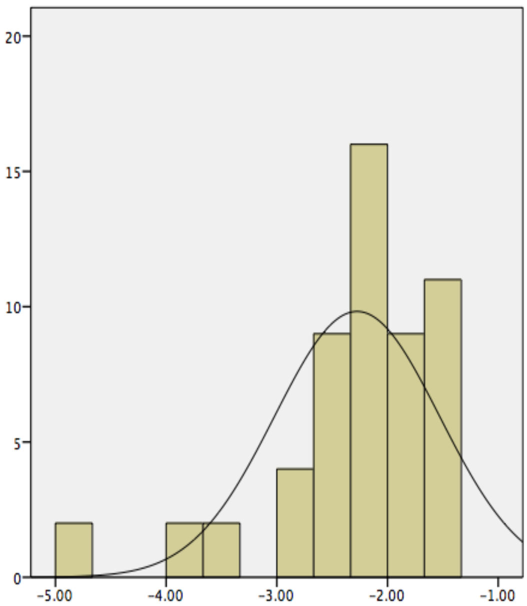


Ankle Dorsiflexion Moment During SLL Task

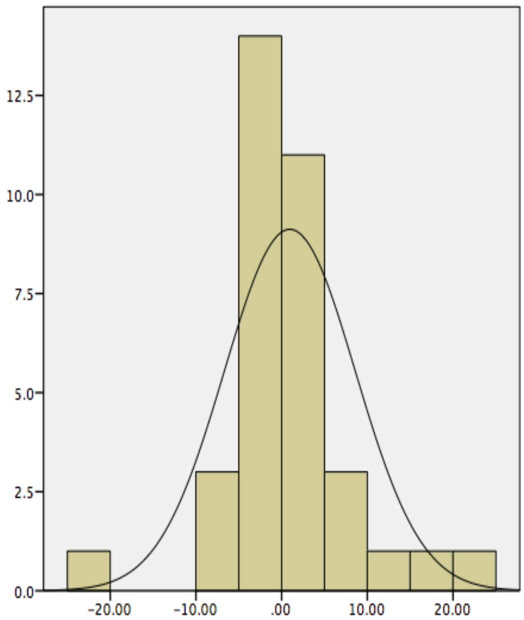
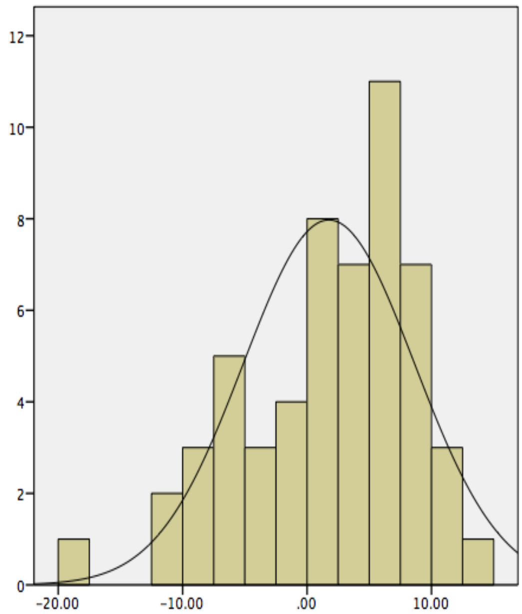
Females



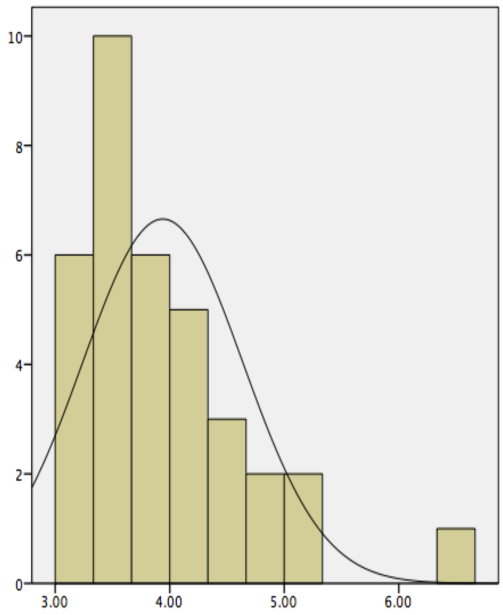
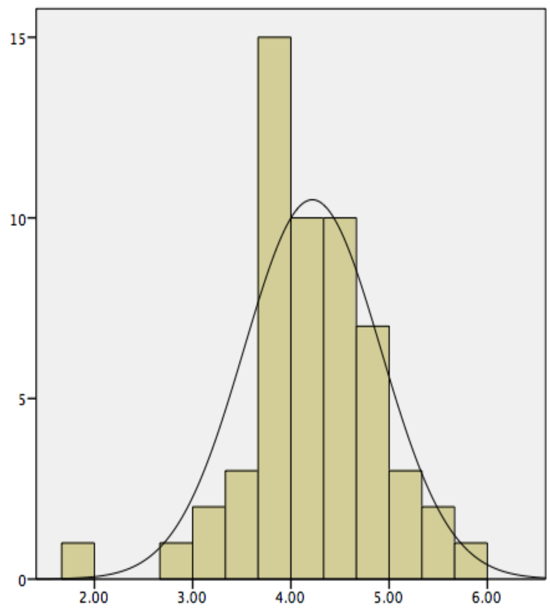
Males



Knee Internal Rotation Angle During SLL Task

Females	Males
	

Vertical Ground Reaction Force During SLL Task

Females	Males
	

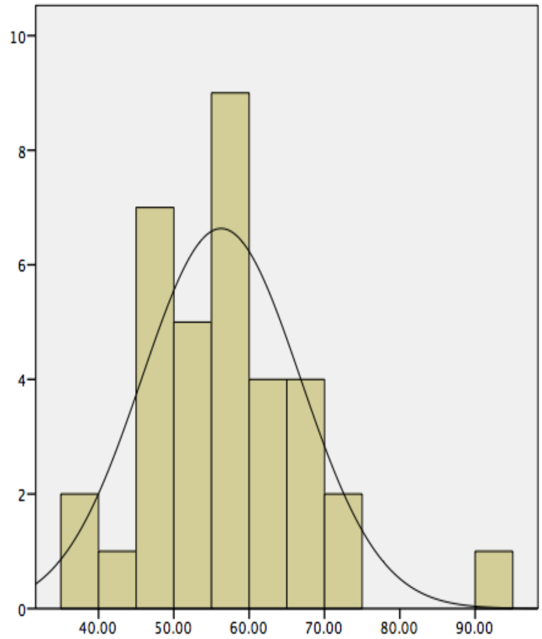
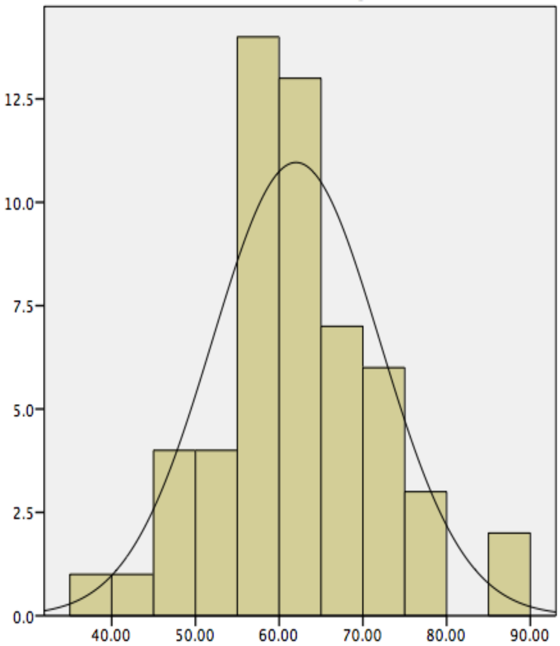
Tests of Normality for RUN variables in Male Participants

	Shapiro-Wilk		
	Statistic	df	Sig.
Hip Flexion Angle	.977	55	.388
Hip Flexion Moment	.981	55	.538
Hip Adduction Angle	.979	55	.444
Hip Adduction Moment	.965	55	.104
Hip Internal Rotation Angle	.977	55	.375
Hip Internal Rotation Moment	.876	55	.000
Knee Flexion angle	.991	55	.954
Knee Flexion Moment	.984	55	.680
Knee Valgus Angle	.975	55	.313
Knee Valgus Moment	.961	55	.071
Knee Internal Rotation Angle	.989	55	.882
Ankle Dorsiflexion Angle	.989	55	.884
Ankle Dorsiflexion Moment	.966	55	.127
Vertical GRF	.836	55	.000

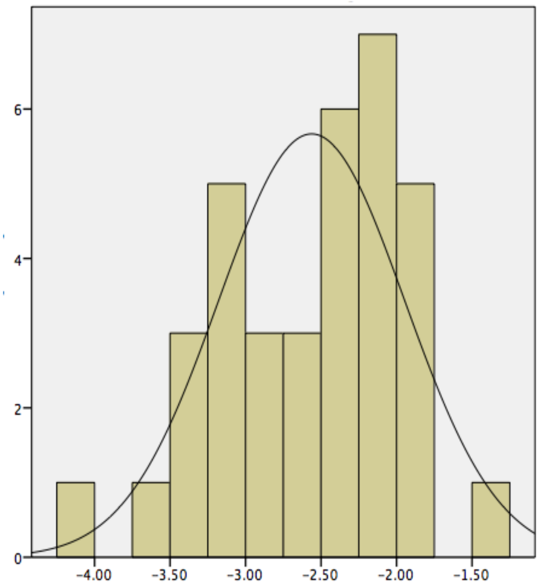
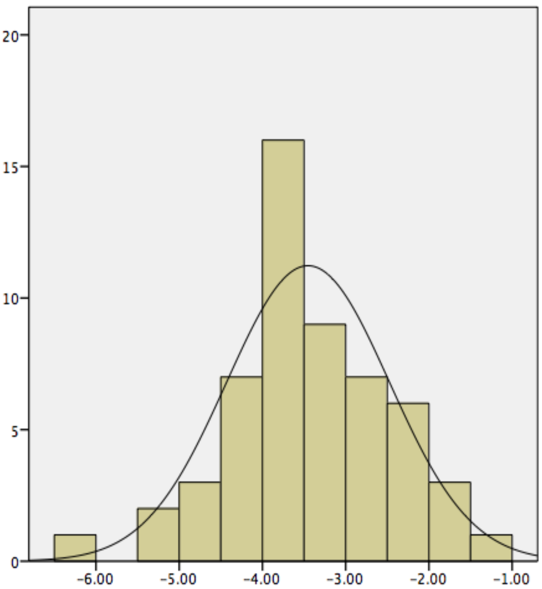
Tests of Normality for RUN variables in Female Participants

	Shapiro-Wilk		
	Statistic	df	Sig.
Hip Flexion Angle	.960	35	.222
Hip Flexion Moment	.962	35	.254
Hip Adduction Angle	.964	35	.301
Hip Adduction Moment	.970	35	.433
Hip Internal Rotation Angle	.987	35	.941
Hip Internal Rotation Moment	.976	35	.640
Knee Flexion angle	.979	35	.733
Knee Flexion Moment	.972	35	.507
Knee Valgus Angle	.932	35	.031
Knee Valgus Moment	.768	35	.000
Knee Internal Rotation Angle	.923	35	.017
Ankle Dorsiflexion Angle	.967	35	.377
Ankle Dorsiflexion Moment	.910	35	.007
Vertical GRF	.967	35	.377

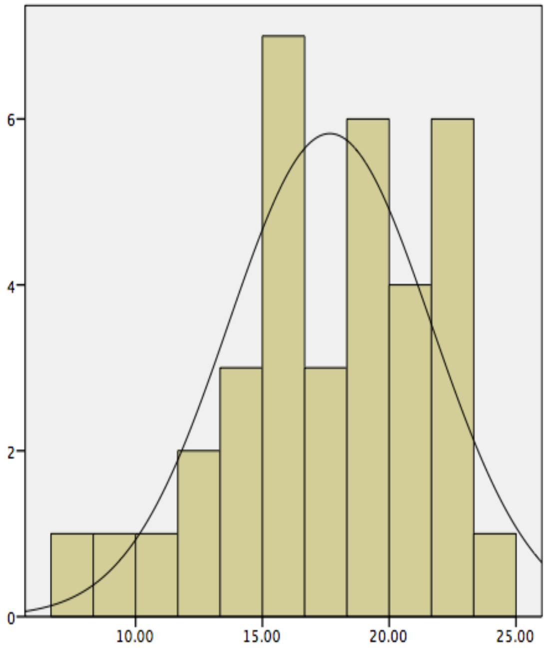
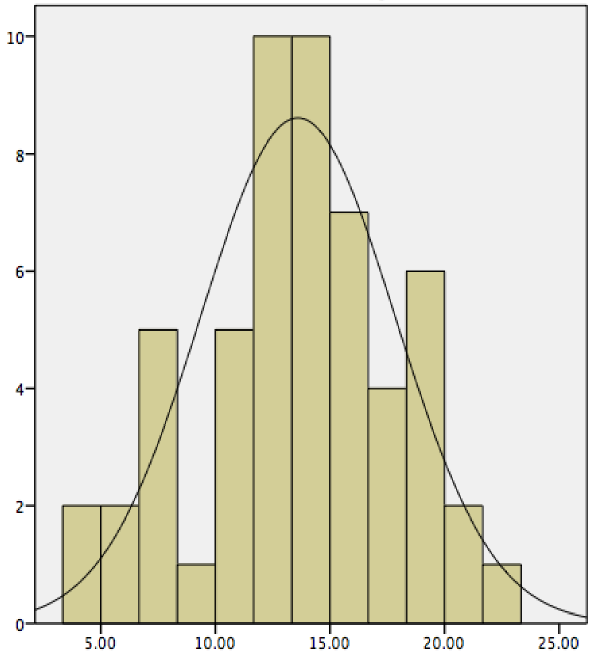
Hip Flexion Angle During RUN Task

Females	Males
	

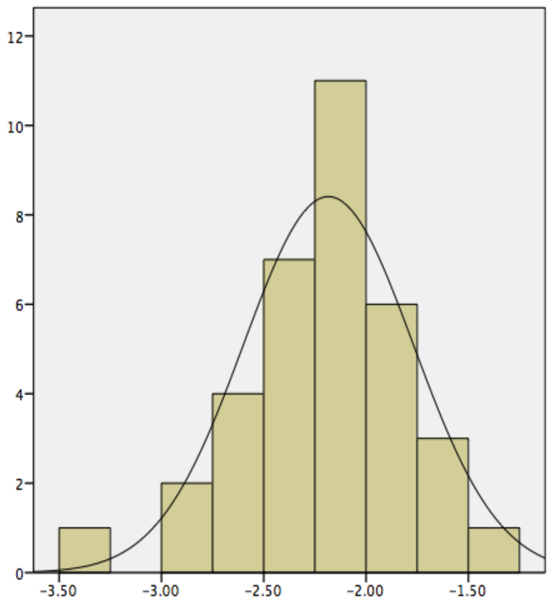
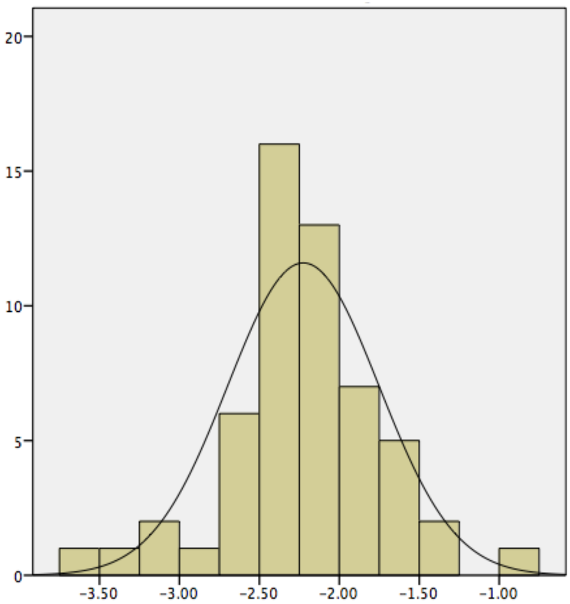
Hip Flexion Moment During RUN Task

Females	Males
	

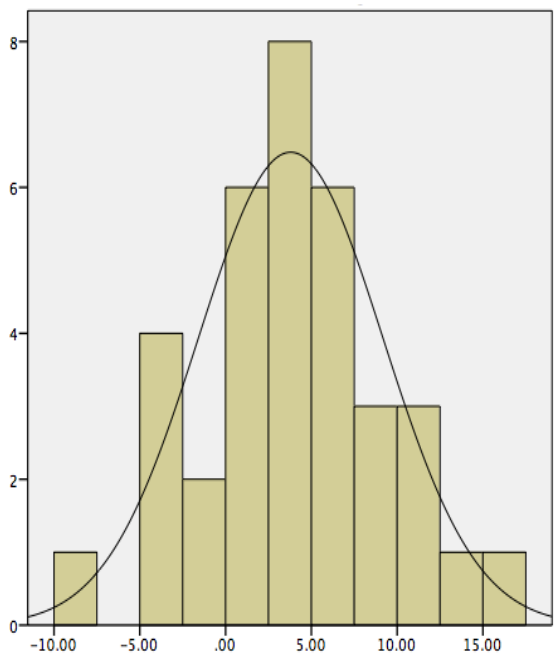
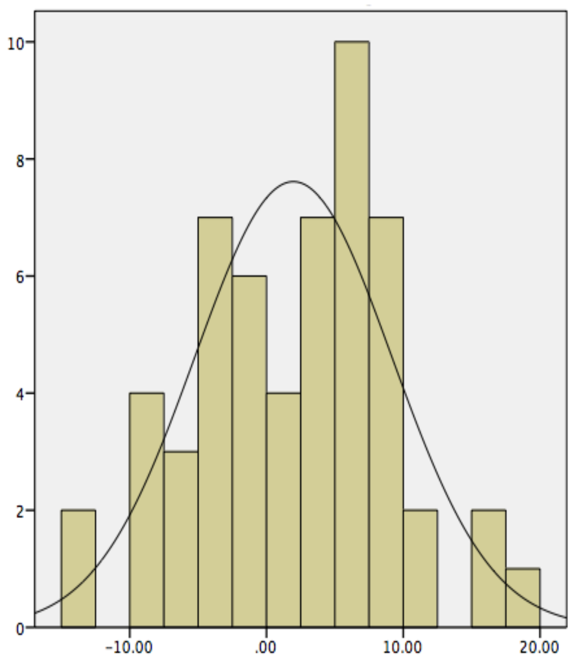
Hip Adduction Angle During RUN Task

Females	Males
	

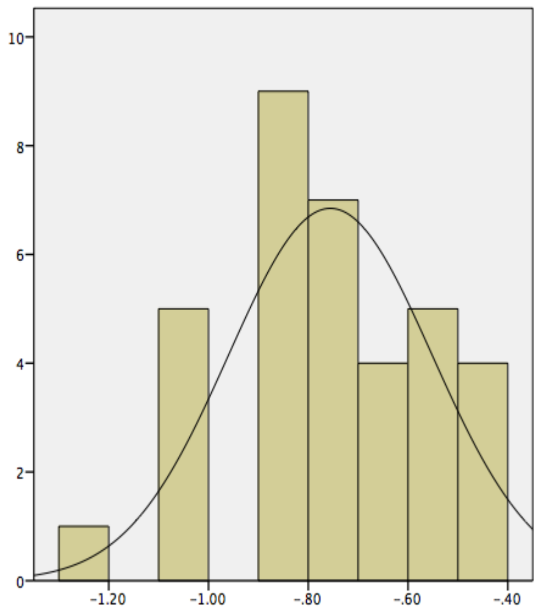
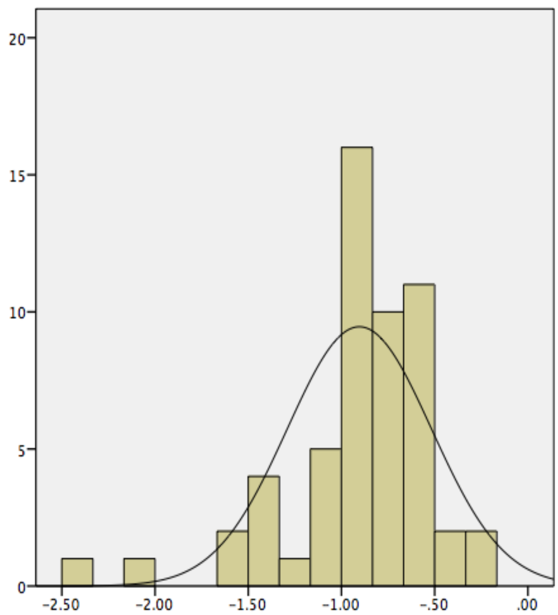
Hip Adduction Moment During RUN Task

Females	Males
	

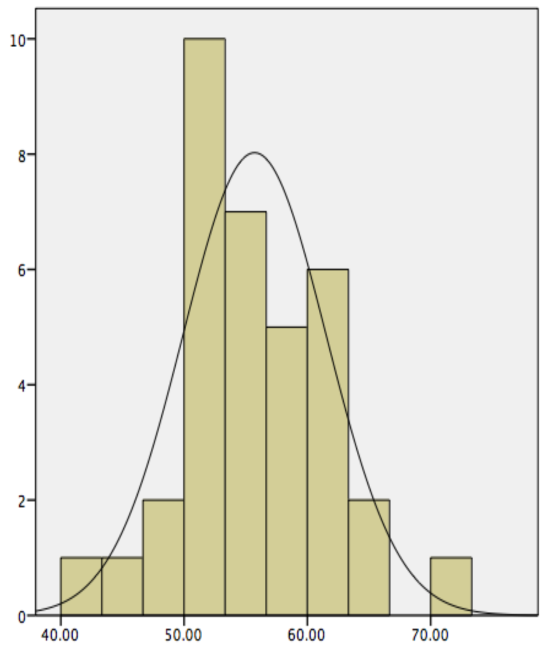
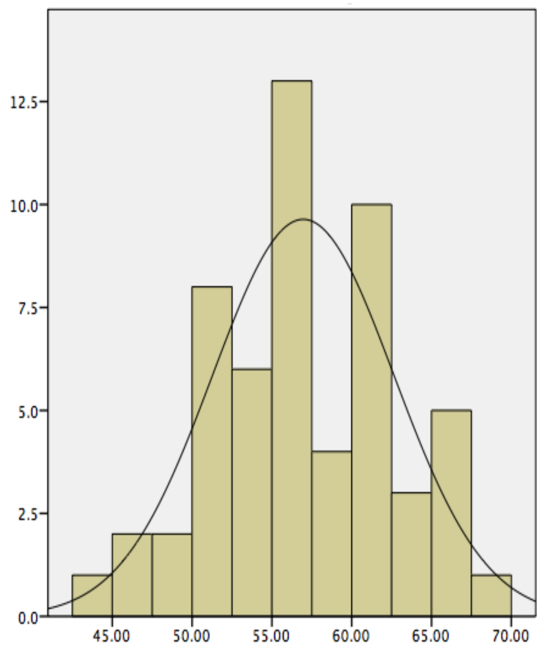
Hip Internal Rotation Angle During RUN Task

Females	Males
	

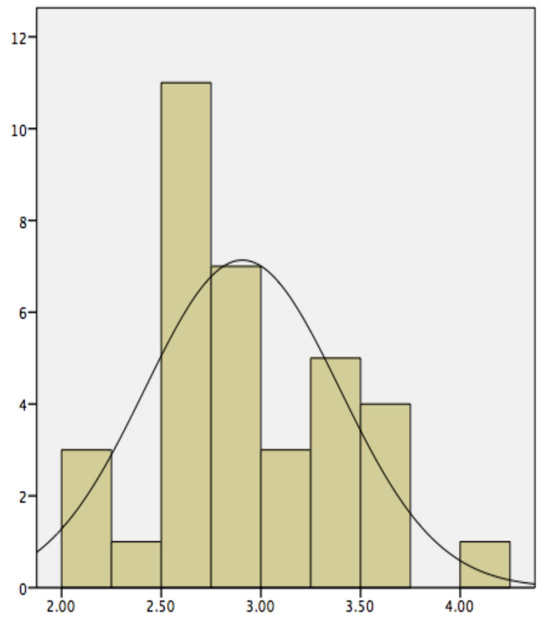
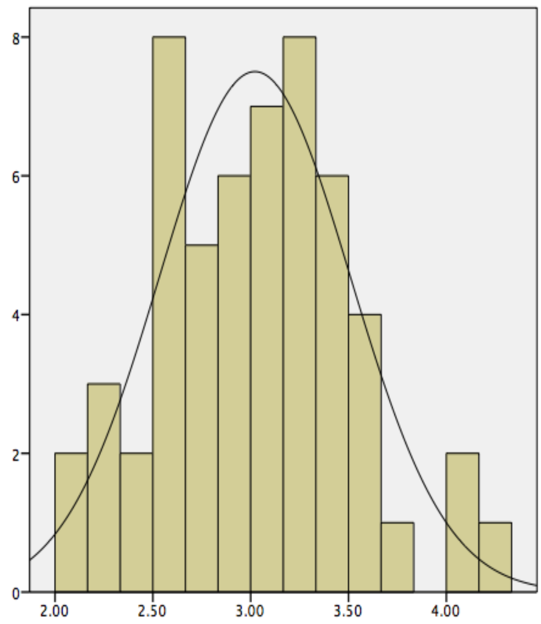
Hip Internal Rotation Moment During RUN Task

Females	Males
	

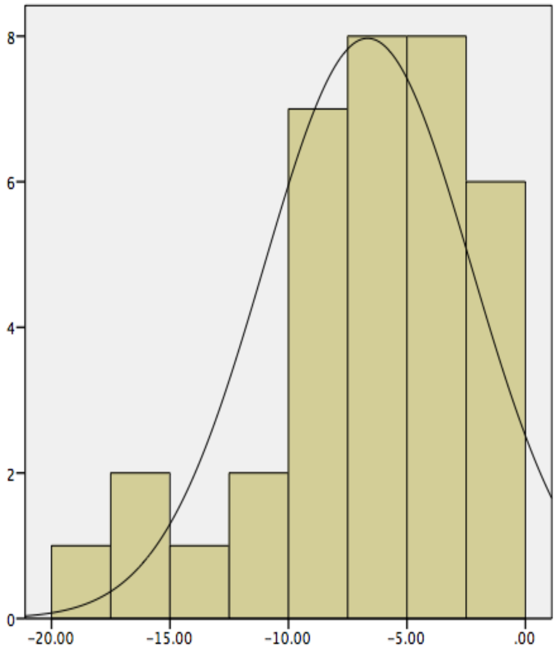
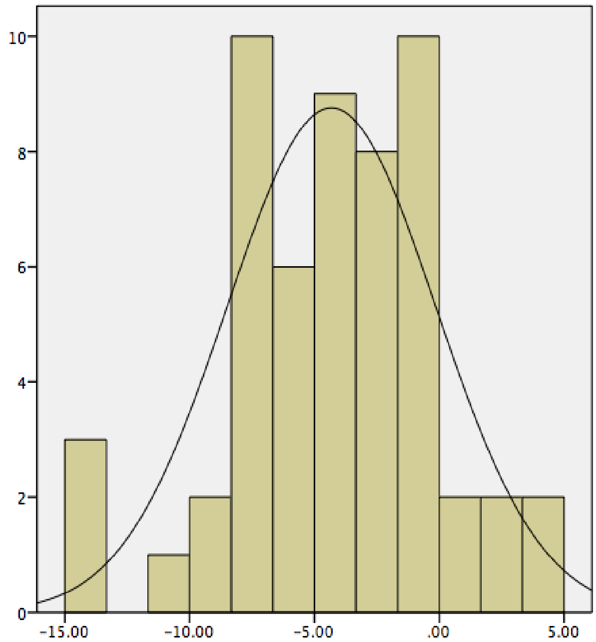
Knee Flexion Angle During RUN Task

Females	Males
	

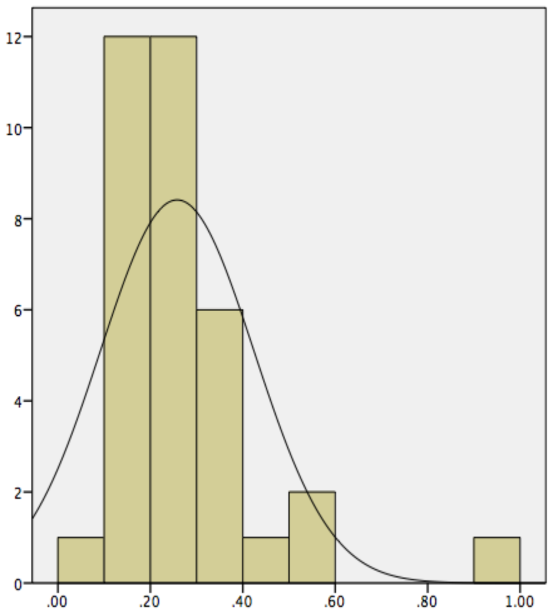
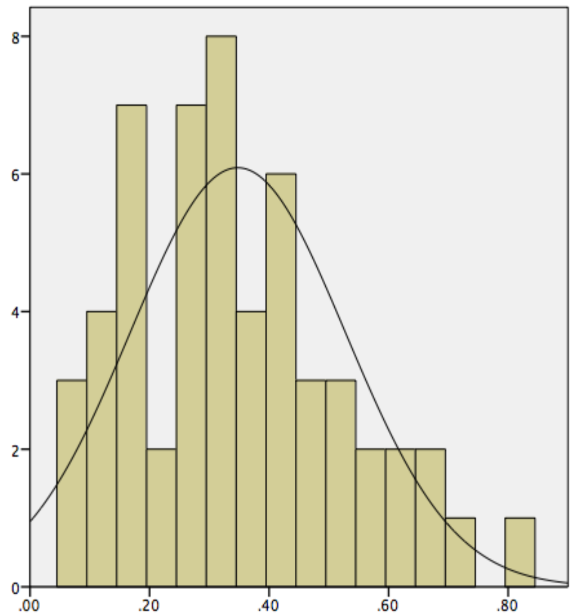
Knee Flexion Moment During RUN Task

Females	Males
	

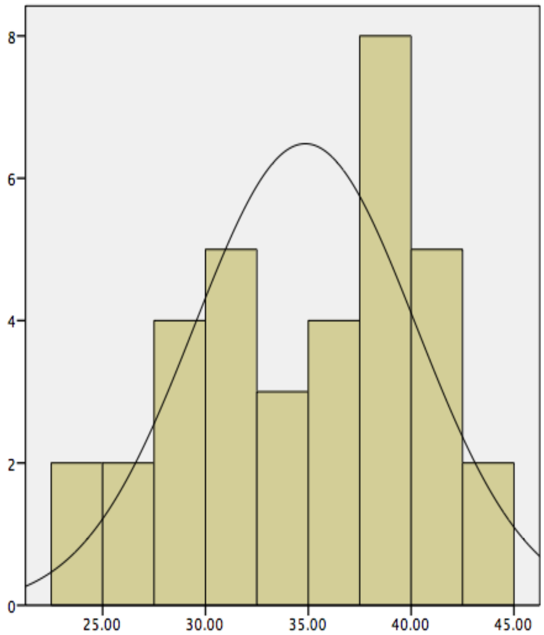
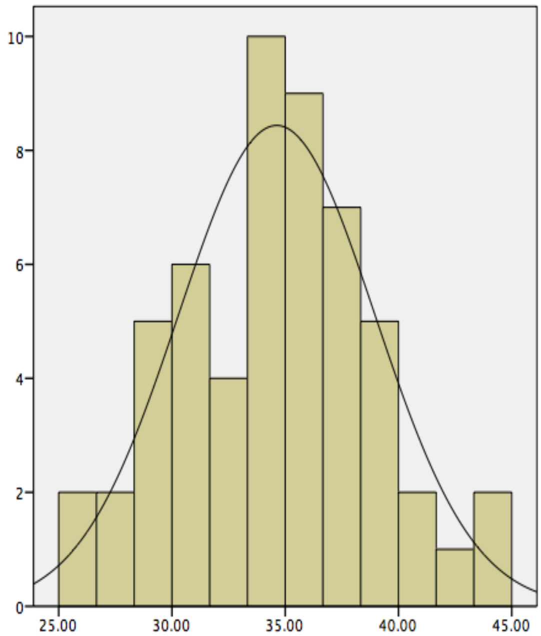
Knee Valgus Angle During RUN Task

Females	Males
	

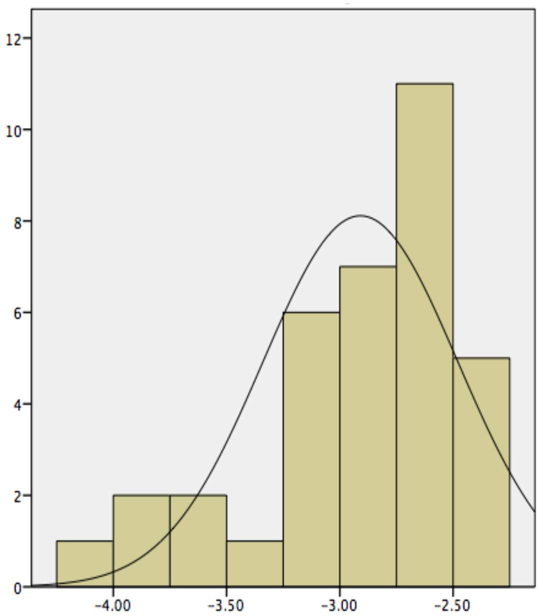
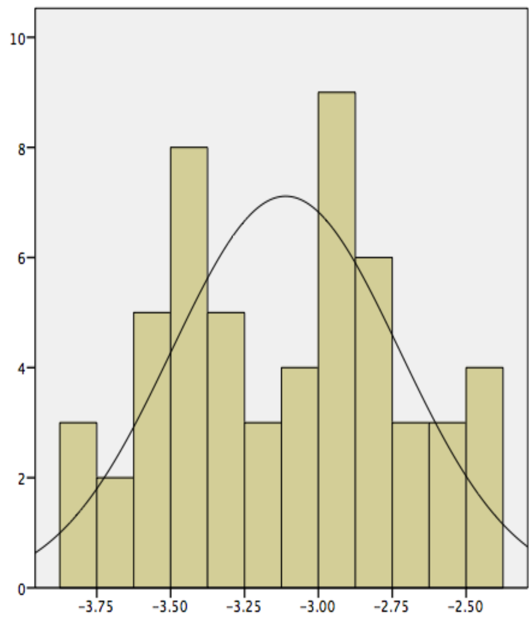
Knee Valgus Moment During RUN Task

Females	Males
	

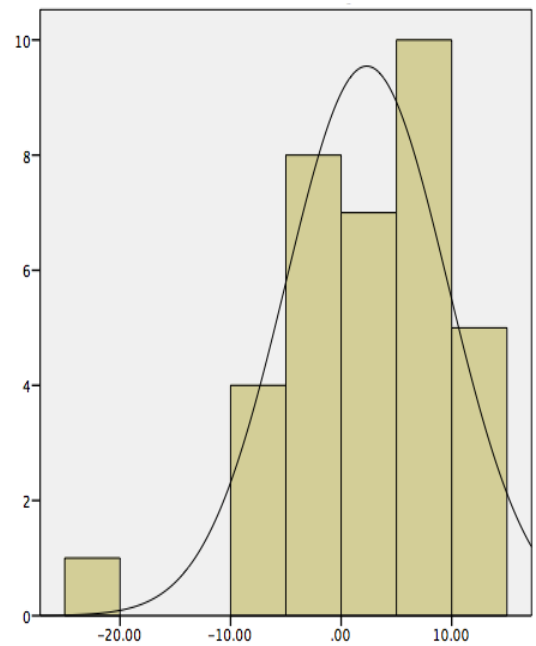
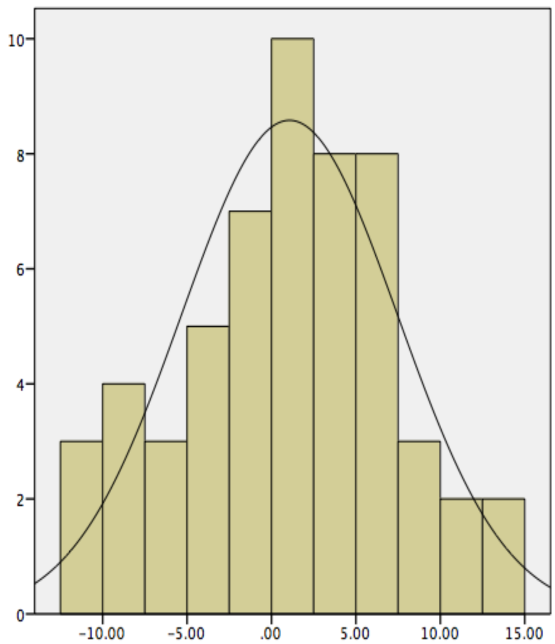
Ankle Dorsiflexion Angle During RUN Task

Females	Males
	

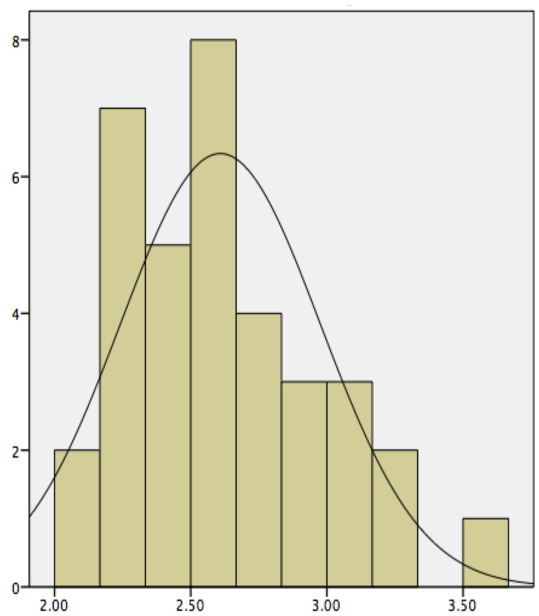
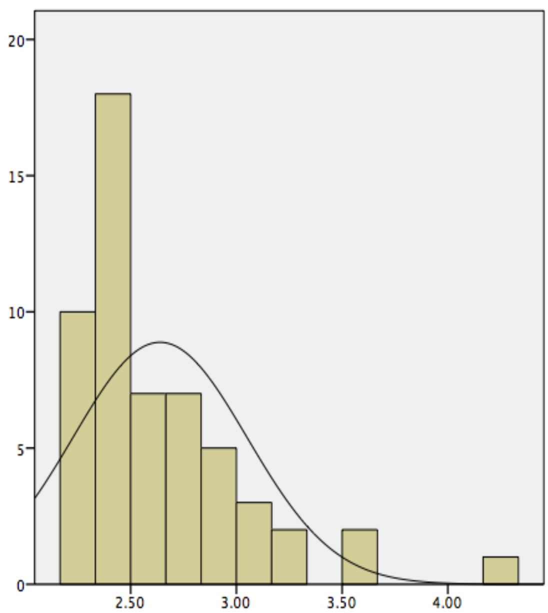
Ankle Dorsiflexion Moment During RUN Task

Females	Males
	

Knee Internal Rotation Angle During RUN Task

Females	Males
	

Vertical Ground Reaction Force During RUN Task

Females	Males
	

Tests of Normality for CUT variables in Male Participants

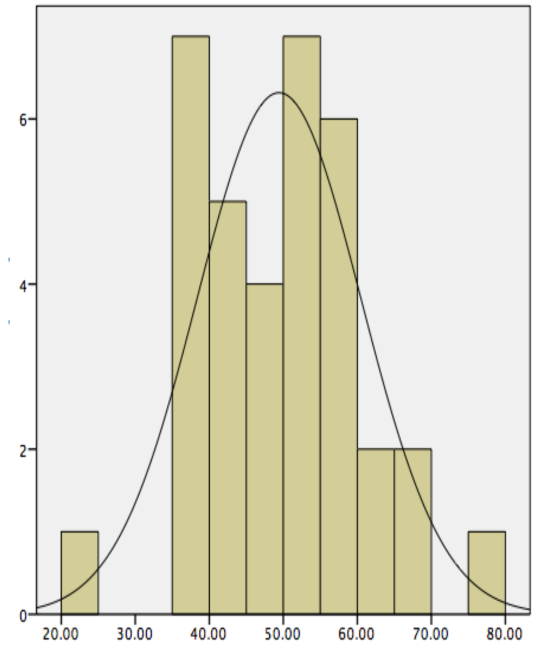
	Shapiro-Wilk		
	Statistic	df	Sig.
Hip Flexion Angle	.991	55	.944
Hip Flexion Moment	.953	55	.030
Hip Adduction Angle	.990	55	.928
Hip Adduction Moment	.986	55	.769
Hip Internal Rotation Angle	.983	55	.626
Hip Internal Rotation Moment	.808	55	.000
Knee Flexion angle	.936	55	.006
Knee Flexion Moment	.980	55	.506
Knee Valgus Angle	.986	55	.790
Knee Valgus Moment	.954	55	.033
Knee Internal Rotation Angle	.984	55	.683
Ankle Dorsiflexion Angle	.975	55	.293
Ankle Dorsiflexion Moment	.978	55	.422
Vertical GRF	.959	55	.058

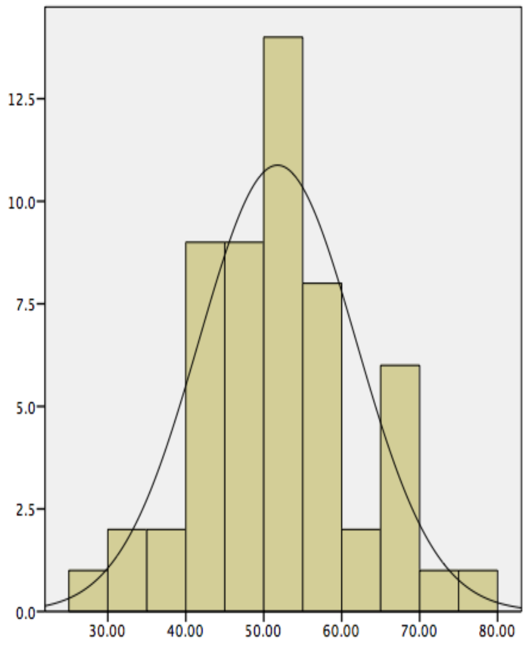
Tests of Normality for CUT variables in Female Participants

	Shapiro-Wilk		
	Statistic	df	Sig.
Hip Flexion Angle	.983	35	.849
Hip Flexion Moment	.883	35	.001
Hip Adduction Angle	.975	35	.593
Hip Adduction Moment	.965	35	.320
Hip Internal Rotation Angle	.983	35	.863
Hip Internal Rotation Moment	.792	35	.000
Knee Flexion angle	.992	35	.995
Knee Flexion Moment	.991	35	.993
Knee Valgus Angle	.979	35	.737
Knee Valgus Moment	.947	35	.089
Knee Internal Rotation Angle	.969	35	.422
Ankle Dorsiflexion Angle	.975	35	.597
Ankle Dorsiflexion Moment	.918	35	.013
Vertical GRF	.952	35	.133

Hip Flexion Angle During CUT Task

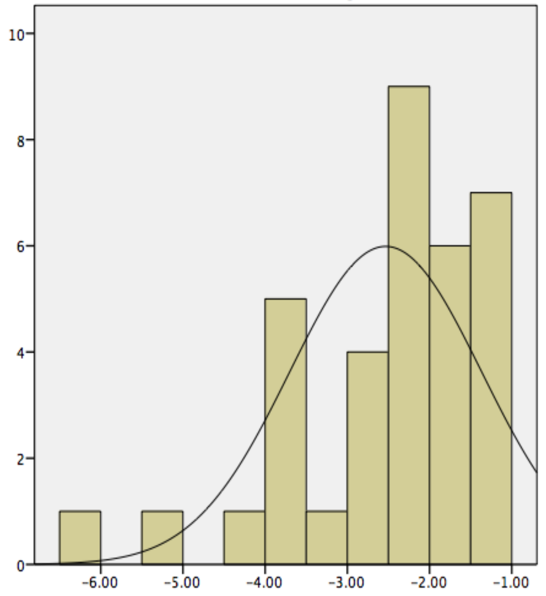
Females	Males

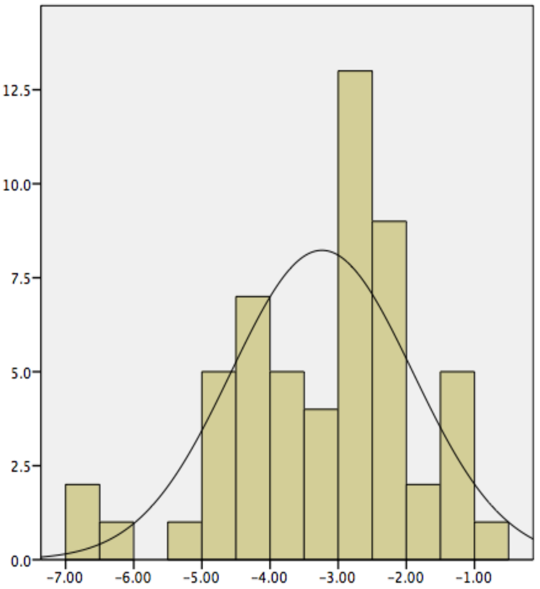




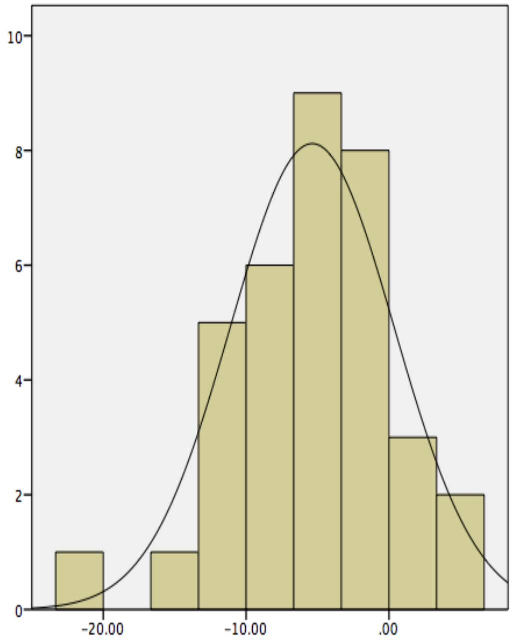
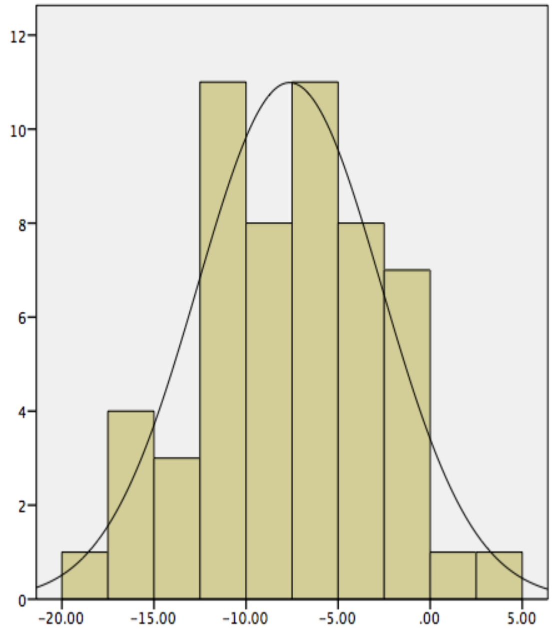
Hip Flexion Moment During CUT Task

Females	Males

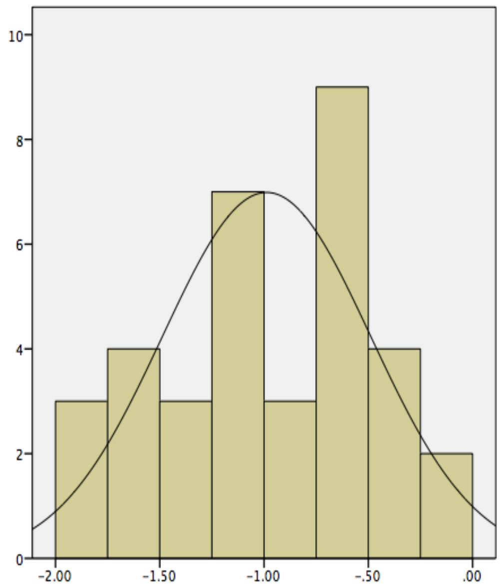
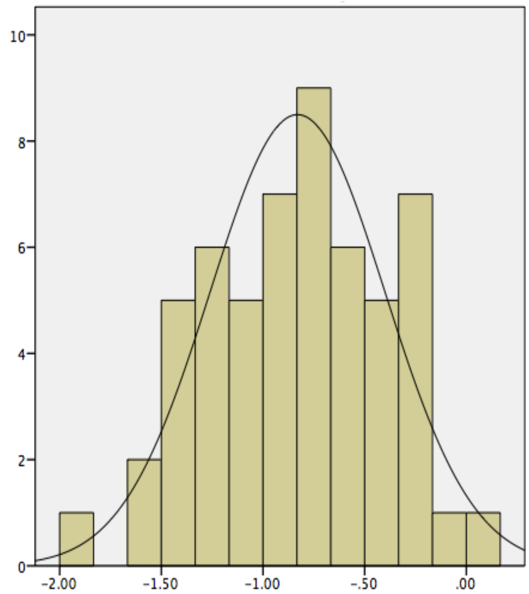




Hip Adduction Angle During CUT Task

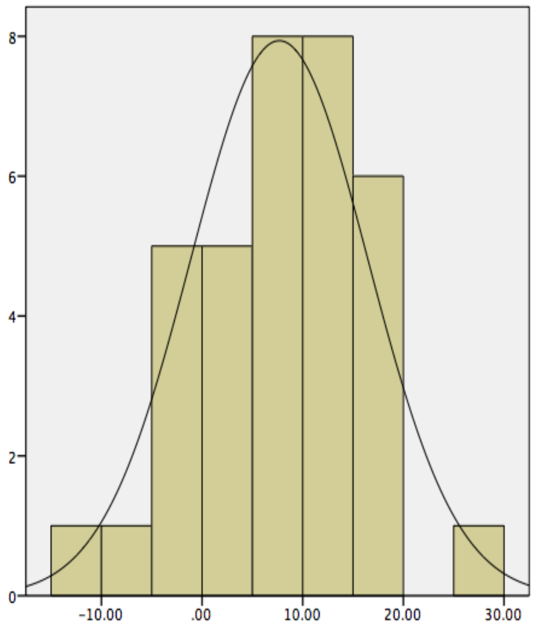
Females	Males
	

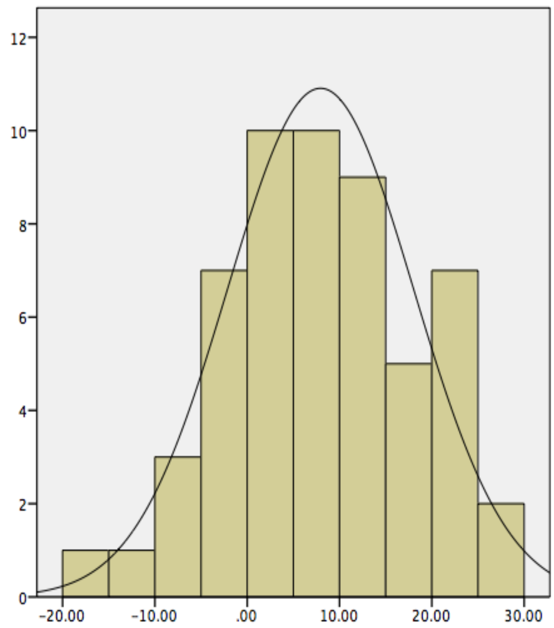
Hip Adduction Moment During CUT Task

Females	Males
	

Hip Internal Rotation Angle During CUT Task

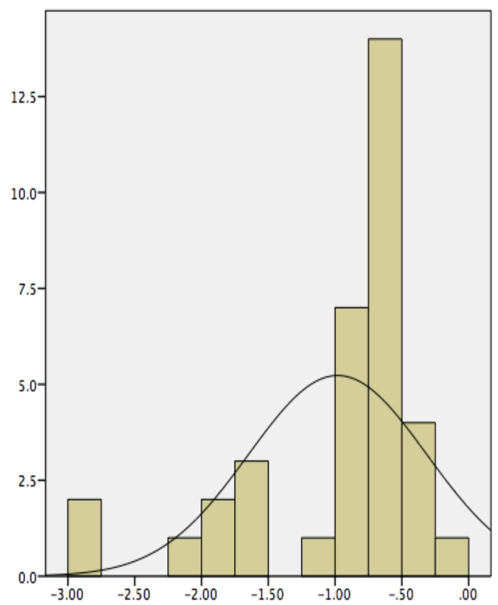
Females	Males

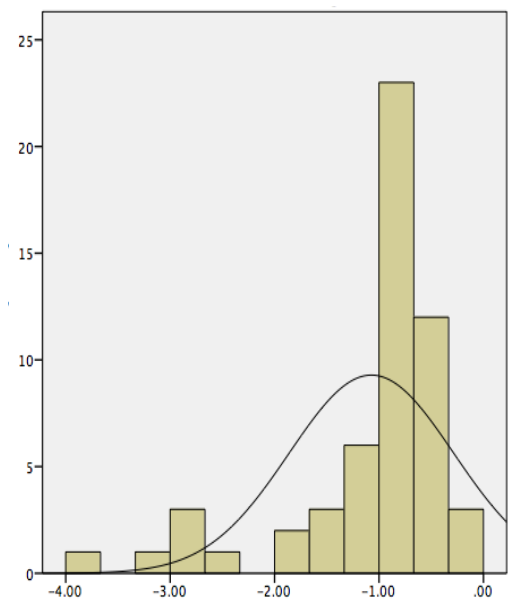




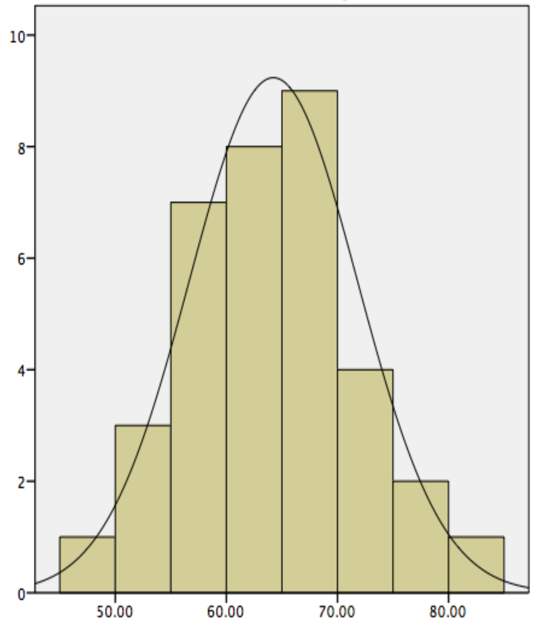
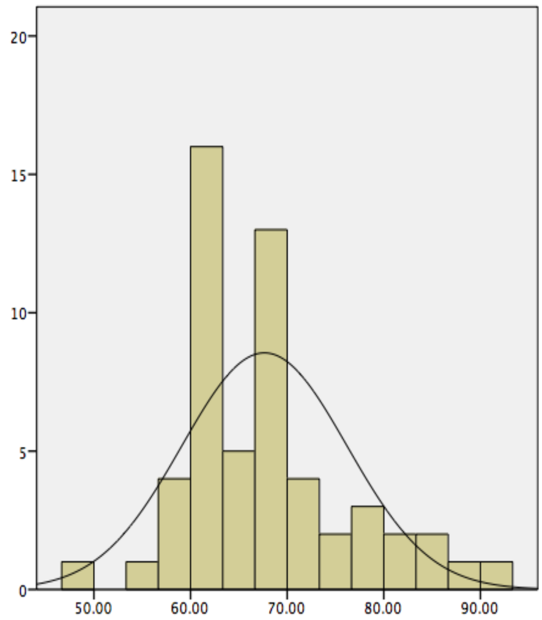
Hip Internal Rotation Moment During CUT Task

Females	Males

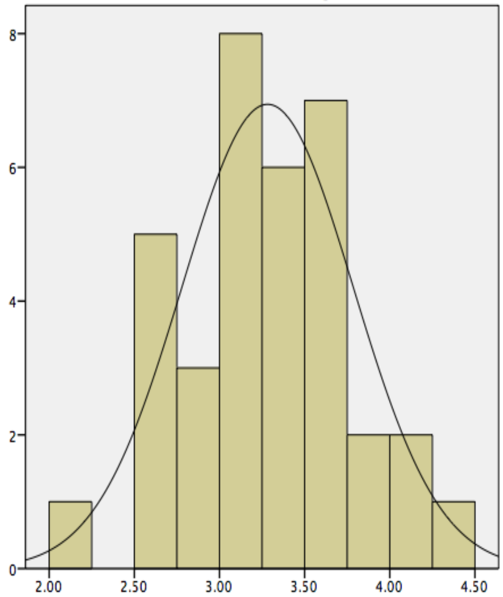
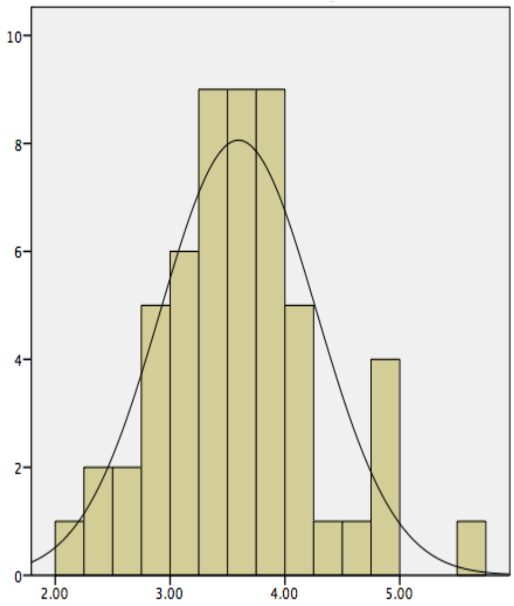




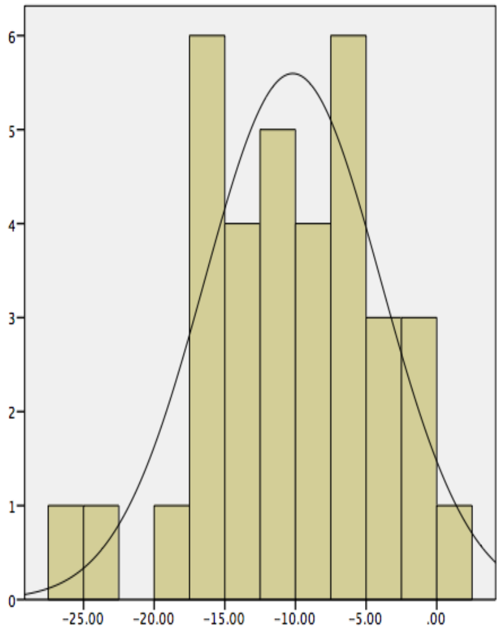
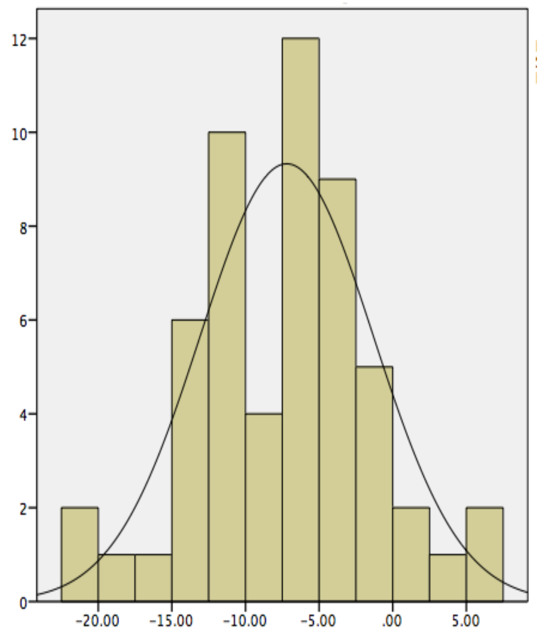
Knee Flexion Angle During CUT Task

Females	Males
	

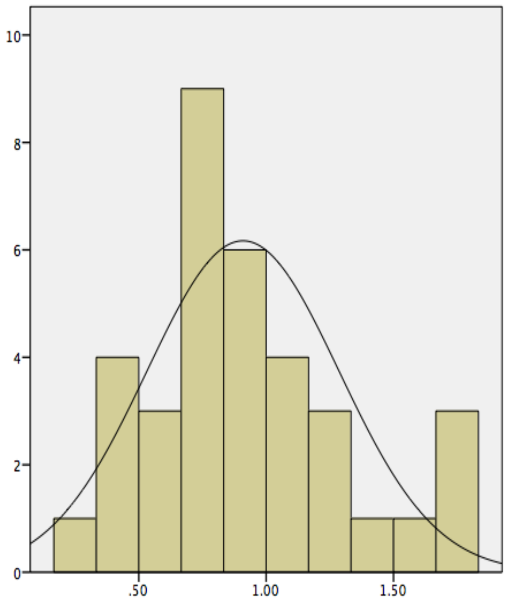
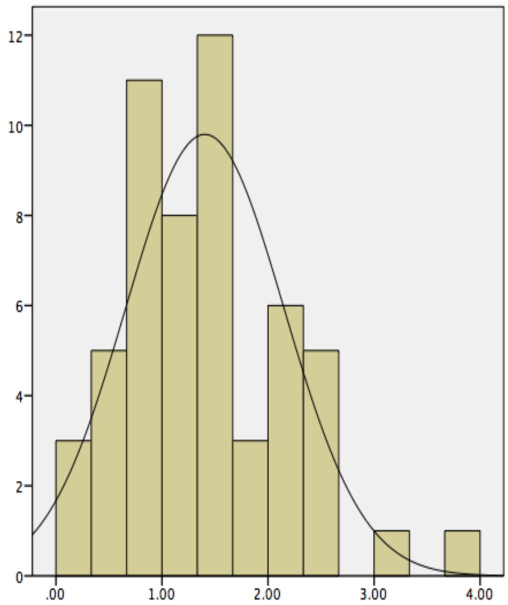
Knee Flexion Moment During CUT Task

Females	Males
	

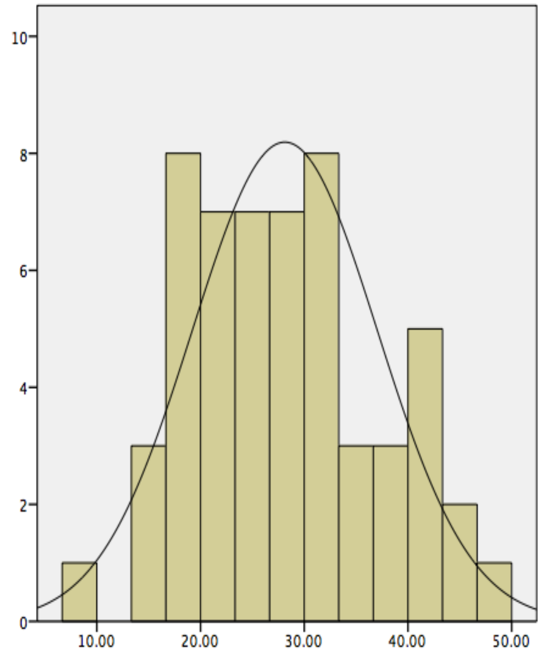
Knee Valgus Angle During CUT Task

Females	Males
	

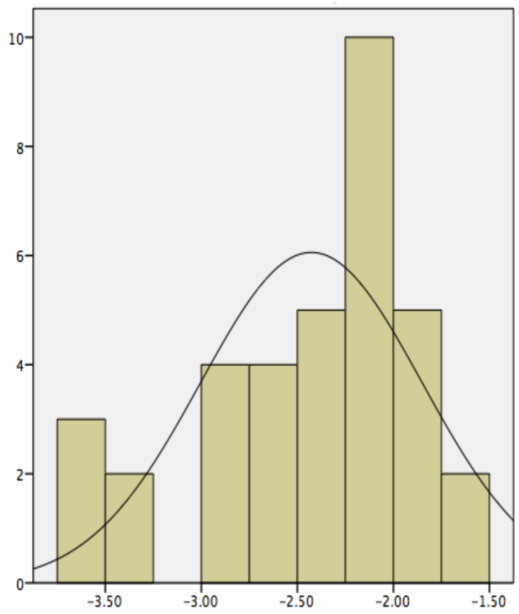
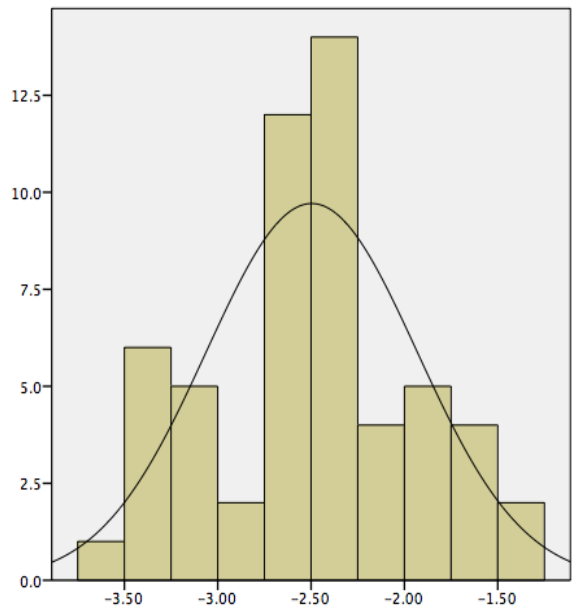
Knee Valgus Moment During CUT Task

Females	Males
	

Ankle Dorsiflexion Angle During CUT Task

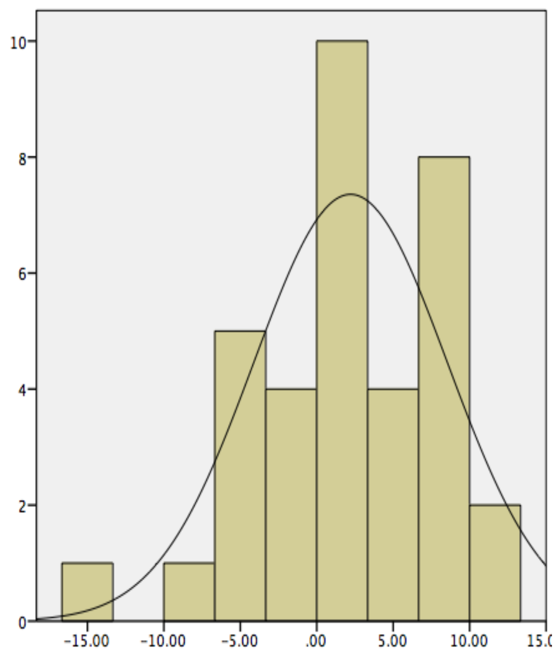
Females	Males
	

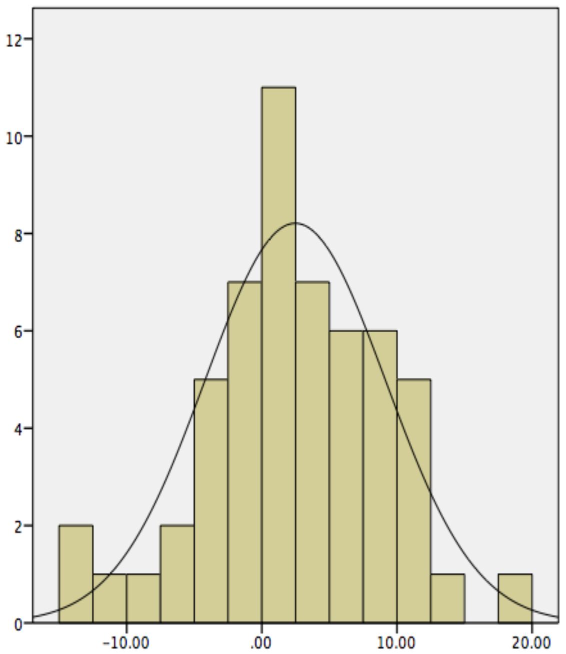
Knee Valgus Moment During CUT Task

Females	Males
	

Knee Internal Rotation Angle During CUT Task

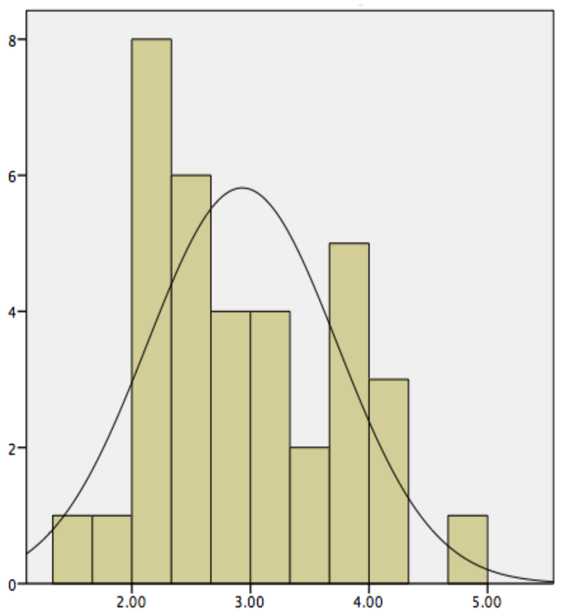
Females | Males

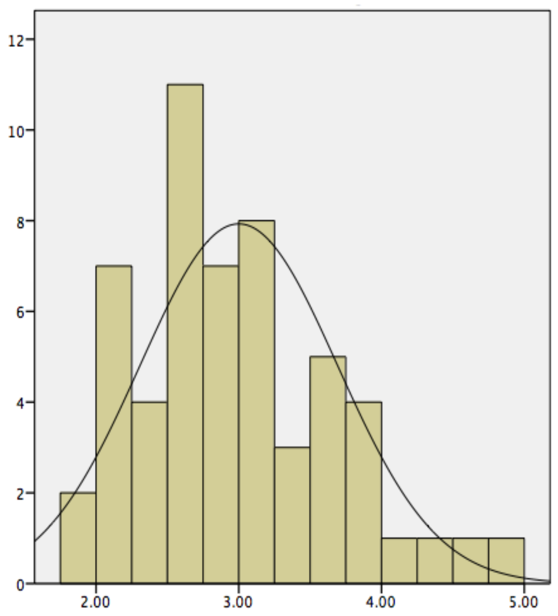




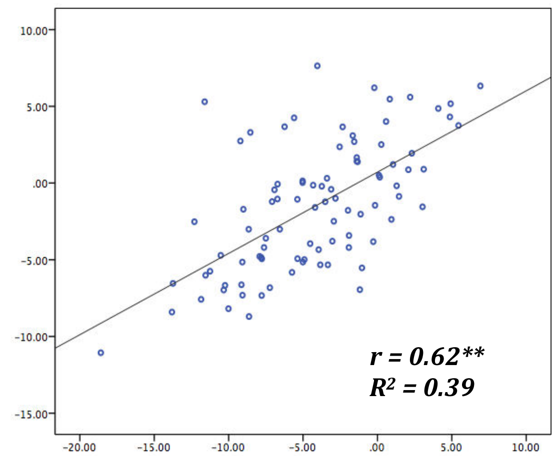
Vertical Ground Reaction Force During CUT Task

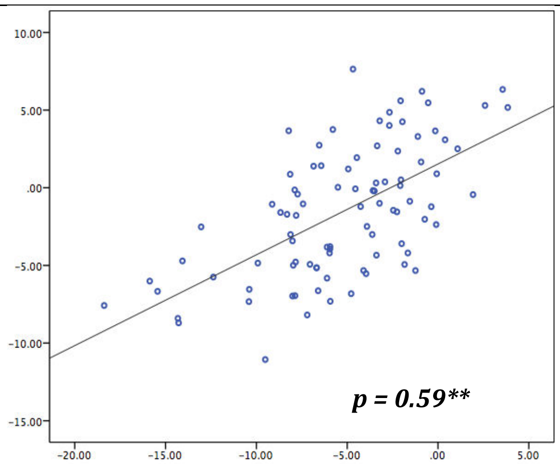
Females | Males





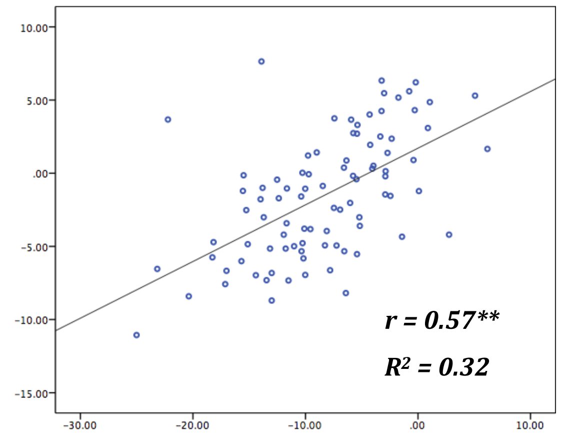
Appendix (D)

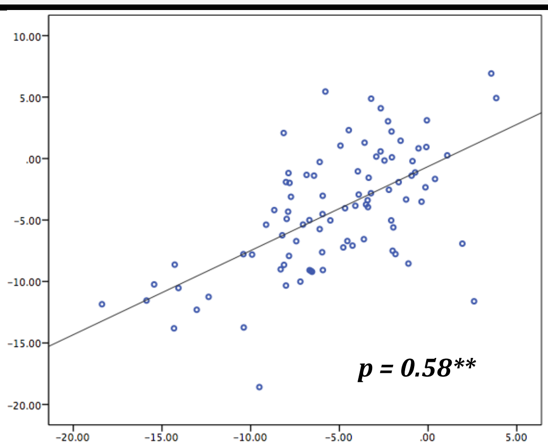




SLS vs. SLL

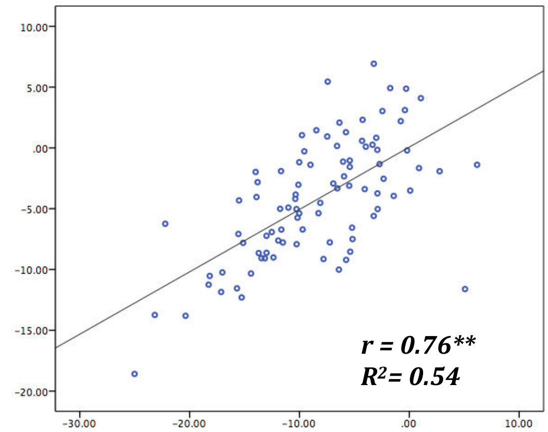
SLS vs. RUN

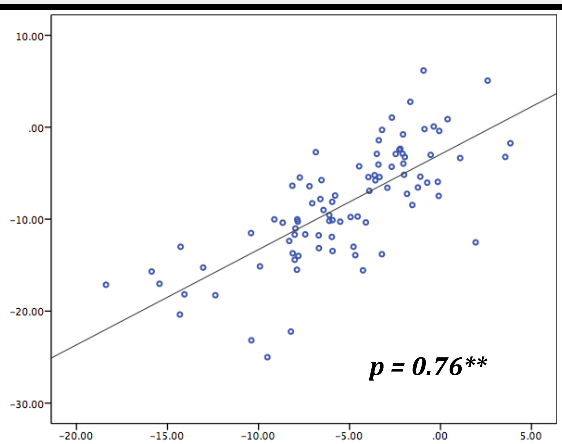




SLS vs. CUT

SLL vs. RUN

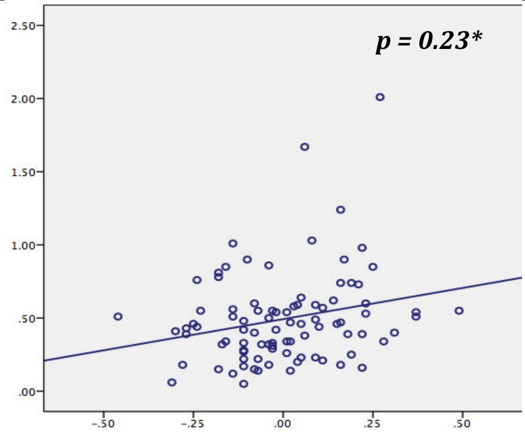


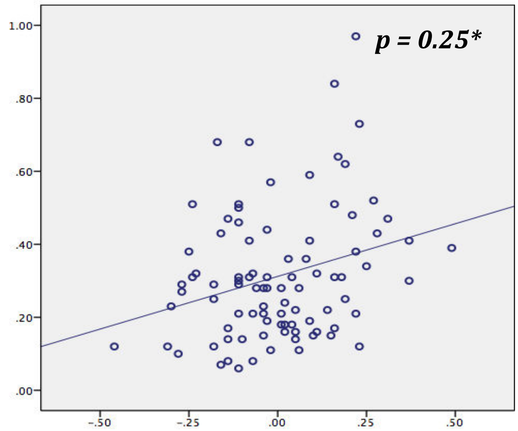


SLL vs. CUT

RUN vs. CUT

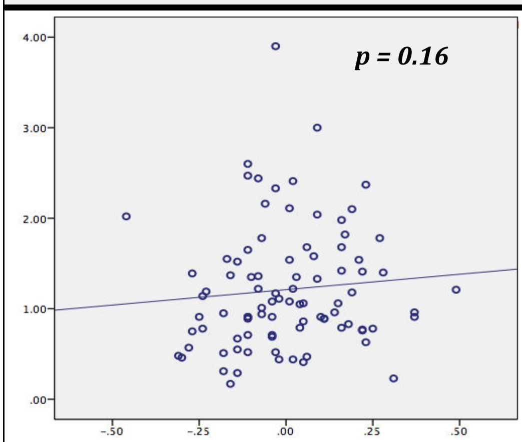
Appendix D-1. Scatterplot illustrating the relationship in Knee valgus angles between SLS, SLL, RUN and Cut tasks

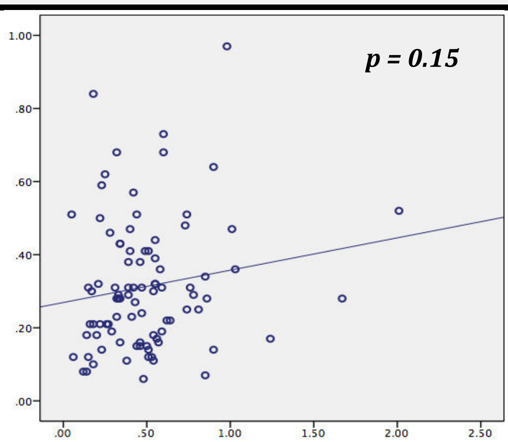




SLS vs. SLL

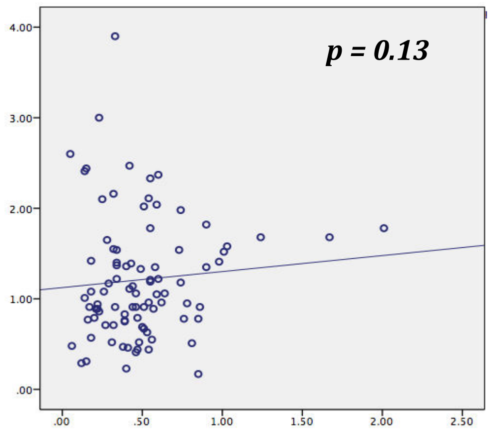
SLS vs. RUN

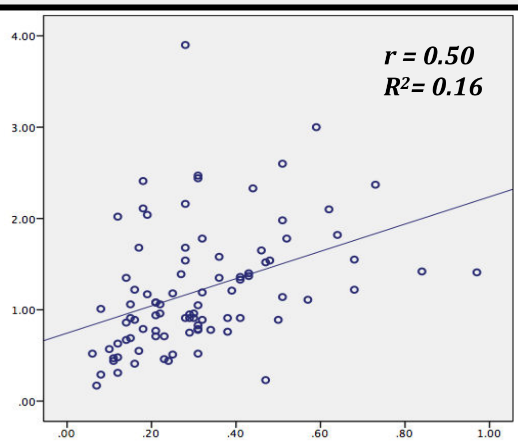




SLS vs. CUT

SLL vs. RUN

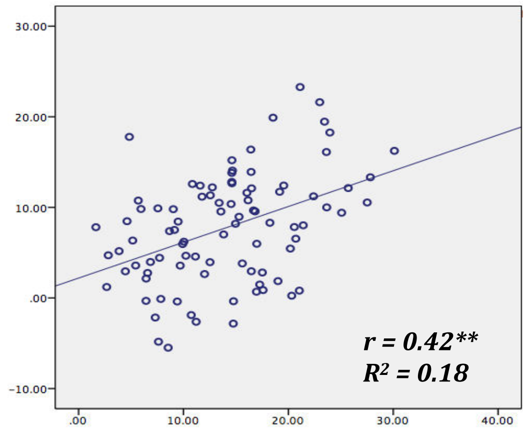


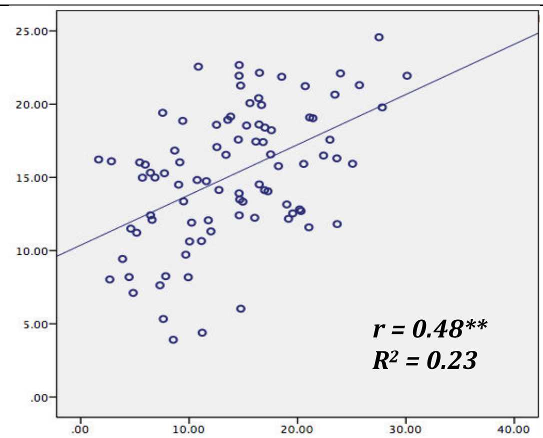


SLL vs. CUT

RUN vs. CUT

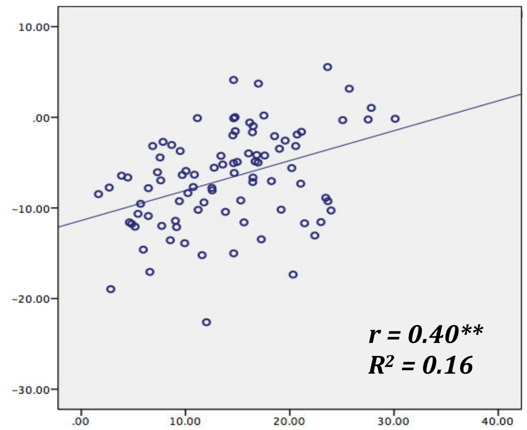
Appendix D-2. Scatterplot illustrating the relationship in Knee valgus moment between SLS, SLL, RUN and Cut tasks

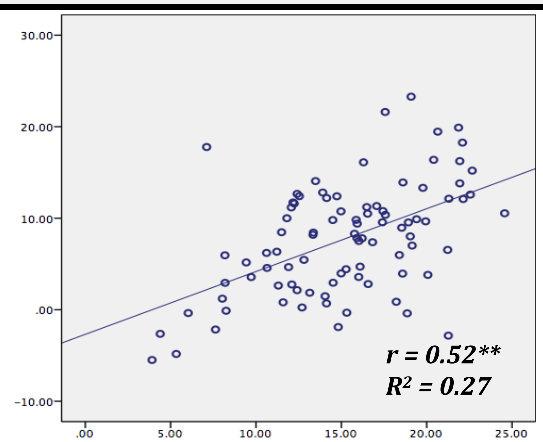




SLS vs. SLL

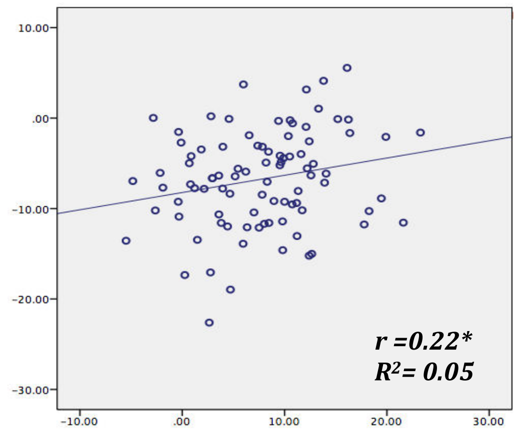
SLS vs. RUN

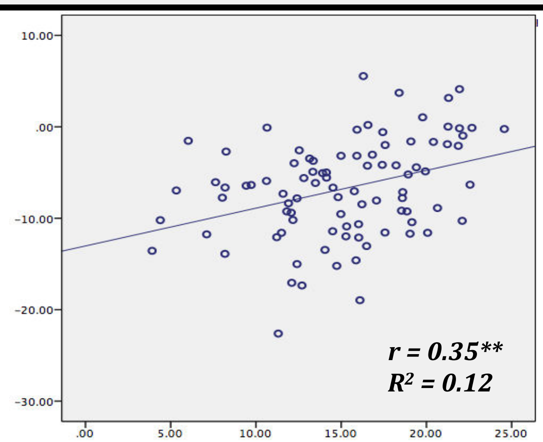




SLS vs. CUT

SLL vs. RUN

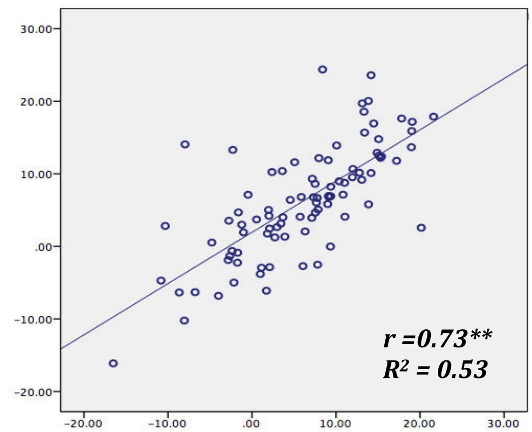


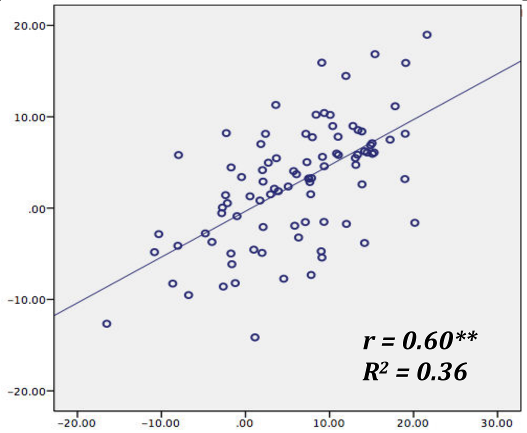


SLL vs. CUT

RUN vs. CUT

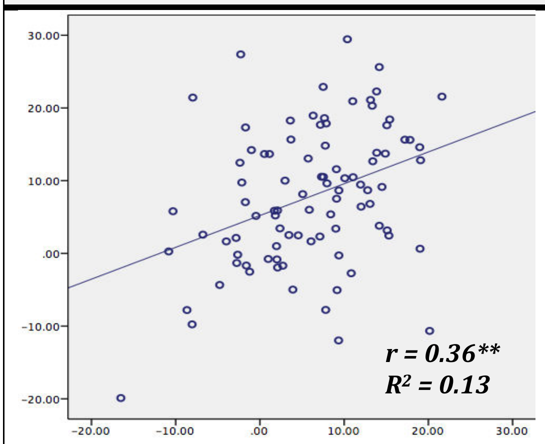
Appendix D-3. Scatterplot illustrating the relationship hip adduction angles between SLS, SLL, RUN and Cut tasks

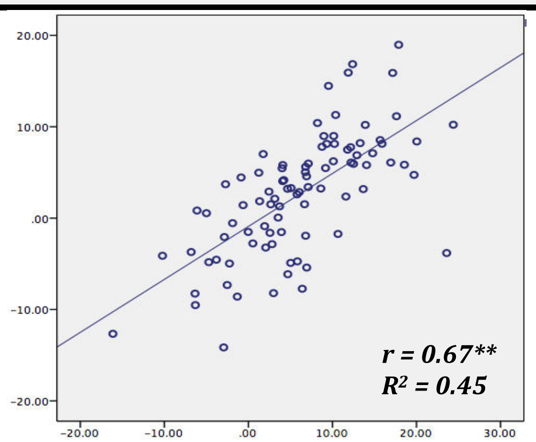




SLS vs. SLL

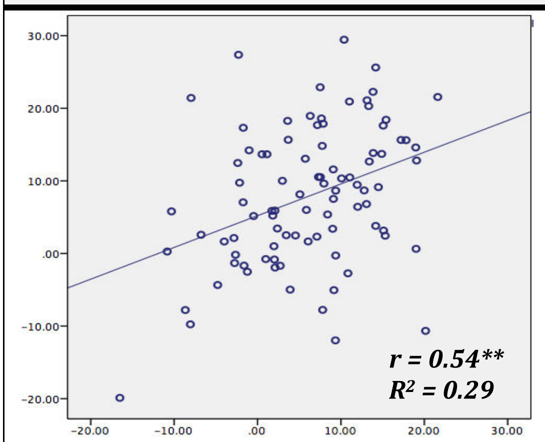
SLS vs. RUN

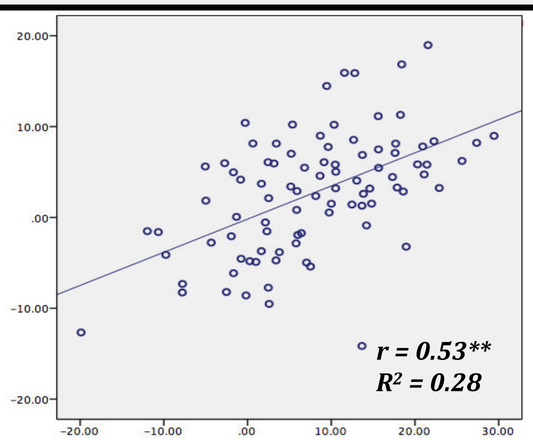




SLS vs. CUT

SLL vs. RUN





SLL vs. CUT

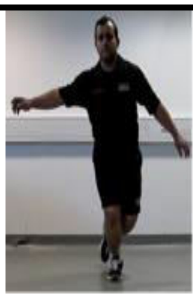
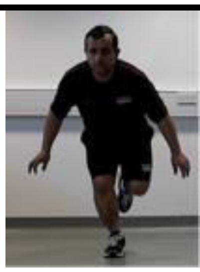
RUN vs. CUT

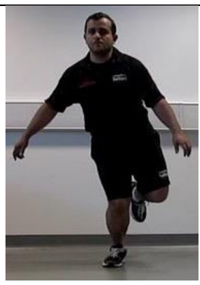
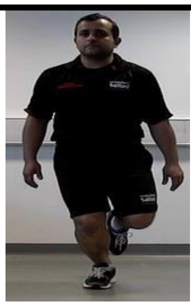
Appendix D-4. Scatterplot illustrating the relationship in hip internal rotation angles between SLS, SLL, RUN and Cut tasks

Appendix (E)

Appendix (E-1)

QASLS

Optimal	*Sub-optimal*	*Sub-optimal pictures*
<u>Arm Strategy:</u> *Arms stays relaxed by sides*	*Excessive arm movement to balance*	 
<u>Trunk Alignment:</u> *Trunk remains in neutral or slightly flexed position*	*Leaning in any direction*	 
<u>Pelvic Plane Goal:</u> *Pelvic maintains horizontal position, doesn't rotate relative to thigh*	*Loss of horizontal plane* *<u>OR</u>* *Excessive tilt or rotation*	 
<u>Thigh Motion Goal:</u> *WB thigh remains in neutral position, and NWB thigh remains parallel to WB thigh*	*Weight Bearing thigh moves into hip adduction* *<u>OR</u>* *Non weight bearing thigh not held in neutral*	 

Knee Position Goal: *Patella stays aligned over middle of foot*	*Patella pointing towards 2nd toe* *(Noticed valgus)* *<u>OR</u>* *Patella pointing past inside of foot* *(Significant valgus)*	 
Steady Stance Goal: *Stance leg is held still for 3 seconds and NWB doesn't touch down*	*Touches down with NWB foot* *<u>OR</u>* *Stance leg wobbles noticeably*	
Optimal		

Appendix E-2
(Feedback Study)
Tests of Normality for SLS Task

	Shapiro-Wilk		
	Statistic	DF	Sig.
SLS_1st session_knee _valgus angle	.949	11	.629
SLS_2nd session_knee_valgus_angle	.987	11	.992
SLS_3rd session_knee_valgus_angle	.874	11	.087
SLS_1st session_knee_flexion_angle	.908	11	.229
SLS_2nd session_knee_flexion_angle	.925	11	.364
SLS_3rd session_knee_flexion_angle	.953	11	.680
SLS_1st session_knee_valgus_moment	.955	11	.706
SLS_2nd session_knee_valgus_moment	.790	11	.007
SLS_3rd session_knee_valgus_moment	.951	11	.658
SLS_1st session_hip_adduction_angle	.936	11	.473
SLS_2nd session_hip_adduction_angle	.826	11	.021
SLS_3rd session_hip_adduction_angle	.888	11	.130
SLS_1st session_hip_flexion_angle	.923	11	.346
SLS_2nd session_hip_flexion_angle	.948	11	.621
SLS_3rd session_hip_flexion_angle	.941	11	.537
SLS_1st session_hip_internal_rotation_angle	.914	11	.275
SLS_2nd session_hip_internal_rotation_angle	.931	11	.426
SLS_3rd session_hip_internal_rotation_angle	.938	11	.498
SLS_1st session_knee_internal_rotation_angle	.754	11	.002
SLS_2nd session_knee_internal_rotation_angle	.855	11	.050
SLS_3rd session_knee_internal_rotation_angle	.969	11	.880

Baseline (1st session), Post Feedback (2nd session), & Follow-up (3rd session)

Appendix E-3
(Feedback Study)
Tests of Normality for SLL Task

	Shapiro-Wilk		
	Statistic	DF	Sig.
SLS_1st session_knee _valgus angle	.944	11	.566
SLS_2nd session_knee_valgus_angle	.971	11	.898
SLS_3rd session_knee_valgus_angle	.924	11	.349
SLS_1st session_knee_flexion_angle	.900	11	.185
SLS_2nd session_knee_flexion_angle	.941	11	.533
SLS_3rd session_knee_flexion_angle	.928	11	.396
SLS_1st session_knee_valgus_moment	.949	11	.633
SLS_2nd session_knee_valgus_moment	.914	11	.269
SLS_3rd session_knee_valgus_moment	.849	11	.042
SLS_1st session_hip_adduction_angle	.896	11	.164
SLS_2nd session_hip_adduction_angle	.886	11	.125
SLS_3rd session_hip_adduction_angle	.968	11	.871
SLS_1st session_hip_flexion_angle	.944	11	.566
SLS_2nd session_hip_flexion_angle	.971	11	.898
SLS_3rd session_hip_flexion_angle	.924	11	.349
SLS_1st session_hip_internal_rotation_angle	.941	11	.529
SLS_2nd session_hip_internal_rotation_angle	.975	11	.933
SLS_3rd session_hip_internal_rotation_angle	.965	11	.831
SLS_1st session_knee_internal_rotation_angle	.722	11	.001
SLS_2nd session_knee_internal_rotation_angle	.789	11	.007
SLS_3rd session_knee_internal_rotation_angle	.946	11	.599

Appendix E-4

(Feedback Study)

Tests of Normality for RUN Task

	Shapiro-Wilk		
	Statistic	DF	Sig.
SLS_1st session_knee _valgus angle	.946	11	.597
SLS_2nd session_knee_valgus_angle	.972	11	.911
SLS_3rd session_knee_valgus_angle	.965	11	.827
SLS_1st session_knee_flexion_angle	.891	11	.143
SLS_2nd session_knee_flexion_angle	.883	11	.114
SLS_3rd session_knee_flexion_angle	.936	11	.479
SLS_1st session_knee_valgus_moment	.960	11	.776
SLS_2nd session_knee_valgus_moment	.918	11	.305
SLS_3rd session_knee_valgus_moment	.929	11	.398
SLS_1st session_hip_adduction_angle	.967	11	.858
SLS_2nd session_hip_adduction_angle	.957	11	.729
SLS_3rd session_hip_adduction_angle	.974	11	.924
SLS_1st session_hip_flexion_angle	.872	11	.082
SLS_2nd session_hip_flexion_angle	.905	11	.213
SLS_3rd session_hip_flexion_angle	.906	11	.220
SLS_1st session_hip_internal_rotation_angle	.941	11	.534
SLS_2nd session_hip_internal_rotation_angle	.895	11	.159
SLS_3rd session_hip_internal_rotation_angle	.935	11	.465
SLS_1st session_knee_internal_rotation_angle	.824	11	.019
SLS_2nd session_knee_internal_rotation_angle	.852	11	.045
SLS_3rd session_knee_internal_rotation_angle	.876	11	.091

Appendix E-5

(Feedback Study)

Tests of Normality for CUT Task

	Shapiro-Wilk		
	Statistic	DF	Sig.
SLS_1st session_knee _valgus angle	.958	11	.744
SLS_2nd session_knee_valgus_angle	.949	11	.630
SLS_3rd session_knee_valgus_angle	.978	11	.955
SLS_1st session_knee_flexion_angle	.988	11	.995
SLS_2nd session_knee_flexion_angle	.984	11	.986
SLS_3rd session_knee_flexion_angle	.976	11	.939
SLS_1st session_knee_valgus_moment	.932	11	.427
SLS_2nd session_knee_valgus_moment	.845	11	.037
SLS_3rd session_knee_valgus_moment	.872	11	.083
SLS_1st session_hip_adduction_angle	.863	11	.063
SLS_2nd session_hip_adduction_angle	.913	11	.262
SLS_3rd session_hip_adduction_angle	.978	11	.955
SLS_1st session_hip_flexion_angle	.918	11	.299
SLS_2nd session_hip_flexion_angle	.907	11	.224
SLS_3rd session_hip_flexion_angle	.925	11	.366
SLS_1st session_hip_internal_rotation_angle	.950	11	.643
SLS_2nd session_hip_internal_rotation_angle	.922	11	.335
SLS_3rd session_hip_internal_rotation_angle	.941	11	.531
SLS_1st session_knee_internal_rotation_angle	.950	11	.643
SLS_2nd session_knee_internal_rotation_angle	.831	11	.024
SLS_3rd session_knee_internal_rotation_angle	.938	11	.493

Appendix E-6

(Feedback Study)

Within-day Means & SEMs values for 3D variables during SLS, SLL, RUN, CUT task in females participants (n=8)

Variables	SLS			SLL			RUN			CUT		
	Mean	**SEM**	**SDD**	**Mean**	**SEM**	**SDD**	**Mean**	**SEM**	**SDD**	**Mean**	**SEM**	**SDD**
						Joint Angles (°)						
Hip ADD	19.61	1.50	5.29	8.80	1.75	4.85	17.3	1.99	5.51	-7.15	3.37	9.14
Hip Flexion	66.50	3.56	12.41	49.2	1.83	5.07	54.7	5.14	14.2	48.4	2.49	6.90
Hip Int. Rot	6.08	2.86	7.92	7.14	3.13	8.67	2.54	2.46	6.81	6.84	3.81	10.56
Knee Valgus	-6.04	1.70	4.82	-5.84	1.71	4.73	-7.04	0.98	2.71	-11.8	1.73	4.79
Knee Flex	90.54	2.56	6.40	70.2	2.62	7.26	53.5	3.68	10.2	66.2	2.04	5.65
Knee Int. Rot	3.75	2.05	5.68	5.21	2.95	8.17	5.25	2.84	7.87	7.31	2.71	7.51
Dorsiflexion	42.83	1.18	3.27	28.6	1.10	3.04	33.1	1.98	5.48	30.9	2.24	6.20
						Moments (Nm/kg)						
Hip ADD	-1.08	0.06	0.16	-1.84	0.58	1.60	-2.38	0.39	1.08	-0.76	0.22	0.60
Hip Flex	-0.75	0.09	0.24	-2.25	0.28	0.77	-2.84	0.44	1.21	-2.70	0.27	0.74
Knee valgus	0.16	0.09	0.19	0.64	0.18	0.49	0.36	0.07	0.19	1.43	0.18	0.49
Knee Flex	1.95	0.09	0.24	3.35	0.09	0.24	2.63	0.22	0.60	3.30	0.16	0.44
Dorsi-Flex	-1.08	0.13	0.36	-2.41	0.24	0.66	-3.06	0.15	0.41	-2.46	0.14	0.38
VGRF (*bw)	1.13	0.02	0.05	4.36	0.12	0.33	2.69	0.14	0.38	3.09	0.18	0.49

SED= Standard Error of Measurement, SDD= Smallest Detectable Difference; ADD= Adduction Int-Rot= Internal Rotation; FLEX= Flexion

Appendix E-7

(Feedback Study)

Between-day Means & SEMs values for 3D variables during SLS, SLL, RUN, CUT task in females participants (n=8)

Variables	SLS			SLL			RUN			CUT		
	Mean	SEM	SDD	Mean	SEM	SDD	Mean	SEM	SDD	Mean	SEM	SDD
Joint Angles (°)												
Hip ADD	17.89	0.90	4.21	7.90	2.38	6.59	17.14	2.49	6.90	-7.84	3.02	8.37
Hip Flexion	67.63	3.74	15.02	49.77	3.77	10.44	55.39	4.74	13.13	49.19	4.98	13.80
Hip Int. Rot	6.58	3.29	9.11	6.48	5.20	14.41	3.03	3.08	8.53	6.51	5.15	14.27
Knee Valgus	-5.59	2.40	5.04	-6.15	3.60	9.97	-7.23	2.41	6.68	-11.6	3.02	8.37
Knee Flex	90.37	2.03	9.64	70.07	2.92	8.98	53.71	3.23	8.95	65.9	4.16	11.53
Knee Int. Rot	3.15	2.58	7.15	4.64	3.97	11.00	3.47	3.62	10.03	5.48	4.09	11.33
Dorsiflexion	43.19	1.11	3.07	28.40	1.26	3.49	33.09	2.42	6.70	30.24	3.82	10.58
Moments (Nm/kg)												
Hip ADD	-1.09	0.13	0.36	-2.03	0.24	0.66	-2.36	0.30	0.83	-0.81	0.13	0.36
Hip Flex	-0.79	0.18	0.49	-2.43	0.51	1.41	-2.84	0.38	1.05	-2.91	0.56	1.55
Knee valgus	0.15	0.08	0.22	0.59	0.22	0.60	0.35	0.09	0.24	1.40	0.20	0.55
Knee Flex	1.96	0.06	0.16	3.37	0.12	0.33	2.67	0.25	0.69	3.25	0.18	0.49
Dorsi-Flex	-1.06	0.11	0.30	-2.47	0.75	2.07	-3.04	0.14	0.38	-2.46	0.16	0.44
VGRF (*bw)	1.12	0.02	0.05	4.42	0.20	0.55	2.66	0.18	0.49	3.08	0.28	0.77

SED= Standard Error of Measurement, SDD= Smallest Detectable Difference; ADD= Adduction Int-Rot= Internal Rotation; FLEX= Flexion