The effect of gender and load type on lower extremity biomechanics during jump landings

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This work is dedicated to the memory of Ga – I'm sure I bored you to death explaining the project to you. Thank you for your unconditional kindness and love. Your vivacious character and sense of humour regardless of circumstance are qualities I always admired. You are so dearly missed.

Abstract

Introduction: Both gender and additional load have been shown to affect landing mechanics. Females are at an increased risk of sustaining an anterior cruciate ligament injury and patellofemoral pain syndrome, whereas males are at a greater risk of developing patellar tendinopathy. There is a paucity of literature examining the effects of load position or the comparison between genders during jump landings with additional load.

Purpose: To examine the effects of different positions of loading on lower extremity biomechanics during a jump-landing task. An additional purpose was to compare lower extremity biomechanics between genders during loaded and unloaded jump landings.

Method: Twenty-four resistance trained males (n = 12, age 21.1 ± 1.4 years, body mass 76.2 ± 10.3 kg, height 1.77 ± 0.08 m) and females (n = 12, age 20.3 ± 1.4 years, body mass 64.4 ± 7.2 kg, height 1.70 ± 0.03 m) were recruited. Three-dimensional lower-limb-joint kinematics and kinetics were measured during 5 bilateral maximal jumps were performed in a randomised order in each of four conditions: unloaded (UL), holding dumbbells (DB), wearing a weighted vest (WV), and with a barbell placed across the shoulders (BB). All loaded conditions were performed with 10% of body weight. A two-way analysis of variance (type of load * gender) was performed on kinetic and kinematic variables. Cohen's d effect sizes were calculated for differences between load types.

Results: Significant differences were shown between genders with the male group exhibiting increased jump height, lower time to peak knee flexion, smaller sagittal plane excursion (ROM), and smaller hip adduction angles at ground contact (IC) and smaller peak hip adduction angles when compared to females. Males were also shown to have significantly greater peak vertical ground reaction force (vGRF), significantly smaller time to peak vGRF, and significantly greater peak knee and hip extension moments. There was no significant interaction between load and gender in all variables measured. All loaded jumps resulted in a significant increase in the time to reach peak knee flexion. Significant decreases were observed in peak hip adduction moment and peak knee valgus moment in the male BB condition and female BB condition respectively when compared to UL jump landings. There was a significant decrease exhibited in vGRF in both genders in the BB condition when compared to the UL condition.

Conclusion: Both gender and additional loading alter landing biomechanics in maximal vertical jump landings. Males and females responded similarly to each of the loaded conditions. Findings suggest use of the BB may decrease the risk of injury relative to DB and WV during maximal countermovement jumps in both genders.

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Abbreviations

ACL Anterior cruciate ligament

BB Barbell loaded jump

COM Centre of mass

DB Dumbbell loaded jump

EMG Electromyography

IC Initial contact

PFPS Patellofemoral pain syndrome

PT Patellar tendinopathy

ROM Range of motion

TSM Total support moment

UL Unloaded jump

vGRF Vertical ground reaction force

VL Vastus lateralis

VMO Vastus medialis oblique

WV Weighted vest loaded jump

Chapter One: Introduction

1.1 Rationale

Jumping and therefore landing manoeuvres commonly occur in a variety of sports and are used as part of many training methods (Lees & Nolan, 1998; Hori, Newton, Kawamori, McGuigan, Andrews, Chapman & Nosaka, 2008; Erculj, Blas & Bracic, 2010; de Villarreal, Requena, Izquierdo & Gonzalez-Badillo, 2013). Jump-landings have commonly been associated with many injuries such as patellar tendinopathy (Bisseling, Hof, Bredeweg, Zwerver & Mulder, 2008), patellofemoral pain syndrome (Boling, Padua, Marshall, Guskiewicz, Pyne & Beutler, 2009), anterior cruciate ligament (ACL) injury (Shimokochi & Shultz, 2008), and ankle sprains (Verhagen, Van der Beek, Bouter, Bahr & Van Mechelen, 2004). Existing literature has sought to link biomechanical and anatomical risk factors of landings with lower extremity injury. Over a period of 10 years, a study examining admittance to a sports injury clinic reported that the knee was the most commonly injured region with 38.9% of all injuries occurring at the knee (Majewski, Susanne & Klaus, 2006). More conservative incidence rates of knee injuries of 27.5% have been reported by another epidemiological study examining admittance to another sports injury clinic (Baquie & Brukner, 1997). Knee injuries can lead to permanent discontinuation of sport (Kujala, Taimela, Antti-Poika, Orava, Tuominen & Myllynen, 1995).

Existing literature has demonstrated higher injury incidence at the knee in females relative to males. This may be due to an adoption of a disadvantageous landing strategy involving a more extended landing posture, increased GRF, greater valgus moment and valgus angle, and higher knee extensor moments (Kernozek, Van Hoof, Torry, Cowley & Tanner, 2004; Pappas, Hagins, Sheikhzadeh, Nordin & Rose, 2007; Gehring, Melnyk & Gollhofer, 2009). Anatomical factors such as intercondylar notch width and Q-angle (Rizzo, Holler & Bassett, 2001; Elias, Wilson, Adamson & Cosgarea, 2004), and physiological factors such as hormonal differences (Shultz, Kirk, Johnson, Sander & Perrin, 2004) contribute to the discrepancies in injury rate. Studies examining landings with additional load have shown additional loading during landing affects landing strategies but have differences in methodology and findings (Kulas, Zalewski, Hortobagyi & DeVita, 2008; Kulas Hortobágyi & DeVita, 2010; Janssen, Sheppard, Dingley, Chapman & Spratford, 2012). Although females have been examined within one of the studies examining additional load, no direct comparison between genders using external load has been attempted and no previous research has examined the effect of different methods of performing loaded jumps (weighted vest, barbell, and dumbbell) on lower limb biomechanics. Due to the paucity of literature investigating jump landings with additional load and the potential devastating effects of knee injuries, further investigation is warranted.

1.2 Aims

The aims of this study were:

1. To evaluate the effects of different methods of loading (barbell, dumbbell, and weighted vest) on lower limb kinematics and kinetics during the landing phase of a countermovement jump.
2. To compare lower limb kinematics and kinetics during landing between genders in loaded and unloaded conditions.

1.3 Objectives

The objectives of this study include:

1. Using three-dimensional (3D) motion analysis coupled with data from a force plate to examine differences in landing biomechanics when different methods of loading are used.
2. Use 3D motion analysis coupled with data from a force plate to examine differences in landing biomechanics between genders in different conditions.

1.4 Hypotheses

H_1 – There will be significant differences in landing kinematics and kinetics between male and female participants

H_2 – There will be significant differences in landing kinematics and kinetics between loading conditions

H_3 – Females will exhibit greater changes to landing kinematics and kinetics in loaded jump conditions when compared to males

H_4 – Different loading positions will affect landing kinematics and kinetics by different magnitudes.

Chapter Two: Review of Literature

2.1 Anatomy of the knee

The knee is a modified synovial hinge joint that consists of three articulations: the tibia and femur (tibiofemoral joint), the femur and patella (patellofemoral joint), and the tibia and fibula (superior tibio-fibular joint). At the tibiofemoral joint, motion occurs in all three planes (Figure 1). The joint rotates in the sagittal plane by flexion and extension, the transverse plane by external rotation and internal rotation, and the frontal plane by valgus and varus (adduction and abduction). Motion can also occur by translation: anteriorly and posteriorly, medially and laterally, and by compression and distraction. Excessive joint loading at the knee in either three planes has the potential to damage internal structures.

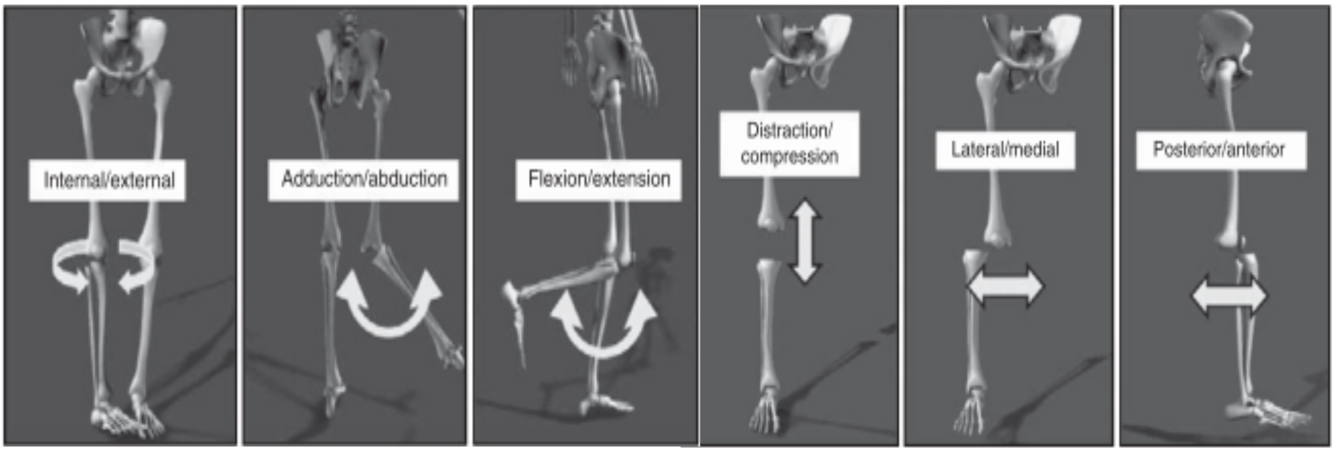

Figure 1. Rotations (left 3) and translations (right 3) at the knee joint (Quatman, Quatman-Yates & Hewett, 2010)

The motion of the knee is stabilised by ligaments including the anterior cruciate ligament (ACL), the posterior cruciate ligament, the lateral collateral ligament, and the medial collateral ligament (Fleming et al., 2001). The ACL provides approximately 85% of the resistance to anterior tibial translation (Russell, Palmeiri, Zinder & Ingersoll, 2006). The ACL is located within the intercondylar notch of the femur and inserts at the anterior intercondylar area of the tibia and the femur on the medial side of its lateral condyle (Figure 2). The ACL is comprised of two components: the anteromedial bundle (resists anterior translation of the tibia in extension) and the larger posterolateral bundle (resists anterior translation of the tibia in flexion) (Jacinda & Mandelbaum, 2007).

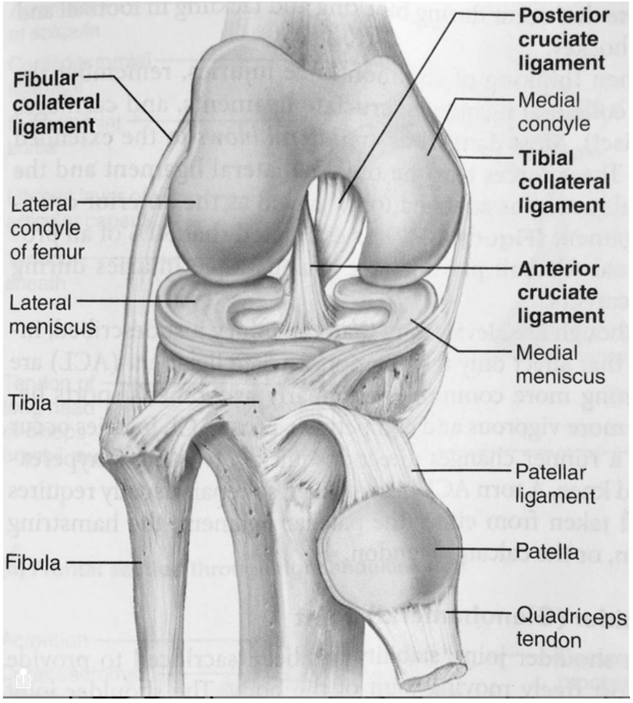

Figure 2. Anterior view of the knee (knee capsule and quadriceps tendon cut) (Marieb & Hoehn, 2007)

The patellofemoral joint is found at the anterior portion of the knee joint where the posterior aspect of the patella glides over the femoral trochlear or patella surface between the femoral condyles (Figure 3). The alignment of the patella is stabilised by the knee capsule, the rectus femoris via the quadriceps tendon, patella ligament, the medial and lateral patellar retinaculum, vastus medialis obliques (VMO), vastus lateralis (VL), and iliotibial band. The main role of the patella is to increase extension forces at the knee by increasing the distance between the quadriceps tendon and the sagittal axis of rotation of the knee.

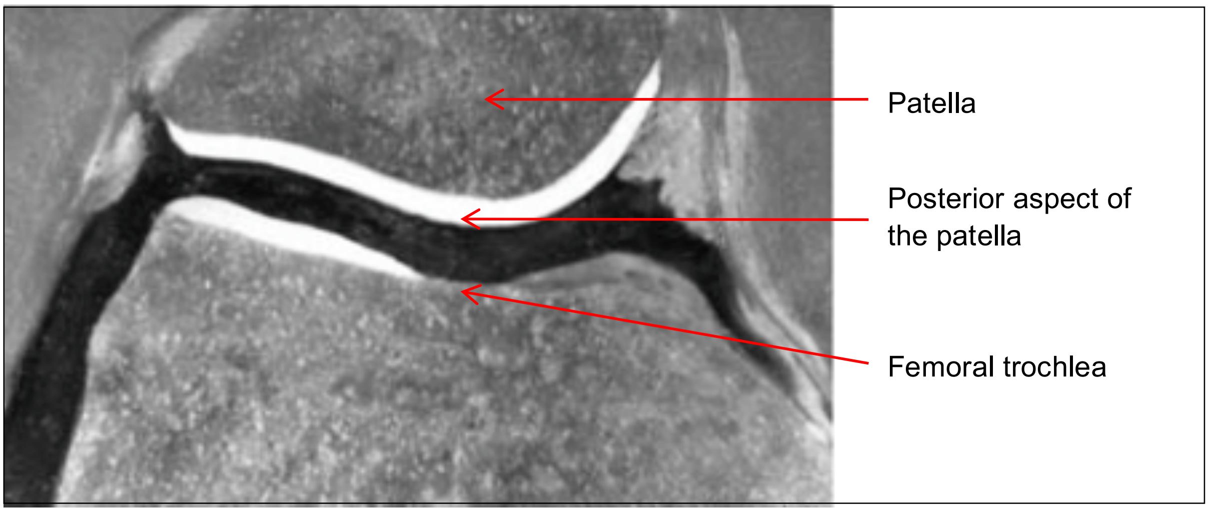

Figure 3. Axial cryosection of the right patellofemoral joint (Stäubli, Dürrenmatt, Porcellini & Rauschning, 1999)

2.2 Patellar Tendinopathy

Patellar tendinopathy (PT) is a common overuse injury, which is strongly associated with athletes involved in sports with a high frequency of jumping and landing. The prevalence of PT has been reported to be as high as 45% in elite volleyball players (Ferretti, Papandrea & Conteduca, 1990) and 32% in elite basketball players (Lian, Engebretsen & Bahr, 2005). The etiology of PT is yet to be fully elucidated, but a combination of high forces acting on the extensor mechanism at the knee and high frequency of dynamic movements are thought to be the main factors involved (Stanish, Curwin & Rubinovich, 1985). This repeated strain on the patellar tendon beyond its tensile strength leads to cumulative microtrauma and degradation of the tendon (Archambault, Wiley & Bray, 1995). The exact cause of PT is disputed as previous studies have used participants with either a history of PT or participants who had PT at the time of testing (Richards, Ajemian, Wiley & Zernicke, 1996; Bisseling et al., 2008; Edwards, Steele, McGhee, Beattie, Purdam & Cook, 2010). It could be argued that participants with a history of PT or that manifest symptoms of PT employ different landing strategies to mitigate symptoms, such as greater amounts of knee flexion, which skew results. It has been suggested that there is an increased risk of PT to male athletes (Vries, Worp, Diercks, Akker-Scheek & Zwerver, 2015), athletes with a higher training volume (Janssen, Steele, Munro & Brown, 2015), and athletes with a greater BMI (Crossley, Thancanamootoo, Metcalf, Cook, Purdam & Warden, 2007) Further prospective studies are required in order to determine which biomechanical risk factors are associated with PT.

To assess relationships between the lower extremity and PT, Richards et al. (1996) examined ankle and knee moments coupled with vertical ground reaction forces (GRF) during landing from volleyball spike jumps. Ten participants took part, 3 of which had patellar tendon pain associated with PT. It was concluded that patellar tendon pain was significantly associated with less knee flexion at the point of maximum vertical GRF, a stiffer landing technique, greater knee extensor moments, and increased plantarflexion. It was suggested that reduced knee flexion decreased the amount of shock absorption at the knee. Bisseling et al. (2008) presented similar findings that showed athletes with previous symptoms of PT demonstrated smaller knee joint flexion angles on initial contact and higher knee extension moments during the eccentric phase of landing. A similar study compared jump-landing technique in athletes with asymptomatic patellar tendon abnormality (a precursor to PT) and athletes with healthy tendons (Edwards et al., 2010). It was shown that athletes with asymptomatic patellar tendon abnormality landed with significantly reduced hip and knee flexion during drop-landings when compared to the healthy athletes. Although a more extended landing position has been associated with PT, the exact mechanisms remain equivocal, as it isn't clear whether the landing strategies employed are the cause or effect of PT.

2.3 Patellofemoral Pain Syndrome

Patellofemoral pain syndrome (PFPS), sometimes known as anterior knee pain or chondromalacia patellae syndrome, is a condition in which pain is identified on the anterior aspect of the knee and is aggravated by movements which increase compressive forces on the patellofemoral joint (Earl, Hertel & Denegar, 2005). Movements that increase compressive forces on the patellofemoral joint include jump-landings and running, which can exacerbate pain (Weiss & Whatman, 2015). The prevalence of PFPS has not been accurately evaluated but a commonly cited figure in the literature is 25%, which is supported only by studies on university level athletes and military recruits (Callaghan & Selfe, 2007).

In a cross-sectional prospective study seeking to examine potential risk factors for developing PFPS, 1597 Naval Midshipmen were tested using three-dimensional (3D) motion analysis, lower extremity isometric strength tests, and postural alignment tests (Q-angle and navicular drop) at the beginning of a 2.5 year prospective follow up (Boling et al., 2009). It was shown that the 40 participants who developed PFPS over the following 2.5 years demonstrated significantly less knee flexion during drop-landings when compared to uninjured participants (Boling et al., 2009). This finding suggests decreased knee flexion at landing is a risk factor for developing PFPS. This contradicts previous suggestions that decreased knee flexion is a compensatory mechanism in response to PFPS to decrease the amount of pressure on the patellofemoral joint (Crossley, Cowan, Bennell & McConnell, 2004).

In order for the patella to align with the femoral trochlea during movement, dynamic stability is provided by the quadriceps tendon, patella tendon, VMO, VL, and iliotibial band. Decreased knee flexion lessens the amount of dynamic stability of the alignment of the patellar in the frontal plain, leading to malalignment of the patella and increased pressure on the articular cartilage surfaces in the patellofemoral joint (MacIntyre, Hill, Fellows, Ellis & Wilson, 2006). Patella tracking (the movement of the patella within the femoral trochlea) may be affected by the magnitude and timing of the force produced by these stabilisers. The VMO is of significance as it is the only stabiliser to produce a medial force on the patella. Electromyography (EMG) onset of the VMO has been shown to be delayed in comparison to the vastus lateralis in PFPS patients in stair-stepping and a postural control task (Cowan, Bennell, Hodges, Crossley & McConnell, 2001; Cowan, Hodges, Bennell & Crossley, 2002). The changes in muscle recruitment patterns seen in both studies were suggested to be a protective mechanism as a result of pain. The results of Powers (2000a) may be due to a type II statistical error as the variability between participants with PFPS was 2 to 3 times higher than that of the control group. Further prospective research is necessary to expound the link between muscle activation patterns and PFPS.

In a study by Werner (1995) participants with PFPS were tested using a knee extension movement on an isokinetic dynamometer and demonstrated significantly lower knee extension torque than healthy controls. These findings were also demonstrated in prospective studies (Van Tiggelen, Witvrouw, Coorevits, Croisier & Roget, 2004; Duvigneaud, Bernard, Stevens, Witvrouw & Van Tiggelen, 2008) and suggest lower knee extensor moments are a risk factor for developing PFPS. The results of these studies suggest athletes with weaker quadriceps are at a higher risk of developing PFPS. Boling et al. (2009) showed that an increase in hip internal rotation, which is likely to be due to weakness of hip abductors and external rotators, is a biomechanical risk factor for developing PFPS. It has been reported that 30 degrees of femoral internal rotation can cause significant increase in contact pressures on the lateral facets of the patellar (Lee, Anzel, Bennett, Pang & Kim, 1994). Weakness of the gluteus medius and gluteus maximus have been suggested to increase hip internal rotation and increase the risk of developing PFPS. Souza and Powers (2009) examined 21 females with PFPS and 20 pain-free controls. It was found that females with PFPS exhibited weaker hip musculature when tested on a dynamometer (14% less isometric hip abductor strength and 17% less isometric hip extensor strength). These findings are supported by a study examining 13 female athletes with PFPS who demonstrated a 16% deficit in isometric hip extension torque when measured by a handheld dynamometer (Cichanowski, Schmitt, Johnson & Niemuth, 2007). As the gluteus maximus is the primary mover in hip extension and hip external rotation, a weakness of the hip extensors may lead to an increase in hip internal rotation (Lee & Powers, 2013). Further prospective studies are required to establish whether changes to hip musculature are a cause rather than a result of PFPS.

There are several conflicting findings regarding the relationship between foot posture and PFPS. One factor postulated to increase rotation of the femur is excessive foot pronation (Boling et al., 2009). Pronation may lead to excessive external tibial and internal femoral rotation, and misalignment of the patellar. Therefore, excessive pronation of the foot may be a risk factor for PFPS (Tiberio, 1987). Other studies examining the association between excessive foot pronation and PFPS have demonstrated no significant evidence to support the hypothesis that excessive foot pronation increases risk of PFPS (Powers, Maffucci &

Hampton, 1995; Witvrouw, Lysens, Bellemans, Cambier & Vanderstraeten, 2000; Thijs, De Clercq, Roosen & Witvrouw, 2008).

The quadriceps angle (Q angle) is commonly defined as the angle formed between a line from the anterior superior iliac spine to the patella center and a line from the tibial tuberosity to the patella center (Figure 4) (Livingston & Mandigo, 1996). An increased Q angle is thought to increase lateral pull of the quadriceps, leading to malalignment of the patella and abnormal distribution of force across the knee leading to PFPS (Elias et al., 2004). Research examining the influence of Q angle on PFPS have produced conflicting findings with some studies showing a relationship between Q angle and PFPS (Mizuno et al., 2001), whereas other investigations have found no significant link (Witvrouw, Lysens, Bellemans, Cambier & Vanderstraeten, 2000). There is also disagreement on the reliability of the measurement of Q angle and so more research is required to develop a standardised protocol (Dierks, Manal, Hamill & Davis, 2011).

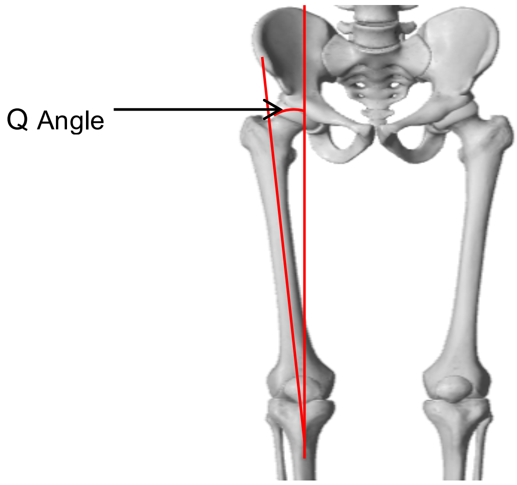

Figure 4. Measurement of Q-angle

2.4 Anterior Cruciate Ligament Injuries

It has been estimated that 80,000 ACL injuries occur annually in the United States of America (Griffin, Agel, Albohm, Arendt, Dick, Garrett, et al. 2000). There have been numerous studies aiming to elucidate potential mechanisms of excessive ACL loading using: retrospective interviews (Boden et al., 2000; Faunø & Wulff, 2006), observational analyses (Seward & McGivern, 2007; Krosshaug, Nakamae, Boden, Engebretsen, Smith, Slauterbeck & Bahr, 2007), computer simulations (McLean, Huang, Su, & Van Den Bogert, 2004), and measurements of ACL loading in vitro and in vivo (Dürselen, Claes & Kiefer, 1995; Li et al., 1999; DeMorat, Weinhold, Blackburn, Chudik & Garrett, 2004). Almost three quarters of ACL injuries are non-contact injuries and usually occur during sudden deceleration, landing, and pivoting manoeuvres (Boden, Feagin Jr. & Garrett Jr., 2000). There are several anatomical factors that have been proposed to increase the risk of ACL injury which are discussed in section 2.5.2.

A common risk factor associated with ACL injury cited by interviews and observational analyses of when ACL injury occurs is low levels of knee flexion (0-30°) or hyperextension during landing (Ferretti et al., 1990; Cochrane, Lloyd, Buttfield, Seward & McGivern, 2007; Krosshaug et al., 2007). The force produced by the quadriceps that loads the ACL is related to the patella tendon-tibia shaft angle (the angle between the patella tendon and the longitudinal axis of the tibia). In low levels of knee flexion, the patella tendon-tibia shift angle is increased, leading to increased strain on the ACL by quadriceps force via the patella tendon (Nunley, Wright, Renner, Yu & Garrett Jr., 2003). This is supported by both *in vivo* and *in vitro* studies that have shown that anterior tibial displacement is greater at shallow levels of knee flexion (Fukubayashi, Torzilli, Sherman & Warren, 1982; Daniel, Malcom, Losse, Stone, Sachs & Burks, 1985). As the ACL provides 85% of the total restraint to anterior tibial displacement, reduced knee flexion results in increased ACL loading (Markolf, Mensch & Amstutz, 1976). During landing movements, the quadriceps muscle produces anterior shear force at the proximal end of the tibia through the patellar tendon (Li, Rudy, Sakane, Kanamori, Ma & Woo, 1999). It has been hypothesised that the hamstring force may increase the anterior shear force at the tibia because of a reduction in the hamstring tendon-tibia shaft angle at reduced knee flexion (Lin, Liu, Gros, Weinhold, Garrett & Yu, 2012).

It has also been suggested that a more extended, or erect, landing technique prevents the musculature around the knee from dissipating landing forces, which increase the risk of ACL injury (Boden, Torg, Knowles & Hewett, 2009). As well as flexion at the knee, reduced flexion at the hip has been implicated with an increased risk of ACL injury (Hewett, Torg & Boden, 2009). It has been shown that landing with a greater amount of hip flexion reduces quadriceps force through the patella tendon, decreasing the load placed on the ACL from the quadriceps (Blackburn & Padua, 2009). Landing with a greater amount of hip flexion has also been shown to increase hamstring force production, which can decrease knee anterior shear force (Kulas et al., 2010).

A factor proposed to increase ACL load in the sagittal plane is a quadriceps dominant landing strategy (Shultz, Nguyen, Leonard & Schmitz, 2009). If the quadriceps exert a force greater than that of the hamstrings, this can lead to an increase in anteriorly directed shear force acting at the proximal end of the tibia (Withrow, Huston, Wojtys & Ashton-Miller, 2006).

It has also been shown that in isolation, excessive quadriceps force at low levels of knee flexion can lead to significant anterior tibial shear force and ACL injury in cadaveric knees (DeMorat et al., 2004). It should be noted that a quadriceps muscle force of 4500 N was used to load the ACL, which is considerably larger than the force used in comparable studies (Dürselen et al., 1995; Li et al., 1999). Hamstrings have been shown to counter ACL loading as a result of quadriceps forces (Li et al., 1999). Therefore, weak hamstrings or decreased coactivation of the hamstrings during quadriceps contraction may also induce increased loading of the ACL during landings (Li et al., 1999).

In addition to sagittal plane changes, there has been evidence that frontal plane mechanisms have a significant role in ACL ruptures, particularly valgus movement at the knee (Boden et al., 2009). A valgus movement of the knee involves femoral adduction, internal rotation of the femur, tibial external rotation, and ankle eversion or foot pronation (Shin, Chaudhari & Andriacchi, 2009). In a prospective cohort study involving 205 female athletes, it was shown that the 9 athletes who had an ACL rupture exhibited greater knee valgus angle and moment, increased knee valgus angle, and decreased knee flexion angles during jump-landings (Hewett et al., 2005). The association in this study between knee valgus alone and ACL rupture should be treated with caution. When taking in to consideration height and weight of the 9 athletes with ACL injuries, the valgus moments were less than 12.5 Nm (Yu & Garrett, 2007). In a majority of cadaver studies, knee valgus moment has been shown to not significantly affect ACL loading unless a proximal shear force is simultaneously applied to the tibia (Berns, Hull & Patterson, 1992; Markolf et al., 1995; Bendjaballah, Shirazi-Adl & Zukor, 1997). These results are in line with findings of retrospective interviews which report ACL-injured athletes recall the knee moving into valgus coupled with shallow knee flexion (Ferretti et al., 1990; McNair, Marshall & Matheson, 1990).

A factor proposed to be associated with increased valgus is a weakness of hip musculature (Jacobs & Mattacola, 2005). It has been shown that females with greater hip abductor strength demonstrate significantly less hip adduction (Jacobs & Mattacola, 2005). This is pertinent to ACL loading, as a reduction in hip adduction has been suggested to reduce knee valgus (Ireland, 1999). However, there are a number of studies that found no relationship between hip strength and knee valgus angle during dynamic tasks such as landing and cutting manoeuvres (Jacobs & Mattacola, 2005; Jacobs, Uhl, Mattacola, Shapiro & Rayens, 2007; Sigward, Ota & Powers, 2008). Further research examining the effects of hip strength on knee valgus angle and moment is required as the findings of existing studies are mixed due to discrepancies in methodology (Cashman, 2012).

2.5.1 Gender differences

When compared to males, female athletes are at greater risk of ACL injury or developing PFPS. There have been numerous studies attempting to elucidate the risk factors behind the discrepancy in injury incidence between males and females. Females have been reported to be 2 to 8 times more likely to sustain an ACL injury and over 2 times more likely to develop PFPS than males (Arendt & Dick, 1995; Boling, Padua, Marshall, Guskiewicz, Pyne & Beutler, 2010). However, it has been suggested that males are twice more likely to develop PT than females (Lian et al., 2005). While discussions of physiological factors are beyond the scope of this report it is worth noting that sex hormones, particularly oestrogen, progesterone, and relaxin, may play a role in increasing joint laxity, and therefore injury incidence, in female athletes (Shultz et al., 2004).

2.5.2 Anatomical differences between genders

It has been suggested that a larger Q angle may exacerbate the risk of injuries including PFPS (Elias et al., 2004). It has been shown that females have a significantly greater Q angle when compared to males (Horton & Hall, 1989). Therefore, females may be at a greater risk of injury due to increase Q angle. Due to discrepancies pertaining to the measurement of Q angle, it would be imprudent to draw conclusions based on a small number of studies. A theory that addresses ACL injury directly is that females have smaller intercondylar notch widths relative to the ACL (Rizzo et al., 2001). This may increase the probability of impingement of the ACL on the intercondylar notch. This has not been unequivocally determined due to differences in techniques of measurement. The width of the intercondylar notch is more likely to be determined by the height than gender, and varies in proportion to the size of the ACL (Fayad, Rosenthal, Morrison & Carrino, 2008). This counters the suggestion that a mismatch between ACL size and notch width is a cause of ACL tears in females. Another factor that has been postulated to exacerbate risk of ACL injury in females is the notion that ACL cross-sectional area, width, and volume have been shown to be larger in males than females (arc, Dome, Gautam, Awh & Rennirt, 2001). However, it has been shown that gender differences in ACL volume are strongly correlated to height differences between males and females (Fayad et al., 2008). Overall, the evidence linking anatomical factors and the risk of knee injury is weak.

2.5.3 Neuromuscular differences between genders

Dynamic neuromuscular control of musculature at the knee and hip when landing is likely to affect lower extremity alignment and therefore joint loading (Hewett, Lindenfeld, Riccobene, & Noyes, 1999). Males have been shown to have a peak hamstrings to quadriceps (H:Q) torque ratio which is greater when compared to females (Hewett, Stroupe, Nance & Noyes, 1996). The agonist of the ACL is the hamstring group, and the antagonist is the quadriceps. A lower H:Q ratio would lead to a diminished ability to resist anterior tibial shear force and an increased risk of ACL injury. The hamstring group also play a role in stabilising the knee in the frontal plane, and so decreased strength may lead to less dynamic control at the knee in the frontal plane (Lloyd & Buchanan, 2001).

Females have been shown to exhibit lower activation of the gluteus medius and gluteus maximus when compared to males during drop landings as measured by EMG (Zazulak, Ponce, Straub, Medvecky, Avedisian & Hewett, 2005). This has been linked as a risk factor of ACL injuries and PFPS due to the potential for hip adduction and knee valgus with weaker hip musculature (Jacobs & Mattacola, 2005; Boling et al., 2009; Lee & Powers, 2013). Females also tend to exhibit a quadriceps dominant landing strategies, in which females activate their quadriceps earlier relative to the hamstrings (Huston & Wojtys, 1996) and land with greater quadriceps activation (Malinzak, Colby, Kirkendall, Yu & Garrett, 2001; Chappell, Creighton, Giuliani, Yu & Garrett, 2007) in comparison to males. Although increased quadriceps activation may increase strain on the passive musculature around the knee, it may not significantly affect excursions or moments in the sagittal plane (Shultz et al., 2009).

Due to strength being modifiable, training intervention studies have been shown to reduce the risk of ACL injury with mixed efficacy (Stojanovic & Ostojic, 2012). These programs are comprised of strengthening, stretching, and feedback training to enhance jump-landing technique in athletes. Although many of these interventions lead to changes in lower extremity movement patterns during dynamic movements, the mechanisms by which this happens are largely unclear (Stojanovic & Ostojic, 2012). Further research is required to determine the efficacy of training programs on landing strategies adopted by males and females.

2.5.4 Gender differences in landing kinematics

There have been several studies examining the differences in landing kinematics between genders (Table 1). Although kinematic variables provide no direct measurement of forces, the positioning of body segments may give an indication of demand on ligaments and tendons. There is evidence to suggest that females land in a more erect landing position than males (Decker, Torry, Wyland, Sterett & Steadman, 2003; Salci, Kentel, Heycan, Akin & Korkusuz, 2004; Yu, Lin & Garrett, 2006). However, it has been suggested that a greater range of motion (ROM) and greater angular velocities in the sagittal plane seen in females may reduce the risk of injury to female athletes, by increasing knee flexion at the time of peak knee extensor power (Decker et al., 2003).

Due to the association of valgus movements and injury, there has been a substantial amount of research examining gender difference in the frontal plane during landing (Kernozek, Torry, Van Hoof, Cowley & Tanner, 2005; Hughes, Watkins & Owen, 2008). Studies have established differences between genders, with females exhibiting greater maximum knee valgus angles and larger frontal ROM at the knee (Kerzonek et al., 2005, Hughes et al., 2008). This may increase the risk of ACL injury in females relative to males due to the strain placed on the ACL during dynamic valgus (Markolf et al., 1995). Contrary to the findings of previous research, it has been suggested that gender differences in landing are due to discrepancies in balance and landing technique rather than innate differences between males and females (Orishimo, Kremenic, Pappas, Hagins & Liederbach, 2009). A correlation was also shown between the age at the start of ballet dance training and hip adduction. Further research is required to determine whether adequate training from a young age negates the increased risk of injury to females.

Table 1. Studies examining differences between genders in landing kinematics.

Study	Participants	Task	Findings
Decker et al. (2003)	Twelve males and 9 females	Two-legged drop landing from 60 cm platform	Females landed with smaller angles of hip, knee and ankle flexion at IC.
Salci et al. (2004)	Eight male and 8 female volleyball players	Volleyball spike and block landings from 40 cm and 60 cm platforms	Females exhibited significantly less knee and hip flexion in 40 cm spike and 40 cm block landings.
Kerzonek et al. (2005)	Fifteen males and 15 females	Two-legged drop landing from 60 cm	No significant difference at IC. Greater peak ankle dorsiflexion, peak ankle pronation, and peak knee valgus in females. Greater frontal ROM in females.
Yu et al. (2006)	Thirty males and 30 females	Vertical stop-jump	Hip flexion and knee flexion at IC was significantly lower in females. Knee flexion at peak anterior tibial shear force was lower in females when compared to males.
McLean et al. (2007)	Ten males and 10 females	Two-legged drop landings from 50 cm before and after fatiguing protocol	Females landed with greater plantarflexion angle at IC, and greater peak ankle supination, knee valgus, and knee internal rotation than males.

Hughes et al. (2008)	Six male and 6 female volleyball players	Volleyball block landings	Females exhibited significantly greater knee valgus angle and frontal ROM.
Orishimo et al. (2009)	Twelve male and 21 female ballet dancers	Single-leg drop landings from 30 cm platform	No gender differences in joint kinematics.

IC = Initial contact, ROM = Range of motion.

2.5.5 Gender differences in landing kinetics

Although kinematic variables can give an indication of how internal and external forces are mediated, it would be imprudent to draw conclusions based solely on the position of the lower extremity (Markolf et al., 1995). For this reason, a number of studies have compared kinetic variables between genders during landings (Table 2). It has been found by many studies that females demonstrate lower levels of normalised hip extension moment and increased normalised knee extension moment when compared to males (Chappell, Yu, Kirkendall & Garrett, 2002; Salci, Kentel, Heycan, Akin & Korkusuz, 2004). It has also been shown that females demonstrate greater knee valgus moments during landing when compared to males (Chappell et al., 2002). A combination of high knee extension and knee valgus moment increases the risk of ACL injury due to increased strain on the ACL (Markolf et al., 1995) Increased valgus moment may increase lateral compressive forces acting on the patellofemoral joint, thus increasing the risk of PFPS (Powers, 2010b).

Table 2. Studies examining differences between genders in landing kinetics.

Study	Participants	Task	Findings
Chappell et al. (2002)	Ten males and 10 females	Forward, vertical, and backward stop-jumps	Females exhibited greater knee extension and knee valgus moments during landing in each task with the exception of extension moment in the backward stop-jump
Decker et al. (2003)	Twelve males and 9 females	Two-legged drop landing from 60 cm platform	Females demonstrated greater knee extensor and ankle plantarflexor power than males.
Salci et al. (2004)	Eight male and 8 female volleyball players	Volleyball spike and block landings from 40 cm and 60 cm platforms	Females' peak knee extensor moment from 60 cm platform was greater than males. Females landed with greater normalised vGRF.
Yu et al. (2006)	Thirty males and 30 females	Vertical stop-jump	Females demonstrated higher vGRF, proximal tibia shear force, and knee extension moment during landing.
McLean et al. (2007)	Ten males and 10 females	Two-legged drop landings from 50	Females exhibited larger knee valgus, knee varus, and internal rotation

		cm before and after fatiguing protocol	moments, and smaller ankle dorsiflexion moments.
Orishimo et al. (2009)	Twelve male and 21 female ballet dancers	Single-leg drop landings from 30 cm platform	No gender differences in joint kinetics. Both groups exhibited minimal peak knee valgus moments and similar peak hip extension and abduction moments.

vGRF = Vertical ground reaction force

Previous research has explicated the potential link between higher GRF and increased injury risk (Kerzonek et al., 2005; Schmitz et al., 2007). Existing investigations exploring drop landings have resulted in varied findings. Studies have shown no difference in peak vertical GRF when normalised to body weight between genders (Decker et al, 2003; Orishimo et al., 2009). Decker (2003) suggested that females compensated for a more erect posture by employing a muscular strategy at the ankle to better dissipate force. However, other studies conclude that females tend to exhibit greater peak normalised vertical GRF when compared to males (Salci et al., 2004; Kernozek et al., 2005; Schmitz et al., 2007). The findings of Salci (2004) and Schmitz (2007) are likely to be explained by the adoption of a more extended posture during the landing phase by females in both studies. Furthermore, it was found that males exhibited a significantly (24%) greater amount of total (sum of hip, knee, and ankle) energy absorption of per unit of bodyweight when compared to the female group (Schmitz et al., 2007). The results from this study should be treated with caution when generalising to two-legged landings due to differences between single- and double-legged landings in the shape of the GRF curve. Caution should be taken when attributing the risk of injury to GRF in isolation, as there are many confounding factors such as the direction of GRF, joint angles, and muscle activity. These factors affect the direction of loading and may even decrease the risk of injury if the force is applied posteriorly at the knee (Myers & Hawkins, 2010).

2.6.1 Loaded Jumps

The ability to express power is essential in many sports and can be used to distinguish between levels of ability in athletes (Kawamori & Haff, 2004). It has been suggested that to increase power, it may be advantageous to train with a load that maximises mechanical power output (Wilson, Newton, Murphy, & Humphries, 1993). A popular exercise used to load athletes in order to maximise mechanical power output is the loaded, or weighted, countermovement jump (de Villarreal et al., 2009). During a loaded countermovement jump, an athlete descends to a comfortable depth, and then immediately jumps for maximum vertical height. The athlete and weight then descends due to gravity and then makes contact with the ground at the start of the landing phase (Hori et al., 2008). There are numerous means of loading countermovement jumps including: holding dumbbells or a hexagonal barbell, the use of a weighted vest or resistance bands, or placing a barbell across the shoulders (Schuna Jr. & Christensen, 2010; Argus, Gill, Keogh, Blazevich & Hopkins, 2011; Swinton, Stewart, Lloyd, Agouris & Keogh, 2012). Previous studies have reported landing from a jump as a common injury mechanism. Despite the popularity of loaded countermovement jumps and the risk of injuries associated with jump landings, there is limited research examining the effects of additional loading on landing biomechanics.

The efficacy of loaded countermovement jumps in training programmes to develop power and athletic ability has been corroborated by the findings of a number of studies (McBride, Triplett-McBride, Davie & Newton, 2002; Hoffman, Ratamess, Cooper, Kang, Chilakos & Faigenbaum, 2005; Cormie, McGuigan & Newton, 2010). It has been suggested that the inclusion of ballistic training exercises, such as loaded jump squats, leads to improvements in power due to the high velocity nature of the exercises (Kraemer & Newton, 2000). The use of added load is thought to increase the number of muscle fibers recruited compared with an unloaded jump and therefore is thought to require increased neural activation and give a greater training stimulus (Faccioni, 1994). The majority of existing literature on the effect of loaded countermovement jumps has examined kinetics and kinematics during the take-off phase (Moir, Sanders, Button & Glaister, 2005; Swinton et al., 2012). To the author's knowledge, only three studies have observed the effects of traditional methods of loading (e.g. weighted vests) on landings (Kulas et al., 2008; Kulas et al., 2010; Janssen et al., 2012). Other experiments have examined the effects of additional load on landing performance by the addition of body armour to military and law enforcement personnel (Sell, Chu, Abt, Nagai, Deluzio, McGrail & Lephart, 2010; Sell, Pederson, Abt, Nagai, Deluzio, Wirt & Lephart, 2013; Dempsey, Handcock & Rehrer, 2014; Brown, O'Donovan, Hasselquist, Corner & Schiffman, 2016).

2.6.2 Effects of Traditional Methods of Loading on Landings

Kulas et al. (2008) investigated changes in landing strategies during either loaded or unloaded drop landings. Male and female participants were required to perform a drop landing from a 0.45m box in an unloaded and a loaded condition (10% of body mass in a fitted vest). Participants were then grouped into a trunk-flexor or trunk-extensor group based on the strategy employed in the weighted landing condition. It was also found that when compared to the unloaded condition, a load of 10% of body mass increased knee angular impulse (18%) and energy absorption (14%), regardless of trunk landing adaptation. It was found that participants who landed with a more extended trunk exhibited greater increases in knee extensor moment and work (24% and 28% respectively) when compared to participants who adopted more trunk flexion (4% and 9% respectively) during loaded landings. This finding suggests that hip control strategies affect knee joint forces during loaded jump landings.

Data from the previous experiment (Kulas et al., 2008) was then used in a subsequent study to evaluate the effect of trunk load and trunk landing strategies on knee anterior shear forces and knee muscle forces during landing (Kulas et al., 2010). In the first step, joint moments and pre-existing physiological data were used to estimate knee muscle forces. The knee muscle forces are then applied in conjunction with joint reaction forces to the tibia to determine knee joint forces. It was found that in the loaded condition, peak and average knee anterior shear forces increased by 17% and 35% respectively in the trunk extensor group. In the trunk flexor group, there were more modest increases in peak (2%) and average knee anterior shear forces (1%). Quadriceps and gastrocnemius forces increased during loaded conditions in both groups, but average hamstring force decreased in the trunk extensor group. The results suggest the effect of load on landings can be offset through adaptation of landing strategies at the trunk, which allows for augmentation of hamstring force to offset increases in quadriceps force. This would theoretically lead to less knee anterior shear force and decrease ACL injury risk. The results of this study should be treated with caution as the biomechanical model used was only used for estimating muscle and shear forces, and did not directly calculate forces acting on the ACL. Trunk position was not manipulated directly and so the results cannot be attributed entirely to this. Another limitation of both studies is that the drop landing provides greater reliability but may not be representative of landings from dynamic movements (Kulas et al. 2008; Kulas et al., 2010).

Alterations to lower limb kinetics and kinematics have been investigated during the landing phase of loaded and unloaded volleyball block jumps (Janssen et al., 2012). Ten male participants performed a series of maximal loaded (fitted vest = 9.89kg – equal to 8-12% of bodyweight) or unloaded jumps. Results showed significantly greater hip flexion at initial contact in the loaded condition compared to the unloaded jumps. No other significant differences were observed for the ankle, knee, hip, or trunk kinematics between the loaded and unloaded conditions. No significant kinetic differences were identified between unloaded and loaded conditions. The investigation only reported sagittal plane variables, but it has been suggested that frontal variables are associated with an increased risk of ACL injuries (Boden et al., 2009). The participants were all highly trained volleyball players who had considerable experience performing unloaded and loaded jumps and may be more proficient in landing techniques when compared to novice athletes. No existing literature has compared different methods of loading on landing strategies (e.g. weighted vest, dumbbells, and barbell).

2.6.3 Effects of Body Armour on Landings

Due to the serious nature of musculoskeletal injuries in law enforcement and military environments, the effect of load in the form of body armour on landing biomechanics has been examined by a small number of studies. The findings of these studies are displayed in Table 3. The main findings of these studies suggest that additional load alter landing strategies that may result in an increased risk of injury. All three studies that reported vGRF showed significant increases in peak vGRF with additional load (Sell et al., 2010; Dempsey et al., 2014; Brown et al., 2016). In addition, when a range of loads were compared, the increase in GRF was greater when a heavier load was used (Brown et al., 2016). Greater vGRF during landing is likely to increase the loading of the lower extremity joints and thus requires considerable muscular strength in order to dissipate landing forces. Although the greater force acting on the lower limb alone may not be sufficient to cause injury, the risk to athletes with poor landing technique, previous injuries, or neuromuscular imbalances may be exacerbated with additional loads (Louw, Grimmer & Vaughan 2006). Brown and colleagues (2016) also reported no difference in energy absorption at the hip joint despite significant increase of angular impulse from light loads (0.9 $kg.m^2.s^{-1}$) relative to medium (1.2 $kg.m^2.s^{-1}$) and heavy loads (1.4 $kg.m^2.s^{-1}$) relative to medium loads. This could be due to the use of a more erect landing posture. Brown et al. (2016) found that although angular impulse in knee flexion increased by 18.75% from light load to medium load and by 11.59% from medium load to heavy load, there was no significant difference between energy absorption at the knees.

Table 3. Summary of studies examining effects of body armour on landing strategies.

	Load (% BW)	**Task**	**Main findings**
Sell et al. (2010)	18.00 ± 4.30	Two-legged drop landing from 50 cm with and without load.	Maximum knee flexion angle, maximum GRF, and time taken to reach maximum values increased with additional load.
Sell et al. (2013)	15.55 ± 4.18	Anterior single-leg jump landing of 30 cm with and without load.	Significantly reduced anterior-posterior, medial-lateral, and vertical dynamic postural stability with additional load.
Dempsey et al. (2014)	8.40 ± 0.80	Following a 5-minute run at 13 $km.h^{-1}$ participants performed: (1) a countermovement vertical jump, (2) a drop landing from 75 cm, (3) a depth jump, and (4) a drop landing with a distraction. All performed with and without load.	Peak GRF increased as a result of additional load and ground contact time increased for depth jump. Peak GRF was increased further by prior exercise and distraction during landing.

Brown et al. (2016)	7.93 ± 1.22 26.46 ± 3.07 52.91 ± 6.14	Two-legged drop landings followed by a cut and run at 45° from a 30 cm platform with light (6%), medium (20%), or heavy (40%) load.	Heavy load increased GRF to a greater extent than medium load, and for the medium relative to light load. Participants exhibited significantly less hip flexion with the medium and heavy load relative to the light load. Peak knee flexion was similar between light and medium load, but significantly lower with heavy load. Sagittal angular impulse in the hip, knee, and ankle increased in the medium load relative to light loads, and increased further with heavy load. No substantial differences in energy absorption at the hip, knee, or ankle.

GRF = Ground reaction force

Only one of the studies reported variables pertaining to the hip joint (Brown et al., 2016). It was shown that when compared to use of a light load (~8% bodyweight), both medium load (~27% bodyweight) and heavy load (~53% bodyweight) decreased mean (± SD) peak hip flexion from 35.9° (±10.5°) to 27.1° (±9.6°) and 27.8° (±8.8°) respectively. This is of significance as it has been suggested that landing with less hip flexion can place the ACL under greater strain when compared to a strategy that adopts a great amount of hip flexion (Kulas et al., 2010). It has been suggested that by increasing hip flexion, the gluteus maximus and hamstrings are in a more advantageous position to exert force, reducing knee extension and valgus moments (Kulas et al., 2010). In a study observing 70 air assault soldiers performing drop-landings with and without load (18% bodyweight), additional load significantly increased the maximum knee flexion angle by 8% (Sell et al., 2010). The findings of Brown and colleagues (2016) suggest that heavier loads decrease peak knee flexion angles to a greater extent than lighter loads. There is a large discrepancy between the loads used in the studies by Brown et al. (2016) and Kulas et al. (2010), which makes comparing the findings of the two studies problematic. Brown and colleagues (2010) also did not use an unloaded control group. Subsequently, any conclusion regarding knee flexion and load will be equivocal at best. Further research is required to better clarify the effect of different loads and landing biomechanics.

When considering valgus and varus angles, it has been shown that there are no significant differences at initial contact between loaded and unloaded conditions (Sell at al., 2010). This may result in a diminished ability to attenuate vertical GRF and may increase the risk of injury. Sell et al. (2010) examined the effects of body armour on dynamic postural stability. It was shown that the addition of body armour resulted in diminished postural stability, which could increase the risk of injury at the ankle (Wikstrom, Tillman, Chmielewski, Cauraugh & Borsa, 2007). The use of a 30-cm hurdle jump to single-leg landing would rarely be used under loaded conditions in a performance environment so the findings may not pertain to coaches and athletes.

2.6.4 Effects of Jump Height on Landings

The external load the body experiences during landings can also be altered through the manipulation of jump or drop height. Existing literature has sought to examine changes to landing strategies at increased velocities by altering drop-landing height (McNitt-Gray, 1991; Ford, Myer, Schmitt, Uhl & Hewett, 2011; Peng, 2011; Ali, Rouhi & Robertson, 2012; Dickin, Johann, Wang & Popp, 2015). Summary of the key findings in each of the studies are displayed in Table 4. All the aforementioned studies demonstrated statistically significant increases in vGRF as a result of dropping from increased heights. Greater vGRF during landing is likely to increase the demand placed on lower extremity musculature to attenuate landing forces, and when combined with suboptimal landing techniques could result in increased injury risk.

Significant increases in peak knee flexion angles occurred as a result of increased drop height in all studies, as well as an increase in peak hip flexion in studies where it was reported (McNitt-Gray, 1991; Ali et al., 2012; Dickin et al., 2015). Three studies (Ford et al., 2011; Peng, 2011; Dickin et al., 2015) utilised a drop jump and so the increases in knee and hip flexion angles may have been a strategy to generate sufficient power for the subsequent jump. It could be suggested that an increase in lower extremity peak flexion angles at the hip and knee may be required to dissipate increased landing forces more effectively. The findings from the studies examining drop jumps may not be generalisable to jump landings due to the differences in landing strategy adopted (Butler et al., 2013). As stated previously, the quadriceps, through the anterior pull of the patellar tendon on the tibia, contributes to ACL loading when knee flexion is less than 45° (Markolf et al., 1995). Ford and colleagues (2014) demonstrated a preferential activation of the quadriceps group in greater drop heights. The increase in activation of the quadriceps group was not accompanied by an increase in the recruitment of hamstring musculature. This suggests that increased landing intensity may increase shear load on the ACL due to differences in feed-forward neuromuscular control. This study only used females and so the strategy employed may not be utilised by males.

In the one study to report frontal kinematics, it was shown that increased drop height did not significantly change peak valgus angle or hip adduction angle at IC (Dickin et al., 2015). A trend was evident for increased ankle inversion moment, which increased with greater drop heights. The consistent peak valgus angles may be due to the significantly greater peak flexion angles at the hip, knee, and ankle during landings from greater heights. This strategy may have attenuated landing forces more effectively in the sagittal plane, thus reducing the need to dissipate force in the frontal plane. The increased intensity of jump landings through the manipulation of drop height has been shown to cause differences in landing strategies in a variety of populations. Due to differences in task, it is difficult to compare the demands on landings in response to increased jump height and additional load.

Table 4. Summary of studies examining effects of drop height on landing strategies

	Height	Task	Main findings
McNitt-Gray (1991)	0.32m 0.72m 1.28m	Two-legged drop landing from various heights performed by gymnasts and recreational athletes.	Increased peak vGRF and peak knee and hip flexion angles and angular velocities as a result of greater drop heights.
Ford et al. (2011)	0.20m 0.30m 0.40m 0.50m 0.60m	Maximal drop jumps from various heights by recreationally active participants.	Hip flexion decreased at IC as a result of greater drop heights. Increased vGRF in landings from greater heights. Increased activation of the quadriceps with greater drop height but no significant change in hamstring activation.
Peng (2011)	0.20m 0.30m 0.40m 0.50m 0.60m	Drop jumps from various heights by physically active college students.	Increased impulse, vGRF, and negative joint work at the ankle and knee as a result of greater drop heights.
Ali et al. (2012)	0.30m 0.50m 0.70m	Single-leg landings from various heights recreationally active participants.	Increases in height were associated with significantly greater vGRF, peak knee flexion angle, peak trunk flexion angle, and power and work at the knee.
Dickin et al. (2015)	0.30m 0.40m 0.50m	Double-leg drop jumps from various heights before and after fatiguing protocol.	Increased drop height caused decreases in hip flexion at IC and increases in peak knee and hip flexion angles. Increased height was also associated with greater vGRF and joint moments and powers.

2.7 Conclusion

The knee has been cited to be the most frequently injured site of the body and the catastrophic consequences of serious injury to the joint are well documented. Maladaptive jump landing strategies which cause excessive loading at the knee joint have been strongly associated with the occurrence of injury at the knee. Previous literature has demonstrated that gender and additional landing demands, either through the addition load or alterations of drop-landing height, can significantly affect landing strategies which may increase the risk of injury during drop or jump landings. These include alterations to sagittal joint angles and increases in vGRF. Thus far, no literature has examined the interplay of increased landing demands through the addition of load and gender during jump landings. Moreover, no existing literature has compared landing techniques when using different methods of loading.

Chapter Three: Reliability Testing

3.1 Introduction

Prior to the collection of data for the main study, the Helen Hayes marker set (Kadaba, Ramakrishnan & Wootten, 1990) to be used for data collection was tested to ensure reliability between days. This was to ensure the researcher was competent at placement of the markers, as well as processing the data collected from trials.

3.2 Method

3.2.1 Participants

Five participants (2 female, 3 male, age 20.6 ± 0.5 years, body mass 84.8 ± 18.7 kg, height 1.81 ± 0.08 m) volunteered for the study after giving written informed consent. To be included in the reliability testing, participants were required to be free of previous serious lower extremity and lower back injury, or current musculoskeletal injury. Institutional ethical approval was granted prior to data collection (protocol number: LMS/PGR/UH/02335).

3.2.2 Testing Procedure

Participants were required to visit the laboratory on two separate days, which were separated by at least 48 hours but no more than 96 hours. The procedure was kept the same for both visits. The height and mass of each subject were recorded prior to data collection and used for normalisation of kinetic variables. Participants were taken through a standardised dynamic warm-up prior to testing consisting of a 5-minute warm up at 100W on a cycle ergometer (Monark 874E Ergomedic, Sweden) followed by lower limb mobility movements (reverse lunges, side lunges, and bodyweight squats). The warm up was in line with recommendations outlined by the American College of Sports Medicine (Haff & Triplett, 2016). On completion of the warm up participants were instructed to practice maximal countermovement jumps with no arm swing until they felt comfortable with landing. Markers were then positioned on the participant and a static trial was performed with the participant standing in a neutral position.

The task tested in this study was a landing from maximal vertical unloaded countermovement jumps. The participants were required to perform 5 successful maximal countermovement jumps. A trial was considered successful once the participant made complete foot contact on the force plate with their preferred foot. The preferred foot was defined as the foot that the participants would use to kick a ball with (Ford, Myer & Hewett, 2003). Participants were not instructed to target the force plate with their preferred foot. To increase the likelihood of a successful landing each trial began with the dominant foot on the force plate. Countermovement jump technique including depth or speed of descent was not standardised but participants were instructed to jump to achieve maximal height. Participants were instructed to not use arm swing. Participants were given 45 seconds rest in between each repetition.

3.2.3 Data Collection

Participants were labelled with 29 passive retro-reflective markers (20mm in diameter) in accordance with the Helen Hayes marker set (Kadaba et al., 1990). Markers were placed on the following: head (top, front and back), right scapula, right and left acromion process, epicondyles of right and left humeri, right and left styloid processes of each radius, the right and left anterior superior iliac spine (ASIS), sacrum, right and left thigh approximately 10 cm superior to tibiofemoral joint line, right and left lateral and medial femoral condyles, right and left lateral tibia (half way between ankle and knee), right and left lateral and medial malleoli, right and left posterior portion of the calcaneus, and the right and left web space between metatarsals 1 and 2. Markers placed on the right and left medial malleoli and right and left medial femoral condyles were used for an initial static trial with the participant standing in the neutral position. All medial markers on the lower limb were removed for landing trials. The three-dimensional (3D) coordinates of these markers were recorded using a set of 8 high-speed Owl cameras (Motion Analysis Corp., Santa Rosa, CA, USA) sampling at 250Hz. Prior to data collection the researcher performed a static and dynamic calibration. The static calibration was performed using an L-frame with 4 retroreflective markers affixed placed on the corner of the force plate. The dynamic calibration was performed by waving a wand with 3 retroreflective markers within the movement space. One AMTI force plate (Advanced Mechanical Technology Inc., Watertown, MA) sampling at 1000Hz was used to gather ground reaction force data. The force plate was surrounded by a customised platform of the same height to ensure an even landing surface.

3.2.4 Data analysis

From the standing trial, a kinematic model consisting of 12 skeletal segments (head, trunk, bilateral upper arm, bilateral lower arm and hand, bilateral upper leg, bilateral shank, and bilateral foot) was created using Cortex software (Motion Analysis Corp., Santa Rosa, CA, USA). Ankle and knee joint centers were defined as the midpoint of the medial and lateral malleolus and the medial and lateral femoral epicondyle markers, respectively. The 3D coordinates of the hip joint centers were approximated using the coordinates of the reflective markers at the left and right ASISs and the marker at the sacrum (Bell, Pedersen & Brand, 1990). Marker coordinates were filtered at 6 Hz using a fourth-order Butterworth filter. Joint angular positions were calculated based on a right-hand convention using Euler angles in a

Y (flexion/extension), X (adduction/abduction), Z (internal/external rotation) rotation sequence. The landing phase was defined as the period of time from initial contact to peak knee flexion. Initial contact was defined as the point at which vertical ground reaction force (vGRF) exceeded 15 N to disregard the effects of noise on the force plate (Cortes et al., 2007). Peak angles were defined as the maximum angle during the landing phase. Total angular excursion or ROM was defined as the difference between the minimum joint angle and peak angle.

Force plate data were low pass filtered at 60 Hz (4th order zero-phase lag Butterworth). Internal joint moment values for each joint were calculated by combining the kinematic and force plate data with anthropometric data in an inverse dynamics solution. All force values and all joint moment parameters were scaled Newton per kilogram of body mass and Newton-meter per kilogram of body mass, respectively.

3.2.5 Statistical analysis

All dependant variables were calculated for each trial. Between-day reliability and consistency was assessed by calculating the Intraclass Correlation Coefficient (ICC (2,k)). ICC (2,k) was chosen as it has been suggested that a two-way model addresses both systematic and random error (Weir, 2005). An ICC (2,k) was used in lieu of an ICC (2,1) due to 5 trials being used in calculations instead of one trial. An ICC above 0.90 was considered very high, between 0.70 and 0.89 as high, between 0.50 and 0.69 as moderate, and below 0.49 as low (Munro, 2005). From calculating the ICC the Standard Error of Measurement was calculated using the following formula:

$$\mathrm{SEM} = \mathrm{SD}\sqrt{(1-\mathrm{ICC})}$$

Where SEM = standard error of measurement, SD = standard deviation of the sample, and ICC = the intraclass correlation coefficient (ICC (2,K)).

3.3 Results

Acceptable interclass correlation (ICC) values were established for peak sagittal angles ranging from high to very high (0.81 - 0.92) and peak frontal angles ranging from moderate to very high (0.66 - 0.96). ICC for measurements of frontal ROM at the hip (0.67) and knee (0.67) were both moderate with between-day reliability of measurement of hip and knee sagittal ROM being very high (0.92 - 0.95). Reliability of sagittal ROM of the ankle was found to be moderate (0.69). Between-day measurement of joint moments in all three planes was found to reliable ranging from moderate to very high (0.69 - 0.85). Reliability of both peak vertical ground reaction force (0.84) and time to peak vertical ground reaction force (0.81) was shown to be high. Table 5 displays the ICC and standard error of the mean (SEM) for kinetic and kinematic variables.

Table 5. Between-day measurement - ICC and SEM of kinematic and kinetic variables.

Variable	ICC	SEM
Hip		
Peak Flexion Angle (º)	0.94	3.57
Peak Adduction/Abduction Angle (º)	0.66	0.84
Frontal ROM (º)	0.67	0.23
Sagittal ROM (º)	0.92	3.50
Peak Extension Moment (Nm/kg.m)	0.85	0.14
Peak Adduction/Abduction Moment (Nm/kg.m)	0.78	0.04
Knee		
Peak Flexion Angle (º)	0.92	2.29
Peak Valgus/Varus Angle (º)	0.96	0.83
Frontal ROM (º)	0.67	0.34
Sagittal ROM (º)	0.95	2.35
Peak Extension Moment (Nm/kg.m)	0.75	0.07
Peak Valgus/Varus Moment (Nm/kg.m)	0.82	0.07
Ankle		
Peak Dorsiflexion / Plantarflexion Angle (º)	0.81	0.51
Sagittal ROM (º)	0.69	0.84
Peak Dorsiflexion / Plantarflexion Moment (Nm/kg.m)	0.77	0.03
vGRF		
Peak (N/kg)	0.84	0.54
Time to peak vGRF (ms)	0.81	4.00

3.4 Discussion

The results presented in table 5 show that between-day reliability ranged from good to excellent in all peak kinematic and peak kinetic variables when using the Helen Hayes marker set. Between-day reliability of the data collected from the force plate was excellent. This means that in the main testing procedure a Helen Hayes marker set will be used in conjunction with a force plate to collect kinematic and kinetic data.

Chapter Four: Method

4.1 Study Design

This study used a repeated measures design to compare landings in different loading conditions (unloaded, dumbbell, weighted vest, and barbell) in male and female moderately trained athletes. Participants attended one 60-minute testing session.

4.2 Participants

Priori power calculation at 80% based on the findings of Yu et al. (2006) determined a sample size of 16 in total was required. Twelve male (age 21.1 ± 1.4 years, body mass 76.2 ± 10.3 kg, height 1.77 ± 0.08 m) and 12 female (age 20.3 ± 1.4 years, body mass 64.4 ± 7.2 kg, height 1.70 ± 0.03 m) moderately trained participants volunteered for the study after giving written informed consent. Moderately trained individuals were defined as the population that had been strength training twice per week for at least one year prior to the study. To be included in the study, participants were required to be free of previous serious lower extremity and lower back injury, or current musculoskeletal injury. Institutional ethical approval was granted prior to the study (protocol number: LMS/PGR/UH/02335).

4.3 Testing Procedure

The height and mass of each subject were recorded prior to data collection and used for normalisation of the kinetic variables and calculation of inertial properties of segments. Participants were taken through a standardised dynamic warm-up prior to testing consisting of a 5-minute warm up at 100W on a cycle ergometer (Monark 874E Ergomedic, Sweden) followed by lower limb mobility movements (reverse lunges, side lunges, and bodyweight squats). The warm up was in line with recommendations outlined by the American College of Sports Medicine (Haff & Triplett, 2016). On completion of the warm up participants were instructed to practice maximal countermovement jumps with no arm swing until they felt comfortable with landing. Markers were then positioned on the participant and a static trial was performed with the participant standing in a neutral position.

The task tested in this study was a landing from maximal vertical countermovement jumps in four conditions: loaded with a barbell (BB), with dumbbells (DB), with a weighted vest (WV), and unloaded (UL) (Figure 5). The weight used in each of the loaded conditions was adjusted to closely match 10% of the participants' body mass (Table 6). The participants were required to perform 3 successful maximal countermovement jumps. A trial was considered successful once the participant made complete foot contact on the force plate with their preferred foot. The preferred foot was defined as the foot that the participants would use to kick a ball with (Ford et al., 2003).

Participants were not instructed to target the force plate with their preferred foot. To increase the likelihood of a successful landing each trial began with the dominant foot on the force plate. Countermovement jump technique including depth or speed of descent was not standardised but participants were instructed to jump to achieve maximal height. Participants were instructed to not use arm swing in all conditions. In the UL and WV conditions participants were instructed to place their hands on their waist throughout the jump and landing. In the DB condition, participants were instructed to keep their arms relaxed. In the BB condition, participants were required to grasp the barbell firmly throughout the jump and landing. Each condition was performed in a randomised order to reduce order effects. Participants were given 45 seconds rest in between each repetition and 3 minutes rest in between conditions to avoid effects of fatigue. All participants were blinded to the hypotheses of the study. The researcher demonstrated and explained a countermovement jump once in each condition to avoid coaching effects.

Table 6. Additional mass used in loaded jump conditions

	Males	**Females**
Average Load (% body mass)	10.08	10.17
Minimum Load (% body mass)	8.57	9.38
Maximum Load (% body mass)	11.27	11.11

4.4 Data Collection

Participants were labelled with 29 passive retro-reflective markers (20mm in diameter) in accordance with the Helen Hayes marker set as previously tested in chapter 3 (Kadaba et al., 1990). For BB condition trials, an additional segment was created by the placement of 3 markers: one on both ends of the barbell, and one on the superior aspect of the weight plate. When testing the WV condition, the marker placed on the right scapula was removed and placed on the weighted vest in approximately the same position relative to the scapula. The three-dimensional (3D) coordinates of these markers were recorded using a set of 8 high-speed Owl cameras (Motion Analysis Corp., Santa Rosa, CA, USA) sampling at 250Hz. Prior to data collection the researcher performed a static and dynamic calibration. The static calibration was performed using an L-frame with 4 retroreflective markers affixed placed on the corner of the force plate. The dynamic calibration was performed by waving a wand with 3 retroreflective markers within the movement space. One AMTI force plate (Advanced Mechanical Technology Inc., Watertown, MA) sampling at 1000Hz was used to gather ground reaction force data. The force plate was surrounded by a customised platform of the same height to ensure an even landing surface.

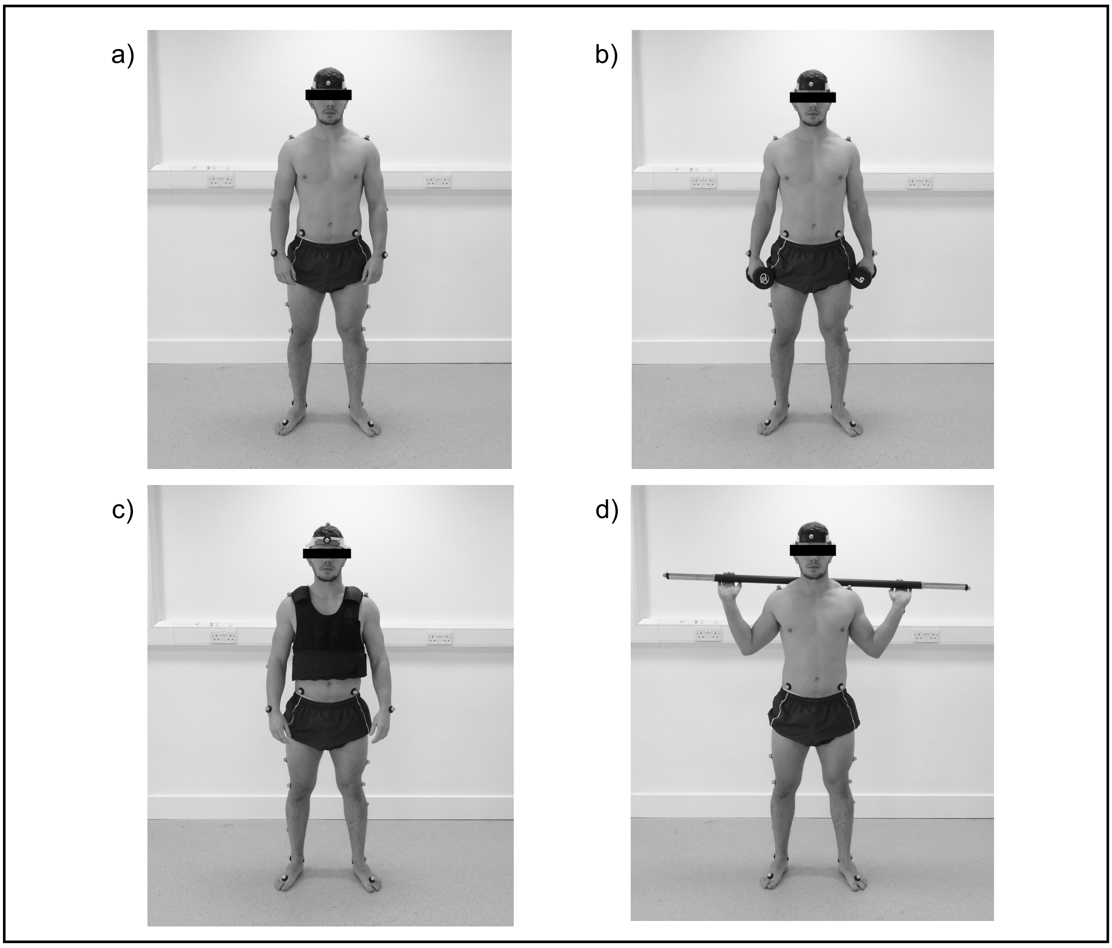

Figure 5. Participant labelled in the: a) UL condition, b) DB condition, c) WV condition, and d) BB condition.

4.5 Data analysis

From the standing trial, a kinematic model consisting of 12 skeletal segments (head, trunk, bilateral upper arm, bilateral lower arm and hand, bilateral upper leg, bilateral shank, and bilateral foot) was created using Cortex software (Motion Analysis Corp., Santa Rosa, CA, USA). Ankle and knee joint centres were defined as the midpoint of the medial and lateral malleolus and the medial and lateral femoral epicondyle markers, respectively. The 3D coordinates of the hip joint centres were approximated using the coordinates of the reflective markers at the left and right ASISs and the marker at the sacrum (Bell et al., 1990). Marker coordinates were filtered at 6 Hz using a fourth-order Butterworth filter. Joint angular positions were calculated based on a right-hand convention using Euler angles in a Y (flexion/extension), X (adduction/abduction), Z (internal/external rotation) rotation sequence. The landing phase was defined as the period from initial contact to peak knee flexion. Initial contact was defined as the point at which vertical ground reaction force (vGRF) exceeded 15

N to disregard the effects of noise on the force plate (Cortes et al., 2007). Time to peak knee flexion was calculated as the amount of time from initial contact to peak knee flexion. Peak angles were defined as the maximum angle during the landing phase. Total angular excursion or ROM was defined as the difference between the minimum joint angle and peak angle.

Force plate data were low pass filtered at 60 Hz (4th order zero-phase lag Butterworth). Internal joint moment values for each joint were calculated by combining the kinematic and force plate data with anthropometric data in an inverse dynamics solution. To allow for additional mass to body segments during weight trials to be included in calculations (e.g. additional mass to trunk segment during WV condition), existing segment inertial parameter data from De Leva (1996) was edited to account for the additional mass in the respective segments. In the WV condition, mass equal to the mass of the weighted vest was added to the trunk segment of the model. In the DB condition, the mass of each dumbbell was added to each of the hand segments in the model. In the BB condition, the mass equal to the mass of the barbell and weight plates was assigned to the additional barbell segment. All force values were scaled Newton per kilogram of body mass ($N.kg^{-1}$). Joint moments were normalised to body mass x height ($N.m.kg^{-1}.m^{-1}$), respectively. Additional load was disregarded when normalising joint moments and force values. Jump height was calculated using the flight time method by using the following equation of acceleration:

$$JH=\frac{1}{2} \times g \times \left(\frac{t}{2}\right)^2$$

where g = 9.81 $m.s^{-2}$ and t is the difference in time between the instant of take-off and the instant of landing. From this value, velocity of the COM at IC was determined using

$$v_{IC}= 2 \times g \text{ x } JH$$

where JH is jump height. Momentum at IC was subsequently calculated as the product of velocity of the COM at IC and the participant's mass.

4.6 Statistical analysis

All dependant variables were calculated for each successful trial and then averaged across the 3 trials. Repeated measures ANOVA (load type * gender) was used to test for differences in peak joint angles, joint angles at initial contact (IC), joint excursion (ROM), time to peak knee flexion, peak vertical ground reaction force (vGRF), and time to peak vGRF. The alpha level was set at $p < 0.05$. A residual analysis was performed to ensure assumptions for the two-way ANOVA were met. Normality of data was confirmed using Shapiro-Wilk's test of normality. Sphericity was tested for using Mauchly's test of sphericity. Post hoc testing was performed using Tukey's HSD test. Cohen's D was used to determine effect sizes between loading conditions within genders and, where appropriate, was used to determine effect sizes between loading conditions across genders. The effect size was defined as trivial if it was <0.2, small 0.2-0.5, medium 0.5- 0.8, and large>0.8 (Portney & Watkins, 2000).

Chapter Five: Results

A two-way repeated measures ANOVA (load type * gender) was conducted to examine the effects of gender and load type on kinematic and kinetic variables. A residual analysis was performed to ensure assumptions for the two-way ANOVA were met. Normality and sphericity of data was confirmed using Shapiro-Wilk's test of normality ($p > 0.05$) and Mauchly's test of sphericity respectively.

5.1.1 Sagittal Kinematic Variables

The two-way repeated measures ANOVA showed no statistically significant interaction effects between gender and load type for lower limb joint rotations at initial contact (IC) or peak joint rotations. There was no statistically significant interaction effect between gender and load for lower limb joint sagittal ROM. There were statistically significant differences in peak ankle dorsiflexion found between genders in unloaded ($p = 0.011$), dumbbell loaded ($p = 0.009$), weighted vest loaded ($p = 0.027$), and barbell loaded ($p = 0.027$) jump landings. Means, standard deviations, and effect sizes for lower limb sagittal plane rotations are displayed in Table 7.

Table 7. Mean sagittal plane peak angles, initial contact (IC) angles, and excursions (ROM) during landing phase (± standard deviation, effect size in brackets)

Rotation [1] (º)	Male				Female			
	UL	DB	WV	BB	UL	DB	WV	BB
Ankle Plantar / Dorsiflexion								
IC	-30.4±5.8	-29.5±5.3 (0.161)	-30.8±6.8 (0.063)	-30.6±4.4 (0.038)	-30.1±5.5	-29.4±5.4 (0.128)	-31.0±5.3 (0.167)	-30.7±6.6 (0.100)
ROM	43.5±5.9	41.4±4.5 (0.400)	43.9±7.1 (0.061)	45.5±4.3 (0.387)	47.6±7.5	45.9±5.8 (0.253)	48.1±5.6 (0.076)	49.5±4.3 (0.310)
Peak	13.0±4.5*	11.9±4.5* (0.244)	13.2±4.4* (0.045)	14.9±2.7* (0.512)	17.5±4.7*	16.5±3.3* (0.246)	17.1±4.2* (0.090)	18.8±5.0* (0.268)
Knee Flexion / Extension								
IC	20.3±6.0	18.8±6.8 (0.233)	18.6±6.5 (0.240)	19.6±1.9 (0.157)	20.5±4.0	18.4±4.0 (0.525)	19.1±1.6 (0.460)	21.0±4.0 (0.125)
ROM	53.5±15.8	52.2±11.6 (0.094)	55.2±11.9 (0.122)	59.3±19.0 (0.332)	63.9±10.5	62.4±11.9 (0.668)	64.1±11.7 (0.522)	66.2±12.6 (0.319)
Peak	73.7±17.8	71.0±10.3 (0.186)	73.8±13.2 (0.006)	78.9±20.3 (0.272)	84.4±10.2	80.8±11.8 (0.326)	83.2±11.2 (0.112)	87.3±13.0 (0.248)
Hip Flexion/ Extension								
IC	26.7±3.1	24.3±9.7 (0.333)	23.6±8.4 (0.490)	27.4±2.2 (0.260)	25.3±5.0	23.6±2.8 (0.420)	21.0±2.8 (1.061)	26.2±3.3 (0.212)
ROM	46.4±20.2	48.7±19.7 (0.115)	50.7±17.0 (0.230)	50.6±17.1 (0.224)	56.4±18.2	58.5±19.4 (0.112)	63.2±22.5 (0.332)	61.2±16.9 (0.273)
Peak	72.6±24.5	73.1±24.4 (0.020)	74.3±21.1 (0.074)	78.0±17.6 (0.253)	81.6±20.7	82.1±20.2 (0.024)	84.2±21.6 (0.113)	87.4±13.4 (0.333)

[1] First of two parameters listed is a positive joint rotation relative to the neutral position
* Indicates a significant difference between genders. # Indicates a significant difference from unloaded condition.

Where appropriate, Cohen's d effect size was calculated between combined averages for both genders in each loading condition. Trivial effect sizes were noted when comparing dorsiflexion at IC between dumbbell (Cohen's d = 0.155, *p* = 0.614), weighted vest (Cohen's d = 0.106, *p* = 0.712), or barbell conditions (Cohen's d = 0.050, *p* = 0.847). Small effect sizes were noted between UL and DB (Cohen's d = 0.270, *p* = 0.895), as well as between UL and BB (Cohen's d = 0.247, *p* = 0.384) in ankle sagittal ROM. Trivial effect sizes were noted between UL and WV in ankle sagittal ROM (Cohen's d = 0.773, *p* = 0.805). Effect sizes for DB (Cohen's d = 0.217, *p* = 0.398) and BB (Cohen's d = 0.348, *p* = 0.455) were found to be small for peak dorsiflexion with changes in the WV condition being trivial (Cohen's d = 0.025, *p* = 0.181).

Knee flexion angle at IC was decreased, with small effect sizes evident, in both DB (Cohen's d = 0.359, *p* = 0.181) and WV (Cohen's d = 0.317, *p* = 0.248) conditions when compared to UL. Differences in knee flexion at IC were trivial between UL and BB (Cohen's d = 0.017, *p* = 0.955). Peak knee flexion did not change as a result of loading with DB (Cohen's d = 0.232, *p* = 0.435), WV (Cohen's d = 0.042, *p* = 0.890), and BB condition (Cohen's d = 0.283, *p* = 0.316). There was no significant difference in sagittal ROM at the knee for DB (Cohen's d = 0.108, *p* = 0.723), WV (Cohen's d = 0.074, *p* = 0.802) and BB condition (Cohen's d = 0.299, *p* = 0.288). Figure 6 shows the knee flexion angle from representative trials in each condition and gender throughout the landing phase.

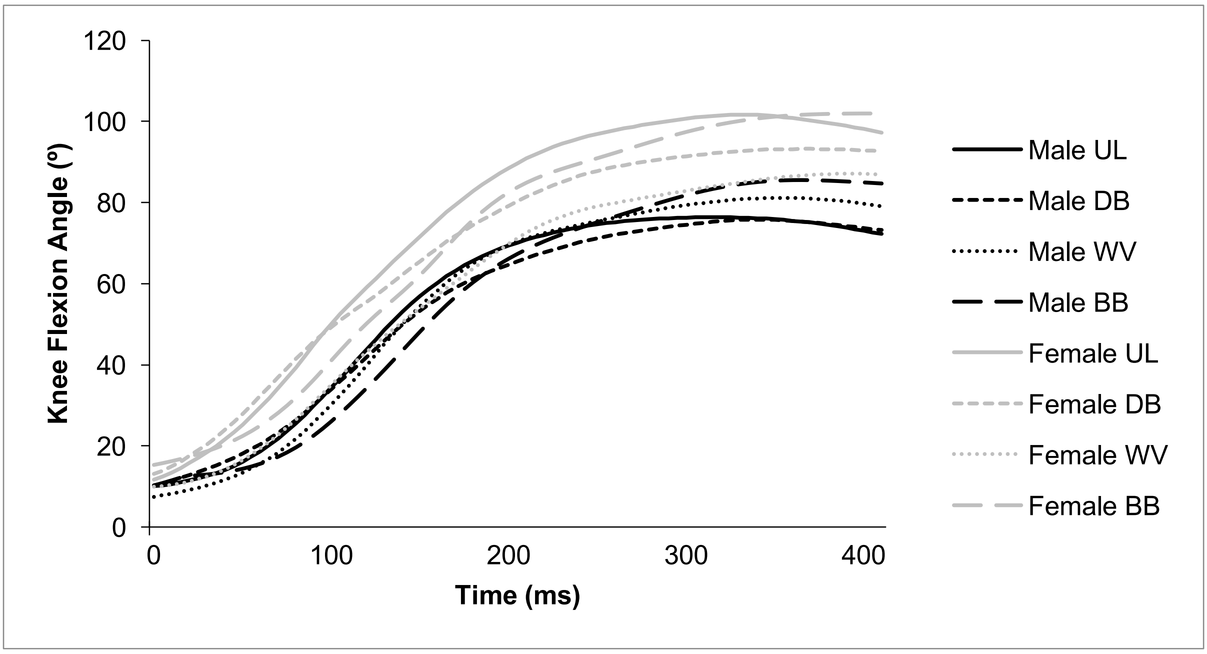

Figure 6. Knee flexion angles during representative jump landings in different loading conditions and genders.

There was a significant difference between genders in the time to reach peak flexion in the barbell loaded jump landing (p = 0.025) with females taking more time to achieve peak knee flexion. Compared to the UL condition, there was no significant difference in the time to peak knee flexion in the DB condition (Cohen's d = 0.450, p = 0.148), a significant moderate effect size was calculated for the 49 ms increase in the WV condition (Cohen's d = 0.618, p = 0.029), and a significant large effect size was calculated for the 70 ms increase in the BB condition (Cohen's d = 0.870, p = 0.002).

The DB condition no change hip flexion at IC compared to UL (Cohen's d = 0.370, p = 0.187). There was a medium effect size (Cohen's d = 0.646, p = 0.018) evident between the WV and UL with an average decrease of 3.75° which was significant in hip flexion at IC. There was a no significant difference between BB and UL in hip flexion at IC (Cohen's d = 0.159, p = 0.605). There were no significant differences in peak hip flexion in DB (Cohen's d = 0.021, p = 0.936), WV (Cohen's d = 0.096, p = 0.722) and BB condition (Cohen's d = 0.273, p = 0.349). No significant difference was found between in hip sagittal ROM in the DB (Cohen's d = 0.115, p = 0.680), WV (Cohen's d = 0.280, p = 0.307), and BB (Cohen's d = 0.235, p = 0.408) conditions. Figure 7 illustrates sagittal joint angles at IC and during peak knee flexion.

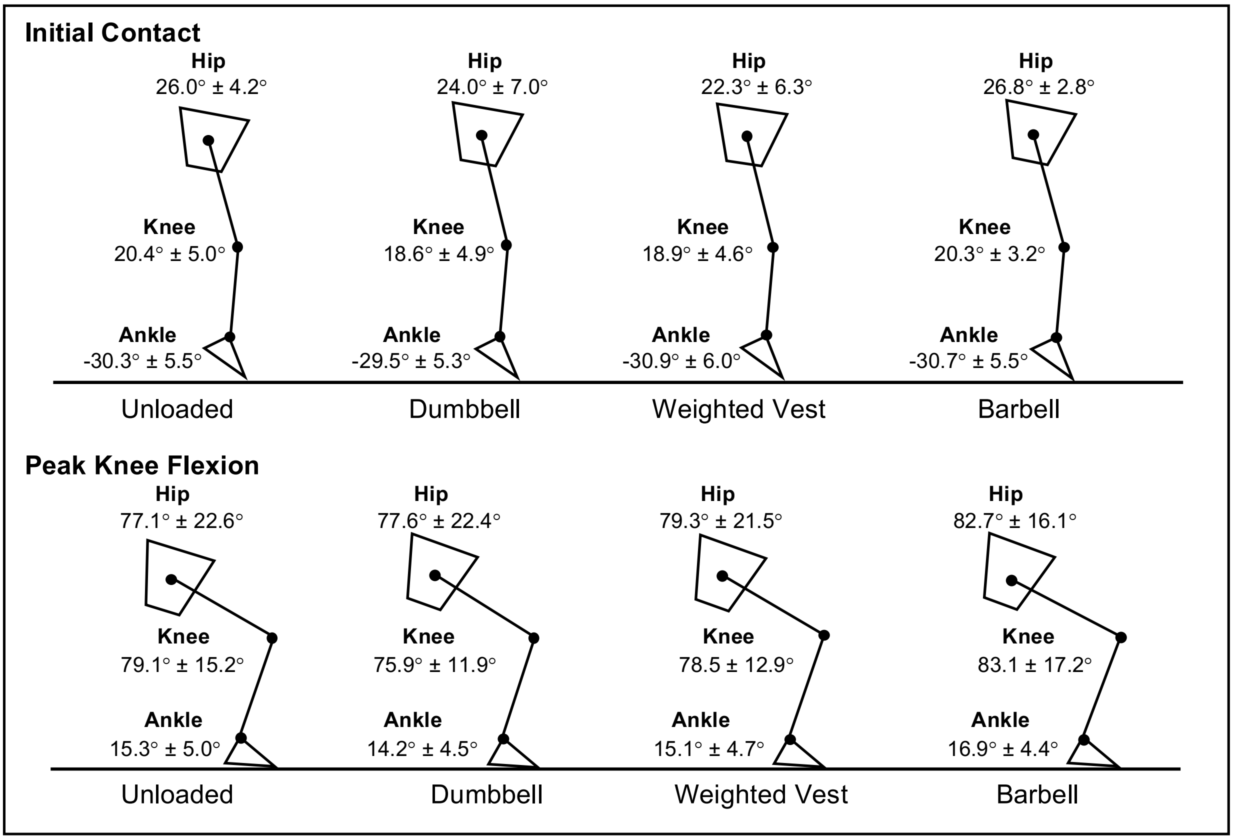

Figure 7. Sagittal joint angles in different conditions during initial contact and peak knee flexion averaged across both genders.

5.1.2 Frontal Kinematic Variables

Table 8 displays mean frontal plane kinematics across all loading types and genders. There were statistically significant differences between genders in hip adduction angle at initial contact with females showing increased adduction in UL ($p < 0.001$), DB ($p = 0.012$), and BB ($p = 0.007$) jump landings. Significant differences between genders were not evident in the WV condition. Gender differences in peak hip adduction angle were shown to be statistically significant in unloaded ($p = 0.025$), weighted vest ($p = 0.008$), and barbell ($p = 0.011$) conditions with females demonstrating greater peak hip adduction angles. Males in the BB condition demonstrated an increased frontal ROM at the knee when compared to females, which was significant ($p = 0.023$).

The was no significant difference observed in knee valgus angle at IC due to additional loading in DB (Cohen's d = 0.564, $p = 0.109$). Both WV (Cohen's d = 0.638, $p = 0.047$), and BB (Cohen's d = 0.647, $p = 0.039$) conditions showed a statistically significant decrease in knee valgus at IC compared to the UL condition. Differences in peak knee valgus angles between loading conditions were found to be trivial across all loading conditions. No significant differences were seen between conditions with trivial effect sizes in DB (Cohen's d = 0.009, $p = 0.979$), WV (Cohen's d = 0.151, $p = 0.637$), and BB conditions (Cohen's d = 0.160, $p = 0.581$) when compared to the UL condition.

Table 8. Average frontal plane peak angles, initial contact (IC) angles, and excursions (ROM) during landing phase (± standard deviation, effect size in brackets)

Rotation [1] (º)	Male				Female			
	UL	DB	WV	BB	UL	DB	WV	BB
Hip Abduction/ Adduction								
IC	-6.9±1.3*	-7.1±1.2* (0.333)	-8.6±1.5 (1.333)	-8.6±0.8* (1.414)	-10.7±2.9*	-9.4±1.6* (0.555)	-9.5±3.4 (0.380)	-11.1±3.6* (0.122)
ROM	5.6±1.0	5.7±2.3 (0.056)	6.3±1.0 (0.700)	7.4±4.3 (0.577)	6.9±2.4	7.7±3.2 (0.283)	7.4±2.6 (0.200)	6.4±1.2 (0.264)
Peak	-11.5±4.1*	-11.3±2.5 (0.082)	-11.9±2.3* (0.170)	-13.2±3.2* (0.601)	-14.0±1.8*	-13.3±2.6 (0.313)	-14.8±1.9* (0.432)	-16.0±1.5* (1.207)
Knee Varus / Valgus								
IC	-4.1±1.0	-2.8±4.6 (0.391)	-2.6±3.8 (0.540)	-2.8±3.4 (0.520)	-5.0±1.2	-3.7±2.0 (0.788)	-3.3±1.9 (1.070)	-2.9±2.4 (1.107)
ROM	14.1±7.3	13.9±5.8 (0.030)	15.1±8.3 (0.128)	13.4±7.3* (0.096)	13.8±5.2	14.1±7.0 (0.049)	14.9±7.5 (0.170)	12.5±3.7* (0.288)
Peak	-15.1±8.8	-16.1±8.6 (0.115)	-15.8±9.2 (0.056)	-16.4±4.6 (0.185)	-15.3±8.0	-16.4±8.2 (0.136)	-15.7±8.0 (0.050)	-15.1±6.0 (0.028)

[1] First of two parameters listed is a positive joint rotation relative to the neutral position
* Indicates a significant difference between genders. # Indicates a significant difference from unloaded condition.

There were no significant changes in hip adduction angle at IC in the DB (Cohen's d = 0.225, p = 0.415), WV condition (Cohen's d = 0.089, p = 0.738), and BB condition (Cohen's d = 0.417, p = 0.108). There was no significant change in peak adduction angle seen in the DB condition (Cohen's d = 0.133, p = 0.593), WV condition (Cohen's d = 0.216, p = 0.414), with a significant moderate effect size calculated for increase observed in the BB condition (Cohen's d = 0.646, p = 0.017). No significant differences were observed in hip frontal ROM during landing in all loaded conditions when compared to the UL condition. Effect sizes of hip frontal ROM in the conditions were calculated to be trivial in the DB condition (Cohen's d = 0.191, p = 0.527), and small in both WV (Cohen's d = 0.281, p = 0.382) and BB (Cohen's d = 0.260, p = 0.375) conditions with no significant difference evident.

5.2.1 Jump Height and Momentum

Differences in jump height were shown to be significant between genders (Table 9) with males demonstrating higher jump heights than females across all conditions (p < 0.001). Jump heights decreased in all loaded conditions. There were significant decreases in jump height in the BB condition in males (Cohen's d = 1.500, p = 0.003) and females (Cohen's d = 3.500, p = 0.043), and in the WV condition in males (Cohen's d = 1.250, p = 0.002). Females demonstrated decreases in jump height in the DB condition (Cohen's d = 3.000, p = 0.322) and WV condition (Cohen's d = 3.500, p = 0.061) when compared to the UL condition. The decrease in jump height between UL and DB conditions in male was found to be large but not significant (Cohen's d = 0.750, p = 0.186). Momentum at IC was shown to increase significant in the male DB group only (Cohen's d = 0.938, p = 0.041) with no other significant differences evident. There was a significantly higher momentum at IC in the males across all conditions when compared to the female group (p < 0.001).

Table 9. Summary of jump height, velocity at IC, and momentum at IC (± standard deviation, effect size in brackets).

	UL		DB		WV		BB	
	Male	Female	Male	Female	Male	Female	Male	Female
Jump height (m)	0.32±0.04*	0.21±0.04*	0.29±0.04* (0.750)	0.20±0.04* (3.000)	0.27±0.04*# (1.250)	0.18±0.04* (3.500)	0.26±0.04*# (1.500)	0.18±0.04*# (3.500)
Momentum at IC ($kg/m.s^{-1}$)	189.3±11.5*	131.4±12.8*	200.7±12.8*# (0.938)	139.3±14.1* (0.587)	190.9±12.7* (0.132)	133.8±13.6* (0.182)	190.2±15.1* (0.067)	133.0±14.4* (0.118)

* Indicates a significant difference between genders. # Indicates a significant difference from unloaded condition.

5.2.2 Sagittal Kinetic Findings

No statistically significant interaction effects were found for kinetic variables between load condition and gender. Statistically significant gender differences in peak hip extension moment were demonstrated in all loading conditions (unloaded, p = 0.002; dumbbell, p = 0.001; weighted vest, p = 0.016; barbell, p = 0.025). There were no significant differences in peak ankle dorsiflexion moment for DB (Cohen's d = 0.200, p = 0.527), WV (Cohen's d =

0.287, p = 0.345), and BB (Cohen's d = 0.415, p = 0.178) conditions compared to UL. Despite no significant interaction effects between gender and load type, peak knee extension moments followed separate patterns in both genders. Due to large differences in mean changes, pooled effect sizes for both genders are not reported. A summary of peak joint moments during landing and effect sizes in each gender are displayed in Table 10. There were no significant differences between conditions. Total support moment is comprised of the peak hip extension, peak knee extension, and peak plantarflexion moments as a percentage of the total lower extremity support moments. Females increased the contribution from peak knee extension moments to total support moment (TSM) in all loaded conditions (Figure 8). Differences in the percentage contribution of the knee extension moment, hip extension moment, and ankle plantar flexion moment were found not to be significant across conditions.

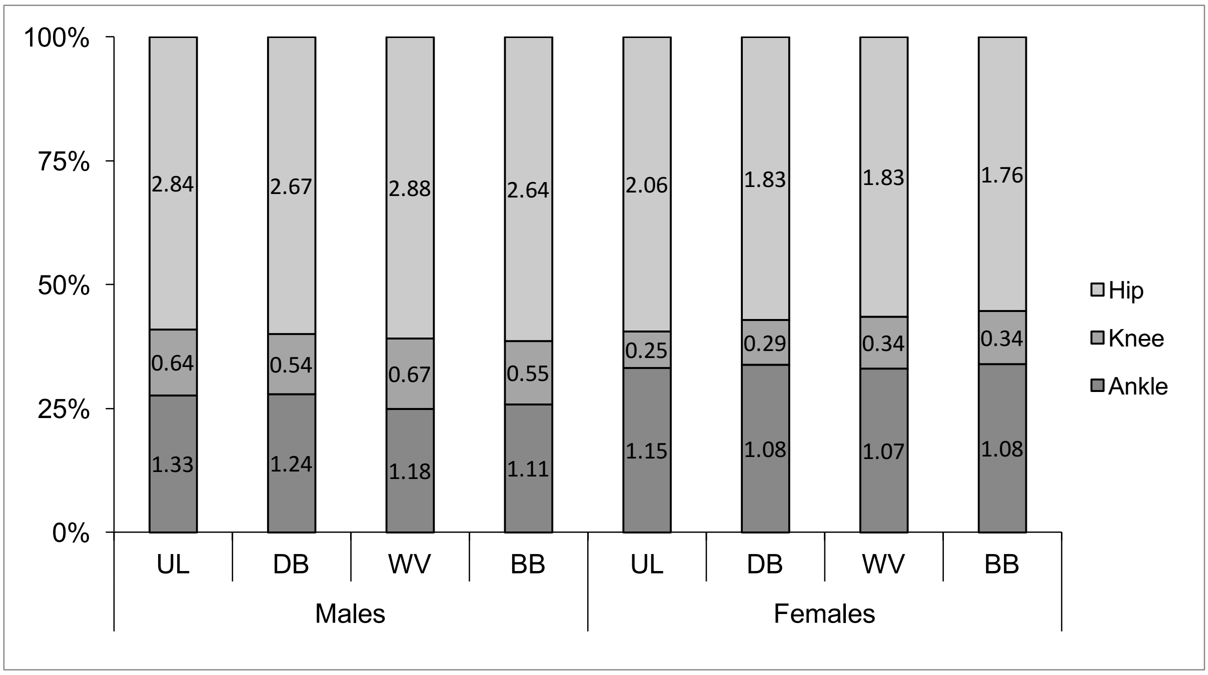

Figure 8. Comparison of peak landing moment and joint contributions between gender and load type.

5.2.3 Frontal Kinetic Findings

Peak knee valgus moment showed no significant differences when compared to the UL condition in DB (Cohen's d = 0.368, p = 0.276) and WV (Cohen's d = 0.160, p = 0.590). There was a moderate effect size for the significant decrease in peak knee valgus moment in the BB condition (Cohen's d = 0.779, p = 0.010). There was a significant decrease in hip adduction moment observed in the BB condition with a large effect size (Cohen's d = 0.839, p = 0.012). Males showed a significant decrease in peak hip adduction moment in the BB condition (Cohen's d = 1.043, p = 0.018). There was no significant difference in the peak hip adduction moment exhibited in the WV condition with a small effect size (Cohen's d = 0.483, p = 0.126) or DB condition (Cohen's d = 0.169, p = 0.496) when compared to the UL condition.

Table 10. Average peak moments landing phase (± standard deviation, effect size in brackets)

Moment	Male				Female			
($Nm.kg^{-1}.m^{-1}$)	UL	DB	WV	BB	UL	DB	WV	BB
Hip								
Extension	2.80±1.15*	2.70±0.88* (0.098)	2.88±1.18* (0.068)	2.64±0.87* (0.157)	2.06±0.52*	1.87±0.43* (0.398)	1.83±0.39* (0.500)	1.76±0.59* (0.539)
Adduction	1.40±0.45	1.30±1.25 (0.106)	1.06±0.28 (0.907)	0.92±0.47# (1.043)	1.13±0.15	1.02±0.26 (0.518)	0.99±0.18 (0.845)	0.81±0.33 (1.248)
Knee								
Extension	0.57±0.18*	0.58±0.08* (0.071)	0.58±0.08* (0.071)	0.54±0.13* (0.191)	0.26±0.09*	0.29±0.13* (0.268)	0.33±0.17* (0.515)	0.33±0.11* (0.697)
Valgus	0.75±0.14	0.70±0.34 (0.192)	0.78±0.40 (0.100)	0.56±0.40 (0.634)	0.84±0.41	0.66±0.28 (0.513)	0.69±0.47 (0.340)	0.49±0.29# (0.986)
Ankle								
Plantarflexion	1.33±0.20	1.24±0.78 (0.158)	1.18±0.59 (0.340)	1.11±0.51 (0.568)	1.15±0.17	1.08±0.16 (0.412)	1.07±0.18 (0.457)	1.08±0.18 (0.400)

* Indicates a significant difference between genders. # Indicates a significant difference from unloaded condition.

5.2.4 Vertical Ground Reaction Force Findings

The female group showed a greater time to peak vertical ground reaction force (vGRF) in all conditions when compared to the male group (p < 0.001). These differences are displayed in Figure 9. Time to peak ground reaction force showed no significant differences between different loading conditions which were trivial when considering Cohen's d effect sizes.

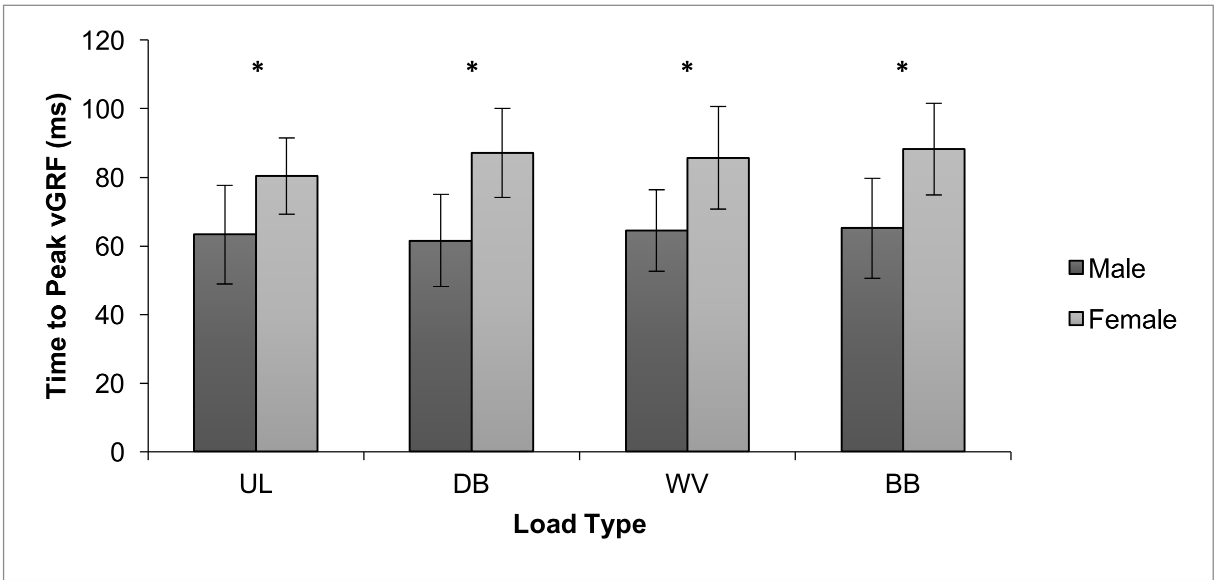

Figure 9. Average time to reach peak vGRF in different loading conditions and genders. * Denotes significance between genders.

The female group demonstrated a statistically significant lower normalised peak vGRF in UL ($p < 0.001$), DB ($p < 0.001$), WV ($p < 0.001$), and BB ($p = 0.007$) conditions compared to the male group. There was a statistically significant decrease in vGRF during the BB condition when compared to the UL condition in males ($p = 0.020$). There was no significant difference in pooled averages across both genders in peak normalised vGRF in the DB condition (Cohen's d = 0.264, $p = 0.252$) and the WV condition (Cohen's d = 0.045, $p = 0.845$) when compared to the UL condition. The average decrease of 3.1 N/kg in normalised vGRF exhibited in the BB condition was significant (Cohen's d = 0.652, $p = 0.004$). Average normalised vGRF for both genders in all conditions are displayed in Figure 10.

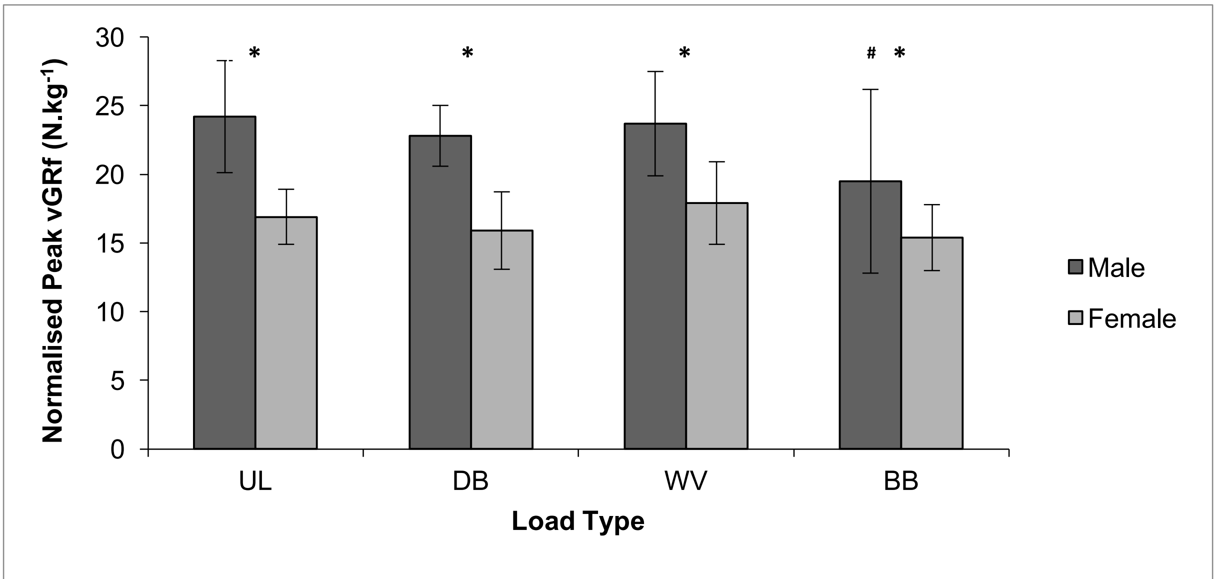

Figure 10. Mean peak normalised vGRF during landing in different conditions and genders.
* Denotes significance between genders. # Indicates a significant difference from unloaded condition.

Chapter Six: Discussion

6.1 Gender Findings

In the current study, there was no significant difference between genders in sagittal ROM and peak flexion angles at the hip and knee, though this was not significant. Sagittal ROM at the ankle was significantly different between genders. Females took significantly longer to reach maximum knee flexion when compared to males. Females exhibited higher hip adduction angles at IC and peak hip adduction angles. This was found to be significant in all loading conditions apart from the DB condition. No significant differences were shown between genders for frontal kinematics at the knee. Males demonstrated significantly greater peak hip and knee extension moments during landing in all loading conditions. There was no significant difference between genders in peak hip adduction moment or peak knee valgus moment. Peak plantarflexion moments in all conditions was not significantly different between genders. Females demonstrated significantly greater times to reach peak vGRF in all conditions compared to males, which was accompanied with significantly lower peak normalised vGRF across all conditions.

There have been several studies that have sought to examine gender differences in lower extremity biomechanics during jump landings (Table 1 and 2). The results presented in this study are in part consistent with existing literature, which have shown that gender differences in lower extremity biomechanics do exist during jump landings. For example: the increased knee and hip extension moment in males; and greater ROM in the sagittal plane in females. However, there are findings presented which contradict and challenge existing literature demonstrating no gender differences in some variables during jump landings. These include an increase in peak normalised vGRF in males, similarities between genders in frontal knee kinematics, and a more flexed posture adopted by females during IC. The discrepancies from the existing literature observed may be due to differences in the task examined in the current investigation (Butler et al., 2013).

Due to the significant differences between genders in some variables the results of this study support the original hypothesis that there would be significant differences in landing kinematics and kinetics between male and female participants. Due to the lack of significant interaction effects between load type and gender, the hypothesis that females will exhibit greater changes to landing kinematics and kinetics in loaded jump conditions when compared to males is rejected.

6.1.1 Sagittal Kinematic Findings: Gender

There was no significant difference between genders in plantarflexion angle at initial contact (IC) during landings, which agrees with the findings of Kernozek et al. (2005) and Cortes et al. (2007). Total sagittal ankle excursion and peak dorsiflexion angles were higher in females than males in all conditions, however there was no significant difference. Existing literature has demonstrated similar increases in ankle sagittal excursion and increased dorsiflexion angles in females compared to males (Decker et al., 2003; Kernozek et al., 2005; McLean et al., 2007). This strategy employed during landing has been previously been suggested to decrease forces at the knee by means of dissipating a greater amount of force at the ankle joint (Decker et al., 2003). This could lead to a decreased risk of injury at the knee due to lower forces acting more proximally in the lower extremity. Energy absorption at each joint was not measured in this investigation so it is not possible to elucidate the relationship between ankle dorsiflexion angle and energy absorption at more proximal joints.

Similarly to previous findings by Gehring et al. (2009) and McLean et al. (2007), males and females landed with similar angles of knee flexion at IC in all conditions. This finding counters previous literature that females adopt a more erect posture during initial contact of jump landings with knees close to full extension (Decker et al., 2003). In previous studies, it has been shown that females utilised an ankle dominant strategy at IC by exhibiting greater amounts of plantarflexion (Decker et al., 2003; Huston et al., 2001). In the current study, average plantarflexion angles at IC in the male group across all conditions (30.3°) were higher than in the study by Decker et al. (2003) where males exhibited an average of 11.3 degrees of plantarflexion. In addition to this, the male group exhibited smaller knee flexion angles at IC of 19.3 degrees when compared to an average 30 degrees of knee flexion in the study by Decker et al. (2003). When considering both studies, it is apparent that the males in the current study exhibited comparably smaller amounts of knee flexion due to a larger amount of plantarflexion. It could be suggested that the males in this study exhibited similar amounts of knee flexion at IC to females due to a greater reliance on ankle musculature to attenuate landing forces towards IC. This may be due to differences in the task examined as both previous studies showing gender differences in knee flexion at IC utilised drop landings (Decker et al., 2003; Huston et al., 2001).

Knee flexion at IC is of significance as ACL injuries are thought to occur during the first 50 milliseconds of landing when the leg is closer to full extension (Krosshaug et al., 2007). Females exhibited greater total excursion and peak angles in knee flexion than males in all loading conditions, though both were not significantly different. These findings agree with several existing studies (Decker et al., 2003; Fagenbaum & Darling, 2003; Gehring et al., 2009). When peak knee flexion values were pooled across all jump conditions, females exhibited an average 9.5° more knee flexion than males which was not significant. There is some conjecture in existing literature regarding the significance of sagittal plane kinematics at the knee on ACL injury risk. Some authors suggest that a greater amount of knee flexion during the landing phase may protect the ACL and patellar tendon against excessive shear loading at the proximal end of the tibia (Fagenbaum & Darling, 2003). Contrary to these postulations, it has been suggested that loading mechanisms in the sagittal plane in isolation are not large enough to be injurious to the ACL (Krosshaug et al., 2007). Increases in knee flexion during landing may lead to decreased risk of suffering from other injuries such as patellar tendinopathy (PT) and patellofemoral pain syndrome (PFPS) (Edwards et al., 2010; Boling et al., 2009). An increase in knee flexion may also change the angle of pull of the hamstring group relative to the tibia, reducing strain on the ACL (Li et al., 1999). Females demonstrated a significantly longer time to reach peak knee flexion when compared to males. This finding contradicts a small number of studies that have reported a shorter time to reach peak knee flexion in females when compared to males in single leg landings (Lephart et al., 2002; Schmitz et al., 2007). This finding suggests that females exhibit altered strategies in knee flexion angular velocity during bilateral landings. This allows for more energy to be absorbed by the knee extensors in the lower extremity during landing and may explain the difference in vGRF between genders.

No significant differences in hip flexion at IC between genders were evident across all loading conditions. This contrasts with several investigations which has shown females land in a more erect posture at the hip (Yu et al., 2006, Salci et al., 2004; Decker et al., 2003). Both sagittal hip excursion and peak hip flexion angles were greater in the female group when compared to males, though these differences were not significant. Flexion of the hip is associated with anterior pelvic tilt. Increased anterior pelvic tilt lengthens the gluteus maximus and hamstring group, affecting the force-length relationship. As the muscles are lengthened with pelvic tilt, their ability to exert a force is augmented which may decrease the requirement on the knee extensors during landing (Kulas et al., 2010). Pelvic tilt was not directly measured in the current study. This has the potential to decrease shear forces caused by the quadriceps at the proximal end of the tibia, and therefore decrease the load placed on the ACL (Yu et al., 2006). Another result of anterior pelvic tilt during landing is the lengthening of the hamstrings, which increases passive tension on the hamstrings group (Blackburn & Padua, 2008). This passive tension created during the lengthening of the hamstrings may act to resist anterior tibial translation, thus reducing the risk of an ACL rupture.

During landing, the knee and hip extensors eccentrically contract to decelerate the center of mass (COM), and act to dissipate the landing forces (Blackburn & Padua, 2009). The gender differences in peak knee flexion, peak hip flexion, and the time to reach peak knee flexion, may be due to a greater amount of eccentric knee and hip extensor strength in the females. If a larger amount of knee and hip extensor eccentric strength is present, it may take longer to stop the downward movement of the body. The greater excursion shown in sagittal angles at the ankle, knee, and hip in females may also act to dissipate forces over greater angular displacement.

6.1.2 Frontal Kinematic Findings: Gender

Both genders demonstrated similar peak knee valgus angles, valgus angles at IC, and similar frontal plane excursions at the knee across all loading conditions. Existing investigations examining knee valgus angles during landings have produced inconsistent results, which may be due to the differences in the task examined (Butler et al., 2013). It would therefore be imprudent to compare frontal kinematic findings of bilateral jump landings examined in this study to unilateral landings or stop jumps. It remains unclear as to why there are inconsistencies in the amount of knee valgus exhibited in difference tasks and this may warrant further investigation. The findings presented herein contradict previous findings that show gender differences in knee valgus during bilateral jump landings (McLean et al., 2007; Pappas et al., 2007; Hughes et al., 2008). The valgus angles presented within this study are high compared to those presented in previous literature (McLean et al., 2007; Hughes et al., 2008). This may have been due to poor strength in the hamstrings or hip abductors in both genders. Measures of strength were not recorded in this study so the mechanism behind the high knee valgus angles is unclear.

The similar valgus knee angles shown agree with three existing studies examining the effects of gender on knee valgus (Cortes, Onate, Abrantes, Gagen, Dowling & Van Lunen, 2007; Wallace, Kernozek, Mikat, Wright, Simons & Wallace, 2008; Orishimo, Kremenic, Pappas, Hagins & Liederbach, 2009). In the studies by Wallace et al. (2008) and Orishimo et al. (2009) it was suggested that the consistency of the knee valgus angles between males and females was due to the training history of the athletes tested. Wallace et al (2008) used males and who could maximally squat a mass of at least 1.5 times their body mass and females who could maximally squat a mass at least equal to bodyweight. Orishimo et al. (2009) used a sample consisting of professional dancers who had an average training age of 12.5 years. It was suggested that years of jump and landing specific training lead to a hip dominant landing strategy which lead to a near linear decrease in knee valgus with increased training age. The current study used a sample of resistance trained recreational athletes. Lower extremity strength was not measured but it is unlikely that the participants possessed similar strength levels to the sample in the study by Wallace et al (2008). The sagittal ankle excursion demonstrated by the females in this study may explain the unexpected similarity in frontal plane kinematics at the knee between males and females. It has been previously been shown that an increase in sagittal ankle excursion leads to decreases in peak knee valgus during a drop jump task (Sigward, Ota & Powers, 2008). In the current investigation, females demonstrated greater sagittal ankle, knee, and hip excursion compared to males, which may have led to a decreased need for compensatory frontal movement at the knee to dissipate landing force.

Females exhibited significantly greater hip adduction angles at IC in all conditions except for the WV condition. This confounds existing literature showing no significant difference of hip adduction angle at IC (Kerzonek et al., 2005; McLean et al., 2007; Orishimo et al., 2009). Feed forward neuromuscular control, which is developed prior to movement, activates muscles around a joint before excessive loading to absorb force or protect passive structures such as ligaments (Beard, Kyberd, Fergusson & Dodd, 1993). It has previously been shown that females exhibit altered neuromuscular control at the hip prior to landing, particularly of the gluteus maximus (Zazulak et al., 2005). The difference in hip frontal angle at IC may be due to altered neuromuscular timings. Although no significance was found, females exhibited greater frontal ROM at the hip. Peak hip adduction angles were significantly higher in females across all conditions except for the DB condition. These findings are in agreement with existing literature (Kerzonek et al., 2005).

Similar to knee valgus angles, levels of hip adduction may be task specific and so the extrapolation of the data from loaded jumps to unloaded jumps may be limited. It has been often suggested that the increased hip adduction angles found in females during dynamic movements is due to inadequate hip abductor strength or recruitment (Wallace et al., 2008). Neither measures of hip abductor strength nor activation levels through the use of EMG were recorded during this investigation so it is not possible to draw conclusions on why there was increased peak hip adduction or hip adduction at IC. Although the significantly greater hip adduction shown in females was in line with the hypothesis, it was unexpected to observe no significant differences in valgus knee angle across all conditions. Increased hip adduction with internal rotation of the femur causes valgus collapse at the knee and therefore an increased risk of injury. This could indicate that females landed with feet closer together, though this was not measured. This could lead to a less stable landing due to a smaller base of support.

6.1.3 Sagittal Kinetic Findings: Gender

Females demonstrated lower peak ankle plantarflexion moments during landing compared to males across all loading condition, though these differences between genders were not significant. Similar findings have previously been demonstrated in studies by Kerzonek et al (2005) and McLean et al (2007). The increase in the plantarflexion moment produced by the male group could be due increased strength of the gastrocnemius. This may have resulted of an altered activation strategy of the ankle musculature leading to an increase in joint stiffness. It has previously been suggested that males may exhibit greater joint stiffness during landing (Butler et al., 2013). Due to the association of joint stiffness and bony injuries it could be theorised that this increased joint stiffness may result in a greater amount of force being dissipated by bony structures (Butler, Crowell & Davis, 2003). This in turn could lead to less stress placed on ligamentous and muscular structures and a decrease in the risk of injury to soft tissue. Future research should more closely examine the energetics of skeletal and non-skeletal structures and the effect on injury risk.

Knee extension moment was found to be significantly higher in males across all conditions. This unexpected finding contradicts existing literature examining jump landings (Chappell et al., 2002; Yu et al., 2006) but is agreement with others (Butler et al. 2013). As joint moments were normalised to bodyweight rather than bodyweight and additional weight in all loaded conditions, this may have resulted in greater normalised moments for the males in all loaded conditions. Due to greater bodyweight, males had greater additional load added in the loaded conditions which was not considered. Knee extension moment has previously been shown to correlate well with proximal tibia shear force during a stop jump task (Yu et al., 2006). This may increase the risk of injury to the ACL. Large knee extension moments have been associated with increased risk of ACL injury (Stearns, Keim & Powers, 2013) and PT (Bisseling et al., 2008). It has previously been suggested that a higher knee extension moment to hip extension moment ratio may contribute to the risk of injury at the knee (Stearns, Keim & Powers, 2013). This measure gives a clear indication of the relative contribution of knee and hip extensors in decelerating the COM (centre of mass) during landings. In this study females demonstrated a smaller knee extension to hip extension ratio, which suggests they used a technique that relied on the hip extensors more heavily than the males. This is contrary to suggestions that females use a knee or quadriceps dominant landing strategy. The smaller knee extension moment exhibited in females relative to hip extension moment may be due to increased hip flexion as explained in section 6.2.1. The increased contribution from the plantarflexion moment as seen in the females may also decrease the reliance on the knee extensors. This finding is consistent with the findings by Decker et al. (2003) that showed females preferred to use ankle musculature to absorb energy to a greater extent than males. The main contributors towards an increase in sagittal plane joint moments are amount of flexion and vertical ground reaction force. In every condition, males demonstrated larger normalised peak vGRF, which was a contributing factor to the increased knee extension moment. It has been suggested previously that due to increased strength of knee extensor musculature, men exhibit greater knee extension moments during landing (Stearns et al., 2013). Though strength was not measured in this current investigation, the difference in strength between genders is well documented (Stearns et al., 2013). Contrary to existing literature, males exhibited more extended postures during landings when compared to females, which may have contributed to the increase in knee extension moment.

Peak normalised hip extension moment was found to be significantly higher in males when compared to females during the landing phase. The corroboration for this finding within existing literature is sparse (Decker et al., 2003; Butler et al., 2013). It has been suggested that the lower extensor moments at the hip present in women are a risk of ACL injury (Hewett et al., 2002), as the large hip extensor musculature absorbs less energy. This consequently leads to more strain being placed on non-contractile structures such as ligaments. Although females exhibited more total hip flexion during jump landings, they adopted a slightly more extended posture at IC. Hip flexion influences hip extension moments by affecting the force-length relationship of hip musculature (Kulas et al., 2010) as well as altering the moment arm from the hip (Blackburn & Padua, 2009). A more extended posture brings the line of action of the trunk and upper body's weight closer to the hip joint centre and therefore decreases the hip extension moment required to maintain the upright position of the trunk. This would be consistent with findings by Kulas et al. (2008) who found that participants who landed with a more extended trunk reduced the ability of hip extensors to perform work. Males also exhibited significantly higher absolute vGRF, which is a key modulator in sagittal plane joint moments. It could also be hypothesised that due to greater levels of strength, males were able to generate greater hip extension moments. This may also explain the decreased sagittal hip excursion shown in females, as males were able to exert a greater hip extension force to decelerate the COM earlier in the landing phase.

6.1.4 Frontal Kinetic Findings: Gender

There were no statistically significant differences in peak normalised valgus moments between genders. These findings are in agreement with the findings of previous investigations (Garrison, Hart, Palmieri, Kerrigan & Ingersoll, 2005; Kerzonek et al., 2008). Conflicting research has found that valgus moments are greater (Chappel et al., 2002; McLean et al., 2007) or lower (Kerzonek et al., 2005) in females when compared to males.

The similarity of knee valgus angles during landing may explain the similarities in frontal knee moments between genders in this study. Increases in knee valgus moments have been shown to contribute to ACL loading and therefore injury (Markolf et al., 1995), as well as increase the risk of developing PFPS (Powers, 2010b). It has been suggested that the risk of an ACL rupture increases when peak knee extensor moment is combined with a peak knee valgus moment or low internal knee varus moment. Due to the paucity of temporal data to accompany these variables, it is not possible to comment on whether gender affects the degree of coupling between knee kinetics.

There were no significant differences evident between genders in peak normalised hip adduction moments during landing, although a trend showed males demonstrated slightly higher peak adduction moments. These findings are in agreement with the findings of Kerzonek et al. (2005) and McLean et al. (2007). This finding is contrary to previous findings of Sigward and Powers (2006) and Pollard et al. (2006) that showed females exhibit a higher adduction moment during side cut manoeuvres. It has previously been suggested that an increased hip adduction moment when combined with weak hip abductors can lead to increased hip adduction and therefore knee valgus (Hewett et al., 2006). Changes in hip adduction moment are typically associated with an ipsilateral lean of the trunk and contralateral pelvic drop in single leg manoeuvres (Pollard et al., 2006). All landings within the current study were bilateral and induced minimal ipsilateral lean of the trunk. Therefore, hip adduction in this study was mainly modulated by the magnitude of vGRF, which was higher in males.

6.1.5 Ground Reaction Force Findings: Gender

Normalised vertical ground reaction force was significantly greater in males in all conditions when compared to females. This finding was unanticipated due to previous findings that showed females demonstrated greater normalised peak vGRF when compared to males (Salci et al., 2004; Kerzonek et al., 2005). The findings agree with a trend found by Decker et al. (2003) showing greater peak normalised vGRF in males but the corroboration for this finding is sparse. The differences in vGRF between genders may be explained in this study by the discrepancies in landing strategies adopted by males and females. Females exhibited larger total joint excursions in the sagittal plane at the ankle, knee, and hip. A more flexed posture during the landing phase has been shown to enhance the ability of the lower body to attenuate landing forces (Blackburn & Padua, 2008). Increased flexion in the ankle, knee, and hip augments the potential of lower body musculature to dissipate landing forces and may act to reduce stress to capsuloligamentous structures. A more flexed posture may also reduce quadriceps activity and decrease anteriorly directed shear force at the proximal end of the tibia (Decker et al., 2003). The greater normalised vGRF shown in the male group may be due to the greater absolute load used by the male participants, which was not included in the normalisation of kinetic variables.

Females also demonstrated a statistically significant increase in the time to reach peak vertical ground reaction force in all conditions. There have been a limited number of studies examining the difference in time to peak vGRF between genders but have found similar findings, though not significant (Afifi & Hinrichs, 2012; Decker et al., 2003). Although the female group landed with a more erect posture than the male group in all conditions, they demonstrated greater ROM in the sagittal plane at the ankle, knee, and hip. This may have led to a more gradual deceleration of the centre of mass, leading to a delayed peak vGRF.

6.2 Load Type Findings

When considering the data in chapter 5 the results suggest that additional load has the potential to affect landing strategies in moderately trained athletes. Furthermore, the data suggests that different methods of adding load during jumps may affect landing strategies by different magnitudes. There was no significant interaction effect between different load type and genders with both males and females changing landing strategies in similar ways for a majority of variables. To the author's knowledge, this is the first investigation seeking to elucidate differences in landing strategies as a result of using different loading modalities. All loaded jumps resulted in a significant increase in the time to reach peak knee flexion. Significant decreases were observed in peak hip adduction moment and peak knee valgus moment in the male BB condition and female BB condition respectively when compared to UL jump landings. There was a significant decrease exhibited in vGRF in both genders in the BB condition when compared to the UL condition.

The results presented in the current study mainly contradict existing literature examining the effects that additional load has on landing strategies (Table 3). Examples of incongruous findings in this study include: similar sagittal plane variables at peak and IC; decreased peak normalised vGRF in the BB condition; and decreases in demand on joints as measured by moderate to large decreases in peak normalised joint moments, particularly in the BB condition. The findings of increased time to peak knee flexion and increased time to peak vGRF were consistent with existing literature. No previous literature has investigated frontal kinematics or kinetics and so comparison to existing findings is not possible. The findings presented suggest that additional loading may alter frontal kinetic and kinematic variables.

Due to the few significant differences between load types in some variables the results of this study in part support the original hypothesis that there would be significant differences in landing kinematics and kinetics between load conditions. Due to differences in effect size between conditions in all variables and significant differences evident in only the BB condition, the hypothesis that different loading positions will affect landing kinematics and kinetics by significantly different magnitudes is accepted.

6.2.1 Sagittal Kinematic Findings: Load Type

Across both genders, there was no significant difference in sagittal angles at IC at the ankle, knee, and hip across all conditions. It was unexpected to find no significant difference in sagittal angles at IC or peak in the dumbbell (DB) condition when considering momentum at IC in the DB condition in both genders. Both showed large effect sizes between DB and UL conditions with momentum at IC being significantly greater in males. It has previously been shown that participants who land with greater momentum at IC by manipulation of increasing drop height exhibit increased hip and knee ROM as well as increased joint stiffness during drop landings (Ford, Myer, Schmitt, Uhl & Hewett, 2011). The more extended posture of the lower extremity adopted was thought to be an anticipatory technique to avoid collapse of the lower extremity during jump landings with greater demand (Brown et al., 2016). This strategy is likely a result of feed-forward neuromuscular control, which activates muscles around the ankle and knee to absorb force and decrease stress on the ligaments (Beard et al., 1993). This strategy was not apparent in the current study and may suggest that the additional load using dumbbells was not sufficient to alter landing technique in the sagittal plane.

Jump landings in the weighted vest (WV) condition also showed similar sagittal angles at IC and peak to the UL condition. This was an unexpected finding as it was thought the participants would use a greater amount of flexion at the hip and knee, as well as dorsiflexion to attenuate the probable increase in landing forces due to additional load. There were similar amounts of hip flexion shown in both DB and WV conditions when compared to UL which suggests the risk of ACL injury may not be increased in these conditions. Low levels of hip flexion may result in a greater risk due to higher compensatory knee moments from the quadriceps. This may act in combination with potentially higher vGRF to anteriorly accelerate the tibia and increase shear forces acting on the ACL (Ball, 1999). Both genders showed no significant differences in momentum at IC with trivial effect sizes due to smaller jump heights achieved in the WV condition when compared to the UL condition. It is proposed that when compared to the UL condition, the WV condition did not elicit greater landing demand and so altered strategies in the sagittal plane were not required to dissipate greater amounts of energy.

Barbell (BB) loaded jump landings demonstrated a more flexed posture during the landing phase when compared to the UL condition, particularly at the knee and ankle, across both genders without large differences at IC. The more flexed position of the body during jump landings in the BB condition was in line with the original hypothesis. When taking into consideration jump height, the BB condition elicited the lowest jump height, so the anticipatory extension of the lower limb as seen in the DB and WV condition may not have been necessitated at IC. The increase in ROM may effectively dissipate additional forces through a greater range of angular displacement, thus decreasing risk of excessive and potentially injurious forces acting on the lower extremity (McNitt-Gray, 1991). As discussed in section 6.2.1, increases in hip and knee flexion can both act to decrease proximal tibial shear force and therefore decrease the strain placed on the ACL (Li et al., 1999; Blackburn & Padua, 2008).

Time to peak knee flexion increased in all loaded conditions when compared to the UL condition in both genders with larger increases evident in the BB condition. Increases in the time to peak knee flexion in WV and BB conditions were significant. This finding corroborates previous literature examining the effects of body armour (Sell et al., 2010). With marginal increases in momentum as shown in all loaded conditions, it takes a greater amount of angular displacement to stop downward movement of the COM. Previous literature examining the effects of additional load on sagittal kinematics during jump and drop landings has produced inconsistent findings due to differences in methodology. Additional load has been shown to cause a more extended posture in the sagittal plane (Janssen et al., 2012; Brown et al., 2016). Findings from Kulas et al. (2008) suggest that responses to additional load are intensified by trunk position adaptation. In the current study, participants were grouped by gender and so comparison is not possible. The current findings agree with the findings by Sell et al. (2010), where air assault soldiers exhibited a more extended posture at IC followed by greater flexion angles in the lower extremity with additional load during drop-landings. The discrepancies shown in peak sagittal plane kinematics between load conditions may be explained by the placement of the load. During the BB and WV conditions, the trunk COM shifted superiorly when compare to UL and DB conditions. In order to maintain stability during landing, the body must use greater amounts of knee and hip flexion to allow the additional load to sit over the midfoot. The more flexed posture during landing could be due to the position of the barbell being more familiar to the participants. The posture adopted during a BB loaded jump landing is similar to that of a back squat. Jump landing technique involving a greater amount of flexion may have been better coordinated in the BB condition due to greater familiarity with the back squat.

6.2.2 Frontal Kinematic Findings: Load Type

Both genders exhibited trivial increases in peak knee valgus angle in all loaded conditions compared to the UL condition. These were unanticipated findings as it was previously hypothesised that increased load would lead to an increase in peak valgus angles. It was thought that an increase in additional load to the body would increase landing forces through increasing the overall mass, and therefore momentum of the body as it makes contact with the ground. If larger landing forces are not sufficiently dissipated through the sagittal plane, movements in the frontal plane may be used to attenuate landing force. In this study, this strategy was not evident and could be due to similar momentum values at IC (Table 9). There were no significant differences found in knee frontal ROM between UL and both DB and WV conditions. There were small decreases in frontal knee ROM in the BB condition when compared to the UL condition. This may have been due to a more efficient dissipation of energy and landing forces through the sagittal plane as both genders exhibited a more flexed landing posture. The large standard deviation in peak knee valgus angles and frontal knee ROM recorded in all conditions and both genders may have masked potential differences between conditions. Males and females showed decreased knee valgus angle at IC across all loaded conditions with greater decreases evident in females when compared to males. This contradicts existing research by Sell et al. (2010) which demonstrated trivial differences in knee valgus between soldiers landing with and without body armour.

Additional loading in the loaded conditions affected the frontal hip kinematics in females differently compared to males (Table 8). To the author's knowledge, this is the first study to examine differences to hip adduction caused by additional loading. Both genders exhibited increases in hip frontal excursion. This may be due to the increased demand on hip musculature to decelerate additional mass, which could not be accommodated in the sagittal plane. Females exhibited smaller hip adduction angles at IC in WV and DB conditions when compared to UL condition. This finding is consistent with previous literature examining the effects of increased demands on lower body musculature by increasing drop height (Dickin et al., 2015). This strategy seen in females may have been an anticipatory mechanism that recruited the hip abductors before ground contact to prevent inwards collapse of the knees during landing. Increased hip flexion decreases capacity of the gluteus medius to exert an abduction moment. As the hip flexes, the moment arm of the gluteus medius reduces, leading to a reduction in the abduction torque it can produce (Dostal, Soderberg & Andrews, 1986). Both males and females demonstrated more flexed positions at the hip, which could have led to the reduced capability of the gluteus medius to abduct. This may expound the increases in hip adduction at IC and peak hip adduction in the BB condition.

6.2.3 Sagittal Kinetic Findings: Load Type

Internal ankle plantarflexion moments decreased in both genders as a result of additional load. This finding conflicts with existing research by Kulas et al. (2008). The participants in the study by Kulas et al. (2008) increased vGRF by an average 12.5% during loaded landings. The magnitude of vGRF is a key modulator of joint moments during jump landing manoeuvres. All load types in this study elicited a decrease in peak normalised vGRF in both genders which was more pronounced in DB and BB conditions, and could explain the decrease in plantarflexion moment. The decrease in plantar-flexion moment may explain the increases in ankle dorsiflexion and sagittal ankle ROM seen in some of the loaded conditions. A decreased plantarflexion moment during landing may be a strategy employed to attenuate greater mass by permitting a greater ROM at the ankle. It is unclear from the data gathered in this study whether the decreases in plantarflexion moment shown may be advantageous in the prevention of injuries. When considering the relative contribution of ankle plantarflexion moment, males decreased the percentage contribution to total support moment (TSM) in all loaded conditions. This may allow for greater contribution from larger proximal musculature at the hip to absorb more energy during landing. The percentage contribution of plantarflexion moment by females to TSM remained unchanged.

Males and females demonstrated different peak knee extension moments as a result of additional loading. Males showed no change in the amount of peak knee extension they used during landing in any loaded jump conditions. Females increased the relative contribution of the knee extensors to a greater extent than males in all loaded conditions. A quadriceps dominant strategy has been cited as a contributing factor to increased risk of ACL rupture in females previously (Salci et al., 2004; Zazulak et al., 2005). Although muscular activity was not quantified, the increase in internal knee extension moment relative to hip extension moment strongly suggests that females utilise a quadriceps dominant strategy when landing with additional load. This is in agreement with literature showing a greater utilisation of the quadriceps group in females when compared to males in dynamic manoeuvres (Malinzak et al., 2001). This may be due to lower levels of hip extension strength relative to knee extensor strength in females, which has previously been demonstrated (Stearns et al., 2013). No previous literature examining additional load has reported knee extension moments and so comparison is not possible.

In all loaded conditions, males utilised a proportionately greater contribution of peak hip extension moment to TSM when compared to the UL condition (Figure 8). This strategy makes use of larger musculature around the hip and act to attenuate landing forces more proximally in the lower extremity (Decker et al., 2003). This strategy could act to decrease demand on musculotendinous and ligamentous structures at the knee and ankle, decreasing the risk of injury. Further research should be carried out to examine the relationship between joint contributions during dynamic manoeuvres and injury risk. There was a decrease in hip extension moment in the DB condition in males and females. The author associates this finding with the decreased hip flexion shown at IC. As described in section 6.2.2, the more extended position at the hip can affect the extension moment but alterations to hip muscle function, and changes in the moment arm length between the COM of the trunk and the hip joint centre. The decreased hip extension moment in the BB condition may be due to a delay in the onset of peak vGRF. This strategy may have decreased the demand of the hip extensors by increasing the angular displacement in which landing forces were attenuated.

6.2.4 Frontal Kinetic Findings: Load Type

Females and males exhibited similar alterations to external hip adduction moments with additional loading, but exhibited differences in the changes to external knee valgus moments. To the author's knowledge, this is the first study to examine changes to lower extremity frontal kinetics as a result of additional load. As such, comparisons to existing literature cannot be made. External knee valgus moments decreased in all loaded conditions with the exception of the male WV condition where a small trivial increase was noted. Females exhibited a greater decrease in knee valgus moment in the DB, WV, and BB condition from the UL condition when compared to males. Both genders exhibited the lower valgus moments in the BB condition. High knee valgus moments have previously been shown to increase tensile strain on the ACL (Garrett & Yu, 2007) and have been associated with the development of PFPS (Myer et al., 2015). The external knee valgus moment during landing is a product of vGRF and the moment arm between the knee joint centre and the vGRF vector (Figure 11). A decrease in knee valgus angle during landing, would decrease valgus moment assuming vGRF is constant. In the current study vGRF decreased in all loaded conditions, which created a greater decrease in knee valgus moment. Although both genders exhibited largely greater peak valgus angles during landing, there were smaller knee valgus at IC. Due to the probable temporal proximity of IC and peak vGRF, external knee valgus decreased as a result of a smaller moment arm due to smaller knee valgus angles where vGRF was greater. Greater decreases of peak knee valgus moment were shown in BB were the result of a combination of smaller knee valgus angles near the beginning of the landing phase and smaller peak vGRF.

Peak hip adduction moment was shown to decrease in conditions with additional loading in all conditions with greater decreases in the WV and BB condition. As mentioned in section 6.2.4, increased hip adduction moment when combined with weak hip abductors can lead to increased hip adduction and therefore knee valgus (Hewett et al., 2006). The findings of this study suggest that the addition of loading does not increase demand placed on the hip abductor group and may decrease it. Section 6.2.4 describes the effect of ipsilateral lean on hip adduction moment. All landings within the current study were bilateral and induced minimal ipsilateral lean of the trunk. Therefore, hip adduction in this study was mainly modulated by the magnitude and vector direction of vGRF. As peak vGRF was decreased in all loaded conditions, it is likely that hip adduction moment was decreased as a result of this.

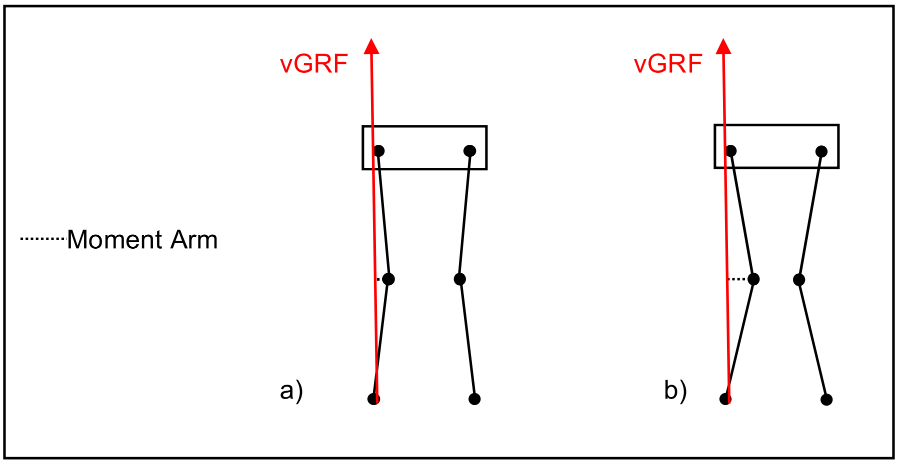

Figure 11. The influence of knee valgus angle and frontal plane loading during landing. Smaller knee valgus angle in figure a) decreases the moment arm and therefore valgus moment when compared to figure b).

6.2.5 Ground Reaction Force Findings: Load Type

It was unexpected to observe smaller vGRF across all loaded conditions in both genders. It was anticipated that increased momentum due to additional load would increase vGRF during landing, as a greater impulse would need to be applied by the lower extremity to stop the COM. As is shown in Table 9, momentum at IC was not increased significantly as a result of additional loading due to smaller displacement of the COM in all loaded conditions. A majority of previous studies have demonstrated increases in vGRF as a result of additional loading (Sell at al., 2010; Dempsey et al., 2014; Brown et al., 2016). All the aforementioned studies controlled for drop height by using drop-landings from the same height. Jenssen et al. (2012) found a non-significant decrease in vGRF, which, similarly to the current study, was displayed alongside a decrease in vertical jump height in the loaded condition. The reason for the decrease in peak vGRF in all loaded conditions is most likely due to an increase in the time to peak vGRF. All loaded conditions caused a delay in the onset of peak vGRF, though this was not significant. According to the impulse-momentum theorem, momentum is equal to impulse (m.v = F.t). By increasing the time (t) in which the momentum of the body was slowed during landing, the average force (F) required to slow the COM upon landing proportionately decreased. This subsequently led to the observed decrease in peak vGRF during landing. Changes to the vGRF in the BB condition were the result of a significantly lower jump height and a greater time to peak vGRF.

6.3 Limitations

The lack of significance in some instances between gender and load type could be attributed to the relatively small sample size used. No electromyographic or strength measures to quantify muscular activity were used to supplement the kinematic and kinetic measures observed. Jump height was not controlled to allow for greater ecological validity, but was significantly different between conditions. A constant load was used, which is comparatively low when considering additional weight used in a performance or military setting (Stone, O'Bryant, McCoy, Celanese, Lehmkuhl & Schilling, 2003; Sell et al., 2010). The study used an inverse dynamics solution, which has numerous limitations that have previously been documented (Riemer, Hsiao-Wecksler & Zhang, 2008). Specific to the current study, the inverse dynamics model assumes that the mass of a segment is concentrated around the centre of mass. In this study, the centre of mass of segments was modified by the addition of load that may not have been accurately accounted for by inverse dynamics. Temporal data was not collected for kinetic variables, which may have been important in discussing injury risk. It has previously been strongly suggested that knee injury mechanisms are multiplanar and the cumulative effect of peak variables may increase injury risk to a greater extent than single planar kinetics (Quatman et al., 2010). As mentioned previously, the extrapolation of this data to other dynamic movements may be limited due to task specific differences in landing biomechanics of the lower extremity.

6.4 Practical Implications

This study provides an examination of the effects of different methods of loading on lower extremity biomechanics during jump landings. This information can be used by strength and conditioning coaches and athletes looking to use loaded jumps in training to inform the prescription of different loading methodologies. It is evident from the results that alterations to technique are consistent between males and females, so gender should not be taken into consideration when deciding which method to use. From the data collected, it is recommended that loaded jumps are performed using a barbell due to the more flexed position during landing and lower peak vGRF. Due to the large standard deviation sizes evident in some variables, practitioners should use the results presented alongside what is observed in the field to determine exercise selection.

6.5 Conclusions

This study has added to a considerable number of studies examining gender differences in lower extremity biomechanics during jump landings. Consistent with existing literature, the existing investigation has demonstrated that gender influences lower extremity biomechanics during jump landings. The findings of this study did not demonstrate potentially injurious landing techniques in the female group that have been previously reported such as: increased knee valgus; a more erect landing posture at IC; and increased normalised peak vGRF. There was no significant interaction between load and gender in all variables measured which suggests that existing gender differences are not exacerbated when load is added by DB, WV, or BB. The findings suggest that the risk of injury is similar for both males or females in each loaded condition.

The data suggest in the current study that the addition load during can alter lower extremity biomechanics in landing from maximal vertical countermovement jumps. Generally, the alteration to landing technique was minimal when comparing between loading conditions and unloaded jump landings. The small alterations to landing strategy may be a result of small alterations to momentum at IC due to smaller jump heights in most loaded conditions. It is suggested that potentially negative implications of additional load during jump landings are for the most part unfounded due to alterations in jump height and therefore similar demands to unloaded jump landings. Based on the findings of the current study, in order to decrease the risk of injury during loaded jumps, the BB is recommended over the DB and WV due to significantly lower peak vGRF and a more flexed posture. As is the case for many applied studies, coaches or athletes should use the results presented alongside what is observed in the field to determine exercise prescription.

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Appendices

Appendix One – Consent Form

UNIVERSITY OF HERTFORDSHIRE
ETHICS COMMITTEE FOR STUDIES INVOLVING THE USE OF HUMAN PARTICIPANTS
('ETHICS COMMITTEE')

FORM EC3
CONSENT FORM FOR STUDIES INVOLVING HUMAN PARTICIPANTS

I, the undersigned *[please give your name here*, *in BLOCK CAPITALS]*

………………………………………………………………………………………………………..…
of *[please give email address]*

…..……………………………………………………………………………………………………
hereby freely agree to take part in the study entitled:

"The effects of gender and load position on lower extremity biomechanics during jump-landings"

1 I confirm that I have been given a Participant Information Sheet (a copy of which is attached to this form) giving particulars of the study, including its aim(s), methods and design, the names and contact details of key people and, as appropriate, the risks and potential benefits, and any plans for follow-up studies that might involve further approaches to participants. I have been given details of my involvement in the study. I have been told that in the event of any significant change to the aim(s) or design of the study I will be informed, and asked to renew my consent to participate in it.

2 I have been assured that I may withdraw from the study at any time without disadvantage or having to give a reason.

3 I have been given information about the risks of my suffering harm or adverse effects. I have been told about the aftercare and support that will be offered to me in the event of this happening, and I have been assured that all such aftercare or support would be provided at no cost to myself.

4 I have been told how information relating to me (data obtained in the course of the study, and data provided by me about myself) will be handled: how it will be kept secure, who will have access to it, and how it will or may be used.

5 I understand that my participation in this study may reveal findings that could indicate that I might require medical advice. In that event, I will be informed and advised to consult my GP. If, during the study, evidence comes to light that I may have a pre-existing medical condition that may put others at risk, I understand that the University will refer me to the appropriate authorities and that I will not be allowed to take any further part in the study.

6 I understand that if there is any revelation of unlawful activity or any indication of non-medical circumstances that would or has put others at risk, the University may refer the matter to the appropriate authorities.

7 I have been told that I may at some time in the future be contacted again in connection with this or another study.

Signature of participant……………………………………..…Date…………………………

Signature of (principal)
investigator……………………………………………………Date…………………………

Name of (principal) investigator *[in BLOCK CAPITALS please]*

Appendix Two – Health Screen

UNIVERSITY OF HERTFORDSHIRE **Researcher:** Ben Hunter
SCHOOL OF LIFE SCIENCE
HEALTH SCREEN 1

Title of Study: **“**The effects of gender and load position on lower extremity biomechanics during jump-landings”
Subject Name:

It is important when having volunteered as subject for this study, and having read the briefing sheet for subjects that you answer the following questions. Please do not answer any questions if you consider them intrusive.

1) Do you suffer from high blood pressure, or any heart problems?

Yes	No

2) Do you often get dizzy, or do you know that you have low blood pressure?

Yes	No

3) When and what did you last eat?

4) Are you under the influence of alcohol or any other psycho-active substance?

Yes	No

5) Have you had a cold or flu in the last two weeks?

Yes	No

6) Are you suffering from any musculo-skeletal injury?

Yes	No

7) Are you currently taking any medication (over the counter, or prescription)?

Yes	No

(you do not need to answer “Yes” if you are only taking oral contraceptives, or if you are an asthmatic with an inhaler available)

8) Have you ever been told that you should not exercise?

Yes	No

9) Do you feel fully fit, and eager to act as subject?

Yes	No

10) Is there any reason, not stated above, why you cannot take part as a subject in this practical?

Yes	No

Signature………………………………………………….. **Date:**

Checked by (Name): **Date:**

Appendix Three – Participant Briefing Sheet

UNIVERSITY OF HERTFORDSHIRE

ETHICS COMMITTEE FOR STUDIES INVOLVING THE USE OF HUMAN PARTICIPANTS ('ETHICS COMMITTEE')

FORM EC6: PARTICIPANT INFORMATION SHEET

Title of study
The effects of gender and load position on lower extremity biomechanics during jump-landings

Introduction

You are being invited to take part in a study. Before you decide whether to do so, it is important that you understand the research that is being done and what your involvement will include. Please take the time to read the following information carefully and discuss it with others if you wish. Do not hesitate to ask us anything that is not clear or for any further information you would like to help you make your decision. Please do take your time to decide whether or not you wish to take part. The University's regulations governing the conduct of studies involving human participants can be accessed via this link:

http://sitem.herts.ac.uk/secreg/upr/RE01.htm

Thank you for reading this.

What is the purpose of this study?

Countermovement jumps loaded or weighted with barbells, dumbbells, and weighted vests are commonly used to improve lower body power and performance. The purpose of this study is twofold: (1) to see whether different methods of weighting countermovement jumps have an effect on landing strategies, and (2) to determine whether different methods of weighting countermovement jumps alter landing strategies of genders in different ways.

Do I have to take part?

It is completely up to you whether or not you decide to take part in this study. If you do decide to take part you will be given this information sheet to keep and be asked to sign a consent form. Agreeing to join the study does not mean that you have to complete it. You are free to withdraw at any stage without giving a reason. A decision to withdraw at any time, or a decision not to take part at all, will not affect any treatment/care that you may receive (should this be relevant).

Are there any age or other restrictions that may prevent me from participating?

You will not be able to participate if you are currently ill or have any disease. You will not be able to participate if you have any musculoskeletal injury. You will be prevented from taking part if you have any history of serious lower extremity injuries or lower back injuries. Those without at least one year of resistance training will also not be able to take part.

How long will my part in the study take?

If you decide to take part in this study, you will be involved in it for a maximum of 2 hours.

What will happen to me if I take part?

On a date of your choosing, you will be asked to come to the laboratory. Here, you will be fully briefed on the procedure, and your height and weight will be taken. Retro-reflective markers will then be placed on you with double-sided tape. You will then be asked to perform 3-5 maximal countermovement jumps in each of the following conditions: hands on hips, holding dumbbells, with a barbell placed across the shoulders, and wearing a weighted vest. The weight of any method of loading will equate to 10% to 20% of your bodyweight.

What are the possible disadvantages, risks or side effects of taking part?

There is a possibility of injuring yourself during the countermovement jumps test, but this will be in a lab-controlled and very safe environment, therefore the likelihood of which happening is rare. You may also experience muscle soreness as a result of the testing or training but this usually subsides within 48 hours.

What are the possible benefits of taking part?

How will my taking part in this study be kept confidential?

Participants will be number coded, all their data will be stored on my password protected personal computer, and access to all data will be restricted to the researcher and supervision team.

What will happen to the data collected within this study?

Results of the study will be used for data analysis, a final report, and publishing. Data will be destroyed 6 months after the examination process has concluded.

Who has reviewed this study?

This study has been reviewed by:

The University of Hertfordshire Health and Human Sciences Ethics Committee with Delegated Authority

The UH protocol number is LMS/PGR/UH/02335

Who can I contact if I have any questions?

If you would like further information or would like to discuss any details personally, please get in touch with me by email: *benhunter92@hotmail.co.uk*

Although we hope it is not the case, if you have any complaints or concerns about any aspect of the way you have been approached or treated during the course of this study, please write to the University's Secretary and Registrar.

Thank you very much for reading this information and giving consideration to taking part in this study.

Appendix Four – Raw Data

Gender	Load Type	HipFlexAngle_Peak	Hipaddabdangle_peak	KneeValgusAngle_Peak	AnklePFDFAngle_Peak	HipFrontal_ROM	KneeSagittal_ROM	KneeFrontal_ROM	Ankle_Sagittal_ROM
1	1	55.181	-8.622	-22.619	11.161	4.967	44.5	21.969	35.983
1	1	78.271	-13.824	-14.133	11.796	4.931	53.112	7.482	42.681
1	1	78.612	-6.215	-29.872	11.474	4.044	54.311	26.354	49.24
1	1	79.056	-10.235	-11.372	15.052	4.744	61.888	7.717	46.235
1	1	101.384	-18.416	-13.841	9.419	4.165	56.46	19.578	31.899
1	1	96.481	-19.797	-20.53	8.233	6.032	57.769	18.769	43.778
1	1	39.34	-8.643	-16.76	13.501	6.704	30.648	4.296	52.111
1	1	44.374	-8.741	-18.317	15.817	6.918	27.396	16.936	39.284
1	1	104.514	-9.033	-7.128	16.404	6.94	73.42	12.821	43.249
1	1	89.482	-12.89	-12.49	23.392	6.431	83.884	12.812	48.389
1	1	70.772	-10.262	-5.03	6.156	6.083	47.58	3.062	40.17
1	1	33.136	-11.701	-26.401	13.665	5.621	50.624	17.641	48.449
1	2	55.681	-11.66	-22.321	10.061	4.704	47.776	22.21	37.252
1	2	78.771	-14.518	-13.805	10.696	8.047	58.279	5.927	37.497
1	2	79.112	-9.713	-31.2	10.374	3.678	51.572	22.668	43.248
1	2	79.556	-8.045	-15.477	13.952	4.49	62.517	9.288	45.645
1	2	101.884	-12.579	-13.256	8.319	6.545	48.373	16.024	35.954
1	2	96.981	-15.724	-23.12	7.133	10.664	76.768	19.894	41.023
1	2	39.84	-8.103	-17.08	12.401	2.724	33.73	8.816	48.926
1	2	44.874	-9.182	-20.976	14.717	5.237	36.366	17.223	37.426

1	2	105.014	-13.279	-7.697	15.304	7.226	47.604	9.71	36.838
1	2	89.982	-12.682	-9.513	22.292	7.701	61.338	11.821	47.526
1	2	71.272	-9.413	-11.562	5.056	4.337	52.74	8.009	43.712
1	2	33.636	-11.191	-24.737	12.565	3.754	49.882	15.043	41.666
1	3	62.053	-11.784	-25.996	12.101	6.502	46.215	26.096	43.06
1	3	56.459	-10.524	-13.123	11.57	5.406	48.257	7.873	40.461
1	3	78.003	-12.224	-34.162	14.822	4.876	55.44	31.181	52.344
1	3	79.584	-9.569	-11.199	14.192	5.869	62.714	6.7	48.874
1	3	93.53	-14.741	-13.927	12.105	6.388	48.959	18.849	35.409
1	3	102.809	-14.714	-19.435	7.501	8.165	65.992	19.08	43.78
1	3	43.623	-9.484	-18.562	12.828	5.172	45.045	5.632	55.584
1	3	56.029	-11.013	-19.337	12.859	7.072	40.171	17.291	33.132
1	3	95.879	-12.274	-11.17	14.849	8.075	69.038	16.677	38.45
1	3	100.244	-16.34	-11.196	23.415	6.077	80.457	15.064	47.897
1	3	76.216	-9.675	-6.481	5.623	6.132	47.332	4.197	38.113
1	3	46.59	-10.735	-23.022	16.136	6.439	53.017	12.267	50.117
1	4	65.262	-9.457	-19.866	17.963	4.159	51.145	21.335	44.212
1	4	96.627	-13.194	-22.519	13.048	9.856	68.359	9.47	46.364
1	4	79.859	-12.184	-17.6564	13.902	4.277	50.465	26.635	49.79
1	4	81.609	-9.801	-20.149	13.118	5.067	66.655	8.193	43.778
1	4	96.129	-15.008	-5.283	13.195	8.523	63.132	19.523	42.324
1	4	97	-18.72	-14.564	15.088	17.148	59.333	39.771	51.012
1	4	49.277	-10.076	-16.288	16.164	2.058	33.81	13.982	54.366
1	4	52.577	-13.207	-12.46	12.313	8.318	35.652	33.614	37.169
1	4	90.171	-17.195	-20.827	17.814	9.284	88.086	36.425	45.919

1	4	93.134	-17.24	-13.713	19.467	11.638	96.957	14.272	46.396
1	4	74.766	-10.594	-17.887	16.685	2.828	46.732	7.681	48.019
1	4	59.785	-11.717	-15.304	10.484	5.186	51.665	4.366	36.512
2	1	112.506	-14.712	-4.379	14.676	11.196	76.327	7.449	45.855
2	1	77.056	-13.251	-10.086	9.189	3.643	51.567	7.217	32.487
2	1	79.567	-17.009	-34.047	11.391	6.556	58.942	18.624	47.539
2	1	46.291	-15.564	-4.1	21.325	4.233	44.299	15.613	55.473
2	1	81.06	-14.111	-16.495	18.342	6.542	62.027	18.884	47.584
2	1	112.657	-16.08	-15.105	14.571	8.439	77.321	8.114	46.124
2	1	111.588	-10.403	-18.626	16.194	9.475	74.914	13.029	35.499
2	1	62.552	-12.279	-18.585	19.187	4.435	56.523	13.904	46.36
2	1	70.886	-13.159	-11.794	21.039	4.511	60.654	7.999	54.459
2	1	72.371	-12.58	-21.236	26.371	8.794	75.147	22.07	51.929
2	1	76.133	-13.262	-13.185	20.055	8.188	66.84	13.424	57.773
2	1	78.32	-15.127	-16.479	16.734	6.451	62.161	19.108	50.148
2	2	108.957	-16.213	-4.094	16.018	6.665	75.269	3.931	46.497
2	2	85.059	-9.539	-14.169	12.78	10.357	57.071	7.49	35.378
2	2	83.682	-15.721	-32.884	10.848	4.56	50.708	26.739	46.296
2	2	55.954	-12.621	-3.113	19.558	5.674	45.835	8.773	53.006
2	2	79.088	-13.076	-18.212	17.969	9.72	65.916	16.198	46.683
2	2	104.676	-13.295	-15.342	14.275	14.815	76.056	9.138	47.153
2	2	109.27	-9.211	-18.109	15.675	9.005	80.097	16.12	34.28
2	2	50.691	-10.722	-21.171	18.415	4.684	46.669	16.564	44.888
2	2	95.029	-14.161	-9.353	17.721	5.75	70.318	8.609	50.441
2	2	74.601	-17.563	-21.744	23.582	11.046	69.68	24.378	48.44

2	2	57.891	-15.229	-20.629	15.387	3.948	53.043	19.466	52.405
2	2	80.283	-12.782	-17.552	15.628	5.72	58.697	12.198	45.778
2	3	109.07	-17.233	-4.178	15.693	6.419	80.053	5.537	44.755
2	3	87.803	-14.785	-14.686	9.007	7.517	59.588	12.956	36.455
2	3	96.494	-17.632	-33.218	10.633	7.656	47.608	26.691	48.558
2	3	49.924	-11.904	-5.49	21.051	3.68	49.419	10.006	52.779
2	3	81.028	-14.707	-17.699	18.459	7.267	62.758	16.053	49.873
2	3	118.257	-14.129	-15.489	15.165	12.757	76.395	9.575	48.952
2	3	105.601	-12.808	-20.591	16.514	9.824	74.748	18.222	40.98
2	3	47.236	-12.21	-20.66	20.123	5.214	47.885	16.122	49.071
2	3	83.923	-14.28	-6.69	17.98	4.317	61.596	5.338	53.405
2	3	75.219	-16.952	-19.707	24.395	10.129	78.172	29.846	45.783
2	3	73.706	-16.285	-12.676	17.982	6.683	67.718	12.293	57.572
2	3	82.109	-14.762	-17.187	17.648	7.645	63.073	15.847	49.01
2	4	106.265	-16.392	-6.232	21.241	5.164	81.446	7.687	47.029
2	4	85.566	-13.96	-13.179	14.385	5.967	54.621	9.628	43.641
2	4	86.449	-19.444	-27.159	8.845	7.007	59.732	14.099	51.517
2	4	57.72	-15.621	-14.261	15.743	5.372	38.821	10.895	48.738
2	4	96.933	-14.018	-8.577	20.616	5.292	68.53	11.508	53.939
2	4	98.312	-14.477	-18.72	15.248	8.8	74.544	9.997	50.478
2	4	108.235	-15.539	-10.569	21.138	8.604	85.608	19.399	47.942
2	4	74.462	-15.576	-21.549	24.565	6.825	59.567	11.611	45.623
2	4	89.048	-14.47	-19.358	17.169	5.695	63.933	13.916	58.124
2	4	86.173	-16.357	-18.64	22.891	6.349	74.39	19.171	45.079
2	4	83.051	-17.802	-9.87	27.24	4.824	69.935	8.975	55.954

2	4	75.768	-18.647	-12.765	16.342	7.267	64.327	13.411	46.837

Gender	Load Type	TimeToPeakKneeFlex	TimeToPeakGRF	GRF_Peak	HipExtMom_Peak	HipAddMom_Peak	KneeExtMom_Peak	KneeValVarMom_Peak	AnklePFDPMom_Peak
1	1	33.8	13.2	22.308	-2.813	-1.571	-0.582	-0.901	1.431
1	1	53	13.8	19.982	-2.802	-0.916	-0.795	-0.765	1.049
1	1	37.2	13	22.752	-1.821	-0.92	-0.882	-0.7	1.256
1	1	57.5	13	17.223	-3.067	-1.661	-0.626	-0.641	1.084
1	1	45.5	9.5	28.288	-2.196	-1.809	-0.521	-0.861	1.241
1	1	58.25	13.5	20.906	-3.314	-2.14	-0.272	-0.817	1.27
1	1	40.2	15.2	29.735	-2.314	-1.911	-0.355	-0.79	1.42
1	1	38	8.6	30.253	-4.185	-0.939	-0.497	-0.842	1.578
1	1	64.5	16	22.14	-2.036	-1.445	-0.414	-0.541	1.27
1	1	62	14	23.59	-1.791	-1.376	-0.439	-0.475	1.159
1	1	43.75	10.5	26.979	-3.552	-0.732	-0.635	-0.872	1.482
1	1	41.75	11.75	26.72	-3.351	-1.392	-0.794	-0.812	1.703
1	2	53.6	10.6	22.266	-2.329	-1.032	-0.571	-0.711	0.505
1	2	55.667	11.667	20.334	-3.365	-1.942	-0.55	-0.581	1.12
1	2	34	13.6	26.237	-1.419	-0.279	-0.569	-0.454	-0.035
1	2	53.8	13.6	21.433	-2.393	0.08	-0.558	-0.739	1.054
1	2	59.25	11	24.245	-1.559	-2.924	-0.493	-0.714	1.154
1	2	53.4	14	22.806	-2.962	-1.338	-0.625	-1.081	1.503
1	2	43.4	15.6	23.618	-2.486	-2.492	-0.565	-0.15	2.753
1	2	47.6	7.8	26.305	-3.426	-1.771	-0.486	-0.57	2.572
1	2	79.5	16.75	21.09	-1.893	-1.863	-0.605	-0.6	1.128

1	2	58.75	13	19.041	-1.218	-1.492	-0.669	-0.393	0.686
1	2	49.2	11	23.889	-3.238	1.612	-0.55	-1.479	1.321
1	2	53.25	9.25	22.778	-2.293	-2.101	-0.767	-0.933	1.142
1	3	47.512	11	21.789	-1.445	-1.572	-0.673	-1.167	1.261
1	3	45.18	10	27.941	-2.783	-1.35	-0.577	-1.325	2.336
1	3	43.287	14	23.907	-1.864	-0.98	-0.589	-1.361	0.398
1	3	56.715	14	21.535	-2.171	-0.756	-0.486	-0.698	0.27
1	3	61.267	10.75	23.595	-2.032	-1.089	-0.6	-0.859	1.049
1	3	85.046	15.2	18.676	-2.341	-1.032	-0.515	-0.629	1.217
1	3	34.022	16.6	29.396	-2.137	-1.449	-0.508	-0.515	1.508
1	3	42.09	8	24.131	-2.553	-0.916	-0.411	-0.756	1.144
1	3	79.188	17.333	20.224	-1.62	-0.812	-0.634	-0.235	0.676
1	3	76.689	15.2	18.38	-1.308	-1.068	-0.665	-0.315	1.076
1	3	58.9	9.6	29.292	-2.679	-0.612	-0.63	-1.184	1.3
1	3	38.456	13	25.467	-2.137	-1.12	-0.607	-0.358	1.968
1	4	52	11.8	12.861	-1.564	-0.804	-0.537	-0.636	0.759
1	4	63.6	15.8	17.68	-1.85	-0.825	-0.549	-1.166	1.161
1	4	57.5	14	18.966	-1.127	-1.065	-0.597	-0.243	0.12
1	4	61	13.6	17.366	-1.351	-0.661	-0.564	-0.81	1.209
1	4	61.6	10.4	31.236	-2.32	-1.325	-0.455	-0.554	1.425
1	4	69.333	13.333	10.95	-1.503	-2.004	-0.398	-0.611	0.877
1	4	54.8	18.4	24.307	-2.044	-0.775	-0.726	-0.575	1.678
1	4	53	8.25	29.724	-2.58	-0.384	-0.629	-0.385	0.607
1	4	61.667	10.667	11.494	-1.672	-0.613	-0.327	0.347	0.711
1	4	62.8	15	16	-1.239	-0.383	-0.622	-0.302	1.449

1	4	52.25	12.75	19.411	-2.859	-0.794	-0.37	-1.03	1.346
1	4	51.25	12.5	24.287	-2.2	-1.454	-0.683	-0.773	1.948
2	1	91.25	15.25	14.075	-1.449	-0.914	-0.217	-0.835	0.892
2	1	35.5	15.5	14.708	-2.104	-1.208	-0.243	-1.242	1.517
2	1	57.25	18.5	14.45	-2.36	-1.161	-0.478	-0.593	1.238
2	1	34.4	18.8	17.192	-2.131	-1.17	-0.188	-1.186	1.349
2	1	54.587	18	14.203	-2.612	-1.363	-0.282	-0.668	0.963
2	1	61	15	18.003	-2.239	-1.184	-0.186	-1.257	1.039
2	1	70	12.8	17.672	-2.063	-1.09	-0.053	-0.598	1.116
2	1	40.6	12.8	21.125	-2.406	-1.433	-0.242	-0.735	1.021
2	1	41.2	23	16.082	-3.07	-0.838	-0.103	-1.4	1.104
2	1	67.75	17	16.662	-1.733	-1.165	-0.176	-0.522	1.237
2	1	61	19.25	18.527	-0.957	-1.028	-0.197	0.075	0.938
2	1	49.829	17	20.203	-1.512	-0.963	-0.082	-1.068	1.363
2	2	92.8	12.2	15.466	-1.609	-0.999	-0.383	-0.999	0.928
2	2	60	18.2	12.008	-1.272	-0.874	-0.052	-0.923	1.125
2	2	51.2	16.4	19.749	-2.376	-1.461	-0.337	-0.415	1.359
2	2	35	17.4	15.135	-1.205	-1.016	-0.311	-1.013	1.174
2	2	66.385	17.5	17.872	-1.44	-1.321	-0.242	-0.785	1.281
2	2	71.8	20	16.387	-1.878	-1.084	-0.275	-0.974	1.137
2	2	90.8	16.4	20.695	-2.015	-0.987	-0.024	-0.528	1.304
2	2	39.5	13.75	17.853	-2.1	-1.267	-0.224	-0.558	0.915
2	2	66.8	22.4	12.497	-1.64	-0.494	-0.109	-0.661	0.815
2	2	66.2	18	16.931	-1.497	-1.318	-0.442	-0.5	1.13
2	2	54.25	19.5	11.992	-0.506	-0.705	-0.07	-0.086	0.924

2	2	60.385	17	13.908	-1.87	-0.856	-0.188	-0.565	0.895
2	3	84.5	15.25	15.546	-1.38	-0.615	-0.269	-0.728	0.712
2	3	61	15.5	15.283	-1.124	-0.917	-0.372	-1.509	1.182
2	3	109.4	19.6	17.353	-1.337	-1.122	-0.526	-1.148	1.412
2	3	35.2	18.6	16.812	-0.927	-0.917	-0.131	-0.003	1.188
2	3	65.798	17.5	18.031	-1.447	-0.842	-0.184	-0.89	0.993
2	3	69.8	17.4	15.844	-2.211	-1.104	-0.192	-1.21	1.009
2	3	82.75	13.25	22.605	-1.806	-0.955	-0.015	-0.485	0.946
2	3	38	14	23.579	-2.467	-1.226	-0.427	-0.664	1.072
2	3	52	20	13.265	-1.874	-0.883	-0.145	-0.783	0.936
2	3	78.6	16.4	19.125	-1.193	-1.206	-0.27	-0.562	1.255
2	3	67.4	20.8	19.177	-1.012	-1.122	-0.481	0.157	0.907
2	3	66.144	17	17.786	-1.571	-0.929	-0.399	-0.496	1.168
2	4	98	14	10.296	-1.469	-0.592	-0.265	-0.37	0.557
2	4	52.5	20.5	11.917	-1.248	-0.876	-0.118	-0.611	0.974
2	4	73.4	18.2	17.869	-1.023	-1.372	-0.349	0.059	1.113
2	4	34.6	19.6	14.649	-1.352	-0.671	-0.102	-0.473	1.259
2	4	70.619	17	16.402	-1.326	-1.312	-0.309	-0.359	1.243
2	4	81.4	17	16.806	-1.632	-0.503	-0.106	-0.985	1.227
2	4	104.2	15.2	17.303	-1.574	-1.453	-0.16	-0.351	1.297
2	4	51.6	12.6	18.51	-1.033	-0.392	-0.356	-0.147	1.037
2	4	77.6	20.6	13.797	-1.515	-0.596	-0.259	-0.926	0.89
2	4	75.4	15.8	16.462	-1.238	-1.049	-0.312	-0.556	1.122
2	4	84	22.5	16.048	-1.302	-0.619	-0.162	-0.537	0.954
2	4	71.126	19	15.361	-1.426	-0.519	-0.398	-0.59	0.853

Male Jump Height

UL	DB	WV	BB
0.38	0.29	0.24	0.23
0.31	0.26	0.26	0.24
0.28	0.27	0.26	0.23
0.30	0.30	0.32	0.32
0.30	0.26	0.27	0.26
0.35	0.35	0.30	0.24
0.32	0.31	0.24	0.27
0.26	0.30	0.23	0.23
0.32	0.35	0.27	0.31
0.38	0.34	0.34	0.36
0.32	0.24	0.23	0.25
0.28	0.27	0.24	0.23

Female Jump Height

UL	DB	WV	BB
0.20	0.17	0.16	0.16
0.28	0.23	0.23	0.21
0.19	0.19	0.19	0.19
0.18	0.17	0.18	0.16
0.31	0.31	0.26	0.28
0.20	0.19	0.18	0.19
0.22	0.22	0.22	0.21
0.16	0.15	0.14	0.14
0.23	0.16	0.15	0.15
0.21	0.21	0.18	0.17
0.21	0.20	0.18	0.18
0.18	0.18	0.13	0.13

Male Momentum at IC

UL	DB	WV	BB
207.9576041	199.9817787	181.9269542	178.0964841
187.4727707	187.5683255	187.5683255	182.5656019
179.1364351	192.4410415	187.5683255	177.4218738
183.3519868	201.8338671	210.8085988	210.8085988
183.3519868	187.5683255	192.4410415	187.5683255
199.3246242	219.4165463	201.8338671	182.5656019
191.5049045	206.370026	182.5656019	192.4410415
170.3927376	201.8338671	177.4218738	177.4218738

Female Momentum at IC

UL	DB	WV	BB
127.953098	131.2321372	126.0837591	126.0837591
151.3961473	150.0696902	150.0696902	145.5889845
123.6144366	136.1860248	136.1860248	136.1860248
120.9623333	129.5095258	133.2642026	125.6426951
158.7430608	174.8872301	160.1636386	166.2096627
127.5054948	136.9159528	133.2642026	136.9159528
133.7288911	147.3291193	147.3291193	143.9417902
114.0443815	121.6530165	117.5279796	117.5279796

191.5049045	219.4165463	192.4410415	206.370026	136.73441	125.6426951	121.6530165	121.6530165
207.9576041	216.5363872	216.5363872	222.8141042	130.6542527	143.9417902	133.2642026	129.5095258
190.8349855	181.9269542	178.0964841	185.6784201	130.6542527	140.4728036	133.2642026	133.2642026
178.5097833	192.9626745	181.9269542	178.0964841	120.9623333	133.2642026	113.2527949	113.2527949