Restrictions in Ankle Dorsiflexion Range of Motion and its Effect on Landing Mechanics

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Declaration of Authorship

I, Louis Philip Howe, declare that this thesis entitled, `Restrictions in ankle dorsiflexion range of motion and its effect on landing mechanics' and the work presented in it are my own, and has not been submitted in substantially the same form for the award of a higher degree elsewhere, and that the word count does not exceed 80,000 words.

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Abstract

Bilateral landings are a common daily activity, yet the effect of restricted ankle dorsiflexion range of motion (DF ROM) on landing mechanics is not well established. The purpose of this thesis was to investigate the effects of ankle DF ROM restriction on bilateral drop-landing performance. In Chapter 3, the trigonometric calculation method for establishing tibia angle during the weight-bearing lunge test (WBLT) reliably measured ankle DF ROM for both single-limb and the between-limb differences. Chapter 4 provided reliability data for kinematic and kinetic measures of bilateral drop-landings from drop heights equal to 50%, 100% and 150% of a maximal countermovement jump (CMJ), which supported interpretation of the subsequent studies. Chapter 5 investigated the relationship between ankle DF ROM and bilateral drop-landing performance from three heights (50%, 100% and 150% CMJ). Although several *moderate* to *large* relationships were identified between ankle DF ROM and kinematic measures of landing mechanics, drop height did not moderate these correlations. Chapter 6 identified differences in measures of landing performance between individuals with functionally restricted and normal ankle DF ROM. The restricted group were unable to increase ankle dorsiflexion at peak flexion following a fatiguing task, relative to the normal group. However, between-group differences reported in this chapter may represent measurement error and demonstrated little functionally relevance. In the final data chapter, a 4-week combined mobility and strength-training intervention was found to improve WBLT performance more than a strength-training only intervention, resulting in greater ankle plantar flexion at initial contact, ankle dorsiflexion at peak flexion, and sagittal plane ankle joint displacement during bilateral drop-landings. Performing ankle mobility exercises, in combination with a strength-training, facilitated greater reliance on the ankle joint during landing tasks, while strength-training alone placed greater emphasis on the hip to attenuate landing forces.

Key words: ankle dorsiflexion, landing, joint mechanics, mobility, fatigue, weight-bearing lunge test

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Poster presentations

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Chapter 1

Preface

1.1 Introduction

Bilateral landings commonly occur in court sports (Lian et al., 1996; McClay et al., 1994), team sports (Bloomfield, Polman and O'Donoghue, 2007) and winter sports (Bere et al., 2014). Similarly, activities of daily living and occupational tasks require bilateral landings be performed by non-athlete populations (Knapik et al., 2003). During bilateral landings, individuals are exposed to peak vertical ground reaction forces (vGRF) equating to multiples of bodyweight, which must be attenuated throughout the lower extremity to diminish injury risk (Zhang, Bates and Dufek, 2000). In order to dissipate these forces across larger joint surface areas, simultaneous flexion at the ankle, knee, and hip joints following impact with the ground must occur (Yeow, Lee and Goh, 2011; Zhang, Bates and Dufek, 2000). This movement strategy allows the performer to gradually decelerate their centre of mass over a greater range of motion, effectively managing exposure to peak vGRF (Zhang, Bates and Dufek, 2000).

When suboptimal movement strategies are employed during landing tasks, injury risk increases (Carlson, Sheehan and Boden, 2016; De Bleecker et al., 2020; Weiss and Whatman, 2015). Owing to the knee joint role as the primary shock absorber during landings (Devita and Skelly, 1992; Yeow, Lee and Goh, 2009), knee joint pathologies are associated with landings (Weiss and Whatman, 2015), either occurring as part of a traumatic episode (Waldén et al., 2015) or developing over time resulting in overuse injury (Boling and Padua, 2013). Two injuries commonly related to the performance of landing tasks are anterior cruciate ligament injury (i.e. traumatic) and patellofemoral pain syndrome (i.e. overuse)

(Weiss and Whatman, 2015). As each pathology results in prolonged disability (Blønd and Hansen, 1998; Simon, Grooms and Docherty, 2019), landing strategies that emphasise the safe attenuation of vGRF are critical to preserve an individual's short- and long-term health.

A number of prospective studies have identified biomechanical variables in landing technique that elevate injury risk (De Bleecker et al., 2020; Weiss and Whatman, 2015). A consistent finding across the literature is the role 'stiff' landing strategies have on increasing injury risk (Boling et al., 2009; Boling et al., 2019; Hewett et al., 2005; Leppänen et al., 2016; Padua et al., 2015). Individuals who subsequently incur an anterior cruciate ligament injury have been shown to generate approximately 20% greater peak vGRF and display 10.5° less peak knee flexion angles during bilateral landings (Hewett et al., 2005). The axial compression at the knee joint caused by increased peak vGRF with diminished knee flexion results in greater anterior tibiofemoral translation, as the posteriorly declined tibial slope causes the femoral condyle to glide posteriorly relative to the tibial plateau as the joint is compressed (Meyer and Haut, 2005). As a result, a positive relationship ($r = 0.51$) has been identified between peak vGRF and anterior shear forces at the tibiofermoral joint during bilateral landings (Yu, Lin and Garrett, 2006). Therefore, individuals exposed to high peak vGRF when the knee joint is less flexed will experience greater loading to the anterior cruciate ligament due to the structural architecture of the tibiofemoral joint that may cause tissue failure (Laughlin et al., 2011).

Another potential mechanism responsible for increasing anterior cruciate ligament strain during landings is the altered quadriceps function and activation when reduced knee flexion at initial ground contact and at the moment of peak flexion is demonstrated. Decreased knee

flexion angle has been shown to increase the patellar tendon-tibial shaft angle (Beynnon and Fleming, 1998), causing a contraction of the quadriceps to increase anterior tibiofemoral shear forces and, in turn, loading on the anterior cruciate ligament (DeMorat et al., 2004). Importantly, this function of the quadriceps diminishes as knee flexion increases (Draganich and Vahey, 1990). To compound the strain imposed upon the anterior cruciate ligament by the quadriceps when a 'stiff' landing strategy is employed, reduced knee flexion angles at initial ground contact ($r = -0.38$) and peak flexion ($r = -0.68$) is also significantly correlated with greater quadriceps activation (Walsh et al., 2012). Therefore, reduced knee flexion at initial ground contact and peak flexion during landings may elevate anterior cruciate ligament injury risk, secondary to an increased patellar tendon-tibial shaft angle and greater activation of the quadriceps muscle.

The hamstring musculature is an important synergist to the anterior cruciate ligament in resisting anterior tibiofemoral translation (Withrow et al., 2008). However, when landings are performed with less knee flexion, the architecture of the hamstring musculature precludes them from buttressing the tibiofemoral anterior shear forces (Pandy and Shelburne, 1997), with the line of pull of the hamstrings being more orthogonal relative to the tibia (Herzog and Read, 1993). Reduced knee flexion throughout the landing cycle consequently diminishes the effectiveness of the hamstrings in supporting the anterior cruciate ligament for managing the anterior shear forces present during landings (Laughlin et al., 2011; Podraza and White, 2010). Additionally, while activity of the quadriceps increases when stiffer landing strategies are adopted, activation of the hamstring shows no difference (Walsh et al., 2012). Therefore, the quadriceps-to-hamstrings ratio imbalance during a landing associated with decreased knee flexion is compounded by the reduced capability of the hamstrings to support the anterior cruciate ligament in managing tibiofemoral shear forces.

Assessment of knee flexion angle at initial ground contact and peak flexion is also an important consideration when establishing risk for overuse injuries. Military cadets who perform bilateral landings with < 20° knee flexion at initial ground contact have been shown to have twice the risk of developing patellofemoral pain syndrome (Boling et al., 2019). Additionally, decreased peak knee flexion angles during bilateral landings have also been shown to be a risk factor for patellofemoral pain syndrome within a similar prospective study (Boling et al., 2009). Decreased sagittal plane knee joint displacement during landings causes the dissipation of joint reaction forces across a reduced contact area, resulting in greater patellofemoral joint stress (Olbrantz et al., 2017). As patellofemoral joint stress during dynamic activities is a contributing factor for the development of patellofemoral pain syndrome (Brechter and Powers, 2002; Farrokhi, Keyak and Powers, 2011; Powers et al., 2017), landings that incorporate sufficient knee joint displacement have the potential to reduce overuse injury risk (Olbrantz et al., 2017). Therefore, strategies that employ greater sagittal plane knee joint displacement may reduce the risk of overuse injuries from occurring.

Greater peak knee abduction angle (commonly described as knee valgus) during bilateral landings is associated with increased anterior cruciate ligament (Hewett et al., 2005) and patellofemroal pain syndrome (Holden et al., 2017) injury risk within prospective study designs. Knee valgus increases lateral tibiofemoral compression, while decreasing the axial load on the medial side of the joint (Boden et al., 2010). As a result, altered tension in the medial and lateral collateral ligaments allow tibiofemoral joint rotation to occur that causes the lateral tibial condyle to translate anteriorly relative to the femoral condyles and increase anterior cruciate ligament strain (Markolf et al., 1995). Therefore, landings performed with

greater knee valgus increase the risk of anterior cruciate ligament injury (Kiapour et al., 2015).

Excessive knee valgus also increases patellofemoral joint stress (Lee, Morris and Csintalan, 2003) and has been identified as a risk factor for the development of patellofemoral pain syndrome during bilateral landings (Holden et al., 2017; Myer et al., 2010). Knee valgus is associated with femoral internal rotation in closed-chain activities (McLean, Huang and Van Den Bogert, 2005), causing the patella to displace and tilt laterally (Powers, 2003; Powers et al., 2017). Consequently, the lateral facet of the patellofemoral joint compresses against the lateral femoral condyle, while the contact area on the medial side of the joint decreases (Liao et al., 2015). Due to this imbalance in load sharing, valgus collapse during landing tasks contributes to overuse pathologies at the patellofemoral joint by elevating joint stress secondary to decreased articular contact surface area.

Given the importance of knee joint kinematics in drop landing injury risk, the ankle joint can play an important role in minimising the load that proximal joints are exposed to during landings following ground contact. With the ankle joint contacting the ground in a plantar flexed alignment, forefoot contact allows the plantar flexor muscles (e.g. gastrocnemius-soleus complex) to initiate the deceleration of the centre of mass prior to heel contact, dampening the shock absorption demands for the knee and hip joint (Kovács et al., 1999; Rowley and Richards, 2015). In support of the ankle joint's role in ensuring safe landing strategies, Boden et al. (2009) found athletes who subsequently incurred an anterior cruciate ligament injury during landings exhibited significantly less ankle plantar flexion at initial ground contact when compared to control athletes performing similar manoeuvres that did

not result in injury (10.7° and 22.9°, respectively). Following heel contact with the ground, ankle dorsiflexion is also required to facilitate the necessary knee and hip flexion for effective shock absorption (Yeow, Lee and Goh, 2011). Accordingly, a statistical trend ($P = 0.07$) for reduced ankle joint displacement during bilateral landings has been identified prospectively as a risk factor for anterior cruciate ligament injury in athlete populations (Leppänen et al., 2017). Therefore, the ankle joint may have a direct role in ensuring safety during landings by contacting the ground in a plantar flexed position to act as the initial shock absorption mechanism, prior to accessing sufficient dorsiflexion that enables ample sagittal plane knee and hip joint displacement.

The hip joint also contributes to minimising injury risk during landings. Increased hip flexion angle at the moment of peak flexion during bilateral landings is correlated with reduced external knee flexion and abduction moments (Dingenen et al., 2015), the latter being associated with both anterior cruciate ligament injury (Hewett et al., 2005) and patellofemoral pain syndrome (Myer et al., 2010). Coordination strategies that employ additional hip flexion during bilateral landings also result in reduced peak vGRF and quadriceps activation (Blackburn and Padua, 2009). As a consequence, decreased sagittal plane hip joint displacement during bilateral landings has been identified as a risk factor for traumatic knee injuries (Leppänen et al., 2017). Therefore, landings incorporating sufficient sagittal plane hip joint displacement minimises injury risk by supporting the ankle and knee joints in safely attenuating vGRF.

Another factor suggested to increase injury risk is asymmetries in landing strategies (Hewett et al., 2010; Pappas and Carpes, 2012; Schot et al., 1994) and commonly occurs during

bilateral landing tasks in uninjured (Pappas et al., 2012; Schot, Bates and Dufek, 1994) and injured populations (Meyer et al., 2018). Individuals with large inter-limb asymmetry in peak vGRF during bilateral landings may excessively load one leg, thereby increasing the potential risk for traumatic (Hewett et al., 2010) and overuse injury (Schot, Bates and Dufek, 1994). Inter-limb asymmetries in force profiles during bilateral landings are particularly important metrics among post-rehabilitated athletes and non-athletes alike. For example, Paterno et al. (2007) found that a group of female athletes who had returned to sport two years after anterior cruciate ligament reconstructive surgery demonstrated side-to-side vGRF asymmetries during a drop vertical jump. Asymmetries were in favour of the uninvolved limb and resulted in a mean difference of 0.5 x bodyweight in peak vGRF, representing a mean asymmetry index score of 14.3%. Likewise, between-limb differences in lower extremity joint angles at initial contact and peak knee flexion during bilateral drop-landings have been identified in patients six-months following anterior cruciate ligament reconstruction (Meyer et al., 2018).

Collectively, the research evidence discussed above suggests that coordination strategies may exist that minimise injury risk during bilateral landings. Importantly, peak vGRF should remain under a tolerable threshold that prevents excessive loading be imposed on any single anatomical structure (Hewett et al., 2005; Leppänen et al., 2016). To achieve this objective, the ankle joint should contact the ground in a plantar flexed position to allow the lower limb musculature time to attenuate vGRF (Boden et al., 2010; Rowley and Richards, 2015). This should occur with moderate amounts of knee flexion to reduce anterior cruciate ligament loading (Laughlin et al., 2011) and decrease the risk of developing patellofemoral pain syndrome (Boling et al., 2019). Following ground contact with the rearfoot, the ankle joint should dorsiflex to facilitate sagittal plane knee and hip joint displacement, resulting in the

dissipation of joint reaction forces over greater joint surface areas (Olbrantz et al., 2017), while diminishing peak vGRF (Zhang, Bates and Dufek, 2000). This motion should happen whilst preventing the excessive frontal and transverse plane motion of the lower extremity that results in valgus collapse. Figure 1.1 provides a demonstration of sagittal plane joint angles at the ankle, knee and hip joints during bilateral landings.

Figure 1.1. Sagittal plane lower extremity alignment at (A) initial ground contact and (B) peak flexion.

Assessment of landing technique allows practitioners to establish an individual's coordinative competency, potentially identifying individuals who would benefit from the prescription of an exercise programme designed to mitigate injury risk (Myer et al., 2007). While three-dimensional (3D) motion capture is considered the gold standard, two-dimensional (2D)

video analysis is commonly used in practice to measure kinematic variables associated with bilateral drop-landing performance, allowing for the assessment of landing mechanics on a larger scale (de Oliveira et al., 2019; Puig-Diví et al., 2019). Additionally, portable force platforms are widely accessible for clinical use in determining the kinetic consequences of the coordination strategies employed during landings. Prior to identifying differences or changes following intervention in kinematic and kinetic variables associated with bilateral drop-landing performance, reliability of testing measures must be reported. Reliability of 2D video analysis for calculating lower extremity joint angles at initial ground contact and the moment of peak flexion is yet to be quantified during bilateral drop-landings. Additionally, no data exists surrounding sagittal plane joint displacement for the ankle, knee and hip, as well as the influence drop height has on the reliability of 2D video analysis for bilateral drop-landings. Similarly, given that variables such as peak vGRF, time to peak vGRF and loading rate during landings have been associated with injury risk (Bisseling et al., 2008, Hewett et al., 2005; Radin et al., 1991), the inherent error related with these measures must also be identified.

Restrictions in ankle dorsiflexion range of motion (DF ROM) are commonly reported in the literature among healthy individuals (Dill et al., 2014; Fong et al., 2011; Dowling, McPherson and Paci, 2018; Malloy et al., 2011; Smith et al., 2019). Limited ankle DF ROM may act as an organismic constraint during landing activities, resulting in the emergence of suboptimal movement strategies. As the knee joint's capacity to flex during landings is a primary function used for shock absorption (Zhang, Bates and Dufek, 2000), the ankle joint facilitates knee flexion through concurrent dorsiflexion as a strategy to manage vGRF (Yeow, Lee and Goh, 2009). Theoretically, a restriction in ankle DF ROM would prevent the forward displacement of the knee joint and limit knee flexion from occurring. This suggestion is

supported by evidence that ankle DF ROM positively correlates with peak knee flexion (Dowling, McPherson and Paci, 2018; Malloy et al., 2015) and sagittal plane knee joint displacement (Fong et al., 2011) during landings. Furthermore, studies have reported a negative relationship between ankle DF ROM and peak vGRF in both healthy (Fong et al., 2011) and previously injured (Hoch et al., 2015) populations. However, the relationship between ankle DF ROM and peak vGRF is not consistent (Malloy et al., 2015; Whitting et al., 2013). Whilst this has been partly attributed to differences in landing tasks (Dill et al., 2015), at present, it remains unclear why contradictory findings exist between studies.

For individuals with restricted ankle DF ROM, modifications in coordination strategies at various time points during landings may provide an opportunity to prevent excessive loading of the musculoskeletal system. At initial ground contact, acutely altering lower extremity joint angles has been highlighted as a strategy to effectively dissipate forces (Blackburn and Padua, 2009; Rowley and Richards, 2015), with greater ankle plantar flexion at initial contact resulting in lower peak vGRF and loading rates (Rowley and Richards, 2015). Landing with greater ankle plantar flexion at initial contact potentially offsets deficits in dorsiflexion at peak flexion to maintain total sagittal plane joint displacement. This strategy may offer individuals with reduced ankle DF ROM a solution to maintaining peak vGRF at a manageable level. However, there are some associated risks with adopting this strategy, as landing with greater ankle plantar flexion at initial contact can increase the risk of injuring ankle and knee ligaments. This occurs as increased ankle plantar flexion at initial ground contact results in excessive supination of the foot (Wright et al., 2000) and reduced knee flexion (Rowley and Richards, 2015), replicating the injury mechanism for an inversion ankle sprain (Terada and Gribble, 2015) and anterior cruciate ligament injury (Laughlin et al., 2011) respectively. Therefore, if restrictions in ankle DF ROM result in the ankle contacting

the ground with greater ankle plantar flexion, ankle DF ROM may be a modifiable risk factor for lower extremity injury.

Although there have been a number of investigations examining the relationship between ankle DF ROM and knee joint kinematics for landing tasks, few have studied the effects of ankle mobility restrictions on hip joint performance. During bilateral drop-landings, eccentric work performed by the hip joint musculature increases as a function of drop height, indicating that greater demands for force attenuation resulted in elevated hip joint contribution for maintaining peak vGRF below a tolerable threshold (Zhang, Bates and Dufek, 2000). Indeed, cueing of 'softer landings' elicited the same effect on eccentric work done by the hip (Zhang, Bates and Dufek, 2000). Collectively, these findings demonstrate that the hip joint is a major contributor to the dissipation of forces during landing tasks. Therefore, it is possible that, in the presence of reduced ankle DF ROM, the hip joint has the capacity to offer a compensation strategy that effectively attenuates peak vGRF. Consistent with this suggestion, hip flexion angle at initial ground contact increases in the presence of reduced ankle joint angular displacement during bilateral landings (Begalle et al., 2015), a coordination strategy known to lower peak vGRF (Blackburn and Padua, 2008). Given that restricted ankle DF ROM results in reduced ankle joint displacement during landings (Dowling, McPherson and Paci, 2018), there is a feasible relationship between ankle DF ROM and hip joint kinematics.

At present, little evidence exists regarding compensations in landing mechanics caused by restricted ankle DF ROM and the interaction of factors associated with bilateral landing performance under varying conditions. As sagittal plane joint displacement at the ankle, knee

and hip have been shown to significantly rise when landing from greater drop heights (Zhang, Bates and Dufek, 2000; McNitt-Gray, 1991), restrictions in ankle DF ROM may present as a more compelling constraint when landing from greater elevations, resulting in increased demand for compensatory strategies. Additionally, no studies to date have investigated the influence of fatigue on landing mechanics for individuals with ankle hypomobility. Exercise-induced fatigue has been shown to cause changes in landings mechanics at initial ground contact that are comparable to those identified for individuals with restricted ankle DF ROM (Weinhandl, Smith and Dugan, 2011). Additionally, peak ankle dorsiflexion (Madigan and Pidcoe, 2003) and knee flexion angles (McNeal, Sands and Stone, 2010) have been shown to increase when landings are performed in a fatigued state. However, an ankle DF ROM restriction may prevent this adaptive strategy from being accessed and as a result, individuals with ankle hypomobility may be limited in their ability to alter their landing strategy when fatigued. Collectively, this evidence suggests that landings performed from elevated drop heights or in the presence of exercise-induced fatigue, may present as a greater challenge to individuals with restricted ankle DF ROM, potentially increasing injury risk.

Suboptimal landing mechanics caused by restricted ankle DF ROM may be modifiable. Exercise-based interventions aimed at improving ankle flexibility have been shown to increase ankle DF ROM in $\leq$ 4-weeks (Aune et al., 2019; Jeon et al., 2015; Nakamura et al., 2017). For individuals with ankle DF ROM restriction, increased ankle mobility may offer an opportunity to improve landing mechanics by enhancing the potential for sagittal plane joint displacement at the ankle, knee and hip. In turn, compensations in coordination strategies caused by restricted ankle DF ROM may become redundant, resulting in reduced injury risk during landing tasks. However, to date, no studies have investigated the effect an exercise intervention attempting to increase ankle mobility may have on landing mechanics.

1.2 Aims of the thesis

Taking in to account the literature summarised in the preceding Preface and subsequent Literature Review, the specific aims of the thesis were to:

1) Establish a reliable method for identifying ankle DF ROM, with special consideration towards determining the reliability for inter-limb asymmetries in ankle DF ROM.

2) (i) Establish the reliability of kinematic and kinetic measurements during bilateral drop-landing performance from varying drop heights. (ii) Establish the reliability of measures for inter-limb asymmetries in landing performance.

3) Determine the relationship between ankle DF ROM and landing mechanics during bilateral drop-landings from varying drop heights.

4) Evaluate the effect of acute exercise-induced fatigue on landing mechanics for individuals identified with limitations in ankle DF ROM.

5) Evaluate the effectiveness of an intervention designed to improve ankle DF ROM and its effect on landing mechanics.

1.3 Thesis structure

The primary focus of this thesis was to further the understanding of ankle DF ROM assessment techniques and their relationship to selected aspects of landing mechanics. To accomplish this, the thesis was structured to; firstly, provide the reader with background

understanding of lower-limb assessment techniques and selected biomechanical principles of landing tasks, via critical review and synthesis of the extant literature, leading to the development of research questions. *Chapter 2,* therefore, presents a review of the available literature. To address the identified knowledge gaps, *Chapters 3 to 7* comprise a series of original investigations that, collectively, present the empirical data of the thesis. Specifically, *Chapter 3* establishes the reliability for a measure of ankle DF ROM and inter-limb asymmetries for ankle DF ROM. This chapter presents preliminary normative data for the magnitude of inter-limb asymmetries in ankle DF ROM for a healthy adult population. *Chapter 4* presents reliability data for kinetic and kinematic measures associated with bilateral drop-landing performance from drop heights equating to 50%, 100% and 150% of countermovement jump (CMJ) height. *Chapters 3* and *4* provide the theoretical foundation by presenting the associated measurement error for all variables from which the remaining chapters are interpreted. *Chapter 5* establishes the relationship between ankle DF ROM and bilateral drop-landing performance from drop heights equating to 50%, 100% and 150% of CMJ height using variables deemed reliable from the preceding chapters. *Chapter 6* identifies differences in landing mechanics between groups of recreational athletes with or without a restriction in ankle DF ROM and presents between-group differences for compensations in landing strategies when acutely fatigued. Finally, *Chapter 7* examines the changes in landing mechanics during bilateral drop-landings when ankle DF ROM is improved following the performance of a 4-week intervention aimed to increase ankle mobility. A summary of findings and concluding discussion, including practical implications, limitations and suggestions for future research are presented in *Chapter 8*.

1.4 Ethical procedures and considerations

Prior to data collection for each investigation, an ethics application was submitted and approved by the University of Cumbria Research Ethics Panel (see Appendix 1 for evidence of ethics approval). All participants were verbally informed prior to their involvement of the risks associated with the experimental procedures. Any participant with a history of lower extremity surgery or had lower extremity injury six-months prior to testing was excluded from data collection. Participants were also provided with an information sheet outlining the purpose and procedures of the investigation (Appendix 2), along with a detailed explanation for their right to withdraw, the handling of data collected and the methods used for research data to be reported. Additionally, prior to data collection all participants completed a Participant Consent Form (Appendix 3) and Physical Activity Readiness Questionnaire (Appendix 4) that were approved by the University of Cumbria Research Ethics Panel.

All data were collected at the University of Cumbria Lancaster campus. A certified first-aider was present for all testing and training sessions. During testing sessions, participants were familiarised with the testing procedures prior to data collection. To ensure the safe execution of the strength and mobility exercises by all participants, training sessions (*Chapter 7*) were supervised by an Accredited Member of the UK Strength and Conditioning Association with > 10 years' experience as a Strength and Conditioning coach.

Chapter 2

Literature Review

2.1 Anatomy of the ankle joint and rearfoot

The tibia, fibula and the trochlear surface of the talus form the ankle joint, anatomically defined as the talocrural joint. Distally, the tibia broadens from the shaft resulting in tibial plafond and the medial malleolus covering the majority of the dome of the talus and the medial facet of the talus (Pretterklieber, 1999). As such, the tibia acts as the primary source for transferring loads distally to the foot complex, while providing medial stability to the ankle. Laterally, the slender shaft of the fibula travels distally to form the lateral malleolus and articulate with the lateral facet of the talus (Pretterklieber, 1999). In contrast to the tibia, the relatively smaller contribution of the fibula to the articulation results in less capacity for the transference of forces across the joint due to its narrower structure (Brockett and Chapman, 2016). However, because of the length of the lateral malleolus and its more distal articulation with the talus, it does provide a high degree of meaningful frontal plane stability for the ankle joint (Van Den Bekerom et al., 2008).

Owing to the described structure of the talocrural joint, it is often referred to as the "mortise" joint, as it possesses a strong resemblance to the carpenter's mortise joint. Classified as a synovial uniaxial modified hinge joint, the ankle joint possesses primarily one degree of movement, allowing for plantar flexion and dorsiflexion in the sagittal plane (Brockett and Chapman, 2016; Palastanga, Field and Soames, 2006). Motion in the frontal and transverse plane at the ankle joint is further limited by the collateral ligaments located on the lateral and medial aspect of the articulation (Palastanga, Field and Soames, 2006). Laterally, inversion

motion is checked at the ankle by the lateral collateral ligaments, comprised of the anterior and posterior talofibular ligament and the calcaneofibular ligament (Brockett and Chapman, 2016; Palastanga, Field and Soames, 2006). Medially, the deltoid ligament group, formed by the tibionavicular ligament, the anterior and posterior tibiotalar ligaments, and the tibiocalcaneal ligament, provides passive stability for the ankle joint by limiting eversion motion (Brockett and Chapman, 2016; Hertel, 2002; Palastanga, Field and Soames, 2006).

Although not regarded as part of the ankle joint anatomically, the tibia and fibula articulate to form the distal tibiofibular joint proximal to the ankle joint. This articulation is stabilised by the anterior and posterior tibiofibular ligaments. This joint is further reinforced and bound together through the interosseous membrane, which connects the shafts of the tibia and fibula and acts as an attachment site for many of the muscles that are located in the lower leg. This joint possesses minimal movement and is commonly classified as a synarthrosis joint (Norkus and Floyd, 2001; Palastanga, Field and Soames, 2006). Functionally, this articulation provides the ankle complex with structural stability.

Distal to the ankle joint, the concave undersurface of the talus and the superior convex surface of the calcaneus form the subtalar joint through the articulation of the posterior, middle and anterior facets (Neumann, 2010; Palastanga, Field and Soames, 2006). The subtalar joint is classified as a synovial plane joint, allowing for sliding of joint surfaces across each of the three facets (Palastanga, Field and Soames, 2006). The motion provided by the subtalar joint allows for the foot to move primarily in the frontal and transverse plane through eversion and inversion, as well as abduction and adduction, respectively (Lewis, Kirby and Piazza, 2007). As such, the combined subtalar joint motion of eversion and

abduction are major components of pronation, while subtalar joint inversion and adduction contribute to supination (Lewis, Kirby and Piazza, 2007; Neumann, 2010). Pronation causes 'unlocking' of the talonavicular and calcaneocuboid joints (collectively known as the midtarsal joint), resulting in a mobile and adaptable structure ideal for adjusting to uneven terrain and shock absorption (Blackwood et al., 2005). Conversely, supination leads to 'locking' of the midtarsal joint, producing a rigid structure for the transfer of forces during propulsion (Blackwood et al., 2005).

Muscles that cross the ankle joint anterior to the axis of rotation contribute to dorsiflexion of the ankle (Neumann, 2010). Such muscles include the tibialis anterior, extensor digitorum longus and extensor hallucis longus. Likewise, the gastrocnemius, soleus, flexor digitorum longus, flexor hallucis longus, tibialis posterior, peroneus longus and peroneus brevis cross the ankle joint posterior to the mediolateral axis of rotation and are therefore, plantar flexors of the ankle joint (Neptune, Kautz and Zajac, 2001; Sutherland, Cooper and Daniel, 1980). These muscles function synergistically as plantar flexors at the ankle joint, while the gastrocnemius also crosses the knee joint with its origination on the posterior aspect of the femoral condyles. As such, the gastrocnemius also acts as a knee flexor (Fleming et al., 2001).

2.2 Osteokinematics and arthrokinematics of the ankle complex

The mediolateral axis of rotation for the ankle joint passes through both the medial and lateral malleolus (Neumann, 2010). As the lateral malleolus is located inferiorly and posteriorly relative to the medial malleolus, the axis of rotation diverges marginally away from the sagittal plane (Singh et al., 1992). As a consequence, the mediolateral axis of rotation

deviates in the frontal and transverse plane 10˚ and 6˚, respectively (Neumann, 2010). This results in dorsiflexion being combined with minor, yet significant, amounts of abduction and eversion, while plantar flexion occurs in combination with adduction and inversion (Singh et al., 1992; Neumann, 2010;).

When describing ankle joint motion, consideration should be given to the unobservable rolling, gliding and spinning of the articular surfaces relative to each other (Loundon and Bell, 1996). These arthrokinematic motions allow for a joint segment to display the visible osteokinematic movements. During open-chain ankle dorsiflexion, the convex talus rolls forward as it simultaneously glides in a posterior direction relative to the concave mortise formed by the tibia and fibula (Loundon and Bell, 1996; Palastanga, Field and Soames, 2006; Neumann, 2010). As per the convex-concave relationship (Edmond, 2016), this arthrokinematic motion is reversed during closed-chain activities, with the concave tibia rolling anteriorly while simultaneously gliding in an anterior direction to allow for ankle dorsiflexion to occur (Loundon and Bell, 1996; Neumann, 2010; Palastanga, Field and Soames, 2006). During both open- and closed-chain movements, full ankle DF ROM is achieved with a 1–4 mm spread of the distal tibiofibular joint that allows for the wider anterior trochlea of the talus to achieve its posterior glide within the talocrural joint (Loudon and Bell, 1996). Simultaneously, the distal fibula must also glide in a posterior-superior direction for the mortise to widen (Delahunt et al., 2013). This mechanism of spreading the distal tibiofibular joint not only allows for ankle dorsiflexion to occur, but also acts to tension the interosseous membrane and tibiofibular ligaments resulting in greater joint stability to facilitate the attenuation of large forces (Neumann, 2010).

2.3 Techniques for measuring ankle dorsiflexion range of motion

Traditionally, ankle DF ROM has been measured passively with the knee in either a flexed and extended position (Mosely and Adams, 1991). Manipulating knee joint alignment allows practitioners to determine potential structures that may be limiting ankle DF ROM (Baumbach et al, 2014). This is due to the biarticular gastrocnemius muscle crossing both the ankle and the knee joint, with an extended knee position lengthening the gastrocnemius muscle proximally, causing it to become the primary regulator of ankle DF ROM (Baumbach et al, 2014). With the knee flexed > 20˚, the gastrocnemius muscle slackens and the length of the uniarticular plantar flexors (e.g. soleus, tibialis posterior, flexor digitorum longus and flexor hallucis longus) and joint capsule become the determining factor in ankle DF ROM (Baumbach et al, 2014). When considering knee alignment as part of the testing procedures for measurements of ankle DF ROM, the functional tasks an individual is required to perform should be a factor. For example, during squatting (Swinton et al., 2000) or landing (Zhang, Bates and Dufek, 2000), the knee flexes > 20° prior to peak ankle dorsiflexion being reached. As such, testing ankle DF ROM with a bent knee would likely be more relevant and provide greater insight for the capacity of the ankle joint to dorsiflex during these movements.

To measure passive ankle DF ROM, a goniometer is used, with the central axis aligned with the lateral malleolus and the stationary arm aligned with the fibula shaft (Martin and McPoil, 2005). The mobile arm can be aligned with either the head of the fifth metatarsal (van Gheluwe et al., 2002) or the lateral midline of the foot (Dill et al., 2014). Measuring ankle dorsiflexion ROM using the technique outlined has been shown to possess good reliability, with intraclass correlation coefficients (ICC) for intra-rater reliability > 0.90 (Mecagni et al., 2000; Diamond et al., 1989; Clapper and Wolf, 1988; van Gheluwe et al., 2003) and standard

error of measurement (SEM) values between 0.6° (van Gheluwe et al., 2002) and 2.0° (Diamond et al., 1989). As a result, passive range of motion testing presents as an acceptable means for measuring ankle DF ROM (Martin and McPoil, 2005).

Although ankle DF ROM can be reliably assessed using passive measurement techniques, measures of passive ankle DF ROM do not provide a strong representation of the functional capabilities of the ankle joint to dorsiflex during closed-chain activities (Rabin and Kozol, 2012). Consequently, passive measures of ankle DF ROM show a poor association with lower extremity coordination patterns during functional activities representing daily actions (Dill et al., 2014). As an alternative option, measuring ankle DF ROM can also be accomplished using the weight-bearing lunge test (WBLT), which is simple to administer, requires minimal palpation skills and provides greater validity for functional ankle DF ROM (Dill et al., 2014; Rabin and Kozol, 2012). The WBLT is performed with the participant standing in a split stance facing a bare-wall, with the back foot heel elevated and both hands against the wall to maintain balance. The forward leg is aligned to be perpendicular to the wall, with either the greater toe (Langarika-Rocafort et al., 2017) or second toe (Rabin and Kozol, 2015) in line with the centre of the calcaneus. The participant is instructed to move the knee of the forward leg towards the wall while maintaining contact between the heel and the ground. In order to control for a neutral subtalar joint alignment, the participant is instructed to make contact with the centre of the patella against a vertical line marked on the wall perpendicular to the foot alignment (Konor et al., 2012). To establish an objective value to represent ankle DF ROM during the WBLT, practitioners can measure a variety of distances or angles at the maximum point of dorsiflexion, prior to heel lift. These include:

- Toe-wall distance using a standard tape measure: the distance between the greater toe and the wall (Bennell et al., 1998).

- Tibia angle relative to vertical using either an inclinometer or goniometer: the inclinometer is positioned 15 cm below the tibia tuberosity (Bennell et al., 1998). The goniometer is aligned with the ground (stationary arm) and the shaft of the fibula (mobile arm) (Konor et al., 2012).

- Achilles tendon angle relative to vertical using an inclinometer: the inclinometer is placed 7.5 cm from the most distal portion of the heel vertically along the Achilles tendon (Langarika-Rocafort et al., 2017).

- Tibia angle relative to vertical using the heel-wall distance (horizontal distance between the calcaneal tuberosity and the wall) and the knee-ground distance (vertical distance between the contact point of the anterosuperior edge of the patella against the wall and the ground): tibia angular displacement from vertical is calculated using trigonometric function (tibia angle = 90 – arctangent [knee-ground distance/heel-wall distance]) (Pope, Herbert and Kirwan, 1998).

Comparing ankle DF ROM using the goniometer technique for measuring passive ankle DF ROM and the WBLT technique using an inclinometer to measure tibia angle, Rabin and Kozol (2012) found the two techniques provided significantly different values (passive ankle DF ROM = $24.6 \pm 5.0°$ and WBLT = $49.3 \pm 5.9°$). In their analysis, the two tests shared a moderate association ($r = 0.64$), with passive ankle DF ROM explaining only 36–40% of the variance for the WBLT. When measuring moments applied to the ankle during each measurement method, the WBLT involved a moment 3–4 times greater than the passive ankle DF ROM test (58.9 N·m vs. 16 N·m) explaining the variance in outcome measures between tests (Rabin and Kozol, 2012). Therefore, passive ankle DF ROM fails to provide the ankle

joint with sufficient forces to access full range of motion. As a result, the WBLT provides a more accurate indication as to the ankle joint's capacity to dorsiflex under load as is the case during landing tasks.

Supporting the findings of Rabin and Kozol (2012), Dill et al. (2014) compared the passive ankle DF ROM test against the WBLT for detecting restrictions in ankle DF ROM that negatively affected lower extremity kinematic strategies during functional activities in 40 healthy recreational athletes. Participants were assigned to either a limited or normal group using an arbitrary threshold based on their passive ankle DF ROM and again on their performance for the WBLT. Participants were deemed limited for passive ankle DF ROM when possessing $\leq 5°$ and normal if possessing $\geq 15°$ ankle DF ROM. Using an inclinometer to measure tibia angle for the WBLT, limited ankle DF ROM participants were $\leq 43°$ and normal $\geq 44°$. Interestingly, the same participants did not comprise both groups, with only 70% remaining in the same group (limited or normal) when allocation procedures were changed between testing measures. No differences were found between-groups for lower extremity kinematics during overhead squat, single-leg squat and a jump-landing task using passive ankle DF ROM to distinguish ankle mobility. However, participants with WBLT scores of $\leq 43°$ displayed significantly lower ankle and knee joint displacement in the sagittal plane during the overhead squat and single-leg squat. Therefore, although techniques to measure passive ankle DF ROM may be reliable (Mecagni et al., 2000; Diamond et al., 1989; Clapper and Wolf, 1988; van Gheluwe et al., 2003), values derived from these tests do not provide an accurate representation of the ankle joint's capacity to dorsiflex during loadbearing closed-chain activities.

When determining the concurrent validity of the WBLT as a representative measure of ankle DF ROM, Hall and Docherty (2017) found that toe-wall distance (r = 0.74) and tibia angle measured using an inclinometer (r = 0.76) were positively correlated with 2D motion video analysis. Likewise, Smith et al. (2019) measured tibia angle during the WBLT using an inclinometer for 20 healthy participants, with concurrent validity determined using digital radiographs measuring talar rotation and tibial inclination as the gold standard. Mean ankle DF ROM for tibia angle using the inclinometer was 35.7 ± 6.4°, while the average tibia inclination from radiographs was 36.9 ± 7.6°. The two methods were positively correlated (r = 0.94), whilst at the point of maximum dorsiflexion during the WBLT, 91.8% of the motion were provided by ankle joint motion, with the remaining 8.2% of the movement occurring at distal joint segments within the foot complex. Given that neighbouring joint segments contributed only a small proportion of motion, the WBLT can be considered as a valid measure of ankle DF ROM.

Intra-rater reliability for the WBLT was first established by Bennell et al. (1998) using the toe-wall distance and an inclinometer to measure tibia angle in 13 healthy participants. With seven days between testing sessions, ICCs for the toe-wall distance and tibia angle were 0.97 and 0.98, respectively. The SEM reported for the toe-wall distance and tibia angle were 0.4 cm and 1.4˚, respectively. Bennell et al. (1998) concluded that both measurement methods could be reliably applied to collect objective values representing ankle DF ROM during the WBLT. Konor et al. (2012) investigated the within-session reliability for measuring ankle DF ROM using the WBLT in 20 healthy participants, comparing toe-wall distance and tibia angle measured using both the inclinometer and goniometer. Intra-rater reliability produced ICC values of 0.98–0.99, 0.96–0.97 and 0.85–0.96 for toe-wall distance, digital inclinometer and goniometer, respectively. SEM values were similar to those reported by Bennell et al. (1998)

for toe-wall distance and tibia angle using the inclinometer, ranging from 0.4–0.6 cm and 1.3–1.4˚, respectively (Konor et al., 2012). However, the SEM values for tibia angle using the goniometer were higher than those reported for the inclinometer, ranging between 1.8–2.8˚ (Konor et al., 2012). As a result, Konor et al. (2012) proposed that the inclinometer might be a more sensitive tool for measuring tibia angle during the WBLT. In a systematic review, Powden, Hoch and Hoch (2015) presented intra-rater reliability for measures of tibia angle and toe-wall distance in healthy populations. Twelve studies met the eligibility criteria, with ICC for intra-rater reliability ranging from 0.65 to 0.99 for all measures of ankle DF ROM. Pooled, minimal detectable change (MDC) values for toe-wall distance and tibia angle were 1.9 cm and 4.7° respectively. As such, the authors concluded that the WBLT demonstrated sufficient sensitivity to clinically detect changes in functional ankle DF ROM.

Another method to calculate tibia angle is the trigonometric calculation method first described by Pope, Herbert and Kirwan (1998). As previously described, this method requires the heel-wall distance and knee-ground distance to calculate the tibia angle relative to vertical at the maximum point of ankle dorsiflexion. Langarika-Rocafort et al. (2017) compared the intra-rater reliability for the measurements of WBLT performance using the toe-wall distance, Achilles tendon angle, tibia angle using an inclinometer and trigonometric calculation method with 25 healthy athletes. Across two testing sessions separated seven days apart, ICCs ranged from 0.87 to 0.95 for all methods (Langarika-Rocafort et al., 2017). However, the relative reliability appeared to be superior (ICC > 0.90) for toe-wall distance and the trigonometric calculation method compared to the measurements taken using the inclinometer (ICC < 0.90) (Langarika-Rocafort et al., 2017). Furthermore, the trigonometric calculation method produced lower SEM (1.2°) and MDC (3.3°) values compared to tibia angle (SEM = 2.2°, MDC = 6.0°) and Achilles tendon angle (SEM = 2.3°, MDC = 6.3°)

measured using an inclinometer, indicating smaller measurement error associated with the test (Langarika-Rocafort et al., 2017). Therefore, WBLT performance measured using the trigonometric calculation method may be more reliable than other techniques measuring tibia angle during the test.

Inter-limb asymmetries in ankle DF ROM are most commonly reported as between-limb differences in test scores (Hoch and McKeon, 2011; Rabin and Kozol, 2015). However, reliability of between-limb differences in ankle DF ROM is yet to be reported. As two measurements (e.g. right and left limb) are required for the calculation of asymmetry, the measurement error associated with bilateral differences for ankle DF ROM during the WBLT will likely be greater than the error of measuring a single-limb. At present, there is a lack of evidence as to the reliability of measuring between-limb differences using potentially sensitive methods of determining ankle DF ROM, such as tibia angle using the trigonometric calculation method.

2.4 Restricted ankle dorsiflexion range of motion and possible causes

Restrictions in ankle DF ROM are commonly reported in injured populations (Reid, Birmingham and Alcock, 2007). Localised damage to structures surrounding the ankle joint limits range of motion due to the symptoms associated with the pathology (Reid, Birmingham and Alcock, 2007). However, following return to sport or physical activity, limitations in ankle range of motion may persist even after symptoms have ceased (Cross et al., 2002). A systematic review investigating the epidemiology of ankle injuries amongst elite and recreational athletes found that ankle injuries are the second most common injury across 70 sports, with a weighted percentage of 11.2–20.8% contribution to total injuries incurred, of

which ankle sprains accounted for 33.0–73.0% of all ankle injuries (Fong et al., 2007). The majority of ankle sprains involve pathology of the lateral collateral ligaments (anterior talofibular, calcaneofibular and posterior talofibular ligaments) and result from an inversion mechanism during tasks such as changing direction or landings (Ferran and Maffuli, 2006). Lateral ankle sprains may cause a loss of gastrocnemius and soleus extensibility (Youdas et al., 2009), tightness of the talocrural posterior capsule (Mattacola and Dwyer, 2002), positional faults of the fibula (Hubbard and Hertal, 2008) and reduced posterior glide of the talus relative to the ankle mortise (Denegar, Hertel and Fonseca, 2002). As a result, lateral ankle sprains have been shown to decrease ankle DF ROM as a by-product of reduced musculotendinous extensibility and disrupted arthrokinematics. Consistent with this suggestion, Hoch et al. (2012) found a significant reduction in toe-wall distance during the WBLT for individuals diagnosed with chronic ankle instability when compared to healthy participants (mean difference = 1.74 cm, Cohen's *d* effect size (ES) = 0.52). Furthermore, the loss of ankle DF ROM following the lateral ankle sprain impaired the performance of functional activities in the injured group. Similarly, Crosbie, Green and Refshauge (1999) found individuals who had incurred a lateral ankle sprain presented with reduced ankle DF ROM, causing reduced contralateral step length during jogging. Consequently, individuals who suffered a lateral ankle sprain resulting in a passive measure of ankle DF ROM of < 4° were significantly more asymmetrical in step length than participants with > 4° (Crosbie, Green and Refshauge, 1999). Therefore, the current body of literature suggests that the occurrence of an ankle injury may cause deficits in ankle DF ROM.

Limited ankle DF ROM may also occur in the absence of injury possibly due to functional demands that are imposed on the ankle complex. Rabin and Kozol (2015) identified inter-limb asymmetry in ankle DF ROM among 64 male military recruits with no history of ankle

or foot injury two years prior to data collection. Using the WBLT to measure between-limb differences for ankle DF ROM, the non-dominant limb was found to present with significantly greater ankle mobility than the dominant limb (mean difference = 5.8°, ES = 0.83). Based on their previous work establishing a single-limb measurement error for the WBLT (see Rabin and Kozol, 2012), 45 of the 64 participants (70%) presented with a side-to-side difference in favour of the non-dominant limb that exceeded the MDC value of 4.5°. It was suggested that as none of the participants presented with an injury, the likely cause of the large side-to-side discrepancies was due to differences in functional demands between the two sides. Although limb dominance has not been consistently reported to affect ankle DF ROM (see Lopes et al., 2018), Rabin and Kozol (2015) proposed the non-dominant limb has a greater role in balance and stability, which requires it to possess greater ankle mobility than the dominant limb.

To identify whether changes in mobility occur in response to the demands placed on the ankle joint, variations in ankle DF ROM have been investigated following competition in professional football players. Wollin, Thorborg and Pizzari (2017) examined the acute effects of match play on range of motion measures immediately post-match as well as 24, 48 and 72 h post-match in youth international players. Determined using toe-wall distance during the WBLT, ankle DF ROM decreased following the match, with mean differences ranging from -0.5 cm to -1.0 cm. However, the findings were not statistically significant and as such, fail to demonstrate acute adaptations in ankle DF ROM following competition. In a similar design, Moreno-Pérez et al. (2019) investigated changes in ankle DF ROM using the toe-wall distance established during WBLT immediately following a match and 48 h post-match in professional male footballers over 12 matches. Compared to pre-match values, ankle DF ROM increased in the dominant limb (i.e. kicking leg) immediately following match play

(mean difference = 0.5 cm, ES = 0.16), though no significant increase was observed in the non-dominant limb. While these findings support the suggestion that a task's functional demands may impair ankle DF ROM, it is questionable whether such small differences are detectable using the WBLT. Although Moreno-Pérez et al. (2019) reported acute adaptations, the mean differences did not surpass the MDC values commonly reported for the WBLT within the literature (Powden, Hoch and Hoch, 2015). As such, the WBLT may not be sufficiently sensitive to identify acute changes in ankle DF ROM following competition.

Whilst acute adaptations in ankle DF ROM reported following competition may be questionable, chronic adaptations to functional demands in ankle mobility have been observed. In the same investigation previously described, Moreno-Pérez et al. (2019) also recorded ankle DF ROM during pre-season, mid-season and post-season. Across the course of the season, each professional player completed an average 30.8 ± 9.9 games. When comparing pre-season values to mid-season and post-season, significant reductions in ankle DF ROM were observed for the dominant (mid-season: mean difference = -0.6 cm, ES = 0.25; post-season: mean difference = -1.1 cm, ES = 0.43) and non-dominant limb (mid-season: mean difference = -0.7 cm, ES = 0.25; post-season: mean difference = -0.9 cm, ES = 0.33), respectively. It was proposed that chronic musculotendinous adaptations might have occurred due to the eccentric muscular contractions regularly performed by the ankle musculature during acceleration, deceleration, change of direction and landing tasks over the course of the season. As high-intensity, eccentric muscle actions have been shown to increase musculotendinous stiffness (Seymore et al., 2017), ankle DF ROM could consequently decrease (Moreno-Pérez et al., 2019). Therefore, chronic loading placed on the ankle complex may also result in limited ankle DF ROM.

2.5 Regional interdependence and ankle dorsiflexion range of motion

The concept of regional interdependence was first introduced in literature describing the musculoskeletal injury examination process (Wainner et al., 2007). The premise of regional interdependence is that during a clinical examination, seemingly unrelated impairments at separate anatomical regions have the potential to cause a primary complaint (see Cheatham and Kreiswirth, 2014; Sueki, Cleland and Wainner, 2013; Vaugh, 2008; Wainner et al., 2007). Although clinical assessments should include a principle focus on the local area relevant to the injury, regional interdependence suggests clinicians should widen their scope of assessments to incorporate neighbouring regions that may identify potential deficiencies in physical qualities that could contribute to the pathology (Cheatham and Kreiswirth, 2014; Sueki, Cleland and Wainner, 2013; Vaughn, 2008; Wainner et al., 2007).

When assessing movement, regional interdependence has also been applied as a model that can support practitioners in identifying the primary cause of suboptimal movement patterns (defined as a movement strategy with reduced efficiency and effectiveness, or a strategy that may increase injury risk) (Cook et al., 2014). In this context, a neighbouring joint complex may cause suboptimal joint alignment at another segment during functional movement tasks (Howe and Cushion, 2017). For example, poor thoracic spine posture may disrupt shoulder kinematics during overhead activities (Barrett et al., 2016; Young et al., 1996). Likewise, abnormal hip mechanics can negatively affect knee kinematics during functional movements such as single-leg squatting or gait (Hollman et al., 2014; Heinert et al., 2008; Powers, 2010).

Consistent with the regional interdependence model, restrictions in ankle DF ROM have been shown to affect knee, hip and spine mechanics during closed-chain activities. Ota et al. (2014) showed acute limitations in ankle DF ROM through the application of a custom-made brace resulted in increased knee varus moments during gait in healthy individuals. Likewise, Mauntel et al. (2013) identified a group of recreational athletes with medial knee displacement had significantly less ankle DF ROM when compared to athletes with no medial knee displacement (mean difference between groups = 4.7°, ES = 0.70). During squatting, limited ankle DF ROM has also been shown to reduce squat depth by decreasing sagittal plane knee flexion (Kim et al., 2015). Furthermore, when ankle DF ROM is restricted, increased peak knee valgus angle (Macrum et al., 2012), elevated hip torque (Fry, Smith and Schilling, 2003) and greater spinal flexion (List et al., 2013) have all been observed during bilateral squatting. Therefore, in support of the regional interdependence model, limitations in ankle DF ROM may cause suboptimal coordination patterns throughout the lower extremity during functional activities in order to compensate for ankle joint hypomobility.

2.6 Ankle dorsiflexion range of motion and ankle kinematics during landings

As the ankle joint supports lower extremity strategies to dissipate forces during landings (Zhang, Bates and Dufek, 2000), restrictions in ankle DF ROM have the potential to cause altered landing mechanics (Mason-Mackay, Whatman and Reid, 2017). During landings, sagittal plane coupling between the ankle, knee and hip joint occurs that manipulates the vertical location of centre of mass to allow for the effective dissipation of vGRF. Theoretically, limited ankle dorsiflexion contribution during landings may constrain the capacity of the knee to flex by preventing the forward rotation of the proximal tibia (Dill et

al., 2014; Fong et al., 2011). As the centre of mass must maintain its location over the base of support to prevent falling, knee flexion without concurrent ankle dorsiflexion directs the centre of mass to travel posteriorly relative to the base of support. If knee flexion were to continue without concomitant ankle dorsiflexion, the centre of mass would move outside of the base of support, resulting in a loss of balance. As maintaining balance is a prerequisite for an effective landing strategy, restricted ankle DF ROM may limit ankle joint contribution and prevent neighbouring joints in the lower extremity from flexing (i.e. knee and hip).

To challenge this theoretical hypothesis, it must first be established whether limited ankle DF ROM results in reduced ankle dorsiflexion during landings. Dowling, McPherson and Paci (2018) found a positive relationship between ankle DF ROM and ankle dorsiflexion during landings. In their investigation, participants performed the WBLT followed by single-leg drop jumps from a drop height of 0.14 m, with data for ankle DF ROM and lower extremity kinematics during landings collected using 3D motion capture. Peak ankle dorsiflexion angle during the single-leg drop jump positively correlated with WBLT performance for the dominant ($r = 0.58$) and non-dominant limb ($r = 0.61$), whilst similarly, ankle DF ROM was associated with sagittal plane ankle joint displacement during the landing task for the dominant ($r = 0.38$) and non-dominant limb ($r = 0.42$). Hoch et al. (2015) found comparable relationships between ankle DF ROM measured with the WBLT and ankle joint motion during a single-leg drop-landing from a 0.40 m drop height in participants suffering chronic ankle instability, supporting the association between ankle DF ROM and its influence on ankle joint contribution during landings. Collectively, these findings suggest that ankle DF ROM is related to ankle joint motion during landings, in that those with greater initial joint ROM will utilise this during a landing task.

While the above results describe the association between ankle DF ROM and ankle landing mechanics, these findings have not been consistently reported. For example, Fong et al. (2011) found a non-significant relationship ($r = 0.15$) between a straight knee passive ankle DF ROM and ankle joint displacement during a bilateral jump-landing task from a 0.30 m drop height. Dill et al. (2014) also investigated the influence of ankle DF ROM on ankle joint kinematics during a similar jump-landing task using the methods previously described (see section 2.3). No significant differences were found between groups for ankle joint displacement during the jump-landings, with a *trivial* ES reported (ES = 0.04). These conflicting findings with regards to ankle DF ROM and its influence on ankle joint kinematics during landings may be explained by the different variables analysed between studies. Unlike Dowling, McPherson and Paci (2018), Fong et al. (2011) and Dill et al. (2014) exclusively investigated the effects of ankle DF ROM on ankle joint displacement (as opposed to peak ankle dorsiflexion angle) during landings. As ankle joint displacement accounts for both ankle joint angle at initial ground contact and peak ankle dorsiflexion, it may be that in an attempt to maintain ankle joint contribution, those with restricted ankle DF ROM land with greater ankle plantar flexion at initial ground contact, in turn, preserving ankle joint displacement.

Increasing ankle plantar flexion at initial ground contact to sustain joint displacement is effectively illustrated by Dowling, McPherson and Paci (2018), where no significant relationship between ankle DF ROM and ankle joint angle at initial ground contact was found for either the dominant ($r = -0.10$) or non-dominant ($r = -0.07$) limb during a single-leg drop jump. However, as previously highlighted, the same investigation reported a positive

relationship between ankle DF ROM and ankle joint displacement. As such, it is likely that the participants presenting with reduced ankle DF ROM for Fong et al. (2011) and Dill et al. (2014) compensated by contacting the ground with greater ankle joint plantar flexion. Interestingly, participants with ankle DF ROM restriction did not adopt this strategy for Dowling, McPherson and Paci (2018). One potential explanation for these differences is the landing tasks used. Both Fong et al. (2011) and Dill et al. (2014) had participants perform the bilateral jump-landings from a drop height of 0.30 m, whereas Dowling, McPherson and Paci (2018) used a drop height of 0.14 m for the single-leg drop jump. With increased drop heights resulting in greater time in the air, landing tasks from higher elevations may permit participants additional time to alter ankle alignment prior to ground contact. This preparatory strategy is also consistent with the anticipated increase in vGRF when landing from greater drop heights. This allows participants to maintain ankle joint displacement through contacting the ground with greater ankle plantar flexion. Indeed, ankle joint angle at initial contact and ankle joint displacement are related ($r = 0.95$), thus, greater plantar flexion angles at initial contact increase ankle joint displacement (Begalle et al., 2015). Therefore, ankle DF ROM may affect peak ankle dorsiflexion angle, although ankle joint displacement can be maintained with alterations in ankle joint alignment prior to ground contact. As such, individuals with restricted ankle DF ROM may attempt to maintain sagittal plane ankle joint displacement during bilateral drop-landings by increasing ankle plantar flexion at initial ground contact as a compensatory strategy.

Some reports have questioned the relationship between ankle DF ROM and peak ankle dorsiflexion during landings, as described by Dowling, McPherson and Paci (2018). Whitting et al. (2011) investigated the effect of ankle DF ROM limitations on landing mechanics using an independent groups design. In their study, ankle DF ROM using the WBLT was

determined for 48 physically active males, with participants ranked from the highest to lowest with regards their ankle DF ROM. The 15 middle-ranked participants were excluded from data analysis to establish a distinctive division for the high and low ankle DF ROM groups. Following the performance of a single-leg drop-landing from a 0.32 m and 0.72 m drop height, no differences were found between groups for peak ankle dorsiflexion angle. An explanation for the lack of difference between groups in sagittal plane ankle kinematics may be the technique used for group allocation. Peak ankle dorsiflexion during the landing task was considerably lower than ankle DF ROM capacity (measured via the WBLT) for both groups. It is likely both groups had sufficient ankle DF ROM to complete the landing task, without needing to compensate during the single-leg drop-landings. Studies investigating landing mechanics in individuals with restricted ankle DF ROM should therefore consider the landing task demands before performing group allocation for participants. This may be accomplished by diagnosing participants as being restricted in ankle DF ROM based on whether they possess sufficient ankle DF ROM to perform activities relevant to the task being investigated.

Changes in frontal plane foot mechanics may occur during landings when restricted ankle DF ROM is present. Although both groups in the study by Whitting et al. (2011) possessed sufficient ankle DF ROM to perform the drop-landing task, frontal plane compensations were observed at the foot complex for the low DF ROM group. Subtalar joint eversion angle at the moment of peak ankle dorsiflexion during the single-leg drop-landing was significantly greater for the low DF ROM (14.3 ± 5.0˚) compared to the high DF ROM group (10.1 ± 6.1˚). Furthermore, this was consistent for both drop heights. Although there are a number of possibilities to explain why those with restricted ankle DF ROM compensated their movement in this way, it is possible that this strategy allows for additional knee joint

displacement to facilitate the attenuation of landing forces. Subtalar joint eversion has been shown to "unlock" the midtarsal joint by aligning the longitudinal and oblique axis of rotation of the talonavicular and calcaneocuboid joints parallel to each other (Blackwood et al., 2005). This results in a mobile foot that permits midtarsal and forefoot dorsiflexion to occur (Blackwood et al., 2005), while aiding shock absorption, secondary to foot pronation (Pratt, 1989). As midtarsal and forefoot dorsiflexion would contribute to rotating the proximal tibia forwards, a certain amount of knee flexion would, theoretically, be maintained to support the attenuation of vGRF. However, increased subtalar joint eversion elevates the strain distribution towards the medial portion of the Achilles tendon (Lersch et al., 2012), potentially contributing to elevated risk for the development of tendinopathy (Lorimer and Hume, 2014). Accordingly, ankle DF ROM has been identified as a modifiable risk factor for mid-portion Achilles tendinopathy in a six-month prospective study of 70 healthy male military recruits (Rabin, Kozol and Finestone, 2014).

In summary, restriction in ankle DF ROM is associated with reduced peak ankle dorsiflexion angles during landing tasks (Dowling, McPherson and Paci, 2018) but not necessarily ankle joint displacement (Fong et al., 2011; Dill et al., 2015). This may be due to individuals with restricted ankle DF ROM increasing plantar flexion angle at the ankle at initial ground contact to preserve joint displacement values. At present, little evidence exists correlating ankle DF ROM to initial contact angles at the ankle joint during bilateral landings.

2.7 Ankle dorsiflexion range of motion and knee kinematics during landings

Sagittal plane knee flexion possesses a primary role in reducing injury risk during landings. Reduced peak knee flexion angles during bilateral landings has been shown to increase peak

vGRF (Zhang, Bates and Dufek, 2000), external knee flexor moments (Devita and Skelly, 1992), quadriceps activation (Blackburn and Padua, 2009) and anterior shear forces at the tibiofemoral joint (Chappell, Kirkenall and Garrett, 2002). With each of these variables being associated with knee ligament injury (Griffin et al., 2000), reduced peak knee flexion may lead to greater injury risk (Walsh et al., 2012). As such, reduced knee flexion during landings has been identified as a mechanism for anterior cruciate ligament injuries (Krosshaug et al., 2007).

Ankle DF ROM has been associated with knee joint kinematics during a variety of landing tasks. During bilateral jump-landings, a positive relationship has been reported between ankle DF ROM and knee joint displacement ($r = 0.46$) (Fong et al., 2011) and peak knee flexion angle ($r = 0.39$) (Malloy et al., 2015), indicating limitations in ankle DF ROM are associated with less sagittal plane knee joint displacement. The relationship between ankle DF ROM and knee joint kinematics during single-leg drop jumps were investigated by Dowling, McPherson and Paci (2018), with knee flexion angle at initial contact ($r = 0.33$), peak knee flexion angle ($r = 0.52$), and knee joint displacement ($r = 0.29$) all significantly positively correlated with WBLT performance for the dominant limb. These findings indicate that individuals with restricted ankle DF ROM perform landings by contacting the ground with the knee in a more extended position to compensate for the lack of available peak knee flexion. This results in a somewhat diminished effect for ankle hypomobility on knee joint displacement during landings, with individuals possessing ankle DF ROM restriction pre-empting their limitations by compensating ahead of ground contact. This is similar to the strategy previously described for the ankle joint (see section 2.6).

As the knee joint's capacity to flex during landings is a primary factor in the attenuation of vGRF (Yeow, Lee and Goh, 2009), reduced knee flexion results in stiffer landing strategies that increase peak vGRF (Zhang, Bates and Dufek, 2000). With restricted ankle DF ROM reducing the ability of the knee to flex during landings, ankle DF ROM has been negatively correlated with peak vGRF (r = -0.41) (Fong et al., 2011). However, the relationship between ankle DF ROM and peak vGRF during landings has not been consistently identified (Malloy et al., 2015; Whitting et al., 2011). A possible explanation for these inconsistent findings may be due to the compensations observed in the frontal plane. Fong et al. (2011) identified a negative association between ankle DF ROM and peak vGRF, but found no relationship between measures of ankle DF ROM and knee valgus displacement (r = -0.29). However, Malloy et al. (2015) identified a significant correlation between ankle DF ROM and peak knee abduction angle (r = 0.36), but no association with peak vGRF. It has been suggested that during landing activities, frontal plane compensations in the lower extremity may be employed to enable individuals with restricted ankle DF ROM to access a movement strategy that allows for the continued lowering of the centre of mass to attenuate peak vGRF (Mason-Mackay, Whatman and Reid, 2017). Restricted ankle DF ROM may increase knee valgus as a consequence of subtalar joint motion reported by Whitting et al. (2011). Subtalar joint eversion results in tibial rotation, which displaces the knee joint medially causing knee valgus (Rodrigues et al., 2015). Although this offers an individual with restricted ankle DF ROM an opportunity to manage peak vGRF during landings, a disadvantage to this strategy would be the potential for excessive loading on the passive structures supporting the knee joint (Salsich and Perman, 2007; Stärke et al., 2013; Yu and Garrett, 2007). As a result, prospective studies have identified individuals demonstrating high amounts of knee valgus during landing tasks are at greater risk for incurring anterior cruciate ligament injury (Hewett et al., 2005; Padua et al., 2009) and patellofemoral pain syndrome (Holden et al., 2017). If restrictions in ankle DF

ROM do impact frontal plane knee alignment, addressing the hypomobility may be a key consideration for reducing knee injury risk in populations regularly performing landing tasks.

In conclusion, restriction in ankle DF ROM is positively associated with peak knee flexion angle during landings (Fong et al., 2011; Malloy et al., 2015). This occurs as ankle DF ROM restriction limits the forward linear displacement of the knee joint, preventing the available knee flexion range of motion from being accessed. To compensate for the lack of peak knee flexion angle, individuals may land with the knee in an extended position at initial ground contact in an attempt to maintain sagittal plane knee joint displacement values (Dowling, McPherson and Paci, 2018). Additionally, as ankle DF ROM is significantly associated with knee abduction angle during landings (Malloy et al., 2015), greater dynamic knee valgus may be employed to allow for the continued lowering of the centre of mass. Although these compensatory strategies have the potential to be beneficial for preventing peak vGRF from becoming excessive, the elevated injury risk for passive structures surrounding the knee joint likely offsets any advantages this strategy offers.

2.8 Ankle dorsiflexion range of motion and hip kinematics during landings

Greater peak hip flexion angles during landings have been proposed to reduce the risk of anterior cruciate ligament injury (Griffin et al., 2000) by supporting the attenuation of peak vGRF and optimising muscle activation strategies at the knee (Blackburn and Padua, 2008; Blackburn and Padua, 2009). Blackburn and Padua (2009) investigated the impact of increasing peak hip flexion angles during bilateral drop-landing performance from a drop height of 0.60 m in which participants performed landings using their preferred strategy *and* following instructions to *"actively flex the trunk during landing"*. There was an increase in

peak hip flexion during the cued condition alongside a reduction in peak vGRF and electromyographic amplitude in the quadriceps. Furthermore, when individuals are cued to perform 'soft' landings resulting in reduced peak vGRF, sagittal plane hip joint displacement and work performed by the hip extensor musculature significantly increases (Zhang, Bates and Dufek, 2000). As such, when individuals incorporate large amounts of hip flexion into their landing strategy, peak vGRF significantly diminish. Therefore, the hip joint's ability to flex supports the successful management of landing forces.

Restrictions in ankle DF ROM may also impact hip joint kinematics during landing tasks. Ankle DF ROM has been found to positively correlate ($r = 0.36$) with sagittal plane hip joint displacement during bilateral jump-landings (Fong et al., 2011). Similarly, Dowling, McPherson and Paci (2018) found a small yet significant positive relationship between WBLT performance and hip angle at the moment of peak knee flexion ($r = 0.25$), as well as sagittal plane hip joint displacement ($r = 0.30$) during single-leg drop jumps. A potential mechanism for limitations in ankle DF ROM influencing hip kinematics is the sagittal plane coupling that occurs in the lower extremity during landing tasks. Yeow, Lee and Goh (2011) examined the sagittal plane coordination pattern between the ankle, knee and hip joints during bilateral drop-landings from a drop height of 0.60 m. An exponential relationship was identified between the knee and hip joints, indicating that the hip flexes at a greater rate relative to the knee. It was suggested that this strategy might have been selected by participants to allow for the transfer of power from distal to proximal joints, as to include the large hip joint musculature to assist with the dissipation of shock. As limitations in ankle DF ROM may inhibit knee flexion from occurring during landings, sagittal plane hip flexion may also be prevented due to the coordinative relationship between the knee and hip joints.

Leporace et al. (2018) investigated the relationship between ankle DF ROM using the WBLT and the knee-to-hip flexion ratio during a single-leg vertical hopping test. No correlation was found between WBLT performance and knee-to-hip ratio, indicating restrictions in ankle DF ROM do not influence the sagittal plane coupling of the knee and hip joints. As ankle hypomobility limits sagittal plane knee joint displacement during landings, reduced hip flexion angle at the moment of peak flexion also occurs as a consequence (Dowling, McPherson and Paci, 2018). Therefore, in order to preserve the sagittal plane coupling of the lower extremity joints during landings (Leporace et al., 2018), restricted ankle DF ROM likely diminishes sagittal plane knee and hip joint contribution.

In summary, ankle DF ROM restriction alters hip joint involvement during landings. This likely occurs as the sagittal plane coupling of knee and hip flexion is a stable feature of landing technique. As restricted ankle DF ROM is associated with decreased peak knee flexion angle and knee joint displacement during landings, hip flexion capacity concurrently diminishes (Dowling, McPherson and Paci, 2018; Fong et al., 2011).

2.9 Ankle dorsiflexion range of motion and inter-limb asymmetries during bilateral landings

Inter-limb asymmetries in landing mechanics have been previously found during bilateral landing tasks in healthy (Edwards et al., 2012; Harry et al., 2017; Niu et al., 2011; Pappas and Carpes, 2012) and injured populations (Paterno et al., 2007). Differences in between-limb performance during bilateral landings have been speculated to contribute to the development of overuse injury (Schot, Bates and Dufek, 1994), with the limb subjected to larger peak

vGRF at greater risk due to the higher loading (Bressel and Cronin, 2005; Harry et al., 2017). Schot, Bates and Dufek (1994) identified mean asymmetries in peak vGRF during bilateral drop-landings from 0.60 m of 14.8% in ten healthy adults. Although differences were not statistically different, Harry et al. (2018) found the average bilateral differences in peak vGRF to be 2.81 $N \cdot kg^{-1}$ during bilateral drop-landings from drop heights equivalent to participants' maximum CMJ height. Inter-limb asymmetries have also been found for joint displacement during bilateral drop-landing tasks. Pappas and Carpes (2012) observed between-limb differences for sagittal plane ankle, knee and hip joint displacement during bilateral drop-landings from 0.40 m for male recreational athletes of 3.4˚, 3.6˚ and 2.1˚, respectively. Although the prevalence of asymmetries during bilateral landings in healthy populations is apparent, it is currently unclear what factors cause inter-limb asymmetries in landing mechanics.

Inter-limb asymmetries in ankle DF ROM have been reported in healthy populations (Rabin et al., 2015), with evidence showing asymmetries in ankle DF ROM negatively affecting performance in change of direction tasks (Gonzalo-Skok et al., 2015). While ankle DF ROM has been shown to negatively influence landing mechanics (Fong et al., 2011; Malloy et al., 2015; Dowling, McPherson and Paci, 2018), evidence for between-limb differences in ankle DF ROM causing asymmetries in landing mechanics is limited. It may be that compensatory strategies for unilateral restrictions in ankle DF ROM cause inter-limb asymmetries in peak vGRF and kinematic measures of landing performance. Consistent with this hypothesis, Crowe et al. (2019) investigated the effect of inter-limb asymmetries in ankle DF ROM on vertical force symmetry during bilateral squatting. Participants with < 5° inter-limb asymmetry on the WBLT performed three bodyweight squats to below parallel in barefoot, with and without a 10° forefoot wedge under the right foot aimed to acutely limit ankle DF

ROM for the right side only. Asymmetries in vertical force were calculated for the upper half and lower half of the descent and ascent phases of the squat for each participant using the impulse-momentum relationship. The wedge condition increased inter-limb vertical force asymmetry for each phase of the movement (ES = 0.7–1.1). As the lower extremity joint angles during the upper phases of the squat are similar to those observed during landings, it was suggested that inter-limb asymmetries in ankle DF ROM could also affect symmetry during landing tasks (Crowe et al., 2019). However, whether asymmetry in ankle DF ROM has the potential to cause asymmetry in landing mechanics is currently unknown.

2.10 Ankle dorsiflexion range of motion and drop height during landings

Investigations establishing the relationship between ankle DF ROM and landing mechanics during bilateral landings have used arbitrary drop heights ranging from 0.30 m (Fong et al., 2011; Dill et al., 2014) to 0.46 m (Sigward et al., 2008). As greater knee extensor strength has been associated with reduced knee joint displacement during landings (Fisher et al., 2016; Howard et al., 2011; Nagai et al., 2013), the use of an arbitrary drop height fails to consider other variables that may determine the landing strategy adopted, such as jump performance and lower extremity strength. To overcome this concern, Malloy et al. (2015) individualised drop height relative to each participant's maximal CMJ height, finding a significant correlation between ankle DF ROM and a number of variables associated with landing mechanics. However, in many sporting activities, jump height will vary as a function of the task, with some sport skills requiring landing from heights that far exceed an individual's maximal CMJ height. For example, performing a CMJ with an arm swing (Slinde et al., 2008) or with a run-up that precedes the jump (Young, Wilson and Byrne, 1999) results in jump heights of 114% and 122% of CMJ height, respectively. Therefore, establishing the

influence that drop height may have on the relationship between ankle DF ROM and landing mechanics will help practitioners assess injury risk when exposing an individual with restricted ankle DF ROM to landing activities that exceed their jump height.

Elevation in drop height has shown to increase the sagittal plane displacement for the ankle, knee and hip joints (Nordin and Dufek, 2017). This occurs as a coping mechanism for the rise in peak vGRF that is present when landing from greater drop heights (McNitt-Gray, 1991; Yeow, Lee and Goh, 2009). During bilateral drop-landings, Zhang, Bates and Dufek (2000) found that between drop heights of 0.32 m and 1.03 m, ankle, knee and hip peak flexion angles increased by 4˚, 12˚ and 25˚, respectively. Increases in peak flexion angles contribute to greater sagittal plane joint displacement, supporting the increased energy absorption demands associated with landings from greater drop heights (Yeow, Lee and Goh, 2010). As restricted ankle DF ROM causes a decrease in peak flexion angles for the ankle, knee and hip joints during landing tasks (Dowling, McPherson and Paci, 2018; Fong et al., 2011; Malloy et al., 2015), individuals with ankle hypomobility may lack the capacity to adjust their coordination strategy to allow for a lower descent of the centre of mass when landing from elevated drop heights. As a result, injury risk may increase when landing from greater drop heights, due to a diminished capacity to access increased flexion at the knee and hip joints required for the management of landing forces.

Ankle plantar flexion angle at initial contact has also shown to increase when landing from higher drop heights (Whitting et al., 2007). Altering initial contact angle is, therefore, also an important strategy for increasing joint displacement to cope with the increased peak vGRF associated with elevated drop heights (Begalle et al., 2015; Rowley and Richards, 2015).

Individuals with restricted ankle DF ROM may increase ankle plantar flexion angle (see section 2.6) while decreasing knee flexion angle (Dowling, McPherson and Paci, 2018) at initial contact as a strategy to maintain ankle and knee joint displacement during landings. As such, it may be that these individuals exhaust this compensation strategy at lower drop heights, removing their capacity to adjust initial contact angles to cope when landing from greater drop heights. With restrictions in ankle DF ROM already limiting ankle, knee and hip peak flexion angles during landings (Dowling, McPherson and Paci, 2018; Fong et al., 2011; Malloy et al., 2015), a diminished ability to alter initial contact angles could potentially result in excessive forces, beyond what is expected as drop height rises. The influence of drop height on the relationship between ankle DF ROM and sagittal plane landing mechanics is yet to be investigated.

In the frontal plane, peak knee valgus angles may also increase with greater drop height. Yeow, Lee and Goh (2009) showed a rise in drop height from 0.30 m to 0.60 m for bilateral drop-landings significantly increased knee abduction angles at the moment of peak knee flexion in 18 recreational male athletes. As restricted ankle DF ROM is associated with greater knee valgus during bilateral landing (Malloy et al., 2015; Sigward, Ota and Powers, 2008), elevations in drop height could exaggerate this compensation. This may result in greater injury risk for individuals with restricted ankle DF ROM when landing from higher drop heights. Whether drop height increases the relationship between ankle DF ROM and measures of peak knee valgus is, however, yet to be established.

2.11 Ankle dorsiflexion range of motion and fatigue during landings

Neuromuscular fatigue, which is defined as the inability for the neuromuscular system to maintain mechanical work for a given task (Fousekis, Tsepis and Vagenas, 2012), has been demonstrated as a risk factor for lower extremity injury (Ekstrand, Hägglund and Waldén, 2011). Neuromuscular fatigue caused by exercise results in an acute accumulation of metabolites, impaired excitation-contraction coupling and a decrease in stretch-reflex sensitivity that reduces muscle stiffness (Gathercole et al., 2015). In response to these changes in physiological status, fatigue decreases muscle strength (Maffiuletti et al., 2007), movement control (Chappell et al., 2005), proprioception (Skinner et al., 1986), muscle reaction times (Hakkinen and Komi, 1986) and balance – both static (Howard, Cawley and Losse, 1998) and dynamic (Johnston et al, 2018). As these qualities underpin optimal landing performance, fatigue has the potential to negatively affect landing mechanics and, subsequently, increase injury risk (Coventry et al., 2006; Ortiz et al., 2010; Thomas, McLean and Palmieri-Smith, 2010).

When performing movement tasks in a fatigued state, the neuromuscular system must reorganise the coordination pattern to offset functional deficits in performance caused by fatigue, while successfully accomplishing the outcome goal (James, Scheuermann and Smith, 2010). Increased injury risk during landing tasks, as a result of prior fatiguing exercise, is likely due to the inability to adopt alternate coordination strategies that effectively compensate for diminished performance of physical qualities. During landings, investigations have shown that fatigue acutely increases peak ankle dorsiflexion (Madigan and Pidcoe, 2003), knee flexion (McNeal, Sands and Stone, 2010) and sagittal plane knee joint displacement (James, Scheuermann and Smith, 2010) in healthy populations. These short-term adaptations in coordination strategy reduce peak vGRF during landings, compensating for the acute decline in force production caused by diminished muscle stiffness (Smith, Sizer

and James, 2009). This has been proposed to occur as a protective mechanism to reduce injury risk by decreasing musculotendinous loading during landings when exhaustion is present (Zadpoor and Nikooyan, 2012). Theoretically, limitations in ankle DF ROM may impact the landing strategies adopted by an individual during states of fatigue, due to the positive correlation of ankle DF ROM with peak ankle dorsiflexion and knee flexion angles during landings (Dowling, McPherson and Paci, 2018; Fong et al., 2011; Hoch et al., 2015; Malloy et al., 2015). Individuals with restrictions in ankle DF ROM may be unable to access greater sagittal plane ankle and knee joint displacement when fatigued, resulting in a failure to increase pliability during landings as a strategy to decrease peak vGRF. As a result, peak vGRF may remain unchanged or even increase at a time when the neuromuscular system's ability to produce high forces required to decelerate the centre of mass is compromised. Therefore, restricted ankle DF ROM has the potential to elevate injury risk when performing landings in a fatigued state, by impairing muscle function and, thus, preventing coordination strategies that are necessary to attenuate vGRF during demanding dynamic tasks.

Altering joint angles at initial ground contact presents another adaptation in coordination strategy that may support the attenuation of peak vGRF when performing landings while acutely fatigued. At initial ground contact, increases in ankle plantar flexion (mean difference = 10.6°, ES = 1.27) and reduced knee flexion (mean difference = 7.0°, ES = 1.06) have been shown to occur when performing bilateral landings in a fatigued state (Weinhandl, Smith and Dugan, 2011). These changes allow for increased sagittal plane joint displacement at the ankle and knee (Begalle et al., 2015) that facilitate lower peak vGRF (Rowley and Richards, 2015). As individuals with restrictions in ankle DF ROM have been shown to compensate by increasing ankle plantar flexion and reducing knee flexion angles at initial ground contact (Dowling, McPherson and Paci, 2018; Fong et al., 2011), it is feasible further exploitation of

this strategy will be inaccessible for these individuals when fatigue is induced. As a consequence, failure to adapt the pre-landing strategy when fatigued may result in elevated injury risk, secondary to suboptimal management of landing forces (Zadpoor and Nikooyan, 2012). Alternatively, the combination of restricted ankle DF ROM and the presence of neuromuscular fatigue may cause an individual to contact the ground during a landing with additional ankle plantar flexion and knee extension, a position associated with the mechanism of ankle (Wright et al., 2000) and knee ligament injury (Boden et al., 2010) respectively. At present, the effects of fatigue on lower extremity joint angles at initial ground contact in populations presenting with restricted ankle DF ROM requires investigation.

In the frontal plane, fatigue may elevate the injury risk at the knee joint during landings by increasing knee valgus (Hewett et al., 2005; Renstrom et al., 2008). McLean et al. (2007) found that male and female athletes demonstrated greater peak knee abduction angle during bilateral drop jumps performed from a 0.50 m drop height following a fatigue protocol consisting of step-ups. This is consistent with Pappas et al. (2007), who found greater peak knee valgus angle during bilateral drop-landings from a 0.40 m drop height after the performance of a high volume of jump exercises. Similarly, Dickin et al. (2015) found peak knee valgus angle increased during bilateral drop jumps following the performance of a fatigue protocol used to induce a 20% deficit in CMJ height. Theoretically, increases in peak knee valgus caused by fatigue may be exaggerated when ankle DF ROM is present. As both restrictions in ankle DF ROM (Malloy et al., 2015) and the presence of fatigue (Dickin et al., 2015; McLean et al., 2007) have been shown to increase peak knee valgus angle, individuals with restricted ankle DF ROM may further compound knee valgus when performing landings in a fatigued state. At present, no evidence exists regarding the relationship between ankle DF ROM and landing mechanics under acute fatigue.

The effects of fatigue have been shown to be task dependant and that should be considered during the conception of study design (James, Scheuermann and Smith, 2010). Potentially due to a lack of representative task design, studies that have induced fatigue in a single muscle group (Hollman et al., 2012) or with unrelated movement patterns (James, Scheuermann and Smith, 2010), demonstrate a diminished effect on kinematic variables associated with landing performance. As such, fatigue protocols that incorporate the stretch-shortening cycle through repetitive jumping actions induce deficits in muscle stiffness that offer the optimal conditions to investigate the effects of lower extremity fatigue on landing performance (Edwards et al., 2014). Furthermore, as movements relying on the stretch-shortening cycle commonly occur during sport (Bloomfield, Polman and O'Donoghue, 2007; Lian et al., 1996; McClay et al., 1994), activities of leisure (Maté-Muñoz et al., 2017) and occupational tasks (Knapik et al., 2003), it would be more ecologically valid to induce fatigue using tasks that incorporate the stretch-shortening cycle (Edwards et al., 2014).

One factor often unrecognised in laboratory-based studies of fatigue is the individuality of the onset and time-course of performance decrements during physical tasks. For example, several investigations have used an arbitrary (fixed across participants) prescription of exercise to induce fatigue (e.g. Brazen et al., 2010; James, Scheuremann and Smith, 2010). This approach can be problematic, as it does not account for the inter-individual variance in performance changes as a function of time during a fatiguing task. As a result, this approach fails to control for the level of fatigue that has been induced, which should be regarded as a necessity when establishing the effects of fatigue on landing mechanics. In turn, the lack of control for the magnitude of fatigue induced may act as a confounding variable when

analysing the effects of fatigue on landing mechanics in different populations. An alternative approach has been to control for the level of induced fatigue during performance by monitoring each individual until a ~ 20–30% reduction of maximum performance (such as CMJ height) has occurred (Edwards, Steele and McGhee, 2010; Edwards et al., 2017; Weeks, Carty and Horan, 2015; Weinhandl, Smith and Dugan, 2011). Therefore, research that is designed to produce fatigue through the implementation of an exercise protocol, should attempt to standardise the level of fatigue induced for each participant prior to retesting landing mechanics.

2.12 Reliability for kinetic measurements associated with bilateral landing performance

Reliability in testing procedures is defined as the consistency of a measurement (Riemann and Lininger, 2018) and is important to establish for all outcome variables to facilitate interpretation of data. In the sports and exercise sciences, this is important, as statistically significant changes in test results may, in fact, be the result of misinterpreted measurement error (i.e. variability) (Atkinson and Nevill, 1998). To account for the associated error, an 'analytical goal' can be pre-defined, which is typically chosen based on the minimally important change in the testing variable. As such, when deciding if a variable is reliable for testing purposes, practitioners can look to determine if the error is less than typical changes observed following commonly applied interventions. For example, studies identifying intra-rater reliability for measuring tibia angle during the WBLT typically produce MDC values of 4.7° (Powden, Hoch and Hoch, 2015). This value represents the boundaries for error, providing 90–95% confidence (depending on the calculation used) that changes outside of this value should be regarded as 'real' (Riemann and Lininger, 2018). Using the WBLT example, changes in performance following stretching and mobilisation programmes

generally exceed 4.7° (Aune et al., 2019; Jeon et al., 2015), indicating that the WBLT is a reliable tool for detecting changes in ankle DF ROM following chronic interventions. As this method for analysing data can support clarification of findings (Atkinson and Nevill, 1998), this method of interpretation will be used throughout this thesis to enhance the depth of the analysis.

Given the practical and clinical importance of kinetic variables, such as peak vGRF, time to peak vGRF and loading rate, including their reported association with injury risk factors during bilateral landing (Bisseling et al., 2008; Hewett et al., 2005; Zadpoor and Nikooyan, 2012), it is important to understand the inherent error (or lack thereof) associated with such testing procedures. To date, reliability of kinetic measures of jump-landing tasks have focused on the propulsive phase of bilateral jumping (Hori et al., 2009; Markovic et al., 2004; Slinde et al., 2008), with limited reliability data for the kinetic factors associated with bilateral drop-landings in healthy populations.

The reliability of force-time data has been investigated in bilateral landing tasks where the landing is immediately followed by a jump (i.e. drop jumps). Ortiz et al. (2007) found a mean of four trials were required to maximise ICC values for peak vGRF during a 0.40 m bilateral drop jump (ICC = 0.93). The SEM for normalised peak vGRF when using the mean value of four trials was reported as 0.28 N·kg^{-1}. This value was lower than when calculating normalised peak vGRF using a single trial (0.57 N·kg^{-1}) and three trials (0.40 N·kg^{-1}). During bilateral landings from a stop-jump task, Milner, Westlake and Tate (2011) reported the relative reliability for within-session (ICC = 0.63) and between-session (ICC = 0.96) reliability as *large* and *near perfect*, respectively. Schwartz et al. (2017) investigated the

reliability of force-time measures associated with single-limb tests during the landing phase of a bilateral CMJ. Relative reliability for the dominant and non-dominant limb were reported as *very large*, with ICC ranging from 0.76–0.78 and 0.85–0.86 for peak vGRF and time to peak vGRF, respectively (Schwartz et al., 2017). Values of absolute reliability were also reported, with SEM for peak vGRF ranging from 355–394 N (not normalised to body mass), while SEM for time to peak vGRF was 0.006 s (Schwartz et al., 2017). These findings support the reliability of force-time measures associated with landings; however, they lack direct application to bilateral drop-landings due to differences in the performance of each task (Collings et al., 2019).

Specific to the performance of bilateral drop-landings in healthy populations, James et al. (2007) reported relative reliability as *very large* for bilateral measures of peak vGRF (ICC = 0.77) and loading rate (ICC = 0.87) from a drop height of 0.61 m. Similar to Ortiz et al. (2007), James et al. (2007) found four trials were required to achieve performance stability for these measures when seeking a maximum ICC value. Using a within-session design, Walsh et al. (2006) reported *near perfect* reliability for peak vGRF (ICC = 0.98) and time to peak vGRF (ICC = 0.92) following a bilateral drop-landing from a 0.31 m box. However, both James et al. (2007) and Walsh et al. (2006) measured peak vGRF bilaterally without considering the force-time variables associated with single-limb performance outcome measures during bilateral drop-landings. Although researchers have attempted to identify changes in a single-limb measure of normalised peak vGRF during bilateral drop-landings following an intervention (Czasche et al., 2017), there is limited evidence for the reliability of this variable. Furthermore, reliability of measures of inter-limb asymmetries associated with bilateral drop-landing performance has not been established in healthy populations. Research

is required to quantify the error associated with kinetic measures of bilateral drop-landing performance from varying drop heights.

2.13 Validity and reliability for kinematic measurements associated with bilateral landing performance

Although 3D motion capture is regarded as the gold standard, in practice 2D video analysis is much more widely accessible to practitioners. Evidence relating to the validity and reliability of kinematic variables used to quantify bilateral drop-landing performance is scarce. This section will provide a summary of findings using a variety of landing tasks.

Frontal plane projection angle (FPPA), a frontal plane representation of knee valgus/varus angle at the knee in the deepest landing position, is commonly employed to assess frontal plane landing mechanics using 2D video analysis. During bilateral landings, McLean et al. (2005) reported a *very large* relationship between 2D video analysis and a 3D motion capture system ($r = 0.80$). This value is similar to the association found by Myer et al. (2011) ($r = 0.87$) and Belyea et al. (2015) ($r = 0.69$) during drop jumps performed from a 0.31 m box. Additionally, FPPA measured using 2D video analysis during drop jumps is negatively correlated with knee abduction moments ($r = -0.59$), indicating lower FPPA values (representing greater knee valgus) increase external knee abduction moments during landings (Mizner et al., 2012). It would appear, therefore that despite the 2D FPPA measurement being unable to fully capture the tri-planar nature of knee valgus, 2D video analysis can be a valid tool to identify individuals at greater injury risk when assessed during landings (Lopes et al., 2012).

For various landing tasks, 2D video analysis has been shown to be a reliable tool for measuring FPPA (Dingenen et al., 2005; Miller and Callister, 2009; Mizner et al., 2012; Munro, Herrington and Carolan, 2012). At present, however, reliability for FPPA has not been evidenced for bilateral drop-landing performance. To support the use of FPPA when quantifying landing mechanics, FPPA presents as a reliable measure during many landing tasks. Herrington (2014) reported FPPA during single-leg drop-landings from a 0.30 m drop height to possess MDC of 3.1˚. However, these values are considerably lower than those reported by Munro, Herrington and Carolan (2012) for bilateral drop jumps from a 0.28 m drop height. SEM for FPPA was 3.0˚, while MDC value was reported as 8.3˚ for between-session reliability. As these values are less than the changes reported during bilateral landings following adherence to a well-designed exercise programme (> 10˚) (Herrington, Munro and Comfort, 2015), FPPA appears to possess sufficient sensitivity to detect changes in frontal plane knee kinematics following a chronic intervention. For relative reliability, ICCs ranged from *very large* (ICC = 0.89) to *near perfect* (ICC = 0.91) for male and female participants respectively. These values are similar to what has been presented elsewhere in the literature for FPPA during unilateral and bilateral landing tasks (for example, see Strensrud et al., 2010). Although FPPA presents as a reliable tool for measuring frontal plane knee mechanics during a variety of landing tasks, this has yet to be shown during bilateral drop-landings.

Researchers investigating the validity and reliability of 2D analysis have typically focussed less on sagittal plane measurement variables that impact load dissipation during landings, such as initial contact angles, joint displacement for the ankle, knee and hip joints or inter-limb asymmetries in lower extremity coordination patterns (e.g., see Begalle et al., 2015; Chappell et al., 2005; Pappas and Carpes, 2012; Rowley and Richards, 2015). For measures of ankle joint kinematics during bilateral drop-landings, 2D video analysis is yet to be

validated against 3D motion capture. For functional tasks, Schurr et al. (2017) found a *large* relationship ($r = 0.51$) for ankle joint displacement during single-leg squatting. Similarly, ankle joint angle measured using 2D video analysis at the bottom of a bilateral squat has shown to average a 3.1° difference when compared to 3D motion capture (Krause et al., 2015). During running, ankle joint angle at toe-off during running measured using 2D video analysis has been shown to have a 1° difference with 3D motion capture, while at touchdown this difference was 4° (Mousavi et al., 2020). These findings have led to 2D video analysis being seen as a valid tool for measuring ankle joint mechanics during a variety of fundamental movement patterns (Krause et al., 2015; Mousavi et al., 2020; Schurr et al., 2017). Although this evidence is limited due to the differing demands of each task, 2D video analysis does appear to provide a valid measure of ankle joint angle during functional activities.

The validity of 2D video analysis has been established for knee and hip joint kinematics during landing tasks. Myer et al. (2011) compared 2D video analysis against 3D motion capture for measures of sagittal plane knee joint displacement during bilateral landings, reporting a positive relationship ($r = 0.95$) with a mean difference of 3.0°. Belyea et al. (2015) also found a positive relationship between measurement tools for hip flexion angle at initial ground contact ($r = 0.48$) and peak flexion ($r = 0.51$), while 2D measures of sagittal plane hip joint displacement were also positively correlated with 3D motion capture ($r = 0.73$). Additionally, Dingenen et al. (2015) reported a negative relationship between peak hip flexion angle measured using 2D video analysis with knee joint moment during bilateral drop jumps, indicating greater peak hip flexion angles result in reduced knee flexion joint moments. The same investigation found peak hip flexion angle was also negatively correlated with knee abduction moments, illustrating the hip joints contribution to facilitating the safe

attenuation of vGRF during landings. In summary, knee and hip joint kinematics observed using 2D video analysis have been shown to provide a valid measure of landing performance.

Investigations to determine the reliability of 2D video analysis for measuring sagittal plane joint angles have used bilateral drop jumps from a drop height of 0.30 m (Beardt et al., 2018; Dingenen et al., 2015; King and Belyea, 2015). Amongst these investigations, SEM values for measures of joint angle at initial ground contact and peak flexion are reported to be < 2.5˚, while relative reliability ranges between 0.71–1.00. As interventions to alter landing mechanics result in modifications to joint angles at specific time points that far exceed the values of measurement error presented in the literature (e.g. Zhang, Bates and Dufek, 2000), 2D video analysis likely possesses the sensitivity to detect clinically meaningful changes in landing mechanics following acute (i.e. fatiguing protocol) and chronic (i.e. training programme) interventions. Although evidence suggests 2D analysis to be suitable for reliably measuring lower extremity joint alignment at various time points during landings, investigations are required to quantify the error associated with measuring sagittal plane joint kinematics during bilateral drop-landings.

To date, reliability of 2D kinematic measures has been established for bilateral vertical drop jumps with drop heights of approximately 0.30 m (Dingenen et al., 2015; King and Belyea, 2015; Miller and Callister, 2009; Mizner et al., 2012; Munro, Herrington and Carolan, 2012). However, normative data for CMJ height suggests that many athletes have the potential to regularly perform bilateral landings that will exceed a 0.30 m drop height (Wisløff et al., 2003). Given that greater drop heights result in greater joint excursions for the ankle, knee and hip joints in the sagittal plane (Zhang, Bates and Dufek, 2000), the reliability of

kinematic measurements should be established for bilateral landings across various heights. Furthermore, when performing landings from varying drop heights relative to CMJ performance, the magnitude of variability in kinematic variables associated with bilateral drop-landing performance has been shown to differ. Nordin and Dufek (2017) investigated the variability in the performance of bilateral drop-landings from 20%, 60%, 100%, 140% and 180% of maximum CMJ height using 3D motion capture analysis. Reduced variability, expressed as standard deviations and coefficient of variation (CV%), was observed for peak joint angles when landing from greater drop heights for the ankle, knee and hip joints in the sagittal plane, indicating variability decreased as drop height increased relative to CMJ performance. As such, it is likely that the error related to measurements of bilateral drop-landing performance decrease as drop height increases. However, researchers have yet to investigate the effect of drop height on force-time and 2D video analysis variables associated with bilateral drop-landing performance with reference to clinically relevant changes.

2.14 Strategies to increase ankle dorsiflexion range of motion

A version of this section has been published in The Sport and Exercise Scientist and the printed version can be found in Appendix 6.

Limitations in ankle DF ROM can be caused by a lack of extensibility in myofascial tissues surrounding the ankle joint (Medeiros and Martinin, 2018) or disruption in ankle joint arthrokinematics (Delahunt et al., 2013; Loudon and Bell, 1996). A variety of exercise-based strategies have been reported to restore ankle DF ROM in individuals with restricted motion (Aune et al., 2019; Jeon et al., 2015; Medeiros and Martinin, 2018). Interventions that are aimed at altering the flexibility of the plantar flexor muscles and related connective tissue

include different forms of stretching (Medeiros and Martinin, 2018), strengthening (Aune et al., 2019; Mahieu et al., 2008) and self-massage (Aune et al., 2019) exercises or some combination of each modality.

Static stretching of the ankle plantar flexors involves lengthening the muscles until a sensation of stretching is reached, and then holding this position for sustained periods (Behm, 2018). Although a variety of forms of stretching can be prescribed, the effect of static stretching on the ankle plantar flexors has been most commonly investigated (Medeiros and Martinin, 2018). The advantage of this technique is that minimal equipment and expertise is required and stretches can be employed without assistance. Additionally, evidence for chronic effects of static stretching on flexibility is greater when compared to other techniques such as proprioceptive neuromuscular facilitation or ballistic stretching (Thomas et al., 2018). As such, static stretching appears to be a suitable modality to increase flexibility.

Nakamura et al. (2012) found static stretching performed by healthy males for a total duration of 2 minutes daily over 4-weeks significantly increased ankle DF ROM. In a follow-up investigation, Nakamura et al. (2017) showed that static stretching of the ankle plantar flexors for a total duration of 2 minutes, three times a week for 4-weeks significantly increased ankle DF ROM whereas a control group showed no change. These findings have been consistently reported across similar studies irrespective of measurement technique (see Blazevich et al., 2014; Johanson et al., 2006; Johanson et al., 2008; Peres et al., 2002). Together these findings suggest that small doses of static stretching for the ankle plantar flexors results in significant improvements in ankle DF ROM that may lead to improved landing technique.

The frequency of stretching between investigations varies, with the performance of static stretching being prescribed ranging between three times per week (Nakamura et al., 2017), and seven times per week (Blazevich et al., 2014; Johanson et al., 2006; Johanson et al., 2008; Nakamura et al., 2012). Collectively, research suggests that the frequency of stretching does not influence gains in flexibility, assuming that the minimum volume threshold is surpassed (Thomas et al., 2018). For example, Marques et al. (2009) reported that static stretching of the hamstring muscles, performed for a total duration of 60 s, three times per week, resulted in significant increases in active knee extension test performance. However, no differences were found between groups performing static stretching three times per week, compared to five times per week (Marques et al., 2009). In a review on programming considerations associated with stretching to increase flexibility, Thomas et al. (2018) identified 5 minutes in total duration over the course of a week was required to increase flexibility, with no greater gains found when time exceeded this duration. Furthermore, no differences were found in flexibility when the total time spent stretching was divided into shorter (< 60 s), moderate (60–120 s) or longer durations (> 120 s). These findings appear to apply to the ankle plantar flexors, with a meta-analysis establishing no difference between protocols using higher volumes of stretching within a study duration (> 5000 s) compared to shorter volumes (< 3000 s) for increasing ankle DF ROM (Medeiros and Martini, 2018). As such, it appears that static stretching performed three times per week for a total weekly duration 5 minutes is sufficient to increase ankle DF ROM.

Eccentric strength-training may also be employed to induce chronic gains in flexibility (Nelson and Brandy, 2004; Potier, Alexander and Seynnes, 2009). Eccentric strength-training involves the active lengthening of the musculotendinous unit under loaded conditions and can facilitate increases in flexibility through the addition of sacromeres in series (Mahieu et al.,

2008; Potier, Alexander and Seynnes, 2009). For the ankle plantar flexors, Mahieu et al. (2008) found an eccentric strengthening exercise performed daily for 6-weeks resulted in significant increases in ankle DF ROM during the WBLT. Of note, ankle DF ROM when tested with both an extended (mean difference from baseline = 6.0°) and flexed knee (mean difference from baseline = 5.6°) alignment increased significantly, with no difference between testing positions. These findings suggest that performance of eccentric strength exercises increases flexibility of both the biarticular and uniarticular plantar flexors equally, rendering a specific protocol to target each muscle group unnecessary. Additionally, the same investigation found passive resistive torque of the plantar flexors significantly decreased from 16.4 ± 0.8 N·m to 12.7 ± 0.6 N·m, suggesting structural adaptations in muscle occurred as a result of the intervention. Using the same exercise protocol over a 4-week duration, Aune et al. (2019) found academy football players' significantly improved WBLT performance by 5.1°. This value exceeded the 2.0° MDC value established for the WBLT measurement technique (Aune et al., 2019). Therefore, performing eccentric strength-training for the ankle plantar flexors provides a suitable stimulus for improving ankle DF ROM and should be included in a well-rounded mobility intervention.

Self-massage, also known as self-myofascial release, can be prescribed to increase flexibility (Behm and Wilke, 2019). Devices such as foam rollers, custom-made balls and roller sticks are specialised equipment that are used to self-induce acute and chronic changes in range of motion (Beardsley and Škarabot, 2015). Halperin et al. (2014) found a significant improvement in WBLT performance following self-massage with a roller stick for three sets of 30 s at a perceived pain level of 7 out of 10 to the calf muscles that was comparable to static stretching. Similarly, de Souza et al. (2019) found an increase in ankle DF ROM during the WBLT for both the non-dominant and dominant limb following 2 sets of 10 repetitions

for the calf muscles using a roller stick. When the dosage for self-massage was doubled, no additional acute gains in ankle DF ROM were found beyond the increases observed following the 2 sets of 10 repetition protocol. Collectively, this evidence demonstrates the application of self-massage can result in acute increases in ankle DF ROM.

An advantage of self-massage over other modalities to improve flexibility is the lack of influence on various performance measures. Although static stretching results in improved flexibility (Medeiros and Martinin, 2018), it also causes an acute negative impact on the performance of athletic activities, such as sprinting (Winchester et al., 2008), jumping (Bradley, Olsen and Portas, 2007) and tasks that require maximal strength (Bacurau et al., 2009). Likewise, eccentric strength exercises induce fatigue that may negatively affect performance acutely (Sakamoto et al., 2010). However, self-massage leads to acute increases in flexibility that is comparable to static stretching, without impairing performance (Halperin et al., 2014). As flexibility exercises are traditionally performed as part of a structured warm-up routine, employing self-massage exercises may be a logical approach to increasing mobility prior to training, without causing losses in performance.

There is a lack of research investigating the chronic effects of self-massage to increase ankle DF ROM. Aune et al. (2019) examined the effects of self-massaging the plantar flexor musculature on WBLT performance. Four weeks of daily self-massage of the calf muscles for 3 sets of 60 s using a foam roller resulted in increased performance in the WBLT (mean difference = 2.6°) that exceeded error associated with the testing procedures (MDC = 2.0°). However, the observed increase did not reach statistical significance ($P = 0.09$). The authors proposed that the lack of significance might have occurred due improper technique, as the

execution of foam rolling was not monitored during the study. To support this suggestion, Smith et al. (2019) investigated the effects of self-massage on ankle DF ROM following the performance of 12 supervised sessions consisting of foam rolling the ankle plantar flexors for 3 sets of 30 s over a 6-week period in healthy university-aged participants. Ankle DF ROM measured during the WBLT increased by 7.9° following the intervention (ES = 3.1), indicating self-massage with appropriate technique can be prescribed as a modality for healthy individuals to stimulate chronic improvements in ankle DF ROM.

Disrupted joint arthrokinematics may also cause restricted ankle DF ROM (Fujii et al., 2010). Ankle DF ROM increased after manual joint mobilisation in both previously injured (Collins, Teys and Vicenzino, 2004; Delahunt et al., 2013; Green et al., 2001) and healthy populations (Howe, 2017; Guo et al., 2006). Mulligan (1993) suggested that the limited ability of the talus to posteriorly glide relative to the tibia and fibula reduces ankle DF ROM, secondary to a disruption in joint arthrokinematics. This is supported by evidence that talar positional faults are common among people with chronic ankle instability (Wikstrom and Hubbard, 2010), with studies showing joint mobilisations increase posterior talar glide after treatment (Vicenzino et al., 2006). Furthermore, mobilisations aimed at improving ankle joint arthrokinematics have shown a significant positive relationship between improved WBLT performance and increases in posterior glide following treatment (Vicenzino et al., 2006). However, whether improved arthrokinematics following mobilisation explains the increased ankle DF ROM observed during post-intervention testing is unclear (Kosik and Gribble, 2016), with other mechanisms such as modulation of the central and peripheral nervous system also suggested as a mechanism for improved function following mobilisation (Bialosky, Bishop and George, 2008).

Although manual therapy is outside the remit of many professionals and logistically challenging to perform regularly, self-mobilisation is often recommended for individuals with limited ankle DF ROM (Cosby and Grindstaff, 2012). In support of this recommendation, Jeon et al. (2015) showed that a stretching technique, using a strap positioned to improve the posterior glide of the talus while concurrently stretching the plantar flexor musculature, increased ankle DF ROM after a 3-week intervention that was performed five times per week in healthy participants. Although arthrokinematic changes after the intervention were not measured, differences in ROM during the WBLT were greater in the group performing stretches with a strap (mean difference from baseline = 5.1°, ES = 0.9) compared with a static stretching only group (mean difference from baseline = 1.3°, ES = 0.3). These findings are comparable to other studies investigating the effects of mobilisations on ankle DF ROM in injured populations (e.g., see Vicenzino et al., 2006). This evidence indicates that mobilisation of the ankle joint can be achieved using a self-mobilisation technique.

There is a need to establish whether chronic improvements in ankle DF ROM lead to altered coordination strategies. Only scant evidence currently exists to indicate whether chronic alterations in range of motion for any joint segment also result in alterations to coordination strategies when performing functional movements. Moreside and McGill (2013) found increases in passive hip range of motion following a 6-week intervention did not result in a change in the performance of active standing hip extension, a standing twist movement or lunging. It was suggested that an additional focus on performing the functional tasks alongside the mobility programme might have led to integration of the improved flexibility into the movement patterns assessed. Such an interpretation is consistent with the findings

reported by Lephart et al. (2007), where flexibility exercises combined with general strength-training increased swing mechanics and measures of golf performance in recreational golfers. The programme was performed alongside golf practice, meaning it is likely that combining flexibility exercises with general strength-training, whilst practicing the sport, afforded participants the opportunity to integrate their improved range of motion into the specific sport skill.

In practice, practitioners rarely prescribe exercises to increase joint range of motion in isolation when a mobility restriction has been identified. More commonly, flexibility exercises to remove mobility restrictions are performed in combination with general strength exercises aimed at improving physical qualities that underpin the performance of daily activities and athletic skills. Therefore, for individuals with restricted ankle motion, researchers should seek to establish the effects of improving ankle DF ROM on landing mechanics following a combined mobility and general conditioning intervention.

2.15 Conclusion

Ankle DF ROM contributes to efficient energy dissipation during moderately demanding, everyday tasks, such as bilateral landings. However, restrictions in ankle DF ROM could impair the ability to effectively dissipate forces. Given the implied clinical relevance of the ankle DF ROM to movement efficiency, it is important to firstly identify the most suitable and reliable tests, which is currently uncertain. At present, reliability for kinematic and kinetic variables is not available within the literature to facilitate the interpretation of bilateral drop-landing performance. Whilst there is some empirical evidence to support the notion that restrictions in ankle DF ROM result in suboptimal landing mechanics, a more detailed

understanding of lower-limb kinetics and kinematics is necessary. Furthermore, it is important to determine whether inter-limb asymmetries in ankle DF ROM relate to landing mechanics, aspects of which are known to correlate with injury risk. Establishing the influence of task constraints, including drop height, on the relationship between ankle DF ROM and landing mechanics will also help to understand whether landing forces moderate this relationship. Indeed, based on the literature review of this thesis, it is feasible that the loading associated with the landing task influences the effect of the restriction on landing mechanics. This understanding would provide practitioners with more information when assessing risk for individuals performing landing tasks from various drop heights. Additionally, compensations in coordination strategies during landings caused by local muscle fatigue are similar to those observed in individuals with restricted ankle DF ROM. As such, it is possible that existing restrictions in ankle DF ROM may limit the landing strategy options for those individuals, resulting in greater injury risk when landing in a fatigued state. Based on the preceding literature review, it is apparent that ankle DF ROM can be chronically increased in as little as 3-weeks. However, the effects of increased ankle DF ROM on landing mechanics is currently not known but could potentially contribute to enhancing landing performance and reducing injury risk factors.

Chapter 3

Reliability for inter-limb asymmetries in ankle dorsiflexion range of motion during the weight-bearing lunge test

A version of this Chapter has been published in International Journal of Sports Physical Therapy and the printed version can be found in Appendix 7.

3.1 Introduction

During bilateral landings, ankle DF ROM is required for efficient energy dissipation (Fong et al., 2011; Yeow, Lee and Goh, 2011). Limited ankle DF ROM has been reported to affect lower-limb force profiles during landings, as ankle DF ROM restriction has been shown to correlate with greater peak vGRF caused by stiffer landing strategies (Fong et al., 2011). As a result, individuals with limited ankle DF ROM may exhibit movement strategies with gross technical errors during bilateral (Fong et al., 2011; Dill et al., 2014; Macrum et al., 2012) and unilateral (Dill et al., 2014; Mauntel et al., 2013) squatting and landing tasks, as well as during gait (Ota et al., 2014). Reduced ankle DF ROM during weight-bearing has been identified as being a modifiable risk factor for many lower limb injuries, with weight-bearing ankle DF ROM of 34° being associated with 2.5 times greater injury risk in military recruits (Pope, Herbert and Kirwan, 1998). Limitations in weight-bearing ankle DF ROM has also been shown to present as a risk factor for hamstring strains in Australian football athletes (relative risk = 2.32) (Gabbe et al., 2006). Furthermore, elite junior basketball players with weight-bearing ankle DF ROM values < 36.5° possess an 18.5% to 29.4% risk of developing patella tendinopathy within a year (Backman and Danielson, 2011). This risk is significantly

greater than the 1.8% to 2.1% for players with > 36.5° ankle DF ROM (Backman and Danielson, 2011). Therefore, restrictions in weight-bearing ankle DF ROM may increase injury risk through the development of mechanical compensations during athletic activities.

Unilateral restrictions in ankle DF ROM may result from injury to the rearfoot complex and have been identified using the WBLT (toe-wall distance) in patients with a history of unilateral ankle sprain (mean difference between injured and uninjured limb = -3.4 cm) (Reid, Birmingham and Alcock, 2007). Furthermore, inter-limb asymmetries in ankle DF ROM have been suggested to occur in response to the functional demands placed on the ankle complex (Moreno-Pérez et al., 2019; Rabin et al., 2015). As such, athletes with a history of lower-leg injury or those exposed to asymmetrical loading might have an inter-limb asymmetry in ankle DF ROM. Although current literature does not provide a clear understanding of the influence inter-limb asymmetries may have on an athlete's performance (Bishop, Turner and Read, 2018), asymmetries in ankle DF ROM have been negatively correlated with performance deficits during change of direction tests ($r = -0.52$) (Gonzalo-Skok et al, 2015). Therefore, asymmetries in ankle DF ROM may have implications for the performance of athletic movements and should be established in healthy populations during initial screening.

However, research investigating normative values for weight-bearing ankle DF ROM has provided conflicting evidence regarding the extent of asymmetries (Cosby and Hertel, 2011; Hoch and McKeon, 2011; Gonzalo-Skok et al, 2015; Rabin et al., 2015). Cosby and Hertel (2011) showed only a 0.8˚ difference in weight-bearing ankle DF ROM using a lunge test with a bent knee. Similarly, Konor et al. (2012) found no difference between left and right

sides during the WBLT in healthy adults. However, normative data from Hoch and McKeon (2011) demonstrated inter-limb asymmetries for ankle DF ROM in healthy participants frequently reached 1.5 cm when measuring toe-wall distance. Furthermore, Rabin et al. (2015) identified greater ankle DF ROM for the non-dominant leg exceeding 10° in 23% of male military recruits. Together, these findings suggest that more evidence is required indicating the prevalence of asymmetries is ankle DF ROM for healthy populations.

Better delineation of relative ankle DF ROM symmetry as measured in a weight-bearing position has several potential clinical and research purposes. Clinically, this information could be used to inform the course of treatment during the rehabilitation process or while prescribing interventions to increase ankle DF ROM. Furthermore, it is common practice to perform bilateral comparisons when assessing deficits in ankle DF ROM, which might lead to diagnostic errors if symmetry is assumed but not present. Without prior assessment and knowledge of normative ankle DF ROM asymmetries, the rehabilitation programme for an athlete with an asymmetry could be misjudged through a lack of consideration for the functional demands placed on the ankle joint.

In order to identify asymmetries in ankle DF ROM that are relevant to functional activities, it has been suggested that using an active weight-bearing assessment provides the most valid representation of ankle DF ROM capacity during dynamic tasks such as squatting and landing (Dill et al., 2014; Whitting et al., 2011). As such, the WBLT has been the subject of many recent investigations (Konor et al., 2012; Langarika-Rocafort et al., 2017; Powden, Hoch and Hoch, 2015). However, a number of different measurement methods can be used to quantify ankle DF ROM during the WBLT, including measuring tibia angle with either a

standard goniometer or inclinometer (Konor et al., 2012; Langarika-Rocafort et al., 2017), Achilles tendon angle with an inclinometer (Langarika-Rocafort et al., 2017), or the toe-wall distance measured using a tape measure (Bennell et al., 1998; Langarika-Rocafort et al., 2017). In an attempt to establish the most reliable method to measure ankle DF ROM during the WBLT, Langarika-Rocafort et al. (2017) compared five commonly used techniques; heel-wall distance, toe-wall distance, tibia angle, Achilles tendon angle and a trigonometric angle derived from heel-wall distance and knee-ground distance. The trigonometric calculation method had the highest between-session intra-rater reliability (ICC = 0.95, SEM = 1.18˚) compared to measurements of tibia angle (ICC = 0.87, SEM = 2.17˚) and Achilles angle (ICC = 0.87, SEM = 2.28˚) (Langarika-Rocafort et al., 2017). The trigonometric calculation method may present as a more reliable tool for clinicians to establish ankle DF ROM during the WBLT.

While the between-session intra-rater reliability of the trigonometric calculation method has been established (Langarika-Rocafort et al., 2017), the within-session intra-rater reliability has yet to be determined. Furthermore, the extent of inter-limb asymmetries in a young, healthy, and active cohort has yet to be established. The aims of this study, therefore, were: i) to establish values of ankle DF ROM asymmetry, ii) identify the influence of leg dominance on ankle DF ROM and iii) to determine the within-session, intra-rater reliability of the trigonometric calculation method during the WBLT in healthy and recreationally active participants for both a single-limb and the symmetry values measured.

3.2 Methods

3.2.1 Study design

Participants reported to the laboratory for a single testing session. Testing was conducted by the lead investigator who had 10 years' experience measuring ankle DF ROM using the WBLT and an accredited member of the British Association of Sport Rehabilitators and Trainers. Prior to data collection, all participants completed a pre-exercise questionnaire and provided written informed consent. Following the recording of height and body mass, participants reported their dominant leg, defined as their preferred leg for kicking a ball. Ankle DF ROM for both legs was then measured using the WBLT with no prior warm-up using a randomised counterbalanced design. Following a 10 minute rest, participants were re-tested to determine within-session reliability of the WBLT using the trigonometric calculation method.

3.2.2 Participants

Using the findings of Rabin et al. (2015) for inter-limb asymmetries for ankle DF ROM between the dominant and non-dominant limb (ES = 0.83), a representative analysis to determine the appropriate sample size was performed. Calculations indicated that to achieve 80% statistical power, a minimum of 39 participants were required to detect inter-limb asymmetries. A total of 50 participants volunteered for the study (28 men, 22 women, age = 22 ± 4 years, height = 1.73 ± 0.11 m, body mass 71.5 ± 15.1 kg). All participants self-reported to be physically active, defined as regularly performing at least 30 minutes of moderate intensity physical activity 3 times per week for at least 6 months prior to testing (Mauntel et al., 2013). Participants were excluded if they had a history of a lower extremity surgical procedure or injury to the lower extremity in the six-months prior to testing. Ethical approval was provided by the University of Cumbria Research Ethics Panel (Appendix 1).

3.2.3 Procedures

In order to measure the heel-wall distance (see Figure 3.1), a 0.70 m tape measure was fixed to the floor, perpendicular to the wall used for testing. Measurements of knee-ground distance were obtained with a 0.70 m tape measure fixed vertically to the wall and perpendicular to the tape measure on the ground (see Figure 3.1). A longitudinal line was marked down on each of the scales for testing purposes. Prior to performing the test, participants were provided with a demonstration and standardised instructions. Participants then completed three familiarisation trials for each leg before performing three trials on each limb, with the mean value from the three attempts from each foot being used for data analysis.

To ensure neither the participant nor investigator could target a specific outcome on subsequent attempts, no markings were made on the tape measure that would indicate the previous attempt. Following a 10 minute break participants were re-tested using the same procedures on both legs to establish within-session reliability. The results were recorded on a separate sheet to blind the investigator from previous distances and participants were not informed of their previous scores. For all participants, leg order was randomised for both trial 1 and 2. The testing procedures and measurements used for the trigonometric calculation are shown in Figure 3.1.
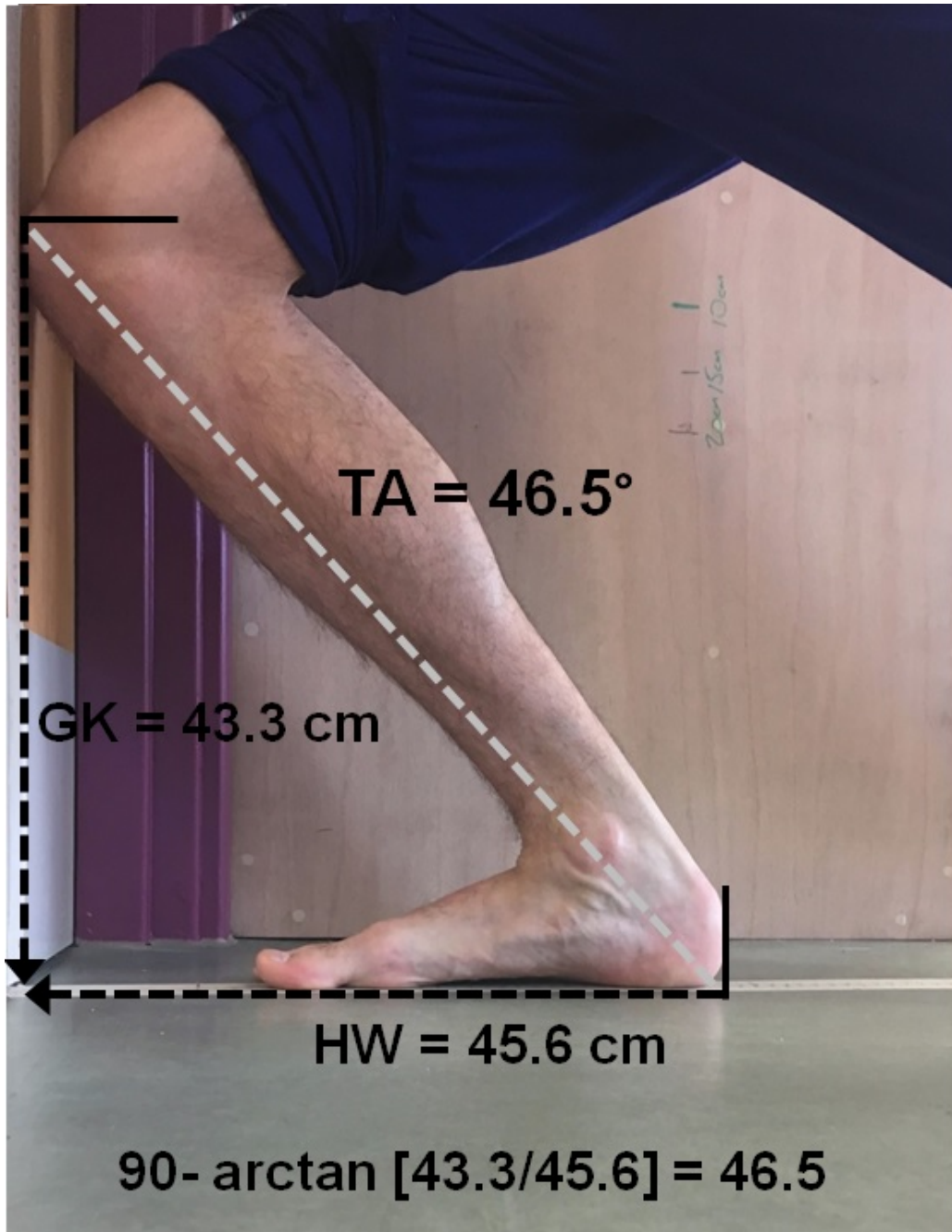
Figure 3.1. Participant performing the WBLT with example distances and calculation. Abbreviations: GK, ground-knee distance; HW, heel-wall distance; TA, trigonometric angle.

Participants began the test by facing a bare wall, with the greater toe of the test leg positioned against the wall. The greater toe and the centre of the heel were aligned using the marked line on the ground. Participants were instructed to place the non-test foot behind them, with the

heel raised and at a distance that they felt helped maximise their performance on the test. This position was established during familiarisation. In order to maintain balance, participants were asked to keep both hands firmly against the wall throughout. The participants were then instructed to slowly lunge forward by simultaneously flexing at the ankle, knee and hip on the test leg in an attempt to make contact between the centre of the patella and the vertical marked line on the wall. No attempt was made to control trunk alignment. Subtalar joint position was controlled by keeping the test foot in the standardised position and ensuring the patella contact with the vertical line was accurate (Konor et al., 2012).

The aim of the test was for the participant to get their heel as far away as possible from the wall, while making contact between the patella and the wall and maintaining firm pressure between the heel and the ground. Throughout the test, the investigator was positioned behind the participant in a low crouched position in order to visually monitor heel lift. Heel lift was defined as the visual elevation of the calcaneus, resulting in a greater ground surface area observed under the rearfoot. Any elevation of the heel during the test was regarded as a failed attempt and feedback was provided to the participants regarding their inability to prevent the heel from rising.

Upon successful completion of an attempt, where contact between the patella and the wall was made with no change in heel position relative to the ground, participants were instructed to move the test foot further away from the wall by approximately 0.5 cm. Although participants were not restricted to the number of attempts they were permitted at a given distance, no more than three attempts were performed by any participant. At the last successful attempt, the distances between the heel and the wall, and the distance between the

anterosuperior edge of the patella and the ground were recorded to the nearest 0.1 cm. Tibia angle for each attempt was calculated with the heel-wall and ground-knee distances, using the trigonometric function outlined by Langarika-Rocafort et al. (2017) (DF ROM = 90 – arctangent [knee-ground distance/heel-wall distance]). Inter-limb differences for ankle DF ROM were calculated by subtracting the left value from the right value. A positive value indicated the right had greater joint displacement for the corresponding segment and *vice versa* for a negative value.

3.2.4 Statistical analysis

The assumption of normality for data sets was checked and confirmed using the Shapiro-Wilk test. Normative data for the inter-limb mean difference for ankle DF ROM from the first test was graphically presented using a frequency-distribution histogram. An independent *t*-test was performed to establish the difference between the dominant and non-dominant limb for ankle DF ROM during the WBLT using the data from the first test. ES were calculated for each comparison, with 0.20 being considered *small*, 0.50 *moderate* and 0.80 or greater *large* (Cohen, 1988).

The within-session intra-rater reliability for single-limb measurements of ankle DF ROM and inter-limb differences in ankle DF ROM was initially assessed using a paired samples *t*-test to calculate systematic bias between trial 1 and 2 (Atkinson and Nevill, 1998). To examine for heteroscedastic errors, the relationship between the mean values between tests and the difference between repeat tests was evaluated using Pearson's correlation coefficient. Relative reliability was determined using ICC as described by Hopkins (2016) and reported with 95% confidence intervals, with ICCs interpreted as follows: 0.01–0.3 *poor*, 0.3–0.5

moderate, 0.5–0.7 *large*, 0.7–0.9 *very large*, and > 0.9 *nearly perfect* (Hopkins, 2016). Absolute reliability was calculated using the CV% (SD / mean *100), the 95% limits of agreement (LOA), SEM (SD√1-ICC) (Atkinson and Nevill, 1998) and MDC (SEM*1.96*√2) (Riemann and Lininger, 2018). All statistical tests were performed using SPSS® statistical software package (v.24; SPSS Inc., Chicago, IL, USA), with the *a-priori* level of significance set at $P < 0.05$. ICC and CV% were calculated using a customised spreadsheet (Hopkins, 2015). Due to the between-limb differences for asymmetries in ankle DF ROM being interval data, CV% was not calculated.

3.3 Results

Forty-one participants (82%) reported their dominant leg to be their right, with the remaining nine participants (18%) reporting their left leg as dominant. WBLT values are summarised in Table 3.1. The mean difference for inter-limb asymmetries in ankle DF ROM from the first test was -0.8˚ ± 3.0˚ (Figure 3.2). Mean WBLT values for the dominant and non-dominant limb were 36.5 ± 4.5˚ and 36.5 ± 4.3˚, respectively. No statistical difference ($P = 0.862$) was identified between the dominant and non-dominant limb.
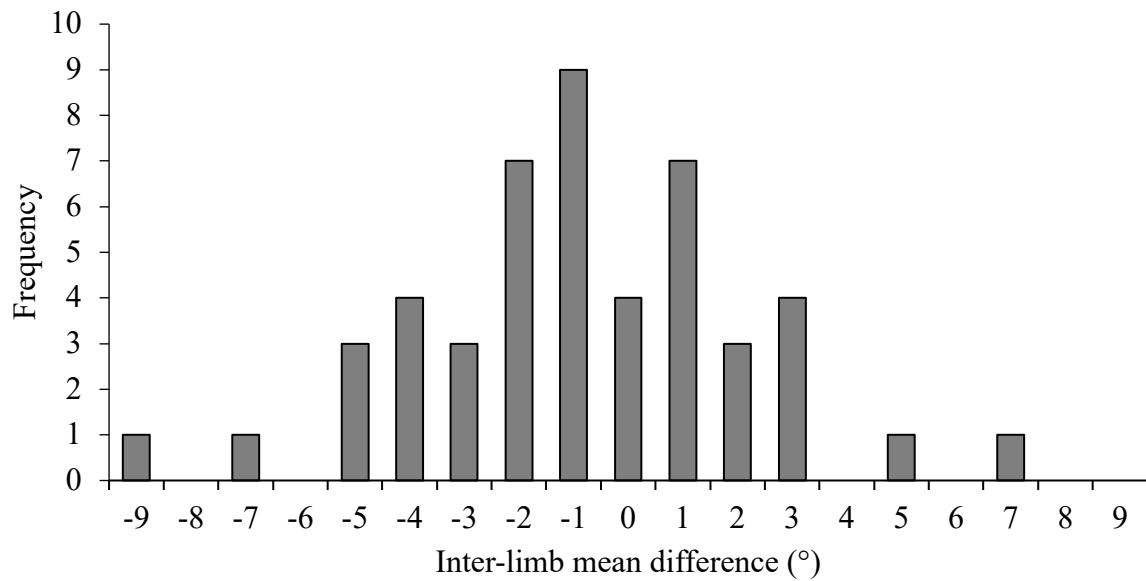
Figure 3.2. Inter-limb differences measured from the WBLT (*n* = 50). Positive values indicate greater ankle DF ROM for the right limb and *vice versa* for a negative value. Zero values indicate no difference.

The within-session reliability of the WBLT is summarised in Table 3.2. There were no systematic biases or heteroscedasticity found for the WBLT using the trigonometric calculation method between trials for either a single measure of ankle DF ROM or ankle DF ROM asymmetry ($P > 0.05$). The relative reliability was established as *nearly perfect* for within-session reliability for a single measure (ICC = 0.98) and inter-limb asymmetries in ankle DF ROM (ICC = 0.94). All values representing relative and absolute reliability are reported in Table 3.2.

Table 3.1. Asymmetry within the WBLT for dominant-to-non-dominant limb comparison ($n = 50$).

Ankle dorsiflexion	Range of motion (˚) (mean ± SD)	Difference (˚) (95% confidence interval)	Effect size
Dominant side	36.5 ± 4.5		
		-0.1 (-1.0, 0.8)	0.02
Non-dominant side	36.5 ± 4.3		

Table 3.2. Within-session intra-rater reliability for the WBLT using the trigonometric calculation method for testing ankle DF ROM for a single-limb and ankle DF ROM between-limb difference ($n = 50$). For ankle DF ROM inter-limb differences, positive values indicate greater ankle DF ROM for the right limb and *vice versa* for a negative value.

Reliability measure	Change in mean (˚)	ICC (95% confidence interval)	95% LOA (˚)	CV% (95% confidence interval)	SEM (˚)	MDC (˚)
Ankle DF ROM	0.1	0.98 (0.97, 0.99)	0.1 ± 1.8	1.7 (1.5, 2.0)	0.6	1.7
Ankle DF ROM inter-limb difference	-0.6	0.94 (0.89, 0.97)	-0.6 ± 2.4	Not applicable	0.8	2.3

3.4 Discussion

The primary aim of this study was to establish values for the inter-limb asymmetries of ankle DF ROM during the WBLT among healthy recreationally active individuals. Of all participants, 44% presented asymmetries in ankle DF ROM exceeding the MDC of 2.3° found in this investigation (Table 3.2), with 8% of participants demonstrating an inter-limb asymmetry greater than 5°, with the largest asymmetry being 8.8°. Therefore, with 44% of this sample having asymmetry values greater than the MDC, these findings suggest that the clinician should not assume symmetry without conducting thorough *a-priori* assessments.

This data supports the findings of Hoch and McKeon (2011) and Rabin et al. (2015), by identifying the existence of inter-limb asymmetries in ankle DF ROM during the WBLT in healthy populations. Using the toe-wall distance during the WBLT, Hoch and McKeon (2011) reported that 68% of participants exhibited an asymmetry of 1.5 cm or less, with some participants approaching asymmetries of approximately 3 cm. Using the conversion calculation suggested by Konor et al. (2012) where 1 cm in toe-wall distance corresponds with approximately 3.6˚ of ankle DF ROM, 32% of the sample in Hoch and McKeon (2011) demonstrated ankle DF ROM asymmetries of > 5.4°, with some participants approaching asymmetries of 10.8°. This is similar to that of Rabin et al. (2015), where 64 healthy male military recruits possessed a bilateral mean difference of 5.8° in favour of the non-dominant leg during the WBLT. Equally, 23% of participants had asymmetries > 10° (Rabin et al., 2015). Although the findings from this Chapter support the notion that bilateral differences are present in healthy populations, this data indicate that the magnitude of inter-limb asymmetry for ankle DF ROM is likely less than previously reported. These findings identify a much smaller mean asymmetry in comparison to previous investigations (Hoch and

McKeon, 2011; Rabin et al., 2015), with 56% of the population in this Chapter possessing inter-limb asymmetries on the WBLT of less than the MDC of 2.3°. Furthermore, none of the participants who volunteered for this study exceeded an asymmetry of 10°, with the greatest asymmetry recorded being 8.8° between limbs.

One possible reason for not observing a similar magnitude in asymmetry may be the method used to measure ankle DF ROM. Both measurement methods adopted by Hoch and McKeon (2011) and Rabin et al. (2015) used to record ankle DF ROM during the WBLT (toe-wall distance and tibia angle using an inclinometer, respectively) have been shown to possess a greater MDC for a single-limb than the 1.7˚ found in this investigation (Table 3.2) (Langarika-Rocafort et al., 2017). As the MDC represents the boundaries of measurement error (Riemann and Lininger, 2018), it is possible that the testing procedures used by both Hoch and McKeon (2011) and Rabin et al. (2015) may have contributed to the level of inter-limb asymmetry observed. For example, the MDC for the measurement method used by Rabin et al. (2015) has been reported to be > 6.0˚ for testing a single-limb (Langarika-Rocafort et al., 2017). Although it is unclear why the trigonometric calculation method provides greater reliability than other measurements of ankle DF ROM during the WBLT, it may be that measuring distances produces superior repeatability than measurements or calculations of angles. This suggestion is supported by Langarika-Rocafort et al. (2017), where ICC values for all distances associated with the trigonometric calculation method were much higher (ranging 0.95–0.96) than measuring tibia (0.87) and Achilles angle (0.87) during the WBLT.

To date, as far as can be ascertained through reviewing the literature, no previously published investigation has established the within-session intra-rater reliability for measuring asymmetries in tibia angle during the WBLT. The findings reported in this Chapter indicate that the error in measurement for inter-limb differences in ankle DF ROM (MDC = 2.3˚) is greater than the error associated with testing a single-limb (MDC = 1.7˚). Measurements of tibia angle using an inclinometer for single-limb ankle DF ROM during the WBLT have previously been shown to possess MDC values > 6.0˚ (Langarika-Rocafort et al., 2017). As the data reported in this Chapter showed greater error to be associated with measures of inter-limb asymmetries in ankle DF ROM, the mean inter-limb difference of 5.8˚ in ankle DF ROM (measured as tibia angle using an inclinometer) reported by Rabin et al. (2015) may represent error in the measurement technique that is compounded by testing both limbs. Although previous investigations have reported intra-rater MDC values as low as 3.2˚ when measuring tibia angle for a single-limb (Powden, Hoch and Hoch, 2015), none have established the reliability for measuring asymmetry. Therefore, it remains possible that the difference between the findings of Rabin et al. (2015) and that reported in this Chapter is due to measurement error associated with the techniques employed to establish inter-limb differences in ankle DF ROM.

Within this study, the MDC for a single-limb measurement for ankle DF ROM during the WBLT was identified as 1.7°, with a SEM of 0.6° (Table 3.2). These values for reliability are lower than reported for alternative measurement methods of tibia angle during the WBLT, with MDC and SEM values ranging between 3.1˚ to 6.4˚ and 1.0˚ to 2.4˚, respectively (Powden, Hoch and Hoch, 2015). Although all reported methods for measuring ankle DF ROM during the WBLT have been identified as having *very large* reliability (all ICC > 0.7) (Powden, Hoch and Hoch, 2015), Langarika-Rocafort et al. (2017) demonstrated that the

trigonometric calculation method used in this study possessed the highest intra-rater reliability and smaller MDC value in comparison to four other measurement methods. Based on the results reported in this Chapter, and those reported by Langarika-Rocafort et al. (2017), it is proposed that the trigonometric calculation method should be used when measuring ankle DF ROM asymmetries, as it appears to be a more sensitive measure. Practically, the trigonometric calculation method does not require specialised equipment, is time efficient, and presents a simple method for calculating ankle DF ROM (Langarika-Rocafort et al., 2017).

Despite the study reported in this Chapter using the same measurement technique as Langarika-Rocafort et al. (2017), the data reported here shows improved reliability. It can be tentatively speculated that one potential reason may be due to the administration of the WBLT. In order to identify peak ankle DF ROM angle during the WBLT, Langarika-Rocafort et al. (2017) relied upon participants informing the investigator of when they had reached maximum distance from the wall. In contrast, the measurement employed in this study was taken at the last successful attempt, which was defined as the furthest distance away from the wall where the participant could make contact between the patella and the wall and prior to the point of heel lift. Heel lift was carefully monitored by the investigator and defined as the visual lifting of the heel, where a greater surface area of the ground could be seen under the rearfoot. This is an important distinction, as it is questionable whether participants can identify at what point ankle DF ROM has terminated and compensatory strategies will be adopted, thus influencing the outcome measurement through a lack of standardisation. This may be especially problematic during the WBLT, as participants are unable to observe ankle motion on the test leg and thus, rely primarily on the sensorimotor system to provide information pertaining to when maximum ankle dorsiflexion angle has

been achieved. As the precision of the sensorimotor system's ability to determine joint position has been shown to vary considerably during similar tasks (Proske and Gandevia, 2012), the technique used by Langarika-Rocafort et al. (2017) for establishing end range of motion during the WBLT is not recommended and may explain the improved reliability found for the procedures in this study.

Leg dominance has previously been shown to possess a relationship with inter-limb asymmetry in ankle DF ROM, with greater ankle DF ROM observed in the non-dominant limb (Rabin et al., 2015). However, the results reported in this Chapter did not identify a difference in ankle DF ROM during the WBLT between the dominant and non-dominant leg. Although it remains unclear why a similar finding was not observed in this study, a few possibilities exist. Firstly, Rabin et al. (2015) proposed that asymmetries in ankle DF ROM between the dominant and non-dominant leg may exist due to the mechanical loading placed on the ankle complex during habitual activities. This is based on a rationale that the ankle joint complex adapts to the demands imposed upon it, with the non-dominant leg being subjected to larger requirements for balance and stability, resulting in greater joint range of motion (Rabin et al., 2015). As all participants in Rabin et al. (2015) were military recruits, it may be that specific physical activities undertaken by the participants in preparation for basic military training resulted in the ankle DF ROM asymmetries identified between the dominant and non-dominant leg, as opposed to the sample that participated in this study, who were physically active but not military trained. Another more likely explanation for the lack of agreement may be due to difference in procedures when conducting the WBLT. Unlike the trigonometric calculation method for measuring ankle DF ROM that was used in this study, Rabin et al. (2015) used an inclinometer placed on the tibia, 15 cm below the tibial tuberosity. As previously discussed, intra-rater reliability for this method has been reported to

be inferior to the trigonometric calculation method (see Langarika-Rocafort et al., 2017). As an analysis of intra-rater reliability was not included in Rabin et al. (2015), it is possible that the procedures used may have contributed to the contrast in findings.

Whether the asymmetry in ankle DF ROM observed in this investigation is clinically meaningful is, at present, unknown. Limitations in ankle DF ROM have been linked to reduced peak knee flexion angles (Fong et al., 2011) and increased peak knee abduction angles (Malloy et al., 2015) during landing activities and these suboptimal movement strategies are associated with anterior cruciate ligament injuries (Hewett et al., 2005; Renstrom et al., 2008). Large asymmetries in ankle DF ROM may, therefore, present as a modifiable variable for reducing risk factors associated with lower extremity injury during dynamic activities. Future studies should establish the relationship between asymmetries in ankle DF ROM and landing mechanics. This evidence would assist in determining the role asymmetries in ankle DF ROM has as a potential contributor to injury.

The results reported in this Chapter indicate that clinicians should not assume ankle DF ROM symmetry. The assumption of symmetry in ankle DF ROM during the rehabilitation of a patient would be inappropriate for restoring function. Instead, it may be more reasonable to identify whether the patient possesses sufficient ankle DF ROM to cope with the movement demands placed on them by the sport and activities of daily living. As activities such as squatting (Kasuyama, Sakamoto and Nakazawa, 2009), landing (Zhang, Bates and Dufek, 2000), running (Novacheck, 1998) and change of direction tasks (Riley et al., 2013) may all require large quantities of ankle DF ROM, ensuring an individual possesses sufficient range of motion to cope with these demands appears to be a more logical guide.

3.5 Conclusion

Recreationally active individuals may present with asymmetrical weight-bearing ankle DF ROM during the WBLT that is normal and not necessarily associated with leg dominance. The findings reported in this Chapter suggest the extent of asymmetry found using this technique is less than what has been previously reported in the literature. Furthermore, calculating weight-bearing ankle DF ROM for a single-limb using the trigonometric calculation method presents as a simple and reliable tool; however, the error associated with identifying asymmetries in weight-bearing ankle DF ROM may exceed the absolute inter-limb difference. Therefore, asymmetries in weight-bearing ankle DF ROM may be error associated with the testing procedures and not a true inter-limb difference. Future investigations should look to establish the mechanical implications of ankle DF ROM asymmetry during functionally relevant activities such as landings. As such, *Chapter 5* will investigate the association for between-limb differences in ankle DF ROM and asymmetries in landing mechanics.

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Chapter 4

Reliability of kinetic and 2D kinematic variables associated with bilateral drop-landing performance

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A version of this Chapter has been published in International Journal of Physical Education, Fitness and Sports and Movement & Sport Sciences/Science & Motricité and the printed versions can be found in Appendix 8 and 9 respectively.

4.1 Introduction

The performance of bilateral landings can expose an individual to peak vGRF equivalent to multiples of bodyweight (Bates et al., 2013; Ortega, Bíes and de la Rosa, 2010; Yanci and Camara, 2016) and have been identified as a movement pattern associated with lower extremity injury (Hewett, Myer and Ford, 2006). Those individuals at greater risk of injury during landings tend to use less effective movement strategies to dissipate forces in multiple planes (Aerts et al., 2013; Boling et al., 2009; Hewett et al., 2005; Padua et al., 2009). For example, in the sagittal plane, decreased ankle plantar flexion (Rowley and Richards, 2015) and knee flexion (Chappell et al., 2005) angle at initial contact, reduced hip (Blackburn and Padua, 2009) and knee flexion angle at the lowest point of the landing (Yu, Lin, and Garrett, 2006), and less ankle joint displacement following ground contact (Begalle et al., 2015) have all been shown to increase mechanical loading throughout the lower extremity. In the frontal and transverse plane, greater peak knee valgus angle during landing tasks has also been shown to increase lower extremity injury risk, secondary to greater knee abduction moments (Hewett et al., 2005). Given the established relationship with injury risk, it is common for

practitioners to pre-screen the bilateral landing strategy adopted by athletes (Bird and Markwick, 2016; Ludgren et al., 2015).

When coordination strategies to decelerate the centre of mass over a large range of motion are either not accessed as a movement solution (Zhang, Bates and Dufek, 2000) or are unavailable to an athlete due to ankle DF ROM restriction (Fong et al., 2011), the result is a higher peak vGRF. As outlined earlier in this Chapter, athletes who are exposed to greater peak vGRF during landings have an increased lower extremity injury risk (Dufek and Bates, 1991). For example, Hewett et al. (2005) showed that pre-screened female athletes who subsequently experienced anterior cruciate ligament injuries, produced normalised peak vGRF 20% higher than non-injured athletes during drop-landing tasks. Additionally, individuals who displayed higher peak vGRF in the 100 ms following ground contact, place greater strain on ligamentous structures located at the tibiofemoral joint (Norcross et al., 2010). Given that variables such as peak vGRF, time to peak vGRF and loading rate are commonly reported as being associated with injury risk during landings (Bisseling et al., 2008, Hewett et al., 2005; Radin et al., 1991), practitioners should be aware of the inherent error associated with testing procedures. This includes error on behalf of the athlete while performing a given protocol (biological error) and that of the equipment (technical error) (Atkinson and Nevill, 1998). Although previous investigations have reported the reliability for outcome measures relating to the propulsive phase of bilateral jumping tasks in various populations (Cormack et al., 2008; Hori et al., 2009; Markovic et al., 2004; Slinde et al., 2008), there is limited information on the kinetic factors associated with bilateral drop-landings (James et al., 2007).

Although 3D analysis is regarded as the gold standard in exploring lower limb kinematics, in practice 2D video analysis is more accessible to practitioners (Munro, Herrington and Carolan, 2012). However, before kinematic measurements of bilateral landing tasks can be used for the purpose of screening, reliability must first be established. It is therefore important to quantify the noise (error) of the proposed field-based measurements. As discussed in Section 2.13, for various landing tasks, 2D video analysis has been shown to be a reliable tool for measuring FPPA; a frontal plane representation of knee valgus/varus angle at the knee in the deepest landing position (Dingenen et al., 2014; McLean et al., 2005; Mizner et al., 2012; Munro, Herrington and Carolan, 2012). However, for joint angle measurements in the sagittal plane, only Dingenen et al. (2015) and King and Belyea (2015) have investigated the reliability of 2D analysis for measurements during bilateral landing activities. Studies investigating the reliability of 2D analysis have not considered key variables during landings that may be affected by ankle DF ROM restriction, such as lower extremity joint angles at initial contact and sagittal plane joint displacement for the ankle, knee, and hip joints (Dowling, McPherson and Paci, 2018; Fong et al., 2011). Furthermore, as the influence of drop height on the relationship between ankle DF ROM and landing mechanics has not been investigated, the reliability of 2D analysis from varying drop heights relative to an individual's force development capacity must also be established. Additionally, it is currently unknown whether 2D analysis possesses the sensitivity to detect acute or chronic changes in landing strategies during drop-landings that can occur in response to an exercise-induced fatigue protocol (i.e. *Chapter 6*) or training programme (i.e. *Chapter 7*), respectively. Therefore, this Chapter will provide the foundation for the interpretation of the subsequent studies.

An additional consideration when analysing kinetic and kinematic measures associated with bilateral drop-landings is inter-limb asymmetries in coordination. Asymmetries in peak vGRF during landings are commonly identified in healthy populations (Britto et al., 2015; Schott, Bates and Dufek, 1994). These asymmetries are an important consideration when working with an individual who performs a high volume of bilateral landings. Large asymmetry in peak vGRF during bilateral landings may subject the leg exposed to higher forces to excessive loading, thereby increasing the potential risk for overuse injury (Schot, Bates and Dufek, 1994). Similarly, asymmetry in landing strategies has been suggested to increase injury risk (Schot, Bates and Dufek, 1994) and commonly exist during bilateral landing tasks in uninjured (Pappas and Carpes, 2012) and injured populations (Meyer et al., 2018). However, the test re-test reliability for force platforms and 2D video analysis to detect inter-limb asymmetries has not been established for mechanical parameters of drop-landings. In such instances, reliable identification of bilateral asymmetry and subsequent interventions to reduce the magnitude of the asymmetry might be warranted and thus, in the first instance, it is necessary to investigate the reliability of asymmetries in kinetic and kinematic variables during bilateral landings. Therefore, in order to identify the effect inter-limb differences in ankle DF ROM may have on inter-limb asymmetries in landing mechanics (i.e. *Chapter 5*), reliability for asymmetries during bilateral drop-landing must first be determined.

The aim of the study reported in this Chapter was to assess the reliability of kinetic and kinematic variables associated with landing performance from varying drop heights. Furthermore, the study also aimed to assess the reliability of kinetic and kinematic measures of inter-limb asymmetries during bilateral drop-landings.

4.2 Methods

4.2.1 Study Design

A within-session repeated measures design was used to establish the reliability for all kinetic variables related to bilateral drop-landings. Participants were required to report to the laboratory for a single testing session. All test sessions were conducted between 10:00 am and 1:00 pm to control for circadian variation. Participants wore wearing spandex shorts and vest so that key landmarks were recognisable by all cameras. After familiarisation, participants performed three CMJ to establish maximum jump height for the landing task. Subsequently, participants performed five bilateral drop-landings from three heights: 50% of their maximum CMJ, 100% of their maximum CMJ and 150% of their maximum CMJ. The participants then repeated the bilateral drop-landings from each height following a 10 minute recovery.

4.2.2 Participants

Thirty-nine male ($n = 22$; age = 22 ± 4 years; height = 1.80 ± 0.06 m; mass = 77.9 ± 14.0 kg) and female ($n = 17$; age = 20 ± 4 years; height = 1.65 ± 0.09 m; mass = 60.3 ± 9.8 kg) recreational athletes volunteered to participate. Recreational athletes were defined as a person who regularly competes 1–3 times per week in sport events involving landings activities, such as court, racquet or team sports (Chappell et al., 2002). Participants were excluded if they had a history of lower extremity surgery or had lower extremity injury six-months prior to testing. All participants were informed of the risks associated with the testing, prior to completing a pre-exercise questionnaire and providing informed written consent. Ethical approval was provided by the University of Cumbria Research Ethics Panel (Appendix 1).

4.2.3 Procedures

Participants firstly performed a 5 minute standardised warm-up and three familiarisation CMJ attempts. CMJ were performed from a standing position with each foot placed on a portable force platform recording at 1000 Hz (Pasco, Roseville, CA, USA). The force platforms were positioned side-by-side, 0.05 m apart and embedded in custom built wooden mounts that were level with the force platforms, preventing any extraneous movement that could influence the force trace recorded. In bare feet, participants were informed to stand with their feet hip-width apart and with hands on their hips to eliminate the contribution of the arm swing. Participants were then asked to rapidly descend prior to explosively jumping as high as possible, with no control being placed on the depth or duration of the countermovement (Benjanuvatra et al., 2013). Upon landing, participants were required to ensure that full contact was made between each foot and the respective force platforms, with trials excluded if either foot made contact with the wooden mounts or neighbouring force platform. Following familiarisation, participants performed three CMJ for data analysis with a 60 s recovery between trials. Using a custom-made Microsoft Excel spreadsheet, the force-time data was analysed using the time in the air method to calculate vertical jump height to the nearest 0.01 m using the following equation (Moir, 2008):

$$\text{Time in the air jump height (cm)} = \tfrac{1}{2}\, g(t/2)^2$$

where g represents the acceleration of gravity (9.81 m/s^2) and t represents the time in the air (s). Time in the air was determined as the period where force was less than 10 N. The maximum value of the three attempts was then used to calculate drop height for the bilateral drop-landings.

Following the performance of the CMJ, reflective markers were placed directly onto the participants' skin by the same investigator using the anatomical locations for sagittal plane lower extremity joint movements and FPPA outlined by Dingenen et al. (2015) and Munro, Herrington and Carolan (2012), respectively. For sagittal plane views, reflective markers were placed on both left and right acromioclavicular joints, greater trochanters, lateral femoral condyles, lateral malleolus and 5th metatarsal heads (Dingenen et al., 2015). For FPPA, reflective markers were placed on both left and right centre of the knee joint (midpoint between the femoral condyles), centre of the ankle joint (midpoint between the malleoli) and on the proximal thigh (midpoint between the anterior superior iliac spine and the knee marker). Midpoints for the knee and ankle were measured with a standard tape measure (Seca 201, Seca, United Kingdom), as outlined by Munro, Herrington and Carolan (2012).

Participants were then familiarised with the bilateral drop-landings from drop heights of 50%, 100% and 150% of maximum CMJ height. Bilateral drop-landings were performed with participants standing bare foot with their arms folded across their chest on a height-adjustable platform (to the nearest 0.01 m) 0.15 m away from two force platforms (Munro, Herrington and Carolan, 2012). Participants were then instructed to step off the platform whilst ensuring that they did not modify the height of the centre of mass prior to dropping from the platform. No feedback on landing performance was provided at any point during testing. For each drop height, participants performed five landings for data collection, with 60 s recovery provided between landings. Following the performance of the initial five landings from each drop height (test 1), participants rested for 10 minutes prior to repeating the standardised warm-up and the bilateral drop-landing protocol (test 2). Drop height order was randomised using a

counterbalanced design for both test 1 and 2. Mean values for all variables using all five trials were calculated for test 1 and test 2. Five trials were used to calculate the mean based on previous investigations demonstrating a plateau in measures of reliability for landing kinetics and kinematics when > 4 trials were used for data analysis (James et al. 2007; Ortiz et al., 2007).

For 2D video analysis, sagittal and frontal plane joint movements were recorded using three standard digital video cameras sampling at 60 Hz (Panasonic HX-WA30). All cameras were set up using the procedures outlined by Payton (2007). For left and right sagittal plane joint movements, cameras were positioned 3.50 m from the centre of either force platform (Dingenen et al., 2015). To record frontal plane kinematics, a camera was placed 3.50 m in front of the centre of the force platforms (Dingenen et al., 2014). All cameras were placed on a tripod at a height of 0.60 m from the ground (Dingenen et al., 2014; Dingenen et al., 2015).

4.2.4 Data analysis

Raw vGRF data were low-pass filtered using a fourth-order Butterworth filter with a cut-off frequency of 50 Hz (Roewer et al., 2014). Peak vGRF, time to peak vGRF, and loading rate was then calculated unilaterally for the right and left leg, as well as bilaterally. For bilateral measures, both the left and right force data were summed prior to analysis. Peak vGRF data was normalised to body mass ($N \cdot kg^{-1}$). For time to peak vGRF to be determined, initial contact was identified as the point that vGRF exceeded 10 N for each limb and bilaterally (Hoch et al., 2015). Time to peak vGRF was then calculated as the time difference between initial contact and the time point where peak vGRF occurred. Loading rate was calculated as normalised peak vGRF divided by time to peak vGRF (Paterno et al., 2007).

Asymmetries for peak vGRF normalised to body mass were calculated as the percentage difference between limbs using Microsoft Excel[TM] and the method outlined by Bishop et al. (2018) using the following equation:

$$(Right\ Limb - Left\ Limb) / (Right\ Limb + Left\ Limb) * 100$$

A positive value indicated the right limb had the largest score and *vice versa* for a negative asymmetry value.

All video recordings were analysed with free downloadable software (Kinovea for Windows, Version 0.8.15). For sagittal plane joint movements, ankle, knee and hip joint angles were calculated at initial contact and peak flexion for both limbs. These angles were then used to calculate joint displacement for each joint by subtracting the peak flexion angle from the initial contact angle. Initial contact was defined as the frame prior to visual impact between the foot and the ground that led to deformation of the foot complex. Peak flexion was identified visually and defined as the frame where no more downward motion occurred at the ankle, knee and hip joints (Dingenen et al., 2015).

Hip flexion angle was calculated as the angle between a line formed between the acromioclavular joint and the greater trochanter and a line between the greater trochanter and the lateral femoral condyle. Knee flexion angle was calculated as the angle between a line formed between the greater trochanter and the lateral femoral condyle and a line between the

lateral femoral condyle and the lateral malleolus. Ankle dorsiflexion angle was calculated as the angle between a line formed between the lateral femoral condyle and the lateral malleolus and a line between the lateral malleolus and the 5th metatarsal head. FPPA was calculated at the deepest landing position, defined as the frame corresponding to maximum knee flexion (Munro, Herrington and Carolan, 2012). This angle was calculated as the angle between the line formed between the proximal thigh marker and the knee joint marker and a line between the knee joint marker and the ankle joint marker (Munro, Herrington and Carolan, 2012). For initial contact and peak flexion, smaller values represented greater hip flexion, knee flexion and ankle dorsiflexion for the hip, knee and ankle joints, respectively. For FPPA, values < 180° represented knee valgus and values > 180° represented knee varus.

Inter-limb differences for sagittal plane joint displacement were calculated by subtracting the left value from the right value for the ankle, knee and hip joints. A positive value indicated the right limb had greater joint displacement for the corresponding segment and *vice versa* for a negative value.

For establishing intra-rater reliability of the hip, knee and ankle joint angle at initial contact and at peak flexion, along with FPPA, the first trial from drop heights of 150% of CMJ height was examined. Additionally, intra-rater reliability was determined for the time point at which the moment of peak flexion occurred during a landing. Twenty randomly selected participants (11 males and 9 females) were examined twice by the same investigator, seven days apart. To determine intra-rater reliability for all variables, two-way mixed (single measure) ICC and SEM for the same trial was established using a customised spreadsheet (Hopkins, 2015). All 2D kinematic outcome measures showed excellent intra-rater reliability,

with ICC for joint angles at initial contact ranging from 0.96 to 0.98 and all SEM values < 1.2°. ICC for joint angles at the peak flexion ranged from 0.95 to 0.99, with all SEM values < 1.5°. ICC for the moment of peak flexion were 0.99 and SEM were 0.01 s.

4.2.5 Statistical analysis

Descriptive statistics (means ± standard deviation) were calculated for all variables for each limb. Independent variables were calculated for both limbs, with all kinetic measures being also calculated bilaterally. Additionally, inter-limb asymmetry was calculated for peak vGRF, as well as inter-limb differences for ankle, knee and hip joint displacement. The assumption of normality was confirmed using the Shapiro-Wilk test. To examine for heteroscedastic errors, the relationship between the mean values between tests and the difference between repeat tests was evaluated using Pearson's correlation coefficient. The within-session reliability for all variables was initially assessed using a paired samples t-test to calculate systematic bias between test 1 and test 2 from each drop height (Atkinson and Nevill, 1998). The α-priori level of significance was set at $P < 0.05$, with a Bonferroni correction applied. Relative reliability was determined using ICC as described by Hopkins (2016) and reported with 95% confidence intervals, with ICCs interpreted as follows: 0.01–0.3 *poor*, 0.3–0.5 *moderate*, 0.5–0.7 *large*, 0.7–0.9 *very large*, and > 0.9 *nearly perfect* (Hopkins, 2016). Absolute reliability was calculated using the CV%, the 95% LOA, SEM (SD√1-ICC) (Atkinson and Nevill, 1998) and MDC (SEM*1.96*√2) (Riemann and Lininger, 2018). Due to the inter-limb differences in joint displacement and asymmetry in peak vGRF being interval data, CV% was not determined. ICC and CV% were calculated using a customised spreadsheet (Hopkins, 2015). This investigation reported a variety of statistical outcomes to

facilitate interpretation of the results by researchers and practitioners. All statistical tests were performed using SPSS® statistical software package (v.24; SPSS Inc., Chicago, IL, USA).

4.3 Results

4.3.1 Kinetic measures associated with bilateral drop-landing performance

The group mean for CMJ height was 0.30 ± 0.08 m. Relative and absolute values of reliability for all kinetic variables are presented in Tables 4.1–4.4. There was no systematic bias or heteroscedasticity found between test 1 and 2 for any variable for each drop height. For measures of peak vGRF, relative reliability was *nearly perfect* (ICC $\geq$ 0.90) for all variables except peak vGRF on the right extremity from the 50% CMJ drop height, which had *very large* relative reliability (ICC = 0.87). Measures of absolute reliability for peak vGRF are reported in Table 4.1, with CV% ranging from 7.1–13.0% for all variables. Time to peak vGRF demonstrated relative reliability of *large* to *near perfect* across all drop heights (ICC = 0.57–0.92). However, absolute reliability was improved for drop heights of 150% CMJ height (CV% = 6.6–9.5%) when compared to drop heights of 100% CMJ height (CV% = 10.5–13.1%) and 50% CMJ height (CV% = 14.9–27.6%) for time to peak vGRF (Table 4.2). Loading rate possessed *very large* to *near perfect* relative reliability (ICC = 0.86–0.95) across all drop heights, and absolute reliability establishing CV% ranging between 13.0–27.6% (Table 4.3). Measures of reliability for asymmetries in peak vGRF are shown in Table 4.4, with relative reliability shown to be *very large* (ICC = 0.72–0.74).

Table 4.1. Within-session reliability for normalised peak vGRF for bilateral drop-landing from all drop heights ($n = 39$).

	Test 1 Mean ± SD ($N \cdot kg^{-1}$)	Test 2 Mean ± SD ($N \cdot kg^{-1}$)	Change in mean ($N \cdot kg^{-1}$)	95% LOA ($N \cdot kg^{-1}$)	ICC (95% CI)	CV%	SEM ($N \cdot kg^{-1}$)	MDC ($N \cdot kg^{-1}$)
Drop height 50% of maximum CMJ height								
Total peak vGRF	2.74 ± 0.91	2.71 ± 0.91	-0.03	-0.03 ± 0.79	0.90 (0.84 – 0.94)	9.4	0.28	0.78
Right peak vGRF	1.76 ± 0.64	1.70 ± 0.54	-0.06	-0.06 ± 0.61	0.87 (0.78 – 0.92)	13.0	0.21	0.60
Left peak vGRF	1.23 ± 0.41	1.22 ± 0.44	0.01	0.01 ± 0.33	0.92 (0.87 – 0.96)	10.0	0.12	0.32
Drop height 100% of maximum CMJ height								
Total peak vGRF	3.41 ± 1.17	3.21 ± 0.95	-0.20	-0.20 ± 0.85	0.92 (0.87 – 0.95)	8.8	0.30	0.83
Right peak vGRF	2.02 ± 0.75	1.93 ± 0.63	-0.10	-0.10 ± 0.56	0.92 (0.86 – 0.95)	10.1	0.20	0.55
Left peak vGRF	1.62 ± 0.58	1.54 ± 0.51	-0.09	-0.09 ± 0.46	0.91 (0.86 – 0.95)	11.2	0.16	0.45
Drop height 150% of maximum CMJ height								
Total peak vGRF	4.18 ± 1.27	3.99 ± 1.28	-0.18	-0.18 ± 0.77	0.95 (0.92 – 0.97)	7.1	0.27	0.75
Right peak vGRF	2.43 ± 0.80	2.32 ± 0.78	-0.11	-0.11 ± 0.65	0.92 (0.86 – 0.95)	9.6	0.23	0.63
Left peak vGRF	2.11 ± 0.75	2.06 ± 0.76	-0.06	-0.06 ± 0.49	0.95 (0.91 – 0.97)	9.7	0.17	0.47

Table 4.2. Within-session reliability for time to peak vGRF for bilateral drop-landing from all drop heights ($n = 39$).

	Test 1 Mean ± SD (s)	Test 2 Mean ± SD (s)	Change in mean (s)	95% LOA (s)	ICC (95% CI)	CV%	SEM (s)	MDC (s)
Drop height 50% of maximum CMJ height								
Total time to peak vGRF	0.088 ± 0.031	0.092 ± 0.035	0.004	0.004 ± 0.038	0.84 (0.74 – 0.90)	15.9	0.013	0.037
Right time to peak vGRF	0.077 ± 0.022	0.081 ± 0.025	0.005	0.005 ± 0.033	0.75 (0.61 – 0.85)	14.9	0.012	0.033
Left time to peak vGRF	0.114 ± 0.057	0.108 ± 0.045	-0.006	-0.006 ± 0.094	0.57 (0.37 – 0.73)	27.6	0.034	0.093
Drop height 100% of maximum CMJ height								
Total time to peak vGRF	0.068 ± 0.023	0.068 ± 0.022	0.004	0.004 ± 0.034	0.91 (0.84 – 0.94)	10.7	0.007	0.019
Right time to peak vGRF	0.065 ± 0.021	0.064 ± 0.015	-0.001	-0.001 ± 0.021	0.84 (0.74 – 0.90)	10.5	0.007	0.020
Left time to peak vGRF	0.080 ± 0.035	0.080 ± 0.035	0.000	0.000 ± 0.033	0.89 (0.82 – 0.94)	13.1	0.011	0.032
Drop height 150% of maximum CMJ height								
Total time to peak vGRF	0.055 ± 0.014	0.056 ± 0.014	0.001	0.001 ± 0.017	0.82 (0.72 – 0.89)	9.5	0.006	0.016
Right time to peak vGRF	0.053 ± 0.012	0.054 ± 0.012	0.001	0.001 ± 0.010	0.91 (0.85 – 0.95)	6.6	0.004	0.010
Left time to peak vGRF	0.063 ± 0.027	0.063 ± 0.023	0.000	0.000 ± 0.021	0.92 (0.86 – 0.95)	8.7	0.007	0.020

Table 4.3. Within-session reliability for loading rate for bilateral drop-landing from all drop heights ($n = 39$).

	Test 1 Mean ± SD (N·s^{-1})	Test 2 Mean ± SD (N·s^{-1})	Change in mean (N·s^{-1})	95% LOA (N·s^{-1})	ICC (95% CI)	CV%	SEM (N·s^{-1})	MDC (N·s^{-1})
Drop height 50% of maximum CMJ height								
Total loading rate	40.3 ± 25.3	38.7 ± 27.9	-1.6	-1.6 ± 26.3	0.88 (0.80 – 0.93)	20.9	9.3	25.7
Right loading rate	28.1 ± 18.0	25.8 ± 16.2	-2.3	-2.3 ± 16.8	0.88 (0.80 – 0.93)	23.4	5.9	16.4
Left loading rate	16.2 ± 11.6	16.2 ± 13.7	0.0	0.0 ± 13.4	0.86 (0.77 – 0.92)	27.6	4.7	13.2
Drop height 100% of maximum CMJ height								
Total loading rate	61.5 ± 37.9	54.8 ± 27.3	-6.7	-6.7 ± 30.9	0.89 (0.82 – 0.94)	16.1	10.9	30.2
Right loading rate	38.0 ± 24.0	35.0 ± 19.3	-3.0	-3.0 ± 17.3	0.92 (0.87 – 0.95)	16.7	6.1	16.8
Left loading rate	27.1 ± 18.9	24.0 ± 14.0	-3.1	-3.1 ± 15.6	0.89 (0.82 – 0.94)	22.8	5.5	15.2
Drop height 150% of maximum CMJ height								
Total loading rate	86.6 ± 42.5	81.1 ± 41.7	-5.5	-5.5 ± 26.7	0.95 (0.92 – 0.97)	13.0	9.4	26.0
Right loading rate	52.0 ± 27.4	49.3 ± 27.4	-2.7	-2.7 ± 19.1	0.94 (0.90 – 0.96)	14.0	6.7	18.7
Left loading rate	41.3 ± 24.1	40.1 ± 24.5	-1.3	-1.3 ± 15.1	0.95 (0.92 – 0.97)	17.0	5.3	14.7

Table 4.4. Within-session reliability for peak vGRF asymmetry for bilateral drop-landing from all drop heights ($n = 39$).

	Test 1	Test 2	Change in mean	95% LOA	ICC (95% CI)	SEM	MDC
	Mean ± SD	Mean ± SD	(%)	(%)		(%)	(%)
	(%)	(%)					
Peak vGRF asymmetry at 50% CMJ	17.4 ± 10.6	16.5 ± 11.6	-0.9	-0.9 ± 16.5	0.72 (0.57 – 0.83)	5.9	16.2
Peak vGRF asymmetry at 100% CMJ	10.9 ± 9.8	11.3 ± 10.9	0.4	0.4 ± 14.8	0.74 (0.60 – 0.84)	5.3	14.6
Peak vGRF asymmetry at 150% CMJ	7.7 ± 9.8	6.7 ± 10.8	-0.9	-0.9 ± 15.3	0.73 (0.57 – 0.83)	5.4	15.0

4.3.2 Kinematic measures associated with bilateral drop-landing performance

Relative and absolute values of reliability for all kinematic variables are presented in Tables 4.5–4.9. There was no systematic bias or heteroscedasticity found between test 1 and 2 for any variable for any drop height. The relative and absolute reliability values are presented in Table 4.5 for initial contact angles for both the right and left side for each drop height. Relative reliability ranged from *very large* to *near perfect* (ICC = 0.81–0.93) and CV% for initial contact variables ranged from 1.0–2.0% across all drop heights.

Table 4.6 presents reliability measures for peak flexion angles between tests, with relative reliability being *near perfect* (ICC = 0.92–0.97) and absolute reliability ranging between 1.9–7.9% for CV% for the hip, knee and ankle joints for all drop heights. Relative reliability for joint displacement ranged from *very large* to *near perfect* (ICC = 0.76–0.97) (Table 4.7). At drop heights of 50% CMJ height, greater absolute variability was identified for joint displacement values (CV% = 10.0–27.8%), but at a drop height of 100% CMJ height, joint displacements values all possessed CV% < 10%. However, at drop heights of 150% of CMJ height, joint displacement for hip exceeded CV% > 10% for both the right and left limb.

Table 4.8 presents the relative and absolute reliability values for FPPA for both the right and left limb for each drop height. Relative reliability ranged from *very large* to *perfect* (ICC = 0.88–0.94). CV% ranged from 1.2–2.3% across each drop height. Mean difference, 95% LOA, relative and absolute reliability values are presented in Table 4.9 for inter-limb differences in sagittal plane joint displacements for the ankle, knee and hip. Relative reliability for between-limb differences in sagittal-plane joint displacement ranged from *large* to *very large* (ICC = 0.50–0.84) with MDC values ranging between 6.0–13.2°.

Table 4.5. Within-session reliability for initial contact angles for bilateral drop-landing from all drop heights ($n = 39$).

	Test 1 Mean ± SD (°)	Test 2 Mean ± SD (°)	Change in mean (°)	95% LOA (°)	ICC (95% CI)	CV%	SEM (°)	MDC (°)
Drop height 50% of maximum CMJ height								
Right ankle	148.6 ± 6.9	147.6 ± 7.5	-0.9	-0.9 ± 6.5	0.90 (0.82 – 0.95)	1.6	2.3	6.3
Left ankle	133.8 ± 8.3	134.6 ± 8.6	0.8	0.8 ± 7.9	0.92 (0.85 – 0.95)	1.9	2.4	6.8
Right knee	169.4 ± 5.0	168.4 ± 5.6	-1.0	-1.0 ± 4.6	0.91 (0.83 – 0.95)	1.0	1.6	4.5
Left knee	158.2 ± 7.4	157.9 ± 7.6	-0.3	-0.3 ± 6.0	0.92 (0.86 – 0.96)	1.4	2.1	5.8
Right hip	161.6 ± 7.0	161.0 ± 7.7	-0.6	-0.6 ± 6.6	0.90 (0.82 – 0.95)	1.5	2.3	6.5
Left hip	153.6 ± 8.7	153.0 ± 9.0	-0.6	-0.6 ± 6.9	0.92 (0.86 – 0.96)	1.7	2.4	6.8
Drop height 100% of maximum CMJ height								
Right ankle	149.3 ± 7.6	148.5 ± 7.5	-0.7	-0.7 ± 5.7	0.93 (0.87 – 0.96)	1.4	2.0	5.6
Left ankle	141.0 ± 6.9	140.7 ± 7.2	-0.4	-0.4 ± 6.9	0.88 (0.78 – 0.93)	1.8	2.4	6.8
Right knee	167.6 ± 4.8	166.1 ± 5.3	-1.6	-1.6 ± 5.1	0.87 (0.77 – 0.93)	1.1	1.6	5.0
Left knee	159.9 ± 5.7	158.9 ± 6.7	-0.9	-0.9 ± 7.8	0.81 (0.66 – 0.89)	1.8	2.7	7.6
Right hip	161.5 ± 6.9	160.2 ± 7.5	-1.3	-1.3 ± 6.0	0.92 (0.85 – 0.95)	1.4	2.1	5.8
Left hip	155.7 ± 8.0	154.4 ± 8.5	-1.2	-1.3 ± 8.4	0.87 (0.77 – 0.93)	2.0	2.9	8.2
Drop height 150% of maximum CMJ height								
Right ankle	149.6 ± 7.0	148.7 ± 7.4	-0.9	-0.9 ± 5.2	0.93 (0.86 – 0.97)	1.3	1.8	5.1
Left ankle	144.6 ± 6.3	143.4 ± 6.9	-1.2	-1.2 ± 5.4	0.92 (0.85 – 0.96)	1.4	1.9	5.3
Right knee	165.4 ± 4.5	164.3 ± 5.1	-1.1	-1.1 ± 4.9	0.87 (0.77 – 0.93)	1.1	1.7	4.8
Left knee	160.9 ± 4.8	159.0 ± 8.2	-1.8	-1.8 ± 6.8	0.80 (0.66 – 0.89)	1.6	2.4	6.6
Right hip	160.4 ± 6.9	159.1 ± 7.1	-1.2	-1.2 ± 6.2	0.90 (0.82 – 0.95)	1.4	2.2	6.0
Left hip	156.3 ± 7.3	155.1 ± 8.2	-1.2	-1.2 ± 7.4	0.89 (0.80 – 0.94)	1.8	2.6	7.2

Table 4.6. Within-session reliability for peak flexion angles for bilateral drop-landing from all drop heights ($n = 39$).

	Test 1 Mean ± SD (°)	Test 2 Mean ± SD (°)	Change in mean (°)	95% LOA (°)	ICC (95% CI)	CV%	SEM (°)	MDC (°)
Drop height 50% of maximum CMJ height								
Right ankle	105.5 ± 9.7	104.7 ± 8.9	-0.7	-0.7 ± 6.7	0.94 (0.88 – 0.97)	2.3	2.3	6.5
Left ankle	102.4 ± 8.4	102.3 ± 7.7	0.0	0.0 ± 5.7	0.94 (0.88 – 0.97)	2.1	2.0	5.6
Right knee	117.6 ± 17.3	117.0 ± 16.7	-0.6	-0.6 ± 11.2	0.95 (0.90 – 0.97)	3.7	3.9	10.9
Left knee	116.8 ± 16.7	116.0 ± 16.1	-0.8	-0.8 ± 11.2	0.94 (0.89 – 0.97)	3.7	3.9	10.9
Right hip	127.1 ± 24.0	126.6 ± 24.6	-0.5	-0.5 ± 18.5	0.93 (0.87 – 0.96)	5.6	6.5	18.0
Left hip	125.4 ± 25.0	124.4 ± 24.9	-1.3	-1.3 ± 19.9	0.92 (0.85 – 0.96)	5.9	7.0	19.4
Drop height 100% of maximum CMJ height								
Right ankle	104.7 ± 9.1	103.5 ± 8.7	-1.2	-1.2 ± 5.5	0.95 (0.91 – 0.97)	1.9	2.0	5.5
Left ankle	100.8 ± 7.6	99.8 ± 7.7	-1.0	-1.0 ± 6.0	0.92 (0.86 – 0.96)	2.2	2.1	5.9
Right knee	107.5 ± 17.6	105.1 ± 16.1	-2.4	-2.4 ± 11.6	0.94 (0.89 – 0.97)	4.5	3.1	10.5
Left knee	106.2 ± 16.9	104.6 ± 16.0	-1.6	-1.6 ± 10.8	0.95 (0.90 – 0.97)	4.1	3.8	11.3
Right hip	114.4 ± 26.6	112.0 ± 25.6	-2.4	-2.4 ± 11.6	0.96 (0.93 – 0.98)	6.0	5.0	13.8
Left hip	112.4 ± 26.6	111.1 ± 25.7	-1.3	-1.3 ± 12.8	0.97 (0.94 – 0.98)	5.5	4.5	12.4
Drop height 150% of maximum CMJ height								
Right ankle	104.6 ± 8.4	103.9 ± 8.9	-0.8	-0.8 ± 7.0	0.92 (0.85 – 0.96)	2.5	2.5	6.8
Left ankle	100.5 ± 7.7	99.7 ± 7.3	-0.7	-0.8 ± 5.5	0.94 (0.88 – 0.97)	2.0	1.9	5.3
Right knee	101.7 ± 14.6	99.4 ± 15.2	-2.4	-2.4 ± 11.1	0.93 (0.87 – 0.96)	4.6	3.9	10.8
Left knee	100.6 ± 14.5	99.0 ± 15.0	-1.7	-2.8 ± 14.5	0.94 (0.89 – 0.97)	4.3	3.6	10.0
Right hip	104.6 ± 26.4	102.1 ± 25.8	-2.6	-2.6 ± 18.8	0.94 (0.88 – 0.97)	7.9	6.6	18.3
Left hip	102.6 ± 26.7	101.6 ± 24.9	-1.0	-1.0 ± 17.6	0.94 (0.89 – 0.97)	7.3	6.2	17.2

Table 4.7. Within-session reliability for sagittal plane joint displacement for bilateral drop-landing from all drop heights ($n = 39$).

	Test 1 Mean ± SD (°)	Test 2 Mean ± SD (°)	Change in mean (°)	95% LOA (°)	ICC (95% CI)	CV%	SEM (°)	MDC (°)
Drop height 50% of maximum CMJ height								
Right ankle	43.1 ± 7.5	42.2 ± 9.1	-1.0	-1.0 ± 11.5	0.76 (0.59 – 0.87)	15.5	4.1	11.3
Left ankle	31.5 ± 7.9	32.3 ± 7.7	0.8	0.8 ± 9.1	0.83 (0.70 – 0.91)	13.2	3.2	8.9
Right knee	51.8 ± 14.2	51.4 ± 14.1	-0.4	-0.4 ± 11.6	0.92 (0.85 – 0.96)	10.0	4.1	11.3
Left knee	41.5 ± 13.7	41.9 ± 13.5	0.5	0.5 ± 11.3	0.91 (0.84 – 0.95)	12.1	4.0	11.0
Right hip	34.4 ± 19.6	34.3 ± 20.1	-0.1	-0.1 ± 15.6	0.92 (0.86 – 0.96)	22.4	5.5	15.2
Left hip	28.2 ± 19.6	28.6 ± 20.2	0.4	0.4 ± 14.5	0.93 (0.88 – 0.96)	27.8	5.0	13.9
Drop height 100% of maximum CMJ height								
Right ankle	44.5 ± 7.1	45.0 ± 6.9	0.5	0.5 ± 7.3	0.86 (0.76 – 0.93)	6.8	2.6	7.1
Left ankle	40.3 ± 5.6	40.9 ± 6.2	0.6	0.6 ± 7.0	0.82 (0.69 – 0.90)	6.4	2.5	6.8
Right knee	60.1 ± 14.9	60.9 ± 13.0	0.9	0.9 ± 10.7	0.93 (0.86 – 0.96)	6.6	3.8	10.5
Left knee	53.7 ± 14.1	54.4 ± 12.8	0.7	0.7 ± 10.4	0.93 (0.86 – 0.96)	7.0	3.7	10.1
Right hip	47.1 ± 22.2	48.2 ± 20.8	1.1	1.1 ± 12.3	0.96 (0.92 – 0.98)	9.6	4.3	11.9
Left hip	43.3 ± 22.5	43.4 ± 20.9	0.1	0.1 ± 11.5	0.97 (0.93 – 0.98)	9.9	4.0	11.2
Drop height 150% of maximum CMJ height								
Right ankle	45.0 ± 6.4	44.9 ± 6.2	-0.1	-0.1 ± 6.1	0.88 (0.79 – 0.94)	5.3	2.2	6.0
Left ankle	44.1 ± 5.5	43.7 ± 4.9	-0.4	-0.5 ± 6.3	0.81 (0.67 – 0.90)	5.5	2.2	6.2
Right knee	63.6 ± 12.5	64.9 ± 12.4	1.3	1.3 ± 10.6	0.91 (0.83 – 0.95)	6.3	3.7	10.4
Left knee	60.2 ± 12.4	60.1 ± 12.5	-0.2	-0.2 ± 10.1	0.92 (0.85 – 0.96)	6.0	3.5	9.8
Right hip	55.7 ± 22.2	57.1 ± 21.6	1.3	1.3 ± 16.9	0.93 (0.86 – 0.96)	11.4	6.0	16.5
Left hip	53.7 ± 22.7	53.5 ± 21.5	-0.2	-0.2 ± 15.2	0.94 (0.86 – 0.96)	10.8	5.3	14.8

Table 4.8. Within-session reliability for FPPA for bilateral drop-landing from all drop heights ($n = 39$).

	Test 1 Mean ± SD (°)	Test 2 Mean ± SD (°)	Change in mean (°)	95% LOA (°)	ICC (95% CI)	CV%	SEM (°)	MDC (°)
Drop height 50% of maximum CMJ height								
Right FPPA	184.4 ± 10.7	184.2 ± 10.8	-0.1	-0.1 ± 7.7	0.94 (0.88 – 0.97)	1.6	2.7	7.5
Left FPPA	184.9 ± 9.3	185.2 ± 9.3	0.3	0.3 ± 6.3	0.94 (0.89 – 0.97)	1.2	2.2	6.1
Drop height 100% of maximum CMJ height								
Right FPPA	186.7 ± 14.0	187.8 ± 13.1	1.1	1.1 ± 9.1	0.94 (0.90 – 0.97)	1.8	3.2	8.9
Left FPPA	186.9 ± 10.5	186.8 ± 11.1	-0.1	-0.1 ± 10.6	0.88 (0.78 – 0.93)	2.0	3.7	10.3
Drop height 150% of maximum CMJ height								
Right FPPA	187.5 ± 14.3	188.3 ± 15.5	0.9	0.9 ± 12.3	0.92 (0.85 – 0.95)	2.3	4.3	12.0
Left FPPA	187.4 ± 12.4	187.9 ± 13.2	0.5	0.5 ± 10.0	0.92 (0.86 – 0.96)	1.9	3.5	9.7

Table 4.9. Within-session reliability for inter-limb differences sagittal plane joint displacement for bilateral drop-landing from all drop heights (*n* = 39).

	Test 1 Mean ± SD (°)	Test 2 Mean ± SD (°)	Change in mean (°)	95% LOA (°)	ICC (95% CI)	SEM (°)	MDC (°)
Drop height 50% of maximum CMJ height							
Ankle displacement	11.7 ± 7.6	9.9 ± 10.1	-1.8	-1.8 ± 13.4	0.72 (0.56 – 0.83)	4.8	13.2
Knee displacement	10.3 ± 6.2	9.5 ± 7.0	-0.9	-0.9 ± 8.8	0.78 (0.65 – 0.86)	3.1	8.7
Hip displacement	6.2 ± 4.2	5.8 ± 5.3	-0.4	-0.4 ± 6.1	0.80 (0.67 – 0.80)	2.1	6.0
Drop height 100% of maximum CMJ height							
Ankle displacement	4.4 ± 7.3	4.1 ± 6.7	-0.1	-0.1 ± 8.8	0.81 (0.69 – 0.88)	3.1	8.6
Knee displacement	6.4 ± 5.9	6.6 ± 6.0	0.2	0.2 ± 8.8	0.73 (0.57 – 0.83)	3.1	8.7
Hip displacement	3.9 ± 4.8	4.9 ± 4.7	1.0	1.0 ± 8.1	0.63 (0.44 – 0.77)	2.9	8.0
Drop height 150% of maximum CMJ height							
Ankle displacement	0.8 ± 6.5	1.2 ± 6.5	0.4	0.4 ± 7.2	0.84 (0.75 – 0.91)	2.7	7.1
Knee displacement	3.4 ± 5.3	4.9 ± 6.0	1.5	1.5 ± 7.2	0.80 (0.67 – 0.88)	2.5	7.1
Hip displacement	2.1 ± 4.8	3.6 ± 4.6	1.5	1.5 ± 7.2	0.50 (0.27 – 0.67)	3.3	9.3

4.4 Discussion

4.4.1 Reliability for kinetic measures associated with bilateral drop landing performance

The first aim of the study reported in this Chapter was to establish the within-session reliability for force-time measures of the bilateral drop-landing from drop heights of 50%, 100% and 150% of maximum CMJ height. The data show that kinetic measures of bilateral drop-landing performance have relative reliability ranging from *large* to *near perfect*, with absolute reliability (represented by CV%) ranging from 6.6–27.6%. Therefore, bilateral drop-landings can be reliably used as a screening tool to determine the management of landing forces, although the error associated with the measure will be strongly influenced by the force-time variable analysed and the magnitude of change being detected (Atkinson and Nevill, 1998).

Importantly, no systematic bias was detected between tests using the within-session design, indicating that no learning effect, participant bias, or acute adaptations were present between tests (Atkinson and Nevill, 1998). These findings suggest that the procedures used in this investigation are appropriate for diminishing the effects of systematic error. Practitioners should be aware of such considerations when designing procedures for testing an athlete's landing capabilities in order to reduce error and allow for better interpretation of their data (Riemann and Lininger, 2018).

James et al. (2007) have reported relative reliability as *very large* for bilateral measures of peak vGRF (ICC = 0.77) and loading rate (ICC = 0.87) for bilateral drop-landings from a 0.61 m drop height. Similarly, using a within-session design, Walsh et al. (2006) reported *near perfect* reliability for peak vGRF (ICC = 0.98) and time to peak vGRF (ICC = 0.92)

following a bilateral drop-landing from a 0.31 m drop height. Collectively, the findings reported in this Chapter support previous investigations, whilst also extending the interpretation of measurement error by quantifying absolute reliability (i.e. agreement) for all variables, across varying drop heights for both unilateral and bilateral measures.

The ICC's for bilateral and unilateral measures of peak vGRF across each drop height ranged from 0.87–0.95, with CV% between 7.1–13.0% (Table 4.1). Although the ICC values suggested peak vGRF during bilateral landings to be arbitrarily reliable, it has been suggested that < 10% for CV% is the acceptable threshold for a test measure to be deemed reliable (Stokes, 1985). This practice for determining absolute reliability would indicate that unilateral measures of peak vGRF during the bilateral drop-landing from heights of 50% and 100% of an individual's CMJ height should be considered to lack the necessary reliability (Table 4.1). Similarly, CV% for time to peak vGRF ranged from 10.5–27.6% for bilateral drop-landings at 50% and 100% of CMJ height, both bilaterally and unilaterally (Table 4.2), resulting in the same arbitrary outcome of unacceptable reliability. However, the use of this arbitrary cut-off point has been contested on the basis that it is not based on a well-defined analytical goal (Atkinson and Nevill, 1998). Therefore, for this study it was purposely decided not to apply an arbitrary 10% threshold for CV% to determine reliability. Instead, practitioners should appreciate that measurements of peak vGRF and time to peak vGRF during bilateral drop-landings, are likely to be more variable at lower drop heights and evaluate this in conjunction with the anticipated or likely signal changes. For example, Vu et al. (2017) showed that firefighters performing bilateral drop-landings from a 0.41 m drop height wearing restrictive firefighting boots were exposed to 10.8% greater peak vGRF bilaterally, when compared to landings in athletic footwear. Based on the data presented in this Chapter, the increase in peak vGRF associated with wearing firefighting boots would be

defined as 'real' from any drop-height between the individuals' 50–150% CMJ height. However, in a study by Milner et al. (2012) investigating the effects of verbal instruction on a bilateral landing task, an instructional cue to land *"with knees over your toes"* led to a 9.0% mean reduction in bilateral peak vGRF across the cohort. Had this landing been performed from a drop height equalling 50% of each individual's maximum CMJ height, this reduction in peak vGRF would reside within the boundaries of measurement error and could not be defined as real change. As changes in landing mechanics have been shown to invoke an increase in peak vGRF of up to 29.6% bilaterally (Zhang, Bates and Dufek, 2000), it can be suggested that CV% reported in this Chapter for peak vGRF may still be low enough to identify changes in an athlete's capacity to successfully attenuate forces across all drop heights. Similarly, differences in time to peak vGRF have been previously shown to differ by approximately 12.3% bilaterally between gymnasts and recreational athletes from a drop landing of 0.30 m (Seegmiller and McCaw, 2003). If this drop height equated to the participants 100% CMJ height, this difference in time to peak vGRF would exceed the CV% of 10.7% established in this Chapter, and therefore present as a meaningful difference between cohorts. Therefore, it is recommended that practitioners consider the measurement error for kinetic measures associated with bilateral landings that is reported in this Chapter when interpreting an individual's competency to attenuate landing forces. This interpretation must be made relative to an individuals's maximum CMJ height, as lower drop heights produce greater variability in measurement error.

Loading rate has been suggested to be an important mechanical variable to consider during landing activities, due to its association with injury risk (Mason-Mackay, Whatman and Reid, 2017). Mean loading rates increased proportionally with drop height, however, the CV% for loading rate observed was among the largest, particularly at lower drop-heights. Yet loading

rate measured bilaterally during drop-landings from 0.61 m has been shown to acutely decrease by 23% following a fatigue protocol (James, Scheuermann and Smith, 2010). Furthermore, significant reductions in ankle plantar flexion angles at initial contact have been shown to increase loading rate bilaterally by 711%, rising from 47.99 N·s^{-1} to 341.16 N·s^{-1} (Rowley and Richards, 2015). When compared to the data reported in this Chapter, such changes would be regarded as meaningful across all drop heights relative to the CV% reported in Table 4.3. With such large changes acutely observed, it is likely that differences in loading rate can be detected, although the magnitude of change will need to be relatively large from lower drop heights.

The change reported herein between drop height and the reliability of landing kinetics supports the findings of recent investigations (Nordin and Dufek, 2017), where the variability (CV%) in lower-limb joint moments were reduced as a function of drop height, which ranged from 20% to 180% of CMJ height. It was suggested that the reduced variability in joint moments observed with increased landing heights indicated a more consistent, yet potentially harmful, reliance on selected joint structures during more demanding tasks, which may increase injury risk (Nordin and Dufek, 2017). The data reported in this Chapter has expanded upon these findings by reporting the reduced variability of kinetic drop-landing profiles at greater drop heights. More specifically, the results indicate that the relative variability for peak vGRF, time to peak vGRF, and loading rate measured both bilaterally and unilaterally, all decreased with greater drop heights.

To support the interpretation of findings from this thesis, the MDC values for all force-time variables were established. These values allow for the practitioner to identify whether an

intervention has resulted in 'meaningful' change (Riemann and Lininger, 2018). An example of this could be a reduction in the peak vGRF an individual is exposed to during bilateral drop-landings. An athlete performing a bilateral drop-landing from a drop height of 50% CMJ height with the bilateral peak vGRF of 2.5 N·kg^{-1} would need to reduce peak vGRF by > 0.78 N·kg^{-1} for the change to be defined as meaningful. Likewise, if the same athlete presented with bilateral peak vGRF of 4.8 N·kg^{-1} from a drop height of 150% CMJ height, a reduction of > 0.75 N·kg^{-1} would be required for the intervention to be deemed successful. These MDC values represent changes in peak vGRF of 31% and 16% from drop heights equating to 50% and 150% of CMJ height, respectively. These examples further illustrate the need to identify drop heights for screening landing mechanics relative to the athlete's CMJ height when interpreting force-time data. However, practitioners should be aware that the use of MDC values to define a change as meaningful for an individual remains somewhat arbitrary and is based on a number of assumptions, such as data being distributed normally (Atkinson and Nevill, 1998).

Asymmetries during athletic activities have been suggested to impair performance outcomes (Bishop, Turner and Read, 2018) and increase injury risk (Pappas and Carpes, 2012; Schot, Bates and Dufek, 1994). This study has shown that a large amount of variability in peak vGRF asymmetry exists during the bilateral drop-landings, with MDC values larger than, or approaching, the mean asymmetry observed in participants across all drop heights (Table 4.4). This is similar to previous findings (Schot, Bates and Dufek, 1994), with the asymmetries in peak vGRF during bilateral landings appearing to vary greatly between tests. Inter-limb asymmetries in force profiles during bilateral landings are particularly important metrics among post-rehabilitation athletes. For example, Paterno et al. (2007) found that a

group of female athletes, who had returned to sport two years after anterior cruciate ligament reconstructive surgery demonstrated side-to-side vGRF asymmetries during a drop vertical jump. These asymmetries were in favour of the uninvolved limb and resulted in a mean difference of 0.5 x bodyweight in peak vGRF, representing a mean asymmetry index score of 14.3% (Paterno et al., 2007). If this magnitude of asymmetry was found during the performance of a bilateral drop-landing task, based on the MDC values presented in Table 4.4, this asymmetry value would not present as meaningful, regardless of drop height. Therefore, when screening for asymmetries during bilateral drop-landings, the findings reported in this Chapter suggest that peak vGRF should be analysed with caution due to the error associated with this outcome variable.

4.4.2 Reliability of kinematic measures associated with bilateral drop landing performance

A second aim of this Chapter was to determine the within-session reliability of kinematic variables using 2D video analysis during bilateral drop-landings from drop heights equating to 50%, 100%, and 150% of an individual's maximum CMJ height. No systematic bias was identified, indicating no evidence of a learning effect, participant bias, or acute adaptations in movement strategies between tests using a within-session design (Atkinson and Nevill, 1998). The *large* to *near perfect* ICC values and CV% ranging between 1.0–27.8% found in this study suggest that kinematic variables can be reliably measured in both the sagittal and frontal plane for bilateral drop-landings from all heights. The findings therefore indicate that 2D video analysis is a reliable method for establishing coordination strategies during bilateral drop-landings, although variability in error will be influenced by the kinematic measurement analysed and the drop height. Previously, 2D video analysis has been validated against 3D motion analysis for both sagittal and frontal plane lower extremity peak joint angles during

landing tasks (Dingenen et al., 2014; Dingenen et al., 2015; Holden et al., 2017; McClean et al., 2005; Mizner et al., 2012; Myer et al., 2011). The findings reported in this Chapter, alongside these previous investigations, mean 2D video analysis can be considered a viable and affordable tool for practitioners to use when assessing injury risk during bilateral drop-landings.

The findings reported in this Chapter have shown that initial contact angles for both limbs can be reliably measured using 2D video analysis, with ICCs ranging from 0.80–0.93 and CV% between 1.0–2.0% across all drop heights (Table 4.5). These findings are similar to the absolute reliability values previously reported for measures of knee and hip joint angles at initial contact during drop jumps using 2D analysis, with SEM values ranging between 1.4–4.1˚ and 1.2–1.3˚, respectively (King and Belyea, 2015). To identify a preferred landing strategy, initial contact angles provide valuable information regarding the athlete's efficiency for attenuating vGRF. Rowley and Richards (2015) showed that when participants consciously increased their ankle plantar flexion angle from 10˚ to 30˚ at initial contact, peak vGRF and loading rates significantly reduced during a bilateral drop-landing from 100% of maximum CMJ height. These findings are supported by that of Kovács et al. (1999), who demonstrated that bilateral landings with reduced ankle plantar flexion at initial contact led to greater force dissipation via the knee and hip joint during the landing phase of a drop jump. Furthermore, following ankle injury, Delahunt et al. (2013) showed that individuals with chronic ankle instability landed with 8.6% less plantar flexion following ankle mobilisation. Based on the absolute reliability values presented in Table 4.5, the data from this study indicates that regardless of drop height, such subtle changes in ankle, knee, and hip joint alignment at initial contact can be detected using 2D video analysis due to the negligible error

of this kinematic measure. Therefore, this test can be used to assess discrete kinematic changes that may influence landing mechanics.

Peak flexion angles for the ankle, knee and hip joints demonstrated *nearly perfect* relative reliability across all drop heights, with ICCs ranging from 0.92–0.97 and CV% between 1.9–7.9% (Table 4.6). Similar to these findings, Beardt et al. (2018) reported ICC values for measuring peak hip and knee flexion angles using 2D analysis during bilateral drop jumps as 0.98 and 0.92, respectively. Likewise, King and Belyea (2015) reported comparable SEM values for peak flexion angles for the knee (SEM = 3.1˚) and hip joint (SEM = 2.4˚) to that of this investigation. Individuals with limited sagittal plane flexion strategies throughout the lower extremity have been suggested to lack the necessary shock absorption to safely attenuate vGRF during landing tasks (Blackburn and Padua, 2009; Sigward, Pollard and Powers, 2012; Zhang, Bates and Dufek, 2000). Zhang, Bates and Dufek (2000) showed that a 5.9°, 22.1° and 25.4° reduction in peak ankle dorsiflexion, knee and hip flexion angles, respectively, between normal and 'stiff' landings, resulted in significantly greater peak vGRF during bilateral drop-landings from drop heights of 0.62 m. With greater peak vGRF during landing tasks potentially increasing lower extremity injury risk (Hewett et al., 2005; Zadpoor and Nikooyan, 2011), individuals using a stiff landing strategy should be identified and an intervention provided to attenuate injury risk (Lopes et al., 2018). Based on CV% presented in Table 4.6, the findings from this study indicate that changes in landing strategies for peak angles of ankle dorsiflexion, knee and hip flexion such as that shown by Zhang, Bates and Dufek (2000), may be reliably identified using 2D video analysis. The findings presented in this Chapter provide clinicians with practically relevant information that may guide the interpretation of bilateral landing tasks, with margins for error in the test measures presented (Riemann and Lininger, 2018).

Sagittal plane joint displacement provides a general overview of the contribution from each joint towards force attenuation during landing tasks (Decker et al., 2003). The results presented in this Chapter indicate that measurements of joint displacement are reliable to detect differences between- and within-participants in joint contribution from drop heights of 100% and 150% of maximum CMJ height, with ICCs ranging from 0.81–0.97 and CV% between 5.3–11.4%. For example, when investigating gender differences in joint displacement angles during bilateral drop-landings from a 0.60 m drop height, mean differences between male and female participants for the ankle, knee and hip joints were 28.3%, 16.4% and 13.0%, respectively (Decker et al., 2003). Similarly, with the application of a prophylactic ankle brace to provide external support, Cordova et al., (2010) found ankle joint displacement reduced by 19.5% during a drop-landing task. Based on the absolute reliability established in this study (Table 4.7), such differences can be detected using 2D video analysis from drop heights equating to 100% and 150% of an individual's maximum CMJ height. However, absolute reliability for joint displacement angles at the ankle, knee and hip were much greater from drop heights of 50% of maximum CMJ height, with CV% ranging between 10.0–27.8%. It is possible that at lower drop heights, the lower mechanical demand, and thus relative ease of the task, increases degrees of movement freedom for participants, facilitating greater variability in joint displacement angles for all segments (Nordin and Dufek, 2017). The findings reported in this Chapter suggest that greater change is required for joint displacement angles at the ankle, knee, and hip following an intervention when lower relative drop heights are used for assessing differences in coordination strategies during bilateral drop-landings.

As a result of an individual displaying limited sagittal plane contribution to attenuating load, compensation may occur through excessive frontal and/or transverse plane lower extremity motion to lower their centre of mass for energy dissipation (Sigward, Pollard and Powers, 2012). The development of compensation strategies most likely results in greater external knee valgus or varus moments occurring (Kernozek et al., 2005). External knee valgus moments and peak angles have been shown to identify athletes at greater risk for anterior cruciate ligament injury (Hewett et al., 2005). With peak FPPA measured using 2D video analysis during landing tasks correlating with 3D measures of peak knee abduction angle and knee abduction moment (Mizner et al., 2012), the findings from this study indicate that individuals at greater risk of anterior cruciate ligament injury may be reliably identified during bilateral drop-landings across various drop heights. For this investigation, CV% and SEM for FPPA across all drop heights ranged from 1.2–2.3% and 2.7–4.3˚, respectively (Table 4.8). These results are similar to the SEM values reported by Munro et al. (2012) for FPPA during single-leg drop-landings (SEM = 2.7–2.9˚) and bilateral drop jumps (SEM = 3.0˚) performed from a 0.28 m drop height. Therefore, using 2D video analysis for identifying peak FPPA is a reliable means for assessing frontal-plane lower extremity kinematics during bilateral drop-landings from heights ranging between 50–150% of maximum CMJ height.

Inter-limb differences in coordination strategies during bilateral drop-landing have been identified in healthy (Harry et al., 2017; Niu et al., 2011; Pappas and Carpes, 2012) and previously injured populations (Meyer et al., 2018). The relative reliability for inter-limb asymmetries in sagittal plane joint displacements in this study was determined as *large* to *very large* (ICC = 0.50–0.84). However, the absolute reliability values observed in this study indicated this measurement to be highly variable. For example, the MDC values for inter-

limb asymmetries in ankle, knee and hip joint displacement across each drop height ranged from 7.1–13.2°, 7.1–8.7° and 6.0–9.3°, respectively (Table 4.9). Pappas and Carpes (2012) investigated gender differences for inter-limb joint kinematics during bilateral drop-landings from a 0.40 m drop height in healthy recreational athletes. Inter-limb differences for sagittal plane joint displacement at the ankle (male = 3.4°, females = 3.8°), knee (male = 3.6°, females = 3.8°) and hip joints (male = 5.6°, females = 5.6°) would not exceed the MDC values presented in this Chapter. This is similar for inter-limb differences observed in injured populations. Using 3D analysis, Meyer et al. (2018) examined side-to-side differences during the landing phase of a bilateral drop vertical jump from a 0.40 m drop height in 17 patients who had undergone unilateral anterior cruciate ligament reconstructive surgery. For sagittal plane knee joint displacement, a 2.5° difference was found between the involved and uninvolved limb. Based on the findings reported in this Chapter, it is likely that this difference would not be detectable using 2D video analysis, irrespective of drop height. On the basis of the findings from this Chapter, it is suggested that measurements of inter-limb differences in sagittal plane joint displacement during bilateral drop-landings cannot be used to detect smaller, yet clinically meaningful, changes.

4.5 Conclusion

During bilateral landings, individuals are exposed to very high loads with the potential for injury. As such, it is imperative that tests to identify coordination strategies during bilateral landings are reliable across a range of loads. With portable force platforms and 2D video analysis equipment being affordable and accessible to practitioners, the reliability of kinetic and kinematic variables related to landing performance has been presented in this Chapter. This investigation showed that peak vGRF, time to peak vGRF and loading rate possessed

relative reliability values ranging from *large* to *near perfect*. Similarly, this Chapter has demonstrated that the use of 2D video analysis is a reliable tool for measuring kinematic variables associated with lower extremity joint angles at initial contact and peak flexion in both limbs during bilateral drop-landings. However, the signal to noise values suggest that drop height will likely influence the variability observed in kinetic and kinematic measures of bilateral landing performance. As such, kinetic and kinematic data for landing mechanics from drop heights equating to 100% and 150% of maximum CMJ are associated with reduced measurement error. Additionally, the error reported in this Chapter for inter-limb asymmetries in peak vGRF and sagittal plane joint displacement during bilateral drop-landings likely hinders detection of clinically relevant changes following interventions. Therefore, the results of subsequent studies presented within this thesis will be interpreted based on the findings presented in this Chapter.

Chapter 5

Ankle dorsiflexion range of motion is associated with kinematic but not kinetic variables related to bilateral drop-landing performance at various drop heights

A version of this Chapter has been published in Human Movement Science and the printed version can be found in Appendix 10.

5.1 Introduction

Bilateral landings are performed by athletes in training and competition (Bloomfield, Polman and O'Donoghue, 2007; McClay et al., 1994) and are also part of daily life during leisure activities (Maté-Muñoz et al., 2017) and occupational tasks (Knapik et al., 2003). To manage the large vertical forces that can equate to multiples of body weight, the ankle, knee and hip joints must be coordinated to provide a movement strategy that facilitates effective energy dissipation (Yeow, Lee and Goh, 2011). In athletic populations, the forces experienced during landings have been identified as a cause for acute (Hewett, Myer and Ford, 2006) and chronic (Dierks et al., 2011) lower extremity injuries. When suboptimal coordination strategies are adopted during landing tasks, higher peak vGRF (Fong et al., 2011) and greater risk of injury may occur (Herrington, 2014; Hewett et al., 2005). For example, sagittal plane initial contact angles (Rowley and Richards, 2015), peak flexion angles (Blackburn and Padua, 2009; Yu, Lin and Garrett, 2006) and sagittal plane joint displacement (Begalle et al., 2015; Podraza and White, 2010) at the ankle, knee and hip joints have all been associated with greater peak vGRF. Likewise, in the frontal and transverse plane, greater peak knee valgus angle during landing tasks have been found to increase injury risk (Hewett et al., 2005).

One modifiable factor able to cause suboptimal landing mechanics is restriction in ankle DF ROM, which is inversely related to peak vGRF during bilateral jump-landings (Fong et al., 2011). The relationship between ankle DF ROM and peak vGRF is likely to be the result of limitations in ankle DF ROM inhibiting knee flexion joint displacement during the shock absorption phase of landing (Fong et al., 2011). This results in a stiffer landing strategy known to increase peak vGRF (Zhang, Bates and Dufek, 2000) and undesirable load being placed on passive structures of the knee (Yu and Garrett, 2007). Furthermore, restrictions in ankle DF ROM have been shown to correspond with frontal plane kinematic compensations throughout the lower extremity during both unilateral (Whitting et al., 2011) and bilateral landings (Malloy et al., 2015; Sigward, Ota and Power, 2008). For example, Malloy et al. (2015) observed that soccer players with restricted ankle DF ROM performed a bilateral landing task with greater peak knee abduction angles. Given that increased peak knee abduction angle during landings has been highlighted as a risk factor for anterior cruciate ligament injury (Hewett et al., 2005) and patellofemoral pain syndrome (Holden et al., 2017), reduced ankle DF ROM can be an important risk factor for injury. However, there is little evidence of other compensatory strategies that may be adopted to manage vGRF when ankle DF ROM is limited, such as increasing ankle plantar flexion at initial contact (Rowley and Richards, 2015) and peak hip flexion angle (Blackburn and Padua, 2009) during bilateral landings.

Although ankle DF ROM has been shown to influence coordination strategies during bilateral landings (Fong et al., 2011; Malloy et al., 2015), little data exists regarding the functional consequences of inter-limb asymmetries in ankle DF ROM during such tasks. With inter-limb asymmetries in ankle DF ROM being present among uninjured populations (Hoch and McKeon, 2011; Rabin and Kozol, 2015), it may be the presence of a unilateral restriction in

ankle DF ROM causes asymmetries in landing mechanics for healthy individuals. During all movement phases of bilateral squatting, unilateral restrictions in ankle DF ROM have been shown to decrease force production in the hypomobile limb (Crowe et al., 2019). As the bilateral squat possesses similar mechanical features to bilateral landings, potential exists for inter-limb asymmetries in ankle DF ROM to produce inter-limb asymmetries in peak vGRF and lower extremity kinematics. However, the relationship between inter-limb asymmetries in ankle DF ROM and bilateral landing performance is yet to be investigated.

Many jumping activities involve landing from a variety of heights that are significantly lower or higher than an individual's standard CMJ height, such as when eliminating the countermovement phase (i.e. squat jump) (Young, Wilson and Byrne, 1999), jumping with an arm swing (Slinde et al., 2008) or performing a run-up immediately prior to the jump (Young, Wilson and Byrne, 1999). As greater initial contact velocities produce elevated peak vGRF and directly influences the coordination strategies adopted (Zhang, Bates and Dufek, 2000), research is required to determine how restrictions in ankle DF ROM alter the movement demands of these tasks at varying drop heights. With peak knee flexion angle increasing when landing from a higher drop height (Zhang, Bates and Dufek, 2000), it may be that restricted ankle DF ROM has an increased effect on landing mechanics when performing landings from greater elevations. Therefore, the aim of this investigation was to determine the relationship between ankle DF ROM and both kinetic and kinematic variables measured during bilateral drop-landings from a range of heights individualised to CMJ performance. A further aim was to establish the relationship between inter-limb differences in ankle DF ROM and asymmetries in landing mechanics. It was hypothesised that: i) reduced ankle DF ROM would correlate with greater peak vGRF caused by reduced ankle dorsiflexion and knee flexion being available for energy absorption, ii) restricted ankle DF ROM would cause

compensations in coordination strategies at other time points (i.e. initial contact) and separate joint segments (i.e. the hip), iii) inter-limb differences in ankle DF ROM would correlate with inter-limb asymmetries in kinetic and kinematic variables associated with bilateral drop-landing performance, iv) landings from higher drop heights would strengthen the relationship between ankle DF ROM and the compensatory strategies in coordination patterns.

5.2 Methods

5.2.1 Study design

Using a cross-sectional design, participants reported for a single test session wearing spandex shorts and vest to evaluate the relationship between ankle DF ROM and the performance of bilateral drop-landings from drop heights of 50%, 100% and 150% of maximum CMJ height. All test sessions were conducted between 10:00 am and 1:00 pm to control for circadian variation.

5.2.2 Participants

Using the findings of Fong et al. (2011), a representative analysis was performed to determine the appropriate sample size based on measures of ankle DF ROM and its relationship with peak vGRF (r = -0.41). Calculations indicated that to achieve 80% statistical power, a minimum of 32 participants was required to detect a significant ($P < 0.05$) correlation between ankle DF ROM and peak vGRF. Thirty-nine recreational athletes (22 men, 17 women, age = 22 ± 4 years, height = 1.74 ± 0.15 m, body mass 70.2 ± 15.1 kg) volunteered to participate in this study. Recreational athletes were defined as a person who regularly competed 1–3 times per week in sport events involving landings activities, such as

court, racquet or team sports (Chappell et al., 2002). Any participant with a history of lower extremity surgery or who had lower extremity injury six-months prior to testing was excluded. All participants were informed of the risks associated with the testing, prior to completing a pre-exercise questionnaire and providing informed written consent. Ethical approval was provided by the University of Cumbria Research Ethics Panel (Appendix 1).

5.2.3 Procedures

Following the recording of height and body mass, ankle DF ROM was measured for both the right and left limb in barefoot using the WBLT (see *Chapter 3*). This procedure was repeated three times, with the mean value for the right limb from the three attempts used for data analysis. For asymmetries in ankle DF ROM, between-limb difference was calculated as the mean WBLT for the right limb minus the mean value for the left limb. A positive value indicated the right limb had the largest score and *vice versa* for a negative asymmetry value.

Following a standardised warm-up, participants were familiarised with the CMJ (see *Chapter 4*) to determine drop height for the performance of bilateral drop-landings. For data collection, three maximal effort CMJs were performed, with 60 s recovery between attempts. The maximum value of the three attempts was then used to calculate box height for the bilateral drop-landings.

Following the performance of the CMJ, reflective markers were placed on each participant by the same investigator (see *Chapter 4*). Participants were then familiarised with the bilateral drop-landings from drop heights of 50%, 100% and 150% of their maximum CMJ height.

Bilateral drop-landings were performed as described in *Chapter 4*. For each drop height, participants performed five landings for data collection, with 60 s recovery provided between landings. Participants completed each block of five bilateral drop-landings from the same drop height in succession, with drop height order randomised using a counterbalanced design. For 2D video analysis, lower extremity sagittal and frontal plane joint movements were recorded using the same methods as detailed in *Chapter 4* of this thesis.

5.2.6 Data analysis

Raw vGRF data for both legs were low-pass filtered using a fourth-order Butterworth filter with a cut-off frequency of 50 Hz (Roewer et al., 2014). Peak vGRF, time to peak vGRF and loading rate was then calculated. Peak vGRF data were normalised to body mass and initial contact velocity ($N \cdot kg^{-1} \cdot m \cdot s^{-1}$). To normalise peak vGRF to drop height, initial contact velocity was calculated using the following equation (Niu et al., 2014):

$$\text{Initial contact velocity } (m \cdot s^{-1}) = \sqrt{2g \cdot DH}$$

where g is the gravitational acceleration and *DH* is drop height. Time to peak vGRF, loading rate and asymmetries in peak vGRF was calculated as described *Chapter 4*. For asymmetries in peak vGRF, a positive value indicated the right limb had the largest score and *vice versa* for a negative asymmetry value.

All video recordings were analysed using the methods described in *Chapter 4*. For hip flexion, knee flexion and ankle dorsiflexion, smaller values represented greater flexion and

ankle dorsiflexion. For FPPA, values < 180° represented knee valgus and values > 180° represented knee varus. Inter-limb asymmetries for sagittal plane joint displacement were calculated by subtracting the left value from the right value for the ankle, knee and hip joints. A positive value indicated the right limb had greater joint displacement for the corresponding segment and *vice versa* for a negative value.

5.2.7 Statistical analysis

Descriptive statistics (means ± standard deviation) were calculated for all dependent variables. The assumption of normality was checked using the Shapiro-Wilk test and confirmed. Pearson bivariate correlation analysis were used to examine the relationship between right ankle DF ROM and kinetic and kinematic dependant variables associated with bilateral drop-landing performance for the right limb for each drop height. Correlations for between-limb differences in ankle DF ROM with inter-limb asymmetries in peak vGRF and sagittal plane joint displacement were also examined. Pearson bivariate correlations were interpreted as *trivial* (0.0–0.1), *small* (0.1–0.3), *moderate* (0.3–0.5), *large* (0.5–0.7), *very large* (0.7–0.9), *nearly perfect* (0.9–1.0) and *perfect* (1.0) (Hopkins, 2016). 95% confidence intervals were calculated for all bivariate correlations to determine the influence of drop height on the relationship between ankle DF ROM and landing mechanics. The *α-priori* level of significance was set at $P < 0.05$. All statistical tests were performed using SPSS® statistical software package (v.24; SPSS Inc., Chicago, IL, USA).

5.3 Results

Mean ankle DF ROM for the right WBLT was 36.3 ± 3.9°, whilst mean inter-limb differences for the WBLT were -0.9 ± 3.0°. Descriptive statistics for dependent variables associated with bilateral drop-landing performance from each drop height, along with correlation coefficients and probability statistics are presented in Table 5.1, 5.2 and 5.3, respectively. Normalised peak vGRF, time to peak vGRF and loading rate for all drop heights was not related to ankle DF ROM, with values ranging from *trivial* to *small* (Table 5.1, 5.2 and 5.3).

From a drop height of 50% (0.15 ± 0.04 m) of maximum CMJ height, a significant negative *moderate* relationship was found between ankle DF ROM and peak knee flexion angle. Significant positive *moderate* relationships were found for FPPA and sagittal plane knee joint displacement (Table 5.1). From drop heights of 100% (0.30 ± 0.08 m) and 150% (0.44 ± 0.12 m) of maximum CMJ height, ankle DF ROM was negatively related (*moderate* to *large*) to knee flexion angle at initial contact, peak ankle dorsiflexion and peak knee flexion angle. Positive *moderate* associations were found for FPPA and sagittal plane knee joint displacement (Table 5.2 and 5.3). Ankle DF ROM showed a negative *moderate* relationship to initial contact angles at the ankle at 100% of maximum CMJ height (Table 5.2). 95% confidence intervals for all bivariate correlations demonstrated overlap across all drop heights. All other relationships were not significant. Relationships for inter-limb differences in ankle DF ROM and inter-limb asymmetries in peak vGRF and sagittal plane joint displacement were non-significant (Figure 5.1, 5.2 and 5.3). Correlations between asymmetry variables ranged from *trivial* to *moderate*.

Table 5.1. Descriptive and correlational statistics for the relationship between ankle DF ROM and kinetic and kinematic variables from drop heights of 50% of maximum CMJ height.

Variable	Mean ± SD	*r*	95% Confidence intervals	*P* value
Peak vGRF, $N \cdot kg^{-1} \cdot m \cdot s^{-1}$	1.06 ± 0.39	-0.28	-0.55, 0.04	0.08
Time to peak vGRF, s	0.077 ± 0.022	-0.12	-0.42, 0.20	0.47
Loading rate, $N \cdot s^{-1}$	28.1 ± 18.01	0.01	-0.31, 0.32	0.95
Initial contact angle, °				
Ankle	148.6 ± 6.9	-0.18	-0.47, 0.14	0.28
Knee	169.4 ± 5.0	-0.15	-0.44, 0.17	0.37
Hip	161.6 ± 7.0	-0.06	-0.37, 0.26	0.73
Peak angle, °				
Ankle	105.5 ± 9.7	-0.27	-0.54, 0.05	0.10
Knee	117.6 ± 17.3	-0.37*	-0.61, -0.06	0.02
Hip	127.1 ± 24.0	-0.23	-0.51, 0.09	0.16
Frontal plane projection	184.4 ± 10.7	0.40*	0.10, 0.64	0.01
Sagittal plane joint displacement, °				
Ankle	43.1 ± 7.5	0.18	-0.14, 0.47	0.26
Knee	51.8 ± 14.2	0.39*	0.08, 0.63	0.01
Hip	34.4 ± 19.6	0.26	-0.06, 0.53	0.11

* Significant correlation between ankle dorsiflexion range of motion and variable.

Table 5.2. Descriptive and correlational statistics for the relationship between ankle DF ROM and kinetic and kinematic variables from drop heights of 100% of maximum CMJ height.

Variable	Mean ± SD	r	95% Confidence intervals	P value
Peak vGRF, $N \cdot kg^{-1} \cdot m \cdot s^{-1}$	0.85 ± 0.30	-0.15	-0.44, 0.17	0.36
Time to peak vGRF, s	0.065 ± 0.021	-0.18	-0.47, 0.14	0.27
Loading rate, $N \cdot s^{-1}$	38.0 ± 24.0	0.10	-0.22, 0.40	0.55
Initial contact angle, °				
Ankle	149.3 ± 7.6	-0.34*	-0.59, -0.03	0.03
Knee	167.6 ± 4.8	-0.37*	-0.61, -0.06	0.02
Hip	161.5 ± 6.9	-0.07	-0.38, 0.25	0.69
Peak angle, °				
Ankle	104.7 ± 9.1	-0.44*	-0.66, -0.14	0.01
Knee	107.5 ± 17.6	-0.42*	-0.65, -0.12	0.01
Hip	114.4 ± 26.6	-0.26	-0.53, 0.06	0.10
Frontal plane projection	186.7 ± 14.0	0.37*	0.06, 0.61	0.02
Sagittal plane joint displacement, °				
Ankle	44.5 ± 7.1	0.19	-0.13, 0.48	0.24
Knee	60.1 ± 14.9	0.39*	0.08, 0.63	0.02
Hip	47.1 ± 22.2	0.30	-0.02, 0.56	0.07

* Significant correlation between ankle dorsiflexion range of motion and variable.

Table 5.3. Descriptive and correlational statistics for the relationship between ankle DF ROM and kinetic and kinematic variables from drop heights of 150% of maximum CMJ height.

Variable	Mean ± SD	r	95% Confidence intervals	P value
Peak vGRF, N·kg^{-1}· m·s^{-1}	0.83 ± 0.24	-0.11	-0.41, 0.21	0.53
Time to peak vGRF, s	0.053 ± 0.012	-0.21	-0.49, 0.11	0.19
Loading rate, N·s^{-1}	52.0 ± 27.4	0.15	-0.17, 0.44	0.36
Initial contact angle, °				
Ankle	149.6 ± 7.0	-0.31	-0.57, 0.01	0.06
Knee	165.6 ± 4.5	-0.40*	-0.64, -0.10	0.01
Hip	160.4 ± 6.9	-0.07	-0.38, 0.25	0.67
Peak angle, °				
Ankle	104.6 ± 8.4	-0.43*	-0.66, -0.13	0.01
Knee	101.7 ± 14.6	-0.52*	-0.72, -0.24	0.001
Hip	104.6 ± 26.4	-0.28	-0.55, 0.04	0.08
Frontal plane projection	187.5 ± 14.3	0.37*	0.06, 0.61	0.02
Sagittal plane joint displacement, °				
Ankle	45.0 ± 6.4	0.22	-0.10, 0.50	0.17
Knee	63.6 ± 12.5	0.47*	0.18, 0.68	0.003
Hip	55.7 ± 22.2	0.32	0.00, 0.58	0.05

* Significant correlation between ankle dorsiflexion range of motion and variable.

Figure 5.1. Relationships between inter-limb differences in ankle DF ROM and inter-limb asymmetries in A) peak vGRF, B) sagittal plane ankle joint displacement, C) sagittal plane knee joint displacement and D) sagittal plane hip joint displacement from drop heights equating to 50% on CMJ height.
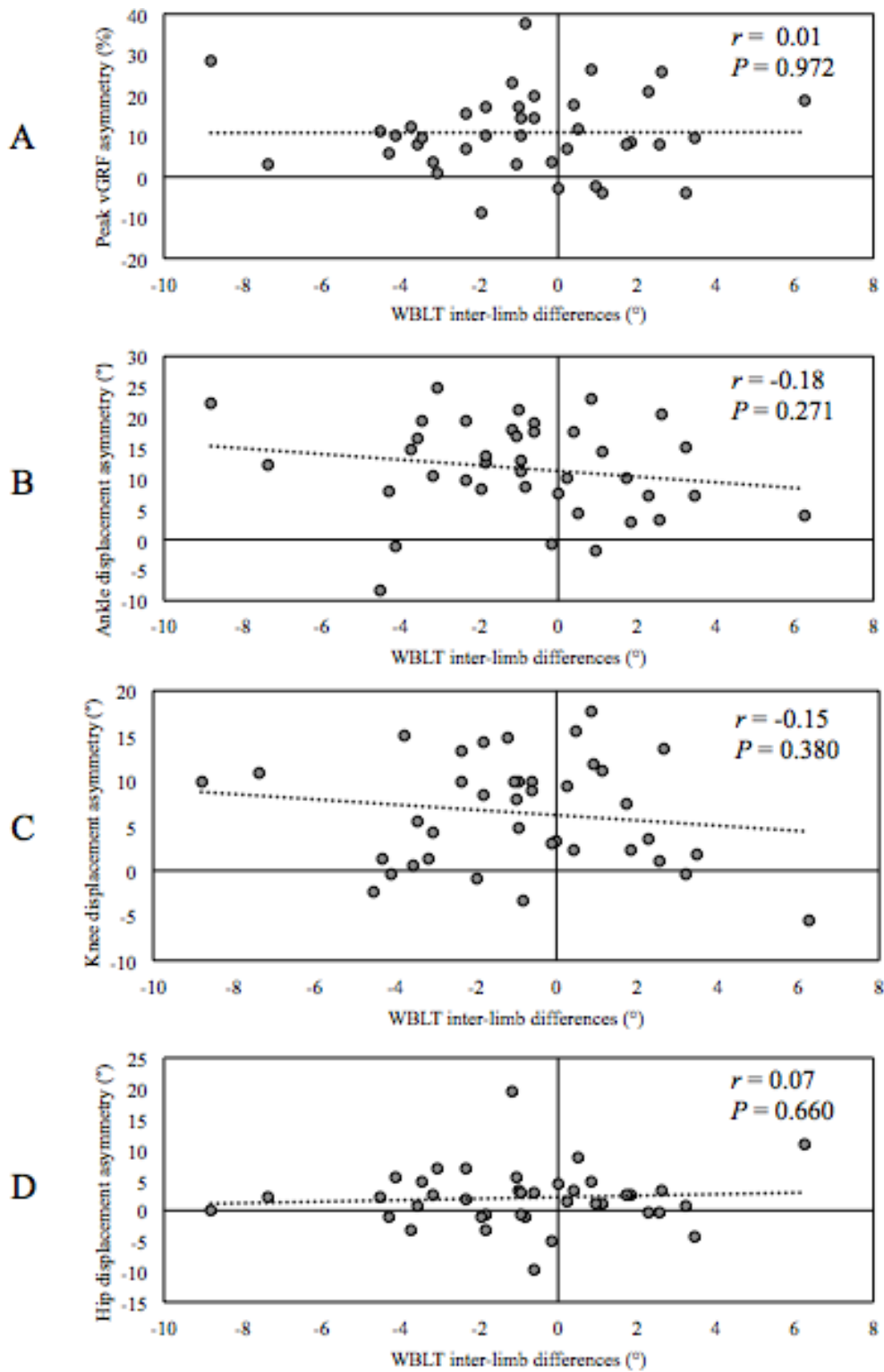
Figure 5.2 Relationships between inter-limb differences in ankle DF ROM and inter-limb asymmetries in A) peak vGRF, B) sagittal plane ankle joint displacement, C) sagittal plane knee joint displacement and D) sagittal plane hip joint displacement from drop heights equating to 100% on CMJ height.
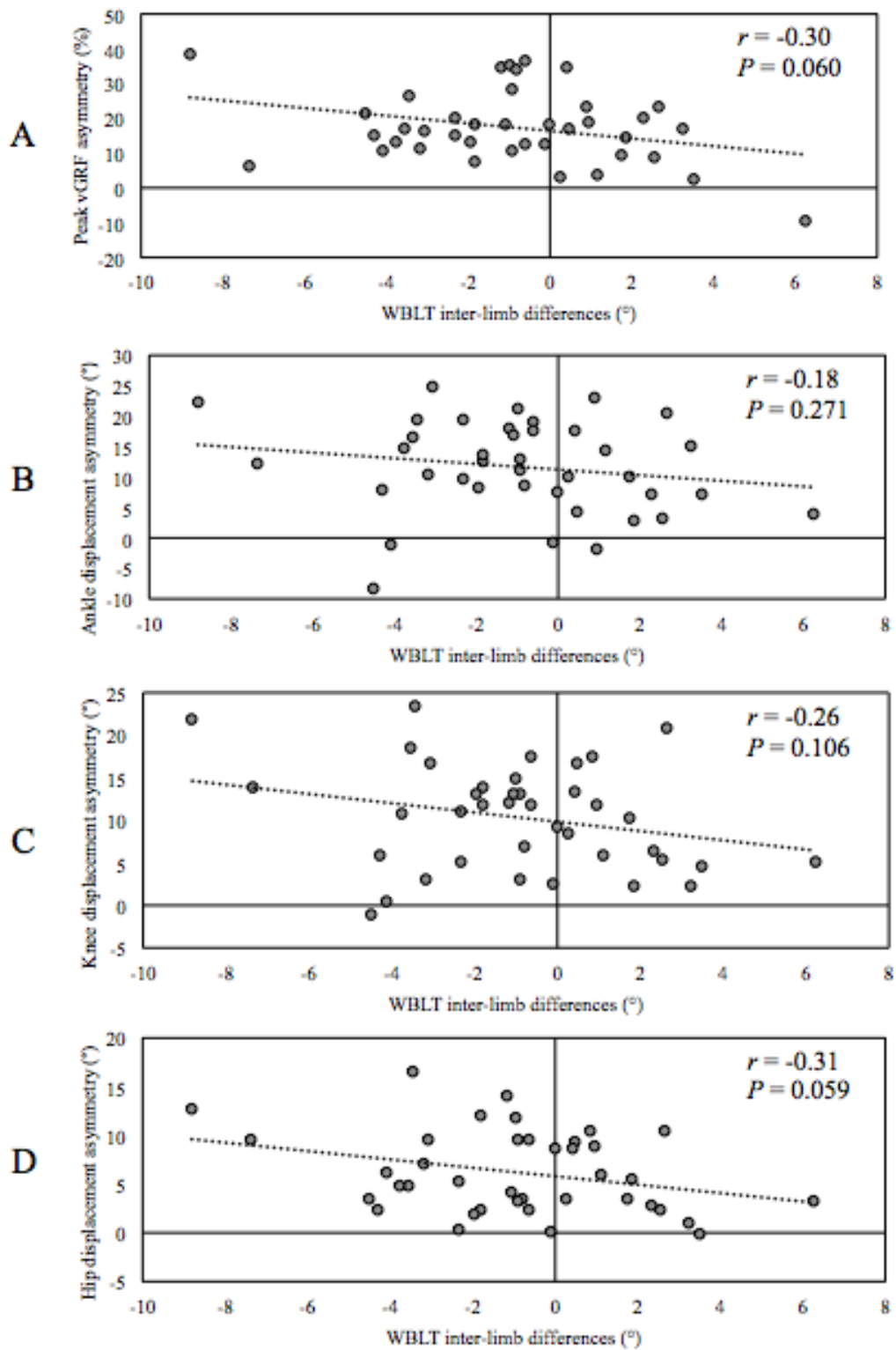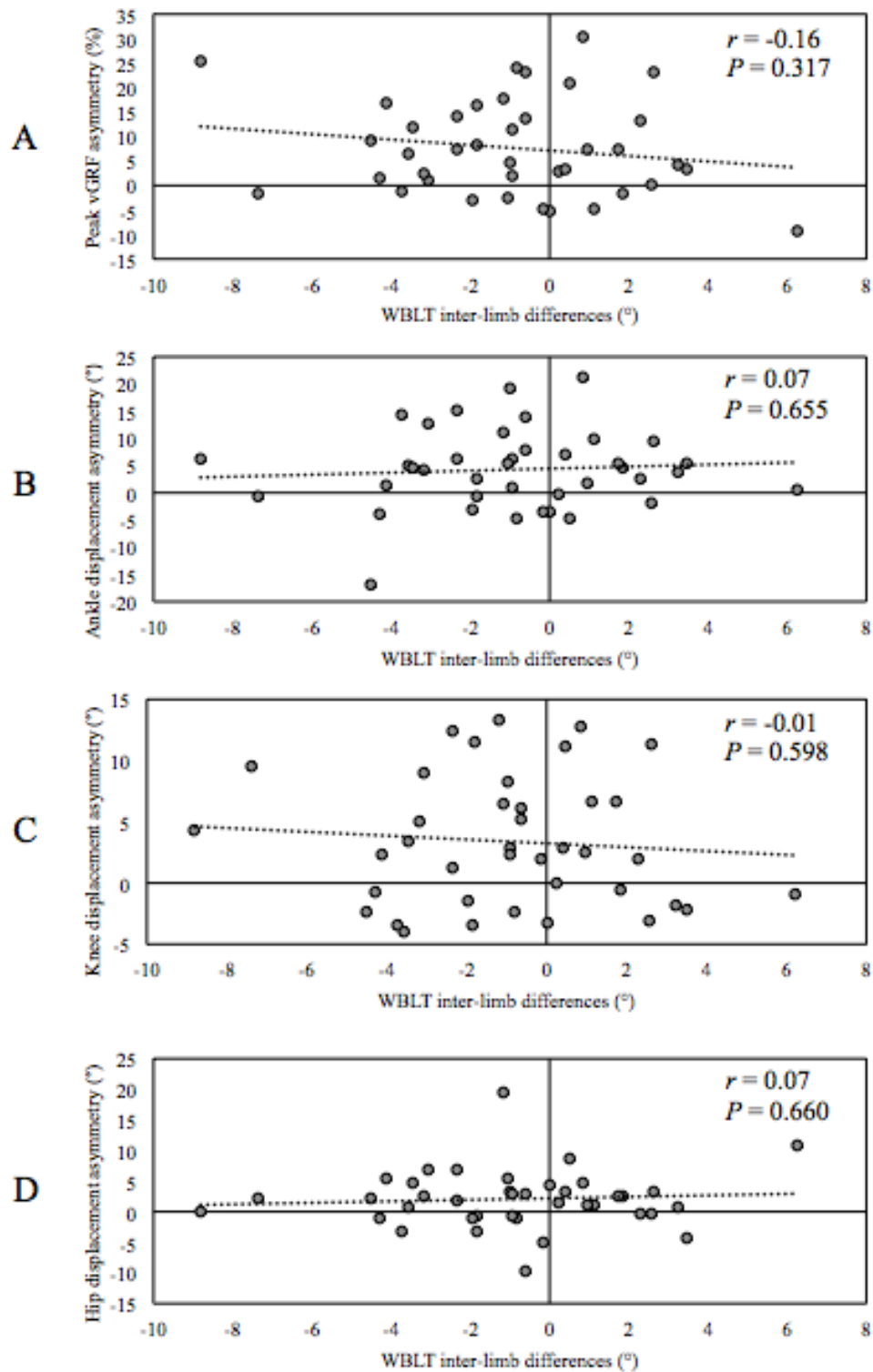
Figure 5.3 Relationships between inter-limb differences in ankle DF ROM and inter-limb asymmetries in A) peak vGRF, B) sagittal plane ankle joint displacement, C) sagittal plane knee joint displacement and D) sagittal plane hip joint displacement from drop heights equating to 150% on CMJ height.

5.4 Discussion

The primary aim of this study was to evaluate the relationship between ankle DF ROM and bilateral drop-landing performance from different drop heights. It was hypothesised that restricted ankle DF ROM would result in greater peak vGRF and altered coordination strategies. However, this hypothesis was only partially supported, as relationships between ankle DF ROM and kinematic variables were found during bilateral drop-landings, without changes in kinetic variables linked to peak vGRF across all drop heights. Ankle DF ROM was mostly *moderately* related to a number of kinematic variables at the ankle and knee joints, indicating a large amount of unexplained variance in the relationship between ankle DF ROM and kinematic variables associated with landing performance. In addition, the relationship between ankle DF ROM and some kinematic variables was only apparent at drop heights of 100% and 150% of CMJ height, indicating that greater mechanical loads may exaggerate the demands for compensatory strategies in coordination during landings. However, there was no association between ankle DF ROM and hip joint kinematics during landings. Therefore, ankle DF ROM is related only to kinematic variables of the ankle and knee joints during drop-landings, with some relationships becoming significant only at higher drop-landing heights.

The main finding of this investigation was that ankle DF ROM did not correlate with peak vGRF, time to peak vGRF or loading rate during landings for all drop heights. In some previous studies, negative relationships between ankle DF ROM and peak vGRF in both healthy (Fong et al., 2011) and previously injured (Hoch et al., 2015) participants have been found during landing tasks. However, consistent with the results reported in this Chapter, Whitting et al. (2011) and Malloy et al. (2015) have reported no relationship between ankle

DF ROM and peak vGRF during landing tasks. One possible reason may be the different compensatory movement patterns observed between studies. For example, participants with limited ankle DF ROM have been shown to compensate in the frontal plane, with increased peak rearfoot eversion (Whittling et al., 2013) and knee abduction angles (Malloy et al., 2015). However, no such relationships were reported by Fong et al. (2011), with knee valgus displacement during a bilateral jump-landing task sharing a non-significant *small* relationship with ankle DF ROM ($r = -0.29$, $P = 0.091$). It has been suggested that during landing tasks, frontal and transverse plane compensations in the lower extremity caused by restrictions in ankle DF ROM may enable individuals to access a movement strategy that allows for the continued lowering of the centre of mass to attenuate peak vGRF and maintain landing forces at a tolerable level (Mason-Mackay, Whatman and Reid, 2017). As participants with restricted ankle DF ROM in Fong et al. (2011) did not demonstrate frontal plane compensations in lower extremity landing strategy, peak vGRF was higher for individuals with ankle hypomobility. The disadvantage of a compensation strategy relying on greater frontal plane compensations would be greater injury risk as the potential rises for excessive loading on the passive structures supporting the knee joint as valgus alignment increases (Yu and Garrett, 2007). Thus, in this Chapter, the *trivial* to *small* relationships found between vGRF and ankle DF ROM are likely to be explained by an altered kinematic profile during landing.

It was also hypothesised that restrictions in ankle DF ROM would correlate with reduced hip joint displacement, decreasing the hip joint's capacity to support the attenuation of vertical forces during bilateral drop-landings. The findings from this Chapter reject this hypothesis, although a statistical trend was present. The hip joint has been shown to possess an important role in the dissipation of vGRF during landings, with hip flexion sharing an exponential

relationship with knee flexion during bilateral landings (Yeow, Lee and Goh, 2011). Additionally, when individuals adopt stiffer landing strategies during bilateral drop-landings, decreased peak flexion angles at both the knee and hip joint result in significant elevations in peak vGRF (Zhang, Bates and Dufek, 2000). Collectively, this evidence indicates that the hip joint supports the knee joint in the dissipation of vGRF during bilateral landing tasks. The findings from this Chapter suggest restrictions in ankle DF ROM diminishes knee joint contribution to lowering the centre of mass during landings, with ankle DF ROM significantly correlated with peak knee flexion angle and sagittal plane knee joint displacement at all drop heights (Table 5.1, 5.2 and 5.3). With no significant relationships found between ankle DF ROM and peak vGRF, it may be expected that the hip joint would increase its contribution through greater sagittal plane joint displacement to offset reduced knee joint involvement. If this were the case, then one would expect a negative relationship to have been present between ankle DF ROM and sagittal plane hip joint displacement in this study. However, although these correlations only demonstrated a trend towards significance (P values ranging between 0.05 to 0.07), the *moderate* relationships between ankle DF ROM and sagittal plane hip joint displacement from drop heights equating to $\geq$ 100% of CMJ height were positive (r = 0.30–0.32). These findings indicate that individuals with restricted ankle DF ROM may land with reduced sagittal plane hip joint displacement, though this suggestion should be interpreted with caution due to the lack of statistical significance. Similar to the magnitude of relationships presented in this Chapter, Dowling, McPherson and Paci (2018) found a *moderate* positive relationship between WBLT performance and sagittal plane hip joint displacement during single-leg drop jumps among 73 male recreational athletes (r = 0.30). Furthermore, Leporace et al. (2018) reported no relationship between WBLT scores and knee-to-hip flexion ratio at the moment of peak flexion during a single-leg vertical hopping test in 24 male professional football players, indicating the sagittal plane

coupling between the knee and hip joints remains stable irrespective of ankle DF ROM. Although the study reported in this Chapter did not find a significant relationship for ankle DF ROM and any variable associated with sagittal plane hip joint kinematics, the trend towards significance suggests that the hip joint may be limited in its ability to compensate for restricted ankle DF ROM during bilateral drop-landings from drop heights of $\geq 100\%$ of CMJ height. This would occur as the hip joint shares a stable relationship with the knee joint to maintain sagittal plane coupling during landings. Further research is required to determine the effect ankle DF ROM may have on hip joint mechanics during bilateral drop-landings.

An alternative explanation for the findings reported in this Chapter may be the negative relationships found between ankle DF ROM and initial contact angles at the ankle (r = -0.31 – -0.34) and knee (r = -0.37 – -0.40) joints. These relationships indicate that individuals with reduced ankle DF ROM compensate during landing tasks by altering their posture at initial contact, with reduced ankle dorsiflexion (i.e. greater ankle plantar flexion) and knee flexion. Altering initial contact angles at the lower extremity have previously been highlighted as a strategy for force dissipation, with greater ankle plantar flexion and reduced knee flexion at initial contact resulting in lower peak vGRF and loading rates during landings (Rowley and Richards, 2015). Landing with greater ankle plantar flexion at initial contact potentially offsets deficits in dorsiflexion at peak flexion to maintain total sagittal plane joint displacement. This strategy offers individuals with reduced ankle DF ROM a solution to maintaining peak vGRF at a manageable level. In support of this suggestion, no relationship was observed between ankle DF ROM and initial contact angles at drop heights of 50% of maximum CMJ height. As elevated drop height results in greater peak vGRF (Zhang, Bates and Dufek, 2000), it is likely that compensatory strategies incorporating increased ankle plantar flexion were only required for the dissipation of landing forces when performing

landings from greater drop heights. However, landing with greater ankle plantarflexion at initial contact has been shown to result in greater risk for ankle ligament injury (Wright et al., 2000). Additionally, landing with reduced knee flexion at initial ground contact would maximise anterior cruciate ligament elevation angle (defined as the angle between the longitudinal axis of the anterior cruciate ligament and the tibia plateau), resulting in increased anterior cruciate ligament loading (Yu and Garrett, 2007) with the shear forces associated with landing in this position (Chappell et al., 2002). Therefore, the findings reported here indicate that deficits in ankle DF ROM elicit movement compensations at initial contact during landings, which could result in increased injury risk (Aerts et al., 2013; Delahunt et al., 2013).

Ankle DF ROM was negatively associated with peak flexion angles for the ankle and knee joint at all drop heights. Restrictions in ankle DF ROM have been associated with reduced peak ankle dorsiflexion (Dowling, McPherson and Paci, 2018) and knee flexion (Dowling, McPherson and Paci, 2018; Fong et al., 2011; Hoch et al., 2015; Malloy et al., 2015) during various landing tasks. The relationship between ankle DF ROM and peak knee flexion angle during landings is particularly relevant during rehabilitation, or for management of injury risk among populations who regularly perform landing activities. Limited knee flexion during landings has been shown to result in greater peak vGRF (Zhang, Bates and Dufek, 2000), quadriceps activity (Blackburn and Padua, 2009), frontal plane knee abduction moments (Pollard, Sigward and Powers, 2010) and increased patellofemoral joint stress (calculated by dividing patellofemoral joint reaction force by patellofemoral contact area) (Olbrantz et al., 2018). The combined increase in these variables is associated with increased risk of anterior cruciate ligament injury (Renstrom et al., 2008) and patellofemoral pain syndrome (Boling et

al., 2009). As such, limitations in ankle DF ROM may be a modifiable risk factor for knee joint injuries.

A positive relationship between ankle DF ROM and FPPA was found during bilateral drop-landings at all drop heights, suggesting that participants with reduced ankle DF ROM had greater knee valgus at the moment of peak flexion. This is consistent with previous research that limited ankle DF ROM is associated with medial knee displacement during a number of functional closed kinetic chain activities (see Lima et al., 2018). It has been suggested that this compensation occurs in order to allow the proximal tibia to continue its forward rotation over the foot via a pronation strategy at the foot complex (Dill et al., 2014). This strategy for managing vGRF during landings is related to increased lower extremity injury risk (Renstrom et al., 2008) and might be avoidable when interventions are employed to improve ankle DF ROM.

An additional aim in this Chapter was to establish the relationship between asymmetries in ankle DF ROM and landing mechanics. Asymmetries in peak vGRF (Schot, Bates and Dufek, 1994) and sagittal plane ankle, knee and hip joint displacement (Pappas and Carpes, 2012) have been reported during bilateral drop-landings in healthy populations. The findings of this Chapter revealed inter-limb asymmetries in ankle DF ROM were not significantly related to any of these variables. Although there are a number of possible explanations for these null relationships, it is possible that the variability associated with performance of bilateral drop landings is responsible (Schot, Bates and Dufek, 1994). This suggestion is consistent with the findings reported in *Chapter 4*, where MDC values for asymmetries in peak vGRF, as well as ankle, knee and hip joint displacement ranged from 14.6–16.2%, 7.1–13.2°, 7.1–8.7° and 6.0–

9.3°, respectively. As these values exceed those reported in studies that have identified asymmetries in landing mechanics (see Harry et al., 2018; Pappas and Carpes, 2012; Schot, Bates and Dufek, 1994), it appears any influence asymmetries in ankle DF ROM may have on landing mechanics is undetectable using the procedures outlined in this thesis.

It was hypothesised that relationships between ankle DF ROM and landing mechanics would increase in magnitude at greater drop heights. This was based on reports that landings from greater drop heights increased peak flexion angles for the ankle, knee and hip joints (McNitt-Gray, 1991; Zhang, Bates and Dufek, 2000). Therefore, it was hypothesised that participants with reduced ankle DF ROM would utilise less ankle dorsiflexion when dropping from greater heights, displaying exaggerated compensations in their coordination strategies in order to safely attenuate vGRF. While the significant relationships found were descriptively different between drop heights, there was considerable overlap of 95% CIs, thereby inferring no statistical differences. As overlap was present in all relationships, the study reported in this Chapter did not identify a clear influence of drop height on the association between ankle DF ROM and landing strategy.

5.5 Conclusion

Ankle DF ROM did not relate to peak vGRF during bilateral drop-landings. This appears to be due to the compensations in coordination strategies developed by individuals with reduced ankle DF ROM. In particular, findings indicate that individuals with limited ankle DF ROM land with greater ankle plantar flexion and knee extension at initial contact to support the attenuation of GRF and compensate for reduced ankle dorsiflexion and knee flexion at the moment of peak flexion. As the relationships established were predominantly *moderate*,

factors beyond ankle DF ROM likely influence the landing strategy adopted by an individual. Although a significant association between ankle DF ROM and ankle and knee joint kinematics during landings was consistently identified across drop heights ≥ 100% of CMJ height, the relationship between ankle DF ROM and hip kinematics during bilateral drop-landings remains unclear. Further investigation is required to identify the influence ankle DF ROM may have on the hip joint during bilateral drop-landings and will be addressed in *Chapter 6* of this thesis. Furthermore, frontal plane compensations were also observed, with ankle DF ROM also being related with FPPA. Although these alterations in movement strategies allow individuals to manage the vertical forces experienced during landings, they may also lead to a greater injury risk during landing activities. Lastly, due to the considerable overlap of 95% CIs for all of significant relationships between different drop heights, the findings in this Chapter indicate that drop height does not influence the relationship between ankle DF ROM and landing mechanics. Based on the findings from *Chapter 4* and *Chapter 5*, no distinguishable differences can be made for the reliability data and the relationships established between ankle DF ROM and measures of landing performance for drop heights equating to 100% and 150% of CMJ. As such, drop heights for the remaining investigations of this thesis will use 150% of CMJ to further develop an understanding for the effect restrictions in ankle DF ROM have on landing mechanics.

Chapter 6

Restrictions in ankle dorsiflexion range of motion alter landing kinematics but not movement strategy when fatigued

6.1 Introduction

To support the dissipation of high peak vGRF during landings, simultaneous flexion at the ankle, knee and hip joints following ground contact must occur (Yeow, Lee and Goh, 2011; Zhang, Bates and Dufek, 2000). Thus, movement strategies that assist in safely attenuating landing forces and enhancing mechanical efficiency are advantageous for reducing injury risk (Hewett et al., 2005). For example, sagittal plane ankle, knee and hip joint alignment at initial contact (Chappell et al., 2002; Rowley and Richards, 2015; Blackburn and Padua, 2009) and at peak flexion (Zhang, Bates and Dufek, 2000) influence the magnitude of peak vGRF during landings, while greater angular joint displacement for the ankle, knee and hip joint supports the load sharing of peak vGRF across each joint segment (Begalle et al., 2015). Failure to adopt a movement strategy that efficiently maintains peak vGRF below a manageable threshold may result in the development of acute (Hewett, Myer and Ford, 2006) and chronic (Dierks et al., 2011) injuries to the lower extremity.

The knee and hip joints have been identified as primary segments for shock absorption during bilateral drop-landings (Yeow, Lee and Goh, 2011). However, as shown in *Chapter 5*, restrictions in ankle DF ROM can negatively influence the coordination of the proximal segments during landings by imposing a mechanical organismic constraint that can limit an individual's capacity to adopt effective movement strategies. It is therefore possible that reduced ankle DF ROM contributes to the development of compensatory strategies

throughout the lower extremity in an attempt to prevent increases in peak vGRF beyond a tolerable threshold (see *Chapter 5*). Consistent with this suggestion, several studies have reported no relationship between ankle mobility and landing forces (Malloy et al., 2015; Whitting et al., 2011). However, *Chapter 5* found ankle DF ROM measured using the WBLT, is negatively related to knee flexion angles at initial contact, indicating individuals with restricted ankle DF ROM contact the ground with reduced knee flexion during bilateral drop-landings from drop heights equating to 100% and 150% of CMJ height. Significant negative relationships were also found between ankle DF ROM and peak ankle dorsiflexion, knee flexion and FPPA at the moment of peak flexion. Collectively, the findings of *Chapter 5* suggest restrictions in ankle DF ROM cause a stiffer landing strategy through limiting knee flexion, necessitating compensations at initial ground contact and the moment of peak flexion to prevent excessive peak vGRF.

Landing tasks can also be affected by exercise-induced fatigue (defined as the inability for the neuromuscular system to maintain mechanical work for a given task (Fousekis, Tsepis and Vagenas, (2012)), which has been shown to increase injury risk (Borotikar et al., 2008). This may occur as prolonged activities such as repetitive jumping, result in exercise-induced fatigue that reduces lower extremity force producing capabilities (Zadpoor and Nikooyan, 2012). In order to maintain the effective attenuation of peak vGRF, altered movement strategies are required to compensate for diminished muscular force production. When fatigue is present, ankle plantar flexion angle acutely increases while knee flexion angle decreases during bilateral drop-landings at the moment of initial ground contact (Weinhandl, Smith and Dugan, 2011). These alterations in coordination strategies likely help to prevent fatigue-induced elevations in peak vGRF by increasing angular displacement at the ankle and knee joints (Begalle et al., 2015; Rowley and Richards, 2015). Interestingly, such

compensations are similar to those demonstrated at initial contact by individuals with restrictions in ankle DF ROM (see *Chapter 5*). It may be that during a fatigued state, individuals with limited ankle DF ROM are unable to alter joint alignment at initial contact as a strategy to manage peak vGRF due to the mobility restriction already requiring this compensation.

It is also feasible that restrictions in ankle DF ROM reduce degrees of movement freedom across key lower-limb segments at the moment of peak flexion during landings, which might be necessary for individuals to effectively control peak vGRF in a fatigued state. Weinhandl, Smith and Dugan (2011) found that greater peak ankle dorsiflexion occurred at maximum knee flexion following a fatigue protocol that resulted in an acute decline in CMJ performance. Similarly, Madigan and Pidcoe (2003) found peak knee flexion angles increased during landings when participants were acutely fatigued. James, Scheuermann and Smith (2010) also detected increased sagittal plane joint displacement for the knee during bilateral drop-landings after fatiguing exercise. Collectively, these studies show that when individuals are fatigued, attenuation of peak vGRF is achieved by increasing the vertical displacement of their centre of mass. For individuals whose movement is constrained by a restriction in ankle DF ROM, this compensatory strategy may not be fully available and their ability to cope with the addition of fatigue may be compromised.

Therefore, the aims of the study presented in this Chapter were: i) to examine differences in landing performance between individuals with restricted and normal ankle DF ROM and ii) identify the effect of fatigue on the compensations in landing strategies for individuals with restrictions in ankle DF ROM. It was hypothesised that: i) individuals with limitations in

ankle DF ROM will present with detectable differences in landing mechanics and ii) individuals with restricted ankle DF ROM would fail to adopt compensatory strategies during landings in an exercise-induced fatigue state that would result in greater peak vGRF.

6.2 Methods

6.2.1 Study design

A mixed study design was employed in which participants were assigned to independent groups (based on ankle DF ROM) who all performed landing tasks in both a non-fatigued and fatigued state. Participants were classified as either having restricted ankle DF ROM (restricted group) or normal ankle DF ROM (normal group) according to performance on the overhead squat and forward arm squat tests (Rabin and Kozol, 2016). Briefly, participants were required to complete the overhead squat test and forward arm squat test for six and three repetitions, respectively. This screen was employed as combined, the overhead squat test and forward arm squat test possesses perfect sensitivity (1.00) and fairly high specificity (ranging between 0.84 and 0.88) for detecting individuals with functional limitations in ankle DF ROM (Rabin and Kozol, 2017). Performance was graded in real-time against the criteria rating outlined by Rabin and Kozol (2016). Participants who demonstrated a negative overhead squat test were invited to take part in a testing session and assigned to the normal group. Participants who displayed a positive finding for both the overhead squat test and forward arm squat test were invited to participate in a testing session and assigned to the restricted group. Participants who presented with a positive overhead squat test and negative forward arm squat test were excluded from the investigation and did not attend a subsequent testing session.

After completing the tests for group allocation, participants attended a single-test session wearing spandex shorts and vest, where ankle DF ROM was measured bilaterally using the WBLT. Participants then performed three maximal CMJ to establish drop height for the bilateral drop-landings and the threshold for establishing the onset of fatigue. Five bilateral drop-landings were then completed from a drop height of 150% CMJ height, both before and after the performance of a fatiguing protocol. All participants were informed of the risks associated with the testing prior to completing a pre-exercise questionnaire and providing informed written consent. Ethical approval was provided by the University of Cumbria Research Ethics Panel (Appendix 1). All test sessions were conducted between 10:00 am and 1:00 pm to control for circadian variation.

6.2.2 Participants

Using the ES of 0.47 presented by James, Scheuermann and Smith (2010) for differences in sagittal plane knee joint displacement during landings following the performance of a fatigue protocol, a representative analysis was performed using G*power to determine the appropriate sample size. With an alpha of 0.05, calculations indicated that to achieve 80% statistical power, a minimum of eight participants per group were required to determine differences in landing mechanics following the fatigue protocol. All participants were required to meet the following inclusion criteria: (1) between the ages of 18–40; (2) no lower extremity injury six-months prior to testing; (3) no history of lower extremity surgery; (4) regularly compete 1–3 times per week in sport events involving landings activities, such as court, racquet or team sports.

Twenty-eight participants volunteered to take part in the experiment. Following the initial screening session using the criteria previously described, four participants were excluded from the analysis, with 12 participants assigned to the restricted group (6 males, 6 females; age = 21 ± 1 years, height = 1.73 ± 0.10 m, body mass 72.4 ± 10.7 kg) and 12 participants to the normal group (6 males, 6 females; age = 23 ± 5 years, height = 1.70 ± 0.07 m, body mass 63.7 ± 8.0 kg).

6.2.3 Procedures

Following the recording of height and body mass during the test session, ankle DF ROM was measured bilaterally using the WBLT (see *Chapter 3*). This procedure was repeated three times for each limb. To ascertain that inter-limb differences did not exist, an independent *t*-test was used to compare the mean of the three trials for left and right WBLT scores. Mean inter-limb differences (1.3 ± 1.4° and 2.1 ± 1.7° for the restricted and normal group, respectively) were not significant ($P > 0.05$) and the right limb was used for data analysis.

Following a standardised warm-up, participants were then familiarised with the performance of a CMJ. For the CMJ, participants stood bare feet with a hip-width stance with their hands placed on their hips. Participants were then asked to rapidly descend prior to explosively jumping as high as possible, with no control being placed on the depth or duration of the countermovement. Jump height was measured using photoelectric cells (Optojump System, Microgate, Bolzano, Italy). Three maximal effort CMJs were performed, with 60 s recovery between attempts. The maximum value of the three attempts was used to calculate drop height for the bilateral drop-landings as well as to establish the onset of fatigue during the fatigue protocol.

Reflective markers were then placed directly onto the participants' skin by the same investigator using the anatomical locations previously outlined (see *Chapter 4*). For both sagittal and frontal plane views, markers were placed on the participants' right side only. Participants were then familiarised with the bilateral drop-landings. Bilateral drop-landings were performed as described in *Chapter 4* from a drop height of 150% of CMJ height only. For each condition (baseline and post fatigue protocol), participants performed five bilateral drop-landings for data collection. Baseline testing allowed for 60 s recovery between landings, while following the fatigue protocol, no recovery was provided between landings beyond the time it took to ascend the height-adjustable platform. For 2D video analysis, sagittal and frontal plane joint movements were recorded as described in *Chapter 4*.

The fatigue protocol consisted of participants performing 30 successive CMJ, while maintaining the same technique as described above. Participants were instructed to keep their hands on their hips and repeatedly jump as high as possible for 30 repetitions, while spending minimal time on the ground between repetitions. Verbal encouragement was provided to ensure participants demonstrated maximal effort throughout. Following the 30th repetition, participants rested 30 s before performing a maximal CMJ for testing purposes. Participants then repeated the protocol until a > 20% decline in CMJ jump height during testing was demonstrated (Weinhandl, Smith and Dugan, 2011). Once participants were unable to reach > 80% of their maximum CMJ height, five bilateral drop-landings were immediately performed using the procedures previously described, with no recovery between landings so as to maintain a fatigued state. The last maximal CMJ was recorded for data analysis, with the

percentage of fatigue calculated as CMJ height post fatigue protocol divided by CMJ height pre fatigue protocol, multiplied by 100.

6.2.7 Data analysis

Raw vGRF data were low-pass filtered using a fourth-order Butterworth filter with a cut-off frequency of 50 Hz (Roewer et al, 2014). Peak vGRF data were calculated for each leg and an independent *t*-test was performed between mean values of peak vGRF for the right and left leg for each participant, which revealed no difference between limbs ($t_{(46)} = 0.657$, *P* = 0.515). As such, peak vGRF, time to peak vGRF and loading rate were independently calculated for the right leg and used for data collection. Peak vGRF data were normalised to body mass and initial contact velocity ($N \cdot kg^{-1} \cdot m \cdot s^{-1}$). Initial contact velocity was calculated using the following equation (Niu et al., 2014):

$$\text{Initial contact velocity } (m \bullet s^{-1}) = \sqrt{2g \cdot DH}$$

where *g* is the gravitational acceleration and *DH* is drop height. Time to peak vGRF and loading rate was calculated as described in *Chapter 4*.

All video recordings were analysed as described in *Chapter 4* of this thesis. For sagittal plane joint movements, ankle, knee and hip joint angles were calculated at initial contact and the point of peak flexion for the right limb. These angles were then used to calculate joint displacement for each joint by subtracting the peak flexion angle from the initial contact angle. For hip flexion, knee flexion and ankle dorsiflexion, smaller values represented greater

flexion and ankle dorsiflexion. For FPPA, values < 180° represented knee valgus and values > 180° represented knee varus.

6.2.8 Statistical analysis

Descriptive statistics (means ± standard deviation) were calculated for each kinetic and kinematic variable. Normality was confirmed for all dependent variables using the Shapiro-Wilk test. Independent *t*-tests were employed to determine between group differences for WBLT scores, maximum CMJ height and percentage of fatigue for CMJ height following the fatigue protocol. To test the first hypothesis, between-group differences at baseline for landing performance were examined using an independent *t*-test for kinetic and kinematic measures. ES were calculated as the difference between the means divided by the pooled standard deviation for all baseline measures and interpreted using the following criteria: 0.20, a *small* difference; 0.50, a *moderate* difference; 0.80, a *large* difference (Cohen, 1988).

A one-way analysis of covariance (ANCOVA) was performed to test the second hypothesis for between-group differences for landing performance following the fatigue protocol. This statistical analysis was chosen so as to provide greater statistical power and reduce variability, while accounting for between-group differences at baseline caused by the procedures for group allocation (Zhang et al., 2014; de Boer et al., 2015). Values for kinetic and kinematic variables associated with landing performance following the fatigue protocol were used as the dependent variable, with baseline (pre) values used as the covariate. The *α-priori* level of significance was set at *P* < 0.05, with a Bonferroni correction applied *post-hoc* in order to reduce the likelihood of Type I errors. Partial eta squared (η^2) values were calculated to indicate the magnitude of group differences in landing mechanics following the

fatigue protocol using the following criteria: 0.02, a small difference; 0.13, a medium difference; 0.26, a large difference (Cohen, 1988). All statistical tests were performed using SPSS® statistical software package (v.24; SPSS Inc., Chicago, IL, USA).

6.3 Results

6.3.1 Between-group differences at baseline

There was a main effect of group on WBLT scores, with the normal group demonstrating 12.6° more ankle DF ROM ($t_{(22)}$ = -10.19, P < 0.001). However, there was no effect of group on baseline differences in CMJ height ($t_{(22)}$ = -1.96, P = 0.062). Table 6.1 presents baseline differences between groups for kinetic and kinematic measures associated with landing performance. There was no effect of group on kinetic measures associated with landings between groups at baseline (P > 0.05).

At initial contact, there was a main effect of group on knee alignment, with the restricted group landing with less knee flexion ($t_{(22)}$ = 3.12, P = 0.005). A main effect of group was also found for peak flexion angles for all joints in the sagittal plane, with the restricted group displaying less ankle dorsiflexion ($t_{(22)}$ = 4.10, P < 0.001), knee flexion ($t_{(22)}$ = 5.34, P < 0.001) and hip flexion ($t_{(22)}$ = 2.28, P = 0.033). Sagittal plane joint displacement for the ankle ($t_{(22)}$ = -4.35, P < 0.001), knee ($t_{(22)}$ = -4.35, P < 0.001) and hip ($t_{(22)}$ = -2.35, P = 0.028) were also significantly less for the restricted group. No other differences were found between groups.

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Table 6.1. Between-group differences at baseline for kinetic and kinematic measures associated with landing performance.

	Restricted (*n*=12)	Normal (*n*=12)	Mean difference (95% confidence interval)	Effect size	*P*-value
	Mean ± SD	Mean ± SD			
Weight-bearing lunge test (°)	32.0 ± 3.3	44.6 ± 2.7	-12.6 (-15.1–-10.0)	4.17	<0.001*
Countermovement jump height (m)	0.30 ± 0.08	0.37 ± 0.10	-0.07 (-0.14–0.00)	0.80	0.06
Kinetic variables					
Peak force (N·kg^{-1}·m·s^{-1})	0.068 ± 0.021	0.064 ± 0.011	0.004 (-0.010–0.018)	0.24	0.568
Time to peak force (s)	0.058 ± 0.011	0.055 ± 0.010	0.003 (-0.005–0.012)	0.31	0.450
Loading rate (N·s^{-1})	38.7 ± 21.3	38.0 ± 11.3	0.7 (-13.7–15.2)	0.04	0.916
Initial contact angles					
Ankle (°)	153.1 ± 3.7	150.4 ± 4.8	2.9 (-0.8–6.5)	0.67	0.116
Knee (°)	170.2 ± 3.1	164.7 ± 5.3	5.5 (1.9–9.3)	1.27	0.005*
Hip (°)	161.8 ± 4.9	160.3 ± 5.8	1.6 (-3.0–6.1)	0.29	0.486
Peak flexion angles					
Ankle (°)	110.8 ± 7.6	96.8 ± 9.0	14.0 (6.9–21.1)	1.67	<0.001*
Knee (°)	102.1 ± 6.4	79.2 ± 13.4	22.8 (13.8–31.9)	2.18	<0.001*
Hip (°)	95.0 ± 17.1	78.7 ± 17.9	16.3 (1.5–31.1)	0.93	0.033*
Frontal plane projection angles (°)	200.0 ± 20.8	207.1 ± 19.2	-7.1 (-24.1–9.8)	0.36	0.392
Joint displacement					
Ankle dorsiflexion (°)	42.5 ± 5.9	53.6 ± 6.6	-11.1 (-16.4–-5.8)	1.78	<0.001*
Knee flexion (°)	68.2 ± 5.9	85.5 ± 12.8	-17.3 (-25.5–-9.1)	1.78	<0.001*
Hip flexion (°)	66.9 ± 14.0	81.6 ± 16.5	-14.7 (-27.7–-1.7)	0.96	0.028*

* = Significant between-group difference (*P* <0.05).

6.3.2 Effects of fatigue

Figure 6.1 presents between-group differences for post-test kinematic measures of bilateral drop-landing performance. All participants achieved a > 20% decline in CMJ height following the performance of the fatigue protocol (restricted group = 68.2 ± 9.8%; normal group = 71.0 ± 6.9%), with no difference between groups for scores of percentage of fatigue ($t_{(22)}$ = -0.99, P = 0.333). There were no main effects of group on post-test normalised peak vGRF ($F_{(1,21)}$ = 0.59, P = 0.451, η^2 = 0.03), time to peak vGRF ($F_{(1,21)}$ = 1.17, P = 0.291, η^2 = 0.05) and loading rate ($F_{(1,21)}$ = 0.42, P = 0.523, η^2 = 0.02). Furthermore, the ANCOVA revealed no effect of group on post-test ankle ($F_{(1,21)}$ = 0.03, P = 0.868, η^2 = 0.00), knee ($F_{(1,21)}$ = 0.00, P = 0.965, η^2 = 0.00) or hip joint angles ($F_{(1,21)}$ = 2.12, P = 0.160, η^2 = 0.09) at initial contact. There was a main effect of group on peak flexion for ankle dorsiflexion ($F_{(1,21)}$ = 5.80, P = 0.025, η^2 = 0.22). Changes from baseline showed that the restricted group displayed less ankle dorsiflexion (mean difference = 0.3˚) than the normal group (mean difference = 2.7˚) following the fatiguing protocol. There were no main effects of group on peak knee flexion angle ($F_{(1,21)}$ = 0.60, P = 0.809, η^2 = 0.00), peak hip flexion angle ($F_{(1,21)}$ = 0.20, P = 0.661, η^2 = 0.01) and FPPA ($F_{(1,21)}$ = 1.92, P = 0.180, η^2 = 0.08). There was a main effect of group on ankle joint displacement following the fatiguing protocol ($F_{(1,21)}$ = 7.88, P = 0.011, η^2 = 0.27). Pairwise comparisons revealed greater ankle joint displacement for the normal group (mean difference = 2.4°) relative to the restricted group (mean difference = 0.1°). There was no main effect of group on knee joint displacement ($F_{(1,21)}$ = 0.66, P = 0.427, η^2 = 0.03) and hip joint displacement ($F_{(1,21)}$ = 0.37, P = 0.557, η^2 = 0.02) post-test.
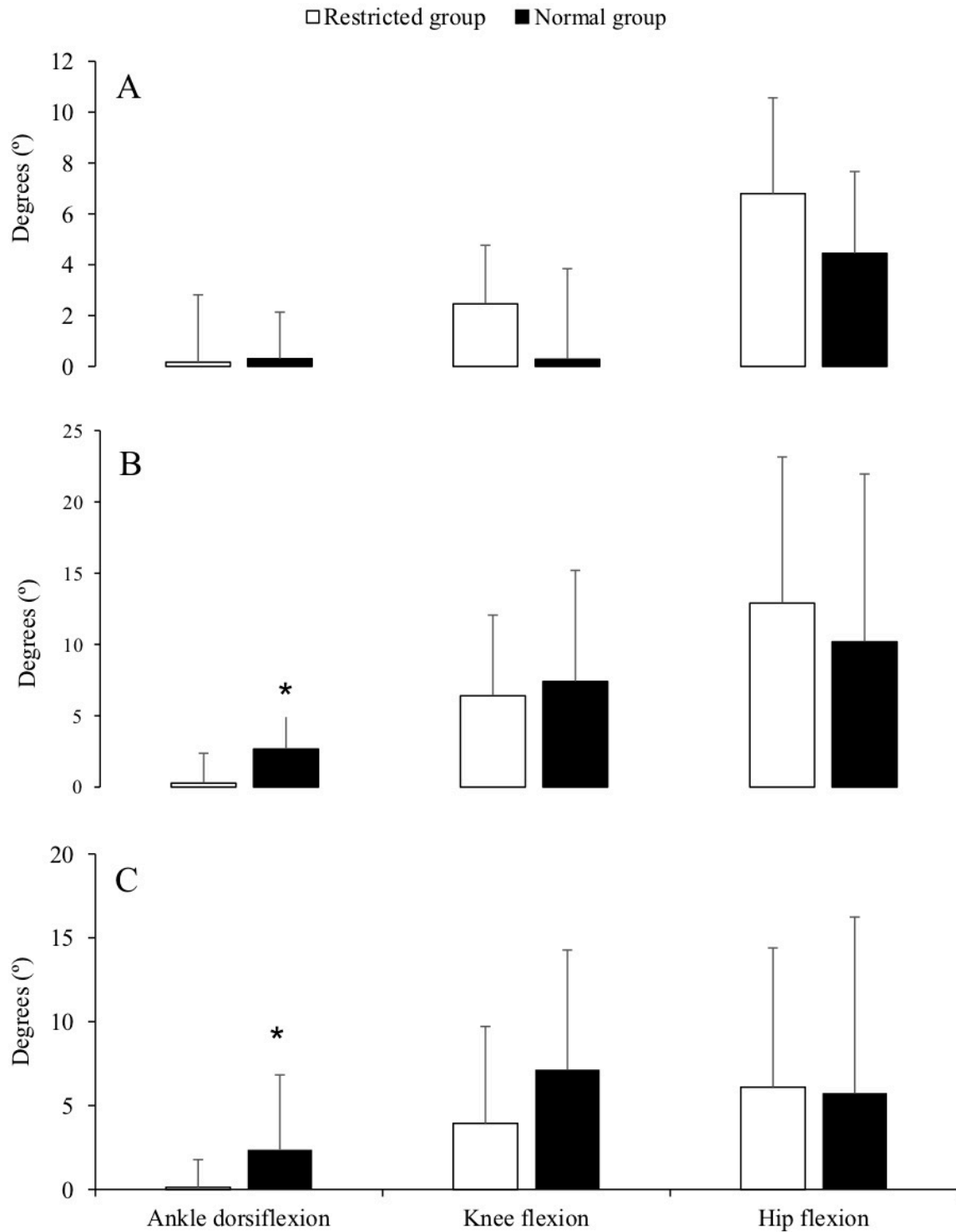
Figure 6.1. Group differences for kinematic measures of bilateral drop-landing performance following the fatigue protocol A) initial contact, B) peak flexion and C) sagittal plane joint displacement. Values represent differences from baseline testing. Means ± SD. * Significant between-group difference ($P < 0.05$).

6.4 Discussion

This study had two main aims; first this Chapter examined the kinetic and kinematic characteristics of landing performance among recreational athletes with either functional restrictions or no restrictions in ankle DF ROM. Secondly, this Chapter assessed the effects of acute fatigue on landing performance between these two groups. It was hypothesised that the restricted group would show reduced sagittal plane joint displacement at the ankle, knee and hip joints relative to the normal group. Further, it was hypothesised that this would affect their ability to compensate for reduced force production capability whilst fatigued, resulting in greater disparities in landing mechanics between groups. Consistent with the first hypothesis, the results revealed that individuals with limited ankle DF ROM landed with less knee flexion at initial contact and reduced ankle, knee and hip flexion at the moment of peak flexion. This resulted in the restricted group displaying significantly less ankle, knee and hip sagittal plane joint displacement relative to the normal group. However, despite these disparities in kinematic patterns, there were no differences in kinetic variables during landing (Table 6.1). Furthermore, these findings show that recreational athletes with limited ankle DF ROM were incapable of utilising greater ankle joint motion when landing in an exercise-induced fatigued state, which was in contrast to the normal group (Figure 6.1). However, this movement compensation did not result in differences between groups for any other kinematic or kinetic variable analysed, meaning that the functional relevance of this finding is uncertain.

A primary finding of the current study was that participants with ankle DF ROM restriction modified their landing mechanics at initial contact and at peak flexion, resulting in significant differences for sagittal plane joint displacement at the ankle, knee and hip joints. Specifically,

at initial contact, participants with restricted ankle DF ROM landed with 5.5° less knee flexion. This is consistent with the findings of Dowling, McPherson and Paci (2018) and those of *Chapter 5*, where relationships between ankle DF ROM and knee flexion angles at initial contact during single-leg ($r = 0.33$) and double-leg landings ($r = -0.31$) were reported, respectively. Furthermore, this value exceeds the 4.8° MDC value presented in *Chapter 4* of this thesis for this measure. Collectively, these results suggest that individuals compensate for restrictions in ankle DF ROM (as measured using the WBLT) by landing with greater knee extension prior to contacting the ground. It is likely that this movement strategy occurs in an attempt to maintain knee joint displacement, as peak knee flexion angles are significantly reduced by restrictions in ankle DF ROM. The majority of acute non-contact knee injuries occur close to the point of initial contact during landings (Krosshaug et al., 2007). Landing with greater knee extension at initial contact has been associated with increased tibia anterior shear forces (Chappell et al., 2002) and is a known mechanism for anterior cruciate ligament injury (Boden et al., 2010). Therefore, reduced ankle DF ROM may expose the knee to greater shear forces during landings, with the potential to increase injury risk.

Compensations at initial contact for restricted ankle DF ROM did not occur at the ankle joint itself. This was an unexpected finding, given that moderate negative relationships were reported between ankle DF ROM and ankle plantar flexion angles at initial contact during bilateral drop-landings from 100% of CMJ height in *Chapter 5*. Increasing ankle plantar flexion at initial contact provides a functional strategy for managing vGRF (Rowley and Richards, 2015), resulting in preservation of ankle joint displacement (Begalle et al., 2015). However, the relationship between ankle DF ROM and ankle plantar flexion angle at initial contact is not always consistent. Dowling, McPherson and Paci (2018) found no such relationship during single-leg drop landings, while a non-significant relationship during

bilateral drop-landings from drop heights equalling 150% of CMJ height is reported in *Chapter 5*. As the study reported here found no difference in ankle plantar flexion angles at initial contact between groups, it is suggested that the ankle does not provide a significant means of movement compensation at this stage of the landings for those with restrictions in ankle DF ROM.

In the current study, ankle DF ROM restriction significantly reduced baseline measures of peak flexion angles and joint displacement for the ankle, knee and hip joints, with large ES found between groups. This is consistent with previous findings presented in this thesis (see *Chapter 5*), where ankle dorsiflexion and knee flexion angles at peak flexion, along with knee joint displacement, have each been related to WBLT performance. Additionally, between-group differences for peak flexion and sagittal plane joint displacement at the ankle and knee exceeded the MDC values presented in *Chapter 4*. The current finding is, therefore, in keeping with the sagittal plane coupling observed between the ankle and knee joints, whereby dorsiflexion at the ankle complex facilitates flexion at the knee joint during landings (Yeow, Lee and Goh, 2011). This coordination pattern allows for greater shock absorption (Devita and Skelly, 1992; Yeow, Lee and Goh, 2011), supporting the management of peak vGRF when loading is greater due to task constraints. Manipulating the demand of a bilateral drop-landing by increasing drop height from 0.32 m to 1.03 m was reported to increase ankle and knee joint peak flexion angles by 4.2˚ and 11.6˚, respectively (Zhang, Bates and Dufek, 2000). Reduced peak knee flexion angle has been shown to increase peak vGRF (Zhang, Bates, and Dufek, 2000), quadriceps muscle activity (Blackburn and Padua, 2009) and frontal plane knee abduction moments (Pollard, Sigward, and Powers, 2010). Each of these variables has been associated with increased anterior cruciate ligament injury risk (Griffin et al., 2000; Renstrom et al., 2008) and patellofemoral pain syndrome (Boling et al., 2009). Therefore,

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limitations in ankle DF ROM may cause individuals to adopt landing strategies that could potentially cause knee ligament injury.

This is the first investigation, to the author's knowledge, that has shown restrictions in ankle DF ROM significantly reduces hip flexion angles at peak flexion and sagittal plane hip joint displacement during bilateral landings in a healthy population. During bilateral drop-landings, the study reported in *Chapter 5* found ankle DF ROM to have a small and non-significant negative relationship with peak hip flexion angle across all drop heights ($r = -0.23$ to -0.28). In this study, it was found that the restricted group had lower peak hip flexion angles, with a mean difference of 16.3° compared to the normal group. Furthermore, mean hip joint displacement was 14.7° less for the restricted group. The hip joint has been shown to provide an important contribution to the dissipation of forces during landing tasks (Yeow, Lee, and Goh, 2011), with a vital role for managing vGRF when landing from higher drop heights (Zhang, Bates, and Dufek, 2000). As a result, restrictions in ankle DF ROM potentially limit the hip joint's capacity to contribute to vGRF attenuation during landings, particularly from greater drop heights. Of note, while the between-group differences exceeded the SEM values reported in *Chapter 4* for these measures, neither surpassed the MDC values. Therefore, these results should be interpreted with caution.

The study reported in this Chapter found no difference for kinetic measures of landing performance between the restricted and normal group. Studies exploring the relationship between ankle DF ROM and kinetic variables have been inconclusive. Consistent with the findings presented in *Chapter 5*, a number of studies have found no significant relationship for ankle DF ROM and peak vGRF, time to peak vGRF and loading rate (e.g. see Malloy et

al., 2015; Whitting et al., 2011). However, Fong et al. (2011) did identify a moderate negative relationship between ankle DF ROM and peak vGRF during a jump-landing task. It has been proposed that the frontal plane compensations in the lower extremity reported by Whitting et al. (2011) and Malloy et al. (2015) may provide a strategy that assists in preserving the descent of the centre of mass to allow for vGRF attenuation (Mason-Mackay, Whatman and Reid, 2017). However, the data reported here challenges this suggestion, with FPPA for both groups showing no significant difference. The present findings indicate kinetic variables associated with landing performance are unlikely to be regulated exclusively by angular joint displacement or postures at specific time points (i.e. peak flexion) in the lower extremity. Peak vGRF has been negatively correlated with angular velocity for the knee ($r = -0.60$) and hip joint ($r = -0.45$) at initial contact during a stop-jump task (Yu, Lin and Garrett, 2006). Similarly, increased eccentric work performed by the knee and hip extensors (Zhang, Bates and Dufek, 2000) and increased muscular activity prior to initial contact (Devita and Skelly, 1992) also contributes to energy dissipation and aids in the reduction of peak vGRF. Therefore, variables such as knee and hip angular velocity at initial contact and the eccentric work performed by the knee and hip extensor musculature may compensate for the reduced lower extremity joint displacement caused by restrictions in ankle DF ROM, resulting in the management of peak vGRF during landings.

The second major aim of this study was to investigate the effect of exercise-induced fatigue on landing mechanics in individuals with restricted ankle DF ROM. In this regard, another primary finding was the difference found between groups in ankle joint coordination during landings after an acute bout of exercise-induced fatigue. Moderate and large effects were found for post-intervention ankle joint angle at peak flexion and sagittal plane ankle joint displacement, respectively. These findings suggest that the restricted group was unable to

access additional ankle dorsiflexion when performing landings in a fatigued state (Figure 6.1). This was in contrast to the normal group, who increased peak ankle dorsiflexion by 2.7° and ankle joint displacement by 2.4° when acutely fatigued. Whether such small differences in peak flexion angles and joint displacement at the ankle are functionally relevant is unknown. However, no differences were found when comparing groups and the effect of fatigue for the knee or hip joints for any kinematic measure associated with landing performance. Furthermore, no differences between groups were identified for any kinetic variable analysed following the fatigue protocol. As both groups were still able to access greater joint displacement at the knee and hip during landings, it seems that the additional ankle joint displacement used by the normal group played no role in the management of peak vGRF through the facilitation of proximal joint segments.

Another consideration is whether 2D video analysis is able to detect such differences in landing strategy. *Chapter 4* reported on the reliability of using 2D video analysis for bilateral drop-landings from drop heights equating to 150% of maximum CMJ height and reported MDC values for ankle dorsiflexion angle at peak flexion and ankle joint displacement were 6.8° and 6.0°, respectively. As differences for the normal group following the fatigue protocol did not exceed these thresholds it is unlikely that the change in joint kinematics for this group can be defined as 'real'. Therefore, individuals with restrictions in ankle DF ROM are no more constrained in their ability to adjust their landing strategy when fatigued than individuals with normal ankle mobility. These findings suggest the presence of ankle DF ROM hypomobility does not increase injury risk when performing landings in a fatigued state.

6.5 Conclusion

Individuals who have restricted ankle DF ROM, based on their performance of closed-chain activities, adopt different landing strategies compared to non-restricted controls. In particular, individuals with functional limitations in ankle DF ROM use less ankle motion relative to controls during bilateral drop-landing landings. This is further exaggerated with the addition of fatigue, although these differences must be interpreted with caution due to the sensitivity of 2D video analysis for detecting changes in landing kinematics. At the knee, individuals compensate for reduced peak knee flexion angles by landing in a more extended posture at initial contact, in an attempt to maintain sagittal plane knee joint displacement and preserve peak vGRF below a tolerable level. This is the first investigation to demonstrate that restrictions in ankle DF ROM affect sagittal plane hip kinematics during bilateral landings, with reduced peak flexion angles and sagittal plane joint displacement at the hip. As restrictions in ankle DF ROM appear to promote landing strategies that are more extended and stiffer in nature, injury risk may be increased during landing tasks for individuals with limited ankle DF ROM. As the findings presented in this Chapter have provided a clearer understanding for how ankle DF ROM impacts bilateral landing strategies, *Chapter 7* will investigate the effects of an intervention to improve ankle DF ROM on landing mechanics in restricted individuals. As the findings of this Chapter demonstrated limited influence of fatigue on the interaction between ankle DF ROM and landing mechanics, fatigue will not be further investigated.

<div align="center">**Chapter 7**</div>

<div align="center">**Improved ankle mobility following a 4-week training programme affects landing**</div>

<div align="center">**mechanics: a randomised controlled trial**</div>

A version of this Chapter has been accepted for publication in The Journal of Strength and Conditioning Research and the printed version can be found in Appendix 11.

7.1 Introduction

During bilateral landings, ankle dorsiflexion aids in attenuating vGRF (Zhang, Bates and Dufek, 2000), whilst facilitating knee and hip flexion via sagittal plane coupling mechanisms to reduce the impact of landing (Yeow, Lee and Goh, 2011). Restrictions in ankle DF ROM is recognised as a modifiable injury risk factor for athletes who perform a high volume of landing activities (Backman and Danielson, 2011). This is likely due to compensations caused by ankle DF ROM restriction during landing tasks, resulting in less effective strategies being used. For example, *Chapter 6* of this thesis showed restricted ankle DF ROM reduces peak flexion angles at the ankle, knee and hip joints. Additionally, during landings where individuals with restricted ankle DF ROM demonstrate reduced sagittal plane joint displacement at the knee, a negative relationship between ankle DF ROM and peak vGRF during bilateral landings has been reported (Fong et al., 2011). These findings suggest that individuals with ankle DF ROM restrictions land using a stiffer strategy that may result in greater landing forces.

Increased ankle mobility may improve landing mechanics by increasing sagittal plane joint displacement at the ankle, knee and hip (Dowling, McPherson and Paci, 2018; Fong et al., 2011), resulting in reduced peak vGRF (Zhang, Bates and Dufek, 2000) and, consequently,

diminished injury risk (Hewett et al., 2005). Ankle DF ROM can be improved in relatively short time periods as significant gains in ankle DF ROM have been shown in ≤ 4-weeks when adhering to interventions designed to increase flexibility of the ankle plantar flexors (Aune et al., 2019; Nakamura et al., 2017) and improve joint arthrokinematics (Jeon et al., 2015). However, little is known regarding the functional consequences of developing ankle mobility as currently no studies have investigated the effect of increasing ankle DF ROM on landing mechanics in individuals identified with a mobility restriction at the ankle joint.

In practice, individuals with restrictions in ankle DF ROM will likely be identified during a pre-exercise screening session prior to initiating a strength and conditioning programme (Howe, Waldron and Read, 2017). When deficits in ankle DF ROM are found, a corrective programme to restore ankle mobility would be prescribed. This would likely be performed as a supplementary intervention alongside a strength-training programme designed to develop relevant physical qualities that will improve athletic performance. However, whether a corrective programme aimed at restoring ankle mobility results in greater sagittal plane ankle, knee and hip joint displacement, which in turn results in reduced peak vGRF during landing tasks is unknown. Therefore, the primary aim of this investigation was to determine the effects of a 4-week ankle mobility programme combined with a strength-training programme on landing mechanics among participants with pre-established ankle restrictions. It was hypothesised that increased ankle mobility would transfer to improved landing mechanics relative to exclusively performing a general strength-training programme. This would occur as a result of the mobility restriction being reduced, allowing for greater sagittal plane joint displacement at the ankle, knee and hip, enhancing shock absorption capacity and rendering compensatory strategies obsolete.

7.2 Methods

7.2.1 Study design

For this investigation, a randomised control trial with an independent groups design was used to investigate the efficacy of a 4-week intervention aimed at improving ankle DF ROM and its associated effects on landing mechanics. The independent variable distinguishing groups was the ankle mobility intervention, with participants either performing a strength-training and ankle mobility programme, or a strength-training programme exclusively. During an initial screening session, participants were required to perform the overhead squat test and forward arm squat test and were graded in real-time against the criteria rating outlined by Rabin and Kozol (2017). Participants with a positive finding for both the overhead squat test *and* forward arm squat test were identified as those demonstrating restricted ankle DF ROM and invited to participate in the study.

Participants that met the inclusion criteria were tested, both before and following the completion of a 4-week intervention, for their performance on the WBLT, maximal CMJ and bilateral drop-landings. Participants were randomly assigned to one of two groups: strength and mobility training; or strength-training only. Group allocation was performed following the initial screening session via an online randomisation system (www.sealedenvelope.com), using stratified randomisation, matched for gender, WBLT scores on the right limb and maximal CMJ height. Both groups performed the same strength-training programme for the lower extremity and trunk musculature, while the strength and mobility group concurrently completed a programme using exercises known to improve ankle DF ROM. Post-testing was performed within seven days of completing the intervention for all participants. All test

sessions were conducted between 10:00 am and 1:00 pm to control for circadian variation. For all testing sessions, participants wore spandex shorts and vest. All participants were informed of the risks associated with the testing and training intervention prior to completing a pre-exercise questionnaire and providing informed written consent. Ethical approval was provided by the University of Cumbria Research Ethics Panel (Appendix 1).

7.2.2 Participants

Using the data from Jeon et al. (2015) who examined differences in ankle DF ROM during the WBLT following self-mobilisation, we performed a representative analysis using G*power to determine the appropriate sample size. With an alpha of 0.05, calculations indicated that to achieve 80% statistical power, a minimum of eight participants per group were required. All participants were required to meet the following inclusion criteria: (1) between the ages of 18–40; (2) no lower extremity injury six-months prior to testing; (3) no history of lower extremity surgery; (4) regularly compete 1–3 times per week in sport events involving landings activities, such as court, racquet or team sports; (5) no previous experience adhering to a structured strength-training programme (6) present with a positive overhead squat and forward arm squat test during the initial screening session, as outlined by Rabin and Kozol (2017). The screen was employed as combined, the overhead squat test and forward arm squat test possesses perfect sensitivity (1.00) and fairly high specificity (ranging between 0.84 and 0.88) for detecting individuals with functional limitations in ankle DF ROM (Rabin and Kozol, 2017). To prevent sport training and competition from influencing outcome measures, data collection and the intervention were completed in the competitive off-season for each participant. Fifty-three participants volunteered for the investigation, with 23 matching the inclusion criteria. Eleven participants were randomly assigned to the strength

and mobility group (6 males, 5 females; age = 21 ± 1 years, height = 1.74 ± 0.10 m, body mass 75.7 ± 15.4 kg) and 12 participants assigned to the strength-training only group (6 males, 6 females; age = 20 ± 1 years, height = 1.72 ± 0.10 m, body mass 71.4 ± 6.8 kg).

7.2.3 Procedures

Testing sessions were structured so that following the recording of height and body mass, ankle DF ROM was measured bilaterally using the WBLT (see *Chapter 3*). This procedure was repeated three times for each limb, with the mean value for the right limb across the three attempts used for data analysis. The greatest inter-limb difference during the WBLT across all participants was 1.1˚, with a mean inter-limb difference of 0.3 ± 0.5° and 0.1 ± 0.4° for the strength and mobility and strength-training only group, respectively.

To establish bilateral drop-height for each participant, three maximal CMJ were performed using the procedures described in *Chapter 6*. The maximum value of the three attempts was used for data analysis and the maximum value from the first test session used to calculate drop height for the bilateral drop-landings for both testing sessions.

Reflective markers were then placed directly onto the participants' skin by the same investigator (see *Chapter 4*). For both sagittal and frontal plane views, markers were placed on the participants right side only. Participants were then familiarised with the bilateral drop-landings from a drop height of 150% of maximum CMJ height (see *Chapter 4*). Participants performed five bilateral drop-landings for data collection, with 60 s recovery between

landings. For 2D video analysis, sagittal and frontal plane joint movements were recorded as outlined in *Chapter 4* of this thesis.

For the interventions, all participants were required to attend three separate training sessions per week for 4-weeks. Sessions involved performing a strength-training programme supplemented with either an intervention to increase ankle DF ROM (strength and mobility group) or the strength-training programme exclusively (strength-training only group). The strength-training programme was designed to develop lower limb and trunk force development capacities (Table 7.1). For all strengthening exercises, loading was progressed on a session-by-session basis depending on participants' individual responses. This was achieved by maintaining the sets and repetition structure for each exercise, while increasing load so that each set was performed 2–3 repetitions from failure whilst maintaining desirable exercise form (Zourdos et al., 2016).

Table 7.1. Strength-training programme performed by all participants.

Exercise	Sets	Reps	Rest (s)
Dynamic warm-up			
Pole squats	2	6	30
Squats with arms forward	2	10	30
Split squats	2	6	30
Single leg box squats	2	6	30
Countermovement jumps	3	6	120
Session 1			
Pause front squat	3	8–10	120
Single-leg box squat	3	10–12	120
Nordic leg curls	3	6–9	120
Session 2			
Romanian deadlifts	3	10–12	120
Reverse lunges	3	8–10	120
Prone bridge	3	30–60 s	60
Session 3			
Pause front squats	3	8–10	120
Step ups	3	10–12	120
Side bridge	3	30–60 s	60

The intervention to increase ankle DF ROM was performed by the strength and mobility group on the same days as the strength-training programme, with exercises completed prior to the dynamic warm-up or following the strength-training programme (Table 7.2). The ankle mobility intervention exercises have previously been shown to increase ankle DF ROM and included self-mobilisation (Jeon et al., 2015), self-massage (Halperin et al., 2014), eccentric strength-training (Aune et al., 2019), and static stretching (Youdas et al., 2003). A brief description is provided for each exercise in Table 7.2 and demonstrated in Figure 7.1. Prescription of all acute variables for the self-mobilisation exercise, self-massage and static stretching exercise remained the same throughout the 4-week intervention. The loading for the eccentric strength-training exercise was progressed using the same format as described for all other strength exercises.

Each training session was separated by at least 48-hours and supervised by a UK Strength and Conditioning Association accredited coach. All participants were consistently provided with coaching to improve movement quality for each exercise. Participants were asked to refrain from performing any other strength exercises for the duration of the intervention.

Table 7.2. Ankle mobility exercises completed by participants in the strength and mobility training group.

| Exercise	Sets	Reps/Duration	Performance
Pre-training session			
Ankle stretch using a strap	3 each leg	20 s	Participant positions their front foot on a 10° incline board (length = 0.30 m, width = 0.10 m) and their rear foot behind the front foot in a short lunge position. A non-elastic looped strap (approximately 0.30 m in length) is positioned so the front of the strap is on the anterior aspect of the talus on the front leg and the back of the strap loops over the medial arch of the rear leg. Participants lunge forward until end ankle DF ROM is achieved for the front leg, whilst both feet remain flat on their respective surfaces. This position is held for the prescribed time, with strap tension modulated by manipulating the distance between the feet (Jeon et al., 2015).
Ankle plantar flexors self-massage	3 each leg	30 s	Participant assumes a seated position, with one knee flexed to 90° and the ankle slightly plantar flexed 10° using a heel support. From this position, participants massage the plantar flexors using a roller massager. The cadence is 1 s to roll the length of the calf muscles, with intensity set at 7/10 using the rate of perceived pain (Halperin et al., 2014).
Post-training session			
Single-leg heel drop	3 each leg	12–15 reps	The participant places their hands on a wall to maintain balance, whilst standing with their heels hanging off a 0.30 m box. Participants plantar flex at both ankles to their end range, then remove one leg off the box before lowering their centre of mass by fully dorsiflexing the ankle on the weight-bearing limb until the point of maximal perceived stretch. The descent phase is performed at a cadence of 6 s and is self-timed (Aune et al., 2019). To load the movement, participants hold a load in one hand. Loading is progressed on a session-by-session basis and is achieved by maintaining the sets and reps structure, while increasing load so that each set is performed 2–3 repetitions from failure.
Bent knee ankle plantar flexor stretch	2 each leg	1 min	The participant places their hands on a wall to maintain balance, whilst standing with one heel (the limb being stretched) hanging off a 0.30 m box. The other foot is positioned so the whole of the foot is on the box. With the knee bent to approximately 30° on the back leg, the participant dorsiflexes the ankle on the stretched foot until a sensation of a substantial stretch is reported (Youdas et al., 2003). This position is held as prescribed.
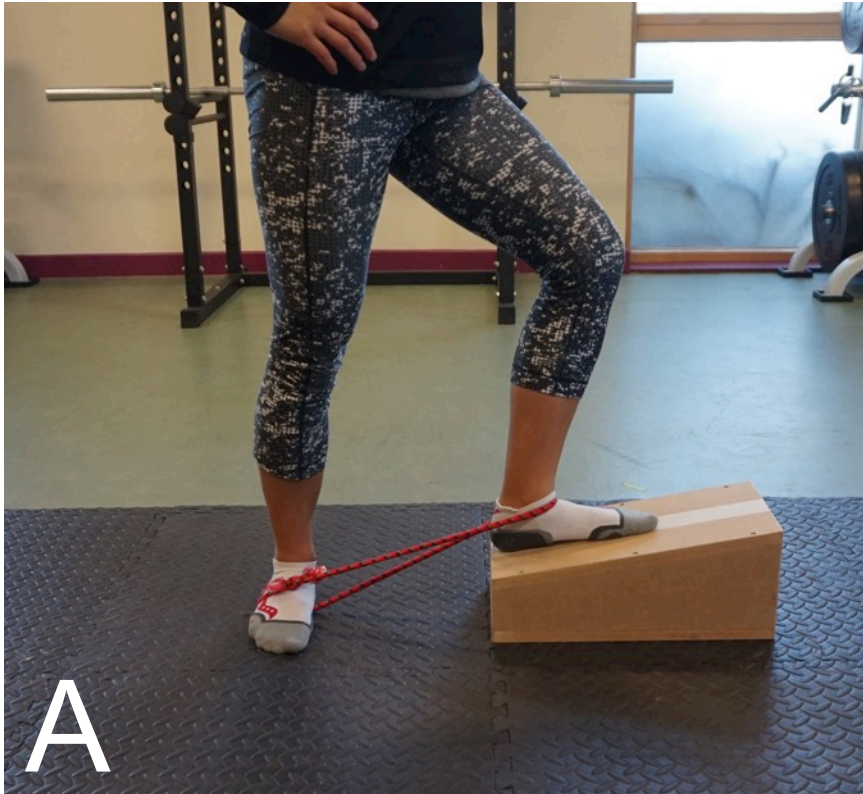
Figure 7.1. Exercises used to increase ankle DF ROM for the strength and mobility group. A) Ankle stretch using a strap; B) Ankle plantar flexors self-massage; C) Single-leg heel drop; D) Bent knee ankle plantar flexor stretch.

7.2.5 Data analysis

Raw vGRF data were low-pass filtered using a fourth-order Butterworth filter with a cut-off frequency of 50 Hz (Roewer et al., 2014). Peak vGRF data were calculated for each leg and normalised to body mass ($N \cdot kg^{-1}$). An independent *t*-test was performed between mean values of peak vGRF for the right and left leg for each participant, with no difference found ($t_{(38)}$ = -0.847, *P* = 0.402). Based on these findings, force-time data was chosen from the right leg to represent kinetic measures of bilateral drop-landing performance. As such, peak vGRF, time to peak vGRF and loading rate were independently calculated for the right leg (see *Chapter 4*).

All video recordings were analysed using the procedures described in *Chapter 4* of this thesis. For sagittal plane joint movements, ankle, knee and hip joint angles were calculated at initial contact and the point of peak flexion for the right limb. These angles were then used to calculate joint displacement for each joint by subtracting the peak flexion angle from the initial contact angle. For hip flexion, knee flexion and ankle dorsiflexion, smaller values represented greater flexion and ankle dorsiflexion. For FPPA, values < 180° represented knee valgus and values > 180° represented knee varus.

7.2.6 Statistical analysis

Descriptive statistics (mean ± standard deviation) were calculated for each kinetic and kinematic variable. The assumption of normality was checked for all dependent variables using the Shapiro-Wilk test. Independent *t*-tests were employed to determine between-group differences for WBLT scores and maximum CMJ height at baseline. A one-way analysis of

covariance (ANCOVA) was used to evaluate difference in WBLT and CMJ performance and between-group differences for landing performance following the training intervention. A one-way ANCOVA was chosen as a statistical tool so as to increase power, reduce variability and account for between-group differences at baseline caused by the procedures for group allocation (de Boer et al., 2015; Zhang et al., 2014). Values for kinetic and kinematic variables associated with landing performance following the training intervention were used as the dependent variable, with baseline values used as the covariate to control for group differences. The α-priori level of significance was set at $P < 0.05$, with a Bonferroni correction applied post-hoc in order to reduce the likelihood of Type I errors. ES were calculated for each comparison, with 0.20 being considered *small*, 0.50 *moderate* and 0.80 or greater *large* (Cohen, 1988). All statistical tests were performed using SPSS® statistical software package (v.24; SPSS Inc., Chicago, IL, USA).

7.3 Results

Three participants from the strength-training only group withdrew from the study (for reasons unrelated to the study), resulting in 20 participants completing both testing sessions (strength and mobility, $n = 11$; strength-training only, $n = 9$). Attendance for the training sessions was 100% for participants included in the data analysis.

At baseline, there was no difference between groups for CMJ height ($t_{(18)} = -0.25$, $P = 0.282$) or WBLT scores ($t_{(18)} = 0.26$, $P = 0.153$). However, there was a main effect of group on WBLT at the post intervention time point ($F_{(1,17)} = 13.94$, $P = 0.002$) (Figure 7.2), with the strength and mobility group (mean difference = 4.1 ± 1.4˚, ES = 1.00) demonstrating greater ankle DF ROM than the strength-training only group (mean difference = 1.0 ± 2.1˚, ES =

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0.18). There was no difference in CMJ height between the strength and mobility (mean difference = 0.04 ± 0.02 m, ES = 0.52) and strength-training only group (mean difference = 0.03 ± 0.02 m, ES = 0.31) following the training intervention ($F_{(1,17)}$ = 3.95, P = 0.063) (Figure 7.3).
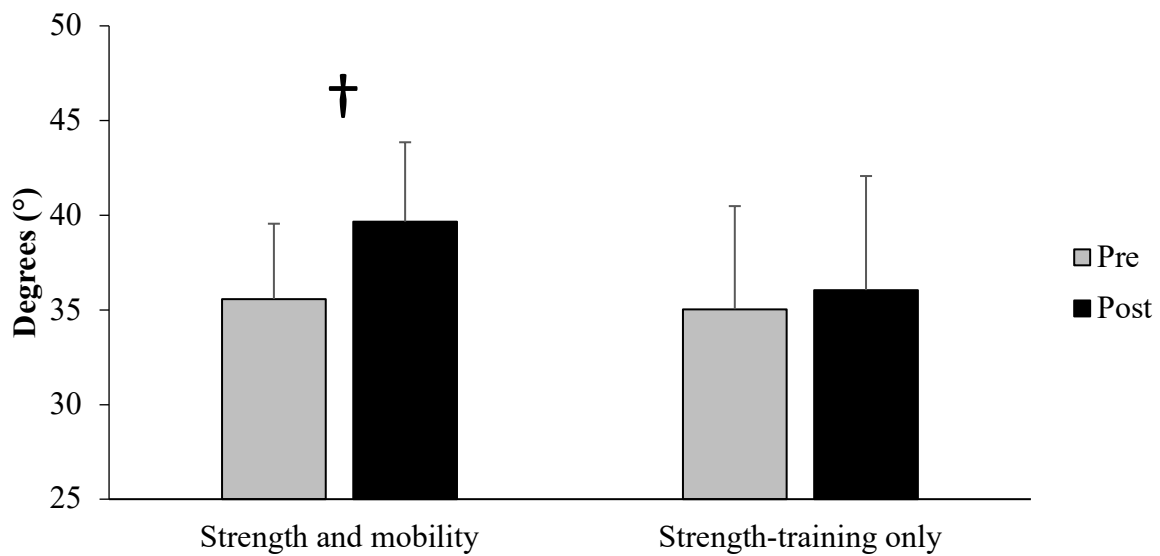

Figure 7.2. WBLT values for both groups (error bars indicate the SD). † indicates a significant between-group difference for post-intervention values (P = 0.002).
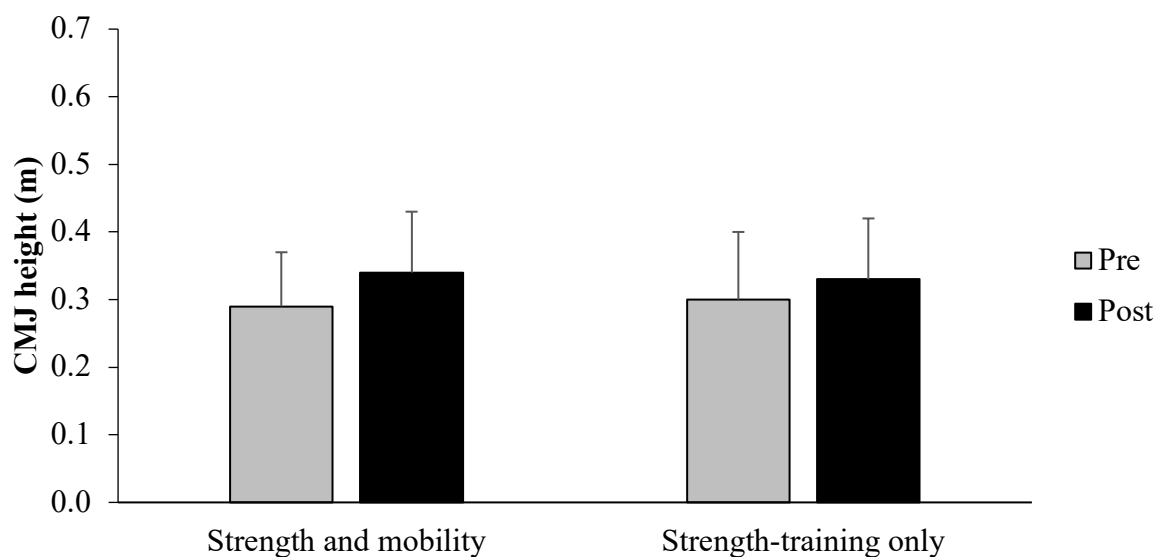
Figure 7.3. CMJ test values for both groups (error bars indicate the SD).

Differences for kinematic and kinetic measures of bilateral drop-landing performance before and after the training intervention are presented in Table 7.3. At initial ground contact a main effect of group was found following the training intervention ($F_{(1,17)}$ = 4.68, P = 0.045), with the strength and mobility group (mean difference = 1.4 ± 2.0˚, ES = 0.46) having less ankle dorsiflexion than the strength-training only group (mean difference = 1.0 ± 2.7˚, ES = 0.22). At peak flexion, there was a main effect of group on ankle dorsiflexion ($F_{(1,17)}$ = 19.14, P < 0.001) and hip flexion ($F_{(1,17)}$ = 4.87, P = 0.041). The strength and mobility group (mean difference = 6.3 ± 2.9˚, ES = 0.74) displayed greater ankle dorsiflexion at peak flexion compared to the strength-training only group (mean difference = -0.4 ± 3.7˚, ES = 0.06), while the strength-training only group showed greater hip flexion at peak flexion (mean difference = 14.4 ± 11.0˚, ES = 0.70) in comparison to the strength and mobility group (mean difference = 4.3 ± 9.0˚, ES = 0.16). Joint displacement for the ankle was significantly greater for the strength and mobility group (mean difference = 7.7 ± 4.0˚, ES = 1.00) than for the strength-training only group (mean difference = -1.4˚ ± 3.3˚, ES = 0.23) following the

training intervention ($F_{(1,17)} = 25.33$, $P < 0.001$). Significant between-group differences were identified post-intervention for hip joint displacement ($F_{(1,17)} = 6.13$, $P = 0.024$), with the strength-training only group showing greater hip joint displacement (mean difference = 8.0 ± 6.6˚, ES = 0.44) than the strength and mobility group (mean difference = 0.7 ± 6.6˚, ES = 0.03). No other between-group differences were found for kinematic measures associated with bilateral drop-landing performance. No significant between-group differences were found for any kinetic measure following the interventions.

Table 7.3. Pre- and post-intervention differences for both groups for kinematic and kinetic measures associated with landing performance.

Variable	Strength and mobility ($n = 11$)				Strength-training only ($n = 9$)			
	Pre-intervention (mean ± SD)	Post-intervention (mean ± SD)	Change in mean (95% confidence interval)	Effect size	Pre-intervention (mean ± SD)	Post-intervention (mean ± SD)	Change in mean (95% confidence interval)	Effect size
Kinetic variables								
Peak force (N·kg^{-1})	2.07 ± 0.69	2.01 ± 0.69	-0.06 (-0.19, 0.08)	0.08	1.86 ± 0.34	1.88 ± 0.48	0.02 (-0.19, 0.15)	0.05
Time to peak force (s)	0.058 ± 0.018	0.058 ± 0.019	0.000 (-0.003, 0.003)	0.01	0.058 ± 0.010	0.064 ± 0.016	0.006 (-0.006, 0.017)	0.41
Loading rate (N·s^{-1})	41.1 ± 22.9	40.5 ± 23.6	-0.6 (-4.4, 5.6)	0.03	34.6 ± 11.8	34.3 ± 14.5	-0.3 (-6.1, 6.8)	0.02
Initial contact angles								
Ankle (°)*	152.2 ± 2.9	153.6 ± 3.1	1.4 (-0.1, 2.8)	0.46	154.1 ± 4.0	153.1 ± 5.2	-1.0 (-2.6, 0.6)	0.22
Knee (°)	169.5 ± 2.3	167.9 ± 2.9	-1.6 (-4.0, 0.9)	0.60	172.0 ± 3.8	168.0 ± 4.1	-3.9 (-6.6, -1.2)	1.00
Hip (°)	161.7 ± 6.4	158.0 ± 6.5	-3.7 (-7.7, 0.3)	0.58	162.8 ± 4.5	156.4 ± 8.5	-6.4 (-10.8, -1.9)	0.94
Peak flexion angles								
Ankle (°)*	108.4 ± 9.0	102.0 ± 8.2	-6.3 (-8.4, -4.3)	0.74	105.8 ± 6.8	106.2 ± 7.3	0.4 (-1.9, 2.7)	0.06
Knee (°)	100.4 ± 16.0	97.0 ± 14.7	-3.4 (-7.7, 0.9)	0.22	99.4 ± 15.6	95.1 ± 15.8	-4.3 (-9.0, 0.5)	0.27
Hip (°)*	96.1 ± 27.0	91.7 ± 28.1	-4.3 (-10.6, 1.9)	0.16	99.4 ± 23.3	85.0 ± 17.7	-14.4 (-21.3, -7.4)	0.70
Frontal plane projection angles (°)	199.3 ± 22.7	204.9 ± 22.3	5.6 (0.0, 11.2)	0.25	195.5 ± 13.2	198.4 ± 14.1	2.8 (-3.3, 8.9)	0.21
Joint displacement								
Ankle dorsiflexion (°)*	43.9 ± 7.3	51.6 ± 8.1	7.7 (5.4, 10.1)	1.00	48.3 ± 5.6	46.9 ± 6.8	-1.4 (-4.0, 1.2)	0.23
Knee flexion (°)	69.1 ± 15.0	70.9 ± 13.8	1.8 (-1.8, 5.5)	0.13	72.6 ± 15.7	72.9 ± 14.8	0.3 (-3.7, 4.4)	0.02
Hip flexion (°)*	65.7 ± 23.6	66.3 ± 23.0	0.7 (-3.6, 4.9)	0.03	63.4 ± 20.2	71.4 ± 16.2	8.0 (3.3, 12.7)	0.44

* = Significant difference between-groups.

7.4 Discussion

The primary aim of this investigation was to identify the effects of a corrective training programme on landing mechanics among participants with restricted ankle DF ROM. It was hypothesised that increasing ankle DF ROM alongside a strength-training programme would transfer to the execution of a landing task when compared to performing a strength-training programme alone. Specifically it was hypothesised that those receiving an intervention to increase ankle DF ROM and a strength-training programme would demonstrate greater sagittal plane joint displacement at the ankle, knee and hip following the removal of the ankle restriction. The findings, however, failed to support this hypothesis, with changes in landing movement strategies during bilateral drop-landings identified for both groups. Specifically, relative to the strength-training only group, increases in ankle DF ROM in the strength and mobility group also resulted in greater ankle plantar flexion at initial ground contact, ankle dorsiflexion at peak flexion, and ankle joint displacement. In contrast, between-group comparisons following the completion of the 4-week programme revealed that the strength-training only group adapted their coordination strategy by increasing hip flexion angle at the moment of peak flexion, resulting in increased sagittal plane hip joint displacement (Table 7.3). As such, it appears that changes in landing strategies following the performance of a strength-training programme are specific to whether restrictions in ankle mobility are considered as part of the design of the intervention.

This is the first investigation to demonstrate that landing mechanics can be altered among individuals who initially present with ankle DF ROM restrictions. Following the intervention, the strength and mobility group increased peak ankle dorsiflexion and ankle joint displacement during bilateral drop-landings by 6.3 and 7.7˚ respectively. These values were

significantly greater than those observed for the strength-training only group (-0.4 and 1.4˚, respectively) and exceed the SEM for both of these kinematic variables previously reported using the same procedures in *Chapter 4*. Along with contributing to shock absorption at initial ground contact (Rowley and Richards, 2015), the ankle joint contributes significantly to angular displacement of the knee joint in the sagittal plane during landings (see *Chapters 5 and 6*). Knee flexion is vital for absorbing shock (Zhang, Bates and Dufek, 2000), with reduced knee flexion diminishing knee extensor power output during landings (Devita and Skelly, 1992). As a result, reduced sagittal plane knee joint displacement may lead to suboptimal landing strategies that increase peak vGRF (Zhang, Bates and Dufek, 2000). Given that the ankle restriction was reduced (i.e. ankle DF ROM increased) following the 4-week intervention, improvements in ankle mobility may facilitate the knee joint's capacity to dissipate vertical forces. In partial support of this suggestion, the strength and mobility group increased their knee flexion at peak flexion by 3.4˚ (Table 7.3). However, this value is less than the SEM of 3.9˚ reported in *Chapter 4* for this variable during bilateral drop-landings using 2D video analysis and should not be interpreted as real change following the intervention. Furthermore, as peak vGRF did not change for the strength and mobility group beyond the error associated with this measure, the modest increase in peak knee flexion angle is unlikely to have provided any functional benefit, as landing forces remained unaffected (Table 7.3).

The findings in this Chapter demonstrate that the strength and mobility group landed with greater ankle plantar flexion at initial ground contact during post-intervention testing. This strategy may be desirable when individuals are attempting to reduce loading associated with a landing task, as 10˚ increases in plantar flexion at initial contact have been shown to decrease peak vGRF and loading rate (Rowley and Richards, 2015). The same investigation also

showed greater plantar flexion at initial contact increased ankle joint contribution to peak support moments. Although this Chapter did not measure changes in plantar flexion strength following the intervention, it may be that elevated strength levels following the performance of the single-leg heel drops allowed the ankle to contribute further to energy dissipation. Although this is possible, the mean difference from baseline for ankle joint angle at initial contact for the strength and mobility group following the intervention was 1.4˚ (Table 7.3). This value is far less than the conscious adjustments in ankle joint alignment used by Rowley and Richards (2015), explaining the lack of difference in kinetic measurements following the intervention. Furthermore, this value is also less than the SEM of 1.8˚ reported in *Chapter 4*. Therefore, the between-group difference for ankle alignment at initial contact could be explained by systematic error and should be interpreted with caution.

Another unexpected finding was the changes at the hip joint for the strength-training only group. These findings show that peak hip flexion angle and hip joint displacement increased by 14.4˚ and 8.0˚, respectively. These values were significantly greater than the 4.3˚ increase in peak hip flexion angle and 0.7˚ for sagittal plane hip joint displacement observed for the strength and mobility group. This finding was surprising, as individuals with functional limitations in ankle DF ROM have been shown to land with reduced peak hip flexion angles and less hip joint displacement when compared to individuals with greater ankle DF ROM (Dowling, McPherson and Paci, 2018). However, this may be beneficial for individuals with limited ankle DF ROM to offset the stiffer landing strategy associated with the presence of an ankle restriction. Increasing hip flexion during bilateral landings has been shown to reduce peak vGRF and quadriceps muscle activity, while increasing peak knee flexion angle (Blackburn and Padua, 2008). Previously, recreational athletes with restrictions in ankle DF ROM have been shown to increase peak hip flexion angle during bilateral drop jumps

following the performance of a hip-strengthening programme (Kondo and Someya, 2016). As the strength-training only group in the current study did not increase ankle DF ROM beyond the error associated with the test (see *Chapter 3*), it seems that an increased involvement of the hip occurred to support the knee in attenuating loading during the bilateral drop-landings. This is likely to have occurred because the strength-training only group were unable to rely on greater ankle joint contribution during landings due to the remaining ankle restriction. Thus, the landing strategy of both groups was altered but in different ways. This finding could be of practical significance to individuals with conditions resulting in chronic (less modifiable) restrictions in ankle DF ROM, such as anterior ankle impingement (Ogilvie-Harris, Mahomed and Demaziere, 1993). Increased hip flexion during landings is associated with increased hip extensor activity, which acts to resist the elevated external flexion moment (Shimokochi et al., 2009). As a result, practitioners working with individuals with a non-modifiable ankle restriction should consider that hip-dominant strategies will be adopted and that training interventions placing greater emphasis on development of the hip musculature could help to tolerate the additional loading that is likely to occur.

In this investigation, no between-group differences were found for any kinetic measure of bilateral drop-landing performance following the 4-week training interventions. Furthermore, neither group demonstrated changes outside of the error previously associated with these measures (*Chapter 4*). Although a number of reasons may exist for these findings, the most likely explanation is the limited evidence for ankle DF ROM influencing landing forces. At present, only Fong et al. (2011) has found a significant correlation between ankle DF ROM and peak vGRF (r = -0.41) in healthy participants. Alternatively, similar to the findings presented in *Chapters 5* and *6*, numerous studies have shown no association between ankle DF ROM and peak vGRF during landing tasks (Malloy et al., 2015; Whitting et al., 2011). As

such, it is likely that other factors influence peak vGRF, such as angular velocity for the knee and hip joints at initial ground contact (Yu, Lin and Garrett, 2000) and the eccentric work performed by the knee and hip extensor musculature (Zhang, Bates and Dufek, 2000). Therefore, the findings presented in this study provide further support for the lack of association between ankle DF ROM and peak vGRF.

In *Chapter 5*, ankle DF ROM was associated with FPPA during landing tasks, indicating that reduced ankle DF ROM increases knee abduction angle. This is suggested to occur as a compensation mechanism for limited ankle DF ROM, whereby increased pronation of the foot complex allows for the continued forward rotation of the proximal tibia (Dill et al., 2014). However, this finding is not consistently reported, with Fong et al. (2011) showing no relationship between ankle DF ROM and knee valgus displacement, while no between-group differences were found in *Chapter 6* for FPPA when comparing individuals with and without ankle DF ROM restriction. In this Chapter, no between-group differences for FPPA were found following the 4-week intervention, with the strength and mobility group and the strength-training only group increasing FPPA angle (increasing knee varus) by 5.6˚ and 2.8˚, respectively (Table 7.3). Both of these values are below the 12.0˚ MDC value previously reported in *Chapter 4* and, consequently, should not be interpreted as a genuine change in frontal plane knee alignment. A possible explanation for why ankle DF ROM did not result in significant reductions in knee valgus (increases in FPPA angle) may be that meaningful medial knee displacement was not found for either group at baseline (strength and mobility group = 199.3 ± 22.7˚; strength-training only group = 195.5 ± 13.2˚). Therefore, supplementing a strength-training programme with an intervention to improve ankle DF ROM, does not appear to reduce peak knee abduction angles during bilateral drop-landings

relative to exclusively performing the strength-training programme in individuals who present with no apparent medial knee displacement.

7.5 Conclusion

This Chapter demonstrated that individuals with a functional restriction in ankle DF ROM were able to change their DF ROM and landing mechanics following a 4-week ankle mobility and strength-training programme. Specifically, those individuals exposed to a strength *and* mobility training programme significantly improved their ankle mobility, resulting in greater ankle dorsiflexion at peak flexion and increased sagittal plane ankle joint displacement when landing relative to those who received a strength-training intervention exclusively. Furthermore, these changes in joint alignment exceeded the error associated with the testing procedures. Conversely, the strength-training only group compensated for their restriction in ankle DF ROM by employing more hip flexion during landings following the strength-training only programme. This is the first investigation to show that improved ankle DF ROM leads to changes in landing mechanics, whereby the newly developed mobility can be integrated into a landing strategy.

Chapter 8

General Conclusion

8.1 Overview

Ankle DF ROM restriction affects landings mechanics in healthy populations during single- (Dowling, McPherson and Paci, 2018; Whitting et al., 2011) and double-leg landing tasks (Fong et al., 2011; Malloy et al., 2015; Sigward, Ota and Powers, 2017). Following an extensive review of the literature, *Chapter 2* highlighted that these changes occurred at initial ground contact (Dowling, McPherson and Paci, 2018) and peak flexion (Malloy et al., 2015; Sigward, Ota and Powers, 2017; Whitting et al., 2011), resulting in reduced angular joint displacement at the ankle, knee and hip joints during landings (Dowling, McPherson and Paci, 2018; Fong et al., 2011). Consistent with evidence that reduced joint displacement at these joint segments is associated with higher peak vGRF during bilateral landings (Zhang, Bates and Dufek, 2000), restrictions in ankle DF ROM have been shown to increase peak vGRF (Fong et al., 2011). However, *Chapter 2* identified a number of areas requiring further research. Specifically, it was not known whether inter-limb asymmetries in ankle DF ROM cause asymmetries in landing mechanics. Furthermore, the influence of drop height or exercise-induced fatigue on compensations in landing mechanics, derived from restrictions in ankle DF ROM, had not been established. Lastly, there was no evidence to support the notion that ankle DF ROM is a modifiable factor for changing landing mechanics following a corrective training programme. Based on the limited evidence identified within the existing body of literature, the programme of work presented in this thesis was designed to provide original insight into the effects of restricted ankle DF ROM on landing mechanics and the influence of acute and chronic constraints (such as exercise-induced fatigue or long-term ankle immobility) on the compensatory strategies derived. Consequently, a number of

research aims were developed to investigate the effects of ankle DF ROM on bilateral drop-landing performance. This Chapter will provide a review of these aims and present the key findings of the current thesis, along with limitations that may direct future research in this area.

Aim 1: Establishing a reliable method for identifying ankle DF ROM. Special consideration was also provided towards determining the reliability for inter-limb asymmetries in ankle DF ROM.

The findings in *Chapter 3* demonstrated that single-limb measures of WBLT performance using the trigonometric calculation method to establish tibia angle provided superior reliability to methods previously described (Powden, Hoch and Hoch, 2015). Therefore, the findings of *Chapter 3* demonstrated that a single-limb measure of WBLT performance using the trigonometric calculation method could be reliably used to identify ankle DF ROM for subsequent investigations in this body of work.

Although there is evidence that inter-limb asymmetries in ankle DF ROM affect the performance of functional activities, such as squatting (Crowe et al., 2019) and change of direction tasks (Gonzalo-Skok et al., 2015), few studies have confirmed the prevalence of inter-limb asymmetries in ankle DF ROM among healthy populations. As identified in *Chapter 2*, there is also limited evidence for the reliability of measuring inter-limb asymmetries in ankle DF ROM using the WBLT. The results of *Chapter 3* revealed that inter-limb asymmetries were less prevalent than previously reported among healthy young adults, with only 8% of participants demonstrating ankle DF ROM asymmetry > 5˚. Although a number of reasons may explain this discrepancy, a likely cause is the error associated with

measuring asymmetries in ankle DF ROM during the WBLT. The findings presented in *Chapter 3* showed MDC values for measures of inter-limb asymmetry were higher than a single measure. Therefore, studies reporting high prevalence and magnitude of asymmetries in ankle DF ROM using less reliable measurement methods during the WBLT (e.g. Rabin et al., 2015), could be confounded by variability in the adopted measurement technique. As a result, the prevalence and magnitude of asymmetries in ankle DF ROM for healthy young adults was much lower than expected and therefore, may not be as common as previously reported.

Aim 2: Examining the reliability of kinetic and kinematic measures associated with bilateral drop-landing performance from varying drop heights. The reliability of inter-limb asymmetry measurements during landings was also established.

Chapter 4 investigated the reliability of kinetic and kinematic variables associated with bilateral drop-landing performance at drop heights equating to 50%, 100% and 150% of maximal CMJ height. For force-time variables, bilateral and unilateral measures of peak vGRF, time to peak vGRF and loading rate demonstrated sufficient reliability, predominantly from drop heights $\geq$ 100% of CMJ height. As such, these variables were used to measure landing performance in the studies that form this thesis. However, regardless of drop height, between-limb asymmetries in peak vGRF demonstrated a high level of measurement noise (i.e. error), with MDC values ranging between 14.6–16.2%. This value far exceeds what is commonly reported in healthy (Yanci and Camara, 2016) and injured populations (Paterno et al., 2007). Therefore, it was reasoned that this measure potentially lacks the necessary sensitivity to detect meaningful changes in landing mechanics or kinetic measures of asymmetry caused by ankle DF ROM during bilateral drop-landings. Nonetheless, as the

influence of asymmetries in ankle DF ROM on landing mechanics had not been investigated prior to this programme of work, it remained feasible that a relationship (of sufficient strength) between ankle DF ROM and kinetic measures could be detected. Consequently, *Chapter 5* investigated the relationship of between-limb differences in ankle DF ROM and asymmetries in kinetic measures during landing performance.

Using 2D video analysis, initial contact and peak flexion angles for the ankle, knee and hip joints on both limbs produced reliable measures with sufficient sensitivity to identify the association between ankle DF ROM and these variables (*Chapter 5*), along with detecting changes following the acute (*Chapter 6*) and chronic (*Chapter 7*) interventions employed. Sagittal plane joint displacement for all joint segments possessed *very large* to *near perfect* reliability, yet demonstrated high variability at a drop height of 50% of CMJ height when compared to greater drop heights. MDC values for between-limb differences in sagittal plane joint displacement for the ankle, knee and hip joints far exceeded what has been reported in the literature for healthy (Pappas and Carpes, 2012) and injured (Meyer et al., 2018) populations. However, it may be that between-limb asymmetries in ankle DF ROM influence between-limb differences in landing mechanics to such a degree that a relationship may be detectable for these variables. The relationship between ankle DF ROM and asymmetries in landing kinematics was investigated in *Chapter 5*.

Aim 3: Determining the relationship between ankle DF ROM and landing mechanics during bilateral drop-landings from varying drop heights.

Ankle DF ROM has been associated with initial contact angles, peak flexion angles and sagittal plane joint displacement at the ankle, knee and hip joints during a variety of landing

tasks (Dowling, McPherson and Paci, 2018; Fong et al., 2011; Malloy et al., 2015; Sigward, Ota and Powers, 2017). Although it has not been consistently reported in the literature (Malloy et al., 2015), ankle DF ROM has also been correlated with peak vGRF during landings (Fong et al., 2011). *Chapter 5* advanced our understanding of these relationships, investigating the influence of drop height on these associations. During bilateral drop-landings from 50%, 100% and 150% of CMJ height, no significant correlations were found between WBLT performance and normalised peak vGRF, time to peak vGRF and loading rate. This likely occurred due to compensations in coordination strategies used by individuals with restricted ankle DF ROM, with relationships found between ankle DF ROM and lower-extremity initial contact angles, peak flexion angles and sagittal plane joint displacement. Specifically, at drop height ≥ 100% of CMJ height, *moderate* to *large* correlations were found between ankle DF ROM and ankle and knee joint angles at initial contact and peak flexion, as well as sagittal plane joint displacement for the knee. It was proposed in *Chapter 5* that these associations highlight coordination strategies that occur as a compensation for restricted ankle DF ROM, allowing peak vGRF to be maintained below a tolerable threshold. This may also explain why additional significant relationships were found at drop heights ≥ 100% of CMJ height, where peak vGRF increase as a result of elevated drop heights (Zhang, Bates and Dufek, 2000). However, while a greater number of significant correlations were identified at greater drop heights, there was considerable overlap for 95% confidence intervals between these relationships. Therefore, *Chapter 5* did not identify a clear effect for drop height on the relationship between ankle DF ROM and landing mechanics. Based on these findings and that of *Chapter 4*, clear differences were not found for the reliability data or the relationships established between ankle DF ROM and bilateral drop-landing performance for drop heights equalling 100% and 150% of CMJ. As a consequence, drop

heights for *Chapter 6* and *Chapter 7* of this thesis used 150% of CMJ to further investigate the effect restrictions in ankle DF ROM have on landing mechanics.

Lastly, no relationship was found between asymmetries in ankle DF ROM and between-limb differences in landing mechanics. Although a number of reasons may be possible for the lack of association, the most likely explanation is the variability related to measures of asymmetries during bilateral landings. This was possibly compounded by the negligible inter-limb differences found in this population for the WBLT (-0.9 ± 3.0°). Therefore, based on the findings described here and that of preceding chapters, the remainder of the thesis excluded further analysis for measures of asymmetry.

Aim 4: Evaluating the effect of acute fatigue on landing mechanics for individuals with limited ankle DF ROM.

Chapter 6 of the current thesis was the first investigation to study the differences in landing mechanics between individuals with functional restrictions in ankle DF ROM and normal ankle DF ROM, as well as the effect of fatigue on landing mechanics in individuals with restricted ankle DF ROM. The results reported in *Chapter 6* showed that in a non-fatigued state, the restricted group had less knee flexion at initial contact relative to the normal group. Peak flexion and sagittal plane joint displacement for the ankle, knee and hip joints were also significantly reduced for the restricted group at baseline when compared to the normal group, with *large* ES (ranging between 0.9 to 2.2) reported for each variable. Additionally, between-group differences for knee flexion angle at initial contact and peak flexion angles for the ankle and knee joints exceeded the MDC values reported in *Chapter 4*. Force-time measures of landing performance were not different between groups, which was consistent with the

results reported in *Chapter 5* and suggests no significant association between ankle DF ROM and peak vGRF, time to peak vGRF and loading rate during bilateral drop-landings. These data develop the understanding of how landing mechanics are altered among individuals identified with restrictions in ankle DF ROM. Furthermore, these results demonstrate the effect of restrictions in ankle DF ROM on sagittal plane hip joint mechanics during bilateral drop-landings. As such, this is the first investigation to establish the role of restrictions in ankle DF ROM in limiting hip joint contribution to vertically lowering the centre of mass during bilateral landing tasks.

No difference was found between-groups for FPPA. As the correlations reported in *Chapter 5* for FPPA were *moderate*, factors beyond ankle DF ROM likely influence frontal plane knee joint mechanics during landings. Such factors may include suboptimal neuromuscular control of the trunk and hip musculature during landings (Myer et al., 2008) and poor eccentric strength of the hip abductor and external rotator muscles (Boling and Padua, 2013). As such, individuals presenting with high peak knee abduction angles during bilateral landings should be screened for deficits in the magnitude and timing of force production around the trunk and hip muscles as a priority ahead of assessing ankle DF ROM.

In summary, the findings presented in *Chapter 6* provide greater insight into the effects of ankle DF ROM restriction on landing mechanics. Specifically, when compared to participants with no functional hypomobility at the ankle joint, individuals with restricted ankle DF ROM perform bilateral drop-landings with less knee flexion at initial ground contact. Additionally, during the descent phase of the landing, individuals with restricted ankle DF ROM display reduced peak flexion angles at the ankle, knee and hip joints relative to their unrestricted

counterpart. As a result, individuals with restricted ankle DF ROM demonstrate kinematic strategies that are associated with an elevated injury risk during landing tasks (Aerts et al., 2013).

For landings following the performance of a fatiguing protocol, *Chapter 6* found only peak ankle dorsiflexion angle and sagittal plane ankle joint displacement differed between groups. For the normal group, peak ankle dorsiflexion angle increased by 2.7°, resulting in 2.4° greater ankle joint displacement. Conversely, the restricted group increased peak ankle dorsiflexion angle and ankle joint displacement by 0.3° and 0.1°, respectively. Although the between-group comparisons for ankle kinematics following the fatigue protocol were found to be significant using the one-way ANCOVA with *moderate* to *large* differences ($\eta^2 = 0.22$–0.27), the mean changes reported did not exceed the MDC values presented for these variables in *Chapter 4*. Therefore, between-group differences reported in this Chapter may represent measurement error and should not be interpreted as 'real' change. Additionally, as no between-group differences were found for any other measure of landing performance under fatigue, such small changes in ankle kinematics are unlikely to provide functional relevance. Consequently, following the acute onset of exercise-induced fatigue, individuals with restricted ankle DF ROM show no difference in landing strategy when compared to individuals with normal ankle mobility.

Aim 5: Investigating the effects of improved ankle DF ROM on landing mechanics following the performance of an intervention aimed to increase ankle mobility.

Chapter 7 presented the first investigation to demonstrate that individuals with restricted ankle DF ROM alter their landing strategy following a 4-week ankle mobility intervention, as

part of a general strength-training programme. Following the ankle mobility intervention, the strength and mobility group increased their WBLT score by 4.1 ± 1.4˚, compared to 1.0 ± 2.1˚ for the strength-training only group. With the addition of an ankle mobility intervention to a strength-training programme, increased peak ankle dorsiflexion angle (mean difference = 6.3 ± 2.9˚) and ankle joint displacement (mean difference = 7.7 ± 4.0˚) during bilateral drop-landings were found for the strength and mobility group. The mean differences for the strength and mobility group in ankle kinematics exceeded the SEM values reported in *Chapter 4*, with increases in ankle joint displacement surpassing the MDC established for this variable. As such, the significant differences in peak ankle dorsiflexion angle and ankle joint displacement were likely to be meaningful based on the error previously found for each measure. This is the first study to demonstrate that restrictions in ankle DF ROM that reduce ankle joint contribution during landings are modifiable following the performance of a corrective programme.

Although participants who received the strength and ankle mobility intervention had significantly greater ankle joint contribution during landings, peak knee and hip flexion angles increased by only 3.4° and 4.3°, respectively. Although these increases in knee and hip flexion may be meaningful, they are less than MDC values reported in *Chapter 4* for each variable and may possibly represent error. Furthermore, changes in peak vGRF, time to peak vGRF and loading rate following the corrective programme were all negligible and lower than the MDC values previously reported in *Chapter 4*. Therefore, although improved ankle mobility resulted in greater ankle joint displacement for the strength and mobility group, it did not result in measurable kinematic changes in landing strategy at the knee and hip joints, resulting in no differences in kinetic measures of landing performance. In the future, researchers should attempt to identify strategies that facilitate proximal joint contribution to

landing performance following improvements in ankle DF ROM in individuals with restricted ankle mobility.

In contrast to the strength and mobility group, individuals with restricted ankle DF ROM performing a strength-training programme exclusively, demonstrated little change during bilateral drop-landings in peak ankle dorsiflexion angle (mean difference = -0.4 ± 3.7˚) and ankle joint displacement (mean difference = = -1.4˚ ± 3.3˚). However, peak hip flexion angle and hip joint displacement increased by 14.4˚ and 8.0˚, respectively. These findings indicate that when ankle DF ROM is not improved, individuals with ankle hypomobility undertaking a strength-training programme adapt their landing strategy by increasing hip joint contribution to support the attenuation on landing forces. This likely occurs as the ankle DF ROM restriction remains, limiting the forward displacement of the knee joint and constraining the individual to rely more on hip joint involvement to lowering the centre of mass during the descent phase of the landing.

8.2 Limitations and future recommendations for research

The programme of work presented in this thesis provides an original and significant contribution to the literature by establishing the effects of ankle DF ROM on landing mechanics across various conditions. However, this research is not without limitations, which are important to acknowledge and which should be addressed in future investigations. Key limitations are addressed below:

- *Selection of mechanical measures used for analysing landing performance:* A consideration for future research using laboratory-based testing equipment is the measures of landing mechanics chosen to represent performance. For this programme

of work, measures were selected based on their association with injury risk and the practicality for data collection and analysis in a clinical or coaching setting. Across the available literature, force-time measures (Boling et al., 2009; Hewett et al., 2005; Leppänen et al., 2016) and lower extremity joint angles at initial ground contact (Boling et al., 2019; Hewett et al., 2005) and peak flexion (Boling et al., 2009; Hewett et al., 2005; Holden et al., 2017; Leppänen et al., 2016; Leppänen et al., 2017) have been consistently associated with injury risk during bilateral landings using prospective study designs. Furthermore, force-time data and joint angles at critical time points during landings take just minutes to process using custom-built spreadsheets and free video analysis software respectively. Consequently, these measures were selected for this programme of work to accomplish the principal objective previously described. However, investigations for the effects of ankle DF ROM on additional kinetic and kinematic variables associated with landing performance would also provide practical significance. For example, identifying the effects of ankle DF ROM restriction on lower extremity joint moments, muscle activation strategies and joint angular velocities would be potentially informative when determining injury risk during landing tasks and developing corrective strategies. Future research establishing the interaction between ankle DF ROM and landing technique should attempt to develop the depth of the analysis to provide greater insight for how ankle DF ROM may influence landing strategies.

- *Population used for analysis:* This thesis aimed to establish the effects of ankle DF ROM in healthy individuals with a history of performing landing tasks via sport participation. For inclusion into data collection, participants were required to present with no history of lower extremity surgery and no lower extremity injury six-months

prior to data collection. As such, the findings presented in this thesis are not applicable to injured populations, but rather, serve as useful information for such population as well as a 'proof of concept' and an impetus for future research in injured populations.

Patients that have undergone lower extremity surgery (Yamazaki et al., 2010) or have history of recurrent injury (Hertel et al., 2006) have been shown to differ in their functional movement patterns relative to healthy populations. Specific to landings, kinematic and kinetic differences have been reported between healthy and previously injured populations (Decker et al., 2002; Haddas, James and Hooper, 2015). As such, individuals with limited ankle DF ROM and a recent history of injury may display compensations that differ to those reported in this thesis. Therefore, the findings of this thesis should not be applied to injured populations that present with restricted ankle DF ROM.

- *The influence of gender on landing performance:* This thesis investigated the relationship between ankle DF ROM and landing mechanics using a convenience participant sample including both male and female recreational athletes. Fundamental biomechanical disparities exist between males and females, owing to differences in neuromuscular control (Landry et al., 2009), resulting in higher incidence rates for non-contact ligament injuries in female athletes (Arendt and Dick, 1995). Differences in landing strategies between males and females are a common finding (Decker et al., 2003; Kernozek et al., 2005; McLean et al., 2007). For example, females have been shown to land with 10.0° greater ankle plantar flexion and 7.2° less knee flexion angles at initial contact (Decker et al., 2003). In the same investigation, 12.4° and

16.4° greater joint displacement for the knee and ankle joints, respectively, have also been found in female recreational athletes (Decker et al., 2003). Additionally, Kernozek et al. (2005) reported 19.4° greater knee valgus, a risk factor for ACL injury (Hewett et al., 2005), in females performing bilateral drop-landings from a 0.60 m drop height. Although the use of an arbitrary drop height questions the external validity of these findings, gender differences still exist in landing strategy when drop height is normalised to maximal CMJ height (Weinhandl, Joshi and O'Connor, 2010). Although differences may exist in landing strategies between genders, the magnitude of these disparities has been shown to be less apparent during bilateral landings (Pappas et al., 2007). Nonetheless, the degree to which ankle DF ROM impacts landing mechanics for each gender is currently unknown and warrants further investigation. Therefore, caution should be shown when generalising the results presented in this thesis to both genders.

Another limitation associated with gender was that this thesis did not consider menstrual cycle status for female participants, which has been shown to influence tendon stiffness and joint laxity (Cesar et al., 2011). As menstrual cycle affects some of the landing variables measured as part of this thesis (Bell et al., 2014), it is possible that the association found in this investigation between ankle DF ROM and landing performance may be influenced by the menstrual cycle. As such, researchers may wish to examine the influence of menstrual cycle on the relationship between ankle DF ROM and landing performance in future research.

- *Footwear:* Another consideration for interpreting the findings of this thesis is the exclusion of footwear. The removal of footwear was incorporated into the study

design in an attempt to control for rearfoot to forefoot elevation that is offered by footwear. For logistical purposes, standard shoes could not be provided for all participants during data collection. Therefore, it was decided that allowing participants to use their own footwear would potentially act as a confounding variable, negatively affecting internal validity.

Landing mechanics during bilateral drop-landings have been shown to differ between shod and barefoot, with significantly greater sagittal plane knee joint displacement observed during shod conditions (Yeow et al., 2011). As such, the findings of this thesis should be interpreted with caution where landings are performed in shod. However, the relationships presented in *Chapter 5* are similar to those reported in studies that have used footwear (Dowling, McPherson and Paci, 2018; Fong et al., 2011; Malloy et al., 2015). Therefore, it is questionable if the use of footwear would have significantly affected the results reported in this thesis.

8.3 Practical applications

The findings from this thesis provide applied practitioners with reliability data that will support their interpretation of results following the assessment of individuals with restricted ankle DF ROM. When measuring ankle DF ROM, the trigonometric calculation method can be used to produce a reliable measure of tibia angle during the WBLT for a single-limb, as well as between-limb differences. Using the methods presented in this thesis, practitioners can identify interventions that successfully improve ankle DF ROM when changes in single-limb measures of WBLT performance surpass the MDC value of 1.7° presented in this thesis (see *Chapter 3*). This will allow practitioners to develop their toolbox for enhancing ankle

mobility by establishing what exercises and the associated prescription create 'real' improvements in ankle DF ROM.

Likewise, this thesis supports the use of portable force platforms and 2D video analysis for measuring kinetic and kinematic variables associated with bilateral drop-landing performance. Data from this equipment derived from the performance of bilateral drop-landings takes minutes to process and provides practitioners insight into an individual's movement skill during this task. Importantly, practitioners screening landing mechanics using bilateral drop-landings should account for drop height when interpreting the findings, as both kinetic and kinematic variables demonstrate greater variability from lower drop heights. As a consequence, it is recommended that practitioners use a drop height of 100% to 150% of maximum CMJ height when designing screening protocols for bilateral drop-landing performance.

The findings presented in this thesis suggest that irrespective of drop height, individuals with restricted ankle DF ROM will likely demonstrate compensatory coordination strategies during bilateral drop-landings so to maintain landing forces below a tolerable threshold. As a consequence of this, injury risk may increase as the strategy adopted mirrors those associated with mechanisms of injury during landings for ankle and knee ligaments (Aerts et al., 2013). Therefore, individual's performing bilateral drop-landings with increased ankle plantar flexion and reduced knee flexion at initial ground contact, reduced ankle, knee and hip flexion angles at peak flexion, and diminished sagittal plane joint displacement throughout the lower extremity, should be assessed for ankle DF ROM restriction using the WBLT. When ankle hypomobility is established, prescription of landing activities as part of a

structured training programme should be carefully considered in an attempt to minimise injury risk whilst the ankle DF ROM restriction is addressed through a corrective programme. This may mean that for individuals with restricted ankle DF ROM, landings should at least initially be performed sparingly as part of a conditioning programme.

Ankle DF ROM can be significantly improved through the prescription of corrective exercises designed to improve ankle plantar flexor flexibility and joint arthokinematics over a 4-week period. Exercise modalities shown to improve ankle DF ROM include static stretching, self-massage, eccentric strength-training and self-mobilisation. When these modalities are performed in combination with a general strength-training programme for the lower extremity and trunk musculature, chronic gains in ankle DF ROM increase ankle joint contribution during bilateral drop-landings. However, greater ankle joint contribution does not lead to increased knee or hip joint contribution and as a result, minimal effect on kinetic measures of bilateral drop-landing performance should be expected. As such, it is suggested methods that have been shown to enhance landing technique by facilitating greater knee and hip joint contribution during landings should be considered following chronic improvements in ankle DF ROM. These may include cueing specific coordination strategies such as flexing at the hip prior to initial ground contact (Blackburn and Padua, 2008), providing real-time feedback on landing performance (Myer et al., 2011) and manipulating task constraints such as arm position during landings (Masters, Johnstone and Hughes, 2016) to facilitate landing mechanics that lower peak vGRF and consequently, minimise injury risk.

Lastly, practitioners may encounter individuals with structural limitations in ankle DF ROM that will not permit improvements in ankle mobility. For example, individuals with anterior

ankle impingement secondary to an osteophyte formation on either the distal tibia or talar neck, will likely demonstrate limited ankle DF ROM that is non-modifiable with non-surgical interventions (Vaseenon and Amendola, 2012). In such instances where individuals with structural limitations in ankle DF ROM routinely perform landing tasks as part of their sport or activities of daily living, practitioners should attempt to help the performer develop strategies that allow them to successfully and safely attenuate landing forces. Similar to what was discussed in the previous paragraph, this would likely involve manipulating constraints during the practice of landings, allowing the individual to self-organise their coordination pattern to minimise injury risk. Based on the findings of this thesis, this may include seeking strategies that support greater hip joint contribution, which should be supplemented with the prescription of strength-training exercises for the hip extensor musculature to support the associated joint moments.

References

AERTS, I., CUMPS, E., VERHAGEN, E.A.L.M., VERSCHUEREN, J., and MEEUSEN, R., 2013. A systematic review of different jump-landing variables in relation to injuries. *The Journal of Sports Medicine and Physical Fitness*. 53 (5), pp. 509-519.

ARENDT, E., and DICK, R., 1995. Knee injury patterns among men and women in collegiate basketball and soccer: NCAA data and review of literature. *The American Journal of Sports Medicine*. 23 (6), pp. 694-701.

ATKINSON, G., and NEVILL, A., 1998. Statistical methods for assessing measurement error (reliability) in variables relevant to sports medicine. *Sports Medicine*. 26 (4), pp. 217-238.

AUNE, A., BISHOP, C., TURNER, A., PAPADOPOULOS, K., BUDD, S., RICHARDSON, M., and MALONEY, S., 2019. Acute and chronic effects of foam rolling vs eccentric exercise on ROM and force output of the plantar flexors. *Journal of Sports Sciences*. 37 (2), pp. 1-8.

BACKMAN, L.J., and DANIELSON, P., 2011. Low range of ankle dorsiflexion predisposes for patellar tendinopathy in junior elite basketball players: a 1-year prospective study. *The American Journal of Sports Medicine.* 39 (12), pp. 2626–2633.

BACURAU, R.F.P., MONTEIRO, G.A., UGRINOWITSCH, C., TRICOLI, V., CABRAL, L.F., and AOKI, M.S., 2009. Acute effect of a ballistic and a static stretching exercise bout on flexibility and maximal strength. *The Journal of Strength and Conditioning Research*. 23 (1), pp. 304-308.

BARRETT, E., O'KEEFFE, M., O'SULLIVAN, K., LEWIS, J., and MCCREESH, K., 2016. Is thoracic spine posture associated with shoulder pain, range of motion and function? A systematic review. *Manual Therapy*. 26, pp. 38-46.

BATES, N., FORD, K., MYER, G., and HEWETT, T., 2013. Kinetic and kinematic differences between first and second landings of a drop vertical jump task: implications for injury risk assessments. *Clinical Biomechanics*. 28 (4), pp. 459-466.

BAUMBACH, S.F., BRUMANN, M., BINDER, J., MUTSCHLER, W., REGAUER, M., and POLZER, H., 2014. The influence of knee position on ankle dorsiflexion - a biometric study. *BMC Musculoskeletal Disorders*. 15 (1), p. 246.

BEARDSLEY, C., and ŠKARABOT, J., 2015. Effects of self-myofascial release: A systematic review. *Journal of Bodywork and Movement Therapies*. 19 (4), pp. 747-758.

BEARDT, B.S., MCCOLLUM, M.R., HINSHAW, T.J., LAYER, J.S., WILSON, M.A., ZHU, Q., and DAI, B., 2018. Lower extremity kinematics differed between a controlled drop-jump and volleyball-takeoffs. *Journal of Applied Biomechanics*. 34 (4), pp. 327-335.

BEGALLE, R., WALSH, M., MCGRATH, M., BOLING, M., BLACKBURN, J., and PADUA, D., 2015. Ankle dorsiflexion displacement during landing is associated with initial contact kinematics but not joint displacement. *Journal of Applied Biomechanics*. 31 (4), pp. 205-210.

BEHM, D.G., 2018. *The science and physiology of flexibility and stretching: implications and applications in sport performance and health*. London, UK: Routledge.

BEHM, D.G., and WILKE, J., 2019. Do self-myofascial release devices release myofascia? Rolling mechanisms: a narrative review. *Sports Medicine*. 49 (8), pp. 1-9.

BELL, D.R., BLACKBURN, J.T., HACKNEY, A.C., MARSHALL, S.W., BEUTLER, A.I., and PADUA, D.A., 2014. Jump-landing biomechanics and knee-laxity change across the menstrual cycle in women with anterior cruciate ligament reconstruction. *Journal of Athletic Training*. 49 (2), pp. 154-162.

BELYEA, B.C., LEWIS, E., GABOR, Z., JACKSON, J., and KING, D.L., 2015. Validity and intrarater reliability of a 2-dimensional motion analysis using a handheld tablet compared with traditional 3-dimensional motion analysis. *Journal of Sport Rehabilitation*. 24 (4), pp. 1-7.

BENJANUVATRA, N., LAY, B.S., ALDERSON, J.A., and BLANKSBY, B.A., 2013. Comparison of ground reaction force asymmetry in one-and two-legged countermovement jumps. *The Journal of Strength and Conditioning Research*. 27 (10), pp. 2700-2707.

BENNELL, K., TALBOT, R., WAJSWELNER, H., TECHOVANICH, W., KELLY, D., and HALL, A.J., 1998. Intra-rater and inter-rater reliability of a weight-bearing lunge measure of ankle dorsiflexion. *Australian Journal of Physiotherapy*. 44 (3), pp. 175–180.

BERE, T., FLØRENES, T., KROSSHAUG, T., HAUGEN, P., SVANDAL, I., NORDSLETTEN, L., and BAHR, R., 2014. A systematic video analysis of 69 injury cases in World Cup alpine skiing. *Scandinavian Journal of Medicine and Science in Sports*. 24 (4), pp. 667-677.

BEYNNON, B.D., and FLEMING, B.C., 1998. Anterior cruciate ligament strain in-vivo: a review of previous work. *Journal of Biomechanics*. 31 (6), pp. 519-525.

BIALOSKY, J.E., BISHOP, M.D., and GEORGE, S.Z., 2008. Regional interdependence: a musculoskeletal examination model whose time has come. *The Journal of Orthopaedic and Sports Physical Therapy*. 38 (3), pp. 159-60.

BIRD, S., and MARKWICK, W., 2016. Musculoskeletal screening and functional testing: considerations for basketball athletes. *International Journal of Sports Physical Therapy*. 11 (5), pp. 784–802.

BISHOP, C., READ, P., LAKE, J., CHAVDA, S., AND TURNER, A., 2018. Interlimb asymmetries: understanding how to calculate differences from bilateral and unilateral tests. *Strength and Conditioning Journal*. 40 (4), pp. 1-6.

BISHOP, C., TURNER, A., and READ, P., 2018. Effects of inter-limb asymmetries on physical and sports performance: a systematic review. *Journal of Sports Sciences.* 36 (10), pp. 1135-1144.

BISSELING, R., HOF, A., BREDEWEG, S., ZWERVER, J., and MULDER, T., 2008. Are the take-off and landing phase dynamics of the volleyball spike jump related to patellar tendinopathy?. *British Journal of Sports Medicine*. 42 (6), pp. 483-489.

BLACKBURN, J.T., and PADUA, D.A., 2008. Influence of trunk flexion on hip and knee joint kinematics during a controlled drop landing. *Clinical Biomechanics*. 23 (3), pp. 313-319.

BLACKBURN, J., and PADUA, D., 2009. Sagittal plane trunk position, landing forces, and quadriceps electromyographic activity. *Journal of Athletic Training*. 44 (2), pp. 174-179.

BLACKWOOD, C.B., YUEN, T.J., SANGEORZAN, B.J., and LEDOUX, W.R., 2005. The midtarsal joint locking mechanism. *Foot and Ankle International*. 26 (12), pp.1074-1080.

BLAZEVICH, A.J., CANNAVAN, D., WAUGH, C.M., MILLER, S.C., THORLUND, J.B., AAGAARD, P., and KAY, A.D., 2014. Range of motion, neuromechanical, and architectural adaptations to plantar flexor stretch training in humans. *Journal of Applied Physiology*. 117 (5), pp. 452-462.

BLØND, L., and HANSEN, L., 1998. Patellofemoral pain syndrome in athletes: a 5.7-year retrospective follow-up study of 250 athletes. *Acta Orthopaedica Belgica*. 64 (4), pp. 393-400.

BLOOMFIELD, J., POLMAN, R., and O'DONOGHUE, P., 2007. Physical demands of different positions in FA Premier League soccer. *Journal of Sports Science and Medicine*. 6 (1), pp. 63-70.

BODEN, B.P., TORG, J.S., KNOWLES, S.B., and HEWETT, T.E., 2009. Video analysis of anterior cruciate ligament injury: abnormalities in hip and ankle kinematics. *The American Journal of Sports Medicine*. 37 (2), pp. 252-259.

BODEN, B.P., SHEEHAN, F.T., TORG, J.S., and HEWETT, T.E., 2010. Noncontact anterior cruciate ligament injuries: mechanisms and risk factors. *The Journal of the American Academy of Orthopaedic Surgeons*. 18 (9), pp. 520–527.

BOLING, M., and PADUA, D., 2013. Relationship between hip strength and trunk, hip, and knee kinematics during a jump-landing task in individuals with patellofemoral pain. *International Journal of Sports Physical Therapy*. 8 (5), pp. 661-669.

BOLING, M.C., PADUA, D.A., MARSHALL, S.W., GUSKIEWICZ, K., PYNE, S., and BEUTLER, A., 2009. A prospective investigation of biomechanical risk factors for

patellofemoral pain syndrome: the Joint Undertaking to Monitor and Prevent ACL Injury (JUMP-ACL) cohort. *The American Journal of Sports Medicine*. 37 (11), pp. 2108-2116.

BOLING, M.C., NGUYEN, A.D., PADUA, D.A., CAMERON, K.L., BEUTLER, A., and MARSHALL, S.W., 2019. Gender-specific risk factor profiles for patellofemoral pain. *Clinical Journal of Sport Medicine: Official Journal of the Canadian Academy of Sport Medicine*. Epub ahead of print.

BOROTIKAR, B.S., NEWCOMER, R., KOPPES, R., and MCLEAN, S.G., 2008. Combined effects of fatigue and decision making on female lower limb landing postures: central and peripheral contributions to ACL injury risk. *Clinical Biomechanics*. 23 (1), pp. 81-92.

BRADLEY, P.S., OLSEN, P.D., and PORTAS, M.D., 2007. The effect of static, ballistic, and proprioceptive neuromuscular facilitation stretching on vertical jump performance. *The Journal of Strength and Conditioning Research*. 21 (1), pp. 223-226.

BRAZEN, D.M., TODD, M.K., AMBEGAONKAR, J.P., WUNDERLICH, R., and PETERSON, C., 2010. The effect of fatigue on landing biomechanics in single-leg drop landings. *Clinical Journal of Sport Medicine*. 20 (4), pp. 286-292.

BRECHTER, H., and POWERS, C.M., 2002. Patellofemoral stress during walking in persons with and without patellofemoral pain. *Medicine and Science in Sports and Exercise*. 34 (10), pp. 1582-1593.

BRESSEL, E., and CRONIN, J., 2005. The landing phase of a jump strategies to minimize injuries. *Journal of Physical Education, Recreation and Dance*. 76 (2), pp. 30-35.

BRITTO, M.A.D, FRANCO, P.S., PAPPAS, E., and CARPES, F.P., 2015. Kinetic asymmetries between forward and drop jump landing tasks. *Revista Brasileira de Cineantropometria and Desempenho Humano.* 17 (6), pp. 661-671.

BROCKETT, C.L., and CHAPMAN, G.J., 2016. Biomechanics of the ankle. *Orthopaedics and Trauma*. 30 (3), pp. 232-238.

CARLSON, V.R., SHEEHAN, F.T., and BODEN, B.P., 2016. Video analysis of anterior cruciate ligament (ACL) injuries: a systematic review. *JBJS reviews*. 4 (11), pp. 1-12.

CESAR, G.M., PEREIRA, V.S., SANTIAGO, P.R.P., BENZE, B.G., DA COSTA, P.H.L., AMORIM, C.F., and SERRÃO, F.V., 2011. Variations in dynamic knee valgus and gluteus medius onset timing in non-athletic females related to hormonal changes during the menstrual cycle. *The Knee*. 18 (4), pp. 224-230.

CHAPPELL, J.D., YU, B., KIRKENDALL, D.T., and GARRETT, W.E., 2002. A comparison of knee kinetics between male and female recreational athletes in stop-jump tasks. *The American Journal of Sports Medicine*. 30 (2), pp. 261-267.

CHAPPELL, J., HERMAN, D., KNIGHT, B., KIRKENDALL, D., GARRETT, W., and YU, B., 2005. Effect of fatigue on knee kinetics and kinematics in stop-jump tasks. *The American Journal of Sports Medicine*. 33 (7), pp. 1022-1029.

CHEATHAM, S., and KREISWIRTH, E., 2014. The regional interdependence model: a clinical examination concept. *International Journal of Athletic Therapy and Training*. 19 (3), pp. 8-14.

CLAPPER, M.P., and WOLF, S.L., 1988. Comparison of the reliability of the Orthoranger and the standard goniometer for assessing active lower extremity range of motion. *Physical Therapy*. 68 (2), pp. 214-218.

COHEN, J., 1988. *Statistical power analysis for the behavioural sciences*. 2nd ed. Hillsdale, NJ: Lawrence Erlbaum Associates, Inc.

COLLINGS, T.J., GORMAN, A.D., STUELCKEN, M.C., MELLIFONT, D. B., and SAYERS, M.G., 2019. Exploring the justifications for selecting a drop landing task to assess injury biomechanics: A narrative review and analysis of landings performed by female netball players. *Sports Medicine*, 49 (3), pp. 385-395.

COLLINS, N., TEYS, P., and VICENZINO, B., 2004. The initial effects of a Mulligan's mobilization with movement technique on dorsiflexion and pain in subacute ankle sprains. *Manual Therapy*. 9 (2), pp. 77–82.

COOK, G., BURTON, L., HOOGENBOOM, B.J., and VOIGHT, M., 2014. Functional movement screening: the use of fundamental movements as an assessment of function - part 1. *International Journal of Sports Physical Therapy*. 9 (3), pp. 396–409.

CORMACK, S., NEWTON, R., MCGUIGAN, M., and DOYLE, T., 2008. Reliability of measures obtained during single and repeated countermovement jumps. *International Journal of Sports Physiology Performance*. 3 (2), pp. 131-144.

COSBY, N.L., and GRINDSTAFF, T.L., 2012. Restricted ankle dorsiflexion self-mobilization. *Strength and Conditioning Journal*. 34 (3), pp. 58–60.

COSBY, N.L., and HERTEL, J., 2011. Relationships between measures of posterior talar glide and ankle dorsiflexion range of motion. *Athletic Training and Sports Health Care.* 3 (2), pp. 76-85.

COVENTRY, E., O'CONNOR, K.M., HART, B.A., EARL, J.E., and EBERSOLE, K.T., 2006. The effect of lower extremity fatigue on shock attenuation during single-leg landing. *Clinical Biomechanics*. 21 (10), pp. 1090-1097.

CROSS, K.M., WORRELL, T.W., LESLIE, J.E., and VAN VELD KHALID, R., 2002. The relationship between self-reported and clinical measures and the number of days to return to sport following acute lateral ankle sprains. *Journal of Orthopaedic and Sports Physical Therapy*. 32 (1), pp. 16-23.

CROWE, M.A., BAMPOURAS, T.M., WALKER-SMALL, K., and HOWE, L.P., 2019. Restricted unilateral ankle dorsiflexion movement increases interlimb vertical force asymmetries in bilateral bodyweight squatting. *The Journal of Strength and Conditioning Research*. 34 (2), pp. 332-336.

DE BLEECKER, C., VERMEULEN, S., DE BLAISER, C., WILLEMS, T., DE RIDDER, R., and ROOSEN, P., 2020. Relationship between jump-landing kinematics and lower extremity overuse injuries in physically active populations: a systematic review and meta-analysis. *Sports Medicine.* Epub ahead of print.

DE BOER, M.R., WATERLANDER, W.E., KUIJPER, L.D., STEENHUIS, I.H., and TWISK, J.W., 2015. Testing for baseline differences in randomized controlled trials: an unhealthy research behavior that is hard to eradicate. *International Journal of Behavioural Nutrition and Physical Activity.* 12 (1), pp. 4.

DE OLIVEIRA, F.C.L., FREDETTE, A., ECHEVERRÍA, S.O., BATCHO, C.S., and ROY, J.S., 2019. Validity and reliability of 2-Dimensional video-based assessment to analyze foot strike pattern and step rate during running: a systematic review. *Sports Health*. 11 (5), pp. 409-415.

DE SOUZA, A., SANCHOTENE, C.G., LOPES, C.M.D.S., BECK, J. A., DA SILVA, A.C.K., PEREIRA, S.M., and RUSCHEL, C., 2019. Acute effect of 2 self-myofascial release protocols on hip and ankle range of motion. *Journal of Sport Rehabilitation*. 28 (2), pp. 159-164.

DECKER, M.J., TORRY, M.R., NOONAN, T.J., RIVIERE, A.M.Y., and STERETT, W.I., 2002. Landing adaptations after ACL reconstruction. *Medicine and Science in Sports and Exercise*. 34 (9), pp. 1408-1413.

DECKER, M.J., TORRY, M.R., WYLAND, D.J., STERETT, W.I., and STEADMAN, J.R., 2003. Gender differences in lower extremity kinematics, kinetics and energy absorption during landing. *Clinical Biomechanics*. 18 (7), pp. 662-669.

DELAHUNT, E., CUSACK, K., WILSON, L., and DOHERTY, C., 2013. Joint mobilization acutely improves landing kinematics in chronic ankle instability. *Medicine and Science in Sports and Exercise*. 45 (3), pp. 514-519.

DEMORAT, G., WEINHOLD, P., BLACKBURN, T., CHUDIK, S., and GARRETT, W., 2004. Aggressive quadriceps loading can induce noncontact anterior cruciate ligament injury. *The American Journal of Sports Medicine*. 32 (2), pp. 477-483.

DENEGAR, C.R., HERTEL, J., and FONSECA, J., 2002. The effect of lateral ankle sprain on dorsiflexion range of motion, posterior talar glide, and joint laxity. *Journal of Orthopaedic and Sports Physical Therapy*. 32 (4), pp. 166-173.

DEVITA, P., and SKELLY, W.A., 1992. Effect of landing stiffness on joint kinetics and energetics in the lower extremity. *Medicine and Science in Sports and Exercise*. 24 (1), pp. 108-115.

DIAMOND, J.E., MUELLER, M.J., DELITTO, A., and SINACORE, D.R., 1989. Reliability of a diabetic foot evaluation. *Physical Therapy*. 69 (10), pp. 797-802.

DIERKS, T. A., MANAL, K. T., HAMILL, J., and DAVIS, I., 2011. Lower extremity kinematics in runners with patellofemoral pain during a prolonged run. *Medicine and Science in Sports and Exercise*. 43 (4), pp. 693–700.

DILL, K.E., BEGALLE, R.L., FRANK, B.S., ZINDER, S.M., and PADUA, D.A., 2014. Altered knee and ankle kinematics during squatting in those with limited weight-bearing–lunge ankle-dorsiflexion range of motion. *Journal of Athletic Training*. 49 (6), pp. 723-732.

DINGENEN, B., MALFAIT, B., VANRENTERGHEM, J., ROBINSON, M., VERSCHUEREN, S., and STAES, F., 2015. Can two-dimensional measured peak sagittal plane excursions during drop vertical jumps help identify three-dimensional measured joint moments?. *The Knee*. 22 (2), pp. 73-79.

DINGENEN, B., MALFAIT, B., VANRENTERGHEM, J., VERSCHUEREN, S., and STAES, F., 2014. The reliability and validity of the measurement of lateral trunk motion in two-dimensional video analysis during unipodal functional screening tests in elite female athletes. *Physical Therapy in Sport*. 15 (2), pp. 117-123.

DRAGANICH, L.F., and VAHEY, J.W., 1990. An in vitro study of anterior cruciate ligament strain induced by quadriceps and hamstrings forces. *Journal of Orthopaedic Research*. 8 (1), pp. 57-63.

DUFEK, J., and BATES, B., 1991. Biomechanical factors associated with injury during landing in jump sports. *Sports Medicine*. 12 (5), pp. 326-337.

EDMOND, S.L., 2016. *Joint mobilization/manipulation*. [eBook] 3rd ed. Elsevier Health Sciences. Available from: https://www.elsevier.com/books/joint-mobilization-manipulation/edmond/978-0-323-29469-0 [Accessed 10 June 2019].

EDWARDS, S., STEELE, J., and MCGHEE, D., 2010. Does a drop-landing represent a whole skill landing and is this moderated by fatigue?. *Scandinavian Journal of Medicine and Science in Sports*. 20 (3), pp. 516-523.

EDWARDS, S., STEELE, J.R., COOK, J.L., PURDAM, C.R., and MCGHEE, D.E., 2012. Lower limb movement *symmetry cannot be assumed when investigating the stop-jump landing. Medicine and Science in Sports and Exercise*. 44 (6). pp. 1123-1130.

EDWARDS, S., STEELE, J.R., PURDAM, C.R., COOK, J.L., and MCGHEE, D.E., 2014. Alterations to landing technique and patellar tendon loading in response to fatigue. *Medicine and Science in Sports and Exercise*. 46 (2), pp. 330-340.

EDWARDS, S., STEELE, J.R., MCGHEE, D.E., PURDAM, C.R., and COOK, J.L., 2017. Asymptomatic players with a patellar tendon abnormality do not adapt their landing mechanics when fatigued. *Journal of Sports Sciences*. 35 (8), pp. 769-776.

FARROKHI, S., KEYAK, J.H., and POWERS, C.M., 2011. Individuals with patellofemoral pain exhibit greater patellofemoral joint stress: a finite element analysis study. *Osteoarthritis and Cartilage*. 19 (3), pp. 287-294.

FERRAN, N.A., and MAFFULLI, N., 2006. Epidemiology of sprains of the lateral ankle ligament complex. *Foot and Ankle Clinics*. 11 (3), pp. 659–662.

FISHER, H., STEPHENSON, M.L., GRAVES, K.K., HINSHAW, T.J., SMITH, D.T., ZHU, Q., WILSON, M.A., and DAI, B., 2016. Relationship between force production during isometric squats and knee flexion angles during landing. *The Journal of Strength and Conditioning Research*. 30 (6), pp. 1670-1679.

FLEMING, B.C., RENSTROM, P.A., OHLEN, G., JOHNSON, R.J., PEURA, G.D., BEYNNON, B.D., and BADGER, G.J., 2001. The gastrocnemius muscle is an antagonist of the anterior cruciate ligament. *Journal of Orthopaedic Research*. 19 (6), pp. 1178-1184.

FONG, C., BLACKBURN, J., NOCROSS, M., MCGRATH, M., and PADUA, D. 2011. Ankle-dorsiflexion range of motion and landing biomechanics. *Journal of Athletic Training*. 46 (1), pp. 5-10.

FONG, D.T.P., HONG, Y., CHAN, L.K., YUNG, P.S.H., and CHAN, K.M., 2007. A systematic review on ankle injury and ankle sprain in sports. *Sports Medicine*. 37 (1), pp. 73-94.

FOUSEKIS, K., TSEPIS, E., and VAGENAS, G. 2012. Intrinsic risk factors of noncontact ankle sprains in soccer: a prospective study on 100 professional players. *The American Journal of Sports Medicine*. 40 (8), pp.1842-1850.

FRY, A.C., SMITH, J.C., and SCHILLING, B.K. 2003. Effect of knee position on hip and knee torques during the barbell squat. *The Journal of Strength and Conditioning Research*. 17 (4), pp. 629-633.

FUJII, M., SUZUKI, D., UCHIYAMA, E., MURAKI, T., TERAMOTO, A., AOKI, M., and MIYAMOTO, S. 2010. Does distal tibiofibular joint mobilization decrease limitation of ankle dorsiflexion?. *Manual therapy*. 15 (1), pp. 117-121.

GABBE, B.J., BENNELL, K.L., FINCH, C.F., WAJSWELNER, H., and ORCHARD, J.W., 2006. Predictors of hamstring injury at the elite level of Australian football. *Scandinavian Journal of Medicine and Science in Sports*. 16 (1), pp. 7-13.

GATHERCOLE, R., SPORER, B., STELLINGWERFF, T., and SLEIVERT, G., 2015. Alternative countermovement-jump analysis to quantify acute neuromuscular fatigue. *International Journal of Sports Physiology and Performance*. 10 (1), pp. 84-92.

GONZALO-SKOK, O., SERNA, J., RHEA, M.R., and MARÍN, P.J., 2015. Relationships between functional movement tests and performance tests in young elite male basketball players. *International Journal of Sports Physical Therapy*. *10* (5), pp. 628-638.

GREEN T, REFSHAUGE K, CROSBIE J, and ADAMS R., 2001. A randomized controlled trial of a passive accessory joint mobilization on acute ankle inversion sprains. *Physical Therapy*. 81 (4), pp. 984–994.

GRIFFIN, L.Y., AGEL, J., ALBOHM, M.J., ARENDT, E.A., DICK, R.W., GARRETT, W.E., GARRICK, J.G., HEWETT, T.E., HUSTON, L., IRELAND, M.L., JOHNSON, R.J., KIBLER, W.B., LEPHART, S., LEWIS, J.L., LINDENFELD, T.N., MANDELBAUM, B.R., MARCHAK, P., TEITZ, C.C., and WOJTYS, E.M., 2000. Noncontact anterior cruciate

ligament injuries: risk factors and prevention strategies. *JAAOS-Journal of the American Academy of Orthopaedic Surgeons*. 8 (3), 141–150.

GUO, L.Y., YANG, C.H., TSAO, H., WANG, C.Y., and LIANG, C.C., 2006. Initial effects of the ankle dorsiflexion mobilization with movement on ankle range of motion and limb coordination in young healthy subjects. *物理治療*. 31 (3), pp.173-181.

HADDAS, R., JAMES, C.R., and HOOPER, T.L., 2015. Lower extremity fatigue, sex, and landing performance in a population with recurrent low back pain. *Journal of Athletic Training*. 50 (4), pp. 378-384.

HAKKINEN K., and KOMI, P.V., 1986. Effects of fatigue and recovery on electromyographic and isometric force- and relaxation-time characteristics of human skeletal muscle. *European Journal of Applied Physiology and Occupational Physiology*. 55 (6), pp. 588-596.

HALL, E. A., and DOCHERTY, C.L., 2017. Validity of clinical outcome measures to evaluate ankle range of motion during the weight-bearing lunge test. *Journal of Science and Medicine in Sport*. 20 (7), pp. 618-621.

HALPERIN I., ABOODARDA S., BUTTON, D., ANDERSEN, L., and BEHM, D., 2014. Roller massager improves range of motion of plantar flexor muscles without subsequent decreases in force parameters. *International Journal of Sports Physical Therapy*. 9 (1), pp. 92-102.

HARRY, J., SILVERNAIL, J., MERCER, J., and DUFEK, J., 2018. Bilateral comparison of vertical jump landings and step-off landings from equal heights. *The Journal of Strength Conditioning Research*. 32 (7), pp. 1937-1947.

HARRY, J., SILVERNAIL, J., MERCER, J., and DUFEK, J., 2017. Comparison of pre-contact joint kinematics and vertical impulse between vertical jump landings and step-off landings from equal heights. *Human Movement Science*. 56, pp. 88-97.

HERRINGTON, L., 2014. Knee valgus angle during single leg squat and landing in patellofemoral pain patients and controls. *The Knee*. 21 (2), pp. 514-517.

HERRINGTON, L., 2014. The effect of hip abductor muscle fatigue on frontal plane knee projection angle during step landing. *International Journal of Athletic Therapy and Training*. 19 (4), pp. 38-43.

HERRINGTON, L., MUNRO, A., and COMFORT, P., 2015. A preliminary study into the effect of jumping–landing training and strength-training on frontal plane projection angle. *Manual Therapy*. 20 (5), pp. 680-685.

HERTEL, J., 2002. Functional anatomy, pathomechanics, and pathophysiology of lateral ankle instability. *Journal of Athletic Training*. 37 (4), pp. 364-375.

HERTEL, J., BRAHAM, R.A., HALE, S.A., and OLMSTED-KRAMER, L.C. (2006). Simplifying the star excursion balance test: analyses of subjects with and without chronic ankle instability. *Journal of Orthopaedic and Sports Physical Therapy*. 36 (3), pp. 131-137.

HERZOG, W., and READ, L.J., 1993. Lines of action and moment arms of the major force-carrying structures crossing the human knee joint. *Journal of Anatomy*. 182 (2), pp. 213-230.

HEWETT, T., STROUPE, A., NANCE, T., and NOYES, F., 1996. Plyometric training in female athletes. Decreased impact forces and hamstring torques. *The American Journal of Sports Medicine*. 24 (6), pp. 765–73.

HEWETT, T., MYER, G., FORD, K., HEIDT JR, R., COLOSIMO, A., MCLEAN, S., VAN DEN BOGERT, A., PATERNO, M., and SUCCOP, P. 2005. Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study. *The American Journal of Sports Medicine*. 33 (4), pp. 492-501.

HEWETT, T., MYER, G., and FORD, K., 2006. Anterior cruciate ligament injuries in female athletes: Part 1, mechanisms and risk factors. *The American Journal of Sports Medicine.* 34 (2), pp. 299-311.

HEWETT, T.E., FORD, K.R., HOOGENBOOM, B.J., and MYER, G.D., 2010. Understanding and preventing acl injuries: current biomechanical and epidemiologic considerations-update 2010. *North American Journal of Sports Physical Therapy*. 5 (4), pp. 234-251.

HOCH, M.C., and MCKEON, P.O., 2011. Normative range of weight-bearing lunge test performance asymmetry in healthy adults. *Manual Therapy.* 16 (5), pp. 516-519.

HOCH, M.C., STATON, G.S., MCKEON, J.M.M., MATTACOLA, C.G., and MCKEON, P.O., 2012. Dorsiflexion and dynamic postural control deficits are present in those with chronic ankle instability. *Journal of Science and Medicine in Sport*. 15 (6), pp. 574-579.

HOCH, M., FARWELL, K., GAVEN, S., and WEINHANDL, J., 2015. Weight-bearing dorsiflexion range of motion and landing biomechanics in individuals with chronic ankle instability. *Journal of Athletic Training.* 50 (8), pp. 833-839.

HOLDEN, S., BOREHAM, C., DOHERTY, C., and DELAHUNT, E., 2017. Two-dimensional knee valgus displacement as a predictor of patellofemoral pain in adolescent females. *Scandinavian Journal of Medicine and Science in Sports*. 27 (2), pp. 188-194.

HOLLMAN, J.H., GALARDI, C.M., LIN, I.H., VOTH, B.C., and WHITMARSH, C.L., 2014. Frontal and transverse plane hip kinematics and gluteus maximus recruitment correlate with frontal plane knee kinematics during single-leg squat tests in women. *Clinical Biomechanics*. 29 (4), pp. 468-474.

HOLLMAN, J.H., HOHL, J.M., KRAFT, J.L., STRAUSS, J.D., and TRAVER, K.J., 2012. Effects of hip extensor fatigue on lower extremity kinematics during a jump-landing task in women: a controlled laboratory study. *Clinical Biomechanics*. 27 (9), pp. 903-909.

HOPKINS, W.G, 2015. Spreadsheets for analysis of validity and reliability [online]. Available from: http://www.sportsci.org/resource/stats/index.html [Accessed January 21 2018].

HOPKINS, W.G., 2016. Precision of measurement [online]. Accessed from: http://sportsci.org/resource/stats [Accessed June 20 2018].

HORI, N., NEWTON, R., KAWAMORI, N., MCGUIGAN, M., KRAEMER, W., and NOSAKA, K., 2009. Reliability of performance measurements derived from ground reaction force data during countermovement jump and the influence of sampling frequency. *The Journal of Strength Conditioning Research*. 23 (3). pp. 874-882.

HOWARD, J.S., FAZIO, M.A., MATTACOLA, C.G., UHL, T.L., and JACOBS, C.A., 2011. Structure, sex, and strength and knee and hip kinematics during landing. *Journal of Athletic Training*. 46 (4), pp. 376-386.

HOWARD, M., CAWLEY, P., and LOSSE, G. 1998. Effect of lower extremity muscular fatigue on motor control performance. *Medicine and Science in Sports and Exercise*. 30 (12), pp.1703-1707.

HOWE, L.P., 2017. The acute effects of ankle mobilisations on lower extremity joint kinematics. *Journal of Bodywork and Movement Therapies*. 21 (4), pp. 775-780.

HOWE, L.P., and CUSHION, E., 2017. A problem-solving process to identify the origins of poor movement. *Professional Strength and Conditioning Journal*. 45, pp. 7-15.

HUBBARD, T.J., and HERTEL, J., 2008. Anterior positional fault of the fibula after sub-acute lateral ankle sprains. *Manual Therapy*. 13 (1), pp. 63-67.

JAMES, C., HERMAN, J., DUFEK, J., and BATES, B., 2007. Number of trials necessary to achieve performance stability of selected ground reaction force variables during landing. *Journal of Sports Science and Medicine*. 6 (1), pp. 126-134.

JAMES, C., SCHEUERMANN, B., and SMITH, M., 2010. Effects of two neuromuscular fatigue protocols on landing performance. *Journal of Electromyography and Kinesiology*. 20 (4), pp. 667-675.

JEON, I., KWON, O., YI, C., CYNN, H., and HWANG, U., 2015. Ankle-dorsiflexion range of motion after ankle self-stretching using a strap. *Journal of Athletic Training*. 50 (12), pp. 1226-1232.

JOHANSON, M.A., WOODEN, M., CATLIN, P.A., HEMARD, L., LOTT, K., ROMALINO, R., and STILLMAN, T., 2006. Effects of gastrocnemius stretching on ankle dorsiflexion and time-to heel-off during the stance phase of gait. *Physical Therapy in Sport*. 7 (2), pp. 93-100.

JOHANSON, M., BAER, J., HOVERMALE, H., and PHOUTHAVONG, P. (2008). Subtalar joint position during gastrocnemius stretching and ankle dorsiflexion range of motion. *Journal of Athletic Training*. 43 (2), pp. 172-178.

JOHNSTON, W., DOLAN, K., REID, N., COUGHLAN, G., and CAULFIELD, B., 2018. Investigating the effects of maximal anaerobic fatigue on dynamic postural control using the Y-Balance Test. *Journal of Science and Medicine in Sport*. 21 (1), pp. 103.108.

JORDAN, M.J., AAGAARD, P., and HERZOG, W., 2015. Lower limb asymmetry in mechanical muscle function: a comparison between ski racers with and without ACL reconstruction. *Scandinavian Journal of Medicine and Science in Sports*. 25 (3), pp. 301-309.

JUNKER, D.H., and STÖGGL, T.L., 2015. The foam roll as a tool to improve hamstring flexibility. *The Journal of Strength and Conditioning Research*. 29 (12), pp. 3480-3485.

KASUYAMA, T., SAKAMOTO, M., and NAKAZAWA, R., 2009. Ankle joint dorsiflexion measurement using the deep squatting posture. *Journal of Physical Therapy Science*. 21 (2), pp.195–199.

KERNOZEK, T.W., TORRY, M.R., VAN HOOF, H., COWLEY, H., and TANNER, S., 2005. Gender differences in frontal and sagittal plane biomechanics during drop landings. *Medicine and Science in Sports and Exercise*. 37 (6), pp. 1003-1012.

KIAPOUR, A.M., KIAPOUR, A., GOEL, V.K., QUATMAN, C.E., WORDEMAN, S.C., HEWETT, T.E., and DEMETROPOULOS, C.K., 2015. Uni-directional coupling between tibiofemoral *frontal and axial plane rotation supports valgus collapse mechanism of ACL injury. Journal of Biomechanics*. 48 (10), pp. 1745-1751.

KIM, S.H., KWON, O.Y., PARK, K.N., JEON, I.C., and WEON, J.H., 2015. Lower extremity strength and the range of motion in relation to squat depth. *Journal of Human Kinetics*. 45 (1), pp. 59-69.

KING, D. and BELYEA, B., 2015. Reliability of using a handheld tablet and application to measure lower extremity alignment angles. *Journal of Sport Rehabilitation*. 24 (4), pp. 2014-0195.

KNAPIK, J.J., CRAIG, S.C., HAURET, K.G., and JONES, B.H., 2003. Risk factors for injuries during military parachuting. *Aviation, Space, and Environmental Medicine.* 74 (7), pp. 768-774.

KONDO, H., and SOMEYA, F., 2016. Changes in ground reaction force during a rebound-jump task after hip strength-training for single-sided ankle dorsiflexion restriction. *Journal of Physical Therapy Science*. 28 (2), pp. 319-325.

KONOR, M.M., MORTON, S., ECKERSON, J.M., and GRINDSTAFF, T.L., 2012. Reliability of three measures of ankle dorsiflexion range of motion. *International Journal of Sports Physical Therapy*. 7 (3), pp. 279-287.

KOSIK, K.B., and GRIBBLE, P.A., 2016. The effect of joint mobilization on dynamic postural control in patients with chronic ankle instability: A critically appraised topic. *Journal of Sport Rehabilitation*. 27 (1), pp. 103-108.

KOVÁCS, I., TIHANYI, J., DEVITA, P., RÁCZ, L., BARRIER, J., and HORTOBÁGYI, T., 1999. Foot placement modifies kinematics and kinetics during drop jumping. *Medicine and Science in Sports and Exercise*. 31 (5), pp. 708-716.

KRAUSE, D.A., BOYD, M.S., HAGER, A.N., SMOYER, E.C., THOMPSON, A.T., and HOLLMAN, J.H., 2015. Reliability and accuracy of a goniometer mobile device application for video measurement of the functional movement screen deep squat test. *International Journal of Sports Physical Therapy. 10* (1), pp. 37-44.

KROSSHAUG, T., NAKAMAE, A., BODEN, B.P., ENGEBRETSEN, L., SMITH, G., SLAUTERBECK, J.R., HEWETT, T.E., and BAHR, R., 2007. Mechanisms of anterior cruciate ligament injury in basketball: video analysis of 39 cases. *The American Journal of Sports Medicine. 35* (3), pp. 359-367.

LANGARIKA-ROCAFORT, A., EMPARANZA, J.I., ARAMENDI, J.F., CASTELLANO, J., and CALLEJA- GONZÁLEZ, J., 2017. Intra-rater reliability and agreement of various methods of measurement to assess dorsiflexion in the Weight Bearing Dorsiflexion Lunge Test (WBLT) among female athletes. *Physical Therapy in Sport. 23*, pp. 37-44.

LAUGHLIN, W.A., WEINHANDL, J.T., KERNOZEK, T.W., COBB, S.C., KEENAN, K.G., and O'CONNOR, K.M., 2011. The effects of single-leg landing technique on ACL loading. *Journal of Biomechanics*. 44 (10), pp. 1845-1851.

LEE, T.Q., MORRIS, G., and CSINTALAN, R.P., 2003. The influence of tibial and femoral rotation on patellofemoral contact area and pressure. *Journal of Orthopaedic and Sports Physical Therapy*. 33 (11), pp. 686-693.

LEPHART, S.M., SMOLIGA, J.M., MYERS, J.B., SELL, T.C., and TSAI, Y.S., 2007. An eight-week golf-specific exercise program improves physical characteristics, swing mechanics, and golf performance in recreational golfers. *The Journal of Strength and Conditioning Research*. 21 (3), pp. 860-869.

LEPORACE, G., TANNURE, M., ZEITOUNE, G., METSAVAHT, L., MAROCOLO, M., and SOUTO MAIOR, A., 2018. Association between knee-to-hip flexion ratio during single-leg vertical landings, and strength and range of motion in professional soccer players. *Sports Biomechanics*. 12, 1-10.

LEPPÄNEN, M., PASANEN, K., KUJALA, U.M., VASANKARI, T., KANNUS, P., ÄYRÄMÖ, S., KROSSHAUG, T., BAHR, R., AVELA, J., PERTTUNEN, J., and PARKKARI, J., 2016. Stiff landings are associated with increased ACL injury risk in young female basketball and floorball players. *The American Journal of Sports Medicine*. 45 (2), pp. 386-393.

LEPPÄNEN, M., PASANEN, K., KROSSHAUG, T., KANNUS, P., VASANKARI, T., and PARKKARI, J., 2017. Landing with less hip flexion is associated with increased risk of ACL injuries in young female team sports players. *British Journal of Sports Medicine*. 51 (4), pp. 350-350.

LERSCH, C., GRÖTSCH, A., SEGESSER, B., KOEBKE, J., BRÜGGEMANN, G.P., and POTTHAST, W., 2012. Influence of calcaneus angle and muscle forces on strain distribution in the human Achilles tendon. *Clinical Biomechanics*. 27 (9), pp. 955-961.

LEWIS, G.S., KIRBY, K.A., and PIAZZA, S.J., 2007. Determination of subtalar joint axis location by restriction of talocrural joint motion. *Gait and Posture*. 25 (1), pp. 63-69.

LIAN, O., ENGEBRETSEN, L., OVREBO, R., and BAHR, R., 1996. Characteristics of the leg extensors in male volleyball players with jumper's knee. *American Journal of Sports Medicine*. 24 (3), pp. 380-385.

LIAO, T.C., YANG, N., HO, K.Y., FARROKHI, S., and POWERS, C., 2015. Femur rotation increases patella cartilage stress in females with patellofemoral pain. *Medicine and Science in Sports and Exercise*. 47 (9), pp. 1775-1780.

LIMA, Y.L., DE PAULA LIMA, P.O., BEZERRA, M.A., DE OLIVEIRA, R.R., and ALMEIDA, G.P.L., 2018. The association of ankle dorsiflexion range of motion and dynamic knee valgus: A systematic review with meta-analysis. *Physical Therapy in Sport.* 29, pp. 61-69.

LINDENBERG, K.M., and CARCIA, C.R., 2013. The influence of heel height on vertical ground reaction force during landing tasks in recreationally active and athletic collegiate females. *International Journal of Sports Physical Therapy*. 8 (1), pp. 1-8.

LIST, R., GÜLAY, T., STOOP, M., and LORENZETTI, S., 2013. Kinematics of the trunk and the lower extremities during restricted and unrestricted squats. *The Journal of Strength and Conditioning Research*. 27 (6), pp. 1529-1538.

LOPES, T.J.A., FERRARI, D., IOANNIDIS, J., SIMIC, M., MÍCOLIS DE AZEVEDO, F. and PAPPAS, E., 2018. Reliability and validity of frontal plane kinematics of the trunk and lower extremity measured with 2-dimensional cameras during athletic tasks: A systematic review with meta-analysis. *Journal of Orthopaedic and Sports Physical Therapy*. 48 (10), pp. 812-822.

LORIMER, A.V., and HUME, P.A., 2014. Achilles tendon injury risk factors associated with running. *Sports Medicine*. 44 (10), pp. 1459-1472.

LOUDON, J.K., and BELL, S.L., 1996. The foot and ankle: an overview of arthrokinematics and selected joint techniques. *Journal of Athletic Training*. 31 (2), pp. 173-178.

LUNDGREN, L.E., TRAN, T.T., NIMPHIUS, S., RAYMOND, E., SECOMB, J.L., FARLEY, O.R., NEWTON, R.U., and SHEPPARD, J.M., 2016. Comparison of impact forces, accelerations and ankle range of motion in surfing-related landing tasks. *Journal of Sports Sciences*. 34 (11), pp. 1051-1057.

MACRUM, E., BELL, D.R., BOLING, M., LEWEK, M., and PADUA, D., 2012. Effect of limiting ankle-dorsiflexion range of motion on lower extremity kinematics and muscle-activation patterns during a squat. *Journal of Sport Rehabilitation*. 21 (2), pp. 144-150.

MADIGAN, M., and PIDCOE, P., 2003. Changes in landing biomechanics during a fatiguing landing activity. *Journal of Electromyography and Kinesiology*. 13 (5), pp. 491-498.

MAFFIULETTI, N., JUBEAU, M., MUNZINGER, U., BIZZINI, M., AGOSTI, F., DE COL, A., LAFORTUNA, C., and SARTORIO, A., 2007. Differences in quadriceps muscle strength and fatigue between lean and obese subjects. *European Journal of Applied Physiology*. 101 (1), pp. 51-59.

MAHIEU, N.N., MCNAIR, P., COOLS, A.N.N., D'HAEN, C., VANDERMEULEN, K., and WITVROUW, E., 2008. Effect of eccentric training on the plantar flexor muscle-tendon tissue properties. *Medicine and Science in Sports and Exercise*. 40 (1), pp. 117-123.

MALLOY, P., MORGAN, A., MEINERZ, C., GEISER, C., and KIPP, K., 2015. The association of dorsiflexion flexibility on knee kinematics and kinetics during a drop vertical jump in healthy female athletes. *Knee Surgery, Sports Traumatology, Arthroscopy*. 23 (12), pp. 3550-3555.

MARKOLF, K.L., BURCHFIELD, D.M., SHAPIRO, M.M., SHEPARD, M.F., FINERMAN, G.A., and SLAUTERBECK, J.L., 1995. Combined knee loading states that

generate high anterior cruciate ligament forces. *Journal of Orthopaedic Research*. 13 (6), pp. 930-935.

MARKOVIC, G., DIZDAR, D., JUKIC, I., and CARDINALE, M., 2004. Reliability and factorial validity of squat and countermovement jump tests. *The Journal of Strength Conditioning Research*. 18 (3), pp. 551-555.

MARQUES, A.P., VASCONCELOS, A.A.P., CABRAL, C.M.N., and SACCO, I.C.N., 2009. Effect of frequency of static stretching on flexibility, hamstring tightness and electromyographic activity. *Brazilian Journal of Medical and Biological Research*. 42 (10), pp. 949-953.

MARTIN, R.L., and MCPOIL, T.G., 2005. Reliability of ankle goniometric measurements: a literature review. *Journal of the American Podiatric Medical Association.* 95 (6), pp. 564-572.

MASON-MACKAY, A., WHATMAN, C. and REID, D., 2017. The effect of reduced ankle dorsiflexion on lower extremity mechanics during landing: A systematic review. *Journal of Science and Medicine in Sport*. 20 (5), pp. 451-458.

MASTERS, C., JOHNSTONE, J., and HUGHES, G., 2016. The effect of arm position on lower extremity kinematics during a single limb drop landing: a preliminary study. *Journal of Functional Morphology and Kinesiology*. 1 (3), pp. 282-288.

MATÉ-MUÑOZ, J.L., LOUGEDO, J.H., BARBA, M., GARCÍA-FERNÁNDEZ, P., GARNACHO-CASTAÑO, M.V., and DOMÍNGUEZ, R., 2017. Muscular fatigue in response to different modalities of CrossFit sessions. *PLOS ONE*. 12 (7), e0181855.

MATTACOLA, C.G., and DWYER, M.K., 2002. Rehabilitation of the ankle after acute sprain or chronic instability. *Journal of Athletic Training*. 37 (4), pp. 413–429.

MAUNTEL, T.C., BEGALLE, R.L., CRAM, T.R., FRANK, B.S., HIRTH, C.J., BLACKBURN, T., and PADUA, D.A., 2013. The effects of lower extremity muscle activation and passive range of motion on single leg squat performance. *The Journal of Strength and Conditioning Research*. 27 (7), pp. 1813-1823.

MCCLAY, I., ROBINSON, J., ANDRIACCHI, T., FREDERIC, E., GROSS, T., MARIN, P., VALIANT, G., WILLIAMS, K., and CAVANAGH, P., 1994. A profile if ground reaction forces in professional basketball. *Journal of Applied Biomechanics*. 10 (3), pp. 222-236.

MCLEAN, S., FELLIN, R., SUEDEKUM, N., CALABRESE, G., PASSERALLO, A., and JOY, S., 2007. Impact of fatigue on gender-based high-risk landing strategies. *Medicine and Science in Sports and Exercise*. 39 (3), pp.502-514.

MCLEAN, S.G., HUANG, X., and VAN DEN BOGERT, A.J., 2005. Association between lower extremity posture at contact and peak knee valgus moment during sidestepping: implications for ACL injury. *Clinical Biomechanics*. 20 (8), pp. 863-870.

MCLEAN, S., WALKER, K., FORD, K., MYER, G., HEWETT, T., and VAN DEN BOGERT, A., 2005. Evaluation of a two dimensional analysis method as a screening and evaluation tool for anterior cruciate ligament injury. *British Journal of Sports Medicine.* 39 (6), pp. 355-362.

MCNEAL, J., SANDS, W., and STONE, M., 2010. Effects of fatigue on kinetic and kinematic variables during a 60-second repeated jumps test. *International Journal of Sports Physiology and Performance*. 5 (2), pp. 218-229.

MCNITT-GRAY, J.L., 1991. Kinematics and impulse characteristics of drop-landings from three heights. *Journal of Applied Biomechanics*. 7 (2), pp. 201-224.

MECAGNI, C., SMITH, J.P., ROBERTS, K.E., and O'SULLIVAN, S.B., 2000. Balance and ankle range of motion in community-dwelling women aged 64 to 87 years: a correlational study. *Physical Therapy*. 80 (10), pp. 1004-1011.

MEDEIROS, D.M., and MARTINI, T.F., 2018. Chronic effect of different types of stretching on ankle dorsiflexion range of motion: systematic review and meta-analysis. *The Foot*. 34, 28-35.

MEYER, C., GETTE, P., MOUTON, C., SEIL, R., and THEISEN, D., 2018. Side-to-side asymmetries in landing mechanics from a drop vertical jump test are not related to asymmetries in knee joint laxity following anterior cruciate ligament reconstruction. *Knee Surgery, Sports Traumatology, Arthroscopy.* 26 (2), pp. 381-390.

MEYER, E.G., and HAUT, R.C., 2005. Excessive compression of the human tibio-femoral joint causes ACL rupture. *Journal of Biomechanics*. 38 (11), pp. 2311-2316.

MILNER, C.E., WESTLAKE, C.G., and TATE, J.J., 2011. Test–retest reliability of knee biomechanics during stop jump landings. *Journal of Biomechanics,* 44 (9), pp.1814-1816.

MILNER, C.E., FAIRBROTHER, J.T., SRIVATSAN, A., and ZHANG, S., 2012. Simple verbal instruction improves knee biomechanics during landing in female athletes. *The Knee.* 19 (4), pp. 399-403.

MIZNER, R., CHMIELEWSKI, T., TOEPKE, J., and TOFTE, K., 2012. Comparison of two-dimensional measurement techniques for predicting knee angle and moment during a drop vertical jump. *Clinical Journal of Sport Medicine.* 22 (3), pp. 221-227.

MOHR, A.R., LONG, B.C., and GOAD, C.L., 2014. Effect of foam rolling and static stretching on passive hip-flexion range of motion. *Journal of Sport Rehabilitation*. 23 (4), pp. 296-299.

MOIR, G.L., 2008. Three different methods of calculating vertical jump height from force platform data in men and women. *Measurement in Physical Education and Exercise Science*. 12 (4), pp. 207-218.

MORENO-PÉREZ, V., SOLER, A., ANSA, A., LÓPEZ-SAMANES, Á., MADRUGA-PARERA, M., BEATO, M., and ROMERO-RODRÍGUEZ, D., 2019. Acute and chronic effects of competition on ankle dorsiflexion ROM in professional football players. *European Journal of Sport Science*. 20 (1), pp. 51-60.

MORESIDE, J.M., and MCGILL, S.M., 2013. Improvements in hip flexibility do not transfer to mobility in functional movement patterns. *The Journal of Strength and Conditioning Research*. 27 (10), pp. 2635-2643.

MOSELEY, A., and ADAMS, R., 1991. Measurement of passive ankle dorsiflexion: procedure and reliability. *Australian Journal of Physiotherapy*. 37 (3), pp. 175-181.

MOUSAVI, S.H., HIJMANS, J.M., MOEINI, F., RAJABI, R., FERBER, R., VAN DER WORP, H., and ZWERVER, J., 2020. Validity and reliability of a smartphone motion analysis app for lower limb kinematics during treadmill running. *Physical Therapy in Sport*. 43, pp. 27-35.

MULLIGAN, B.R., 1993. Mobilisations with movement (MWM's). *Journal of Manual and Manipulative Therapy*. 1 (4), pp. 154–156.

MUNRO, A., HERRINGTON, L., and CAROLAN, M., 2012. Reliability of 2-dimensional video assessment of frontal plane dynamic knee valgus during common athletic screening tasks. *Journal of Sport Rehabilitation*. 21 (1), pp. 7-11.

MYER, G.D., FORD, K.R., BRENT, J.L., and HEWETT, T.E., 2007. Differential neuromuscular training effects on ACL injury risk factors in" high-risk" versus" low-risk" athletes. *BMC Musculoskeletal Disorders*. 8 (1), pp. 39-46.

MYER, G.D., CHU, D.A., BRENT, J.L., and HEWETT, T. E., 2008. Trunk and hip control neuromuscular training for the prevention of knee joint injury. *Clinics in Sports Medicine*. 27 (3), pp. 425–448.

MYER, G.D., BRENT, J.L., FORD, K.R., and HEWETT, T.E., 2011. Real-time assessment and neuromuscular training feedback techniques to prevent ACL injury in female athletes. *Strength and Conditioning Journal*. 33 (3), pp. 21-35.

MYER, G.D., FORD, K.R., FOSS, K.D.B., GOODMAN, A., CEASAR, A., RAUH, M.J., DIVINE, J.G., and HEWETT, T.E., 2010. The incidence and potential pathomechanics of patellofemoral pain in female athletes. *Clinical Biomechanics*. 25 (7), pp. 700-707.

MYER, G.D., FORD, K.R., KHOURY, J., and HEWETT, T.E., 2011. Three-dimensional motion analysis validation of a clinic-based nomogram designed to identify high ACL injury risk in female athletes. *The Physician and Sportsmedicine*. 39 (1), pp.19-28.

NAGAI, T., SELL, T.C., HOUSE, A.J., ABT, J.P., and LEPHART, S.M., 2013. Knee proprioception and strength and landing kinematics during a single-leg stop-jump task. *Journal of Athletic Training*. 48 (1), pp. 31-38.

NAKAMURA, M., IKEZOE, T., TAKENO, Y., and ICHIHASHI, N., 2012. Effects of a 4-week static stretch training program on passive stiffness of human gastrocnemius muscle-tendon unit in vivo. *European Journal of Applied Physiology*. 112 (7), pp. 2749-2755.

NAKAMURA, M., IKEZOE, T., UMEGAKI, H., KOBAYASHI, T., NISHISHITA, S., and ICHIHASHI, N., 2017. Changes in passive properties of the gastrocnemius muscle–tendon unit during a 4-week routine static-stretching program. *Journal of Sport Rehabilitation*. 26 (4), pp. 263-268.

NELSON, R.T., and BANDY, W.D., 2004. Eccentric training and static stretching improve hamstring flexibility of high school males. *Journal of Athletic Training*. 39 (3), pp. 254-258.

NEPTUNE, R.R., KAUTZ, S.A., and ZAJAC, F.E., 2001. Contributions of the individual ankle plantar flexors to support, forward progression and swing initiation during walking. *Journal of Biomechanics*. 34 (11), pp. 1387-1398.

NEUMANN, D.A., 2010. *Kinesiology of the musculoskeletal system: foundation for rehabilitation*. Missouri: Mosby Elsevier.

NIU, W., FENG, T., JIANG, C., and ZHANG, M., 2014. Peak vertical ground reaction force during two-leg landing: A systematic review and mathematical modeling. *BioMed Research International.* 2014, pp. 1-10.

NIU, W., WANG, Y., HE, Y., FAN, Y., and ZHAO, Q., 2011. Kinematics, kinetics, and electromyogram of ankle during drop-landing: a comparison between dominant and non-dominant limb. *Human Movement Science*. 30 (3), pp. 614-623.

NORCROSS, M., BLACKBURN, J., GOERGER, B., and PADUA, D., 2010. The association between lower extremity energy absorption and biomechanical factors related to anterior cruciate ligament injury. *Clinical Biomechanics*. 25 (10), pp. 1031-1036.

NORDIN, A., and DUFEK, J., 2017. Lower extremity variability changes with drop-landing height manipulations. *Research in Sports Medicine*. 25 (2), pp. 144-155.

NORKUS, S.A., and FLOYD, R.T., 2001. The anatomy and mechanisms of syndesmotic ankle sprains. *Journal of Athletic Training*. 36 (1), pp. 68-73.

NOVACHECK, T.K., 1998. The biomechanics of running. *Gait and Posture.* 7 (1), pp. 77–95.

OGILVIE-HARRIS, DJ, MAHOMED, N, and DEMAZIERE, A., 1993. Anterior impingement of the ankle treated by arthroscopic removal of bony spurs. *The Journal of Bone and Joint Surgery. British Volume.* 75 (3), pp. 437-440.

OLBRANTZ, C., BERGELIN, J., ASMUS, J., KERNOZEK, T., RUTHERFORD, D., and GHEIDI, N., 2018. Effect of posttrial visual feedback and fatigue during drop landings on patellofemoral joint stress in healthy female adults. *Journal of Applied Biomechanics*. 34 (1), pp. 82-87.

ORTEGA, D.R., BÍES, E.C.R., and DE LA ROSA, F.J.B., 2010. Analysis of the vertical ground reaction forces and temporal factors in the landing phase of a countermovement jump. *Journal of Sports Science and Medicine*. 9 (2), pp. 282-287.

ORTIZ, A., OLSON, S.L., ETNYRE, B., TRUDELLE-JACKSON, E. E., BARTLETT, W., and VENEGAS-RIOS, H.L., 2010. Fatigue effects on knee joint stability during two jump tasks in women. *The Journal of Strength and Conditioning Research*. 24 (4), pp. 1019-1027.

ORTIZ, A., OLSON, S., LIBBY, C.L., KWON, Y.H., and TRUDELLE-JACKSON, E., 2007. Kinematic and kinetic reliability of two jumping and landing physical performance tasks in young adult women. *North American Journal of Sports Physical Therapy: NAJSP*T. 2 (2), pp. 104-112.

OTA, S., UEDA, M., AIMOTO, K., SUZUKI, Y., and SIGWARD, S.M., 2014. Acute influence of restricted ankle dorsiflexion angle on knee joint mechanics during gait. *The Knee*. 21 (3), pp. 669-675.

PADUA, D.A., DISTEFANO, L.J., BEUTLER, A.I., DE LA MOTTE, S.J., DISTEFANO, M.J., and MARSHALL, S.W., 2015. The landing error scoring system as a screening tool for an anterior cruciate ligament injury–prevention program in elite-youth soccer athletes. *Journal of Athletic Training*. 50 (6), pp. 589-595.

PADUA, D., MARSHALL, S., BOLING, M., THIGPEN, C., GARRETT JR, W., and BEUTLER, A., 2009. The Landing Error Scoring System (LESS) is a valid and reliable clinical assessment tool of jump-landing biomechanics: the JUMP-ACL study. *The American Journal of Sports Medicine*. 37 (10), pp. 1996-2002.

PALASTANGA, N., FIELD, D., and SOAMES, R., 2006. *Anatomy and human movement: structure and function*. 5th ed. London, UK: Elsevier Health Sciences.

PANDY, M.G., and SHELBURNE, K.B., 1997. Dependence of cruciate-ligament loading on muscle forces and external load. *Journal of Biomechanics*. 30 (10), pp. 1015-1024.

PAPPAS, E., and CARPES, F., 2012. Lower extremity kinematic asymmetry in male and female athletes performing jump-landing tasks. *Journal of Science and Medicine in Sport*. 15 (1), pp. 87-92.

PAPPAS, E., HAGINS, M., SHEIKHZADEH, A., NORDIN, M., and ROSE, D., 2007. Biomechanical differences between unilateral and bilateral landings from a jump: gender differences. *Clinical Journal of Sport Medicine*. 17 (4), pp. 263-268.

PAPPAS, E., SHEIKHZADEH, A., HAGINS, M., and NORDIN, M. 2007. The effect of gender and fatigue on the biomechanics of bilateral landings from a jump: peak values. *Journal of Sports Science and Medicine*. 6 (1), pp. 77-84.

PATERNO, M., FORD, K., MYER, G., HEYL, R., and HEWETT, T., 2007. Limb asymmetries in landing and jumping 2 years following anterior cruciate ligament reconstruction. *Clinical Journal of Sport Medicine*. 17 (4), pp. 258-262.

PAYTON, C.J., 2007. Motion analysis using video. In C. J. Payton, and R. M. Bartlett, ed., 2007. *Biomechanical evaluation of movement in sport and exercise*. New York: Routledge. pp. 8-32.

PERES, S.E., DRAPER, D.O., KNIGHT, K.L., and RICARD, M.D., 2002. Pulsed shortwave diathermy and prolonged long-duration stretching increase dorsiflexion range of motion more than identical stretching without diathermy. *Journal of Athletic Training*. 37 (1), pp. 43–50.

PODRAZA, J.T., and WHITE, S.C., 2010. Effect of knee flexion angle on ground reaction forces, knee moments and muscle co-contraction during an impact-like deceleration landing: implications for the non-contact mechanism of ACL injury. *The Knee*. 17 (4), pp. 291-295.

POLLARD, C., SIGWARD, S., and POWERS, C., 2010. Limited hip and knee flexion during landing is associated with increased frontal plane knee motion and moments. *Clinical Biomechanics*. 25 (2), pp. 142-146.

POPE, R., HERBERT, R., and KIRWAN, J., 1998. Effects of ankle dorsiflexion range and pre-exercise calf muscle stretching on injury risk in army recruits. *Australian Journal of Physiotherapy.* 44 (3), pp. 165–172.

POTIER, T.G., ALEXANDER, C.M., and SEYNNES, O.R., 2009. Effects of eccentric strength-training on biceps femoris muscle architecture and knee joint range of movement. *European Journal of Applied Physiology.* 105 (6), pp. 939-944.

POWERS, C.M., 2003. The influence of altered lower-extremity kinematics on patellofemoral joint dysfunction: a theoretical perspective. *Journal of Orthopaedic and Sports Physical Therapy.* 33 (11), pp. 639-646.

POWERS, C.M., 2010. The influence of abnormal hip mechanics on knee injury: a biomechanical perspective. *Journal of Orthopaedic and Sports Physical Therapy.* 40 (2), pp. 42-51.

POWERS, C.M., WITVROUW, E., DAVIS, I.S., and CROSSLEY, K.M., 2017. Evidence-based framework for a pathomechanical model of patellofemoral pain: 2017 patellofemoral pain consensus statement from the 4th International Patellofemoral Pain Research Retreat, Manchester, UK: part 3. *British Journal of Sports Medicine.* 51 (24), pp. 1713-1723.

PRATT, D.J., 1989. Mechanisms of shock attenuation via the lower extremity during running. *Clinical Biomechanics.* 4 (1), pp. 51-57.

PRETTERKLIEBER, M.L., 1999. Anatomy and kinematics of the human ankle joint. *Der Radiologe.* 39 (1), pp. 1-7.

PROSKE, U., and GANDEVIA, S.C., 2012. The proprioceptive senses: their roles in signaling body shape, body position and movement, and muscle force. *Physiological Review.* 92 (3), pp. 1651-1697.

PUIG-DIVÍ, A., ESCALONA-MARFIL, C., PADULLÉS-RIU, J.M., BUSQUETS, A., PADULLÉS-CHANDO, X., and MARCOS-RUIZ, D., 2019. Validity and reliability of the Kinovea program in obtaining angles and distances using coordinates in 4 perspectives. *PLOS ONE*. 14 (6), e0216448.

RABIN, A., and KOZOL, Z., 2017. Utility of the overhead squat and forward arm squat in screening for limited ankle dorsiflexion. *The Journal of Strength and Conditioning Research*. 31 (5), pp. 1251-1258.

RABIN, A., and KOZOL, Z., 2012. Weightbearing and nonweightbearing ankle dorsiflexion range of motion: are we measuring the same thing?. *Journal of the American Podiatric Medical Association*. 102 (5), pp. 406-411.

RABIN, A., KOZOL, Z., and FINESTONE, A.S., 2014. Limited ankle dorsiflexion increases the risk for mid-portion achilles tendinopathy in infantry recruits: a prospective cohort study. *Journal of Foot and Ankle Research*. 7(1), p. 48.

RABIN, A., KOZOL, Z., SPITZER, E., and FINESTONE, A.S., 2015. Weight-bearing ankle dorsiflexion range of motion—can side-to-side symmetry be assumed?. *Journal of Athletic Training*. 50 (1), pp. 30-35.

RADIN, E.L., YANG, K.H., RIEGGER, C., KISH, V.L., and O'CONNOR, J.J., 1991. Relationship between lower limb dynamics and knee joint pain. *Journal of Orthopaedic Research*. 9 (3), pp. 398-405.

REID, A., BIRMINGHAM, T.B., and ALCOCK, G., 2007. Efficacy of mobilization with movement for patients with limited dorsiflexion after ankle sprain: a crossover trial. *Physiotherapy Canada.* 59 (3), pp. 166-172.

RENSTROM, P., LJUNGQVIST, A., ARENDT, E., BEYNNON, B., FUKUBAYASHI, T., GARRETT, W., GEORGOULIS, T., HEWETT, T., JOHNSON, R., KROSSHAUG, T., and MANDELBAUM, B., 2008. Non-contact ACL injuries in female athletes: an International Olympic Committee current concepts statement. *British Journal of Sports Medicine*. 42 (6), pp. 394-412.

RIEMANN, B.L., and LININGER, M.R. 2018. Statistical primer for athletic trainers: the essentials of understanding measures of reliability and minimal important change. *Journal of Athletic Training.* 53 (1), pp. 98-103.

RILEY, P.O., KENT, R.W., DIERKS, T.A., LIEVERS, W.B., FRIMENKO, R.E., and CRANDALL, J.R., 2013. Foot kinematics and loading of professional athletes in American football-specific tasks. *Gait and Posture.* 38 (4), pp. 563-569.

RODRIGUES, P., CHANG, R., TENBROEK, T., VAN EMMERIK, R., and HAMILL, J., 2015. Evaluating the coupling between foot pronation and tibial internal rotation continuously using vector coding. *Journal of Applied Biomechanics*. 31 (2), pp. 88-94.

ROEWER, B.D., FORD, K.R., MYER, G.D., and HEWETT, T.E., 2014. The 'impact' of force filtering cut-off frequency on the peak knee abduction moment during landing: artefact or 'artifiction'?. *British Journal of Sports Medicine*. 48 (6), pp. 464–468.

ROWLEY, M., and RICHARDS, J., 2015. Increasing plantar flexion angle during landing reduces vertical ground reaction forces, loading rates and the hip's contribution to support moment within participants. *Journal of Sports Science*. 33 (18), pp. 1922-1931.

SAKAMOTO, A., MARUYAMA, T., NAITO, H., and SINCLAIR, P.J., 2010. Acute effects of high-intensity dumbbell exercise after isokinetic eccentric damage: interaction between altered pain perception and fatigue on static and dynamic muscle performance. *The Journal of Strength & Conditioning Research*. 24 (8), pp. 2042-2049.

SALSICH, G.B., and PERMAN, W.H., 2007. Patellofemoral joint contact area is influenced by tibiofemoral rotation alignment in individuals who have patellofemoral pain. *Journal of Orthopaedic and Sports Physical Therapy*. 37 (9), pp. 521-528.

SCHOT, P., BATES, B., and DUFEK, J., 1994. Bilateral performance symmetry during drop-landing: a kinetic analysis. *Medicine and Science in Sports and Exercise.* 26 (9), pp. 1153-1159.

SCHWARTZ, C., FORTHOMME, B., PAULUS, J., KAUX, J.F., BRÜLS, O., DENOËL, V., and CROISIER, J.L., 2017. Reliability of unipodal and bipodal counter movement jump landings in a recreational male population. *European Journal of Sport Science*. 17 (9), pp. 1143-1152.

SCHURR, S.A., MARSHALL, A.N., RESCH, J.E., and SALIBA, S.A., 2017. Two-dimensional video analysis is comparable to 3D motion capture in lower extremity movement assessment. *International Journal of Sports Physical Therapy*. 12 (2), pp. 163-172.

SEEGMILLER, J.G., and MCCAW, S.T., 2003. Ground reaction forces among gymnasts and recreational athletes in drop-landings. *Journal of Athletic Training*. 38 (4), pp. 311-314.

SEYMORE, K.D., DOMIRE, Z.J., DEVITA, P., RIDER, P.M., and KULAS, S., 2017. The effect of Nordic hamstring strength-training on muscle architecture, stiffness, and strength. *European Journal of Applied Physiology*. 117 (5), pp. 943–953.

SHIMOKOCHI, Y., YONG LEE, S., SHULTZ, S.J., and SCHMITZ, R.J., 2009. The relationships among sagittal-plane lower extremity moments: implications for landing strategy in anterior cruciate ligament injury prevention. *Journal of Athletic Training.* 44 (1), pp. 33-38.

SIGWARD, S., OTA, S. and POWERS, C., 2008. Predictors of frontal plane knee excursion during a drop land in young female soccer players. *Journal of Orthopaedic Sports and Physical Therapy.* 38 (11), pp. 661-667.

SIMON, J.E., GROOMS, D.R., and DOCHERTY, C.L., 2019. The long-term impact of osteoarthritis following knee surgery in former college athletes. *Journal of Sport Rehabilitation*. 28 (1), pp. 33-38.

SINGH, A.K., STARKWEATHER, K.D., HOLLISTER, A.M., JATANA, S., and LUPICHUK, A.G., 1992. Kinematics of the ankle: a hinge axis model. *Foot and Ankle*. 13 (8), pp. 439-446.

SKINNER, H.B., WYATT, M.P., HODGDON, J.A., CONARD, D.W., and BARRACK, R.L., 1986. Effect of fatigue on joint position sense of the knee. *Journal of Orthopaedic Research*. 4 (1), pp. 112-118.

SLINDE, F., SUBER, C., SUBER, L., EDWÉN, C., and SVANTESSON, U., 2008. Test-retest reliability of three different countermovement jumping tests. *The Journal of Strength and Conditioning Research*. 22 (2), pp. 640-644.

SMITH, M.D., LEE, D., RUSSELL, T., MATTHEWS, M., MACDONALD, D., and VICENZINO, B., 2019. How much does the talocrural joint contribute to ankle dorsiflexion range of motion during the weight-bearing lunge test? A cross-sectional radiographic validity study. *Journal of Orthopaedic and Sports Physical Therapy*, 49 (12), 934-941.

SMITH, M., SIZER, P., and JAMES, C., 2009. Effects of fatigue on frontal plane knee motion, muscle activity, and ground reaction forces in men and women during landing. *Journal of Sports Science and Medicine*. 8 (3), pp. 419-427.

SMITH, J.C., WASHELL, B.R., AINI, M.F., BROWN, S., and HALL, M.C., 2019. Effects of static stretching and foam rolling on ankle dorsiflexion range of motion. *Medicine and Science in Sports and Exercise*. 51 (8), pp. 1752-1758.

STÄRKE, C., KOPF, S., LIPPISCH, R., LOHMANN, C. H., and BECKER, R., 2013. Tensile forces on repaired medial meniscal root tears. Arthroscopy. *The Journal of Arthroscopic and Related Surgery*. 29 (2), pp. 205-212.

STOKES, M., 1985 Reliability and repeatability of methods for measuring muscle in physiotherapy. *Physiotherapy Practice*. 1 (2), pp. 71-76.

SUEKI, D.G., CLELAND, J.A., and WAINNER, R.S., 2013. A regional interdependence model of musculoskeletal dysfunction: research, mechanisms, and clinical implications. *Journal of Manual and Manipulative Therapy*. 21 (2), pp. 90-102.

SUTHERLAND, D.H., COOPER, L., and DANIEL, D., 1980. The role of the ankle plantar flexors in normal walking. *The Journal of Bone and Joint Surgery*. 62 (3), pp. 354-363.

SWINTON, P.A., LLOYD, R., KEOGH, J.W., AGOURIS, I., and STEWART, A.D., 2012. A biomechanical comparison of the traditional squat, powerlifting squat, and box squat. *The Journal of Strength & Conditioning Research*. 26 (7), pp. 1805-1816.

TERADA, M., and GRIBBLE, P.A., 2015. Jump landing biomechanics during a laboratory recorded recurrent ankle sprain. *Foot and Ankle International*. 36 (7), pp. 842-848.

THOMAS, A.C., MCLEAN, S.G., and PALMIERI-SMITH, R.M., 2010. Quadriceps and hamstrings fatigue alters hip and knee mechanics. *Journal of Applied Biomechanics*. 26 (2), pp. 159-170.

THOMAS, E., BIANCO, A., PAOLI, A., and PALMA, A., 2018. The relation between stretching typology and stretching duration: the effects on range of motion. *International Journal of Sports Medicine*. 39 (4), pp. 243-254.

VASEENON, T., and AMENDOLA, A., 2012. Update on anterior ankle impingement. *Current Reviews in Musculoskeletal Medicine*. 5 (2), pp. 145-150.

VAN DEN BEKEROM, M.P., OOSTRA, R.J., ALVAREZ, P.G., and VAN DIJK, C.N., 2008. The anatomy in relation to injury of the lateral collateral ligaments of the ankle: a current concepts review. *Clinical Anatomy*. 21 (7), pp. 619-626.

VAN GHELUWE, B., KIRBY, K.A., ROOSEN, P., and PHILLIPS, R.D., 2002. Reliability and accuracy of biomechanical measurements of the lower extremities. *Journal of the American Podiatric Medical Association*. 92 (6), pp. 317-326.

VAUGHN, D.W., 2008. Isolated knee pain: a case report highlighting regional interdependence. *Journal of Orthopaedic and Sports Physical Therapy*. 38 (10), pp. 616-623.

VICENZINO, B., BRANJERDPORN, M., TEYS, P., and JORDAN, K., 2006. Initial changes in posterior talar glide and dorsiflexion of the ankle after mobilization with movement in individuals with recurrent ankle sprain. *Journal of Orthopaedic and Sports Physical Therapy*. 36 (7), pp. 464-471.

VU, V., WALKER, A., BALL, N., and SPRATFORD, W., 2017. Ankle restrictive firefighting boots alter the lumbar biomechanics during landing tasks. *Applied Ergonomics*. 65, pp. 123-129.

WALDÉN, M., KROSSHAUG, T., BJØRNEBOE, J., ANDERSEN, T.E., FAUL, O., and HÄGGLUND, M., 2015. Three distinct mechanisms predominate in non-contact anterior cruciate ligament injuries in male professional football players: a systematic video analysis of 39 cases. *British Journal of Sports Medicine*. 49 (22), pp. 1452-1460.

WALSH, M., BOLING, M.C., MCGRATH, M., BLACKBURN, J.T., and PADUA, D.A., 2012. Lower extremity muscle activation and knee flexion during a jump-landing task. *Journal of Athletic Training*. 47 (4), pp. 406-413.

WALSH, M., FORD, K., BANGEN, K., MYER, G., and HEWETT, T., 2006. The validation of a portable force plate for measuring force-time data during jumping and landing tasks. *The Journal of Strength and Conditioning Research*. 20 (4), pp. 730-734.

WARD, R.E., FONG YAN, A., ORISHIMO, K.F., KREMENIC, I.J., HAGINS, M., LIEDERBACH, M., HILLER, C.E., and PAPPAS, E., 2019. Comparison of lower limb stiffness between male and female dancers and athletes during drop jump landings. *Scandinavian Journal of Medicine and Science in Sports*. 29 (1), pp. 71-81.

WEEKS, B.K., CARTY, C.P., and HORAN, S.A., 2015. Effect of sex and fatigue on single leg squat kinematics in healthy young adults. *BMC Musculoskeletal Disorders*. 16 (1), pp. 271-280.

WEINHANDL, J.T., JOSHI, M., and O'CONNOR, K.M., 2010. Gender comparisons between unilateral and bilateral landings. *Journal of Applied Biomechanics*. 26 (4), pp. 444-453.

WEINHANDL, J.T., SMITH, J.D., and DUGAN, E.L., 2011. The effects of repetitive drop jumps on impact phase joint kinematics and kinetics. *Journal of Applied Biomechanics.* 27 (2), pp. 108-115

WEISS, K., and WHATMAN, C., 2015. Biomechanics associated with patellofemoral pain and ACL injuries in sports. *Sports Medicine*. 45 (9), pp. 1325-1337.

WHITTING, J.W., STEELE, J.R., JAFFREY, M.A., and MUNRO, B. J., 2007. Parachute landing fall characteristics at three realistic vertical descent velocities. *Aviation, Space, and Environmental Medicine*. 78 (12), pp. 1135-1142.

WHITTING, J.W., STEELE, J.R., MCGHEE, D.E., and MUNRO, B.J., 2011. Dorsiflexion capacity affects Achilles tendon loading during drop-landings. *Medicine and Science in Sports and Exercise.* 43 (4), pp. 706–713.

WHITTING, J., STEELE, J., MCGHEE, D., and MUNRO, B., 2013. Passive dorsiflexion stiffness is poorly correlated with passive dorsiflexion range of motion. *Journal of Science and Medicine in Sport.* 16 (2), pp. 157-161.

WIKSTROM, E.A., and HUBBARD, T.J., 2010. Talar positional fault in persons with chronic ankle instability. *Archives of Physical Medicine and Rehabilitation.* 91 (8), pp. 1267-1271.

WINCHESTER, J.B., NELSON, A.G., LANDIN, D., YOUNG, M.A., and SCHEXNAYDER, I.C., 2008. Static stretching impairs sprint performance in collegiate track and field athletes. *The Journal of Strength and Conditioning Research.* 22 (1), pp. 13-19.

WISLØFF, U., CASTAGNA, C., HELGERUD, J., JONES, R., and HOFF, J., 2004. Strong correlation of maximal squat strength with sprint performance and vertical jump height in elite soccer players. *British Journal of Sports Medicine.* 38 (3), pp. 285-288.

WITHROW, T.J., HUSTON, L.J., WOJTYS, E.M., and ASHTON-MILLER, J.A., 2008. Effect of varying hamstring tension on anterior cruciate ligament strain during in vitro impulsive knee flexion and compression loading. *The Journal of Bone and Joint Surgery.* 90 (4), pp. 815-823.

WOLLIN, M., THORBORG, K., and PIZZARI, T., 2017. The acute effect of match play on hamstring strength and lower limb flexibility in elite youth football players. *Scandinavian Journal of Medicine and Science in Sports.* 27 (3), pp. 282-288.

WRIGHT, I., NEPTUNE, R., VAN DEN BOGERT, A., and NIGG, B., 2000. The influence of foot positioning on ankle sprains. *Journal of Biomechanics.* 33 (5), pp. 513–519.

YAMAZAKI, J., MUNETA, T., JU, Y. J., and SEKIYA, I., 2010. Differences in kinematics of single leg squatting between anterior cruciate ligament-injured patients and healthy controls. *Knee Surgery, Sports Traumatology, Arthroscopy.* 18 (1), pp. 56-63.

YANCI, J., and CAMARA, J., 2016. Bilateral and unilateral vertical ground reaction forces and leg asymmetries in soccer players. *Biology of Sport*. 33 (2), pp. 179-183.

YEOW, C.H., LEE, P.V.S., and GOH, J.C.H., 2009. Effect of landing height on frontal plane kinematics, kinetics and energy dissipation at lower extremity joints. *Journal of Biomechanics*. 42 (12), pp. 1967-1973.

YEOW, C.H., LEE, P.V.S., and GOH, J.C.H., 2009. Regression relationships of landing height with ground reaction forces, knee flexion angles, angular velocities and joint powers during double-leg landing. *The Knee*. 16 (5), pp. 381-386.

YEOW, C.H., LEE, P.V.S., and GOH, J.C.H., 2010. Sagittal knee joint kinematics and energetics in response to different landing heights and techniques. *The Knee*. 17 (2), pp. 127-131.

YEOW, C.H., LEE, P.V.S., and GOH, J.C.H., 2011. Non-linear flexion relationships of the knee with the hip and ankle, and their relative postures during landing. *The Knee*. 18 (5), pp. 323-328.

YEOW, C.H., LEE, P.V.S., and GOH, J.C.H., 2011. An investigation of lower extremity energy dissipation strategies during single-leg and double-leg landing based on sagittal and frontal plane biomechanics. *Human Movement Science.* 30 (3), pp. 624-635.

YOUDAS, J.W., MCLEAN, T.J., KRAUSE, D.A., and HOLLMAN, J.H., 2009. Changes in active ankle dorsiflexion range of motion after acute inversion ankle sprain. *Journal of Sport Rehabilitation*. 18 (3), pp. 358-374.

YOUNG, J.L., HERRING, S.A., PRESS, J.M., and CASAZZA, B.A. 1996. The influence of the spine on the shoulder in the throwing athlete. *Journal of Back and Musculoskeletal Rehabilitation*. 7 (1), pp. 5-17.

YOUNG, W., WILSON, G., and BYNE, C., 1999. Relationship between strength qualities and performance in standing and run-up vertical jumps. *Journal of Sports Medicine and Physical Fitness*. 39 (4), pp. 285-293.

YU, B., and GARRETT, W.E., 2007. Mechanisms of non-contact ACL injuries. *British Journal of Sports Medicine*. 41 (suppl 1), pp. i47-i51.

YU, B., LIN, C., and GARRETT, W., 2006. Lower extremity biomechanics during the landing of a stop-jump task. *Clinical Biomechanics*. 21 (3), pp. 297-305.

ZADPOOR, A., and NIKOOYAN, A., 2012. The effects of lower extremity muscle fatigue on the vertical ground reaction force: a meta-analysis. *Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine*. 226 (8), pp. 579-588.

ZHANG, S., BATES, B., and DUFEK, J., 2000. Contributions of lower extremity joints to energy dissipation during landings. *Medicine and Science in Sports and Exercise*. 32 (4), pp. 812-819.

ZHANG, S., PAUL, J., NANTHA-AREE, M., BUCKLEY, N., SHAHZAD, U., CHENG, J., DEBEER, J., WINEMAKER, M., WISMER, D., PUNTHAKEE, D., and AVRAM, V., 2014. Empirical comparison of four baseline covariate adjustment methods in analysis of continuous outcomes in randomized controlled trials. *Clinical Epidemiology*. 6, pp. 227.-235

ZOURDOS, M.C., KLEMP, A., DOLAN, C., QUILES, J.M., SCHAU, K.A., JO, E., HELMS, E., ESGRO, B., DUNCAN, S., MERINO, S.G., and BLANCO, R., 2016. Novel

resistance training–specific rating of perceived exertion scale measuring repetitions in reserve. *The Journal of Strength and Conditioning Research*. 30 (1), pp. 267-275.

Appendices

Appendix 1: Ethical approval

27 September 2017

Our Ref: DC/SB

Louis Howe
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Bowerham Road

University of
Cumbria

Research Office
University of Cumbria
Lancaster Campus
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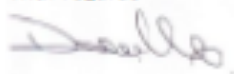Fax: 01524 384385
Email: research.office@cumbria.ac.uk

Dear Louis

Request for Ethical Clearance – Our Ref 16/88
Project: Development and variability of compensatory movement strategies during athletic activities in healthy subjects with side-to-side asymmetry of ankle dorsiflexion range of motion

Thank you for your recent application for ethical review. Approval is granted with no changes or amendments required.

With regards

Professor Diane Cox
Chair
Research Ethics Panel

University of Cumbria
Cumbria

17 October 2018

Our Ref: DC/SB

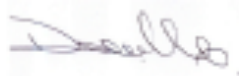Louis Howe
MSS
Lancaster

Dear Louis

Request for Ethical Clearance – Our Ref: 18/11
Project: The effect of an ankle conditioning intervention on unfatigued and fatigued movement strategies during bilateral drop-landings

Thank you for your revised application regarding the issues that required a response.

Approval is granted with no further changes or amendments required.

Kind regards

Professor Diane Cox
Chair
Research Ethics Panel

University of Cumbria

Participant Information Sheet

Research Project: *Development and variability of compensatory movement strategies during athletic activities in healthy subjects with side-to-side asymmetry of ankle dorsiflexion range of motion*

About the study

This research project is investigating the effects of ankle dorsiflexion range of motion (ROM) restrictions on movement strategies during bilateral drop landings. The purpose is to identify if ankle joint restrictions cause compensations in movement.

Some questions you may have about the research project:

Why have you asked me to take part and what will I be required to do?

You have been asked to participate in this research project due to your training history/physical activity level and your injury-free status.

At the beginning of each session, you will have your ankle dorsiflexion ROM measured on both sides and will be required to perform three countermovement jumps for maximal height. Then, you will be asked to perform five bilateral drop landings from box heights of 50%, 100% and 150%. As an example, if you can perform a maximal countermovement jump of 30cm, you will be required to perform drop landings after stepping off a box of 15, 30 and 45cm high.

Across each testing session, you will be required to perform five drop-landings. In the first session, you will have the opportunity to practice the bilateral drop landings with the ankle braces from each box height.

What if I do not wish to take part or change my mind during the study?

Your participation in the study is entirely voluntary. You are free to withdraw from the study at any time without having to provide a reason for doing so.

What happens to the research data?

The data will be used for presentations and publication. Your data will at all times remain anonymised. Your data will be stored using a coded system to protect your identity and will be saved in password protected systems. Upon request, your data will be made available to you at any time.

How will the research be reported?

The research will be reported as part of a PhD thesis. This may result in conference presentations and journal publications. If requested, you will be provided to access any research that your participation in this project has contributed to.

How can I find out more information?

Please contact the principle researcher (Louis Howe) directly via email: louis.howe@cumbria.ac.uk

What if I want to complain about the research?

Initially you should contact the researcher directly. However, if you are not satisfied or wish to make a more formal complaint you should contact Diane Cox, Director of Research Office, University of Cumbria, Bowerham Road, Lancaster, LA1 3JD. diane.cox@cumbria.ac.uk

University of Cumbria

Participant Information Sheet
Research Project: *The impact of an ankle conditioning intervention on unfatigued and fatigued movement strategies during bilateral drop-landings*

About the study
This research project is investigating the impact of ankle dorsiflexion range of motion (ROM) on coordination strategies during bilateral drop-landings before and after an ankle conditioning intervention.

Some questions you may have about the research project:

Why have you asked me to take part and what will I be required to do?
Your participation for this study will require you to complete a familiarisation session, 4 test sessions and potentially a training intervention lasting 6-weeks. The testing sessions will be distributed either side of a 6-week period (i.e. test sessions 1 and 2 before and test sessions 3 and 4 after the 6-week period).

Familiarisation session: For this session, you will have your ankle dorsiflexion ROM measured on both sides and will be required to perform 3 countermovement jumps (CMJ) for maximal height. These tests will be repeated again at the beginning of test session 3. Then, you will be asked to perform 3 bilateral drop-landings from a drop heights of 150% of your CMJ. As an example, if you can perform a maximal countermovement jump of 30cm, you will be required to perform drop-landings from a box height of 45cm.

Test sessions: Across each test session, you will be required to perform 5 bilateral drop-landings from a drop height of 150% of your CMJ (established in the familiarisation session). In test sessions 1 and 3, this will be performed directly after a warm-up. In test session 2 and 4, you will be asked to perform a warm up, then complete an exercise circuit, before performing 5 bilateral drop-landings. The circuit will be stopped once your CMJ height declines by 20% of your maximum height. The exercise circuit will be:
- Bodyweight squats x 30
- Repeat CMJ x 30
- Jump lunges x 30 (15 each side)

Intervention: You may be part of one of two intervention groups, or part of the control group. The intervention will be performed 2-5 times per week and will last up to 6-weeks. The exercises included in the intervention will be general flexibility and lower extremity resistance training exercises. For all of the exercises performed as part of the intervention, you will be coached by an accredited strength and conditioning coach with >10 years of practical experience. Some of the exercises may be performed in your own time, while some will require you to attend the Rehabilitation Gym on the University of Cumbria, Lancaster campus at specific times.

Your performance will be recorded through the landing forces and video analysis of your landings.

What if I do not wish to take part or change my mind during the study?
Your participation in the study is entirely voluntary. You are free to withdraw from the study at any time without having to provide a reason for doing so. You will not be disadvantaged by doing so in any way.

What happens to the research data?
The data will be used for presentations and publications. Your stored data will at all times remain anonymised. Your data will be stored using a coded system to protect your identity and will be saved in password protected systems. Upon request, your data will be made available to you at any time.

How will the research be reported?
The research will be reported as part of a PhD thesis. This may result in conference presentations and journal publications. If requested, you will be provided access to any research that your participation has contributed to. No individual will be identified by the data presentation.

How can I find out more information?
Please contact the principle researcher (Louis Howe) directly via email: louis.howe@cumbria.ac.uk

What if I want to complain about the research?
Initially you should contact the researcher directly. However, if you are not satisfied or wish to make a

more formal complaint you should contact Diane Cox, Director of Research Office, University of

Cumbria, Bowerham Road, Lancaster, LA1 3JD. diane.cox@cumbria.ac.uk

University of
Cumbria

Participant Consent Form

Research Project: The effect of an ankle conditioning intervention on unfatigued and fatigued movement strategies during bilateral drop-landings

Please answer the following questions by circling your responses:

Have you read and understood the information sheet about this study?   YES   NO

Have you been able to ask questions and had enough information?   YES   NO

Do you understand that you are free to withdraw from this study at any time, and without having to give a reason for withdrawal?   YES   NO

Please sign here if you wish to take part in the research and feel you have had enough information about what is involved:

Signature of participant:.. **Date:**.................

Name (block letters):...

Signature of investigator:... **Date:**.................

Name (block letters):...

Appendix 4: Physical activity readiness questionnaire

University of
Cumbria

Department of Medical and Sport Sciences
Physical Activity Readiness Questionnaire (Par-Q)

Participant Full Name: _____ Emergency contact number: _____

Course: _____

This form needs to be completed and returned to the relevant member of staff before participating in any scheduled physical activity during the day.

An indicative (but not exhaustive) list of activities that you are likely to participate in during the day, can be found below. If you are unsure about any of these, or any of the activities you will perform, please ask any member of staff for more information.

Walking	Team games (e.g. football, rugby, basketball, hockey)
Running	Weight lifting
Jumping	Physical conditioning exercises
Cycling	Stretching

An indicative (but not exhaustive) list of conditions that could prevent you from partaking are

Experiencing chest pains after exercise	Regularly taking drugs or medicines
Getting out of breath at rest or slight exertion	Having pain or limited movement in any joints
Often getting headaches, dizziness, or fainting spells	Pregnancy
Having been diagnosed with a heart condition	

If you have any of these conditions, others not mentioned above or you have concerns about your ability to take part in any of the activities, please ask any member of staff for advice.

Are you able to participate in physical, exercise and sporting activities?   Yes   No

If you answered 'No' to the above question, please ask any member of staff for advice.

The information given above is to the best of my knowledge a true and accurate record. I understand that I need to notify staff of any changes in my circumstances. I consent to take part in the study's activities at my own risk, and, if necessary, will have obtained GP clearance to do so.

Signed (participant): _____ Date: _____

Appendix 5: Practical approach to problem-solving movement tasks limited by an ankle dorsiflexion restriction

Practical Approach to Problem-Solving Movement Tasks Limited by an Ankle Dorsiflexion Restriction

Louis Howe, MSc,[1] Mark Waldron, PhD,[2,3] and Jamie North, PhD[2]
[1]Medical and Sport Sciences, University of Cumbria, Lancaster, United Kingdom; [2]School of Sport, Health and Applied Science, St Mary's University, Twickenham, Middlesex, United Kingdom; and [3]School of Science and Technology, University of New England, Armidale, New South Wales, Australia

ABSTRACT

LIMITATIONS IN ANKLE DORSIFLEXION RANGE OF MOTION HAVE BEEN SHOWN TO INCREASE COMPENSATORY MOVEMENTS AT BOTH PROXIMAL AND DISTAL JOINT SEGMENTS IN THE LOWER EXTREMITY. THIS ARTICLE DISCUSSES METHODS TO ASSESS AND CORRECT DEFICIENCIES IN ANKLE DORSIFLEXION RANGE OF MOTION. PREVIOUSLY, HOWEVER, THE REMOVAL OF JOINT RESTRICTIONS HAS NOT BEEN SHOWN TO REDUCE COMPENSATORY STRATEGIES DEVELOPED THROUGH SUCH RESTRICTIONS. THEREFORE, THIS ARTICLE WILL ALSO DISCUSS IMPORTANT CONSIDERATIONS FOR FACILITATING THE RELEARNING PROCESS AND PROPOSE KEY PRINCIPLES FOR DEVELOPING A CORRECTIVE PROGRAM.

INTRODUCTION

During high load activities, failure to control joint segments has the potential to result in excessive loading of both active and passive structures, with injury being a possible outcome (20). Poor movement quality during dynamic activities may be caused by reduced movement control from a stability perspective, whereby suboptimal muscle activation strategies lead to compensatory movements at joints being loaded, resulting in the poor transfer of forces across joint segments (22). Another cause for compensatory movement strategies is joint hypomobility (12,35,36), where joint restrictions reduce movement options for the performer, leading to a suboptimal approach to solving a movement challenge.

During dynamic tasks such as squatting (28), jumping (12), and running (46), ankle dorsiflexion is a natural strategy used by athletes to manipulate the location of the center of mass and dissipate the load in preparation for propulsion. These activities vary in their demands for ankle dorsiflexion range of motion (ROM), with approximately 10° required for walking, increasing to 30° for running (46).

A reduction in ankle dorsiflexion ROM has been identified as a risk factor for numerous injuries. In the lower leg, limited ankle ROM is a risk factor for the development of plantar fasciitis (29), tibial stress syndrome, ankle sprains, and Achilles tendinopathy (48). More proximally, restrictions in ankle joint ROM have been related to the occurrence of hamstring strains (13), iliotibial band syndrome (38), anterior knee pain (39), and patella tendinopathy (1).

Although the exact mechanism through which a restriction in ankle dorsiflexion ROM increases the risk of lower extremity injuries is presently unclear, researchers have identified a number of dysfunctional movement patterns that are developed as a consequence of joint hypomobility. Limitations in ankle dorsiflexion ROM have been shown to result in greater peak vertical ground reaction forces, secondary to reduced peak knee flexion angles during landing tasks (12). This jarring strategy is likely adopted because of reduced ankle dorsiflexion ROM limiting the angular displacement of the proximal tibia, thereby inhibiting knee flexion from occurring (12). During movements such as

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Strategies to increase ankle dorsiflexion range of motion

Louis P. Howe, Dr Mark Waldron, Dr Jamie S. North and Dr Theodoros M. Bampouras review exercise-based strategies to restore ankle dorsiflexion range of motion.

Introduction

Ankle dorsiflexion range of motion (DF ROM) restriction negatively influences the performance of functional activities in healthy populations (Howe et al., 2019). Limitations in ankle DF ROM can be caused by a lack of extensibility in myofascial tissues surrounding the ankle joint or disruption in ankle joint arthrokinematics. A variety of exercise-based strategies have been reported to restore ankle DF ROM in individuals with restricted motion and will be presented within this brief review (an example of an exercise related to each strategy is presented in Figure 1).

Figure 1. Example of exercises used to increase ankle DF ROM. **A)** Static stretch (Bent knee ankle plantar flexor stretch); **B)** Eccentric strength training (Single-leg heel drop); **C)** Self-massage (Ankle plantar flexors); **D)** Self-mobilisation (Ankle stretch using a strap). All images courtesy of Louis Howe.

Static stretching

Although a variety of forms of stretching can be prescribed, the effect of static stretching on the ankle plantar flexors has been most commonly investigated and has a wider evidence base of chronic effects when compared to other techniques. Static stretching of the ankle plantar flexors involves lengthening the muscles until a sensation of stretching is reached, and then holding this position for sustained periods. The advantage of this technique is that minimal equipment and expertise is required, and stretches can be employed without assistance. For example, static stretching of the ankle plantar flexors for a total duration of 2 min, three times a week for 4 weeks significantly increased ankle DF ROM in comparison to a control group, which showed no change (Nakamura et al., 2017). Such findings have been consistently reported across similar studies irrespective of measurement technique.

An interesting aspect is the response of static stretching to various programming considerations. For example, the frequency of stretching between investigations, ranging from three to seven times per week. However, a certain static stretching duration

(60s) performed three times per week increased flexibility gains - but there was no additional benefit to performing it five times a week (Marques et al., 2009). Similarly, 5 minutes in total duration over the course of a week was sufficient to increase flexibility, with additional time dedicated to stretching bringing no greater gains. Furthermore, no differences existed when the total time spent stretching was divided into shorter (< 60 s), moderate (60-120 s) or longer durations (> 120 s), and higher volumes (> 5,000 s) compared to shorter volumes (< 3,000 s) of stretching duration. Thus, as the magnitude of increase in flexibility gains appears relatively uninfluenced by the frequency and volume of static stretching (Medeiros & Martini, 2018), it appears that static stretching performed three times per week for a total weekly duration of 5 minutes is sufficient to increase ankle DF ROM.

Eccentric strength training

Eccentric strength training may also be employed to induce chronic gains in flexibility. Eccentric strength training involves the active lengthening of the musculotendinous unit under loaded conditions and can facilitate increases in flexibility primarily through the addition of sacromeres in series (Mahieu et al., 2008). Performed daily for 6 weeks, eccentric strengthening exercise resulted in significant increases in ankle DF ROM. Interestingly, the increase was not different with the knee extended or flexed, suggesting that eccentric strength increases flexibility of both the biarticular and uniarticular plantar flexors equally, rendering a specific protocol to target each muscle group unnecessary. The same exercise protocol, but over a shorter, 4-week duration programme, resulted in very similar ankle DF ROM increases (Aune et al., 2019), indicating that gains can be realised in a short period of time. Concurrently to the addition of sacromeres in series, eccentric strength training causes another adaptation to the ankle plantar flexors by significantly reducing the passive resistive torque, suggesting structural adaptations occur in the muscle-tendon unit as a result of the intervention. Therefore, performing eccentric strength training provides a suitable stimulus for improving ankle DF ROM and should be included in a well-rounded mobility intervention.

Self-massage

Self-massage, also known as self-myofascial release, can be prescribed to increase flexibility. Devices such as foam rollers, custom-made balls and roller sticks are specialised equipment that are routinely used in practice to self-induce acute and chronic changes in range of motion, but studies on the chronic effects of self-massage to exclusively increase ankle DF ROM are lacking. Self-massage to the calf muscles with a roller stick for three sets of 30 s at a perceived pain level of 7 out of 10, resulted in a significant improvement in ankle DF ROM, and one that was comparable to static stretching (Halperin et al., 2014). Interestingly, and similar to the non "dose-response" of static
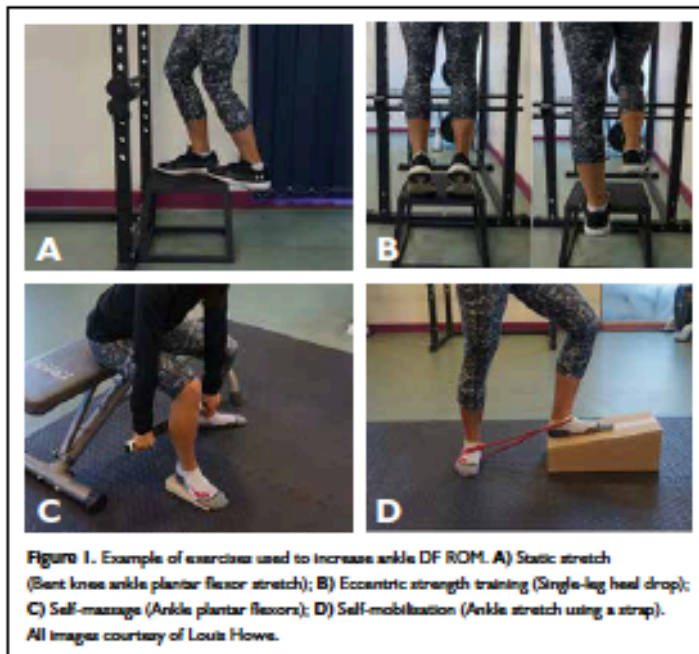
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Appendix 7: Within-session reliability for inter-limb asymmetries in ankle dorsiflexion range of motion during the weight-bearing lunge test

WITHIN-SESSION RELIABILITY FOR INTER-LIMB ASYMMETRIES IN ANKLE DORSIFLEXION RANGE OF MOTION MEASURED DURING THE WEIGHT-BEARING LUNGE TEST

Louis P. Howe[1,2]
Theodoros M. Bampouras[3]
Jamie S. North[4]
Mark Waldron[5,6]

ABSTRACT

Background: The identification of asymmetrical inter-limb ankle dorsiflexion range of motion (DF ROM) has the potential to influence the course of treatment during the rehabilitation process, with limitations in ankle DF ROM potentially increasing injury risk. However, reliability for methods to identify ankle DF ROM asymmetries remain under described in the literature.

Purpose: To determine the reliability of the trigonometric calculation method for measuring ankle DF ROM during the weight-bearing lunge test (WBLT) for both a single limb and the symmetry values. The secondary purpose was to establish values of ankle DF ROM asymmetry and identify the influence of leg dominance on ankle DF ROM.

Study Design: Cross-sectional study.

Methods: Ankle DF ROM was measured bilaterally in 50 healthy and recreationally active participants (28 men, 22 women, age = 22 ± 4 years, height = 172.8 ± 10.8 cm, body mass 71.5 ± 15.1 kg), using the trigonometric measurement method during the WBLT. Each ankle was measured twice in a single testing session to establish within-session reliability.

Results: Values are presented for asymmetries in DF ROM. No differences were identified between the dominant and non-dominant limb (p = 0.862). Within-session reliability for measuring a single limb was classified as 'good' (ICC = 0.98) with a minimal detectable change value of 1.7˚. For measuring ankle DF ROM asymmetry, reliability was established as 'good' (ICC = 0.85) and a minimal detectable change value of 2.1˚ was determined.

Conclusions: Although symmetry in ankle DF ROM may not be assumed, the magnitude of asymmetry may be less than previously reported in a population of recreationally active individuals. Discrepancies between previous research and the findings of the present study may have been impacted by differences in measurement methods. Furthermore, clinicians should be aware that the error associated with measures of asymmetry for ankle DF ROM during the WBLT is greater than that of a single limb.

Level of Evidence: 2b

Key words: ankle dorsiflexion, inter-limb asymmetry, reliability, trigonometric calculation method

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Appendix 8: Reliability of independent kinetic variables and measures of inter-limb asymmetry associated with bilateral drop-landing performance

Reliability of independent kinetic variables and measures of inter-limb asymmetry associated with bilateral drop-landing performance

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Abstract: The purpose of this investigation was to establish the within-session reliability for peak vertical ground reaction force (vGRF), time to peak vGRF, and loading rate, both unilaterally and bilaterally, during a drop-landing task as well as the reliability of inter-limb asymmetry in peak vGRF. Twenty-two men (age = 22 ± 4 years; height = 180.4 ± 6.1 cm; mass = 77.9 ± 14.0 kg) and 17 women (age = 20.4 ± 3.6 years; height = 164.6 ± 9.4 cm; mass = 60.3 ± 9.8 kg) volunteered for a single testing session. Participants completed three countermovement jumps (CMJ) to establish maximum jump height before performing five bilateral drop-landings from 50%, 100%, and 150% of their maximum CMJ height. The bilateral drop-landing protocol was then repeated after a 10 min recovery. Systematic bias, intraclass correlation coefficient (ICC), coefficient of variation (CV%) and minimal detectable change (MDC) values for each kinetic measurement was calculated for the left and right leg, as well as bilaterally. There was no systematic bias present between trials ($P > 0.05$). All kinetic measurements showed relative reliability, ranging from *large* to *near perfect* (ICC = 0.57–0.95). Absolute reliability ranged considerably depending on the measure and drop-height, with peak vGRF and time to peak GRF showing the greatest reliability at higher drop heights (CV% = 6.6–9.7%). Loading rate for all drop heights demonstrated CV% ranging 13.0–27.6%. Furthermore, MDC values for inter-limb asymmetries in peak vGRF ranged between 14.5–16.2% for all drop heights. Overall, many of the kinetic measurements evaluated were sufficiently reliable to detect typical changes in bilateral drop-landing performance when greater drop heights were used.

Key Words: within-session reliability, kinetic variables, landings; inter-limb asymmetry

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Appendix 9: Reliability of two-dimensional measures associated with bilateral drop-landing performance

ARTICLE

Reliability of two-dimensional measures associated with bilateral drop-landing performance

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Abstract – The aim of this study was to establish the within-session reliability for two-dimensional (2D) video analysis of sagittal- and frontal-plane measures during bilateral drop-landing tasks. Thirty-nine recreational athletes (22 men, 17 women, age = 22 ± 4 years, height = 1.74 ± 0.15 m, body mass 70.2 ± 15.1 kg) performed five bilateral drop-landings from 50, 100 and 150% of maximum countermovement jump height, twice on the same day. Measures of reliability for initial contact angle, peak flexion angle and joint displacement for the hip, knee, and ankle joints, frontal-plane projection angles (FPPA), as well as inter-limb asymmetries in joint displacement were assessed. No systematic bias was present between trials ($P > 0.05$). All kinematic measurements showed relative reliability ranging from large to near perfect (ICC = 0.52–0.96). Absolute reliability ranged between measures, with CV% between 1.0–1.6% for initial contact angles, 1.9–7.9% for peak flexion angles, 5.3–22.4% for joint displacement, and 1.6–2.3% for FPPA. Absolute reliability for inter-limb asymmetries in joint displacement were highly variable, with minimal detectable change values ranging from 6.0–13.2°. Therefore, 2D video analysis is a reliable tool for numerous measures related to the performance of bilateral drop-landings.

Keywords: within-session reliability, kinematics, landings

Résumé – **Fiabilité des mesures bidimensionnelles associées aux performances d'atterrissage en chute bilatérale.** Le but de cette étude était d'établir la fiabilité intra-session pour l'analyse vidéo bidimensionnelle (2D) de mesures sur le plan sagittal et frontal lors de tâches d'atterrissage en chute libre bilatérales. Trente-neuf sportifs sportifs (22 hommes et 17 femmes, Age = 22 ± 4 ans, taille = 1,74 ± 0,15 m, masse corporelle 70,2 ± 15,1 kg) ont effectué cinq atterrissages bilatéraux à partir de 50, 100 et 150 % du maximum hauteur du saut en contre-mouvement, deux fois le même jour. Mesures de fiabilité pour l'angle de contact initial, l'angle de flexion maximal et le déplacement articulaire pour les articulations de la hanche, du genou et de la cheville, les angles de projection dans le plan frontal (FPPA), ainsi que les asymétries inter-membres dans le déplacement articulaire. Aucun biais systématique n'était présent entre les essais ($p > 0,05$). Toutes les mesures cinématiques ont montré une fiabilité relative allant de grande à quasi parfaite (ICC = 0,52–0,96). La fiabilité absolue variait d'une mesure à l'autre, avec des CV% compris entre 1,0 et 1,6 % pour les angles de contact initiaux, entre 1,9 et 7,9% pour les angles de flexion maximaux, entre 5,3 et 22,4 % pour les déplacements articulaires et entre 1,6 et 2,3 % pour les FPPA. La fiabilité absolue pour les asymétries inter-membres dans le déplacement articulaire était très variable, avec des valeurs de changement détectables minimales allant de 6,0 à 13,2°. Par conséquent, l'analyse vidéo 2D est un outil fiable pour de nombreuses mesures liées à la performance des atterrissages bilatéraux.

Mots clés : fiabilité intra-session, cinématique, atterrissages

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Appendix 10: Ankle dorsiflexion range of motion is associated with kinematic but not kinetic variables related to bilateral drop-landing performance at various drop heights

Ankle dorsiflexion range of motion is associated with kinematic but not kinetic variables related to bilateral drop-landing performance at various drop heights

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ABSTRACT

Limited evidence is available concerning ankle dorsiflexion range of motion (DF ROM) and its relationship with landing performance from varying drop heights. The aim of this investigation was to determine the relationship between ankle DF ROM and both kinetic and kinematic variables measured during bilateral drop-landings from 50%, 100% and 150% of counter-movement jump height. Thirty-nine participants were measured for their ankle DF ROM using the weight-bearing lunge test, after which five bilateral drop-landings were performed from 50%, 100% and 150% of maximal countermovement jump height. Normalized peak vertical ground reaction force (vGRF), time to peak vGRF and loading rate was calculated for analysis, alongside sagittal-plane initial contact angles, peak angles and joint displacement for the hip, knee and ankle. Frontal-plane projection angles were also calculated. Ankle DF ROM was not related to normalized peak vGRF, time to peak vGRF or loading rate ($P > 0.05$), regardless of the drop height. However, at drop heights of 100% and 150% of countermovement jump height, there were numerous significant ($P < 0.05$) moderate to large correlations between ankle DF ROM and initial contact angles ($r = -0.34$ to -0.40) and peak angles ($r = -0.42$ to -0.52) for the knee and ankle joints. Knee joint displacement ($r = 0.39-0.47$) and frontal-plane projection angle ($r = 0.37-0.40$) had a positive relationship with ankle DF ROM, which was consistent across all drop heights. Ankle DF ROM influences coordination strategies that allow for the management of vGRF during bilateral drop-landings, with alterations in alignment for the knee and ankle joints at both initial contact and peak angles.

1. Introduction

Bilateral landings from a height are performed by athletes in training and competition (Bloomfield, Polman, & O'Donoghue, 2007; McClay et al., 1994) and are also part of daily life during leisure activities and occupational tasks (Knapik, Craig, Hauret, & Jones, 2003). Successfully executing a bilateral landing is necessary to attenuate the large vertical forces that can equate to multiples of body weight thus preserving the integrity of anatomical structures of the lower-limbs (Hewett et al., 2005). To appropriately manage high vertical forces, the hip, knee and ankle joint must be coordinated to provide a movement strategy that facilitates effective

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Improved Ankle Mobility After a 4-Week Training Program Affects Landing Mechanics: A Randomized Controlled Trial

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Abstract

Howe, LP, Bampouras, TM, North, JS, and Waldron, M. Improved ankle mobility after a 4-week training program affects landing mechanics: a randomized controlled trial. *J Strength Cond Res* XX(X): 000–000, 2020—This study examined the effects of a 4-week ankle mobility intervention on landing mechanics. Twenty subjects with restricted ankle dorsiflexion range of motion (DF ROM) were allocated to either a strength training only ($n = 9$) or a strength training and ankle mobility program ($n = 11$). Subjects performed a weight-bearing lunge test and bilateral drop-landings before and after the intervention. Normalized peak vertical ground reaction force (vGRF), time to peak vGRF, and loading rate were calculated, alongside sagittal-plane initial contact angles, peak angles, and sagittal-plane joint displacement for the ankle, knee, and hip. Frontal-plane projection angles were also calculated. After the intervention, only the strength and mobility group improved ankle DF ROM (mean difference $= 4.1°$, effect size [ES] $= 1.00$, $p = 0.002$). A one-way analysis of covariance found group effects for ankle joint angle at initial contact ($p = 0.045$), ankle ($p < 0.001$) and hip joint angle at peak flexion ($p = 0.041$), and sagittal-plane ankle ($p < 0.001$) and hip joint displacement ($p = 0.024$) during bilateral drop-landings. Post hoc analysis revealed that the strength and mobility group landed with greater ankle plantarflexion at initial contact (mean difference $= 1.4 ± 2.0°$, ES $= 0.46$) and ankle dorsiflexion at peak flexion (mean difference $= 6.3 ± 2.9°$, ES $= 0.74$) after the intervention, resulting in a greater ankle joint displacement (mean difference $= 7.7 ± 4.0°$, ES $= 1.00$). However, the strength training only group landed with increased peak hip flexion (mean difference $= 14.4 ± 11.0°$, ES $= 0.70$) and hip joint displacement (mean difference $= 8.0 ± 6.6°$, ES $= 0.44$) during post-testing. The findings suggest that changes in landing strategies following the performance of a strength training program are specific to whether restrictions in ankle mobility are considered as part of the intervention.

Key Words: landing mechanics, ankle dorsiflexion, mobility

Introduction

During landings, ankle dorsiflexion aids in attenuating vertical ground reaction forces (vGRFs) (34), while facilitating knee and hip flexion through sagittal-plane coupling mechanisms to reduce the impact of landing (31). Restrictions in ankle dorsiflexion range of motion (DF ROM) are recognized as a modifiable injury risk factor for athletes who perform a high volume of landing activities (2). This is likely due to compensations caused by ankle DF ROM restriction during landing tasks, resulting in less effective strategies being used. For example, reduced ankle DF ROM has been shown to limit peak ankle, knee, and hip flexion angles (10,14), while increasing peak knee abduction angles during landings (14,22). In addition, during landings where individuals with restricted ankle DF ROM demonstrate reduced knee flexion joint displacement, a negative relationship between ankle DF ROM and peak vGRF during bilateral landings has been reported (11). These findings suggest that individuals with ankle DF ROM restrictions land using a stiffer strategy with greater peak knee abduction angles that may result in elevated landing forces.

Increased ankle mobility may improve landing mechanics by increasing sagittal-plane joint displacement at the ankle, knee, and hip (10,11,14), resulting in reduced peak vGRF (34) and, consequently, diminished injury risk (13). Interestingly, ankle DF ROM can be improved in relatively short time periods, as significant gains in ankle DF ROM have been shown in ≤4 weeks when adhering to interventions designed to increase the flexibility of the ankle plantarflexors (1,19,24). Little is known regarding the functional consequences of developing ankle mobility because currently, no studies have investigated the effect of increasing ankle DF ROM on landing mechanics in individuals identified with a mobility restriction at the ankle joint.

In practice, individuals with restrictions in ankle DF ROM will likely be identified during a pre-exercise screening session before initiating a strength and conditioning program (18). When deficits in ankle DF ROM are found, a corrective program to restore ankle mobility would be prescribed. This would likely be performed as a supplementary intervention alongside a strength training program designed to develop relevant physical qualities that will improve athletic performance. However, whether a corrective program aimed at restoring ankle mobility results in greater sagittal-plane ankle, knee, and hip joint displacement, which in turn results in reduced peak vGRF during landing tasks,

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