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Evaluation of Lower Body Strength and Landing Strategy of Elite Athletes After Anterior Cruciate Ligament Reconstruction with Hamstring Tendon Autograft

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Evaluation of Lower Body Strength and Landing Strategy of Elite Athletes After Anterior Cruciate Ligament Reconstruction with Hamstring Tendon Autograft

by

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Abstract

The purpose of this study was to identify the effects of anterior cruciate ligament reconstruction (ACLR) using the semitendinosus tendon autograft on lower body strength capacity and landing strategy in athletes who had returned to competition. Additionally, we sought to identify strength metrics that influenced landing characteristics previously identified as risk factors for primary or subsequent ACL injury. In our first study, plyometric-trained athletes performed bilateral vertical drop landings (VDLs) initiated with a step off technique with each limb serving as the leading leg. Peak vertical ground reaction force (Fz_{peak}) and impulse in the first 100 ms after ground contact ($Impulse_{100ms}$) was calculated for each limb under each lead leg condition. We identified that lead leg selection altered force-time characteristics and between limb symmetry, which may impact return to sport decision making after injury. In our second study, we recruited athletes with ACLR, non-injured, sport performance matched controls and non-injured, sport-matched but development-level controls to perform single leg landings from 25 cm and 50 cm heights and maximum voluntary contractions to assess strength about the knee and hip joints. The ACLR limb had knee flexion strength deficits compared to the contralateral limb and was stronger in hip abduction compared to the contralateral limb. The ACLR limb did not differ in any other comparisons, including across landing kinetics and kinematics. However, we observed main effects of strength on landing variables, highlighting the importance of lower body maximal strength on landing strategy. Together, these results suggest that it is important to use caution when assessing bilateral landing technique using VDL tasks, a common practice in clinical assessments following ACLR. Furthermore, we determined that lower body strength can largely be regained following ACLR, and as such movement strategies after ACLR can mimic that of healthy, elite athlete peers.

Preface

The following chapters are based on scientific manuscripts

Chapter 2 Lawson D., Jordan MJ., Herzog W. The Effects of Lead Leg Selection on Bilateral Vertical Landing Force-Time Characteristics: Implications for Rehabilitation. *Scandinavian Journal of Medicine & Science in Sports* [Under Review]

Chapter 3 Lawson D., Jordan MJ., Herzog W. The Relationship Between Lower Body Strength and Single-Leg Landing Performance of Elite Athletes After ACL Reconstruction. *Medicine & Science in Sports & Exercise* [In Preparation]

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Table of Contents

List of Tables

List of Figures

List of Abbreviations

ACL	anterior cruciate ligament
ACLR	anterior cruciate ligament reconstruction
AMI	arthrogenic muscle inhibition
ASIS	anterior superior iliac spine
BPTB	bone patellar tendon bone
DVJ	drop vertical jump
Fz	vertical ground reaction force
Fz_{peak}	peak vertical ground reaction force
$Impulse_{100ms}$	impulse in the first 100 ms after ground contact
IMU	inertial measurement unit
LME	linear mixed effect model
MVIC	maximum voluntary isometric contraction
RTS	return to sport
SSC	stretch shorten cycle
VDL	vertical drop landing

Chapter 1: Introduction and Literature Review

1.1 Anterior Cruciate Ligament Injury

The anterior cruciate ligament (ACL) serves as a passive restraint to anterior tibial translation relative to the femur, and as a secondary restraint against internal and external rotation of the tibia relative to the femur (Duthon et al., 2006). Additionally, the ACL is a sensory organ, providing proprioceptive feedback of knee joint position via mechanoreceptors. This proprioceptive feedback influences reflexive activation of muscles crossing the knee joint, and provides protection of the joint itself (Duthon et al., 2006). ACL rupture is a debilitating knee joint injury typically resulting from high energy absorption demand events, such as jump landing, pivoting or change of direction maneuvers (Bere et al., 2011; Koga et al., 2010; Krosshaug et al., 2007). ACL rupture often results in substantial pain, reconstructive surgery and a lengthy rehabilitation time period that prevents ongoing participation in sport (Westin et al., 2018). There is also an increased risk of knee joint osteoarthritis and subsequent knee joint traumatic injury following a primary ACL injury, despite reconstructive surgery (Barenius et al., 2014; Laboute et al., 2010; Lind et al., 2012).

Professional winter slope sport athletes (e.g. ski and snowboard alpine and freestyle, ski and snowboard cross, and ski jumping) have high incidences of knee injuries (0.62 per 1000 athlete-days) (Fu et al., 2020). Specifically, ACL injuries account for 13.6% of all injuries and 67.9% of knee injuries in alpine ski racing (Flørenes et al., 2012). Over a 25-year period, the French national alpine team averaged 6.1 ACL ruptures per season (injury rate of 5.7 ACL injuries per 100 skier-seasons) (Pujol et al., 2007). The epidemiology of ACL injury in freestyle skiing and snowboarding has not been studied to the same extent as in alpine ski racing, but the injury rates for freestyle skiing and snowboarding are comparable (Flørenes et al., 2012) or higher than in alpine skiing (Fu et al., 2020). In multi-directional team sports sport (i.e. basketball, soccer, rugby), the yearly prevalence of ACL injury is up to 3.67% (Beynnon et al., 2014; Moses et al., 2012). Contrary to field sports where clear sex-differences in ACL injury rates have been shown (Anderson et al., 2016; Arendt & Dick, 1995; Beynnon et al., 2014), there are no sex-differences in ACL injury rates amongst elite alpine ski racers (Flørenes et al., 2012). Here, sex-specific ACL injury risk factors that are thought to put the female athlete at higher risk for ACL injury in field sports (Anderson et al., 2016; Arendt & Dick, 1995; Beynnon et al., 2014) are superseded by the high-energy injury mechanism found in winter slope sports (Bere et al., 2011; Flørenes et al., 2012).

1.2 Non-Contact ACL Injury Mechanisms

A majority of ACL injuries occur in high energy, non-contact situations (e.g. pivoting, landing from a jump) and these have been thoroughly studied via video analysis in a variety of sports (Bere et al., 2014; Koga et al., 2010; Krosshaug et al., 2007; Waldén et al., 2015). In team field and court sports (i.e. basketball, rugby, handball), less than 20º of knee flexion at initial ground contact is consistently observed in non-contact ACL injury situations (Koga et al., 2010; Krosshaug et al., 2007; Montgomery et al., 2018). A decrease in knee flexion angle during high energy ground interactions can strain the ACL as result of the high compressive forces at the tibiofemoral joint (Beaulieu et al., 2021; Shin et al., 2007). The unique interaction of the tibial plateau and femoral condyles causes anterior translation and internal rotation of the tibia relative to the femur when the joint is compressed (Beaulieu et al., 2021). Further, a flexed hip position coupled with minimal knee flexion (i.e., $<20°$) often results in a posterior shift of the body's center of mass, which must be counteracted by strong internal knee extension moments generated by the quadriceps muscles (Boden et al., 2010). Strong quadriceps contractions with relatively small angles of knee flexion (i.e. <20º) may result in further anterior tibial translation due to the line of action of the patellar tendon (Demorat et al., 2004; Herzog & Read, 1993), but at larger angles of knee flexion the line of action of the patellar tendon provides synergistic protection of the ACL due to a posterior directed line of action (Aune et al., 1997; Bodor, 2001). Taken together, these data highlight the importance of the active stabilizers (i.e., thigh joint muscles) and kinematics (i.e., knee joint angles) in ACL injury events.

The ACL injury mechanism in winter slope sports like alpine skiing are distinct from field sports like soccer and basketball (Bere et al., 2011). For example, a common injury mechanism in alpine skiing includes hyperflexion of the knee joint upon landing, alongside anteriorly directed tibial shear forces coming from the ski (Bere et al., 2011). This has been called the "boot anterior drawer mechanism" in recreational skiers (McConkey, 1986) and more recently the "landing back weighted mechanism" in alpine ski racing (Bere et al., 2011). Here, backward-directed angular momentum at the point of jump takeoff causes the skier to lose balance in the backward direction during flight, resulting in a ski tail first landing. This causes forward rotation of the skis on initial ground contact while the skier continues to fall backward with combined knee valgus and internal rotation. Tibiofemoral compression and anterior motion of the tibia relative to the femur occurs as a result, placing sufficient strain on the ACL to cause a tear (Bere et al., 2011; Spörri et al., 2017).

Knee valgus, a knee joint position consisting of femoral adduction, knee abduction and tibial external rotation (Hewett et al., 2005; Utturkar et al., 2013), is often observed in non-contact ACL injury situations (Koga et al., 2010; Waldén et al., 2015). This body position has been identified as a potential risk factor for ACL injury in female athletes (Hewett et al., 2005) and female professional basketball players had a relative risk of knee valgus collapse 5.3 times greater than their male counterparts (Krosshaug et al., 2007). However, elongation of the ACL in knee valgus positions has been contested (Utturkar et al., 2013). Additionally, the pattern of bone bruising on the articulating surfaces of the tibia and femur shown via magnetic resonance imaging after acute ACL injury is more indicative of an impact (i.e. joint compression) mechanism than of a shear (i.e. knee valgus) mechanism (Boden et al., 2010). It is therefore debated if the knee valgus observed is a mechanism, or risk factor, for ACL injury or if it occurs as a result of a lack of passive resistance that the ACL provides in that movement once the ACL has torn (Utturkar et al., 2013).

1.3 The Relationship Between Vertical Landing Movement Strategy and Muscular Strength

Based on the scientific literature, ACL injury mechanisms are complex and sport-specific but involve failure of the passive restraint system (e.g., ligamentous tissues) (Beaulieu et al., 2021), consequent to high external energy injury mechanisms, inadequate protection from active restraints (i.e., muscles) and lower limb kinematics that load the ACL. As discussed, ACL injuries frequently occur during vertical jump landings; consequently, sport performance and sport medicine practitioners may consider conducting standardized landing assessments in a laboratory environment alongside neuromuscular testing to identify high-risk movement strategies and strength deficits that may predispose athletes to ACL injury, ACL reinjury or determine when they are sufficiently prepared for a high-risk sport environment (Buckthorpe & Della Villa, 2019; Myer, Paterno, et al., 2006a).

The vertical landing movement strategy is characterized by synergistic energy absorption of the ankle, knee and hip joints (McBride & Nimphius, 2020) and is often quantified by a combination of joint angle range of motion and force-time characteristic (James et al., 2003;

Lees, 1981; Minetti, 1998; Nordin et al., 2017). The goal of a landing is to dissipate the body's vertical linear momentum. Based on the impulse momentum relationship, this can be accomplished by decelerating the body's center of mass to a vertical velocity of 0 m/s (Lees, 1981; McNitt-Gray et al., 2001). According to Newton's second law of motion, the rate of change of the linear momentum of an object is directly proportional to the external forces applied to the object. Therefore, to perform a landing the lower extremities must apply force to the ground, and the muscles of the lower body must produce sufficient joint moments to decelerate the body mass (Minetti, 1998). Here, muscular strength at the hip, knee and ankle are thought to influence the movement strategy that is used to dissipate the external energy and the relative involvement of the hip, knee and ankle joints (McBride & Nimphius, 2020). Crucially, as it pertains to knee joint loading, participants with diminished lower body strength measured in a loaded back squat shifted toward a more knee dominant energy absorption strategy under landing conditions with increasing eccentric load. Conversely, stronger individuals shifted toward a hip dominant strategy, with decreased knee joint work (McBride & Nimphius, 2020). These results suggest that landing strategy, as measured by joint work, is related to lower body strength capacity. Further, theoretical modeling has identified that the maximum landing height one can safely land from (i.e., landing height tolerance) is explained by lower body strength capacity, whereby greater lower body maximal strength allows for safe landings from higher heights compared to weaker individuals (Minetti, 1998). Increased lower body eccentric strength capacity also provides a greater number of feasible landing strategies to employ in a given landing situation (i.e. landing at large knee joint flexion angles) (Minetti, 1998). Increased lower body strength capacity is therefore theorized to provide protection against injury in the event of suboptimal body position, like one may encounter in sporting contexts (Dicesare et al., 2020;

Minetti, 1998). Conversely, a mismatch of lower body strength capacity compared to the eccentric demands of a landing task may limit the feasible energy absorption and landing strategies available to an individual (Dicesare et al., 2020). As such, any loss of body position or misjudged timing could result in suboptimal energy absorption strategies and thus higher injury risk (Minetti, 1998).

One such strategy is landing 'stiff'. Stiff landing strategies are characterized by limited knee joint range of motion and high Fz_{peak} magnitudes (Butler et al., 2003; Devita & Skelly, 1992). Due to a short landing phase duration, there is a limited opportunity to attenuate the energy associated with the landing thus requiring large external forces (Fz_{peak}) to reverse the body's linear momentum (Minetti, 1998) . Such landings may be important in sport performance settings to optimize fast stretch shorten cycle (SSC) contractions, such as those that occur during a change of direction maneuver (Brazier et al., 2019; Butler et al., 2003; Maloney et al., 2016), but stiff landing strategies are also associated with ACL injury in untrained participants and youth athletes (Hewett et al., 1999; Leppänen et al., 2017). Conversely, low landing stiffness is also associated with soft tissue injuries, and potentially with knee ligament injury (Butler et al., 2003; Minetti, 1998). As such, a landing strategy characterized by moderate knee joint range of motion, and consequently a small to moderate internal knee extension moments and Fz_{peak}, may be optimal (Butler et al., 2003). However, the interaction between internal and external forces should be viewed in a context-specific manner, and with an aim to optimize the landing strategy for the task demands (e.g., landing in gymnastics versus alpine ski racing), while preventing excessive ACL loading.

Muscle strength influences landing strategy. For example, individuals with high quadriceps (Ithurburn et al., 2015) and hip external rotation maximal strength (Lawrence et al.,

2008) have been shown to land with decreased peak vertical ground reaction forces (Fz_{peak}) compared to their respective lower strength counterparts during single leg drop landings. Additionally, individuals with increased quadriceps (Ithurburn et al., 2015) and hamstring strength (Lephart et al., 2002) moved through greater knee joint range of motion in single leg landing tasks. Theoretically, increasing the knee joint range of motion allows for the total impulse of the landing to be applied over a longer duration compared to landings with limited knee joint range of motion. This strategy likely minimizes Fz_{peak} but places an increased demand on the knee extensors to eccentrically contract to control the body's center of mass (Podraza & White, 2010). For example, it has been shown that increasing knee flexion from less than 25º to greater than 50º during landing resulted in an 85% increase in knee extensor moment (Podraza & White, 2010). Consequently, individuals with higher quadriceps strength can use their strength to gradually decelerate their body through large ranges of motion and land with decreased stiffness (Lephart et al., 2002).

The demands placed on the lower body musculature to decelerate the body's center of mass are decreased in stiff landing strategies, potentially offering a viable landing when the eccentric demand is too large for an individual's strength capacity (Devita & Skelly, 1992). With the knee flexed <25º during single leg landings, the peak knee extensor moment was 0.3 ± 0.3 Nm/kg, while it increased to 1.9 ± 0.6 Nm/kg at knee flexion angles between 50 and 75º (Podraza & White, 2010). Landing with an average of 48º less knee flexion resulted in a 7-fold reduction in peak knee extensor moment, but increased Fz_{peak} by only 3 N/kg (Podraza & White, 2010). Landing in this extended knee position may also increase knee joint compression and anterior tibial translation due to the increase in Fz_{peak} magnitude and the line of action of the quadriceps and hamstrings muscle groups (Beaulieu et al., 2021; Herzog & Read, 1993). This may result in increased strain and risk of rupture of the ACL (Englander et al., 2019).

1.4 ACL Reconstruction Surgery

ACL reconstruction (ACLR) surgery is typically recommended following ACL rupture to restore knee joint stability and improve return to sport outcomes, particularly in young and active populations (Schreiber & van Eck, 2010; Siegel et al., 2012). ACLR using an autograft requires surgical transplantation of tissue from another area of the patient to be used as the new ACL graft. Two of the most common autograft techniques utilize either the medial hamstring tendon from the semitendinosus and gracillis muscles (hamstring autograft) or the middle third of the patellar tendon (bone patellar tendon bone - BPTB). Alternatively, an allograft, tissue transplanted from another person, may be used (Siegel et al., 2012). The use of autografts may be accompanied by comorbidities associated with the tissue harvest, such as diminished knee flexion strength (hamstring tendon autograft), or anterior knee pain (BPTB autograft) (Siegel et al., 2012). However, autografts are associated with up to 4 times lower rates of failure when compared to allografts (Siegel et al., 2012; The MARS Group et al., 2014; Widner et al., 2019). As such, autografts are the preferred graft type, particularly in sporting populations (The MARS Group et al., 2014), and the specific graft chosen is based off surgeon preference, concomitant injuries at time of ACL injury, and sport or lifestyle demands (Cerulli et al., 2013; Siegel et al., 2012; The MARS Group et al., 2014).

1.5 Lower Body Strength Deficits Following ACLR

Quadriceps strength deficits after ACLR have been well documented and its recovery is a key goal of rehabilitation (Buckthorpe et al., 2019). Following surgery, arthrogenic muscle inhibition of the quadriceps is present and diminishes the ability to recruit the quadriceps' high-threshold motor units during contractions (Rice et al., 2014). This manifests as a deficit in maximum voluntary isometric contraction (MVIC) strength following ACLR (Buckthorpe et al., 2019). Morris et al. (2021) found a decrease in quadriceps strength as measured by MVIC at 70º and 90º of knee flexion in ACLR athletes compared to sport-matched controls after semitendinosus autograft ACLR surgery, despite a return to sport. Additionally, quadriceps strength was diminished in the ACLR limb of elite ski racers when compared to the uninjured contralateral limb (22% asymmetry) and uninjured elite ski racers (15% deficit) more than 2 years after ACLR surgery (Jordan et al., 2015a). Beynnon et al. (2014) found that one year post ACLR, isokinetic quadriceps strength was diminished by 12-17% compared to the contralateral leg at 60º/sec, 180º/sec, and 240º/sec, regardless of graft type. These results are supported by a scoping review that identified between-limb quadriceps strength deficits of greater than 20% at 6-months post-ACLR that typically decrease in magnitude but are still apparent several years after ACLR surgery (Palmieri-Smith et al., 2008).

The hamstring muscle group is uniquely positioned to act synergistically with the ACL to resist anterior tibial translation and rotation of the tibia relative to the femur (Serpell et al., 2015). However, ACLR using the semitendinosus tendon autograft has a significant impact on hamstring strength, with deficits exaggerated at deep knee flexion angles (Konrath et al., 2016; Morris et al., 2021; Nomura et al., 2015). Knee flexion strength deficits ranged from 15% to 40% when comparing the ACLR limb to contralateral at 60º and 105º of knee flexion (Nomura et al., 2015). Elite ski racers demonstrated greater than a 14% deficit in knee flexion strength at 70º of knee flexion when the ACLR limb was compared to the contralateral limb despite being greater than 2 years post-ACLR surgery (Jordan et al., 2015a). Similarly, participants that underwent

ACLR with semitendinosus tendon autograft displayed a 10% asymmetry in knee flexion torque at high contraction velocity at 3-years post-ACLR (Beynnon et al., 2014).

Proximal strength at the hip can have downstream effects on control of distal joints, particularly in dynamic movements like those associated with ACL injury (Dix et al., 2018; Reiman et al., 2009). As such, frontal and horizontal plane strength deficits at the hip have been a large focus in the ACLR rehabilitation literature. Deficits in hip adduction strength have been identified in ACLR limbs following semitendinosus autograft surgery compared to control subjects with the leg in 15º and 30º of hip abduction (Hiemstra et al., 2005). The deficits, up to 44%, were identified as particularly hazardous because of the adductor group's role in modulating knee and hip position during pivoting and cutting motions (Hiemstra et al., 2005). Conversely, Thomas et al (2013) did not find post-operative differences in adduction strength between the ACLR limb reconstructed with a BPTB autograft and the contralateral limb. Interestingly, Thomas also found that the ACLR limb had greater hip adduction strength than the control participants (Thomas et al., 2013). The differences in results can likely be attributed to semitendinosus tendon harvest, which affects the pes anserinus where the semitendinosus inserts alongside the gracilis, a primary hip adductor, and sartorius, a weak abductor (Neumann, 2010). Thus, hip adduction strength deficits may be expected after ACLR using the semitendinosus tendon. Hip abduction strength deficits have been identified as a risk factor for primary and secondary ACL injury, and as such regaining strength in the muscle group is a rehabilitation priority (Buckthorpe & Della Villa, 2019; K. Ford et al., 2015; Myer et al., 2006; Reiman et al., 2009). Studies have repeatedly shown no significant deficit in hip abduction strength when comparing the ACLR limb to the contralateral limb (Fryer et al., 2019; Tate et al., 2017; Thomas et al., 2013), and ACLR limb to control subjects (Boo et al., 2018; Fryer et al., 2019; Noehren et al., 2014; Tate et al., 2017; Thomas et al., 2013). Indeed, these studies often find the ACLR to be slightly stronger than their comparison group, though the differences did not reach significance.

There are conflicting results on the impact of ACLR on hip external rotation strength. Noehren at al. (2014) found that there was no difference between the hip external rotation strength of the ACLR limb compared to control subjects (6.9 ± 4.4 Nm/kg vs. 6.4 ± 1.5 Nm/kg) approximately 7 months post-ACLR with either semitendinosus or BPTB autograft (2014). Conversely, Boo et al. (2018) found a significant deficit in hip external rotation strength on the ACLR limb compared to control subjects in young female athletes (mean age: 14.7 ± 1.0 years) 7.3 months after surgery. Graft type was not identified in this study (Boo et al., 2018). Further investigations have linked decreased hip external rotation strength with weak quadriceps after ACLR. Those with decreased quadriceps strength after ACLR using either BPTB or semitendinosus autograft had lower external rotation strength than control subjects (14.4 ± 3.0 vs. 17.5 ± 3.0 %BW). However, those with high quadriceps strength after ACLR did not show a decrease in hip external rotation strength compared to control subjects (Bell et al., 2016). The same study found no difference in hip internal rotation strength between ACLR limbs and control subjects, regardless of quadriceps strength deficits, but the scientific literature evaluating hip internal rotation strength following ACLR is highly limited.

1.6 Vertical Landing Assessment Methodology After ACLR

Assessments of vertical landing kinetics and kinematics following ACLR have been used frequently to identify functional deficits and asymmetries that may persist after ACLR (Buckthorpe & Della Villa, 2019; Burgi et al., 2019; Myer, Paterno, et al., 2006a) along with the drop vertical jump (DVJ) test (Paterno et al., 2010; Pedley et al., 2017). Here, a participant drops off an elevated platform and subsequently performs a maximal vertical jump while minimizing the time between drop landing and jump takeoff, relying greatly on their eccentric-concentric contraction coupling (stretch-shortening cycle - SSC) ability (K. R. Ford et al., 2003; Pedley et al., 2017). Due to the ability to identify risk of second ACL injury and link to sport performance, the DVJ is often used in landing assessment research (Paterno et al., 2010; Pedley et al., 2017). Typically, the kinetics and kinematics of the first landing of the DVJ are examined (Mueske et al., 2018; Paterno et al., 2010). However, muscle activation studies suggest the DVJ is an intensive lower body task (Ambegaonkar et al., 2011). Therefore, the DVJ may not be a suitable assessment technique until late in the rehabilitation period when adequate strength has been acquired (Chmielewski et al., 2006). As such, relying solely on the DVJ during ACLR rehabilitation may limit a practitioner's ability to identify early deficits in landing performance and strategy that could be addressed through targeted neuromuscular training (Nagelli et al., 2020). Vertical drop landings (VDL) are also used to mimic landings seen in sport by requiring the athlete to drop and stick on a single or dual force plate system to assess energy absorption strategy and between-limb force-time characteristics (Schot et al., 1994; Tran et al., 2015b). Due to the unidirectional nature of the task, the VDL may be a more pragmatic assessment to monitor across and inform the entire rehabilitation process, compared to the DVJ (Buckthorpe & Della Villa, 2019).

DVJ and VDL assessments can be performed in bilateral or unilateral conditions (Daoukas et al., 2019; Ithurburn et al., 2015; Johnston et al., 2018; Paterno et al., 2010). It is suggested that bilateral functional movement assessments may provide a better evaluation of movement strategy, whereby asymmetries in bilateral tasks may highlight learned disuse strategies especially following lower extremity injury (Cohen et al., 2020). Conversely, single leg tasks do not provide the opportunity to offload the previously injured limb by shifting load to the contralateral limb as can be seen in bilateral tasks (Cohen et al., 2020). As such, single leg tasks may provide insight into a limb's landing capacity. Therefore, both unilateral and bilateral landing assessments are encouraged throughout the ACLR rehabilitation process to understand both limb-specific landing capacity deficits and altered movement strategy (Cohen et al., 2020; Yeow et al., 2011). Although these tasks vary in their nature (i.e. unilateral versus bilateral; drop jump versus drop landing), each assessment type can provide insight into compensatory movement strategies or diminished capacities of the lower extremity following ACLR (Burgi et al., 2019; Collings et al., 2019).

Three main landing task initiation strategies are utilized in rehabilitation research: dropping off a platform (Ford et al., 2003; Hewett et al., 2005; Mueske et al., 2018), stepping off a platform (Decker et al., 2003; Devita & Skelly, 1992; Harry et al., 2017; Zhang et al., 2018) and dropping from hanging (James et al., 2007; Schot et al., 1994). Dropping off a platform includes a double leg forward hop from the elevated platform, whereby noticeable vertical displacement deems a landing trial invalidated (Ford et al., 2003). The step off technique requires the participant to begin standing on one foot on the platform and step forward. Limited evidence suggests the presence of differences in landing force-time characteristics between the leading and trailing leg when the step off technique is employed (Ball et al., 2010; Lim et al., 2020). However, these results were only observed in DVJ tasks and not in VDLs (Harry et al., 2017). The drop off and step off techniques also introduce the potential of changes in body center of mass position, introducing potential discrepancies between real drop height, determined by the impulse-momentum relationship, and platform height (Afifi & Hinrichs, 2012; Collings et al., 2019). To avoid these issues, a small number of scientific investigations have employed the hanging drop technique, in which participants are suspended over the ground by hanging onto a moveable bar and then release their grip to initiate the landing (James et al., 2007; Schot et al., 1994). Although this technique limits drop height variability, it requires large space and significant set up time.

1.7 Landing Kinetic and Kinematic Deficits Following ACLR

At the time of return to sport, young athletes with mixed ACLR autograft types were divided into high and low quadriceps strength based on between-limb strength symmetry (high symmetry: > 90%; low symmetry: < 85% symmetry) and assessed in the double leg drop jump. Those that had low quadriceps strength on their ACLR limb had large asymmetries in Fz_{peak} compared to ACLR subjects with high quadriceps strength and control subjects (Schmitt et al., 2015). Furthermore, the low quadriceps strength group's contralateral limb had significantly large Fz_{peak} magnitude compared to the contralateral limb of the high quadriceps strength group, suggesting a shift of strategy toward offloading the weaker and previously injured limb. However, 71% of the participants in the low strength group had a BPTB autograft, while only 24% in the high strength group had a BPTB (semitendinosus tendon autograft: n = 25/37) suggesting an effect of graft choice on the landing kinetics (Schmitt et al., 2015). Further, at 6 months post-ACLR participants have demonstrated lower Fz_{peak} on their affected side compared to the contralateral limb during bilateral drop jumps from 40 cm (Mueske et al., 2018). Similarly, Paterno found that female athletes had significantly higher Fz_{peak} magnitude on their contralateral limb (2.0 ± 0.6 N/BW) compared to the ACLR limb (1.5 ± 0.3 N/BW) during the initial landing in a 31 cm drop jump. Interestingly, the ACLR limb produced the same Fz_{peak} as each of the limbs of the control group (dominant: 1.6 ± 0.2 N/BW; nondominant: 1.5 ± 0.3 N/BW) indicating a stiffer landing strategy in the ACLR group (Paterno et al., 2007). Mueske et al. (2018) also found that peak knee joint flexion angle during landing was also diminished on the

ACLR limb versus the contralateral. The ACLR limb also exhibited decreased sagittal plane moments and energy absorption compared to the contralateral limb, and decreased knee and ankle energy absorption compared to control subjects (Mueske et al., 2018).

The ACLR limb has demonstrated kinetic and kinematic changes in single leg landing tasks when compared to contralateral and control limbs (Johnston et al., 2018). Here, reductions in peak knee internal extensor moment and knee joint range of motion have been identified in the ACLR limb (Johnston et al., 2018) alongside elevated Fz_{peak} magnitude compared to control subjects (Rocchi et al., 2018). Additionally, Ithurburn et al. (2015) observed diminished knee flexion range of motion and internal knee extensor moment during single leg landings on the ACLR limb compared to the contralateral limb. Between-limb landing asymmetries were greater in ACLR subjects that also demonstrated large between-limb quadriceps strength asymmetry (Ithurburn et al., 2015). Sixty-four participants from the same cohort were assessed two years following return to sport and showed a significant improvement in knee range of motion asymmetry. However, knee flexion range of motion on the ACLR limb was not increased in the two-year period. Instead, the reduction in asymmetry was due to a decrease in the range of motion of the contralateral limb, which may be a detrimental trend (Ithurburn et al., 2019).

After ACLR, sagittal plane trunk flexion asymmetry was also observed during single leg landing (Ithurburn et al., 2015). ACLR subjects that demonstrated large quadriceps maximal strength deficits presented with more trunk flexion asymmetry during landing than those with less quadriceps strength asymmetry. Here, large trunk flexion range of motion was observed in ACLR subjects that had diminished maximal quadriceps strength on their ACLR limb, suggesting a hip dominant strategy to offload the knee (Ithurburn et al., 2015). At a two year follow up, trunk flexion asymmetry was reduced, suggesting a transition away from the hip dominant, knee offloading strategy that was apparent at RTS (Ithurburn et al., 2019). Conversely, during a single leg drop vertical jump task at 21 months post-ACLR surgery, Vairo et al. (2008) observed increased hip flexion angles at initial ground contact on the ACLR side compared to the contralateral limb and matched controls. Interestingly, there was a trend toward increased knee flexion at ground contact on the ACLR limb versus contralateral limb and controls, and significantly greater knee flexion at Fz_{peak} when comparing the ACLR limb to controls (Vairo et al., 2008). The results from Ithurburn et al. (2019) and Vairo et al. (2008) demonstrate that compensation patterns after ACLR may be specific to the task (i.e. single leg drop landing versus single leg drop jump) and ACLR autograft type. Ithurburn's participants consisted of a mix of graft types (semitendinosus: 61%; BPTB: 30%; allograft: 8%) compared to only ipsilateral semitendinosus plus gracilis tendon autografts in Vairo's study (Ithurburn et al., 2019; Vairo et al., 2008).

Investigations using mixed graft types have also identified significant relationships between quadriceps maximal strength, hip abduction maximal strength and frontal plane compensations during single leg drop landings, measured as frontal plane trunk lean (Fryer et al., 2019). Additionally, frontal plane trunk lean was elevated in the ACLR group compared to controls (ACLR symmetry index: 143 ± 47% vs. Control symmetry index: 106 ± 33%) indicating altered frontal plane control following ACLR (Fryer et al., 2019). Lateral trunk positions may increase knee abduction loads, which are associated with ACL injury (Beaulieu et al., 2021; Myer et al., 2008). Trunk and pelvis control during dynamic tasks has been linked to strength at the hip (Malloy et al., 2016) and this study found a small but significant correlation between hip abduction strength and frontal plane trunk excursion in the ACLR limb (Fryer et al., 2019).

1.8 Return to Sport After ACLR and Subsequent ACL Injury Rates

There is no consensus on when an athlete should return to sport following ACLR and returning to sport after primary ACL injury is a major risk factor for subsequent reinjury (Barber-Westin & Noyes, 2011; Burgi et al., 2019; Grindem et al., 2016), suggesting the relevance of expansive neuromuscular testing, including assessments of muscle power, plyometric ability and landing capacity to identify trainable neuromuscular deficits (Buckthorpe & Della Villa, 2019). Grindem et al. (2019) determined that returning to pivoting, cutting and jumping sports after ACLR increased the risk of future ACL injury by greater than 4 times. Furthermore, the risk of a subsequent ACL injury for athletes returning to sport after primary ACLR was 15 times higher than the risk of previously uninjured athletes suffering their first ACL injury (Paterno et al., 2012). As such, the timing, and criteria for RTS clearance after ACLR has garnered much attention in the sport science community.

RTS prior to the 9-month post-ACLR mark is hazardous for athletes, particularly those competing in jumping and pivoting sports (Grindem et al., 2016; Laboute et al., 2010). Returning to sport within 7 months of ACLR surgery led to a significantly greater risk of future ACL injury than returning after the 7-month mark (15.3% vs. 5.2%) (Laboute et al., 2010), while other studies have identified a 51% reduction in reinjury rate for each month RTS is delayed up to 9 months (Grindem et al., 2016). A high fraction of secondary ACL injuries (ipsilateral or contralateral) occur in the first 2 years following ACLR, representing an especially risky time period for athletes that have returned to sport (Lind et al., 2012). Although time from surgery has a major influence of risk of future injury, RTS decision making is increasingly reliant on objective neuromuscular assessments. Lower body muscle strength and power assessed via dynamometry, single and bilateral jumping and landing performance, and muscle mass measures are frequently cited as assessments used to determine RTS readiness (Barber-Westin & Noyes, 2011; Burgi et al., 2019; Jordan et al., 2020). A between-limb symmetry index of greater than 85-90%, return to pre-injury levels of performance and comparisons against normative data from the same sport and level are often used as criteria to pass RTS testing batteries (Burgi et al., 2019; Jordan et al., 2020).

Despite greater attention on restoring neuromuscular and movement deficits after ACLR, between 20-45% of athletes do not return to sport after ACL surgery (Ardern et al., 2014; King et al., 2020; Lai et al., 2018). Unfortunately, studies assessing RTS success after ACLR often use RTS as a dichotomous outcome, with participation in one game or competition considered a RTS (Lai et al., 2018). Although up to 80% of athletes return to sport following ACLR, they are still at risk of subsequent knee joint injury. Rates of ACL injury following RTS has been reported between 3-33% (King et al., 2020; Laboute et al., 2010; Lind et al., 2012; Webster & Feller, 2018) with an increased risk in younger athletes (Lind et al., 2012; Webster et al., 2014). There are mixed results when comparing second ACL injury risk between graft types. King found an increased risk of graft rupture following hamstring tendon autograft (King, Richter, Franklyn-Miller, Daniels, Wadey, Jackson, et al., 2018), while other studies have concluded that hamstring autografts may have a higher rupture rate than BPTB grafts despite the result failing to reach significance (Laboute et al., 2010; Pinczewski et al., 2007; Widner et al., 2019). However, higher rupture rates following hamstring autograft is not always the case (Webster & Feller, 2018). Risk of contralateral ACL injury is consistently elevated in those that received a BPTB graft (King, Richter, Franklyn-Miller, Daniels, Wadey, Jackson, et al., 2018; Pinczewski et al., 2007).

Additionally, simply returning to sport is not the desired outcome for many athletes. Athletes want to return to their previous, or higher, level of competition following injury (Burgi et al., 2019). Though up to 80% percent of elite athletes return to their sport following ACLR, 60% of elite athletes do not return to their pre-injury level of performance (Lai et al., 2018). In winter slope sports, such as alpine ski racing and freestyle skiing, return to high performance is more common after ACLR. As many as 50% of the skiers that made the top 30 in the world had suffered an ACL injury in their career (Pujol et al., 2007). Assessment of X Games skiers, an invite-only freestyle skiing competition widely regarded as the highest level of the sport outside of the Olympics, found that 87% competitors suffered a previous ACL injury in their career (Erickson et al., 2013). It is certainly possible to have a long and successful career following ACLR. However, approximately 20% of athletes do not return to sport after ACLR and a large percentage of those that do return to sport do not return to their previous level of competition (Lai et al., 2018). As such, a better understanding of the long-term, trainable deficits associated with ACLR is required to improve rehabilitation, RTS outcomes and sport performance following ACLR.

1.9 Summary and Purpose

While lower body strength and energy absorption strategy, as measured by landing performance, following ACLR appears to be a key consideration for safeguarding against future ACL injury, a comprehensive evaluation of strength at the knee and hip joints in conjunction with landing performance in an elite athlete population is lacking.

The primary purpose of this investigation was to first determine the effect of ACLR on lower body strength and landing strategy. A secondary purpose was to quantify the effect of lower body maximal strength at the knee and hip on the landing strategy of elite athletes that have undergone an ACLR and returned to sport. This was done by conducting two separate studies shown below in Chapter 2 and Chapter 3. We first attempted to identify if a bilateral drop landing was a sufficient task to quantify force-time characteristics of the landing phase. We then quantified the hip and knee joint strength and single leg landing strategy at two landing heights of elite ACLR athletes that had returned to sport, healthy elite athletes, and development-level healthy athletes. Further, the relationships between strength and landing strategy, as measured by landing phase knee joint angles and force-time characteristics, were explored in these populations. It was hypothesized that the bilateral drop landing tasks would result in limb biased force-time characteristics. Furthermore, we hypothesized that the ACLR limb would display strength deficits at both the knee and hip in comparison to the contralateral limb and elite control athletes. As such, we also hypothesized the ACLR limb would demonstrate a stiffer landing strategy, characteristic of increased risk of subsequent ACL injury.

Chapter 2: The Effects of Lead Leg Selection on Bilateral Vertical Landing Force-Time Characteristics: Implications for Rehabilitation

2.1 Introduction

The bilateral vertical drop landing (VDL) and drop vertical jump (DVJ) tests are commonly used to assess anterior cruciate ligament (ACL) injury risk (Bates et al., 2013b; Paterno et al., 2010) and return to sport readiness after ACL reconstruction (ACLR) surgery (Myer, Paterno, et al., 2006b). Both tasks require participants to drop off of an elevated platform and either land and stabilize their body as in the VDL (Tran et al., 2015b) or immediately perform a maximal effort vertical jump following landing (DVJ) (K. R. Ford et al., 2003). These tests assess lower extremity energy absorption capacity, landing strategy and in the case of the DVJ, eccentric-concentric contraction coupling or stretch-shorten cycle ability (Paterno et al., 2010; Schmitt et al., 2015).

There are three common techniques used to initiate VDL and DVJ tasks, a step-off, double leg forward "drop off", and hanging drop. The step-off technique requires the participant to begin standing on an elevated platform and take a step forward to initiate the landing task (Collings et al., 2019; Decker et al., 2003; Devita & Skelly, 1992; Harry et al., 2017; Zhang et al., 2018). Alternatively, researchers may implement a double leg forward drop, which includes a small forward hop off the elevated platform (K. R. Ford et al., 2003; Hewett et al., 2005; Mueske et al., 2018). In both cases, there may be a discrepancy between actual drop height, determined by center of mass touchdown velocity, and platform height (Afifi & Hinrichs, 2012). Participants may lower their center of mass while using the step-off technique or may introduce a vertical component to their double leg forward drop (James et al., 2007; Schot et al., 1994). Limited research has been done using the hanging drop technique, where the participant holds onto a bar, suspended over the ground and then releases their grip to initiate the landing (Holsgaard-Larsen et al., 2014; Jordan et al., 2015b; Read et al., 2020). Taken together, the influence of task initiation strategy on VDL and DVJ outcomes has not been elucidated.

To assess between-limb capacities in an ACLR population, variations of the vertical jump and landing have been conducted using a dual force plate system, permitting simultaneous measurement of the vertical ground reaction force (Fz) produced by each limb with a dual force plate system (Jordan et al., 2015b, 2020; Taberner et al., 2020). Here, a comparison of the right versus left kinetic impulse measured as a between-limb asymmetry index has been proposed as a pragmatic and sensitive assessment of the post-injury recovery period after ACLR in an athlete population (King, Richter, Franklyn-Miller, Daniels, Wadey, Moran, et al., 2018; Mueske et al., 2018; Nagelli et al., 2018; Smeets et al., 2020). Kinetic analysis of vertical landings using dual force plate systems (Buckthorpe & Della Villa, 2019) is also commonplace but the effect of the landing test protocol itself on the force-time outcome measures are unclear. As the post-injury recovery period may unfold differently between athletes, and a criteria-based approach for return to sport decision making has been recommended (Buckthorpe & Della Villa, 2019), including the use of expansive neuromuscular testing such as vertical landing capacity (Barber-Westin & Noyes, 2011; Burgi et al., 2019; Myer, Paterno, et al., 2006b), the influence of the landing protocol on measures of between-limb asymmetry is crucial as clinical thresholds are often used to guide return to sport decision making (Ball et al., 2010; Lim et al., 2020).

The influence of the step off initiation technique on kinetic variables has been explored more in DVJ tasks than VDLs. It has been found that the lead leg contributes a large fraction to the force-time characteristics (Fz_{peak}, Impulse) compared to the trail leg during DVJs, although this is not consistent across drop heights (Ambegaonkar et al., 2011). Though these studies add to the understanding the impact of protocol selection on kinetic variables, a DVJ is a different task than a VDL and thus the generalizability of the results to VDLs may be limited (Harry et al., 2017). Harry identified there were no differences in Fz_{peak} between the lead and trail leg during a VDL from 37 cm. However, the authors suggested the discrete time point nature of Fz_{peak} may have masked force-time characteristics during the landing task, and that the landing height may not have been sufficient to draw out kinetic differences between limbs (Bates et al., 2015).

Landing kinetics are also known to independently influence ACL stress and strain (Hewett et al., 2005; Leppänen et al., 2017) and have been proposed as potential risk factors for ACL injury (Norcross et al., 2013). Here, large peak Fz (Fz_{peak}) magnitude during landing is associated with increased strain on the ACL via increased anterior tibial shear force and is associated with ACL rupture (Norcross et al., 2010). The first 100 ms after ground contact has previously been identified as the initial impact phase of landing and high energy absorption in this phase is positively correlated with landing strategies associated with ACL injury (Bates et al., 2015; Hewett et al., 2005; Norcross et al., 2013). Limb-specific, landing-phase impulse metrics allow for the characterization of the demands placed on each limb over the time frame associated with non-contact ACL injury, which occurs less than 100 ms following ground contact (Harry et al., 2017; S. Nigg et al., 2013). While the overall aim of a vertical jump landing is to reduce the jumper's downward momentum to zero, discrete-phase analysis of the landing impulse over time frames relevant to ACL injury (i.e. impulse in the first 100 ms – $Impulse_{100ms}$) may provide additional insight into landing capacity alongside characterization of the force-time curve over a large segment of the landing period compared to discrete timepoint analysis (e.g. the instantaneous peak vertical ground reaction force) (Tran et al., 2015b).

Given the paucity of scientific evidence on the influence of leg selection for vertical landing tasks, the relationship between landing kinetics and ACL injury and the fact landing tests are frequently recommended for return to sport decision making using clinical thresholds, the purpose of this study was to quantify the effect of lead leg selection on the timing of ground contact, peak vertical ground reaction force and early phase kinetic impulse ($Impulse_{100ms}$) during a VDL initiated with a step-off technique in trained population. We hypothesized that the leading leg would make ground contact before the trail leg and contribute more to the peak vertical ground reaction forces and energy attenuation demands.

2.2 Methods

2.2.1 Participants

Thirteen athletes (males: n=8, females: n=5; age = 23±3 years; body mass = 74±11 kg; preferred kicking leg: right: n = 12, left: n = 1) who had regularly participated in plyometric training in the previous 4 months were recruited. Participants with history of a lower body orthopaedic injuries and lower back injuries in the previous 6 months were excluded. The Conjoint Faculties Research Ethics Board at the University of Calgary approved the experimental protocol, and all participants gave written informed consent to participate in this study.

2.2.2 Test Procedure

Prior to testing, participants performed a self-selected lower body and trunk warm up consisting of aerobic activity and dynamic stretching. Participants stood still on a dual force plate system (Accupower Force Platform, AMTI, Watertown, Massachusetts, USA) to obtain a 5 s body weight recording. Participants then stepped onto a 45 cm box located 5 cm behind the force plates with their feet placed approximately hip width apart and performed six vertical drop landings (VDLs) onto the force plates. Participants were instructed to step off the box without bending the trail leg and to achieve a half squat landing position (approximately 90º of knee flexion) "as quickly and as softly" as possible (Peng, 2011). Participants maintained the stationary half squat landing position until they were cued by the tester to step back onto the box. A 60 s rest period was provided between landings. The initial lead-leg order was randomized, and subjects were required to perform 3 trials with one lead-leg prior to switching to the other leg. Previous investigations have shown good reliability of landing kinetic variables utilizing 3 trials (Norcross et al., 2013; Rocchi et al., 2018). All testing was conducted by the same researcher who was a certified exercise professional. The researcher verified the landing position using a high-speed video camera that provided a sagittal plane view of the participant. Trials that did not adhere to the study protocol were discarded and repeated. The brand and style of the footwear worn by the participants was not controlled but athletic training shoes were required. The position of the arms was not controlled.

2.2.3 Data Collection

Simultaneous data collection of the vertical ground reaction forces (Fz) from the right and left force plates was performed at a sampling frequency of 1500 Hz (MyoResearch Version 3.14, Noraxon, Scottsdale, Arizona, USA) and analyzed with custom-built computer scripts (Matlab Version R2019B, Mathworks, Natick, Massachusetts, USA). The raw Fz voltage signals were converted to Newtons. The participant's system mass was determined using the total Fz (sum of left and right Fz) over a 500 ms interval. The initial landing ground contact timepoint was determined separately for the left and right limbs when Fz reached a 10 N threshold (Bates et al., 2013b; Norcross et al., 2013; Rocchi et al., 2018). The termination of the landing period was determined as the timepoint when the total Fz returned within ± 5% of the participant's baseline body weight calculated over a 1000 ms time period (Impellizzeri et al., 2007; Jordan et al., 2015b). The peak vertical ground reaction force (Fz_{peak}) and impulse in the first 100 ms after initial ground contact ($Impulse_{100ms}$) were determined independently for the right and left limbs and body mass normalized.

2.2.4 Statistical Analysis

Data were averaged across the 3 trials with each leg leading. A two factor (2 lead leg conditions x 2 limbs) within-subject factors ANOVA was performed to evaluate the effect of lead leg selection on limb-specific $Impulse_{100ms}$, Fz_{peak} and foot initial ground contact timing. When an interaction effect was identified post hoc comparisons were performed used paired t-tests with Bonferroni adjustment. Shapiro-Wilks tests confirmed normality and Levene's tests assessed homogeneity of variances. Limb symmetry was determined for Fz_{peak} and $Impulse_{100ms}$ (Impellizzeri et al., 2007; Jordan et al., 2015b) as:

$$Limb\ Symmetry\ Index = \frac{1 - (Lead\ Leg - Trail\ Leg)}{Maximum\ of\ Lead\ or\ Trail\ Leg} \times\ 100\%$$

Statistical analysis was carried out using RStudio (Version 1.2.1335). Statistical significance was set at a=0.05 and all data are presented as mean ± 1 SD, unless otherwise noted.

2.3 Results

A main effect for lead leg selection on Fz_{peak} was found [Lead Leg: 2.29 ± 0.58, Trail Leg: 1.98 ± 0.51 N/BW; $F(1,12) = 8.0$, $p = 0.015$] and a significant interaction between lead leg selection and limb-specific Fz_{peak} was found [$F(1,12) = 13.8$, $p = 0.003$] (Table 2.1). Pairwise comparisons using a t-test with Bonferroni adjustment showed a significant difference between right leg (2.49 ± 0.67 N/BW) and left leg (1.97 ± 0.39 N/BW) Fz_{peak} when the right leg was leading ($p = 0.01$). A significant interaction between lead leg selection and limb-specific impulse was found [$F(1, 12) = 27.6$, $p < 0.00$]). Therefore, the effect of lead leg selection was analyzed for each limb via pairwise comparisons. Paired t-tests identified a significant difference between right and left leg $Impulse_{100ms}$ when the right leg was leading (Right Leg: 0.12 ± 0.01; Left Leg: 0.09 ± 0.02; $p<0.01$) (Table 2.2). Visualization of between-limb difference in $Impulse_{100ms}$ and Fz_{peak} based on lead leg selection are presented in Figure 2.1. There was a significant interaction effect between lead leg selection and limb-specific ground contact timing [$F(1,12) = 16.965$, $p = 0.01$] and pairwise comparison showed that the right leg made ground contact significantly prior to the left leg when the right leg was leading (Right Leading: difference: 7.6 ± 5.1 ms; $p<0.001$; Left Leading: mean difference: 4.2 ± 10.1 ms; $p = 0.16$) as presented in Figure 2.2.

Lead leg stratified and pooled limb symmetry indexes are presented in Figure 2.3. Only 2 participants had limb symmetry values between 90 and 110% for $Impulse_{100ms}$ with the left leg leading (mean limb symmetry index = 87.8 ± 24.8%), while zero participants had a symmetry value between 90 and 110% with the right leg leading (mean limb symmetry index = 78.1 ± 16.0%). Similarly, 4 participants had a Fz_{peak} limb symmetry index between 90 and 110% with the left leg leading (mean limb symmetry index = 93.4 ± 20.3%), and 1 participant was within those limits with the right leg leading (mean limb symmetry index = 82.3 ± 21.7%). Pooled landing data resulted in mean limb symmetry indexes of 82.9 ± 12.1% and 87.9 ± 10.2% for $Impulse_{100ms}$ and Fz_{peak}, respectively.

Table 2.1 Comparison of Lead Leg and Trail Leg relative Impulse$_{100ms}$ and Fz$_{peak}$ (mean ± SD).

Variable	Lead Leg	Trail Leg	Effect Size
Impulse$_{100ms}$ (N•s/BW)	0.12 ± 0.01	0.10 ± 0.02	0.13
Fz$_{peak}$ (N/BW)	2.29 ± 0.58	1.98 ± 0.51*	0.40

Effect size calculated as partial Eta squared. * Different from lead leg p<0.05

Table 2.2 Right versus left landing kinetics by lead leg selection

	Impulse$_{100ms}$ (N•s/BW)		**Fz$_{peak}$ (N/BW)**	
	Left Leg	Right Leg	Left Leg	Right Leg
Left Leading	0.11 ± 0.01	0.10 ± 0.02	2.08 ± 0.40	1.99 ± 0.64
Right Leading	0.09 ± 0.02	0.12 ± 0.01**	1.97 ± 0.39	2.49 ± 0.67*

* Difference between left and right legs p<0.05
** Difference between left and right leg p<0.01

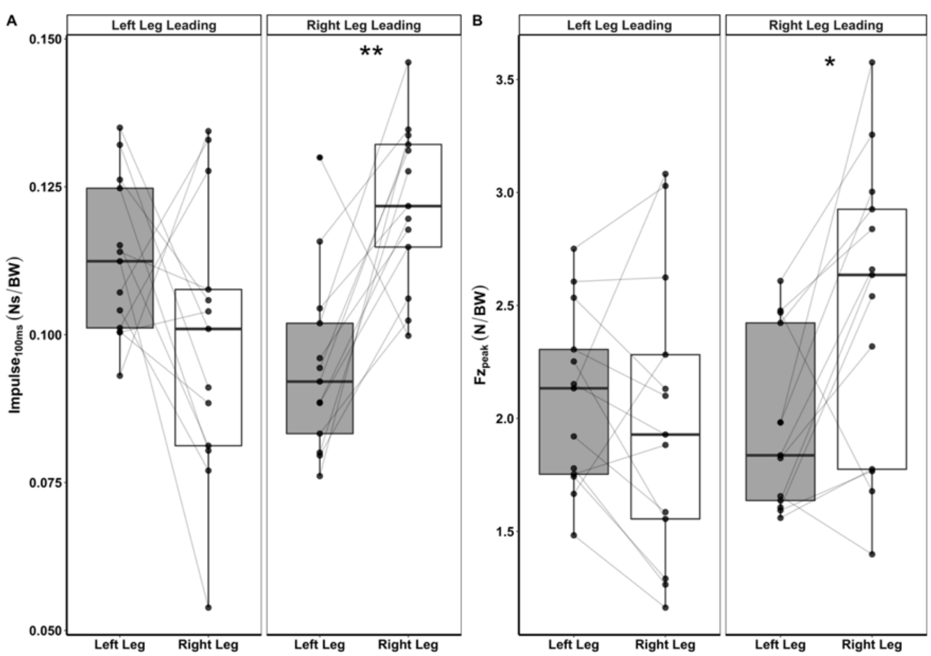

Figure 2.1 Impulse$_{100ms}$ and Fz$_{peak}$ by Lead and Trail Leg Selection. Pairwise comparison of the mean relative Impulse$_{100ms}$ (Panel A) and Fz$_{peak}$ (Panel B) experienced by the right and left legs, separated by lead leg selection for each subject. White boxes represent the right leg and grey boxes represent the left leg. * Difference between left and right leg p<0.05. ** Difference between left and right leg p<0.01

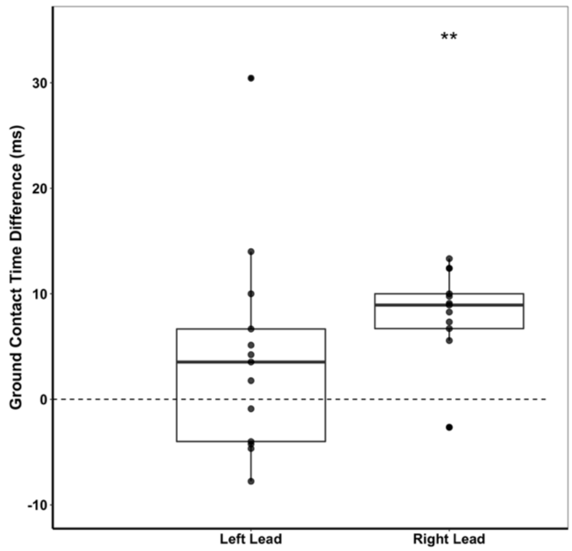

Figure 2.2 Limb Initial Ground Contact Time Difference Separated by Lead Leg Selection. Difference was calculated by lead leg contact time subtract trail leg contact time. A positive value indicates that the lead leg made ground contact prior to the trail leg. Dashed line indicates no difference in contact time between lead and trail limbs. ** A significant difference in ground contact times between left and right limbs with the right leg leading off the box, p<0.01

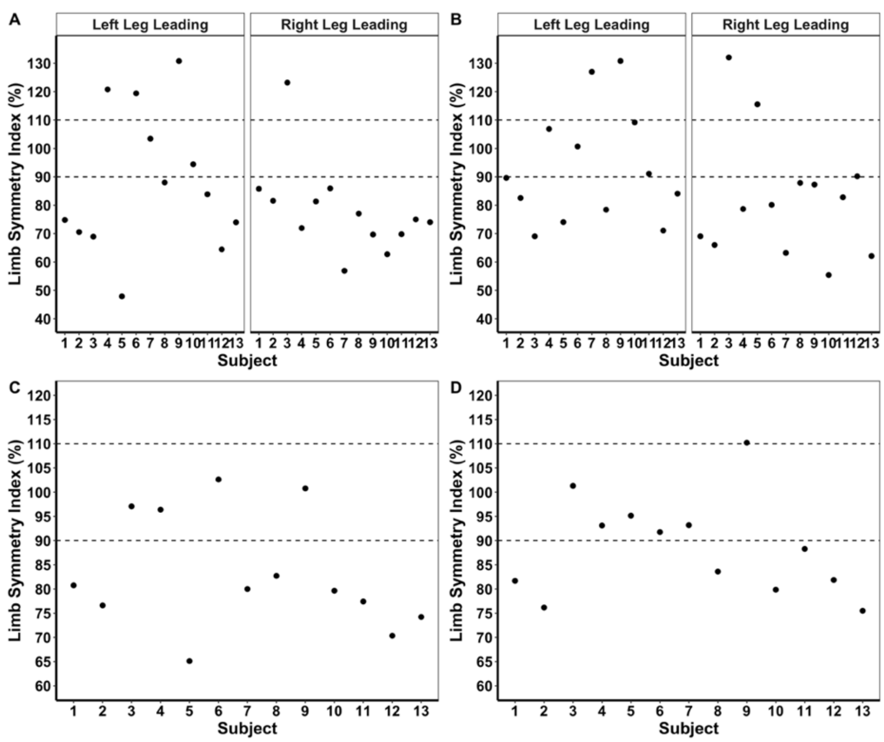

Figure 2.3 Limb Symmetry Index by Lead Leg Separated and Pooled Data. Average limb symmetry index of Impulse$_{100ms}$ (A) and Fz$_{peak}$ (B) separated by lead leg selection. Right and left lead leg pooled Impulse$_{100ms}$ (C) and Fz$_{peak}$ (D) limb symmetry indices are also presented. Values

below 100 indicate lead leg dominance. Limb symmetry values between 90 and 110% suggest non-problematic differences between limbs and are indicated by the dashed lines.

2.4 Discussion

In this study, we quantified the differences in the initial ground contact timing, Fz_{peak} and $Impulse_{100ms}$ of the lead versus the trail legs during a VDL using the step-off technique. Additionally, we assessed the effect of lead leg selection on limb specific Fz_{peak} and $Impulse_{100ms}$ by comparing these metrics between legs in both lead leg scenarios. We showed that the between-limb symmetry index is altered by lead leg selection in an athlete population. The data presented has practical relevance to practitioners and clinicians who employ the VDL as a part of an ACL injury/reinjury risk stratification using a between-limb symmetry index, and the results highlight the importance of controlling for the effects of lead leg selection. We also provide a new perspective on the VDL test by including the VDL landing kinetic impulse alongside Fz_{peak} in the limb symmetry index outcome measure.

The VDL and DVJ are frequently used to assess ACL injury risk and return to sport readiness in athletes (Bates et al., 2013b; Collings et al., 2019; Hewett et al., 2005). Further, a between-limb symmetry index < 90%, or >110% in the in Fz_{peak} and $Impulse_{100ms}$ is often used as a threshold to identify an increased risk of lower-extremity injury or to delay return to sport (Harry et al., 2017). On average the participants in our study displayed a limb symmetry index lower outside of 90-110% (i.e., outside the typical clinical threshold) reflecting lead leg dominance. Consequently, if a practitioner or clinician were not controlling for the lead leg selection in the VDJ, they may incorrectly identify the presence of a concerning limb symmetry index. Not only might this lead to the inappropriate use of neuromuscular training programs designed to address a limb-specific deficit but also there exists the possibility that a true landing capacity deficit is missed.

In agreement with our hypothesis, the lead leg made initial ground contact prior to the trail leg and contributed a higher fraction of the Fz than the trail leg when right and left leading trials were pooled (Fz_{peak}: limb symmetry index = 87.9%; $Impulse_{100ms}$: limb symmetry index = 82.49%). Contrary to our hypothesis, the demands on the lead leg were not consistently greater than the trail leg when the results were stratified by left and right leg leading trials. With the right leg leading, the right leg experienced significantly greater Fz_{peak} (limb symmetry index = 82.3%) and $Impulse_{100ms}$ (limb symmetry index = 78.1%) than the left leg. The right leg also made initial ground contact prior to the left leg when the right leg was leading. However, when the left leg led off the platform there were no significant between limb differences in ground contact timing, $Impulse_{100ms}$ or Fz_{peak}.

Twelve of the 13 participant identified their right leg as their preferred kicking leg. Therefore, it is warranted to present the results of the one left leg preferred participant. This participant followed the results of the rest of the participants. Fz_{peak} was greater on the lead leg compared to the trail leg with the right leg leading (right leg: 2.5 N/BW vs. left leg: 1.6 N/BW). With the left leg leading Fz_{peak} was greater on the trailing (right) leg than on the leading leg. Similarly, $Impulse_{100ms}$ was greater on the leading leg than trailing leg when the right leg led (right leg: 0.13 Ns/BW vs. left leg: 0.08 Ns/BW), but there was no difference between limbs with the left leg leading (left leg: 0.10 Ns/BW vs. right leg: 0.10 Ns/BW). Finally, with the right leg leading it made ground contact prior to the left leg (mean difference: 12.4 ms), while there was a smaller difference in ground contact time with the left leg leading (mean difference: 1.8 ms). As such, despite preferring to kick with the left leg this subject did not differ from the rest of the participants in terms of the impact that lead leg selection had on force-time characteristics in bilateral landing.

Our study contributes practical knowledge to clinicians and practitioners on the effects of lead leg selection on VDL kinetics in a trained population and diverges from the findings of others in the literature. One previous study (Harry et al., 2017; S. Nigg et al., 2013) investigated the kinetic and kinematic differences between the lead and trail leg using a step off technique (mean VDL height: 36.7 cm). The subjects in that study performed 15 VDLs with a consistent lead leg. Contrary to our current findings, the investigation found no differences in Fz_{peak} between lead and trail legs. The authors suggested that null result may be due to the discrete nature of Fz_{peak}. Discrete time point analyses may mask between-limb differences across the force-time curve (Ball et al., 2010; Lim et al., 2020). Previously, it has been shown that Fz_{peak} is asymmetrical between legs at low drop heights and more symmetrical at high drop heights (>40 cm) when utilizing a step off technique (Dufek & Bates, 1990; Podraza & White, 2010). Therefore, the lack of differences shown by Harry and the differences found in the present study are surprising. The contradictory results may be due to participants in Harry's study self-selected their leading leg and it remained consistent throughout. Conversely, in the present investigation participants performed trials with each limb leading. The results of our current investigation suggest that between-limb differences in Fz_{peak} are apparent with one leg leading during a VDL but not the other. Kinematic analysis of trials in the study by Harry et al. revealed that the leading leg made ground contact with a decreased knee flexion angle compared to the trail leg. No between limb differences in hip and ankle angles were apparent. While a decreased knee flexion angle at initial ground contact during landing tasks is associated with larger Fz_{peak} magnitudes (Harry et al., 2017), no differences were found in Fz_{peak} . The authors speculated that

Fz_{peak} between-limb differences may have been masked by lead leg selection (Ambegaonkar et al., 2011; Collings et al., 2019), which is in agreement with the results from our present study.

Studies that analyse the first landing of a drop vertical jump movement also provide indirect comparisons for our results. However, it is acknowledged that a subsequent movement after landing, such as the vertical jump in a DVJ, alters the first landing performance (Ambegaonkar et al., 2011; Collings et al., 2019). Paterno et al. (2020) also assessed Fz_{peak} from the right and left limbs in the landing phase of a DVJ using a drop off technique. ACLR participants landed with a significant between limb difference in Fz_{peak} (ACLR limb: 2.0 ± 0.6 N/body weight (BW), uninvolved limb: 1.5 ± 0.3 N/BW). Paterno found that uninjured controls showed no between limb differences in Fz_{peak}. Our present findings showed between-limb differences when the landing task was initiated with the right leg, but no significant differences were apparent with left leg leading. Lim identified between limb Fz_{peak} differences of 15.5% with the leading leg contributing a greater fraction during drop jumps from 46 cm initiated by a step off (Lim et al., 2020). This result is similar to our results which showed a 12.1% difference in Fz_{peak} when left and right leading trials were combined, 17.7% difference when the right leg led and 6.6% difference with the left leg leading. Lim also identified that the lead leg made ground contact 9 ms prior to the trail leg (Lundgren et al., 2015; Tran et al., 2015b; Yeow et al., 2011), which was similar to our finding when the right leg lead off the platform (mean difference: 7.6 ms) but more than double the difference we observed with the left leg leading. Therefore, initiation of the drop landing via “dropping” rather than stepping off the box may explain the discrepancy between our results and Paterno’s. A number of other studies have assessed Fz_{peak} during VDLs (Bates et al., 2013a; Hewett et al., 2005) and in the landing preceding a vertical jump (Collings et al., 2019) but between limb comparisons of peak forces and impulses were not made.

In the present study we identified that the lead leg during a VDL contributed a larger fraction of the Fz_{peak} and dissipated more energy (measured as the kinetic impulse) in the first 100 ms following ground contact compared to the trail leg. Further stratifying trials into right leg leading showed that the right (lead) leg produced a larger Fz_{peak} and greater $Impulse_{100ms}$ compared to the left leg. In left leg leading trials, there were no between-limb differences in $Impulse_{100ms}$ or Fz_{peak} magnitudes (Figure 1). Consequently, simply altering the lead leg during a VDL could be the difference between an individual passing return to sport testing thresholds or not, despite no changes in strength, movement ability or energy absorption capacity.

Our results suggest that a drop landing using the step-off initiation technique may not be an appropriate strategy for sport science practitioners and clinicians to assess landing task performance, particularly in injury risk assessments or return to sport decision making. If the step-off technique is used, we suggest that practitioners account for the possibility of misrepresentation of symmetry due to lead leg selection. Alternating the leading leg and averaging the results may provide a more accurate representation of movement ability and energy absorption capacity than using the same leading leg for all trials. When landing data were pooled in the present study, the proportion of participants with a limb symmetry index within acceptable limits was increased ($Impulse_{100ms}$: 4 out of 13 participants, Fz_{peak}: 6 out of 13 participants) compared to when data was stratified by lead leg selection (Figure 2.3). Alternatively, practitioners may consider utilizing the double leg forward drop or hanging drop to initiate landing assessments.

Sport science practitioners and clinicians should consider that the isolated landing tasks may not reflect a vertical jump landing (McBride & Nimphius, 2020). Here, isolated landing tasks may be a better indication of an individual's capacity to decelerate the body (VDL) or utilize the stretch-shortening cycle (DVJ) than a way to isolate vertical jump landing strategy.

2.4.1 Limitations

There are limitations with this study. First, we did not assess lower extremity strength and the association with landing strategy. The constraints placed on individuals based on their lower body strength leads to unique problem-solving and joint loading strategies (Duthon et al., 2006; Hassebrock et al., 2020). Between-limb differences in lower extremity strength may have magnified or masked the differences seen in landing characteristics with each leg leading. Incorporating kinematic assessment, via 3D motion capture or inertial measurement units, alongside kinetic characteristics could have provided a clearer picture of an individual's movement strategy. In this study, we were primarily interested in how lead leg selection influences foot contact timing and vertical ground reaction forces and therefore did not include kinematic assessment. Though we found differences in foot contact timing, Fz_{peak} and $Impulse_{100ms}$ between the lead and trail leg, it is unknown how these differences affected joint positions and moments. Finally, we only assessed the influence of lead leg selection on VDL performance from a height of 45 cm. It is unknown if the effects of lead leg selection across smaller and greater heights.

2.4.2 Conclusion

In conclusion, we found that force-time characteristics of the leading limb and trailing limb in a vertical drop landing task were affected by lead leg selection. Additionally, between limb symmetry was below accepted thresholds for variables associated with lower limb injury risk when landing trials were stratified by lead leg selection. Subjects showed greater symmetry when left and right leg leading trials were pooled. These results suggest that the lead leg selection when using a step-off technique during landing tasks may artificially alter limb symmetry in metrics associated with lower limb injury risk. Therefore, it may be suitable to include VDL and DVJ trials with each limb leading. The development and utilization of landing assessment procedures that are not reliant on stepping or hopping off a platform also warrants investigation. Practitioners should also consider the influence of testing procedures on results when interpreting clinical and/or research findings.

Chapter 3: The Relationship Between Lower Body Strength and Single-Leg Landing Performance of Elite Athletes After ACL Reconstruction

3.1 Introduction

The anterior cruciate ligament (ACL) provides knee joint stability by resisting anterior translation and rotation of the tibia relative to the femur, and serves as a proprioceptive organ for the lower extremity (Anderson et al., 2016). The rate of ACL injury is high in elite sport (Jordan et al., 2017; Lai et al., 2018; Lohmander et al., 2007) and ACL injury is associated with a long rehabilitation, knee joint degradation and the development of post-traumatic knee joint osteoarthritis (Waldén et al., 2015).

A majority of ACL injuries occur in non-contact situations characterized by rapid-onset, high energy absorption demands (Koga et al., 2010; Krosshaug et al., 2007). Non-contact ACL injuries in field and court sports are primarily associated with change of direction maneuvers or single leg landings (Krosshaug et al., 2007; Montgomery et al., 2018; Waldén et al., 2015), specifically, when the knee is in an extended position (i.e. 20º of knee flexion) (Beaulieu et al., 2021). Decreased knee flexion angle at ground contact is associated with higher peak vertical ground reaction forces (Fz_{peak}) and increased compressive forces at the knee joint, which can strain the ACL (Dufek & Bates, 1991; Norcross et al., 2013; Schmitz & Shultz, 2010). It is thought that a combination of insufficient lower body strength capacity to rapidly dissipate high magnitudes of energy (Heinrich et al., 2014, 2018; Waldén et al., 2015) and suboptimal body alignment can increase risk of ACL injury or reinjury (Schreiber & van Eck, 2010; Siegel et al., 2012).

Athletes typically undergo surgical reconstruction following ACL injury to restore knee joint stability and improve the likelihood of returning to sport (RTS) (Schreiber & van Eck, 2010). ACLR surgery using an ipsilateral semitendinosus autograft is a common surgical technique (Konrath et al., 2016; Morris et al., 2021) that leads to persistent deficits in knee flexor strength (Baratta et al., 1988; Serpell et al., 2015). This is particularly detrimental because the knee flexors help provide knee joint stability and assist the ACL in resisting anterior tibial translation and rotation of the tibia (Buckthorpe et al., 2019). Quadriceps strength has repeatedly been shown to be diminished following ACLR regardless of autograft used (Grindem et al., 2016) and decreased quadriceps strength is a risk factor for ACL re-injury (Bell et al., 2016; Fryer et al., 2019; Hiemstra et al., 2005; Maestroni et al., 2021; Rostami et al., 2018). Hip adduction and external rotation strength deficits have also been identified following ACLR (McBride & Nimphius, 2020; Minetti, 1998). Lower body strength deficits may impact energy absorption strategies (Lisee et al., 2019), thereby increasing the risk of subsequent ACL injury.

Quadriceps strength is the primary explanatory variable for knee flexion angle during single leg landing tasks following ACLR, with stronger quadriceps associated with greater knee joint range of motion (Lawrence et al., 2008). Additionally, increased hip external rotation strength is associated with reduced peak vertical ground reaction force (Fz_{peak}) (Lawrence et al., 2008). Large Fz_{peak} magnitudes, particularly with decreased knee flexion, are associated with increased risk of ACL injury (Beaulieu et al., 2021; Hewett et al., 2005). Therefore, increased strength at the knee and hip may provide protective measures for the ACL by decreasing Fz_{peak} (Leppänen et al., 2017; Podraza & White, 2010). Although differences in Fz_{peak} between ACLR limbs and control subjects have not been observed during single leg landings, clinically relevant asymmetry in Fz_{peak} between the ACLR and contralateral limb is apparent in 20% of athletes at 7 months post ACLR (Ithurburn et al., 2015). Furthermore, this between-limb asymmetry is still apparent in 15% of athletes greater than 2.5 years post-surgery, suggesting long term changes in landing strategy following ACLR (Ithurburn et al., 2019).

Assessing force characteristics over time periods associated with ACL injury, measured as impulse in the first 100 ms after ground contact ($Impulse_{100ms}$) may provide additional insights into the landing demands placed on the limb (Harry et al., 2017; B. M. Nigg, 1985; Norcross et al., 2013). High energy absorption in the first 100 ms after landing resulted in greater knee extension moment, anterior tibial shear force and posterior directed ground reaction forces which likely increased ACL loading (Norcross et al., 2013). However, $Impulse_{100ms}$ has not been directly measured in ACLR populations that have returned to sport.

Objective testing batteries are typically employed to determine athlete RTS readiness after ACLR (Burgi et al., 2019). Passing criteria include achieving >85% limb symmetry across a number of tests, restoring mechanical muscle function to pre-injury level or reaching the performance level of healthy peers (Barber-Westin & Noyes, 2011; Burgi et al., 2019; Jordan et al., 2020; Webster & Feller, 2018). Despite this, long-term neuromuscular deficits are present in athletes that have returned to sport (Ithurburn et al., 2019; Jordan et al., 2015a, 2020; King, Richter, Franklyn-Miller, Daniels, Wadey, Moran, et al., 2018; Morris et al., 2021; Paterno et al., 2007) and may explain the high reinjury rate in athletes that RTS after ACLR (King et al., 2020; Lind et al., 2012; Paterno et al., 2012). Identifying long-term, trainable neuromuscular deficits following ACLR in athletes that have returned to sport can better inform future rehabilitation focuses and potentially provide a protective mechanism prior to and after primary ACL injury.

Due to the paucity of kinetic and kinematic assessment of single leg landing performance and comprehensive strength tests of elite athletes after ACLR, the purpose of this study was to compare the lower body strength and single leg landing strategy of elite athletes that underwent

ACLR to those of healthy elite athletes and healthy development-level healthy athletes. Additionally, we sought to quantify the effect of knee and hip maximal strength on five key landing variables (Fz_{peak}, $Impulse_{100ms}$, knee joint initial contact angle, knee joint range of motion during landing, initial contact velocity). We hypothesized that the ACLR limb has lower maximal muscle strength compared to the limbs of control subjects, but is stronger than the limbs of development-level athletes. In addition, we hypothesized that there would be no difference between the ACLR limb and the comparison groups for the landings from a 25 cm height, but the ACLR limb would show higher $Impulse_{100ms}$ and $Fz_{peak,}$ but smaller knee initial contact angle and range of motion compared to the contralateral limb, control subject limbs, and the limbs of the development-level athletes when landing from the 50 cm landing condition.

2.2 Methods

2.2.1 Participants

Thirty athletes were recruited for this study and were grouped by competition-level and injury status (anterior cruciate ligament reconstructed group - ACLR, n = 10; elite performer group – ELITE, n = 10; development level group - DEVO, n = 10). Participants in the ACLR group (male: n = 5; female: n = 5) were recruited from Canadian national sport programs, university varsity sport and professional sport teams. Athletes were between 1-4 years post-ACLR and had all returned to competition. All participants underwent a single ACLR surgery using the ipsilateral semitendinosus tendon graft (left ACLR: n = 5; right ACLR: n = 5). Surgery was performed by 7 surgeons from 5 sport medicine clinics, and secondary injuries at the time of ACL injury were not controlled. Sport-matched and sex-matched control participants were recruited for the DEVO and ELITE groups, and participants were performance matched for ELITE. Individuals who were not medically cleared for sport or who sustained a lumbar spine injury or lower limb injury within the previous 6 months were excluded from the study. Subject characteristics are presented in Table 1. The University of Calgary's Conjoint Faculties Research Ethics Board approved the experimental protocol (REB15-1094) and all subjects gave written informed consent to participate in the study.

Table 3.1 Participant Characteristics

	ACLR	ELITE	DEVO
N	10	10	10
	(M: N=5; F: N=5)	(M: N=5; F: N=5)	(M: N=5; F: N=5)
Age (years)	21.9 ± 2.4	22.1 ± 4.8	18.0 ± 2.1
Height (m)	1.74 ± 0.12	1.76 ± 0.14	1.77 ± 0.11
Mass (kg)	72.5 ± 14.4	71.2 ± 15.0	73.0 ± 11.1
Dominant Limb	Left: N=1	Left: N=1	Left: N=1
	Right: N=9	Right: N=9	Right: N=9
Post-Op (months)	27.4 ± 10.9	-	-
Injured Limb	Left: N=5	-	-
	Right: N=5		
Sport			
Winter Slope	5	5	5
Soccer	2	2	2
Basketball	1	1	1
Volleyball	2	2	2
Level			
National	3	3	0
National Next-GEN	3	3	0
Professional	1	1	0
University Varsity	3	3	0
Provincial	0	0	1
National Program Identified	0	0	2
Club/High School	0	0	7

3.2.2 Single Leg Landing

All testing was performed in a single testing session. Participants performed a standardized warm up that included dynamic stretches of the quadriceps, hamstrings, adductors and gluteal muscles and included jumping and landing tasks. IMUs (myoMotion Research Pro

IMU, Noraxon, Scottsdale, Arizona, USA) were then applied to the subject by a single researcher (DL) in accordance with the manufacturer's guidelines using a reinforced elastic strap. Foot IMUs were secured to the subject's shoe at the midpoint. Shank IMUs were fixed to the lower leg 3 cm medial to the tibial tuberosity. A measuring tape was then used to identify a straight line from the midpoint of the base of the patella to the ipsilateral hip crease. Thigh IMUs were placed 8 cm proximal to the base of the patella. Thigh IMUs were affixed to the skin using an adhesive medical bandage. The pelvic IMU was centred 2 cm proximal to a line bisecting the bilateral posterior superior iliac crests (Chakraverty et al., 2007) using an elastic strap.

Prior to the landing protocol, subjects stood still with one foot on each force plate of a dual force plate system for a standing body weight assessment. Subjects were then familiarized with the test by practicing the single leg landing task for a self-determined number of repetitions to a maximum of 5 trials per limb. A standing IMU calibration was then conducted. Subjects stood with their feet hip width apart and heels, buttocks, upper back and back of head against a stadiometer. Subjects were instructed to "stand as tall as possible without locking their knees" while the IMU positions were calibrated (MyoResearch Version 3.14, Noraxon, Scottsdale, Arizona, USA). Lower body alignment was verified prior to each calibration by the same certified exercise professional. The angle between the shank and thigh IMUs during the standing calibration provided a reference for the neutral knee joint position of the subject (i.e., 0° of knee flexion). As such, all knee joint angle measures were then calculated relative to the calibration position. Prior to the initiation of this study, between-calibration reliability of the knee joint angle measure showed good agreement (CV: 3.1%).

Subjects then performed 5 single leg drop landings on each leg (James et al., 2007) in a randomized manner for the 25 cm condition and then the 50 cm condition. Unsuccessful landings, defined as an inability to maintain balance, were discarded and repeated. Subjects performed all landings on one leg then moved to the other. Landings were determined to be successful if the subject stepped straight off the elevated platform without lowering their center of mass, landed on the force plate, maintained contact on the plate and stabilized their body position until cued to stand (approximately 3 seconds) without their non-landing leg touching the ground. Hands remained on the hips throughout the movement. Subjects self-selected their landing depth and strategy. Participants were allowed to wear their preferred athletic training shoes. Each trial was separated by a 30 second rest period.

3.2.3 Hip and Knee Maximal Strength Assessments

Assessment of maximum lower extremity torque was performed on a Cybex dynamometer instrumented with a third-party force transducer (Omega, LC703-500, Stamford, Connecticut, USA) sampling at 1500 Hz. For hip internal and external rotation assessment, subjects were seated upright in the dynamometer chair with a hip angle of 100º. The subject's popliteal fossa was approximately 1 cm from the edge of the seat and subjects were instructed to maintain a 90º bend in their test knee (Figure 3.1 A). A blocking pad was positioned on the lateral aspect of the thigh and a strap was fastened over the blocking pad and distal aspect of the thigh. Two straps also secured the torso and attached to a strap that crossed the hip crease of the subject. The bracket encasing the force transducer was fixed to the medial aspect of the shank 5 cm proximal to the medial malleolus by a non-compliant polyester strap. Hip external rotation compressed the force transducer, while internal rotation tensioned the force transducer. Hip abduction and adduction torque assessment were conducted in a supine position with the tested leg abducted 15º (Figure 3.1 B). A strap was placed across the hips in line with the greater trochanters and an additional strap was across the torso in line with the anterior superior iliac spine (ASIS). Two straps originating above the shoulders attached to the strap in line with the ASIS to secure the torso. The non-tested leg was strapped to the dynamometer chair distal to the knee. The force transducer's bracket was positioned on the lateral thigh 8cm proximal to the lateral knee joint line. Successful trials of hip abduction and adduction torque measurement required the subject to keep the knee straight with the patella pointing superiorly. Additionally, the posterior aspect of the tested leg had to remain in contact with the dynamometer chair. Hip abduction compressed the force transducer and adduction tensioned the force transducer.

For the knee extension test, the subject was seated upright with a hip angle of 100º and a strap over the distal thigh secured the tested limb to the dynamometer chair (Figure 3.1 D). An additional two straps crossed the chest, and one strap crossed the hips to secure the subject. The force transducer was secured to the anterior tibia 2.5 cm proximal to the lateral malleolus. Knee flexion torque was measured in a prone position with a fully extended hip. A strap secured the hips in this position throughout the contractions and the force transducer was secured to the posterior shank 2.5 cm proximal to the lateral malleolus (Figure 3.1 C). The knee angle was set to 70º of knee flexion (full knee extension: 0º) for both knee flexion and knee extension torque assessments.

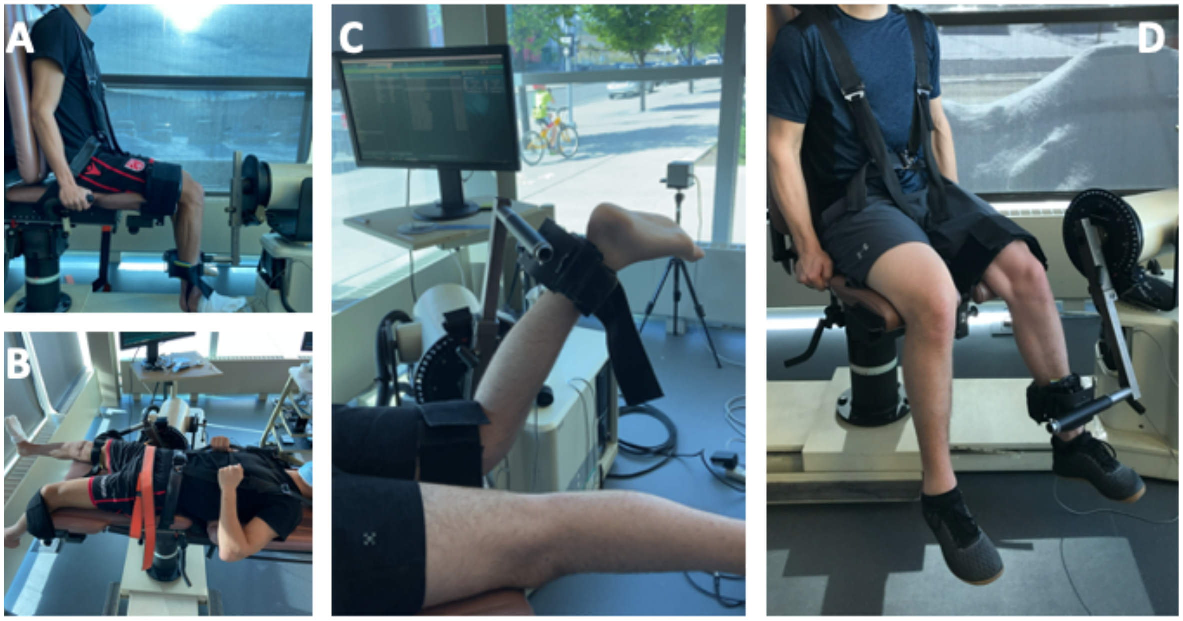

Figure 3.1 Testing Positions for Maximum Voluntary Isometric Contractions at the Knee and Hip Joints. Hip external and internal rotation strength were tested in a seated position (A). Hip abduction and adduction strength were tested in supine lying with the tested leg abducted 15º (B). Knee flexion strength was tested in prone lying with the knee at 70º of knee flexion (C). Knee extension strength was tested in a seated position with the knee at 70º of knee flexion (D).

The isometric leg press extension test was performed with the subject seated at a hip angle of 100º, knee angle set to 60º of knee flexion (full knee extension: 0º) and the foot of the tested limb centered on a vertically mounted force plate (PASPORT Force Platform, PASCO Canada, Oakville, Ontario, Canada) sampling at 1000 Hz. The foot of the non-tested leg was firmly pressed into the ground at a self-selected position (Figure 3.2).

Figure 3.2 Isometric Leg Press Extension Testing Position. Subjects were seated at a hip angle of 100º and knee angle at 60º of knee flexion. The foot of the tested leg was centered on the vertically mounted force plate.

Following 3 submaximal warm up contractions (approximately 25, 50, 75% of maximum effort), the participants performed 3 maximal voluntary isometric contractions (MVIC) for each muscle group. Subjects were instructed to contract as hard as possible from a completely relaxed state. Each contraction was held for 3 s and separated by a 25 s rest period within each contraction type. A 2 min rest period was given between muscle groups. Consistent and strong verbal encouragement was provided along with real-time visual feedback to ensure maximal effort was given. The order of testing was randomized for all subjects.

3.3 Data Acquisition and Processing

3.3.1 Landing Analysis

Vertical ground reaction forces were measured using a dual force plate system (Accupower Force Platform, AMTI, Watertown, Massachusetts, USA) sampling at 1500 Hz and were recorded synchronously with the IMU signals sampling at 100 Hz. The IMUs were re-calibrated every third trial. The participant's system mass was determined during the standing bodyweight assessment using the total Fz (sum of left and right) over a 500 ms interval.

The IMU and force data were stored on a personal computer for subsequent analysis (MyoResearch Version 3.14, Noraxon, Scottsdale, Arizona, USA). Following data collection, the data was exported for further analysis using custom-built computer scripts (Matlab Version R2019B, Mathworks, Nattick, Massachusetts, USA). For landing assessment, alignment between IMU data and force data was verified prior to analysis. Raw Fz voltage signals were converted to Newtons based on force plate calibration analysis. Initial ground contact during landing was identified as Fz > 10 N (Norcross et al., 2013; Rocchi et al., 2018). Stabilization after landing was defined as the point at which Fz stabilized between ±5% of the participant's baseline body weight for a 1000 ms period (Tran et al., 2015a) (Figure 3.4). The peak vertical ground reaction force (Fz_{peak}) was identified for each landing trial. $Impulse_{100ms}$ was calculated as the integration of the Fz-time curve for the first 100 ms after initial ground contact. Velocity at ground contact was determined using the impulse-momentum relationship. Net landing-phase impulse was calculated by subtracting the subjects body weight from the Fz magnitude and then integrating the Fz-time curve from initial ground contact to stabilization. The net impulse was then divided by the subject's mass to determine initial velocity at ground contact. Knee flexion angle was determined by the sagittal plane angle between the shank and thigh IMU at initial ground contact and at the post-landing stabilization position defined above, whereby 0° corresponded to a fully extended knee joint. Joint landing range of motion was calculated as the difference between the stabilized and initial contact joint angles.

3.3.2 Torque Analysis

Torque data was sampled at 1500 Hz using an analogue to digital board and stored on a personal computer (MyoResearch Version 3.14, Noraxon, Scottsdale, Arizona, USA) for subsequent analysis using custom built scripts (Matlab Version R2019B, Mathworks, Nattick, Massachusetts, USA). To determine torque, a straight-line measurement from the joint axis of rotation to the mid-point of the force transducer represented the moment arm. Force was applied to the linear force transducer (Omega, LC703-500, Stamford, Connecticut, USA) perpendicular to moment arm. Raw voltage from the force transducer was converted to torque (Nm). The trial with the highest maximal torque was defined as the MVIC for each muscle group and selected for analysis. Trials initiated with an obvious countermovement were removed from the analysis. MVICs were normalized to body mass for group comparisons ($Nm \cdot kg^{-1}$).

3.3.4 Statistical Analysis

To assess differences in lower body strength measures and landing kinetics and kinematics between the ACLR and contralateral limbs a paired t-test was performed for each of the measures. All relevant assumptions were met, except for $Impulse_{100ms}$ in the 50 cm condition. In this case, a Wilcoxon signed-rank test was performed. Hedge's effect size between the ACLR and contralateral limb was determined for each measure (package: 'effsize') and the magnitude was interpreted as: <0.2: negligible; 0.2-0.49: small; 0.5-0.79: medium; >0.8 large**.** Between-limb differences in strength and landing metrics were assessed for the ELITE and DEVO groups. No differences between the left and right leg were found within the groups. Therefore, a between-limb average was calculated and used in assessing differences in strength and landing metrics between the ACLR limb, ELITE and DEVO groups via Kruskal-Wallis tests by rank. Hedge's effect size was calculated independently for the ACLR-ELITE and ACLR-DEVO comparisons and interpreted as previously described. For all comparisons significance was set to $\alpha = 0.05$.

We identified 5 landing variables associated with ACL injury risk and landing strategy (i.e., Fz_{peak}, $Impulse_{100ms}$, knee joint initial contact angle, knee joint range of motion, initial ground contact velocity) *a priori*. Isometric strength metrics at the hip (i.e., abduction and external rotation strength) and at the knee (i.e., extension and flexion and isometric leg press extension) were identified as potential predictors of the landing variables above based on a review of the literature (Lawrence et al., 2008; Lisee et al., 2019). As such, we built separate hip strength and knee strength linear mixed effects models (LME) (R version 1.2.1335, 'lme4' package) to quantify the effect of strength on each landing outcome. LMEs were fit with fixed effects for strength metrics (Hip model: hip external rotation and hip abduction strength; Knee model: knee extension, knee flexion and leg press extension strength), limb status (ACLR-Injured, ACLR-Contralateral, ELITE-Non-Injured, DEVO-Non-Injured) and landing height (25 cm, 50 cm). Subjects were included in the model as random intercepts, with limb (left and right) nested in athlete to account for the repeated measurement of landings and strength assessments. Here, the left and right limbs for DEVO and ELITE participants were included in the models. The fixed effect estimate of the strength metrics on the landing variables were determined, along with the effects of strength metrics by limb status.

3.4 Results

3.4.1 Lower Body Maximal Strength

The ACLR limb had significantly lower peak knee flexion torques compared to the contralateral limb (1.10 ± 0.19 Nm/kg vs. 1.38 ± 0.17 Nm/kg; P= 0.003; ES= -1.38). The ACLR limb also had higher peak hip abduction torque compared to the contralateral limb (1.21 ± 0.34 vs. 1.13 ± 0.30; P= 0.02; ES= 0.19) (Figure 3.3). There were no other significant differences in peak torques between limbs in the ACLR group (Table 3.2), and no significant differences were found for the lower body strength measures between the ACLR limb, ELITE and DEVO subjects (Table 3.3).

Table 3.2 Comparison of Knee and Hip Strength Between the ACLR Injured and Contralateral Limbs

	ACLR Limb	Contralateral Limb	P-value	Hedge's Effect Size (95% CI)
Knee Flexion (Nm/kg)	1.10 ± 0.19	1.38 ± 0.17	**0.003**	**−1.38 (-2.09, -0.66)**
Knee Extension (Nm/kg)	3.64 ± 0.50	3.58 ± 0.64	0.63	0.08 (-0.27, 0.43)
Hip Internal Rotation (Nm/kg)	1.01 ± 0.20	1.01 ± 0.17	0.99	0 (-0.40, 0.39)
Hip External Rotation (Nm/kg)	0.81 ± 0.17	0.81 ± 0.21	0.95	0.01 (-0.31, 0.33)
Hip Abduction (Nm/kg)	1.21 ± 0.34	1.13 ± 0.30	**0.02**	0.19 (0.04, 0.33)
Hip Adduction (Nm/kg)	1.99 ± 0.51	2.06 ± 0.42	0.43	−0.12 (-0.43, 0.18)
Leg Press Extension (N/kg)	38.17 ± 7.01	38.32 ± 6.98	0.87	−0.02 (-0.27, 0.24)

Table 3.3 Comparison of Knee and Hip Strength between ACLR Injured Limb, ELITE and DEVO

	ACLR Limb	ELITE	DEVO	P-value	Hedge's Effect Size (95% CI)	
					ACLR – ELITE	ACLR - DEVO
Knee Flexion (Nm/kg)	1.10 ± 0.19	1.16 ± 0.20	1.12 ± 0.16	0.76	−0.29 (-1.19, 0.61)	−0.09 (-0.98, 0.81)
Knee Extension (Nm/kg)	3.64 ± 0.50	3.34 ± 0.39	3.14 ± 0.44	0.09	**0.63 (-0.29, 1.55)**	**1.02 (0.06, 1.97)**
Hip Internal Rotation (Nm/kg)	1.01 ± 0.20	1.03 ± 0.16	1.03 ± 0.17	0.98	−0.12 (-1.02, 0.77)	−0.11 (-1.01, 0.79)
Hip External Rotation (Nm/kg)	0.81 ± 0.17	0.73 ± 0.20	0.75 ± 0.20	0.62	0.42 (-0.49, 1.33)	0.32 (-0.58, 1.22)
Hip Abduction (Nm/kg)	1.21 ± 0.34	1.22 ± 0.26	1.18 ± 0.22	0.92	−0.03 (-0.93, 0.87)	0.08 (-0.82, 0.98)
Hip Adduction (Nm/kg)	1.99 ± 0.51	2.08 ± 0.13	1.87 ± 0.38	0.29	−0.23 (-1.32, 0.67)	0.27 (-0.64, 1.17)
Leg Press Extension (N/kg)	38.17 ± 7.01	39.08 ± 6.66	33.26 ± 5.30	0.11	−0.13 (-1.03, 0.77)	**0.76 (-0.18, 1.69)**

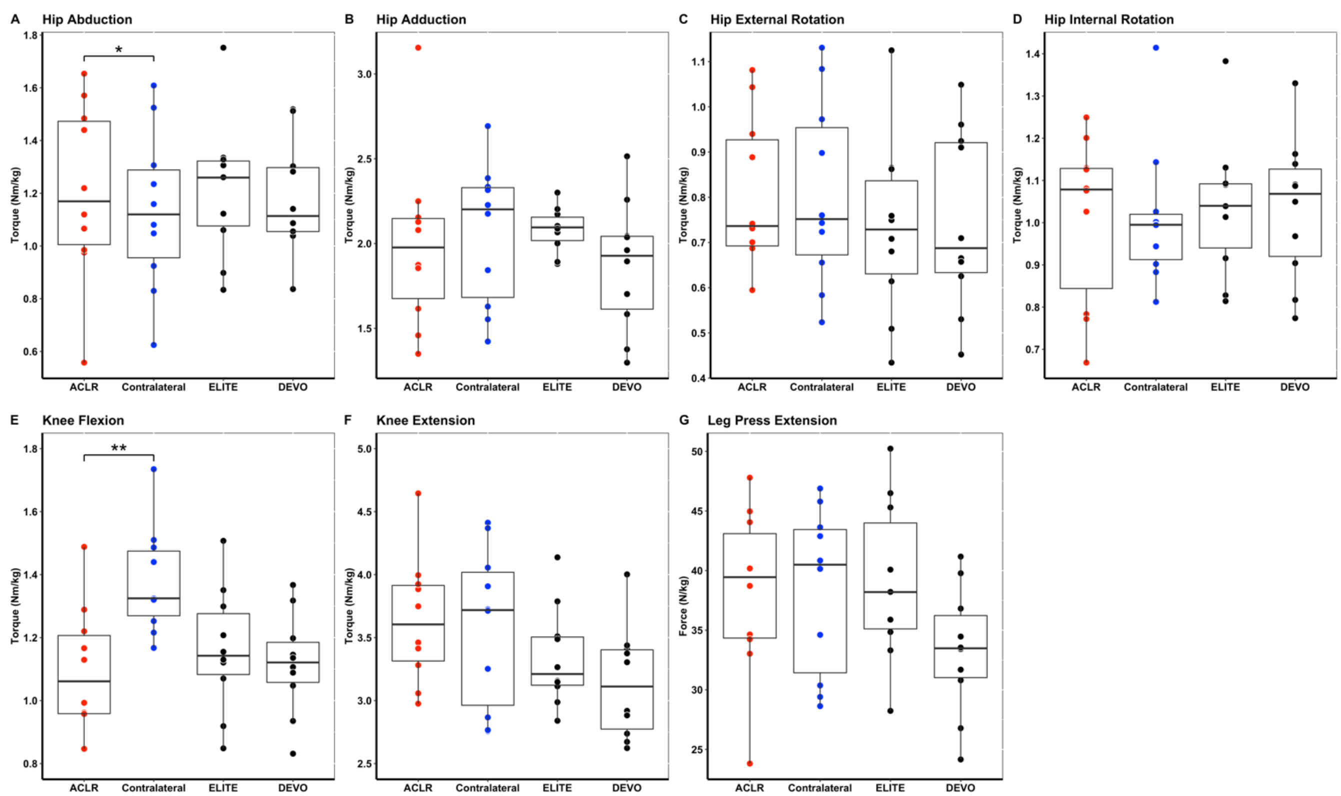

Figure 3.3 A Comparison of Hip and Knee Strength by Limb Status. Lower body strength among ACLR limbs, contralateral limbs and ELITE and DEVO athletes. * A significant difference in hip abduction was found between the ACLR and contralateral limbs P<0.05. ** A significant difference in knee flexion was found between the ACLR and contralateral limbs P<0.01

3.4.2 Landing Kinetics and Kinematics

A sample landing force-time and knee joint angle-flexion representation is presented in Figure 3.4. No between-limb differences in landing kinetics or knee joint kinematics were found when comparing ACLR and contralateral limbs for either the 25 cm or 50 cm single leg landing conditions (Table 3.4). Additionally, there were no group differences found for landing kinetics or kinematics when comparing the ACL limb condition with the ELITE and DEVO groups in either landing condition (Table 3.5).

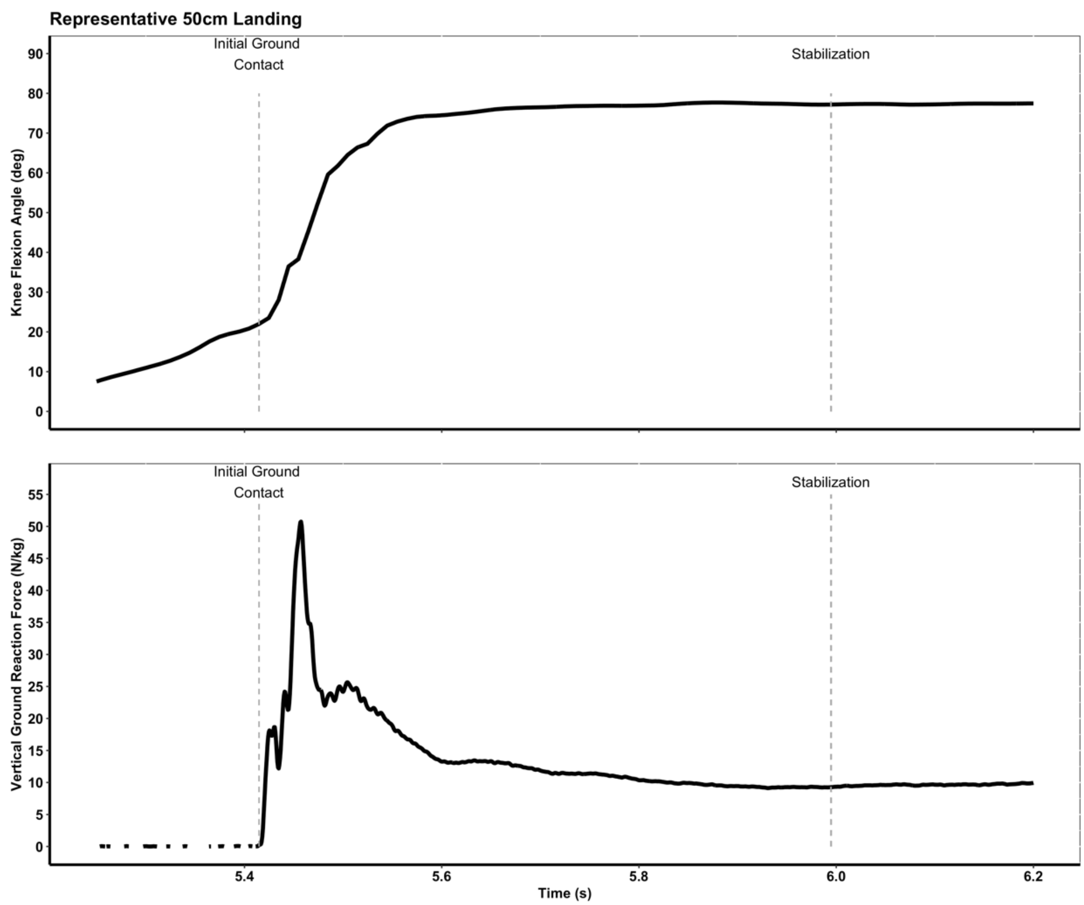

Figure 3.4 Sample Force-Time and Knee Joint Angle-Time Tracing at the 50 cm Landing Condition. Sample landing highlighting initial ground contact timing of the foot (i.e., >10 N of vertical ground reaction force) and stabilization of the body's center of mass (bottom). Knee joint angle at initial ground contact was identified alongside the knee flexion angle at stabilization, where the difference between the two angles was calculated as the knee joint's range of motion (top).

Table 3.4. Landing Kinetics and Kinematics of the ACLR Injured and Contralateral Limbs

	ACLR Limb	Contralateral Limb	P-value	Hedge's Effect Size (95% CI)
25 cm landing				
Fz_{peak} (N/kg)	30.11 ± 5.77	29.75 ± 5.18	0.81	0.06 (-0.44, 0.56)
$Impulse_{100ms}$ (Ns/kg)	1.60 ± 0.19	1.54 ± 0.18	0.24	0.29 (-0.20, 0.78)
Knee Initial Contact Angle (º)	17.3± 7.5	17.4 ± 8.6	0.93	−0.01 (-0.22, 0.22)
Knee Range of Motion (º)	44.5 ± 13.5	44.6 ± 12.8	0.99	−0.00 (-0.31, 0.31)
Initial Contact Velocity (m/s)	1.48 ± 0.27	1.49 ± 0.25	0.53	-0.09 (-0.40, 0.21)
50 cm landing				
Fz_{peak} (N/kg)	48.92 ± 7.08	48.29 ± 7.66	0.68	0.08 (-0.30, 0.46)
$Impulse_{100ms}$ (Ns/kg)	2.22 ± 0.15	2.19 ± 0.23	0.92	0.15 (-0.40, 0.70)
Knee Initial Contact Angle (º)	16.0 ± 5.6	16.8 ± 7.6	0.63	−0.11 (-0.55, 0.34)
Knee Range of Motion (º)	46.9 ± 15.2	48.4 ± 15.5	0.67	−0.09 (-0.52, 0.34)
Initial Contact Velocity (m/s)	2.43 ± 0.24	2.44 ± 0.23	0.84	-0.02 (-0.21, 0.17)

Table 3.5. Comparison of Landing Kinetics and Kinematics between ACLR Injured Limb, ELITE and DEVO

					Hedge's Effect Size (95% CI)	
	ACLR Limb	ELITE	DEVO	P-value	ACLR - ELITE	ACLR - DEVO
25 cm landing						
Fz_{peak} (N/kg)	30.11 ± 5.77	31.84 ± 1.66	33.56 ± 8.52	0.55	−0.39 (-1.30, 0.51)	−0.45 (-1.37, 0.46)
$Impulse_{100ms}$ (Ns/kg)	1.60 ± 0.19	1.68 ± 0.10	1.66 ± 0.21	0.78	−0.52 (-1.44, 0.40)	−0.29 (-1.20, 0.61)
Knee Initial Contact Angle (º)	17.3 ± 7.5	17.7 ± 5.3	15.7 ± 4.9	0.78	−0.06 (-0.96, 0.84)	0.24 (-0.66, 1.14)
Knee Range of Motion (º)	44.5 ± 13.5	42.6 ± 11.9	42.7 ± 11.7	0.88	0.15 (-0.75, 1.05)	0.14 (-0.76, 1.04)
Initial Contact Velocity (m/s)	1.48 ± 0.27	1.62 ± 0.07	1.61 ± 0.24	0.32	-0.70 (-1.63, 0.22)	-0.52 (-1.44, 0.39)
50 cm landing						
Fz_{peak} (N/kg)	48.92 ± 7.08	53.50 ± 8.06	54.77 ± 13.43	0.49	−0.58 (-1.50, 0.34)	−0.52 (-1.44, 0.40)
$Impulse_{100ms}$ (Ns/kg)	2.22 ± 0.15	2.26 ± 0.14	2.20 ± 0.15	0.66	−0.30 (-1.20, 0.61)	0.15 (-0.75, 1.05)
Knee Initial Contact Angle (º)	16.0 ± 5.6	13.8 ± 4.3	13.2 ± 4.4	0.35	0.42 (-0.49, 1.33)	0.53 (-0.38, 1.45)
Knee Range of Motion (º)	46.9 ± 15.2	48.5 ± 12.3	50.2 ± 14.1	0.93	−0.11 (-1.02, 0.79)	−0.22 (-1.12, 0.68)
Initial Contact Velocity (m/s)	2.43 ± 0.24	2.51 ± 0.11	2.53 ± 0.19	0.68	-0.40 (-1.31, 0.50)	-0.41 (-1.32, 0.50)

Relative Fz_{peak} at 25 cm (A) and 50 cm (B) landing conditions for each limb status, and $Impulse_{100ms}$ at the 25 cm (C) and 50 cm (D) landing conditions are also presented in Figure 3.4. There were no significant differences between limb statuses.

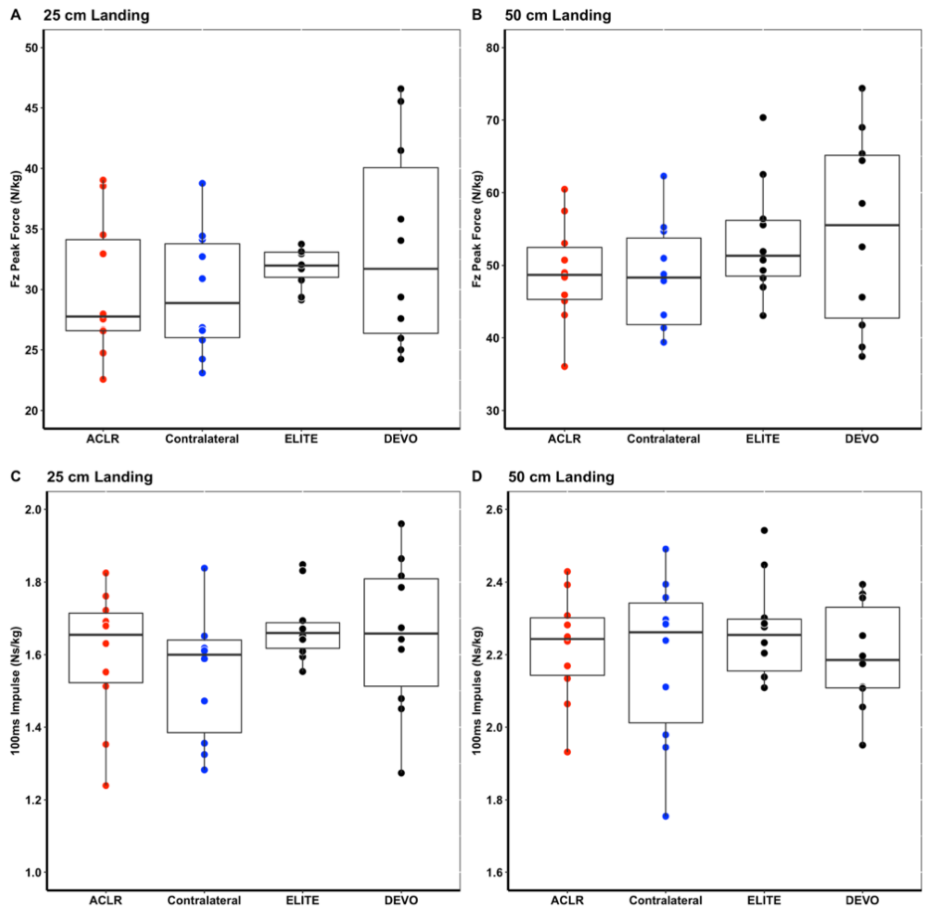

Figure 3.5 Landing Kinetics by Limb Status at 25 cm and 50 cm Landing Conditions. Peak vertical ground reaction force (Fz_{peak}) (Panels A and B) and impulse in the first 100ms after ground contact (Panels C and D) are presented. There were no significant differences between groups.

3.4.3 The Effect of Knee and Hip Strength on Landing Strategy

The main effects of lower body strength on landing strategy kinetic and kinematic variables are presented in Table 3.6. There was a main effect of hip abduction ($P = 0.04$) and leg press extension ($P = 0.04$) strength on knee joint range of motion. Additionally, there was a main effect of leg press extension strength on initial ground contact velocity ($P = 0.02$). The effects of lower body strength on landing strategy by leg status are presented in Figures 3.5-3.9. A correlation analysis was conducted to assesses the relationship between hip and knee strength and the five identified landing variables across all subjects (Figure 3.10). There was a moderate negative to moderate positive relationship between strength variables and landing kinetic and kinematic variables, except for $Impulse_{100ms}$ where there were only weak correlations with strength at both the 25 cm and 50 cm landing conditions.

Table 3.6. Estimate (SE) of The Main Effects of Hip and Knee Strength on Landing Variables

	Fz_{peak}		$Impulse_{100ms}$		Knee IC Angle		Knee ROM		IC Velocity	
	Estimate	P-Value	Estimate	P-Value	Estimate	P-Value	Estimate	P-Value	Estimate	P-Value
Abduction	-11.1 (5.9)	0.06	-0.04 (0.2)	0.76	-2.8 (4.3)	0.51	16.7 (7.7)	**0.04**	0.13 (0.1)	0.18
External Rotation	9.5 (8.4)	0.26	0.06 (0.2)	0.78	0.54 (6.2)	0.93	-13. (11.2)	0.25	0.17 (0.2)	0.24
Extension	1.5 (4.1)	0.71	0.06 (0.1)	0.51	1.1 (2.8)	0.69	-0.18 (4.8)	0.97	-0.04 (0.1)	0.47
Flexion	15.8 (9.3)	0.09	-0.12 (0.2)	0.55	-6.1 (5.9)	0.31	9.7 (12.3)	0.43	0.29 (0.2)	0.06
Leg Press Extension	-0.02 (0.3)	0.95	0.00 (0.01)	0.75	0.19 (0.2)	0.35	0.93 (0.5)	**0.04**	0.01 (0.0)	**0.02**

Point estimates and 95% confidence intervals of the groupwise effect of hip and knee strength on Fz_{peak} are presented in Figure 3.5. Hip abduction (A), hip external rotation (B) and knee extension (C) strength had a significant effect on Fz_{peak} experienced by the DEVO group in 50 cm landings, whereby increased strength resulted in reduced Fz_{peak}. There were no other significant effects of strength on Fz_{peak}.

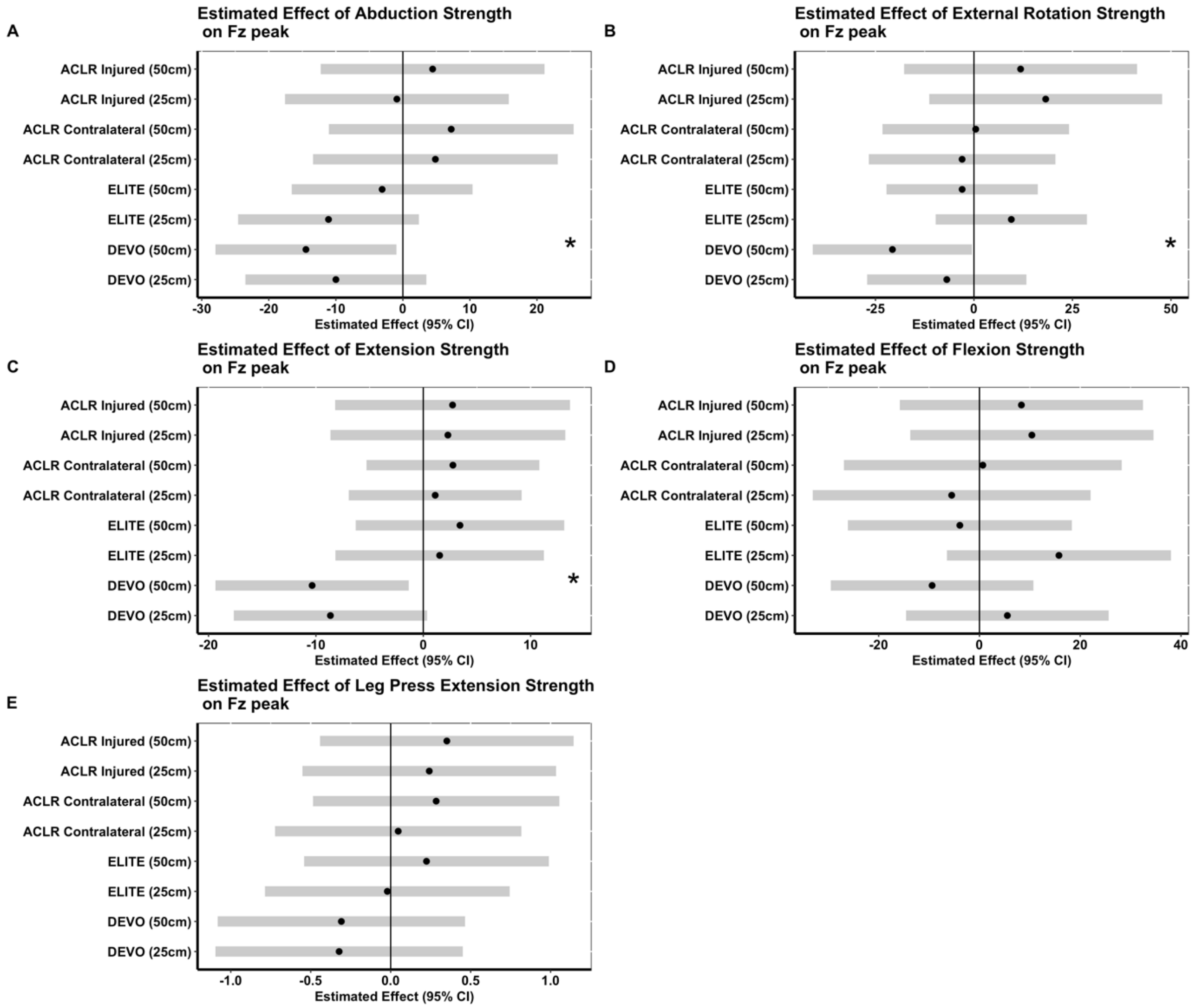

Figure 3.6 The Estimated Effect of Hip and Knee Strength on Peak Vertical Ground Reaction Force (Fz_{peak}) by Leg Status and Landing Height. * Indicates a significant effect. Hip abduction (Panel A), hip external rotation strength (Panel B) and knee extension strength (Panel C) had a significant negative effect on Fz_{peak} for the DEVO group at 50 cm.

The effect of knee and hip strength on $Impulse_{100ms}$ for each limb status and landing condition is presented in Figure 3.6. An increase in hip external rotation strength (B) decreased the $Impulse_{100ms}$ experienced by the ACLR contralateral limb at the 50 cm landing condition. No other strength measure had a significant groupwise effect on $Impulse_{100ms}$ at either landing condition.

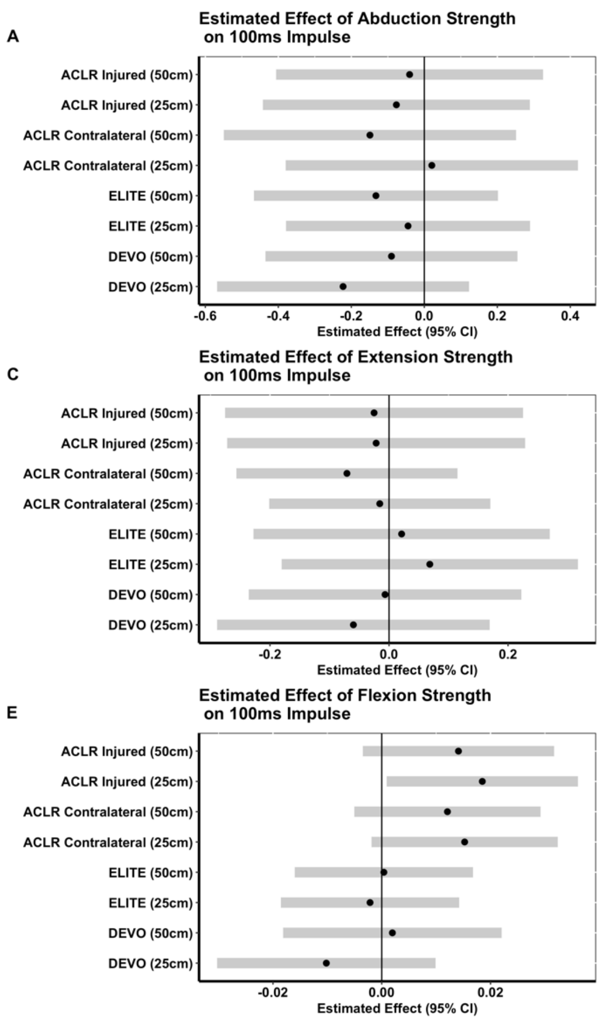

Figure 3.7 The Estimated Effect of Hip and Knee Strength on Impulse in the First 100 ms After Ground Contact by Leg Status and Landing Height. No strength measures had a significant effect**.**

Leg press extension strength had a significant effect on knee initial contact angle for the ACLR injured and contralateral limbs at the 25 cm and 50 cm landing conditions (Figure 3.7, panel E), where an increase in leg press extension strength led to a more flexed knee on initial ground contact under these limb and landing conditions. Conversely, increased knee flexion strength (D) had a significant effect on knee initial contact angle for the ACLR contralateral limb, leading to less knee flexion at ground contact. No strength measured had a significant effect on knee range of motion during landing in any limb status or landing condition (Figure 3.8).

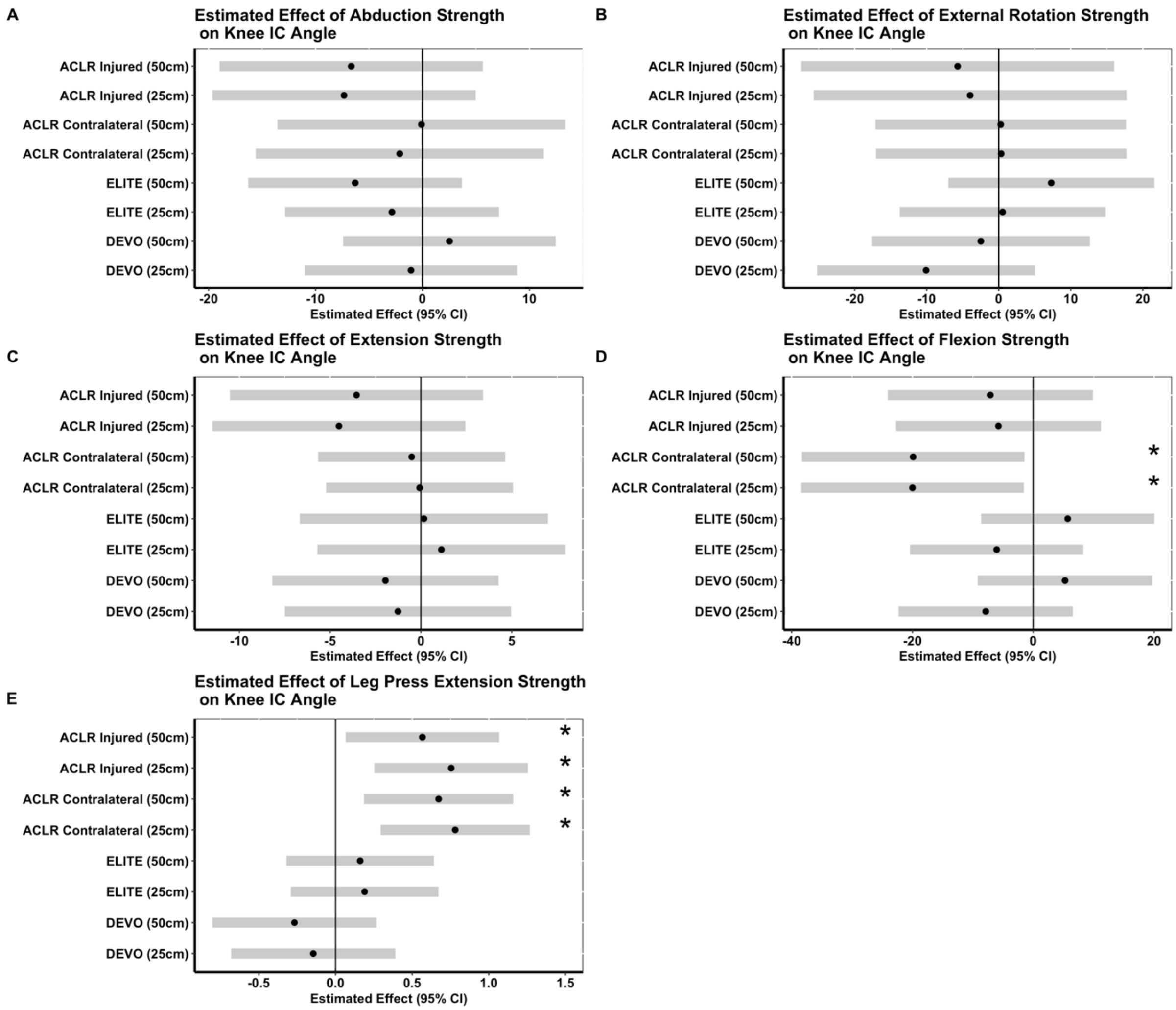

Figure 3.8 The Estimated Effect of Hip and Knee Strength on Knee Joint Angle at Initial Ground Contact (Knee IC) by Leg Status and Landing Height. * Indicates a significant effect. Knee flexion strength (Panel D) had a significant effect on Knee IC angle, with increased strength leading to less knee flexion at initial contact. Increased leg press extension strength led to a more flexed knee for the ACLR contralateral and ACLR injured limbs at both landing condition heights (Panel E).

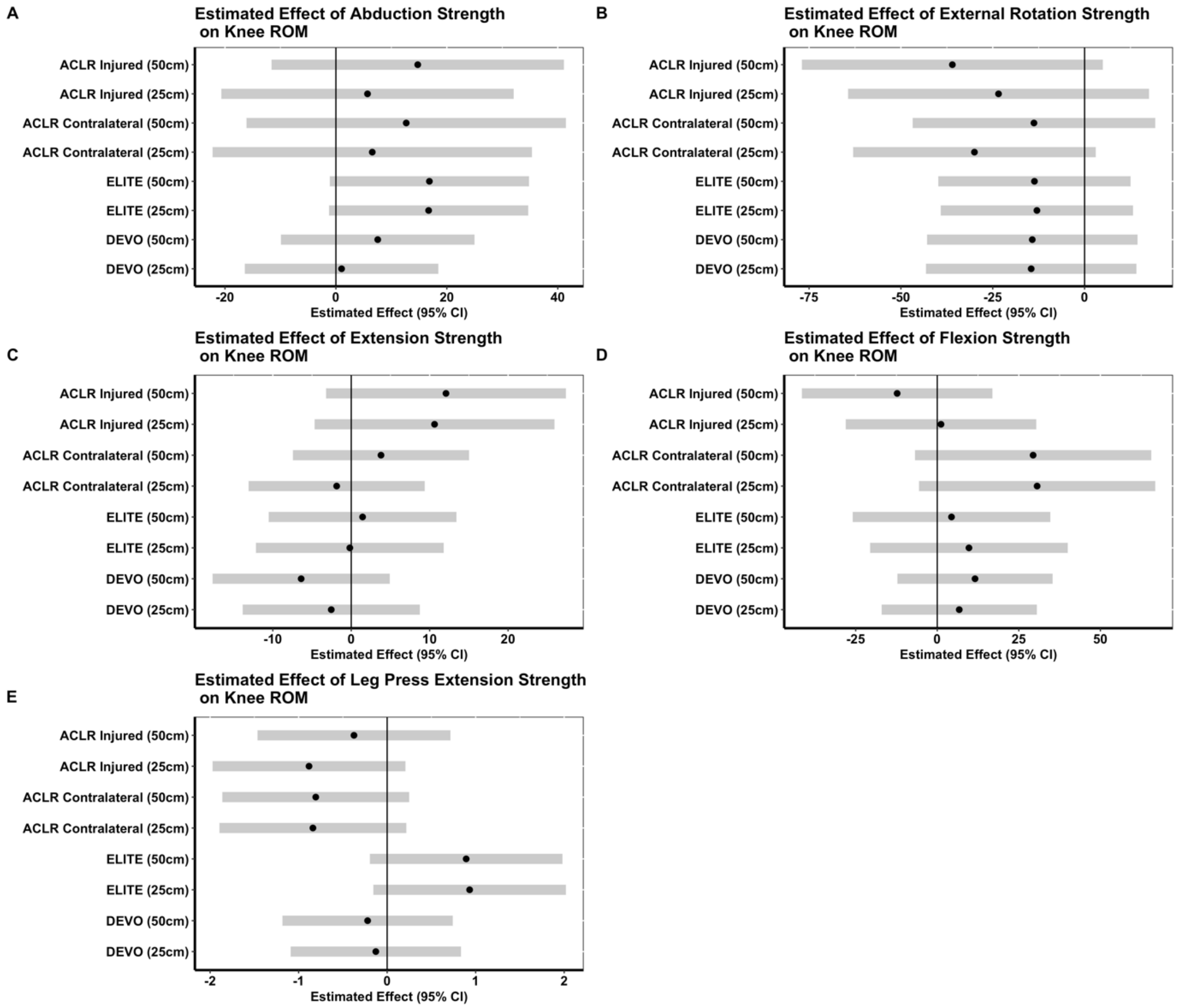

Figure 3.9 The Estimated Effect of Hip and Knee Strength on Knee Joint Range of Motion during Landing (Knee ROM) by Leg Status and Landing Height. No strength measures affected Knee ROM for any group under either landing condition.

Hip strength measures did not have a significant effect on initial ground contact velocity for any limb status or landing condition (Figure 3.9). Knee extension strength had a significant effect on initial contact velocity for the DEVO group at both landing heights, where an increase in strength led to decreased initial ground contact velocity. Knee flexion strength (D) had a significant effect on initial contact velocity for the ACLR injured limb at 25 cm and leg press extension strength (E) had a significant effect on initial contact velocity for the ACLR injured and contralateral limbs at both 25 cm and 50 cm landing conditions and the ELITE athletes at 25 cm. Increase flexion strength and leg press extension strength for the previously described groups led to an increase in initial contact velocity.

Finally, the Pearson's correlation coefficient between all-subject knee and hip strength and landing variables previously identified as related to ACL injury risk are presented in Figure 3.10. Of note, there were moderate to strong correlations between initial ground contact velocity and Fz_{peak} and $Impulse_{100ms}$, whereby increased ground contact velocity was associated with increased Fz_{peak} and $Impulse_{100ms}$ magnitudes in each landing height condition. Additionally, moderate positive relationships were found between initial ground contact velocity and knee initial contact angle, and initial ground contact velocity and knee range of motion.

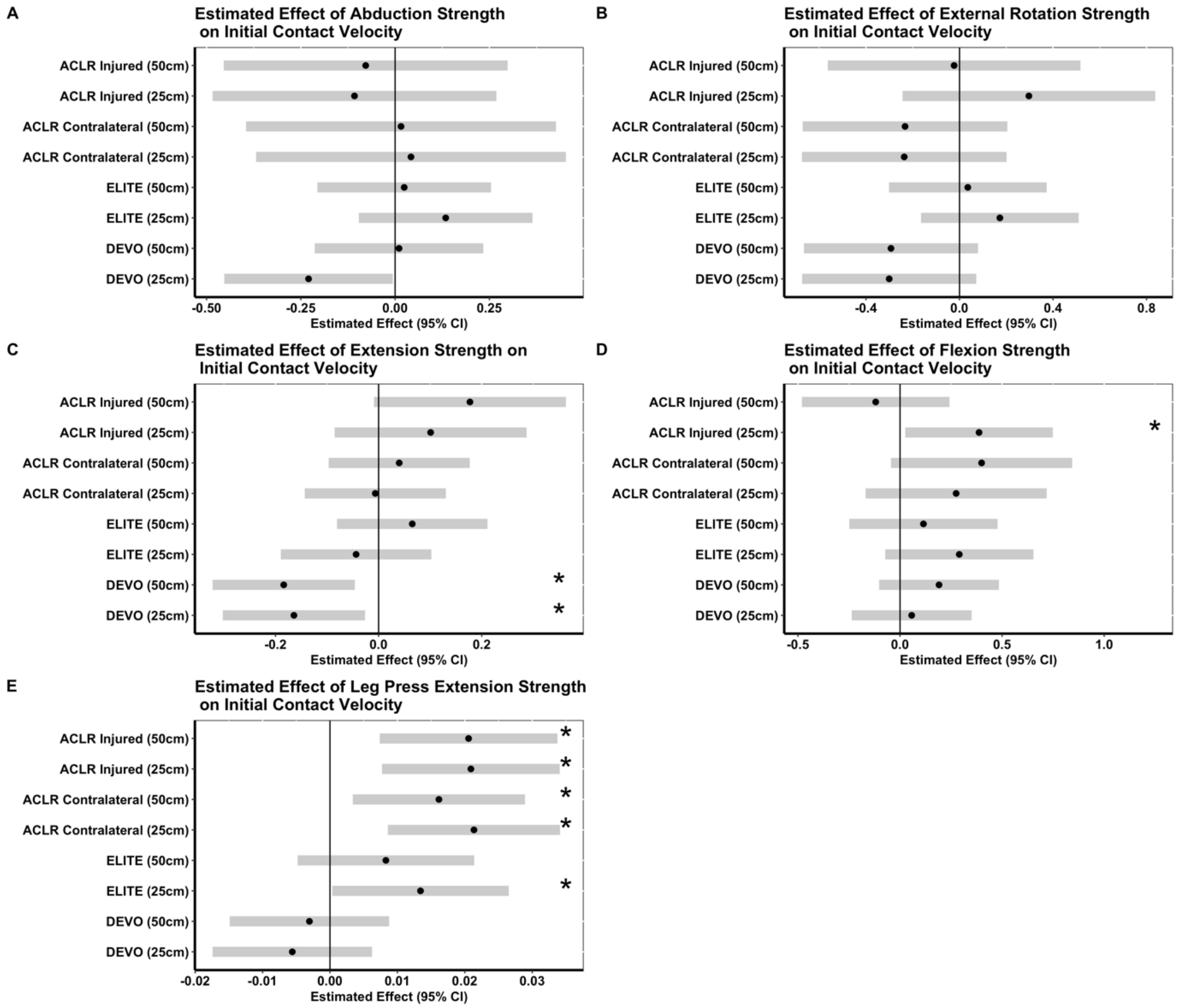

Figure 3.10 The Estimated Effect of Hip and Knee Strength on Initial Ground Contact Velocity by Leg Status and Landing Height. * Indicates a significant effect. Knee extension strength had a significant effect on initial contact velocity for the DEVO group at both landing heights (Panel C), with increased strength leading to decreased velocity. An increase in knee flexion strength led to an increase in initial contact velocity for the ACLR injured limb at 25 cm (Panel D). Increased leg press extension strength led to an increase in initial contact velocity for the ACLR injured and ACLR contralateral limbs at both landing heights, and for the ELITE group in 25 cm landings (Panel E).

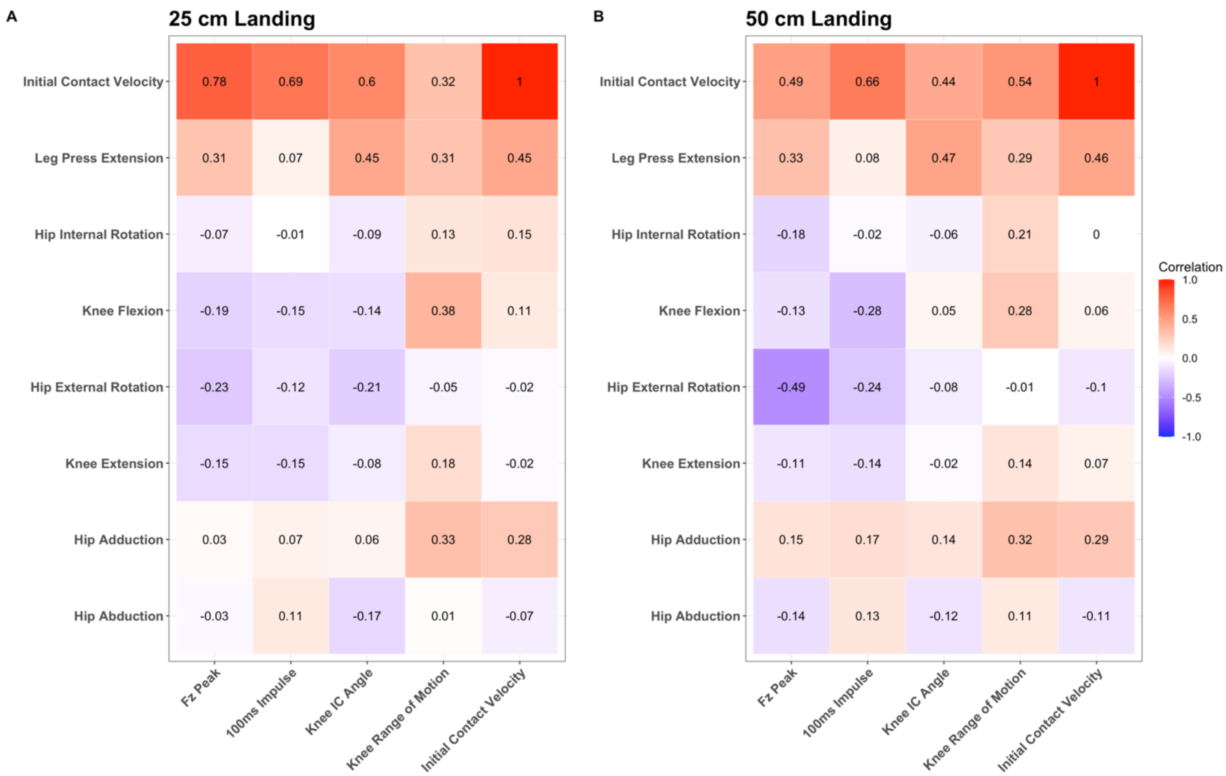

Figure 3.11 Correlation Matrix of Lower Body Strength and Single Leg Landing Kinetic and Kinematic Variables. Pearson's correlation matrix between knee and hip maximal strength and landing variables at the 25 cm (A) and 50 cm (B) landing conditions.

3.5 Discussion

In this study, we quantified the strength of the muscles at the knee and hip joint and the single-leg landing strategy from two heights of athletes that had returned to sport after ACLR and compared them to healthy ELITE and DEVO athletes. Additionally, the effect of strength measures on landing variables associated with ACL injury risk was determined. To the best of our knowledge, this is the first study to assess strength at the knee and hip joints in addition to single leg landing performance from heights of increasing demand. Strength deficits at the knee and hip after ACLR have been well documented in the literature (Hiemstra et al., 2005; Jordan et al., 2015a; Konrath et al., 2016; Tate et al., 2017), and lower body strength is associated with landing strategy (Ithurburn et al., 2015; McBride & Nimphius, 2020). Since many ACL injuries and reinjuries occur during single leg landing situations in sport (Koga et al., 2010; Krosshaug et al., 2007), it is important to understand the effects of ACLR on lower body maximal strength at the knee and hip and single leg landing ability. In this study, we provided a new perspective to assess the effect of knee and hip muscle strength on landing variables previously identified as risk factors for ACL injury across multiple landing heights, sport-performance level, and injury status.

Contrary to our hypothesis, the ACLR limb did not demonstrate consistent strength deficits compared to the contralateral limb and sport-performance matched controls, nor was it stronger than the development-level participants. Rather, the ACLR limb had decreased knee flexion strength compared to the contralateral limb, and increased hip abduction strength compared to the contralateral limb. Additionally, no differences were found in landing kinetic or kinematic variables between the ACLR and contralateral limb, or when comparing the ACLR limb to ELITE and DEVO subjects. However, an effect of hip abduction and leg press extension strength on knee joint range of motion during landing, and an effect of leg press extension strength on initial ground contact velocity were found. Here, participants with higher strength demonstrated increased knee joint range of motion and a higher initial ground contact velocity. As decreased knee joint range of motion is associated with risk of ACL injury (Beaulieu et al., 2021; Hewett et al., 1999; Leppänen et al., 2017), the present findings suggest that increased lower body maximal muscle strength may lead to a more protective movement strategy. Further, variation in the initial ground contact velocity may suggest the presence of a protective movement strategy. Notably, participants with decreased leg press extension strength may lower their center of mass towards the ground in landing tasks utilizing the step-off technique to reduce their downward energy, leading to a lower initial ground contact velocity, and thus reducing the demands of the landing task itself. In the present study, we were unable to control for this possibility, but we did identify that those participants with higher lower body strength also had higher initial ground contact velocities, suggesting that those with less lower body strength may have altered their strategy to reduce the demands of the landing task.

3.5.1 Knee and Hip Maximal Strength

The finding of diminished knee flexion strength in the ACLR limb compared to the contralateral limb after surgical reconstruction using the semitendinosus autograft was consistent with previous literature (Jordan et al., 2015a; Konrath et al., 2016; Morris et al., 2021). Jordan et al. (2015a) found a 14% between-limb deficit in knee flexion strength at 70º in ACLR elite ski racers that had returned to sport. Similarly, Morris et al. (2021) found consistent deficits in knee flexion strength across the joint angle-torque relationship among athletes, and Nomura et al. (2015) identified large deficits at large knee flexion angles after ACLR with a semitendinosus autograft. Semitendinosus tendon harvest for the purpose of ACLR autograft impacts the cross-sectional area of the semitendinosus and tendon regrowth is not guaranteed (Konrath et al., 2016; Morris et al., 2021), therefore knee flexion strength deficits are expected in the ACLR limb. We found no difference in knee extension strength between the ACLR limb and contralateral or ACLR limb, ELITE and DEVO subjects. Quadriceps strength deficits are commonly found after ACLR. Quadriceps strength deficits are often due, in part, to arthrogenic muscle inhibition (AMI) which leads to marked decreases in quadriceps muscle activation due to a combination of knee joint swelling, pain and structural damage after ACLR surgery (Rice et al., 2014). As a consequence, the ability to recruit high-threshold motor units is diminished (Rice et al., 2014). In this present investigation we did not find a statistically significant or clinically relevant difference in quadriceps strength between the ACLR and contralateral limbs. However, we observed clinically relevant differences based on effect sizes between the ACLR limb, ELITE and DEVO groups, despite the comparisons not reaching statistical significance. The ACLR limb was 1.09 times stronger than the ELITE group, while it was 1.16 times stronger than the DEVO participants. This result is compelling due to the AMI phenomenon. The lack of a quadriceps strength deficit in the ACLR limb, and potential increased strength compared to comparison groups, begs the question of if these participants no longer suffer from AMI or if their quadriceps force potential is such that even after the effects of AMI, they are still equal, or slightly stronger, than their comparison groups. However, the answer to that question was beyond the scope of this investigation.

Our study also assessed global lower body strength via isometric leg press extension. To the best of our knowledge this is the first investigation to utilize this test in an ACLR population. We observed negligible differences between the ACLR and contralateral limbs, and between the ACLR limb and ELITE athletes, suggesting a recovery of global lower body strength following ACLR. Additionally, there was not a significant difference between the ACLR and DEVO groups. However the difference was likely clinically relevant to sport science professionals (Redden et al., 2018). As such, the isometric leg press appeared to effectively differentiate between elite athletes, regardless of ACL injury status, and development-level athletes. This standardized and repeatable test of lower body strength has also been used to assess between-limb asymmetry in lower body strength in alpine ski racers, and increased asymmetry was found to be a risk factor for ACL injury (Steidl-Müller et al., 2018)

Previous research showed no difference in hip abduction strength when comparing the ACLR limb to the contralateral limb and to control groups (Fryer et al., 2019; Noehren et al., 2014; Tate et al., 2017; Thomas et al., 2013). Indeed, a trend toward increased abduction strength in the ACLR limb is often found (Fryer et al., 2019; Noehren et al., 2014; Tate et al., 2017). In agreement with previous literature, we found that hip abduction strength was significantly higher in the ACLR limb compared to the contralateral limb. Regaining hip abduction and external rotation strength is a key objective to rehabilitation after ACLR (Myer, Paterno, et al., 2006a; Reiman et al., 2009). As such, it is encouraging that hip abduction strength after ACLR was consistently found to be on par with or greater than the contralateral limb and control subjects. Contrary to the results of Hiemstra et al. (2005), we did not observe a decrease in hip adduction strength on the ACLR side compared to control subjects. Previously, hip adduction strength was observed to be up to 44% lower on the ACLR limb compared to the dominant limb of control subjects (Hiemstra et al., 2005), whereas in the present study, we only observed small, non-significant differences in adduction strength between the ACLR and contralateral limbs and ACLR and ELITE participants. Previously, hip adduction deficits were postulated to be related to semitendinosus tendon harvest during ACLR surgery, due to its shared insertion with the gracilis tendon on the pes anserinus (Hiemstra et al., 2005). However, the ACLR subjects in our study showed no deficit despite all ACLR subjects undergoing surgery using the semitendinosus autograft. This difference in results may be a result of different testing positions as we utilized a supine position, while previous assessments were performed in standing (Hiemstra et al., 2005). Furthermore, our ACLR subjects were elite athletes who had returned to sport, while Hiemstra did not describe the sporting background of the participants (Hiemstra et al., 2005).

We found no difference in hip external rotation and hip internal rotation strength between the ACLR and contralateral limbs. Similarly, there were no significant differences in these metrics between the ACLR limb, and ELITE and DEVO groups. However, the ACLR limb demonstrated 1.10 times greater hip external rotation strength than the ELITE group. This result failed to reach statistical significance, but the observed effect size suggests the result may be clinically relevant. Previously, external rotation strength deficits have been shown in the ACLR limb compared to the control limb (Boo et al., 2018), though this result has not been consistently observed in the literature (Noehren et al., 2014). Low hip external rotation strength has previously been linked to weak quadriceps after ACLR, whereas those with strong quadriceps did not demonstrate hip external rotation strength deficits (Bell et al., 2016). In the present investigation, we did not observe a significant difference in knee extension strength between the ACLR and all comparison groups, with a tendency for the ACLR limb to be stronger. Therefore, our result of no deficit in hip external rotation strength on the ACLR limb agrees with the literature. We also did not find differences in hip internal rotation strength between the ACLR limb and all comparison groups, which is supported by the minimal literature available on hip internal rotation strength after ACLR (Bell et al., 2016).

3.5.2 Single Leg Landing Kinetics and Kinematics

Subjects in this study performed single leg drop landings from heights of 25 cm and 50 cm. Two landing conditions were included in an effort to identify if a high energy demand landing (i.e., 50 cm landing condition) would highlight limits in landing capacity (McBride & Nimphius, 2020). Contrary to this, we observed no group differences in landing kinetic or kinematic variables relevant to future ACL injury risk in either landing condition. No differences in Fz_{peak} or $Impulse_{100ms}$ between groups were seen in either landing height condition. Previously, the ACLR limb has presented with significantly higher Fz_{peak} compared to the contralateral limb in a landing task including both horizontal and vertical displacement (Rocchi et al., 2018). Conversely, no between-limb differences have been shown in Fz_{peak} at 8-month post ACLR (Nagelli et al., 2020) and at RTS (Ithurburn et al., 2015). To the best of our knowledge, this was the first investigation quantifying $Impulse_{100ms}$ in ACLR athletes that had returned to sport. Impulse_{100ms} can be used to quantify the energy absorption demands in a time frame associated with ACL injury (Norcross et al., 2013). Norcross et al. (2013) observed that high energy absorption in the first 100 ms following a bilateral landing, quantified by negative joint work, led to a landing strategy that may increase ACL loading (Norcross et al., 2013). The lack of significant differences in force-time characteristics between groups in this present investigation may be explained by training status. Nagelli identified that 12 sessions of neuromuscular training focused on core and posterior chain muscle activation and landing mechanics reduced Fz_{peak} in athletes after ACLR (Nagelli et al., 2020). The ACLR participants in this present study are elite athletes. As such, they have access to strength and conditioning specialists and neuromuscular training programs. It is speculated that this training history led to improvements in force dissipation strategies during landing (McBride & Nimphius, 2020; Minetti, 1998), whereby we observed no differences between the ACLR limb and contralateral limb, or between the ACLR limb, ELITE and DEVO groups.

In addition, the ACLR athletes demonstrated comparable knee joint initial ground contact angle, range of motion and ground contact velocity to the contralateral limb, ELITE athletes, and DEVO athletes. All limb conditions made initial ground contact with minimal knee flexion (i.e., < 20º of flexion), with a trend toward a decrease in knee flexion angle at initial ground contact in the 50 cm landing condition compared to the 25 cm condition. Consequently, the athletes moved through consistent knee joint range of motion regardless of leg status and drop height (range of motion: 25 cm landing: 42.6 – 44.6º; 50 cm: 46.9 – 50.2º). Moving through large ranges of motion may allow the quadriceps to eccentrically contract to gradually decelerate the body's center of mass (Lephart et al., 2002; Podraza & White, 2010). Previously, a strong relationship between quadriceps maximal strength and knee joint range of motion after ACLR had been established (Lisee et al., 2019). Additionally, ACLR participants with high quadriceps maximal strength have demonstrated increased knee joint range of motion during single leg landings when compared to ACLR participants that demonstrated quadriceps strength deficits compared to the contralateral limb (Ithurburn et al., 2015). Further, coaching cues for landing (Elias et al., 2015) and neuromuscular training programs (Padua et al., 2018) can affect energy absorption strategies utilized during landing after ACLR. It is speculated that elite athletes after ACLR may be instructed to land "softly" and move through large ranges of motion during rehabilitation to counteract the potential deleterious effect of stiff landing strategies on ACL strain and reinjury risk (Butler et al., 2003; Hewett et al., 1999; Norcross et al., 2013; Padua et al., 2018), thus providing context for the similarity in knee joint kinematics between injury status and competition level groups that were observed in this study.

3.5.3 The Effect of Knee and Hip Strength on Landing Strategy

We identified main effects of hip abduction and leg press extension on knee joint range of motion during landing. Here, increased hip abduction strength and in leg press extension strength led to increased knee joint range of motion during single leg landing across two landing height conditions. Previously, high hip external rotation strength had been associated with increased knee joint rotation during landings (Lawrence et al., 2008), but the effect of hip abduction strength has not been observed before. However, several hip muscles contribute to both hip external rotation and hip abduction strength (Neumann, 2010). Therefore, it is likely that a combination of hip external rotation strength and hip abduction strength impacts the knee joint range of motion during single leg landings. Furthermore, leg press extension, utilized as a global indicator of lower body strength, affected knee joint range of motion. Previously, McBride and Nimphius demonstrated that global lower body strength, as measured in a 1-

repetition maximum back squat, significantly impacted landing strategy (McBride & Nimphius, 2020). Further, it is theorized that a lack of global lower body strength leads to a stiff landing strategy, characterized by limited joint range of motion (Devita & Skelly, 1992). Therefore, an increase in global strength may be beneficial to avoid hazardous stiff landings that are associated with ACL injury risk (Hewett et al., 1999; Minetti, 1998). Finally, we found a main effect of leg press extension strength on initial ground contact velocity during landing. Post-hoc analysis revealed that all participants demonstrated decreased landing velocity in comparison to the box height (i.e., participants lowered their center of mass), despite concerted efforts to control the center of mass height during task initiation. Participants with diminished leg press extension strength also demonstrated lower ground contact velocity in the 25 cm and 50 cm landing conditions compared to those with higher leg press extension strength. Decreased muscle strength and power may predict lack of confidence in the ACLR limb after surgery (Ageberg & Roos, 2016). Therefore, it is speculated that stronger athletes in this study, demonstrated by isometric leg press strength, had less deviation from the single landing task protocol, and therefore employed a strategy that increased their vertical landing velocity compared to their weaker counterparts.

Hip abduction, hip external rotation, and knee extension strength had a significant effect on Fz_{peak} experienced by the DEVO group at the 50 cm landing condition. Here, an increase in these strength measures decreased Fz_{peak}. This result is supported by previous literature that found an effect of quadriceps strength (Ithurburn et al., 2019) and external rotation strength (Lawrence et al., 2008) on Fz_{peak} measures. Interestingly, these strength metrics did not have an effect for the remaining experimental conditions or leg groups. It is speculated that the DEVO athletes did not have the same neuromuscular training background as the ELITE and ACLR athletes. As a result, the 50 cm landing condition was difficult and increased strength in these areas provided a protective mechanism that may not have been required in the ELITE and ACLR athletes.

Greater leg press extension strength was associated with a greater knee flexion angle at initial ground contact for the ACLR injured limb and contralateral limb, which may confer a protective effect on the ACL (Padua et al., 2018). Further, landing with an increased knee flexion angle increases the eccentric deceleration demands on the lower body musculature (Podraza & White, 2010). Although the effect was small, this may suggest that increasing global lower body strength (i.e., leg press extension strength) is a feasible strategy to promote landing strategies with greater knee flexion, which is often a preferred outcome of neuromuscular training programs (Padua et al., 2018). Conversely, the ACLR contralateral limb showed a significant effect of knee flexion strength leading to increased knee extension on landing. The ACLR injured limb showed a similar trend, but statistical significance was not reached. This result is in contrast to findings by Lephart at al. (2002) where participants with diminished knee flexion strength also performed a single leg landing with less knee flexion. Making initial ground contact at a shallow knee flexion angle provides a larger percentage of the knee joint's anatomical range of motion to move through during the landing phase. Therefore, one could land in an extended position but travel through a large range of motion to decelerate the body, which could provide an optimal strategy between high external loading and internal loading (Butler et al., 2003). Incidentally, we found no effect of lower body strength on knee joint range of motion during single leg landings.

3.5.4 Limitations

We were limited in this study by a small sample size. As we were interested in the strength and landing strategy of elite athletes, our recruiting pool was small. Additionally, our inclusion criteria for the ACLR subjects (i.e., one ACLR surgery with semitendinosus autograft, >1 year after surgery, and returned to sport) provided additional limitations. Given this challenge, the recruited athletes came from multiple sports and surgeries were performed by multiple surgeons. The small sample size also limited the statistical analyses. For instance, a principal component analysis may have provided an opportunity to group athletes and differentiate between groups based on strength and landing metrics. As highlighted, the landing velocity of participants was decreased compared to the platform height used to initiate the landings. Despite our best efforts to control this, it resulted in a discrepancy between the task we set out to test (e.g., 50 cm single leg landing) and the one performed by the participants. This may have contributed to the null results in landing variables between groups.

3.5.5 Conclusion

The present assessment of knee and hip strength and landing strategy of elite ACLR athletes demonstrated that strength can largely be recovered in the ACLR limb less than 2.5 years after surgery. As such, single leg landing kinetics and kinematics did not differ between the ACLR injured limb, ACLR contralateral limb, ELITE and DEVO groups. Further, we identified that lower body strength has a significant effect on landing variables associated with ACL injury risk. These results add to the previous literature by highlighting the importance of global lower body strength and strength at the hip to aid in the recovery of movement strategies, particularly single leg loading situations which are widely performed in sports, and represent a risk of ACL injury (Krosshaug et al., 2007; Montgomery et al., 2018). Future investigations should consider how strength affects landing strategy after ACLR utilizing the BPTB autograft, in addition to assessing joint moments to better quantify the contributions of joints to the body's deceleration in these tasks.

Chapter 4: Conclusions and Future Directions

The primary aim of this thesis was to quantify the effects of ACLR using the semitendinosus autograft technique on the knee and hip strength and landing strategy of elite athletes following return to sport. This thesis contributed to the understanding in two main areas: (1) identifying methodological considerations and potential outcome biases associated with landing task selection; and (2) quantification of lower body strength at the knee and hip joints and evaluating the subsequent effects of strength at each joint on landing strategy, including variables commonly associated with ACL injury. After ACLR, the assessment and monitoring of neuromuscular capacity is used to objectively progress rehabilitation and to screen athletes for return to sport readiness (Burgi et al., 2019; Jordan et al., 2020; Taberner et al., 2020). In this thesis, we identified knee and hip strength metrics associated with landing strategies linked to ACL injury risk and provided insight into how assessment methodology can impact the effect testing outcomes.

In Chapter 2, the influence of lead leg selection in a bilateral VDL was quantified in plyometric trained athletes. VDL assessments are frequently used to assess ACL injury risk (Collings et al., 2019), whereby between-limb symmetry is quantified. VDLs can be initiated with three main techniques, one of which is a step off (Decker et al., 2003; Devita & Skelly, 1992). Previously, between limb differences were quantified using this technique. However, a systematic assessment of the bias introduced by the initiation technique had not been performed. As such, participants performed VDLs from a 45 cm box with each limb serving as the lead leg (n = 3 per leg) and landed on a dual force plate system. The lead leg contributed a greater fraction of Fz_{peak} compared to the trail leg. When results were stratified by lead leg selection, there were significant between limb differences in Fz_{peak} and $Impulse_{100ms}$ with the right leg leading but not with the left leg leading. Based on the bias introduced by lead leg selection, sport science practitioners are encouraged to perform drop landing assessments trials with each limb leading, or to initiate the task in another manner (e.g., hanging by the arms from an elevated bar).

In Chapter 3, the lower body strength and single leg landing strategy of elite athletes with ACLR, elite athletes without ACLR (ELITE) and development-level athletes without ACLR (DEVO) were quantified and compared. Strength at the knee and hip have been shown to elicit an effect on the landing strategy employed (Ithurburn et al., 2015; Lawrence et al., 2008; Lephart et al., 2002; Minetti, 1998) and revealed long standing deficits after ACLR (Buckthorpe et al., 2019; Morris et al., 2021; Noehren et al., 2014; Thomas et al., 2013). Single leg landings are associated with ACL injury and are frequently used as screening tests for ACLR injury risk (Johnston et al., 2018)**.** Participants performed 5 single leg landings on each leg from heights of 25 cm and 50 cm while wearing lower body IMUs. Subsequently, isometric strength at the knee (i.e., knee flexion, knee extension, leg press extension) and hip (i.e., hip abduction, hip adduction, hip external rotation, hip internal rotation) was quantified. The ACLR limb demonstrated diminished knee flexion strength compared to the contralateral limb and had increased hip abduction strength compared to the contralateral limb. In contradiction to our hypothesis, the ACLR limb demonstrated no other deficits in muscle strength. Additionally, there were no differences in landing kinetics or kinematics between the ACLR limb and its comparison groups at either landing height condition. Linear mixed effects models identified a main effect of hip abduction and leg press extension strength on knee joint range of motion through the landing, indicating increased strength was associated with a greater range of motion. Further, we identified a main effect of leg press extension strength on the initial ground contact velocity. It is speculated that stronger athletes, quantified by the leg press extension strength test, did not attenuate their vertical velocity by lowering their center of mass at the point of step-off compared to their less strong counterparts. This suggests that, (1) practitioners control for the possibility that participants may change the demands of a single leg vertical landing test based on their strength level, and (2) that lower-body strength dependent alterations in movement strategies occurred in our study, whereby the downward energy of the body center of mass was minimized by participants with less strength (i.e., as a protective strategy to reduce the demands of the single leg landing task).

The relatively small and heterogenous sample was a limitation of this, limiting the ability to compare results across sex and sport. However, all ACLR athletes recruited are elite athletes competing professionally, in national team programs or at the university level. Therefore, although the absolute number of ACLR participants is low, it represents a large percentage of the elite athletes in Western Canada that fit our inclusion criteria (i.e., a single ACLR using semitendinosus autograft 1+ year ago and returned to sport). A second limitation of this study is the cross-sectional nature of the study design. We were unable to compare our results with pre-injury data and athletes performed their rehabilitation across multiple centers so neuromuscular training through rehabilitation was not consistent.

Future research should focus on the effect of lower body strength on single leg landing strategy in athletes after ACLR using the BPTB autograft. Differing functional deficits have been identified after BPTB versus semitendinosus autograft ACLR (Mueske et al., 2018; Webster et al., 2004), and thus their impact on landing strategy should be elucidated. Further, investigations incorporating comprehensive lower body strength testing and inverse dynamics to quantify joint work during single leg landings from multiple heights should be performed to identify the effect of lower body strength on energy absorption strategies after ACLR. Despite continuing research on strength and movement deficits after ACLR, the risk of ACL reinjury and contralateral ACL injury remain high (Ardern et al., 2014; King et al., 2020; Lai et al., 2018). It is our hope that the results of this thesis can inform sport science practitioners and rehabilitation specialists of key areas of focus for strength training after ACLR.

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UNIVERSITY OF CALGARY

UNIVERSITY OF CALGARY CONSENT
ADDITIONAL INFORMATION FOR CONTINUING RESEARCH PARTICIPANTS

TITLE: The Long-Term Effects of ACL Injury on Bilateral Limb Asymmetry and Muscle Activation in Elite Alpine Ski Racers

SPONSOR: Own the Podium
Canadian Sport Institute Calgary

INVESTIGATORS: Dr. Walter Herzog
Dr. Matt Jordan (Ph. 403-714-4655)
Nathaniel Morris
Drew Lawson (Ph. 613-246-3739)

You are participating in the above-named research study. When you agreed to participate, the researchers told you they would share any new information about the study that might affect your willingness to continue to participate in the study.

The study now involves new risk information that are described below. The researchers will explain the new risk information and then ask for your consent to participate in the new procedures as well as to continue participating in the study. With the exception of the information provided below, all of the information provided to you previously still applies.

WHAT ARE THE NEW RISKS INVOLVED IN THIS STUDY?

As a result of the COVID-19 pandemic there are increased risks associated with participation in this study. Risks associated with the study include:

- Increased time at the Canadian Sport Institute (CSI) Calgary at WinSport and increased exposure to CSI Calgary staff and athletes

Measures taken to reduce this risk include:

- Screening of all individuals entering CSI Calgary (e.g. Alberta Health Services COVID-19 Risk Screening questionnaire and temperature checks)
- Tracking of entrance and exit timing of all individuals visiting CSI Calgary
- Use of personal protective equipment for staff and participants (e.g. masks)
- Frequent use of Health Canada approved hand sanitizer
- Strict cleaning and disinfection of all testing apparatuses after use

- Physical distancing measures

All other aspects of the study described in the original consent form remain the same.

WHAT OTHER CHOICES DO I HAVE IF I DON'T WANT TO PARTICIPATE?

At any point you can choose to withdraw from the study. Your decision will not affect your status within your sport team.

WHO CAN I CONTACT IF I HAVE QUESTIONS ABOUT CONTINUING IN THIS STUDY?

The Research Team:
You may contact Dr. Matt Jordan at 403-714-4655 with any questions or concerns about the research or your participation in this study.

Conjoint Health Research Ethics Board (CHREB):
If you have any questions concerning your rights as a possible participant in this research, please contact the Chair, Conjoint Health Research Ethics Board, University of Calgary at 403-220-7990.

WHAT ARE MY RIGHTS IF I DECIDE TO CONTINUE TO TAKE PART IN THIS STUDY?

Continuing to take part in this study is your choice. You can choose whether or not you want to participate. Whatever decision you make, there will be no penalty to you.

- You have a right to have all of your questions answered before deciding whether to continue to take part.
- If you decide to continue to take part, you can leave the study at any time.

HOW DO I INDICATE MY AGREEMENT TO PARTICIPATE?

Your signature on this form indicates that you have understood to your satisfaction the information regarding your continued participation in the research project and agree to continue to take part. In no way does this waive your legal rights nor release the investigators or involved institutions from their legal and professional responsibilities.

SIGNATURE OF STUDY PARTICIPANT

____
Name of Participant

Signature of Participant Date

SIGNATURE OF PERSON OBTAINING CONSENT

Name of Person Obtaining Consent Contact Number

Signature of Person Obtaining Consent Date

SIGNATURE OF THE WITNESS

Name of Witness

Signature of Witness Date

A signed copy of this consent form has been given to you to keep for your records and reference.

HUMAN PERFORMANCE LABORATORY

INFORMED CONSENT FORM
University of Calgary
Faculty of Kinesiology
Human Performance Laboratory

Project Title: The Long-Term Effects of ACL Injury on Bilateral Limb Asymmetry and Muscle Activation in Elite Alpine Ski Racers

Investigators: Walter Herzog, Matthew Jordan, Drew Lawson

This consent form, a copy of which has been given to you, is only part of the process of informed consent. It should give you the basic idea of what the research is about and what your participation will involve. If you would like more details about something mentioned here, or information not included here, you should feel free to ask. Please take the time to read this form carefully and to understand any accompanying information.

Background

Anterior cruciate ligament (ACL) injury is very common amongst elite alpine skiers, and despite significant scientific attention and equipment modification injury rates have remained unchanged. Furthermore, after the first ACL injury, elite alpine skiers are at considerable risk for re-injury suggesting that future research be done into the long-term effects of ACL injury.

In other populations it has been shown that over the long-term, ACL injury results in significant bilateral asymmetries during multi-joint movements such as jumping and squatting, and neuromuscular deficits. These asymmetries and deficits are linked to ACL injury and the development of early osteoarthritis. Elite alpine skiers have shown asymmetries in the propulsive phase of jumping, however landing asymmetry has not been assessed. A large percentage of ACL injuries in elite ski racers occur during jump landing. Therefore, asymmetries during the landing phase of jumping warrants investigation.

Given the high rates ACL re-injury in elite alpine skiers, the well known long-term effects of ACL injury on bilateral asymmetries in other populations, and the presence of bilateral symmetry in healthy elite alpine ski racers it is proposed that research be undertaken to better understand the long-term effects of ACL injury on elite alpine ski racers.

Purpose

The purpose of this research project is to investigate the long-term effects of ACL injury on bilateral limb asymmetry during jump landing, isometric hip external rotation and isometric hip abduction. It is hoped that the current research project will improve the understanding of the long-term effects of ACL injury on elite alpine ski racers, and lead to better rehabilitation, screening and testing protocols.

Explanation of Subject's Involvement

As a participant in this study you are required to attend one testing session of approximately 90 minutes in duration. You will perform three maximal effort contractions of hip external rotation, hip internal rotation, hip adduction and hip abduction in an isometric dynamometer to assess hip muscular strength. Both limbs will be assessed. You will then perform a series of drop-and-stick landings from a 0.5-meter box onto a dual force plate system to assess landing ability.

Risk and Discomforts

The risks involved are minimal. There might be some discomfort or post-test joint or muscle pain of short duration. There is some potential for minor muscle strain.

Research Related Injury

In the event that you suffer injury as a result of participating in this research there will be no compensation provided to you by the University of Calgary, the Calgary Health Region or the Researchers. You still have all your legal rights. Nothing said in this consent form alters your right to seek damages.

Benefits to be Expected

If you agree to participate in this study, there may or may not be a direct medical benefit for you. If you have previously suffered an ACL injury, the results of this study may be of benefit to your rehabilitation. Further, this study will assist the researchers in developing a better understanding of the long-term effects of ACL injury on performance and neuromuscular function in elite alpine ski racers. The results of this investigation will lead to further research to optimize and improve rehabilitation protocols, and to develop better screening and testing protocols for alpine ski racers returning from ACL injury.

Do I have to Participate?

Participation in this study is voluntary. You are free to withdraw from the study at any point by informing any of the lead investigators. In no way will your voluntary withdrawal affect you. Furthermore the investigators reserve the right to withdraw from the study should any factor arise that may affect the research question.

Costs for the Participants

There are no costs associated with your participation in this study.

Privacy of Your Records

Information obtained during this research project is confidential. It will not be released without your written consent. The information however, may be used for statistical analysis or scientific purposes with your right to privacy retained. To prevent the invasion of privacy through a digital medium, all computerized data will be saved on a password protected hard drive. All passing of information between computers will be done only with the use of an external hard drive eliminating the need of a network transfer of information. Three years following the final day of data collection all files will be destroyed. Files saved on disk will be erased and hard copy files will be shredded. Identification of subjects through publication will be prevented by the use of the Subject ID Codes.

Freedom of Consent

Your signature on this form indicates that you have understood to your satisfaction the information regarding participation in the research project and agree to participate as a subject. In no way does this waive your legal rights nor release the investigators, or involved institutions from their legal and professional responsibilities. You are free to withdraw from the study at any time without jeopardizing your health care. If you have further questions concerning matters related to this research, please contact:

Drew Lawson (Ph. 613-246-3739 or Dr. Matthew Jordan (Ph. 403-714-4655) or Dr. Walter Herzog (Ph. 403-220-8525)

If you have any questions concerning your rights as a possible participant in this research, please contact the Chair Conjoint Health Research Ethics Board, University of Calgary at 403-220-7990.

Signatures

Participant	Signature and Date
Investigator Name	Signature and Date
Witness' Name	Signature and Date

A signed copy of this consent form has been given to you to keep for your records and reference.

HUMAN PERFORMANCE LABORATORY

INFORMED CONSENT FORM

University of Calgary
Faculty of Kinesiology
Human Performance Laboratory

INFORMED CONSENT FORM

Project Title: The Long-Term Effects of ACL Injury on Bilateral Limb Asymmetry and Muscle Activation in Elite Alpine Ski Racers

Investigators: Walter Herzog, Matthew Jordan

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Background

Anterior cruciate ligament (ACL) injury is very common amongst elite alpine skiers, and despite significant scientific attention and equipment modification injury rates have remained unchanged. Furthermore, after the first ACL injury, elite alpine skiers are at considerable risk for re-injury suggesting that future research be done into the long-term effects of ACL injury.

In other populations it has been shown that over the long-term, ACL injury results in significant bilateral asymmetries during multi-joint movements such as jumping and squatting, and neuromuscular deficits. These asymmetries and deficits are linked to ACL injury and the development of early osteoarthritis. While there have been no studies to date evaluating the bilateral limb asymmetries in elite alpine skiers with a history of ACL injury, healthy skiers display marked symmetry. This may be an important performance indicator given the extreme physical demands of alpine ski racing.

Given the high rates ACL re-injury in elite alpine skiers, the well known long-term effects of ACL injury on bilateral asymmetries in other populations, and the presence of bilateral symmetry in healthy elite alpine ski racers it is proposed that research be undertaken to better understand the long-term effects of ACL injury on elite alpine ski racers.

Purpose

The purpose of this research project is to investigate the long-term effects of ACL injury on bilateral limb asymmetry during the squat jump, isokinetic knee extension, isokinetic knee flexion, and muscle activation in the hamstrings and quadriceps muscles in elite alpine ski racers. It is hoped that the current research project will improve the understanding of the long-term effects of ACL injury on elite alpine ski racers, and lead to better rehabilitation, screening and testing protocols.

Explanation of Subject's Involvement

As a participant in this study you are required to undergo five maximal contractions of knee extension and knee flexion in an isokinetic dynamometer, and perform a series of vertical jumps and landings on a force plate system. Throughout the testing protocol surface measurements will be taken from your quadriceps and hamstrings muscles to assess the degree of muscle activation.

Risk and Discomforts

The risks involved are minimal. There might be some discomfort or post-test joint or muscle pain of short duration. There is some potential for minor muscle strain.

Research Related Injury

In the event that you suffer injury as a result of participating in this research there will be no compensation provided to you by the University of Calgary, the Calgary Health Region or the Researchers. You still have all your legal rights. Nothing said in this consent form alters your right to seek damages.

Benefits to be Expected

This study will assist the researchers in developing a better understanding of the long-term effects of ACL injury on performance and neuromuscular function in elite alpine ski racers. The results of this investigation will lead to further research to optimize and improve rehabilitation protocols, and to develop better screening and testing protocols for alpine ski racers returning from ACL injury.

Obligation to Participate and Withdrawal of Consent

Participation in this study is voluntary. You are free to withdraw from the study at any point by informing any of the lead investigators. In no way will your voluntary withdrawal affect your care. Furthermore the investigators reserve the right to withdraw you from the study should any factor arise that may affect the research question.

Personal Information

Information obtained during this research project is confidential. It will not be released without your written consent. The information however, may be used for statistical analysis or scientific purposes with your right to privacy retained. To prevent the invasion of privacy through a digital medium, all computerized data will be saved on a password protected hard drive. All passing of information between computers will be done only with the use of an external hard drive eliminating the need of a network transfer of information. Three years following the final day of data collection all files will be destroyed. Files saved on disk will be erased and hard copy files will be shredded. Identification of subjects through publication will be prevented by the use of the Subject ID Codes.

Freedom of Consent

Your signature on this form indicates that you have understood to your satisfaction the information regarding participation in the research project and agree to participate as a subject. In no way does this waive your legal right nor release the investigator or involved institutions from their legal and professional responsibilities. You are free to withdraw from the study at any time. Your continued participation should be as informed as your initial consent, so you should feel free to ask for clarification or new information throughout your participation. If you have further questions concerning matters related to this research, please contact:

Matthew Jordan (Ph. 403-714-4655) or Dr. Walter Herzog (Ph. 403-220-8525)

If you have any questions concerning your rights as a possible participant in this research, please contact the Chair Conjoint Health Research Ethics Board, University of Calgary at 403-220-7990.

_______________________________ _______________________________

Participant Date

_______________________________ _______________________________

Investigator Date

_______________________________ _______________________________

Witness Date

A copy of this consent form has been given to you to keep for your records and reference.