The Effect of Quadriceps Strength Symmetry and Task Demands on Lower Extremity Biomechanics

by

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Abstract: Anterior cruciate ligament (ACL) ruptures are a common injury in sports with short and long term consequences. Second ACL injury rates, defined as an injury to the contralateral or healthy limb, or a re-graft tear of the reconstructed ACL, are on the rise and can range from 24-49%. Previous literature suggests that quadriceps strength limb symmetry index (LSI) is a predictor of second ACL injury after returning to sport. Since single leg hop tests have been shown to not be a predictor of future second ACL injuries, using a functional task with systematically increased demands may help identify strength asymmetry thresholds at which ACLR individuals lose the ability to biomechanically adapt and present movement profiles that may increase risk for a second ACL injury. Specific to this thesis project, we hypothesized that: 1) ACL reconstructed (ACLR) individuals with <90% quadriceps strength LSI would exhibit biomechanical asymmetries regardless of task demands compared to ACLR individuals with ≥90% quadriceps strength LSI and healthy controls. 2) ACLR individuals with ≥90% quadriceps strength LSI would exhibit biomechanical asymmetries but only when task demands are highest compared to healthy controls. The two purposes of this study are to determine: 1) the effects of quadriceps strength symmetry and task demands on lower extremity biomechanics in ACLR and healthy controls, 2) evaluate asymmetries in single leg tasks. Data was collected on 10 healthy controls and 10 ACLR individuals that were all physically active and scored on the

Tegner survey. All participants were asked to complete three hopping tasks commonly used in return to sport testing: the single leg hop for distance, the triple leg hop for distance, and the 6-meter timed hop. Quadriceps strength was measured using a dynamometer and the results determined our two ACLR groups: ACLR LSI ≥90% and ACLR LSI <90%. Participants performed three vertical drop jumps each from box heights of 30cm, 45cm, and 60cm. Additionally, participants performed single leg landing and single leg takeoff hops targeting 75% of their maximum single leg hop distance obtained earlier in the protocol. The ACLR LSI <90% group displayed biomechanical asymmetries during both the drop jump and single leg hopping tasks. The uninjured limb of the ACLR LSI <90% group displayed significantly higher knee adduction torques upon initial contact with the forceplate compared to the healthy matched control limbs at 30cm and 60cm heights, and ACLR LSI≥90% uninjured limbs at all three landing heights ($p=0.018$, observed power=0.81). During the single leg hopping trials, the ACLR LSI <90% group displayed a reduction in quadriceps efforts in their injured limb vs non-injured limb as demonstrated by reduced knee energy generation (hop landing) and absorption (hop takeoff), while also maintaining similar ground reaction forces during both single leg landing and takeoff trials ($p=0.005$, observed power= 0.90). The current thesis had several limitations that could have masked the results obtained: 1) the ACLR groups had small sample sizes, 2) the Tegner scale scores were statistically different between groups with the highest level of activity present in the ACLR LSI <90% group. Overall, ACLR LSI<90% exhibited movement characteristics in both the double leg drop jumps and single leg hopping tasks that suggest they are at heightened risk for a second ACL injury. Future research efforts should substantiate these findings as well as attempt to explain why these movement compensations occur post- ACL reconstruction.

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Chapter I. Introduction

The anterior cruciate ligament (ACL) is a commonly injured ligament with incidence rates up to 250,000 in the United States annually.[1] The number of ACL injuries is rising due to increase participation in higher level athletics at younger ages, resulting in an estimated 50% of ACL injuries occurring in young athletes ages 15 to 25.[2,3] ACL injuries have short and long term consequences for the individual and for society. One significant example is that individuals who have sustained an ACL injury are at an increased risk of developing post-traumatic knee osteoarthritis (OA). Over 50% of individuals who suffer an ACL injury develop knee OA 10-20 years after the injury.[4] The economic burden of ACL injuries with their associated consequences is between $7.6 billion annually when treated with anterior cruciate ligament reconstruction (ACLR) and $17.7 billion annually when treated with rehabilitation only.[5] Although ACLR lowers the economic burden of OA, it does not mitigate OA development and second ACL injuries after returning to sport are common.

While the incidence rates of ACL injuries and ensuing reconstruction surgeries have risen, second injury rates are also high and range from 24-49% in active individuals who return to sport (RTS) within two years of ACLR.[6,7] The clinical RTS screening following ACL reconstruction typically includes: time since surgery, strength symmetry, performance-based functional testing, and patient-reported function tests.[8–10] The purpose of the screen is to determine whether the athlete is ready to RTS. While reconstructive surgeries allow athletes to RTS within a year, high second ACL injury rates suggest an inadequacy in the current RTS criteria.

The RTS criteria has been questioned in recent literature due to high second injury rates. While RTS is usually recommended after 6 months, the risk of sustaining a second ACL injury is

highest during the first year after ACLR.[11,12] Welling et al assessed patients at 6 and 9 months after ACLR using functional tests, quadriceps and hamstring strength, and patient-reported outcome questionnaires.[13] The results showed that only 3.2% of patients passed all RTS screenings at 6 months post-ACLR, and only 11.3% of patients passed all RTS screenings 9 months post-ACLR.[13] These low RTS pass rates suggest the athletes are being cleared too early and the risk of a second injury is potentially greater. Grindem et al also found that 39.5% of individuals who RTS earlier than 9 months after ACLR sustained a second injury, while only 19.4% of individuals who waited later than 9 months after ACLR sustained a second injury.[12] Grindem et al also found that for every 1 month delay up to 9 months after ACLR, the second injury rate was reduced by 51%.[12] While length of time since ACLR has been linked to decreasing second ACL injury risk, second injury rates overall are still too high given the short and long term consequences associated with ACL injuries.

Patient-reported outcome(PROs) assessment tools specific to the knee joint, such as the International Knee Documentation Committee (IKDC), have been regarded as an important RTS criteria after ACLR.[14] PROs are used to measure patient perspective on how the knee joint affects daily life and sports activities.[15] Zwolski et al used predicted that individuals who score high on the IKDC would also have quadriceps strength symmetry.[9] The results concluded that having an IKDC score ≥ 94.8 predicted whether the individual possessed the healthy population quadriceps limb symmetry index (LSI) which is ≥90% with a high sensitivity of (.813).[9] Considering the importance of muscular symmetry in the RTS criteria and the cost of a dynamometer to determine muscular strength, Zwolski et al demonstrated how PROs could be used to aid in the RTS process.[9] However, to date, while PROs are used in the RTS screening process, there is no evidence suggesting that PRO scores predict risk of a second ACL injury.[12]

Performance-based tests are utilized in the RTS criteria due to their ability to quantify knee joint ability in a functional environment.[16] The most commonly used performance-based tests are: the single hop test for distance, triple hop test for distance, the crossover hop for distance, and the 6-m timed hop. Hop tests are commonly used to calculate a functional limb symmetry index; however, using the uninvolved limb as a reference to the involved limb is a growing concern.[17] Bilateral strength and functional deficits may occur from inhibition of motor activation, de-conditioning, or inadequate recondition, which could result in a falsely high limb symmetry index of ≥90%.[18] Gokeler et al found that while 83% of individuals after ACLR had a LSI ≥90%, the hop distance was still 16-19% shorter than the control group.[19] While ACLR athletes may have an LSI above 90%, hop tests may underestimate performance deficits.[19] Thus, while performance-based tests are a common screening tool used in the RTS criteria, bilateral muscular and functional deficits could falsely clear athletes too soon to their sport.

Muscular symmetry makes up a large majority of the RTS criteria. One reason athletes may not successfully return to their sport is because they do not regain their pre-injury muscle function.[20] Since bilateral hamstring and quadriceps strength deficits <10% can be seen in the healthy population,[21] current RTS guidelines state limb symmetry indices ≥90% relate to "normal" strength.[14] Out of all of the current RTS criteria, only quadriceps strength symmetry (along with waiting until 9 months post-surgery as indicated earlier) is significantly related to second ACL injury risk.[12] Therefore, understanding the role muscular strength symmetry has on biomechanical function is critical to ultimately refine RTS criteria and mitigate the chances of a second injury.

Despite hop test symmetry, ACLR and ACL deficient (ACLD) populations change the way they accomplish the functional hopping tasks which may mask underlying "risky" lower

extremity biomechanics and predispose them to a second injury. Strategies to off-load the knee joint on the involved limb despite having hop distance deficits <10% compared to uninvolved limbs during hop tests are achieved through increased efforts by the hip extensors and the ankle plantarflexors.[22,23] Thus, because traditional hop tests rely on an individuals' ability level, using a standardized task may be better suited to identify performance asymmetries. Further, using a functional task with systematically increased demands may also help to identify strength asymmetry thresholds at which point ACLR lose the ability to biomechanically adapt and present with movement profiles that may increase risk for a second ACL injury.

Purpose

The two purposes of this study are to determine: 1) the effects of quadriceps strength symmetry and task demands on lower extremity biomechanics in ACLR and healthy controls, 2) evaluate the effect of quadriceps strength symmetry on lower extremity biomechanical symmetries in single leg hop tasks.

Hypothesis

The overall hypothesis is that biomechanical asymmetries are dependent on both quadriceps strength symmetry and task demands.

The hypotheses tested in this thesis are: 1) ACLR individuals with a quadriceps limb symmetry index (LSI) <90% would exhibit biomechanical asymmetries regardless of task demands compared to ACLR individuals with a quadriceps LSI ≥90% and healthy controls. 2) ACLR individuals with a quadriceps LSI ≥90% would exhibit biomechanical asymmetries but only when task demands are highest compared to healthy controls.

Significance

The significance of this study is that the results could suggest a possible need for modifying the RTS criteria. Given that hop test screens have not been shown to predict second injury risk combined with the idea that many clinicians do not have expensive isokinetic equipment to screen for strength asymmetries, assessing how strength asymmetry affects lower extremity biomechanics using a standardized functional task is important towards modifying the functional testing portion of RTS criteria. This thesis provides insight as to what role quadriceps symmetry plays in the RTS criteria, and if increased task demands have an increased effect on lower extremity biomechanics.

Delimitations

1.) ACLR subjects were males and females ages 18-28 and previously cleared for RTS by their surgeon.
2.) ACLR subjects had their reconstruction surgery no longer than 5 years before testing.
3.) Control subjects were males and females ages 18-28 and without a history of lower extremity pain or injury, neuromuscular or musculoskeletal diseases, or other orthopedic problems.
4.) Biomechanical analyses focused on bilateral lower extremity joint positions, moments, and energetics.

Limitations

1.) The analyses are limited to the accuracy of the equipment used: force plate, motion analysis system, and dynamometer.

2.) The thesis results will not be able to provide information related to variables associated with second injury risk because the study follows a retrospective design comparing ACLR to healthy controls.

Assumptions

1.) The equipment placed on the subjects' limb will not interfere with natural movements i.e.. hop testing and drop jumping.

2.) Net joint torque and joint forces are assumed to be applied from a sum of muscle forces, passive tissue forces, and frictional force of joints.

Operational Definitions

1) Muscular Symmetry – The magnitude of side-to-side differences in the variable assessed such as muscular strength, or knee joint torques for example.

2) Muscular Imbalances – The ratio of strength of one muscle relative to its antagonist i.e. quadriceps / hamstring strength ratios.

   *While symmetry and imbalances are often used interchangeably in the literature, these operational definitions will be used for consistency purposes in this thesis.

3) Limb Symmetry Index – A calculation commonly used to determine discrepancies in an individual's limbs. In this thesis, the LSI was obtained by dividing the injured limb (or injured matched for healthy controls) by the uninjured limb (or uninjured matched for controls) and multiplying it by 100 to acquire a percentage. Therefore, a LSI below 100% indicated that the injured or injured matched limb was lower compared to the uninjured limb or uninjured matched-control limb. Conversely, if the LSI was above 100% this

would indicate the injured or injured-matched limb was greater than the uninjured limb or uninjured matched-control limb.

4) Task Demands – For this thesis, we are referencing the increases in height of the drop jump task as reflective of an increase in task demands.
5) Return to Sport – Clinician approved decision for an athlete to return to their sport of choice for unrestricted activities.
6) Second Injury – A second injury is referring to an injury to the contralateral/uninjured limb, or graft tear of the injured limb. This is common terminology in the literature when referring to any ACL injury following the initial ACL reconstruction.
7) Re-injury- A graft tear to the injured and previously reconstructed limb of an individual.

Chapter II: A review of the literature

Introduction to the Review of the Literature

The purpose of this study is to determine the effects of quadriceps strength asymmetry and task demands on lower extremity biomechanics in ACLR and healthy controls. The following literature review will discuss: **1)** the epidemiological problem of Anterior Cruciate Ligament (ACL) injuries and reconstruction in athletes, **2)** the current state of the RTS criteria, **3)** the role muscular symmetries play in return to sport (RTS) criteria, and **4)** identifying the need for understanding how muscular strength symmetry affects lower extremity biomechanics in ACL injured populations.

ACL Injury Incidence and Long-Term Consequences

The anterior cruciate ligament (ACL) is a commonly injured ligament with a high prevalence rate in sports. Up to 250,000 ACL injuries occur in the United States annually, with an estimated 50% of injuries occurring in young athletes ages 15 to 25 years[1–3]. The number of ACL injuries is rising with increased participation in higher level athletics at earlier ages, even with increased awareness of the injury and improved diagnostic methods[24]. Consequently, there has also been an increase in the rate of Anterior Cruciate Ligament reconstruction (ACLR) per 100,000 people from 17.6 in 1990 to 50.9 in 2009 in New York State alone.[25] Despite reconstructive efforts, the long term consequences of associated with ACL injuries, and the following reconstructions, are still poor.

Knee osteoarthritis (OA) is one of the most devastating long-term consequences associated with ACL injuries. Athletes who have sustained an ACL injury are at an increased risk of developing post-traumatic knee OA.[26] OA is described as, “a common, age related, loss

of articular cartilage in synovial joints associated with varying degrees of osteophyte formation, subchondral bone change, and synovitis".[27] OA leads to pain and functional impairment in the young or middle-aged adult.[27] Given that a majority of athletes who suffer ACL tears are less than 30 years of age, ACL injuries are responsible for a large number of individuals with early-onset OA in the ages between 30-50.[28] The reported rates of OA 10-20 years after an ACL injury average over 50%.[27] Among averaged adults aged 25-44, the prevalence of symptomatic knee OA is approximately 1-5% in women and 1-4% in men[36], which is significantly lower than the reported rates of OA among individuals who have suffered an ACL injury 10-20 years prior. Since over 50% of ACL injuries occur in athletes between the ages of 15 to 25, over 50% of these athletes will begin developing symptomatic knee OA by the time they are 30 to 40 years old.

The risk of developing OA post-ACL injury has increased the economic burden associated with ACL tears. The average lifetime cost to society for a typical patient undergoing ACLR is roughly $38,000, while the average cost for rehabilitation is $88,500.[5] The lifetime burden of ACL injuries with the associated consequences of OA is $7.6 billion annually when treated with ACLR and $17.7 billion annually when treated with rehabilitation only.[5] While the economic burden of OA is lower following ACLR, evidence shows that even when reconstructed, this is still not an effective intervention to mitigate OA development and second ACL injuries after RTS are common.[29]

While the incidence rates of ACL injuries and ensuing reconstruction surgeries have risen, second-injury rates range from 24-49% in active individuals who RTS within two years of ACLR.[6,7] Second-injury rates reflect injuries to the contralateral <u>and</u> the operated limb combined. Paterno et al examined incidence rates of second ACL injuries and acknowledged graft tears to

the injured limb or a tear to the contralateral limb as a second ACL injury.[38] While reconstructive surgeries allow athletes to return to sport within a year, clearly the need exists to improve on post-surgical treatment and return to sport criteria such that the risk of a second tear is mitigated.

The Current State of the Return to Sport (RTS) Criteria

The clinical RTS screening following ACL reconstruction typically includes: time since surgery, strength symmetry, performance-based functional testing, and patient-reported function tests.[8–10] The point of the screen is to determine readiness to return to sport and implies: 1) the operated limb is normal, and 2) second injury risk is minimized. Given, second ACL injury rates are as high as 24-49%, this suggests an inadequacy in the current RTS screening and criteria.[6,7]

The RTS criteria has been questioned in recent literature due to increasing second injury rates. Traditionally, RTS is recommended after 6 months. However, the risk of sustaining a second ACL injury is highest during the early period of RTS (6-12 months after surgery).[11,12] In addition, approximately half of all graft ruptures occur within the first postoperative year in athletes 25 years or younger.[30] Welling et al assessed changes in biomechanical function overtime in patients tested at 6 months and 9 months after ACLR.[13] The screening included: a jump-landing task assessed with the Landing Error Scoring System(LESS), three single-leg hop tasks, isokinetic quadriceps and hamstring strength, and two questionnaires (ACL-RSI and the IKDC).[13] The results showed that only 3.2% of patients passed *all* RTS criteria at 6 months after ACLR, and only 11.3% of patients passed all RTS criteria at 9 months after ACLR.[13] In a cohort study, Grindem et al found that for every 1 month delay in RTS, until 9 months after ACLR, the knee second injury rate was reduced by 51%.[12] In addition, patients who participated in level I sports earlier than 9 months after ACLR sustained second injuries rates as high as 39.5%

compared to 19.4% in patients who waited later than 9 months after ACLR to return to level I sports.[12] These results have led to an adoption of 9 months minimum in order to RTS. High second injury rates during the first year after ACLR also highlights why it is critical to standardize all clinical RTS criteria that decreases second injury risk.

Patient-reported outcome (PRO) assessment tools specific to the knee joint, such as the International Knee Documentation Committee (IKDC), have been regarded as an important measure of successful outcome after ACLR and used in RTS screening.[14] PROs measure the patient's perspective on how the knee joint affects daily life and sports activities.[15] PROs are also used to determine the severity of the injury and to track progress over time.[15] The IKDC is a commonly used PRO that assesses symptoms, sports activity, and function, making it highly relevant for individuals after ACLR.[31] The IKDC is scored on a 0-to-100 scale, with a higher score representing higher knee function.[31] Zwolski et al examined whether the IKDC could be used as a screening tool for quadriceps strength deficits.[9] They separated subjects into two groups: a high IKDC score group (≥90) and a low IKDC score group (<90), and predicted that the high IKDC score group would have high quadriceps strength of the involved limb and normal quadriceps strength symmetry (≥90%).[9] The results concluded that having a IKDC score of ≥94.8 predicted whether the patient would possess a quadriceps strength limb symmetry index(Q-LSI) ≥ 90% with a high sensitivity of (.813).[9] Due to cost, skill, and time to administer dynamometer tests, the results from Zwolski et al demonstrate how PROs can be effectively used to aid in the RTS process.[9] However, PROs have not been shown to predict a decrease in second injury rates.[12]

Performance-based tests are utilized in the RTS criteria due to their ability to quantify knee joint function in a functional environment.[16] Hop tests, such as the single hop test for

distance, triple hop test for distance, the crossover hop for distance, and the 6-m timed hop are all performance-based functional tests with extensive research supporting their reliability.[16] Hop tests are commonly used to calculate a functional limb symmetry index (LSI). However, there are concerns regarding the use of the uninvolved limb as a reference to the involved limb. Research indicates that comparison with the contralateral side may not be ideal due to bilateral neuromuscular deficits after ACL injuries and reconstruction.[19] Thigh muscle strength and functional deficits may occur from inhibition of motor activation, de-conditioning, or inadequate reconditioning.[18] Therefore, a bilateral deficit could result in falsely high limb symmetry index, since the LSI is calculated as a ratio between the values of the limbs. Gokeler et al found that athletes had bilateral deficits on four different hop tests post ACLR compared with controls.[19] While 83% of the ACLR subjects had a LSI ≥90%, the jump distances of the ACLR group were 16-19% shorter when compared to the control group.[19] These results highlight that while ACLR subjects may have a LSI above 90%, muscle strength and function is still lower when compared to a healthy population.[19] Wellsandt et al found that patients who met a 90% symmetry criterion for strength and functional tests 6 months post-ACLR would not be cleared for RTS if compared against performance on the contralateral limb ***before*** ACLR.[32] While performance-based function tests are a common screening tool used in the RTS criteria, as stated earlier, high second injury rates and bilateral muscular and functional deficits jeopardize the legitimacy of these tests to clear patients for RTS. In addition, functional hop tests have yet to ever be shown as a significant predictor for a second ACL injury.[12]

Muscular strength symmetry in the RTS criteria

Muscular symmetry makes up a large majority of the RTS criteria. One reason athletes may not successfully return to their sport is because they do not regain their pre-injury muscle function.[20] Muscular strength of quadriceps and hamstrings are primarily obtained using isometric and/or isokinetic dynamometer tests. Limb symmetry indices (LSI) allow researchers to quantify the muscular strength on an athletes operated limb compared to the healthy and non-operated limb. Hamstring and quadriceps strength deficits < 10% can be expected in the normal population.[21] Current RTS guidelines state quadriceps and hamstring limb symmetry indices ≥90% relate to good strength and higher patient-reported outcome scores.[14] Schmitt et al studied quadriceps femoris (QF) asymmetry during functional testing on ACLR patients who had been cleared to RTS.[33] The study divided the ACLR patients into two categories: one group with QF strength deficits >15% (quadriceps LSI ≤85%) and another group with QF strength deficits <10% (quadriceps LSI ≥90%).[33] The results showed that QF strength deficits >15% negatively affect function and performance, while QF strength deficits <10% demonstrate functional performance similar to that of an uninjured individual.[33] These results suggest patients with strength deficits <10% are ready to return to sport at competitive levels because they *appear* to be "functionally equivalent" to healthy controls.[33] Grindem et al found that out of their 69 ACLR subjects, 18 (26%) suffered a second ACL injury after RTS.[12] The average quadriceps strength LSI was 75% in the group that suffered a second injury, while the group that did not suffer a second ACL injury within a year of returning to play had a quadriceps strength LSI of 84.4%.[12] Their analysis revealed that for every 1 percentage increase in quadriceps strength LSI, a 3% reduction in second injury rates can be seen.[12]

Out of all the RTS criteria combined, only length of time since ACL reconstruction and quadriceps strength LSI have been shown to be linked to decreasing second ACL injury risk.[12] In addition, given such high second injury rates, there is a need for a better understanding as to what role muscle strength symmetry plays in the RTS criteria.

Muscular strength symmetry on lower extremity biomechanics in ACL injured populations

Muscle strength asymmetries or imbalances affect biomechanical function in ACLR and ACL deficient individuals. While muscle asymmetries refer to the side to side differences of a particular muscle i.e. quadriceps asymmetry, muscle imbalances generally refer to the ratio of a muscle relative to its antagonist such as the hamstring/quadriceps ratio. While both thigh muscle asymmetries and imbalances pertain to the ACLR literature, this section is focused on how thigh muscle asymmetries affect biomechanical function in both healthy and ACLR populations. Understanding how muscular strength asymmetries affect biomechanical variables associated with ACL injuries could help to refine current RTS criteria and potentially reduce second injury rates.

While asymmetrical hop test performance is clearly detectable in ACLR individuals with gross quadriceps strength deficits (>15%), deficits in hop testing performance in individuals with past ACL injury (muscle strength deficits <10%) are not generally seen because this group of individuals compensate their movements to accomplish the task. In individuals with ACL deficiencies, despite having quadriceps strength deficits and hop distance deficits both of <10% compared to uninvolved limbs, knee extensor torques in the involved limbs during the landing phase of the hop were 167±54Nm in the ACL deficient limb vs 224±80Nm in the non-injured which equates to a knee extensor torque deficit of ~25%.[22] This strategy to off-load the knee

joint during the single leg hop tests was suggested to be compensated by increased efforts by the hip extensors where hip extensor moments were on average ~39% higher than the knee extensor torques in the injured limb while hip extensor moments were only ~4% higher than knee extensor efforts in the noninvolved limb.[22] These compensatory movement patterns to off-load the knee joint during single leg activities are also prevalent in ACLR individuals despite having hop test deficits <10% on average.[23] In the ACLR limbs during landing, peak joint powers were 43% lower in the knee, 19% lower in the hip, but 42% higher at the ankle compared to the uninvolved limb.[23] Because individuals with previous ACL injury clearly compensate how they move to achieve hop test symmetry, it is not surprising that hop testing symmetry has never been shown to be a significant predictor for a second ACL injury.[12] In addition, even in the presence of hop test symmetry, ACLR individuals still do not hop as far as uninjured[17] and healthy individuals. Thus, because traditional hop tests rely on an individuals' ability level, using a standardized task may be better suited to identify performance symmetries. Vertical drop jumps have been shown to be a successful indicator of future ACL injuries.[37] Hewett et al performed a cohort study in which female volleyball players performed vertical drop jumps at a standard height of 31cm before their season.[37] The study concluded that at initial contact, knee valgus moments were an indicator of future ACL injuries when performing vertical drop jumps.[37] Thus, using a functional task with systematically increased demands may also help to identify strength asymmetry thresholds at which point ACLR lose the ability to biomechanically adapt and present with movement profiles that may increase risk for a second ACL injury.

Purpose

The two purposes of this study are to determine: 1) the effects of quadriceps strength symmetry and task demands on lower extremity biomechanics in ACLR and healthy controls, 2) evaluate

the effect of quadriceps strength symmetry on lower extremity biomechanical symmetries in single leg hop tasks.

Hypothesis

The overall hypothesis is that biomechanical asymmetries are dependent on both muscle strength symmetry and task demands.

The hypotheses to be tested in this thesis are: 1) ACLR individuals with <90% quadriceps strength LSI would exhibit biomechanical asymmetries regardless of task demands compared to ACLR individuals with ≥90% quadriceps strength LSI and healthy controls. 2) ACLR individuals with ≥90% quadriceps strength LSI would exhibit biomechanical asymmetries but only when task demands are highest compared to healthy controls.

Summary

ACL injuries have negative long-term consequences, and many individuals suffer second ACL injuries. Because currently used hop testing tests do not predict second ACL injuries, it is critical to develop functional testing screens to be used in RTS criteria that mitigates second injury-rates in athletes. Further, using a functional task with systematically increased demands may also help to identify strength asymmetry thresholds at which point ACLR lose the ability to biomechanically adapt and present with movement profiles that may increase risk for a second ACL injury. We propose that ACLR individuals with <90% quadriceps strength LSI will exhibit biomechanical asymmetries regardless of task demands; while ACLR individuals with ≥90% quadriceps strength LSI will exhibit biomechanical asymmetries only when demands are highest.

Chapter III: Methods

Introduction

This research investigation was a cross-sectional study comparing two groups of ACLR individuals to healthy controls. Based on the reviewed literature, two hypotheses were formulated: ACLR individuals with quadriceps strength LSI <90% will exhibit biomechanical asymmetries regardless of task demands compared to ACLR individuals with quadriceps strength LSI ≥90% and healthy controls; second, ACLR individuals with quadriceps strength LSI ≥90% will exhibit biomechanical asymmetries but only when task demands are highest compared to healthy controls. To test these hypotheses, lower extremity biomechanical analyses were performed while healthy and ACLR subjects complete drop jump tests at 30cm, 45cm, and 60cm. This section provides a summary of the participant characteristics, inclusion/exclusion criteria, equipment, procedures, statistical analysis used to test our hypothesis as well as the secondary purpose. The secondary purpose was to evaluate the effect of quadriceps strength symmetry on lower extremity biomechanical symmetries in single leg hop tasks. Although not included in our hypothesis, we wanted to examine the relationship between quadriceps strength and single leg hopping task in order to compare our results with double leg drop jumps as well as previous research.

Subjects

Participants in this study were college-aged individuals between the ages of 18-25. All of the ACLR subjects had their reconstructive surgery less than five years before testing and were self-reportedly medically cleared for unrestricted activities. All subjects completed the informed consent process approved by the University UMCIRB. ACLR subjects were divided into two

groups based on their quadriceps strength limb symmetry index: ACLR individuals with a quadriceps strength limb symmetry index ≥ 90% and ACLR individuals with a quadriceps strength limb symmetry index <90%.

Inclusion criteria

Control group

1. Recreationally active (as assessed by the Tegner activity scale) healthy adults with no history of knee surgeries or injuries.
2. Aged 18-28

ACLR group

1. ACL reconstructive surgery in the past 5 years.
2. Self-reportedly medically cleared for unrestricted activities.
3. Age 18-28
4. Recreationally active as assessed by the Tegner activity scale with a minimum score of 5.

Exclusion criteria

Control group

1. Lack of physical activity (< 5) based on Tegner scale
2. Previous knee injuries or surgeries.
3. Age < 18 or > 28years.

ACLR group

1. Not medically cleared for unrestricted activities (self-reported)
2. ACLR surgery longer than five years before testing
3. Lack of physical activity (< 5) based on the Tegner scale.

4. Age < 18 or > 28years.

Procedures

Prior to each data collection, all equipment was calibrated and maintained according to the factory protocols. Prior to the participants arrival, the force plate was located using an L frame to mark the coordinates in the system and 4 placement markers were used to define the area of the platform. The testing area and motion capture location were calibrated with a 600mm T-wand with an accepted calibration trial consisting of less than or equal to a 2mm average residual error per camera capture. Before motion analysis, the global coordinate system was calibrated by waving the calibration wand across the area of interest. The calibration wand, which contains two spherical markers on each end, creates the global coordinate system and also calibrates the motion capture software. Upon arrival to the lab on testing day and following completion of informed consent, all participants were administered the Knee Injury and Osteoarthritis Outcome Score (KOOS) questionnaire. The KOOS has been shown to be a reliable and valid test for ACLR individuals to rate their overall knee function.[35] The KOOS questionnaire consists of five subscales: pain, function of daily living, function in sport and recreation, knee related to quality of life, and other symptoms. The previous week is the time period considered when answering the questions. While not specifically related to the hypotheses, collecting data on the KOOS allowed for a better description of the two ACLR groups after partitioning them based on the quadriceps strength LSI criteria.

Once the KOOS questionnaire was completed, all participants height and weight were taken using a Seca 703 digital scale (Seca GMBN & C. Kg, Hamburg, Germany). Next, all participants performed a standard hop testing battery commonly used in RTS evaluation.[19,33] These clinical standard measurements were useful for comparison of our sample to the literature.

All subjects performed three single-leg hop tests, including the single hop for distance (cm), triple hop for distance (cm), and 6-meter timed hop (seconds). The goal of the single and triple hop for distance is for the subject to jump as far as possible while maintaining a controlled landing on the ipsilateral limb. The goal of the timed hop test is for the subjects to hop on a single leg as quickly as possible over a 6-meter distance. Once familiar with each of the tasks, three trials on each leg were measured. 75% of the maximal single-leg hop test for distance was used later for motion analysis.

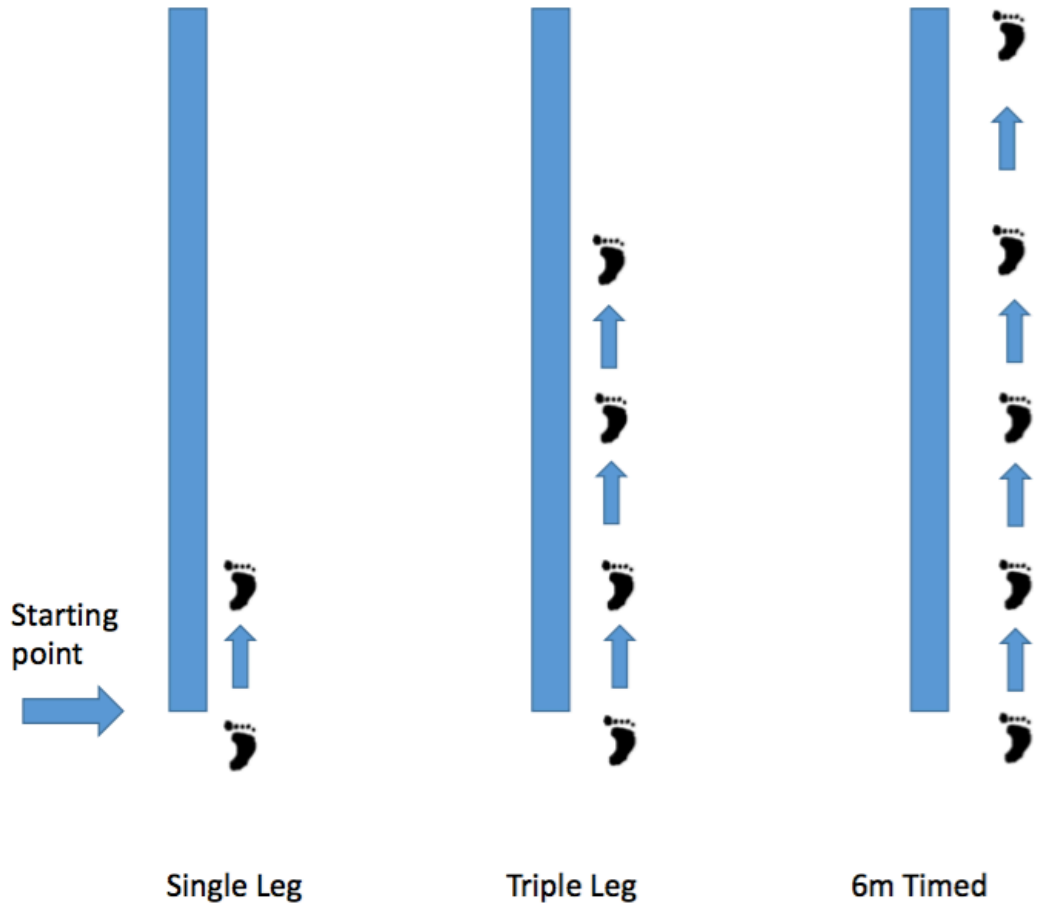

Figure 1: Function Hop Test Diagram

All participants regardless of group then underwent bilateral strength testing using a HUMAC NORM Dynamometer (CSMI, model 502140, Stoughton, MA). With participants

seated in the dynamometer with hip flexed at 90°, the lateral epicondyle of the knee joint was lined up with the axis of rotation of the dynamometer arm. To secure the subject, a strap went across the subject's chest. The lower leg was secured to the dynamometer arm and chair. To familiarize the subjects with these tests, the subjects performed 3 trials at 50% of the participant's expected maximum effort, with a 1-minute rest in between each trial. These trials were used to ensure proper stabilization of the subject in the dynamometer. After the familiarization trials, five maximal concentric knee extensor (quadriceps: vastus lateralis, vastus intermedius, vastus medialis, and the rectus femoris) and flexor (hamstrings: semitendinosus, semimembranosus, and the biceps femoris long-and short-heads) repetitions were tested on each limb at 60 °/second. After isokinetic testing, five repetitions of maximal isometric knee extension and flexion efforts will be tested on each limb at 60° of knee flexion after 3 familiarization trials are completed.[9] Isometric knee extensor and flexor torques were processed to determine the LSI of each participant.

Biomechanical analysis

After completing the bilateral strength testing, spherical markers were placed on the subject to define segments of the lower limbs, pelvis, and trunk. Markers were placed on the left and right iliac crest, left and right ASIS and PSIS, and the right and left greater trochanters to define the pelvis. Trunk markers were placed on the lateral aspect of the acromion processes, the center of the superior aspect of the sternum, and a four-marker shell was worn along the participants spine. The thigh was defined by using the right and left greater trochanters and the right and left medial and lateral femoral epicondyles. The shank was defined using the medial and lateral femoral epicondyles and the medial and lateral malleoli of the right and left leg. The feet were defined using the medial and lateral malleoli and the 1st and 5th metatarsal heads of the

right and left leg and foot. 4-marker rigid plastic shells were placed on the lateral right and left thigh and shank while a 3-marker shell were placed on the top of the right and left foot to capture segment motion during the dynamic trials. A static calibration trial was taken after all of the markers were placed on the correct landmarks. A static calibration is used to define a local coordinate system for each of the segments. After static calibration, the 4-marker and 3-marker shells on the thigh and foot and the markers on the left and right ASIS and PSIS, the left and right greater trochanters, the lateral femoral epicondyles, lateral malleoli, and trunk remained on the subject until completion of the tests, while the other markers were removed.

All participants were instructed to perform a drop jump task from 30cm, 45cm, and 60cm onto a force platform (AMTI, Newton, MA). Following landing with one foot on each forceplate, the subject immediately jumped vertically to attempt to grab a hanging ball from the ceiling. The ball was used to standardize the jump direction in all participants. Participants performed 3 drop-jumping trials at each height. In addition, subjects then performed single leg hops on each limb at the previously determined distance of 75% of their farthest jump obtained during the initial single leg hop test for distance. This included 3 trials where the participant hops onto (i.e. landing on) the forceplate, and 3 trials where the participant hops off (i.e. takeoff) of the forceplate. This was done to collect data on how the participant lands and takes off on each limb during a single leg hop at 75% of their maximum distance. While not directly addressing the purpose and hypotheses of this thesis, the incorporation of a biomechanical analysis of the single leg hop tasks was done to 1) assist with comparison of lower extremity biomechanical variables to the literature and 2) to determine biomechanical sources of compensation for reduced quadriceps strength LSI when an individual is landing on, or taking off from, a single leg. For all drop jump and single leg hopping trials, kinematic data was collected using a 10-

camera motion capture system (Opus 300+ Cameras, Qualisys, Goteborg, Sweden) while kinetic data was captured using force platforms (AMTI, Newton, MA). Each camera was set at a capture frequency of 240 Hz and both force plates were set at 1440 Hz.

Data reduction:

The ACLR groups were then assigned to two groups based on isometric dynamometer results. The limb symmetry index (LSI) for each person was calculated by taking the peak quadriceps torque on the injured and dividing it by the peak quadriceps torque on the uninjured limb and multiplying that value by 100 to obtain a percentage (injured matched and uninjured matched for healthy controls). The symmetric group (ACLR LSI ≥90%) consisted of individuals with a quadriceps strength limb symmetry index ≥ 90% (<10% deficit), while the asymmetric group (ACLR LSI <90%) consisted of individuals with quadriceps strength LSI <90% (>10% deficit). LSI was calculated by the ratio of the surgical limb / nonsurgical limb. In the case of healthy subjects in calculating LSI, the "surgical" limb was a matched limb to the ACLR groups. The 90% threshold was used to define the two ACLR groups because ≥ 90% has been deemed a safe threshold and used in the prediction of second ACL injury.[12]

For motion capture data reduction, once all marker data were identified and tracked using Qualisys Track Manager (QTM), all kinematic and kinetic data was exported for further analysis. Motion capture and force data was processed with Qualisys Track Manager (QTM) Software (Innovation Systems Inc, Columbiaville, MI) and then analyzed using Visual 3-D (V3D) program (C-Motion Inc, Rockville, MD). In Visual 3D, an 8 segment (bilateral feet, shanks, thighs, a pelvis and trunk) model was built from the standing calibration trial. The link-segment model was then applied to all motion trials where all kinematic and kinetic data was calculated. Location of the individual reflective markers within the global coordinate system as well as the

local coordinate system was determined, as well as the virtual joint centers, and each segments center of mass using calibration recordings. Ankle and knee joint centers were determined calculated as 50% of the distance between the medial and lateral malleoli calibration markers and medial and lateral femoral epicondyle calibration markers. A similar method was used for the hip with the joint center being determined by calculating 25% of the distance between the right and left greater trochanters calibration markers. The kinematic data was filtered using a butterworth lowpass filter at 10Hz while force data was low pass filtered at 50Hz. V3D was used to calculate joint reaction forces and joint torques using linear and angular Newtonian equations of motion. Ground reaction force, center of pressure, segmental anthropometrics, kinematic position and acceleration data were used for the calculations to calculate joint toques and joint reaction forces. This approach used inverse dynamics beginning with the segment where the known ground reaction forces come from (foot), moving proximally to the shank, and then the thigh, using the prior segment to calculate the next. An example of how the joint torques are calculated can be seen in Appendix C. Joint powers, or the rate of completing work, are calculated by multiplying the joint torque by angular velocity of the joint. In addition, energy absorption during the landing phase is the area under the power curve and is represented by negative value due to contribution of the eccentric quadriceps muscle effort; while energy generation during the takeoff phase, calculated as the area under the power curve and is represented by a positive value due to the contribution of the concentric quadriceps muscle effort. Both energy absorption and generation are integrations of the power curve and were examined in the drop jumps and single leg landing trials. Joint torque and joint power were low pass filtered at 10hz. Moment impulse, the area under the torque curve, is calculated by taking the torque (Nm) and multiplying it by time (s) was also examined in the drop jumps and single leg landing trials.

Statistical Analyses

Data was expressed as means ± standard deviations. Additionally, box plots were constructed to allow for a transparent visualization of the variance in the data for each of the groups beyond the commonly reported means and standard deviations. Statistical analysis was performed with a 3 (group) x2 (limb) x3 (task) Mixed Model ANOVAs. The three groups were: healthy controls, ACLR quadriceps LSI ≥90%, and ACLR quadriceps LSI <90%. The three task demands were the drop jumps from 30, 45, and 60cm. These analyses allowed for evaluation of significant main effects across the three groups, significant within group effects across tasks and limbs (injured/non-injured) and any interactions. Dependent variables included the following bilaterally: knee valgus angle, knee abduction torque, hip extensor torques, knee extensor torques, ankle plantarflexor torques, ankle energy absorption, knee energy absorption, hip energy absorption. For the single leg hopping tasks (landing and takeoff), 3 (group) x 2 (limb) mixed model ANOVAs were performed using the same dependent variables. Alpha levels were set to 0.05 to determine statistical significance for all analyses.

Matched Controls

The statistical analyses were performed using "injured matched" and "uninjured matched" for the healthy controls. Since there were 7 participants who had ACLR on their left limb, we used seven healthy controls left limb as an "injured matched". For the remaining 3 healthy controls, the right limb was used as an injured matched. This was done to mitigate any bias between the right and left limbs for the healthy controls. To reduce any chance of bias, the first 7 healthy controls were selected to be a part of the left limb injured matched, while the last 3 were selected to be a part of the right injured matched.

Chapter IV: Results

General Demographics and Subject Reported Function Scores

The three groups were not different in terms of their demographics except in two categories (Table 1). Physical activity was measured by the Tegner scale and were significantly different across the three groups (p=0.019, Figure 3). The ACLR LSI <90% group had significantly higher activity levels than both the ACLR LSI ≥90% group and the healthy controls(ACLR LSI<90%=8.4±1.4, ACLR LSI≥90%=6.2±1.5, healthy controls= 6.6±1.4). The criteria that divided the participants into groups, quadriceps limb symmetry index, was also significantly different across the three groups (p=0.030). The ACLR quadriceps LSI <90% group was significantly lower than the ACLR quadriceps LSI ≥90%, while the ACLR quadriceps LSI <90% group was not significantly lower than the control group (ACLR LSI ≥90%=100.18±5.0, ACLR LSI<90%=84.0±5.3). However, there was an outlier in the ACLR LSI <90% group for quadriceps LSI (Figure 2). The range of Tegner scores were from 5-9, meaning participants ranged from recreationally running at least 3 times a week, to 9, meaning he/she participate in competitive sports such as soccer, football, wrestling, or gymnastics. The patient reported scores from the KOOS were statistically different between the healthy (n=10) and the ACLR LSI ≥90% (n=5) group at each of the 5 sub scores (Table 2), while the ACLR LSI<90% group was not significantly different from either group. However, there was an outlier in the KOOS Sport subscale for the ACLR LSI ≥90% group (Figure 4). The healthy KOOS scores were averaged between both limbs. The ACLR participants reconstructed limbs were reported. In addition, no significant differences in the single leg hop for distance, triple leg hop for distance, and the 6-meter timed were found between groups (Table 3). The hop test LSI for the healthy group was

found by the following equation: injured-matched limb / uninjured-matched limb *100. For the ACLR groups, the following equation was used: injured limb distance/healthy limb distance*100.

	Healthy (n=10)	ACLR LSI≥90% (n=5)	ACLR LSI<90% (n=5)	p-value
Sex (male/female)	4/6	2/3	1/4	N/A
Age (yrs)	19.8±0.8	20.4±1.7	19.4±1.1	0.361
Height (cm)	173.8±9.9	176.0±10.2	168.8±5.9	0.457
Mass (kg)	74.1±15.8	73.6±19.9	65.6±7.2	0.588
Tegner	6.6±1.4*	6.2±1.5*	8.4±1.4*	0.019
Quad LSI%	94.7±11.3	100.2±5.0*	84.0±5.3*	0.036
Ham LSI%	99.0±11.3	87.0±11.9	90.4±15.3	0.192

Table 1. Participant Characteristics. ACLR LSI ≥90% represents the ACLR group with a quadriceps LSI ≥90%. ACLR LSI <90% represents the ACLR group with a quadriceps LSI of <90%. Mean±SD % *p<0.05

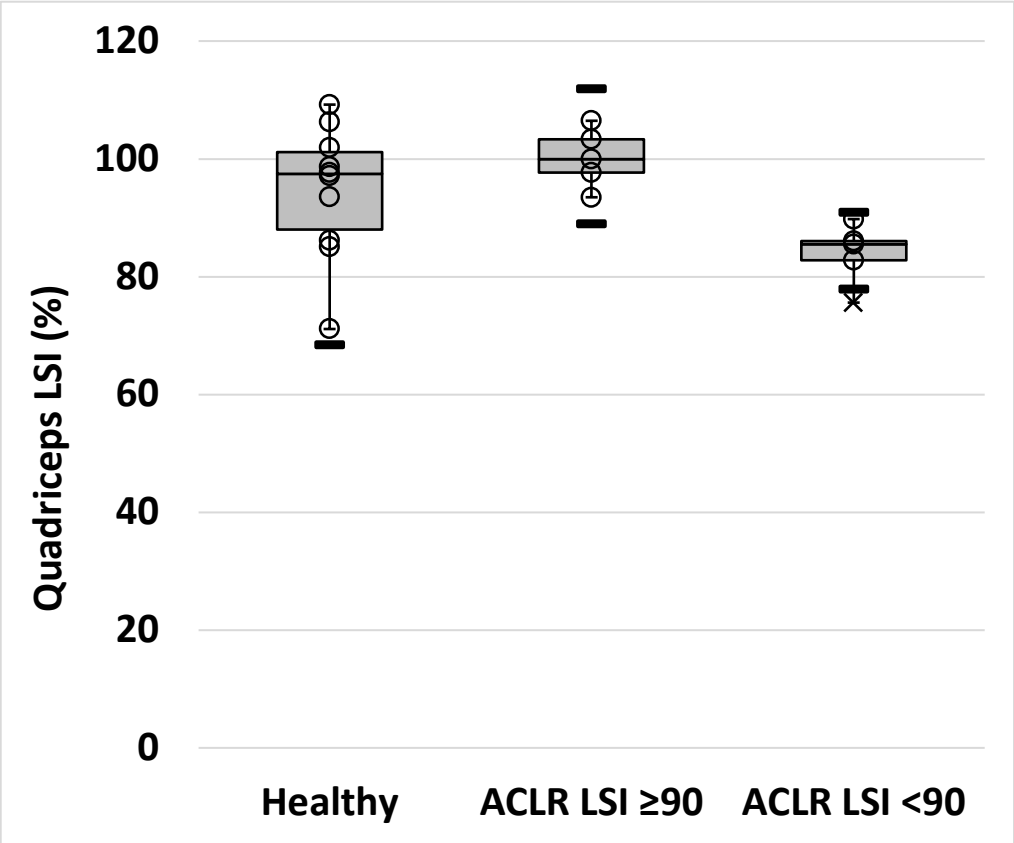

Figure 2. Quadriceps LSI % Box Plot. The quadriceps limb symmetry index of the three groups. The thick horizontal dashed lines represent the threshold detection for the upper and lower outlier detection. The whisker represent the minimum and maximum values for the data points. The horizontal line in the middle of the box represent the median value. The top and bottom of the box represent the 1st and 3rd quartiles, respectively.

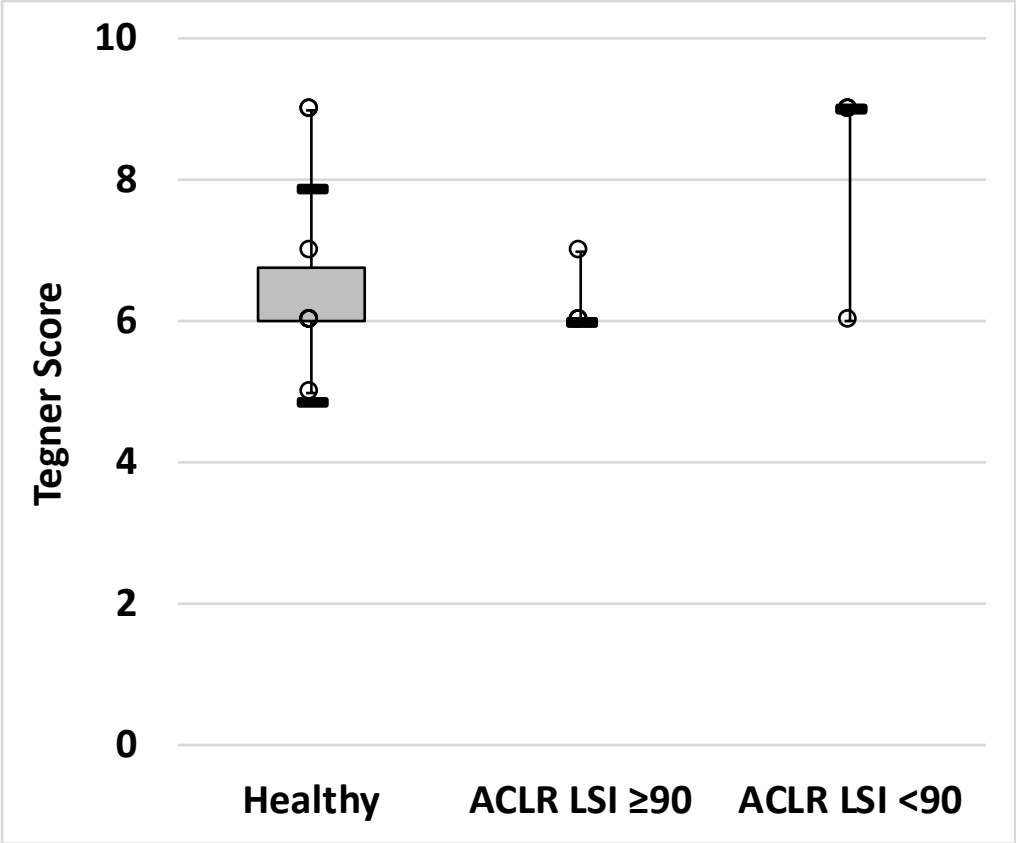

Figure 3. Tegner Score Box Plots. The Tegner Scores of the three groups. The thick horizontal dashed lines represent the threshold detection for the upper and lower outlier detection. The whisker represent the minimum and maximum values for the data points. The horizontal line in the middle of the box represent the median value. The top and bottom of the box represent the 1st and 3rd quartiles, respectively.

	Healthy n=10	ACLR LSI≥90% (n=5)	ACLR LSI <90% (n=5)	p-value
Symptoms %	97.1±3.9*	75.7±22.6*	86.4±7.7	0.014
Pain %	99.0±1.6*	82.2±20.8*	90±7.2	0.032
ADL %	99.9±0.5*	91.8±10.4*	97.9±2.9	0.038
Sport %	100±0*	69.0±26.6*	91±8.9	0.003
QOL %	97.5±3.7*	70±29.8*	77.5±14.4	0.015

Table 2. Patient Reported Function Scores. ACLR LSI ≥90% represents the ACLR group with a quadriceps LSI ≥90%. ACLR LSI <90% represents the ACLR group with a quadriceps LSI of <90%. ADL = Activities of Daily Living, QOL= Quality of life. Mean±SD % *p<0.05

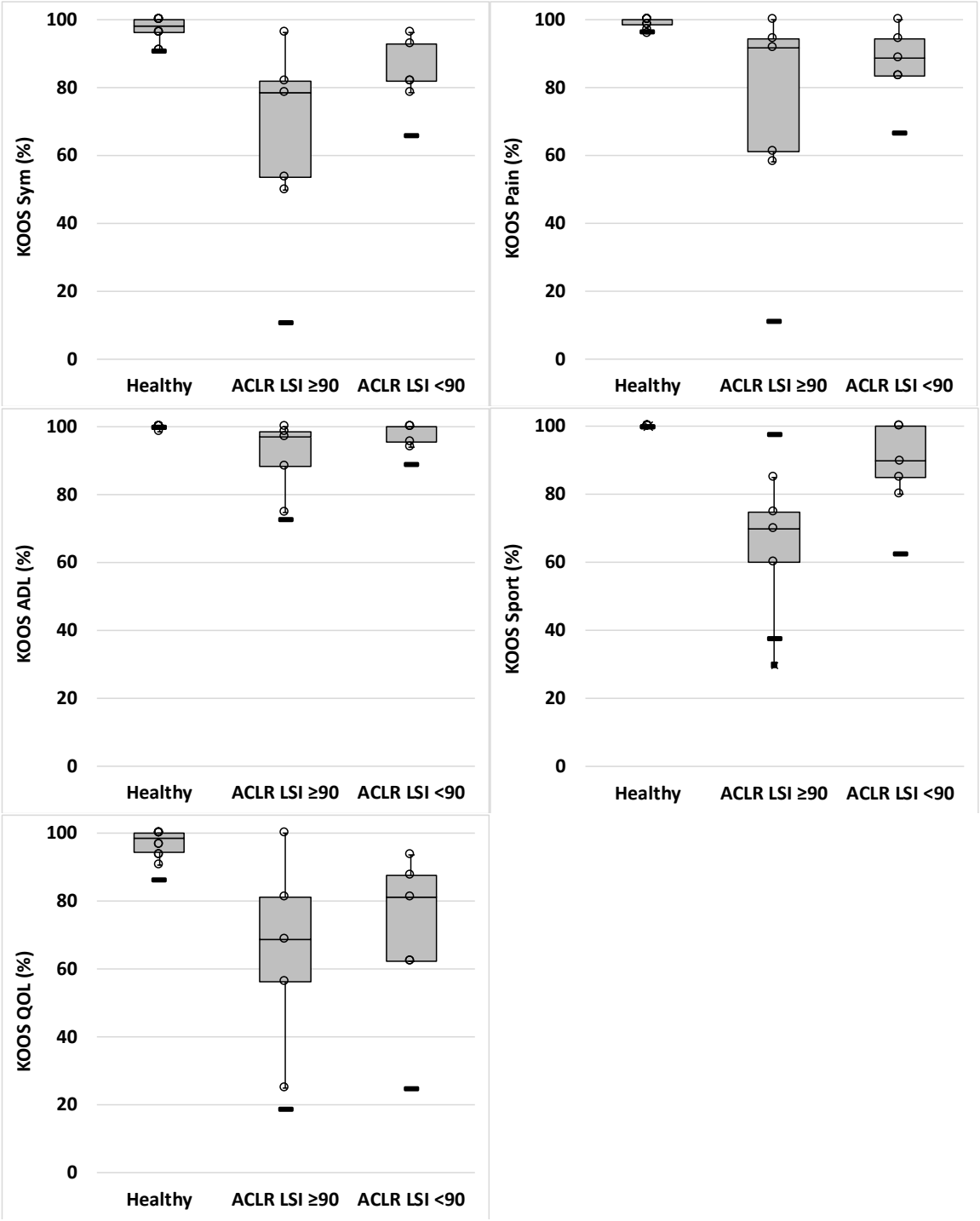

Figure 4. KOOS Box Plots. Sym= Symptom, ADL = Activities of Daily Living, QOL= Quality of Life. The thick horizontal dashed lines represent the threshold detection for the upper and lower outlier detection. The whisker represent the minimum and maximum values for the data

points. The horizontal line in the middle of the box represent the median value. The top and bottom of the box represent the 1st and 3rd quartiles, respectively.

	Healthy	ACLR LSI≥90% (n=5)	ACLR LSI<90% (n=5)	p-value
Single-Leg Hop for Distance LSI%	102.±11.5	100.4±9.9	96.5±8.5	0.626
Triple-Leg Hop for Distance LSI%	98.1±9.7	99.8±9.8	93.4±4.7	0.496
6-Meter Timed LSI %	96.5±8.1	98.5±11.0	101.5±4.8	0.556

Table 3. Single Leg Hop Test Limb Symmetry Index (LSI). ACLR LSI ≥90% represents the ACLR group with a quadriceps LSI ≥90%. ACLR LSI <90% represents the ACLR group with a quadriceps LSI of <90%. Mean±SD %

Effect of Landing Height and Quadriceps LSI on Knee Biomechanics

3x2x3 mixed model ANOVAs were used to assess the effects of landing height, quadriceps LSI, and limb on knee joint biomechanical variables. Within factors were: landing height (30cm,45cm,and 60cm) and limb (injured, uninjured). The three groups (between subjects factor) were healthy controls, ACLR LSI≥90%, and ACLR LSI<90%. These analyses allowed for the detection of main effects for height/task, limb, and group as well as any potential interactions among the factors. Each dependent variable was analyzed at three time points of the drop jumping task associated with the landing phase: initial contact with the force plate, at peak

ground reaction force, and when the center of gravity (COG) vertical minimum was reached indicating the end of the landing phase of the drop jump. Due to having only 5 subjects in each ACLR group, the observed power for all significant results are reported along with the p-values.

Frontal Plane Knee Angles and Torques

For frontal plane knee angles at any timepoint, there were no significant main effects of height or limb or any significant interactions (Table 4). There was a significant group effect ($p=0.031$, observed power: 0.67) for frontal plane knee angle when the center of gravity was at its minimum position. Individuals in the ACLR LSI<90% had significantly more knee valgus, regardless of limb or landing height, when the COG was at the vertical minimum position compared to the ACLR LSI≥90%.

At initial contact, there was a landing height*limb*group interaction ($p=0.018$, observed power=0.81) for frontal plane knee torques. Tukey's post hoc testing revealed significant bilateral differences between the injured vs uninjured limbs of the ACLR LSI<90% but at the 30cm and 60cm heights and not at the 45cm height. Further, the uninjured limb of the ACLR LSI<90% group had significantly higher knee adduction torques upon initial contact with the forceplate compared to the healthy matched control limbs at 30cm and 60cm heights, and ACLR LSI≥90% uninjured limbs at all three landing heights.

At the time of peak ground reaction force, there were significant height and groups main effects for frontal plane knee torques (Table 5). The significant main effect for height revealed differences ($p=0.023$, observed power =0.70) in frontal plane knee torques between the 30cm and 60cm heights (30cm= 2.1±1.02%BW*ht, 45cm= 2.3±1.10%BW*ht, 60cm= 2.6±1.42%BW*ht). Individuals displayed higher knee adduction torques at 60cm compared to 30cm, but not at 45cm. The group main effect shows ACLR LSI ≥90% displayed a significantly

lower knee adduction torque (p=0.017, observed power=0.757) compared to the healthy controls and the ACLR LSI <90% (ACLR LSI≥90=1.16±1.05%BW*ht, ACLR LSI<90%=2.87±1.05%BW*ht, healthy= 2.93±1.04%BW*ht).

At COG vertical minimum, there were significant main effects for landing height and between limbs. The significant main effect for height revealed differences (p=0.001, observed power =0.99) between the 30cm and 60cm heights (30cm= -0.29±1.53%BW*ht, 45cm= -0.79±1.61%BW*ht, 60cm= -1.08±1.28%BW*ht). Thus, individuals displayed significantly higher knee abduction torques at 60cm compared to 30cm, but not at 45cm. A limb main effect (p=0.033, observed power=0.590) showed the uninjured limb displayed higher knee abduction torque at the COG minimum compared to the injured limb (injured=-0.26±1.38, uninjured=-1.18±1.92).

Frontal Knee Angle		Healthy		ACLR LSI ≥ 90%*		ACLR LSI < 90%*	
Timepoint	Drop Jump Height	Injured Limb Matched	Uninjured Limb Matched	Injured Limb	Uninjured Limb	Injured Limb	Uninjured Limb
Initial Contact	30cm	2.2 ± 1.3	1.0 ± 2.6	2.9 ± 2.3	2.8 ± 0.8	1.5 ± 2.5	0.7 ± 1.5
	45cm	2.2 ± 1.4	1.4 ± 2.7	2.6 ± 1.8	3.4 ± 1.1	1.5 ± 2.6	0.6 ± 1.8
	60cm	2.5 ± 2.2	1.9 ± 2.8	2.7 ± 2.8	3.4 ± 1.1	1.0 ± 3.4	0.0 ± 2.2
Pk Ground Reaction Force	30cm	1.8 ± 3.0	-0.0 ± 4.3	3.3 ± 4.6	0.4 ± 3.0	-0.2 ± 5.3	-3.5 ± 4.2
	45cm	1.2 ± 3.5	-0.5 ± 4.0	0.9 ± 0.7	0.9 ± 0.7	-1.2 ± 6.5	-3.7 ± 3.3
	60cm	1.4 ± 3.5	-0.5 ± 4.0	2.5 ± 3.9	1.3 ± 0.8	1.4 ± 7.9	-3.6 ± 3.1
COGy Minimum	30cm	-1.3 ± 6.2	-6.2 ± 6.9	3.6 ± 5.1	0.1 ± 5.7	-3.4 ± 6.4	-9.5 ± 4.0
	45cm	-1.7 ± 6.1	-6.2 ± 8.6	-1.7 ± 6.1	1.2 ± 4.8	-5.6 ± 7.4	-9.3 ± 4.5
	60cm	0.0 ± 6.3	-5.3 ± 8.8	0.0 ± 6.3	2.1 ± 4.8	-5.2 ± 9.0	-8.5 ± 6.6

Table 4. Frontal Plane Knee Angles. ACLR LSI ≥90% represents the ACLR group with a quadriceps LSI ≥90%. ACLR LSI <90% represents the ACLR group with a quadriceps LSI of <90%. COGy Minimum= Center of Gravity Vertical Minimum, Pk Ground Reaction Force= Peak Ground Reaction Force, *Group main effect for Center of Gravity Vertical Minimum. p=0.05

Frontal Knee Torque		Healthy*		ACLR LSI ≥ 90%*		ACLR LSI < 90%	
Timepoint	Drop Jump Height	Injured Limb Matched☆	Uninjured Limb Matched☆	Injured Limb☆	Uninjured Limb☆	Injured Limb☆	Uninjured Limb☆
Initial Contact	30cm	0.27 ± 0.32	0.64 ± 0.60	0.35 ± 0.14	0.32 ± 0.29	0.70 ± 1.24**	1.51 ± 0.67**
	45cm	0.39 ± 0.31	0.56 ± 0.43	0.18 ± 0.25	0.25 ± 0.35	0.56 ± 1.08	1.11 ± 0.73
	60cm	0.58 ± 0.39	0.45 ± 0.39	0.15 ± 0.39	0.29 ± 0.17	0.22 ± 0.56**	1.43 ± 1.04**
Pk Ground Reaction Force	30 cm***	1.83 ± 1.65	2.94 ± 1.74	1.39 ± 1.41	0.85 ± 0.98	1.80 ± 1.25	3.62 ± 2.04
	45cm	2.03 ± 1.18	3.81 ± 2.15	1.22 ± 1.78	1.10 ± 0.88	1.84 ± 1.02	3.93 ± 2.63
	60cm***	2.90 ± 1.82	4.09 ± 2.30	1.34 ± 1.98	1.04 ± 0.71	2.22 ± 1.7	4.30 ± 3.29
COGy Minimum	30cm***	-0.06 ± 1.72	-0.70 ± 1.85	-0.90 ± 0.97	-1.88 ± 1.99	1.22 ± 0.50	0.58 ± 1.95
	45cm	-0.67 ± 1.72	-0.95 ± 1.95	-0.88 ± 0.65	-2.72 ± 2.62	0.60 ± 1.04	-0.15 ± 1.88
	60cm***	-0.63 ± 1.66	-1.01 ± 1.88	-1.57 ± 1.54	-2.69 ± 1.85	0.51 ± 0.96	-1.11 ± 0.85

Table 5. Frontal Plane Knee Torques. ACLR LSI ≥90% represents the ACLR group with a quadriceps LSI ≥90%. ACLR LSI <90% represents the ACLR group with a quadriceps LSI of <90%. COGy Minimum= Center of Gravity Vertical Minimum, Pk Ground Reaction Force= Peak Ground Reaction Force, *Group main effect during Initial Contact and Peak Ground Reaction Force p=0.05, **Group*Limb*Landing Height interaction p=0.05, ***Height Main effect p=0.05, ☆ Limb main effect during Center of Gravity Vertical Minimum p=0.05

Sagittal Plane Knee Angles and Torques

For sagittal plane knee angles, there were no significant main effects of limb or group or any significant interactions at any timepoint. There was a significant main effect of height (p=0.001, observed power: 0.967) for sagittal plane knee angle at the time of peak ground reaction force. Regardless of group or limb, individuals displayed an increased knee flexion angle at 30cm compared to 45cm and 60cm (30cm=35.6±11.1°, 45cm= 32.2±8.1°, 60cm=30.1±6.2°).

For sagittal knee torques, there were no significant main effects of limb or group. There was a significant main effect of height (p=0.001, observed power: 0.99) for sagittal plane knee torques at initial contact. Knee flexion torques at initial contact decreased from 30cm to 45cm and again at 60cm (30cm= -1.1±0.55%BW*ht, 45cm=-0.7±0.43%BW*ht, 60cm= -0.3±0.39 %BW*ht). Similarly, there was a significant main effect of height (p=0.001, observed power=

0.987) for sagittal plane knee torque at peak ground reaction force. Knee extensor torques increased from 30cm to 45cm and again at 60cm (30cm=3.7±2.90%BW*ht, 45cm=4.4±2.39%BW*ht, 60cm=5.2±2.22%BW*ht) At the time of the drop jump when the center of gravity was at its lowest minimum vertical position, there was a three way interaction (group*limb*landing height) for knee extensor torques (p=0.038, observed value=0.72). While there were no changes in knee extensor torque in the injured limbs for any group, the ACLR LSI <90% group displayed significantly lower knee extensor torques in the uninjured limbs at the 60cm landing height (5.56±2.88%BW*ht) compared to both 30cm (6.75±2.53%BW*ht) and 45cm (6.81±2.71%BW*ht). In addition, the ACLR LSI≥90% group displayed significantly higher knee extensor torques for the uninjured limb (6.89±2.88%BW*ht), but only at 60cm, compared to the ACLR LSI<90% (5.56±2.88 %BW*ht) and healthy controls (5.84±2.91%BW*ht).

Knee moment impulse and energy absorption

For knee moment impulse during the landing phase of the drop jump task, no significant main effects of limb or group or any interactions at any timepoint were found. There was a significant main effect of height (p=0.001, observed power: 0.99) for knee moment impulse. Moment impulses increased as the height increased (30cm=1.46±0.46%BW*ht, 45=1.59±0.45%BW*ht, 60cm= 1.7±0.55%BW*ht). In addition, there was a significant main effect of height (p=0.001, observed power: 0.99) for energy absorption. The energy absorption increased as the height increased (30cm=-5.38±2.02%BW*ht, 45cm=-6.88± 2.21%BW*ht, 60cm= -8.05±2.69%BW*ht).

Effect of Single Leg Hops on Knee Biomechanics

2x3 mixed model ANOVAs were used to assess the effects of single leg landings and takeoffs on knee joint biomechanical variables. Within factors were: the two limbs (injured, uninjured) and the between subjects factor was group (healthy controls, ACLR LSI≥90%, and ACLR LSI <90%). These analyses allowed for the detection of main effects for limb and group as well as any potential interactions among the factors. Each dependent variable was analyzed at three time points of the single leg landing task and one time point for the single leg takeoff task. The three time points for the landing task were: initial contact with the force plate, at peak ground reaction force, and at peak knee flexion. The time point for the single leg takeoff task is peak knee flexion.

Single Leg Landing Knee Angles and Torques

All subjects performed single leg hop at 75% of their maximum hop distance. This was performed with each limb independently. For knee flexion angles, there were no significant findings at initial contact (p=0.892, observed power= 0.07) or at peak GRF (p=0.660, observed power=0.11). At peak knee flexion, there was a landing limb*group interaction (p=0.007, observed power=0.86) for knee flexion angles. Individuals in the ACLR LSI <90% displayed less knee flexion angles in their injured limb compared to their uninjured limb (ACLR LSI <90% injured= 40.9±3.7°, ACLR LSI <90% uninjured=45.7±5.5°) whereas the healthy and ACLR LSI ≥90% did not differ between limbs.

During single leg landings there was a significant limb*group interaction for knee moment impulse (p=0.015, observed power=0.775). There was a significant bilateral difference between injured and uninjured limbs in the ACLR LSI <90% group whereas the healthy and ACLR LSI ≥90% did not differ between limbs. The injured limb of the ACLR LSI <90% group

was significantly lower than the contralateral uninjured limb and healthy injured-matched limbs (ACLR LSI <90% injured= 1.23±0.26%BW*ht, ACLR LSI <90% uninjured =1.69±0.53%BW*ht, healthy injured matched= 1.65±0.64%BW*ht).

During single leg landings a significant limb * group interaction (p=0.005, observed power= 0.90) for knee energy absorption. There was a significant bilateral difference between injured and uninjured limbs in the ACLR LSI <90% group whereas the healthy and ACLR LSI≥90% did not differ between limbs (Figure 5). The injured limb of the ACLR LSI <90% group exhibited less knee energy absorption than the contralateral uninjured limb (ACLR LSI <90% injured= -3.99±1.37%BW*ht, ACLR LSI% <90 uninjured= -6.17±1.55%BW*ht).

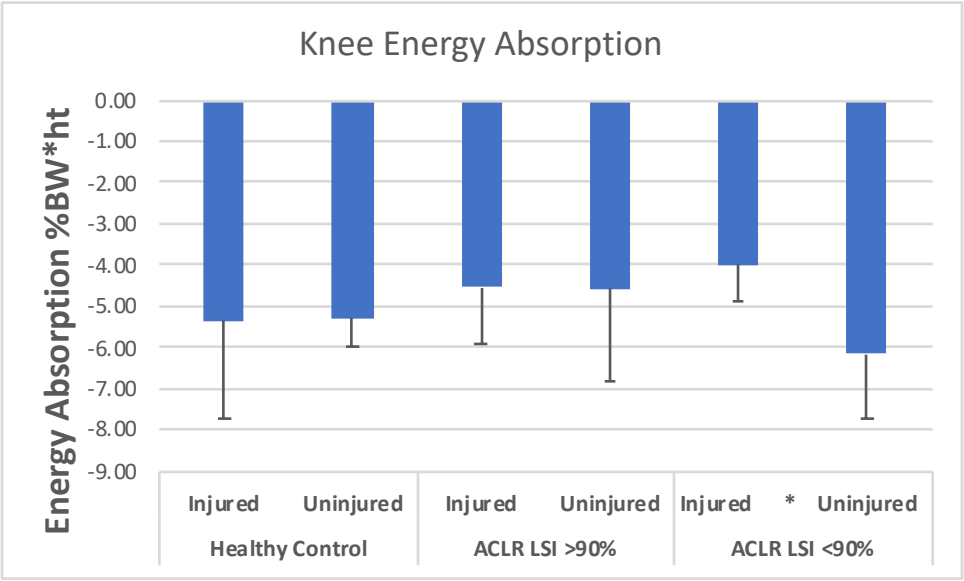

Figure 5. Knee Energy Absorption During Single Leg Landings. ACLR LSI ≥90% represents the ACLR group with a quadriceps LSI ≥90%. ACLR LSI <90% represents the ACLR group with a quadriceps LSI of <90%. For the healthy controls, "Injured" and "Uninjured" represent the injured and uninjured matched limbs. * Represents limb*group interaction in which the injured limb of the ACLR LSI <90% group exhibited less knee energy absorption than the contralateral uninjured limb. For the healthy controls, "Injured" and "Uninjured" represent the injured and uninjured matched limbs.

Hip and Ankle Energy Absorption During Landing

Hip energy absorption revealed no significant main effects for limb or limb*group interaction (p=0.156,observed power=0.37, Figure 6). Similarly, ankle energy absorption revealed no main effects for limb or limb*group interaction (p=0.258, observed power=0.27, Figure 7).

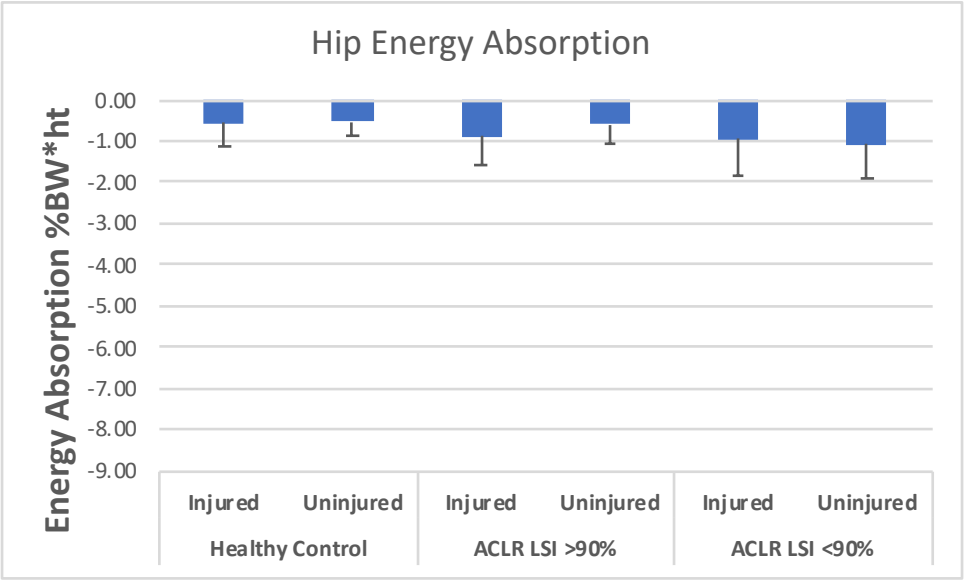

Figure 6. Hip Energy Absorption During Single Leg Landings. ACLR LSI ≥90% represents the ACLR group with a quadriceps LSI ≥90%. ACLR LSI <90% represents the ACLR group with a quadriceps LSI of <90%. For the healthy controls, "Injured" and "Uninjured" represent the injured and uninjured matched limbs.

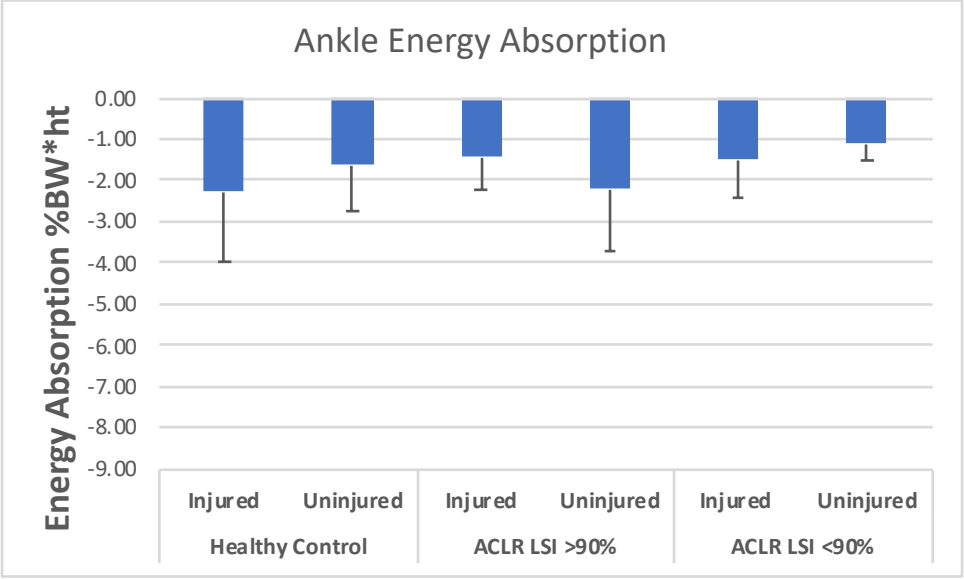

Figure 7. Ankle Energy Absorption During Single Leg Landings. ACLR LSI ≥90% represents the ACLR group with a quadriceps LSI ≥90%. ACLR LSI <90% represents the ACLR group with a quadriceps LSI of <90%. For the healthy controls, "Injured" and "Uninjured" represent the injured and uninjured matched limbs.

Total limb energy absorption, calculated as the sum total energy absorption from the hip extensors, knee extensors, and ankle plantarflexors, revealed a limb * group interaction (p=0.007, observed power=0.862, Figure 8) during single leg landings. There was a significant bilateral difference between injured and uninjured limbs in the ACLR LSI <90% group whereas the healthy and ACLR LSI ≥90% did not differ between limbs (Figure 8). The injured limb of the ACLR LSI <90% group exhibited less total energy absorption than the contralateral uninjured limb and the healthy injured matched control limb (ACLR LSI <90 injured= -6.46±1.11%BW*ht, ACLR LSI <90 uninjured= -8.36±1.72%BW*ht, healthy injured matched=-8.20±2.61%BW*ht)

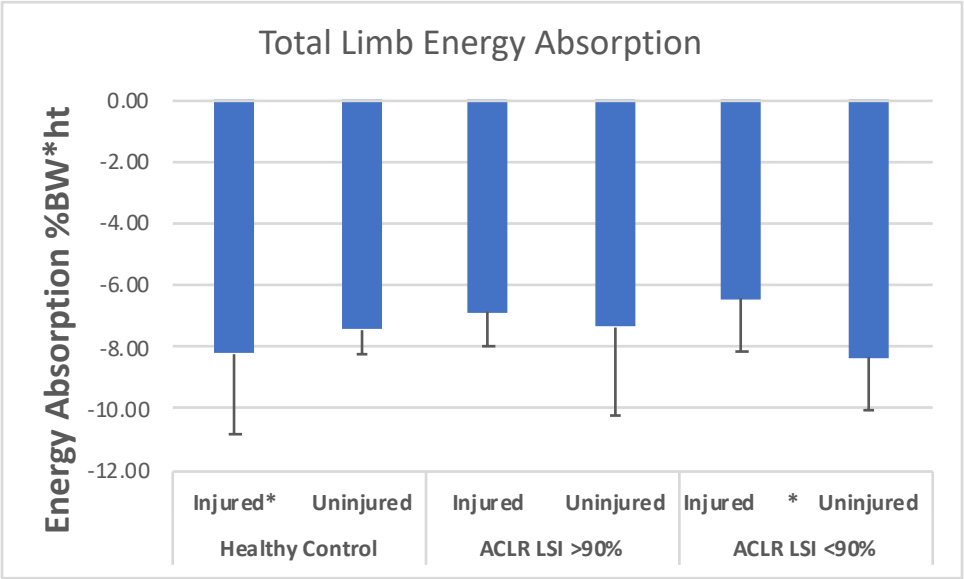

Figure 8. Limb Energy Absorption During Single Leg Landings. ACLR LSI ≥90% represents the ACLR group with a quadriceps LSI ≥90%. ACLR LSI <90% represents the ACLR group with a quadriceps LSI of <90%. For the healthy controls, "Injured" and "Uninjured" represent the injured and uninjured matched limbs. * Represents limb*group interaction in which the

injured limb of the ACLR LSI <90% group exhibited less total energy absorption than the contralateral uninjured limb and the healthy injured matched control limb. For the healthy controls, “Injured” and “Uninjured” represent the injured and uninjured matched limbs.

Single Leg Takeoff Knee Angles, Torques, and Energy Generation

At peak knee flexion, there was a limb*group interaction (p=0.007, observed power=0.86) for knee angles. The ACLR LSI <90% group displayed less knee flexion angles at peak knee flexion (the beginning of the self-initiated single leg hopping task) on their injured limb compared to their uninjured limb (ACLR LSI <90 injured=40.9±4.4°, ACLR LSI <90 uninjured= 45.6±4.3°). There were no bilateral differences for this variable in the healthy control group or the ACLR LSI ≥90% group.

Again at peak knee flexion, there was a limb*group interaction (p=0.01, observed power= 0.83)for knee extensor torques. The ACLR LSI <90% group displayed significant bilateral differences between injured and uninjured limbs. The injured limb of the ACLR LSI <90% group showed significantly lower knee extensor torques at peak knee flexion compared to the ACLR LSI <90% uninjured limb and to the healthy injured matched limb (ACLR LSI <90 injured=4.03±1.61%BW*ht, ACLR LSI <90 uninjured= 4.47±1.77%BW*ht, healthy injured matched=7.33±1.61%BW*ht).

Knee energy generation revealed a limb*group interaction (p=0.041, observed power=0.62, Figure 9). The ACLR LSI <90% group displayed significant bilateral differences between injured and uninjured limbs. The injured limb of the ACLR LSI <90% group showed significantly lower knee energy generation compared to the ACLR LSI <90% uninjured limb and to the healthy injured matched limb (ACLR LSI <90 injured=0.11±0.72%BW*ht, ACLR LSI <90 uninjured= 0.96±0.78%BW*ht, healthy injured matched=1.84±0.72%BW*ht, Figure 9).

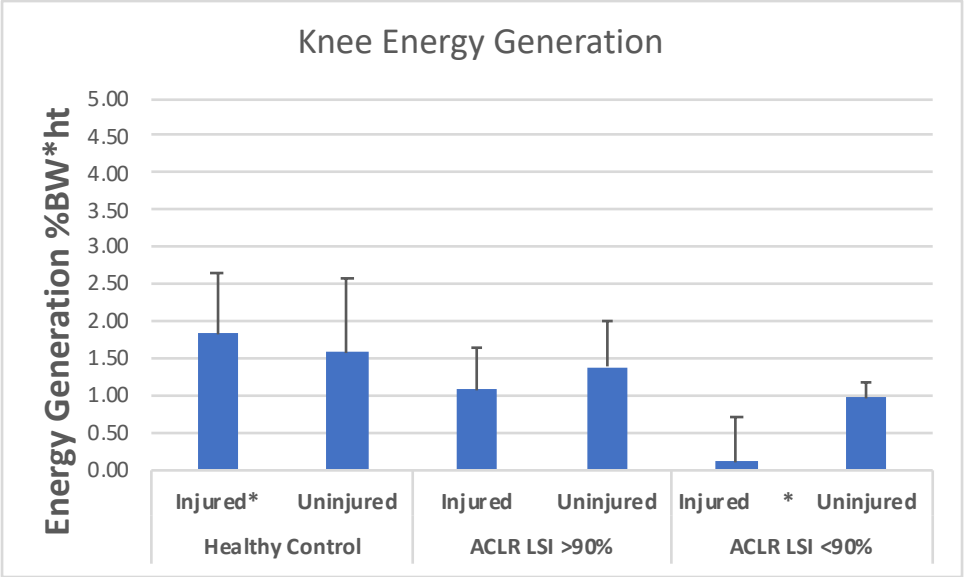

Figure 9. Knee Energy Generation During Single Leg Takeoffs. ACLR LSI ≥90% represents the ACLR group with a quadriceps LSI ≥90%. ACLR LSI <90% represents the ACLR group with a quadriceps LSI of <90%. For the healthy controls, "Injured" and "Uninjured" represent the injured and uninjured matched limbs. * Represents limb*group interaction in which the injured limb of the ACLR LSI <90% group exhibited less total energy absorption than the contralateral uninjured limb and the healthy injured matched control limb. For the healthy controls, "Injured" and "Uninjured" represent the injured and uninjured matched limbs.

Hip and Ankle Energy Generation During Takeoff

Hip energy absorption revealed no significant main effects of interactions (p=0.065,observed power=0.54, Figure 10). Similarly, ankle energy generation revealed no significant main effects or limb*group interaction (p=0.864, observed power=0.07, Figure 11).

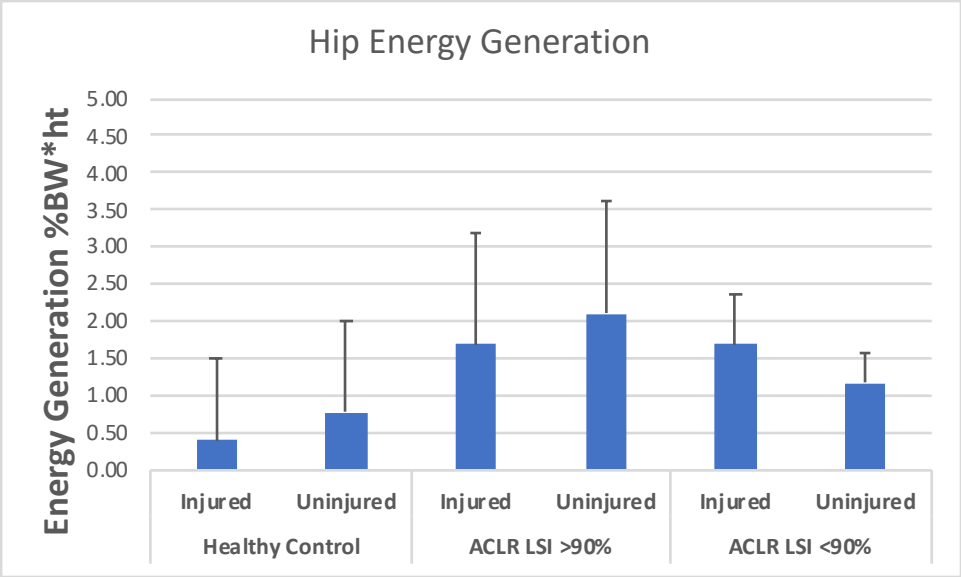

Figure 10. Hip Energy Generation During Single Leg Takeoffs. ACLR LSI ≥90% represents the ACLR group with a quadriceps LSI ≥90%. ACLR LSI <90% represents the ACLR group with a quadriceps LSI of <90%. For the healthy controls, "Injured" and "Uninjured" represent the injured and uninjured matched limbs. For the healthy controls, "Injured" and "Uninjured" represent the injured and uninjured matched limbs.

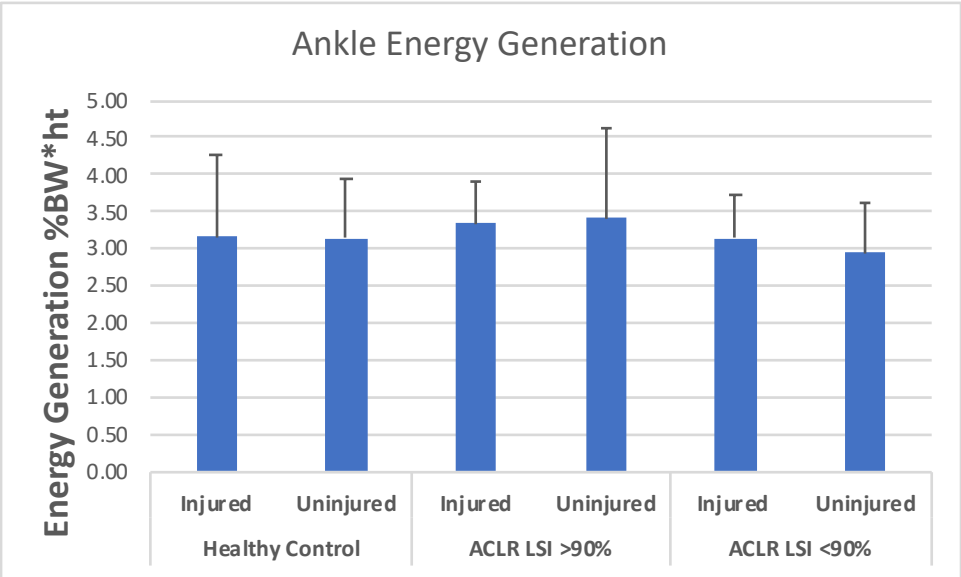

Figure 11. Ankle Energy Generation During Single Leg Takeoffs. ACLR LSI ≥90% represents the ACLR group with a quadriceps LSI ≥90%. ACLR LSI <90% represents the ACLR group with a quadriceps LSI of <90%. For the healthy controls, "Injured" and "Uninjured" represent the injured and uninjured matched limbs. For the healthy controls, "Injured" and "Uninjured" represent the injured and uninjured matched limbs.

Total limb Energy Generation During Takeoff

Total limb energy generation, calculated as the sum total energy generation from the hip extensors, knee extensors, and ankle plantarflexors, did not reveal any significant main effects for limb, group, or a limb * group interaction (p=0.348, observed power=0.215, Figure 12) during single leg takeoffs.

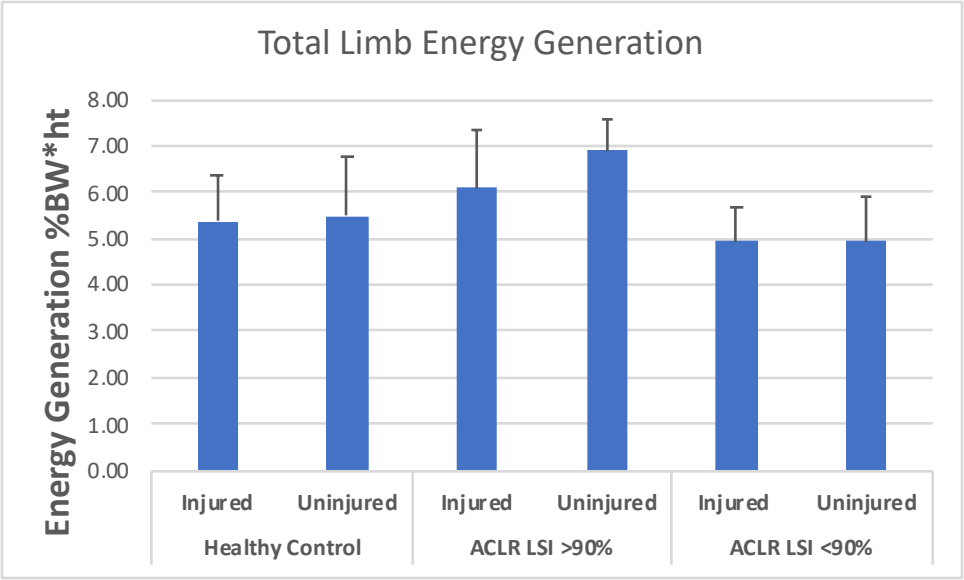

Figure 12. Limb Energy Generation During Single Leg Takeoffs. ACLR LSI ≥90% represents the ACLR group with a quadriceps LSI ≥90%. ACLR LSI <90% represents the ACLR group with a quadriceps LSI of <90%. For the healthy controls, "Injured" and "Uninjured" represent the injured and uninjured matched limbs. For the healthy controls, "Injured" and "Uninjured" represent the injured and uninjured matched limbs.

Chapter V: Discussion

The first hypothesis examined in this thesis explored whether ACLR LSI <90% individuals would exhibit biomechanical asymmetries regardless of task demands i.e. changes in landing height. While there were significant bilateral differences between frontal knee torques of the injured vs uninjured limbs of the ACLR LSI<90% during initial contact, the significant bilateral difference occurred for the 30cm and 60cm conditions but not at 45cm. Thus, the ACLR LSI <90% group exhibited biomechanical asymmetries indicative of heightened ACL injury risk but these findings were not consistent across landing heights and the uninjured limb displayed higher knee adduction torques at initial contact compared to the injured limbs. Further, when center of mass was at its lowest point, the ACLR LSI<90% group displayed lower knee extensor torques for the uninjured limb but only at 60cm when compared to 45cm and 30cm landing heights. Overall, the bilateral differences for the ACLR LSI<90% appear to be driven by higher frontal plane knee torques but lower knee extensor torques in the uninjured limb compared to the injured limb. The second hypothesis stated that the ACLR LSI ≥90% group would exhibit biomechanical asymmetries when task demands are highest (60cm) but not when task demands are at their lowest (30cm). The ACLR LSI ≥90% did not show any significant differences in the dependent variables examined in the analysis across any landing height. Lastly, while several main effects for landing height were found showing that as landing height increased, knee moment impulses and knee energy absorption increased, these adaptations to increasing task demands were not unique to one group or limb.

Do ACLR individuals with quadriceps LSI<90% exhibit biomechanical asymmetries regardless of task demands?

While the ACLR LSI <90% group did exhibit biomechanical asymmetries indicative of heightened ACL injury risk, the asymmetries did not occur consistently across task demands. In addition, the uninjured limb of the ACLR LSI<90% group showed significantly higher knee adduction torques at initial contact with the forceplate compared to the healthy matched control limbs, and ACLR LSI≥90% uninjured limbs at all three landing heights (p=0.018, observed power=0.81). These results suggest that the ACLR LSI <90% group displays heightened risk for a second ACL injury regardless of task demands. However, thigh muscle strength and functional deficits may occur from inhibition of motor activation.[18] Since we did not collect data on these individuals before their ACL injury, it is impossible to know whether motor activation inhibition occurred after their ACL reconstruction, resulting in biomechanical asymmetries. A potential sign of an inhibition of motor activation was suggested when Gokeler et al concluded that ACLR individuals jump 16-19% shorter on both the injured and uninjured limb to achieve symmetry compared to healthy individuals.[18] With the current data, when adjusted for subject height, the ACLR LSI <90% group jumped a shorter distance on both limbs (73.26±0.09%ht), compared to the healthy control group (79.85±0.14%ht) and the ACLR LSI ≥90% group (81.65±0.11%ht). However, a 2x3 mixed model ANOVA revealed no significant differences (p-value=0.652, observed power=0.11) between distances of the three groups. The effect size observed comparing the healthy group to the ACLR LSI<90% group was moderate (0.54) and the effect size comparing ACLR LSI <90% to ACLR LSI ≥90% was large (0.82). While Gokeler et al found that ACLR individuals jump a shorter distance to achieve symmetry, our findings showed that only ACLR LSI<90% individuals jump a moderately shorter distance regardless of limb

compared to healthy individuals and the ACLR LSI ≥90% group. Future studies investigating whether or not individuals with quadriceps LSI <90% also possess quadriceps inhibition that explains a relatively shorter hop distance are warranted.

While single leg hops were not a part of the original hypothesis and purpose, the data is still important to compare with literature as well as to compare with the double leg drop jump results. Single leg hops allowed us to evaluate asymmetries while one limb is loaded independently of the other. Orishimo et al showed that ACL deficient individuals offload their injured knee by compensating with their ipsilateral hip and ankle during single leg takeoff and landing trials, while simultaneously reducing quadriceps muscle efforts.[23] Similarly, reduced knee energy absorption and generation was observed in the ACLR LSI <90% during single leg landing and takeoff trials. The ACLR LSI <90% group displayed a reduction in quadriceps effort shown by their reduced knee energy generation and absorption during both single leg takeoff and landing trials respectively. As a possible explanation for the reduced quadriceps efforts, this suggests that the ground reaction force vector directed through the knee joint of the injured limb of the ACLR LSI <90% group rather than posterior to it which would engage higher quadriceps efforts (Figure 13). Thus, the ACLR LSI <90% injured limb knee extensors performed less work (less negative powers) compared to their uninjured limb (Figure 14).

When an individual decreases knee extensor torque, they are using less quadriceps and therefore placing increased stress on other structures such as ligaments and articular cartilage suggesting that this landing strategy might heighten risk of developing knee OA in individuals with quadriceps LSI <90%. Tsai et al found that during drop landing tasks, ACLR individuals exhibit greater muscle co-contraction along with less knee flexion which resulted in increased tibiofemoral compressive forces, which may be associated with the high risk of developing knee

osteoarthritis in this population.[41] In addition, Tsai et al also found that training ACLR individuals to increase hip and knee flexion during landing can decrease tibiofemoral compressive forces.[42] Similarly, the ACLR LSI <90% group exhibited significantly less knee flexion on their injured limb compared to their uninjured limb (ACLR LSI <90% injured= 40.9±3.7°, ACLR LSI <90% uninjured=45.7±5.5°). Thus, it is possible that this group exhibited increased tibiofemoral compressive forces, which may be associated with the high risk of developing knee osteoarthritis in this population.[41]

Because the ACLR LSI <90% group displayed signs of compensation to off-load the quadriceps muscle, this also suggests that the quadriceps strength LSI and knee energetics LSI are correlated. When we combined both ACLR groups, we saw a strong correlation for both the single leg landing (r=0.74, p-value=0.014, Figure 15) and the single leg takeoff trials (r=0.64, p-value=0.045, Figure 16). Despite no significant differences in hopping distance LSI during the clinical examination of the hop test battery (Table 3), individuals who have undergone ACLR will off-load the quadriceps efforts during these functional tasks depending on the extent of the quadriceps strength deficit.

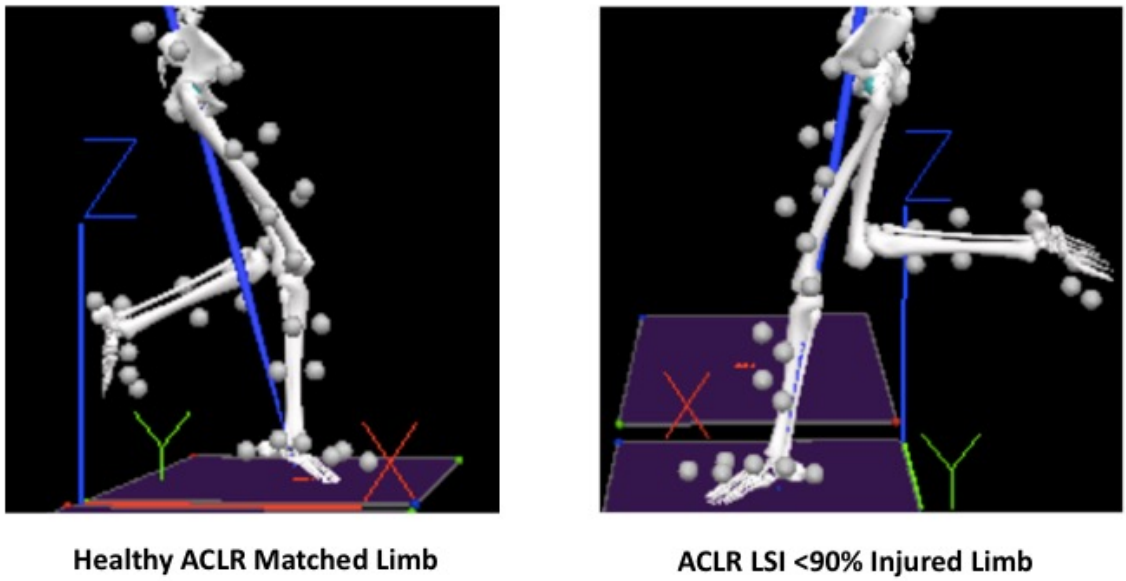

Figure 13. Single Leg Landing Strategies. Visual 3D demonstration of the force vector going through the knee joint in the ACLR <90% group while the force vector on the healthy control participant goes through the center of mass. Each timepoint was taken at initial contact with the forceplate.

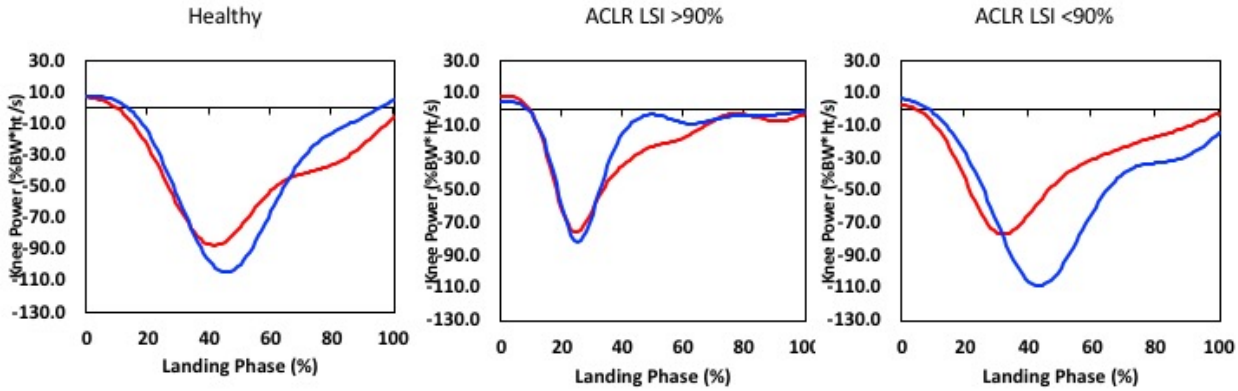

Figure 14. Knee Power During Single Leg Landing. Red line represents the injured or injured matched limb in each group. The blue line represents the uninjured limb in each group. ACLR

LSI ≥90% represents the ACLR group with a quadriceps LSI ≥90%. ACLR LSI <90% represents the ACLR group with a quadriceps LSI of <90%. For the healthy controls, "Injured" and "Uninjured" represent the injured and uninjured matched limbs. For the healthy controls, "Injured" and "Uninjured" represent the injured and uninjured matched limbs. Each group is represented by one participant.

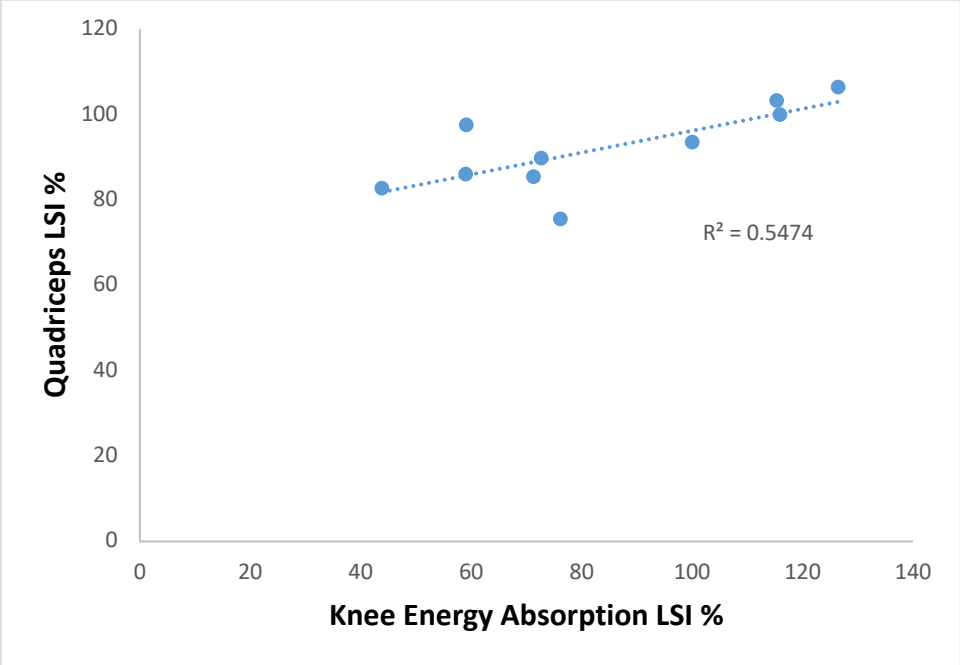

Figure 15. Knee Energy Absorption LSI % and Quadriceps LSI % Correlation During Single Leg Hop Landing

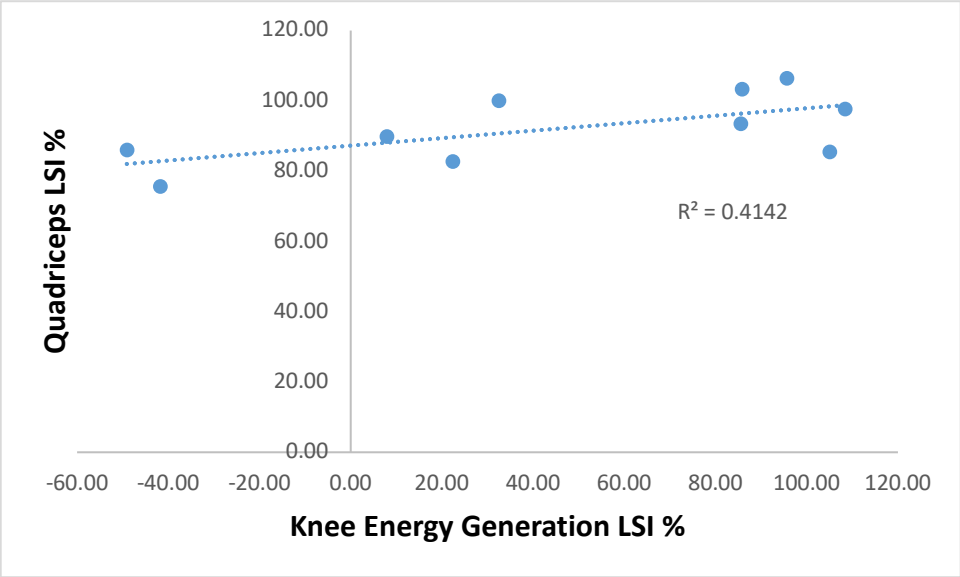

Figure 16. Knee Energy Generation LSI % and Quadriceps LSI % Correlation During Single Leg Hop Takeoff

Collectively, the results of the double-leg landing and single leg hopping tasks suggest that ACLR individuals with quadriceps LSI <90% are at risk for a second ACL injury. Grindem et al found that having a low quadriceps LSI% is a future predictor of second ACL injuries.[12] Out of 69 participants, the 18 that returned to sport and suffered a second ACL injury had an average quadriceps strength LSI of 75.0%.[12] In addition, Schmitt et al found that ACLR individuals who have a quadriceps LSI <85% demonstrate negatively affected performance and function during single leg hopping tasks while ACLR individuals who had an average quadriceps strength LSI ≥90% demonstrate functional equivalency to healthy controls.[33] Based off the results gathered from the ACLR LSI <90% group who had a quadriceps strength LSI of 84%, several biomechanical asymmetries were observed that may heighten their risk for a second ACL injury. During the double leg drop jumps, the ACLR LSI <90% group exhibited frontal plane knee asymmetries that may be indicative of heightened ACL injury risk for the uninjured limb. The uninjured limb of the ACLR LSI<90% group showed significantly higher knee adduction torques at initial contact with the forceplate compared to the healthy matched control limbs at 30cm and 60cm heights, and ACLR LSI≥90% uninjured limbs at all three landing heights. Hewett et al examined using bilateral drop jumps from a standardized height of 31cm to predict future initial ACL injuries.[37] Hewett et all found that knee valgus moments at initial contact were indicators of future ACL injury risk.[37] Thus, the uninjured limb of the ACLR LSI<90% group in the current study exhibited heightened risk for a second ACL injury. This finding is supported in literature by the high second ACL injury percentage in the contralateral and previously uninjured limb. Paterno et al conducted a longitudinal study to examine the incidence rates of ipsilateral and contralateral ACL injuries.[38] Paterno et al discovered 75% of second ACL injuries occurred

in the contralateral, or uninjured limb.[38] Thus, one may conclude that the ACLR LSI <90% group experienced heightened ACL injury risk for their uninjured limb despite having stronger quadriceps muscle strength in the uninjured vs injured limbs.

While the ACLR LSI <90% group primarily exhibited frontal plane asymmetries during the double leg drop jumps, the single leg hops revealed different asymmetries that increased their risk for a second ACL injury but potentially to the reconstructed limb. During single leg landing, the ACLR LSI <90% injured limb displayed lower knee moment impulse, knee energy absorption, and total energy absorption compared to the uninjured limb. Similarly, the single leg take off trials revealed lower knee energy generation of the ACLR LSI <90% injured limb. Thus, the ACLR LSI <90% group exhibited less knee extensor effort during the single leg landing and takeoff trials. A decrease in the quadriceps efforts would increase the load on other knee structures such as the ACL. Thus, the ACLR LSI <90% group exhibited biomechanical asymmetries during two different tasks that would heighten ACL injury. The reconstructed limb could potentially be at heightened risk for a re-tear during the single leg hopping trials, while the uninjured limb appeared to show a heightened risk for injury during the drop jump trials due to the high knee adduction torque.

Do ACLR individuals with quadriceps LSI ≥90% exhibit biomechanical asymmetries when task demands are heightened?

The ACLR LSI≥90% group did not display any asymmetries indicative of ACL injury regardless of height. One possible explanation is that the ACLR LSI ≥90% is functionally equivalent to the control group in regards to their quadriceps symmetry (ACLR LSI ≥90% Quadriceps symmetry= 100.2±5.0, Healthy control group quadriceps symmetry= 94.7±11.3). Since quadriceps asymmetries <10% can be seen in the normal population[21], this could explain

why the ACLR LSI≥90% group displayed no significant asymmetries indicative of heightened ACL injury. In addition, the ACLR LSI≥90% group and the ACLR LSI <90% group did not exhibit significantly lower single leg hop test LSI (Healthy = 102.1±11.5, ACLR LSI ≥90% = 100.4±9.9, ACLR LSI <90%= 96.5±8.5). The ACLR LSI ≥90% also jumped roughly the same distance on both limbs (healthy injured matched limb = 80.72±0.16%ht, healthy uninjured matched= 79.0±0.13%ht, ACLR LSI ≥90% injured=81.55±0.11%ht, ACLR LSI ≥90% uninjured=81.75±0.12%ht). Similar to the findings of Schmitt et al who found that ACLR LSI ≥90% individuals hop equivalent distances on hopping tasks to healthy controls[33], the ACLR LSI ≥90% group in our study was equivalent to the healthy controls in several facets including hop distance which could help explain why this group never displayed any biomechanical asymmetries indicative of heightened ACL injury risk regardless of task demands.

Limitations and Future Directions

The Tegner scale is commonly used to measure the activity level in individuals for research purposes. Across our three groups, the ACLR LSI <90% group was significantly higher than the healthy and the ACLR LSI ≥90% groups (healthy =6.6±1.4, ACLR LSI ≥90%=6.2±1.5, ACLR LSI <90%=8.4±1.4). Thus, the most active group was also the most deficient group based on quadriceps strength symmetry. The ACLR <90% exhibited traits that potentially put them at risk for future second ACL injury to either the ipsilateral (single-leg dominant tasks) or contralateral (double leg dominant tasks) limb. This also highlights that ACLR individuals can be highly functional and compete in sports, yet be at heightened risk to suffer a second ACL injury.

An additional limitation of this study was the small sample size. Although we have a low sample size in each group, we have confidence in our primary results due to the high observed

powers for several interactions including: the frontal plane knee torques during the drop jumps (p=0.018, observed power=0.81), the knee energy absorption differences during the single leg landing (p=0.005, observed power= 0.90), and the total limb energy absorption differences during single leg landings (p=0.007, observed power=0.86). In addition, we saw multiple height main effects which highlight how the participants all experienced increased biomechanical demands to the landings as height increased. Thus, even with smaller sample sizes for the ACL groups, the observed powers give confidence that chances for statistical errors due to low statistical power are minimal. However, it is important to note that non-significant differences with low observed powers does not necessarily mean that the groups are not significantly different in reality. In addition, due to the low sample size, outliers can have a large impact on results which could potentially increase differences between groups or cause there to be no significant differences between the groups. An example of this can be highlighted in the knee flexion angle during single leg landing from a hop at the time of peak ground reaction force, the time at which ACL injury is thought to occur. With this analysis, there were no effects of group (p=0.875) or limb*group interactions (p=0.125, observed power=0.411, Figure 17). However, analysis of the box plots suggest that 1) the dispersion of knee flexion angles withn the healthy group were large and 2) the injured limbs of both ACLR groups appeared to be ~10 degrees of flexion, but an outlier was present in the ACLR LSI <90% group which collectively could have prevented the analysis from detecting a significant limb*group interaction. This is important because low knee flexion angles potentially increase strain on the ACL. Nonetheless, the limited sample size for the ACLR groups (each n=5) coupled with the greater dispersion of knee flexion angles in the healthy group (n=10) prevented such an analysis to produce results with adequate statistical power.

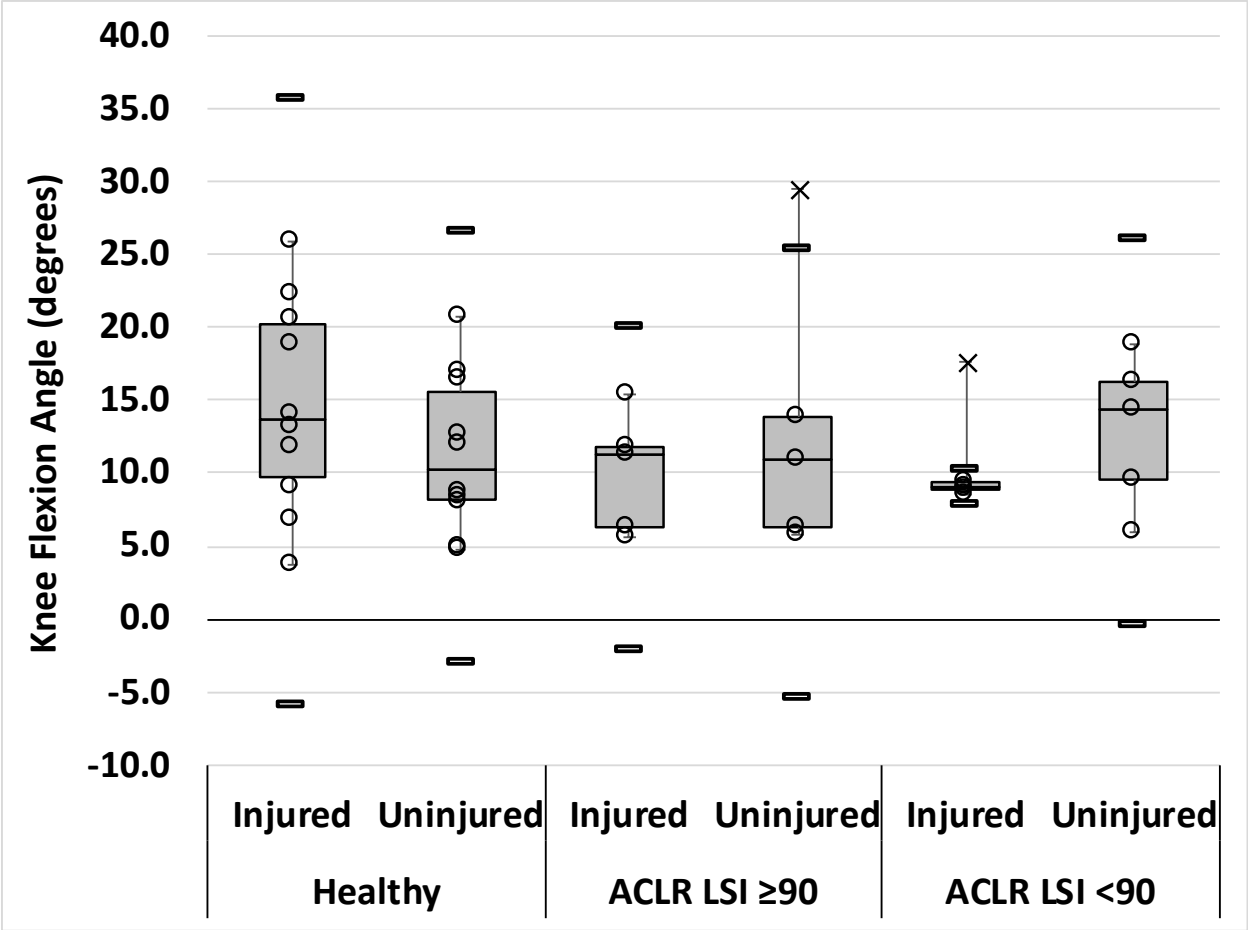

Figure 17: Effect of Group on Knee Flexion Angle at Peak Ground Reaction Force During Landing from a Single Leg Hop. Flexion angles are positive (+), Extension angles are negative (-). X symbols denote outliers from the group.

An additional limitation of this study was the sex makeup across our ACLR groups. The ACLR LSI ≥90% had 3 females and 2 males in their group, while the ACLR LSI <90% group had four females and 1 male in their group. The difference among males to females in each group is important when you consider that females suffer ACL injuries at an alarmingly higher rates than to males.[40] Since Hewett et al found that higher knee abduction torques is associated with increased risk for ACL injury but only studied this in female athletes, the issue of sex-differences poses a confounding factor to the results of this study. Thus, given that the current study was not statistically powered to evaluate the effect of sex on the dependent variables, the

higher female percentage in the ACLR LSI <90% group along with the low sample size in each of the ACLR groups are limitations in this study.

Future directions in this area could examine whether muscular inhibition explains why ACLR individuals suffer from low quadriceps strength LSI in order to explain how the relatively lower quadriceps strength in the reconstructed limb resulted in the biomechanical asymmetries presented. The ACLR LSI <90% group was our most active group according to the Tegner scale, yet suffered from the lowest quadriceps strength LSI. Central activation ratio (CAR) has been shown to provide an outcome measure to quantify gross neuromuscular function of the quadriceps.[39] Future research could examine motor activation of the quadriceps using CAR in ACLR individuals to understand if muscular inhibition plays a role in the increased risk for second ACL injuries.

Conclusions

Quadriceps strength LSI has been shown to be a predictor of second ACL injuries.[12] While the ACLR LSI ≥90% group did not appear to possess biomechanical characteristics indicative of heightened ACL injury risk, the ACLR LSI <90% group appeared to display biomechanical characteristics suggesting they could potentially be at risk for a second ACL injury which includes a future injury to either limb. Despite the ACLR LSI <90% group having the highest Tegner scores, they still exhibited biomechanical asymmetries indicative of heightened ACL second injury risk during single leg hopping trials and double leg drop jumps. ACLR LSI <90% appeared to be at risk for future ACL injury. The biomechanical characteristics presented here that are thought to heighten the risk for ACL injury should be replicated with a larger sample size to substantiate the robustness of the current results.

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Appendix A. Institutional Review Board Approval

EAST CAROLINA UNIVERSITY
University & Medical Center Institutional Review Board
4N-64 Brody Medical Sciences Building· Mail Stop 682
600 Moye Boulevard · Greenville, NC 27834
Office 252-744-2914 · Fax 252-744-2284 · www.ecu.edu/ORIC/irb

Notification of Initial Approval: Expedited

From: Biomedical IRB
To: Anthony Kulas
CC:

Date: 7/3/2019
Re: UMCIRB 17-001075
Thigh Muscle Strength Symmetry and Functional Symmetry

I am pleased to inform you that your Expedited Application was approved. Approval of the study and any consent form(s) occurred on 7/2/2019. The research study is eligible for review under expedited category # 4, 6, 7. The Chairperson (or designee) deemed this study no more than minimal risk.

Changes to this approved research may not be initiated without UMCIRB review except when necessary to eliminate an apparent immediate hazard to the participant. All unanticipated problems involving risks to participants and others must be promptly reported to the UMCIRB. The investigator must submit a Final Report application to the UMCIRB prior to the Expected End Date provided in the IRB application. If the study is not completed by this date, an Amendment will need to be submitted to extend the Expected End Date. The Investigator must adhere to all reporting requirements for this study.

Approved consent documents with the IRB approval date stamped on the document should be used to consent participants (consent documents with the IRB approval date stamp are found under the Documents tab in the study workspace).

The approval includes the following items:

Name	Description
Informed Consent-revised	Consent Forms
KOOS Survery	Surveys and Questionnaires
Participant Measurements	Recruitment Documents/Scripts
protocol	Study Protocol or Grant Application
Recruitment Announcement	Recruitment Documents/Scripts
Screening Questionnaire	Recruitment Documents/Scripts
Tegner Activity Scale	Surveys and Questionnaires

The Chairperson (or designee) does not have a potential for conflict of interest on this study.

Appendix B. Pilot Data

Pilot Test Data

Single Leg Hop LSI	Triple Leg Hop LSI	6m Timed LSI	Isokinetic quadriceps LSI at 60°/s	Isometric quadriceps LSI at 60°
96.77%	97.07%	97.76%	88.50%	101.60%

Table 6. Pilot 1's limb symmetry indices across dynamometer and hop testing batteries.

Pilot 1 displayed ≥90% LSI in four out of the five tests (Figure 2). Pilot 1 produced a quadriceps LSI below 90% during the isokinetic dynamometer test, but produced a quadriceps LSI greater than 100% during the isometric tests. Since pilot 1 was a healthy participant with no history of an ACL tear or ACLR, the LSI was computed by taking the nondominant limbs value and dividing it by the value of the dominant limb and then multiplying by 100. One possible explanation as to why pilot 1 displayed an asymmetric LSI in one dynamometer test and not the other is due to fatigue. One way to combat this is to encourage the participant to take at least a minute break in between each trial.

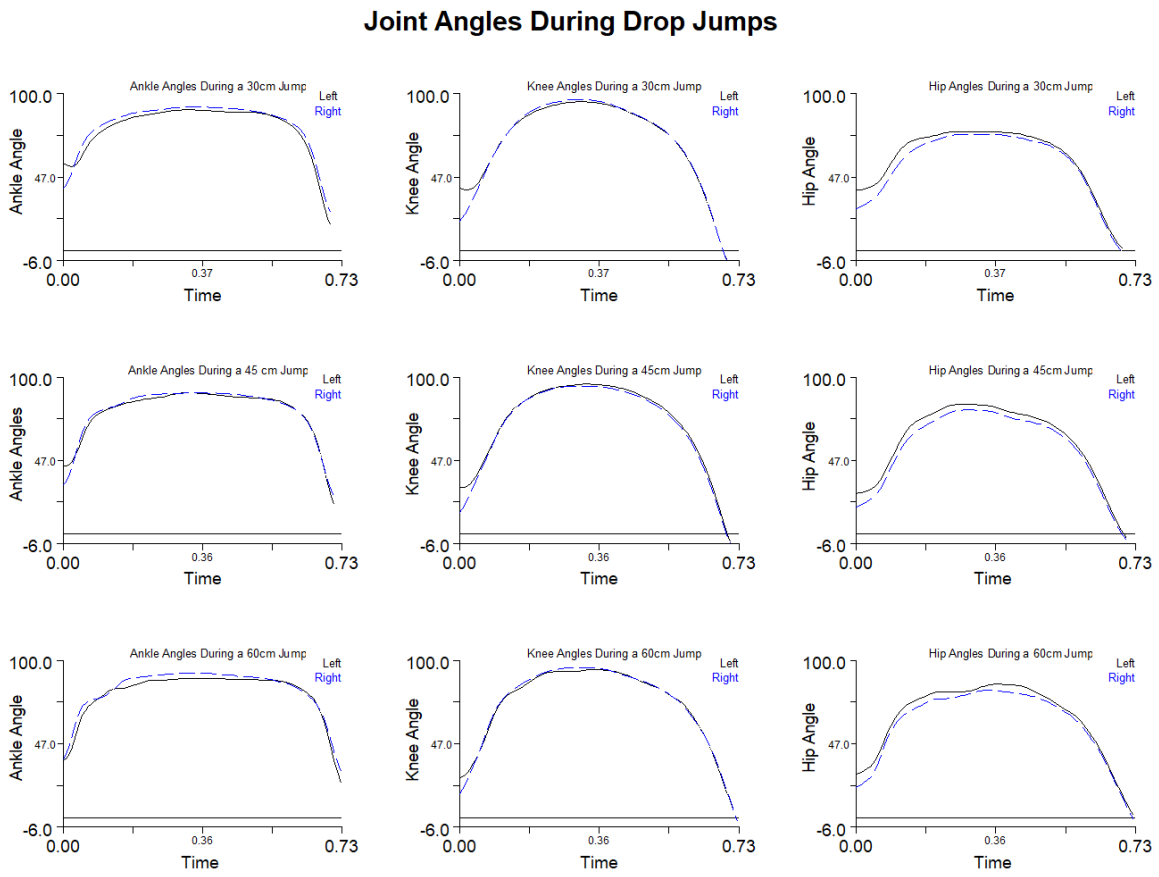

Figure 17. Pilot 1 hip, knee, and ankle joint angles at 30cm, 45cm, and 60cm.

Figure 2 displays the mean hip, knee, ankle joint angles across 5 trials at 30cm, 45cm, and 60cm. The x-axis represents time in seconds while the y-axis represents the joint angle throughout the trial. Each trial begins 200 milliseconds before initial contact and ends with toe-off. Pilot 1 is a healthy and symmetrical participant. Therefore, similar joint angle curves highlight how similar the movements were bilaterally across the three heights.

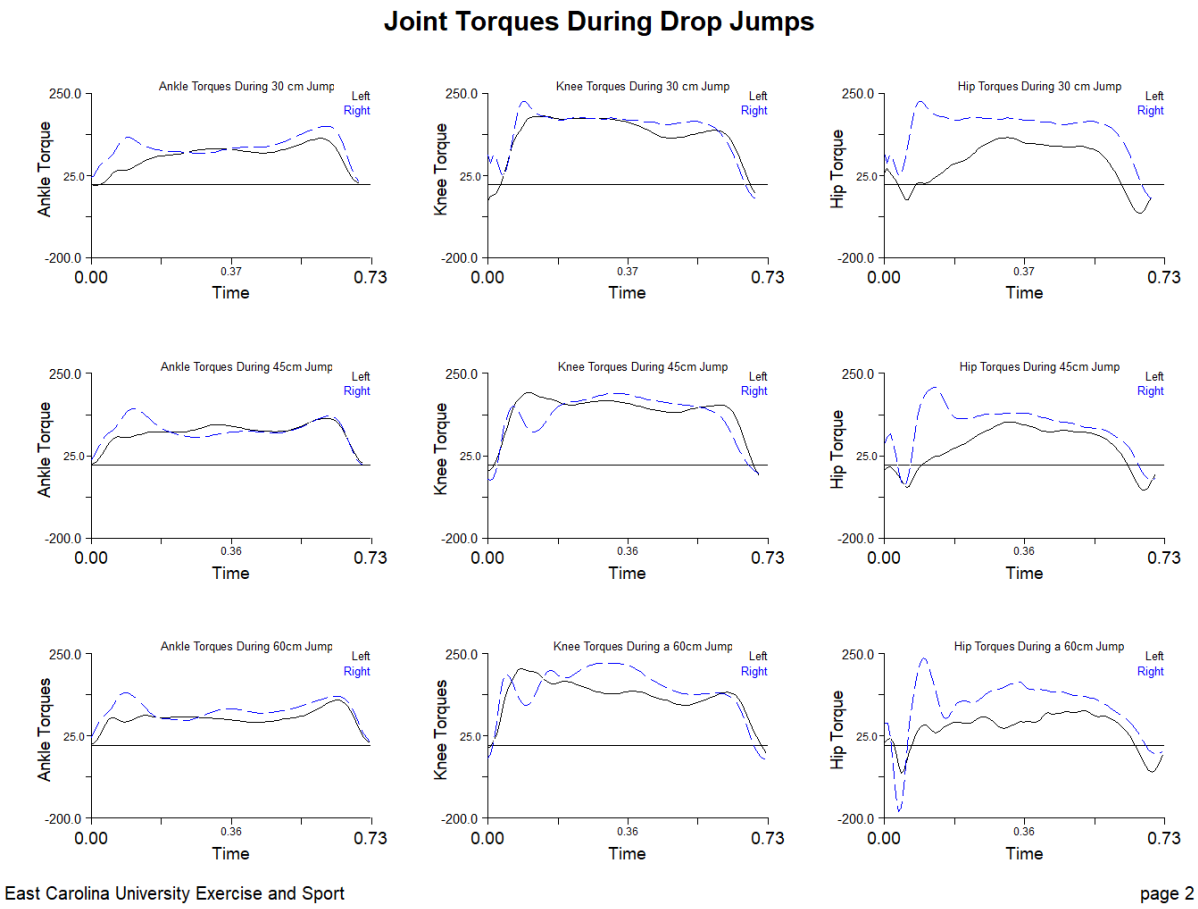

Figure 18. Pilot 1 hip, knee, and ankle joint torques at 30cm, 45cm, and 60cm

The x-axis represents time in seconds while the y-axis represents the joint torque measured in Newton meters (Nm). Pilot 1 produced the greatest torque at the knees, followed by the hips, and lastly the ankles, at each of the three heights. Since pilot 1 is a healthy participant with no history of knee injuries, there was no compensatory adjustments made during the drop jumping trials. Thus, pilot 1 had close to symmetrical bilateral torques at each of the three heights in the three joints examined.

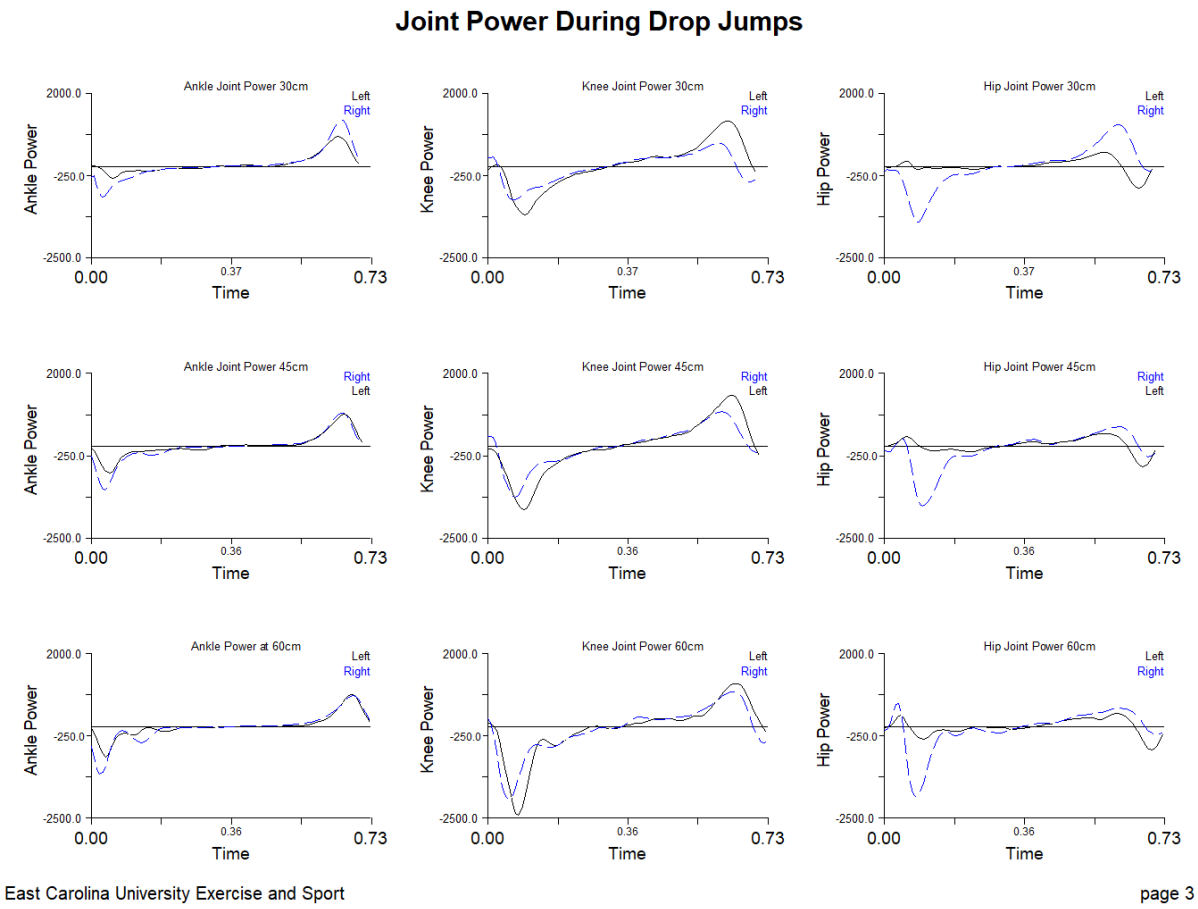

Figure 19. Pilot 1 hip, knee, and ankle joint power at 30cm, 45cm, and 60cm

The x-axis represents time in seconds while the y-axis represents the joint power in Watts (W). The joint power curves show the energy dissipation as pilot 1 uses eccentric thigh muscles to absorb the landing, followed by energy generation as pilot 1 uses concentric thigh muscles to generate power for the jumping phase. The joint power throughout the different heights remain symmetrical for the most part, which can possibly be accounted by pilot 1's symmetrical quadriceps strength and hop testing results. Pilot 1 produced the highest joint power at the knee, followed by the ankle, and lastly at the hip.

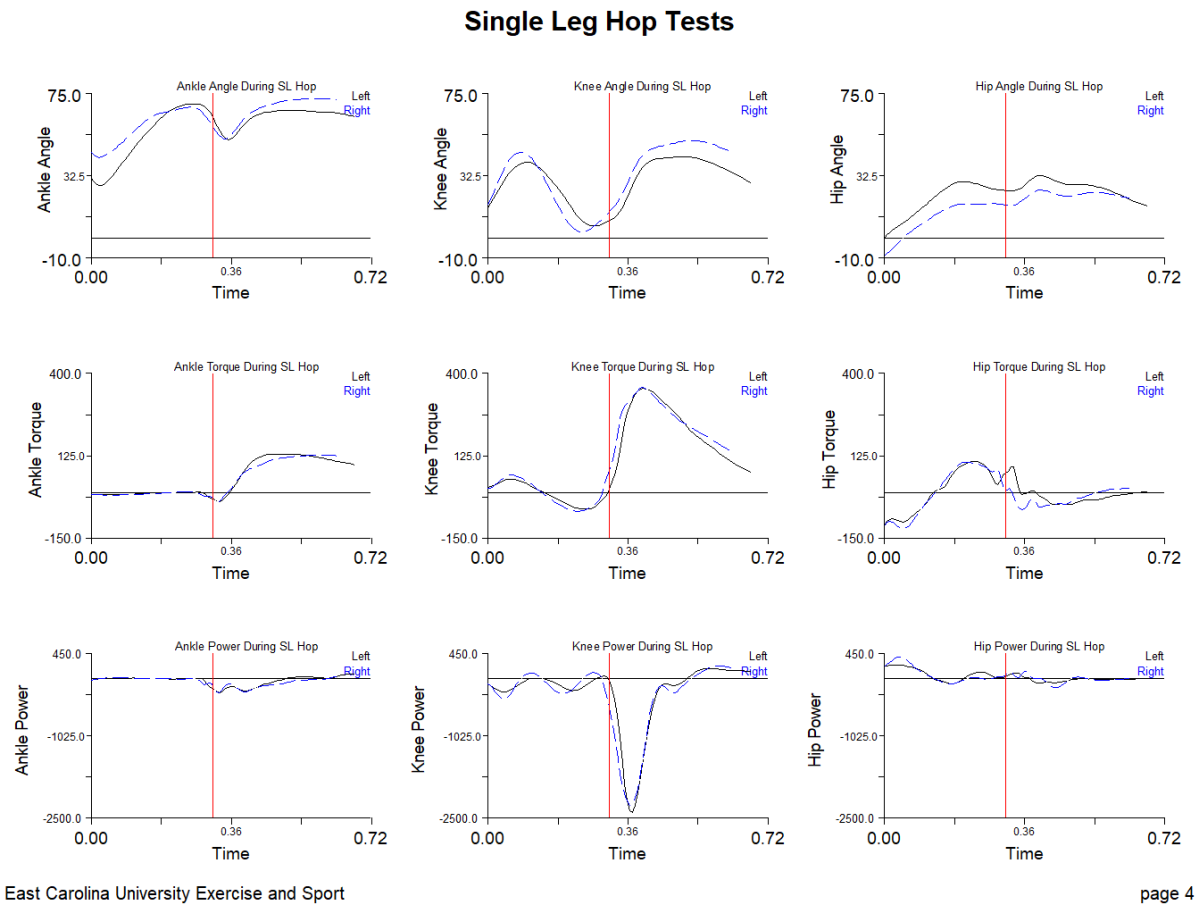

Figure 20. Pilot 1's hip, knee, and ankle joint angle, torque, and power during single leg hop tests

Figure 5 displays the joint angle, torque, and power across three single-leg jumping trials on each limb. The red lines display the initial contact with the forceplate. The x-axis represents time in seconds. The first row represents joint angles in degrees. The second row displays joint torques in Newton meters. Lastly, the third row represents joint powers in Watts. The trial began at toe-off and ended 400 milliseconds after the participant made contact with the forceplate. This timeframe was chosen because it allowed enough time for the participant to successfully land and stabilize themselves. All three of Pilot 1's hop testing results were symmetrical. Thus, it is appropriate that pilot 1 also shows symmetrical joint angle, torque, and power in the single leg

hop tests. These results highlight how pilot 1 used similar hopping techniques to accomplish the task.

Appendix C. Inverse Dynamics

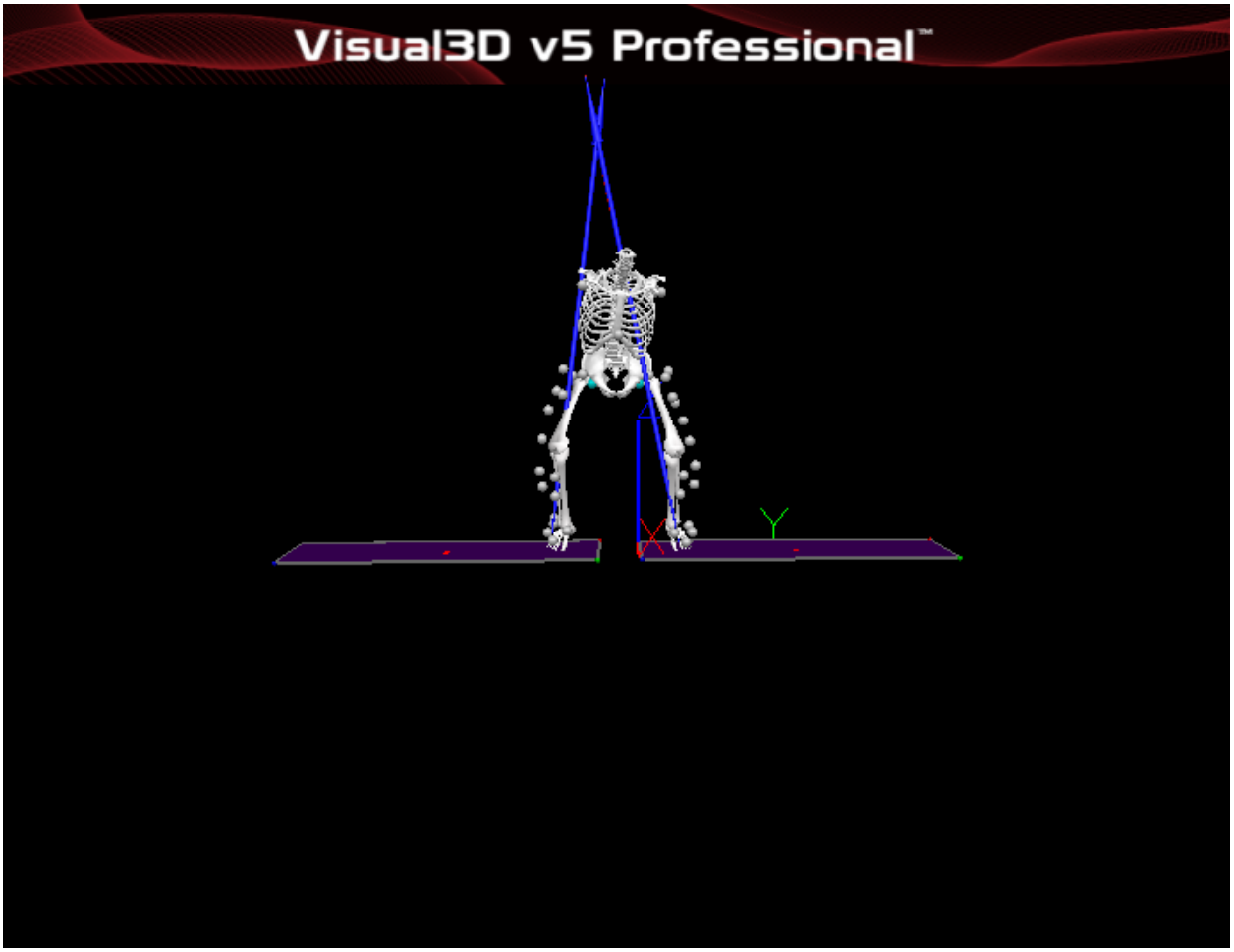

Figure 21. Visual 3D Model

Inverse Dynamics

When calculating joint torques, it is important to understand the assumptions about how these measurements are calculated. For example, each segment has a fixed mass and a segment center of mass. These were calculated using known cadaver measurements and using the segment distances calculated by Qualisys and V3D. Next, the center of mass (COM) of each segment is fixed. Additionally, moment of inertia and segment lengths are constant.

Variables calculated from Qualisys motion capture and Visual 3D. Below is an example as to how torques at a joint are calculated using inverse dynamics. Using Newton's third law, we know that there is an equal and opposite force acting at each hinge in a model. To solve for this, inverse dynamics works distally to proximally, where the proximal calculation carries over to the distal end of the next segment. To solve

Center of gravity acceleration of the foot:

ax=.92 m/s^2

ay=.46 m/s^2

Ground reaction forces:

GRFx=33.73N

GRFy= 629.37N

Inertia of segment: .0046

Angular acceleration: 10.9

Mass of foot= 1kg

Distal radius=0.06m

Proximal radius= 0.04m

JRFy= mass of segment* gravity-GRFy+mass(ay)

JRFy= $1kg(9.81m/s^2) - 629.37N + 1kg(.46m/s^2$= -619.1

JRFx=mass*ax-GRFx

JRFx=$1(.92) - 33.73 = -32.81N$

Moment of Ankle+ $-629.37N(.064m) - 33.73N(.02m) - 32.81N(.04m) - 619.1N(.041m) = .0046(10.9rad/s)$

Calculated Ankle torque= 67.70N/m

Calculated Ankle torque using V3D software= 67.82N/m.

Appendix D. Sagittal Plane Results

Sagittal Knee Angle		Healthy		ACLR - Quadriceps LSI ≥ 90%		ACLR - Quadriceps LSI < 90%		Limb Analysis						
Timepoint	Drop Jump Height	Injured Limb Matched	Uninjured Limb Matched	Injured Limb	Uninjured Limb	Injured Limb	Uninjured Limb	Height	Height * Group	Limb	Limb * Group	Height * Limb	Task * Limb * Group	Group
Initial Contact	30cm	12.7 ± 7.1	11.6 ± 8.0	16.5 ± 9.1	15.2 ± 9.0	12.5 ± 4.9	13.7 ± 6.3	0.07	0.883	0.961	0.716	0.385	0.210	0.517
	45cm	11.7 ± 4.3	11.3 ± 4.9	13.8 ± 8.7	14.0 ± 7.3	12.2 ± 3.4	12.3 ± 4.2							
	60cm	10.6 ± 4.3	9.8 ± 4.3	13.3 ± 6.9	15.1 ± 5.7	9.9± 2 .5	10.5 ± 4.0							
Peak Ground Reaction Force	30cm*	27.7 ± 9.7	32.1 ± 9.7	32.0 ± 16.4	43.8 ± 19.7	39.0 ± 8.4	39.3 ± 7.3	0.001	0.167	0.099	0.658	0.676	0.343	0.296
	45cm	27.2 ± 8.4	32.3 ± 10.7	28.6 ± 14.3	34.1 ± 7.5	34.8 ± 6.9	36.3 ± 6.9							
	60cm*	24.9 ± 8.3	31.1 ± 10.0	27.6 ± 12.3	31.5 ± 3.0	32.1 ± 9.6	33.3 ± 7.0							
Center of Gravity Vertical Minimum	30cm	86.6 ± 14.4	86.2 ± 12.6	81.7 ± 8.2	84.5 ± 9.1	96.1 ± 11.2	95.1 ± 10.9	0.418	0.360	0.600	0.102	0.704	0.323	0.187
	45cm	90.2 ± 13.3	88.7 ± 11.7	82.3 ± 11.0	84.8 ± 11.2	96.5 ± 8.7	96.0 ± 7.3							
	60cm	87.4 ± 12.6	85.5 ± 12.1	81.5 ± 14.4	85.2 ± 12.9	97.9 ± 9.9	98.2 ± 9.4							

Table 7. Sagittal Plane Knee Angles. ACLR LSI ≥90% represents the ACLR group with a quadriceps LSI ≥90%. ACLR LSI <90% represents the ACLR group with a quadriceps LSI of <90%. *Height Main effect p=0.05

Sagittal Knee Torque		Healthy		ACLR - Quadriceps LSI ≥ 90%		ACLR - Quadriceps LSI < 90%		Limb Analysis						
Timepoint	Drop Jump Height	Injured Limb Matched	Uninjured Limb Matched	Injured Limb	Uninjured Limb	Injured Limb	Uninjured Limb	Height	Height * Group	Limb	Limb * Group	Height * Limb	Task * Limb * Group	Group
Initial Contact	30cm	-0.70 ± 0.67	-1.24 ± 0.92	-1.19 ± 0.37	-1.70 ± 0.97	-0.82 ± 0.58	-1.12 ± 0.97	0.001	0.752	0.09	0.894	0.426	0.070	0.124
	45cm	-0.29 ± 0.87	-0.89 ± 0.77	-0.87 ± 0.53	-1.19 ± 1.16	0.10 ± 0.45	-0.99 ± 0.80							
	60cm	0.25 ± 0.95	-0.71 ± 0.86	-0.40 ± 0.92	-0.64 ± 1.25	0.04 ± 0.68	-0.52 ± 0.82							
Peak Ground Reaction Force	30cm	5.00 ± 2.74	3.67 ± 2.24	3.81 ± 3.24	3.52 ± 2.82	3.04 ± 4.04	3.43 ± 3.77	0.001	0.726	0.367	0.85	0.767	0.351	0.639
	45cm	5.19 ± 2.64	4.23 ± 2.30	4.92 ± 2.46	4.99 ± 3.62	4.34 ± 2.53	2.89 ± 2.69							
	60cm	6.08 ± 2.79	5.32 ± 2.45	5.57 ± 2.81	5.41 ± 3.07	4.90 ± 2.15	3.82 ± 2.23							
Center of Gravity Vertical Minimum	30cm	7.03 ± 2.45	5.72 ± 2.64	6.73 ± 2.29	6.54 ± 2.11	6.50 ± 2.83	6.75 ± 2.68 **	0.219	0.595	0.334	0.612	0.364	0.03	0.979
	45cm	7.06 ± 3.10	6.02 ± 2.80	7.13 ± 1.76	6.27 ± 2.12	6.89 ± 3.05	6.81 ± 2.96							
	60cm	6.69 ± 3.15	5.84 ± 2.87 **	6.45 ± 2.56	6.89 ± 2.88 **	6.13 ± 3.58	5.56 ± 3.24 **							

Table 8. Sagittal Plane Knee Torques. ACLR LSI ≥90% represents the ACLR group with a quadriceps LSI ≥90%. ACLR LSI <90% represents the ACLR group with a quadriceps LSI of <90%. **(group*limb*landing height) p=0.05.