Assessing Symmetry in Landing Mechanics During Single-Leg and Bilateral Tasks in Healthy Recreational Athletes

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Thesis submitted to the faculty of the Virginia Polytechnic Institute and State University
in partial fulfillment of the requirements for the degree of

Master of Science
In
Biomedical Engineering

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April 5, 2017
Blacksburg, Virginia

Keywords: ACL, Biomechanics, Landing, Symmetry, Kinematics, Kinetics, Motion Capture, Athletes, Sports

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ACADEMIC ABSTRACT

INTRODUCTION: ACL-reconstructed (ACL-R) patients exhibit side-to-side asymmetries in movement and loading patterns after surgery, some of which are predictive of a secondary ACL injury. These asymmetries have not been fully assessed in healthy athletes. **PURPOSE:** To quantify side-to-side symmetry in secondary injury predictors in healthy athletes and compare these metrics to those measured in previous cohorts of ACL-R patients, as well as to assess differences in these metrics between two landing tasks and between sexes. **METHODS:** 60 healthy recreational athletes performed seven trials of a stop-jump task and seven trials of a single-leg hop for distance on each limb. The kinematics and kinetics of the first landing of the stop-jump and the landing of the single-leg hop were analyzed with a 10-camera motion analysis system (240Hz) and 2 embedded force plates (1920Hz). Limb symmetry indices (LSIs) were calculated for each variable and compared between subject groups, tasks, and sexes with Wilcoxon Signed Rank tests ($p<0.05$). **RESULTS:** Control subjects exhibited asymmetry in hop distance ($p=0.006$). ACL-R subjects displayed greater asymmetry in knee flexion variables, peak forces, and peak knee extension moments during the bilateral landing ($p<0.001$) and in hop distance ($p<0.001$). Control subjects showed greater asymmetry in knee flexion variables during the single-leg hop ($p<0.001$). Males and females showed similar symmetry in both tasks. **CONCLUSIONS:** Symmetry cannot be assumed in control subjects in all metrics. Asymmetries are more prevalent in ACL-R athletes than in healthy controls. Future work will continue to examine the usefulness of each metric in assessing ACL-R rehabilitation.

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GENERAL AUDIENCE ABSTRACT

Up to 200,000 ACL injuries occur in the US annually. Researchers have demonstrated that ACL-reconstructed (ACL-R) patients display differences in movements between their injured leg and their healthy leg during athletic activities. In some cases, these differences, or asymmetries, can increase a person's risk of sustaining a second ACL injury. However, movement symmetry is not well understood in people who have not had an ACL injury. The goal of this work was to better understand asymmetries in healthy people so that we can better assess those who have suffered an ACL injury. We did this by assessing movement in healthy athletes during single- and double-leg landing activities that have been traditionally used to assess recovery in ACL-R patients. We found that the healthy athletes exhibited significant asymmetries in several metrics during both the single- and double-leg landings. These results indicate that movement symmetry should not be assumed in healthy control subjects. We also similarities and differences in symmetry profiles between single- and double-leg landing activities in a control population. The results of this study will enable researchers to better understand movement deficiencies in ACL-R patients when compared to healthy control subjects as we continue to work to minimize re-injury following return to sport in ACL patients.

ACKNOWLEDGEMENTS

I would like to start by thanking Adam Heilmann and Lindsay Maguire, who also participated in the data collection for this project. I would also like to acknowledge my fellow lab members – Jon Gladish, Cherice Hughes-Oliver, and Kristen Renner – who had to put up with me and my testing schedule during the project.

I would like to thank my advisor, Dr. Robin Queen, for giving me the opportunity to pursue quality research in this field, and for giving me the tools necessary for success. You have been a constant source of knowledge and encouragement in the two years I have worked for you, and I feel fully prepared for a career in biomechanics as a result of working in your lab.

To my family and friends: your constant encouragement throughout this process has been much appreciated and has been very important in the success of this project. Katie, you have shown me what true patience looks like, and I couldn't have done this without your support.

Finally, I would like to acknowledge all of the subjects that participated in this project. This project wouldn't exist without your willingness to get involved in biomechanical research.

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1 INTRODUCTION

1.1 Introduction to the ACL

1.1.1 Etiology of ACL Injuries

Injuries to the anterior cruciate ligament (ACL) are common in sports, especially in sports that require cutting and pivoting motions [1,2]. It has been estimated that 130,000-200,000 people sustain an injury to the ACL each year [3,4]. Additionally, 1 of every 60 to 100 females sustain an ACL injury [4]. ACL tears make up 2.6% of all injuries in college sports [5], and 45% of all internal knee injuries involve the ACL [6]. This high incidence of injury results in a significant financial burden. Each ACL reconstruction (ACL-R) procedure carries an expense of over $38,000 [7], and the total cost for these procedures has been estimated at $1-2 billion annually [8].

It is well-known that females are at a higher risk of an ACL injury than males [1,5,9,10]. Females are 4-6 times more likely to tear an ACL while participating in pivoting and cutting sports than males who participate in the same sports [1]. Arendt et al. found that ACL injury rates in females were nearly three times higher than injury rates in males in basketball and soccer [11]. The overall number of ACL injuries in female athletes has also risen significantly in recent years due to increased female participation in sport activities [12,13]. Sports such as women's gymnastics, women's soccer, women's lacrosse, and women's basketball have some of the highest risks of ACL injury in high school and collegiate athletics, along with spring football and men's lacrosse [5,10]. Additionally, females have a 26% chance of returning to pre-injury performance following an ACL injury, while males have a slightly higher chance at 37% [14].

1.1.2 Anatomy of the ACL

The anterior cruciate ligament lies within the knee between the femur and the tibia. It originates in the posteromedial portion of the lateral femoral condyle and descends medially, distally and anteriorly to the anterolateral portion of the medial tibial plateau [15-17]. Average ligaments are 27-38mm in length and 7-12mm in width, with males possessing slightly larger ligaments in terms of length and volume [15,18,19]. The ACL is comprised primarily of Type I and Type III collagen, with small amounts of proteoglycans and elastin [15]. The primary function of the ACL is to resist anterior tibial translation and tibial rotation [15,16,20]. The ACL has several structural components, but these have proven to be difficult to distinguish in vivo. Because of this, scientists have broken down the ACL into specific functional units, or bundles [15,16,20-22]. The two primary bundles are referred to as the anteromedial bundle and the posterolateral bundle [16,20-22]. The anteromedial bundle is primarily active during knee flexion as it protects against anterior tibial translation, while the posterolateral bundle resists tibial rotation while the knee is close to full extension [20]. When the knee is fully extended, the anteromedial bundle is relaxed and the posterolateral bundle is tight and subjected to a high tensile load, but the anteromedial bundle tightens and the posterolateral bundle relaxes as the knee flexes [16,21,22]. As a whole, the ACL is longest during knee extension, and its length decreases by approximately 10mm when the knee is fully flexed [23-26]. It follows that the strain placed on the ACL is also highest at these high length positions [25-27].

1.2 ACL Injuries

1.2.1 Injury Risk Factors

Many researchers have worked to better understand causes and risk factors for ACL injuries. A portion of this involves examining trends in the injuries seen in competitions and athletic activities. 70% of all ACL injuries are non-contact in nature [28]. Common movements that typically result in an injury to the ACL include pivoting, particularly while the knee is near full extension [2,29], landing from a jump with the knee in or near full extension [29.30], or a change of direction or cutting motion combined with a deceleration [29,30]. Beynnon et al. reinforced these ideas by showing that extending the knee increases strains on the ACL [31]. Additionally, McNair et al. reported that ACL injuries typically occur at the point in time when the foot returns to the ground following either a step or a jump [32]. Researchers have also found that ACL injury rates are higher during games or competition than in a practice setting, suggesting an influence of exertion on ACL injury risk [33]. This effect could be due to increased fatigue seen in athletes during games when compared to practices. Many studies have shown that movements are altered by fatigue during jumping and landing tasks [34-40]. However, Dai et al. studied relative effort during landings and found that while a "softer" landing from a jump reduced ACL loading, it also resulted in a decrease in athletic performance [41]. This suggests that the level of exertion expended by an athlete during an athletic competition could have an effect on ACL injury risk. This idea is reinforced by studies that have identified increased quadriceps activity as a risk factor for injury [42-45].

Biomechanical factors have been studied extensively in an attempt to find the specific mechanism that leads to an ACL tear [2,12,27,46-50]. The "valgus collapse" position is one that has been identified as a significant risk factor for injury. This position involves a combination of hip internal rotation, knee valgus, and tibial external rotation, usually with the knee near full extension [51,52]. When this occurs, the athlete's knee seems to "collapse" downwards and inwards. This position was first identified through video analysis of ACL injuries occurring during athletic competitions [51,52]. It has since been verified as a risk factor by a variety of methods, including retrospective studies, in vitro tissue testing, and through mathematical models, as well as through biomechanical analysis using three-dimensional motion capture and force platforms [12,29,31,32,42,43,47,48,53-55]. Other than the valgus collapse position, specific biomechanical risk factors include a large hip extension at ground contact [47] and decreased hip extensor work [56]. Decreased core stability, specifically deficits in lateral trunk control and low back pain, have also been cited as injury risks [50,54]. Another major component seen in many ACL injuries is a large anterior tibial shear force caused by anterior translation of the tibia [42,55,57]. Since the ACL acts as the primary restraint against anterior tibial translation, this motion places large strains on the ACL [57]. It has been shown that a large quadriceps force can exacerbate this anterior tibial translation and increase an athlete's risk of injury, particularly between 50° of flexion and full extension [29,31,42-44]. The hamstrings muscles act as knee stabilizers as they co-contract with the quadriceps muscles to prevent this anterior motion [44,58,59]. Because of this balance between the two major muscle groups, athletes' "quad-ham ratios" are commonly examined as a potential injury risk, with smaller quad-ham ratios being more advantageous than larger ones [44,58,60-62].

It is widely known that female athletes are more susceptible to ACL injuries than male athletes in the same sports. Because of this discrepancy, researchers have investigated differences in movement and loading patterns between males and females to further identify risk factors for ACL injuries. Previous research has shown that when compared to males, females exhibit greater knee valgus angles and moments [53,54,63-65] and decreased knee flexion [64-68] in landing and cutting tasks. These two factors, combined with greater knee external rotation [64,68], show that females are more likely to exhibit the "valgus collapse" position that has been attributed to many ACL injuries. Krosshaug et al. found that the valgus collapse position is 5.3 times more likely to be associated with a female ACL tear than with a male ACL tear [51]. However, this position is not the only risk factor associated with a greater likelihood of ACL tears in female athletes. Other movements, positions and loading patterns identified as risk factors include greater vertical and posterior ground reaction forces (GRFs) in landing tasks [12,43,69], greater hip adduction in both single-leg landings and cutting tasks [12,64], decreased hip flexion [64,66,67], greater anterior tibial shear force [65], and increased variability in knee kinematics and kinetics [64]. Females also exhibit increased quadriceps activity and decreased hamstring and hip muscle activity when compared to males [50,67]. Each of these factors predisposes females to a higher risk of ACL injury.

In addition to these biomechanical risk factors, researchers have explored factors that are intrinsic in each individual athlete. Researchers have shown that an athlete's age can have a significant effect on their risk of ACL injury, particularly in female athletes [12,325,326]. They have reported that postpubertal females carry a greater risk of ACL injury than females that have not yet reached puberty. While some of this increased risk is due to other intrinsic factors associated with puberty, much of the risk is due to changing movement mechanics, as others have shown that movement mechanics change over time in young athletes [322,323,327]. Much work has been done to examine the effects of intercondylar notch width on an athlete's ACL injury risk [70-75]. Since the ACL translates anteriorly, medially and distally through this notch, its width could affect the motions and actions of the ACL. Many studies have found that athletes with a narrower intercondylar notch are at an increased risk of ACL injury [70-73,75]. The geometry of the tibial plateau is also another factor that has been explored. Research has shown that a steeper lateral tibial plateau can predispose athletes to ACL injury [76-78]. Additional risk factors include generalized joint laxity [62,75,79-81], genetic influences [82,83], and hormonal effects [79,84-89]. Limb dominance has also been discussed as a potential factor in ACL injuries. Some have asserted that each limb is equally likely to suffer an ACL injury [90,91], while others have found no differences between limbs in gait and in jumping and landing in healthy individuals [92-94]. However, a study by Brophy et al. reported greater incidences of ACL tears in the dominant legs of males and in the non-dominant legs of females [95]. Wang et al. also found side-to-side asymmetries in knee extension when the non-dominant limb underwent ACL reconstruction and asymmetries in varus rotation and tibial internal rotation when the dominant limb underwent ACL reconstruction [96]. Furthermore, additional studies have found significant effects of limb dominance in normal and fast gait [97,98]. Although multiple methods exist to determine limb dominance, all of these studies determined that the dominant limb was the one that subjects preferred to use to kick a soccer ball [90,91,93,95,96,98]. These suggest that there may be a neuromuscular influence of limb dominance on potential injury mechanisms. However, this idea has not been thoroughly explored.

1.2.2 Side-to-Side Asymmetry

The quantification of side-to-side symmetry in movements and loading patterns was originally completed in ACL-R populations to assess rehabilitation following surgery. Side-to-side symmetry is the quantification of the similarity of the actions of the two lower limbs in any metric during a given task. For example, a person would be said to possess side-to-side symmetry in knee flexion at ground contact if each of the person's knees was flexed at the same angle at the instant of ground contact. The two limbs can be compared by using a simple t-test [99-103], but their symmetry can also be quantified by using a limb symmetry index (LSI), or simply a symmetry index (SI) [104-114]. Some studies employ a simple ratio for use as an LSI, as given below:

$$\frac{X_L}{X_R} * 100\% \tag{1}$$

where X_L and X_R denote data for the left and right limbs, respectively, for the same variable [106,107,109]. This ratio can also be reversed to label the numerator as the right limb variable. In the case of pathologic patients, the ratio can be written as:

$$\frac{X_{Ax}}{X_{NAx}} * 100\% \tag{2}$$

where X_{Ax} and X_{NAx} denote data for the affected and non-affected limbs, respectively [108,112]. Robinson et al. proposed a formula for quantifying side-to-side symmetry:

$$\mathrm{SI} = \frac{X_R - X_L}{0.5 * (X_R + X_L)} * 100\% \tag{3}$$

where X_R and X_L denote data for the right and left limbs, respectively [113]. A large number of studies now employ this formula [105,110,111,114]. This is likely due to the fact that Equation 3 better accounts for differences in variables of small magnitude. In a hypothetical example, if the value of a variable in the affected limb is 0.2, and the value of the same variable in the unaffected limb is 0.004, the original ratio would yield an LSI of 5,000%. However, if these values are plugged into Equation 3, the LSI is found to be 192.2%. This value allows for more reasonable comparisons to other symmetry indices. Side-to-side symmetry can also be assessed by simply examining the numerator of the symmetry index referenced above, as was done by Paterno et al. [162]. This metric can be utilized to examine the magnitude of the differences between limbs and compare these differences between two or more groups.

In addition, researchers have attempted to quantify symmetry in the time-series data of certain variables [115-124]. A large number of studies have employed principal component analysis (PCA) for this quantification [119-122]. This technique involves the separation of several waveforms into components based on the variance in the data. This allows the researcher to identify differences between the waveforms. However, the method requires a complex calculation and is difficult to understand. Other researchers utilize regions of deviation to quantify waveform symmetry [115-117]. This technique essentially compares the values of two waveforms through either a simple t-test or a difference calculation at each time point in the series. While this is a much simpler calculation than what is employed by the PCA method, this technique is limited to the comparison of only two data sets, and it does not account for similarities and differences in the general shapes of the waveforms. Other techniques sometimes used for this purpose include neural networks and eigenvectors [118,123,124].

The coefficient of multiple determination (CMD) is another metric that quantifies the similarity of two or more waveforms [125,126]. Kadaba et al. first proposed this metric for use in biomechanical studies:

$$CMD = 1 - \frac{\sum_{i=1}^{N}\sum_{t=0}^{T}(y_{it}-\bar{y}_t)^2/[(T+1)(N-1)]}{\sum_{i=1}^{N}\sum_{t=0}^{T}(y_{it}-\bar{y})^2/[(T+1)N-1]} \qquad (4)$$

where N = the number of curves to be compared, T = the number of time points making up each curve, y_{it} = each signal, $\bar{y}_t$ = the mean of the signals at each time point, and $\bar{y}$ = the mean of all signals [125]. This metric has been previously utilized to measure the repeatability of subjects' kinematics and kinetics during separate testing sessions [125-129]. It follows that this metric could be used to compare the waveforms of the kinematics or kinetics of each limb throughout a movement task, though this utilization has not been reported in the literature. This metric is useful because it can be found through a relatively simple calculation, and it is visually intuitive and easy to understand. These are important factors for physicians, as the metric can be employed to assess rehabilitation and injury risk in a clinical setting.

Previous work to assess the usefulness of CMD calculations in quantiying side-to-side symmetry has indicated that this metric is most appropriately utilized to compare waveforms that are similar in nature. According to the CMD formula, the value of the CMD can fall between -1 and 1. However, values that fall below 0 are deemed to be inappropriate for analysis. This means that only positive CMD values can be reasonably used to draw conclusions from the data being analyzed. This factor limits the time-series variables that can be analyzed with this metric. Traditionally, kinematics and kinetics in the sagittal plane in each limb display similar profiles, while kinematics and kinetics in the frontal plane in each limb seem to act more independently of each other. Figure 1 shows an example of these differing profiles. This means that the CMD metric seems to be most useful for the analysis of variables that display similar movement or loading patterns between limbs, such as sagittal knee angles or vertical ground reaction forces. This idea is reinforced by previous studies that examined only sagittal plane knee mechanics when analyzing symmetry between waveforms [287,291,314]. Previous work has also shown that the examination of waveform symmetry in this manner yields different information about

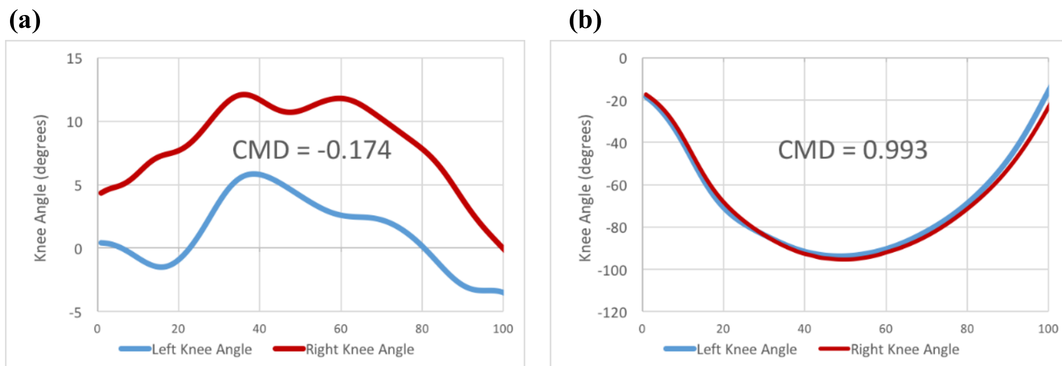


Figure 1a: Frontal Plane Knee Angles and Corresponding CMD Value
Figure 1b: Sagittal Plane Knee Angles and Corresponding CMD Value

movement symmetry in a particular metric than a simple LSI calculation. Preliminary data in an ACL-R population has suggested that athletes may show significant asymmetries in CMD values in some metrics while showing no significant asymmetries in LSI values in the same metrics, and vice versa. This suggests that both symmetry assessment techniques may be useful to gain a better understanding of an athlete's movement patterns after ACL reconstruction.

In addition to these techniques, side-to-side symmetry can also be quantified as the difference between limbs across an entire group [41,102,141,294]. While this method does not give information about side-to-side symmetry within each subject in a group, it does allow for an analysis of the actions of one limb compared to the other in an entire group.

The method has been utilized previously to quantify differences between dominant and non-dominant limbs in healthy control subjects [102] as well as between surgical and non-surgical limbs in ACL-R patients [41,141,294]. This method differs from the quantification of symmetry in each individual subject in a group, but it is unknown if one method is more useful than another when assessing side-to-side symmetry in an entire group.

Previous studies have found significant asymmetries in ACL-injured athletes, particularly in ground reaction forces [41,100,130-134], leg strength [135-138], knee flexion angles [100,101,132], knee extension moments [41,100,101,103,132,133,135,139-142] and knee valgus moments [142,143]. Additionally, some of these asymmetries were found to be significantly greater than asymmetries found in healthy control subjects [101,130,131,134,139,141-143], and while few studies have examined differences in side-to-side symmetry between sexes, females have been shown to exhibit greater asymmetries in frontal plane knee angles during landing [53,265]. Asymmetries in knee moments are of particular importance to researchers. In many cases, lower moments in the knee can lead to higher moments in the hip and the ankle [103,139,140,144], thus altering mechanics and increasing the risk of injury in one or more joints. Additionally, asymmetries in movement and loading patterns can persist for long periods of time, even after the athlete has been released to return to sports (RTS) [131,134,141,143,145,146]. These have been seen specifically in gait mechanics [147-150], and in many cases these asymmetries can lead to knee osteoarthritis [151]. Asymmetries can persist for up to 7 years after ACL reconstruction [100,131,134,143,145,148,150,152-155]. These asymmetries are generally found in loading patterns [100,134,143,152,153,156], energetics (i.e. joint works and powers) [153,157], and leg strength [145,154,155,158,159].

Asymmetries have been found to be predictive of ACL injury [160-162]. Myer et al. found correlations between primary ACL injury risk and the difference in anterior knee laxity between limbs [160]. Additionally, Paterno et al. identified large asymmetries in peak knee extension moment during the initial landing phase as a significant risk factor for a secondary ACL injury [162]. In this study, 56 athletes who had previously undergone ACL reconstruction and had been released to return to sport were tested during a bilateral landing task. Lower extremity joint kinematics and kinetics as well as postural stability were quantified in each athlete. The athletes were followed for one year after the testing, with 13 athletes suffering a secondary ACL within that time. A logistic regression model was used to determine the biomechanical factors most associated with a secondary ACL injury [162]. Furthermore, asymmetries in other metrics, such as quadriceps strength or ground reaction forces, have been utilized to predict additional kinematic and kinetic asymmetries [41,138].

1.3 Treatment Options

1.3.1 Surgical Considerations

Injuries to the ACL are treated through either surgical reconstruction or non-surgical rehabilitation. Previous studies have found poor knee function and stability in patients who underwent non-surgical rehabilitation protocols [163-165] and improved quad strength and stability in patients who underwent a surgical reconstruction [106,166,167]. One study has shown reduced rates of knee osteoarthritis (OA) in patients who underwent ACL reconstruction [166]; however, many studies have reported similar rates of knee OA between ACL-R patients and those who underwent nonsurgical treatment for their injury [168-171], and some studies have even shown larger rates of knee OA in reconstructed patients [167,172-174]. It should be noted, however, that these radiographic changes at the knee may be more influenced by the activity levels seen in each group of patients following ACL injury. Many of the studies that have shown equivalent or higher rates of knee OA in ACL-R populations than in nonsurgical populations also cited reduced activity levels in the nonsurgical groups [169-173]. Von Porat et al. reported that 30% of ACL-injured patients changed their lifestyle severely following injury [170]. In another study examining female soccer players 12 years after ACL injury, more than 50% of the players had never played competitive soccer again after their injury [171]. Furthermore, Neuman et al. advocated for reducing activity levels after injury in order to avoid future complications, and Jomha et al. did not even include ACL-injured patients with low activity levels in their assessment of OA outcomes after injury [172,175]. It can be assumed that the large number of knee OA cases seen in ACL-R patients may be caused by the higher activity following injury in these patients. In addition, the knee stability seen in ACL-R patients is necessary for returning to high levels of activity [106,166,167]. As a result, up to 90% of ACL injuries, particularly in young athletes, are treated through surgical intervention [176,177].

1.3.2 ACL Reconstruction

A typical ACL surgery involves the implantation of a graft that mimics the function of a normal ACL. For these procedures, surgeons choose to use either an autograft or an allograft [178-181]. It has been shown that the outcomes of autografts and allografts are similar [181-183], and one study reported decreased anterior laxity and knee extension torque in knees reconstructed with allografts when compared to autograft reconstructions [184]. However, others have indicated higher rates of failure in allografts than autografts in similar patient populations [178,179,185-187], and studies have found that patients with allografts are more likely to suffer a secondary ACL injury [188,189]. Autografts have been shown to elicit an improved biological response when implanted and are stronger than allografts [180]. In addition, the use of allografts in ACL-R has been reported to enhance the likelihood of single-leg hop distance asymmetry, which is a primary metric used to determine return-to-sport readiness after ACL-R [190]. The most common types of autografts used for these procedures are hamstring tendons and patellar tendons (sometimes referred to as bone-patellar tendon-bone (BTB) grafts [191-196]. Previous research has found similar outcomes among these graft types [191,192,194,197-199]. However, the use of patellar tendon grafts has been shown to lead to increased problems at the donor site, specifically at the patellar tendon, than other graft types [193,198]. Symptoms of this donor site morbidity include anterior knee pain and kneeling pain in patients with a patellar tendon graft [193-195,198,200].

In addition to the option of graft type, surgeons choose to use either single- or double-bundle grafts when performing an ACL reconstruction surgery [201-208]. Since the anteromedial bundle of the ACL is the primary resistor against anterior tibial translation in a healthy ACL, surgeons have used a single-bundle technique to repair the ACL and thus prevent anterior tibial translation in the injured knee [203]. As of 2009, 90% of all surgeons utilized this method [209]. However, this single-bundle graft has been shown to lead to increased rotational instability, also known as pivot shift [201-203,208]. Double-bundle grafts have been proven to limit this pivot shift [201,202,208] and in some cases have also been shown to limit knee laxity and improve anterior stability [202,207,208,210]. However, other studies have reported similar results in these metrics between the two techniques [204,206].

During an ACL reconstruction procedure, surgeons must drill holes in both the femur and the tibia to place the ACL graft. The placement of these holes, or tunnels, has been debated [211-213]. Researchers agree that the tunnels used in the reconstruction should mimic the anatomic location of the original ACL in order to achieve the best surgical outcomes [211,213,214]. The transtibial drilling technique has been used in the past to accelerate the surgery and reduce morbidity [215]. This technique involves drilling the femoral tunnel through the tibial tunnel [216]. However, the technique does not ensure graft placement that is consistent with the original ACL anatomy [212,215-217]. Furthermore, research has shown that incorrect tunnel placement can lead to significant problems postoperatively, including higher levels of knee OA [215], rotational instability [213], or graft failure [213,216,217]. For this reason, surgeons sometimes drill the femoral tunnel through the anteromedial portal of the knee [218-220]. This technique has been shown to improve tibial translational and rotational stability when compared to the transtibial drilling technique [218,219].

1.4 Treatment Outcomes

1.4.1 Knee Osteoarthritis

Post-traumatic knee osteoarthritis (OA) has been identified as a significant long-term complication of ACL injury [2,166,170,171,174,222-224]. Incidences of knee OA in people with a previous ACL injury have been reported at 41-90% [2,170,171,221,222].

One possible reason for this large incidence is the altered movement mechanics seen in people with an ACL injury. Studies have reported altered gait mechanics in ACL-injured people, both when compared to the contralateral limb [147-149,151,224] and when compared to healthy control subjects [150,223,225,226]. Many of these altered mechanics lead to the loss of cartilage at the articular surface of the knee [227-230]. This is likely due to decreased stability in the knee due to the loss of the ACL [166,174,224,228]. Additionally, the altered gait mechanics seen in these people could accelerate the cartilage thinning mechanism that is commonly seen in knee OA conditions [228,229]. ACL reconstruction surgeries have been shown to result in fewer cases of knee OA when compared to conservative treatment strategies [147,150,174,224]. Several studies have indicated that ACL-R surgery improves gait mechanics when compared to the patients' pre-operative gait [150,225]. Additionally, ACL-R procedures have been shown to return patients to normal levels of gait [231], but many believe that deficits still exist in these patients when compared to healthy controls [150,232-236]. Other risk factors include choice of graft type [81,237], knee laxity [238], more than one ACL injury [221], asymmetries in hop distance [81], and accompanying meniscal injuries [2,222,239]. This research shows that the risk of knee OA in ACL-injured people is high regardless of the success of ACL-R surgery or the treatment option chosen [222].

Another potential risk factor for knee OA after an ACL injury is decreased quadriceps strength commonly seen in ACL patients [237,240-242]. In some cases, this decreased strength can affect movement mechanics in these patients [243-245]. Shelburne et al. reported greater anterior tibial translation as a result of this quadriceps weakness [245], while others found decreased knee moments in the ACL-R knee [243,244]. These alterations in movement mechanics could lead to cartilage damage in the knee. Decreased quadriceps strength also limits the muscles' ability to absorb energy. Without this energy absorption capability, cartilage and other structures in the knee must absorb this energy, thus leading to potential problems with knee OA [241].

1.4.2 Secondary ACL Injuries

In recent years, researchers have begun to examine risk factors for secondary ACL injuries, or ACL injuries that occur in athletes who have previously suffered an ACL injury. ACL injury rates have been found to be higher in previously injured athletes than in uninjured athletes [81,177,246-249]. Paterno et al. asserted that ACL-R athletes were six times more likely to suffer a secondary ACL injury than healthy athletes [248], and Salmon et al. reported that the largest risk factor for an ACL injury in athletes was the occurrence of an ACL injury in the previous 12 months [249]. Pinczewski et al. also reported a secondary injury rate of 27% in a cohort of ACL-R patients [81]. Females are more likely to suffer these secondary ACL injuries than males [177,188,250,251]. Paterno et al. reported that females are four times more likely to suffer a secondary injury than males [177], while Maletis et al. quantified this risk at 26% in the contralateral limb [188]. Noojin et al. also reported failure rates of 23% in females and 4% in males [251]. Some of these injuries are due to differences in outcomes of different graft types: allograft use has been shown to result in more secondary injuries than autograft use [188,189], and hamstring grafts have been shown to carry a higher failure risk than patellar tendon grafts [188]. However, secondary ACL injuries are seen at least as commonly in the contralateral limb as in the originally injured limb [247,252-254]. This means that there could be biomechanical factors present in these ACL-injured athletes that predispose them to larger secondary injury rates. These factors could have been contributors to the initial ACL injury or the results of the initial injury. Factors that have been previously explored in secondary ACL injury populations include decreased quadriceps strength, increased knee abduction moments, and increased frontal plane range of motion during landing [162,255,256].

1.5 Testing Methods

1.5.1 Single-Leg Hop Testing

Side-to-side asymmetries have traditionally been associated with an increased risk of secondary injury in ACL-R athletes [130,162]. As such, a large portion of functional testing to determine an athlete's readiness for RTS is based on side-to-side asymmetries. Single-leg hop testing has been extensively utilized to assess functional recovery in ACL-R patients [81,138,140,141,190,226,257]. In this testing, single-leg hop distance is quantified and compared between the two limbs. Previous work has indicated that a hop distance symmetry of greater than 90% represents an adequate level of functional recovery following ACL-R [81,190,258]. Hop testing has been proven to be useful in differentiating between ACL-R athletes and healthy controls [142,259], and studies have indicated deficits in hop distance symmetry in ACL-R populations [141,144,260]. Additionally, several studies have claimed that a unilateral landing carries a greater risk for ACL injury than a bilateral landing, so it is imperative to assess this movement in ACL populations [46,80]. However, it has been suggested that single-leg hop testing does not fully assess functional recovery following ACL-R [136,140]. In one study, Novak et al. found an 11% asymmetry in quadriceps strength but only a 6% asymmetry in hop distance [136]. Orishimo et al. also reported similar results, with a 7% asymmetry in hop distance but significant side-to-side asymmetries in knee moments and powers [140]. It is possible that the compensations being found in many asymmetric ACL-R athletes may allow the athletes to achieve "normal" symmetry in hop distance [103,139,140,144]. Ernst et al. found lower knee extension moments and larger hip and ankle moments in ACL-R athletes during single-leg vertical jumping and landing [139]. Similar compensations have also been reported in subsequent studies [103,140,144]. These results prove that individuals that exhibit symmetric single-leg hop distances may not be moving symmetrically, which may expose them to a higher risk of secondary injury. This has prompted some researchers to begin examining what has been called "movement quality" to more appropriately assess single-leg hopping tasks [99,101,102,261,262]. Movement quality involves investigating how a movement is accomplished rather than just the result of the movement. This enables researchers to identify movement compensations and better assess these movements. Several studies have utilized video analysis to assess the quality of the single-leg hop movement [261,262]. Others have examined this movement with three-dimensional motion capture and force analysis [99-101,242]. These studies have found side-to-side asymmetries in peak knee extension moment [99-101,242], peak vGRF [100-101,242], knee flexion at ground contact [100] and sagittal knee ROM [99]. It should be noted that the athletes participating in most of these studies had been cleared to return to sports-related tasks at the time of their testing [99,101,142]. These results show that ACL-R athletes may not be moving in a safe manner at the time that they return to sports. This reinforces the need to assess movement quality when assessing return-to-sport (RTS) readiness in ACL-R athletes.

Many studies have utilized either single-leg hop testing [12,50,99,102,139,140,142,260,263] or bilateral landing testing [56,65-67,130,264,265] in either ACL-R or control populations. However, only a few studies have examined both movements [144,266-269], and none of these studies have attempted to identify the test that is most relevant for assessing symmetry in ACL-R or control subjects. Several studies have reported no differences between ACL-R and control subjects in side-to-side symmetry during a single-leg hopping task in peak knee flexion [142,159] and peak vGRF [99], indicating that the single-leg hopping task may promote greater asymmetry in uninjured subjects. In addition, with all of the recent emphasis on side-to-side symmetry in athletes who have suffered an ACL injury, it is important to quantify symmetry in healthy athletes. In the past, symmetry has been assumed in healthy people [104,119,270]. However, some studies have suggested that this is an incorrect assumption [105,265,271-273]. Furthermore, since a portion of the healthy population will inevitably suffer an injury to the ACL in the future, it would follow that this portion of the population may exhibit side-to-side asymmetries in movements such as jumping and landing. It is important to quantify this side-to-side symmetry in a healthy population in order to fully understand asymmetries that exist in ACL-R populations, particularly in metrics that may be indicative of a secondary injury. Therefore, one purpose of this study is to quantify side-to-side symmetry in several secondary injury metrics in healthy recreational athletes during both single-leg and bilateral landings and assess whether or not these values can be labeled as normal symmetry. Another purpose is to compare these symmetry values to several cohorts of ACL-R patients in both single-leg and bilateral landing tasks in order to more appropriately assess the risk of secondary injury in these patients. The third purpose is to compare asymmetries seen during the single-leg and bilateral landing tasks to identify which of these tasks is the better indicator of kinematic and kinetic asymmetry. Finally, the fourth purpose is to compare asymmetries seen in healthy athletes between males and females to assess any differences based on sex in these metrics.

1.6 Specific Aims

Aim 1: To measure side-to-side symmetry in the variables of interest in bilateral jumping and landing and single-leg hop testing in healthy control subjects.

Hypothesis: Significant asymmetries in variables of interest would not exist in healthy control subjects.

60 healthy subjects with previous experience in sports involving jumping and landing performed 7 successful trials of a stop-jump task and 7 successful single hops for distance. Three-dimensional kinematics and kinetics were assessed with a 10-camera motion capture system and two multi-axis force platforms. A limb symmetry index (LSI) was calculated to quantify side-to-side symmetry in peak knee extension moment during landing, peak vertical ground reaction force (vGRF), peak knee flexion, knee flexion at ground contact, and frontal plane knee range of motion (ROM) in each landing task. Coefficients of multiple determination (CMDs) were calculated for vGRF and knee flexion angle for each task. 95% confidence intervals were also calculated for the LSIs and CMDs [105]. In addition, the average values for the dominant and non-dominant limbs for each of the variables of interest were compared using Wilcoxon Signed Rank tests. An LSI was also calculated based on these average values.

Aim 2: To compare side-to-side symmetry in the variables of interest between healthy control subjects and two cohorts of previously tested ACL-R patients.

Hypothesis: Significant differences in side-to-side symmetry would be found between ACL-R patients and healthy control subjects.

The same cohort of subjects as described in Aim 1 were compared to two historical cohorts of ACL-R patients – one of these cohorts completed stop-jump tasks and the other completed the hop testing protocol described in this study. Differences between the LSIs and magnitudes of difference between limbs as well as CMDs seen in control subjects and in ACL-R patients were investigated. The five variables of interest discussed above were compared in the stop-jump groups, while symmetry in peak vertical ground reaction forces as well as symmetry in hop distances were compared in the hop testing groups.

Aim 3: To compare the values of the variables of interest discussed in Aim 1 between the single-leg and bilateral landing tasks.

Hypothesis: Greater asymmetries would exist in the variables of interest during the single-leg hopping task.

The same cohort of subjects as described in Aim 1 was assessed as a part of this aim. The LSIs, magnitudes of differences between limbs, and CMDs for the control subjects were compared between the two landing tasks in order to assess any differences that may have existed between the two tasks.

Aim 4: To compare symmetry values for the variables of interest between males and females in the control population to assess the effects of sex on landing strategies.

Hypothesis: The female subjects would show higher levels of asymmetry than male subjects.

The data obtained in Aim 1 was used to test the Aim 4 hypothesis. The sex difference in LSIs, magnitudes of differences between limbs, and CMDs for the control subjects were determined.

2 METHODS

2.1 Subject Information

In order to test the hypotheses outlined here, 60 healthy control subjects (30 males, 30 females) were asked to complete several jumping and landing tasks in the lab. A power analysis was conducted to identify these subject numbers. Previously reported values for each variable of interest in control subjects for both single-leg and bilateral landing tasks were obtained. It was determined that a minimum of 29 subjects per group were needed to find significant differences in all of the variables in question at a power of 0.8. Since males and females have been shown to exhibit differences in landing mechanics, at least 29 subjects of each sex were required to complete the analysis. The inclusion criteria for the study mandated that each subject (1) was between the ages of 18 and 35, (2) had not previously sustained any significant lower extremity injuries (i.e. injuries that required major surgical intervention or implantation of a medical device), (3) did not suffer from chronic ankle instability, and (4) had not sustained any lower extremity injuries in the two months prior to testing. Additionally, all subjects were classified as recreational athletes, which was operationally defined as participating in sports-related activities or workouts at least three times per week. This stipulation was put in place to ensure that all subjects felt comfortable performing the movement tasks of interest and were able to complete the tasks comfortably. Prior to testing, all subjects were asked to sign informed consent that had been approved by the Institutional Review Board (IRB). Subjects also completed a MARX activity level survey prior to participating in the testing. The MARX survey is valid and reliable for comparing activity levels between patient groups in sports medicine research [324].

2.2 Testing Methods

2.2.1 Subject Preparation

Prior to testing, subjects were outfitted with a pair of athletic compression shorts and a pair of standard neutral cushioning running shoes (Air Pegasus 31, Nike Inc., Beaverton, OR). Motion capture data was collected with a 10-camera system (Qualisys

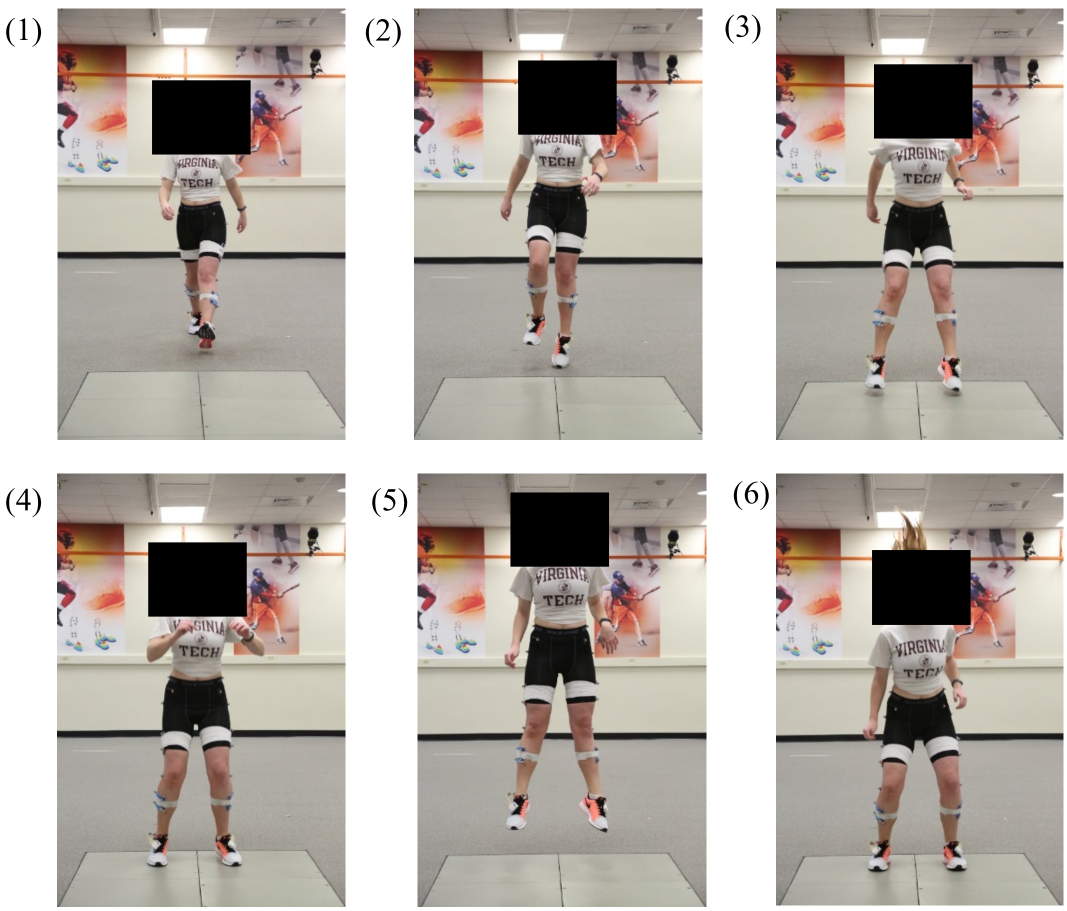
Figure 2: Marker set used during the testing.

AB, Goteborg, Sweden) (240 Hz), and force data was collected with two embedded force plates (AMTI, Watertown, MA) (1920 Hz). Prior to testing, subjects were outfitted with reflective markers placed at specific anatomic landmarks on the lower extremity. The markers were used to define and track the motion of the rearfoot, shank, thigh, and pelvis during the movement tasks. Anatomical markers were placed on the first and fifth metatarsal heads, medial and lateral malleoli, medial and lateral femoral condyles and iliac crests. These markers were used to define the proximal and/or distal ends of segments in the lower extremity. The pelvis was defined using markers placed on the right and left anterior superior iliac spines and right and left posterior superior iliac spines, with an additional marker placed between lumbar vertebrae L4 and L5. Markers placed on rigid clusters were used to track the motion of the thigh and shank. Single markers placed on the left and right greater trochanters were also used to track the motion of the thighs. The rigid clusters on the thighs were rotated towards the posterior aspect of the thigh and the clusters on the shanks were rotated towards the anterior aspect of the shank. These rotations were done to ensure a three-dimensional representation of each lower extremity segment, which allowed for the quantification of both translations and rotations of the segments. The rearfoot was tracked using three single markers placed on the superior and inferior aspects of the calcaneus and the peroneal tubercle (Figure 1). Following a standing calibration of the motion capture system, the markers on the first and fifth metatarsal heads, medial malleoli, and medial femoral condyles were removed, leaving the remainder of the markers to track the subjects' motion during the testing.

2.2.2 Bilateral Stop Jump Testing

Subjects were asked to perform seven repetitions of a stop jump task. The movement involves a running approach of several steps, a one-legged vertical takeoff, a bilateral landing with each foot on a separate force plate, a bilateral vertical jump, and a bilateral landing (Figure 2). Subjects were instructed to move as fast as they were safely able to and desired during the approach and jump as high as they were safely able to and desired during the jump phase. Specific instructions regarding the level of effort required for the test was not given. Motion capture and force plate data were collected simultaneously during each of the seven repetitions.

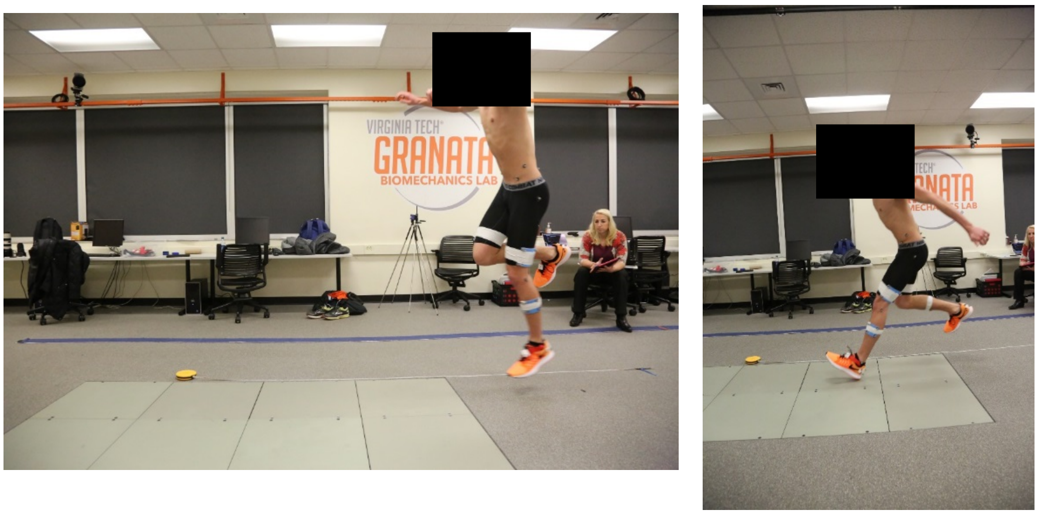
Figure 3: Visual description of stop jump task. (1) Several step approach, (2) lift off from single leg, (3) and (4) bilateral landing with one foot on each force plate, (5) bilateral lift off straight into the air, and (6) bilateral landing.



2.2.3 Single-Leg Single Hop Testing

After the completion of the stop jump tasks, subjects then completed seven repetitions of a single-leg single hop for distance with each limb (Figure 3). The subjects completed the seven hopping trials on their nondominant limb before performing seven

Figure 4: Visual description of single-leg hopping task. Subject lifts off with one foot, hops a maximal distance forward, and lands on the same foot.



trials on their dominant limb. Limb dominance was defined as the limb that the subject preferred to use to kick a soccer ball. To determine this, a soccer ball was placed in front of the subject and the subject was simply asked to kick it. The foot used to kick the ball was determined to be that subject's dominant limb. Subjects performed one to three practice trials on each limb until they were comfortable with the movement before beginning the recorded trials. In order to minimize fatigue, a rest period of up to 30 seconds between repetitions was provided. Restrictions were not placed on arm movement during hop testing and additional instruction on how to complete the movement was not provided. In order for a hop to be defined as successful, the subjects were required to stand on the hopping leg for two seconds without touching the ground with the other foot, touching the ground with an upper extremity, or generally losing balance. If a trial was deemed unsuccessful, that trial was repeated. Subjects hopped forward for a maximal distance and landed with their foot fully on a force plate. In order to achieve a maximum hop distance while still landing on the force plate, each subject's general hop distance was assessed during the practice trials. The subjects' starting point was then placed such that a hop of average distance would result in a correct landing on a force plate. Subjects were instructed to hop for maximum distance while maintaining a balanced single-leg stance at the conclusion of the trial. A trial was considered to be successful if the subject's foot landed fully on the force plate, in addition to the maintenance of balance as described earlier. Trials were repeated until seven successful trials were collected. Hop distance was measured to the furthest landing point of the great toe and recorded to the nearest centimeter. Data was again recorded simultaneously by the motion capture system and the force plate during the single hop for distance.

2.2.4 Data Collection in Previous ACL-R Cohorts

Kinematic and kinetic data was previously collected in a cohort of 20 ACL-R athletes (7 males, 13 females; 6.2 ± 0.2 months after ACL-R). All athletes in the cohort participated in high school or collegiate athletics and intended on returning to a sport that involved jumping and cutting. All subjects signed informed consent prior to any testing procedures were completed. Subjects were provided with a pair of athletic compression shorts and a standard pair of neutral cushioning running shoes (Air Pegasus, Nike Inc., Beaverton, OR) for the testing. 46 reflective markers were placed at specific anatomic landmarks on the lower extremities. Three dimensional coordinate data was collected with an 8-camera motion capture system (Motion Analysis Corporation, Santa Rosa, CA) (120 Hz) and force data was collected with two embedded force plates (AMTI, Watertown, MA) (2400 Hz). Each subject performed five trials of the stop jump task described earlier.

Loading and performance data was previously collected in a cohort of 30 ACL-R subjects (9 males, 21 females, 6.8 ± 1.2 months after ACL-R). All subjects signed either informed consent or informed assent prior to testing. Each subject also completed a MARX activity level survey prior to testing. Subjects were provided with a standard pair of neutral cushioning running shoes (Air Pegasus 31, Nike Inc., Beaverton, OR) for the testing. A pair of single-sensor force-sensing insoles (pedoped, Novel Electronics, St. Paul, MN) (100 Hz) was used to measure the normal force between each subject's feet and shoes during the testing. Subjects performed single-leg single hops for distance as described earlier. Each subject performed two of these hops with the non-surgical limb followed by two hops with the surgical limb. Load data was collected and stored on an iPad via Bluetooth. Hop distance was also measured with a standard tape measure affixed to the floor.

2.3 Analysis

2.3.1 Data Processing

The three-dimensional motion capture coordinate data and tri-axial force plate data was collected in Qualisys Track Manager (Qualisys AB, Goteborg, Sweden). Manual tracking of the markers was performed in this program to ensure proper identification of all markers in the three-dimensional coordinate space for each trial. The coordinate data and force data collected during this project as well as the coordinate data and force data from the previous ACL-R cohort was exported to Visual3D (C-Motion, Inc., Germantown, MD) for further analysis. The three-dimensional data was filtered with a 12 Hz low-pass Butterworth digital filter, and the force data was filtered with this same filter at 100 Hz. Cardan angles were calculated between adjacent segments, with the order of rotation consisting of flexion/extension in the x direction, followed by abduction/adduction in the y direction, and concluding with internal/external rotation in the z direction. Joint moments were calculated through inverse dynamics based on the joint angles and the ground reaction forces. All ground reaction forces were normalized to each subjects' body mass, and all joint moments were normalized to each subjects' body mass and height. The time of the stop-jump trials was normalized to the length of the stance phase, or the time that the subjects' feet were in contact with the force plates during the initial landing of the task. The time of the single-leg hopping task was normalized to the length of the landing phase, or the time from the initial contact of the foot with the force plate to the time of peak knee flexion. In both cases, initial foot contact was labeled as 0% of the task and the end of the task was labeled as 100%. A 10N threshold was used to determine the times of initial foot contact and final toe-off. In the previous ACL-R cohort that performed the single-leg hopping task, the load data was exported and run through a custom written Matlab program (Mathworks, Natick, MA, USA) to extract the peak loads occurring in each trial.

2.3.2 Data Analysis

In order to investigate side-to-side symmetry in these landing tasks, limb symmetry indices (LSIs) and coefficients of multiple determination were calculated for each subject.

All calculations were performed using another custom written Matlab program. LSIs were calculated according to the following formula:

$$\text{LSI} = \left|\frac{X_D - X_{ND}}{0.5*(X_D + X_{ND})}\right| * 100\%, \qquad (5)$$

where X_D and X_{ND} denote data for the dominant and nondominant limbs, respectively. This metric was used to yield information regarding the symmetry seen between the subjects' limbs at specific time points, or discrete points. An LSI of 0% in this metric would indicate perfect symmetry between the two limbs. One LSI for each variable in each landing task was calculated for each subject from each subject's mean data across all trials. Additionally, coefficients of multiple determination (CMDs) were calculated according to the following formula:

$$CMD = 1 - \frac{\sum_{i=1}^{N}\sum_{t=0}^{T}(y_{it} - \bar{y}_t)^2/[(T+1)(N-1)]}{\sum_{i=1}^{N}\sum_{t=0}^{T}(y_{it} - \bar{y})^2/[(T+1)N-1]} \qquad (6)$$

where N = the number of curves to be compared, T = the number of time points making up each curve, y_{it} = each signal, $\bar{y}_t$ = the mean of the signals at each time point, and $\bar{y}$ = the mean of all signals [125]. This metric was utilized to give information regarding the symmetry seen in the two limbs across the entire landing phase of both the stop jump task and the single-leg single hop task. A CMD of 1 in this metric would indicate identical waveforms and thus perfect symmetry between the two limbs. One CMD for each variable in each landing task was calculated for each subject from each subject's mean time-series data across all trials.

For each of the landing tasks, discrete variables to be considered included peak knee extension moment during the first 10% of the stance phase [139,162], peak vGRF [100-102,132,142,280], knee flexion at ground contact [56,100], peak knee flexion [12], and frontal plane knee range of motion (ROM) [162]. LSIs were calculated for each of these variables according to Equation 6 in the control population described here for both the stop jump and single-leg hop tasks, along with an LSI for hop distance in the single-leg hop tasks. In addition, LSIs were calculated for the same variables in the previous cohort of ACL-R patients that performed the stop jump task, and LSIs for peak vGRF and hop distance were calculated for the group of ACL-R patients that performed the single-leg hopping task.

2.3.3 Statistics

All statistical analysis was performed in JMP Pro 12 (SAS, Cary, NC). A significance level of α=0.05 was set for all statistical comparisons. Wilcoxon Signed Rank tests are the nonparametric equivalent of paired t-tests, while Wilcoxon tests are the nonparametric equivalent of independent samples t-tests.

Statistics for Aim 1: In the control group, a mean, standard deviation, and 95% confidence interval were calculated for the LSIs and CMDs in each variable in each landing task. Additionally, the values of the dominant and nondominant limbs of each subject in the control group were compared with Wilcoxon Signed Rank tests to assess statistically significant differences between the two limbs. An LSI was also calculated based on the average values of the two limbs across the entire group.

Statistics for Aim 2: To assess differences in symmetry in each variable of interest between ACL-R and control subjects, the LSIs and CMDs for both the single-leg hop and stop jump tasks were compared between the two groups. A parametric ANCOVA was used to assess these differences, with age being introduced as a covariate. The variable sets were transformed with Box-Cox Power Transformations to make them normally distributed, thus allowing for the parametric ANCOVA to be run. The magnitudes of the differences between limbs in each variable were also compared between the groups in the same fashion. In addition, the values of each variable in the nondominant limbs of the control subjects were compared to the values of the variable in the nonsurgical limbs of the ACL subjects, and the values of each variable in the dominant limbs of the control subjects were compared to the values of the variable in the surgical limbs of the ACL subjects. These comparisons were performed with parametric ANCOVAs with age used as a covariate. These variable sets were also transformed with Box-Cox Power Transformations to create normallly distributed sets.

Statistics for Aim 3: LSIs and CMDs for the single-leg hop and the stop jump task in the control group were compared with Wilcoxon Signed Rank tests to assess differences between the two tasks in control subjects. Additionally, the magnitudes of the differences between limbs in each variable were also compared between the two tasks with Wilcoxon Signed Rank tests. The values of each variable in the nondominant limb were also compared between the two tasks with Wilcoxon Signed Rank tests, and the values of each variable in the dominant limb were compared in the same fashion.

Statistics for Aim 4: Wilcoxon tests were conducted to compare LSIs between males and females in the control group for each variable, with separate comparisons being performed for each landing task. The magnitudes of the differences between limbs in each variable were also compared between the two sexes with the same tests. In addition, the values of each variable in the nondominant limb were compared between the two sexes with Wilcoxon tests, and the values of each variable in the dominant limb were compared in the same fashion. Finally, the values of the dominant and nondominant limbs were compared in each sex and in each landing task with Wilcoxon Signed Rank tests.

3 RESULTS

3.1 Subject Demographics

Subject demographics are listed in Tables 1a, 1b, and 1c. Wilcoxon tests revealed that the mean age of the control cohort was significantly higher than the ACL cohort that performed the bilateral landing task ($p<0.001$) (Table 1a) and the ACL cohort that performed the single-leg hopping task ($p<0.001$) (Table 1b). The control cohort was also significantly taller than the bilateral landing ACL cohort ($p=0.012$) (Table 1a). In the the male subjects (Table 1c).

Table 1a: Demographics Comparisons Between Control Group and Bilateral Landing ACL Group

	Control Group (n=60)	ACL Bilateral Group (n=20)	p-value
Age	21.6±2.9	15.8±1.2	**<0.001**
Height (m)	1.75±0.08	1.69±0.10	**0.012**
Mass (kg)	69.0±10.3	71.7±16.8	0.820

Table 1b: Demographics Comparisons Between Control Group and Single-Leg Landing ACL Group

	Control Group (n=60)	ACL Bilateral Group (n=20)	p-value
Age	21.6±2.9	19.4±4.2	**<0.001**
Height (m)	1.75±0.08	1.73±0.07	0.342
Mass (kg)	69.0±10.3	72.4±13.5	0.336

Table 1c: Demographics Comparisons Between Male Control Group and Female Control Group

	Male Control Group (n=30)	Female Control Group (n=30)	p-value
Age	21.2±2.6	22.0±3.2	0.436
Height (m)	1.81±0.07	1.69±0.05	**<0.001**
Mass (kg)	75.8±8.3	62.2±7.2	**<0.001**

Table 2: MARX Activity Level Survey Results Between Control Group and Single-Leg Hop ACL Group

	ACL (n=30)	Control (n=60)	p-value
Running	3.73±0.58	3.34±0.96	**0.029**
Cutting	3.57±0.86	2.02±1.61	**<0.001**
Deceleration	3.60±0.67	2.29±1.55	**<0.001**
Pivoting	3.53±0.78	2.08±1.59	**<0.001**

Table 2 shows the results of the MARX activity level survey in the control group as well as in the ACL-R group that completed the single-leg hopping tasks. The ACL-R group had increased activity levels in all four tasks, thus showing that the ACL-R group was more regularly physically active than the control group.

3.2 Results of Aim 1

LSI and CMD values and 95% confidence intervals in the control subjects in each variable and each task is shown in Table 3 as well as in Figure 4. Paired comparisons between the dominant and nondominant limbs of each subject in the control group in each metric are shown in Table 4. While some of the LSI values displayed in Table 3 and Figure 4 are higher than the standard 10-15% benchmark for normal movement symmetry, Wilcoxon Signed Rank tests revealed that only hop distance was significantly different between limbs in the control group (p=0.006) (Table 4). No other variables showed significant differences between limbs. Table 4 also includes LSI values calculated from the average values of the dominant and nondominant limbs across all control subjects. Additionally, Figure 5 displays representative graphs to illustrate example CMD calculations in the control group. The graphs are meant to give context to the average CMD values displayed in Table 3.

Table 3: Means and 95% Confidence Intervals for LSIs and CMDs in the Control Group During Both Tasks

		Stop Jump		Single Leg Hop	
		Mean	**95% CI**	**Mean**	**95% CI**
Limb Symmetry Index (%)	**Peak Knee Ext. Moment During Landing (%)**	28.2	23.5, 33.0	32.7	20.7, 44.7
	Frontal Plane ROM (%)	52.4	43.3, 61.4	42.2	33.2, 51.1
	Peak Knee Flexion (%)	2.1	1.6, 2.5	9.9	7.9, 11.9
	Knee Flexion at GC (%)	19.8	15.2, 24.4	43.9	34.5, 53.2
	Peak vGRF (%)	17.4	13.5, 21.3	10.3	8.0, 12.6
	Hop Distance (%)	--	--	8.5	6.5, 10.4
CMD	**vGRF**	0.824	0.779, 0.868	0.843	0.802, 0.884
	Knee Flexion	0.987	0.985, 0.990	0.930	0.909, 0.952

Figure 5: LSIs and 95% Confidence Intervals in the Control Group During Both Tasks

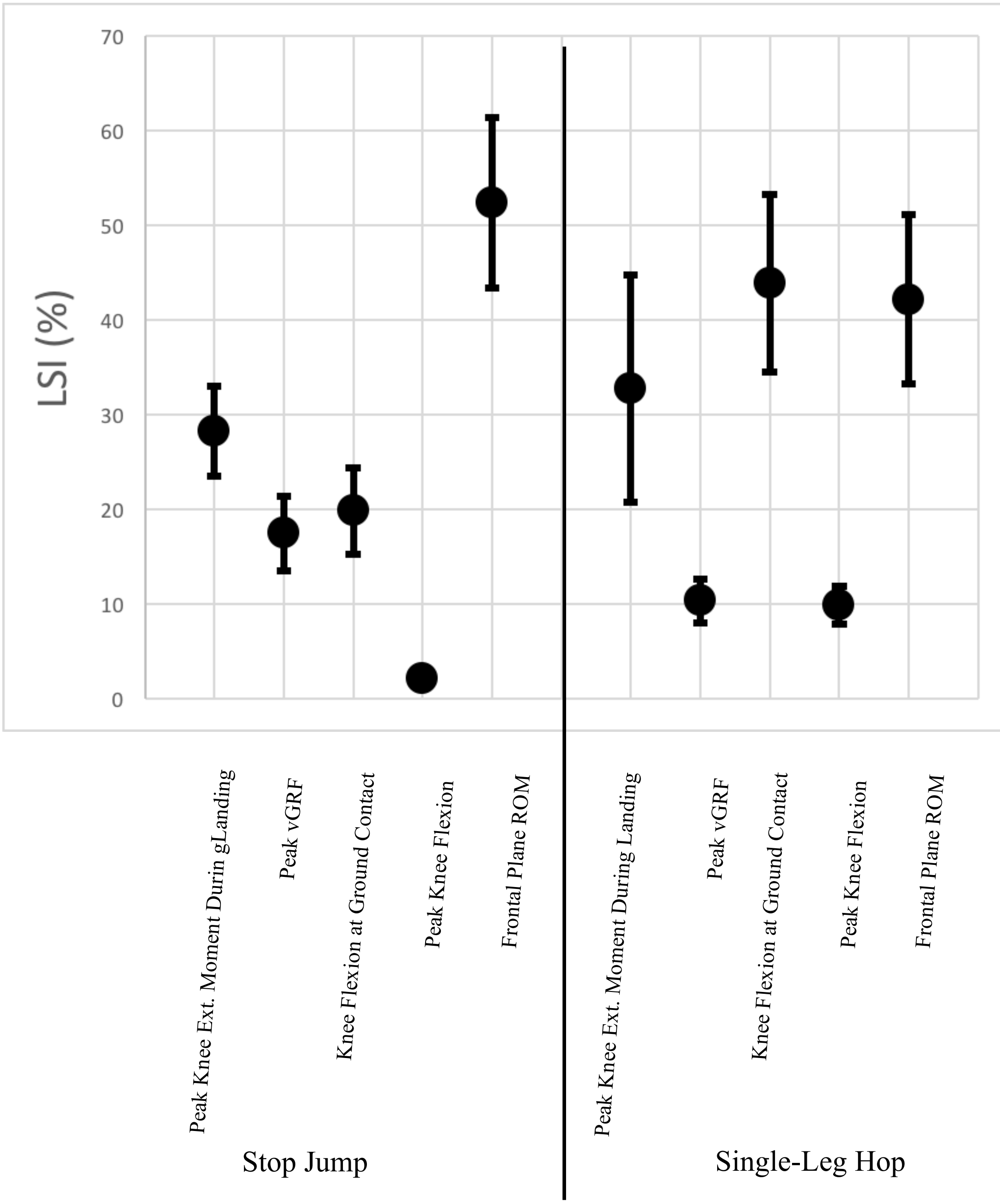
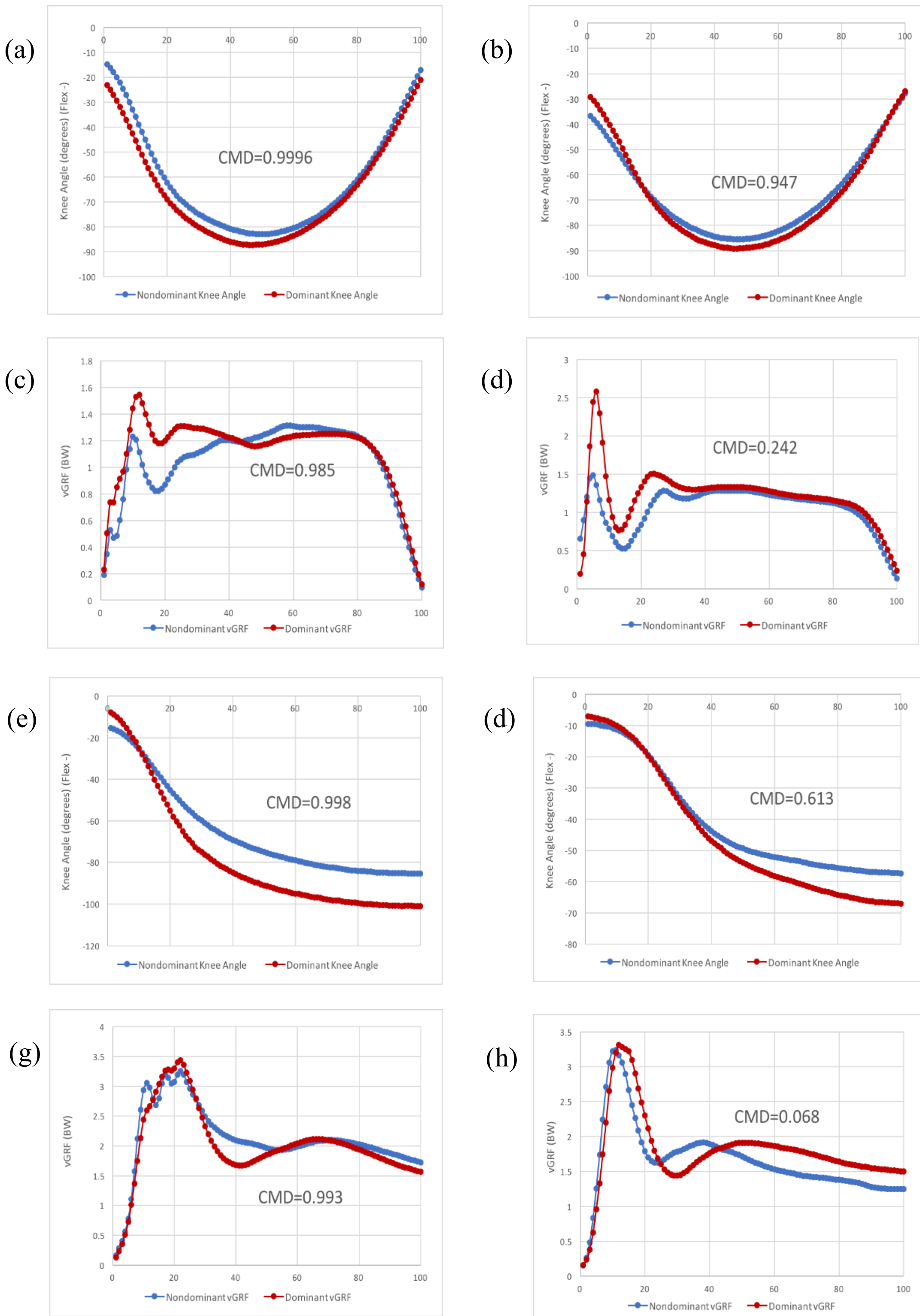


Table 4: Direct Comparisons Between Limbs in Control Subjects

		Mean Value in ND Limb	Mean Value in D Limb	p-value	Mean Values LSI (%)
Stop Jump	**Peak Knee Ext. Moment During Landing (BW*BH)**	0.08±0.03	0.08±0.03	0.684	1.13
	Peak vGRF (BW)	1.66±0.49	1.71±0.45	0.794	2.88
	Knee Flexion at Ground Contact (degrees)	24.20±7.72	23.00±7.56	0.053	5.11
	Peak Knee Flexion (degrees)	86.05±15.99	86.31±16.12	0.362	0.30
	Frontal Plane ROM (degrees)	6.59±4.38	6.60±4.17	0.879	0.17
Single-Leg Hop	**Peak Knee Ext. Moment During Landing (BW*BH)**	0.13±0.06	0.12±0.06	0.314	4.08
	Peak vGRF (BW)	3.38±0.49	3.33±0.51	0.183	1.45
	Knee Flexion at Ground Contact (degrees)	8.74±4.98	8.40±5.35	0.484	3.91
	Peak Knee Flexion (degrees)	66.52±9.82	65.56±9.73	0.264	1.44
	Frontal Plane ROM (degrees)	11.74±6.44	10.54±6.16	0.198	10.83
	Hop Distance (cm)	150.99±39.51	156.89±36.07	**0.006**	3.83

Figure 6: Representative control graphs of time-series data with corresponding CMD values. Graphs on the left showed highest CMD of any subject in that particular metric, and graphs on the right showed lowest CMD of any subject in that metric. (a) and (b) knee flexion during stop-jump, (c) and (d) vGRF during stop-jump, (e) and (f) knee flexion during single-leg hop, (g) and (h) vGRF during single-leg hop.



3.3 Results of Aim 2

The comparisons of landing characteristics between the control subjects and ACL subjects during both tasks and between tasks within the control cohort are found in Table 5. Lower LSI values are indicative of improved symmetry over higher LSI values, and higher CMD values are indicative of improved symmetry over lower CMD values. During

Table 5: LSI and CMD Comparisons between ACL-R and Control Groups for Both Landing Tasks

			ACL (n=20)	Control (n=60)	p-value
Limb Symmetry Index (%)	**Stop Jump**	**Peak Knee Ext. Moment During Landing (%)**	72.9 ± 57.3	28.2 ± 18.4	**<0.001**
		Peak vGRF (%)	22.7 ± 14.6	17.4 ± 15.2	**0.021**
		Knee Flexion at GC (%)	32.3 ± 29.0	19.8 ± 17.8	0.065
		Peak Knee Flexion (%)	6.0 ± 3.6	2.1 ± 1.7	**<0.001**
		Frontal Plane ROM (%)	30.3 ± 17.2	52.4 ± 35.0	0.064
	Single-Leg Hop	**Peak vGRF (%)**	15.6 ± 11.5	7.7 ± 7.5	**<0.001**
		Hop Distance (%)	23.0 ± 15.9	8.5 ± 7.6	**<0.001**
CMD	**Stop Jump**	**vGRF**	0.656 ± 0.272	0.824 ± 0.173	**0.010**
		Knee Flexion	0.951 ± 0.050	0.987 ± 0.011	**<0.001**

the stop-jump task, the ACL subjects exhibited significantly higher LSI values in peak knee flexion ($p<0.001$), peak knee extension moment during landing ($p<0.001$), and peak vGRF ($p=0.021$), and significantly lower CMD values in vGRF ($p=0.010$) and in knee flexion ($p<0.001$). Differences between groups in LSIs in knee flexion at ground contact ($p=0.065$) and frontal plane ROM ($p=0.064$) were not significant. During the single-leg hopping task, the ACL subjects showed higher LSI values in peak vGRF ($p<0.001$) and in hop distance ($p<0.001$). All of these analyses were performed with age as a covariate. Age was shown to significantly affect symmetry in peak vGRF during the single-leg hop task ($p=0.031$) as well as CMD values for both vGRF ($p=0.012$) and knee flexion angle ($p=0.004$), but did not significantly affect symmetry in any of the other variables studied. An independent samples t-test comparing these variables in the older half of the subject group to those in the younger half of the subject group revealed that younger subjects exhibited lower LSI values in peak vGRF during the single-leg hop task and higher CMD values in both variables during the stop jump task.

The comparisons between the magnitudes of the differences between limbs when comparing the ACL and control groups are found in Table 6. Greater magnitudes of differences between limbs were seen in the ACL-R group in peak knee flexion during the stop jump task ($p<0.001$) and in peak vGRF ($p=0.012$) and hop distance ($p<0.001$) during the single-leg hopping task. No other differences in the magnitudes of difference between limbs were observed. In addition, age was shown to significantly affect the magnitude of the difference between limbs in knee flexion at ground contact ($p=0.034$) and frontal plane ROM ($p=0.037$) during the stop jump task and peak vGRF ($p=0.040$) during the single-leg hop task. The same independent samples t-test mentioned above showed that younger subjects exhibited higher magnitudes of differences between limbs than older subjects in knee flexion at ground contact during the stop jump task and in peak vGRF during the single-leg hop task, and they exhibited lower magnitudes of differences between limbs in frontal plane ROM during the stop jump task. The results show that the two symmetry metrics are in agreement for the variables studied during the single-leg hopping task, but little agreement exists for the variables studied during the bilateral landing task.

Table 6: Magnitudes of Differences Between Limbs in ACL and Control groups

			ACL (n=20)	Control (n=60)	p-value
Magnitues of Differences Between Limbs	**Stop Jump**	**Peak Knee Ext. Moment During Landing (BW*BH)**	0.03 ± 0.02	0.02 ± 0.01	0.078
		Peak vGRF (BW)	0.41 ± 0.29	0.28 ± 0.26	0.362
		Knee Flexion at GC (degrees)	6.40 ± 6.23	4.21 ± 3.47	0.674
		Peak Knee Flexion (degrees)	4.40 ± 2.60	1.80 ± 1.51	**<0.001**
		Frontal Plane ROM (degrees)	2.30 ± 1.61	3.30 ± 2.62	0.596
	Single-Leg Hop	**Peak vGRF (BW)**	0.43 ± 0.36	0.23 ± 0.21	**0.012**
		Hop Distance (cm)	32.45 ± 22.29	12.30 ± 10.53	**<0.001**

3.4 Results of Aim 3

Aim 3 was conducted to identify any differences in side-to-side symmetry between the two landing tasks in the control population. Comparisons between LSIs calculated during the stop jump and single-leg hop tests in the control group are shown in Table 7. The single-leg hopping task resulted in higher LSI values in knee flexion at ground contact ($p<0.001$) and in peak knee flexion ($p<0.001$). CMD values in knee flexion were lower ($p<0.001$) during the single-leg hopping task. Significantly lower LSI values in peak vGRF ($p=0.009$) during the single-leg hop task were also identified. Differences in LSIs in frontal plane ROM ($p=0.085$) and peak knee extension moment during landing ($p=0.346$) as well as in CMDs in vGRF ($p=0.779$) were not significant. These results show that asymmetries were mixed between the two tasks and that one landing task did not promote more asymmetry than the other.

Table 7: LSI and CMD Comparisons Between Stop Jump and Single-Leg Hop tasks in Control Group

		Stop Jump (n=60)	Single Leg Hop (n=60)	p-value
Limb Symmetry Index (%)	Peak Knee Ext. Moment During Landing (%)	28.2 ± 18.4	32.7 ± 46.5	0.346
	Peak vGRF (%)	17.4 ± 15.2	10.3 ± 8.9	**0.009**
	Knee Flexion at GC (%)	19.8 ± 17.8	43.9 ± 34.9	**<0.001**
	Peak Knee Flexion (%)	2.1 ± 1.7	9.9 ± 7.7	**<0.001**
	Frontal Plane ROM (%)	52.4 ± 34.9	42.1 ± 34.6	0.085
CMD	vGRF	0.824 ± 0.173	0.843 ± 0.158	0.779
	Knee Flexion	0.987 ± 0.011	0.930 ± 0.082	**<0.001**

The comparisons of the magnitudes of the differences between limbs and between tasks (stop jump and single-leg hop) in the control group are found in Table 8.

Greater magnitudes of differences between limbs were seen in the single-leg hop in peak knee flexion ($p<0.001$) and in peak knee extension moment during landing ($p=0.031$), but no other differences in the magnitudes of difference between limbs were seen. These results show little agreement between the LSIs and the magnitudes of diifferences between limbs in the determination of symmetry when comparing tasks, as only peak knee flexion and frontal plane ROM showed the same differences between groups in each symmetry metric.

Table 8: Magnitudes of Differences Between Limbs in Stop Jump and Single-Leg Hop Tasks in Control Group

		Stop Jump	Single-Leg Hop	p-value
Magnitues of Differences Between Limbs	**Peak Knee Ext. Moment During Landing (BW*BH)**	0.02 ± 0.01	0.03 ± 0.02	**0.031**
	Peak vGRF (BW)	0.29 ± 0.26	0.34 ± 0.29	0.288
	Knee Flexion at GC (degrees)	4.21 ± 3.47	3.67 ± 2.93	0.467
	Peak Knee Flexion (degrees)	1.80 ± 1.51	6.53 ± 5.28	**<0.001**
	Frontal Plane ROM (degrees)	3.30 ± 2.62	4.35 ± 4.09	0.302

3.5 Results of Aim 4

Aim 4 was conducted to explore sex differences in side-to-side symmetry in the variables of interest in both tasks. Comparisons between sexes in the control group are displayed in Tables 9 and 10. The data shows that females exhibited higher LSI values in knee flexion at ground contact during the stop-jump task (p=0.028) (Table 9) and showed higher CMD values in knee flexion during the single-leg hopping task (p=0.042) (Table 10). No other significant differences between sexes were observed.

Table 9: Comparisons between sexes during the stop jump task in the control group

		Male (n=30)	Female (n=30)	p-value
Limb Symmetry Index (%)	Peak Knee Ext. Moment During Landing (%)	26.2 ± 21.6	30.2 ± 14.6	0.172
	Peak vGRF (%)	17.3 ± 15.2	15.6 ± 11.5	0.900
	Knee Flexion at GC (%)	14.7 ± 14.3	24.8 ± 19.6	**0.028**
	Peak Knee Flexion (%)	2.2 ± 1.7	1.9 ± 1.7	0.395
	Frontal Plane ROM (%)	58.4 ± 37.1	46.3 ± 32.1	0.246
CMD	vGRF	0.797 ± 0.198	0.850 ± 0.142	0.363
	Knee Flexion	0.988 ± 0.010	0.987 ± 0.011	0.610

Table 10: LSI and CMD Comparisons Between Sexes During the Single-Leg Hop Task in the Control Group

		Male (n=30)	Female (n=30)	p-value
Limb Symmetry Index (%)	**Peak Knee Ext. Moment During Landing (%)**	27.6 ± 24.5	37.7 ± 61.3	0.728
	Peak vGRF (%)	10.6 ± 9.4	9.9 ± 8.5	0.819
	Knee Flexion at GC (%)	39.6 ± 29.3	47.8 ± 39.6	0.682
	Peak Knee Flexion (%)	12.1 ± 8.9	12.2 ± 9.0	0.058
	Frontal Plane ROM (%)	45.6 ± 36.5	38.7 ± 32.8	0.473
	Hop Distance (%)	8.1 ± 7.7	8.8 ± 7.5	0.587
CMD	**vGRF**	0.832 ± 0.181	0.855 ± 0.134	0.728
	Knee Flexion	0.903 ± 0.104	0.958 ± 0.037	**0.042**

The comparisons between the magnitudes of the differences between limbs between sexes in the control group are found in Tables 11 and 12. No differences between magnitudes of difference between limbs were seen in any of the variables during either of the landing tasks.

Table 11: Comparisons of the Magnitudes of Differences Between Limbs Between Sexes During the Stop Jump Task in the Control Group

		Male (n=30)	**Female (n=30)**	**p-value**
Magnitudes of Differences Between Limbs	**Peak Knee Ext. Moment During Landing (BW*BH)**	0.02 ± 0.02	0.02 ± 0.01	0.959
	Peak vGRF (BW)	0.32 ± 0.28	0.25 ± 0.23	0.333
	Knee Flexion at GC (degrees)	5.11 ±3.98	3.31 ± 2.64	0.077
	Peak Knee Flexion (degrees)	1.59 ± 1.52	2.02 ± 1.50	0.201
	Frontal Plane ROM (degrees)	3.76 ± 2.74	2.84 ± 2.46	0.176

Table 12: Comparisons of the Magnitudes of Differences Between Limbs Between Sexes During the Single-Leg Hop Task in the Control Group

		Male (n=30)	Female (n=30)	p-value
Magnitudes of Differences Between Limbs	**Peak Knee Ext. Moment During Landing (BW*BH)**	0.04 ± 0.03	0.03 ± 0.02	0.137
	Peak vGRF (BW)	0.27 ± 0.26	0.41 ± 0.32	0.056
	Knee Flexion at GC (degrees)	3.77 ± 3.58	2.34 ± 3.45	0.326
	Peak Knee Flexion (degrees)	5.67 ± 4.34	7.39 ± 6.03	0.363
	Frontal Plane ROM (degrees)	4.12 ± 4.39	4.57 ± 3.83	0.446
	Hop Distance (cm)	13.73 ± 12.47	10.87 ± 8.10	0.529

In addition to these results, kinematic and kinetic values can be found in Appendix A at the conclusion of the document.

4 DISCUSSION

This study had multiple purposes. The first was to assess levels of symmetry in healthy recreational athletes in kinematic and kinetic parameters normally associated with secondary ACL injury. We also wanted to compare these levels of symmetry to levels seen in several separate cohorts of ACL-reconstructed (ACL-R) patients. We wanted to compare kinematic and kinetic parameters seen during single-leg and bilateral tasks in healthy recreational athletes. We also wanted to compare these parameters between male and female healthy recreational athletes. Each of these purposes shared the common goal of assessing tools that are commonly used to measure rehabilitation following ACL reconstruction.

4.1 Discussion of Aim 1

The first portion of the project examined symmetry in movement mechanics in the control population to determine normal levels of symmetry in healthy control subjects. The upper limit for "correct" movement symmetry has been quantified between 10-15% for hop distance [160,284,286], peak vertical ground reaction force (vGRF) during a bilateral landing [160], and quadriceps strength [285,286], with some researchers even claiming that an LSI of 0% (perfect symmetry) in muscle strength is required for return to sports involving cutting and pivoting [257]. As shown in Table 3 and Figure 4, many of the variables explored in this study exhibited higher side-to-side asymmetry values than this 10-15% benchmark, including peak vGRF during the stop jump task and knee extension moment at ground contact, knee flexion at ground contact, and frontal plane knee range of motion (ROM) in the stop-jump and single-leg hop tasks. However, the paired Wilcoxon Signed Rank tests revealed that only hop distance measured during the single-leg hopping task showed a statistical difference between limbs in the control population. Therefore, our original hypothesis was proven to be incorrect, as no statistical differences existed between limbs in the original variables of interest.

The asymmetry values reported here agree with previous literature that has reported significant asymmetries in healthy control subjects in knee extension moment at ground contact [287], knee flexion at ground contact [272], and peak frontal plane angles [288] during jumping and landing activities. The control subjects in the present study showed low LSIs in peak vGRF during the single-leg hop task and in peak knee flexion during both tasks. This agrees with previous work by Holsgaard-Larsen et al. that reported normal symmetry in peak knee flexion in healthy controls during both types of landing [159], as well as with work by Van der Harst et al. that reported peak knee flexion values for each limb that would equate to an LSI of 1.67% in healthy control subjects during a single-leg landing task [141]. Using this method to quantify symmetry, the subjects in the present study displayed peak knee flexion values that equated to an LSI of 1.44%.

The concept of control subjects possessing movement and loading asymmetries is in agreement with previous work that examined movement mechanics in walking. Herzog et al. and Robinson et al. found asymmetries in ground reaction forces during normal human gait [105,113], and Gunderson et al. reported side-to-side differences in maximum knee extension as well as several spatiotemporal parameters during walking in healthy control subjects [271]. Additionally, Ferber et al. showed that healthy subjects demonstrated asymmetries in hip work during normal walking [273]. These studies along with the present one show that movement mechanics in healthy control subjects may not be perfectly symmetric, as has been assumed in previous work [104,119,270]. As such, asymmetries seen during rehabilitation of ACL-R athletes may not be indicative of incomplete rehabilitation, and it may be inappropriate to assume that these asymmetries are entirely caused by the ACL injury and surgery. However, since it has been clearly shown that asymmetries in several of these metrics are risk factors for secondary ACL injury [162], it is important that ACL-R athletes continue to take steps to achieve improved movement symmetry to reduce the likelihood of further injuries to the ACL graft or to the contralateral ACL.

Table 4 displays the mean values in each limb across the entire group and an LSI based on these mean values in each variable of interest. The data demonstrates that despite fairly large average LSI values for some of the variables, the only statistical difference between the average values for each limb was in hop distance. Additionally, the LSIs calculated from the mean values in each limb were much lower than the average LSIs across the subjects. These results indicate that these varying methods of assessing symmetry do not provide the same information. Studies in the past have utilized the method of comparing the mean values of each limb in order to assess side-to-side symmetry in one or more groups [41,102,130,132,133,141,142,143,294]. This method allows for an understanding of movement patterns seen across an entire group. However, LSIs have also been commonly used to quantify symmetry [99,105,110,111,114,134,159], and some researchers have reported differences in side-to-side symmetry between ACL-R patients and healthy control subjects using this metric [99,134,159]. This proves that the calculation of a LSI for each subject can be relevant for the assessment of healthy and injured athletes, and thus can be used to quantify side-to-side symmetry in a control population. Furthermore, the calculation of a LSI, either through the metric utilized in this study or through a simple ratio, can be easily implemented in a clinical setting and can give physicians instant feedback regarding a patient's rehabilitation.

Table 3 also displays the coefficients of multiple determination (CMDs) measured in the control subjects during both tasks. Since no studies have utilized this metric to assess side-to-side symmetry during a landing task, the true relevance of these CMD values is unknown. However, we used effect sizes previously determined in the literature to give context to these CMD values [289,290]. Cohen asserted that an effect size of 0.1 or lower should be deemed a weak effect, while an effect size of 0.3 represents a moderate effect and an effect size of 0.5 or higher is a large effect [290]. These have been updated to further dichotomize large effects, with an effect size of 0.35 or less representing weak correlations, 0.36-0.67 representing moderate correlations, 0.68-0.9 representing strong correlations, and 0.90-1.0 representing very strong correlations [289]. While there is no statistical significance connected with these scores, they can give an idea of the similarities between two waveforms, or in our case, the symmetry between limbs. Using these correlation sizes, the CMDs for peak vGRF across the landing phase of both tasks can be labeled as showing strong correlations between limbs, and the CMDs for knee flexion angles across the landing phase of both tasks can be labeled as showing very strong correlations between limbs. These indicate that the healthy control subjects showed movement and loading patterns that were fairly symmetrical, but this cannot be verified statistically from this data.

4.2 Discussion of Aim 2

The second portion of the study aimed to compare the levels of symmetry seen in the healthy control subjects in this study to levels of symmetry found in two previous cohorts of ACL-R patients. Table 5 shows that ACL-R patients possessed larger asymmetries in peak knee extension moment during the initial landing phase, peak vertical ground reaction force, and peak knee flexion during the bilateral landing task and in peak vGRF and hop distance during the single-leg landing task. Therefore, our original hypothesis for this aim was supported, as the ACL-R groups displayed larger asymmetries than the control group in all but one of the variables of interest.

The peak knee flexion data is in agreement with previous work that has reported greater asymmetries in sagittal knee ROM in a group of ACL-R patients than in healthy control subjects during bilateral landing tasks [159]. The larger asymmetries seen in ACL-R patients in peak vGRF during each landing task are in agreement with Paterno et al., who reported a larger asymmetry in ACL-R patients during a bilateral landing task [130], and with Myer et al., who reported a larger asymmetry in ACL-R patients during a single-leg landing task [134]. Paterno et al. reported vGRF values of 2.0 body weights (BW) in the uninvolved limb and 1.5 BW in the involved limb of the ACL-R patients as well as values of 1.6 BW in the nondominant limb and 1.5 BW in the dominant limb of healthy control subjects [130]. In the present study, ACL-R subjects displayed vGRF values of 1.98 BW and 1.59 BW in their uninvolved and involved limbs, respectively, and control subjects displayed vGRF values of 1.66 BW and 1.71 BW in their nondominant and dominant limbs, respectively, during the bilateral landing task. Myer et al. reported limb symmetry ratios of 95% in an ACL-R group and 102% in a healthy control group during a single-leg landing task [134], which are similar to the ratios of 90.56% in the ACL-R group and 101.46% in the control group seen in the present study. However, the greater asymmetry seen in the ACL-R group in peak vGRF during the single-leg landing contrasts with Ithurburn et al., who reported no differences between ACL-R and control groups in this metric [99].

Previous work in healthy control subjects has reported asymmetry in knee extension moment at ground contact as well as no asymmetry in peak knee extension moment [287,310]. Additionally, Ernst et al. reported a significant decrease in knee extension moment in the involved limb of ACL-R patients compared to the uninvolved limb and both limbs of a control group [139], and a peak knee extension moment LSI of 6.93% has been reported in an ACL-R cohort [159]. Butler et al. also reported average values for each limb that would give an LSI of 26.53% when using the metric used in the present study [294]. This is lower than the 49.49% LSI given in the present study, but this difference could be attributed to the fact that the present study examined peak knee extension moments only during the first 10% of the stance phase and Butler et al. examined these peak moments across the entire stance phase. To our knowledge, the present study is the first to compare symmetry in peak knee extension moments during the initial portion of landing between ACL-R subjects and healthy controls.

Asymmetries in single-leg hop distance have been widely investigated in the literature, with many studies reporting greater asymmetries in ACL-R populations [101,136,138,141,159,295,296]. LSIs for control groups have ranged from 95.5% to 98.6% [101,102,159], and LSIs for ACL-R groups have ranged from 83.8% to 94.1% [101,282]. These results suggest that the previous ACL-R groups exhibited decreased hop distance asymmetry compared to the ACL-R cohort described in the present study. It is worth mentioning that symmetry in single-leg hop distance has been shown to improve as the time since surgical reconstruction increases [282,296-298], with one study reporting changes in LSIs over a period of 10 days [296]. This suggests that because even small differences in the time since surgery can significantly affect LSI values, the time since surgery is an important factor to consider when comparing the results of single-leg hop tests between ACL-R groups as well as between ACL-R and control groups. This assertion is reinforced by results from Novak et al., who reported a 6% LSI in ACL-R patients at least two years following reconstruction [136], and from Trigsted et al., who reported no differences in hop distance symmetry between a control group and an ACL-R group at an average of 31.9 months following reconstruction [293].

The ACL-R subjects displayed decreased CMD values in both vGRF ($p=0.010$) and knee flexion angles ($p<0.001$) when compared to the healthy control subjects (Table 5). The CMD metric has been utilized previously to assess both inter-trial and inter-session variability by comparing the waveforms for a particular variable from multiple trials or testing sessions [125-129]. Variations of the CMD metric have also been used recently to quantify differences between limbs in knee extension moments across a trial [287] as well as to quantify differences between ACL-R and control subjects in knee flexion angles across a trial [291]. The lower values seen in the ACL-R group when compared to the control group in both variables indicate that the use of a CMD for quantifying side-to-side symmetry across time may be a useful measure. Additionally, preliminary data has suggested that CMD and LSI calculations may provide different information regarding side-to-side symmetry in an ACL-R population. While an athlete may exhibit side-to-side symmetry at one particular point in time during a landing task, they may be employing an asymmetrical movement pattern across the entire task. This idea was reinforced by the results of this study, as no differences were seen in symmetry between the groups in knee flexion at ground contact (p=0.065) but CMD values for knee flexion across the entire landing phase indicated a significant difference in symmetry between the groups. This demonstrates that it may be important to include both CMD and LSI values in an assessment of symmetry in either an ACL-R athlete or a healthy athlete.

The control subjects displayed symmetry values in hop distance that can be labeled as normal according to previous work [257,285,286], while the ACL-R subjects exhibited significantly higher asymmetry values. Previous work has reported success in distinguishing between healthy and unhealthy athletes with the single-leg hopping task [315,226,284]. Logerstedt et al. reported that asymmetries in single-leg hop distance 6 months following ACL-R were predictive of decreased knee function 1 year postoperatively [260]. In addition, Barber et al. reported that control subjects who exhibited a hop distance LSI of less than 85% (or greater than 15%, with our LSI metric) had a higher rate of self-reported difficulty in pivoting, cutting, and twisting during athletic activities [284]. Furthermore, this metric has been used previously to successfully measure functional improvements during rehabilitation in ACL-R patients [282,296-298]. The results of this study reinforce the idea that an LSI of 90% (or 10%, using our LSI metric) is an appropriate benchmark for proper hop distance symmetry, as the control population satisfied this benchmark while the ACL-R population did not.

The relative ages of the subject groups was an important factor to consider in this analysis. Previous work has demonstarted differences in outcomes based on the age of the patients at the time of the ACL injury [224,299], and others have reported differences in movement mechanics based on age [322,323,327]. The ages of the control group studied here and the ACL-R groups collected previously were significantly different, with each ACL-R group being younger than the control group (Table 1). This difference in age was accounted for by using age as a covariate in the statistical comparisons between groups (Table 5). When this age difference was accounted for, the ACL subjects still displayed higher levels of asymmetry in many of the variables of interest when compared to the control group. This indicates that even though movement mechanics can change as an athlete ages, the presence of a previous ACL injury is a more important factor in the alteration of movement mechanics than the athlete's age. Despite this, subject age was shown to significantly affect several variables, including LSIs in peak vGRF during the single-leg landing task ($p=0.031$), magnitudes of differences between limbs in knee flexion at ground contact ($p=0.034$) and frontal plane ROM ($p=0.037$) during the stop jump task, magnitudes of differences between limbs in peak vGRF during the single-leg task ($p=0.040$), and CMDs in vGRF ($p=0.012$) and knee flexion angle ($p=0.004$) during the stop jump task. In addition, the magnitudes of differences between limbs in knee flexion at ground contact ($p=0.674$) and frontal plane ROM ($p=0.596$) were not significantly affected by the presence of an ACL injury and thus were only affected by age. Younger subjects displayed better symmetry than older subjects in all of these variables except for the magnitudes of differences between limbs in knee flexion at ground contact during the stop jump task, where older subjects exhibited lower magnitudes of differences than younger subjects. These improved symmetry values are in agreement with Ford et al., who found that post-pubertal females displayed greater frontal plane knee angles than pubertal females, which placed the post-pubertal females at a greater risk for an ACL injury [322].

Despite the significant difference in LSIs in peak knee flexion during the bilateral landing task between the two subject groups, the ACL-R group still displayed "normal" levels of symmetry in this metric. Holsgaard-Larsen et al. reported a peak knee flexion limb symmetry ratio of 98.6% - 100% represents perfect symmetry - in an ACL-R group during a bilateral landing task [159]. The ACL-R subjects in this study displayed a limb symmetry ratio of 99.92% in this metric. Additionally, the average values of the surgical and nonsurgical limbs were not significantly different (Table A2, Appendix A), which agrees with the results presented by Butler et al. that showed no differences in average peak knee flexion values between limbs in an ACL-R group [294]. The ACL-R subjects in the present study also demonstrated a significant difference between limbs in peak vGRF during the bilateral landing task ($p<0.0001$) (Table A2, Appendix A), which agrees with the results from multiple previous studies [41,100,294]. Studies that have examined the kinematics of only one limb during these landing tasks have reported decreased levels of peak knee flexion in ACL-R patients when compared to healthy controls [291-293]. To compare our results to these studies, we also examined the differences between groups in the data for individual limbs. The nondominant and dominant limbs of the control subjects were compared to the nonsurgical and surgical limbs, respectively, of the ACL-R subjects, as has been done previously [284,292]. Interestingly, the subjects in the present study displayed no differences in peak knee flexion angles between the nondominant/uninvolved limbs ($p=0.214$) or the dominant/involved limbs ($p=0.149$) during the bilateral landing task (Table A4, Appendix A). This disagrees with the results presented by Delahunt et al., who reported a higher peak knee flexion in control subjects than in ACL-R subjects during a bilateral landing task [291]. The results in the present study were likely affected by the age of the subjects, as a previous analysis that didn't account for age demonstrated a significant difference in peak knee flexion between the groups.

No differencs were seen in frontal plane range of motion (ROM) symmetry between the two groups during the bilateral landing task. To our knowledge, only Roos et al. has compared frontal plane knee angles during landing between ACL-R and control groups. They reported that ACL-R patients exhibit higher levels of frontal plane knee excursion than control subjects during a single-leg landing task [292]. Most of the literature that has examined frontal plane knee mechanics has focused on varus and valgus moments, as these have been shown to place stress on the ACL during landing and cutting tasks [49,68,300,301]. However, Paterno et al. reported that ACL-R patients who displayed higher levels of frontal plane knee motion on the operative limb were over three times more likely to suffer a secondary ACL injury than ACL-R patients who displayed lower levels of frontal plane knee motion [162], and Hewett et al. reported that female athletes who suffered an ACL tear displayed higher peak valgus angles prior to injury than athletes who did not suffer an injury [12]. Additionally, Torry et al. suggested that an increase in frontal plane knee motion is indicative of increased anterior tibial translation [302], which has been shown to be a significant risk factor for ACL injury [42,55,57]. These studies reflect the need for further analysis of frontal plane knee angles in ACL-R populations. The large asymmetry values recorded in the control population would suggest that the limbs are naturally moving independently of the other and thus should be examined in isolation rather than with an LSI. This point is strengthened by the large number of studies that have reported peak valgus angles in only one limb in both ACL-R and control populations [69,269,305-307]. Additionally, the large number of studies that have examined symmetry in sagittal plane knee angles demonstrates that potentially side-to-side symmetry is more appropriate for the assessment of those variables than variables in the frontal plane [99,102,141,142,159,272,294].

The method of analyzing the magnitude of the differences between limbs as a way of assessing side-to-side symmetry was first utilized by Paterno et al., as he examined peak knee extension moments during the initial portion of the landing phase. Using this metric, the authors found that asymmetry in this peak knee extension moment was indicative of an increased risk of secondary injury [162]. To assess the usefulness of this symmetry measure, we compared the differences seen in this measure between the bilateral landing ACL and control groups to the differences seen in LSIs between the same two groups. Interestingly, the two metrics did not show equivalent differences in each variable of interest (Tables 5 and 6). While both measures of symmetry indicated differences in symmetry between groups in both variables of interest in the single-leg hop task, the magnitude of differences between limbs was only different between groups in peak knee flexion during the stop jump task. Since it is generally accepted that ACL-R athletes show increased asymmetry when compared to healthy athletes [101,130,131,134,139,141-143], the LSI measure may be more accurately indicating the differences in side-to-side symmetry than the magnitude of differences measure.

The MARX activity level survey administered during this study revealed that the control subjects showed less frequent participation in running ($p=0.029$), cutting ($p<0.001$), deceleration ($p<0.001$) and pivoting ($p<0.001$) activities than the ACL-R cohort that completed the single-leg hop testing (Table 2). This difference in athletic participation may have impacted the results of the study. Previous work has reported differences in movement mechanics between high-level athletes and inexperienced athletes [274-278]. However, the effects of athletic experience on side-to-side symmetry are unknown.

4.3 Discussion of Aim 3

The third portion of the study aimed to compare symmetry values seen in the single-leg and bilateral tasks in the control group. As shown in Table 3, greater asymmetry was seen in the single-leg hop task in knee flexion at ground contact, peak knee flexion, and knee flexion angles across the entire landing phase ($p<0.001$ in all cases), and greater asymmetry was seen in the stop-jump task in peak vGRF ($p=0.009$). Our hypothesis was partially supported, as more asymmetry was observed during the single-leg hop task than during the stop-jump task, but the mixed results do not allow for a conclusive statement regarding the effects of a specific landing task on side-to-side symmetry in control subjects.

The greater asymmetry in knee flexion at ground contact during the single-leg hop task reported here disagrees with results reported by McPherson et al., who suggested that control subjects exhibit greater asymmetry in knee flexion at ground contact during bilateral landings [311]. That study reported higher knee flexion angles than those found in the present study, particularly in knee flexion at ground contact (38.2° in the dominant limb and 39.9° in the nondominant limb in a bilateral landing task, compared to 23.0° and 24.2° in the dominant and nondominant limbs, respectively, in the present study) [311]. Holsgaard-Larsen et al. reported peak knee flexion limb symmetry ratios of 101.3% during a bilateral landing task and 97.8% during a single-leg task, although it is unknown if these ratio values were significantly different [159]. In addition, Van der Harst et al. reported average peak knee flexion values for each limb during a single-leg landing task that would give a limb symmetry ratio of 98.34% [102]. The subjects in the present study displayed peak knee flexion limb symmetry ratios of 101.45% in the bilateral landing task and 99.70% in the single-leg landing task.

Many studies that have compared the two landing techniques have analyzed the mechanics of only one limb during each task, particularly in knee flexion angles. These studies agree that control subjects exhibit significantly increased peak knee flexion and knee flexion at ground contact during a bilateral landing task [25,266,269,307,312]. In addition, multiple studies have reported differences in peak vGRF between the two landing tasks [266,307], with Yeow et al. reporting greater peak vGRF values in single leg landings when compared to bilateral landings [266]. An examination of the differences between the tasks in individual limbs demonstrated that all variables were significantly different in each limb between the two landing tasks ($p<0.001$ in all cases) (Table A6, Appendix A). Subjects displayed higher values in peak knee extension moment during landing, higher vGRF values, decreased knee flexion at ground contact, decreased peak knee flexion, and increased frontal plane ROM in the single-leg hop task when compared to the stop jump task. Bates et al. reported vGRF values of 2 BW in the left limb and 2.17 BW in the right limb in a control population during a bilateral landing task (Bates 2013), and Paterno et al. reported values of 1.6 BW and 1.5 BW in the same task [130]. Additionally, Harty et al. reported a single leg vGRF of 3.40 in a control population during a single-leg landing task [307]. The subjects in the present study also displayed this behavior, with peak vGRF values of 3.38 and 3.33 in the nondominant and dominant limbs respectively in the single-leg landing task and values of 1.66 and 1.71 in the nondominant and dominant limbs respectively in the bilateral landing task.

Several studies have also previously examined differences in frontal plane knee angles between the two landing tasks. Nagano et al. and Weinhandl et al. asserted that the frontal plane knee ROM is greater during a bilateral landing than during a single-leg landing in control populations [312,313], with Nagano et al. reporting frontal plane ROM values of 11.2° during a bilateral landing task and 6.6° during a single-leg landing task [313]. Other studies in either male or female control subjects also showed higher frontal plane ROM values during the bilateral landing task [53,69,312]. The results of the present study do not agree with these results, as the subjects in the present study exhibited frontal plane ROM values of 6.6° in each limb during the stop jump task and values of 11.74° and 10.54° during the single-leg landing task. The differences in frontal plane ROM in the single-leg landing task could be due to the fact that the other studies quantified knee ROM during a drop landing, meaning that subjects stood on a platform and dropped onto a force plate on one foot [312,313]. The subjects in the present study completed a maximum distance forward hop, which combines a forward hop with a single-leg landing on a force plate. It is possible that this difference in landing technique could have resulted in different ROM values in the frontal plane. At least two studies have utilized both single-leg drop landings and maximum forward hops to examine differences between genders [319] and between ACL-R and control groups [328], but neither study examined the differences between the two tasks. More work is needed to explore this idea further.

4.4 Discussion of Aim 4

The final goal of the study was to assess sex differences in the control group during each of the two landing tasks. Table 8 shows that females exhibited greater asymmetry in knee flexion at ground contact during the stop jump task (p=0.028) and that males displayed more asymmetry in knee flexion across the landing phase of the single-leg hop task (0.0421). No other differences in LSIs were observed. Our hypothesis was not supported, as few differences in side-to-side symmetry between sexes were identified.

To our knowledge, only three studies have previously examined sex-specific symmetry in landing mechanics. Ford et al. and Pappas et al. both reported that females displayed greater asymmetries in peak valgus knee angles during bilateral landing tasks than their male counterparts [53,265]. The subjects in the present study did not show any differences between sexes in symmetry in this metric (p=0.420). Additionally, Bell et al. suggested that no differences existed between sexes in peak vGRF symmetry during a bilateral landing task [318]. This agrees with the results of the present study, as we found no differences between sexes in this metric (p=0.900).

The male subjects in this study displayed knee flexion angles at ground contact of 25.38° in the nondominant limb and 24.54° in the dominant limb during the stop jump task. This is in line with other previous work, as studies have reported values in this metric between 10° and 39.9° in either limb [43,69,269,311,312,316,317]. The female subjects displayed similar values of 23.03° in the nondominant limb and 21.46° in the dominant limb during the stop jump task. Previous work has reported values of between 5.4° and 31.9° in either limb [43,69,269,312,316,317]. Peak knee flexion values were also in line with previous work. The male subjects in the present study exhibited values of exactly

84.45° and 84.97° in the nondominant and dominant limbs respectively during the stop jump task, while females showed values of exactly 87.66° in both limbs. Previous literature has reported peak knee flexion values during this task of 77.36° to 95° in male subjects and 68.54° to 93° in female subjects [12,43,69,291,311,317].

Much of the previous work examining sexes has largely explored biomechanical measures in only one limb [65,66,69,269,280,312,316,317,319,320]. Multiple studies have reported reduced knee flexion angles at ground contact in females performing a bilateral landing task [66,316,317]. The present study did not reinforce this assertion, as a comparison of male and female dominant (p=0.204) and nondominant (p=0.252) limbs yielded no differences between the sexes. In fact, the results of this study showed that only peak knee extension moment during landing in both landing tasks and hop distance in both limbs significantly differed between sexes (Table A10, Table A11, Appendix A). Several studies have also reported no significant differences in peak knee flexion during a bilateral landing task [317], knee flexion at ground contact in a bilateral landing task [69], peak vGRF during a bilateral landing task [316], and frontal plane ROM during both landing tasks [312]. However, many previous studies contradict these results. Multiple studies have found sex differences in knee flexion at ground contact during bilateral landings [66,316,317], peak vGRF during bilateral [69] and both landing tasks [269], and frontal plane ROM during both landing tasks [53,69,321]. The level of experience in the athletes tested in this study, particularly in the male athletes, could have impacted these results. Each study that agreed with the lack of differences between sexes also utilized a healthy recreational athlete population for testing [69,312,316,317]. It may follow that since both sexes were less experienced athletically than a population of true athletes, the two sexes may have moved more similarly than is commonly seen in such a comparison.

A large portion of the sex comparisons in landing has been focused on knee angles in the frontal plane [53,69,269,305,306,312,321]. This large emphasis on frontal plane knee mechanics is based on previous studies by Hewett et al., who reported a greater risk for primary ACL injury in female athletes who displayed large knee valgus angles and moments [12], and by Paterno et al., who reported a greater risk for secondary ACL injury in females ACL-R patients who displayed greater frontal plane ROM in the involved limb [162]. Multiple studies have reported larger peak valgus angles than females in both tasks when compared to males [53,69,269,305]. This agrees with the results of this study, as females showed significantly larger peak valgus angles in the nondominant ($p<0.001$) and dominant ($p<0.001$) limbs during the stop jump task and in the nondominant ($p=0.027$) limb during the single-leg landing task.

4.5 Limitations

One limitation of the study was the population of healthy recreational athletes that participated in the study. Even though all subjects possessed at least some experience in jumping and landing activities, the unavoidable differences in experience among the subject population could have affected the movement mechanics measured in this study. Table 2 shows that MARX activity level scores differed between the control group and the ACL-R group that performed the single-leg hop testing (this data was not available in the ACL-R group that performed the bilateral stop jump testing). Many studies have previously shown that differences in athletic experience can affect movement mechanics in a variety of ways [274-278]. Additionally, the level of strength and neuromuscular training of the participants in this study was unknown. Previous work has highlighted the positive effects of neuromuscular training programs on movement mechanics, mostly as a strategy for reducing the incidence of primary or secondary ACL injuries [106,160,279-282]. It has been shown that plyometric training improves peak knee flexion and knee flexion at ground contact in single-leg landings [279]. In addition, strength training has been shown to reduce asymmetries in movements [160,281,283] and decrease peak landing forces [280,281]. The level of neuromuscular and strength training was not noted for the subjects in this study, so a large distribution of levels of strength and neuromuscular coordination among the subjects is possible. Because of these factors, the movement mechanics measured in this study may not provide a perfect comparison for mechanics measured in an ACL-R population. Another limitation was that only loading data was collected in the ACL-R population that completed the single-leg hopping tasks described in this study, meaning that kinematics could not be compared between the control and ACL-R populations. The final limitation was the reliability of these loading data. These data were measured with in-shoe pressure sensors (pedoped, Novel Electronics, St. Paul, MN) that have been validated for measuring load in walking and running but not yet in jumping and landing. Future work will attempt to validate these devices so that more accurate comparisons can be made.

5 CONCLUSIONS

The first conclusion of this study is that healthy control subjects show significant asymmetries in several metrics, including some that have been cited as predictors for secondary ACL injury in ACL-R patients. The second is that despite the presence of these asymmetries in a control population, control subjects still land more symmetrically than athletes who have previously suffered an ACL injury. The third conclusion is that bilateral and single-leg hopping tasks can both be used to assess movement symmetry, as one of these tasks does not promote more symmetrical movements than the other. The final conclusion of the study is that very few differences exist in symmetry between sexes in a control population during these tasks. In order to continue to explore each of these conclusions, future work must include a kinematic and kinetic analysis of a single-leg hopping task in a group of ACL-R patients. The results of this analysis would allow researchers to compare asymmetries in kinematic and kinetic parameters between ACL-R and control subjects during a single-leg hopping task. It would also allow for a comparison of these metrics between landing tasks in ACL-R subjects. Other future work must include a comparison of symmetry values to true values in the variables discussed in this study. This would allow us to better understand which measures of symmetry are important to consider when assessing injury risks in an ACL-R population.

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APPENDIX A

Table A1: Direct Comparisons Between Limbs in Control Group During Both Tasks

		Direct Comparison Between Limbs			
		Mean Value in Nondominant Limb	Mean Value in Dominant Limb	p-value	LSI from Mean Values
Stop Jump	Peak Knee Ext. Moment During Landing (BW*BH)	0.08±0.03	0.08±0.03	0.684	1.13
	Peak vGRF (BW)	1.66±0.49	1.71±0.45	0.794	2.88
	Knee Flexion at Ground Contact (degrees)	24.20±7.72	23.00±7.56	0.053	5.11
	Peak Knee Flexion (degrees)	86.05±15.99	86.31±16.12	0.362	0.30
	Frontal Plane ROM (degrees)	6.59±4.38	6.60±4.17	0.879	0.17
Single-Leg Hop	Peak Knee Ext. Moment During Landing (BW*BH)	0.13±0.06	0.12±0.06	0.314	4.08
	Peak vGRF (BW)	3.38±0.49	3.33±0.51	0.183	1.45
	Knee Flexion at Ground Contact (degrees)	8.74±4.98	8.40±5.35	0.484	3.91
	Peak Knee Flexion (degrees)	66.52±9.82	65.56±9.73	0.264	1.44
	Frontal Plane ROM (degrees)	11.74±6.44	10.54±6.16	0.198	10.83
	Hop Distance (cm)	150.99±39.51	156.89±36.07	**0.006**	3.83

Table A2: LSIs and Direct Comparisons Between Limbs in ACL-R Groups

		LSI	Direct Comparison Between Limbs			
		LSI (%)	Mean Value in Nondominant Limb	Mean Value in Dominant Limb	p-value	LSI from Mean Values
Stop Jump	Peak Knee Ext. Moment During Landing	72.90±57.27	0.06±0.03	0.04±0.03	**<0.001**	49.49
	Peak vGRF	22.70±14.62	1.98±0.43	1.59±0.33	**<0.001**	22.15
	Knee Flexion at Ground Contact	32.29±29.02	19.97±8.07	20.00±6.34	0.648	0.14
	Peak Knee Flexion	6.00±3.58	73.86±7.96	73.79±7.72	0.927	0.08
	Frontal Plane ROM	30.32±17.16	7.64±2.73	7.11±2.32	0.388	7.14
Single-Leg Hop	Peak vGRF	15.34±10.79	3.13±0.53	2.84±0.48	**0.003**	9.91
	Hop Distance	23.02±15.88	147±28.97	117±28.62	**<0.001**	22.40

Table A3: CMD Values in Both Groups During Both Tasks

		ACL	Control	p-value
Stop Jump	vGRF	0.656±0.272	0.824±0.173	**0.005**
	Knee Flexion	0.951±0.050	0.987±0.011	**<0.001**
Single-Leg Hop	vGRF	--	0.843±0.158	--
	Knee Flexion	--	0.930±0.082	--

Table A4: LSIs and Magnitudes of Differences Between Limbs Compared Between ACL-R and Control Groups

			ACL	Control	p-value
LSI (%)	Stop Jump	Peak Knee Extension Moment During Landing (%)	72.90±57.27	28.22±18.39	**<0.001**
		Peak vGRF (%)	22.70±14.62	17.39±15.16	**0.021**
		Knee Flexion at Ground Contact (%)	32.29±29.02	19.77±17.76	0.065
		Peak Knee Flexion (%)	6.00±3.58	2.06±1.68	**<0.001**
		Frontal Plane ROM (%)	30.32±17.16	52.36±34.95	0.064
	Single -Leg Hop	Peak vGRF (%)	15.34±10.79	7.70±7.47	**<0.001**
		Hop Distance (%)	23.02±15.88	8.45±7.61	**<0.001**
Difference Between Limbs	Stop Jump	Peak Knee Extension Moment During Landing (BW*BH)	0.03±0.02	0.02±0.01	0.078
		Peak vGRF (BW)	0.41±0.29	0.29±0.26	0.362
		Knee Flexion at Ground Contact (degrees)	6.40±6.23	4.21±3.47	0.674
		Peak Knee Flexion (degrees)	4.40±2.60	1.80±1.51	**<0.001**
		Frontal Plane ROM (degrees)	2.30±1.61	3.30±2.62	0.596
	Single -Leg Hop	Peak vGRF (BW)	0.43±0.36	0.23±0.21	**0.012**
		Hop Distance (cm)	32.45±22.29	12.30±10.53	**<0.001**

Table A5: Differences Between ACL-R and Control Groups in Each Limb During Each Landing Task

			ACL	Control	p-value
Stop Jump	Nondominant/ Nonsurgical Limb	Peak Knee Ext. Moment During Landing (BW*BH)	0.06±0.03	0.08±0.03	0.214
		Peak vGRF (BW)	1.98±0.43	1.66±0.49	0.206
		Knee Flexion at Ground Contact (degrees)	19.97±8.07	24.20±7.72	0.136
		Peak Knee Flexion (degrees)	73.86±7.96	86.05±15.99	0.214
		Frontal Plane ROM (degrees)	7.64±2.73	6.59±4.38	**0.012**
	Dominant/ Surgical Limb	Peak Knee Ext. Moment During Landing (BW*BH)	0.04±0.03	0.08±0.03	**<0.001**
		Peak vGRF (BW)	1.59±0.33	1.71±0.45	0.085
		Knee Flexion at Ground Contact (degrees)	20.00±6.34	23.00±7.56	0.664
		Peak Knee Flexion (degrees)	73.79±7.72	86.31±16.12	0.149
		Frontal Plane ROM (degrees)	7.11±2.32	6.60±4.17	0.128
Single-Leg Hop	Nond omina	Peak vGRF (BW)	3.13±0.53	3.18±0.57	0.750
		Hop Distance (cm)	147±28.97	151±39.51	0.161
	Domi nant/	Peak vGRF (BW)	2.84±0.48	3.12±0.63	0.077
		Hop Distance (cm)	117±28.62	157±36.07	**<0.001**

Table A6: LSIs and Magnitudes of Differences Between Limbs Compared Between Stop Jump and Single-Leg Hop Tasks in Control Group

		Stop Jump	Single-Leg Hop	p-value
LSI (%)	Peak Knee Extension Moment During Landing (%)	28.22±18.39	32.68±46.53	0.346
	Peak vGRF (%)	17.39±15.16	10.26±8.88	**0.009**
	Knee Flexion at Ground Contact (%)	19.77±17.76	43.87±34.91	**<0.001**
	Peak Knee Flexion (%)	2.06±1.68	9.88±7.74	**<0.001**
	Frontal Plane ROM (%)	52.36±34.95	42.15±34.61	0.085
Difference Between Limbs	Peak Knee Extension Moment During Landing (%)	0.02±0.01	0.03±0.02	**0.031**
	Peak vGRF (%)	0.29±0.26	0.34±0.29	0.288
	Knee Flexion at Ground Contact (%)	4.21±3.47	3.67±2.93	0.467
	Peak Knee Flexion (%)	1.80±1.51	6.53±5.28	**<0.001**
	Frontal Plane ROM (%)	3.30±2.62	4.35±4.09	0.302

Table A7: LSIs and Direct Comparisons Between Tasks in Nondominant and Dominant Limbs in Control Group During Both Tasks

		Stop Jump	Single-Leg Hop	p-value
Nondominant Limb	Peak Knee Ext. Moment During Landing (BW*BH)	0.08±0.03	0.13±0.06	**<0.001**
	Peak vGRF (BW)	1.66±0.49	3.38±0.49	**<0.001**
	Knee Flexion at Ground Contact (degrees)	24.20±7.72	8.74±4.98	**<0.001**
	Peak Knee Flexion (degrees)	86.05±15.99	66.52±9.82	**<0.001**
	Frontal Plane ROM (degrees)	6.59±4.38	11.74±6.44	**<0.001**
Dominant Limb	Peak Knee Ext. Moment During Landing (BW*BH)	0.08±0.03	0.12±0.06	**<0.001**
	Peak vGRF (BW)	1.71±0.45	3.33±0.51	**<0.001**
	Knee Flexion at Ground Contact (degrees)	23.00±7.56	8.40±5.35	**<0.001**
	Peak Knee Flexion (degrees)	86.31±16.12	65.56±9.73	**<0.001**
	Frontal Plane ROM (degrees)	6.60±4.17	10.54±6.16	**<0.001**

Table A8: LSIs and Magnitudes of Differences Between Limbs Compared Between Genders in Control Group

			Males	Females	p-value
LSI (%)	Stop Jump	Peak Knee Extension Moment During Landing (BW*BH)	26.23±21.58	30.21±14.64	0.172
		Peak vGRF (BW)	17.28±15.23	17.50±15.35	0.900
		Knee Flexion at Ground Contact (degrees)	14.73±14.35	24.82±19.59	**0.028**
		Peak Knee Flexion (degrees)	2.20±1.72	1.92±1.67	0.395
		Frontal Plane ROM (degrees)	58.40±37.12	46.32±32.11	0.246
	Single-Leg Hop	Peak Knee Extension Moment During Landing (BW*BH)	27.64±24.48	37.72±61.26	0.728
		Peak vGRF (BW)	10.61±9.43	9.91±8.45	0.819
		Knee Flexion at Ground Contact (degrees)	39.64±29.27	47.81±39.56	0.682
		Peak Knee Flexion (degrees)	7.62±5.58	12.15±8.95	0.058
		Frontal Plane ROM (degrees)	45.63±36.52	38.67±32.84	0.473
		Hop Distance (cm)	8.12±7.70	8.78±7.51	0.587
Difference Between Limbs	Stop Jump	Peak Knee Extension Moment During Landing (BW*BH)	0.02±0.02	0.02±0.01	0.959
		Peak vGRF (BW)	0.32±0.28	0.25±0.23	0.333
		Knee Flexion at Ground Contact (degrees)	5.11±3.98	3.31±2.64	0.077
		Peak Knee Flexion (degrees)	1.59±1.52	2.02±1.50	0.201
		Frontal Plane ROM (degrees)	3.76±2.74	2.84±2.46	0.176
	Single-Leg Hop	Peak Knee Extension Moment During Landing (BW*BH)	0.04±0.03	0.03±0.02	0.137
		Peak vGRF (BW)	0.27±0.26	0.41±0.32	0.056
		Knee Flexion at Ground Contact (degrees)	3.77±3.58	2.34±3.45	0.326
		Peak Knee Flexion (degrees)	5.67±4.34	7.39±6.03	0.363
		Frontal Plane ROM (degrees)	4.12±4.39	4.57±3.84	0.446
		Hop Distance (cm)	13.73±12.47	10.87±8.10	0.529

Table A9: LSIs and Direct Comparisons Between Limbs in Females in Control Group During Both Tasks

		LSI	Direct Comparison Between Limbs			
		LSI (%)	Mean Value in ND Limb	Mean Value in D Limb	p-value	LSI from Mean Values
Stop Jump	Peak Knee Ext. Moment During Landing	30.21±14.64	0.07±0.02	0.07±0.03	0.513	3.89
	Peak vGRF	17.50±15.35	1.58±0.42	1.65±0.34	0.391	4.25
	Knee Flexion at Ground Contact	24.82±19.59	23.03±6.29	21.46±7.35	**0.008**	7.05
	Peak Knee Flexion	1.92±1.67	87.66±14.99	87.66±14.91	0.928	0.005
	Frontal Plane ROM	46.32±32.11	6.95±4.94	7.15±4.45	0.833	2.90
Single-Leg Hop	Peak Knee Ext. Moment During Landing	37.72±61.26	0.11±0.05	0.11±0.06	0.906	0.70
	Peak vGRF	9.91±8.45	3.30±0.50	3.33±0.61	0.695	1.12
	Knee Flexion at Ground Contact	47.81±39.56	9.03±5.07	8.40±6.38	0.358	7.16
	Peak Knee Flexion	12.15±8.95	65.53±9.43	65.41±9.12	0.787	0.19
	Frontal Plane ROM	38.67±34.61	11.70±6.11	10.35±6.10	0.269	12.22
	Hop Distance	8.78±7.51	129.9±31.34	136.98±27.59	**0.004**	5.30

Table A10: LSIs and Direct Comparisons Between Limbs in Males in Control Group During Both Tasks

		LSI	Direct Comparison Between Limbs			
		LSI (%)	Mean Value in ND Limb	Mean Value in D Limb	p-value	LSI from Mean Values
Stop Jump	Peak Knee Ext. Moment During Landing	26.23±21.58	0.09±0.03	0.09±0.03	0.778	1.71
	Peak vGRF	17.28±15.23	1.74±0.55	1.76±0.54	0.710	1.61
	Knee Flexion at Ground Contact	14.73±14.35	25.38±8.87	24.54±7.58	0.580	3.37
	Peak Knee Flexion	2.20±1.72	84.45±17.03	84.97±17.39	0.261	0.61
	Frontal Plane ROM	58.40±37.11	6.22±3.79	6.04±3.86	0.992	2.97
Single-Leg Hop	Peak Knee Ext. Moment During Landing	27.64±24.48	0.15±0.05	0.14±0.05	0.178	7.58
	Peak vGRF	10.61±9.43	3.45±0.48	3.32±0.40	**0.010**	3.97
	Knee Flexion at Ground Contact	39.64±29.27	8.45±4.96	8.40±4.19	0.896	0.56
	Peak Knee Flexion	7.62±5.58	67.50±10.27	65.72±10.45	0.138	2.67
	Frontal Plane ROM	45.63±36.52	11.79±6.87	10.73±6.31	0.462	9.46
	Hop Distance	8.12±7.70	172.08±35.68	176.80±32.65	0.265	2.70

Table A11: Direct Comparisons Between Genders in Nondominant Limb in Control Group During Both Tasks

		Male	Female	p-value
Stop Jump	Peak Knee Ext. Moment During Landing (BW*BH)	0.09±0.03	0.07±0.02	**0.004**
	Peak vGRF (BW)	1.74±0.55	1.58±0.42	0.261
	Knee Flexion at Ground Contact (degrees)	25.38±8.87	23.03±6.29	0.252
	Peak Knee Flexion (degrees)	84.45±17.03	87.66±14.99	0.623
	Frontal Plane ROM (degrees)	6.22±3.79	6.95±4.94	0.403
	Peak Valgus Angle (degrees) (- indicates valgus)	1.37±5.22	-7.19±7.24	**<0.001**
Single-Leg Hop	Peak Knee Ext. Moment During Landing (BW*BH)	0.15±0.05	0.11±0.05	**0.010**
	Peak vGRF (BW)	3.45±0.48	3.30±0.50	0.317
	Knee Flexion at Ground Contact (degrees)	8.45±4.96	9.03±5.07	0.513
	Peak Knee Flexion (degrees)	67.50±10.27	65.53±9.43	0.741
	Frontal Plane ROM (degrees)	11.79±6.87	11.70±6.11	0.651
	Hop Distance (cm)	172.08±35.68	129.9±31.34	**<0.001**
	Peak Valgus Angle (degrees) (- indicates valgus)	-1.94±5.52	-4.71±4.56	**0.027**

Table A12: Direct Comparisons Between Genders in Dominant Limb in Control Group During Both Tasks

		Male	Female	p-value
Stop Jump	Peak Knee Ext. Moment During Landing (BW*BH)	0.09±0.03	0.07±0.03	**0.020**
	Peak vGRF (BW)	1.76±0.54	1.65±0.34	0.475
	Knee Flexion at Ground Contact (degrees)	24.54±7.58	21.46±7.35	0.204
	Peak Knee Flexion (degrees)	84.97±17.39	87.66±14.91	0.741
	Frontal Plane ROM (degrees)	6.04±3.86	7.15±4.45	0.317
	Peak Valgus Angle (degrees) (- indicates valgus)	1.95±6.38	-7.48±5.47	**<0.001**
Single-Leg Hop	Peak Knee Ext. Moment During Landing (BW*BH)	0.14±0.05	0.11±0.06	**0.041**
	Peak vGRF (BW)	3.32±0.40	3.33±0.61	0.944
	Knee Flexion at Ground Contact (degrees)	8.40±4.19	8.40±6.38	0.912
	Peak Knee Flexion (degrees)	65.72±10.45	65.41±9.12	0.992
	Frontal Plane ROM (degrees)	10.73±6.31	10.35±6.10	0.695
	Hop Distance (cm)	176.80±32.65	136.98±27.59	**<0.001**
	Peak Valgus Angle (degrees) (- indicates valgus)	-2.50±6.36	-4.21±4.05	0.154