A Principal Component Analysis Investigation of Drop Landings for Defining Anterior Cruciate Ligament Injury Risk Factors

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A PRINCIPAL COMPONENT ANALYSIS INVESTIGATION OF DROP LANDINGS FOR DEFINING ANTERIOR CRUCIATE LIGAMENT INJURY RISK FACTORS

by

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ABSTRACT

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Injury to the anterior cruciate ligament (ACL) has been widely investigated through observational video analysis and laboratory based cadaveric, motion capture and computer simulation models. With the greater incidence of injury in the female population, recent emphasis has been placed on understanding ACL injury mechanisms in females. By using our understanding of injury mechanisms and prospective studies, injury prediction methods can be created. Once injury can be reliably predicted, training methods can be implemented to reduce likelihood of injury and avoid devastating consequences. There is a need for a reliable way to reduce motion capture data obtained in a laboratory setting to viable measures that characterize the entire data set and correlate such measures to clinically relevant tests.

The present study performed motion analysis on healthy active young adult females during drop jump landings to characterize normal jump landing dynamics. Kinematic and kinetic data was reduced using principal component analysis to objectively determine variables of importance. Five principal components represented a cumulative 87.41% of the data set variance. Using principal component scores, significant associations were identified between principal component four (base of support at initial contact, peak knee abduction moment and 100 ms after initial contact) and knee flexion to extension isokinetic strength ratio. Additional significant correlation was found between principal component five (initial contact coronal knee moment and transverse knee moment) and abduction to adduction isokinetic strength ratio tested at 90°/sec. These results suggest principal component analysis is a viable method to reducing dynamic motion capture data. Further, principal component scores are a possible way to predict isokinetic strength ratios obtained in the clinic.

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1 Introduction

1.1 Overview

Anterior cruciate ligament (ACL) injury has been widely studied in an effort to better understand injury mechanisms and injury prevention. Injury to the ACL is one of the most commonly occurring athletic injuries with a large portion of injury occurring with noncontact mechanisms. With the growing number of female athletes, ACL injuries in the female population has surpassed the amount of injuries in the male population (LaBella, 2014). Given this gender discrepancy, much research has been devoted to determining risk factors associated with injury through gender comparison and female-specific studies. Once injury mechanisms are identified, training programs can be implemented in susceptible populations to reduce risk and avoid the long term repercussions associated with ACL injury.

1.2 ACL Injury

1.2.1 Knee Anatomy

The knee is the joint comprised of the articulation between the femur and tibia (Neumann, 2010). Given the structure of the articulating bones, the surrounding ligaments, muscles and articular cartilage of the knee are vital in maintaining stability of the joint. Function and mechanics of the knee joint are largely dependent on the action of the surrounding joints; the hip and ankle, as well as the muscular strength and control of these joints.

There are four major ligaments within the knee connecting the femur to the tibia to provide stability to the joint. These ligaments include the medial collateral ligament (MCL), lateral collateral ligament (LCL), posterior cruciate ligament (PCL) and the anterior cruciate ligament (ACL). When the knee is extended, the MCL and LCL are taut and provide the primary resistance to coronal motion in abduction and adduction, respectively. In a flexed knee position, the MCL and LCL are slack allowing a greater tibial rotation range of motion without stressing these ligaments. Given the change in laxity with flexion, the collateral ligaments experience increased vulnerability to injury due to coronal motion when the knee is in an extended position.

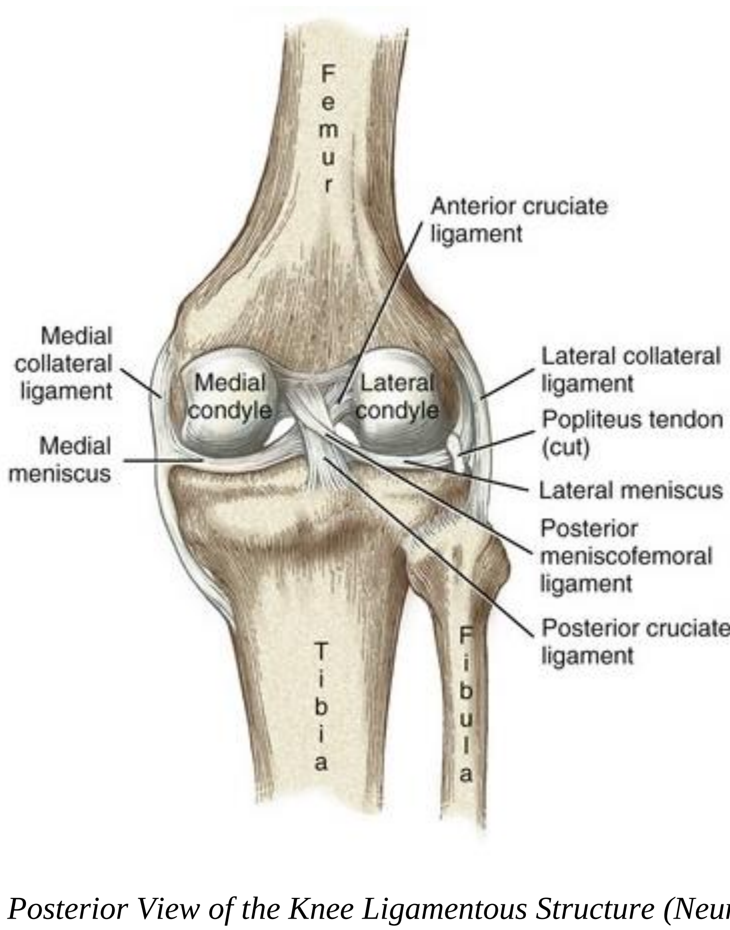

Figure 1: *Posterior View of the Knee Ligamentous Structure (Neumann, 2010)*

The cruciate ligaments, the ACL and PCL, cross within the intercondylar notch of the femur to connect the tibia to the femur. The ACL attaches to the anterior intercondylar area of the tibia, extends posteriorly in a lateral direction to attach to the medial side of the lateral femoral condyle. The cruciate ligaments are able to provide multiplanar stability to the knee joint. Similar to the collateral ligaments, the ACL and PCL are most taut when approaching an extended position. This makes them most vulnerable to injury when the knee is in extension. The ACL plays a large role in the preventing anterior translation of the tibia and posterior translation of the femur. These translations occur with the internal force produced from quadriceps flexion in a near extended position. Additionally, the ACL is able to assist the collateral ligaments in resisting varus, valgus and axial rotation of the knee (Neumann, 2010). Injury to the ligaments of the knee are most likely to occur with a high velocity stretch of the ligament while it is already experiencing tension (in an extended position). Additional considerations for injury include the ground reaction force (GRF), muscle forces, joint alignment, and surrounding tissue.

Contained within the knee joint at the medial and lateral plateaus of the proximal tibia, the medial and lateral meniscus are cartilaginous regions providing reduced compressive stress between the femur and tibia by increasing the contact surface area. When damage to the ligaments of the knee occur, the menisci are frequently injured due to the articular trauma endured. Damage patterns to this articular cartilage have been suggested as a way to determine injury mechanisms to the ACL (Levine, 2013).

In addition to the passive structures within the knee, lower extremity musculature surrounding the knee is an important contributor to knee stability and must be considered when evaluating biomechanics and neuromuscular control of the joint. The knee musculature, when appropriately activated, provides reduced likelihood of injury to the passive interarticular structures. Musculature surrounding the knee of importance in stability include the knee extensors and flexor-rotators. Involved in knee extension, the quadriceps femoris muscle group is made up of the rectus femoris, vastus lateralis, vastus medialis, and vastus intermedius. Of these four muscles, the rectus femoris is the only biarthrodial joint involved in both hip flexion and knee extension. The function of the quadriceps muscles are to stabilize the knee and provide controlled resistance to gravity on the body's center of mass.

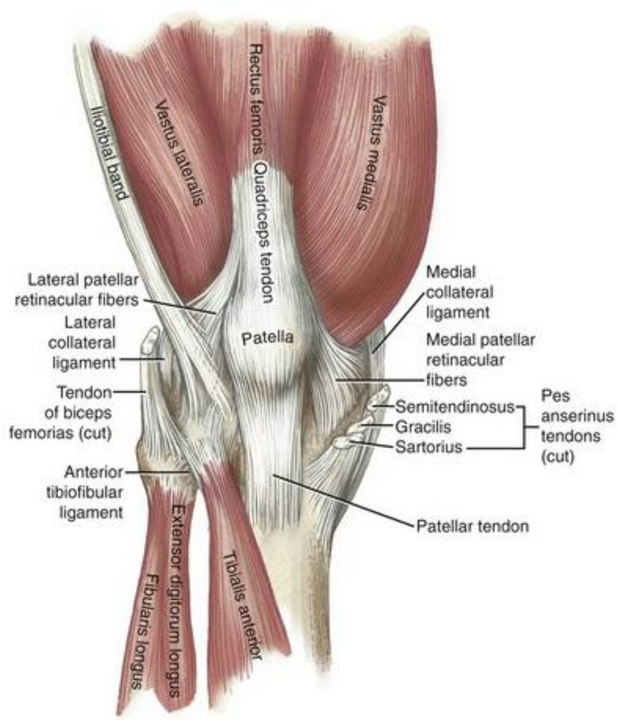

Figure 2: *Anterior View of the Lower Extremity Musculature Surrounding the Knee (Neumann, 2010).*

Muscles involved in knee flexion include the hamstrings (semimembranosus, semitendonosus, and biceps femoris), sartorius, gracilis, popliteus, and the gastrocnemius. Several of these muscles have both a flexion and rotation action at the knee. With insertions at the posterior tibia, the knee flexor group is able to provide assistance to the ACL in applying a posterior force on the tibia to resist anterior translation at an extended position. While it does not cross the knee joint, the soleus has been identified to help in combating anterior tibial translation given its origin at the proximal tibia (Mokhtarzadeh, 2013). Proper activation and strength of the muscles involved in knee motion is necessary to prevent injury to the lower extremities.

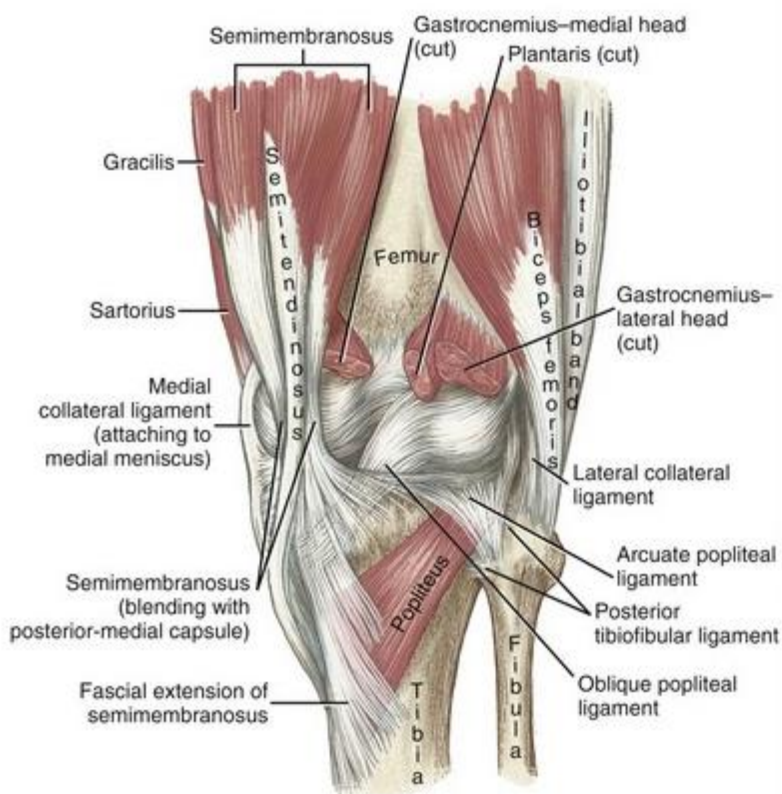

Figure 3: *Posterior View of the Lower Extremity Musculature Surrounding the Knee (Neumann, 2010).*

The musculature of the hip is of importance in ACL injury prevention due to its role in controlling the torso and upper body relative to the lower extremities. Hip musculature, mainly in the coronal plane, is of large concern given improper coronal alignment of the femur can produce unfavorable motion at the knee. Primary hip adductors include the pectineus, adductor longus, gracilis, adductor brevis, and adductor magnus. Primary hip abductors include the gluteus medius, gluteus minimus, and tensor fasciae latae.

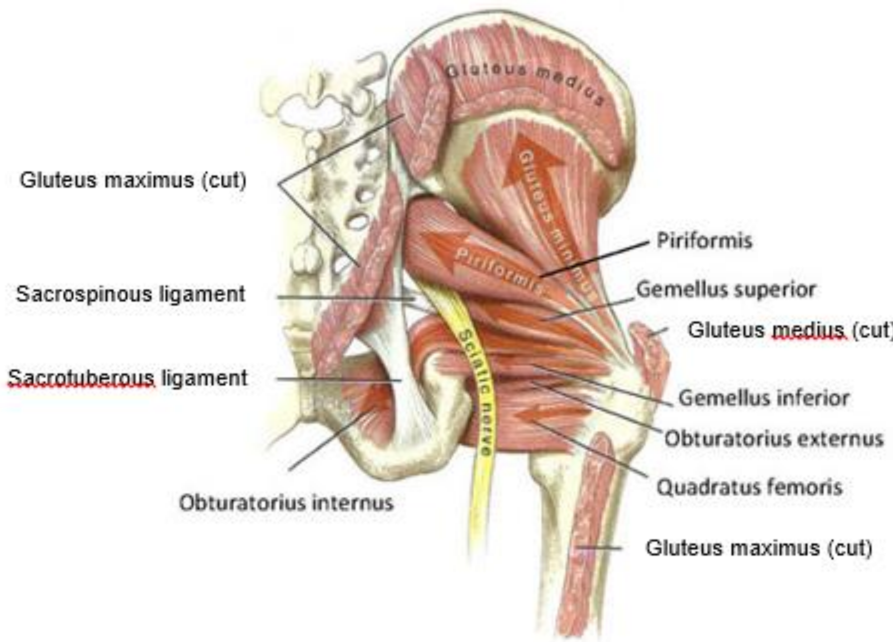

Figure 4: *Posterior View of the Muscles of the Hip Region (Neumann, 2010)*

These muscles are vital in producing hip stability in the coronal plane. Greatest abduction torque output is produced at an adducted angle or in neutral hip alignment when the muscle is longest and decreases with hip abduction resulting in decreased muscle lengths. The hip adductor strength potential is highest when the hip is in a greater degree of hip abduction, placing the adductors at their optimal force production length. Weakness of the hip abductors can contribute to adduction and internal rotation of the femur resulting in a valgus knee angle and externally rotated tibia relative to the femur.

1.2.2 Prevalence

Anterior cruciate ligament injuries compose up to 4.9% of total sports injuries. The greatest incidence of injury occurs in the collegiate athlete population reaching 15 injuries per 100,000 athlete-exposures (LaBella, 2014). Around 70% of all ACL injuries are due to noncontact mechanisms (PJ McNair, 1990). Comparing noncontact ACL injury rates between genders, collegiate females are 2-4 times more likely to suffer injury than are males (Agel, 2005; Arendt, 1999).

1.2.3 Repercussions

Suffering ACL injury results in both long and short term consequences to the athlete. In over 50% of ACL injuries, there is injury to another structure of the knee (Lohmander, 2007). Without surgical intervention, the individual may be forced to deal with instability for the rest of their life. Those having to endure the cost and trauma of surgical intervention, even with modern surgical techniques, may still never return to their pre-injury ability. Due to articular cartilage damage suffered during and after injury, the athlete is ten times more likely to develop degenerative knee osteoarthritis, likely within 10 to 20 years of injury, which will continue to affect them for the rest of their lives (Lohmander, 2007; LaBella, 2014; Levine, 2013).

1.3 Injury Mechanisms

Mechanisms of ACL injury are important to determine prior to understanding risk of ACL injury. Observational analysis of injuries occurring during athletic play have played a vital role in the understanding of mechanisms of injury. Additionally, cadaveric models and three dimensional computer models provide insights into ACL loading when strained in various planes of motion.

1.3.1 Observational Analysis

Observational analysis has been extremely helpful in determining mechanisms of noncontact ACL injury. Although with a limited amount of video sources to evaluate, such studies are hard to come by. Furthermore, these studies are restricted in accuracy by the camera angle and video quality.

Olsen et al. identified two main maneuvers resulting in ACL injury: a plant-and-cut movement and a one-legged jump landing. Injuries during a jump landing typically resulted in a forceful valgus knee angle and an external rotation at the tibia with the knee in a near fully extended position (Olsen, 2004). Through video analysis of in-game situations, knee injuries were compared with similar movements by athletes that did not result in injury. Injured athletes exhibited greater knee abduction (valgus) angles when landing than was observed in uninjured participants (Boden, 2009; Hewett, 2009). Contributing to knee valgus, greater lateral trunk motion has been observed in injured individuals through observational in-game study (Hewett, 2009). Additionally, injured athletes were reported to land in a more flat-footed position with limited ankle range of motion. Injuries also correlated to greater hip flexion at initial contact just prior to injury

(Boden, 2009). These results suggest the need to investigate all joints of the lower extremities to truly understand injury mechanisms and risk.

While these investigations provide insight into injury mechanisms, video analysis is limited due to the amount of quality video sources available for evaluation. In these circumstances, injuries are observed from a single plane making accurate measurement of dynamic motion difficult. Moreover, kinetic evaluation results are limited given the inability to measure ground reaction force during in-game situations.

1.3.2 Cadaveric Models

A closer investigation can be conducted using cadaveric study and may bring into question the validity of video analysis. Various studies have identified differences in mechanical response to loading with changes in loading orientation (Fukuda, 2003; Meyer, 2008) as well as with age (Woo, 1991). For a set of specimen 22-35 years of age, the ultimate load of the ACL when tested at an angle of 30 degree of flexion was found to be 2,160 ± 157 N (Woo, 1991). Ultimate load decreased with age of the specimen tested. When incrementally loading a cadaveric femur-knee-tibia specimen in compression and torsion, pre and post ACL failure was assessed by Meyer et al. When loading in compression, the knee responded with internal tibial rotation and anterior tibial translation prior to rupture at 5.4 kN. After rupture, the knee exhibited external rotation. When a torsional load was applied, failure occurred at 58 degrees of internal tibial rotation at a torque of 33 Nm and after rupture responded with increasing valgus knee angle (Meyer, 2008). While both internal and external rotation have been identified in video analysis of ACL injury (Olsen, 2004), these results suggest that timing associated with a perceived failure from observational analysis and the actual timing of ACL rupture may differ. Dynamics identified to be associated with injury through observation may actually have resulted from post-failure knee instability (Meyer, 2008). Nevertheless, findings of this study continue to suggest transverse and coronal motion involvement in ACL rupture.

Additional research has been focused on assessing cadaveric models during simulated drop landings while reporting knee abduction, internal rotation, and anterior tibial translation (Levine, 2013). The findings of Levine et al. were able to determine increases in ACL strain were significantly associated with knee abduction. They also reported patterns of damage to the tibia plateau cartilage depending on an abduction injury pattern or an internal rotation injury patter (Levine, 2013). Such results are consistent with the suggestion that ACL injury contributes to chronic repercussions due to articular trauma. Given that timing of injury cannot be entirely identified through injury observation, post-injury cartilage damage has the ability to identify specific injury mechanisms. Further investigation of the multiplanar nature of ACL injury is necessary to determine risk factors in loading strategies.

1.4 Injury Risk Factors

Several studies have attempted to identify the risk factors for ACL injury in an effort to detect athletes who are at an increased risk of injury. Once high-risk athletes are identified, preventative training procedures can be implemented. Methods to more easily determine risk in a clinical setting have been suggested (Myer, 2010a; Myer, 2010b; Myer, 2011b) in an attempt to eliminate the use of large, expensive three dimensional motion analysis systems. However, three dimensional motion analysis systems are able to provide incredibly accurate kinematic and kinetic evaluations. Given the accuracy of three dimensional motion capture systems, a method to reducing the large amount of data assessed through motion analysis is desired. Once reduced, a correlation of such results to clinical measures is can be investigated. Injury risk factors defined in the literature range from biomechanical (kinematic and kinetic), neuromuscular, environmental and passive factors such as anatomy, hormonal changes, and age.

1.4.1 Biomechanical: Kinematics

Biomechanical evaluation of kinematics in athletes has been largely used to explore risk of ACL injury. Video analysis and three dimensional motion capture evaluation of controlled tasks in a laboratory setting are used to determine differences between groups with greater risk of injury.

Drop jumps have become an important evaluation tool in determining biomechanical factors relating to ACL injury risk (Earl, 2007; Hewett, 2005). A drop jump begins with an individual standing on a platform. Next, the individual steps from the platform landing on both feet and subsequently performs a maximal height jump. A drop jump maneuver has been widely accepted as a controlled laboratory method to assess a similar scenario as would be seen during athletic competition. Altering ground contact time when landing from a drop has significant effect on mechanics, specifically on sagittal plane kinetics of landing and should be monitored in evaluation (Bobbert, 1987; Young, 1995; Walsh, 2004). While contact time is important to consider, the height at which a subject begins the jump has little effect on resulting power, work, and moments calculated throughout landing (Young, 1995). Variations in drop jumps have become evident in the literature making consistency within a study and between studies essential factors to consider in test design. Such inconsistencies cause altered mechanics as seen through various studies assessing differing types of drop jumps techniques.

In addition, landing unilaterally compared with bilateral landing alters biomechanics. A unilateral tasks produced significantly different kinematic results, including increased valgus angles at the knee (Pappas, 2007b; Nagano, 2008). During unilateral landings, females demonstrated increased valgus angles, at landing with greater vertical ground reaction forces than bilateral landings (Pappas, 2007b). While significant differences were seen between unilateral and bilateral landings, females consistently show increased knee valgus angles and vertical ground reaction forces during both types of landings compared with males. Comparing coronal motion during a step down and double leg drop landings, bilateral drop landings produced greater coronal motion in both genders (Earl, 2007). These results suggest the use of bilateral landing evaluation provides a safe, but effective, dynamic alternative to unilateral assessments when evaluating ACL injury risk, specifically during drop landings performed by females.

1.4.1.1 Video Analysis

Video analysis is able to provide a basic understanding of landing patterns viewing one plane of motion. The simplicity of a video system is desirable but limits the comprehensive analysis that can be obtained through three dimensional motion capture systems. In a normative data set using two dimensional video analysis, Harrington et al. assessed young adult females reporting valgus knee angles ranging from 7-13 degrees at peak knee flexion during a bilateral drop landing and indicated that there was not a significant difference between dominant and non-dominant legs during landing (Harrington, 2010). The group compared results with a population of recreationally active males reporting valgus angles from 3-5 degrees. The difference in coronal motion between male and female athletes completing a controlled landing task is suggested to be a contributing factor to the increase in injury incidence in the female collegiate population.

Reducing the need for large, expensive, equipment to determine ACL injury risk in athletes is desirable. Using predictive algorithms to reduce the many factors contributing to risk of injury has recently increased in popularity. Through the use of video analysis and clinical measurements, Bittencourt et al. was able to identify contributions to frontal plane knee projection angle at landing. Contributions included shank-forefoot alignment, hip abductor isometric torque, and passive hip internal rotation range of motion (Bittencourt, 2012).

1.4.1.2 Three Dimensional Motion Analysis

Biomechanics laboratory technology has largely influenced the method of identifying injury risk. Prospective studies using three dimensional (3D) motion capture have been able to identify subtle differences between those athletes who would later suffer injury and those who did not. Many studies have also been able to accurately determine differences between genders through controlled laboratory evaluation while others have identified age related changes in mechanics also contributing to likelihood of injury. Three dimensional motion analysis is able to quantitatively assess each joint, in each plane of motion to define minute differences in mechanics indeterminable through video observations.

Few prospective studies have been completed due to limitations in subject recruitment and testing prior to ACL injury. Such studies attempt to test large groups of active, healthy individuals using drop jump evaluations at the beginning of an athlete's season. The athletes are then followed throughout the season and injuries are reported. Once injuries take place, the preseason evaluation of injured participants can be compared to the uninjured individuals to determine preseason susceptibility to injury.

Hewett et al. tested 205 adolescent female athletes prior to their athletic season (Hewett, 2005). Throughout the season nine athletes suffered ACL injury. Kinematic results from 3D motion analysis of drop jumps taken prior to the season revealed those suffering injury had 8 degrees greater knee abduction angle at initial contact. Temporal analysis indicated a 16% shorter stance time between initial contact and take-off. Through an injury prediction analysis, Hewett et al. was able to determine injury with 73% specificity and 78% sensitivity. This study shows the immense importance of the relation of coronal dynamics of the knee to ACL injury.

Similar coronal knee motion trends have been found in laboratory assessments comparing males with females. On average, females show a larger valgus motion at the knee when landing (Ford, 2003). Ford et al. suggests a greater chance of injury to the dominant leg of a female given greater valgus knee angles on the dominant side compared to the non-dominant (Ford, 2003). In addition to coronal knee motion, transverse motion at the knee defined as tibial rotation is commonly associated with ACL injury. When landing unilaterally, females show less time to peak internal rotation than do males (Lephard, 2002a). Such findings suggest females may land in a manner providing insufficient time to appropriately dissipate and adapt to the ground reaction force.

Based on the change in ligament laxity due to anatomy throughout the sagittal knee range of motion, reduced knee flexion at initial contact and through landing has been suggested an identifier of ACL injury risk. When controlling knee flexion, low knee flexion angles at landing result in an increased valgus angle, increased internal knee adduction moments and decreased energy absorption at the hip and knee (Pollard, 2010). Knee flexion angle has the ability to alter muscle activation of the hamstrings and quadriceps with the potential to reduce anterior tibial translation due to muscle forces (Pollard, 2010; Podraza, 2010). With females typically exhibiting less knee flexion than males at landing, the connection between injury and these mechanics (extended knee and increasing valgus knee angles) is likely to be a reason for the gender difference in injury rates (Lephart, 2002a).

While motion at the knee may seem the most likely contributor, foot placement, ankle motion, hip motion and torso alignment are also of concern. Neutral coronal alignment of the hip and ankle allows the knee to sustain greatest load without injury (Chaudhari, 2006). However, in the sagittal plane, increased flexion of the hips is suggested to reduce risk of injury by forcing a greater knee flexion and a hamstring demand (Shimokochi, 2012). This is not consistent with the landing pattern chosen by females as was reported by Decker et al. When landing, females choose a more erect body position than males (Decker, 2003). Further, recent findings suggest pelvic anterior tilt induces femoral internal rotation throughout hip flexion range of motion (Bagwell, 2015). Excessive internal rotation of the hip is potentially damaging to the ACL. Training of proper landing technique should take these findings into account to prevent potentially dangerous internal rotation of the femur.

Further focus has been placed on the relation of landing kinematics to foot-landing techniques. Such analysis is important given the foot placement at contact could create instability or alter kinematics at the knee. Foot landing technique changes have been suggested to be correlated to kinematics by reducing hip flexion with a forefoot impact and decreasing knee flexion and increasing valgus with rear foot impact (Cortes, 2007). A recent study investigating change in toe direction (foot progression angle) on resulting drop vertical jump landing mechanics revealed with the feet internally rotated, subjects demonstrated increased knee abduction angle, tibial internal rotation, and knee abduction moment (Ishida, 2015). Given these results, toe direction may be an important factor in favorably altering at-risk mechanics as well as the foot progression angle throughout landing.

1.4.2 Biomechanical: Kinetics

Kinetic patterns of landing are important to investigate force, moment and energy propagation through the lower extremities. Few prospective studies exist that incorporate kinetic evaluation. One study with 205 adolescent athletes revealed a 2.5 times greater knee abduction moment and a 20% greater ground reaction force in the injured population (Hewett, 2005).

Anterior-posterior forces on the knee are of great concern given the main role of the ACL in knee joint stability is to resist anterior translation of the proximal tibia or posterior translation of the distal femur. When comparing anterior-posterior shear forces between previously injured women and uninjured women, women with previous ACL reconstruction had significantly less anterior-posterior shear force than uninjured women likely due to increased co-contraction of the quadriceps and hamstrings upon single-leg drop landing (Ortiz, 2008). While anterior-posterior shear differed, similar joint angles at the hip and knee were produced suggesting the need for kinetic analysis.

Sagittal plane joint moments are of concern given the role of the quadriceps on anterior force at the proximal tibia. Overcoming this force must be done by the ACL as well as through hamstring co-contraction. Internal knee extensor moments have been correlated to sagittal plane ankle and hip moments (Shimokochi, 2009). Less knee extensor moment was produced through landing mechanics that demonstrated greater torso flexion. The results of this study suggest that leaning forward at landing would produce favorable mechanics, as seen through joint moment evaluation, and knee stability by increasing activation of the hamstrings and thus reducing likelihood of injury (Shimokochi, 2009).

In-game observational data clearly suggests importance of coronal and transverse kinematics, moments applied in these two planes are just as important to injury. While transverse and coronal knee moments apply risky joint mechanics independently, these factors combined create a much greater risk of injury (Shin, 2011). Based on a 3D computer model, the greatest strain on the ACL has been found to occur with increased valgus and internal rotation moments at the knee (Shin, 2011).

Lower extremity energetics are an important factor in assessing ACL injury mechanics. If the musculature is unable to adequately absorb impact energy, passive structures of the musculoskeletal system (i.e. ligaments) must absorb energy exposing them to greater risk of injury (Norcross, 2010). Females have been identified as having a preferred landing strategy of greater erectness at initial contact and an increased sagittal range of motion of the ankle and hip joints (Decker, 2003). Such a landing pattern allows females to absorb greatest energy through the distal joints; the knee and ankle. This study revealed both the male and female populations used the knees as a primary shock absorber, the female population secondarily utilized the ankles while the male population chose the hips as a secondary energy absorber (Decker, 2003). Even though both groups use the knee as a primary absorber, females have been identified as absorbing more energy at the knee than do men (Schmitz, 2010). During the impact phase of a jump landing, Norcross et al. reported more favorable biomechanics result from greater energy absorption at the knee with less absorbed at the hip and ankle (Norcross, 2010).

1.4.3 Muscular (Strength and EMG)

While biomechanics may be the most obvious display of injury risk prediction, the musculature of the lower extremities are the control units of the skeletal system allowing for mechanics to occur. Muscular strength, activation, and timing of the lower extremity muscles may contribute to injury and have been extensively studied. A study by Ortiz et al. represents the importance of muscular contributions. This study compared muscle activity of the quadriceps and hamstrings of ACL injured and healthy individuals during a single leg landing. While muscle activation differed significantly between groups, joint angles did not (Oritz, 2008).

Quadriceps and hamstring contribution is of principal focus as these muscles are the primary controllers of knee extension and flexion. Through investigation of strength Lephart et al. demonstrated weaker thigh muscles normalized to body weight in a female population resulting in a more rapid stiffening of the knee when landing compared with males (Lephart, 2002a). Additionally, females have been found to be more susceptible to anterior tibial translation due to muscle activation. Females tend to land with a lower hamstring activation and greater quadriceps activation leading to a quadriceps dominant landing strategy (Urabe, 2005; Ebben, 2010). Additional study suggests greatest anterior shear force occurs with greater peak quadriceps activation and knee extension moments (Schultz, 2009). Quadriceps dominance in females is of concern because with a greater activation of the quadriceps muscles, specifically in an extended knee position, an anterior force is placed at the proximal end of the tibia causing stress on the ACL. When in an extended position, the ACL and MCL are taut thus decreasing injury threshold. In this position any force absorbed by the ligaments due to coronal motion leaves the ligaments at greater risk of injury. Knee extension strength has been shown to predict increased energy absorption at the knee while a greater knee flexion strength was indicative of greater hip torsional strength (Schmitz, 2010). Schmitz et al. stated "large proportion of variance in lower extremity mechanics was not explained by thigh strength" indicating a need for further investigation of the variance of lower extremity mechanics related to lower extremity musculature.

In addition to strength and activation, timing of muscle activation is important in determining factors contributing to injury. Men compared with women have shown to activate the quadriceps (vastus medialis and vastus lateralis) earlier prior to ground contact when landing from a jump. Additionally, after contact men demonstrate a greater hamstring to quadriceps activation ratio compared to women (Ebben, 2010).

While few studies have focused on the musculature of the lower leg in contributing to ACL injury risk and injury prevention, investigation of the gastrocnemius and soleus has suggested importance. A study using OpenSimm to model the muscle forces of the hamstrings, quadriceps, gastrocnemius, and soleus indicated that during a single leg landing, the soleus is able to contribute nearly a third of the posterior muscle force applied to the tibia (Mokhtarzadeh, 2013). The large amount of posterior force applied by the soleus is intriguing given the vulnerability of the ACL to injury when the tibia is translated anteriorly. Athletic training may be able to alter athletes pre-landing and landing strategies to incorporation proper activation of the soleus, in addition to other lower extremity muscles, to prevent anterior translation at detrimental amounts.

Hip abduction and adduction musculature is important to consider given the extensive amount of literature suggesting coronal motion as an injury risk factor. Such effects are supported by an investigation of anticipatory muscle contraction of the hip abductors/adductors that indicated a reduction in coronal motion at the knee in an effort to prevent injury to the ACL (Chaudhari, 2006). Increased peak eccentric abduction torque of the hip has shown to be correlated with lower peak valgus angles. This suggests strengthening of the hip abductors could reduce change in knee valgus angles and reduce risk of injury (Jacobs, 2005). Additionally, hip abductor peak torque in adult females has been found to be less than that of adult males when normalized to body weight and height. Peak torque in females has shown to be moderately correlated with hip flexion, hip adduction and knee valgus peak joint displacement when landing a jump suggesting hip abductor strength plays an important role in neuromuscular control for women (Jacobs, 2007). A prospective study of competitive athletes identified that abductor hip isometric strength tested using a handheld dynamometer with the leg at 30 degrees of abduction was able to predict ACL injury in a mixed male and female population (Khayambashi, 2016). In addition to strength assessments of the hip abductors/adductors, electromyography (EMG) activity has provided vital information for determining differences in neuromuscular control between genders. Carcia et al. investigated gluteus medius EMG activity in females noting greater variability than males when landing. However, no gender difference was evident in mean and peak amplitude before and after landing (Carcia, 2007).

1.4.4 Passive Characteristics (Hormonal, Anatomical, Age)

Several other unalterable factors have been discussed in the literature. These “passive” factors may include but are not limited to hormonal changes, age related changes, and anatomical differences.

Onset of puberty has been largely investigated to understand why female injury rates increase after puberty. Swartz et al. was unable to identify gender differences in landing mechanics but was able to recognize a difference in landing patterns between children aged 9.5 years and adults aged 24 years (Swartz, 2005). This study suggests children produce greater knee valgus angles, less hip flexion, less knee flexion at max ground reaction force and greater maximum ground reaction force (Swartz, 2005). While these results indicate a possible higher risk of injury in a population of children, this is inconsistent with the incidence of injury relative to age in the general population (LaBella, 2014). Focusing on a female pre- and postpubescent comparison, the postpubescent population was found to produce decreased knee flexion angles at initial contact, increased medial-lateral knee forces and decreased knee extension moments during landing (Hass, 2005). In addition, hormonal changes throughout menses are a possible contributor to female increased risk by altering neuromuscular control patterns and altering muscle activation timing in landings (Derick, 2008). A rapid increased in injury incidence of females soon after the age of puberty onset appears to relate to hormonal changes and likely the inability to adapt to neuromuscular control patterns experienced throughout menses.

Anatomical factors have been suggested in the literature as a clinically measureable factor that could contribute to ACL injury risk. The anatomical structure of the knee joint itself has been suggested as a factor. The lateral wall of the intercondylar notch has been a suggested location of ACL impingement when the joint is in an externally rotate and abducted position (Park, 2005). A smaller intercondylar notch, typically found in females, may cause greater risk of impingement in the female population. An investigation of the static posture and anatomical build of ACL injured females indicated a greater occurrence of genu recurvatum, increased navicular drop and excessive subtalar joint pronation compared with uninjured females (Loudon, 1996; Beckett, 1992). While several factors have been suggested, investigators recognize static characteristics can be overcome during dynamic movements (Chaudhari, 2006). Women have been found to have greater joint laxity, lower joint stiffness and greater energy loss when evaluating tibial rotation compared with males suggesting a possible cause of increased injury incidence in females (Park, 2005). Schultz et al. also investigated joint laxity tying the results to jump landing technique. This study indicated individuals with increased anterior knee laxity, general joint laxity and decreased genu recurvatum demonstrated increased energy absorption, increased knee stiffness, and decreased ankle stiffness when landing from a jump (Schultz, 2010). Further joint stiffness characterization at the ankle has suggested a relation of dorsiflexion flexibility to the kinematics and kinetics of landing (Malloy, 2014).

1.5 Injury Prevention

Prevention of ACL injury has traditionally included strengthening of lower extremity muscles to stabilize the knee. Instructional training with self-observation and expert feedback may be able to provide long term altered mechanics to reduce injury risk (Barber-Westin, 2010; Etnoyer, 2013). Additional neuromuscular training programs have shown promise to dynamically alter mechanics in female athletes (Hewett, 1996; Hewett, 1999; Myer, 2005; Noyes, 2005).

1.5.1 Strength Training

Strength training procedures have been suggested to decrease risk of ACL injury. While this remains a popular intervention tool, studies have shown that strength training alone may not be sufficient to alter biomechanics. A strength training protocol implementing quadriceps, hamstring, gluteus medius and gluteus maximus strengthening was unable to alter mechanics of a stop-jump task (Herman, 2008).

1.5.2 Neuromuscular Training

Evaluation of training techniques suggests neuromuscular training can reduce risk of ACL injury in predisposed individuals. Neuromuscular training may be able to improve the ability to dynamically stabilize the knee which has been attributed to the increased risk of ACL injury in females (Ford, 2003). Immediate changes in mechanics have been suggested as well as long term, sustained reduced risk. Through video analysis of drop jump landings as an assessment tool, a neuromuscular training regimen was able to show increased knee separation in females suggesting a reduced knee abduction at landing (Noyes, 2005). Additionally, flexion range of motion has been altered resulting in a decreased exposure to torque in the coronal plane (Myer, 2005). High school athletes were able to sustain improvements gained through neuromuscular training when retested 12 months after implementation of training as indicated by an improved knee alignment measured through knee separation distance (Barber-Westin, 2010). While neuromuscular training programs appear to be a promising method of injury prevention, a technique to clearly determine which athletes require training has not been defined.

1.6 Principal Component Analysis

Principal component analysis (PCA) is a statistical method of multivariate analysis used to reduce data set dimensionality (Jolliffe, 2002). PCA has several applications including its use in interpreting and reducing data obtained through gait analysis (Carriero, 2008; Krzak, 2015). PCA has been shown to be a useful tool in objectively determine the most relevant parameters that should be used in traditionally subjective clinical tests to determine knee stability after ACL injury (Labbe, 2010). Further, PCA has been implemented in the determination of which force and temporal variables are most valuable in the prediction of jump height (Laffaye, 2014). With the large amount of parameters that have been previously identified as ACL injury mechanisms and risk factors, PCA may be able to provide an objective method to determining the most influential variables obtained from motion analysis.

1.7 Purpose of Study

Given the state of current ACL injury risk investigation, it is evident there are several contributing factors that must be considered when evaluating an athlete for injury risk. Current investigators and clinicians are in need of an all-encompassing method of interpreting the vast array of variables obtained from three dimensional motion capture evaluation to more easily identify ACL injury risk. This study attempts to identify such a method through evaluation of bilateral drop jump landings in a physically active, young adult female population. Using principal component analysis (PCA), the most salient variables, representing the variance of the entire data set, can be isolated. This study characterizes hip abduction/adduction isometric and isokinetic strength as well as knee flexion/extension isometric and isokinetic strength. Additionally, investigation of the association between principal component scores and lower extremity strength measures are defined to determine clinically relevant strength assessments for identification of ACL injury risk.

2 Methods and Materials

2.1 Participants

Twenty healthy, recreationally active young adult females (Table 1) provided written informed consent to participate in the study protocol as approved by the Medical College of Wisconsin's (MCW) Institutional Review Board (IRB). Participants were screened to ensure no previous knee injury and no current lower extremity injuries. Testing took place at the MCW Center for Motion Analysis (CMA).

Table 1: *Subject Demographic Data including age (years), height (mm), weight (kg), and hours of activity participated in weekly (hours/week).*

	Age (years)	**Height (mm)**	**Weight (kg)**	**Activity (hours/week)**
Mean	21.0 ± 1.78	1698 ± 55	63.7 ± 5.48	8.0 ± 3.7
Min	18	1610	57	2
Max	24	1785	74	15

2.2 Procedures

2.2.1 Subject Preparation

Anthropometric measurements were obtained including height, weight, inter-anterior superior iliac spine (ASIS) distance, bilateral leg length, bilateral knee width, and bilateral ankle width. Participants were affixed with seventeen reflective markers (Table 2) to coincide with a modified Helen Hayes marker set used for the Plug-in Gait model (Vicon; Oxford Metrics Ltd., Oxford, England).

Table 2: *List of Markers used in dynamic trials for Plug-in Gait model*

Marker Name	**Location**	**Description**
LASI	Left ASIS	Placed directly over the left anterior superior iliac spine
RASI	Right ASIS	Placed directly over the right anterior superior iliac spine
SACR	Sacral Marker	Placed mid-way between the posterior superior iliac spines
LTHI	Left Thigh	Placed on the lateral surface of the left thigh, along the femur
RTHI	Right Thigh	Placed on the lateral surface of the right thigh, along the femur
LKNE	Left Knee	Placed on the lateral epicondyle of the left knee
RKNE	Right Knee	Placed on the lateral epicondyle of the right knee
LTIB	Left tibia	Placed on the lateral left shank, along the tibia
RTIB	Right tibia	Placed on the lateral right shank, along the tibia
LANK	Left Ankle	Placed on the lateral malleolus of the left ankle
RANK	Right Ankle	Placed on the lateral malleolus of the right ankle
LTOE	Left Toe	Placed on the shoe over the second metatarsal head of the left foot
RTOE	Right Toe	Placed on the shoe over the second metatarsal head of the right foot
LHEE	Left Heel	Placed on the shoe over the calcaneus at the same height as the left toe marker
RHEE	Right Heel	Placed on the shoe over the calcaneus at the same height as the right toe marker

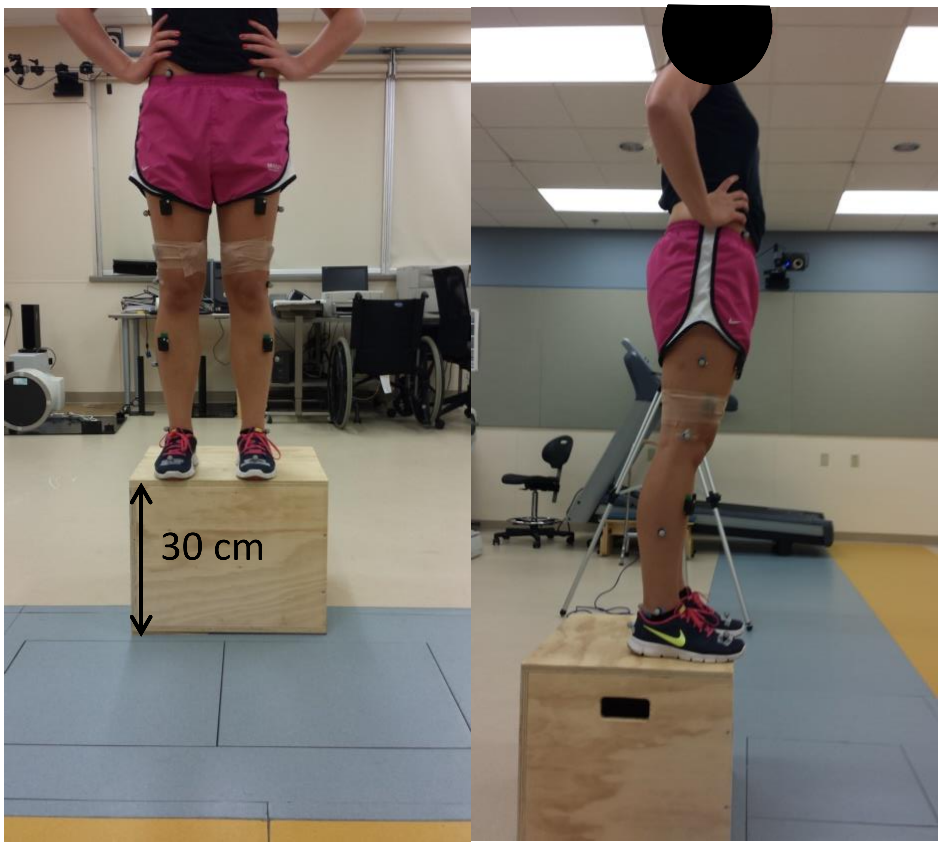

Figure 5: *Marker set and marker locations used in Plug-in Gait model.*

Markers were placed bilaterally on the head of the second metatarsal, calcaneus, lateral malleolus, lateral shank, lateral femoral epicondyle, and thigh (Figure 5). To track pelvis motion and locate the hip joint center, markers were placed on the left and right ASIS and on the superior sacrum in line with the posterior suprerior iliac spines. Additionally, subjects were outfitted with wireless surface EMG electrodes (Delsys, Inc., Natick, Massachusetts). Sensors were placed bilaterally at the anterior tibialis, medial gastrocnemius, rectus femoris, medial hamstring, and gluteus medius. Each subject was tested wearing their own athletic shoes. The metatarsal and heel markers were taped to the shoes over the anatomical landmarks mentioned above.

2.2.2 Motion Analysis

Motion capture was performed using a Vicon MX twelve-camera, three-dimensional motion analysis system (Vicon; Oxford Metrics Ltd., Oxford, England) at a sampling frequency of 250 Hz. Ground reaction forces were obtained through integration of force plates (Bertec Corporation,Columbus, OH) embedded in the floor sampling at 3000 Hz (Figure 6).

Figure 6: *Motion Capture Laboratory Twelve Camera Setup with 30 cm plyometric box placed just behind floor-embedded force platforms.*

Three-dimensional motion data was collected while the subject performed a drop jump task (Figure 7). Each participant was asked to begin standing on a 30 cm plyometric box then step from the box landing on both feet at the same time with one foot on each force plate. Directly after landing, the subject performed a maximal height jump. Throughout the jump, the subject was required to maintain her hands at her waist.

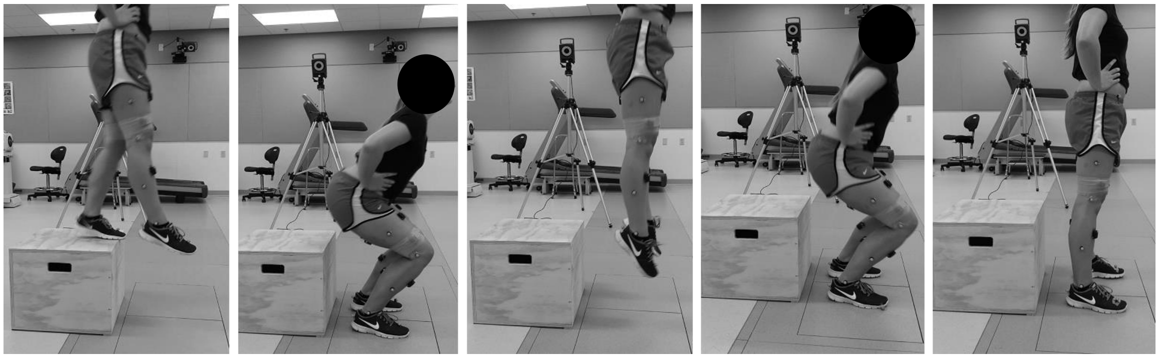

Figure 7: *Double leg drop jump test progression*

Prior to motion capture, subjects were given a demonstration of the activity and allowed practice jumps to become familiar with the procedure. Ten trials with clean bilateral force plate landings were taken of each subject. If the subject failed to produce a clean force plate strike or hands were removed from her waist, the subject was asked to redo the trial. Marker positions and force plate data were run through Vicon's Plug-In-Gait model for kinematic and kinetic results throughout each drop jump. Data from the first 100 ms of landing, also considered the impact phase, was used for analysis. Points of interest included initial contact (IC), peak knee abduction moment (KAM) and 100 ms after IC (the end of the impact phase). Initial contact was chosen as a time point of interest given the consistencies in observed mechanics of injured athletes at landing (Hewett, 2009; Boden, 2009). Position at IC sets the lower extremities up for load acceptance at landing. If poor alignment exists at IC, kinetics up the kinematic chain can be detrimental. Peak KAM is of interest given results from Hewett et al. suggesting greater peak KAM during drop landings in individuals subsequently suffering injury

(Hewett, 2005). An understanding of the factor that contribute to such large moments is desired. “It appears that increased valgus motion and valgus moments at the knee joint during the impact phase of jump-landing tasks are key predictors of an increased potential for ACL injury in females” (Hewett, 2005). The “impact phase” was chosen as the time frame over which the data would be evaluated given previous study (Devita, 1992; Schot, 1994; Weinhandl, 2011). The impact phase is defined as the first 100 ms after initial contact. This has been identified as a region after two vertical ground reaction force peaks, once the vertical ground reaction force plateaus. Minimal change in the vertical ground reaction force would result in little variation in the kinetic results. Further, through video analysis of ACL injuries occurring during in-game situations, the estimated time of injury occurred between 17 and 50 ms after initial contact (Krosshaug, 2007). Through preliminary evaluation of the data, this region is sufficient for characterizing mechanics of drop landing.

2.2.3 Biodex Strength Assessment

After motion capture, an isometric and isokinetic strength assessment was performed using a Biodex System (Biodex Medical Systems, Shirley, New York). Strength evaluation was done on bilateral knees in flexion and extension and bilateral hips in adduction and abduction.

Knee flexion/extension was tested from a seated position where the subject was secured to the Biodex chair sitting upright such that the knees were allowed to freely flex and extend (Figure 8).

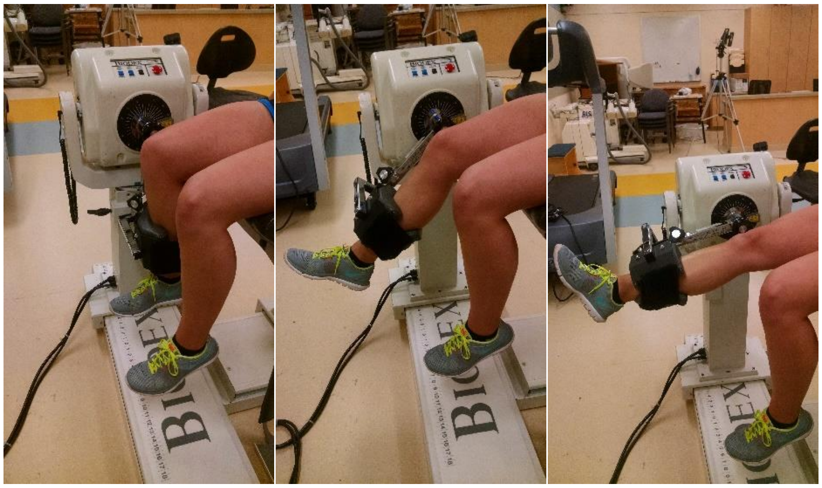

Figure 8: *Biodex set up for knee flexion/extension at 90, 60, and 30 degrees of flexion*

Prior to testing, the subject's range of motion at the knee in flexion/extension was measured and used to set software stops. Flexion and extension of the knee was tested isometrically at 90, 60, and 30 degrees of flexion relative to each subject's maximum extension. Subjects were instructed to apply maximum force against the test arm for 5 seconds in extension first, then were given 5 seconds to relax. After the rest period, maximum force in flexion was applied to the test arm by the subject, then was given a 5 second rest period. The process was then repeated five times and the average peak torque of the five trials was obtained. Between isometric tests, the subject was given five minutes to rest to avoid fatigue. Isokinetic tests were done at a rate of 60, 75 and 90 deg/s. Testing began with concentric quadriceps contraction through the subject's range of motion then immediately following, a concentric hamstring contraction was tested. Five repetitions were performed at each rate with a five minute rest period between each test to avoid fatigue.

Subsequently, strength evaluation was performed on bilateral hips in adduction and abduction from a standing position (Figure 9). Subjects were allowed to hold the top of the Biodex system to maintain balance but were instructed to refrain from using it gain leverage against the test arm.

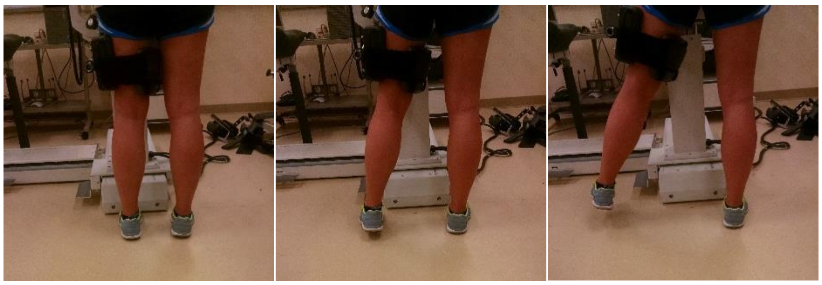

Figure 9: *Biodex set up for hip adduction/abduction at 0, 15, and 30 degrees of abduction*

Prior to testing, the subject's range of motion at the hip in abduction/adduction was measured and used as a software safety stop. Isometric strength was evaluated at 0 degrees (standing upright), 15 degrees and 30 degrees of abduction. Similar to knee flexion extension isometric tests, five abduction and adduction repetitions were performed with five seconds between abduction and adduction tests. With change in test angle, the subject was given 5 minutes of rest to prevent fatigue. If the subject's range of motion did not include the preselected test angle in isometric tests, the subject was excused from that test (one case). Isokinetic strength was measured against the test arm at a rate of 60, 75 and 90 deg/sec. Testing began with concentric abductor contraction through the subject's range of motion then immediately following, a concentric adductor contraction was performed. Five repetitions were performed at each rate with a five minute rest period between each test to avoid fatigue.

2.3 Statistics

Motion capture data was reduced through the statistical method of PCA using SPSS software (SPSS, Chicago, Illinois) and Matlab (MathWorks, Inc, Natick, Massachusetts). Subsequently, an association analysis was performed between principal components and isometric hip adduction and abduction at each angle tested. Power of significant results was then performed.

2.3.1 Principal Component Analysis

Principal component analysis (PCA) is a method of multivariate analysis used in data reduction (Jolliffe, 2002). This method of analysis converts a data set of variables using an orthogonal transformation into linearly uncorrelated principal components. The number of principal components produced is less than or equal to the original number of variables. Principal components are produced such that the first principal component (PC1) represents the greatest amount of variance of the original data set and is quantitatively represented by an eigenvalue. The following principal components represent the maximum amount of variance possible while adhering to the constraints of the previous components. The following components attempt to represent the next largest percent of variance until reaching the same number of principal components as the number of variables in the initial data set. With each successive principal component less data variance is represented. The results of PCA provide a component loading score for each variable in the data set that represents the correlation between that variable and the principal component. Variable with greatest correlation (loading scores magnitude > 0.4) are retained in that component. Any variables with significant loading on more than one component are said to be complex and are removed from the analysis. PCA is valuable in data reduction as it is able to keep information contained in the entire data set while reducing the number of variables of interest.

The first step in PCA is the creation of a correlation matrix. For this data set a correlation matrix is used (as opposed to a covariance matrix) because the variables within the data set have different measurement units (Jolliffe, 2002). With different units of measurement, the magnitude of measurement values differ and thus variance within that data set differs. Using the correlation matrix eliminates variance difference due to variable magnitude by centering the group mean at zero and scaling the data set from -1 to +1. For PCA to be a justified method of factor analysis the entire data set must have a Kaiser Meyer-Olkin measure of sampling adequacy (Kaiser, 1974) greater than 0.5 as well as demonstrate Bartlett's Test of Sphericity significance ($p>0.001$) (Bartlett, 1954). PCA is an iterative process requiring data elimination if specific retention criteria are not met. The following retention criteria were used: (1) variables used in PCA must show some correlation between one another, (2) communalities extraction values must indicate a 50% variance representation by all retained variables ($h^2 \geq 0.5$), and (3) variables retained must show simple structure as determined by weighting scores of great than 0.4 or less than -0.4 on only one principal component (Jolliffe, 2002; Krzak, 2015). If any retention criteria were not met by a specific variable, that variable was removed and PCA was re-run. This process was repeated until all retention criteria were met. The final number of principal components was determined based on the number of components with eigenvalues greater than 1. Variables with weighting scores less than 0.4 and greater than -0.4 loading on any principal component were removed form that principal component.

In this study, PCA was performed on dynamic motion results of previously determined important factors contributing to ACL injury including coronal hip angle, coronal knee angle, coronal knee moment, and transverse knee moment. Additionally, factors hypothesized to contribute to ACL injury were included in PCA including base of support and foot progression angle. PCA was performed on the set of six variables at initial contact (IC), at peak knee abduction moment (KAM) and at the end of the impact phase of the landing (100 ms after IC). Following PCA, PC scores were calculated for each sample taken. PC scores were used to assess associations between motion capture data and strength assessment results.

To calculate a PC score, a z-score was first calculated on the raw kinematic and kinetic data set (D'Agostino, 2006). This was done to standardize the data and eliminate magnitude differences in kinematic and kinetic data. Next, a weighted sum was calculated using component loading scores multiplied by the z-score for each variable contained in a given principal component. The following equation was used for PC score calculation (Krzak, 2015; Jolliffe, 2002):

$$PC\ Score_{ij} = \sum_{k} X_{ik}\alpha_{jk}$$

Where, PC $Score_{ij}$ is the score for the *i*th person and *j*th principal component. X_{ik} represents the z-score of the *k*th original variable while α_{jk} represents a matrix weighting score coefficients. This calculation results in a weighted sum of z-scores for each component for each sample based on variable weight represented in a specific principal component.

2.3.2 Association Analysis

Subsequently, Pearson correlation coefficients of principal component scores for each component and isokinetic strength ratio results were calculated and a p-value was obtained. Pearson correlation is useful in determining linear correlation between variables. Assumptions made when using a Pearson correlation are that the data is normally distributed and is linearly related.

3 Results

Data collected consisted of the results from motion capture evaluation using a standard lower extremity model. These included kinematic and kinetic results measured at the ankle, knee and hip of bilateral lower extremities. Strength measures of the knee flexor/extensors and hip adductors/abductors were also collected. Data was collected bilaterally. Initial data evaluation revealed insignificant differences between left and right legs of the twenty participants. Final evaluation was performed on collected data independent of leg side resulting in a sample size of 34 legs after outlier removal. Statistical results include PCA and association analysis between PC score and strength measures.

3.1 Motion Capture

3.1.1 Kinematics

Kinematic results of each joint of the lower extremities were determined throughout the impact phase of a drop jump landing. Each joint was evaluated in the sagittal, coronal and transverse planes. Additionally, base of support and foot progression angle are reported.

3.1.1.1 Base of Support

Base of support was determined using distance between the centers of pressure on each force plate (Figure 10). At initial contact, mean base of support was 488 ± 72 mm. At peak KAM, base of support reached 559 ± 61 mm and at the end of the impact phase reached 564 ± 45 mm.

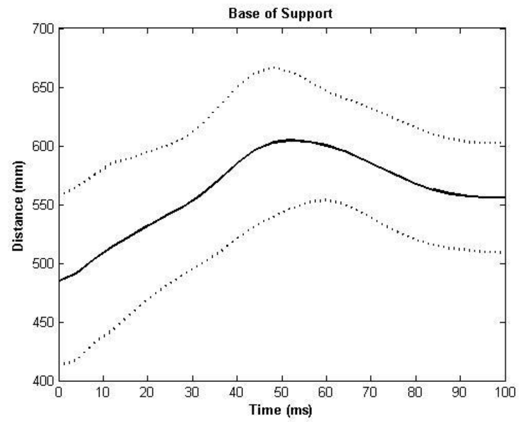

Figure 10: *Base of support (mm) group mean (solid line) ± one standard deviation (dashed line) plotted against impact phase time from 0-100 ms.*

3.1.1.2 Foot Progression Angle

Foot progression angle is the angle of the foot relative to direction of forward progression (Figure 11). In the present study, the direction of motion is in the anterior direction. Group mean foot progression throughout the entire impact phase was internally rotated at an angle of 3.4 ± 4.7 degrees, 5.5 ± 6.1 degrees and 9.9 ± 6.3 degrees at IC, peak KAM and 100 ms, respectively.

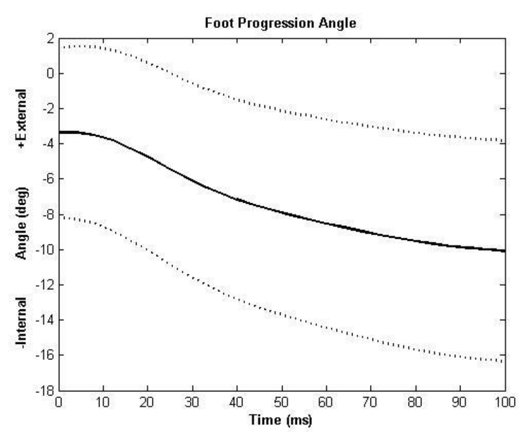

Figure 11: *Foot progression angle (deg) group mean (solid line) ± one standard deviation (dashed line) plotted against impact phase time form 0-100 ms.*

3.1.1.3 Joint Angles

Each joint of the lower extremities were kinematically evaluated in each plane of motion (Figure 12). At initial contact coronal hip angle was adducted and remained so throughout the impact phase. Group mean coronal hip adduction at IC, peak KAM, and 100 ms were 9.5 ± 4.4 degrees, 9.9 ± 4.9 degrees, and 9.6 ± 6.0 degrees, respectively. Group mean coronal knee angle remained abducted throughout landing at an angle of 6.5 ± 5.1 degrees at IC, 6.9 ± 6.4 degrees at peak KAM, and 4.6 ± 9.4 degrees at 100 ms. Mean and standard deviation of all other mean kinematic values at points of interest can be found in Appendix A.

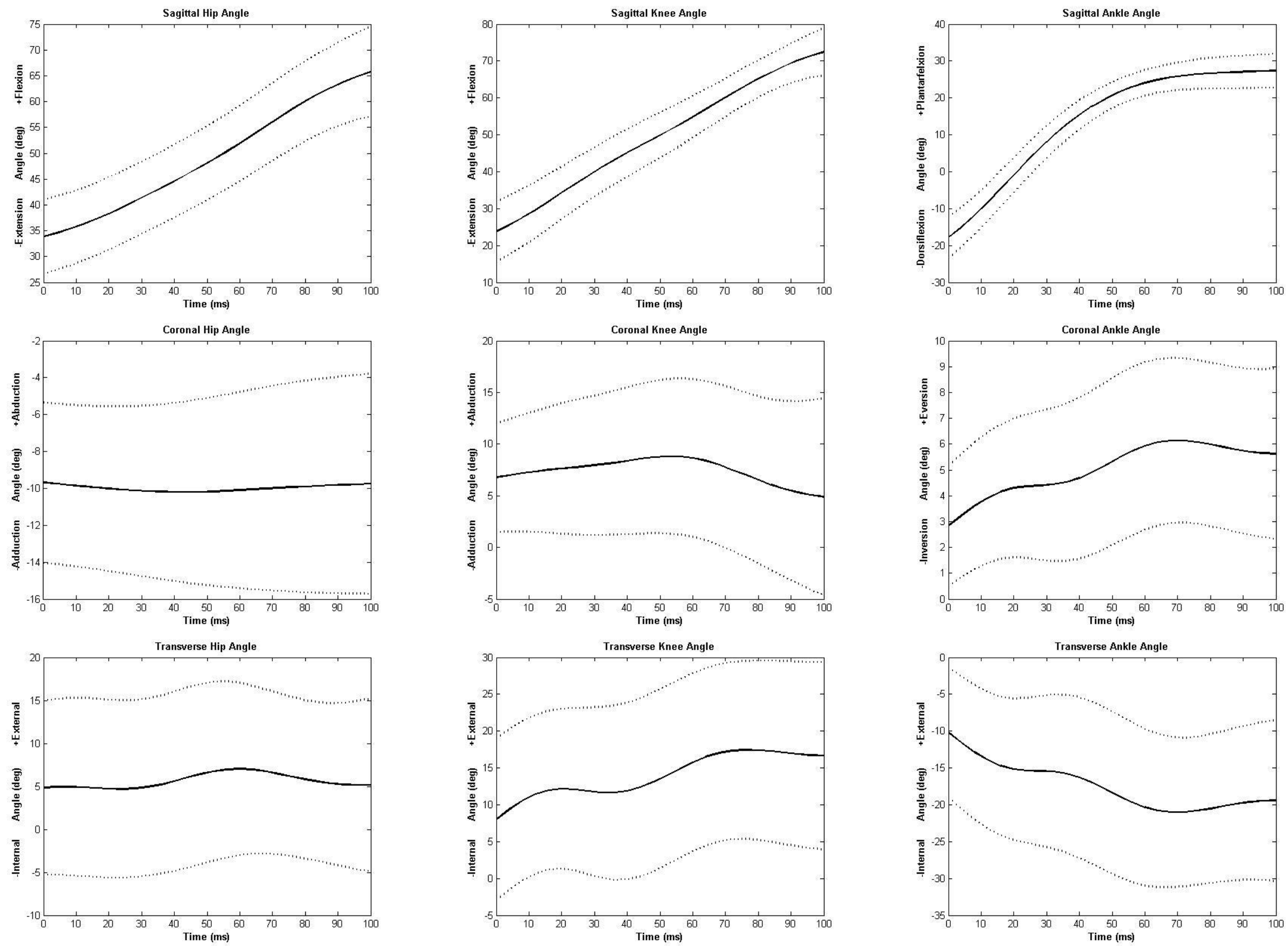

Figure 12: *Group mean (solid line) and ± one standard deviation (dashed lines) of joint angles (deg) of the hip (left column), knee (middle column), and ankle (right column) in the sagittal (top row), coronal (middle row), and transverse (bottom row) planes. The joint angles are plotted verse impact phase time from 0-100 ms.*

3.1.2 Kinetics

Kinetic results for each subject were determined. Mean and standard deviation of each variable, regardless of leg side and dominance was calculated for the population of active, young adult females. Results presented below include ground reaction force, joint forces in three planes of motion, external joint moments in three planes of motion and joint power.

3.1.2.1 Ground Reaction Force

Group mean vertical ground reaction force at peak KAM was 629 ± 373 N. At 100 ms, vertical ground reaction force reached a value of 711 ± 241 N (Figure 13). The group mean vertical ground reaction force curve exhibited two peaks with the minimum value between occurring just prior to peak KAM.

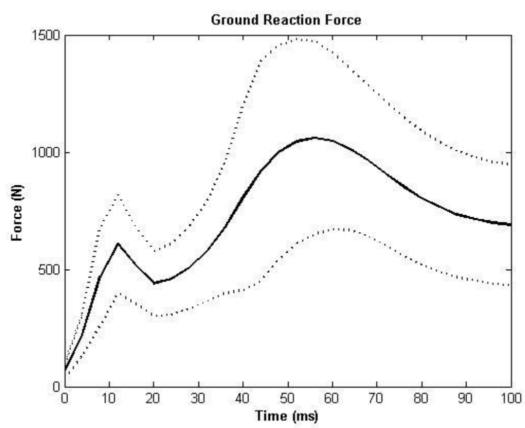

Figure 13: *Group mean (solid line) and ± one standard deviation (dashed lines) of vertical ground reaction force (N). Vertical ground reaction force is plotted verse impact phase time from 0-100 ms.*

3.1.2.2 Joint Forces

Forces on each joint were determined using Vicon's Plug-in Gait model. Mean group force plots can be found in Figure 14.

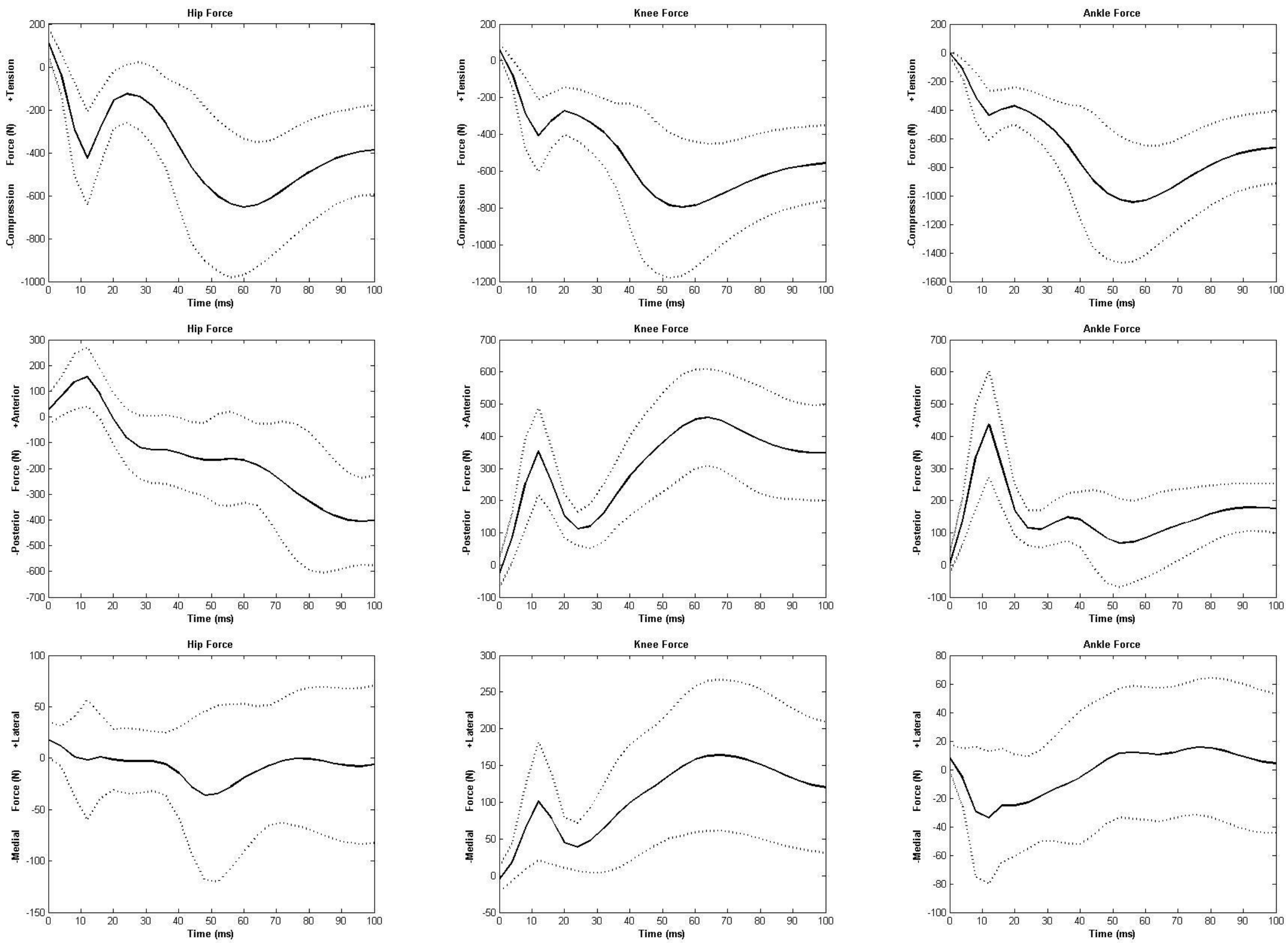

Figure 14: *Group mean (solid line) and ± one standard deviation (dashed lines) of joint forces (N) of the hip (left column), knee (middle column), and ankle (right column) in tension/compression (top row), anterior/posterior shear (middle row), and medial/lateral shear (bottom row) directions. The joint forces are plotted verse impact phase time from 0-100 ms.*

3.1.2.3 Joint Moments

Applied moments on each joint were determined through three dimensional motion analysis and plotted versus time for the impact phase (100 ms) (Figure 15). Average coronal knee moment at IC was an adduction moment of 2.0 ± 6.0 Nm. Peak KAM occurred on average at a value of 10.9 ± 9.4 Nm in abduction. At 100 ms, an adduction moment of 20.5 ± 20.5 Nm was applied to the knee. Average transverse knee moments applied in internal rotation to the knee at IC and 100 ms were 2.3 ± 2.2 Nm and 1.5 ± 4.1 Nm, respectively. At peak KAM an average external rotation knee moment of 1.0 ± 4.4 Nm was applied to the knee. Mean and standard deviation of all other mean moment values at points of interest can be found in Appendix A

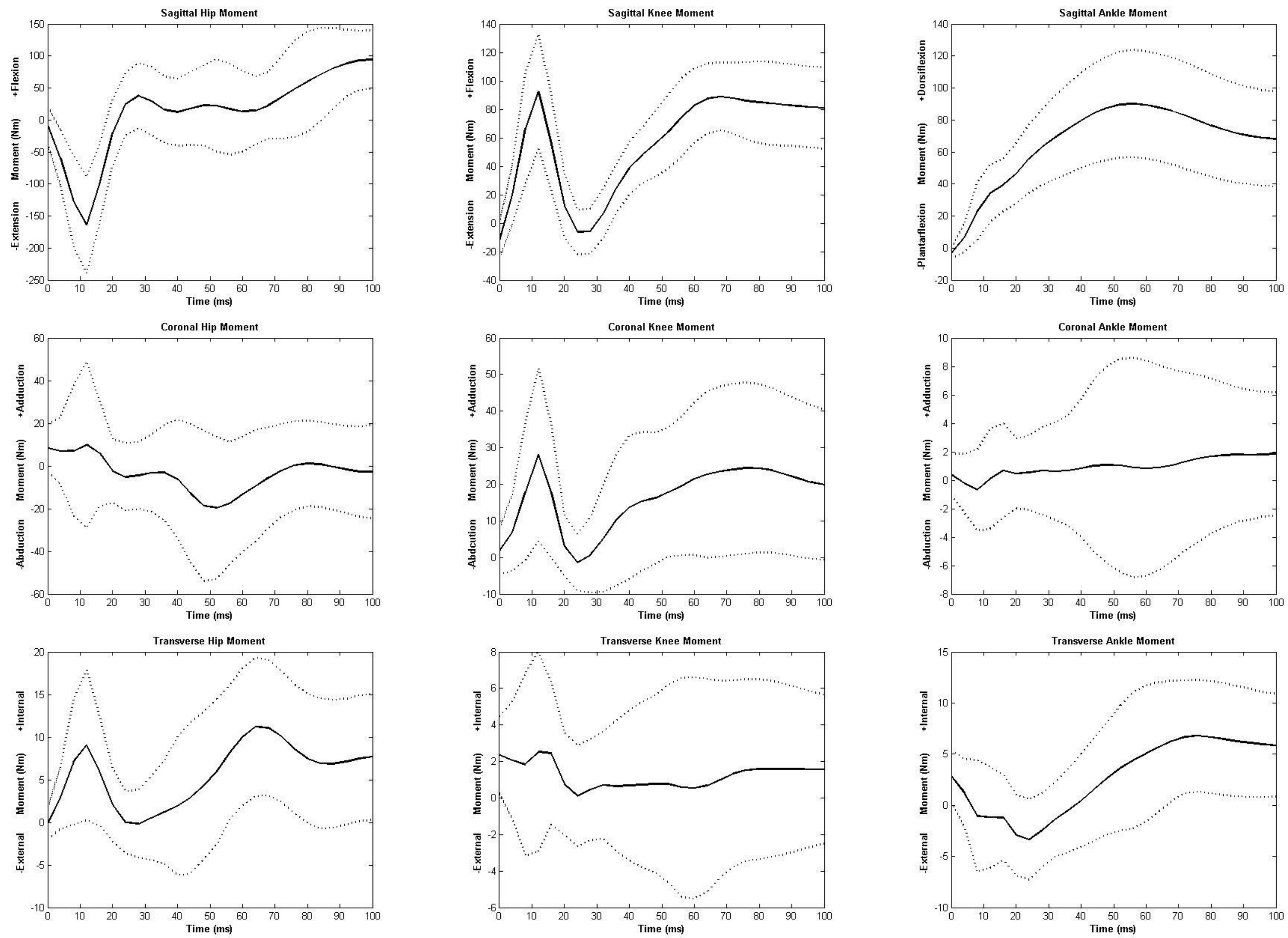

Figure 15: *Group mean (solid line) and ± one standard deviation (dashed lines) of external joint moments (Nm) of the hip (left column), knee (middle column), and ankle (right column) in the sagittal (top row), coronal (middle row), and transverse (bottom row) planes. The joint angles are plotted verse impact phase time from 0-100 ms.*

3.1.2.4 Power

Group mean power plots can be found in Figure 16. In the first 20 ms after IC, the hip generated power (Figure 16). Just after (4 ms) the region of hip power generation, peak KAM was achieved. The knee exhibited a large power absorption prior to peak KAM. At peak KAM, the group mean knee power was in a slight state of power production at 1.5 ± 5.0 W/kg. After peak KAM, the group mean knee power exhibited energy absorption. Throughout the entire impact phase, the group mean ankle power showed power absorption reaching a peak just after peak KAM.

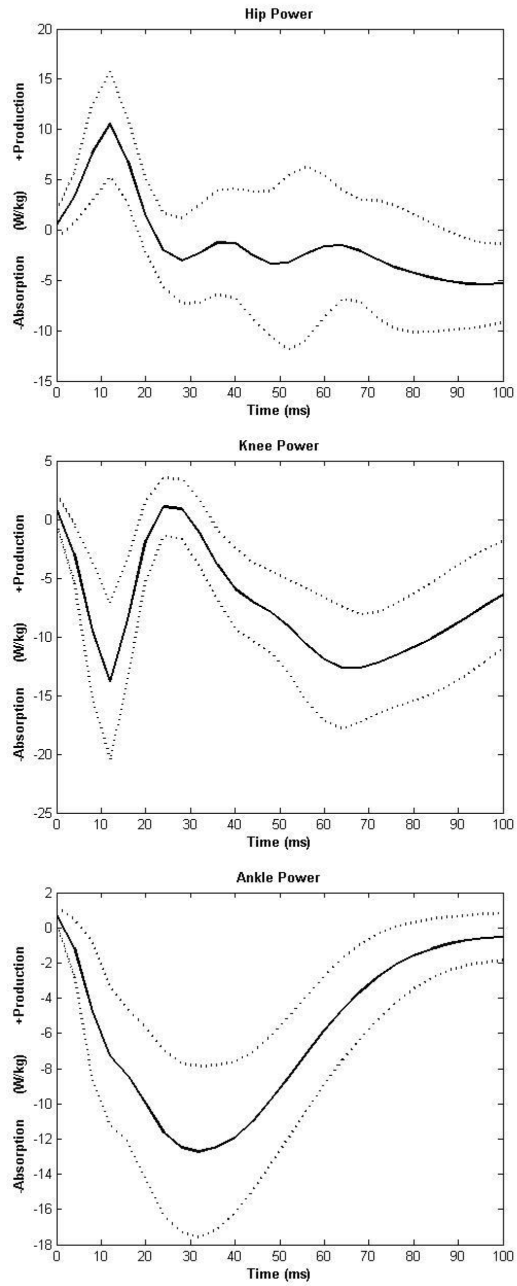

Figure 16: *Group mean (solid line) and ± one standard deviation (dashed lines) of joint power (W/kg) of the hip (top), knee (middle), and ankle (bottom). The joint powers are plotted verse impact phase time from 0-100 ms.*

3.2 Strength Assessment

Biodex strength assessments were performed on isokinetic and isometric hip abduction/adduction as well as isokinetic and isometric knee flexion/extension (Appendix B). Isokinetic strengths were evaluated by looking at strength ratios of extension to flexion and abduction to adduction at each rate tested.

3.2.1 Knee Extension-Flexion

Strength results revealed a knee extension (quadriceps) strength dominance as indicated by a mean strength ratio less than 1 for each isokinetic test administered (Table 3).

3.2.2 Hip Abduction-Adduction

Isokinetic coronal hip strength ratios revealed similar results at each rate tested. At each rate, the ratio of abduction to adduction was greater than 1 revealing an abduction (gluteus medius) dominant strength throughout range of motion (Table 3).

Table 3: *Group Mean (Standard Deviation) of Average Isokinetic Torque Ratios of Flexion to Extension and Abduction to Adduction at Rates Tested*

	Isokinetic Strength Test Rate		
	60º/sec	**75º/sec**	**90º/sec**
Ratio of Flexion to Extension	0.43 (0.15)	0.45 (0.14)	0.44 (0.15)
Ratio of Abduction to Adduction	1.37 (1.09)	1.42 (1.25)	1.49 (1.27)

3.3 Principal Component Analysis

Principal component analysis was performed on six variables previously identified as ACL injury risk factors or variables identified as possible factors through initial data evaluation. After removal of outliers (± two standard deviations from the group mean) as suggested by the study statistician, PCA was performed. Six variables measured at three different time points throughout landing were investigated using PCA resulting in 18 initial variables. Time points selected for evaluation were IC, peak KAM, and 100 ms after IC. Peak KAM occurred at an average 32.3 ± 14.8 ms after IC. Four variables were eliminated after failing to meet retention criteria. A correlation matrix made up of Pearson correlation coefficients was obtained and significant correlation between retained variable were identified (Table 4). Retained variable mean and standard deviation over the impact phase are presented in Figure 17. PCA results can be found in Table 5.

Table 4: *Pearson correlation coefficient matrix for variables retained in Principal Component Analysis (* denotes significant correlation of p<0.05). Matrix variables include Coronal Knee Angle (AKY), Coronal Hip Angle (AHY), Coronal Knee Moment (MKY), Transverse Knee Moment (MKZ), Foot Progression Angle (FP), and Base of Support (BOS).*

		IC					Peak KAM				100 ms after IC				
Time	**Variable**	**AKY**	**AHY**	**MKY**	**MKZ**	**FP**	**AKY**	**AHY**	**FP**	**BOS**	**AKY**	**AHY**	**MKY**	**FP**	**BOS**
IC	**AKY**	1.0													
	AHY	-.14	1.0												
	MKY	-.13	-.01	1.0											
	MKZ	.40*	-.10	.59*	1.0										
	FP	-.14	.05	.01	-.14	1.0									
Peak KAM	**AKY**	.89*	.02	.03	.44*	-.20	1.0								
	AHY	-.14	.94*	.02	-.18	.01	.05	1.0							
	FP	-.2	.10	-.14	-.35*	.95*	-.26	.06	1.0						
	BOS	-.14	.28	-.01	-.14	-.33	-.08	.24	-.23	1.0					
100 ms after IC	**AKY**	.60*	.01	.03	.24	-.42*	.80*	.06	-.40*	-.00	1.0				
	AHY	-.11	.75*	-.01	-.12	.02	.10	.90*	.07	.13	.12	1.0			
	MKY	-.02	-.06	.11	.16	.59*	.07	-.07	.53*	-.39*	-.18	-.02	1.0		
	FP	-.33	.22	-.06	-.38*	.83*	-.42*	.13	.90*	-.08	-.44*	.04	.28	1.0	
	BOS	-.21	.24	-.04	-.30	-.03	-.19	.23	.13	.77*	-.10	.14	-.11	.17	1.0

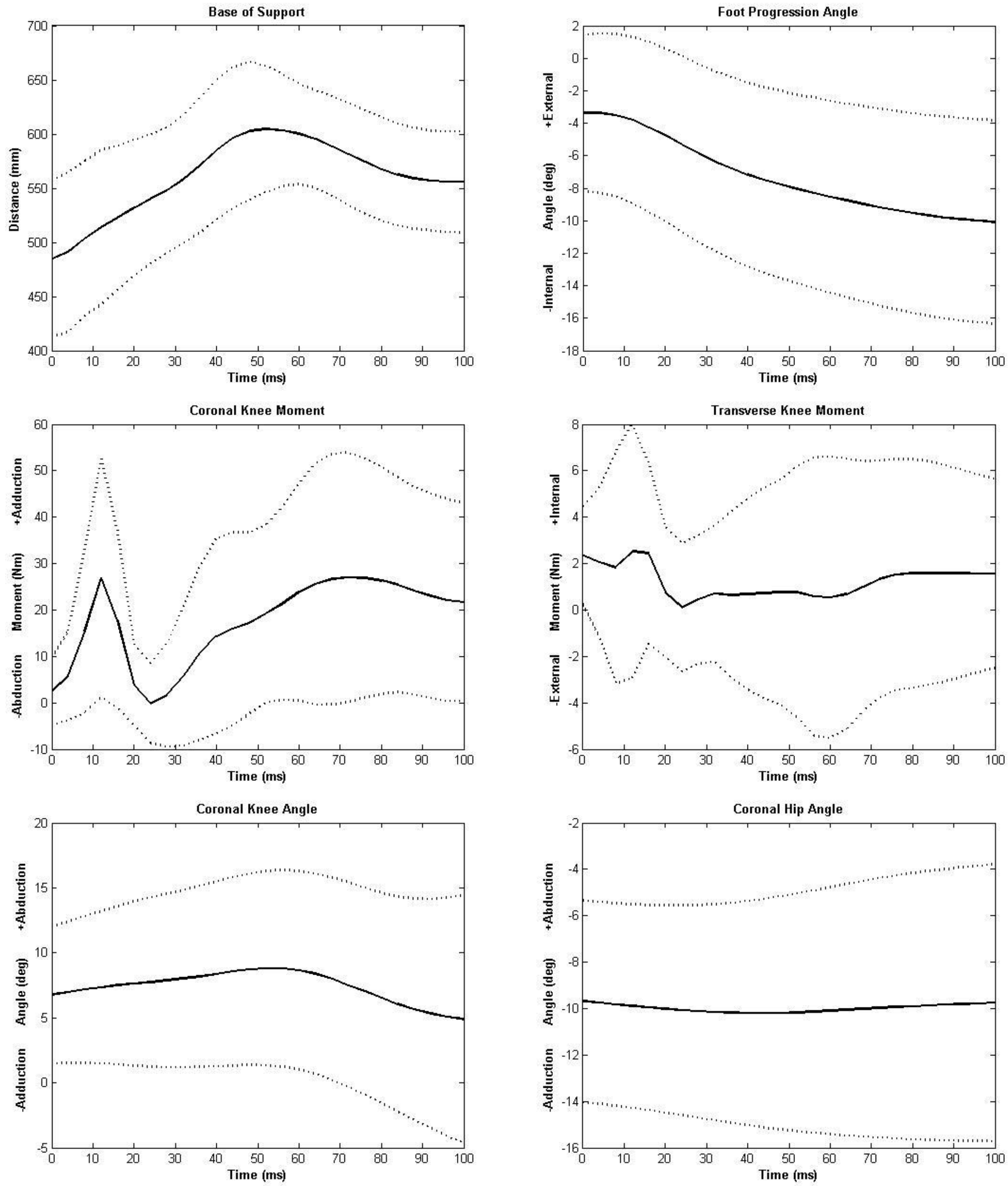

***Figure 17**: Group mean (solid line) and ± one standard deviation (dashed lines) of the six variables including in PCA plotted against impact phase time from 0-100 ms. Variables included are base of support in mm (top left), foot progression angle in degrees (top right), coronal knee moment in Nm (middle left), transverse knee moment in Nm (middle right), coronal knee angle in degrees (bottom left) and corona hip angle in degrees (bottom right).*

Results of Kaiser-Meyer-Olkin (KMO) measure of sampling adequacy was 0.522 and Bartlett's Test of Sphericity indicated a significance value of 0.000 supporting the use of PCA on the present data set. Principal component analysis resulted in five principal components representing a cumulative 87.41% of the variance of the initial data set. The first principal component (PC1) is made up of foot progression angle at the three points of IC, peak KAM and 100 ms after IC as well as the coronal knee moment at 100 ms. PC1 represented 29.00% variance of the original data set. Principal component two (PC2) represents 22.07% of the original data set with contributing variables of coronal hip angle at IC, peak KAM and 100 ms after IC. The third component (PC3) represents 16.51% of the variance in the original data set and is made up of coronal knee angle at all time points. Principal component four (PC4) represents 10.75% of the entire data set variance and is made up of base of support measures at peak KAM and 100 ms. The final principal component (PC5) represents 9.08% variance of the original data set with contributing variables of coronal moment and transverse moment at the knee measured at IC. Communalities extraction (h^2) was calculated which represents the amount of variance accounted by each retained variable (Table 5).

Table 5: *Principal Component Analysis Results containing eigenvalue, percent variance, cumulative variance, component loading score and communalities extraction coefficient of retained variables.*

	Principal Component					
	PC1	**PC2**	**PC3**	**PC4**	**PC5**	
Eigenvalue	4.06	3.09	2.31	1.51	1.27	
% Variance	29.00	22.07	16.51	10.75	9.08	
% Cumulative Variance	29.00	51.07	67.58	78.33	87.41	
Variable Name	**Component Loading**					**h^2**
Foot Progression Angle at IC	0.960					0.951
Foot Progression Angle at Peak KAM	0.954					0.977
Foot Progression Angle at 100 ms	0.826					0.842
Coronal Knee Moment at 100 ms	0.690					0.606
Coronal Hip Angle at IC		0.920				0.881
Coronal Hip Angle at Peak KAM		0.985				0.984
Coronal Hip Angle at 100 ms		0.931				0.873
Coronal Knee Angle at IC			0.923			0.880
Coronal Knee Angle at Peak KAM			0.970			0.978
Coronal Knee Angle at 100 ms			0.798			0.745
Base of Support at Peak KAM				0.904		0.911
Base of Support at 100 ms				0.929		0.900
Coronal Knee Moment at IC					0.915	0.853
Transverse Knee Moment at IC					0.819	0.855

From the PCA results, a PC score was calculated for each leg evaluated for each principal component. PC scores were used to perform association analysis between motion capture dynamics and strength assessment results.

3.4 Association Analysis

Following PCA, Pearson's correlations were performed to assess association between PC scores and strength results (Table 6). Pearson correlation coefficients were calculated for associations between hip isokinetic strength abduction to adduction ratios at 60º/sec, 75º/sec, and 90º/sec and PC scores 1 through 5. Additional Pearson correlation coefficients were calculated to determine association between knee isokinetic strength flexion to extension ratios at 60º/sec, 75º/sec, and 90º/sec and the five PC scores.

Table 6: *Pearson Correlation Coefficients Between Principal Component Scores and Hip Abduction:Adduction Strength Ratio and Knee Flexion:Extesion Strength Ratio (* denotes significance p<0.05)*

	Hip Strength Abduction to Adduction Ratio			**Knee Strength Flexion to Extension Ratio**		
	60º/sec	**75º/sec**	**90º/sec**	**60º/sec**	**75º/sec**	**90º/sec**
PC1 Score	0.17	0.20	0.07	0.01	0.05	0.06
PC2 Score	-0.02	0.05	0.03	-0.14	-0.13	-0.15
PC3 Score	-0.12	-0.28	-0.32	0.10	0.04	-0.06
PC4 Score	0.18	0.14	0.19	-0.43*	-0.38*	-0.41*
PC5 Score	-0.28	-0.26	-0.34*	0.06	0.21	0.16

Association results indicated a significant correlation ($P < 0.05$) between hip abduction to adduction strength at a rate of 90º/sec and PC5 score, made up of base of support at peak KAM and 100 ms after IC. PC4 score, made up of coronal knee moment and transverse knee moment at IC, was significantly correlated to isokinetic knee flexion to extension ratios at all rates tested.

4 Discussion

The results of this study reveal valuable insight into using PCA to reduce kinematic and kinetic data obtained through motion analysis of drop landings that can be further investigated to better understand and, potentially, predict ACL injury risk.

4.1 Kinematics

The female population tested employed a landing base of support that increased through the first 50% of the impact phase. Base of support peaked at approximately 50 ms after IC, and then slightly decreased until reaching the end of the impact phase at 100 ms. Misalignment of the foot with the knee while the hip is locked in neutral alignment has been suggested to decrease the injury threshold of the knee (Chaudhari, 2006). A widening stance at ground contact of landing, with the feet outside of hip width, would place the knees in a position most vulnerable to a valgus applied force. This position would induce a valgus knee angle if the hip abductor muscles are unable to overcome the valgus force to shift the knee into alignment. This is likely the case in the population tested given the abducted knee angle at IC and through landing with an adducted hip angle. However, other dynamic factors are likely to be at play given the lack of significant correlation between base of support and coronal angles of the hip and knee.

Foot progression angle was found to progress internally through landing. This parameter may be an important, relatively uninvestigated factor contributing to ACL injury and injury risk as it would directly relate to tibial rotation through the kinetic chain. Any rotation of the foot must be compensated for through the lower extremities. Recent investigation by Ishida et al. illuminated the importance of foot angle identifying a relationship to knee abduction angle, tibial rotation, and knee abduction moment. With internally rotated toes, increased knee abduction angle, internal tibial rotation, and knee abduction moment were reported. This study revealed naturally selected landing strategies resulted in an external foot rotation of 8.9 ± 6.4 degrees at IC that progressed externally reaching an angle of 11.0 ± 5.6 degrees at peak knee flexion (Ishida, 2015). Ishida et al. did not provide foot progression angle throughout landing, only at IC and peak knee flexion, thus temporal characteristics as they relate to full lower extremity dynamics cannot be compared. Natural landing, as was assessed in the present study, revealed a mean foot internal rotation at IC that increased through the impact phase. Similar to the findings of Ishida et al., a significant correlation was identified between foot progression angle at peak KAM and transverse knee moment at IC as well as foot progression angle at 100 ms after CI and transverse knee moment at IC. Additionally, significant correlation between foot progression angle at 100 ms after IC and coronal knee angle at two different time points (peak KAM and 100 ms after IC) were identified.

By attempting to restrict change in foot progression angle or aiming for a certain foot progression angle, a change in transverse knee moments and coronal knee angles may be seen. It is possible that foot progression angle is a way of compensating for motion that is forced upon the lower extremities from the upper body due to gravity during the deceleration of landing. In an attempt to slow the upper body at landing with restricted knee flexion, the coronal and transverse planes must respond to applied force (Meyer, 2008). Then, the feet are rotated to reduce rotational motion at the knee. Future investigation should look at correlation between transverse knee angle and foot progression angle. Significance was not identified between foot progression and coronal knee moment. While foot progression angle may not be a method to control coronal knee moment directly, it could reduce the combination of coronal and transverse plane moments at the knee, thus reducing risk of injury based on a simulated model of ACL strain (Shin, 2011).

Coronal hip and knee angle were included in PCA due to the importance of coronal plane motion in ACL injury. The group mean coronal hip angles remained in an adducted angle throughout the impact phase. Based on the group mean, an abducted angle at the knee was observed throughout the impact phase. While the two may intuitively seem connected through the kinetic chain, the correlation matrix results did not identify a significant correlation between coronal hip angle and coronal knee angle.

4.2 Kinetics

Ground reaction force results were consistent with previous work (Bates, 2013). The impact phase was a chosen region of interest because the ground reaction force at the end of the 100 ms period after IC begins to stabilize prior to forceful take-off for the concentric, jumping portion of the drop jump task. As is evident in the present study, the group mean exhibited a stabilization of the ground reaction force around 100 ms. Once the ground reaction force stabilizes, changes in moments would be primarily due to changes in the moment arm distance between the ground reaction force vector and the joint experiencing moment application instead of changes in force magnitude. Additionally, it has been determined that ACL injuries typically occur between 17 and 50 ms after initial contact (Krosshaug, 2007), which further supports the use of 100 ms as a definitive end point for the impact phase of drop jumps.

The kinetics of interest used in the PCA included coronal and transverse knee moments. Coronal knee moment increased in adduction just after IC peaking around 10 ms. On average, peak coronal knee abduction moment occurred at 32 ms. After the peak abduction moment occurred, the coronal moment increased in adduction but exhibited little change from 50 ms to 100 ms after IC. Given the relatively steady coronal angle at the knee, changes in coronal knee moment are likely attributed to applied forces on the joint. While the average transverse knee moment across the normal female population appears to remain stable, standard deviation of the data set reaches both internal and external transverse moments throughout the entire impact phase.

Power of the lower extremities during landing is an interesting kinetic measure as its calculation incorporates moment and angular velocity in all three planes of motion. The current study provided a group mean power plot for all three joints of the lower extremities. These results are similar to that of previous studies (DeVita, 1992; Decker, 2003; Norcross, 2010; Schmitz, 2010). It has been suggested that females choose an energy absorption pattern utilizing the knees as a primary absorber and the ankles as a secondary absorber (Decker, 2003). In contrast, males utilize the knees as a primary absorber with the hips as a secondary energy absorber. The difference in absorption strategy may be a contributing factor to the higher injury rates in the female population. The results of the current study suggest a power production during the first 20 ms of landing, after IC. It is possible that this region is the cause of such a result from Decker et al. This region is likely due to the transmission of the upper body weight to the lower through the hip joints. The present study dynamics indicated a peak extension moment occurring prior to 20 ms after IC with a flexion angular velocity at the hip as is evident by the positive slope of the hip flexion angle curve. The large power production region is due to the combination of these factors. However, the presence of the region of power production in female landing but not in males has not been investigated. Upper body motion was not assessed in the present study so definitive conclusions cannot be made. It is possible females require a different absorption pattern than males at the hips due to anthropometric and weight distribution differences between females and males. Measurements of the upper body may be used in the future to determine if such factors contribute to the female hip power production just after IC.

4.3 Strength

Isometric and isokinetic strength of knee flexion and extension was assessed. A ratio of the average peak torque for hamstring to quadricep strength was obtained for each isokinetic speed. The results of this study indicate a quadriceps dominance at each speed assessed in the normal healthy female population. The findings of quadriceps dominance is consistent with other studies that assessed isokinetic strength at similar rates (Rosene, 2001). However, the isokinetic strength ratios tested at 60 º/sec in the current study (0.43 ± 0.15) are even less than those reported by Rosene et al. (right leg: 0.51 ± 0.11, left leg: 0.49 ± 0.12), indicating a greater quadriceps dominance in the population of the current study. The difference in reported values may be due to the population difference between the current study and that of Rosene et al. The current study tested recreationally active females while Rosene et al. tested intercollegiate athletes. It is possible the level of training and athletic experience affects the outcome of the ratio tests, which in this case would suggest greater susceptibility to injury for recreationally active females given the decreased hamstrings to quadriceps ratio.

Isometric and isokinetic strength of the hip abductors and adductors was assessed. A ratio of average peak torque for abduction to adduction was obtained at each isokinetic speed. The results suggest a dynamic abduction strength dominance at the hips for all speeds. These results are similar to results reported by Sugimoto et al. when abduction to adduction ratio is calculated from the abductor peak torques reported. This suggests an abductor dominance when testing isokinetic abductor:adductor peak torque ratios in collegiate female athletes (Sugimoto, 2014). Abductor to adductor ratios for collegiate athletes were greater than the ratio of the recreationally active population in this study. This may suggest a greater likelihood of injury in the recreationally active population due to dynamic abductor weakness. However, recreationally active individuals are less frequently exposed to the highly competitive environments compared to collegiate athletes. Given the greater exposure to competitive situations, the continued study of collegiate athletes is encouraged.

4.4 Principal Component Analysis

Principal component analysis resulted in five PC's made up of kinematic and kinetic variables at three different time points throughout landing. From the eighteen variables input into PCA, four were eliminated because they failed to meet retention criteria, leaving fourteen variables contained within the final results. The first five PC's made up a cumulative 87.4% variance of the original data set. Variance is a desirable measure to explore. If the majority of variance is represented in only a few variables, the rest of the data contributes little to subtle changes in motion results. Principal components representing a larger data set of motion capture data can be correlated to clinical assessments to create a simpler assessment tool that fully encompasses the kinetic and kinematic motion of a drop landing. Comprehensive clinical assessment tools are desired to more easily determine risk of ACL injury and, in turn, implement injury prevention methods. Principal component analysis is a statistical method able to reduce the vast amount of ACL injury risk parameters down to statistically relevant components.

The variables contributing to each principal component were found to be significantly correlated to each variable within that principal component, excluding foot progression angle and coronal knee moment at 100 ms after IC, both contained in PC1. The first principal component variables of coronal knee moment at 100 ms and foot progression angle are likely have a positive correlated due to a changing coronal moment arm with change in foot progression angle. Given the results of PCA revealed components made up of the same variables at different time points for three of the five principal components, it may be desirable to perform separate PCA's at each time point independently as this is likely the cause of component grouping. PC2 is a representation of the coronal hip angle throughout the landing phase. Similarly, PC3 and PC4 are representations of the coronal knee angle and base of support throughout the landing phase, respectively. Additionally, these results suggest that the population produced a consistent landing pattern across individuals because of the strong correlation between the same measurements taken at different times through landing. PC5 represents the coronal knee moment and transverse knee moment at initial contact. The positive correlation between these two variables may be attributed to a shared moment arm in the coronal plane.

Principal component analysis suggests foot progression angle at all time points and coronal knee moment at 100 ms after IC are the most salient variables in the original data set representing 29.0% of the total data set variance. These variables are identified as being important contributors to the variance of the original data set. Given the greater variance, these variables may be more sensitive to kinematic and kinetic changes in the lower extremities during jump landing. Coronal hip angle is the only contributing variable to PC2, making up 22.1% of the total data set variance. Given these results, attempting to alter or control foot progression and coronal hip angle, may be able to steer the knee in a desired motion pattern to reduce likelihood of ACL injury. Kinematic and kinetic variables at the knee contributed to PC3 and PC5 representing 16.5% and 9.1% of data variance, respectively. Contribution from base of support was seen in PC4 representing 10.8% of data variance.

Association analysis between PC scores and isokinetic hip and knee strength revealed few significant correlations. A significant negative Pearson correlation was observed between PC5 scores and isokinetic hip abduction to adduction ratio tested at a rate of 90°/sec. Contributing variables to PC5 were coronal and transverse knee moments at IC. The ratio of abduction to adduction for this test was 1.49 ± 1.27 suggesting stronger abduction contribution than adduction. This suggests with increased PC score, thus increased coronal and transverse knee moments at initial contact, a decrease in isokinetic abduction to adduction hip strength ratio is observed. This suggests a low isokinetic strength ratio of abduction to adduction at the hip, or dynamic abductor weakness relative to adductor strength could be used as a predictor for coronal and transverse knee moments during a drop jump.

Significant negative Pearson correlation results were found between hamstrings to quadriceps isokinetic strength ratio at all speeds and PC4 scores. Contributing variables to PC4 were base of support at peak KAM and at 100 ms after IC. Increased PC4 scores are representative of increased (wider) base of support. The correlation results suggest a greater base of support correlates to a decrease in knee flexor to extensor ratio. Thus, a wider base of support relates to a more quadriceps dominant strength ratio. Biomechanically, the relation between the sagittal strength measures and base of support width is difficult to explain. It is possible with less quadriceps involvement thus a greater ratio of flexion to extension, greater knee flexion motion is produced during landing. Such a landing method is able to better contain motion in the sagittal plane thus affecting base of support. However, another factor must contribute to this connection as sagittal plane knee motion is observed to be constantly increasing while base of support reaches a peak half way through landing.

Principal component analysis appears to be a reliable method of reducing the amount of data obtained through motion analysis of drop landings. The results of this study are promising and suggest the use of PCA could be employed to create predictive algorithms using regression modeling to more completely predict ACL injury. Currently, only one such algorithm exists utilizing measures of body mass index, tibial length, knee flexion range of motion, knee abduction angle, and knee extensor torque to predict high KAM (Myer, 2010b; Myer, 2011a; Myer, 2011b). The present study using PCA was able to identify knee abduction angle and coronal knee moment as important variables in representing data variance as is consistent with the variables contained in Myer et al.'s prediction algorithm. This algorithm was able to predict high sensitivity and specificity but did not include transverse plane metrics. Given the current results and the multiplanar nature of ACL injury, inclusion of transverse plane dynamics is desirable.

The current study, and the findings of Ishida et al. suggest foot progression angle may be a way of assisting in the prediction of ACL injury. Additionally, transverse knee moment and base of support were contained within the PCA results suggesting their importance in drop landing assessments.

4.5 Future Directions

The present study provides valuable preliminary work from which ACL injury prediction investigation can expand. Given these results, PCA proves to be a viable method to reducing the vast amount of kinematic and kinetic data that is obtained through motion analysis. The present study was limited in the population size thus limiting the amount of variables that could be included in PCA while continuing to meet the KMO measure of sampling adequacy requirement (KMO > 0.5). Future study with a larger sample size, meeting a suggested ratio of one variable included in PCA to every five samples, should focus on incorporating a larger amount of variables to more fully understand and be able to identify the most salient measures related to drop landing assessments beyond the six measures used in the present study. A future study looking at just one time point for the six variable examined in this study should include 30 samples. Additionally, future investigation should focus on one time point rather than the three time points chosen in the present study. This is advisable due to principal components including several of the same variables measured at different time points in the same principal component.

Principal component analysis was able to identify the importance of coronal variables contained in previous regression models used in prediction of ACL injury.

Future regression models should focus on the incorporation of multiplanar lower extremity joint kinematics and kinetics to more reliably identify risk of injury.

Motion analysis results are able to identify net moments and forces at each joint. However, given the importance of muscle activation, timing and strength on ACL stress, a more precise calculation, taking into account internal muscle forces applied at the joint would be able to more accurately predict ACL injury. One such way of better determine stresses on the ACL for injury prediction would be through the use of OpenSim modeling software.

The present study found few significant associations between abduction to adduction isokinetic strength ratio and principal component scores. Khayambashi et al. was able to predict subsequent ACL injury using isometric abduction strength (Khayambashi, 2016). Future direction may focus on investigate the relation of PC scores to isometric abduction strength rather than isokinetic abduction to adduction ratios given the promising results of Khayambashi et al. If isometric abductor strength is able to predict injury to the ACL, it may also be able to predict motion associated with ACL injury through PC score calculation.

The current study sets up the process through which future studies can use PCA to reduce data obtained through three dimensional motion capture. Given the results of drop jump and ACL injury prediction presented by Krosshaug et al., it may be advisable to investigate other screening tools such as a single leg drop landing or cutting maneuvers (Krosshaug, 2016). However, the vertical drop jump cannot be entirely dismissed given the regression results from Krosshaug et al.'s study only took into account five variables taken from the kinematics and kinetics of the drop jump performed. Principal component analysis used in a prospective study to predict ACL injury provides a promising method to incorporating a large amount of motion dynamic data and statistically determining variables of greatest importance.

5 Conclusion

With the current understanding of ACL injury mechanisms and risk factors, it is ever important to develop methods of predicting injury. Through reliable injury risk prediction methods, training programs can be implemented to prevent injury to the ACL in athletes exhibiting high risk. Such prediction methods should take into account the multiplanar and multi-joint contributions to ACL injury. Principal component analysis is one method of data reduction that has the capacity to objectively reduce the large amount of data obtained through three dimensional motion analysis to be used in injury prediction statistical models.

The purpose of this paper was to investigate the mechanics of jump landings in a physically active, young adult female population. Secondly, to apply PCA as a data reduction technique on kinematic and kinetic measures obtained during a drop jump landing. Finally, it was hypothesized that a PC score calculated from the z-score of the raw data values and principal component loadings would show significant association with isokinetic hamstring to quadriceps strength ratio and isokinetic hip abduction to adduction strength ratio.

This study was able to identify five principal components capable of representing 87.41% of the variance in the original data set. Previously holding little importance, foot progression angle was identified as a variable with strong correlation to the first principal component representing 29.00% of the data set variance. This study was able to statistically relate principal component results of dynamic movement data to strength measures of the musculature at the knee and hip joints.

Further investigation of additional clinical measures related to principal component scores and regression modeling has the potential to improve reliability of previous ACL injury prediction models. This method allows incorporation of potentially important additional dynamic motion data, such as foot progression angle, currently not used in prediction models. Improvements to current prediction techniques will allow effective and efficient implementation of training procedures for those individuals identified as having high risk of injury.

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APPENDIX A

Table 7: *Group Mean (± standard deviation) Kinematic Data for the Hip, Knee and Ankle in the Sagittal, Coronal and Transverse Planes at Three Time Points (IC, Peak KAM, 100 ms after IC) throughout the Impact Phase of Landing.*

Joint	**Measure**	**Time**		
		IC	**Peak KAM**	**100 ms**
Hip	Flexion Angle (°)	33.5 ± 7.0	41.2 ± 8.3	65.4 ± 8.8
	Abduction Angle (°)	-9.6 ± 4.4	-9.9 ± 4.9	-9.6 ± 6.0
	Transverse Angle (°)	4.5 ± 10.1	4.5 ± 9.8	4.9 ± 10.0
Knee	Flexion Angle (°)	23.7 ± 8.4	38.2 ± 9.9	72.2 ± 6.7
	Abduction Angle (°)	6.5 ± 5.1	6.9 ± 6.4	4.6 ± 9.4
	Transverse Angle (°)	8.2 ± 11.1	10.9 ± 10.7	16.9 ± 12.8
Ankle	Dorsiflexion Angle (°)	-18.0 ± 5.6	3.5 ± 9.9	27.3 ± 4.6
	Eversion Angle (°)	2.8 ± 2.4	4.0 ± 2.6	5.7 ± 3.3
	Rotation Angle (°)	-10.3 ± 9.0	-14.3 ± 9.2	-19.8 ± 11.1
External Foot Progression Angle (°)		-3.4 ± 4.7	-5.5 ± 6.1	-9.9 ± 6.3
Base of Support (mm)		488.3 ± 72.9	559.8 ± 61.2	564.0 ± 45.5

Table 8: *Group Mean (± standard deviation) Force Data for the Hip, Knee and Ankle in the Sagittal, Coronal and Transverse Planes at Three Time Points (IC, Peak KAM, 100 ms after IC) throughout the Impact Phase of Landing.*

Joint	**Measure**	**Time**		
		IC	**Peak KAM**	**100 ms**
Hip	Anterior (+)/Posterior (-) (N)	29.4 ± 56.2	-154.8 ± 119.2	-405.5 ± 163.3
	Lateral (+)/Medial (-) (N)	17.8 ± 17.1	-28.3 ± 64.3	-5.2 ± 75.9
	Tension (+)/Compression (-) (N)	110.3 ± 58.3	-268.6 ± 282.9	-404.9 ± 197.0
Knee	Anterior (+)/Posterior (-) (N)	-23.2 ± 38.5	161.7 ± 141.7	361.9 ± 139.9
	Lateral (+)/Medial (-) (N)	-4.9 ± 16.8	19.6 ± 30.4	117.5 ± 88.2
	Tension (+)/Compression (-) (N)	58.7 ± 30.6	-455.8 ± 331.0	-571.6 ± 189.5
Ankle	Anterior (+)/ Posterior (-) (N)	4.5 ± 26.1	58.2 ± 100.5	180.3 ± 71.2
	Lateral (+)/Medial (-) (N)	8.4 ± 13.0	-28.7 ± 36.9	7.0 ± 46.2
	Tension (+)/Compression(-) (N)	5.0 ± 19.3	580.5 ± 375.7	684.7 ± 236.6
Vertical Ground Reaction Force (N)		66.4 ± 28.0	629.5 ± 373.1	711.5 ± 241.9

Table 9: *Group Mean (± standard deviation) Moment Data for the Hip, Knee and Ankle in the Sagittal, Coronal and Transverse Planes at Three Time Points (IC, Peak KAM, 100 ms after IC) throughout the Impact Phase of Landing.*

Joint	**Measure**	**Time**		
		IC	**Peak KAM**	**100 ms**
Hip	Flexion Moment (Nm)	-10.0 ± 25.8	47.4 ± 53.8	92.8 ± 45.6
	Adduction Moment (Nm)	8.8 ± 10.7	-25.0 ± 29.6	-3.8 ± 22.2
	Internal Rotation Moment (Nm)	-0.1 ± 1.8	6.7 ± 7.1	8.1 ± 7.5
Knee	Flexion Moment (Nm)	-11.0 ± 10.0	13.0 ± 35.6	83.4 ± 25.7
	Adduction Moment (Nm)	2.0 ± 6.0	-10.4 ± 9.4	20.5 ± 20.6
	Internal Rotation Moment (Nm)	2.3 ± 2.1	-1.0 ± 4.4	1.6 ± 4.1
Ankle	Dorsiflexion Moment (Nm)	-3.0 ± 3.1	54.1 ± 28.6	70.0 ± 28.4
	Inversion Moment (Nm)	0.4 ± 1.5	-0.7 ± 3.5	1.8 ± 4.5
	Internal Rotation Moment (Nm)	2.8 ± 2.7	-1.8 ± 4.9	6.1 ± 5.0

Table 10: *Group Mean (± standard deviation) Power Data for the Hip, Knee and Ankle at Three Time Points (IC, Peak KAM, 100 ms after IC) throughout the Impact Phase of Landing.*

Joint	**Time**		
	IC	**Peak KAM**	**100 ms**
Hip Power Production (W/kg)	0.59 ± 1.28	-4.87 ± 5.58	-5.31 ± 3.94
Knee Power Production (W/kg)	0.77 ± 1.28	-1.54 ± 5.00	-6.78 ± 4.64
Ankle Power Production (W/kg)	0.70 ± 0.49	-7.93 ± 4.73	-0.61 ± 1.45

APPENDIX B

Table 11: *Group Mean (± Standard Deviation) Average Peak Torque per Body Weight (%) Isometric Strength Assessment Results from Hip Abduction/Adduction Tested at 0, 15 and 30 degrees of Hip Abduction and Knee Flexion/Extension Tested at 60, 90, and 90 degrees of Knee Flexion from Full Extension.*

Test Angle	**Hip Abduction Isometric Strength (Torque/BW, %)**	**Hip Adduction Isometric Strength (Torque/BW, %)**
0°	86.8 ± 20.5	70.9 ± 27.5
15°	79.8 ± 21.4	109.1 ± 33.1
30°	65.1 ± 22.2	151.7 ± 35.5
	Knee Extension Isometric Strength (Torque/BW, %)	**Knee Flexion Isometric Strength (Torque/BW, %)**
30°	145.9 ± 33.3	103.3 ± 27.5
60°	220.6 ± 57.4	98.6 ± 25.8
90°	176.9 ± 40.8	78.4 ± 24.0

Table 12: *Group Mean (± Standard Deviation) Average Peak Torque per Body Weight (%) Isokinetic Strength Assessment Results from Hip Abduction/Adduction Tested at 60°sec, 75°sec, and 90°sec and Knee Flexion/Extension Tested at 60°sec, 75°sec, and 90°sec.*

Test Rate	**Hip Abduction Isokinetic Strength (Torque/BW, %)**	**Hip Adduction Isokinetic Strength (Torque/BW, %)**
60°sec	82.5 ± 26.2	87.3 ± 63.7
75°sec	85.5 ± 30.4	90.5 ± 63.8
90°sec	88.0 ± 33.4	90.4 ± 70.0
	Knee Extension Isokinetic Strength (Torque/BW, %)	**Knee Flexion Isokinetic Strength (Torque/BW, %)**
60°sec	78.4. ± 37.0	79.3 ± 26.3
75°sec	181.7 ± 28.1	80.1 ± 25.7
90°sec	178.1 ± 31.9	79.2 ± 28.1

Table 13: *Group Mean (± Standard Deviation) Average Peak Torque (Nm) Isometric Strength Assessment Results from Hip Abduction/Adduction Tested at 0, 15 and 30 degrees of Hip Abduction and Knee Flexion/Extension Tested at 60, 90, and 90 degrees of Knee Flexion from Full Extension.*

Test Angle	**Hip Abduction Isometric Strength (Nm)**	**Hip Adduction Isometric Strength (Nm)**
0°	54.9 ± 14.9	44.1 ± 19.0
15°	50.7 ± 15.0	69.0 ± 23.7
30°	42.0 ± 14.3	95.6 ± 26.3
	Knee Extension Isometric Strength (Nm)	**Knee Flexion Isometric Strength (Nm)**
30°	93.2 ± 24.4	65.6 ± 19.0
60°	140.6 ± 40.8	62.2 ± 17.8
90°	122.1 ± 66.3	49.4 ± 16.3

Table 14: *Group Mean (± Standard Deviation)) Average Peak Torque (Nm) Isokinetic Strength Assessment Results from Hip Abduction/Adduction Tested at 60°sec, 75°sec, and 90°sec and Knee Flexion/Extension Tested at 60°sec, 75°sec, and 90°sec.*

Test Rate	**Hip Abduction Isokinetic Strength (Nm)**	**Hip Adduction Isokinetic Strength (Nm)**
60°sec	42.5 ± 18.3	45.4 ± 39.1
75°sec	45.7 ± 19.1	48.1 ± 42.3
90°sec	47.0 ± 21.8	45.7 ± 37.4
	Knee Extension Isokinetic Strength (Nm)	**Knee Flexion Isokinetic Strength (Nm)**
60°sec	101.3 ± 27.1	42.6 ± 17.8
75°sec	102.2 ± 21.5	45.8 ± 16.9
90°sec	99.5 ± 24.9	43.2 ± 17.5