THE EFFECT OF ANTERIOR CRUCIATE LIGAMENT INJURY AND RECONSTRUCTION ON LOWER EXTREMITY BIOMECHANICS, COORDINATION, AND VARIABILITY

Benjamin McMillen Goerger

A dissertation submitted to the faculty of the University of North Carolina at Chapel Hill in partial fulfillment of the requirements for the degree of Doctor of Philosophy in the Department of Interdisciplinary Human Movement Science (School of Medicine)

Chapel Hill
2012

Approved by:

Darin A. Padua, PhD, ATC

Anthony I. Beutler, MD

J. Troy Blackburn, PhD, ATC

Stephen W. Marshall, PhD

John H. Wilckens, MD

ABSTRACT

BENJAMIN MCMILLEN GOERGER: The Effect of Anterior Cruciate Ligament Injury and Reconstruction on Lower Extremity Biomechanics, Coordination, and Variability
(Under the direction of Darin A. Padua)

Individuals that suffer an ACL injury, and undergo reconstructive surgery are at an increased risk for the development of osteoarthritis and a secondary ACL injury. However, there is no information other than case studies, that has documented the effect of ACL injury on lower extremity biomechanics and coordination, and few studies have assessed asymmetry in these measures relative to noninjured individuals. The purpose of this study was to determine if lower extremity biomechanics and coordination are altered by ACL injury. A second purpose was to determine if differences in bilateral asymmetry in lower extremity biomechanics, coordination, and variability for those with ACL injury are greater than those with no injury. Following ACL injury, we observed an increase in knee valgus ($F_{(2,66)} = 3.957$, $p = 0.024$) and hip adduction ($F_{(2,66)} = 3.773$, $p = 0.028$) at Initial Ground Contact for both the injured and noninjured limb, as well as a decrease in peak knee varus ($F_{(2, 66)} = 5.198$, $p = 0.008$). An increase in peak knee valgus was also observed in the noninjured limb ($F_{(2,66)} = 3.768$, $p = 0.028$). This was associated with a decrease in peak knee extension moment ($F_{(2,66)} = 4.509$, $p = 0.015$), peak hip flexion moment ($F_{(2,66)} = 3.847$, $p = 0.026$), and peak anterior tibial shear force ($F_{(2,66)} = 4.530$, $p = 0.014$) for the injured limb. In addition, we observed an alteration in coordination of hip and knee transverse plane motion for the injured limb following ACL injury ($F_{(2,65)} = 4.398$, $p = 0.016$). The only differences we observed for asymmetry was for those with

ACLR to have greater asymmetry for peak internal knee flexion moment (Z_{KS} = 1.42, p = 0.035) and peak vertical ground reaction force (Z_{KS} = 1.45, p = 0.031). We did not observe any significant difference in asymmetry of lower extremity coordination or variability between groups. Our findings provide evidence for how ACL injury and ACLR may increase an individual's risk for a second ACL injury and the development of osteoarthritis.

ACKNOWLEDGEMENTS

This project would not have been possible without the help and guidance of several people, first being my committee of Drs. Darin Padua, Anthony Beutler, Troy Blackburn, Stephen Marshall, and John Wilckens. I am grateful for their time and effort to make this a successful project, and consider myself very lucky to have had the opportunity to learn from them. I would like to specifically acknowledge Dr. Marshall, Dr. Beutler, and Dr. Padua for their encouragement to pursue this project and their time and effort in conducting the JUMP ACL study, without which this project would never have been possible. Last, I am very thankful to Dr. Padua. If it were not for his mentorship that began so many summers ago in Fishkill, NY, I would not have the ability or the opportunities that I have today.

I would also like to acknowledge others who have helped make this project a success along the way. A big thank you goes to Sue Wolf for helping me with everything from IRB forms and data, to shipping equipment and reimbursements. Thank you also to Barnett Frank, Rebecca Begalle, and Molly Grasso for their assistance with data collection. They all braved a different adversity (i.e. extreme weather, disastrous travel, moving heavy equipment) to help with this project, and I am thankful.

I am truly thankful for the support my friends and family have given me. My parents, Philip and Suzanne Goerger, are not only great parents, but were also excellent role models for how to balance parenthood and graduate school as I was growing up. Last, none of this would have truly been possible without the unconditional love and support of my wife Candice, and son Garrett. I will never be able to thank them enough for the love and joy they bring into my life every day.

TABLE OF CONTENTS

LIST OF TABLES

Table **Page**

LIST OF FIGURES

CHAPTER ONE

INTRODUCTION

Injury to the anterior cruciate ligament (ACL) can be a devastating injury that requires a significant amount of time and effort to allow for recovery and return to one's respective level of activity or sport. Anterior cruciate ligament reconstruction (ACLR) is often performed to restore mechanical stability to the knee and facilitate return to participation in physical activity and sport. The true number of ACL injuries that occur each year in the United States is unknown, but recent projections have estimated a 67.8% increase in the number of ACLRs performed over a ten year period.[1] For the most part, ACLR does meet the goals of restoring mechanical stability allowing persons to continue participation in physical activity.[2,3]

Returning to participation in a sport or high demand physical activity is a short-term goal. Reconstructive surgery and rehabilitation are often conducted with the goal of returning function to the injured limb. The marker for adequate function is returning the limb to how it functioned before injury, but this may be a flawed benchmark as this may reestablish factors that predisposed the person to injury in the first place. These sentiments are reflected in the literature, as those with a previous ACL injury have an increased incidence of subsequent knee injury[4], and particularly an increased risk for a second ACL injury, regardless of the initial leg injured.[5-7] Those with a previous ACL injury reportedly have a four fold[5] to ten fold [7] increased risk for injury compared to those with no previous ACL injury. The risk for reinjury is independent of the type of graft used for ACLR, and nonspecific to the previously injured knee.[8] In fact, returning to high demand sports or activities, the goal of ACLR, is associated with risk for injury of the contralateral uninjured ACL.[8] Faude et al[6] postulated factors that predispose those with ACL injury to subsequent injury may be person specific and not knee specific. Whether the increased risk for reinjury is the product of ACL injury and reconstruction, or residual biomechanical risk factors that caused the initial injury is unknown.

Significant research has been devoted to understanding the factors that cause initial injury to the ACL. Cadaveric studies have demonstrated that the ACL is loaded when an anterior force is applied to the proximal tibia.[9] Additional factors such as frontal plane loading[10,11], transverse plane loading[11], weight bearing[12,13], and a combination of these factors can also increase the ACL strain.[14-17] Going a step further, analyses of actual ACL injuries captured on video, it appears that a majority of these injuries occur during a landing from a jump[18-20] and during a planting, cutting maneuver.[20] The position at the time of injury has also been described as landing with less plantar flexion[21], with the foot placed outside the knee creating a valgus alignment while the knee is extended[20], and having greater hip flexion.[21] Descriptions of ACL loading and the mechanism of injury provide limited information, as they describe how the ACL can be injured, but they do not necessarily provide information as to what places a person at greater risk for suffering injury. In addition, these descriptions are relevant to describing a first injury, not necessarily a subsequent injury after reconstruction, rehabilitation, and return to activity.

There is, however, limited information on the prospective risk factors for non-contact ACL injury. Hewett et al[22] demonstrated in a prospective analysis of adolescent female athletes that those that went on to suffer a non-contact ACL injury displayed greater knee valgus angle, lower maximum peak knee flexion angle, greater external knee valgus moment, greater peak vertical ground reaction force, and greater external hip flexion moment.[22] While the information is limited in scope, because of the population studied and a relatively low number of injuries, it does identify biomechanical risk factors that may be important for non-contact ACL injury. One finding of particular use for understanding risk factors for subsequent ACL injury, is that the athletes that went on to suffer injury displayed bilateral differences in external knee valgus moment.[22]

Other bilateral differences in lower extremity biomechanics have been identified in persons after ACLR.[23-31] The presence of asymmetries in movement and loading are thought to be the result of injury to the ACL, but because they have also been identified as a prospective risk factor for initial injury, it may indicate the risk factors for initial injury are not being adequately addressed in rehabilitation. Further evidence of this problem was provided by Paterno et al[23], when they observed that asymmetrical loading of the lower extremities during a double leg jump landing, particularly bilateral differences in sagittal plane knee moments, was predictive of a second ACL injury. Other biomechanical variables that were predictive of injury included transverse plane moment impulse of the noninjured leg, and frontal plane knee displacement of the injured limb.[23] Asymmetrical movement and loading of the lower extremity may be important factors for understanding the increased risk for injury for those with ACLR. The importance of asymmetry is still unclear though, as no evidence has been provided to indicate that magnitude of asymmetry for those with ACLR is any greater than those observed in healthy noninjured persons. Only this type of information can provide an indication of risk factors that are not being adequately addressed during reconstruction and rehabilitation.

As informative as biomechanical analyses are, they only provide a limited view on the changes in movement after ACL injury and ACLR. Recent work has been done to characterize the coordination and variability of movement within the lower extremity in those with ACLR.[32-35] Coordination and variability have been researched under the notion that quantifying these measures may provide an assessment of an individual's risk for injury. Persons with ACLR, when compared to those without injury, have demonstrated increased variability[33,34], as well as altered coordination between the shank and thigh.[32,35] There is, however, a lack of prospective data to indicate if these differences are inherent to the individual and were present prior to injury, or if they are the result from ACL injury and reconstruction. Again, research that incorporates repeated measures taken prior to ACL injury and after subsequent ACLR is needed to confirm the role that this type of information has in understanding risk for the initial and subsequent ACL injuries.

Because those with ACLR have an increased risk for suffering a subsequent ACL injury as compared to those with no injury, they are a unique population for study to add to the current understanding of ACL injury risk factors. It is possible that ACLR alters lower extremity biomechanics in such a way that increases their risk for reinjury, or that biomechanical factors that placed them at risk for the first injury are still present and compounded in the presences of ACLR. Unfortunately, this cannot be confirmed based on the current information that is available for this population. The same is true for measures of lower extremity coordination and variability, which is thought to influence the risk for injury, but has not been used prospectively to identify risk factors for injury. The only way to determine if the above postulation is correct is to conduct research that incorporates repeated assessment that includes measures prior to the initial ACL injury and after the subsequent ACLR. In addition, bilateral differences in lower extremity biomechanics, coordination, and variability have been documented in those with ACLR, but there is little evidence to determine if these differences are greater than those with no previous ACL injury. Therefore, the purpose of this study is to determine if lower extremity biomechanics and coordination obtained prior to ACL injury change following ACL injury and subsequent ACLR. A second purpose is to determine if between limb differences in lower extremity biomechanics, coordination, and variability are different as compared to individuals with no history of ACL injury.

1.1 Research Questions

Research Question 1: Are lower extremity biomechanics during a double leg jump landing changed following ACL injury and subsequent ACLR?

> *Research Question 1a:* Are lower extremity biomechanics of the injured limb changed following ACL injury and subsequent ACLR?

> *Research Question 1b:* Are lower extremity biomechanics of the noninjured limb changed following ACL injury and subsequent ACLR?

Research Question 2: Is lower extremity joint coordination during a double leg jump landing changed following ACL injury and subsequent ACLR?

> *Research Question 2a:* Is lower extremity joint coordination of the injured limb changed following ACL injury and subsequent ACLR?
>
> *Research Question 2b:* Is lower extremity joint coordination of the noninjured limb changed following ACL injury and subsequent ACLR?

Research Question 3: Is the magnitude of between limb differences in lower extremity biomechanics during a double leg jump landing different for persons with ACLR as compared to those with no history of ACL injury at Follow-Up?

Research Question 4: Is the magnitude of between limb differences in lower extremity joint coordination and variability during a double leg jump landing different for persons with ACLR as compared to those with no history of ACL injury at Follow-Up?

1.2 Research Hypotheses

Research Hypothesis 1a: There will be no change in lower extremity biomechanics of the injured limb during a double leg jump landing from Baseline to Follow-Up for both the ACLR and Control Groups.

Research Hypothesis 1b: There will be no change in lower extremity biomechanics of the noninjured limb during a double leg jump landing from Baseline to Follow-Up for both the ACLR and Control groups.

Research Hypothesis 2a: There will be no change in lower extremity joint coordination of the injured limb during a double leg jump landing from Baseline to Follow-Up for both the ACLR and Control groups.

Research Hypothesis 2b: There will be no change in lower extremity joint coordination of the noninjured limb during a double leg jump landing from Baseline to Follow-Up for both the ACLR and Control groups.

Research Hypothesis 3: The ACLR Group will demonstrate greater differences in between limb lower extremity biomechanics at Follow-Up as compared to the Control Group.

Research Hypothesis 4: The ACLR Group will demonstrate greater differences in between limb lower extremity joint coordination and variability at Follow-Up as compared to the Control Group.

1.3 Operational Definitions

ACL Injury: Disruption of the anterior cruciate ligament injury as confirmed by examination of medical records.

ACLR: Any surgery performed to replace the injured anterior cruciate ligament for the purpose of restoring stability to the knee.

Dominant Limb: The response given by a participant when asked, “Which leg would you use to kick a ball for maximal distance?”.

Injured Limb: For the ACLR Group, the limb in which an ACL injury occurred. For the Control Group, the limb that was tested during initial testing for the JUMP ACL study.

Noninjured Limb: For the ACLR Group, the limb in which an ACL injury did not occur. For the Control Group, the limb that was not tested during initial testing for the JUMP ACL study.

Joint Coordination: The relationship in angular position between two joints, quantified by determining the average coupling angle across the phase of interest during the double leg jump landing task as measured using vector coding.

Joint Coordination Variability: The between trials variability of joint coordination during the double leg jump landing task as measured using vector coding.

Baseline: Testing that occurred during the JUMP ACL study, and resulted in collection of biomechanical data on the participant's dominant leg at the time of their enrollment in their respective service academy.

Follow-Up: Testing that occurred in the Spring of 2011 in which lower extremity biomechanical data were collected on those identified for the purposes of this study that were enrolled in the JUMP ACL study.

Double Leg Jump Landing: A task in which the participant stands atop a 30cm box, located a distance equal to half their body height from the front of a force plate, jumps down, landing on both feet and immediately jumping for maximum height.

Preparatory Phase: The 100 milliseconds before Initial Ground Contact, during a double leg jump landing.

Initial Ground Contact: The first time point at which the vertical ground reaction force exceeds 10 N during a double leg jump landing.

Landing Phase: A time period during a double leg jump landing defined as the point from Initial Ground Contact to the time point at which the maximum knee flexion angle is reached.

Anterior Knee Laxity: The anterior displacement of the tibia, measured in millimeters, under a 30lb force using a knee arthrometer.

1.4 Assumptions & Limitations

- All participants will perform tasks with their best effort.
- The type of graft or reconstructive procedure performed for the Injured Group will not be considered for this study.
- Concomitant injury associated with ACL injury will not be considered for this study.
- The amount of time since ACLR will not be considered for this study.
- Rehabilitation programs after ACLR are similar across service academies and time for those in the Injured Group.
- The results of this study may be limited to those that are young and physically active.
- The Control Group is representative of the cohort at the time of Follow-Up.

1.5 Delimitations

- All participants were enrolled in the JUMP ACL study and had complete biomechanical data collected at baseline testing.
- All participants were still enrolled in their respective service academy and the JUMP ACL study.
- All participants were participating in physical activity.
- All participants were members of the 2007 or 2008 JUMP ACL cohort.

1.6 Independent Variables

Time:

- Baseline: Testing performed for the JUMP ACL study in the Summers of 2007 and 2008 upon each participant's entry in to the respective service academy.
- Follow-Up: Testing conducted in the Spring of 2011.

Group:

- ACLR Group: Participants that had suffered one ACL injury and undergone subsequent ACLR since Baseline, had no prior history of ACL injury at Baseline, and had complete biomechanical data from Baseline.
- Control Group: Participants that have no history of ACL injury, and had complete biomechanical data from Baseline.

1.7 Dependent Variables

Kinematics: Sagittal, frontal, and transvers plane peak angles of the knee and hip

Kinetics: Sagittal, frontal, and transverse plane moments of the knee and hip; anterior tibial shear force; vertical ground reaction force

Joint Coordination and Variability: Knee sagittal plane - Hip sagittal plane; Knee frontal plane – Hip frontal plane; Knee transverse plane – Hip transverse plane; Knee frontal plane – Hip transverse plane; Knee transverse plane – Hip frontal plane

1.8 Significance

Information about the changes that occur after ACL injury and ACLR is important for developing better methods during rehabilitation to reduce a person's risk for reinjury and secondary complications. The evidence to date has been limited to isolated observation of movement patterns prior to injury, or following injury. This study may provide more insight into risk factors for subsequent ACL injury, particularly whether there are inherent characteristics within a person that place them at risk for initial injury that are not being addressed with rehabilitation and are keeping them at risk as they return to physical activity.

CHAPTER TWO

REVIEW OF THE LITERATURE

The purpose of this study is to examine the effect of ACL injury and ACLR on a sample of young, physically active persons. This review of the available literature will indicate what additional information this study will provide to the current literature regarding the biomechanics and coordination of those with ACLR.

2.1 ACL Injury as a Risk Factor for Subsequent ACL Injury and Knee OA

It is important to study and understand the effect of ACL injury on lower extremity biomechanics because persons that have previously suffered an ACL injury are more likely to suffer subsequent knee injury.[4] Specifically, those with a previous ACL injury are at a greater risk for suffering a subsequent ACL injury.[5-7] The risk ranges between a four[5] and ten fold[7] increased risk for reinjury in those that report previous ACL injury. What is interesting, though, is that the risk for subsequent injury is not limited to the reconstructed leg.[8] Salmon et al[8] reported that only the time to reinjury was different between those that suffered a second injury in their reconstructed or healthy limb, with the reconstructed limb being more likely to be injured in the first twelve months after returning to sport activity. In addition, suffering an initial ACL injury by a contact mechanism of injury was a significant predictor for reinjury of the reconstructed limb.[8] This is despite no difference in median graft diameter, or any difference in the presence of articular cartilage or meniscal damage at the time of reconstruction.[8] Graft type does not seem to influence the risk of reinjury either, as time to injury was not different between those that had a BPTB or HS autograft used for the initial reconstruction.[8] Injury to the healthy or uninjured ACL was predicted by a return to sports or activities that were classified as strenuous or moderate in demand after initial injury and reconstruction.[8] The lack of any difference in risk for subsequent ACL injury between limbs may be the result of these factors being person specific rather than limb specific, as was suggested by Faude et al[6] after they noted that when treating limbs as individual cases, the risk for reinjury was no longer associated with that of previous ACL injury.

The goal of ACLR is to provide mechanical stability to the knee and allow for return to a person's previous level of physical activity or sport. Reconstructive surgery has been shown to accomplish these goals, allowing persons to continue participating in sports and recreational activity.[2,3] It also helps to protect against subsequent injuries to the menisci and articular cartilage that would require additional surgery.[2,36] As noted previously though, ACLR and returning to sport may increase the risk for reinjury. With ACLR, however, there is also a high incidence of knee OA and degenerative changes of the articular cartilage.[2,37-39] Even within subjects, the incidence of OA is significantly higher [38], and there is less subchondral bone area[40] in the reconstructed knee as compared to the contralateral healthy knee. The presence of degenerative changes and severity of knee OA after ACLR appears to be highly influenced by concomitant meniscal injury[41-43], and/or articular cartilage injury.[2,37] Graft type used and the amount of anterior-posterior laxity that remains after reconstructive surgery does not determine the severity of knee OA in those with ACLR.[37,38,41] Salmon et al[43] did note, however, poorer radiographic findings for those with a loss of knee extension or greater laxity during a lachman test.[43]

There are contrasting findings for mensical involvement.[2,39] Jarvela et al[39] found that the presence of accompanying injuries, which included meniscal injuries, to an ACL injury only increased the number of subsequent surgical procedures performed on the knee, and did not affect radiologic findings of articular cartilage degradation. Similar findings were noted by Lebel et al[2], indicating no difference in the rate of OA for those with meniscal involvement. They did find however, that delaying ACLR with the presence of a meniscal injury for more than one year was associated with an increased incidence of degenerative changes at the medial knee.[2] The contradictory findings give some indication that degenerative changes are influenced by the integrity of the meniscus, but they may also be influenced by other extraneous factors, such as how a person moves and loads the knee after ACLR.

For long term outcomes, ACLR may still be better than conservative treatment though, as Mihelic et al[3] found that severity of knee OA was greater for those who received conservative treatment relative to those who had surgical reconstruction. Therefore, an interaction between providing mechanical stability and properly using and loading the knee may be best to manage the sequela of reinjury and joint degeneration following initial ACL injury and ACLR.

Summary of ACL Injury as a Risk Factor for Subsequent ACL Injury and Knee OA

To summarize the findings of the risk for subsequent injury, those with previous ACL injury are at a greater risk for subsequent ACL injury, the risk for reinjury is not knee dependent, nor graft dependent. Risk factors for subsequent injury are return to sport within 12 months for reinjury of the reconstructed knee, and return to participating in more demanding sports or activities. These findings suggest that it may be important to examine bilateral lower extremity biomechanics within those with ACLR and compared to those with no history of ACL injury, as risk factors may be person dependent and not limb dependent. Also, these findings suggest that risk factors for ACL injury are not being corrected with surgical intervention or rehabilitation. Therefore, understanding the lower extremity biomechanics before and after initial ACL injury and reconstruction may help identify factors that increase the risk for subsequent injury.

The true number of ACL injuries that occur in the United States is unknown, however, the number of surgeries to reconstruct the ACL has been estimated to have grown by 67.8% in 10 years.[1] This means that there is a growing population of persons with ACLR that are going to be at greater risk for recurrent ACL injury and knee OA which, in the long term, could put a financial burden on the health care system. Besides the financial costs associated with these injuries, there are also expected to be declines in physical capacity, which may limit one's ability to engage in physical activity to prevent disease and reduce quality of life.

2.2 Factors Contributing to the Loading of the ACL

Injury to the ACL, as with any tissue, occurs when the loads placed on it exceed the mechanical properties of the tissue. Excessive loading of the ACL, therefore, can increase the risk of injury. Alignment of the tibia relative to the femur, and forces, whether internally generated by soft tissue or externally applied by interaction with the environment, can influence the loading of the ACL during dynamic activity. A review of known factors that load the ACL may help identify factors that are important for understanding why those with ACLR are at an increased risk for subsequent ACL injury, this is with the assumption that ACLR is performed with a proper technique to replicate the *in situ* behavior of the native ACL. The purpose of this section of the literature review is to identify factors that increase loading placed on the ACL, are associated with non-contact ACL injury, and are predictive of non-contact ACL injury.

ACL Loading

Butler et al[9] demonstrated that the ACL is the primary restraint to anterior tibial displacement. This contribution is present regardless of knee flexion angle, as it provided approximately 87% of the restraint at 30° of knee flexion, and approximately 85% at 90° of knee flexion.[9] In addition, the ACL is able to withstand a significant amount of loading before rupture, approximately 2,100 Newtons of force in young cadaveric specimens.[44]

An anterior direct force placed on the tibia is able to load the ACL at all angles of knee flexion.[15] The extent of loading though, is sensitive to the angle of knee flexion.[45] The ACL undergoes greater strain and loading at lower degrees of knee flexion, particularly 30° of knee flexion or less.[14,15,45,46] The majority of these findings have been demonstrated using cadaveric models, but Beynnon et al[46] were able to measure strain of the anteromedial bundle of the ACL *in vivo*. Strain was significantly greater during a Lachman test at 30° of knee flexion as compared to an anterior drawer at 90° of knee flexion.[46] These findings indicate that the amount of knee flexion can affect the strain of the ACL or the load placed on it; with lower knee flexion angles being more likely to increase loading of the ACL. It can be inferred that the combination of an anterior directed force and lower knee flexion angle might place greater load on the ACL and increase the likelihood of injury during dynamic tasks.

The ACL is not sensitive to the conditions of the sagittal plane alone. Fukuda et al[10] noted a relationship between knee flexion angle and *in situ* force in the ACL when an external valgus torque was applied. In accordance with the finding associated with an anteriorly directed force, they noted that *in situ* force was particularly high at knee flexion angles closer to full extension.[10] A similar pattern has been found with the application of an external varus torque increasing ACL tension through the knee flexion range of motion.[11] Again, the greatest strain was noted at knee flexion angles between 0° and 30° of knee flexion.[11] Miyasaka et al[11] noted, in agreement with Fukuda et al[10], a significant increase in ACL strain with the application of an external valgus torque, they did note that this increase was relatively small. In contrast to both of these studies, Berns et al[14] found that neither isolated varus nor valgus torque had any effect on straining the ACL in cadaveric models. The differences between the findings of these studies may be explained by a relatively large amount of time between studies, and advancement in techniques in assessing cadaveric models that may make it easier to find changes in ACL loading from frontal plane torques. Therefore, it is likely that external frontal plane torques that are applied to the tibia increase strain on the ACL, particularly at low knee flexion angles as the more recent studies have demonstrated.

There is a limited amount of research that has examined the effect of isolated axial torque on loading of the ACL. Miyasaka et al[11] demonstrated that the application of an internal rotational torque increased loading of the ACL at knee flexion angles between 0° and 45° of knee flexion. Loading of the ACL with the application of an external rotational torque was relatively small though.[11] These findings indicate that internal rotation of the tibia relative to the femur increase loading of the ACL. Again, this loading appears to be limited to relatively low knee flexion angles, when the knee is closer to an extended position.

Understanding the effects of isolated external loading, as have been described up to now, may not be a proper analog to in vivo conditions during dynamic tasks that occur during sport or exercise participation. Representations of multiplane loading and muscle forces are more appropriate for identifying how the ACL is loaded during dynamic activity. The evidence for the effects of multiplane loading on ACL strain are less consistent than the findings for isolated, uniplanar loading. Berns et al[14] originally found that the addition of an external valgus torque to an anterior directed force on the tibia significantly increased ACL strain as compared to the anterior force alone. The addition of an axial torque, internal rotation and external rotation, though had no significant effect on ACL strain.[14] This is particularly interesting because the same authors found that isolated transerve or frontal plane loading significantly increased ACL strain.[14] These findings are in opposition to those of Markolf et al[15], noting the combination of an anterior tibial force and internal rotation torque near full extension was one of the key loading combination for high ACL forces. They also found that anterior tibial force with a valgus moment also produced high ACL forces, as well as the application of valgus moment to a knee already loaded with internal rotation torque also significantly increased ACL strain.[15] Kanamori et al[16] had complimentary results, in which a simulated pivot shift, valgus torque with internal rotational torque, produced greater anterior tibial translation near full knee extension as compared to the application of internal rotational torque alone. Internal rotational torque and valgus moment appear to primarily load the anteromedial bundle of the ACL as compared to the posterolateral bundle, although elevated forces are still present in both.[47] Durselen et al[17] though, found that combined varus and internal rotation torque strained the ACL between 20° and 40° of knee flexion. There was no significant effect for the combination of valgus and external rotational torque.[17] These findings seem to indicate that axial loading of the knee may have more of an effect on ACL loading than frontal plane loading. This is what Markolf et al[15] concluded, citing that the risk of ACL injury from the addition of valgus moment in combination with internal rotational torque was no different than the risk of injury from internal rotational torque alone.

Simulated muscle forces acting at the knee can have a significant effect on the loading of the ACL.[17,46,48-50] Beynnon and colleagues[46] were able to measure ACL strain in vivo during isometric quadriceps contractions. They found that an isometric quadriceps contraction significantly increased strain on the ACL at 30° of knee flexion when compared to the strain observed at 90°.[46] This relationship also occurs for in vitro cadaveric models as well.[17,49,50] Not only does isolated quadriceps contraction increase strain in the ACL, but DeMorat et al[48] demonstrated that high quadriceps loading has the potential to injure the ACL under circumstances that simulate a non-contact ACL injury. In addition, Li et al[50] observed that a simulated isolated quadriceps muscle force produced tibial internal rotation at lower levels of knee flexion (0°-30°) in a cadaveric model. Hamstrings muscle force, acting as the antagonistic muscle to the quadriceps muscle force, is able to counteract loading of the ACL.[49,50] Draganich and Vahey[49] first demonstrated this in 1990 using a cadaveric model. They found that simulated a hamstrings muscle force at lower knee flexion angles was able to significantly reduce the strain placed on the ACL by an isolated quadriceps muscle force.[49] Li et al[50] found complimentary findings nine years later, noting the addition of a hamstrings muscle force reduced the in situ forces of the ACL at 15°, 30°, and 60° of knee flexion.

These findings discussed so far have demonstrated ACL loading in non-weight bearing, but to understand factors that load the ACL to determine risk for injury it would be best to incorporate the component of weight bearing in the analysis. Weight bearing alone increases the strain on the ACL when measured *in vivo*.[13] In fact, Cerulli et al[12] reported in a single subject case report, that *in vivo* ACL strain was greatest at the point of initial contact with the ground during a single leg landing when the knee was relatively extended. In the sagittal plane, the effect of knee flexion angle on the *in situ* forces of the ACL are similar to those seen in non-weight bearing, with forces peaking at 15° of knee flexion.[51] There appears, however, to be no difference in the effect of an anterior shear force on ACL strain between weight bearing and non-weight bearing,[13] suggesting that weight bearing does not change the loading characteristics of the ACL in the sagittal plane.

In the frontal plane, Fleming et al[13] that frontal plane loading of the knee with a varus and valgus torque was significantly higher when weight bearing was simulated as compared to a non-weight bearing condition. The authors noted, however, that loading of the ACL did not differ with the application of a varus or valgus torque in weight bearing, concluding that the ACL is not a primary restraint to frontal plane loading in weight bearing.[13] The differences were likely the result of weight bearing alone.[13] When Withrow et al[52] simulated a jump landing with impulsive loading, they found the addition of a valgus moment increased *in vitro* ACL strain. These results may be more informative as they are representative of a more dynamic condition that may more accurately simulate *in vivo* loading of the ACL.

In contrast to the findings reported in non-weight bearing, internal rotational torque of the tibia does not appear to load the ACL in weight bearing.[13,51,53] Fleming et al[13] found that the strain of the ACL in the presence of internal rotational torque did not change between non-weight bearing and weight bearing. They did, though, find that there was a significant increase in ACL strain with weight bearing when an external rotational torque was applied. Similar findings are reported by Lo et al[51] and Wunschel et al[53]. Lo et al[51] concluded that the ACL plays a limited role in resisting axial rotation during weight bearing after noting that internal rotational torque did not change the in situ forces of the ACL. Wunschel et al[53] came to a similar conclusion after finding that the transection of the ACL had no effect on tibial rotation during weight bearing knee flexion. A similar finding is described during a simulated pivot landing by Oh et al[54], in which ACL transection resulted in a small increase in tibial internal rotation. The increase in internal tibial rotation was significant after ACL transection, but the authors noted that the relative increase was so small that it was not meaningful given the rather large increase in anterior tibial translation.[54] This is an important note as the cadaveric model used was placed under large muscle loads that would be expected with a pivot landing, and as seen with the non-weight bearing models, muscle forces can affect loading of the ACL.

Summary of ACL Loading

The ACL appears to be most sensitive to loading from an anteriorly directed force on the tibia. Although an anterior force applied throughout the range of knee flexion loads the ACL, loading is greatest when the knee is in a relatively extended position. These conditions are not sensitive to weight bearing, indicating they are consistent factors for loading of the ACL. Externally applied valgus torque and internal rotational torque load the ACL in a non-weight bearing condition. These trends are not as consistent when weight bearing is included in the cadaveric model, and in fact, for axial loading it reverses with external rotational torque loading the ACL in weight bearing. Regardless, the key factors for ACL loading appear to be an anteriorly directed force that can be produced by the quadriceps, with the knee in a relatively extended position, with external moments applied in the frontal and/or transverse plane. Therefore, it would be inferred that a person who demonstrated these lower extremity patterns while performing dynamic tasks would place greater load on their ACL and have a greater risk for injury.

The techniques used for assessing ACL loading are limited by the lack of kinematic and kinetic factors acting at the thigh and hip. Cadaveric models provide much of the evidence, but understanding movements associated with ACL injury may be more informative. Inclusion of such factors may improve our understanding of risk for ACL injury.

2.3 Observed Mechanism of Non-Contact ACL Injury

Another mean by which to assess what movement patterns load the ACL and place a person at greater risk for injury, is to examine the kinematics of individuals while suffering a non-contact ACL injury. The most common mechanism of injuries observed involve landing from a jump[18-20] or a plant-cut maneuver.[20] At the ankle, those that have injured their ACL have been described as landing with less plantar flexion at initial contact with the ground and maintaining this position during the time of injury as compared to selected controls.[21] In addition, Olsen et al[20] described those injuring their ACL as having the foot firmly fixed to the floor and placed outside of their knee during a plant-cut or foot firmly fixed to the floor and externally rotated during one-legged landing in female handball players.

At the knee, Olsen et al[20] described female handball players as having the knee in extension with valgus alignment tibial rotation during injuries that occurred both during plant-cut and one-legged landings. Those with ACL injury have similar amounts of knee flexion as those who do not suffer injury during similar movements.[21,55] Within those who suffered ACL injury, females had greater knee flexion as compared to males at a point shortly after initial contact with the ground.[19] When examining frontal plane knee kinematics, Boden et al[21] found no difference in knee valgus angle at the point of initial foot contact between those who did and did not suffer an ACL injury. They did note, however, that those who injured their ACL moved into more knee valgus after initial contact.[21] Of those who suffer ACL injury, Hewett et al[55] found that females had more knee valgus than males who injured their ACL, and went through more knee valgus as compared to females who did not injure their ACL. This is similar to the findings of Krosshaugh et al[19], who noted that females went through more knee valgus as compared to males even though there were no differences at the time of initial contact with the ground.

Description of the position of the hip and thigh during ACL injury is limited. Those suffering ACL injury have been described as having grater hip flexion as compared to a sample of controls but demonstrating no difference in hip abduction.[21] More information regarding the description of hip motion is provided by Krosshaug et al[19], as in their description of ACL injury they noted that females displayed a "valgus collapse" mechanism more often than males. What they describe as "valgus collapse" consists of hip internal rotation, knee valgus, and external rotation of the tibia.[19]

Summary of Observed Mechanism of Non-Contact ACL Injury

Of the limited number of non-contact ACL injuries that have been captured on videotape and analyzed, an attempt, though limited, at describing lower extremity movement patterns associated with the injury can be developed. Injuries appear to occur during tasks that involve landing from a jump and planting the foot and cutting or changing direction. Unfortunately the majority of research related to the identification of movement patterns associated with ACL injury has been directed at identifying differences between genders during injury. It is unfortunate because the rationale follows that the motion patterns that females display are associated with ACL injury and disregard the fact that both males and females injure their ACL by non-contact means. A complete picture of ACL injury, therefore, may not be developed.

At the ankle, those who injure their ACL land with less plantar flexion of the ankle, and with the ankle placed outside of the knee with the shank rotated. The knee tends to be in an extended position, though the amount of knee flexion may be no different than those who do not suffer an ACL injury. The amount of frontal plane motion at the knee also appears to be important, and may have some gender implications in those who injure their ACL. This emphasis has been placed on positioning of the hip, but it appears that the amount of hip flexion and internal rotation may be important factors for the non-contact mechanism of ACL injury. Again, what is missing from this body of literature is a full description of hip kinematics that are associated with ACL injury. Not only may factors related to the hip influence risk for initial ACL injury but may also add to the understanding of why those with ACLR have an increased risk for subsequent ACL injury.

These findings agree relatively well with factors identified in vivo and in vitro to load the ACL. However, obtaining accurate quantitative data of lower extremity kinematics from videotape data is hampered by technical limitations that may skew the findings because of the inability to confirm that motions occurred within the plane of view. The time of ACL injury is also not known with these videotape descriptions. This is particularly important as Meyer et al[56] recently noted that the motions observed immediately after ACL rupture are not representative of the relative motions that induce strain in the ACL. In a cadaveric study, the authors found that frontal and transverse plane motion of the tibia changed immediately after ACL injury induced by internal tibial torsion and tibiofemoral compression.[56] This evidence indicates that descriptions of ACL injury from videotape analysis may not actually describe factors that load the ACL but may describe the kinematics that result from tibiofemoral compression following ACL injury.

2.4 Prospective Risk Factors for ACL Injury

Describing the loading of the ACL and mechanisms by which non-contact ACL injury occur may not be helpful in understanding what places those with ACLR at greater risk for subsequent injury and knee OA. These factors describe the mechanism, by which the ACL is injured, but they have not been demonstrated in those with ACLR and they do not describe risk factors for injury. Identification of risk factors can begin by analyzing how those with ACLR differ from a matched healthy population and a longitudinal analysis of how ACL injury and ACLR change biomechanics.

For initial ACL injury there is preliminary evidence that specific biomechanical factors place individuals at an increased risk for ACL injury.[22] Hewett et al[22] found that young female athletes who went on to suffer a non-contact ACL injury displayed greater knee valgus (at initial contact and peak values), less maximum knee flexion, greater peak external knee valgus moment, greater peak vertical ground reaction force, and greater peak external hip flexion moment during a double leg drop vertical jump. In addition, a logistic regression analysis indicated that peak knee external valgus moment and knee valgus angles were predictive of non-contact ACL injury.[22] These results may be considered preliminary because the authors captured a limited number of injuries in a very specific population. In addition, the methodology and statistical analyses performed may have influenced some results as kinetic measures were not normalized to body mass or height, and the control group consisted of pooling the data from both limbs of those that did not suffer injury which likely reduced the within subjects variance and made the statistical analysis of between group comparisons more powerful.

One factor that is often over looked in this study however, is the presence of between limb differences in those that went on to suffer an ACL injury and the lack of such a difference in the healthy controls. Specifically, the injured group had a significant bilateral difference in external knee valgus moment prior to ACL injury.[22] This is of important note because bilateral differences are commonly observed in persons after ACLR, and this finding provides some evidence that such differences may not result solely from ACLR, but may be present before initial injury.

Bilateral differences in lower extremity biomechanics, as described by Paterno et al[23], were predictive of subsequent ACL injury in those with ACLR. Differences in internal knee extension moment between the two limbs, with the uninvolved limb demonstrating less internal knee extension moment as compared to the involved limb, during a double leg drop vertical jump was predictive of a second ACL injury.[23] Other factors that were predictive of a second injury included having less internal hip external rotation moment in the uninvolved limb, having greater frontal plane knee motion in the involved limb, and having deficits in single leg postural stability of the involved limb as compared to those who did not experience subsequent injury.[23] These results, again, are preliminary as a relatively small young sample was used for this analysis, and these results may only be applicable to the specific population this sample represents.

Summary of Prospective Risk Factors for ACL Injury

These findings indicate that predictors for initial non-contact ACL injury are similar to the predictors of a subsequent non-contact ACL injury after ACLR. The findings do not agree completely, which may be the result of assessing different populations, but it appears that the biomechanics of the hip and biomechanics of the knee in the frontal and sagittal planes influence the risk for non-contact ACL injury. In addition, there is preliminary evidence that bilateral asymmetries related to these factors may also be important for understanding the risk for ACL injury as they have been found in separate studies to be present before initial ACL injury and before a second ACL injury. It is unknown though if biomechanics change after ACL injury and ACLR. Therefore, it is not possible to determine if what makes a person more susceptible to a second ACL injury after ACLR is a product of the loss of the native ACL and surgical technique, or is the result of underlying neuromuscular factors that are not being adequately addressed with rehabilitation and conditioning.

2.5 Biomechanical Characteristics of ACLR

Persons that undergo anterior cruciate ligament reconstruction surgery after ACL injury display differences in kinematics and kinetics, both as compared to healthy individuals with no injury and contralateral differences between the reconstructed and noninjured leg.[24-31,57-61] These differences are consistent across a variety of tasks as well. Often those with ACLR demonstrate differences despite completion of a formalized rehabilitation program, and return to a level of physical activity or sport. The consistency in differences and their presence across multiple tasks indicate that persons with ACLR use their reconstructed limb differently, which may be what places them at a greater risk of subsequent injury or abnormal joint loading that predisposes them to knee OA.

Kinematic Characteristics of ACLR

As would be expected, an ACLR knee displays kinematic differences when compared to its contralateral healthy knee and healthy individuals with no history of ACL injury. During gait, persons with ACLR demonstrate differences in sagittal plane kinematics as compared to noninjured persons.[61] Webster et al[61] examined gait patterns in those with ACLR and compared those who had a bone patellar tendon bone (BPTB) autograft, a 4-strand hamstring tendon (HS) autograft, and healthy individuals with no lower extremity abnormalities. As a group, those with ACLR demonstrated similar sagittal plane hip and ankle kinematics as the noninjured group.[61] Those with a BPTB autograft, however, displayed less peak knee flexion during midstance when compared to the healthy controls. This difference in knee flexion was not present in the ACLR members with a HS autograft.[61] The kinematic patterns observed in the ACLR knee, though, may be dependent on the nature of the movement performed. During a relatively low demand double leg squat, Salem et al[24] noted only differences in the amount of ankle dorsiflexion between the reconstructed and healthy leg. When performing a double leg weighted squat, persons with ACLR had less dorsiflexion in the reconstructed limb, but displayed similar sagittal plane motion at the hip and knee.[24] This analysis was limited to the sagittal plane only though.

Differences observed in the ACLR knee appear to be greater when the physical demand of the movement is greater or the limbs are required to act independently.

Individuals with ACLR that have been cleared by a physician to return to light sport activity demonstrate significant differences in frontal and transverse plane motion of the ACLR knee as compared to the noninjured knee during downhill running.[25] Tashman et al[25] utilized dynamic radiographic stereophotogrammetric analysis (RSA) to assess between limb differences in tibiofermoral kinematics. They found that the ACLR knee was significantly more externally rotated and adducted as compared to the noninjured knee, but found no differences in knee flexion.[25] The authors noted that the absolute differences in frontal and transverse plane were small, but when considered relative to the total motion in these planes at the tibiofemoral joint they are very large differences.[25]

Using the same RSA technique, Deneweth et al[26] assessed tibiofemoral joint kinematics during a single-legged hop landing. The participants in this study were relatively similar to those Tashman et al[25] analyzed; graft type used for ACLR was not restricted, and all participants had returned to light physical activity. When the single-legged hop landing was performed with the ACLR knee, individuals demonstrated less knee flexion at initial contact with the ground, as well as less internal rotation, and less lateral displacement of the tibia relative to the femur.[26] Significantly less maximum knee flexion and less maximum internal rotation during the landing phase of the task was also noted in the ACLR knee.[26] The ACLR knee also demonstrated greater maximum anterior tibial displacement relative to the femur. This work suggests that when using the ACLR knee to land from a jump, individuals land and stay more extended and with more tibial external rotation. These studies are also beneficial, as they assessed all three planes of motion, not just the sagittal plane. Incorporation of movement from the other planes gives a more complete indication of kinematic differences associated with ACLR.

These findings are consistent with other studies that have examined kinematics of ACLR knees during single-leg landing.[27,60] Orishimo et al[27] assessed between limb differences in persons with ACLR during takeoff and landing from a single-leg hop test. When the participants performed the single-leg hop with the ACLR knee, they displayed significantly less sagittal plane motion at the hip, knee and ankle during takeoff and landing.[27] Not only did the participants move differently when landing, but also used less sagittal plane motion to perform the hop. All the participants were classified clinically as having a normal single-leg hop test, where the distance they could hop with the ACLR leg was within 85% of the noninjured leg.[27] This may indicate that whether an individual can perform a task with their ACLR knee isn't important, but it is how they perform it. Graft type was not considered by Orishimo et al[27], but Webster et al[60] found that between limb differences in knee flexion during single leg hop tests was present for those with BPTB autograft but not those with a HS autograft. The differences in knee flexion of the BPTB group were consistent for both a horizontal single-leg hop test and vertical horizontal single-leg hop test.[60]

In contrast to these findings, Vairo et al[28] found that when persons landed on their ACLR leg they landed in a more flexed position and achieved more sagittal plane motion as compared to the noninjured leg and a group of healthy matched controls. When landing on the ACLR leg, they landed with greater hip flexion at initial contact with the ground, and had greater hip and knee flexion at the moment of peak vertical ground reaction force (vGRF) as compared to when they landed on the noninjured leg. When compared to healthy uninjured matched controls, they had greater hip flexion at initial ground contact, and greater hip flexion, knee flexion, and ankle dorsiflexion at the moment of peak vGRF.[28]

Ortiz et al[59] found that females with a longer time period, an average of approximately 7 years, after ACLR had no between limb differences in peak hip and knee kinematics during a single-leg drop jump and single-leg hop task. The lack of differences were consistent in all three planes of motion.[59] In addition, the peak hip and knee kineamtics of those with ACLR were similar to healthy noninjured individuals. Nyland et al[58] also examined single-leg landing adaptations in a group of individuals with ACLR that were farther removed from the time of their injury, 2-11 years. They found that when individuals performed a repetitive single-leg counter movement jump, there was no difference in how persons moved their leg when performing the task on the reconstructed and noninjured leg.[58] They found, in particular, that the peak flexion angles of the hip, knee, and ankle, as well as the mean knee flexion velocity at each joint were not different between legs.[58] Gender did not appear to influence how the reconstructed limb was used either.[58] While in contradiction to the previous studies, these studies appear to indicate that sagittal plane kinematics during single-leg movements are not consistent across groups of individuals with ACLR. In addition, the differences are not present in individuals who are farther removed from the time of injury and ACLR.[58,59]

The evidence of bilateral differences in sagittal plane kinematics in persons with ACLR appears relatively equal for and against. These studies though are conducted with single-leg movements and limit the analyses to sagittal plane motion. To understand how ACLR impacts more realistic activities it may be best to assess the individual to use both limbs as between limb kinematics may not be present in the sagittal plane during double leg tasks.[24]

In a comparison of differences in sagittal plane biomechanics between those with ACLR and healthy noninjured matched controls, Decker et al[29] found that those with ACLR landed with the hip and ankle more extended initially, but went through greater sagittal plane motion and velocity at the ankle during landing. Interestingly, participants in both group were matched for how softly they landed, and even with this factor accounted for, those with ACLR tended to land in a more extended position.[29] Again, analysis was limited to sagittal plane motion.

Summary of Kinematic Characteristics of ACLR

The literature is not conclusive, but it appears that persons with ACLR display bilateral differences in kinematics. They are also kinematicallly different from persons who have never suffered an ACL injury. The literature demonstrates a consistent use of averaged values at specific time points. This approach limits the interpretation of movement differences as dynamic motion is reduced to a single number.

The lack of extensive research outside of the sagittal plane is a major limitation of the current research devoted to the kinematics after ACLR. Other planes of motion need to be assessed, particularly since recent evidence indicates that frontal plane motion of the knee is predictive of secondary injury in persons with ACLR.[23] Paterno et al[23] tested and prospectively followed a cohort of athletes who had completed rehabilitation and returned to their prior level of activity. They found that persons that later suffered an additional ACLR injury had increased frontal plane motion of the reconstructed knee during the landing phase of a double leg drop vertical jump as compared to those with ACLR who did not suffer an additional ACL injury. To better understand the effect of ACLR on risk factors for additional ACL injury and knee osteoarthritis, motion in the frontal and transverse plane at the knee and hip should be analyzed.

Kinetic Characteristics of ACLR

Not only do those with ACLR display differences in how they, both bilaterally and as compared to noninjured persons, but they also display some differences in the kinetics experienced during dynamic tasks. The kinetic differences between noninjured and persons with ACLR are more evident, though with more demanding tasks.[57]

Bush-Joseph et al[57] compared sagittal plane moments of the knee across low and high demand tasks between a group with ACLR and a group of noninjured participants. The authors noted that the kinetic profiles of the two groups were very low demand tasks such as walking and stair-climbing. The ALCR group demonstrated a slight decrease in external knee flexion moment during gait, when the effect of walking speed was introduced as a covariate.[57] These findings are somewhat similar to those of Webster et al[61] who noted minimal differences in sagittal plane kinetics when comparing ACLRs based on graft type to healthy noninjured persons; only those with a HS autograft showed reduced external knee extension moment during the terminal stance of gait. The more demanding tasks of jogging and jog-cut demonstrated more distinguishable group differences, as those with ACLR had significantly less peak external knee flexion moment than the noninjured group.[57] While these findings are again limited to the sagittal plane, they warrant the use of more demanding tasks to characterize the kinetic differences of those with ACLR.

Other authors have also noted differences in the kinetic profiles of those with ACLR. During single-leg movements there appears to be trend towards individuals trying to protect the reconstructed knee by minimizing the forces acting on the limb, reducing the internal moments produced at the knee, and increasing the internal moments produced at the hip and ankle.[27,28,31,58-60] Ernst et al[31] found that the reconstructed limb of those with ACLR had less internal knee extension moment as compared to the noninjured limb during a single-leg vertical jump. This was due to less internal extension moment of the reconstructed limb and not greater internal extension moment in the noninjured limb, as the noninjured limb did not differ in the magnitude of internal knee extension moment as compared to a healthy noninjured group.[31] In addition, when the authors summated the extension moment of the entire reconstructed limb, it was significantly less than that of both the contralateral limb and the noninjured healthy group.[31] These differences in internal knee extension moment were also found by Ortiz et al[59], who noted lower values in the reconstructed knees as compared to the noninjured limb during a single-leg vertical hop task. These findings weren't consistent within the sample across tasks, because there were no bilateral differences between limbs during a single-leg drop jump for internal knee extension moment.[59]

Both Nyland et al[58] and Vairo et al[28] found that when landing on a single-leg, those with ACLR produced significantly less peak vGRF when using the reconstructed limb as compared to the noninjured limb. However, Vairo et al[28] also noted that there was no bilateral difference in internal hip extension moment or summated extension moment for the ACLR group. These findings are in conflict with those of Orishimo et al[27] who found that there was no bilateral difference in peak vGRF during single-leg landing or takeoff. However, they also found no bilateral difference for internal extension moment at either the hip, knee, or ankle during landing.[27] During takeoff, however, the noninjured knee did produce greater internal knee extension moment as compared to the reconstructed knee.[27] Another interesting finding was the lack of a significant difference in anterior tibial shear force experienced by the reconstructed knee, as this force is believed to represent loading of the ACL.[27]

Graft type used for reconstructed was not directly compared in these studies, but the findings of Webster et al[60] provide some evidence that the kinetic profiles of persons with ACLR may be in some respects influenced by graft type. They found that when comparing persons with BPTB autograft to those with HS autograft, the BPTB group had bilateral differences in external knee flexion moments and peak vGRF, with higher peak vGRF and lower external knee flexiom moment occurring when landing on the reconstructed leg.[60] This was despite the lack of bilateral differences in summated external flexion moment.[60] The HS autograft demonstrated no bilateral differences in external knee flexion moment, peak vGRF, or summated external flexion moment.[60]

These differences in kinetics seem to be present when persons with ACLR perform double-leg tasks as well. In contrast to the other findings of decreased internal knee extension moment in the reconstructed leg, Salem et al[24] noted greater internal knee extension moment in the reconstructed limb during a weighted parallel squat. Also noted was a greater ratio in regards to the peak internal hip extension moment relative to that of the knee, which they explained as those with ACLR using more hip extension rather than knee extension moment in the reconstructed limb to complete the squat.[24] These bilateral differences were present despite no difference in peak vGRF.[24]

During a double-leg drop vertical jump, Paterno et al[30] observed similar bilateral differences in young female athletes with ACLR as was noted during single-leg tasks. There was significantly more peak vGRF with the noninjured limb during landing, which was also greater when compared to a group of healthy participants that had never suffered an ACL injury.[30] The bilateral differences in peak vGRF were also present during the takeoff phase of the task.[30] The authors took into account the rate at which the vGRF was applied to the body, and noted that the noninjured limb had a greater loading rate as compared to the reconstructed limb and healthy noninjured controls.[30] These findings suggest that this sample of female athletes with ACLR unevenly loaded their lower extremities when landing from a jump; placing greater force on the noninjured leg and loading it at a greater rate than the reconstructed leg.[30] In this case the compensation was to place greater load on the noninjured leg, as the vGRFs of the reconstructed leg were no different than those of the healthy noninjured controls.

Similar patterns were found by Decker et al[29] when they compared those with ACLR to a control group. As mentioned previously, the authors matched for whether subjects performed a soft or stiff landing.[29] Despite this they noted that the ACLR group had significantly lower peak internal hip extension and knee extension moments during landing.[29] While there was no difference between groups for peak vGRF, the ACLR group did reduce the loading rate of the peak vGRF as compared to the healthy control group.

With the presence of some inconsistencies in the literature there appears to be a trend in the research available indicating that kinetic bilateral differences exist in those with ACLR. The major deficit of the research thus far, is its limitation to analyzing sagittal plane kinetics and not exploring kinetic differences in the frontal or transverse plane. This deficit in the research is highlighted by the recent evidence provided by Paterno et al[23] that transverse plane kinetics at the hip are predictive of secondary injury in those with ACLR. Those who went on to suffer a second ACL injury after initial ACLR and return to pre-injury physical activity/sport level had less hip external rotation moment in the noninjured leg during landing.[23] Bilateral differences in internal knee extension moment were also predictive of subsequent injury, with the reconstructed limb producing significantly greater moment than the uninvolved being associated with injury.[23] The differences in internal hip external rotation moment was the variable that was most predictive of secondary injury.[23] This provides evidence for exploring bilateral difference of kinetics at the hip and knee, as well as all planes of motion to better understand why those with ACLR may be at increased risk for subsequent injury.

Summary of Kinetic Characteristics of ACLR

The research overall indicates some general trends in the kinetic profiles of those with ACLR: unloading or protecting of the reconstructed knee and placing greater load on the other joints of the lower extremity. As noted, it may be more effective to assess the kinetic variables in all three planes of motion, both the hip and the knee, under relatively demanding tasks.

Summary of Biomechanical Characteristics of ACLR

A lack in understanding of the biomechanical profile of those with ACLR before ACL injury and reconstruction makes it difficult to determine if the differences previously mentioned are inherent to the individual and important to understanding their risk for future injury, or simply a residual effect of the reconstruction itself that has not adequately been addressed during rehabilitation. The ability to compare individuals pre-injury and post-injury would provide better understanding of the deficits not being adequately addressed in rehabilitation, which may be placing those with ACLR at greater risk for subsequent injury.

2.6 Coordination and Variability Characteristics of ACLR

The limits of using traditional biomechanical measures to describe the movements of those with ACLR have been overcome by the use of some of techniques from Dynamical Systems Theory. These techniques include measures of coordination and non-linear measures that can describe the characteristics of an entire time-series rather than discrete point estimates.[62]

Measures of coordination and variability of human motion have been mainly limited to the lower extremity. These measures have been used to describe the movements of those with ACL deficiency[63-66] and ACLR.[32-35] The variability, stability, and coordination of sagittal plane motion of the thigh, shank, and foot has been characterized by examining differences in values of Approximate Entropy (ApEn), Lyapunov's Exponent (LyE), and continuous relative phase dynamics.

After ACL injury, bilateral differences in sagittal plane variability occur in those that do not undergo ACLR and remain ACL deficient (ACLD). [63,65,66] This difference in variability is consistent in the presence of small perturbations such as changes in walking speed[65] and rather novel constraints such as walking backwards.[66] Stergiou et al[65] first characterized the bilateral differences of the dynamic stability of sagittal plane motion of the thigh and shank of persons with ACLD. The authors used the LyE of the knee flexion-extension time series to characterize the local stability, defined as the sensitivity of the knee to small perturbations, of both knees during walking at different speeds.[65] The ACLD knee demonstrated a larger LyE value, indicating less local stability, as compared to the contralateral noninjured knee. The difference was consistent regardless of walking speed.[65] The authors concluded based on their results that the ACLD knee is less sensitive to perturbations. This indicates that the ACLD knee is more locally unstable, and that the lack of differences across walking speed may indicate that those with ACLD alter their movement patterns to maintain what local stability they do have.[65] One of the interesting factors of the sample of subjects used for this study was the length of time between their injury and the date of testing. The sample mean for the time between injury and testing was 33.5 months, indicating that the bilateral differences were resistant to time. This is contrary to what is observed in traditional biomechanical measures, where studies using ACLR tended to note less bilateral differences in those who were farther removed from the time of their injury and reconstruction.

Georgoulis et al[63] found similar bilateral differences in variability in a group of persons with ACLD that were on average 19.9 months removed from the time of their injury when tested. The authors used a different nonlinear measure to characterize variability in the knee flexion-extension time series during gait across different walking speeds, ApEn.[63] The sample of participants had significantly different amounts of variability between knees; the ACLD knee demonstrated more regular, less variable movement in the sagittal plane as indicated by a lower ApEn value. Again, there was no limb by speed interaction, although ApEn values did increase as walking speed increased.[63] The lower ApEn values of the ACLD knee were interpreted as more regular, predictable motion in the sagittal plane and a decreased ability to adjust to unpredictable perturbations during walking.

The fact that ACL injury can alter movement patterns in the noninjured contralateral knee, prompted Moraiti et al[64] to examine stride-to-stride differences in variability of the knee flexion-extension time series in those who were ACLD and healthy noninjured persons. As with the sample studied by Stergiou and colleagues[65], this group of persons with ACLD were on average 33.5 months past their date of injury when tested. The authors also used the LyE to characterize the sensitivity of the system to the initial conditions of the internal and external environment.[64] The healthy noninjured group had significantly greater LyE values as compared to the involved limb of the ACLD group.[65] The authors interpreted their findings as the ACLD knee being less sensitive to initial conditions, decreased variability in the knee flexion-extension time series, and overall, less complexity in their motion.[64] This may indicate that those with an ACLD knee may constrain the movement of their shank and thigh, but this decreasing the complexity of their motion, making them less able to appropriately handle natural perturbations during walking.[64]

The evidence of these studies indicates that under a very familiar task such as walking, the ACLD knee demonstrates differences in variability in the sagittal plane as compared to their bilateral noninjured knee, and those with no history of ACL injury. Zampeli et al[66] though found that these patterns were consistent when persons with ACLD performed a relatively novel task, backwards walking. Again, they used the LyE value of stride-to-stride variability in the knee flexion-extension time series to perform a within subjects comparison of those persons with ACLD and between subjects comparison of those with no history of ACL injury.[66] Unlike the previous studies, this sample of persons with ACLD was tested at a time close to the date of their injury, a mean of 8.1 months. Their findings were in agreement with the previous works; the ACLD knee exhibited a higher value for LyE as compared to the noninjured contralateral knee, but a lower value than the healthy noninjured control group.[66] The ACLD was again shown to be less variable and less complex in the knee flexion-extension time series as compared to those with no ACL injury, but to be more variable and complex than the contralateral noninjured knee. As may be assumed, the differences in the contralateral limb of the ACLD group may represent an attempt by this population to maintain some level of symmetry between the two limbs during walking.[66]

A logical extension of this research is to determine if those with ACLR exhibit differences in variability as well, as reconstruction is performed to return mechanical stability and normal motion to the knee. Moraiti et al[33] assessed the stride-to-stride variability of those with a BPTB and HS autograft. The authors wanted to determine the functional outcome 2 years after reconstructive surgery. The ApEn of the knee flexion-extension time series during walking was compared to a group of healthy individuals who had never suffered an ACL injury. There was no difference in clinical measures or activity levels between those with BPTB and HS autograft. In addition, there was no difference in stride-to-stride variability between the two graft types. However, those with ACLR, regardless of graft type, demonstrated significantly greater variability, indicated by a larger ApEn value, as compared to the healthy noninjured control group.[33] The authors explained their findings as ACLD results in a decrease in variability, but reconstructive surgery may provide individuals with the comfort to add accessory motion to their movement patterns.[33]

Moraiti and colleagues essentially repeated their previous study in 2010, but examined stride-to-stride variability using LyE value.[34] Again, they compared those with BPTB autograft, HS autograft, and a healthy noninjured control group. This study, however, also included bilateral comparisons for those with ACLR. They confirmed the lack of difference in stride-to-stride variability of the knee flexion-extension time series between the two types of grafts. They also confirmed that those with ACLR, regardless of graft type, had significantly greater variability, as indicated by a larger LyE value. Besides confirming their previous findings, they also noted that the contralateral noninjured knee of the ACLR group was significantly more variable than the reconstructed knee. Thus, it appears that not only does ACLR reconstruction increase sagittal plane variability about the knee, but it also creates increased variability in the contralateral noninjured knee. The authors attributed this finding to those with ACLR trying to maintain some symmetry in the amount of variability between the two limbs.

Identifying differences in the variability of movement may not adequately describe the movement patterns that are associated with ACLR. Kurz et al[32] attempted to better describe the motion after ACLR by using relative phase dynamics to assess the coordination of shank and thigh movement while walking and jogging. This technique provides information about the degree to which two segments are in-phase (move in the same direction) or out-of-phase (move in the opposite direction). Shank-thigh and foot-shank coordination in the sagittal plane was assessed during walking and jogging at a self-selected speed. The mean absolute relative phase (MARP) value was calculated to provide a quantitative comparison. Participants with ACLR were tested on average 3.4 years after the date of surgery, and were compared to a group of gender and age matched control participants. Significant differences in the MARP values for the foot-shank and shank-thigh were present during walking. During the stance phase, the foot-shank motion was more in-phase for the ACLR group, while shank-thigh motion in the sagittal plane was more out-of-phase when compared to that of the healthy noninjured matched controls. Running only produced differences in the coordination between the foot-shank, with the ACLR group demonstrating greater in-phase motion during the stance phase of running. These findings, in addition to the previously mentioned, indicate that not only do those with ACLR have different amounts of variability in the sagittal plane time series, but they also coordinate the movement differently between the foot and shank, and shank and thigh.

Similar differences in relative phase dynamics of the knee and ankle were found by van Uden et al[35] in a small sample of persons with ACLR, approximately one year out from surgery, during a continuous single leg hop task. Those with ACLR did not display bilateral differences in movement coordination, but the reconstructed limb was found to be more in-phase as compared to a healthy noninjured control group. In addition, there was no difference between the coordination of the healthy control group, and the noninjured limb of the ACLR group. Additional analyses were performed to calculate the standard deviation of the relative phase, which is equivalent to the deviation phase (DP), to determine the variability or stability of movement coordination of each limb. Again, the reconstructed limb of the ACLR group had a significantly higher DP value as compared to the healthy controls, and was also higher than the noninjured limb of the ACLR group.[35] This suggests that those with ACLR have different movement coordination, and the coordination is less stable than those with no ACL injury.

Summary of Coordination and Variability Characteristics of ACLR

The research related to movement coordination and variability of the ACLD population indicates that they have bilateral differences in the variability of the knee flexion-extension time series during gait, and that they differ from those that have never suffered an ACL injury as well. In general, the ACLD knee has less complexity in the organization of movement in the sagittal plane. This may cause them to be less able to adapt to unexpected perturbations or changes that occur in the internal or external environment during walking. These differences are present during both relatively long and short periods of time after injury, suggesting that these differences are resistant to change over time, though no longitudinal studies have been performed. These differences also appear to be present despite ACLR.

Assessment of the coordination of the lower extremity segments about the knee has not been as extensively analyzed as measures of variability. These assessments though can provide a more detailed analysis of movement, but also an indication of the stability of the coordination, as demonstrated by van Uden et al[35]. These techniques may also be more appropriate to used in more dynamic tasks, other than gait.

One limitation of the available literature is the lack of analyses assessing coordination outside of the sagittal plane. Kurz et al[32] addressed this as a limitation to their study, indicating that the associated noise in marker motion associated with the other planes prevented them from performing these analyses in multiple planes. Still these planes are worthy of assessing as they may help identify additional information about the movement patterns of persons with ACLR. However, these analyses have a major advantage over traditional biomechanical measures because they provide more detailed information about motion and its sensitivity to changing conditions. A more thorough interpretation can then be generated about the differences in how those with ACLR perform a task, rather than inferring differences from a point estimate.

The previous studies that have examined variability associated with ACLD and ACLR are limited in the amount of information they provide with regards to movement by the analyses used. As the majority of these studies have quantified lower extremity variability with the measures of ApEn and LyE, they are only able to describe the magnitude of the variability associated with these measures. While variability is important to understanding the potential for further injury in this population, the addition of how movement is coordinated in this population would provide greater detail and the potential to understand what interventions might be effective in augmenting injury risk. Only Kurz et al[32] and van Uden[35] have quantified both coordination and variability in this population by measuring Continuous Relative Phase (CRP). However, there are other ways to measure both coordination and variability that may be more appropriate for understanding how ACLR changes intra-limb and inter-limb coordination and variability.

2.7 Coordination and Variability Methodology Review

The two most common techniques for assessing coordination and variability in biomechanical data appear to be Continuous Relative Phase (CRP) and Vector Coding. Both techniques are continuous techniques that allow for the measurement of coordination and variability over an entire time period of interest, providing both spatial and temporal information.[67]

Continuous Relative Phase

Continuous Relative Phase measures the relative phase, or difference in phase angle, of two segments oscillating about a joint at each point across a time period of interest during movement.[62,68,69] To obtain the phase angle of a segment, a phase-plane is created by plotting the angular position of a joint relative to its first derivative, angular velocity, for the given movement cycle of interest.[68]The Cartesian coordinates of the movement trace are then converted to Polar coordinates, in which the position of each data point can be redefined based on the length of a radius from the origin and an angle relative to the right-horizontal.[62,69] The angular portion of the Polar coordinates represents the phase angle of that given data point.[62,69] This relationship can be expressed mathematically as:

$$\theta_i = \tan^{-1}\left[\frac{y_i}{x_i}\right]$$

where θ_i represents the phase angle for a particular segement at time point i, and y_i, x_i represent the Cartesian coordinates for the segments angular velocity and angular position, respectively, at time point i .[62,69] To obtain the relative phase angle, the phase angle of the proximal segment is subtracted from the distal segment (1)[62], or the phase angle of the distal segment is subtracted from the proximal segment (2)[69].

$$(1)\ \Phi_{\text{relative phase angle}} = \theta_{\text{distal segment}} - \theta_{\text{proximal segment}}$$

$$(2)\ \Phi_{\text{relative phase angle}} = \theta_{\text{proximal segment}} - \theta_{\text{distal segment}}$$

The relative phase angle is then calculated for each point for the time of interest during the movement to obtain CRP. Plotting the resulting CRP provides a description of how the movement is coordinated. However, the CRP has to be quantified in some manner to allow for objective comparison between groups. The mean absolute relative phase (MARP) has been previously used for this purpose and can indicate the relationship between two segments, whether their movement is generally out-of-phase or in-phase.[62] Kurz and Stergiou[62] provided the calculation of MARP as:

$$\text{MARP} = \sum_{i=1}^{N} \frac{\left|\Phi_{\text{relative phase angle}}\right|}{N}$$

where N represents the number of data points in the time period of the movement cycle of interest.

Defining the variability of coordination parameters between two segments may provide additional information as to the risk for orthopaedic injury.[68] Quantifying the variability of coordination between two segments, defined as the deviation phase (DP), indicates how stable the coordinative patterns are.[62] Kurz and Stergiou[62] defined the calculation of DP as:

$$DP = \frac{\sum_{i=1}^{N} |SD_i|}{N}$$

where N represents the number of data points in the time period of the movement cycle of interest, and SD represents the standard deviation of the ensemble at point i.

As a measure, CRP analyses provide a way to quantify the coordination and stability of movement between two interacting segments.[62,68] Because it provides information related to angular position and its first derivative, angular velocity, it may give an indication of how the neuromuscular control of movement is organized.[62]

A number of studies have previously used this measure to examine coordination of the lower extremity, mostly during gait.[70-74] In addition, all of theses studies used healthy participants in their studies.[70-74] The focus of these studies has been to examine how different conditions change coordination and variability within the lower extremity.[70,73,74] Li et al[70] examined CRP during walking and running, in a small sample of healthy male participants. The effect of having to clear obstacles during gait has been examined as well.[73,74] These studies have indicated that changes in the external environment have an effect on lower extremity coordination,[73,74] as well as changing the performance constraints of the task can have an effect as well.[70] However, these studies do not provide evidence that coordination and the stability of that coordination is associated with lower extremity injury. Heiderscheit et al[71] did examine differences in coordination and variability of lower extremity coordination between those with high and low quadriceps angle (Q-angle). The authors found that there was no difference in lower extremity coordination variability related to Q-angle during running.[71] Again, these results are limited because it was a healthy sample of participants. These findings do though, give an indication that variability coordination of the lower extremity may not be dependent on structure of the lower extremity, but may provide instead an indication of how movement is coordinated by the neuromuscular system.

Review of Technical Aspects of Data Collection: Continuous Relative Phase

Studies that have previously used CRP as a measure of coordination and variability for biomechanical analysis have consisted of relatively small sample sizes, ranging from 6 to 40 participants.[70-74] An important characteristic of these studies, as well, is that all participants used have been healthy, free of any orthopaedic injury.[70-74] A greater number of trials have been performed than is done with more traditional biomechanical studies, when employing these techniques, 5[70,72] or 10 trials.[71,73-75] The studies that have used 10 trials for analysis have listed variability of the coordination as a primary interest of the study.[71,73-75]CRP appears to have been performed exclusively with tasks that have included analysis of gait, walking or running.[70-75] Coordination and variability analyses have included examination of the interaction of the foot and shank[71-75], and shank and thigh[70-75] in various combinations of plane orientation. These data have been collected with sampling frequencies that are typically used for biomechanical analyses (120-240 Hz) but have been limited to video based systems.[70-75] During data processing, low pass filters have been used to filter the kinematic data, with cutoff frequencies of 8-20 Hz.[70-74]

Limitations of Continuous Relative Phase

Sinusoidal Pattern of Movement

The motion of interest has to be sinusoidal in nature in order to properly employ CRP as a measure of coordination.[67,69,76] In addition, a matching frequency between the two segments is required as well.[67] When analyzing motion in the sagittal plane this may not be an issue during such activities as gait and jumping. Analyzing motion in the other planes, or inter-plane comparisons between segments, may be inappropriate using this measure. These technical issues of applying CRP to various motions may be addressed by normalizing the phase-plane curves, but there is no consistency.[69]

Normalization of Phase-Plane Trajectory

There is some contention in the literature as to proper normalization with regards to CRP. While some authors have argued that there is no need for normalization of the phase-plane data during calculation of CRP due to the processes of the calculation itself[77], it is generally accepted that normalization must be performed to obtain proper interpretation of these measures.[67,69,78] However, Peters et al[78] contended that normalization procedures are not necessary if the movement of interest fulfilled the first assumption of sinusoidal motion.

Normalization procedures have the effect of placing the phase-plane at the center of the graph about the origin, accounting for differences in the magnitude of motion each segment experiences during the time of interest, and resulting in a more circular phase plane.[67,69,78] Although there is no consensus on appropriate normalization[69], Hamill et al[67] described the effect of two common normalization techniques; normalization based on a unit circle, and normalization based on a maximum velocity. Within those techniques normalization was also based on the maximum value per separate trial or based on the maximum value over several trials.[67] They observed that normalization will have an effect on the resulting description of coordination and variability as assessed using CRP, but left the selection of appropriate normalization parameters up to the investigator to decide based on their question of interest.[67]

Determining the Range of the Phase Angles

For CRP to accurately describe relative motion between two segments, a range for the possibility of relative phase values is established. Wheat and Glazier[69] describe the debate as to the appropriateness of specific ranges. As they note, based on the calculations provided earlier, the calculation of the phase angle of a segment will produce values ranging between ±90°.[69] This range of values may not capture the nature of the interaction between two segments and may need manipulating. The authors note that previous biomechanical studies performed by Hamill et al[67] have manipulated the input values for the phase angle calculation to achieve a range between 0° and 180°.[69]The calculation of the phase angle remains the same for all data points where the value of y if y is greater than zero, but if the value of y is less than zero the formula is reformatted as: $y = 180 + \tan^{-1}\left(\frac{y}{x}\right)$.[67] Hamill et al[67] manipulated the calculation in this manner to avoid redundancy in the values produced when a range of 0° and 360° is used. However, there is some controversy as to whether important information is lost with respect to the variability of a system when the range of values is not set to include a full 360° of possible values.[69] The benefit of limiting the range of value to only 180° is the ability to use linear statistics to evaluate differences and calculate descriptive statistics.[69] The benefits of using a range of 360° can be used with the implementation of circular statistics.[69]

Vector Coding

Where CRP uses phase-plane plots to describe coordination between two segments, angle-angle plots are used for vector coding, a separate continuous technique to quantify the coordination between two segments.[69] The technique is based on angle-angle plots of two joints of interest, and unlike CRP does not require normalization of the of the data plots.[67,69] Sparrow et al[79] described procedures to determine the similarities of these plots to allow for comparison across time, populations, or conditions. Since, Tepavac and Field-Fote[80], Hamill et al[67], and Heiderscheit et al[81] presented variations of their methods.[69] The equations presented by Heiderscheit et al[81] appear to be more commonly used for biomechanical studies.[72,82,83] This technique uses the following equation to determine the angle created by the vector connecting two adjacent points on the angle-angle plot relative to the right horizontal,

$$\theta_i = \tan^{-1}\left[\frac{y_{i+1} - y_i}{x_{i+1} - x_i}\right]$$

where i represents the succession of data points in the time of interest.[81] As Wheat and Glazier[69] noted in their assessment of techniques of vector coding, this technique is limited to only provide a description of the orientation of the vectors along the angle-angle plot. They noted that the equations presented by Tepavac and Field-Fote[80] allowed for this assessment as well as the assessment of the length of the vector between data points and the overall shape of the plot. While this additional information may be useful, the overall coordination between two segments seems to be captured in the assessment of the coupling angle.

The range of values for the coupling angle can be 0°-360°.[67] Although, the range of values has been more limited, 0°-90°, in studies that have examined coupling during specific time intervals of a movement.[72,82,83] The value of the coupling angle provides an indication of the magnitude of the relative motion between the two joints.[67] Specifically, a value of 0°, 90°, 180°, or 270° indicates that only one of the segments is moving relative to the other; 0° and 180° indicate movement of the proximal segment, 90° and 270° indicate the distal segment is moving.[67] Values that fall between these, indicate a combination of movement is occurring in both the proximal and distal segment. Hamill et al[67] went on to state that the amount of relative motion is equal when the coupling angle equals values of 45°, 135°, 225°, and 315°, this movement is in the same direction when the coupling angle equals 45° or 225°, and in the opposite direction when the coupling angle equals 135° and 315°.

Review of Technical Aspects of Data Collection: Vector Coding

Vector coding has been previously used to assess coordination and variability of the lower extremity.[72,81-83] The tasks examined have included walking[75], running[72,81,83], and unanticipated cutting[82]. However, the majority of assessments have been performed on healthy participants[72,82,83]. Heiderscheit et al[81] used a pathologic sample when they examined the variability of joint coordination in a group of females with unilateral patellofemoral pain as compared to those without. While Ferber et al[83] did conduct a comparison of runners who had been fitted with orthotics due to injury to healthy runners, both groups were injury free at the time of data analysis.

A review of the methods used for these particular studies provides general guidelines for how to appropriately capture kinematic data that can be used for this type of analysis. A relatively moderate sample size has been for these analyses, and has ranged from 16 participants to 40 participants.[72,81-83] During testing, multiple trials of the movement of interest have been used for data analysis. The number of trials used ranges from 5 to 15, with 5 trials being most commonly used.[72,75,81-83] Kinematic data have been sampled at 120 Hz[72,83] and 240 Hz[75,81,82], although the preference of kinematic sampling appears to be at the discretion of the researcher rather than based on any empirical evidence. These kinematic data are generally filtered using a low pass zero lag filter with a relatively low cutoff frequency, 8-9 Hz.[72,81-83] The kinematic data is usually interpolated to 101 data points that coincide with the time point of interest[72,81-83], such as manipulating the data so that the stance period of gait is consistent across trials and subjects, allowing for consistent comparisons.

Limitations of Vector Coding

Because there is the possibility that values of the coupling angle will range from 0°-360° and these values indicate direction, the use of circular statistics may be necessary to determine the descriptive statistics.[67] This is a minor limitation, as previous researchers have employed and described such methods.[67,81] The use of circular statistics may not be necessary though, as indicated by previous authors who have assessed specific time periods during a movement, and have demonstrated a range of coupling values from 0°-90°.[72,82,83]

The examination of specific time points, or segmenting a movement into particular sections may be necessary to minimize the error in calculation of the coupling angle when there is minimal displacement of either joint.[81] Heiderscheit et al[81] commented that when there is minimal joint displacement the coupling angle can become more sensitive to small changes in joint position and increase variability in the calculation. This situation may arise, in particular, when a joint is changing the relative direction it is moving. Therefore, it may be necessary to segment a movement in to multiple sections to allow for a more accurate assessment of coordination.

Summary of Coordination and Variability Methodology Review

The question becomes is one technique, CRP or vector coding, better than the other. To answer this, Miller et al[75] performed a direct comparison of the two techniques to assess variability in kinematic data. What the authors noted, was essentially a trade off between the two techniques; vector coding being more sensitive to biologic variability, CRP more sensitive to variability derived from theoretical data; CRP being a more conservative measure, but vector coding having greater clinical implication as it is derived from kinematic data that is not normalized or altered.[75] The authors suggested that any decision for the use of one technique be based on goals of the research.[75]

Ultimately, both techniques provide unique information about lower extremity coordination, but the benefits of measures obtained using vector coding outweigh those of CRP. Joint coordination characterized by vector coding, in particular, is relatively easy to compute and interpret, has more clinical relevance, and is more sensitive to actual biological motion. The methodological limitations of CRP that were previously described (normalization, required sinusoidal motion, determining range of phase angles) detract from the use of CRP as a measure of coordination, and are not offset by the ability to include the time dependent component of velocity in reference to measuring coordination relative to the information gained by using vector coding.

2.8 Summary of the Literature Review

The purpose of this study is to examine the effect of ACL injury and ACLR on a sample of young, physically active persons. This study will expand the information regarding risk factors for subsequent ACL injury and the development of knee OA by analyzing hip and knee biomechanics in all three planes of motion. Measures of coordination and variability of the thigh and shank will also be included. This study will be unique with regards to previous studies examining those with ACLR, in that information collected prior to the initial injury will be included.

CHAPTER THREE

METHODOLOGY

3.1 Experimental Design

This study employed a repeated measures, case-cohort research design. The purposes of this study were to (1) determine if lower extremity biomechanics at Baseline are changed by ACL injury and subsequent ACLR, for both the injured and noninjured limbs (2) determine if lower extremity joint coordination at Baseline is changed by ACL injury and subsequent ACLR, for both the injured and noninjured limbs, (3) determine if there are differences in the magnitude of between limb differences in lower extremity biomechanics at Follow-Up for those with ACLR and those with no history of ACL injury, and (4) determine if there are differences in the magnitude of between limb differences in lower extremity joint coordination and variability at Follow-Up for those with ACLR and those with no history of ACL injury. Participants were recruited from a larger on-going study being conducted at the United States service academies (Joint Undertaking to Monitor and Prevent ACL injury; JUMP ACL). Participants who completed biomechanical testing for the JUMP ACL study, suffered one ACL injury since enrollment in the service academies, had underwent reconstructive surgery, returned to full participation, and still enrolled in the service academies, were recruited for this study along with a sample of healthy matched controls. The primary independent variables for this study were injury status (ACLR Group, Control Group) and time

(Baseline, Follow-Up).

3.2 Participants

Participants were recruited from the JUMP ACL study at the three major service academies of the United States (United States Air Force Academy, United States Naval Academy, United States Military Academy). Selection of participants was limited to those that had complete biomechanical data at the time of baseline testing for the JUMP ACL study, were still enrolled at their respective service academy, and were still enrolled in the JUMP ACL study. Forty-five participants were identified as having suffered an ACL injury since Baseline testing during the summer of their enrollment in an academy. Matched controls were also identified at each academy based on the matching factors of gender, service academy, and year of enrollment in the service academy. A list of matched controls was generated with a 3:1 ratio of those with ACL injury, and given a randomly assigned priority number for recruitment. Participants were recruited by means of email, and personal contact on site. Following initial testing for Follow-Up, 4 additional participants who had suffered an ACL injury from the 2009 cohort enrolled in the study, and completed testing. A total of 88 participants were enrolled and completed testing for this study; 38 with an ACL injury since Baseline and 50 matched control participants.

3.3 Instrumentation

Marx Activity Scale

The Marx activity scale was developed to allow for a quick assessment of a patients level of activity without being specific to sport participation as with the Tegner scale.[84] It does not assess the ability of a person to engage in these specific activities, but assesses what they are actually doing. The scale was derived out of review of the limitations of other activity scales commonly used in orthopaedics, and from information provided by patients and sports medicine practioners.[84] Its' measures are correlated with other well-established surveys used to measure activity level in patients with knee injuries.[84] Because the scale quantifies specific activities and not just participation in particular sports that are believed to have higher rates of these types of movements, the authors suggest that it may be a better assessment of activity.[84] This scale has previously been used to assess long-term outcomes of those with ACLR.[85,86]

Knee Injury and Osteoarthritis Outcome Score

The Knee Injury and Osteoarthritis Outcome Score (KOOS) is a self-administered questionnaire for patients to indicate their perception of the functioning of their injured knee.[87,88] The KOOS captures information on five dimensions; pain, symptoms, activities of daily life function, sport and recreation function, and knee-related quality of life.[87] Each dimension provides a series of questions in which the patient is asked to rate their response on a 5 point likert scale.[87] Calculating a KOOS score consists of scoring the responses so they are transformed to a 100 point scale; with a lower score value indicating the patient has a perception of more severe knee problems and a higher score indicating the opposite.[87]

Three-Dimensional Motion Capture System

A three-dimensional electromagnetic motion tracking system (Ascension Technologies, Inc., Burlington, VT) was used to collect bilateral kinematic data. The system was integrated with the Motion Monitor Software (Innovative Sports Training, Inc., Chicago, IL) and consisted of a short-range electromagnetic transmitter and eight tethered receivers. All kinematic data were sampled at 144 Hz, consistent with the previous methods of the JUMP ACL study.

Force Plates

A non-conductive force plate was used (Bertec Corporation, Columbus, OH) to collect all kinetic data. To be consistent with the previous methods of the JUMP ACL study, all kinetic data were sampled at 1,440 Hz.

Knee Arthrometer

A KT-1000 knee arthrometer (MEDmetric Corp, San Diego, CA) was used to assess anterior knee laxity bilaterally. Measures were recorded at a 30lbs force level.

3.4 Procedures

Upon arrival for testing each participant read and signed an informed consent form from each of the respective academies. Participants also filled out a questionnaire regarding their physical activity and sport participation during their enrollment at the academies. The questionnaire also included the Marx Activity scale, and the KOOS.

(**Appendix A**) This information was included in data analysis, but instead used as a description of the participant samples.

Information regarding the leg that was injured and the dominant leg, determined by asking the participant which leg they would use to kick a ball for maximal distance, was recorded as well. Following completion, participants performed two testing procedures: double leg jump landing, and an assessment of anterior knee laxity. The order of testing was standardized, so that the double-leg jump landing was performed first and the assessment of anterior knee laxity assessed second. Measures were performed bilaterally, and the order of testing for leg was counter-balanced by alternating whether measures were performed first on the right or left leg, regardless of leg dominance. Anterior knee laxity measures were included as a description of the samples and were not included in any statistical analyses for the purposes of this study.

Double Leg Jump Landing

Electromagnetic receivers were fixed to the inner shank and outer thigh of both legs, and the sacrum using double sided tape. A prewrap dressing and athletic tape were applied over the sensors on the thigh and shank, and an elastic belt was placed over the sacral sensor to minimize sensor movement during testing to minimize motion artifact. After the receivers were placed, each participant was asked to stand within the range of the electromagnetic transmitter in a neutral position with their feet shoulder width apart and arms by their side. The points of the medial and lateral femoral epicondyle, medial and lateral malleoli, and the anterior superior iliac spine (ASIS) were palpated and digitized bilaterally using a calibrated stylus. The primary investigator performed all digitization and gave final approval of the digitized model for all participants.

After successful digitization, the procedures of the double leg jump landing were explained to each participant. Consistent with procedures performed in the JUMP ACL study, participants performed a double leg jump landing. Participants were required to stand atop a box with a height of 30cm, located a distance equal to half their body height from the edge of the force plate (**Figure 1**). They jumped forward from the box landing with their test leg completely on the force plate, and immediately jumped vertically for maximum height. A trial was considered successful if the participant landed with the foot of the test leg completely on the force plate, with the non-test leg making no contact with the force plate during the initial landing, and the participant not pausing in between the initial landing and vertical jump. The primary investigator determined if each trial was successfully performed for all participants, and was also responsible for determining if the data captured for each trial was accurate and free of error.

Each participant successfully completed five trials per leg. Because the equipment used in this study was limited to the use of one force plate, a total of ten successful trials were recorded to permit bilateral analyses.

Anterior Knee Laxity Assessment

The KT-1000, as described previously, was used to assess anterior knee laxity bilaterally. Participants were positioned in a supine position with an adjustable bolster placed underneath their distal thigh, just proximal to the knee joint line. The bolster was adjusted to achieve a knee flexion angle between 20° and 30°. A foot support was placed underneath their heels to ensure that legs stayed in a fixed position during testing. The KT-1000 device was aligned with the knee joint line and secured to the shank using two Velcro straps. A small compressible piece of foam padding was placed between the lower shank and the device to ensure comfort of the participant. The device was adjusted accordingly to ensure that resulting displacement of the tibia would be in the sagittal plane of the knee, the point of resistance was over the participant's patella, and the height of the pull arm was adjusted so that the arthrometer measured zero millimeters of displacement at rest. The participant was instructed to relax during the procedures. An anterior force was applied on the tibia, while a stabilizing force was placed on the patella. The force was increased until the audio tone indicating a force of 30lbs was being applied was heard. The anterior tibial displacement was recorded to the closest half-millimeter.

Three successful trials were completed on each leg. To be considered a successful trial, the 30lb tone had to be achieved, the displacement of the arthrometer had to be in the sagittal plane, and the dial of the arthrometer had to return to a position of 0±0.5 mm at rest to ensure the participant was relaxed during the assessment.[89] Additional trials were performed if the investigator subjectively felt that the participant was not relaxed during the trial. These procedures have a high intrarater reliability ($ICC_{(3,1)}$ = 0.969, SEM = 0.32 mm).

3.5 Data Capture, Processing and Reduction

Prior to each data collection session, a global axis system was established along with integration of the force plates according to the manufacturer's guidelines using the Motion Monitor software. The axes of the global axis system were defined using a right-hand convention, with the positive x-axis corresponding with the forward direction of the double leg jump movement, the positive y-axis defined by a vector located with a positive 90° rotation about the z-axis relative to the x-axis, and the positive z-axis defined by a vector located with a positive 90° rotation about the x-axis from the position of the y-axis.

Segments of the shank, thigh, and pelvis were defined within the data collection software. The shank segments were defined by the segment endpoints of the ankle joint center and knee joint center, and a third non-collinear point of the shank electromagnetic receiver. The thigh segments were defined by the segment endpoints of the knee joint cent and hip joint center, and a third non-collinear point of the thigh electromagnetic receiver. The thigh joint center was estimated within the software using the Bell method.[90] The pelvis segment was defined based on the of the right and left ASIS, and the third non-collinear point of the sacrum electromagnetic receiver.

Local coordinate systems were established based on a right-hand convention and coincided with the orientation of the global axis system such that, the positive x-axis corresponded with the anterior direction, the positive y-axis as the medial direction for the right leg and lateral direction for the left leg, and the positive z-axis as the superior direction. Cardan angles using an Euler sequence were used to calculate joint angles for the knee and hip using the Motion Monitor software. The Euler sequence for both knee and hip angle was defined by a first rotation about the y-axis, a second rotation about the x-axis, and a third rotation about the z-axis. The first rotation about the y-axis corresponded with sagittal plane motion of the knee (+ flexion / - extension) and hip (+ extension / - flexion). The second rotation about the x-axis corresponded with frontal plane motion of the knee (+ varus / - valgus) and hip (+ adduction / - abduction). The third rotation about the z-axis corresponded with transverse plane motion of the knee (+ internal rotation / - external rotation) and hip (+ internal rotation / - external rotation). The sign conventions to define the direction of anatomical motion for the first rotation about the y-axis were consistent for the left leg, but the inverse for the second and third rotations. This factor was corrected in the data reduction process to make all sign conventions indicative of the relative motions listed previously.

Moments for the knee and hip were calculated within the Motion Monitor software using a standard inverse dynamics approach. These moments were calculated as internal moments and are representative of the moment produced within the body to resist the external moments generated on the body by interaction with the environment. Moments in each plane of motion were calculated for the knee and hip, as well as the proximal anterior tibial shear force. The proximal anterior tibial shear force was defined as the resultant force acting on the shank segment calculated at the point of the knee.

Prior to exportation of the data, the data were filtered within the Motion Monitor software using a Butterworth filter with a cutoff frequency of 14.5 Hz. The data were filtered prior to exportation to ensure no introduction of a time shift in the data between the kinematic and kinetic data. All data were exported at the sampling rate of 1440 Hz, consistent with that of the sampling frequency of the kinetic data, and the highest sampling frequency of the data collected. This required up-sampling of the kinematic data that was performed during exportation by the Motion Monitor software.

Definition and Calculation of Dependent Variables

The vertical ground reaction force was used to define the stance phase for the double leg jump landing. The point at which the vertical ground reaction force first exceeded 10N was defined as Initial Ground Contact, and the subsequent point at which the vertical ground reaction force value fell below 10N was used to define toe-off. The time period between Initial Ground Contact and toe-off was defined as the stance phase of the double leg jump landing. The time from Initial Ground Contact until peak knee flexion was defined as the Landing Phase of the double leg jump landing. The time period of the 100 milliseconds preceding Initial Ground Contact was defined at the Preparatory Phase. Dependent variables were calculated relative to these time points of interest.

Kinematic Variables

The assessment of kinematic values was performed for the initial trials collected for the original JUMP ACL study (Baseline), and from the follow up data collection (Follow-Up) for all repeated measures analyses. Values for the kinematic variables were determined for the following phases of the double leg jump landing: Preparatory Phase, Initial Ground Contact, and Landing Phase. Kinematic values for all three planes of knee joint motion and hip joint motion were determined for the Preparatory Phase and Initial Ground Contact. These values were averaged across trials. Maximum and minimum values for all three planes of knee joint motion and hip joint motion were determined for the Landing Phase and averaged across trials.

Kinetic Variables

The assessment of kinetic values was performed for the initial trials collected at Baseline, and from the Follow-Up data collection for all repeated measures analyses. Values for the kinetic variables were determined for the following phases of the double leg jump landing: Initial Ground Contact, and Landing Phase. Values for the moments of the knee and hip for all three planes of motion, as well as anterior tibial shear force were recorded at the point of Initial Ground Contact and averaged across trials. The maximum and minimum values during the Landing Phase for the knee and hip, for all three planes of motion, as well as the maximum anterior tibial shear and vertical ground reaction force was determined, recorded, and averaged across trials.

All moments were reported as internal moments and normalized by the product of the participant's body weight (N) and body height (m). Vertical ground reaction force was normalized to the participant's body weight.

Joint Coordination and Variability

Joint coordination was determined using a vector coding method as described by Heiderscheit et al[81] and Ferber et al[83] as an alteration to the technique proposed by Sparrow et al.[79] Angle-angle plots were constructed for each joint coupling of interest during the Landing Phase of the double leg jump landing (**Figure 2**). The coupling angle for each point was calculated as:

$$\theta_i = \text{abs}\{\tan^{-1}\left[\frac{y_{i+1} - y_i}{x_{i+1} - x_i}\right]]$$

The resulting values were converted from radians to degrees, a mean value was calculated for each trial, and a between trials mean was calculated to represent a joint coordination value. Variability of joint coordination was calculated as the average between trial standard deviation.[82] The joint coordination pairs that were investigated included: Hip Sagittal Plane – Knee Sagittal Plane, Hip Frontal Plane – Knee Frontal Plane, Hip Transverse Plane – Knee Transverse Plane, Hip Frontal Plane – Knee Transverse Plane, and Hip Transverse Plane – Knee Frontal Plane.

All data reduction was conducted using customized MATLAB software programs (Mathworks, Natick, MA, v7.10).

3.6 Statistical Analysis

An a priori alpha level of 0.05 was set to determine statistical significance for all statistical analyses. All statistical analyses were performed using IBM SPSS Statistics (IBM Corp., Armonk, NY, v.19.0) software package. Prior to the completion of all statistical analyses, each dependent variable was assessed for statistical outliers and to ensure normal distribution.

Research Question 1: Are lower extremity biomechanics during a double leg jump landing at Baseline changed following ACL injury and subsequent ACLR?

Statistical Procedure Research Question 1a & 1b: A 3x2 (group, time) mixed model analysis of covariance (ANCOVA) to adjust for the influence of differences in proportions of gender between groups was performed for each dependent variable. Tukey's post hoc analysis was performed for all significant interactions and group main effects.

Research Question 2: Is lower extremity joint coordination during a double leg jump landing at Baseline changed following ACL injury and subsequent ACLR?

Statistical Procedure Research Question 2a & 2b: A 3x2 (group, time) mixed model analysis of covariance (ANCOVA) to adjust for the influence of differences in proportions of gender between groups was performed for each dependent variable. Tukey's post hoc analysis was performed for all significant interactions and group main effects.

Research Question 3: Is the magnitude of between limb differences in lower extremity biomechanics during a double leg jump landing at Follow-Up, different in persons with ACLR as compared to those with no history of ACL injury at Follow-Up?

Statistical Procedure Research Question 3: The absolute value of a difference score was calculated between the right and left leg for each dependent variable. Between group differences (ACLR v. Control) were determined using two-sample Kolmogorov-Smirnov tests.

Research Question 4: Is the magnitude of between limb differences in lower extremity joint coordination and variability during a double leg jump landing at Follow-Up different at in persons with ACLR as compared to those with no history of ACL injury at Follow-Up?

Statistical Procedure Research Question 4: The absolute value of a difference score was calculated between the right and left leg for each dependent variable. Between group differences (ACLR v. Control) were determined using two-sample Kolmogorov-Smirnov tests.

CHAPTER FOUR

SUMMARY OF RESULTS

4.1 Introduction

The purpose of this chapter is to provide a summary of the results for this project, and will be addressed according to each research question. A more detailed presentation of the results for Research Question 1, 2, and 3 is provided in manuscripts attached to this document as appendices. Manuscript 1 (**Appendix B**) will address Research Question 1, Manuscript 2 (**Appendix C**) will address Research Question 2, and Manuscript 3 (**Appendix D**) will address Research Question 3. This chapter will provide a more detailed presentation of results concerning Research Question 4 as well as information from Research Question 1 and 3 regarding kinematic analyses during the Preparatory Phase that were not included in the respective manuscripts. In addition, the discussion and interpretation for the results not addressed in a manuscript will be provided in the next chapter.

4.2 Overview of Study and Participant Demographics

Eighty-eight participants were enrolled in this study for further biomechanical testing at Follow-Up. Thirty-eight of the participants had suffered an ACL injury, undergone reconstructive surgery, and returned to physical activity during their enrollment in the JUMP ACL study. Of this group, 6 were identified as reporting a prior ACL injury at the time of Baseline testing and 3 had suffered multiple ACL injuries since Baseline. Because of a low number of participants that suffered injury to the leg on which biomechanical data at Baseline was captured, 2 participants that indicated a prior ACL injury to Baseline testing were retained in the analyses for Research Question 1 and 2 as biomechanical data was recorded for their noninjured limb at Baseline, and they later suffered injury to that same limb. Fifty participants were enrolled as healthy matched controls, of which, 5 did not complete biomechanical testing at Baseline, 5 did not complete biomechanical testing at Follow-Up, and 1 reported an ACL injury prior to Baseline testing.

Because of the nature of each research question, participant inclusion and demographics were not the same across all analyses. Detailed information regarding participant inclusion criteria and demographics are provided in each manuscript. Overall there were a total of 70 participants (31 ACLR, 39 Controls) available to address Research Questions 1 and 2, and 72 (28 ACLR, 44 Controls) available to address Research Questions 3 and 4.

4.3 Results

4.3.1 Results Research Question 1

The purpose of this analysis was to determine the effect of ACL injury and subsequent ACLR on lower extremity biomechanics. Kinematics at the Preparatory Phase, Initial Ground Contact, and Landing Phase were compared within groups across time and between groups. Kinetic variables were compared at Initial Ground Contact and Landing Phase. Group demographics and descriptions are provided in **Tables 4-6**, and summary of descriptive statistics and statistical outcomes in **Tables 7-17**. Ensemble plots for each dependent variable during the Landing Phase are presented for each group in **Figures 3-44**.

Preparatory Phase: Comparison of kinematics at Preparatory Phase are not included in any of the accompanying manuscripts, and will be presented in detail here. No significant interactions for time and group were observed for any kinematic variables during the Preparatory Phase: knee sagittal plane angle ($F_{(2,66)} = 0.933$, $p = 0.399$), knee frontal plane angle ($F_{(2,66)} = 0.179$, $p = 0.837$), knee transverse plane angle ($F_{(2,66)} = 0.027$, $p = 0.973$), hip sagittal plane angle ($F_{(2,66)} =1.989$, $p = 0.145$), hip frontal plane angle ($F_{(2,66)} = 2.122$, $p = 0.128$), and hip transverse plane angle ($F_{(2,66)} = 0.006$, $p = 0.994$). No significant group main effects were observed either at the Preparatory Phase: knee sagittal plane angle ($F_{(2,66)} = 0.284$, $p = 0.754$), knee frontal plane angle ($F_{(2,66)} = 0.300$, $p = 0.742$), knee transverse plane angle ($F_{(2,66)} = 0.594$, $p = 0.555$), hip sagittal plane angle ($F_{(2,66)} = 1.316$, $p = 0.275$), hip frontal plane angle ($F_{(2,66)} = 1.405$, $p = 0.253$), and hip transverse plane angle ($F_{(2,66)} = 0.145$, $p = 0.865$). A significant time main effect was observed for hip frontal plane angle ($F_{(1,66)} = 4.521$, $p = 0.037$) during the Preparatory Phase, indicating an increase in hip adduction over time regardless of group. No other significant time main effects were observed: knee sagittal plane angle ($F_{(1,66)} = 0.930$, $p = 0.338$), knee frontal plane angle ($F_{(1,66)} = 0.021$, $p = 0.884$), knee transverse plane angle ($F_{(1,66)} = 0.341$, $p = 0.561$), hip sagittal plane angle ($F_{(1,66)} = 0.051$, $p = 0.822$), and hip transverse plane angle ($F_{(1,66)} = 0.360$, $p = 0.550$).

A detailed description of statistical comparison of biomechanical variables at Initial Ground Contact and Landing Phase are presented in Manuscript one. A summary of the significant findings from these analyses will be presented here.

Initial Ground Contact: A significant interaction for time and group was observed for knee frontal plane ($F_{(2,66)} = 3.957$, $p = 0.024$) and hip frontal plane ($F_{(2,66)} = 3.773$, $p = 0.028$) angles at Initial Ground Contact. Following post hoc analysis, we observed no significant difference among groups at Baseline or Follow-Up for knee frontal plane angle. However, both the ACLR–Injured Limb and ACLR-Noninjured Limb groups had a significant increase in knee valgus angle at Initial Ground Contact compared to Baseline. A similar pattern of change was observed for the hip frontal plane angle, as there was no difference among groups at Baseline, but both the ACLR-Injured Limb and ACLR-Noninjured Limb groups significantly increased hip adduction at Initial Ground Contact from Baseline to Follow-Up. At Follow-Up, both ACLR groups displayed significantly greater hip adduction compared to the Control group. No significant interactions were observed for any other kinematic variable at Initial Ground Contact. No significant group main effects were observed either, though a significant time effect was present for transverse plane hip angle ($F_{(1,66)} = 4.731$, $p = 0.033$), with an increase in hip external rotation across time regardless of group.

A significant interaction for time and group was observed for transverse plane knee moment ($F_{(2,66)} = 3.373$, $p = 0.040$), sagittal plane hip moment ($F_{(2,66)} = 4.266$, $p = 0.018$), and transverse plane hip moment ($F_{(2,66)} = 3.226$, $p = 0.046$) at Initial Ground Contact. Post hoc analyses indicated no difference among groups across time for transverse plane knee moment. In addition, groups were not different at Baseline for sagittal or transverse plane hip moment. For sagittal plane hip moment, we observed an increase in hip extension moment at Initial Ground Contact over time for the Control group. At Follow-Up the ACLR-Injured Limb group demonstrated a hip flexion moment and the ACLR-

Noninjured Limb demonstrated a hip extension moment that was significantly between the two groups. For the transverse plane hip moment interaction, both the ACLR-Injured Limb and Control groups had a significant decrease for hip internal rotation moment across time. The associated changes resulted in both being significantly less than the ACLR-Noninjured Limb group at Follow-Up. No other significant interactions were observed for kinetic measures at Initial Ground Contact.

A significant group main effect was observed for frontal plane hip moment ($F_{(1,66)} = 3.178$, $p = 0.048$) with the ACLR-Noninjured Limb group having significantly higher internal hip adduction moment than the Control group regardless of time. Time main effects for frontal plane knee moment ($F_{(1,66)} = 16.802$, $p < 0.001$), frontal plane hip moment ($F_{(1,66)} = 11.684$, $p = 0.001$), and vertical ground reaction force ($F_{(1,66)} = 6.401$, $p = 0.014$) were observed, with each variable decreasing across time regardless of group. No other significant group or time main effects were observed.

Landing Phase: During the Landing Phase of the double leg jump landing we observed significant interactions for peak knee varus angle ($F_{(2, 66)} = 5.198$, $p = 0.008$), peak knee valgus angle ($F_{(2,66)} = 3.768$, $p = 0.028$), and peak knee internal rotation angle ($F_{(2,66)} = 4.204$, $p = 0.019$). There was no difference among groups for any of these variables at Baseline. Both the ACLR-Injured Limb and ACLR-Noninjured Limb groups had a significant decrease in peak knee varus angle from Baseline to Follow-Up, with no difference among groups at Follow-Up. The ACLR-Noninjured Limb group also demonstrated a significant increase in peak knee valgus angle over time. A similar increase was observed for the ACLR-Injured Limb group as well, but it was not significant, and there was no difference among groups at Follow-Up. In addition, the

Control group demonstrated a significant increase in peak knee internal rotation angle. This change resulted in greater peak knee internal rotation for the Control group as compared to the ACLR-Noninjured Limb group at Follow-Up. No other significant interactions for peak kinematic variables during the Landing Phase were observed.

Time main effects indicated an increase in peak knee flexion ($F_{(1,66)} = 25.168$, $p < 0.001$), increase in peak hip flexion ($F_{(1,66)} = 25.326$, $p < 0.001$), decrease in peak hip internal rotation ($F_{(1,66)} = 5.263$, $p = 0.025$), and increase in peak hip external rotation ($F_{(1,66)} = 3.986$, $p = 0.050$) from Baseline to Follow-Up regardless of group. No group main effects were observed.

For peak kinetic variables, significant interactions for peak knee extension moment ($F_{(2,66)} = 4.509$, $p = 0.015$), peak hip flexion moment ($F_{(2,66)} = 3.847$, $p = 0.026$) and peak anterior tibial shear force ($F_{(2,66)} = 4.530$, $p = 0.014$) were observed during the Landing Phase. These interaction effects were the result of changes for the ACLR-Injured Limb group only, as there was no difference among groups at Baseline for these variables, but the ACLR-Injured Limb group demonstrated a significant decrease in peak knee extension moment, peak hip flexion moment, and peak anterior tibial shear force. This change resulted in the ACLR-Injured Limb group demonstrating lower peak values for each variable as compared to the ACLR-Noninjured Limb group at Follow-Up. No other significant interactions were present for peak kinetic variables during the Landing Phase.

Group main effects included peak knee flexion moment ($F_{(1,66)} = 3.508$, $p = 0.036$) and peak knee valgus moment ($F_{(1,66)} = 3.501$, $p = 0.036$), but post hoc analysis did not indicate significant group differences. Time main effects for peak knee valgus moment ($F_{(1,66)} = 25.659$, $p < 0.001$), peak hip abduction moment ($F_{(1,66)} = 5.723$, $p = 0.020$), and peak hip external rotation moment ($F_{(1,66)} = 6.804$, $p = 0.011$) were observed, with an associated decrease in each variable from Baseline to Follow-Up regardless of group.

4.3.2 Results Research Question 2

The purpose of this analysis was to determine the effect of ACL injury and subsequent ACLR on lower extremity joint coordination. Joint coordination, quantified as the mean coupling angle, was calculated to characterize relative motion between the hip and knee in the sagittal plane, hip and knee in the frontal plane, hip and knee in the transverse plane, hip in the frontal plane and knee in the transverse plane, and hip in the transverse plane and knee in the frontal plane. Measures were analyzed during the Landing Phase, and compared within groups across time and between groups. Group demographics and descriptions are provided in **Tables 18-20**, and summary of descriptive statistics and statistical outcomes in **Tables 21 & 22**.

Our main finding for this analysis was a significant interaction for Hip Transverse Plane – Knee Transverse Plane ($F_{(2,65)} = 4.398$, $p = 0.016$) coupling angle. We observed no differences among the groups at Baseline. However, only the ACLR-Injured Limb group had a significant decrease in Hip Transverse Plane – Knee Transverse Plane coordination over time, indicating more equal knee rotation relative to hip rotation during the Landing Phase. At Follow-Up, the mean coupling angle for the ACLR-Injured Limb group was significantly less than the Control. We observed no other significant interactions for this analysis, though a time main effect for Hip Frontal Plane – Knee Transverse Plane ($F_{(1,65)} = 4.789$, $p = 0.032$) was observed. The relative change over time indicated a shift towards more hip frontal plane motion relative to knee transverse plane motion at Follow-Up as compared to Baseline. No other time main effects were observed, and no group main effects were observed.

4.3.3 Results Research Question 3

The purpose of this analysis was to quantify and compare between limb asymmetry in biomechanics during a double leg jump landing. Asymmetry was quantified as the absolute value of the difference between values for the right and left leg of each variable of interest. Measures were analyzed during Preparatory Phase, Initial Ground Contact, and Landing Phase. Group demographics and descriptions are provided in **Tables 23 & 24**, and summary of descriptive statistics and statistical outcomes in **Tables 25-27**.

Preparatory Phase: We observed no difference between groups in the amount of bilateral asymmetry for knee sagittal plane angle ($Z_{KS} = 0.815$, $p = 0.519$), knee frontal plane angle ($Z_{KS} = 0.630$, $p = 0.822$), or knee transverse plane angle ($Z_{KS} = 0.494$, $p = 0.968$). No difference in hip sagittal plane angle ($Z_{KS} = 0.593$, $p = 0.873$), hip frontal plane angle ($Z_{KS} = 0.704$, $p = 0.704$), or hip transverse plane angle ($Z_{KS} = 0.655$, $p = 0.785$) was observed either.

Initial Ground Contact: We observed no difference in asymmetry between groups for any kinematic or kinetic variables at Initial Ground Contact.

Landing Phase: No between group differences for asymmetry in peak hip and knee kinematics were observed during the Landing Phase. We did observe greater asymmetry in peak knee flexion moment for the ACLR group as compared to the Control group during the Landing Phase ($Z_{KS} = 1.42$, p = 0.035), as well as greater asymmetry in peak vertical ground reaction force ($Z_{KS} = 1.45$, p = 0.031). Follow up analysis indicated that increased asymmetries for the ACLR group were the result of greater knee flexion moment for the injured limb (0.058 ± 0.032 Nm/BWxBH) as compared to the noninjured limb (0.053 ± 0.031 Nm/BWxBH), but greater peak vertical ground reaction force for the noninjured limb (2.97 ± 0.069 N/BW) as compared to the injured limb (2.55 ± 0.80 N/BW). No other differences in asymmetry of the peak kinetic variables during the Landing Phase were observed.

4.3.4 Results Research Question 4

The purpose of this analysis was to quantify and compare between limb asymmetry in joint coordination and variability during a double leg jump landing. Demographics and anthropometrics for the two groups are provided in **Tables 28-29**. We were unable to collect a date of surgery for three members of the ACLR group, and graft information for 9. The ACLR group was on average 1.88 ± 0.66 years post-surgery. Eight participants had a bone-patella tendon-bone autograft, 10 had a hamstrings autograft, and one an Achilles tendon allograft. Asymmetry in joint coordination was quantified as the absolute difference in mean coupling angle between the right and left limb for each variable of interest. This method was also used to quantify asymmetry for the between trials standard deviation used to represent joint coordination variability. All measures were analyzed during the Landing Phase of the double leg jump landing. Descriptive statistics for each variable of interest are presented in **Tables 30 & 31**.

We did not observe any between group differences in asymmetry of joint coordination between the hip and knee: Hip Sagittal Plane – Knee Sagittal Plane (Z_{KS} = 0.778, p = 0.580), Hip Frontal Plane – Knee Frontal Plane (Z_{KS} = 0.482, p = 0.974), Hip Transverse Plane – Knee Transverse Plane (Z_{KS} = 0.497, p = 0.966), Hip Frontal

Plane – Knee Transverse Plane ($Z_{KS} = 0.435$, $p = 0.991$), and Hip Transverse Plane – Knee Frontal Plane ($Z_{KS} = 0.728$, $p = 0.664$). No between group difference was observed for asymmetry in joint coordination variability either: Hip Sagittal Plane – Knee Sagittal Plane ($Z_{KS} = 0.632$, $p = 0.819$), Hip Frontal Plane – Knee Frontal Plane ($Z_{KS} = 0.566$, $p = 0.905$), Hip Transverse Plane – Knee Transverse Plane ($Z_{KS} = 0.705$, $p = 0.703$), Hip Frontal Plane – Knee Transverse Plane ($Z_{KS} = 0.882$, $p = 0.417$), and Hip Transverse Plane – Knee Frontal Plane ($Z_{KS} = 0.724$, $p = 0.670$).

CHAPTER FIVE

DISCUSSION OF RESULTS

5.1 Introduction

Results not discussed in the manuscripts will be discussed in this chapter. This includes analysis of kinematics during the Preparatory Phase for Research Question 1, kinematic asymmetry during the Preparatory Phase for Research Question 3, and analysis of bilateral asymmetry in joint coordination and variability for Research Question 4.

5.2 Preparatory Phase Kinematics

Previous research has indicated that lower extremity biomechanics at Initial Ground Contact are predictive of an initial ACL injury for healthy individuals[22] and secondary ACL injury for those with ACLR.[91] There is also preliminary evidence that peak ACL strain may occur prior to contact with the ground when landing from a jump.[92] For these reasons, how an individual prepares to land from a jump may influence their risk for ACL injury. In addition, between limb differences in preparation for landing may be important, as asymmetry in sagittal plane knee loading at initial contact is predictive of a second ACL injury for those with ACLR. [91] We examined the effect of ACL injury and ACLR on kinematics during the Preparatory Phase, and analyzed between asymmetry in kinematics during the Preparatory Phase.

Our observations indicate that ACL injury and subsequent ACLR have no effect on kinematics prior to landing. The only significant change we observed was an increase in hip adduction over time for all groups. It is likely that this change was the result of a factor other than ACL injury, such as engaging in physical activity at the service academies. In addition, we observed no significant difference in asymmetry of bilateral kinematics prior to landing for those with ACLR as compared to healthy control participants. Therefore, it appears that ACL injury did not affect how our participants prepared to land during the double leg jump landing, and did not induce asymmetry in landing preparation.

Few studies have previously examined kinematics prior to Initial Ground Contact[93,94], and we are unaware of any studies that have been conducted using individuals with ACLR. Taylor et al[94] examined the preparation phase of a jump landing in their study to characterize ACL length and strain during a dynamic task. This analysis observed that ACL strain may be greatest at 55 miliseconds prior to Initial Ground Contact.[94] Their analysis was limited to this purpose only though and provide little help in reference to interpreting our results. Chappell et al[93] examined gender differences in preparatory lower extremity kinematics and electromyography during a stop-jump task. They observed females to have decreased knee and hip flexion as compared to males when preparing for the landing phase of the stop-jump.[93] Other differences included, decreased hip abduction, decreased hip external rotation, and increased knee internal rotation for females.[93] These kinematic differences were accompanied by a reported increased quadriceps electromyography amplitude for the female participants as well.[93]These findings are not directly applicable to our findings, but a gender comparison was performed as females are thought to represent an at risk population for ACL injury, much like those with ACLR.

The only study to examine any variables related to preparation for landing in those with ACLR was performed by Vairo et al[28] in which they examined differences in preparatory muscle activation for a sample of participants with ipsilateral semitendinosus and gracilis autografts. They examined the preparatory muscle activation at a time point of 128 miliseconds prior to Initial Ground Contact, very similar to our definition of the Preparatory Phase (100 miliseconds prior to Initial Ground Contact). In this analysis, they observed an increase in quadriceps and hamstrings co-contraction as compared to healthy matched controls prior to landing.[28] They also observed between limb differences for those with ACLR, in which they observed decreased gastrocnemius activation of the injured limb as compare to the noninjured limb.[28] The authors, unfortunately, did not examine kinematics prior to landing, as such information would have benefitted the interpretation of our current results. They did examine sagittal plane angles of the hip, knee, and ankle at the point of Initial Ground Contact and observed both between limb and between group differences for hip flexion at this time point.[28] They noted increased hip flexion for the injured limb as compared to the noninjured limb, and as compared to the healthy matched control.[28] This may not mean that preparatory kinematics were altered though, as we observed differences in frontal plane knee and hip kinematics at Initial Ground Contact with no difference in these measures at Preparatory Phase. Also, we observed no greater between limb asymmetry at both Preparatory Phase and Initial Ground Contact.

One limitation of the analysis of kinematics at Preparatory Phase for this study was the use of an electromagnetic tracking system to collect biomechanical data. Chappell et al[93] employed a video based system to collect kinematic data during their analysis of the Preparatory Phase for a stop-jump task. These systems generally have a larger capture volume than the system we utilized. As such, our ability to assess kinematics at 100 miliseconds prior to Initial Ground Contact may have been hampered as some participants may not have been fully in the capture volume of the transmitter for this system because of subject variability in height and resultant distance they were required to jump from for the double leg jump landing. This analysis, however, was not identified as a primary analysis for this study, and we felt that it was more important to maintain consistency between the Baseline and Follow-Up testing sessions to allow for proper comparison of biomechanics during landing than it was to capture better data for this analysis. Future research may address this limitation by altering the double leg jump landing task or assessing a time point closer to Initial Ground Contact.

5.3 Research Question 4

Using a vector coding technique to quantify joint coordination as the mean coupling angle between the two joints of interest, we observed no group differences for asymmetry during the Landing Phase of the double leg jump landing. We also observed no differences between groups for bilateral asymmetry in movement variability. These findings suggest that for our sample of participants with ACLR, the relative difference between movement coordination of the knee and hip in the injured and noninjured limb, is no different than the between limb difference for our sample of healthy matched control participants. These results were unexpected, as it is generally believed that orthopedic injury can induce alterations in movement coordination and variability, and our findings are in contrast with previous observations of between group and between limb differences for those with ACLR.[32,33,35,95]

For movement coordination, Kurz et al[32] observed altered coordination about the knee and ankle during gait for those with ACLR. The authors noted that the relative difference in movement coordination at the knee were not present during running, and only coordination at the ankle was significantly different than a group of healthy controls.[32] The authors do acknowledge that the lack of difference may have been the result of a relatively low sample size relative to the amount of variability present for the coordination measure during running.[32] It may though provide some evidence that differences in coordination may be somewhat dependent on the nature of the task. Van Uden et al[35] observed altered coordination of the knee and ankle in the sagittal plane during the more physically demanding task of single leg hopping. Their observation of altered coordination was accompanied by increased variability for the injured limb of those with ACLR as compared to healthy controls as well. Like our findings they observed no between limb difference in coordination for either the ACLR or healthy control group, but did note increased variability of the injured limb as compared to the noninjured limb for those with ACLR.[35]

Increased variability of for knee flexion-extension of the reconstructed limb of those with ACLR has also been previously reported by Moraiti et al[33] using measures of approximate entropy. They also, found no differences between autograft type, bone-patella tendon-bone or semitendinosus-gracilis.[33] In a follow-up analysis to describe variability of the noninjured limb, they observed even greater variability for the noninjured limb as compared to the injured in their sample of individuals with ACLR. It should be noted that the task the participants were required to complete was walking on a motorized treadmill, and it can be assumed required much less physical effort of the participant than the double leg jump landing we used.

It should also be noted that the authors used alternative methods to measure joint coordination than the ones we employed for this analysis. The above studies were conducted using measures of continuous relative phase[32,35] to quantify coordination, and other nonlinear measure to quantify variability[33,95]. Because these methods require a continuous series of data, we were unable to use any of these methods as each jump landing was separated into a separate trial and therefore did not represent continuous motion on the participants' part. The method we employed to quantify coordination and variability have been previously used though. Pollard et al[82] used this method to compare joint coordination variability between healthy male and female subjects during an unanticipated cutting maneuver to provide a better understanding for the higher incidence of noncontact ACL injuries observed for female athletes.[82] For this study they noted a decrease in variability of coordination between the hip and knee in all planes of motion for females.[82] Relative differences in joint coordination variability using this technique have been observed in a patient population as well, as Heiderscheit et al[81] observed differences in transverse plane coordination variability of the thigh and shank for participants with patella-femoral pain syndrome. Similar to our findings though, the authors found no between limb differences in variability for either group, though they did not compare asymmetry.

Because of the limitations associated with how we chose to conduct this analysis we cannot directly compare the values we observed for each group to their findings, as we only compared the absolute difference between limbs, and did not examine raw values. In a previous analysis we did observe a significant alteration in joint coordination of the hip and knee in the transverse plane for the injured leg following ACL injury. This indicates that joint coordination is altered following ACL injury, and in combination with the findings of this analysis, suggest that this may induce alterations in the noninjured limb to minimize between limb differences and maintain symmetry. This notion was proposed by Moraiti et al[95]when they observed increased variability of the noninjured limb of those with ACLR. Therefore, alterations in coordination and variability induced by ACL injury may be transferred to the noninjured limb to minimize asymmetry. Assuming that movement coordination and variability affect risk for reinjury in those with ACLR, this may help to explain why the increased incidence of reinjury is not isolated to the injured limb.[96] The limitations in our analysis do not allow us to detect or comment on any between group differences for these measures, and future studies should give consideration to this drawback.

Table 1. List of kinematic dependent variables

PREPARATORY PHASE	INITIAL GROUND CONTACT	LANDING PHASE
Knee Sagittal Plane Angle	Knee Sagittal Plane Angle	Knee Flexion
Knee Frontal Plane Angle	Knee Frontal Plane Angle	Knee Extension
Knee Transverse Plane Angle	Knee Transverse Plane Angle	Knee Varus
Hip Sagittal Plane Angle	Hip Sagittal Plane Angle	Knee Valgus
Hip Frontal Plane Angle	Hip Frontal Plane Angle	Knee Internal Rotation
Hip Transverse Plane Angle	Hip Transverse Plane Angle	Knee External Rotation
		Hip Extension
		Hip Flexion
		Hip Adduction
		Hip Abduction
		Hip Internal Rotation
		Hip External Rotation

Table 2. List of kinetic dependent variables

INITIAL GROUND CONTACT	LANDING PHASE
Knee Sagittal Plane Moment	Knee Flexion Moment
Knee Frontal Plane Moment	Knee Extension Moment
Knee Transverse Plane Moment	Knee Varus Moment
Hip Sagittal Plane Moment	Knee Valgus Moment
Hip Frontal Plane Moment	Knee IR Moment
Hip Transverse Plane Moment	Knee ER Moment
Anterior Tibial Shear Force	Hip Extension Moment
Vertical Ground Reaction Force	Hip Flexion Moment
	Hip Adduction Moment
	Hip Abduction Moment
	Hip IR Moment
	Hip ER Moment
	Max Anterior Tibial Shear Force
	Max Vertical Ground Reaction Force

Table 3. List of joint coordination and variability dependent variables

LANDING PHASE
[redacted]
Knee Sagittal Plane - Hip Sagittal Plane
Knee Frontal Plane - Hip Frontal Plane
Knee Transverse Plane - Hip Transverse Plane
Knee Frontal Plane - Hip Transverse Plane
Knee Transverse Plane - Hip Frontal Plane

Table 4. Participant demographics and anthropometrics for Research Question #1. Units of measure are Age (years), Height (cm), and Mass (kg). Values represent means ± standard deviation.

		Baseline			Follow-Up		
	n	Age	Height	Mass	Age	Height	Mass
ACLR-INJ	12 (8 m, 4 f)	18.64 ± 0.50	174.10 ± 7.31	72.64 ± 9.48	21.42 ± 0.79	174.29 ± 7.56	76.25 ± 9.95
ACLR-NINJ	19 (9 m, 10 f)	18.52 ± 0.58	170.06 ± 9.26	68.99 ± 10.93	21.47 ± 0.77	170.05 ± 9.13	72.87 ± 12.78
Control	39 (20 m, 19 f)	18.48 ± 0.46	172.56 ± 9.10	70.17 ± 12.96	20.98 ± 0.73	172.73 ± 8.99	73.11 ± 13.16

Table 5. Group chronological descriptive statistics for Research Question #1. (Mean ± SD, Days)

	Baseline to Follow-Up	Baseline to Injury	Injury to Surgery	Surgery to Follow-Up
ACLR-Injured Limb	1,074.42 ± 197.28	367.73 ± 156.06	33.70 ± 20.29	666.90 ± 209.24
ACLR-Noninjured Limb	1,247.68 ± 179.04	533.33 ± 267.97	40.39 ± 24.92	691.06 ± 243.01
Control	1,077.59 ± 180.34	-	-	-

Table 6. Descriptive statistics for Research Question #1. Anterior knee laxity assessed using KT-1000, Marx Activity score, and KOOS for each group. Values represent mean ± standard deviation, and units of measure other than anterior laxity (mm) are scale dependent.

			KOOS				
	Bilateral Difference Anterior Laxity (mm)	Marx Activity Total Score	Pain	Symptom	ADL	Sport/Rec	QOL
ACLR-INJ	1.9 ± 1.1	13.92 ± 2.75	84.75 ± 9.97	71.50 ± 13.16	93.00 ± 7.20	80.42 ± 16.16	65.63 ± 20.90
ALCR-NINJ	2.5 ± 1.8	11.21 ± 4.57	86.95 ± 11.62	73.42 ± 15.85	96.74 ± 5.05	78.42 ± 17.72	71.71 ± 22.57
Control	1.1 ± 0.8	11.05 ± 3.03	96.26 ± 5.06	92.54 ± 8.71	98.82 ± 2.09	93.97 ± 10.27	90.87 ± 14.04

*Laxity measures were not obtained for 3 members of the Control group, 1 member of the ACLR-Noninjured Limb group

Table 7. Descriptive statistics for knee and hip kinematics (°) at Initial Ground Contact for Research Question #1. Baseline (PRE) and Follow-Up (POST)

		Control		ACLR-Noninjured Limb		ACLR-Injured Limb	
		Mean ± SD	95TH CI	Mean ± SD	95TH CI	Mean ± SD	95TH CI
Knee Sag	PRE	19.81 ± 5.69	(17.99, 21.63)	17.90 ± 5.69	(15.29, 20.51)	18.82 ± 5.72	(15.52, 22.12)
	POST	18.02 ± 7.21	(15.72, 20.33)	16.69 ± 7.22	(13.38, 19.99)	16.28 ± 7.26	(12.10, 20.46)
Knee Frt	PRE	0.96 ± 6.50	(-1.12, 3.04)	1.33 ± 6.51	(-1.65, 4.31)	2.61 ± 6.54	(-1.16, 6.38)
	POST	-0.49 ± 5.75	(-2.33, 1.35)	-4.84 ± 5.76	(-7.48, -2.20)	-4.22 ± 5.78	(-7.55, -0.89)
Knee Trv	PRE	-2.56 ± 7.88	(-5.08, -0.04)	-3.43 ± 7.89	(-7.05, 0.18)	-0.80 ± 7.93	(-5.37, 3.77)
	POST	-2.74 ± 7.30	(-5.07, -0.41)	-4.75 ± 7.31	(-8.10, -1.40)	-2.23 ± 7.35	(-6.46, 2.00)
Hip Sag	PRE	-33.16 ± 9.01	(-36.04, -30.28)	-26.33 ± 9.02	(-30.47, -22.20)	-31.42 ± 9.07	(-36.64, -26.19)
	POST	-29.95 ± 10.49	(-33.30, -26.60)	-29.67 ± 10.51	(-34.48, -24.85)	-30.07 ± 10.56	(-36.16, -23.99)
Hip Frt	PRE	-9.95 ± 6.60	(-12.06, -7.84)	-9.00 ± 6.61	(-12.03, -5.98)	-11.12 ± 6.64	(-14.95, -7.30)
	POST	-6.81 ± 9.59	(-9.87, -3.74)	0.41 ± 9.61	(-3.99, 4.81)	0.02 ± 9.65	(-5.54, 5.59)
Hip Trv	PRE	-1.85 ± 6.28	(-3.86, 0.16)	-1.99 ± 6.29	(-4.87, 0.89)	-2.12 ± 6.32	(-5.77, 1.52)
	POST	-3.17 ± 6.40	(-5.22, -1.13)	-4.13 ± 6.41	(-7.06, -1.19)	-4.18 ± 6.44	(-7.89, -0.47)

*Values for descriptive statistics are based on Gender entered as a covariate in the statistical model at a value of 0.53

Table 8. Descriptive statistics for knee and hip moments (Nm/BHxBW) and kinetics (N/BW) at Initial Ground Contact for Research Question #1. Baseline (PRE) and Follow-Up (POST)

		CONTROL		ACLR-Noninjured Limb		ACLR-Injured Limb	
		Mean ± SD	95TH CI	Mean ± SD	95TH CI	Mean ± SD	95TH CI
Knee Sag	PRE	0.00 ± 0.02	(-0.01, 0.01)	0.01 ± 0.02	(0.00, 0.03)	0.01 ± 0.02	(-0.00, 0.02)
	POST	0.03 ± 0.03	(0.02, 0.04)	0.04 ± 0.03	(0.02, 0.05)	0.01 ± 0.03	(-0.01, 0.03)
Knee Frt	PRE	0.02 ± 0.02	(0.01, 0.03)	0.03 ± 0.02	(0.02, 0.03)	0.02 ± 0.02	(0.01, 0.03)
	POST	0.01 ± 0.02	(0.00, 0.01)	0.01 ± 0.02	(0.01, 0.02)	0.00 ± 0.02	(-0.01, 0.01)
Knee Trv	PRE	-0.01 ± 0.01	(-0.01, -0.00)	-0.01 ± 0.01	(-0.01, -0.00)	-0.01 ± 0.01	(-0.01, -0.00)
	POST	-0.00 ± 0.01	(-0.01, 0.00)	-0.01 ± 0.01	(-0.01, -0.00)	-0.00 ± 0.01	(-0.01, 0.00)
Hip Sag	PRE	-0.02 ± 0.07	(-0.04, 0.00)	0.03 ± 0.07	(0.00, 0.06)	0.03 ± 0.07	(-0.01, 0.07)
	POST	0.04 ± 0.08	(0.02, 0.07)	0.07 ± 0.08	(0.04, 0.11)	-0.00 ± 0.08	(-0.05, 0.05)
Hip Frt	PRE	0.03 ± 0.05	(0.02, 0.05)	0.06 ± 0.05	(0.03, 0.080)	0.05 ± 0.05	(0.02, 0.08)
	POST	0.01 ± 0.05	(-0.00, 0.03)	0.04 ± 0.05	(0.02, 0.06)	0.02 ± 0.05	(-0.01, 0.04)
Hip Trv	PRE	0.02 ± 0.02	(0.01, 0.03)	0.02 ± 0.02	(0.01, 0.03)	0.02 ± 0.02	(0.01, 0.03)
	POST	0.00 ± 0.02	(-0.00, 0.01)	0.02 ± 0.02	(0.02, 0.03)	0.01 ± 0.02	(-0.00, 0.02)
ATSF	PRE	-0.03 ± 0.17	(-0.08, 0.02)	-0.12 ± 0.17	(-0.20, -0.05)	-0.10 ± 0.17	(-0.20, -0.01)
	POST	-0.16 ± 0.22	(-0.23, -0.09)	-0.24 ± 0.22	(-0.34, -0.14)	-0.06 ± 0.22	(-0.18, 0.07)
VGRF	PRE	0.12 ± 0.05	(0.10, 0.13)	0.11 ± 0.05	(0.09, 0.13)	0.12 ± 0.05	(0.09, 0.150)
	POST	0.09 ± 0.03	(0.08, 0.10)	0.08 ± 0.03	(0.06, 0.09)	0.09 ± 0.03	(0.08, 0.11)

*Values for descriptive statistics are based on Gender entered as a covariate in the statistical model at a value of 0.53

Table 9. Descriptive statistics for peak knee and hip kinematics (°) at Landing Phase for Research Question #1. Baseline (PRE) and Follow-Up (POST)

		CONTROL		ACLR-Noninjured Limb		ACLR-Injured Limb	
		Mean ± SD	95TH CI	Mean ± SD	95TH CI	Mean ± SD	95TH CI
Knee FLX	PRE	84.26 ± 13.49	(79.95, 88.58)	78.51 ± 13.51	(72.32, 84.69)	79.65 ± 13.58	(71.82, 87.47)
	POST	92.58 ± 16.23	(87.39, 97.77)	90.86 ± 16.25	(83.41, 98.30)	85.55 ± 16.33	(76.14, 94.96)
Knee EXT	PRE	19.82 ± 5.66	(18.01, 21.63)	16.84 ± 5.67	(14.25, 19.43)	18.64 ± 5.69	(15.36, 21.92)
	POST	18.16 ± 7.20	(15.85, 20.46)	16.69 ± 7.21	(13.39, 20.00)	16.24 ± 7.25	(12.06, 20.42)
Knee VRS	PRE	6.69 ± 7.43	(4.32, 9.07)	9.27 ± 7.44	(5.86, 12.67)	7.48 ± 7.48	(3.17, 11.79)
	POST	6.21 ± 7.10	(3.94, 8.48)	0.55 ± 7.11	(-2.71, 3.81)	1.28 ± 7.14	(-2.84, 5.40)
Knee VLG	PRE	-7.90 ± 8.18	(-10.52, -5.28)	-5.26 ± 8.19	(-9.01, -1.51)	-6.57 ± 8.23	(-11.32, -1.83)
	POST	-6.97 ± 7.73	(-9.44, -4.50)	-11.61 ± 7.74	(-15.16, -8.07)	-11.70 ± 7.78	(-16.18, -7.22)
Knee IR	PRE	9.18 ± 8.66	(6.41, 11.95)	9.29 ± 8.67	(5.32, 13.26)	12.33 ± 8.72	(7.31, 17.35)
	POST	17.10 ± 8.23	(14.47, 19.73)	9.30 ± 8.24	(5.53, 13.07)	13.24 ± 8.28	(8.47, 18.01)
Knee ER	PRE	-7.31 ± 8.84	(-10.14, -4.48)	-9.28 ± 8.85	(-13.34, -5.23)	-4.95 ± 8.90	(-10.08, 0.17)
	POST	-4.21 ± 6.45	(-6.27, -2.15)	-8.24 ± 6.45	(-11.19, -5.28)	-4.61 ± 6.49	(-8.35, -0.87)
Hip EXT	PRE	-33.12 ± 9.02	(-36.00, -30.23)	-26.09 ± 9.03	(-30.22, -21.95)	-31.10 ± 9.07	(-36.32, -25.87)
	POST	-29.41 ± 10.42	(-32.74, -26.08)	-29.61 ± 10.43	(-34.39, -24.83)	-30.00 ± 10.48	(-36.04, -23.96)
Hip FLX	PRE	-72.63 ± 18.71	(-78.61, -66.64)	-63.46 ± 18.74	(-72.04, -54.88)	-67.15 ± 18.83	(-78.00, -56.30)
	POST	-79.14 ± 21.23	(-85.93, -72.36)	-76.86 ± 21.26	(-86.60, -67.12)	-73.80 ± 21.37	(-86.12, -61.48)
Hip ADD	PRE	1.49 ± 8.58	(-1.26, 4.23)	0.88 ± 8.59	(-3.06, 4.81)	0.49 ± 8.63	(-4.49, 5.47)
	POST	-1.72 ± 8.14	(-4.32, 0.89)	4.18 ± 8.15	(0.44, 7.91)	4.24 ± 8.19	(-0.49, 8.96)
Hip ABD	PRE	-13.11 ± 8.18	(-15.72, -10.49)	-12.06 ± 8.19	(-15.81, -8.30)	-14.10 ± 8.23	(-18.85, -9.36)
	POST	-15.32 ± 8.74	(-18.11, -12.52)	-9.96 ± 8.76	(-13.97, -5.95)	-8.90 ± 8.80	(-13.97, -3.83)
Hip IR	PRE	6.89 ± 6.34	(4.87, 8.92)	8.86 ± 6.35	(5.95, 11.77)	6.31 ± 6.38	(2.64, 9.99)
	POST	7.19 ± 9.14	(4.27, 10.12)	2.93 ± 9.15	(-1.27, 7.12)	3.89 ± 9.20	(-1.41, 9.19)
Hip ER	PRE	-6.01 ± 6.37	(-8.05, -3.98)	-5.05 ± 6.38	(-7.97, -2.12)	-6.84 ± 6.41	(-10.53, -3.14)
	POST	-6.50 ± 7.19	(-8.80, -4.20)	-10.69 ± 7.20	(-13.99, -7.40)	-7.13 ± 7.24	(-11.30, -2.96)

*Values for descriptive statistics are based on Gender entered as a covariate in the statistical model at a value of 0.53

Table 10. Descriptive statistics for knee moments (Nm/BHxBW) and kinetics (N/BW) at Landing Phase for Research Question #1. Baseline (PRE) and Follow-Up (POST)

		CONTROL		ACLR-Noninjured Limb		ACLR-Injured Limb	
		Mean ± SD	95TH CI	Mean ± SD	95TH CI	Mean ± SD	95TH CI
Knee EXT	PRE	-0.206 ± 0.055	(-0.224, -0.189)	-0.219 ± 0.055	(-0.245, -0.194	-0.234 ± 0.056	(-0.266, -0.202)
	POST	-0.204 ± 0.043	(-0.217, -0.190)	-0.214 ± 0.043	(-0.234, -0.194)	-0.169 ± 0.044	(-0.194, -0.144)
Knee FLX	PRE	0.021 ± 0.043	(0.008, 0.035)	0.057 ± 0.043	(0.037, 0.077)	0.038 ± 0.044	(0.013, 0.063)
	POST	0.046 ± 0.034	(0.035, 0.057)	0.053 ± 0.034	(0.037, 0.069)	0.050 ± 0.034	(0.030, 0.070)
Knee VLG	PRE	-0.064 ± 0.037	(-0.076, -0.052)	-0.087 ± 0.037	(-0.104, -0.070)	-0.074 ± 0.038	(-0.095, -0.052)
	POST	-0.045 ± 0.027	(-0.053, -0.036)	-0.055 ± 0.027	(-0.067, -0.042)	-0.044 ± 0.027	(-0.060, -0.029)
Knee VRS	PRE	0.061 ± 0.026	(0.053, 0.069)	0.057 ± 0.026	(0.045, 0.069)	0.064 ± 0.026	(0.049, 0.079)
	POST	0.063 ± 0.027	(0.055, 0.072)	0.067 ± 0.027	(0.054, 0.079)	0.055 ± 0.027	(0.039, 0.070)
Knee ER	PRE	-0.046 ± 0.023	(-0.053, -0.038)	-0.040 ± 0.023	(-0.050, -0.029)	-0.044 ± 0.024	(-0.058, -0.030)
	POST	-0.056 ± 0.019	(-0.063, -0.050)	-0.049 ± 0.019	(-0.058, -0.040)	-0.037 ± 0.020	(-0.049, -0.026)
Knee IR	PRE	0.043 ± 0.022	(0.036, 0.050)	0.045 ± 0.022	(0.035, 0.055)	0.049 ± 0.022	(0.037, 0.062)
	POST	0.031 ± 0.021	(0.024, 0.038)	0.044 ± 0.021	(0.034, 0.053)	0.037 ± 0.021	(0.025, 0.049)
ATSF	PRE	1.070 ± 0.319	(0.968, 1.172)	1.192 ± 0.319	(1.046, 1.339)	1.247 ± 0.321	(1.062, 1.432)
	POST	1.134 ± 0.244	(1.056, 1.212)	1.185 ± 0.245	(1.073, 1.297)	0.923 ± 0.246	(0.781, 1.065)
VGRF	PRE	3.094 ± 1.016	(2.769, 3.418)	3.254 ± 1.017	(2.788, 3.720)	3.188 ± 1.022	(2.599, 3.777)
	POST	2.726 ± 0.761	(2.482, 2.969)	3.084 ± 0.762	(2.735, 3.433)	3.114 ± 0.765	(2.673, 3.555)

*Values for descriptive statistics are based on Gender entered as a covariate in the statistical model at a value of 0.53

Table 11. Descriptive statistics for hip moments (Nm/BHxBW) at Landing Phase for Research Question #1. Baseline (PRE) and Follow-Up (POST)

		CONTROL		ACLR-Noninjured Limb		ACLR-Injured Limb	
		Mean ± SD	95TH CI	Mean ± SD	95TH CI	Mean ± SD	95TH CI
Hip FLX	PRE	-0.214 ± 0.103	(-0.247, -0.181)	-0.254 ± 0.103	(-0.301, -0.206)	-0.256 ± 0.104	(-0.316, -0.196)
	POST	-0.205 ± 0.067	(-0.227, -0.184)	-0.216 ± 0.067	(-0.247, -0.185)	-0.138 ± 0.068	(-0.177, -0.099)
Hip EXT	PRE	0.251 ± 0.127	(0.210, 0.292)	0.255 ± 0.128	(0.197, 0.314)	0.229 ± 0.128	(0.155, 0.303)
	POST	0.202 ± 0.091	(0.172, 0.231)	0.193 ± 0.091	(0.151, 0.235)	0.240 ± 0.092	(0.187, 0.293)
Hip ABD	PRE	-0.131 ± 0.065	(-0.152, -0.111)	-0.153 ± 0.065	(-0.183, -0.123)	-0.171 ± 0.065	(-0.209, -0.134)
	POST	-0.108 ± 0.068	(-0.130, -0.086)	-0.141 ± 0.068	(-0.173, -0.110)	-0.102 ± 0.068	(-0.142, -0.063)
Hip ADD	PRE	0.137 ± 0.067	(0.115, 0.158)	0.149 ± 0.068	(0.118, 0.180)	0.145 ± 0.068	(0.106, 0.184)
	POST	0.140 ± 0.054	(0.123, 0.157)	0.124 ± 0.054	(0.099, 0.149)	0.091 ± 0.054	(0.060, 0.122)
Hip ER	PRE	-0.085 ± 0.052	(-0.101, -0.068)	-0.095 ± 0.052	(-0.119, -0.071)	-0.099 ± 0.053	(-0.129, -0.069)
	POST	-0.052 ± 0.033	(-0.063, -0.042)	-0.078 ± 0.033	(-0.093, -0.063)	-0.055 ± 0.033	(-0.074, -0.036)
Hip IR	PRE	0.076 ± 0.029	(0.067, 0.086)	0.068 ± 0.029	(0.054, 0.081)	0.069 ± 0.029	(0.052, 0.086)
	POST	0.060 ± 0.031	(0.050, 0.069)	0.068 ± 0.031	(0.054, 0.082)	0.048 ± 0.031	(0.030, 0.065)

*Values for descriptive statistics are based on Gender entered as a covariate in the statistical model at a value of 0.53

Table 12. Observed power and effect size (η_p^2) for analyses of knee and hip kinematics an Initial Ground Contact for Research Question #1.

Variable	Observed Power	Effect Size (η_p^2)
Knee Sagittal Plane Angle		
Time	0.051	0.000
Group	0.173	0.022
Time x Group	0.065	0.003
Knee Frontal Plane Angle		
Time	0.959	0.176
Group	0.233	0.032
Time x Group	0.691	0.107
Knee Transverse Plane Angle		
Time	0.062	0.002
Group	0.168	0.022
Time x Group	0.070	0.004
Hip Sagittal Plane Angle		
Time	0.220	0.021
Group	0.274	0.038
Time x Group	0.403	0.058
Hip Frontal Plane Angle		
Time	0.969	0.186
Group	0.560	0.083
Time x Group	0.669	0.103
Hip Transverse Plane Angle		
Time	0.573	0.067
Group	0.067	0.004
Time x Group	0.063	0.003

Table 13. Observed power and effect size (η_p^2) for analyses of knee and hip moments and kinetics at Initial Ground Contact for Research Question #1.

Variable	Observed Power	Effect Size (η_p^2)
Knee Sagittal Plane Moment		
Time	0.297	0.031
Group	0.580	0.086
Time x Group	0.478	0.069
Knee Frontal Plane Moment		
Time	0.981	0.203
Group	0.309	0.041
Time x Group	0.125	0.014
Knee Transverse Plane Moment		
Time	0.426	0.042
Group	0.127	0.015
Time x Group	0.617	0.093
Hip Sagittal Plane Moment		
Time	0.069	0.003
Group	0.597	0.089
Time x Group	0.726	0.114
Hip Frontal Plane Moment		
Time	0.920	0.150
Group	0.590	0.088
Time x Group	0.130	0.015
Hip Transverse Plane Moment		
Time	0.352	0.038
Group	0.683	0.105
Time x Group	0.596	0.089
Vertical Ground Reaction Force		
Time	0.704	0.089
Group	0.151	0.019
Time x Group	0.086	0.007
Anterior Tibial Shear Force		
Time	0.053	0.000
Group	0.551	0.081
Time x Group	0.401	0.057

Table 14. Observed power and effect size (η_p^2) for analyses of knee kinematics during Landing Phase for Research Question #1.

Variable	Observed Power	Effect Size (η_p^2)
Knee Flexion		
Time	0.999	0.276
Group	0.234	0.032
Time x Group	0.207	0.028
Knee Extension		
Time	0.061	0.001
Group	0.278	0.039
Time x Group	0.104	0.010
Knee Varus		
Time	0.940	0.162
Group	0.203	0.027
Time x Group	0.813	0.136
Knee Valgus		
Time	0.651	0.079
Group	0.117	0.013
Time x Group	0.669	0.102
Knee Internal Rotation		
Time	0.248	0.025
Group	0.460	0.066
Time x Group	0.719	0.113
Knee External Rotation		
Time	0.136	0.011
Group	0.484	0.070
Time x Group	0.114	0.012

Table 15. Observed power and effect size (η_p^2) for analyses of hip kinematics during Landing Phase for Research Question #1.

Variable	Observed Power	Effect Size (η_p^2)
Hip Extension		
Time	0.232	0.023
Group	0.258	0.036
Time x Group	0.483	0.070
Hip Flexion		
Time	0.999	0.277
Group	0.184	0.024
Time x Group	0.252	0.035
Hip Adduction		
Time	0.075	0.003
Group	0.336	0.048
Time x Group	0.524	0.077
Hip Abduction		
Time	0.077	0.004
Group	0.407	0.058
Time x Group	0.444	0.064
Hip Internal Rotation		
Time	0.618	0.074
Group	0.137	0.016
Time x Group	0.497	0.072
Hip External Rotation		
Time	0.503	0.057
Group	0.158	0.020
Time x Group	0.435	0.063

Table 16. Observed power and effect size (η_p^2) for analyses of knee moments and kinetics during Landing Phase for Research Question #1.

Variable	Observed Power	Effect Size (η_p^2)
Knee Flexion Moment		
Time	0.319	0.033
Group	0.635	0.096
Time x Group	0.400	0.057
Knee Extension Moment		
Time	0.720	0.092
Group	0.178	0.023
Time x Group	0.751	0.120
Knee Varus Moment		
Time	0.208	0.020
Group	0.062	0.002
Time x Group	0.264	0.037
Knee Valgus Moment		
Time	0.999	0.280
Group	0.634	0.096
Time x Group	0.145	0.018
Knee Internal Rotation Moment		
Time	0.326	0.034
Group	0.383	0.055
Time x Group	0.176	0.023
Knee External Rotation Moment		
Time	0.238	0.024
Group	0.484	0.070
Time x Group	0.333	0.047
Anterior Tibial Shear Force		
Time	0.406	0.044
Group	0.280	0.039
Time x Group	0.753	0.121
Vertical Ground Reaction Force		
Time	0.289	0.030
Group	0.202	0.027
Time x Group	0.151	0.019

Table 17. Observed power and effect size (η_p^2) for analyses of hip moments during Landing Phase for Research Question #1.

Variable	Observed Power	Effect Size (η_p^2)
Hip Flexion Moment		
Time	0.609	0.072
Group	0.311	0.044
Time x Group	0.678	0.104
Hip Extension Moment		
Time	0.057	0.001
Group	0.059	0.002
Time x Group	0.231	0.032
Hip Abduction Moment		
Time	0.654	0.080
Group	0.487	0.071
Time x Group	0.285	0.040
Hip Adduction Moment		
Time	0.149	0.013
Group	0.232	0.032
Time x Group	0.433	0.062
Hip External Rotation Moment		
Time	0.729	0.093
Group	0.454	0.066
Time x Group	0.163	0.021
Hip Internal Rotation Moment		
Time	0.239	0.024
Group	0.205	0.028
Time x Group	0.314	0.044

Table 18. Participant demographics and anthroopometrics for Research Question #2. Units of measure are Age (years), Height (cm), and Mass (kg). Values represent means ± standard deviation.

		Baseline			Follow-Up		
	n	Age	Height	Mass	Age	Height	Mass
ACLR-INJ	12 (8 m, 4 f)	18.64 ± 0.50	174.10 ± 7.31	72.64 ± 9.48	21.42 ± 0.79	174.29 ± 7.56	76.25 ± 9.95
ACLR-NINJ	19 (9 m, 10 f)	18.52 ± 0.58	170.06 ± 9.26	68.99 ± 10.93	21.47 ± 0.77	170.05 ± 9.13	72.87 ± 12.78
Control	38 (19 m, 19 f)	18.47 ± 0.46	172.05 ± 8.65	69.16 ± 11.47	20.95 ± 0.73	172.16 ± 8.71	72.35 ± 12.37

Table 19. Group chronological descriptive statistics for Research Question #2. (Mean ± SD, Days)

	Baseline to Follow-Up	Baseline to Injury	Injury to Surgery	Surgery to Follow-Up
ACLR-Injured Limb	1,074.42 ± 197.28	367.73 ± 156.06	33.70 ± 20.29	666.90 ± 209.24
ACLR-Noninjured Limb	1,247.68 ± 179.04	533.33 ± 267.97	40.39 ± 24.92	691.06 ± 243.01
Control	1,071.76 ± 179.00	-	-	-

Table 20. Descriptive statistics for Research Question #2. Bilateral difference of anterior knee laxity assessed using KT-1000, Marx Activity score, and KOOS for each group. Values represent mean ± standard deviation, and units of measure other than anterior laxity (mm) are scale dependent.

			KOOS				
	Bilateral Difference Anterior Laxity (mm)	Marx Activity Total Score	Pain	Symptom	ADL	Sport/Rec	QOL
ACLR-INJ	1.9 ± 1.1	13.92 ± 2.75	84.75 ± 9.97	71.50 ± 13.16	93.00 ± 7.20	80.42 ± 16.16	65.63 ± 20.90
ALCR-NINJ	2.5 ± 1.8	11.21 ± 4.57	86.95 ± 11.62	73.42 ± 15.85	96.74 ± 5.05	78.42 ± 17.72	71.71 ± 22.57
Control	1.1 ± 0.8	10.92 ± 2.95	96.24 ± 5.12	92.63 ± 8.81	98.79 ± 2.11	93.95 ± 10.41	91.12 ± 14.14

*Laxity measures were not obtained for 1 member of the ACLR-Noninjured Limb group

Table 21. Average coupling angle for Research Question #2. (Mean, SD, 95th Confidence Interval) Control, ACLR-Noninjured Limb Group, and ACLR-Injured Limb Group at Baseline (Pre) and Follow-Up (Post). *P* values represent time main effects, group main effects, and time x group interactions.

		Control		ACLR-Noninjured Limb		ACLR-Injured Limb	
		Mean ± SD	95TH CI	Mean ± SD	95TH CI	Mean ± SD	95TH CI
Hip & Knee Sagittal Plane	PRE	53.76 ± 6.75	(51.57, 55.95)	54.76 ± 6.75	(51.67, 57.85)	53.99 ± 6.79	(50.08, 57.91)
	POST	52.48 ± 8.90	(49.60, 55.36)	51.64 ± 8.90	(47.56, 55.71)	55.32 ± 8.96	(50.16, 60.49)
Hip & Knee Frontal Plane	PRE	43.23 ± 7.29	(40.86, 45.59)	43.42 ± 7.30	(40.07, 46.76)	44.31 ± 7.34	(40.07, 48.54)
	POST	42.26 ± 7.42	(39.86, 44.67)	41.29 ± 7.43	(37.89, 44.70)	44.52 ± 7.48	(40.21, 48.83)
Hip & Knee Transverse Plane	PRE	52.76 ± 6.91	(50.52, 55.00)	54.98 ± 6.91	(51.81, 58.14)	54.85 ± 6.96	(50.84, 58.86)
	POST	54.78 ± 6.59	(52.64, 56.92)	53.46 ± 6.60	(50.44, 56.48)	47.97 ± 6.64	(44.15, 51.80)
Hip Frontal & Knee Transverse Plane	PRE	49.15 ± 7.67	(46.67, 51.64)	52.58 ± 7.67	(49.06, 56.09)	51.66 ± 7.72	(47.21, 56.11)
	POST	49.63 ± 6.04	(47.67, 51.58)	49.05 ± 6.04	(46.29, 51.82)	45.42 ± 6.07	(41.92, 48.92)
Hip Transverse & Knee Frontal Plane	PRE	47.51 ± 6.57	(45.38, 49.64)	46.45 ± 6.58	(43.44, 49.47)	48.35 ± 6.62	(44.53, 52.16)
	POST	47.49 ± 8.05	(44.88, 50.10)	45.99 ± 8.06	(42.29, 49.68)	47.71 ± 8.11	(43.04, 52.38)

*Values for descriptive statistics are based on Gender entered as a covariate in the statistical model at a value of 0.52

Table 22. Observed power and effect size (η_p^2) for analyses of mean coupling angle during Landing Phase for Research Question #2.

Variable	Observed Power	Effect Size (η_p^2)
Hip & Knee Sagittal Plane		
Time	0.366	0.040
Group	0.091	0.008
Time x Group	0.190	0.025
Hip & Knee Frontal Plane		
Time	0.055	0.001
Group	0.128	0.015
Time x Group	0.087	0.008
Hip & Knee Transverse Plane		
Time	0.514	0.059
Group	0.272	0.038
Time x Group	0.740	0.119
Hip Frontal & Knee Transverse Plane		
Time	0.578	0.069
Group	0.191	0.026
Time x Group	0.513	0.076
Hip Transverse & Knee Frontal Plane		
Time	0.050	0.000
Group	0.125	0.015
Time x Group	0.054	0.001

Table 23. Participant demographics and anthropometrics for Research Question #3. Units of measure are Age (years), Height (cm), and Mass (kg). Values represent means ± standard deviation.

	n	Age	Height	Mass
ACLR	24 (14 m, 10 f)	21.58 ± 0.78	172.01 ± 8.85	74.69 ± 12.50
Control	39 (20 m, 19 f)	21.00 ± 0.77	172.25 ± 8.94	72.27 ± 13.72

Table 24. Descriptive statistics for Research Question #3. (Mean ± SD) ACLR and Control groups for bilateral difference in anterior knee laxity, Marx activity score, and KOOS scores.

			KOOS				
	Bilateral Difference Anterior Laxity (mm)	Marx Activity Total Score	Pain	Symptom	ADL	Sport/Rec	QOL
ACLR	2.4 ± 1.7	12.08 ± 4.43	87.15 ± 10.98	72.17 ± 14.71	96.20 ± 5.41	79.58 ± 18.17	69.53 ± 22.75
Control	1.2 ± 1.0	10.87 ± 3.16	95.37 ± 6.29	92.49 ± 8.89	98.23 ± 3.19	92.44 ± 11.52	90.87 ± 14.04

Table 25. Asymmetry of kinematics (°) and moments (Nm/BHxBW) and kinetics (N/BW) for groups at Initial Ground Contact for Research Question #3.

	ACLR			CONTROL			Z_{KS}	P value
	Mean	(SD)	Median	Mean	(SD)	Median		
Knee Sagittal	6.982	(4.439)	6.179	7.337	(5.505)	6.260	0.568	0.903
Knee Frontal	8.484	(6.339)	6.350	7.970	(5.686)	6.348	0.432	0.993
Knee Transverse	7.120	(4.818)	6.154	6.340	(5.211)	5.444	0.507	0.960
Hip Sagittal	7.571	(4.189)	7.475	7.213	(5.464)	5.944	0.630	0.822
Hip Frontal	11.271	(9.178)	8.129	13.047	(9.431)	12.793	0.690	0.725
Hip Transverse	8.858	(6.502)	7.567	9.423	(7.653)	7.617	0.395	0.998
Knee Sagittal Moment	0.025	(0.023)	0.021	0.026	(0.022)	0.020	0.432	0.992
Knee Frontal Moment	0.021	(0.022)	0.016	0.023	(0.015)	0.021	0.939	0.341
Knee Transverse Moment	0.011	(0.008)	0.008	0.010	(0.008)	0.008	1.075	0.198
Hip Sagittal Moment	0.070	(0.048)	0.058	0.070	(0.058)	0.059	0.729	0.663
Hip Frontal Moment	0.058	(0.047)	0.045	0.061	(0.041)	0.055	0.618	0.840
Hip Transverse Moment	0.032	(0.027)	0.026	0.021	(0.015)	0.019	0.791	0.559
Anterior Tibial Shear Force	0.109	(0.083)	0.080	0.114	(0.097)	0.094	0.704	0.704
Vertical Ground Reaction Force	0.024	(0.028)	0.017	0.031	(0.028)	0.024	0.988	0.283

Table 26. Asymmetry of peak kinematics (°) during Landing Phase for Research Question #3.

	ACLR			CONTROL			Z_{KS}	P value
	Mean	(SD)	Median	Mean	(SD)	Median		
Knee Flexion	8.836	(6.409)	7.181	9.857	(6.604)	9.193	0.704	0.704
Knee Extension	6.982	(4.439)	6.179	7.337	(5.505)	6.260	0.568	0.903
Knee Varus	8.643	(8.499)	6.876	9.974	(6.573)	8.477	1.129	0.075
Knee Valgus	9.939	(9.767)	5.763	11.308	(7.983)	8.628	1.161	0.135
Knee Internal Rotation	8.704	(5.643)	8.114	7.579	(4.739)	7.254	0.877	0.425
Knee External Rotation	8.070	(5.009)	6.642	6.793	(5.152)	5.858	0.889	0.407
Hip Extension	7.528	(4.172)	7.488	7.043	(5.593)	6.004	0.927	0.357
Hip Flexion	8.621	(5.505)	8.668	8.971	(6.058)	8.400	0.519	0.951
Hip Adduction	9.441	(8.723)	7.556	12.998	(9.481)	11.474	0.988	0.283
Hip Abduction	9.052	(7.040	8.884	11.461	(7.867)	9.431	0.692	0.725
Hip Internal Rotation	12.314	(8.073)	11.090	10.794	(8.393)	8.273	0.828	0.500
Hip External Rotation	11.349	(8.470)	8.668	10.306	(8.551)	8.591	0.778	0.580

Table 27. Asymmetry of peak moment and kinetic variables (Nm/BHxBW) during Landing Phase for Research Question #3.

	ACLR			CONTROL			Z_{KS}	P value
	Mean (SD)		Median	Mean (SD)		Median		
Knee Extension Mom	0.050	(0.041)	0.037	0.043	(0.026)	0.042	0.667	0.765
Knee Flexion Mom	0.039	(0.029)	0.032	0.026	(0.030)	0.015	1.421	0.035
Knee Valgus Mom	0.026	(0.021)	0.020	0.032	(0.019)	0.028	1.186	0.120
Knee Varus Mom	0.033	(0.024)	0.027	0.036	(0.028)	0.032	0.507	0.960
Knee ER Mom	0.025	(0.020)	0.019	0.023	(0.017)	0.020	0.469	0.980
Knee IR Mom	0.020	(0.024)	0.012	0.017	(0.019)	0.013	0.692	0.725
Hip Flexion Mom	0.084	(0.047)	0.072	0.065	(0.055)	0.050	1.137	0.151
Hip Extension Mom	0.088	(0.078)	0.069	0.079	(0.086)	0.054	0.939	0.341
Hip Abduction Mom	0.070	(0.059)	0.058	0.057	(0.050)	0.047	0.902	0.390
Hip Adduction Mom	0.060	(0.054)	0.055	0.045	(0.037)	0.033	0.803	0.539
Hip ER Mom	0.031	(0.027)	0.023	0.026	(0.024)	0.023	0.914	0.373
Hip IR Mom	0.023	(0.019)	0.017	0.021	(0.022)	0.013	0.791	0.559
Anterior Tibial Shear Force	0.190	(0.154)	0.156	0.170	(0.132)	0.148	0.605	0.857
Posterior Tibial Shear Force	0.139	(0.095)	0.133	0.131	(0.125)	0.101	0.865	0.443
Vertical Ground Reaction Force	0.682	(0.508)	0.554	0.465	(0.279)	0.388	1.445	0.031

Table 28. Participant demographics and anthropometrics for Research Question #4. Units of measure are Age (years), Height (cm), and Mass (kg). Values represent means ± standard deviation.

	n	Age	Height	Mass
ACLR	28 (17 m, 11 f)	21.46 ± 0.79	172.49 ± 8.73	75.30 ± 11.77
Control	37 (20 m, 17 f)	21.03 ± 0.74	172.66 ± 9.69	73.06 ± 14.16

Table 29. Descriptive statistics for Research Question #4. Bilateral difference of anterior knee laxity assessed using KT-1000, Marx Activity score, and KOOS for each group. Values represent mean ± standard deviation, and units of measure other than anterior laxity (mm) are scale dependent.

			KOOS				
	Bilateral Difference Anterior Laxity (mm)	Marx Activity Total Score	Pain	Symptom	ADL	Sport/Rec	QOL
ACLR	2.4 ± 1.6	12.25 ± 4.21	86.57 ± 11.32	72.57 ± 14.65	95.68 ± 6.11	79.46 ± 17.71	69.87 ± 21.99
Control	1.0 ± 0.7	10.84 ± 3.10	95.38 ± 6.38	92.65 ± 9.06	98.11 ± 3.23	92.70 ± 11.88	89.86 ± 15.33

*Laxity measures were not obtained for 1 member of the ACLR group

Table 30. Asymmetry of mean coupling angles during Landing Phase for Research Question #4. (Mean (Standard Deviation), Median, Kolmogorov-Smirnov Z score (Z_{KS}))

	ACLR			CONTROL			Z_{KS}	P value
	Mean (SD)		Median	Mean (SD)		Median		
Hip Sagittal - Knee Sagittal	2.371	(1.891)	1.860	1.961	(1.164)	1.767	0.778	0.580
Hip Frontal – Knee Frontal	9.031	(5.938)	7.454	8.649	(6.452)	6.542	0.482	0.974
Hip Transverse – Knee Transverse	7.377	(6.828)	5.646	7.201	(5.237)	6.374	0.497	0.966
Hip Frontal – Knee Transvers	6.095	(5.381)	3.598	7.015	(6.547)	5.024	0.435	0.991
Hip Transverse – Knee Frontal	4.084	(3.368)	3.353	5.165	(4.988)	3.848	0.728	0.664

Table 31. Asymmetry of between trial variability during Landing Phase for Research Question #4. (Mean (Standard Deviation), Median, Kolmogorov-Smirnov Z score (Z_{KS}))

	ACLR		CONTROL		Z_{KS}	P value
	Mean (SD)	Median	Mean (SD)	Median		
Hip Sagittal - Knee Sagittal	1.228 (1.023)	0.897	1.219 (0.874)	1.047	0.632	0.819
Hip Frontal – Knee Frontal	2.820 (2.006)	2.531	2.683 (1.674)	2.496	0.566	0.905
Hip Transverse – Knee Transverse	2.362 (2.294)	1.726	2.333 (1.717)	2.177	0.705	0.703
Hip Frontal – Knee Transverse	2.807 (2.444)	2.452	2.179 (1.727)	2.218	0.882	0.417
Hip Transverse – Knee Frontal	3.058 (2.424)	2.667	2.229 (1.801)	1.918	0.724	0.670

Figure 1. Double leg jump landing. Participants were required to stand atop a box located a distance equal to one half of their body height from the front edge of the force plate, jump forward, land with their foot completely on the force plate, and then immediately make a vertical jump for maximum height.

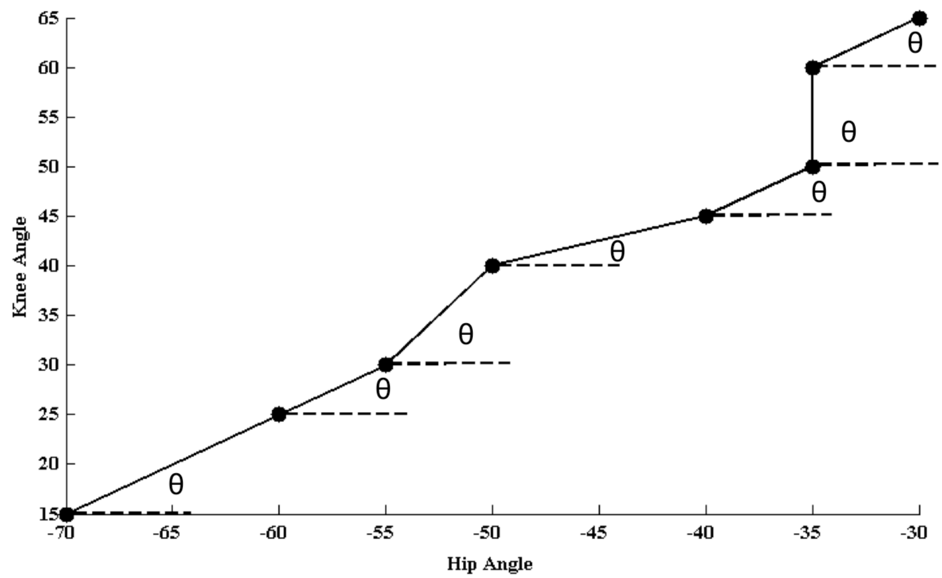

Figure 2. Visual representation of the calculation of coupling angles. Determined using vector coding to characterize joint coordination.

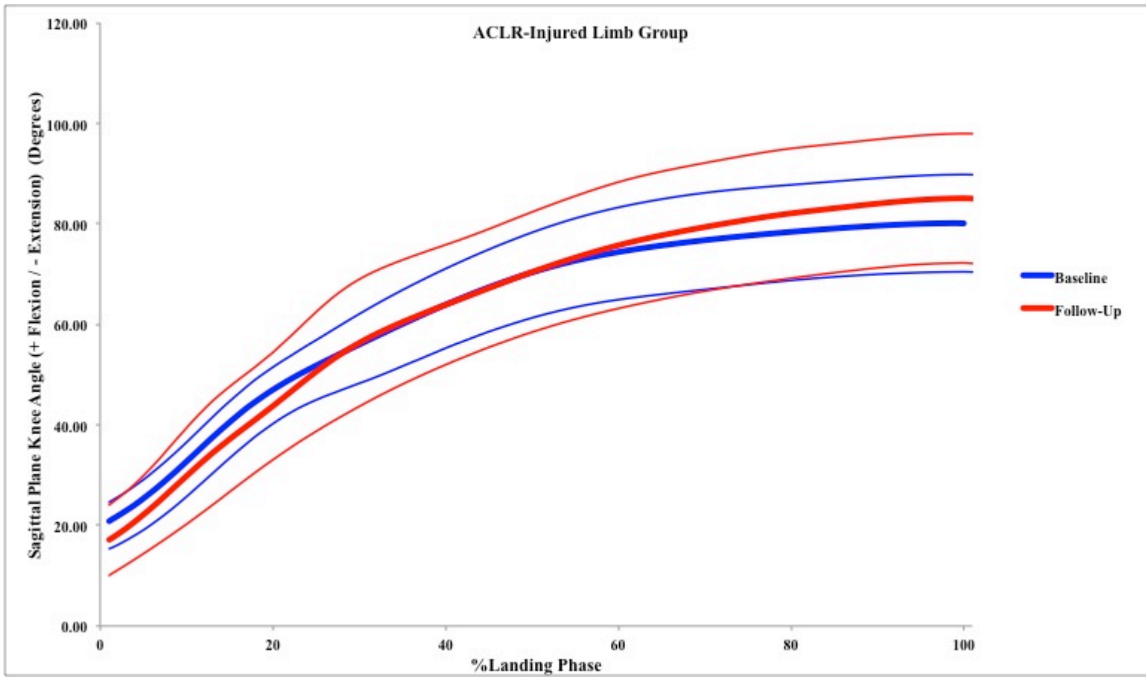

Figure 3. Ensemble average plot of sagittal plane knee angle at Baseline and Follow-Up during the Landing Phase for the ACLR-Injured Limb Group.

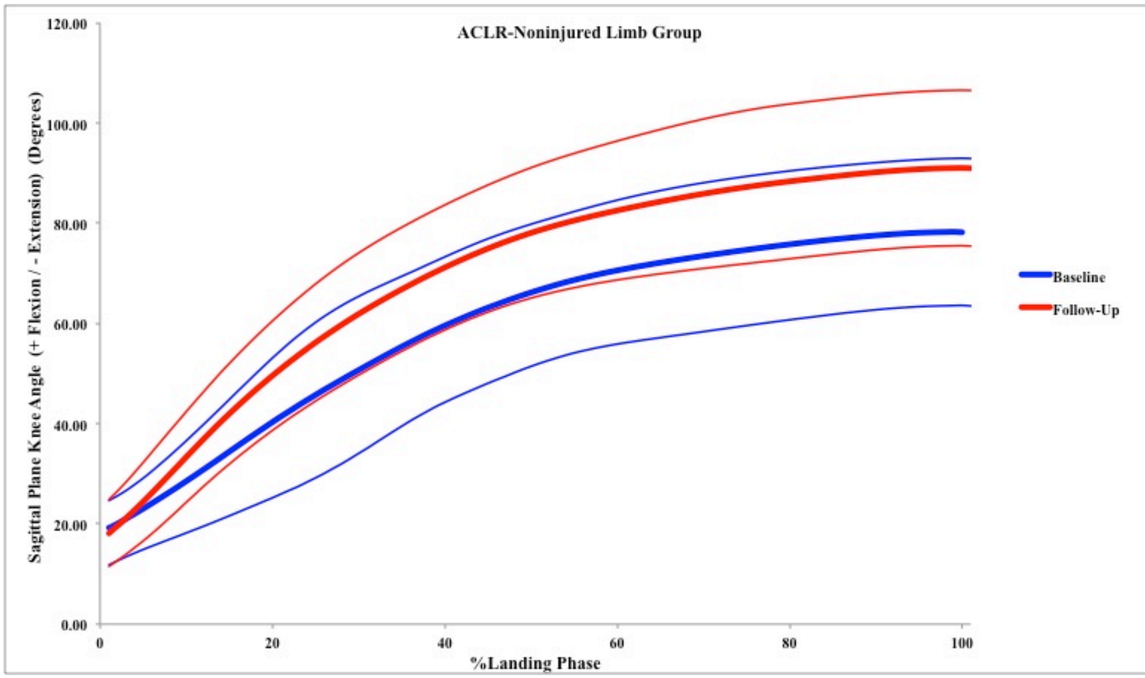

Figure 4. Ensemble average plot of sagittal plane knee angle at Baseline and Follow-Up during the Landing Phase for the ACLR-Noninjured Limb Group.

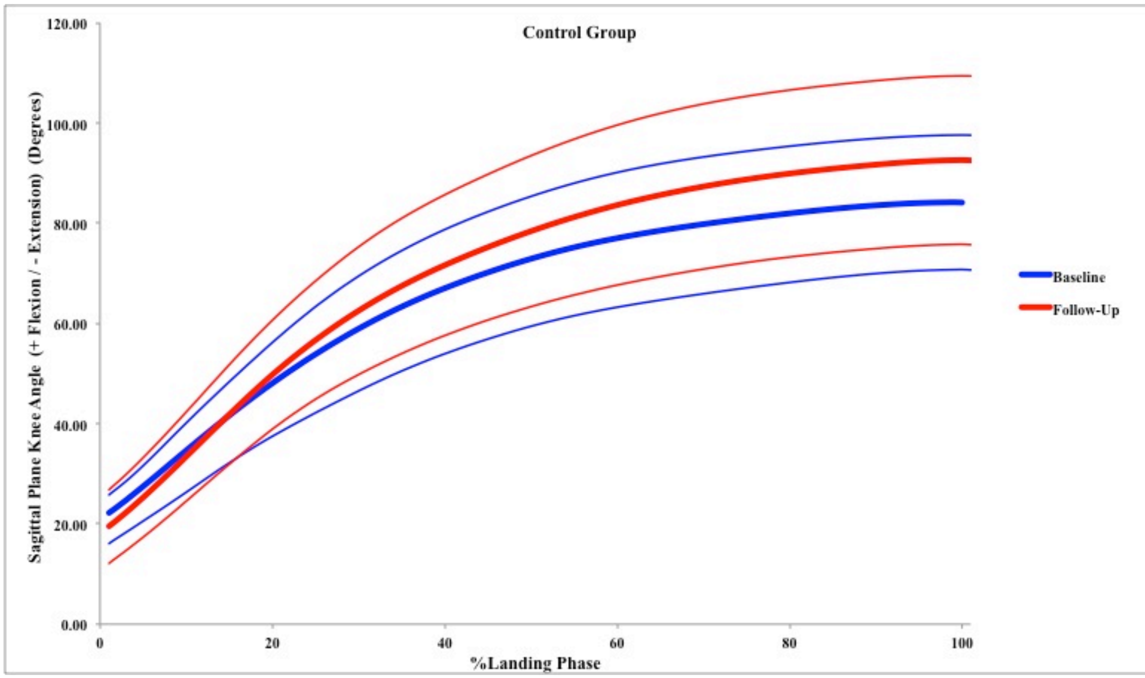

Figure 5. Ensemble average plot of sagittal plane knee angle at Baseline and Follow-Up during the Landing Phase for the Control Group.

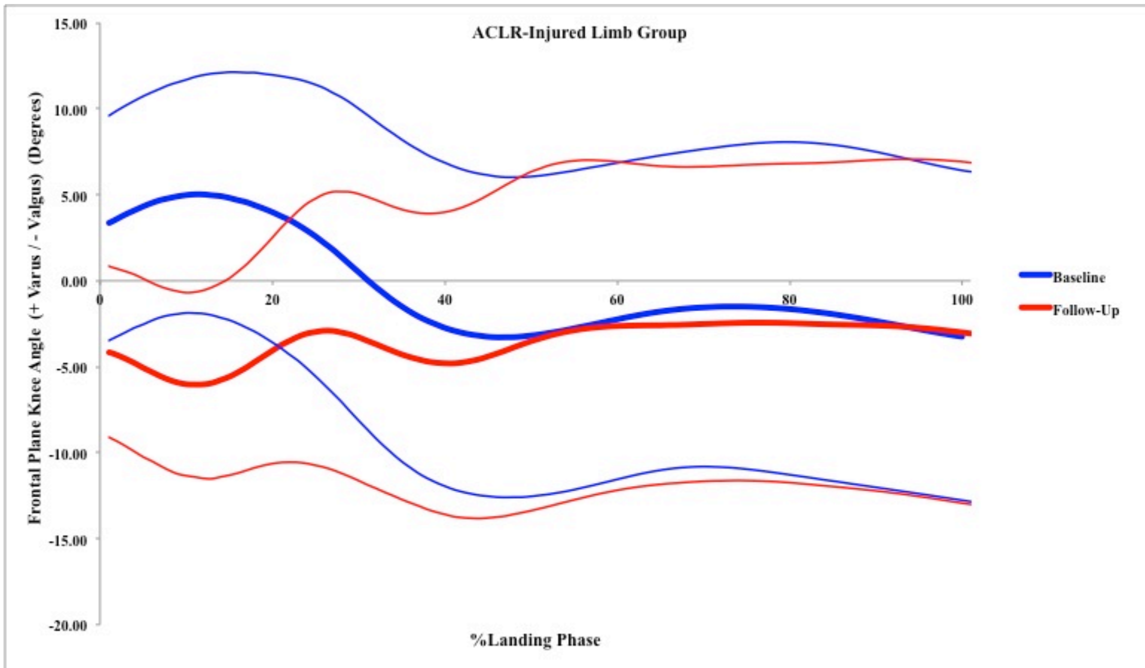

Figure 6. Ensemble average plot of frontal plane knee angle at Baseline and Follow-Up during the Landing Phase for the ACLR-Injured Limb Group.

Figure 7. Ensemble average plot of frontal plane knee angle at Baseline and Follow-Up during the Landing Phase for the ACLR-Noninjured Limb Group.

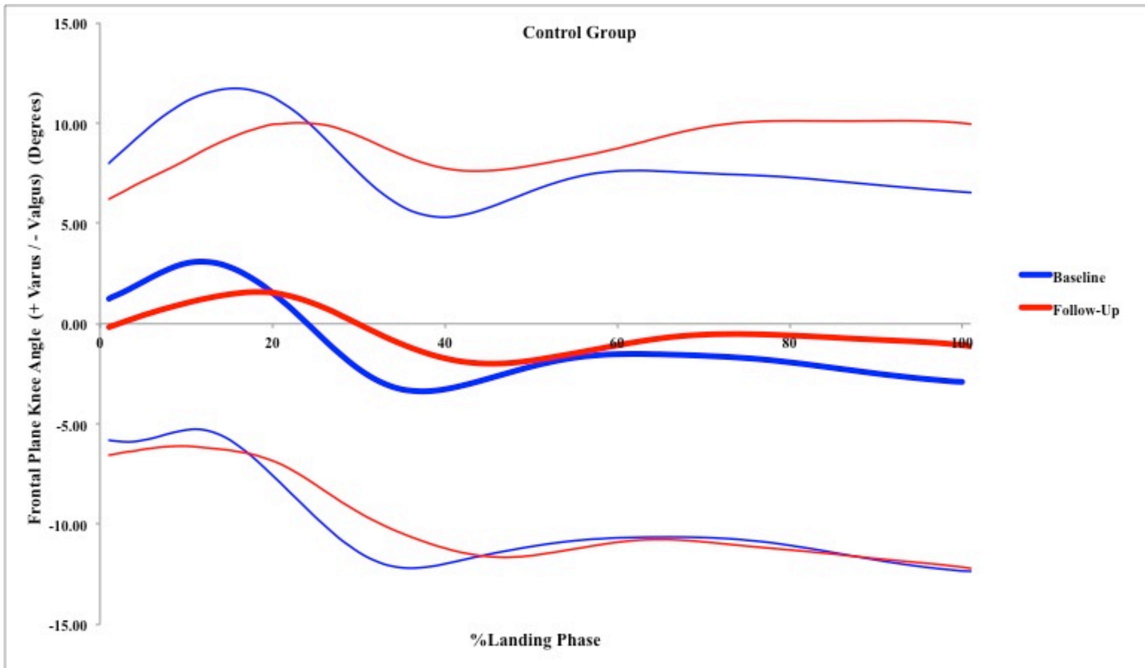

Figure 8. Ensemble average plot of frontal plane knee angle at Baseline and Follow-Up during the Landing Phase for the Control Group.

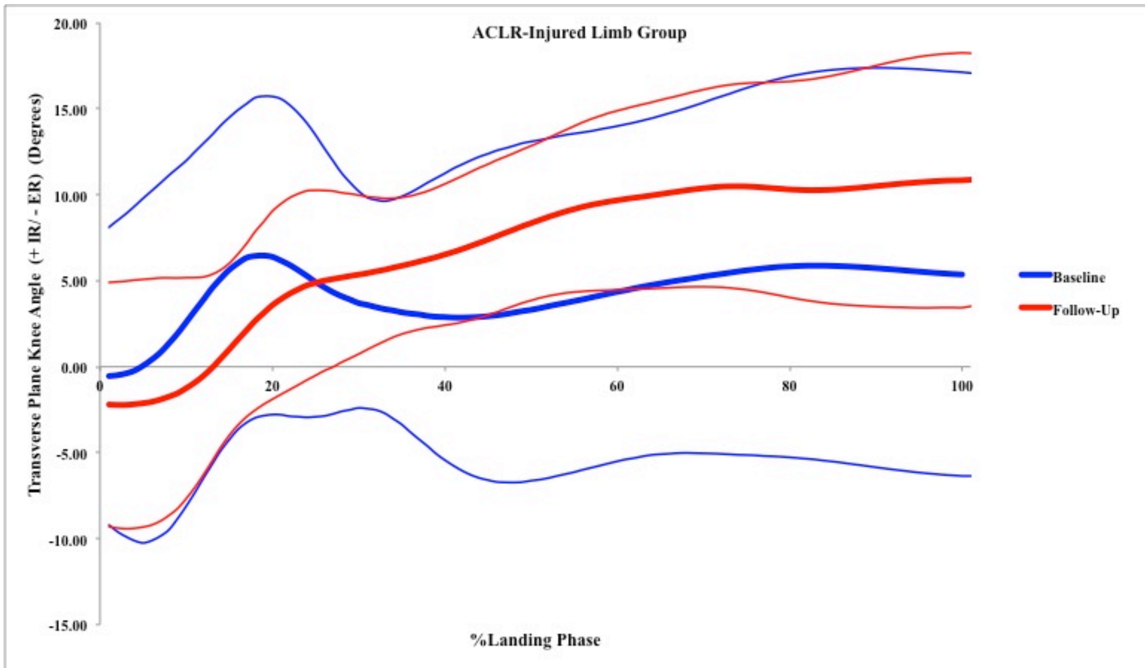

Figure 9. Ensemble average plot of transverse plane knee angle at Baseline and Follow-Up during the Landing Phase for the ACLR-Injured Limb Group.

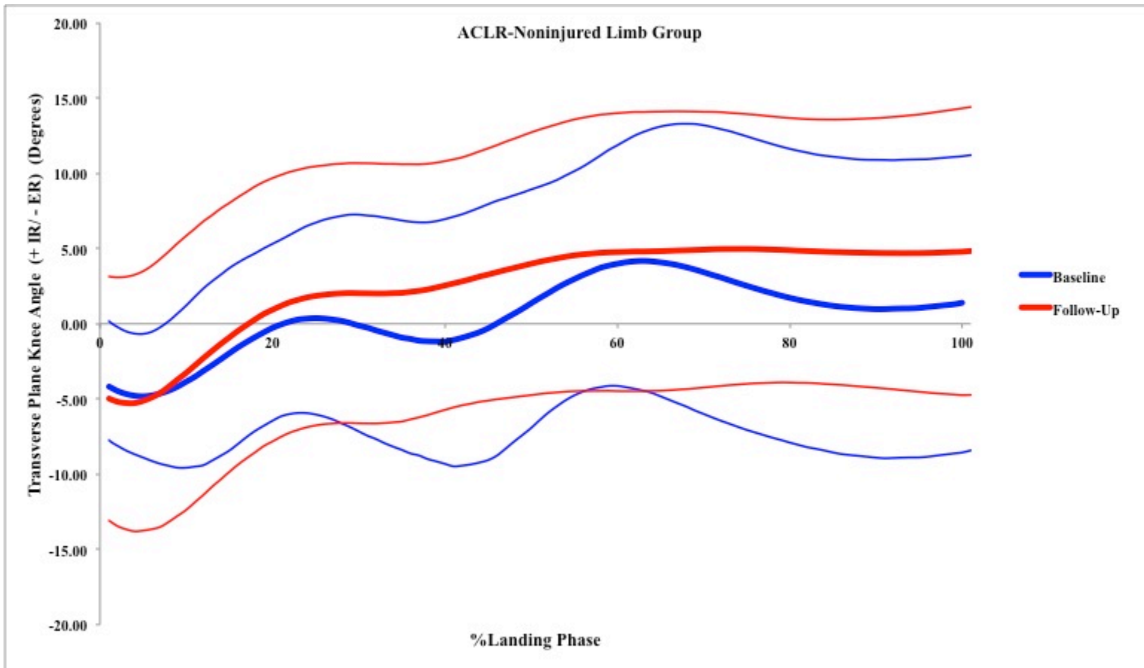

Figure 10. Ensemble average plot of transverse plane knee angle at Baseline and Follow-Up during the Landing Phase for the ACLR-Noninjured Limb Group.

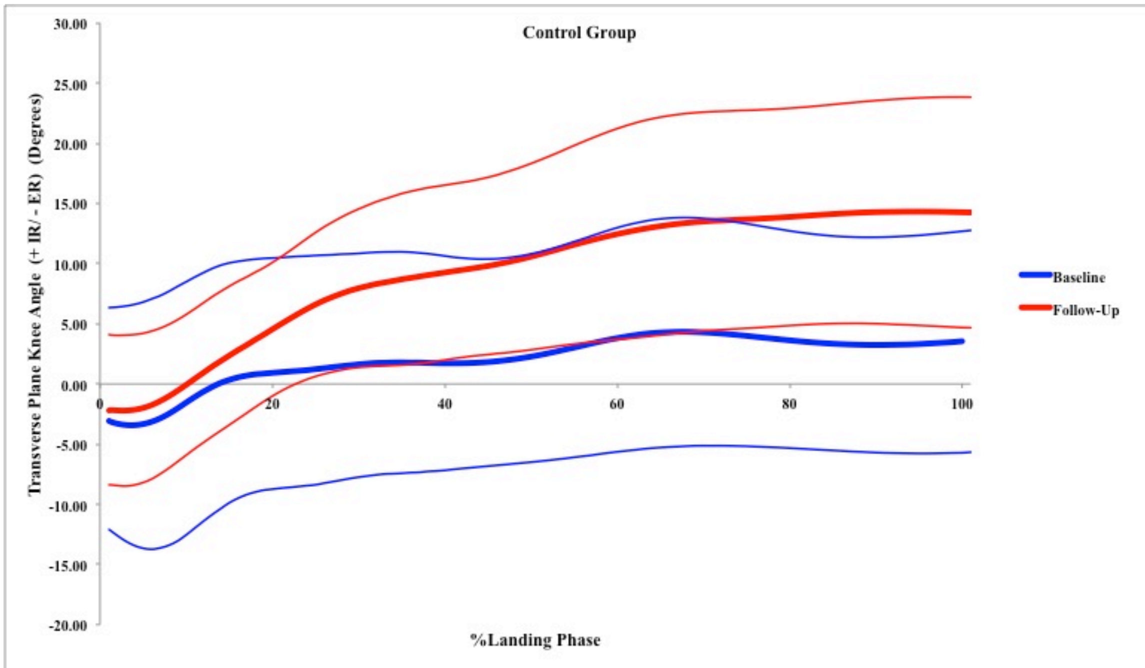

Figure 11. Ensemble average plot of transverse plane knee angle at Baseline and Follow-Up during the Landing Phase for the Control Group.

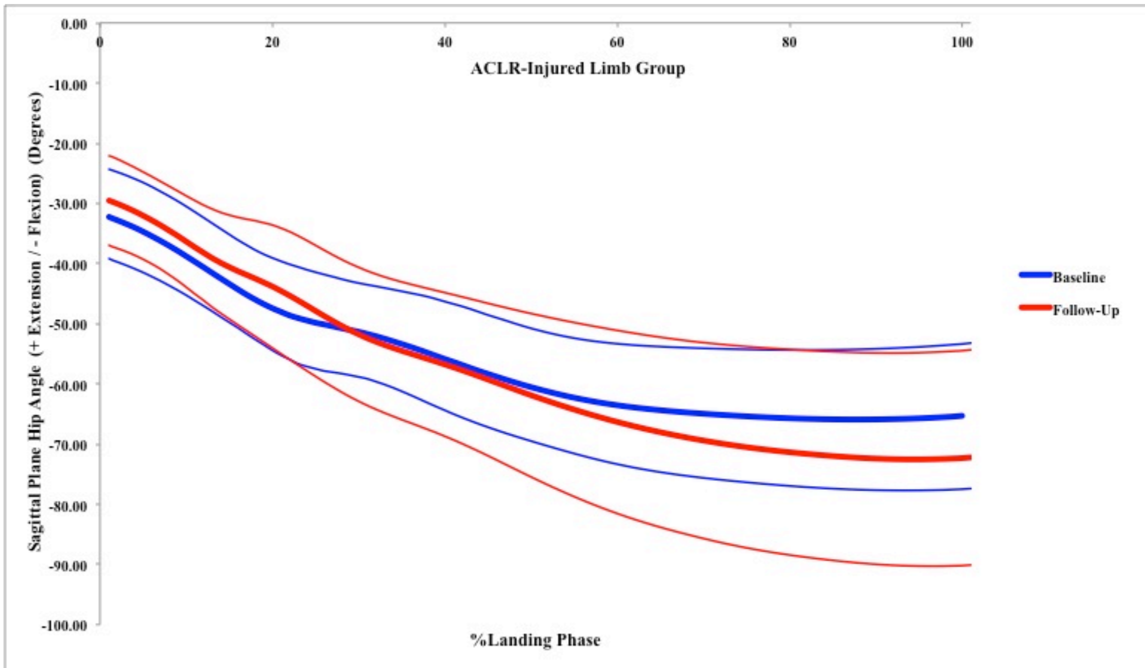

Figure 12. Ensemble average plot of sagittal plane hip angle at Baseline and Follow-Up during the Landing Phase for the ACLR-Injured Limb Group.

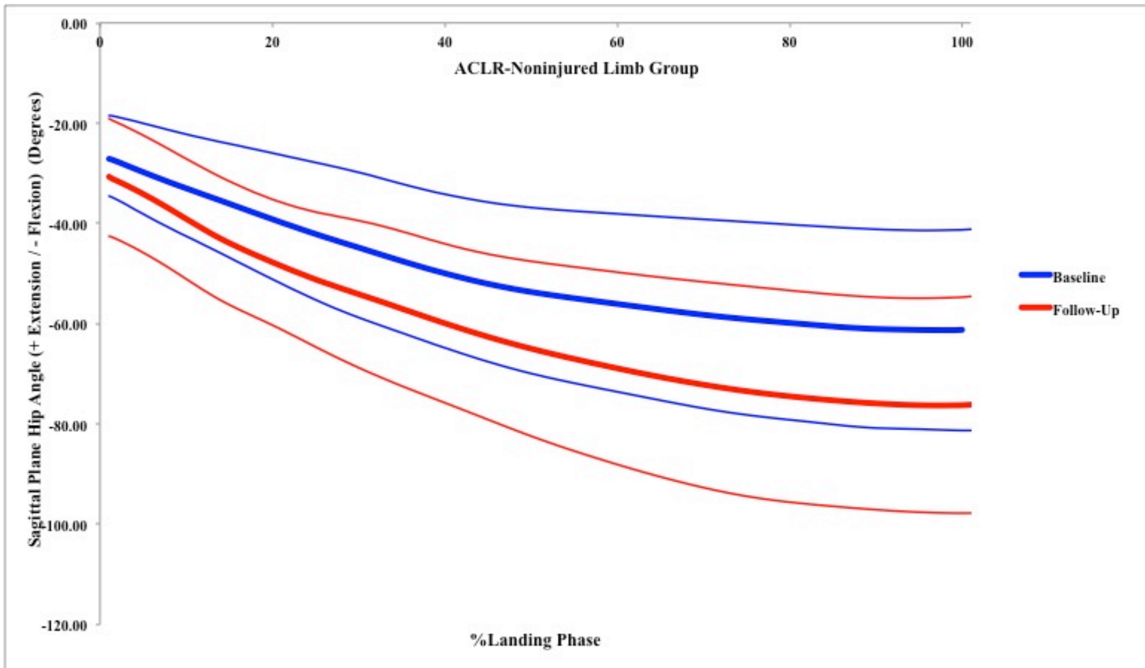

Figure 13. Ensemble average plot of sagittal plane hip angle at Baseline and Follow-Up during the Landing Phase for the ACLR-Noninjured Limb Group.

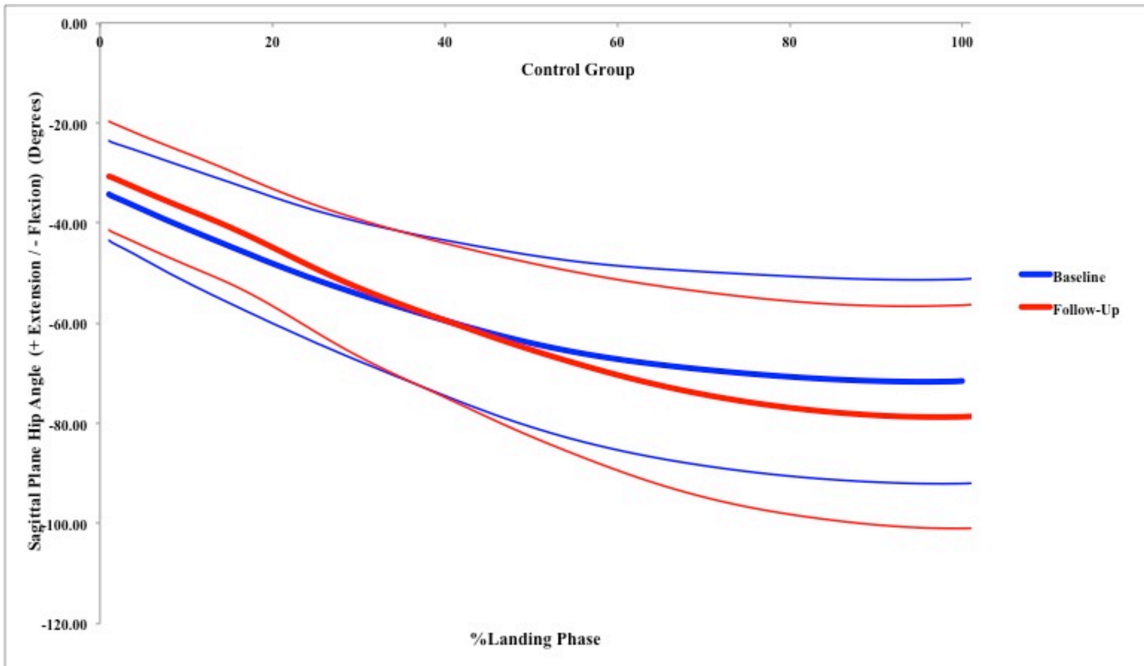

Figure 14. Ensemble average plot of sagittal plane hip angle at Baseline and Follow-Up during the Landing Phase for the Control Group.

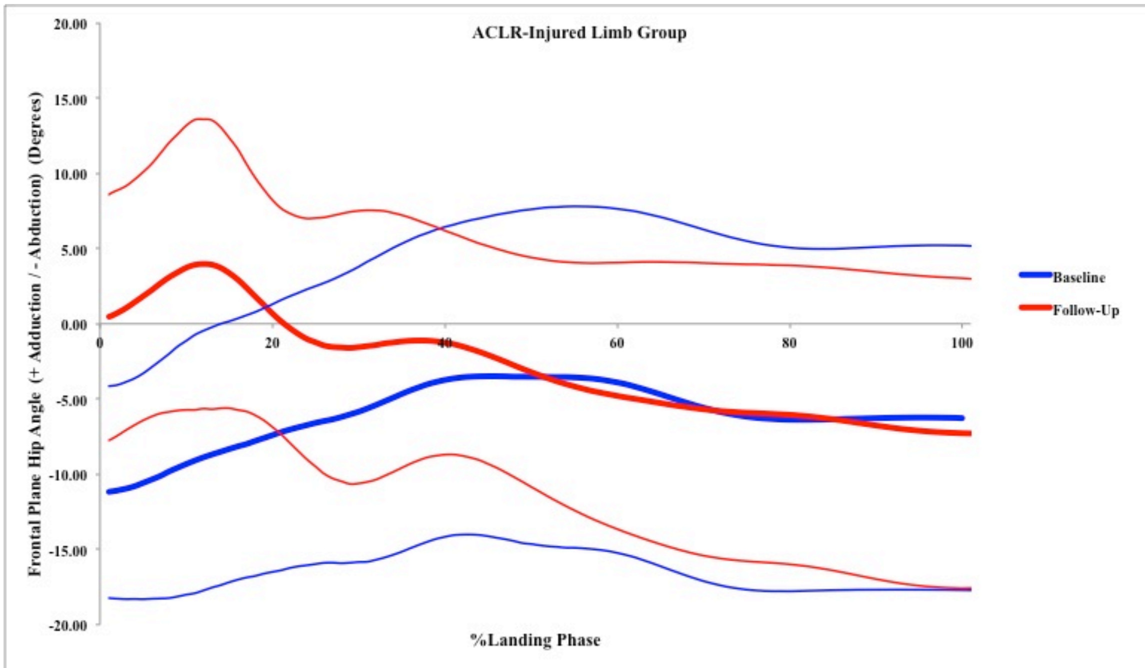

Figure 15. Ensemble average plot of frontal plane hip angle at Baseline and Follow-Up during the Landing Phase for the ACLR-Injured Limb Group.

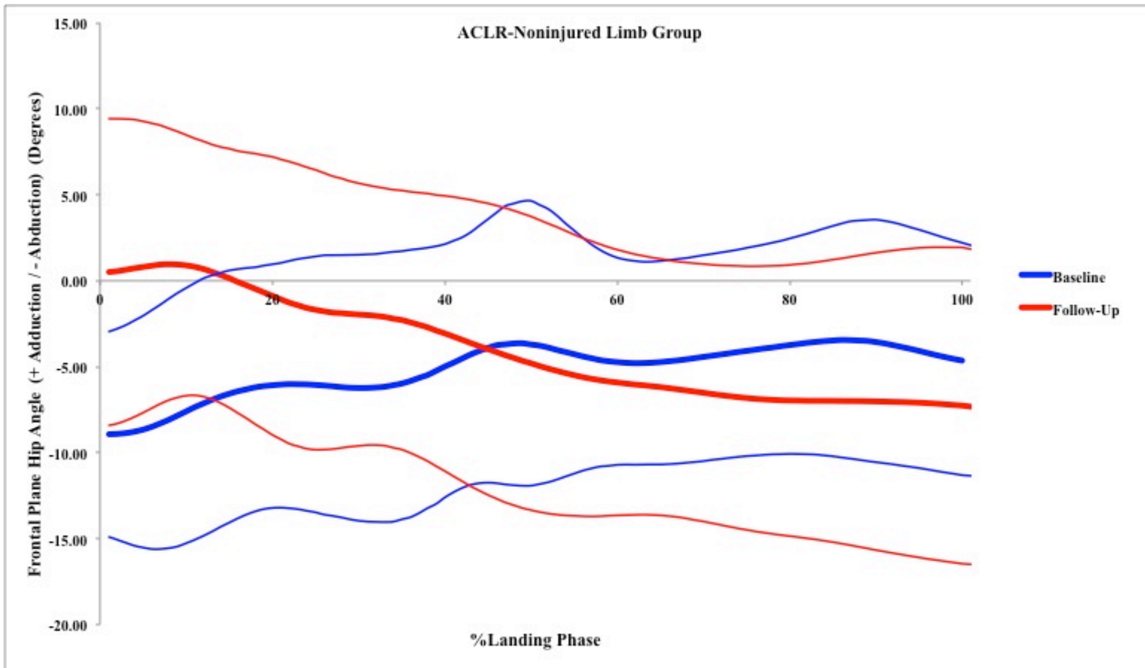

Figure 16. Ensemble average plot of frontal plane hip angle at Baseline and Follow-Up during the Landing Phase for the ACLR-Noninjured Limb Group.

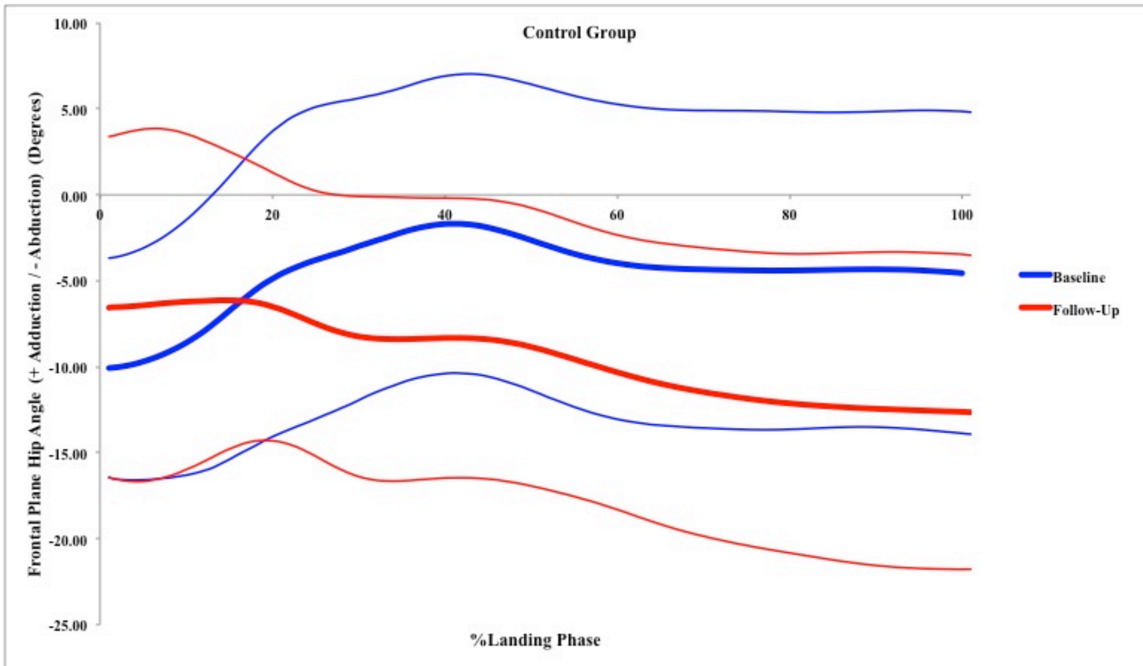

Figure 17. Ensemble average plot of frontal plane hip angle at Baseline and Follow-Up during the Landing Phase for the Control Group.

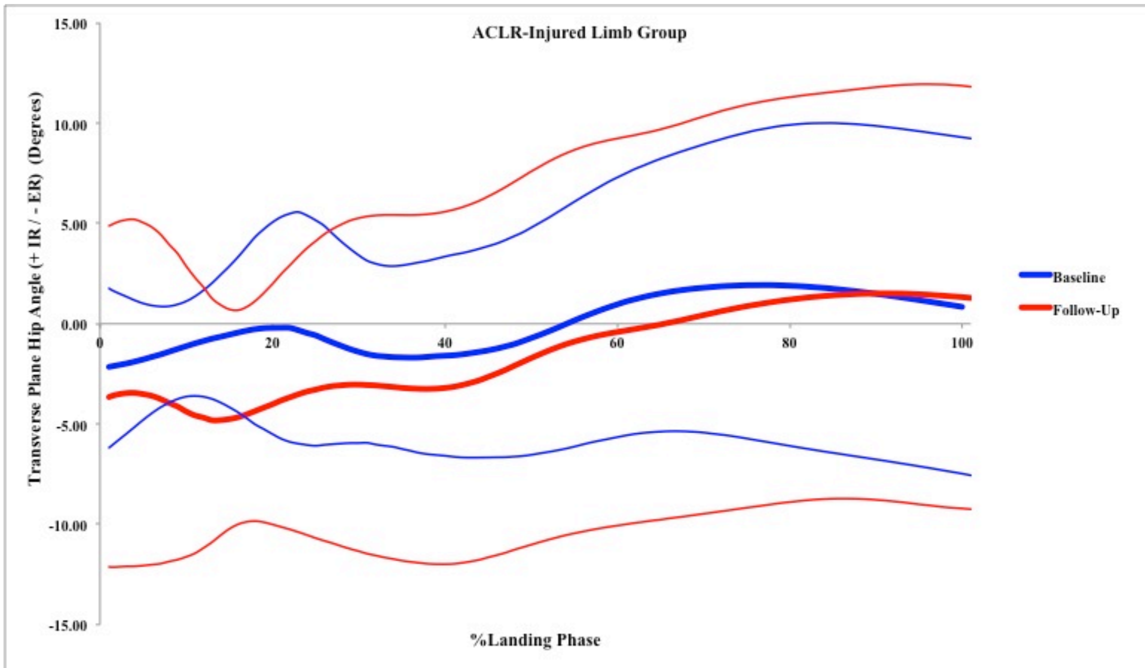

Figure 18. Ensemble average plot of transverse plane hip angle at Baseline and Follow-Up during the Landing Phase for the ACLR-Injured Limb Group.

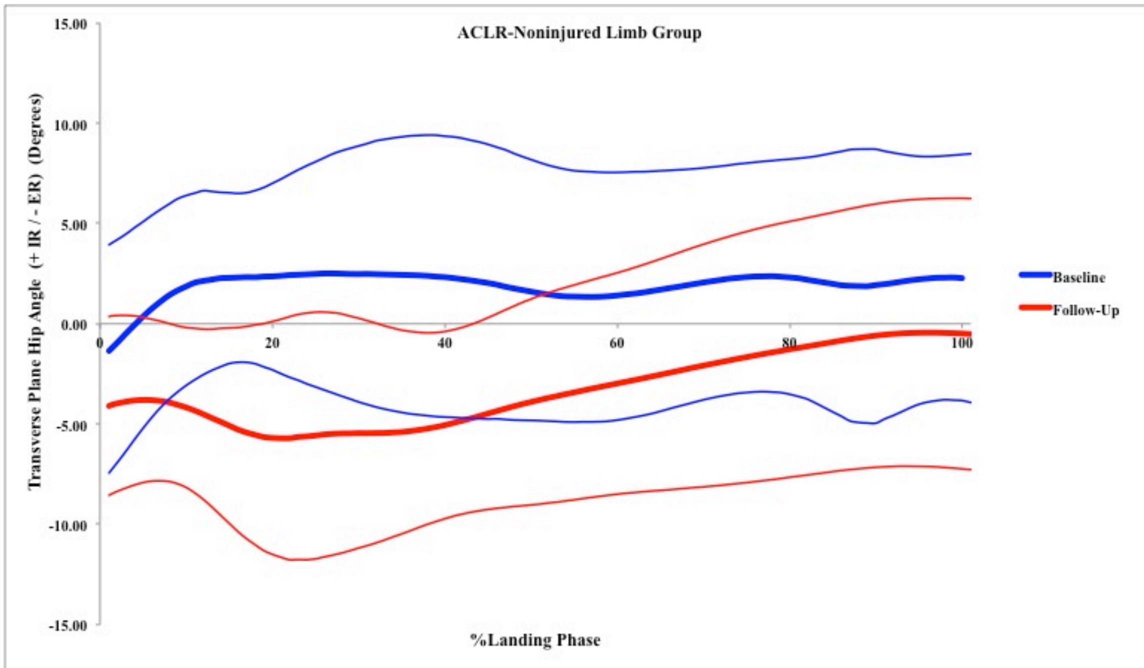

Figure 19. Ensemble average plot of transverse plane hip angle at Baseline and Follow-Up during the Landing Phase for the ACLR-Noninjured Limb Group.

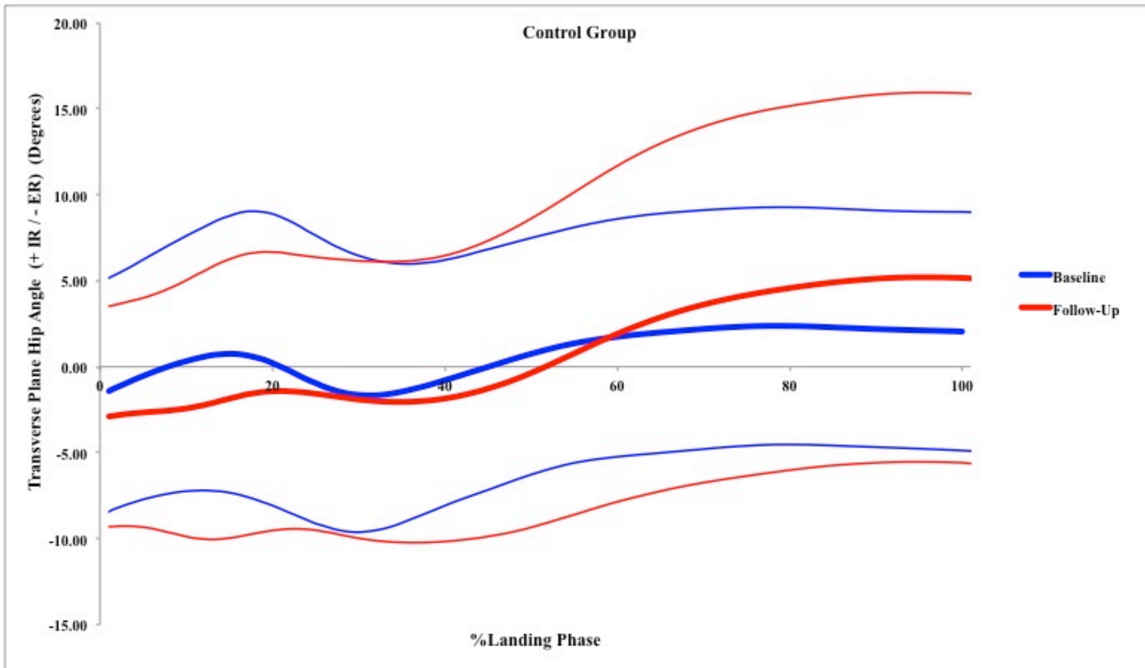

Figure 20. Ensemble average plot of transverse plane hip angle at Baseline and Follow-Up during the Landing Phase for the Control Group.

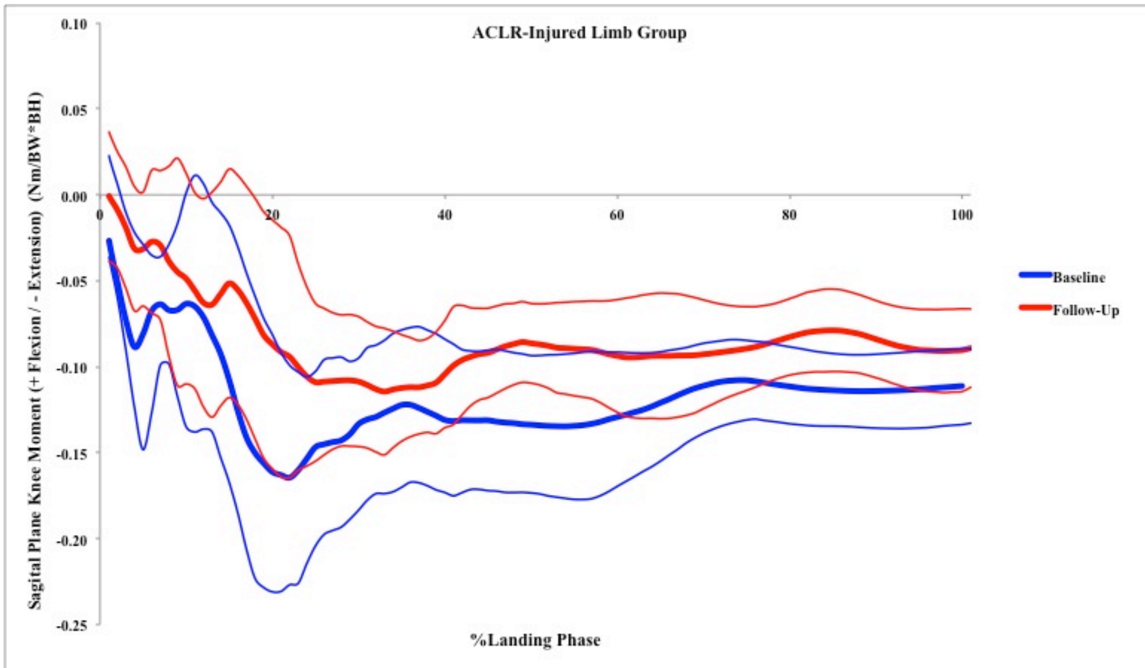

Figure 21. Ensemble average plot of sagittal plane knee moment at Baseline and Follow-Up during the Landing Phase for the ACLR-Injured Limb Group.

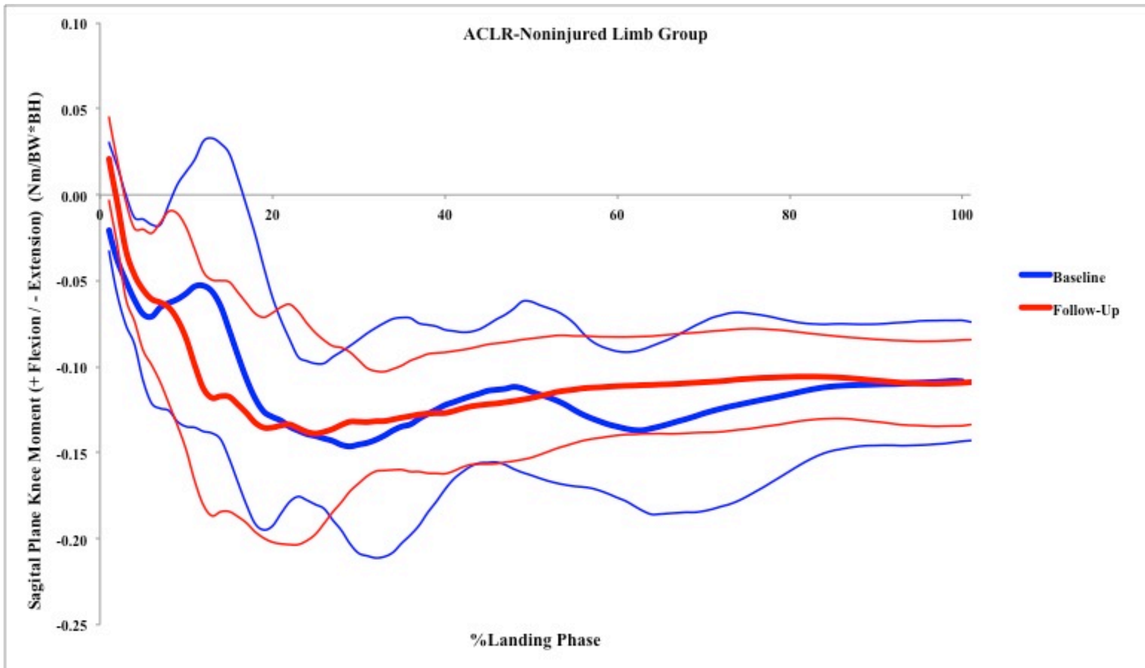

Figure 22. Ensemble average plot of sagittal plane knee moment at Baseline and Follow-Up during the Landing Phase for the ACLR-Noninjured Limb Group.

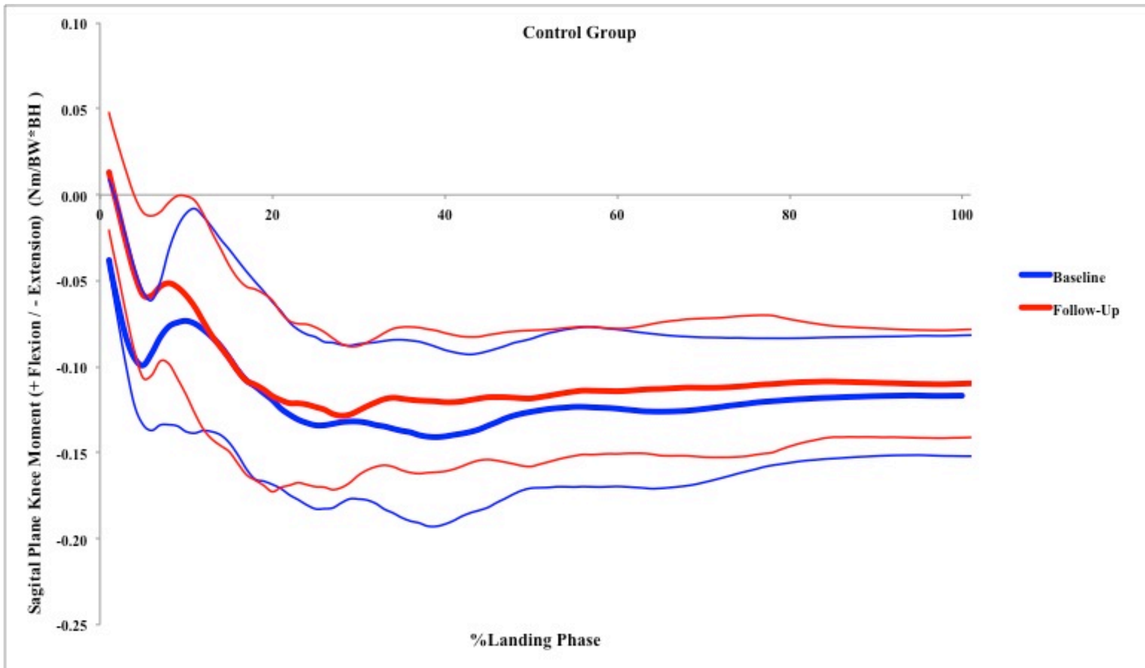

Figure 23. Ensemble average plot of sagittal plane knee moment at Baseline and Follow-Up during the Landing Phase for the Control Group.

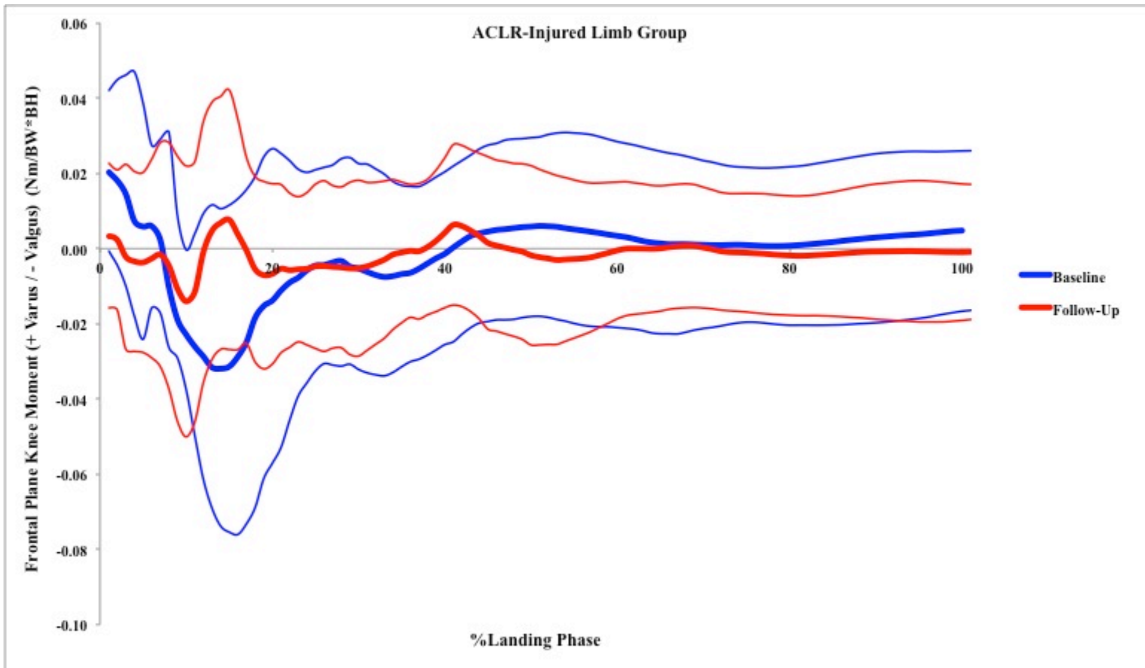

Figure 24. Ensemble average plot of frontal plane knee moment at Baseline and Follow-Up during the Landing Phase for the ACLR-Injured Limb Group.

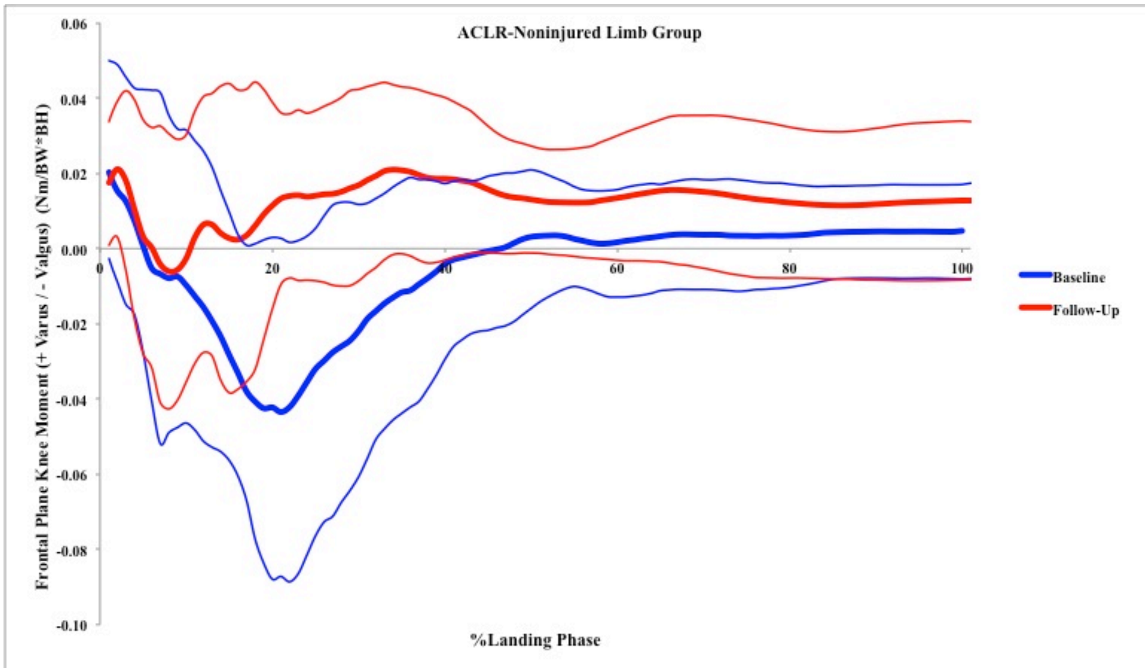

Figure 25. Ensemble average plot of frontal plane knee moment at Baseline and Follow-Up during the Landing Phase for the ACLR-Noninjured Limb Group.

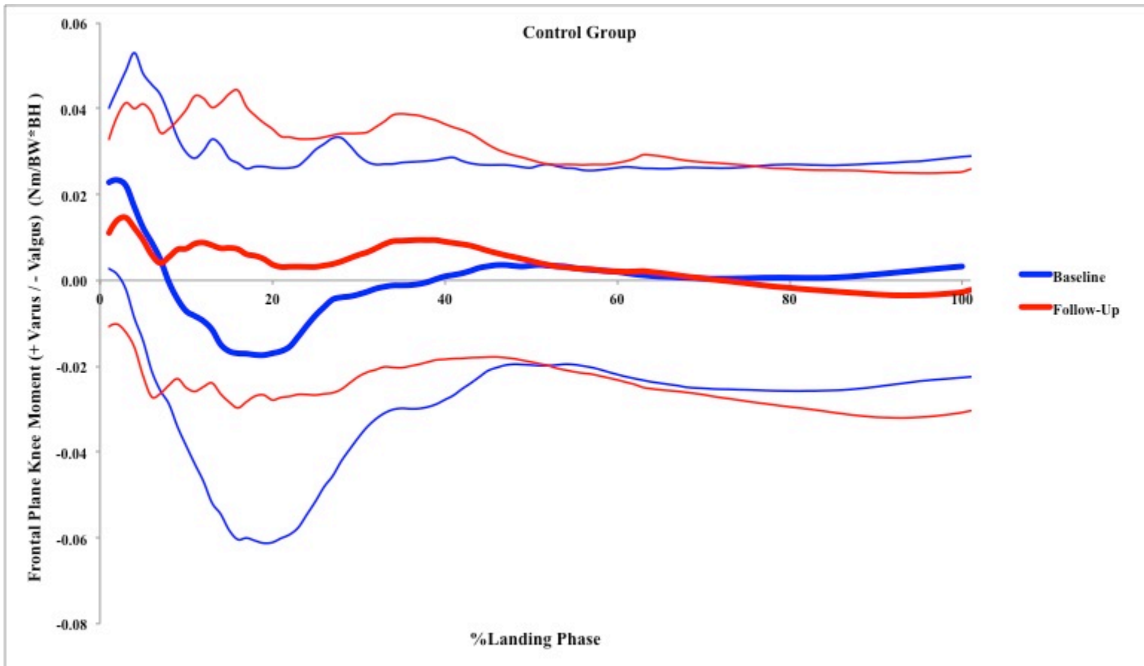

Figure 26. Ensemble average plot of frontal plane knee moment at Baseline and Follow-Up during the Landing Phase for the Control Group.

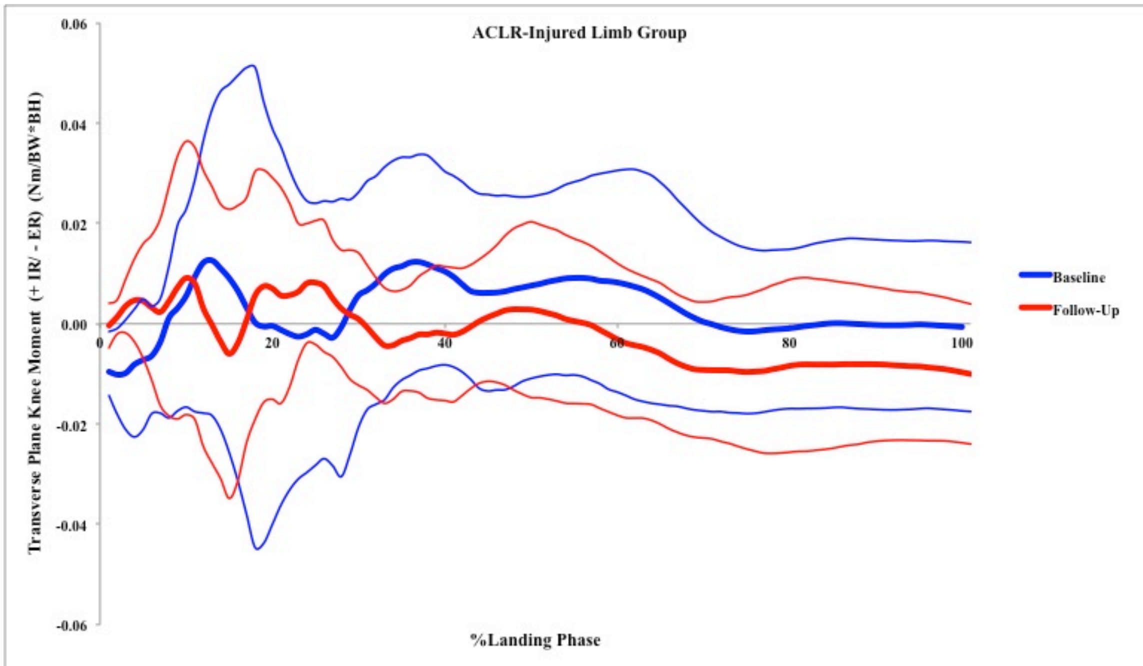

Figure 27. Ensemble average plot of transverse plane knee moment at Baseline and Follow-Up during the Landing Phase for the ACLR-Injured Limb Group.

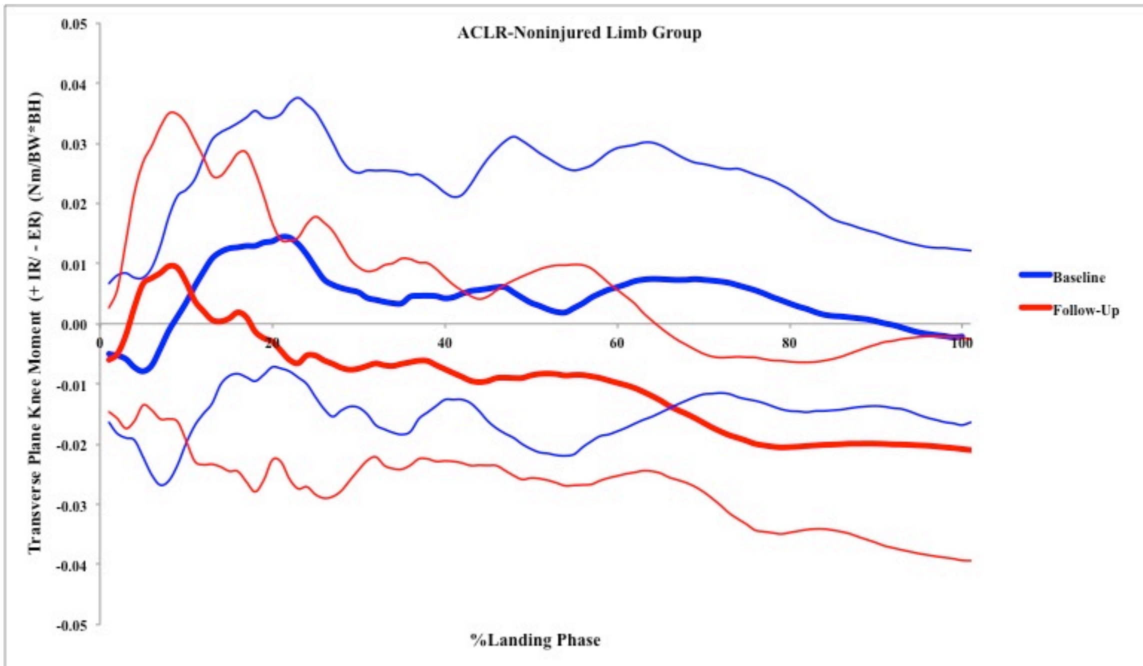

Figure 28. Ensemble average plot of transverse plane knee moment at Baseline and Follow-Up during the Landing Phase for the ACLR-Noninjured Limb Group.

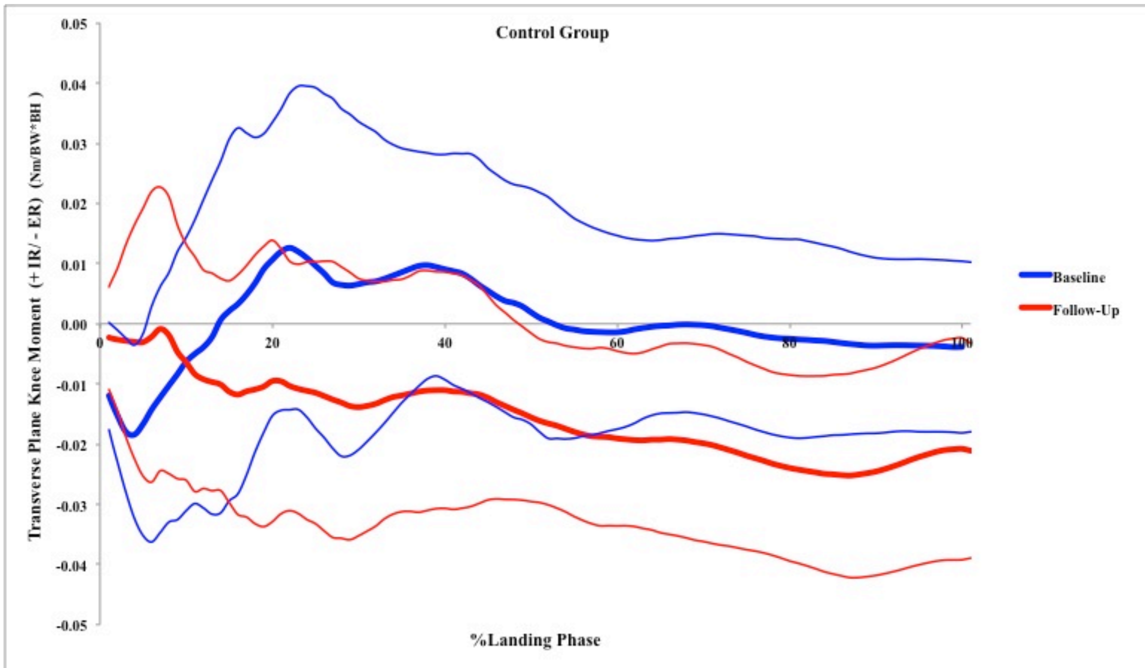

Figure 29. Ensemble average plot of transverse plane knee moment at Baseline and Follow-Up during the Landing Phase for the Control Group.

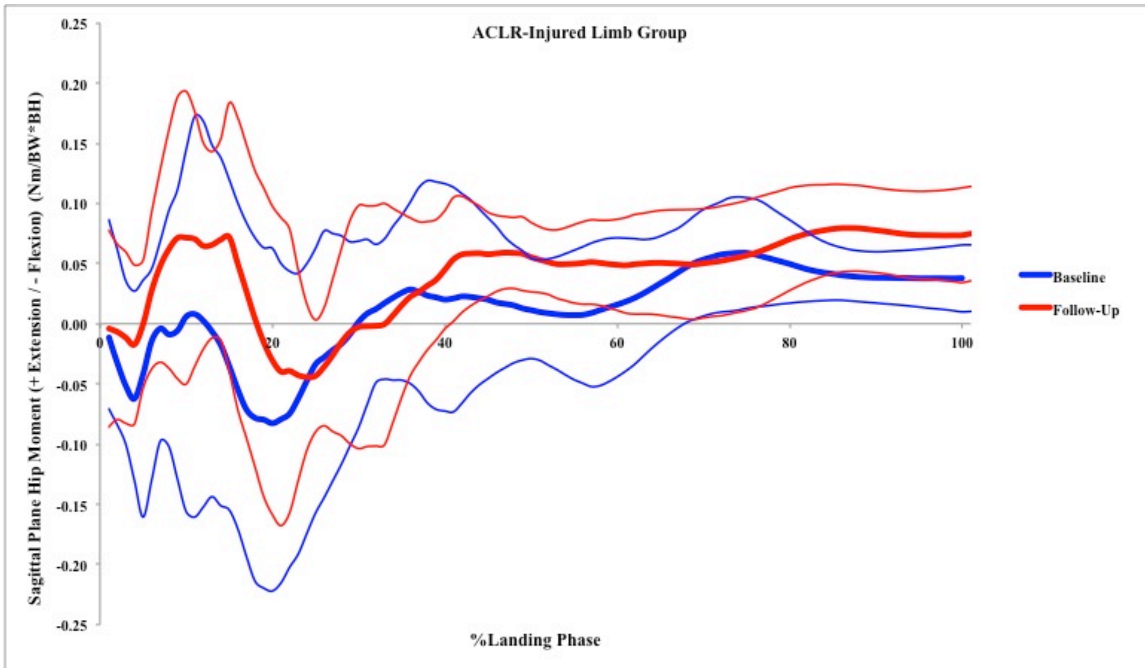

Figure 30. Ensemble average plot of sagittal plane hip moment at Baseline and Follow-Up during the Landing Phase for the ACLR-Injured Limb Group.

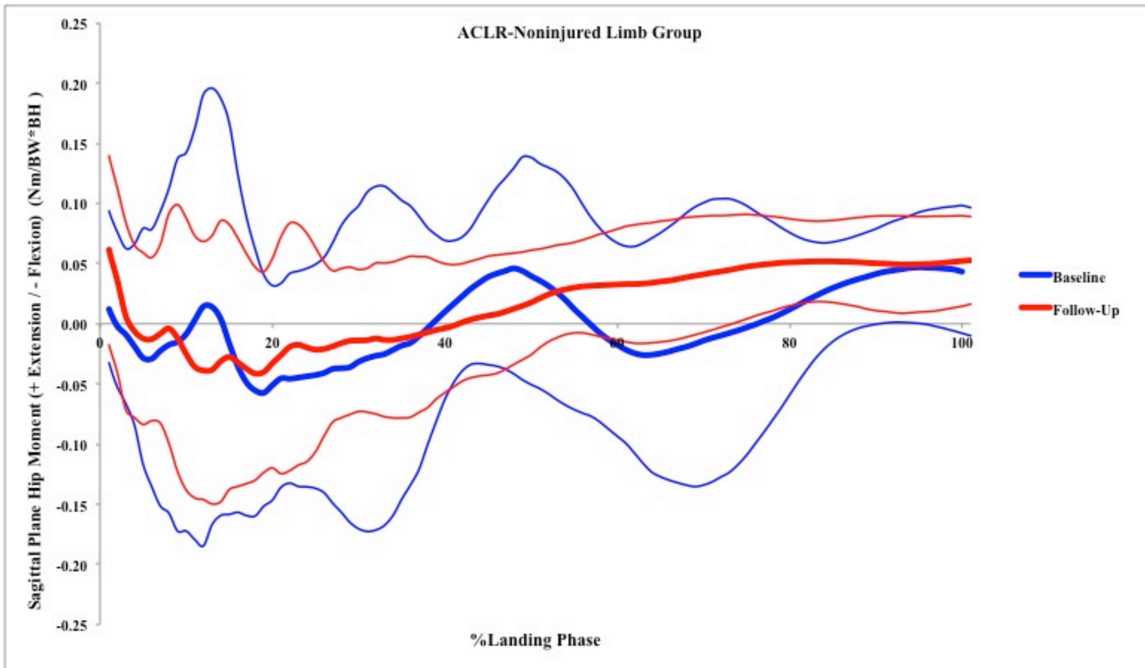

Figure 31. Ensemble average plot of sagittal plane hip moment at Baseline and Follow-Up during the Landing Phase for the ACLR-Noninjured Limb Group.

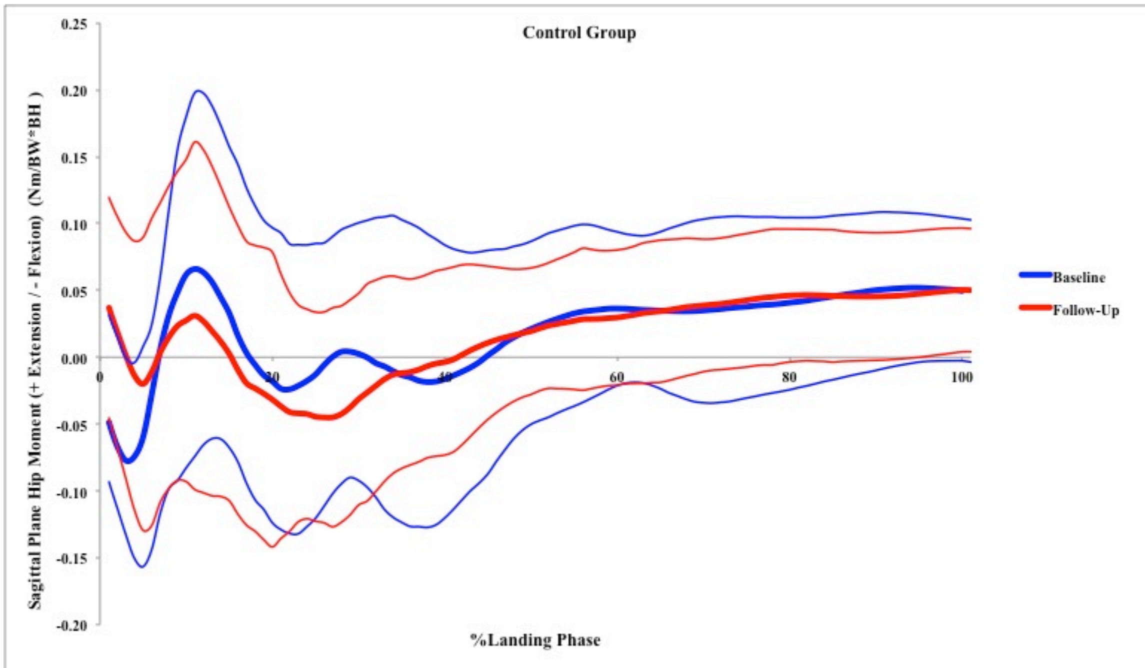

Figure 32. Ensemble average plot of sagittal plane hip moment at Baseline and Follow-Up during the Landing Phase for the Control Group.

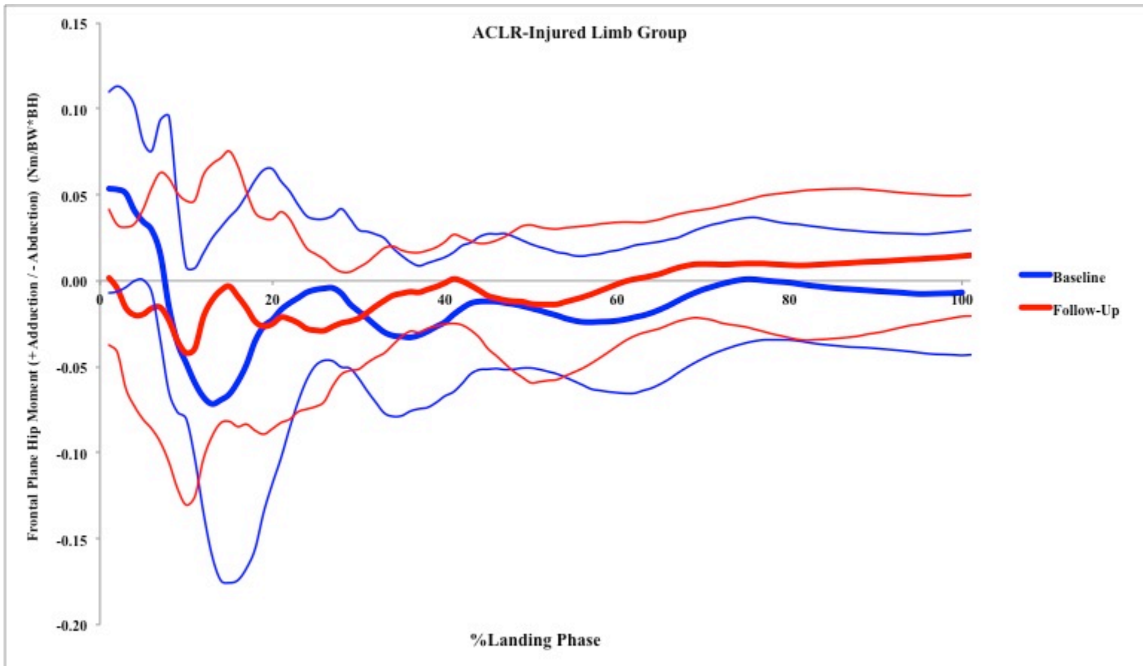

Figure 33. Ensemble average plot of frontal plane hip moment at Baseline and Follow-Up during the Landing Phase for the ACLR-Injured Limb Group.

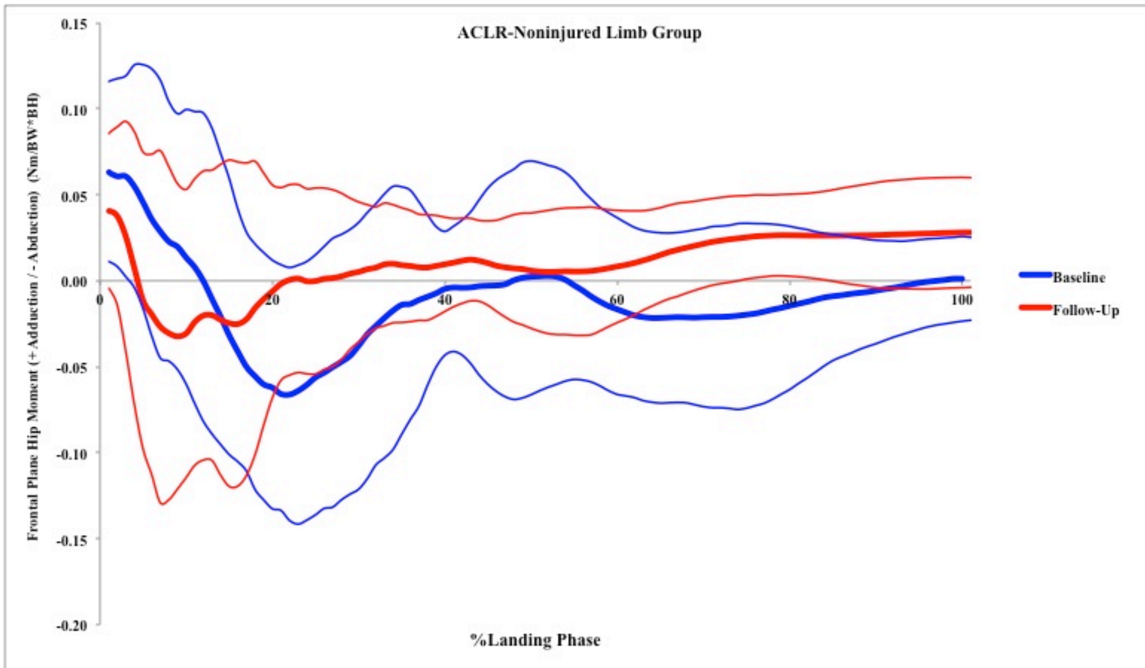

Figure 34. Ensemble average plot of frontal plane hip moment at Baseline and Follow-Up during the Landing Phase for the ACLR-Noninjured Limb Group.

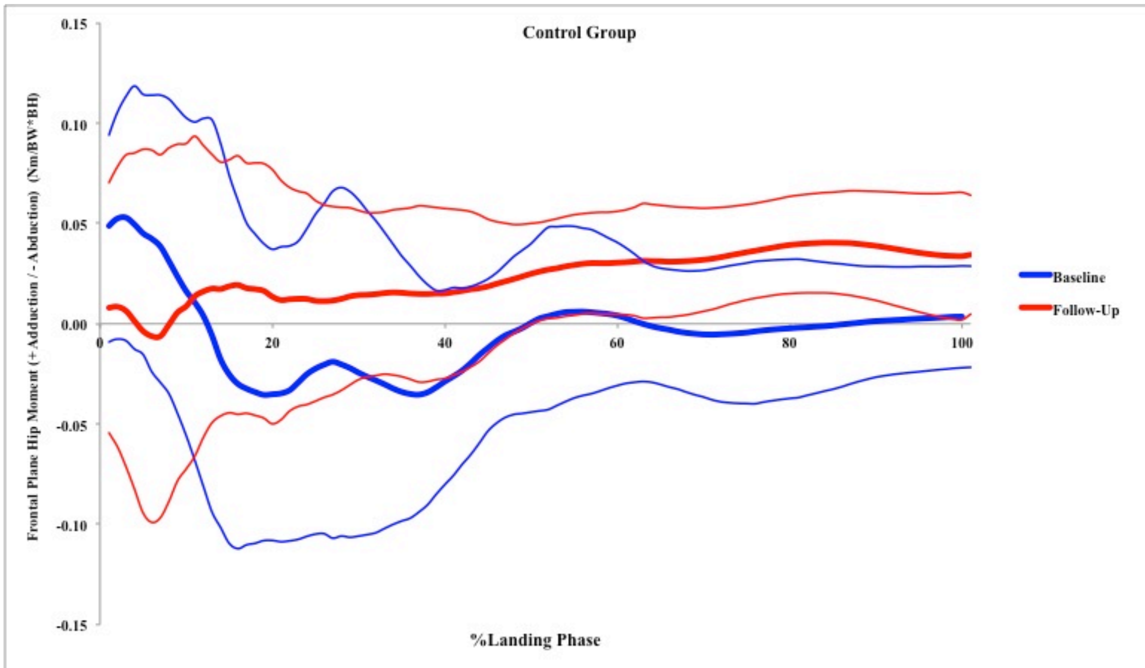

Figure 35. Ensemble average plot of frontal plane hip moment at Baseline and Follow-Up during the Landing Phase for the Control Group.

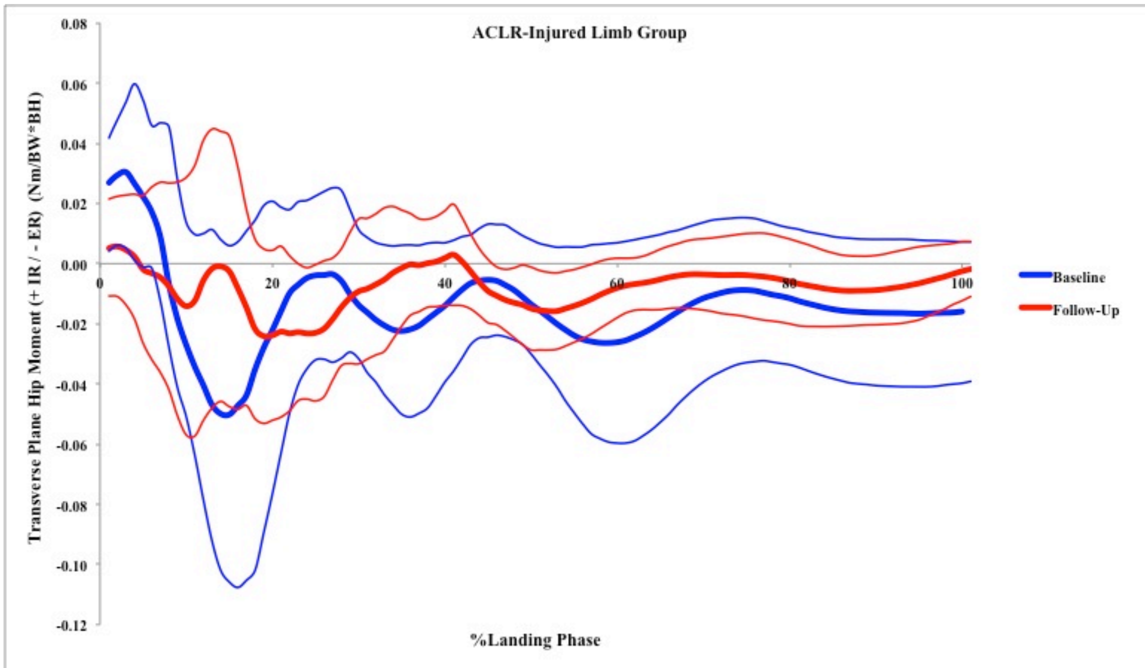

Figure 36. Ensemble average plot of transverse plane hip moment at Baseline and Follow-Up during the Landing Phase for the ACLR-Injured Limb Group.

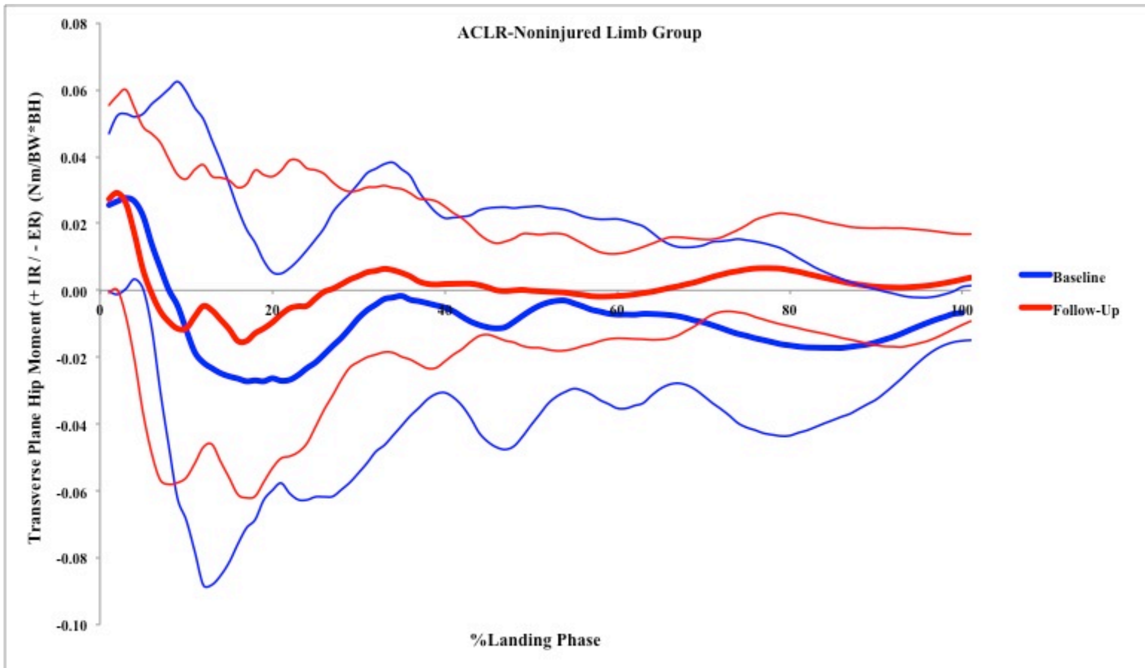

Figure 37. Ensemble average plot of transverse plane hip moment at Baseline and Follow-Up during the Landing Phase for the ACLR-Noninjured Limb Group.

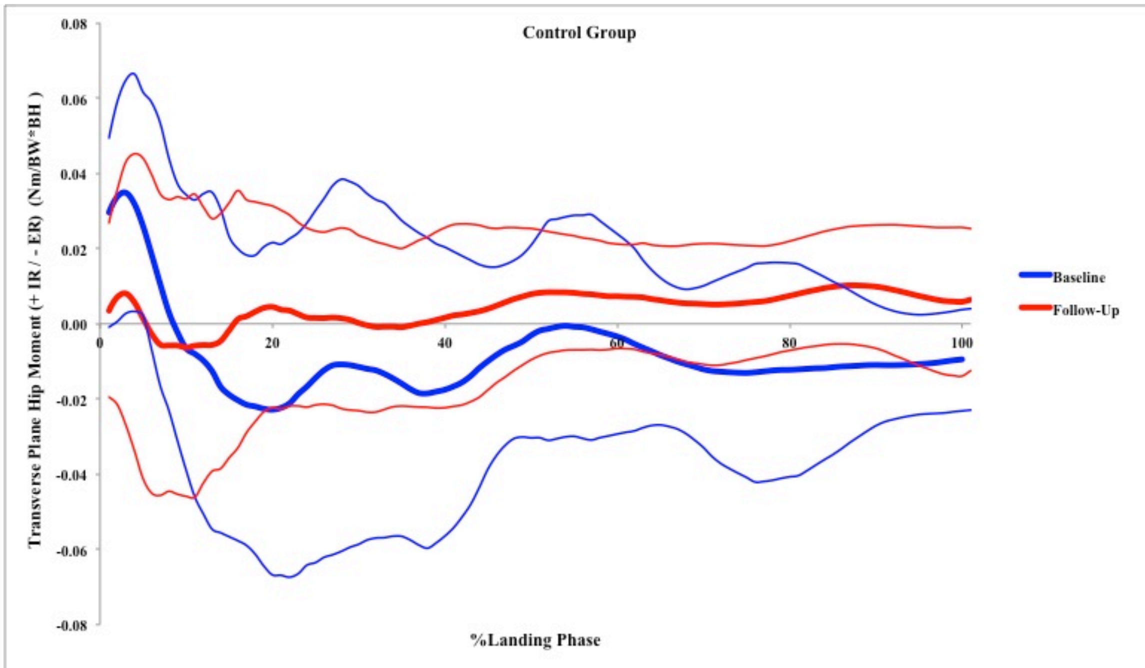

Figure 38. Ensemble average plot of transverse plane hip moment at Baseline and Follow-Up during the Landing Phase for the Control Group.

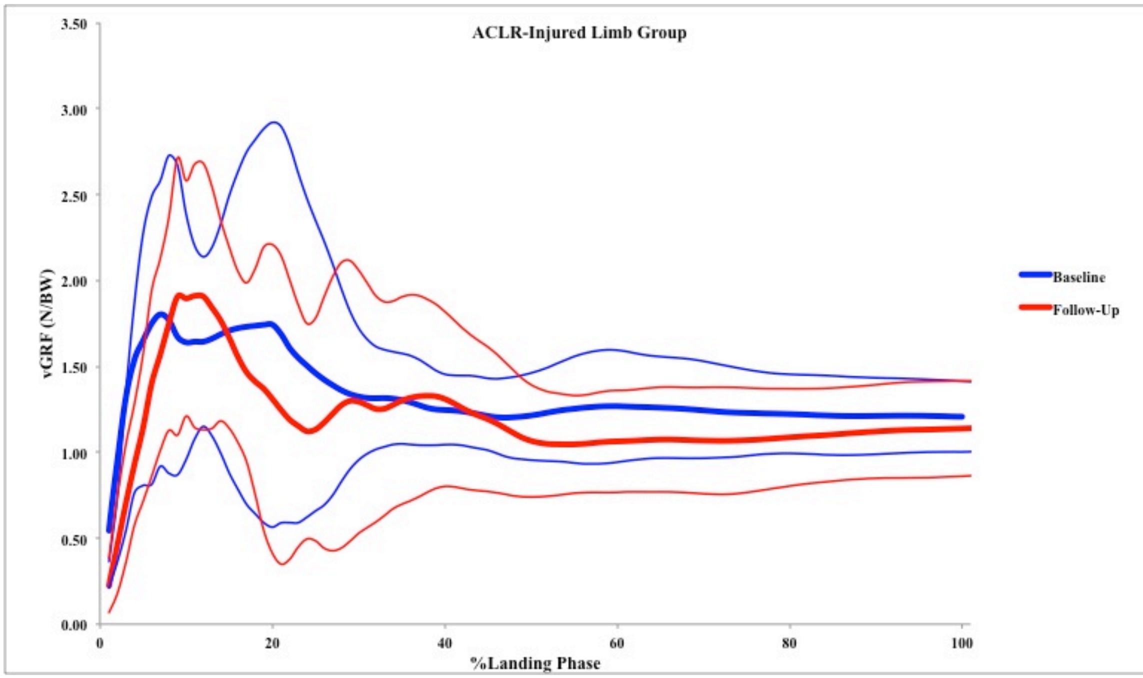

Figure 39. Ensemble average plot of vertical ground reaction force (vGRF) at Baseline and Follow-Up during the Landing Phase for the ACLR-Injured Limb Group.

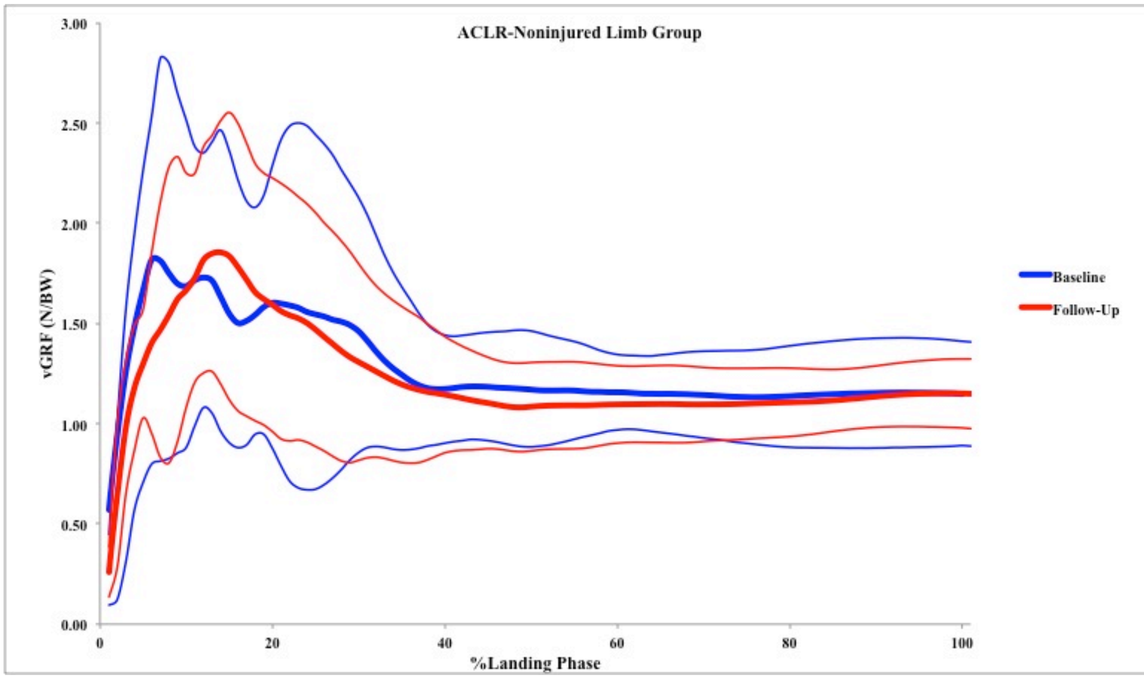

Figure 40. Ensemble average plot of vertical ground reaction force (vGRF) at Baseline and Follow-Up during the Landing Phase for the ACLR-Noninjured Limb Group.

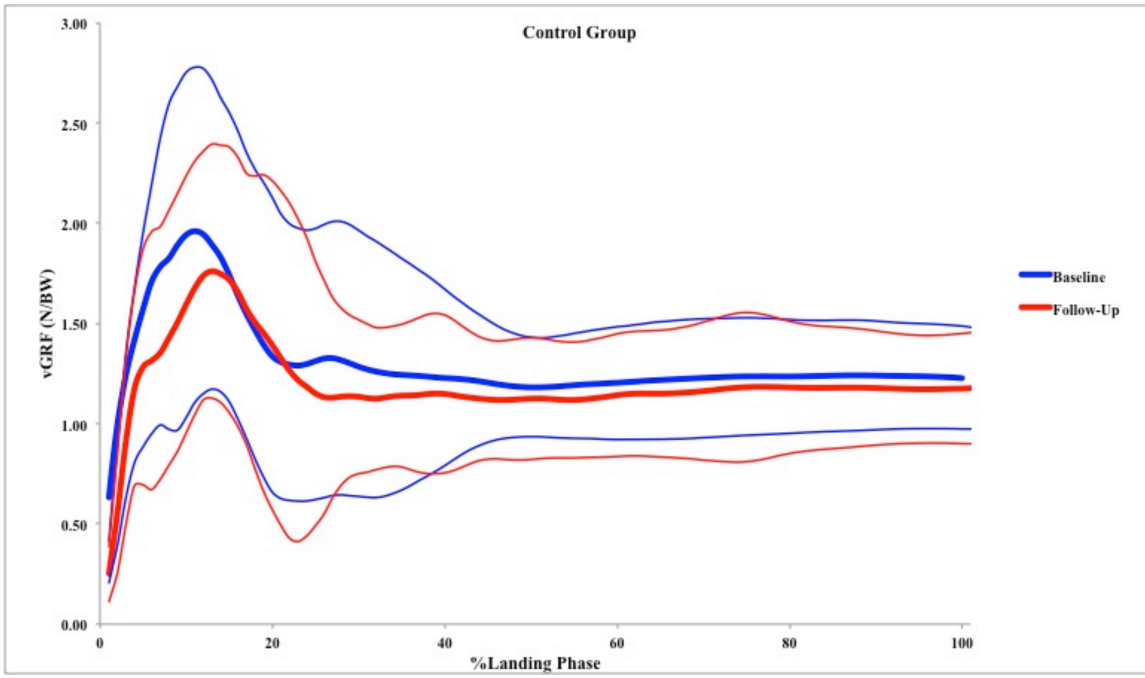

Figure 41. Ensemble average plot of vertical ground reaction force (vGRF) at Baseline and Follow-Up during the Landing Phase for the Control Group.

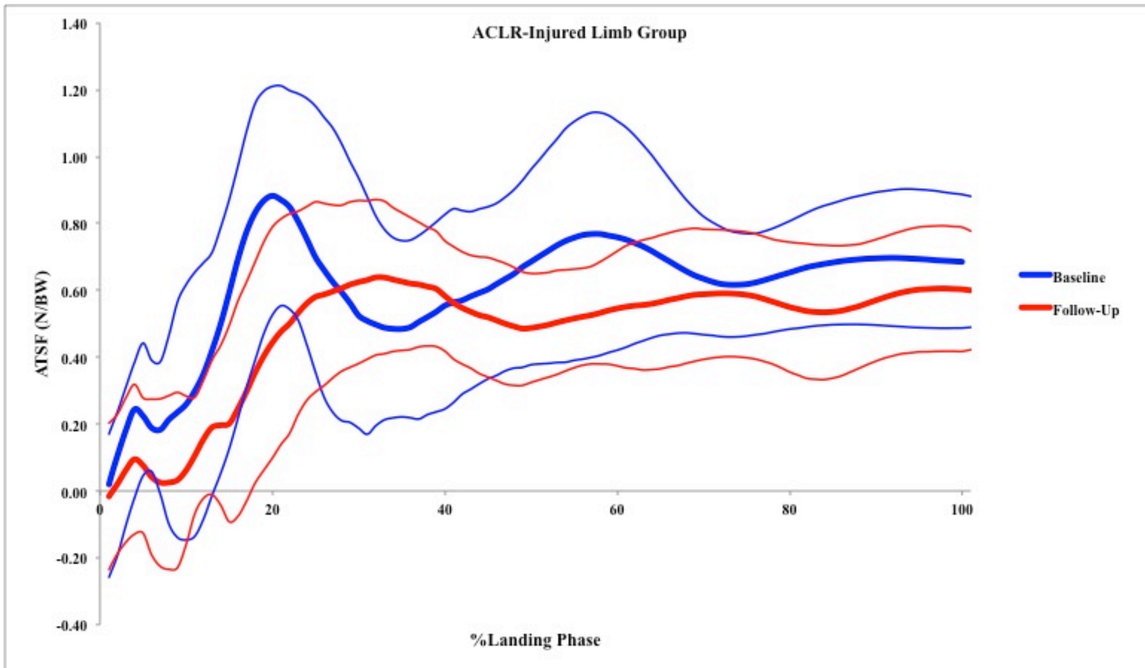

Figure 42. Ensemble average plot of anterior tibial shear force (ATSF) at Baseline and Follow-Up during the Landing Phase for the ACLR-Injured Limb Group.

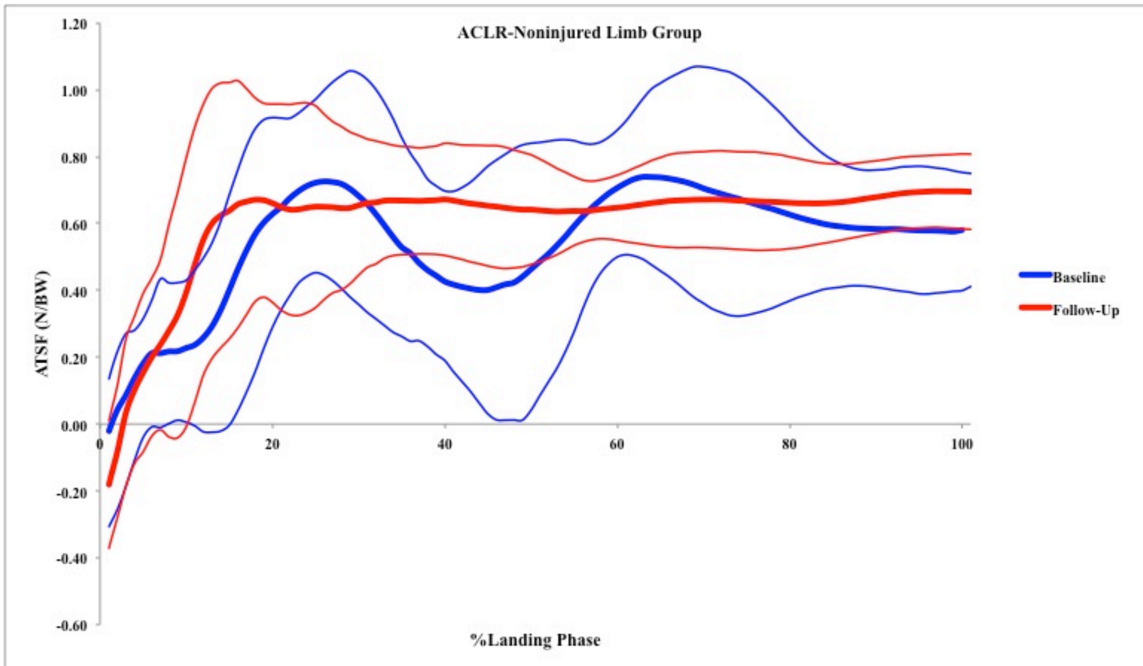

Figure 43. Ensemble average plot of anterior tibial shear force (ATSF) at Baseline and Follow-Up during the Landing Phase for the ACLR-Noninjured Limb Group.

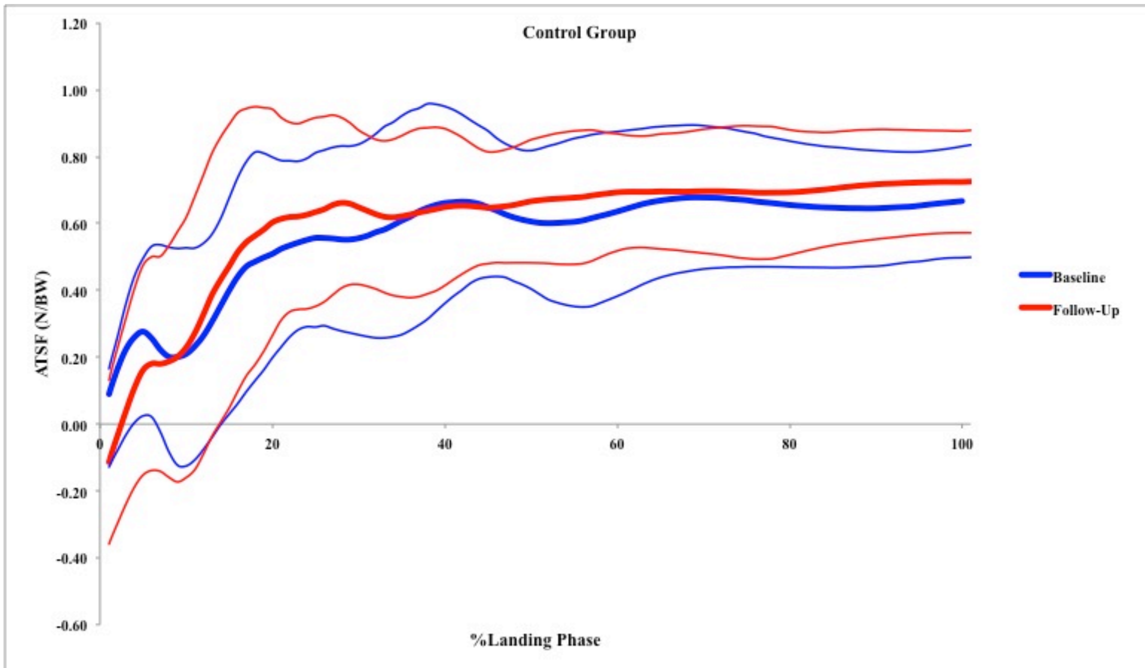

Figure 44. Ensemble average plot of anterior tibial shear force (ATSF) at Baseline and Follow-Up during the Landing Phase for the Control Group.

APPENDIX A. SUBJECT QUESTIONAIRE

ACLID: ______________________ Date:_____/______/_______

Sport Participation: Please list the sport and level at which you participated in while at the USAFA.

	Sport	**Level** (i.e. varisty, IM, club, etc.)
First Year:	____________________	____
Second Year:	____________________	____
Third Year:	____________________	____
Fourth Year:	____________________	____

Activity: Please indicate how often you performed each activity in your healiest and most active state, **in the past year.**

	Less than one time in a month	**One time in a month**	**One time in a week**	**2 or 3 times in a week**	**4 or more times in a week**
M1. Running: running while playing a sport or jogging	O	O	O	O	O
M2. Cutting: changing directions while running	O	O	O	O	O
M3. Decelerating: coming to a quick stop while running	O	O	O	O	O
M4. Pivoting: turning your body with your foot planted while playing a sport; For example: skiing, skating, kicking, throwing, hitting a ball (golf, tennis, squash), etc.	O	O	O	O	O

KOOS KNEE SURVEY

Today's date: _____/_____/_____

ACLID: ______________________________

INSTRUCTIONS: This survey asks for your view about your knee. This information will help us keep track of how you feel about your knee and how well you are able to perform your usual activities.
Answer every question by ticking the appropriate box, only one box for each question. If you are unsure about how to answer a question, please give the best answer you can.

Symptoms
These questions should be answered thinking of your knee symptoms during the **last week**.

S1. Do you have swelling in your knee?

Never	Rarely	Sometimes	Often	Always
☐	☐	☐	☐	☐

S2. Do you feel grinding, hear clicking or any other type of noise when your knee moves?

Never	Rarely	Sometimes	Often	Always
☐	☐	☐	☐	☐

S3. Does your knee catch or hang up when moving?

Never	Rarely	Sometimes	Often	Always
☐	☐	☐	☐	☐

S4. Can you straighten your knee fully?

Always	Often	Sometimes	Rarely	Never
☐	☐	☐	☐	☐

S5. Can you bend your knee fully?

Always	Often	Sometimes	Rarely	Never
☐	☐	☐	☐	☐

Stiffness
The following questions concern the amount of joint stiffness you have experienced during the **last week** in your knee. Stiffness is a sensation of restriction or slowness in the ease with which you move your knee joint.

S6. How severe is your knee joint stiffness after first wakening in the morning?

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

S7. How severe is your knee stiffness after sitting, lying or resting **later in the day**?

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Pain

P1. How often do you experience knee pain?

Never	Monthly	Weekly	Daily	Always
☐	☐	☐	☐	☐

What amount of knee pain have you experienced the **last week** during the following activities?

P2. Twisting/pivoting on your knee

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P3. Straightening knee fully

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P4. Bending knee fully

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P5. Walking on flat surface

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P6. Going up or down stairs

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P7. At night while in bed

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P8. Sitting or lying

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P9. Standing upright

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Function, daily living

The following questions concern your physical function. By this we mean your ability to move around and to look after yourself. For each of the following activities please indicate the degree of difficulty you have experienced in the **last week** due to your knee.

A1. Descending stairs

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A2. Ascending stairs

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

For each of the following activities please indicate the degree of difficulty you have experienced in the **last week** due to your knee.

A3. Rising from sitting

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A4. Standing

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A5. Bending to floor/pick up an object

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A6. Walking on flat surface

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A7. Getting in/out of car

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A8. Going shopping

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A9. Putting on socks/stockings

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A10. Rising from bed

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A11. Taking off socks/stockings

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A12. Lying in bed (turning over, maintaining knee position)

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A13. Getting in/out of bath

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A14. Sitting

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A15. Getting on/off toilet

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

For each of the following activities please indicate the degree of difficulty you have experienced in the **last week** due to your knee.

A16. Heavy domestic duties (moving heavy boxes, scrubbing floors, etc)

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A17. Light domestic duties (cooking, dusting, etc)

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Function, sports and recreational activities

The following questions concern your physical function when being active on a higher level. The questions should be answered thinking of what degree of difficulty you have experienced during the **last week** due to your knee.

SP1. Squatting

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

SP2. Running

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

SP3. Jumping

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

SP4. Twisting/pivoting on your injured knee

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

SP5. Kneeling

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Quality of Life

Q1. How often are you aware of your knee problem?

Never	Monthly	Weekly	Daily	Constantly
☐	☐	☐	☐	☐

Q2. Have you modified your life style to avoid potentially damaging activities to your knee?

Not at all	Mildly	Moderately	Severely	Totally
☐	☐	☐	☐	☐

Q3. How much are you troubled with lack of confidence in your knee?

Not at all	Mildly	Moderately	Severely	Extremely
☐	☐	☐	☐	☐

Q4. In general, how much difficulty do you have with your knee?

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Thank you very much for completing all the questions in this questionnaire.

APPENDIX B. MANUSCRIPT 1

Lower Extremity Biomechanics Do Not Return to Pre-Injury Status Following Anterior Cruciate Ligament Injury: The JUMP ACL Study

(British Medical Journal)

ABSTRACT

Background: Information as to how ACL injury and ACLR alter lower extremity biomechanics may help improve rehabilitation and return to physical activity guidelines to reduce the risk for secondary ACL injury.

Aim: To determine the effect of ACL injury and subsequent ACLR on lower extremity biomechanics of the injured and noninjured limb.

Methods: 70 participants (12 ACLR-Injured Limb, 19 ACLR-Noninjured Limb, 39 Control) who were part of the JUMP ACL study were included in this analysis. Lower extremity biomechanics during a double-leg jump landing were analyzed at two time points (Baseline, Follow-Up) that coincided with times prior to and following ACL injury. Variables of interest included hip and knee kinematics in three planes of motion at Initial Ground Contact, and peak values for each during the Landing Phase.

Results: Knee valgus angle and hip adduction angle at Initial Ground Contact increased for both ACLR groups. They also demonstrated a decrease in peak knee varus angle, and an increase in peak knee valgus and hip adduction during the Landing Phase. The ACLR-Injured Limb group demonstrated a decrease in peak internal knee extension moment and peak anterior tibial shear force during the Landing Phase.

Conclusion: Following ACL injury we observed an increase in medial displacement of the hip and knee during a double leg jump landing. This change in movement pattern may

be the result of quadriceps dysfunction of the injured limb and may increase their risk for noncontact ACL injury.

INTRODUCTION

Anterior cruciate ligament injury and subsequent reconstructive surgery (ACLR) is associated with a decrease in physical activity[1,2] and the onset of osteoarthritis.[3,4] They are also at an increased risk for a subsequent ACL injury as compared to individuals with no history of ACL injury.[5-7] Previous reports put the risk for reinjury at a level ranging from a 5 to 15 fold increased risk in this population[5-7] with the incidence rates ranging from 6-25%.[7-10] While several factors may ultimately contribute to the increased risk for reinjury for those with ACLR; lower extremity biomechanics have been identified as prospective risk factors for reinjury.[11]

Differences in lower extremity biomechanics are altered for those with ACLR. These individuals display differences in the movement and loading of their limbs when compared to those who have never suffered an ACL injury.[12-15] They also display bilateral differences between the injured and uninjured limb[14-18], which is consistent with the observation that the risk for reinjury is nonspecific to the injured limb.[19] This has led to the belief that following ACL injury, individuals adopt movement and loading patterns that predispose them to risk for reinjury, and that these altered patterns are present despite successful treatment and rehabilitation.

Unfortunately, there is limited evidence to support this assertion as no studies have captured data prior to and following ACL injury and subsequent ACLR. The evidence that is available is limited to individual case reports[20,21], which are valuable for the information they provide, but lack an adequate number of observations to draw anything but tentative conclusions. The ability to observe what biomechanical factors are altered following ACL injury, for both the injured and noninjured limb, may provide valuable information about how to improve rehabilitation and ensure a safer return to physical activity for these individuals.

Therefore, the purpose of this study was to determine if and how, lower extremity biomechanics change following ACL injury and subsequent ACLR. Because of previously observed bilateral differences in lower extremity biomechanics following ACL injury, this analysis will include examination of the effect of injury on the injured and noninjured limb.

METHODS

Participants

This study employed a repeated measures, case-cohort research design. All participants were recruited from the Joint Undertaking to Monitor and Prevent ACL Injury (JUMP ACL) Study, a multi-year, multi-site prospective study to identify risk factors for ACL Injury. Participants were enrolled at the United States service academies; United States Air Force Academy, United States Military Academy, and United States Naval Academy, and completed initial biomechanical testing (Baseline) for the JUMP ACL Study during the summer of their enrollment year. Each participant was prospectively followed during their career at the service academy for ACL injuries.

Participants identified for enrollment in this study (Follow-Up), was limited to those with complete Baseline biomechanical data from the 2007 and 2008 cohorts. ACL injured (Cases) were identified as having suffered an ACL injury during their enrollment in the study, had complete biomechanical data at Baseline, and were still currently enrolled in the JUMP ACL study and respective service academy. In addition, for each

Case, three Controls were identified for Follow-Up and were matched based on the sex, cohort year, and service academy of each Case. Based on these criteria, individuals were identified and assigned a random priority number to contact for enrollment in this Follow-Up analysis. To maximize enrollment for Cases, three from the 2009 cohort were identified and completed testing for this study. A schematic depicting enrollment for this study and the 2007 and 2008 cohorts is provided in **Figure 1**.

Thirty-eight Cases, and 50 Controls, were enrolled for Follow-Up testing. Of the 38 Cases, 6 self-reported an ACL injury prior to Baseline data collection, and 3 suffered more than one ACL injury since Baseline. Of the 6 that had a prior ACL injury, two were retained in the data set as they had complete biomechanical data on the noninjured limb at baseline and later suffered an injury to that same limb. Of the Controls, 10 were unable to complete biomechanical testing at Follow-Up due to time constraints and 1 reported a prior ACL injury at Baseline testing. Therefore, 31 Cases and 39 Controls had adequate biomechanical data for both the Baseline and Follow Up testing to be included in this analysis. Baseline testing for the JUMP ACL study only captured unilateral lower extremity biomechanics and not all ACL injuries for the Cases occurred on the tested limb; 12 individuals injured the tested limb, 19 injured the non-tested limb. Therefore, the Cases (n=31) were further sub-divided into two separate groups; ACLR-Injured Limb (n=12) for the participants who injured the limb that biomechanical data were collected on and ACLR-Noninjured Limb (n=19) for participants who injured the limb that data were not collected on. This provided us with three groups for analysis, and allowed us to assess the affects of ACL injury on both the injured and noninjured limbs.

Procedures

Prior to biomechanical testing at Follow-Up each participant read and signed an informed consent and was asked to complete the Marx Activity Scale[22], Knee Injury and Osteoarthritis Outcome Score (KOOS)[23], and have anterior knee laxity assessed bilaterally.[24] At Baseline and Follow-Up, each participant performed a double leg jump landing maneuver (**Figure 2**).[25,26] For this task participants were required to stand atop a 30cm jump box located a distance from the front edge of the force plate equal to half of their body height. They then jumped forward towards the force plate, landed, and immediately performed a maximum effort vertical jump. To be considered a successful trial the participant had to land with only the foot of the test leg making contact with the force plate, the foot of the test leg being completely on the force plate, and perform the task with no hesitation between the landing and jumping phase of the task.

Biomechanical data were collected during the task using an electomagentic tracking system (Ascension Technologies Inc., Burlington, VT) integrated with a non-conductive force plate (Bertec Co., Columbus, OH). Prior to data collection electromagnetic sensors were attached to the shank and thigh of the participant's test limb, as well their pelvis. The position of the medial and lateral malleoli, medial and lateral femoral epicondyles, and the anterior superior iliac spines relative to the segment sensors was recorded using a movable sensor. The ankle joint center and knee joint centers were estimated as the midpoint between the malleoli and femoral epicondyles, respectively, and the hip joint center was estimated based on the location of the anterior superior iliac spines according to the Bell Method.[27] A model of the shank, thigh, and pelvis was determined based on these points, with the shank segment defined by the two endpoints of the ankle joint and knee joint centers and the shank sensor, the thigh segment defined by the two endpoints of the knee joint center and hip joint center and the thigh sensor, and the pelvis as the anterior superior iliac spines and the pelvis sensor. Local right-handed axis systems were embedded in each segment, and the orientation of each coincided with the global axis system. The global axis system was established and defined prior to data collection and coincided with the positive x-axis oriented with the direction the participant faced during the double leg jump landing, the positive z-axis defined as a vector oriented along the true vertical, and the y-axis defined as a vector located at a positive 90° rotation about the z-axis relative to the positive x-axis.[25,26]

Data analysis

All kinematic data were collected at a sampling frequency of 144 Hz and kinetic data were sampled at a sampling frequency of 1,444 Hz. Kinematic data were filtered using a 4th order Butterworth filter (14.5 Hz) and all biomechanical data were exported using the Motion Monitor Software (Innovative Sports Training, Inc., Chicago, IL). Prior to exportation, the kinematic data of the hip and knee joint in all three planes of motion was defined using an Euler sequence Y,X,Z; such that the first rotation was defined about the y-axis, second rotation about the x-axis, and third rotation about the z-axis. Internal joint moments in all three planes of motion at the hip and knee were calculated using inverse dynamics within the software. All moments were normalized to by the product of body height (m) and body weight (N), and reported here as normalized internal joint moments.

The vertical ground reaction force and anterior tibial shear force data were normalized to body weight (N), and reported as such.

Following data exportation, the data were reduced to calculate all dependent variables during the time points of interest using a customized MATLAB (Mathworks, Inc., Natick, MA) program. These variables included hip and knee biomechanics (kinematics and moments) in all three planes of motion, as well as the vertical ground reaction force and anterior tibial shear force. Values for each variable were recorded at the time of Initial Ground Contact, defined as the first time point at which the vertical ground reaction force exceeded 10 N, and peak values for each variable in each direction during the Landing Phase of the double leg jump landing, defined as the time from Initial Ground Contact to peak knee flexion was achieved.

Before statistical analyses were performed each variable was assessed for normality. Z-scores for the skewness and kurtosis of each dependent variable was calculated, and variables with values greater than 1.96 were identified for further analysis to identify statistical outliers within the data. These procedures consisted of plotting the standardized residuals and generating box plots to identify outliers. When outliers were identified the statistical analysis to compare the specific variable was computed with and without the outliers included in the data set. If removal of the outliers did not create any change in the significance of the analysis, the data were retained in the final analysis. When removal resulted in a change in the significance of the analysis the data for the individual trials were further analyzed. This procedure consisted of analyzing the values for each specific trial, and if one or more values appeared out of normal range the biomechanical data were analyzed for data collection errors. Any trials that exhibited collection errors were removed from the computed within-participant average, and was performed for all dependent variables, so as to retain as many participants and information as possible. This resulted in some participants having fewer than three trials used to calculate between trial average for the dependent variables in this analysis.

Statistical analyses were performed to determine changes in lower extremity biomechanics following ACL injury. These analyses consisted of 3x2 (Group:ACLR-Injured Limb, ACLR-Noninjured Limb, Control x Time: Baseline, Follow-Up) mixed model analysis of covariance to adjust for sex, for each dependent variable. Post hoc analyses consisted of Tukey's HSD, and were performed for any significant interaction effect. An alpha level of 0.05 was set a priori to determine statistical significance for all analyses. All statistical analyses were performed using IBM SPSS v19 (SPSS, Inc., an IBM company, Chicago, IL).

RESULTS

Participant Demographics

Participant demographics and anthropometrics for Baseline and Follow-Up testing sessions are summarized in **Table 1**. Group chronological data for testing sessions, ACL injury, and surgery are presented in **Table 2**. Graft type was not obtained for 5 members of the ACLR-Injured Limb group, and 6 of the ACLR-Noninjured Limb group. For the ACLR-Injured Limb group 3 had a bone-patella tendon-bone autograft and 4 had a hamstrings autograft. For the ACLR-Noninjured Limb group 5 had a bone-patella tendon-bone graft, 7 had a hamstring autograft, and one had an Achilles tendon allograft.

Descriptive statistics for anterior knee laxity, Marx Activity Scale, and KOOS are provided in **Table 3**.

The normality and outlier procedures identified 9 dependent variables changed in the significance of the analysis when outliers were removed. For the total 9 dependent variables, 27 participants were identified as contributing values considered as outliers. Nine participants had data collection errors in one or more of the three trials used to calculate within-participant averages. All had at least one trial with no errors, and all participants were retained for the final analysis.

All descriptive statistics for each dependent variable and each group are provided in **Tables 4-8**.

Initial Ground Contact

Kinematics

We observed a significant Time x Group interaction for frontal plane knee ($F_{(2,66)}$ = 3.957, $p = 0.024$) and hip ($F_{(2,66)}$ = 3.773, $p = 0.028$) angles at Initial Ground Contact. Post hoc analysis indicated that there was no significant difference in knee angle in the frontal plane among the groups at Baseline or Follow-Up. Following ACL injury, both the ACLR–Injured Limb and ACLR-Noninjured Limb groups had a significant increase in knee valgus angle at Initial Ground Contact compared to Baseline. However, there was no change in knee valgus angle for the Control group between Baseline and Follow-Up.

For the frontal plane hip angle, post hoc analysis indicated a similar pattern of change. There was no difference among groups at Baseline, but both the ACLR-Injured Limb and ACLR-Noninjured Limb groups significantly increased hip adduction at Initial

Ground Contact from Baseline to Follow-Up. There was no difference in the hip adduction angle between the ACLR groups, but both ACLR groups displayed significantly greater hip adduction compared to the Control group at Follow-Up.

We observed no other significant interaction effects for kinematics at Initial Ground Contact: sagittal plane knee angle ($F_{(2,66)} = 0.104$, $p = 0.901$), transverse plane knee angle ($F_{(2,66)} = 0.135$, $p = 0.874$), sagittal plane hip angle ($F_{(2,66)} = 2.023$, $p = 0.140$), and transverse plane hip angle ($F_{(2,66)} = 0.089$, $p = 0.915$). A significant time effect was present for transverse plane hip angle ($F_{(1,66)} = 4.731$, $p = 0.033$) indicating an increase in hip external rotation from Baseline to Follow-Up regardless of group.

Kinetics

We observed a significant Time x Group interaction for transverse plane knee moment ($F_{(2,66)} = 3.373$, $p = 0.040$), sagittal plane hip moment ($F_{(2,66)} = 4.266$, $p = 0.018$), and transverse plane hip moment ($F_{(2,66)} = 3.226$, $p = 0.046$) at Initial Ground Contact. Post hoc analysis for transverse plane knee moment indicated no significant difference among or between the groups at Baseline or Follow-Up. However, significant differences were present for sagittal plane hip moment and transverse plane hip moment following post hoc analysis. There were no differences among groups for sagittal plane hip moment at Baseline, but the Control group demonstrated a significant increase in hip extension moment from Baseline to Follow-Up. Also, at Follow-Up the ACLR groups were significantly different with the ACLR-Injured Limb group demonstrating an internal hip flexion moment and the ACLR-Noninjured Limb group demonstrating an internal hip extension moment at Initial Ground Contact. At Baseline there were no significant differences among groups for transverse plane hip moment, but both the ACLR-Injured Limb and Control groups had a significant decrease of internal hip internal rotation moment from Baseline to Follow-Up. At Follow-Up, both groups demonstrated significantly less internal hip internal rotation moment than the ACLR-Noninjured Limb group.

No other significant interaction effects were observed for kinetic variables at Initial Ground Contact: sagittal plane knee moment ($F_{(2,66)} = 2.458$, $p = 0.093$), frontal plane knee moment ($F_{(2,66)} = 0.476$, $p = 0.623$), frontal plane hip moment ($F_{(2,66)} = 0.506$, $p = 0.605$), anterior tibial shear force ($F_{(2,66)} = 2.011$, $p = 0.142$), and vertical ground reaction force ($F_{(2,66)} = 0.238$, $p = 0.789$). A significant group main effect was observed for frontal plane hip moment ($F_{(1,66)} = 3.178$, $p = 0.048$) with post hoc analysis indicating the ACLR-Noninjured Limb group had significantly higher internal hip adduction moment than the Control group regardless of time. There were no group differences between the ACLR groups or the ACLR-Injured Limb and Control groups.

Time main effects for frontal plane knee moment ($F_{(1,66)} = 16.802$, $p < 0.001$), frontal plane hip moment ($F_{(1,66)} = 11.684$, $p = 0.001$), and vertical ground reaction force ($F_{(1,66)} = 6.401$, $p = 0.014$) were present. There was a decrease in internal knee varus moment, decrease in internal hip adduction moment, and decrease in vertical ground reaction force at Initial Ground Contact from Baseline to Follow-Up regardless of group.

Landing Phase

Kinematics

Statistical analysis of the peak kinematic values indicated significant Time x Group interactions for peak knee varus angle ($F_{(2, 66)} = 5.198$, $p = 0.008$), peak knee valgus angle ($F_{(2,66)} = 3.768$, $p = 0.028$), and peak knee internal rotation angle ($F_{(2,66)} = 4.204$, $p = 0.019$). Post hoc analyses demonstrated no difference in peak knee varus angle among groups at Baseline or Follow-Up. However, both the ACLR-Injured Limb and ACLR-Noninjured Limb groups demonstrated a significant decrease in peak knee varus angle over time. There were similar findings for peak knee valgus angle, as there was no difference among groups at Baseline or Follow-Up, but the ACLR-Noninjured Limb group demonstrated a significant increase in peak knee valgus angle from Baseline to Follow-Up. There was a similar increase in peak knee valgus angle for the ACLR-Injured Limb group but post hoc analysis indicated it was not significant. No group differences for peak knee internal rotation angle during the Landing Phase were observed at Baseline, however the Control group demonstrated a significant increase in peak knee internal rotation angle from Baseline to Follow-Up. Peak knee internal rotation angle was also greater than the ACLR-Noninjured Limb group at Follow-Up. No significant change in knee rotation angle was observed for the ACLR groups following ACL injury.

No other significant interactions were observed for peak kinematic variables during the Landing phase: knee flexion ($F_{(2,66)} = 0.944$, $p = 0.394$), knee extension ($F_{(2,66)} = 0.349$, $p = 0.706$), knee external rotation ($F_{(2,66)} = 0.412$, $p = 0.664$), hip flexion ($F_{(2,66)} = 1.190$, $p = 0.311$), hip extension ($F_{(2,66)} = 2.488$, $p = 0.091$), hip adduction ($F_{(2,66)} = 2.745$, $p = 0.072$), hip abduction ($F_{(2,66)} = 2.256$, $p = 0.113$), hip internal rotation ($F_{(2,66)} = 2.576$,

$p = 0.084$), and hip external rotation ($F_{(2,66)} = 2.205$, $p = 0.118$). Time main effects were observed for peak knee flexion ($F_{(1,66)} = 25.168$, $p < 0.001$), hip flexion ($F_{(1,66)} = 25.326$, $p < 0.001$), hip internal rotation ($F_{(1,66)} = 5.263$, $p = 0.025$), and hip external rotation ($F_{(1,66)} = 3.986$, $p = 0.050$). There was an increase in peak knee flexion, increase in peak hip flexion, decrease in peak hip internal rotation, and increase in hip external rotation from Baseline to Follow-Up regardless of group. No group main effects were observed.

Kinetics

Significant Time x Group interactions for peak knee extension moment ($F_{(2,66)} = 4.509$, $p = 0.015$), peak hip flexion moment ($F_{(2,66)} = 3.847$, $p = 0.026$) and peak anterior tibial shear force ($F_{(2,66)} = 4.530$, $p = 0.014$) during the Landing phase were observed. Post hoc analyses revealed that these interaction effects were the result of changes from Baseline to Follow-Up for the ACLR-Injured Limb group only. There was no significant difference among groups at Baseline for any of the variables, but the ACLR-Injured Limb group demonstrated a significant decrease in peak internal knee extension moment, peak internal hip flexion moment, and peak anterior tibial shear force from Baseline to Follow-Up. This change resulted in the ACLR-Injured Limb group having lower values for each variable as compared to the ACLR-Noninjured Limb group at Follow-Up.

No other significant interactions were present for peak kinetic variables during the Landing Phase: knee flexion moment ($F_{(2,66)} = 2.003$, $p = 0.143$), knee valgus moment ($F_{(2,66)} = 0.595$, $p = 0.555$), knee varus moment ($F_{(2,66)} = 1.255$, $p = 0.292$), knee external rotation moment ($F_{(2,66)} = 1.632$, $p = 0.203$), knee internal rotation moment ($F_{(2,66)} = 0.769$, $p = 0.467$), hip extension moment ($F_{(2,66)} = 1.076$, $p = 0.347$), hip abduction moment ($F_{(2,66)} = 1.371$, $p = 0.261$), hip adduction moment ($F_{(2,66)} = 2.193$, $p = 0.120$), hip external rotation moment ($F_{(2,66)} = 0.699$, $p = 0.501$), hip internal rotation moment ($F_{(2,66)} = 1.527$, $p = 0.225$), posterior tibial shear force ($F_{(2,66)} = 0.348$, $p = 0.708$), and vertical ground reaction force ($F_{(2,66)} = 0.630$, $p = 0.536$). Group main effects for peak knee flexion moment ($F_{(1,66)} = 3.508$, $p = 0.036$) and peak knee valgus moment were found ($F_{(1,66)} = 3.501$, $p = 0.036$), but post hoc analysis did not indicate significant group differences.

In addition, time main effects for peak knee valgus moment ($F_{(1,66)} = 25.659$, $p < 0.001$), peak hip abduction moment ($F_{(1,66)} = 5.723$, $p = 0.020$), and peak hip external rotation moment ($F_{(1,66)} = 6.804$, $p = 0.011$), indicating a significant decrease in each variable from Baseline to Follow-Up, regardless of group.

DISCUSSION

The primary finding of this study was the observation that ACL injury and ACLR cause specific changes in movement and loading of both the injured and noninjured limb that may help to explain the increased risk for reinjury associated with this population. Specifically, we observed increased medial displacement of the knee and hip for both the ACLR-Injured Limb group and ACLR-Noninjured Limb group. These alterations were observed in relation to very little change in lower extremity biomechanics for the Control group. This may indicate that ACL injury changes what would otherwise be relatively stable lower extremity movement and loading patterns. An explanation for these results will be provided in the following paragraphs.

We observed a number of changes among the groups following ACL injury and subsequent ACLR. At Initial Ground Contact we observed a difference in sagittal plane moment at the hip, with the ACLR groups being different at Follow-Up. This was the result of an increase in internal hip extension moment for the ACLR-Noninjured Limb group and a decrease in internal hip extension moment for the ACLR-Injured Limb group. However, we also observed an increase in internal hip extension moment for the Control group from Baseline and Follow-Up. Neither ACLR groups were significantly different from the Control group at Follow-Up for internal hip extension moment. We also observed a decrease in internal hip internal rotation moment for both the ACLR-Injured Limb group and Control group, with no change for the ACLR-Noninjured Limb group. In addition, both demonstrated less internal hip internal rotation moment at Initial Ground Contact than the ACLR-Noninjured Limb group at Follow-Up. These findings may be supported in part by the previous observations of Decker et al[13], as they noted decreased energy absorption at the hip as compared to the knee and ankle for the injured limb of those with ACLR. They did not observe such differences for their sample of healthy control subjects, but did not examine the noninjured limb of the ACLR group.[13] Our results compliment these findings, as we observed differences in loading at the hip for those with ACLR, in which the injured limb tended to be unloaded at Initial Ground Contact, but the noninjured limb maintained or increased loading. The significance of these findings should be interpreted with caution, as we made these observations at a point in the jump landing where there is minimal loading occurring on the lower extremity.

We also observed an increase in peak knee internal rotation angle for the Control group, but no change for either ACLR groups. This is of important note as knee internal rotation is thought to be altered following ACL injury and ACLR.[17,28-31] In addition, alterations in knee internal rotation have been proposed as a mechanism for the progression of knee osteoarthritis following ACLR.[32] Our findings suggest that peak knee internal rotation was restored following ACLR, and only the Control group had a significant change. We though, only assessed peak values during a task that generally does not require as much transverse plane motion as other tasks previously described. Therefore, our task may not be indicative of more common tasks that are performed repeatedly throughout the course of a day such as walking and turning, which may more greatly contribute to articular cartilage loading and the progression of osteoarthritis.

We believe that the most important finding of this study is the increased medial displacement of the knee and hip for those with ACLR as this may provide evidence for the increased risk for ACL reinjury in this population. As noted previously we observed an increase in hip adduction and knee valgus at Initial Ground Contact for both the ACLR-Injured Limb group and ACLR-Noninjured Limb groups. We also observed a decrease in peak knee varus angle for both groups, and an increase in peak knee valgus angle for the ACLR-Noninjred Limb group. We did not observe a significant change for peak knee valgus for the ACLR-Injured Limb, but their magnitude of increase was very similar to the ACLR-Noninjured Limb group. This observation is in agreement with Delahunt et al[33], as they observed greater hip adduction and decreased knee varus motion for a group of females with ACLR, as compared to a healthy control group. In addition, we believe that these alterations in movement may be explained by a continued avoidance of loading the reconstructed ACL, as the ACLR-Injured limb group demonstrated a significant decrease in anterior tibial shear force, even though they had returned to full activity and were on average more than 20 months months post surgery.

An anteriorly directed force on the tibia primarily loads the ACL.[34,35] As the anterior tibial shear force is meant to represent this force; we may interpret the observed decrease in this variable for the ACLR-Injured Limb group as a decrease in loading of the reconstructed ACL following injury. This effect was limited to the injured limb of those with ACL injury, as we observed no significant change in the ACLR-Noninjured Limb group or the Control group over time. Sell et al[36] identified significant predictors of anterior tibial shear force during a double leg stop jump in a sample of healthy individuals. The variables included peak posterior ground reaction force, normalized external knee flexion/internal knee extension moment, knee flexion angle, quadriceps activation, and gender; increases in these variables as well as being female were associated with an increased anterior tibial shear force.[36] These findings are important, as they can be used to better describe the factors responsible for the decrease in anterior tibial shear force associated with the ACLR-Injured limb group.

Of the variables identified by Sell et al[36], the decrease in anterior tibial shear force we observed for the ACLR-Injured Limb group is likely the result of the decreased internal knee extension moment we observed for this group as well. We have come to this conclusion as we observed no significant difference among groups for peak knee flexion angle, and we attempted to control for gender differences in our statistical model. In addition, we performed a post hoc analysis of the peak posterior ground reaction force and found no difference among groups either ($F_{(2,66)} = 0.210$, $p = 0.811$). Therefore, of the factors predictive of anterior tibial shear force, we can acknowledge that internal knee extension moment and quadriceps activation may have resulted in the alterations we observed.

Decreased internal knee extension moment has been previously observed in those with ACLR.[12,18,37,38] As internal knee extension moment is primarily generated by the quadriceps muscles to balance an external knee flexion moment during landing, this decrease may be an active strategy to reduce loading of the reconstructed ACL or the result of reduced capacity of the quadriceps muscles. Internal knee extension moment can be altered by landing strategy, as Blackburn and Padua[39] demonstrated that landing with greater trunk flexion reduced internal knee extension moment in a sample of healthy individuals during a single leg drop landing. This was observed in conjunction with a decrease in quadriceps activity and peak vertical ground reaction force.[39] We did not observe any difference among groups for peak vertical ground reaction force, and believe, based on the definition of our biomechanical model, any differences in trunk flexion would have been observed as an increase in peak hip flexion, which we did not observe. Therefore, we believe that the decrease in internal knee extension moment is not the result of this specific technique change, and is likely due to deficits in the capacity of the quadriceps. Berchuck et al[40] first described an avoidance in the use of the quadriceps in a group of ACL deficient patients as an attempt minimize anterior loading of the knee. Deficits in the ability to activate the quadriceps has been previously observed for those with ACLR[41] and quadriceps strength has been observed to be important for the restoration of normal biomechanics following ACLR.[42] Because of the limitations of our data collection, we are unable to determine if our observations are the result of either of these. However, we did observe an accompanying decrease in peak internal hip flexion moment for the ACLR-Injured Limb group. The rectus femoris is a member of the quadriceps group that is responsible for both knee extension and hip flexion. It may be reasonable to assume that decreased capacity of the quadriceps, including the rectus femoris, would produce the observed deficits in internal knee extension moment and hip flexion moment.

Whatever the cause for decreased internal knee extension moment, it has important implications that can help to explain our observed alterations in frontal plane knee kinematics for those with ACLR. The quadriceps and hamstrings muscle groups and their resultant knee extension and flexion moments have been demonstrated to be able to resist frontal plane loading of the knee.[43-46] In particular, co-contraction between the quadriceps and hamstrings is able to resist varus and valgus moments acting at the knee, and frontal plane loading is decrease when these muscles generate greater knee extension and flexion moments.[44,45] Given these findings in reference to our observations, a decrease in knee extension moment for the ACLR-Injured Limb group could compromise their ability to resist frontal plane loading at the knee. As we did not observe any alteration in frontal plane loading or peak vertical ground reaction force for this group, we can assume that the resultant change in frontal plane motion, decreased peak knee varus angle and a trend towards increased peak knee valgus angle, was the result of a compromised ability to resist frontal plane moments induced by a decrease in knee extension moment. The effect of quadriceps dysfunction on frontal plane motion has been previously observed in another population at risk for ACL injury.[47] Palmieri-Smith et al[47] observed a significant association between decreased preparatory activation of the quadriceps and an increased peak knee valgus angle in females during landing. We did not observe a significant increase in peak knee valgus angle for the ACLR-injured limb group, but our mean values following injury were greater than those reported by the authors.[47] Though we did not observe differences in knee extension moment at Initial Ground Contact, we believe this same mechanism may have been responsible for the increase in knee valgus angle at Initial Ground Contact as well. As we mentioned previously, the reported moments at Initial Ground Contact should be interpreted with caution as there is minimal loading acting on the lower extremity.

There was also an increase in hip adduction at Initial Ground Contact for the ACLR-Injured Limb group. This alteration may be of importance as well, as it could represent an attempt to use frontal plane muscles to compensate for the decreased ability of the quadriceps to produce knee extension moment. Pollard et al[48] made a similar conclusion when they observed increased peak knee valgus in a group of healthy females who landed with less knee and hip flexion. The less flexed group also demonstrated greater knee extension moment and greater knee adduction moment though.[48] Our observed differences in frontal plane motion both at the hip and knee were present without changes in frontal plane moments, and we can only propose this a tentative theory.

One component of our results that is not encompassed in this explanation is the observed alterations in frontal plane kinematics for the ACLR-Noninjured limb group. It is possible that the increase in frontal plane motion at the knee and hip for the ACLR-Noninjured Limb group is the result of an attempt to match motion between limbs. This notion is tempered by the fact that data collected for both ACLR groups consists of two separate groups of individuals; therefore, the observed movement patterns may be unique to each sample. However, the fact that the kinematic patterns of increased medial displacement of the hip and knee at Initial Ground Contact for both the ACLR-Injured Limb and ACLR-Noninjured Limb groups, as well as decreased peak knee varus suggests some underlying cause that is driving similar changes in both groups. The ACLR groups did not differ in any kinematic variable, other than peak knee valgus angle, however our ability to observe changes in peak knee valgus angle in the ACLR-Injured Limb group was likely the result of being under powered as a post hoc analysis indicated we would have needed to include 14 more participants for the ACLR-Injured Limb group to be properly powered (0.80) to detect significant differences.

This concept of matching between limbs is supported by previous findings related to those with ACLR. In their analysis of variability between the shank and thigh during gait, Moraiti et al[49] observed alterations in both the injured and noninjured limbs of a sample of persons with ACLR. The authors explained these bilateral alterations as an attempt to maintain symmetry in movement and variability between limbs for those with ACLR. To apply this explanation of our findings, if deficits in the ability to produce internal knee extension moment result in alterations in frontal plane motion of injured limb, then to maintain symmetry the noninjured limb may do the same despite a no deficit in quadriceps capacity of the noninjured limb. This attempt may be supported by our observation of a group main effect for internal hip adduction and internal rotation moment at Initial Ground Contact for the ACLR-Noninjured Limb group as increases in these internal moments would likely result in increased hip adduction and possibly knee valgus, as they are attempting to pull the femur in towards midline with the foot planted.

Ultimately, we believe that these findings have very important clinical implications. This is the first study, other than case reports[20,21], that has described measures prior to and following ACL injury and subsequent ACLR. Most importantly we have observed that not all biomechanical factors return to normal following ACLR, and that alterations in key parameters may increase the risk for subsequent ACL injury. Increased medial knee displacement of the knee has been previously identified as a prospective risk factor for noncontact ACL injury for both healthy individuals and individuals with ACLR.[11,50] This is important, as individuals who return to physical activity may be doing so with poor movement patterns that are placing them at risk for further injury. Our findings though, provide insight into how rehabilitation and intervention efforts could be improved to minimize reinjury risk. First, our findings indicate that rehabilitation efforts should not be focused solely on the injured limb, as the noninjured limb displays alterations in movement following injury. Because alterations occur in both limbs, it may not be appropriate to solely focus on symmetrical movement between limbs, but rather overall quality of motion for both limbs. Second, we believe that promoting efforts to regain proper loading of the injured limb after proper graft healing may help mitigate the development of movement patterns that may increase the risk for secondary injury. Future research will be needed to confirm this notion. Also, we believe that our findings provide strong evidence for the need of serial assessment of those at risk for ACL injury and those with ACLR. Many of the alterations we observed for the ACLR groups would not have been found if they were just compared to the Control group post-injury only. Our sample of participants with ACLR were often several months out from surgery, and had returned to full physical activity but still displayed altered movement patterns. These two factors highlight the need for serial testing prior to and following ACL injury and ACLR, and may indicate that those with ACLR need to engage in intermittent programs to address faulty movement patterns, as the retention of movement patterns is dependent on the duration of the intervention.[51,52] Last, we believe that our findings support the proposition that those with ACLR should be considered as an at risk population for ACL injury. We observed consistent alterations in movement and loading patterns, despite a relatively heterogenous sample related to surgical graft type, and gender. Therefore, it appears that once an individual suffers an ACL injury, this factor may trump any others thought to predispose these individuals to subsequent injury.

Our study is not without limitations. The limitations of our study design prevent us from definitive conclusion that the changes we observed for the ACLR groups occurred following ACL injury. There were periods of time both between Baseline and ACL injury, as well time between the return to physical activity and Follow-Up. There is a possibility that the changes we observed occurred prior to ACL injury. The lack of change in biomechanics of the Control group, however, strongly supports that the changes we observed with this study are the result of ACL injury and ACLR only. The consistency of our findings among a relatively heterogenous group of ACLRs also strengthens the evidence that these changes are due to injury. In addition, as the ability to collect repeated measures on individuals prior to and following injury is quite valuable and quite rare, we did not control for the mechanism of injury, type of surgery, or rehabilitation program. This was done intentionally in hopes of being able to include as many participants as possible without negatively affecting the internal validity of this study. The heterogeneity of our study though does increase the ability to apply our findings to larger populations.

Data collection methods also imposed some limitations for this current study. Because biomechanical data were limited to the knee and hip, we cannot comment on the potential for changes at the ankle and trunk that may help provide more detailed information regarding our findings. In addition, the collection of unilateral biomechanical data of the original JUMP ACL study at Baseline prevented us from being able to capture data on the injured limb prior to injury for all participants. Therefore, the information we present for the injured and noninjured limb is from two separate groups. The conclusions we have drawn from this study are based on the assumption that the patterns observed for the injured and noninjured limb are representative of the population of those with ACLR.

To conclude, our observations indicate that following ACL injury, physically active individuals with ACLR demonstrate an increase in frontal plane displacement of the hip and knee during a double leg jump landing. This increase in medial displacement is present in both the injured and noninjured limb. In addition, we believe these movement patterns are the result of decreased knee extension moment of the injured limb. These alterations in movement have been previously identified as prospective risk factors for noncontact ACL injury and may help to explain why those with ACLR are at an increased risk for secondary ACL injury.

What is already known about this topic

- Those who suffer an ACL injury, undergo ACLR, and return to physical activity are at a increased risk for secondary ACL injury
- The risk for reinjury is not dependent on the limb previously injured
- Biomechanical differences of those with ACLR have been observed, and prospective risk factors for this population have been identified

What this study adds

- This study provides for the first time, evidence of how lower extremity biomechanics change following ACL injury and ACLR in a sample of participants
- Frontal plane displacement of the hip and knee increases following ACL injury in both the injured and noninjured limb. This is accompanied by a decrease in knee extension moment of the injured limb.
- These motions have been previously identified as prospective risk factors for noncontact ACL injury, and provide initial evidence for why those with ACL injury are at an increased risk for secondary injury.

Table 1. Participant demographics and antrhopometrics. Units of measure are Age (years), Height (cm), and Mass (kg). Values represent means ± standard deviation.

		Baseline			Follow-Up		
	n	Age	Height	Mass	Age	Height	Mass
ACLR-INJ	12 (8 m, 4 f)	18.64 ± 0.50	174.10 ± 7.31	72.64 ± 9.48	21.42 ± 0.79	174.29 ± 7.56	76.25 ± 9.95
ACLR-NINJ	19 (9 m, 10 f)	18.52 ± 0.58	170.06 ± 9.26	68.99 ± 10.93	21.47 ± 0.77	170.05 ± 9.13	72.87 ± 12.78
Control	39 (20 m, 19 f)	18.48 ± 0.46	172.56 ± 9.10	70.17 ± 12.96	20.98 ± 0.73	172.73 ± 8.99	73.11 ± 13.16

Table 2. Group chronological descriptives. Unit of measure is days and values presented represent group mean ± standard deviation

	Baseline to Follow-Up	Baseline to Injury	Injury to Surgery	Surgery to Follow-Up
ACLR-Injured Limb	1,074.42 ± 197.28	367.73 ± 156.06	33.70 ± 20.29	666.90 ± 209.24
ACLR-Noninjured Limb	1,247.68 ± 179.04	533.33 ± 267.97	40.39 ± 24.92	691.06 ± 243.01
Control	1,077.59 ± 180.34	-	-	-

Table 3. Descriptive statistics for bilateral difference of anterior knee laxity assessed using KT-1000, Marx Activity score, and KOOS for each group. Values represent mean ± standard deviation, and units of measure other than anterior laxity (mm) are scale dependent.

			KOOS				
	Bilateral Difference Anterior Laxity (mm)	Marx Activity Total Score	Pain	Symptom	ADL	Sport/Rec	QOL
ACLR-INJ	1.9 ± 1.1	13.92 ± 2.75	84.75 ± 9.97	71.50 ± 13.16	93.00 ± 7.20	80.42 ± 16.16	65.63 ± 20.90
ALCR-NINJ	2.5 ± 1.8	11.21 ± 4.57	86.95 ± 11.62	73.42 ± 15.85	96.74 ± 5.05	78.42 ± 17.72	71.71 ± 22.57
Control	1.1 ± 0.8	11.05 ± 3.03	96.26 ± 5.06	92.54 ± 8.71	98.82 ± 2.09	93.97 ± 10.27	90.87 ± 14.04

*Laxity measures were not obtained for 3 members of the Control group, 1 member of the ACLR-Noninjured Limb group

Table 4. Descriptive statistics for knee and hip kinematics (°) at Initial Ground Contact for Baseline (PRE) and Follow-Up (POST)

		Control		ACLR-Noninjured Limb		ACLR-Injured Limb	
		Mean ± SD	95TH CI	Mean ± SD	95TH CI	Mean ± SD	95TH CI
Knee Sag	PRE	19.81 ± 5.69	(17.99, 21.63)	17.90 ± 5.69	(15.29, 20.51)	18.82 ± 5.72	(15.52, 22.12)
	POST	18.02 ± 7.21	(15.72, 20.33)	16.69 ± 7.22	(13.38, 19.99)	16.28 ± 7.26	(12.10, 20.46)
Knee Frt	PRE	0.96 ± 6.50	(-1.12, 3.04)	1.33 ± 6.51	(-1.65, 4.31)	2.61 ± 6.54	(-1.16, 6.38)
	POST	-0.49 ± 5.75	(-2.33, 1.35)	-4.84 ± 5.76	(-7.48, -2.20)	-4.22 ± 5.78	(-7.55, -0.89)
Knee Trv	PRE	-2.56 ± 7.88	(-5.08, -0.04)	-3.43 ± 7.89	(-7.05, 0.18)	-0.80 ± 7.93	(-5.37, 3.77)
	POST	-2.74 ± 7.30	(-5.07, -0.41)	-4.75 ± 7.31	(-8.10, -1.40)	-2.23 ± 7.35	(-6.46, 2.00)
Hip Sag	PRE	-33.16 ± 9.01	(-36.04, -30.28)	-26.33 ± 9.02	(-30.47, -22.20)	-31.42 ± 9.07	(-36.64, -26.19)
	POST	-29.95 ± 10.49	(-33.30, -26.60)	-29.67 ± 10.51	(-34.48, -24.85)	-30.07 ± 10.56	(-36.16, -23.99)
Hip Frt	PRE	-9.95 ± 6.60	(-12.06, -7.84)	-9.00 ± 6.61	(-12.03, -5.98)	-11.12 ± 6.64	(-14.95, -7.30)
	POST	-6.81 ± 9.59	(-9.87, -3.74)	0.41 ± 9.61	(-3.99, 4.81)	0.02 ± 9.65	(-5.54, 5.59)
Hip Trv	PRE	-1.85 ± 6.28	(-3.86, 0.16)	-1.99 ± 6.29	(-4.87, 0.89)	-2.12 ± 6.32	(-5.77, 1.52)
	POST	-3.17 ± 6.40	(-5.22, -1.13)	-4.13 ± 6.41	(-7.06, -1.19)	-4.18 ± 6.44	(-7.89, -0.47)

*Values for descriptive statistics are based on Gender entered as a covariate in the statistical model at a value of 0.53

Table 5. Descriptive statistics for knee and hip moments (Nm/BHxBW) and kinetics (N/BW) at Initial Ground Contact for Baseline (PRE) and Follow-Up (POST)

		CONTROL		ACLR-Noninjured Limb		ACLR-Injured Limb	
		Mean ± SD	95TH CI	Mean ± SD	95TH CI	Mean ± SD	95TH CI
Knee Sag	PRE	0.00 ± 0.02	(-0.01, 0.01)	0.01 ± 0.02	(0.00, 0.03)	0.01 ± 0.02	(-0.00, 0.02)
	POST	0.03 ± 0.03	(0.02, 0.04)	0.04 ± 0.03	(0.02, 0.05)	0.01 ± 0.03	(-0.01, 0.03)
Knee Frt	PRE	0.02 ± 0.02	(0.01, 0.03)	0.03 ± 0.02	(0.02, 0.03)	0.02 ± 0.02	(0.01, 0.03)
	POST	0.01 ± 0.02	(0.00, 0.01)	0.01 ± 0.02	(0.01, 0.02)	0.00 ± 0.02	(-0.01, 0.01)
Knee Trv	PRE	-0.01 ± 0.01	(-0.01, -0.00)	-0.01 ± 0.01	(-0.01, -0.00)	-0.01 ± 0.01	(-0.01, -0.00)
	POST	-0.00 ± 0.01	(-0.01, 0.00)	-0.01 ± 0.01	(-0.01, -0.00)	-0.00 ± 0.01	(-0.01, 0.00)
Hip Sag	PRE	-0.02 ± 0.07	(-0.04, 0.00)	0.03 ± 0.07	(0.00, 0.06)	0.03 ± 0.07	(-0.01, 0.07)
	POST	0.04 ± 0.08	(0.02, 0.07)	0.07 ± 0.08	(0.04, 0.11)	-0.00 ± 0.08	(-0.05, 0.05)
Hip Frt	PRE	0.03 ± 0.05	(0.02, 0.05)	0.06 ± 0.05	(0.03, 0.080)	0.05 ± 0.05	(0.02, 0.08)
	POST	0.01 ± 0.05	(-0.00, 0.03)	0.04 ± 0.05	(0.02, 0.06)	0.02 ± 0.05	(-0.01, 0.04)
Hip Trv	PRE	0.02 ± 0.02	(0.01, 0.03)	0.02 ± 0.02	(0.01, 0.03)	0.02 ± 0.02	(0.01, 0.03)
	POST	0.00 ± 0.02	(-0.00, 0.01)	0.02 ± 0.02	(0.02, 0.03)	0.01 ± 0.02	(-0.00, 0.02)
ATSF	PRE	-0.03 ± 0.17	(-0.08, 0.02)	-0.12 ± 0.17	(-0.20, -0.05)	-0.10 ± 0.17	(-0.20, -0.01)
	POST	-0.16 ± 0.22	(-0.23, -0.09)	-0.24 ± 0.22	(-0.34, -0.14)	-0.06 ± 0.22	(-0.18, 0.07)
VGRF	PRE	0.12 ± 0.05	(0.10, 0.13)	0.11 ± 0.05	(0.09, 0.13)	0.12 ± 0.05	(0.09, 0.150)
	POST	0.09 ± 0.03	(0.08, 0.10)	0.08 ± 0.03	(0.06, 0.09)	0.09 ± 0.03	(0.08, 0.11)

*Values for descriptive statistics are based on Gender entered as a covariate in the statistical model at a value of 0.53

Table 6. Descriptive statistics for peak knee and hip kinematics (°) at Landing Phase for Baseline (PRE) and Follow-Up (POST)

		CONTROL		ACLR-Noninjured Limb		ACLR-Injured Limb	
		Mean ± SD	95TH CI	Mean ± SD	95TH CI	Mean ± SD	95TH CI
Knee FLX	PRE	84.26 ± 13.49	(79.95, 88.58)	78.51 ± 13.51	(72.32, 84.69)	79.65 ± 13.58	(71.82, 87.47)
	POST	92.58 ± 16.23	(87.39, 97.77)	90.86 ± 16.25	(83.41, 98.30)	85.55 ± 16.33	(76.14, 94.96)
Knee EXT	PRE	19.82 ± 5.66	(18.01, 21.63)	16.84 ± 5.67	(14.25, 19.43)	18.64 ± 5.69	(15.36, 21.92)
	POST	18.16 ± 7.20	(15.85, 20.46)	16.69 ± 7.21	(13.39, 20.00)	16.24 ± 7.25	(12.06, 20.42)
Knee VRS	PRE	6.69 ± 7.43	(4.32, 9.07)	9.27 ± 7.44	(5.86, 12.67)	7.48 ± 7.48	(3.17, 11.79)
	POST	6.21 ± 7.10	(3.94, 8.48)	0.55 ± 7.11	(-2.71, 3.81)	1.28 ± 7.14	(-2.84, 5.40)
Knee VLG	PRE	-7.90 ± 8.18	(-10.52, -5.28)	-5.26 ± 8.19	(-9.01, -1.51)	-6.57 ± 8.23	(-11.32, -1.83)
	POST	-6.97 ± 7.73	(-9.44, -4.50)	-11.61 ± 7.74	(-15.16, -8.07)	-11.70 ± 7.78	(-16.18, -7.22)
Knee IR	PRE	9.18 ± 8.66	(6.41, 11.95)	9.29 ± 8.67	(5.32, 13.26)	12.33 ± 8.72	(7.31, 17.35)
	POST	17.10 ± 8.23	(14.47, 19.73)	9.30 ± 8.24	(5.53, 13.07)	13.24 ± 8.28	(8.47, 18.01)
Knee ER	PRE	-7.31 ± 8.84	(-10.14, -4.48)	-9.28 ± 8.85	(-13.34, -5.23)	-4.95 ± 8.90	(-10.08, 0.17)
	POST	-4.21 ± 6.45	(-6.27, -2.15)	-8.24 ± 6.45	(-11.19, -5.28)	-4.61 ± 6.49	(-8.35, -0.87)
Hip EXT	PRE	-33.12 ± 9.02	(-36.00, -30.23)	-26.09 ± 9.03	(-30.22, -21.95)	-31.10 ± 9.07	(-36.32, -25.87)
	POST	-29.41 ± 10.42	(-32.74, -26.08)	-29.61 ± 10.43	(-34.39, -24.83)	-30.00 ± 10.48	(-36.04, -23.96)
Hip FLX	PRE	-72.63 ± 18.71	(-78.61, -66.64)	-63.46 ± 18.74	(-72.04, -54.88)	-67.15 ± 18.83	(-78.00, -56.30)
	POST	-79.14 ± 21.23	(-85.93, -72.36)	-76.86 ± 21.26	(-86.60, -67.12)	-73.80 ± 21.37	(-86.12, -61.48)
Hip ADD	PRE	1.49 ± 8.58	(-1.26, 4.23)	0.88 ± 8.59	(-3.06, 4.81)	0.49 ± 8.63	(-4.49, 5.47)
	POST	-1.72 ± 8.14	(-4.32, 0.89)	4.18 ± 8.15	(0.44, 7.91)	4.24 ± 8.19	(-0.49, 8.96)
Hip ABD	PRE	-13.11 ± 8.18	(-15.72, -10.49)	-12.06 ± 8.19	(-15.81, -8.30)	-14.10 ± 8.23	(-18.85, -9.36)
	POST	-15.32 ± 8.74	(-18.11, -12.52)	-9.96 ± 8.76	(-13.97, -5.95)	-8.90 ± 8.80	(-13.97, -3.83)
Hip IR	PRE	6.89 ± 6.34	(4.87, 8.92)	8.86 ± 6.35	(5.95, 11.77)	6.31 ± 6.38	(2.64, 9.99)
	POST	7.19 ± 9.14	(4.27, 10.12)	2.93 ± 9.15	(-1.27, 7.12)	3.89 ± 9.20	(-1.41, 9.19)
Hip ER	PRE	-6.01 ± 6.37	(-8.05, -3.98)	-5.05 ± 6.38	(-7.97, -2.12)	-6.84 ± 6.41	(-10.53, -3.14)
	POST	-6.50 ± 7.19	(-8.80, -4.20)	-10.69 ± 7.20	(-13.99, -7.40)	-7.13 ± 7.24	(-11.30, -2.96)

*Values for descriptive statistics are based on Gender entered as a covariate in the statistical model at a value of 0.53

Table 7. Descriptive statistics for knee moments (Nm/BHxBW) and kinetics (N/BW) at Landing Phase (PRE) and Follow-Up (POST)

		CONTROL		ACLR-Noninjured Limb		ACLR-Injured Limb	
		Mean ± SD	95TH CI	Mean ± SD	95TH CI	Mean ± SD	95TH CI
Knee EXT	PRE	-0.206 ± 0.055	(-0.224, -0.189)	-0.219 ± 0.055	(-0.245, -0.194	-0.234 ± 0.056	(-0.266, -0.202)
	POST	-0.204 ± 0.043	(-0.217, -0.190)	-0.214 ± 0.043	(-0.234, -0.194)	-0.169 ± 0.044	(-0.194, -0.144)
Knee FLX	PRE	0.021 ± 0.043	(0.008, 0.035)	0.057 ± 0.043	(0.037, 0.077)	0.038 ± 0.044	(0.013, 0.063)
	POST	0.046 ± 0.034	(0.035, 0.057)	0.053 ± 0.034	(0.037, 0.069)	0.050 ± 0.034	(0.030, 0.070)
Knee VLG	PRE	-0.064 ± 0.037	(-0.076, -0.052)	-0.087 ± 0.037	(-0.104, -0.070)	-0.074 ± 0.038	(-0.095, -0.052)
	POST	-0.045 ± 0.027	(-0.053, -0.036)	-0.055 ± 0.027	(-0.067, -0.042)	-0.044 ± 0.027	(-0.060, -0.029)
Knee VRS	PRE	0.061 ± 0.026	(0.053, 0.069)	0.057 ± 0.026	(0.045, 0.069)	0.064 ± 0.026	(0.049, 0.079)
	POST	0.063 ± 0.027	(0.055, 0.072)	0.067 ± 0.027	(0.054, 0.079)	0.055 ± 0.027	(0.039, 0.070)
Knee ER	PRE	-0.046 ± 0.023	(-0.053, -0.038)	-0.040 ± 0.023	(-0.050, -0.029)	-0.044 ± 0.024	(-0.058, -0.030)
	POST	-0.056 ± 0.019	(-0.063, -0.050)	-0.049 ± 0.019	(-0.058, -0.040)	-0.037 ± 0.020	(-0.049, -0.026)
Knee IR	PRE	0.043 ± 0.022	(0.036, 0.050)	0.045 ± 0.022	(0.035, 0.055)	0.049 ± 0.022	(0.037, 0.062)
	POST	0.031 ± 0.021	(0.024, 0.038)	0.044 ± 0.021	(0.034, 0.053)	0.037 ± 0.021	(0.025, 0.049)
ATSF	PRE	1.070 ± 0.319	(0.968, 1.172)	1.192 ± 0.319	(1.046, 1.339)	1.247 ± 0.321	(1.062, 1.432)
	POST	1.134 ± 0.244	(1.056, 1.212)	1.185 ± 0.245	(1.073, 1.297)	0.923 ± 0.246	(0.781, 1.065)
VGRF	PRE	3.094 ± 1.016	(2.769, 3.418)	3.254 ± 1.017	(2.788, 3.720)	3.188 ± 1.022	(2.599, 3.777)
	POST	2.726 ± 0.761	(2.482, 2.969)	3.084 ± 0.762	(2.735, 3.433)	3.114 ± 0.765	(2.673, 3.555)

*Values for descriptive statistics are based on Gender entered as a covariate in the statistical model at a value of 0.53

Table 8. Descriptive statistics for hip moments (Nm/BHxBW) at Landing Phase (PRE) and Follow-Up (POST)

		CONTROL		ACLR-Noninjured Limb		ACLR-Injured Limb	
		Mean ± SD	95TH CI	Mean ± SD	95TH CI	Mean ± SD	95TH CI
Hip Flexion	PRE	-0.214 ± 0.103	(-0.247, -0.181)	-0.254 ± 0.103	(-0.301, -0.206)	-0.256 ± 0.104	(-0.316, -0.196)
	POST	-0.205 ± 0.067	(-0.227, -0.184)	-0.216 ± 0.067	(-0.247, -0.185)	-0.138 ± 0.068	(-0.177, -0.099)
Hip Extension	PRE	0.251 ± 0.127	(0.210, 0.292)	0.255 ± 0.128	(0.197, 0.314)	0.229 ± 0.128	(0.155, 0.303)
	POST	0.202 ± 0.091	(0.172, 0.231)	0.193 ± 0.091	(0.151, 0.235)	0.240 ± 0.092	(0.187, 0.293)
Hip Abduction	PRE	-0.131 ± 0.065	(-0.152, -0.111)	-0.153 ± 0.065	(-0.183, -0.123)	-0.171 ± 0.065	(-0.209, -0.134)
	POST	-0.108 ± 0.068	(-0.130, -0.086)	-0.141 ± 0.068	(-0.173, -0.110)	-0.102 ± 0.068	(-0.142, -0.063)
Hip Adduction	PRE	0.137 ± 0.067	(0.115, 0.158)	0.149 ± 0.068	(0.118, 0.180)	0.145 ± 0.068	(0.106, 0.184)
	POST	0.140 ± 0.054	(0.123, 0.157)	0.124 ± 0.054	(0.099, 0.149)	0.091 ± 0.054	(0.060, 0.122)
Hip ER	PRE	-0.085 ± 0.052	(-0.101, -0.068)	-0.095 ± 0.052	(-0.119, -0.071)	-0.099 ± 0.053	(-0.129, -0.069)
	POST	-0.052 ± 0.033	(-0.063, -0.042)	-0.078 ± 0.033	(-0.093, -0.063)	-0.055 ± 0.033	(-0.074, -0.036)
Hip IR	PRE	0.076 ± 0.029	(0.067, 0.086)	0.068 ± 0.029	(0.054, 0.081)	0.069 ± 0.029	(0.052, 0.086)
	POST	0.060 ± 0.031	(0.050, 0.069)	0.068 ± 0.031	(0.054, 0.082)	0.048 ± 0.031	(0.030, 0.065)

*Values for descriptive statistics are based on Gender entered as a covariate in the statistical model at a value of 0.53

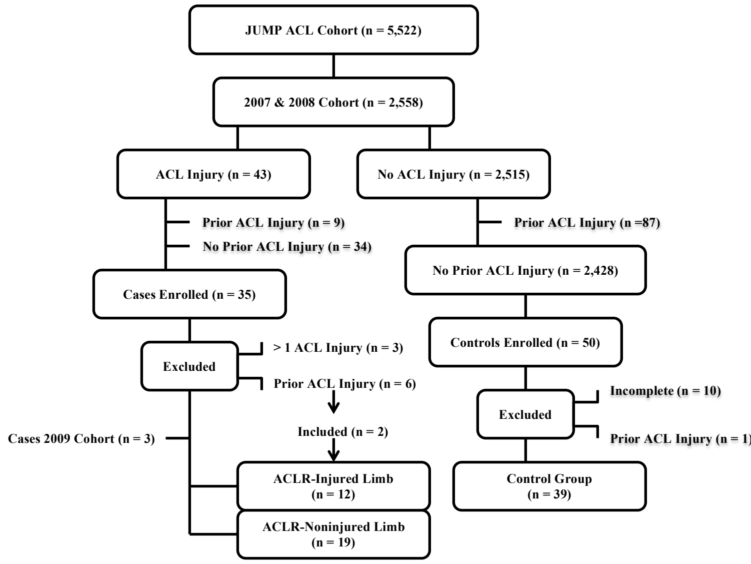

Figure 1. Description of Participant Enrollment from the JUMP ACL Cohorts of 2007 and 2008

Figure 2. Double Leg Jump Landing. Participants were required to stand atop a box located a distance equal to one half of their body height from the front edge of the force plate, jump forward, land with their foot completely on the force plate, and then immediately make a vertical jump for maximum height.

REFERENCES

1. Spindler KP, Huston LJ, Wright RW, et al. The prognosis and predictors of sports function and activity at minimum 6 years after anterior cruciate ligament reconstruction: a population cohort study. *Am J Sports Med.* Feb 2011;39(2):348-359.

2. Ardern CL, Taylor NF, Feller JA, Webster KE. Return-to-sport outcomes at 2 to 7 years after anterior cruciate ligament reconstruction surgery. *Am J Sports Med.* Jan 2012;40(1):41-48.

3. von Porat A, Roos EM, Roos H. High prevalence of osteoarthritis 14 years after an anterior cruciate ligament tear in male soccer players: a study of radiographic and patient relevant outcomes. *Ann Rheum Dis.* Mar 2004;63(3):269-273.

4. Lohmander LS, Ostenberg A, Englund M, Roos H. High prevalence of knee osteoarthritis, pain, and functional limitations in female soccer players twelve years after anterior cruciate ligament injury. *Arthritis Rheum.* Oct 2004;50(10):3145-3152.

5. Orchard J, Seward H, McGivern J, Hood S. Intrinsic and extrinsic risk factors for anterior cruciate ligament injury in Australian footballers. *Am J Sports Med.* Mar-Apr 2001;29(2):196-200.

6. Faude O, Junge A, Kindermann W, Dvorak J. Risk factors for injuries in elite female soccer players. *Br J Sports Med.* Sep 2006;40(9):785-790.

7. Paterno MV, Rauh MJ, Schmitt LC, Ford KR, Hewett TE. Incidence of contralateral and ipsilateral anterior cruciate ligament (ACL) injury after primary ACL reconstruction and return to sport. *Clin J Sport Med.* Mar 2012;22(2):116-121.

8. Salmon L, Russell V, Musgrove T, Pinczewski L, Refshauge K. Incidence and risk factors for graft rupture and contralateral rupture after anterior cruciate ligament reconstruction. *Arthroscopy.* Aug 2005;21(8):948-957.

9. Shelbourne KD, Gray T, Haro M. Incidence of subsequent injury to either knee within 5 years after anterior cruciate ligament reconstruction with patellar tendon autograft. *Am J Sports Med.* Feb 2009;37(2):246-251.

10. Wright RW, Dunn WR, Amendola A, et al. Risk of tearing the intact anterior cruciate ligament in the contralateral knee and rupturing the anterior cruciate ligament graft during the first 2 years after anterior cruciate ligament reconstruction: a prospective MOON cohort study. *Am J Sports Med.* Jul 2007;35(7):1131-1134.

11. Paterno MV, Schmitt LC, Ford KR, et al. Biomechanical measures during landing and postural stability predict second anterior cruciate ligament injury after anterior cruciate ligament reconstruction and return to sport. *Am J Sports Med.* Oct 2010;38(10):1968-1978.

12. Ernst GP, Saliba E, Diduch DR, Hurwitz SR, Ball DW. Lower extremity compensations following anterior cruciate ligament reconstruction. *Phys Ther.* Mar 2000;80(3):251-260.

13. Decker MJ, Torry MR, Noonan TJ, Riviere A, Sterett WI. Landing adaptations after ACL reconstruction. *Med Sci Sports Exerc.* Sep 2002;34(9):1408-1413.

14. Paterno MV, Ford KR, Myer GD, Heyl R, Hewett TE. Limb asymmetries in landing and jumping 2 years following anterior cruciate ligament reconstruction. *Clin J Sport Med.* Jul 2007;17(4):258-262.

15. Vairo GL, Myers JB, Sell TC, Fu FH, Harner CD, Lephart SM. Neuromuscular and biomechanical landing performance subsequent to ipsilateral semitendinosus and gracilis autograft anterior cruciate ligament reconstruction. *Knee Surg Sports Traumatol Arthrosc.* Jan 2008;16(1):2-14.

16. Salem GJ, Salinas R, Harding FV. Bilateral kinematic and kinetic analysis of the squat exercise after anterior cruciate ligament reconstruction. *Arch Phys Med Rehabil.* Aug 2003;84(8):1211-1216.

17. Tashman S, Collon D, Anderson K, Kolowich P, Anderst W. Abnormal rotational knee motion during running after anterior cruciate ligament reconstruction. *Am J Sports Med.* Jun 2004;32(4):975-983.

18. Gokeler A, Hof AL, Arnold MP, Dijkstra PU, Postema K, Otten E. Abnormal landing strategies after ACL reconstruction. *Scand J Med Sci Sports.* Feb 2010;20(1):e12-19.

19. Brophy RH, Wright RW, Powell JW, Matava MJ. Injuries to kickers in American football: the National Football League experience. *Am J Sports Med.* Jun 2010;38(6):1166-1173.

20. Grindstaff TL, Jackson KR, Garrison JC, Diduch DR, Ingersoll CD. Decreased quadriceps activation measured hours prior to a noncontact anterior cruciate ligament tear. *J Orthop Sports Phys Ther.* Aug 2008;38(8):508-516.

21. Robbins SM, Clark JM, Maly MR. Longitudinal gait and strength changes prior to and following an anterior cruciate ligament rupture and surgical reconstruction: a case report. *J Orthop Sports Phys Ther.* Mar 2011;41(3):191-199.

22. Marx RG, Stump TJ, Jones EC, Wickiewicz TL, Warren RF. Development and evaluation of an activity rating scale for disorders of the knee. *Am J Sports Med.* Mar-Apr 2001;29(2):213-218.

23. Roos EM, Roos HP, Lohmander LS, Ekdahl C, Beynnon BD. Knee Injury and Osteoarthritis Outcome Score (KOOS)--development of a self-administered outcome measure. *J Orthop Sports Phys Ther.* Aug 1998;28(2):88-96.

24. Ahlden M, Kartus J, Ejerhed L, Karlsson J, Sernert N. Knee laxity measurements after anterior cruciate ligament reconstruction, using either bone-patellar-tendon-bone or hamstring tendon autografts, with special emphasis on comparison over time. *Knee Surg Sports Traumatol Arthrosc.* Sep 2009;17(9):1117-1124.

25. Padua DA, Marshall SW, Boling MC, Thigpen CA, Garrett WE, Jr., Beutler AI. The Landing Error Scoring System (LESS) Is a valid and reliable clinical assessment tool of jump-landing biomechanics: The JUMP-ACL study. *Am J Sports Med.* Oct 2009;37(10):1996-2002.

26. Boling MC, Padua DA, Marshall SW, Guskiewicz K, Pyne S, Beutler A. A prospective investigation of biomechanical risk factors for patellofemoral pain syndrome: the Joint Undertaking to Monitor and Prevent ACL Injury (JUMP-ACL) cohort. *Am J Sports Med.* Nov 2009;37(11):2108-2116.

27. Bell AL, Pedersen DR, Brand RA. A comparison of the accuracy of several hip center location prediction methods. *J Biomech.* 1990;23(6):617-621.

28. Deneweth JM, Bey MJ, McLean SG, Lock TR, Kolowich PA, Tashman S. Tibiofemoral joint kinematics of the anterior cruciate ligament-reconstructed knee during a single-legged hop landing. *Am J Sports Med.* Sep 2010;38(9):1820-1828.

29. Webster KE, Feller JA. Tibial rotation in anterior cruciate ligament reconstructed knees during single limb hop and drop landings. *Clin Biomech (Bristol, Avon).* Jan 11 2012.

30. Chouliaras V, Ristanis S, Moraiti C, Stergiou N, Georgoulis AD. Effectiveness of reconstruction of the anterior cruciate ligament with quadrupled hamstrings and bone-patellar tendon-bone autografts: an in vivo study comparing tibial internal-external rotation. *Am J Sports Med.* Feb 2007;35(2):189-196.

31. Ristanis S, Stergiou N, Patras K, Vasiliadis HS, Giakas G, Georgoulis AD. Excessive tibial rotation during high-demand activities is not restored by anterior cruciate ligament reconstruction. *Arthroscopy.* Nov 2005;21(11):1323-1329.

32. Andriacchi TP, Briant PL, Bevill SL, Koo S. Rotational changes at the knee after ACL injury cause cartilage thinning. *Clin Orthop Relat Res.* Jan 2006;442:39-44.

33. Delahunt E, Sweeney L, Chawke M, et al. Lower limb kinematic alterations during drop vertical jumps in female athletes who have undergone anterior cruciate ligament reconstruction. *J Orthop Res.* Jan 2012;30(1):72-78.

34. Beynnon B, Howe JG, Pope MH, Johnson RJ, Fleming BC. The measurement of anterior cruciate ligament strain in vivo. *Int Orthop.* 1992;16(1):1-12.

35. Markolf KL, Burchfield DM, Shapiro MM, Shepard MF, Finerman GA, Slauterbeck JL. Combined knee loading states that generate high anterior cruciate ligament forces. *J Orthop Res.* Nov 1995;13(6):930-935.

36. Sell TC, Ferris CM, Abt JP, et al. Predictors of proximal tibia anterior shear force during a vertical stop-jump. *J Orthop Res.* Dec 2007;25(12):1589-1597.

37. Bush-Joseph CA, Hurwitz DE, Patel RR, et al. Dynamic function after anterior cruciate ligament reconstruction with autologous patellar tendon. *Am J Sports Med.* Jan-Feb 2001;29(1):36-41.

38. Orishimo KF, Kremenic IJ, Mullaney MJ, McHugh MP, Nicholas SJ. Adaptations in single-leg hop biomechanics following anterior cruciate ligament reconstruction. *Knee Surg Sports Traumatol Arthrosc.* Nov 2010;18(11):1587-1593.

39. Blackburn JT, Padua DA. Sagittal-plane trunk position, landing forces, and quadriceps electromyographic activity. *Journal of athletic training.* Mar-Apr 2009;44(2):174-179.

40. Berchuck M, Andriacchi TP, Bach BR, Reider B. Gait adaptations by patients who have a deficient anterior cruciate ligament. *J Bone Joint Surg Am.* Jul 1990;72(6):871-877.

41. Hart JM, Pietrosimone B, Hertel J, Ingersoll CD. Quadriceps activation following knee injuries: a systematic review. *Journal of athletic training.* Jan-Feb 2010;45(1):87-97.

42. Lewek M, Rudolph K, Axe M, Snyder-Mackler L. The effect of insufficient quadriceps strength on gait after anterior cruciate ligament reconstruction. *Clin Biomech (Bristol, Avon).* Jan 2002;17(1):56-63.

43. Olmstead TG, Wevers HW, Bryant JT, Gouw GJ. Effect of muscular activity on valgus/varus laxity and stiffness of the knee. *J Biomech.* 1986;19(8):565-577.

44. Lloyd DG, Buchanan TS. Strategies of muscular support of varus and valgus isometric loads at the human knee. *J Biomech.* Oct 2001;34(10):1257-1267.

45. Lloyd DG, Buchanan TS, Besier TF. Neuromuscular biomechanical modeling to understand knee ligament loading. *Med Sci Sports Exerc.* Nov 2005;37(11):1939-1947.

46. Andriacchi TP, Andersson GB, Ortengren R, Mikosz RP. A study of factors influencing muscle activity about the knee joint. *J Orthop Res.* 1984;1(3):266-275.

47. Palmieri-Smith RM, Wojtys EM, Ashton-Miller JA. Association between preparatory muscle activation and peak valgus knee angle. *J Electromyogr Kinesiol.* Dec 2008;18(6):973-979.

48. Pollard CD, Sigward SM, Powers CM. Limited hip and knee flexion during landing is associated with increased frontal plane knee motion and moments. *Clin Biomech (Bristol, Avon).* Feb 2010;25(2):142-146.

49. Moraiti CO, Stergiou N, Vasiliadis HS, Motsis E, Georgoulis A. Anterior cruciate ligament reconstruction results in alterations in gait variability. *Gait Posture.* Jun 2010;32(2):169-175.

50. Hewett TE, Myer GD, Ford KR, et al. Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study. *Am J Sports Med.* Apr 2005;33(4):492-501.

51. Padua DA, DiStefano LJ, Marshall SW, Beutler AI, de la Motte SJ, DiStefano MJ. Retention of movement pattern changes after a lower extremity injury prevention program is affected by program duration. *Am J Sports Med.* Feb 2012;40(2):300-306.

52. Myklebust G, Bahr R. Return to play guidelines after anterior cruciate ligament surgery. *Br J Sports Med.* Mar 2005;39(3):127-131.

APPENDIX C. MANUSCRIPT 2

Coordination of the Hip and Knee During a Double Leg Jump Landing Are Altered By ACL Injury: The JUMP ACL Study

(Clinical Biomechanics)

Background: Those with ACLR are at an increased risk for secondary ACL injury and the development of osteoarthritis. Previous work has identified differences in lower extremity biomechanics and joint coordination in this population. The purpose of this study was to compare joint coordination prior to and following ACL injury to better understand the effect of ACL injury on lower extremity joint coordination.

Methods: 69 participants (12 ACLR-Injured Limb, 19 ACLR-Noninjured Limb, 38 Control) who were part of the JUMP ACL cohort were included in this analysis. Average coupling angle between the hip and knee in the sagittal, frontal, and transverse plane was calculated, as well as interaction in the frontal and transverse planes during a double-leg jump landing prior to (Baseline) and following ACL injury (Follow-Up).

Findings: We observed a significant change in the average coupling angle of the hip and knee in the transverse plane that indicated the ACLR-Injured Limb group had more equal relative motion between the two joints following ACL injury and ACLR.

Interpretation: Based on the findings we observed, it appears that ACL injury and ACLR result in altered coordination of the hip and knee rotation, most likely associated with decreased independence of knee rotation. This is in agreement with previous research that has identified alteration in knee rotation of those with ACLR and may have implications for the progression of knee osteoarthritis.

INTRODUCTION

An increased incidence of secondary ACL injury has been reported for individuals with a previous history of ACL injury and subsequent reconstructive surgery (ACLR).[1-3] The incidence of injury for these individuals when they return to physical activity is between 6-25%.[3-6] Their risk for reinjury is 5-15 times higher for ACL injury as compared to those who have no history of ACL injury[1-3], and the risk for reinjury is not dependent on the previously injured limb.[6] In addition to the risk of secondary ACL injury, an increased prevalence of knee osteoarthritis has also been previously observed for this population.[7,8] The risk for reinjury and the potential sequelae of poor outcomes leading to the development of knee osteoarthritis highlights the need for a better understanding of how to improve care and treatment for these injuries.

Previous work has been conducted to identify the unique characteristics of those with ACLR and have identified differences when compared to healthy individuals[9-12] as well as between limbs.[11-16] Those with ACLR, in general demonstrate differences associated with decreased loading of the reconstructed knee. These studies provide useful information, but have isolated particular biomechanical variables for analysis. This isolation may provide a limited view, as movement between multiple joints in multiple planes has to be coordinated for successful execution of movement. Assessment of the relative coordination of movement between joints or segments may provide a more global view of how individuals with ACLR coordinate movement.

Work in the area of movement coordination and ACL injury has been previously conducted. This work has focused on individuals after ACLR[17,18] or for individuals thought to be at an increased risk for injury (i.e. females).[19] These studies are again useful, but limited, as they have to be interpreted with the assumption that any differences that are present in those with ACLR are the result of ACL injury and ACLR. As such, the influence of ACL injury and subsequent ACLR on movement coordination is not truly known. Such analyses would provide better evidence for the coordination patterns that are the result of ACL injury itself, and not unique to the individual. Therefore, the purpose of this study was to address this limitation by comparing the coordination of the hip and knee during a double leg jump landing prior to and following ACL injury and subsequent ACLR.

METHODS

Participants

Participants for this study were recruited from the Joint Undertaking to Monitor and Prevent ACL Injury (JUMP ACL) Project. The JUMP ACL Project is a multi-year prospective study conducted with members of the United States' service academies. The purpose of this project was to identify risk factors for noncontact ACL injury and included collection of lower extremity biomechanics, isometric strength, and posture, as well as detailed orthopaedic and physical activity information. Measures were collected on a sample of each incoming class between 2004 and 2009, and participants were followed during their enrollment at their respective service academy for ACL injury.

For the purposes of this study, we wanted to collect biomechanical measures on participants that had injured their ACL during enrollment in the JUMP ACL study, as well as a group of healthy matched control participants. Identification of participants for follow up testing was limited to those members of the 2007 and 2008 cohorts. Selection was also limited to those that had complete biomechanical data from their initial enrollment in the study, and were currently enrolled in their respective service academy and the JUMP ACL project. In addition, healthy matched controls were identified based on the criteria of matching the Cases for cohort year, service academy, and sex. Details regarding the enrollment of participants can be found in **Figure 1**.

Procedures

At the time of initial enrollment for the JUMP ACL project (Baseline) and during data collection for follow up analysis (Follow-Up), each participant read and signed an informed consent, had anthropometric information (height and mass) and demographic information recorded (age, sex). They were also asked at Follow-Up to complete the Marx Activity Scale[20], Knee Injury and Osteoarthritis Outcome Score (KOOS)[21], and have anterior knee laxity assessed bilaterally[22], prior to performing a double leg jump landing (**Figure 2**) during which data for lower extremity biomechanics were collected. The task required participants to stand atop a 30 cm high box located a distance from the edge of a force plate equal to half their height, jump forward from the box, landing with the foot of their instrumented leg completely on the force plate, and upon landing immediately make a maximal effort vertical jump. Trials were excluded and repeated if participants didn't land with the foot of the instrumented limb completely on the force plate, the foot of the other limb made contact with the force plate, or if the participant hesitated between the landing and vertical jump phases of the task.

Biomechanical data during the double leg jump landing was collected on the instrumented limb using an electromagnetic tracking system (Ascension Technologies Inc., Burlington, VT) integrated with a non-conductive force plate (Bertec Co., Columbus, OH, USA). All biomechanical data collection was facilitated by the use of the Motion Monitor Software (Innovative Sports Training, Inc., Chicago, IL, USA). Prior to completion of the double leg jump landing, all participants were instrumented so that electromagnetic sensors were affixed to the shank and thigh of the test limb, and pelvis. The sensor for the shank was placed on the skin overlying the medial tibia, the thigh sensor on the mid portion of the lateral thigh and affixed with double sided tape and secured with overlying athletic tape. The pelvis sensor was affixed to the sacrum at the mid point between the posterior superior iliac spines with double sided tape and held in place with an elastic belt. Sensor placement and procedures were performed to minimize the introduction of motion artifact during the jump landing and were consistent between Baseline and Follow-Up testing sessions.

Following placement of the electromagnetic sensors, a model of the lower extremity was constructed through digitization of the medial and lateral malleoli, medial and lateral femoral epicondyles, and the anterior superior iliac spines using a moveable sensor. These landmarks were used to define the ankle joint center, knee joint center, and hip joint center respectively.[23] Shank, thigh, and pelvis segments were defined based on the use of the joints as segment endpoints and the respective electromagnetic sensors. Local right-handed segment axis systems were embedded into each segment, the orientation of which coincided with the global axis system (positive z-axis coincided with the vertical direction, positive x-axis coincided with the anterior direction of the participant, and the positive y-axis was defined relative to the positive x-axis as a vector coinciding with a 90° positive rotation about the z-axis).

Kinematic and kinetic data were sampled at a frequency of 144 Hz and 1,444 Hz, respectively. All data acquisition procedures were consistent between Baseline and Follow-Up sessions.

Data Reduction

Prior to data exportation, all trials of interest from Baseline and Follow-Up were visually inspected for data collection errors not detected at the time of collection to ensure the quality of the biomechanical model. Three trials from Baseline and Follow-Up data collection were selected for calculation of measures of interest. The measures of interest for this analysis included sagittal, frontal, and transverse plane motion at the hip and knee, as well as the vertical ground reaction force data. Hip kinematics were defined as the motion of the thigh segment relative to the pelvis, and knee kinematics as the shank segment relative to the thigh. An Euler sequence of Y, X, Z was used to define orientation of the relative segments and calculate hip and knee angles, such that a first rotation occurred about the y-axis, second rotation about the x-axis, and the third rotation about the z-axis. All kinematic data were filtered using a 4^{th} order Butterworth filter (14.5 Hz), and all data were exported using the Motion Monitor software.

For the purposes of this study we were interested in describing the change in the relative coordination of the movement of the hip and knee prior to and following ACL injury during a double leg jump landing task. We were interested in the time period of landing during the jump landing, defined as the time point from initial ground contact (vertical ground reaction force first exceeded 10N) to the time point of maximum knee flexion. Time series data for hip and knee angles in all three planes were extracted for each trial during the landing phase. Each series of data was normalized to 101 data points.

Angle-angle plots were generated for each coordination interaction of interest: Hip Sagittal Plane – Knee Sagittal Plane, Hip Frontal Plane – Knee Frontal Plane, Hip Transverse Plane – Knee Transverse Plane, Hip Frontal Plane – Knee Transverse Plane, and Hip Transverse Plane – Knee frontal Plane. Plots were generated such that the horizontal axis corresponded with the relative motion of the proximal joint (hip) and the vertical axis as the relative motion of the distal joint (knee) (**Figure 3**). Joint coordination was defined based on quantification of the coupling angle between the hip and knee for each angle-angle plot based on techniques previously used by Heiderscheit et al[24] and Ferber et al.[25] This technique calculated an angle between the horizontal axis of the graph and a vector connecting two subsequent data points of the angle-angle plot. The absolute value of this angle was calculated so that the resulting coupling angles ranged from 0°-90° to allow the use of standard parametric statistics rather than circular statistics.[25] Using our conventions, an average coupling angle value of 45° represents equal relative motion between the two joints, a value greater than 45° represents greater relative motion of the knee, and a value less than 45° represents greater relative motion of the hip. This procedure was completed for each subsequent data point of the angle-angle plot and the average of the values was calculated. This procedure was performed for each trial, and an average value was calculated for the three trials. All calculations were performed using a customized MATLAB program (Mathworks, Inc., Natick, MA).

Statistical Analyses

Prior to statistical analysis, each variable was assessed for normality to ensure proper use of parametric tests and to identify potential statistical outliers. This included the calculation of Z-scores for the skewness and kurtosis of each variable, and those with values with a value greater than 1.96 were identified for further analysis including plotting of standardized residuals and box plots to characterize normality and identify outliers. If statistical outliers were identified statistical procedures were conducted with and without values for the outliers included in the analysis. The potential outliers remained in the final data set if no change in statistical significance of main effects or interaction effects resulted. Results that were altered by removal were further analyzed for any potential errors in data collection or inaccuracy in the biomechanical model. To retain as many participants in the analysis as possible, any trials that demonstrated collection errors were removed such that some participants may have a mean value calculated from fewer than 3 trials.

To assess the effect of ACL injury on coordination of the hip and knee five 3x2 (Group: ACLR-Injured Limb, ACLR-Noninjured Limb, Control; Time: Baseline, Follow-Up) mixed model analyses of covariance were performed. Because of the relative difference in proportions of males and females in each group, sex was entered as a covariate for each analysis. Post hoc analyses consisted of Tukey’s HSD and were implemented for any significant interaction effect or group main effect. An a priori alpha level of 0.05 was set for all analyses, and all statistical analyses were conducted using IBM SPSS v19 (SPSS, Inc., an IBM company, Chicago, IL).

RESULTS

Data Outlier & Normality Check

Our procedures to identify outliers and check normality of the dependent variables to ensure proper application of parametric statistics identified two variables as potentially nonparametric, Hip Sagittal Plane – Knee Sagittal Plane at Follow-Up and Hip Transverse Plane – Knee Frontal Plane at Follow-Up. Potential statistical outliers were identified, and when the values for the respective participants were removed it did not change the interpretation of the findings for this analysis. Therefore, the participants and their representative three trial averages were retained in the analysis. One participant from the Control group was eliminated from data analysis because of excessive motion artifact at the time of initial ground contact for all three trials, leaving 38 participants for the Control Group. Demographics and anthropometrics for each group at Baseline and Follow-Up are summarized in **Table 1** and **Table 2**. Graft type was not obtained for 5 members of the ACLR-Injured Limb group, and 6 of the ACLR-Noninjured Limb group. For the ACLR-Injured Limb group 3 had a bone-patella tendon-bone autograft and 4 had a hamstrings autograft. For the ACLR-Noninjured Limb group 5 had a bone-patella tendon-bone graft, 7 had a hamstring autograft, and one had an Achilles tendon allograft. Descriptive statistics for anterior knee laxity, Marx Activity Score, and KOOS are provided in **Table 3**.

Lower Extremity Coordination

We observed a significant Time x Group interaction for Hip Transverse Plane – Knee Transverse Plane ($F_{(2,65)} = 4.398$, $p = 0.016$) coupling angle. Post hoc analyses revealed there were no differences among the groups at the time of Baseline testing. However, the ACLR-Injured Limb group had a significant change in Hip Transverse Plane – Knee Transverse Plane coordination from Baseline to Follow-Up. Specifically, there was a shift of less knee rotation relative to hip rotation (Baseline: 54.847 (50.837, 58.858); Follow-Up: 47.973 (44.147, 51.800)). The relative coordination of hip and knee motion in the transverse plane for the ACLR-Injured Limb group at Follow-Up was significantly different than the Control group (54.779 (52.644, 56.915)). After ACL injury, the ACLR-Injured Limb group demonstrated a decrease in the relative knee motion, and demonstrated more equal transverse plane motion of the hip and knee. There were no other significant interactions: Hip Sagittal Plane – Knee Sagittal Plane ($F_{(2,65)} = 0.850$, $p = 0.432$), Hip Frontal Plane – Knee Frontal Plane ($F_{(2,65)} = 0.247$, $p = 0.782$), Hip Frontal Plane – Knee Transverse Plane ($F_{(2,65)} = 2.678$, $p = 0.076$), and Hip Transverse Plane – Knee Frontal Plane ($F_{(2,65)} = 0.025$, $p = 0.975$).

A time main effect was observed for Hip Frontal Plane – Knee Transverse Plane ($F_{(1,65)} = 4.789$, $p = 0.032$). When groups were collapsed across time, the change in values for each coordination variable indicated that there was a shift towards more hip motion relative to knee motion at Follow-Up as compared to Baseline. No other time main effects were observed, and no group main effects were observed. Descriptive statistics for each variable of interest is provided in **Table 4**.

DISCUSSION

The primary finding of this study indicates that ACL injury and subsequent ACLR causes a change in the relative coordination of hip rotation and knee rotation during a double leg jump landing. We observed no other changes in coordination across the hip and knee joint planes. In addition, these alterations were limited to the injured limb. This observation cannot be directly compared to any previous findings, as this is the first time, to our knowledge, that coordination measures have been compared prior to and following ACL injury and subsequent ACLR. Alterations in coordination of those with ACLR have been previously observed, though.[17,18] Both Van Uden et al[17] and Kurz et al[18] observed alterations in sagittal plane coordination of the lower extremity as assessed by continuous relative phase methods for those with ACLR during single leg hopping and gait, respectively. Differences in the variability of coordination between the thigh and shank in the transverse plane were previously observed between healthy males and females by Pollard et al[19] during a side-step cutting task. Direct comparison of our findings is not appropriate as we did not compare differences between genders and we did not measure coordination variability between the two joints. Their findings do though provide some support for our findings of alterations in transverse plane coordination of the hip and knee after ACL injury.

This shift in transverse plane coordination may help to expand our current understanding as to why those with ACLR are at an increased risk for reinjury and knee osteoarthritis. These findings are of clinical importance as alterations in knee rotation observed following ACL injury have been thought to influence loading of articular cartilage and influence the risk for the development of osteoarthritis.[26] Following ACL injury, we observed more equal motion between the hip and knee in the transverse plane, with a shift towards more hip rotation relative to knee rotation for the members of the ACLR-Injured Limb group following ACL injury. Examining the angle-angle plots for these measures at Baseline (**Figure 4**) and Follow-Up (**Figure 5**) for the ACLR-Injured Limb group, some inferences about the relative change in coordination can be made. At Baseline, it appears that a majority of the initial transverse plane motion during landing was produced by the knee, indicated by the sharp near vertical trajectory of the plot. Greater knee rotation is present for the plot at Follow-Up as well, but the rise is not as dramatic because of an accompanying increase along the horizontal axis. This indicates that the ACLR-Injured Limb group used more hip rotation during initial landing at Follow-Up. This greater hip rotation is present throughout landing phase at Follow-Up, and is indicated by an increase in the length of the plot along the horizontal axis. We believe that this indicates greater coordination between the hip and knee in the transverse plane, and may represent a loss of independence of knee rotation post ACLR.

Differences in the magnitude of knee rotation among those with ACLR have been previously reported.[14,15,27-32] There has been conflicting observations for alterations in tibial rotation, with several authors observing a decrease in tibial internal rotation[15,27,28] or greater external rotation offset[14,29], and increased tibial internal rotation.[30-32] These studies have included a mix of tasks that have incorporated large transverse plane motion with a pivoting aspect, and more horizontally directed tasks such as ours. Of those that have quantified the amount of tibial rotation over a specified time period[27,28,30-32], the majority have observed an increase in tibial rotation associated with ACLR. The exceptions are Webster et al[27] as they observed a decrease in tibial rotation range of those with ACLR, and Webster and Feller[28] who observed no difference among groups for tibial internal rotation range. This is of importance for considering our results as Webster and Feller[28] analyzed tibial rotation during a single leg horizontal hop and single leg vertical drop, and the other studies included pivoting components that required transverse plane motion to complete. The authors did note that internal rotation range was less for those with ACLR than healthy control groups, but their findings did not reach statistical significance. Examination of the angle-angle plots, our findings appear to contradict these previous findings, as it appears that post-injury there was a tendency for the ACLR-Injured Limb group to land in slightly more knee external rotation, and obtain a greater amount of knee internal rotation as compared to their pre-injury condition. This observation, however, is based on the averaged ensemble plot of the group, and we cannot make any statement as to whether this difference would be significant given the variability between participants. It does appear that the relatively sharp increase in knee internal rotation during the initial stages of landing, are mitigated post-injury by an associate increase in hip rotation as well. This may indicate that the overall transverse plane motion is shared by the knee and hip, and may represent a loss of independence in knee rotation after ACLR. Unfortunately, Webster et al[27] and Webster and Feller[28] did not report hip rotation in their analysis and such information would be of benefit in the future to determine if alterations in tibial rotation they observed are better interpreted relative to the mount of hip rotation as well. There is previous evidence that peak hip internal rotation for those with ACLR is no different than healthy control subjects during a drop jump[33], but may be of little use in the interpretation of our findings as relatively equal amounts of hip internal rotation were obtained prior to and following injury.

The alterations in the coordination of the transverse plane motion of the hip and knee that we observed may be explained as occurring either as result of ACLR or as an adopted movement strategy. Cadaveric studies have previously demonstrated that tibial rotation may not be restored following ACLR resulting in reduced internal tibial rotation during dynamic activities.[34] Tibial rotation has also been demonstrated to be particularly sensitive to femoral tunnel placement, with more oblique tunnel placements allowing for more similar rotational patterns of an intact knee.[35,36] Therefore, the loss of independence in knee internal rotation we observed may be in part due to restrictions on tibial rotation imposed by ACLR. It is, however, also possible that the alterations in coordination represent an adopted movement strategy to increase stability at the knee and avoid loading the reconstructed ACL during dynamic activities. This proposition is supported by the findings of Vairo et al[12] as they observed alteration in muscle activity for those with ACLR associated with increasing stability at the knee.[12]

Another observation that we feel is of importance, is that we observed no other change in the relative coordination of the hip and knee following ACL injury. This is contrary to our initial expectations, particularly for the sagittal plane coordination, as previous studies have identified alterations in sagittal plane coordination for those with ACLR.[17,18] The lack of differences we observed in comparison to previous findings may be attributed to the population we sampled from, the technique we employed to assess coordination, and the task we chose. Our sample may have exhibited differences in coordination of the hip and knee in additional planes had we chosen different tasks to analyze. In addition, a lack of change in coordination among the groups for the other planes of interest does not mean that kinematics were unaltered by ACLR. As this technique assesses the relative motion of one joint to the other in a given plane, it is not sensitive to alterations in kinematics if the change is the same for both the hip and knee. Therefore, following ACL injury the ACL injured group may have demonstrated an increase in medial displacement of the knee, however, if there were a subsequent increase in medial displacement of the hip as well, the average coupling angle would not indicate the change over time.

Our study is not without limitations that must be acknowledged to enhance interpretation of our findings, the first being that we did not control for mechanism of injury, graft selection, concomitant joint injury, or gender, though we attempted to control for this in our statistical model, when identifying ACL injured participants for inclusion in this study. Because of the unique opportunity to compare coordination measures prior to and following ACL injury and the limited population we had to select from, we decided to use relatively open inclusion criteria for our ACL injured groups for fear that strict inclusion criteria would limit our sample group numbers. This increased the heterogeneity of our sample though, and may increase the external validity of our findings to a broader population of those with ACLR. In addition, due to the limitations of our study design there was a period of time that passed between when participants were first tested at Baseline and when injury occurred. Therefore, it is possible that the alterations we observed occurred during this time period and were not the result of ACL injury and ACLR. However, this is highly unlikely due to the lack of changes in the matched control group, and we are confident that the results we observed were due to ACL injury and ACLR alone.

CONCLUSIONS

Following ACL injury and ACLR, coordination of the hip and knee joints in the transverse are altered so that there is a more equal contribution of rotational motion between the hip and knee for the ACL injured limb. This alteration is likely the result of a decrease in the independence of rotation at the knee, and may contribute the progression of knee osteoarthritis for those with ACLR. Those with ACLR may be helped by rehabilitation and intervention strategies aimed at restoring coordination of hip and knee rotation post surgery. Future research should assess techniques to accomplish this.

REFERENCES

1. Orchard J, Seward H, McGivern J, Hood S. Intrinsic and extrinsic risk factors for anterior cruciate ligament injury in Australian footballers. *Am J Sports Med.* Mar-Apr 2001;29(2):196-200.

2. Faude O, Junge A, Kindermann W, Dvorak J. Risk factors for injuries in elite female soccer players. *Br J Sports Med.* Sep 2006;40(9):785-790.

3. Paterno MV, Rauh MJ, Schmitt LC, Ford KR, Hewett TE. Incidence of contralateral and ipsilateral anterior cruciate ligament (ACL) injury after primary ACL reconstruction and return to sport. *Clin J Sport Med.* Mar 2012;22(2):116-121.

4. Salmon L, Russell V, Musgrove T, Pinczewski L, Refshauge K. Incidence and risk factors for graft rupture and contralateral rupture after anterior cruciate ligament reconstruction. *Arthroscopy.* Aug 2005;21(8):948-957.

5. Shelbourne KD, Gray T, Haro M. Incidence of subsequent injury to either knee within 5 years after anterior cruciate ligament reconstruction with patellar tendon autograft. *Am J Sports Med.* Feb 2009;37(2):246-251.

6. Wright RW, Dunn WR, Amendola A, et al. Risk of tearing the intact anterior cruciate ligament in the contralateral knee and rupturing the anterior cruciate ligament graft during the first 2 years after anterior cruciate ligament reconstruction: a prospective MOON cohort study. *Am J Sports Med.* Jul 2007;35(7):1131-1134.

7. von Porat A, Roos EM, Roos H. High prevalence of osteoarthritis 14 years after an anterior cruciate ligament tear in male soccer players: a study of radiographic and patient relevant outcomes. *Ann Rheum Dis.* Mar 2004;63(3):269-273.

8. Lohmander LS, Ostenberg A, Englund M, Roos H. High prevalence of knee osteoarthritis, pain, and functional limitations in female soccer players twelve years after anterior cruciate ligament injury. *Arthritis Rheum.* Oct 2004;50(10):3145-3152.

9. Ernst GP, Saliba E, Diduch DR, Hurwitz SR, Ball DW. Lower extremity compensations following anterior cruciate ligament reconstruction. *Phys Ther.* Mar 2000;80(3):251-260.

10. Decker MJ, Torry MR, Noonan TJ, Riviere A, Sterett WI. Landing adaptations after ACL reconstruction. *Med Sci Sports Exerc.* Sep 2002;34(9):1408-1413.

11. Paterno MV, Ford KR, Myer GD, Heyl R, Hewett TE. Limb asymmetries in landing and jumping 2 years following anterior cruciate ligament reconstruction. *Clin J Sport Med.* Jul 2007;17(4):258-262.

12. Vairo GL, Myers JB, Sell TC, Fu FH, Harner CD, Lephart SM. Neuromuscular and biomechanical landing performance subsequent to ipsilateral semitendinosus and gracilis autograft anterior cruciate ligament reconstruction. *Knee Surg Sports Traumatol Arthrosc.* Jan 2008;16(1):2-14.

13. Salem GJ, Salinas R, Harding FV. Bilateral kinematic and kinetic analysis of the squat exercise after anterior cruciate ligament reconstruction. *Arch Phys Med Rehabil.* Aug 2003;84(8):1211-1216.

14. Tashman S, Collon D, Anderson K, Kolowich P, Anderst W. Abnormal rotational knee motion during running after anterior cruciate ligament reconstruction. *Am J Sports Med.* Jun 2004;32(4):975-983.

15. Deneweth JM, Bey MJ, McLean SG, Lock TR, Kolowich PA, Tashman S. Tibiofemoral joint kinematics of the anterior cruciate ligament-reconstructed knee during a single-legged hop landing. *Am J Sports Med.* Sep 2010;38(9):1820-1828.

16. Gokeler A, Hof AL, Arnold MP, Dijkstra PU, Postema K, Otten E. Abnormal landing strategies after ACL reconstruction. *Scand J Med Sci Sports.* Feb 2010;20(1):e12-19.

17. van Uden CJ, Bloo JK, Kooloos JG, van Kampen A, de Witte J, Wagenaar RC. Coordination and stability of one-legged hopping patterns in patients with anterior cruciate ligament reconstruction: preliminary results. *Clin Biomech (Bristol, Avon).* Jan 2003;18(1):84-87.

18. Kurz MJ, Stergiou N, Buzzi UH, Georgoulis AD. The effect of anterior cruciate ligament reconstruction on lower extremity relative phase dynamics during walking and running. *Knee Surg Sports Traumatol Arthrosc.* Mar 2005;13(2):107-115.

19. Pollard CD, Heiderscheit BC, van Emmerik RE, Hamill J. Gender differences in lower extremity coupling variability during an unanticipated cutting maneuver. *J Appl Biomech.* May 2005;21(2):143-152.

20. Marx RG, Stump TJ, Jones EC, Wickiewicz TL, Warren RF. Development and evaluation of an activity rating scale for disorders of the knee. *Am J Sports Med.* Mar-Apr 2001;29(2):213-218.

21. Roos EM, Roos HP, Lohmander LS, Ekdahl C, Beynnon BD. Knee Injury and Osteoarthritis Outcome Score (KOOS)--development of a self-administered outcome measure. *J Orthop Sports Phys Ther.* Aug 1998;28(2):88-96.

22. Ahlden M, Kartus J, Ejerhed L, Karlsson J, Sernert N. Knee laxity measurements after anterior cruciate ligament reconstruction, using either bone-patellar-tendon-bone or hamstring tendon autografts, with special emphasis on comparison over time. *Knee Surg Sports Traumatol Arthrosc.* Sep 2009;17(9):1117-1124.

23. Bell AL, Pedersen DR, Brand RA. A comparison of the accuracy of several hip center location prediction methods. *J Biomech.* 1990;23(6):617-621.

24. Heiderscheit BC, Hamill J, van Emmerik RE. Variability of stride characteristics and joint coordination among individuals with unilateral patellofemoral pain. *J Appl Biomech.* 2002;18:110-121.

25. Ferber R, Davis IM, Williams DS, 3rd. Effect of foot orthotics on rearfoot and tibia joint coupling patterns and variability. *J Biomech.* Mar 2005;38(3):477-483.

26. Andriacchi TP, Briant PL, Bevill SL, Koo S. Rotational changes at the knee after ACL injury cause cartilage thinning. *Clin Orthop Relat Res.* Jan 2006;442:39-44.

27. Webster KE, Palazzolo SE, McClelland JA, Feller JA. Tibial rotation during pivoting in anterior cruciate ligament reconstructed knees using a single bundle technique. *Clin Biomech (Bristol, Avon).* Dec 10 2011.

28. Webster KE, Feller JA. Tibial rotation in anterior cruciate ligament reconstructed knees during single limb hop and drop landings. *Clin Biomech (Bristol, Avon).* Jan 11 2012.

29. Scanlan SF, Chaudhari AM, Dyrby CO, Andriacchi TP. Differences in tibial rotation during walking in ACL reconstructed and healthy contralateral knees. *J Biomech.* Jun 18 2010;43(9):1817-1822.

30. Chouliaras V, Ristanis S, Moraiti C, Stergiou N, Georgoulis AD. Effectiveness of reconstruction of the anterior cruciate ligament with quadrupled hamstrings and bone-patellar tendon-bone autografts: an in vivo study comparing tibial internal-external rotation. *Am J Sports Med.* Feb 2007;35(2):189-196.

31. Ristanis S, Giakas G, Papageorgiou CD, Moraiti T, Stergiou N, Georgoulis AD. The effects of anterior cruciate ligament reconstruction on tibial rotation during pivoting after descending stairs. *Knee Surg Sports Traumatol Arthrosc.* Nov 2003;11(6):360-365.

32. Ristanis S, Stergiou N, Patras K, Vasiliadis HS, Giakas G, Georgoulis AD. Excessive tibial rotation during high-demand activities is not restored by anterior cruciate ligament reconstruction. *Arthroscopy.* Nov 2005;21(11):1323-1329.

33. Ortiz A, Olson S, Libby CL, et al. Landing mechanics between noninjured women and women with anterior cruciate ligament reconstruction during 2 jump tasks. *Am J Sports Med.* Jan 2008;36(1):149-157.

34. Yoo JD, Papannagari R, Park SE, DeFrate LE, Gill TJ, Li G. The effect of anterior cruciate ligament reconstruction on knee joint kinematics under simulated muscle loads. *Am J Sports Med.* Feb 2005;33(2):240-246.

35. Scopp JM, Jasper LE, Belkoff SM, Moorman CT, 3rd. The effect of oblique femoral tunnel placement on rotational constraint of the knee reconstructed using patellar tendon autografts. *Arthroscopy.* Mar 2004;20(3):294-299.

36. Loh JC, Fukuda Y, Tsuda E, Steadman RJ, Fu FH, Woo SL. Knee stability and graft function following anterior cruciate ligament reconstruction: Comparison between 11 o'clock and 10 o'clock femoral tunnel placement. 2002 Richard O'Connor Award paper. *Arthroscopy.* Mar 2003;19(3):297-304.

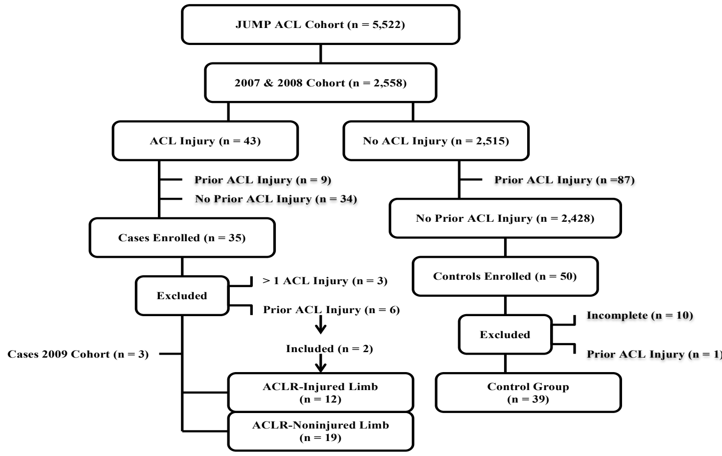

Figure 1. Description of the JUMP ACL Cohorts of 2007 and 2008, as well as participant enrollment

Figure 2. Double Leg Jump Landing. Participants were required to stand atop a box located a distance equal to one half of their body height from the front edge of the force plate, jump forward, land with their foot completely on the force plate, and then immediately make a vertical jump for maximum height.

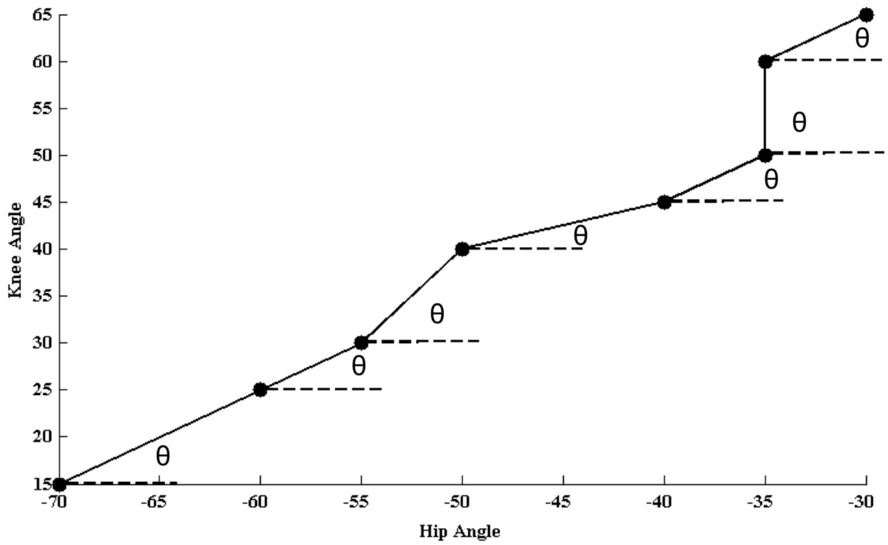

Figure 3. Graphic representation depicting the calculation of coupling angles between subsequent points of an angle-angle plot for the hip and knee.

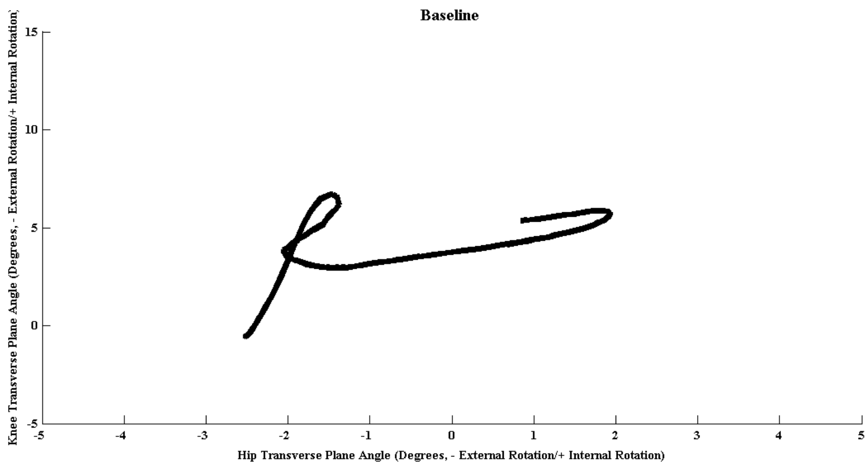

Figure 4. Angle-Angle plot of Hip Transverse Plane - Knee Transverse Plane for the ACLR-Injured Limb group at Baseline.

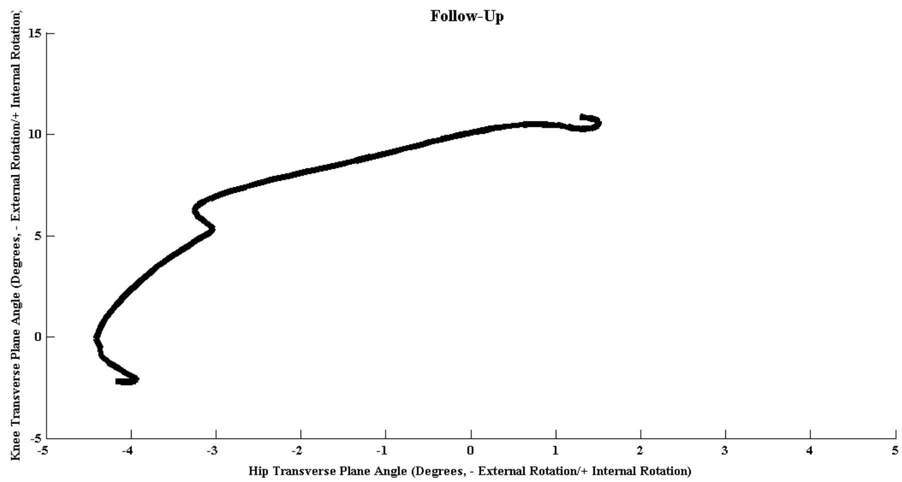

Figure 5. Angle-Angle plot of Hip Transverse Plane - Knee Transverse Plane for the ACLR-Injured Limb group at Follow-Up.

Table 1. Participant demographics and antrhopometrics. Units of measure are Age (years), Height (cm), and Mass (kg). Values represent means ± standard deviation.

		Baseline			Follow-Up		
	n	Age	Height	Mass	Age	Height	Mass
ACLR-INJ	12 (8 m, 4 f)	18.64 ± 0.50	174.10 ± 7.31	72.64 ± 9.48	21.42 ± 0.79	174.29 ± 7.56	76.25 ± 9.95
ACLR-NINJ	19 (9 m, 10 f)	18.52 ± 0.58	170.06 ± 9.26	68.99 ± 10.93	21.47 ± 0.77	170.05 ± 9.13	72.87 ± 12.78
Control	38 (19 m, 19 f)	18.47 ± 0.46	172.05 ± 8.65	69.16 ± 11.47	20.95 ± 0.73	172.16 ± 8.71	72.35 ± 12.37

Table 2. Group chronological descriptive statistics (Mean ± Standard deviation, Days)

	Baseline to Follow-Up	Baseline to Injury	Injury to Surgery	Surgery to Follow-Up
ACLR-Injured Limb	1,074.42 ± 197.28	367.73 ± 156.06	33.70 ± 20.29	666.90 ± 209.24
ACLR-Noninjured Limb	1,247.68 ± 179.04	533.33 ± 267.97	40.39 ± 24.92	691.06 ± 243.01
Control	1,071.76 ± 179.00	-	-	-

Table 3. Descriptive statistics for bilateral difference of anterior knee laxity assessed using KT-1000, Marx Activity score, and KOOS for each group. Values represent mean ± standard deviation, and units of measure other than anterior laxity (mm) are scale dependent.

			KOOS				
	Bilateral Difference Anterior Laxity (mm)	Marx Activity Total Score	Pain	Symptom	ADL	Sport/Rec	QOL
ACLR-INJ	1.9 ± 1.1	13.92 ± 2.75	84.75 ± 9.97	71.50 ± 13.16	93.00 ± 7.20	80.42 ± 16.16	65.63 ± 20.90
ALCR-NINJ	2.5 ± 1.8	11.21 ± 4.57	86.95 ± 11.62	73.42 ± 15.85	96.74 ± 5.05	78.42 ± 17.72	71.71 ± 22.57
Control	1.1 ± 0.8	10.92 ± 2.95	96.24 ± 5.12	92.63 ± 8.81	98.79 ± 2.11	93.95 ± 10.41	91.12 ± 14.14

*Laxity measures were not obtained for 1 member of the ACLR-Noninjured Limb group

Table 4. Average Coupling Angle (Mean, SD, 95th Confidence Interval) for the Control, ACLR-Noninjured Limb Group, and ACLR-Injured Limb Group at Baseline (Pre) and Follow-Up (Post)

		Control		ACLR-Noninjured Limb		ACLR-Injured Limb	
		Mean ± SD	95TH CI	Mean ± SD	95TH CI	Mean ± SD	95TH CI
Hip & Knee Sagittal Plane	PRE	53.76 ± 6.75	(51.57, 55.95)	54.76 ± 6.75	(51.67, 57.85)	53.99 ± 6.79	(50.08, 57.91)
	POST	52.48 ± 8.90	(49.60, 55.36)	51.64 ± 8.90	(47.56, 55.71)	55.32 ± 8.96	(50.16, 60.49)
Hip & Knee Frontal Plane	PRE	43.23 ± 7.29	(40.86, 45.59)	43.42 ± 7.30	(40.07, 46.76)	44.31 ± 7.34	(40.07, 48.54)
	POST	42.26 ± 7.42	(39.86, 44.67)	41.29 ± 7.43	(37.89, 44.70)	44.52 ± 7.48	(40.21, 48.83)
Hip & Knee Transverse Plane	PRE	52.76 ± 6.91	(50.52, 55.00)	54.98 ± 6.91	(51.81, 58.14)	54.85 ± 6.96	(50.84, 58.86)
	POST	54.78 ± 6.59	(52.64, 56.92)	53.46 ± 6.60	(50.44, 56.48)	47.97 ± 6.64	(44.15, 51.80)
Hip Frontal & Knee Transverse Plane	PRE	49.15 ± 7.67	(46.67, 51.64)	52.58 ± 7.67	(49.06, 56.09)	51.66 ± 7.72	(47.21, 56.11)
	POST	49.63 ± 6.04	(47.67, 51.58)	49.05 ± 6.04	(46.29, 51.82)	45.42 ± 6.07	(41.92, 48.92)
Hip Transverse & Knee Frontal Plane	PRE	47.51 ± 6.57	(45.38, 49.64)	46.45 ± 6.58	(43.44, 49.47)	48.35 ± 6.62	(44.53, 52.16)
	POST	47.49 ± 8.05	(44.88, 50.10)	45.99 ± 8.06	(42.29, 49.68)	47.71 ± 8.11	(43.04, 52.38)

*Values for descriptive statistics are based on Gender entered as a covariate in the statistical model at a value of 0.52

APPENDIX D. MANUSCRIPT 3

A Comparison of Lower Extremity Biomechanical Asymmetry of Those With and With Out Anterior Cruciate Ligament Reconstruction: The JUMP ACL Study

(British Journal of Sports Medicine)

ABSTRACT

Background: Those with ACLR are at an increased risk for subsequent ACL injury. Asymmetries in lower extremity biomechanics have been previously observed for these individuals, and may influence their risk for reinjury. Few studies though have compared asymmetry to that present in healthy controls.

Aim: The purpose of this study was to quantify and compare asymmetry in lower extremity biomechanics during a double leg jump landing task for those with and without ACLR.

Methods: Sixty-three participants from the JUMP ACL cohort, 24 participants with ACLR and 39 without completed a double leg jump landing task. Bilateral lower extremity biomechanics were collected during Initial Ground Contact and Landing Phase of the task. The absolute values of difference scores were used to quantify asymmetry for each measure of interest. Average group values were compared using two-sample Kolmogorov-Smirnov tests.

Results: Between group differences in peak internal knee flexion moment and peak vertical ground reaction force were observed, with the ACLR group demonstrating greater between-limb asymmetry. Follow up analysis of mean values for the injured and noninjured limb of the ACLR group demonstrated increased peak internal knee flexion

moment in the injure limb, and increased peak vertical ground reaction force for the noninjured limb. No other between group differences were observed.

Conclusion: The increased difference in loading between the injured and noninjured limb in the absence of asymmetry in kinematics suggests that those with ACLR may employ strategies to reduce loading of the injured limb that are not obvious by assessing movement patterns alone.

INTRODUCTION

Reconstructive surgery following anterior cruciate ligament injury (ACLR) does not guarantee that an individual will be able to return to their same level of physical activity[1,2] or protect against the onset of osteoarthritis.[3,4] In addition, those with ACLR are at an increased risk for a secondary ACL injury as compared to those with no ACL injury.[5-7] The incidence of reinjury for those with a history of ACLR when they return to physical activity is between 6-25%[7-10], and the risk for reinjury is 5-15 times higher as compared to those who have no history of ACL injury.[5-7] In addition, the risk for reinjury does not appear to be isolated to the previously injured limb.[10] Therefore, there is a need for a better understanding of how to improve outcomes following ACLR, particularly for those in this population that wish to remain physically active or engage in sport activities.

Differences in lower extremity biomechanics for those with ACLR have been previously identified. They generally present with differences in the movement and loading of their limbs when compared to those who have never suffered an ACL injury.[11-14] Not only do ACLR patients perform activities differently than healthy individuals but also display differences or asymmetry between injured and contralateral limb following injury[13-17], which provides a rationale for why risk for reinjury is not limited to the previously injured limb.[18] However, few studies have examined asymmetry for those with ACLR in comparison to asymmetry that is present in a healthy noninjured group of individuals. [11,13,17,19] Only Gokeler et al[17] directly quantified and compared asymmetry in individuals with ACLR to a healthy control group during a single leg forward hop, but were unable to make comparisons in biomechanical data because of a limited number of patients. This lack of information with regards to asymmetry present in healthy individuals makes it difficult to know the true importance of asymmetry for the poor outcomes and reinjury associated with ACLR.

This may ultimately lead to better identification and interventions, reducing the risk for injury in this population, as asymmetry related to specific movement and loading patterns has been previously identified as a prospective risk factor for ACL injury in physically active adolescents with ACLR.[20] Therefore, the purpose of this study was to quantify and compare biomechanical asymmetry in a group of healthy, physically active individuals with and without ACLR.

METHODS

Participants

Participants for this study were recruited from the Joint Undertaking to Monitor and Prevent ACL Injury (JUMP ACL) Project, a multi-year prospective study conducted with members of the United States' service academies, including the United States Air Force Academy, the United States Military Academy, and the United States Naval Academy. The purpose of this project is to identify risk factors for noncontact ACL injury. Members of each incoming class between 2004 and 2009 enrolled in the study, and were prospectively followed to identify risk factors for ACL injury. We limited potential enrollment in this study to members of the 2007, 2008, and 2009 cohorts. Participants that had injured their ACL during enrollment in the JUMP ACL study, as well as a group of noninjured controls were matched based on cohort year, service academy, and sex. We attempted to achieve a 2:1 control to case ratio for our sample. Seventy-three participants were enrolled in this study, 29 with ACLR (12 females, 17 males) and 44 healthy matched controls (22 females, 22 males).

Procedures

Prior to data collection, each participant read and signed an informed consent, had anthropometric and demographic information recorded. In addition each participant was asked to complete the Marx Activity Scale[21], Knee Injury and Osteoarthritis Outcome Score (KOOS)[22], and had anterior knee laxity assessed bilaterally.[23] Each participant then performed a double leg jump landing (**Figure 1**). This task required participants to stand atop a 30 cm high box located a distance from the edge of a force plate equal to half their height, jump forward from the box, landing with one foot completely on the force plate, and upon landing immediately make a maximal effort vertical jump. Trials were excluded and repeated if participants didn't land with their foot completely on the force plate, the foot of the other limb made contact with the force plate, or if the participant hesitated between the landing and vertical jump phases of the task. We were limited to one force plate for this analysis, to collect kinetic data on both limbs participants alternated between the right and left leg landing on the force plate for each subsequent trial. The order of alternating was counterbalanced between participants. At least 5 successful trials per limb were performed to ensure that a minimum of three successful trials was collected for each limb per participant. Biomechanical data were collected using an electromagnetic tracking system (Ascension Technologies Inc., Burlington, VT) integrated with a non-conductive force plate (Bertec Co., Columbus, OH). All data collection was conducted using the Motion Monitor Software (Innovative Sports Training, Inc., Chicago, IL)

Prior to completion of the double leg jump landing, all participants were instrumented with electromagnetic sensors.[24,25] The sensors were affixed to each leg, so that one was placed on each shank and thigh, and one was placed on the pelvis. The shank sensors were placed on the skin overlying the medial tibia, the thigh sensor on the mid portion of the lateral thigh and the pelvis sensor was affixed to the sacrum at the mid point between the posterior superior iliac spines. Each sensor was adhered to the participant's skin using double-sided tape with athletic tape placed over it. The pelvis sensor was held in place using double sided tape and an elastic belt. Sensor placement and procedures were performed to minimize motion artifact during the jump landing.

Following placement of the electromagnetic sensors, a model of the lower extremities was constructed through digitization of the medial and lateral malleoli, medial and lateral femoral epicondyles, and the anterior superior iliac spines using a moveable sensor.[24,25] The midpoints of these landmarks were used to define the ankle and knee joint centers, respectively. The hip joint centers were defined based on the Bell Method.[26] Shank, thigh, and pelvis segments were defined based on the joints as segment endpoints and the respective electromagnetic sensors as a third non-collinear point. Local right-handed segment axis systems were embedded into each segment, the orientation of which coincided with the global axis system, positive z-axis along the vertical direction, positive x-axis along the anterior direction of the participant, and the positive y-axis along a vector with a 90° positive rotation about the z-axis relative to the positive x-axis.

Data Analysis

All kinematic and kinetic data were sampled at a frequency of 144 Hz and 1,444 Hz, respectively. Each trial was visually inspected within the Motion Monitor software to identify any trials that included data collection errors that were not identified at the time of data collection. Errors of interest included excessive motion artifact of sensors at the time of contact with the force plate when landing, improperly constructed biomechanical models, and low vertical ground reaction force measures indicative of incomplete foot contact with the force plate. We then identified three trials for each limb per participant for data analysis.

Kinematic and kinetic values for the sagittal, frontal, and transverse plane the hip and knee were analyzed for this study. Measures of vertical ground reaction force and anterior tibial shear force, defined as the resultant anterior force located at the knee, were also included. Kinematics of the hip were defined as the thigh segment relative to the pelvis, and kinematics of the knee as the shank segment relative to the thigh. An Euler sequence of Y, X, Z was used to define orientation of the relative segments and calculate hip and knee angles. All moment data were calculated, and are reported, as internal moments normalized to a product of body height (m) and weight (N). Vertical ground reaction force and anterior tibial shear force data were normalized to body weight. All kinematic data were filtered using a 4^{th} order Butterworth filter (14.5 Hz) prior to data exportation, and all biomechanical measures were calculated within the Motion Monitor software.

Biomechanical asymmetries at the time points of Initial Ground Contact and the Landing Phase of the double leg jump landing were analyzed for this study. Initial Ground Contact was defined as the first time point that the vertical ground reaction force exceeded 10 N, and the Landing Phase as the time frame from Initial Ground Contact to the time point coinciding with peak knee flexion. Peak 3D joint angles and moments as well as anterior tibial shear force and vertical ground reaction force were calculated for the Landing Phase. Values were averaged across three trials or each limb per participant. All data reduction procedures were conducted using a customized MATLAB program (Mathworks, Inc., Natick, MA).

Asymmetry values were calculated as the absolute difference between limbs for each dependent variable. This method was chosen because we were only interested in comparing the relative magnitude of between limb differences for each group. This method is also helpful as it avoids having to determine how to match limbs for the Control group and ACLR group, and prevents any complications with sign conventions denoting direction for kinematics and kinematics. Some detail can be lost performing the comparison this way, but if differences were found between groups the raw values were analyzed to provide a description of the relative between limb biomechanics.

Procedures to assess the normality of data were conducted for each variable of interest to ensure the proper use of parametric statistics and to identify potential statistical outliers. This included calculation of Z-scores for the skewness and kurtosis of each variable. Those variables with a value greater than 1.96 were further analyzed by constructing box plots to identify potential statistical outliers. This assessment procedure indicated that the data did not meet the assumptions of normality for parametric tests. Between group differences (ACLR v. Control) were determined using two-sample Kolmogorov-Smirnov tests. An a priori alpha level of 0.05 was set, and all statistical procedures were conducted using IBM SPSS v19 (SPSS, Inc., an IBM company, Chicago, IL).

RESULTS

Participant Demographics

Demographic and anthropometric data describing each group is located in **Table 1** and **Table 2**. Ten participants were eliminated from data analysis (ACLR group: 5 (2 females, 3 males), Control group: 5 (3 females, 2 males) because they had fewer than 3 acceptable trials for one or both limbs following inspection. This left a total of 63 participants for the final analysis (ACLR group: n=24, Control group: n=39). For the ACLR, 8 participants had a bone-patella tendon-bone autograft, 8 had a hamstrings autograft, and 1 had an Achilles tendon allograft. We were unable to obtain information for graft type for 7 participants. Those with ACLR were on average 1.89 ± 0.68 years post surgery, but we were unable to obtain date of surgery information for 3 participants of the ACLR group.

No between group differences for asymmetry in kinematics or kinetics were observed between groups at Initial Ground Contact (**Table 3**), and no between group differences for asymmetry in kinematics were identified during the Landing Phase (**Table 4**).

We observed an increased asymmetry in peak internal knee flexion moment for the ACLR group (Mdn = 0.040) as compared to the Control group (Mdn = 0.026) during the Landing Phase (Z_{KS} = 1.42, p = 0.035). The ACLR group also had increased asymmetry in peak vertical ground reaction force (Mdn = 0.68) as compared to the Control group (Mdn = 0.46) during the Landing Phase (Z_{KS} = 1.45, p = 0.031). Examining the mean values for each variable as a follow up analysis to better describe the asymmetries for the ACLR group, they had greater peak internal knee flexion moment for the injured limb (0.058 ± 0.032 Nm/BWxBH) as compared to the noninjured limb (0.053 ± 0.031

Nm/BWxBH). This equaled 9.4% greater peak internal knee flexion moment for the injured limb as compared to the noninjured. However, the noninjured limb (2.97 ± 0.069 N/BW) had greater peak vertical ground reaction force as compared to the injured limb (2.55 ± 0.80 N/BW) for the ACLR group, equaling 16.5% greater peak vertical ground reaction force on the noninjured limb. No other differences in asymmetry of the peak kinetic variables during the Landing Phase were observed (**Table 5**).

DISCUSSION

The main finding we observed for this study was increased asymmetry in peak knee flexion moment and peak vertical ground reaction force for the ACLR group compared to the healthy group during the landing phase. Follow up analysis of the values for the injured and noninjured limbs provided evidence that the asymmetry for these factors in the ACLR group was driven by an increased peak knee flexion moment for the injured limb and an increase in peak vertical ground reaction force for the noninjured limb.

The increased peak internal knee flexion moment we observed for the injured limb of the ACLR group is likely an attempt to decrease loading of the reconstructed ACL by increasing use of the hamstring muscles. Previous work has demonstrated that hamstrings force can decrease anterior tibial translation and potential ACL loading.[27,28] Similar attempts to decrease loading of the reconstructed limb for those with ACLR have been previously observed, however, these attempts have been produced with the involved limb demonstrating decreases in external knee flexion or internal knee extension moment.[11,29,30] Deficits in knee extension moment for those with ACLR has been characterized as a 'quadriceps avoidance' by Berchuck et al[31] in which they observed an increase in internal knee flexion moment for those with ACL deficiency during gait. Our observations occurred with those with ACLR and during a task that requires primarily sagittal plane motion and loading. We did not observe asymmetry in peak internal knee extension moment nor asymmetry in anterior tibial shear force, and if we had considered these variables alone to assess sagittal plane loading at the knee we would have observed no difference in asymmetry for the ACLR group. Therefore, it may be worth considering internal knee flexion moment in addition to other variables to characterize residual asymmetries in knee loading.

Our findings of asymmetry in peak vertical ground reaction force are supported by previous studies that have identified increased vertical ground reaction force for the noninjured limb for those with ACLR.[13,14,32] Both Vairo et al[14] and Nyland et al[32] observed similar results with decreased loading of the injured limb relative to the noninjured for those with ACLR when performing single leg jump landing tasks. When compared to a control group though, the difference observed by Vairo et al[14] was suggested to be the result of decreased loading of the injured limb, rather than increased loading of the noninjured limb as we have observed. Our results are very similar to those of Paterno et al[13] as they observed increased vertical ground reaction force for the noninjured limb when compared to the injured limb as well as to a control group during a double leg drop vertical jump. They observed a relative difference in peak vertical ground reaction force of approximately 33% greater loading of the noninjured limb. Our observation of 16.5% is roughly half of this between limb difference. This difference, however, may be explained in part by the use of females only for Paterno et al[13], as females with ACLR have been previously shown to demonstrate differences in vertical ground reaction force as compared to males with ACLR.[32] These previous studies restricted their sampled ACLR group to a specific graft type though; ipsilateral semitendinosous and gracilis for Vairo et al[14], and patellar tendon bone-tendon-bone for Paterno et al[13]. It is possible that the differences between these studies was in part the result of graft type as differences in knee biomechanics have been previously reported for graft type during single leg landing.[29] However, Webster et al[29] did not observe differences in peak vertical ground reaction force between grafts for single leg landings.

Another possibility is that loading for those with ACLR may be dependent on the task, and different strategies may be used to minimize loading of the injured knee. During a single leg landing on the injured limb those with ACLR may alter their movement to reduce loading. This notion is supported in part by the observation of a decreased peak vertical ground reaction force being accompanied by an increased hip flexion observed by Vairo et al[14] when landing on the injured limb. We observed no asymmetry in kinematics during a double leg jump landing, and suggest that instead of using altered movement strategies to reduce loading of the injured limb, they may have chose to shift a greater portion of the vertical ground reaction force to the noninjured limb.

This has important implications, the first being that this strategy may increase risk for a second ACL injury as an increase in peak vertical ground reaction force has been previously identified as a prospective risk factor for noncontact ACL injury.[33] So, in attempting to reduce loading of the injured limb, they may employ a faulty movement strategy that increases their risk for ACL injury. The observation of these altered loading patterns in the absence of asymmetry in kinematics has implications for clinicians as well. It is reasonably simple for clinicians to identify altered movement strategies or asymmetries in movement that may represent dysfunction in those with ACLR. Our results though suggest that movement dysfunction may be present without accompanying asymmetry in kinematics. It may be useful for clinicians to employ methods to identify altered loading when implementing intervention programs for those with ACLR.

Our study is not without limitations that must be consider when interpreting the results. Limitations in our data collection capabilities prevented us from being able to analyze asymmetry in lower extremity biomechanics from the same trials. It is possible that any differences, or lack of differences we observed, may have been due in part to how the individual performed those particular trials and not representative of how the two limbs moved in unison. Also, for the purposes of this study we were solely interested in quantifying asymmetry between limbs. It is possible that we would have observed group main effects had we employed another strategy such as an ANOVA model. Differences in proportion of males and females in our groups may have influence results, as gender differences in biomechanics have been previously reported for those with ACLR.[32,34] Given the nature of the data, we were unable to employ methods to account for this group discrepancy and must accept it as a limitation. Lastly, our inclusion criteria were rather lenient for our sample selection for the ACLR group. Because of limited time and access to the cohort we chose to investigate, we decided to adopt these criteria to ensure adequate numbers for participant enrollment in the ACLR group. However, the heterogeneity of our sample may increase the external validity of our findings.

In summary, we observed greater asymmetry in sagittal plane loading and vertical ground reaction force for those with ACLR as compared to control groups. These greater asymmetries in loading were present despite no difference in asymmetry among the hip or knee kinematics. Our findings, in combination with previous observations, provide evidence that those with ACLR may employ methods to decrease loading of the injured knee during a double leg jump landing. However, the methods they employ to achieve this may increase their risk for a subsequent ACL injury. In addition, these asymmetries in loading may not be easily detected by clinicians, as they are not accompanied by increased asymmetry between limbs.

What Is Already Known About This Topic:

- **Those with ACLR are at an increased risk for reinjury.**
- **Asymmetries in lower extremity biomechanics are present following ACLR and may contribute to the increased risk for reinjury.**

What This Study Adds:

- **Greater asymmetries in loading are present for those with ACLR as compared to healthy individuals.**
- **Asymmetries in loading may be present despite no difference in asymmetry of kinematics.**

REFERENCES

1. Spindler KP, Huston LJ, Wright RW, et al. The prognosis and predictors of sports function and activity at minimum 6 years after anterior cruciate ligament reconstruction: a population cohort study. *Am J Sports Med.* Feb 2011;39(2):348-359.

2. Ardern CL, Taylor NF, Feller JA, Webster KE. Return-to-sport outcomes at 2 to 7 years after anterior cruciate ligament reconstruction surgery. *Am J Sports Med.* Jan 2012;40(1):41-48.

3. von Porat A, Roos EM, Roos H. High prevalence of osteoarthritis 14 years after an anterior cruciate ligament tear in male soccer players: a study of radiographic and patient relevant outcomes. *Ann Rheum Dis.* Mar 2004;63(3):269-273.

4. Lohmander LS, Ostenberg A, Englund M, Roos H. High prevalence of knee osteoarthritis, pain, and functional limitations in female soccer players twelve years after anterior cruciate ligament injury. *Arthritis Rheum.* Oct 2004;50(10):3145-3152.

5. Orchard J, Seward H, McGivern J, Hood S. Intrinsic and extrinsic risk factors for anterior cruciate ligament injury in Australian footballers. *Am J Sports Med.* Mar-Apr 2001;29(2):196-200.

6. Faude O, Junge A, Kindermann W, Dvorak J. Risk factors for injuries in elite female soccer players. *Br J Sports Med.* Sep 2006;40(9):785-790.

7. Paterno MV, Rauh MJ, Schmitt LC, Ford KR, Hewett TE. Incidence of contralateral and ipsilateral anterior cruciate ligament (ACL) injury after primary ACL reconstruction and return to sport. *Clin J Sport Med.* Mar 2012;22(2):116-121.

8. Salmon L, Russell V, Musgrove T, Pinczewski L, Refshauge K. Incidence and risk factors for graft rupture and contralateral rupture after anterior cruciate ligament reconstruction. *Arthroscopy.* Aug 2005;21(8):948-957.

9. Shelbourne KD, Gray T, Haro M. Incidence of subsequent injury to either knee within 5 years after anterior cruciate ligament reconstruction with patellar tendon autograft. *Am J Sports Med.* Feb 2009;37(2):246-251.

10. Wright RW, Dunn WR, Amendola A, et al. Risk of tearing the intact anterior cruciate ligament in the contralateral knee and rupturing the anterior cruciate ligament graft during the first 2 years after anterior cruciate ligament reconstruction: a prospective MOON cohort study. *Am J Sports Med.* Jul 2007;35(7):1131-1134.

11. Ernst GP, Saliba E, Diduch DR, Hurwitz SR, Ball DW. Lower extremity compensations following anterior cruciate ligament reconstruction. *Phys Ther.* Mar 2000;80(3):251-260.

12. Decker MJ, Torry MR, Noonan TJ, Riviere A, Sterett WI. Landing adaptations after ACL reconstruction. *Med Sci Sports Exerc.* Sep 2002;34(9):1408-1413.

13. Paterno MV, Ford KR, Myer GD, Heyl R, Hewett TE. Limb asymmetries in landing and jumping 2 years following anterior cruciate ligament reconstruction. *Clin J Sport Med.* Jul 2007;17(4):258-262.

14. Vairo GL, Myers JB, Sell TC, Fu FH, Harner CD, Lephart SM. Neuromuscular and biomechanical landing performance subsequent to ipsilateral semitendinosus and gracilis autograft anterior cruciate ligament reconstruction. *Knee Surg Sports Traumatol Arthrosc.* Jan 2008;16(1):2-14.

15. Salem GJ, Salinas R, Harding FV. Bilateral kinematic and kinetic analysis of the squat exercise after anterior cruciate ligament reconstruction. *Arch Phys Med Rehabil.* Aug 2003;84(8):1211-1216.

16. Tashman S, Collon D, Anderson K, Kolowich P, Anderst W. Abnormal rotational knee motion during running after anterior cruciate ligament reconstruction. *Am J Sports Med.* Jun 2004;32(4):975-983.

17. Gokeler A, Hof AL, Arnold MP, Dijkstra PU, Postema K, Otten E. Abnormal landing strategies after ACL reconstruction. *Scand J Med Sci Sports.* Feb 2010;20(1):e12-19.

18. Brophy RH, Wright RW, Powell JW, Matava MJ. Injuries to kickers in American football: the National Football League experience. *Am J Sports Med.* Jun 2010;38(6):1166-1173.

19. Neitzel JA, Kernozek TW, Davies GJ. Loading response following anterior cruciate ligament reconstruction during the parallel squat exercise. *Clin Biomech (Bristol, Avon).* Aug 2002;17(7):551-554.

20. Paterno MV, Schmitt LC, Ford KR, et al. Biomechanical measures during landing and postural stability predict second anterior cruciate ligament injury after anterior cruciate ligament reconstruction and return to sport. *Am J Sports Med.* Oct 2010;38(10):1968-1978.

21. Marx RG, Stump TJ, Jones EC, Wickiewicz TL, Warren RF. Development and evaluation of an activity rating scale for disorders of the knee. *Am J Sports Med.* Mar-Apr 2001;29(2):213-218.

22. Roos EM, Roos HP, Lohmander LS, Ekdahl C, Beynnon BD. Knee Injury and Osteoarthritis Outcome Score (KOOS)--development of a self-administered outcome measure. *J Orthop Sports Phys Ther.* Aug 1998;28(2):88-96.

23. Ahlden M, Kartus J, Ejerhed L, Karlsson J, Sernert N. Knee laxity measurements after anterior cruciate ligament reconstruction, using either bone-patellar-tendon-bone or hamstring tendon autografts, with special emphasis on comparison over time. *Knee Surg Sports Traumatol Arthrosc.* Sep 2009;17(9):1117-1124.

24. Padua DA, Marshall SW, Boling MC, Thigpen CA, Garrett WE, Jr., Beutler AI. The Landing Error Scoring System (LESS) Is a valid and reliable clinical assessment tool of jump-landing biomechanics: The JUMP-ACL study. *Am J Sports Med.* Oct 2009;37(10):1996-2002.

25. Boling MC, Padua DA, Marshall SW, Guskiewicz K, Pyne S, Beutler A. A prospective investigation of biomechanical risk factors for patellofemoral pain syndrome: the Joint Undertaking to Monitor and Prevent ACL Injury (JUMP-ACL) cohort. *Am J Sports Med.* Nov 2009;37(11):2108-2116.

26. Bell AL, Pedersen DR, Brand RA. A comparison of the accuracy of several hip center location prediction methods. *J Biomech.* 1990;23(6):617-621.

27. Li G, Rudy TW, Sakane M, Kanamori A, Ma CB, Woo SL. The importance of quadriceps and hamstring muscle loading on knee kinematics and in-situ forces in the ACL. *J Biomech.* Apr 1999;32(4):395-400.

28. Markolf KL, O'Neill G, Jackson SR, McAllister DR. Effects of applied quadriceps and hamstrings muscle loads on forces in the anterior and posterior cruciate ligaments. *Am J Sports Med.* Jul-Aug 2004;32(5):1144-1149.

29. Webster KE, Gonzalez-Adrio R, Feller JA. Dynamic joint loading following hamstring and patellar tendon anterior cruciate ligament reconstruction. *Knee Surg Sports Traumatol Arthrosc.* Jan 2004;12(1):15-21.

30. Ortiz A, Olson S, Libby CL, et al. Landing mechanics between noninjured women and women with anterior cruciate ligament reconstruction during 2 jump tasks. *Am J Sports Med.* Jan 2008;36(1):149-157.

31. Berchuck M, Andriacchi TP, Bach BR, Reider B. Gait adaptations by patients who have a deficient anterior cruciate ligament. *J Bone Joint Surg Am.* Jul 1990;72(6):871-877.

32. Nyland J, Klein S, Caborn DN. Lower extremity compensatory neuromuscular and biomechanical adaptations 2 to 11 years after anterior cruciate ligament reconstruction. *Arthroscopy.* Sep 2010;26(9):1212-1225.

33. Hewett TE, Myer GD, Ford KR, et al. Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study. *Am J Sports Med.* Apr 2005;33(4):492-501.

34. Webster KE, Feller JA. Tibial rotation in anterior cruciate ligament reconstructed knees during single limb hop and drop landings. *Clin Biomech (Bristol, Avon).* Jan 11 2012.

Table 1. Participant demographics and antrhopometrics. Units of measure are Age (years), Height (cm), and Mass (kg). Values represent means ± standard deviation.

	n	Age	Height	Mass
ACLR	24 (14 m, 10 f)	21.58 ± 0.78	172.01 ± 8.85	74.69 ± 12.50
Control	39 (20 m, 19 f)	21.00 ± 0.77	172.25 ± 8.94	72.27 ± 13.72

Table 2. Descriptive statistics (Mean ± SD) of ACLR and Control groups for bilateral difference in anterior knee laxity, Marx activity score, and KOOS scores

			KOOS				
	Bilateral Difference Anterior Laxity (mm)	Marx Activity Total Score	Pain	Symptom	ADL	Sport/Rec	QOL
ACLR	2.4 ± 1.7	12.08 ± 4.43	87.15 ± 10.98	72.17 ± 14.71	96.20 ± 5.41	79.58 ± 18.17	69.53 ± 22.75
Control	1.2 ± 1.0	10.87 ± 3.16	95.37 ± 6.29	92.49 ± 8.89	98.23 ± 3.19	92.44 ± 11.52	90.87 ± 14.04

Table 3. Asymmetry of kinematics (°), moments (Nm/BHxBW) and kinetics (N/BW) for groups at Initial Ground Contact

	ACLR			CONTROL			Z_{KS}	P value
	Mean	(SD)	Median	Mean	(SD)	Median		
Knee Sagittal	6.982	(4.439)	6.179	7.337	(5.505)	6.260	0.568	0.903
Knee Frontal	8.484	(6.339)	6.350	7.970	(5.686)	6.348	0.432	0.993
Knee Transverse	7.120	(4.818)	6.154	6.340	(5.211)	5.444	0.507	0.960
Hip Sagittal	7.571	(4.189)	7.475	7.213	(5.464)	5.944	0.630	0.822
Hip Frontal	11.271	(9.178)	8.129	13.047	(9.431)	12.793	0.690	0.725
Hip Transverse	8.858	(6.502)	7.567	9.423	(7.653)	7.617	0.395	0.998
Knee Sagittal Moment	0.025	(0.023)	0.021	0.026	(0.022)	0.020	0.432	0.992
Knee Frontal Moment	0.021	(0.022)	0.016	0.023	(0.015)	0.021	0.939	0.341
Knee Transverse Moment	0.011	(0.008)	0.008	0.010	(0.008)	0.008	1.075	0.198
Hip Sagittal Moment	0.070	(0.048)	0.058	0.070	(0.058)	0.059	0.729	0.663
Hip Frontal Moment	0.058	(0.047)	0.045	0.061	(0.041)	0.055	0.618	0.840
Hip Transverse Moment	0.032	(0.027)	0.026	0.021	(0.015)	0.019	0.791	0.559
Anterior Tibial Shear Force	0.109	(0.083)	0.080	0.114	(0.097)	0.094	0.704	0.704
Vertical Ground Reaction Force	0.024	(0.028)	0.017	0.031	(0.028)	0.024	0.988	0.283

Table 4. Asymmetry of peak kinematics (°) during Landing Phase

	ACLR			CONTROL			Z_{KS}	P value
	Mean (SD)		Median	Mean (SD)		Median		
Knee Flexion	8.836	(6.409)	7.181	9.857	(6.604)	9.193	0.704	0.704
Knee Extension	6.982	(4.439)	6.179	7.337	(5.505)	6.260	0.568	0.903
Knee Varus	8.643	(8.499)	6.876	9.974	(6.573)	8.477	1.129	0.075
Knee Valgus	9.939	(9.767)	5.763	11.308	(7.983)	8.628	1.161	0.135
Knee Internal Rotation	8.704	(5.643)	8.114	7.579	(4.739)	7.254	0.877	0.425
Knee External Rotation	8.070	(5.009)	6.642	6.793	(5.152)	5.858	0.889	0.407
Hip Extension	7.528	(4.172)	7.488	7.043	(5.593)	6.004	0.927	0.357
Hip Flexion	8.621	(5.505)	8.668	8.971	(6.058)	8.400	0.519	0.951
Hip Adduction	9.441	(8.723)	7.556	12.998	(9.481)	11.474	0.988	0.283
Hip Abduction	9.052	(7.040	8.884	11.461	(7.867)	9.431	0.692	0.725
Hip Internal Rotation	12.314	(8.073)	11.090	10.794	(8.393)	8.273	0.828	0.500
Hip External Rotation	11.349	(8.470)	8.668	10.306	(8.551)	8.591	0.778	0.580

Table 5. Asymmetry of peak moments (Nm/BHxBW) and kinetics (N/BW) during Landing Phase

	ACLR			CONTROL			Z_{KS}	P value
	Mean	(SD)	Median	Mean	(SD)	Median		
Knee Extension Mom	0.050	(0.041)	0.037	0.043	(0.026)	0.042	0.667	0.765
Knee Flexion Mom	0.039	(0.029)	0.032	0.026	(0.030)	0.015	1.421	0.035
Knee Valgus Mom	0.026	(0.021)	0.020	0.032	(0.019)	0.028	1.186	0.120
Knee Varus Mom	0.033	(0.024)	0.027	0.036	(0.028)	0.032	0.507	0.960
Knee ER Mom	0.025	(0.020)	0.019	0.023	(0.017)	0.020	0.469	0.980
Knee IR Mom	0.020	(0.024)	0.012	0.017	(0.019)	0.013	0.692	0.725
Hip Flexion Mom	0.084	(0.047)	0.072	0.065	(0.055)	0.050	1.137	0.151
Hip Extension Mom	0.088	(0.078)	0.069	0.079	(0.086)	0.054	0.939	0.341
Hip Abduction Mom	0.070	(0.059)	0.058	0.057	(0.050)	0.047	0.902	0.390
Hip Adduction Mom	0.060	(0.054)	0.055	0.045	(0.037)	0.033	0.803	0.539
Hip ER Mom	0.031	(0.027)	0.023	0.026	(0.024)	0.023	0.914	0.373
Hip IR Mom	0.023	(0.019)	0.017	0.021	(0.022)	0.013	0.791	0.559
Anterior Tibial Shear Force	0.190	(0.154)	0.156	0.170	(0.132)	0.148	0.605	0.857
Posterior Tibial Shear Force	0.139	(0.095)	0.133	0.131	(0.125)	0.101	0.865	0.443
Vertical Ground Reaction Force	0.682	(0.508)	0.554	0.465	(0.279)	0.388	1.445	0.031

Figure 1. Double Leg Jump Landing. Participants were required to stand atop a box located a distance equal to one half of their body height from the front edge of the force plate, jump forward, land with their foot completely on the force plate, and then immediately make a vertical jump for maximum height.

REFERENCES

1. Lyman S, Koulouvaris P, Sherman S, Do H, Mandl LA, Marx RG. Epidemiology of anterior cruciate ligament reconstruction: trends, readmissions, and subsequent knee surgery. *J Bone Joint Surg Am.* Oct 2009;91(10):2321-2328.

2. Lebel B, Hulet C, Galaud B, Burdin G, Locker B, Vielpeau C. Arthroscopic reconstruction of the anterior cruciate ligament using bone-patellar tendon-bone autograft: a minimum 10-year follow-up. *Am J Sports Med.* Jul 2008;36(7):1275-1282.

3. Mihelic R, Jurdana H, Jotanovic Z, Madjarevic T, Tudor A. Long-term results of anterior cruciate ligament reconstruction: a comparison with non-operative treatment with a follow-up of 17-20 years. *Int Orthop.* Feb 2.

4. Walden M, Hagglund M, Ekstrand J. High risk of new knee injury in elite footballers with previous anterior cruciate ligament injury. *Br J Sports Med.* Feb 2006;40(2):158-162; discussion 158-162.

5. Orchard J, Seward H, McGivern J, Hood S. Intrinsic and extrinsic risk factors for anterior cruciate ligament injury in Australian footballers. *Am J Sports Med.* Mar-Apr 2001;29(2):196-200.

6. Faude O, Junge A, Kindermann W, Dvorak J. Risk factors for injuries in elite female soccer players. *Br J Sports Med.* Sep 2006;40(9):785-790.

7. Marshall SW, Padua DA, Beutler AI, Boling MC, Garrett W. History of prior ACL injury as a risk factor for incident ACL injury. *Journal of athletic training.* 2008;43(5):557.

8. Salmon L, Russell V, Musgrove T, Pinczewski L, Refshauge K. Incidence and risk factors for graft rupture and contralateral rupture after anterior cruciate ligament reconstruction. *Arthroscopy.* Aug 2005;21(8):948-957.

9. Butler DL, Noyes FR, Grood ES. Ligamentous restraints to anterior-posterior drawer in the human knee. A biomechanical study. *J Bone Joint Surg Am.* Mar 1980;62(2):259-270.

10. Fukuda Y, Woo SL, Loh JC, et al. A quantitative analysis of valgus torque on the ACL: a human cadaveric study. *J Orthop Res.* Nov 2003;21(6):1107-1112.

11. Miyasaka T, Matsumoto H, Suda Y, Otani T, Toyama Y. Coordination of the anterior and posterior cruciate ligaments in constraining the varus-valgus and internal-external rotatory instability of the knee. *J Orthop Sci.* 2002;7(3):348-353.

12. Cerulli G, Benoit DL, Lamontagne M, Caraffa A, Liti A. In vivo anterior cruciate ligament strain behaviour during a rapid deceleration movement: case report. *Knee Surg Sports Traumatol Arthrosc.* Sep 2003;11(5):307-311.

13. Fleming BC, Renstrom PA, Beynnon BD, et al. The effect of weightbearing and external loading on anterior cruciate ligament strain. *J Biomech.* Feb 2001;34(2):163-170.

14. Berns GS, Hull ML, Patterson HA. Strain in the anteromedial bundle of the anterior cruciate ligament under combination loading. *J Orthop Res.* Mar 1992;10(2):167-176.

15. Markolf KL, Burchfield DM, Shapiro MM, Shepard MF, Finerman GA, Slauterbeck JL. Combined knee loading states that generate high anterior cruciate ligament forces. *J Orthop Res.* Nov 1995;13(6):930-935.

16. Kanamori A, Woo SL, Ma CB, et al. The forces in the anterior cruciate ligament and knee kinematics during a simulated pivot shift test: A human cadaveric study using robotic technology. *Arthroscopy.* Sep 2000;16(6):633-639.

17. Durselen L, Claes L, Kiefer H. The influence of muscle forces and external loads on cruciate ligament strain. *Am J Sports Med.* Jan-Feb 1995;23(1):129-136.

18. Ferretti A, Papandrea P, Conteduca F, Mariani PP. Knee ligament injuries in volleyball players. *Am J Sports Med.* Mar-Apr 1992;20(2):203-207.

19. Krosshaug T, Nakamae A, Boden BP, et al. Mechanisms of anterior cruciate ligament injury in basketball: video analysis of 39 cases. *Am J Sports Med.* Mar 2007;35(3):359-367.

20. Olsen OE, Myklebust G, Engebretsen L, Bahr R. Injury mechanisms for anterior cruciate ligament injuries in team handball: a systematic video analysis. *Am J Sports Med.* Jun 2004;32(4):1002-1012.

21. Boden BP, Torg JS, Knowles SB, Hewett TE. Video analysis of anterior cruciate ligament injury: abnormalities in hip and ankle kinematics. *Am J Sports Med.* Feb 2009;37(2):252-259.

22. Hewett TE, Myer GD, Ford KR, et al. Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study. *Am J Sports Med.* Apr 2005;33(4):492-501.

23. Paterno MV, Schmitt LC, Ford KR, et al. Biomechanical measures during landing and postural stability predict second anterior cruciate ligament injury after anterior cruciate ligament reconstruction and return to sport. *Am J Sports Med.* Oct;38(10):1968-1978.

24. Salem GJ, Salinas R, Harding FV. Bilateral kinematic and kinetic analysis of the squat exercise after anterior cruciate ligament reconstruction. *Arch Phys Med Rehabil.* Aug 2003;84(8):1211-1216.

25. Tashman S, Collon D, Anderson K, Kolowich P, Anderst W. Abnormal rotational knee motion during running after anterior cruciate ligament reconstruction. *Am J Sports Med.* Jun 2004;32(4):975-983.

26. Deneweth JM, Bey MJ, McLean SG, Lock TR, Kolowich PA, Tashman S. Tibiofemoral joint kinematics of the anterior cruciate ligament-reconstructed knee during a single-legged hop landing. *Am J Sports Med.* Sep;38(9):1820-1828.

27. Orishimo KF, Kremenic IJ, Mullaney MJ, McHugh MP, Nicholas SJ. Adaptations in single-leg hop biomechanics following anterior cruciate ligament reconstruction. *Knee Surg Sports Traumatol Arthrosc.* Nov;18(11):1587-1593.

28. Vairo GL, Myers JB, Sell TC, Fu FH, Harner CD, Lephart SM. Neuromuscular and biomechanical landing performance subsequent to ipsilateral semitendinosus and gracilis autograft anterior cruciate ligament reconstruction. *Knee Surg Sports Traumatol Arthrosc.* Jan 2008;16(1):2-14.

29. Decker MJ, Torry MR, Noonan TJ, Riviere A, Sterett WI. Landing adaptations after ACL reconstruction. *Med Sci Sports Exerc.* Sep 2002;34(9):1408-1413.

30. Paterno MV, Ford KR, Myer GD, Heyl R, Hewett TE. Limb asymmetries in landing and jumping 2 years following anterior cruciate ligament reconstruction. *Clin J Sport Med.* Jul 2007;17(4):258-262.

31. Ernst GP, Saliba E, Diduch DR, Hurwitz SR, Ball DW. Lower extremity compensations following anterior cruciate ligament reconstruction. *Phys Ther.* Mar 2000;80(3):251-260.

32. Kurz MJ, Stergiou N, Buzzi UH, Georgoulis AD. The effect of anterior cruciate ligament reconstruction on lower extremity relative phase dynamics during walking and running. *Knee Surg Sports Traumatol Arthrosc.* Mar 2005;13(2):107-115.

33. Moraiti CO, Stergiou N, Ristanis S, et al. The effect of anterior cruciate ligament reconstruction on stride-to-stride variability. *Arthroscopy.* Jul 2009;25(7):742-749.

34. Moraiti CO, Stergiou N, Vasiliadis HS, Motsis E, Georgoulis A. Anterior cruciate ligament reconstruction results in alterations in gait variability. *Gait Posture.* Jun;32(2):169-175.

35. van Uden CJ, Bloo JK, Kooloos JG, van Kampen A, de Witte J, Wagenaar RC. Coordination and stability of one-legged hopping patterns in patients with anterior cruciate ligament reconstruction: preliminary results. *Clin Biomech (Bristol, Avon).* Jan 2003;18(1):84-87.

36. Dunn WR, Lyman S, Lincoln AE, Amoroso PJ, Wickiewicz T, Marx RG. The effect of anterior cruciate ligament reconstruction on the risk of knee reinjury. *Am J Sports Med.* Dec 2004;32(8):1906-1914.

37. Keays SL, Newcombe PA, Bullock-Saxton JE, Bullock MI, Keays AC. Factors involved in the development of osteoarthritis after anterior cruciate ligament surgery. *Am J Sports Med.* Mar;38(3):455-463.

38. van der Hart CP, van den Bekerom MP, Patt TW. The occurrence of osteoarthritis at a minimum of ten years after reconstruction of the anterior cruciate ligament. *J Orthop Surg Res.* 2008;3:24.

39. Jarvela T, Kannus P, Jarvinen M. Anterior cruciate ligament reconstruction in patients with or without accompanying injuries: A re-examination of subjects 5 to 9 years after reconstruction. *Arthroscopy.* Oct 2001;17(8):818-825.

40. Andreisek G, White LM, Sussman MS, et al. Quantitative MR imaging evaluation of the cartilage thickness and subchondral bone area in patients with ACL-reconstructions 7 years after surgery. *Osteoarthritis Cartilage.* Jul 2009;17(7):871-878.

41. Liden M, Sernert N, Rostgard-Christensen L, Kartus C, Ejerhed L. Osteoarthritic changes after anterior cruciate ligament reconstruction using bone-patellar tendon-bone or hamstring tendon autografts: a retrospective, 7-year radiographic and clinical follow-up study. *Arthroscopy.* Aug 2008;24(8):899-908.

42. Cohen M, Amaro JT, Ejnisman B, et al. Anterior cruciate ligament reconstruction after 10 to 15 years: association between meniscectomy and osteoarthrosis. *Arthroscopy.* Jun 2007;23(6):629-634.

43. Salmon LJ, Russell VJ, Refshauge K, et al. Long-term outcome of endoscopic anterior cruciate ligament reconstruction with patellar tendon autograft: minimum 13-year review. *Am J Sports Med.* May 2006;34(5):721-732.

44. Woo SL, Hollis JM, Adams DJ, Lyon RM, Takai S. Tensile properties of the human femur-anterior cruciate ligament-tibia complex. The effects of specimen age and orientation. *Am J Sports Med.* May-Jun 1991;19(3):217-225.

45. Takai S, Woo SL, Livesay GA, Adams DJ, Fu FH. Determination of the in situ loads on the human anterior cruciate ligament. *J Orthop Res.* Sep 1993;11(5):686-695.

46. Beynnon B, Howe JG, Pope MH, Johnson RJ, Fleming BC. The measurement of anterior cruciate ligament strain in vivo. *Int Orthop.* 1992;16(1):1-12.

47. Gabriel MT, Wong EK, Woo SL, Yagi M, Debski RE. Distribution of in situ forces in the anterior cruciate ligament in response to rotatory loads. *J Orthop Res.* Jan 2004;22(1):85-89.

48. DeMorat G, Weinhold P, Blackburn T, Chudik S, Garrett W. Aggressive quadriceps loading can induce noncontact anterior cruciate ligament injury. *Am J Sports Med.* Mar 2004;32(2):477-483.

49. Draganich LF, Reider B, Ling M, Samuelson M. An in vitro study of an intraarticular and extraarticular reconstruction in the anterior cruciate ligament deficient knee. *Am J Sports Med.* May-Jun 1990;18(3):262-266.

50. Li G, Rudy TW, Sakane M, Kanamori A, Ma CB, Woo SL. The importance of quadriceps and hamstring muscle loading on knee kinematics and in-situ forces in the ACL. *J Biomech.* Apr 1999;32(4):395-400.

51. Lo J, Muller O, Wunschel M, Bauer S, Wulker N. Forces in anterior cruciate ligament during simulated weight-bearing flexion with anterior and internal rotational tibial load. *J Biomech.* 2008;41(9):1855-1861.

52. Withrow TJ, Huston LJ, Wojtys EM, Ashton-Miller JA. The effect of an impulsive knee valgus moment on in vitro relative ACL strain during a simulated jump landing. *Clin Biomech (Bristol, Avon).* Nov 2006;21(9):977-983.

53. Wunschel M, Muller O, Lo J, Obloh C, Wulker N. The anterior cruciate ligament provides resistance to externally applied anterior tibial force but not to internal rotational torque during simulated weight-bearing flexion. *Arthroscopy.* Nov;26(11):1520-1527.

54. Oh YK, Kreinbrink JL, Ashton-Miller JA, Wojtys EM. Effect of ACL Transection on Internal Tibial Rotation in an in Vitro Simulated Pivot Landing. *J Bone Joint Surg Am.* Feb;93(4):372-380.

55. Hewett TE, Torg JS, Boden BP. Video analysis of trunk and knee motion during non-contact anterior cruciate ligament injury in female athletes: lateral trunk and knee abduction motion are combined components of the injury mechanism. *Br J Sports Med.* Jun 2009;43(6):417-422.

56. Meyer EG, Haut RC. Anterior cruciate ligament injury induced by internal tibial torsion or tibiofemoral compression. *J Biomech.* Dec 5 2008;41(16):3377-3383.

57. Bush-Joseph CA, Hurwitz DE, Patel RR, et al. Dynamic function after anterior cruciate ligament reconstruction with autologous patellar tendon. *Am J Sports Med.* Jan-Feb 2001;29(1):36-41.

58. Nyland J, Klein S, Caborn DN. Lower extremity compensatory neuromuscular and biomechanical adaptations 2 to 11 years after anterior cruciate ligament reconstruction. *Arthroscopy.* Sep;26(9):1212-1225.

59. Ortiz A, Olson S, Libby CL, et al. Landing mechanics between noninjured women and women with anterior cruciate ligament reconstruction during 2 jump tasks. *Am J Sports Med.* Jan 2008;36(1):149-157.

60. Webster KE, Gonzalez-Adrio R, Feller JA. Dynamic joint loading following hamstring and patellar tendon anterior cruciate ligament reconstruction. *Knee Surg Sports Traumatol Arthrosc.* Jan 2004;12(1):15-21.

61. Webster KE, Wittwer JE, O'Brien J, Feller JA. Gait patterns after anterior cruciate ligament reconstruction are related to graft type. *Am J Sports Med.* Feb 2005;33(2):247-254.

62. Kurz MJ, Stergiou N. Applied dynamic systems theory for the analysis of movement. In: Stergiou N, ed. *Innovative analyses of human movement: analytical tools for human movement research*. Champaign, IL: Human Kinetics; 2004:93-120.

63. Georgoulis AD, Moraiti C, Ristanis S, Stergiou N. A novel approach to measure variability in the anterior cruciate ligament deficient knee during walking: the use of the approximate entropy in orthopaedics. *J Clin Monit Comput.* Feb 2006;20(1):11-18.

64. Moraiti C, Stergiou N, Ristanis S, Georgoulis AD. ACL deficiency affects stride-to-stride variability as measured using nonlinear methodology. *Knee Surg Sports Traumatol Arthrosc.* Dec 2007;15(12):1406-1413.

65. Stergiou N, Moraiti C, Giakas G, Ristanis S, Georgoulis AD. The effect of the walking speed on the stability of the anterior cruciate ligament deficient knee. *Clin Biomech (Bristol, Avon).* Nov 2004;19(9):957-963.

66. Zampeli F, Moraiti CO, Xergia S, Tsiaras VA, Stergiou N, Georgoulis AD. Stride-to-stride variability is altered during backward walking in anterior cruciate ligament deficient patients. *Clin Biomech (Bristol, Avon).* Dec;25(10):1037-1041.

67. Hamill J, Haddad JM, McDermott WJ. Issues in quantifying variability from a dynamical systems perspective. *J Appl Biomech.* 2000;16:407-418.

68. Hamill J, van Emmerik RE, Heiderscheit BC, Li L. A dynamical systems approach to lower extremity running injuries. *Clin Biomech (Bristol, Avon).* Jun 1999;14(5):297-308.

69. Wheat JS, Glazier PS. Measuring coordination and variability in coordination. In: Davids K, Bennett S, Newell K, eds. *Movement system variability*. Champaign, IL: Human Kinetics; 2006:167-181.

70. Li L, van den Bogert EC, Caldwell GE, van Emmerik RE, Hamill J. Coordination patterns of walking and running at similar speed and stride frequency. *Human movement science.* 1999;18:67-85.

71. Heiderscheit BC, Hamill J, Van Emmerik RE. Q-angle influences on the variability of lower extremity coordination during running. *Med Sci Sports Exerc.* Sep 1999;31(9):1313-1319.

72. Dierks TA, Davis I. Discrete and continuous joint coupling relationships in uninjured recreational runners. *Clin Biomech (Bristol, Avon).* Jun 2007;22(5):581-591.

73. Stergiou N, Jensen JL, Bates BT, Scholten SD, Tzetzis G. A dynamical systems investigation of lower extremity coordination during running over obstacles. *Clin Biomech (Bristol, Avon).* Mar 2001;16(3):213-221.

74. Stergiou N, Scholten SD, Jensen JL, Blanke D. Intralimb coordination following obstacle clearance during running: the effect of obstacle height. *Gait Posture.* May 2001;13(3):210-220.

75. Miller RH, Chang R, Baird JL, Van Emmerik RE, Hamill J. Variability in kinematic coupling assessed by vector coding and continuous relative phase. *J Biomech.* Sep 17 2010;43(13):2554-2560.

76. Diedrich FJ, Warren WH, Jr. Why change gaits? Dynamics of the walk-run transition. *J Exp Psychol Hum Percept Perform.* Feb 1995;21(1):183-202.

77. Kurz MJ, Stergiou N. Effect of normalization and phase angle calculations on continuous relative phase. *J Biomech.* Mar 2002;35(3):369-374.

78. Peters BT, Haddad JM, Heiderscheit BC, Van Emmerik RE, Hamill J. Limitations in the use and interpretation of continuous relative phase. *J Biomech.* Feb 2003;36(2):271-274.

79. Sparrow WA, Donovan E, van Emmerik RE, Barry EB. Using relative motion plots to measure changes in intra-limb and inter-limb coordination. *Journal of motor behavior.* 1987;19(1):115-129.

80. Tepavac D, Field-Fote EC. Vector coding: a technique for quantification of intersegmental coupling in multicyclic behaviors. *J Appl Biomech.* 2001;17:259-270.

81. Heiderscheit BC, Hamill J, van Emmerik RE. Variability of stride characteristics and joint coordination among individuals with unilateral patellofemoral pain. *J Appl Biomech.* 2002;18:110-121.

82. Pollard CD, Heiderscheit BC, van Emmerik RE, Hamill J. Gender differences in lower extremity coupling variability during an unanticipated cutting maneuver. *J Appl Biomech.* May 2005;21(2):143-152.

83. Ferber R, Davis IM, Williams DS, 3rd. Effect of foot orthotics on rearfoot and tibia joint coupling patterns and variability. *J Biomech.* Mar 2005;38(3):477-483.

84. Marx RG, Stump TJ, Jones EC, Wickiewicz TL, Warren RF. Development and evaluation of an activity rating scale for disorders of the knee. *Am J Sports Med.* Mar-Apr 2001;29(2):213-218.

85. Spindler KP, Huston LJ, Wright RW, et al. The prognosis and predictors of sports function and activity at minimum 6 years after anterior cruciate ligament reconstruction: a population cohort study. *Am J Sports Med.* Feb 2011;39(2):348-359.

86. Dunn WR, Spindler KP. Predictors of activity level 2 years after anterior cruciate ligament reconstruction (ACLR): a Multicenter Orthopaedic Outcomes Network (MOON) ACLR cohort study. *Am J Sports Med.* Oct 2010;38(10):2040-2050.

87. Roos EM, Roos HP, Lohmander LS, Ekdahl C, Beynnon BD. Knee Injury and Osteoarthritis Outcome Score (KOOS)--development of a self-administered outcome measure. *J Orthop Sports Phys Ther.* Aug 1998;28(2):88-96.

88. Roos EM, Lohmander LS. The Knee injury and Osteoarthritis Outcome Score (KOOS): from joint injury to osteoarthritis. *Health Qual Life Outcomes.* 2003;1:64.

89. Ahlden M, Kartus J, Ejerhed L, Karlsson J, Sernert N. Knee laxity measurements after anterior cruciate ligament reconstruction, using either bone-patellar-tendon-bone or hamstring tendon autografts, with special emphasis on comparison over time. *Knee Surg Sports Traumatol Arthrosc.* Sep 2009;17(9):1117-1124.

90. Bell AL, Pedersen DR, Brand RA. A comparison of the accuracy of several hip center location prediction methods. *J Biomech.* 1990;23(6):617-621.

91. Paterno MV, Schmitt LC, Ford KR, et al. Biomechanical measures during landing and postural stability predict second anterior cruciate ligament injury after anterior cruciate ligament reconstruction and return to sport. *Am J Sports Med.* Oct 2010;38(10):1968-1978.

92. Ardern CL, Webster KE, Taylor NF, Feller JA. Return to the preinjury level of competitive sport after anterior cruciate ligament reconstruction surgery: two-thirds of patients have not returned by 12 months after surgery. *Am J Sports Med.* Mar 2011;39(3):538-543.

93. Chappell JD, Creighton RA, Giuliani C, Yu B, Garrett WE. Kinematics and electromyography of landing preparation in vertical stop-jump: risks for noncontact anterior cruciate ligament injury. *Am J Sports Med.* Feb 2007;35(2):235-241.

94. Taylor KA, Terry ME, Utturkar GM, et al. Measurement of in vivo anterior cruciate ligament strain during dynamic jump landing. *J Biomech.* Feb 3 2011;44(3):365-371.

95. Moraiti CO, Stergiou N, Vasiliadis HS, Motsis E, Georgoulis A. Anterior cruciate ligament reconstruction results in alterations in gait variability. *Gait Posture.* Jun 2010;32(2):169-175.

96. Wright RW, Dunn WR, Amendola A, et al. Risk of tearing the intact anterior cruciate ligament in the contralateral knee and rupturing the anterior cruciate ligament graft during the first 2 years after anterior cruciate ligament reconstruction: a prospective MOON cohort study. *Am J Sports Med.* Jul 2007;35(7):1131-1134.