THE RELATIONSHIP BETWEEN KNEE FLEXION, HIP FLEXION, AND TRUNK FLEXION ANGLES AND ANTERIOR TIBIAL SHEAR FORCE DURING A JUMP-LANDING TASK

Hollie Janine Walusz

A thesis submitted to the faculty of the University of North Carolina at Chapel Hill in partial fulfillment of the requirements for the degree of Master of Arts in the Department of Exercise and Sport Science (Athletic Training)

Chapel Hill
2007

Approved by:
Darin Padua
Troy Blackburn
Chris Hirth
Michelle Boling
Melanie McGrath
ABSTRACT

Hollie Walusz: The relationship between knee flexion, hip flexion, and trunk flexion angles and anterior tibial shear force during a jump-landing task
(Under the direction of Darin Padua)

Objective: To evaluate the relationships between sagittal plane knee, hip, and trunk angles with peak anterior tibial shear force (ATSF) during a jump-landing task. Subjects: Thirty-three healthy female recreational athletes with no prior history of anterior cruciate ligament (ACL) injury volunteered to participate in this study. Measurements: Knee, hip, and trunk kinematic data were collected with an electromagnetic motion capture system during a jump-landing task. Within- and between-subject data were analyzed using correlation and multiple regression statistical analyses. Results: Knee flexion angle at peak ATSF was found to be significantly correlated with peak ATSF (p=0.021, r = -0.367). Additionally, trunk flexion (world) at peak ATSF was found to have a significant positive relationship with peak ATSF (p=0.046, r =0.309). Knee flexion angle at peak ATSF was found to predict approximately 13.5% of the variance in ATSF (p=.042, R²=.135). Multiple within-subject correlations and regressions were found, however they varied amongst all angles between subjects.

Conclusion: Knee flexion angles appear to predict ATSF in female athletes. While hip and trunk flexion angles did not predict ATSF in this study, training should emphasize increased flexion at all three joints in order to decrease ACL injury. Sagittal plane joint angles did not explain all the variability in peak ATSF, therefore other factors need to be examined.
ACKNOWLEDGEMENTS

I want to first and foremost thank my thesis committee for helping me complete this research study. Their guidance, support, and words of encouragement were paramount in the completion of this project.

To Darin, Troy, Chris, Michelle, and Melanie: Your immense help with this study was priceless and I cannot thank you enough for all of your time and effort. Without your willingness to answer the endless questions, fix the equipment whenever it decided not to work, or have patience with all the revisions and meetings, this project could not have been a success. Your input was invaluable and for that I will always be grateful.

To my friends and classmates: I am forever thankful for your understanding and support during this project as well as at this point in my life. These last two years have been truly amazing.

Finally, I would like to thank my family. Your unconditional love and constant support has helped bring me where I am today. You continue to inspire me to achieve greater things, and for that I thank you.
# TABLE OF CONTENTS

<table>
<thead>
<tr>
<th>LIST OF TABLES</th>
<th>vi</th>
</tr>
</thead>
<tbody>
<tr>
<td>Chapter</td>
<td></td>
</tr>
<tr>
<td>I. INTRODUCTION</td>
<td>1</td>
</tr>
<tr>
<td>Purpose</td>
<td>4</td>
</tr>
<tr>
<td>Independent/Predictor Variables</td>
<td>4</td>
</tr>
<tr>
<td>Dependent/Criterion Variable</td>
<td>5</td>
</tr>
<tr>
<td>Research Questions</td>
<td>5</td>
</tr>
<tr>
<td>Null Hypothesis</td>
<td>6</td>
</tr>
<tr>
<td>Alternate Hypothesis</td>
<td>7</td>
</tr>
<tr>
<td>Research Hypothesis</td>
<td>8</td>
</tr>
<tr>
<td>Operational Definitions</td>
<td>8</td>
</tr>
<tr>
<td>Assumptions</td>
<td>9</td>
</tr>
<tr>
<td>Delimitation</td>
<td>10</td>
</tr>
<tr>
<td>Limitations</td>
<td>10</td>
</tr>
<tr>
<td>II. LITERATURE REVIEW</td>
<td>11</td>
</tr>
<tr>
<td>Anatomy of the ACL</td>
<td>12</td>
</tr>
<tr>
<td>Epidemiology of ACL Injuries</td>
<td>13</td>
</tr>
<tr>
<td>Etiology of ACL Injuries</td>
<td>14</td>
</tr>
<tr>
<td>Sports and ACL Injury</td>
<td>21</td>
</tr>
</tbody>
</table>
Appendix B: Figures .................................................................58
Appendix C: Manuscript .........................................................60
Appendix D: Informed Consent Form.........................................81
REFERENCES............................................................................89
## LIST OF TABLES

<table>
<thead>
<tr>
<th>Table</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>1. Data Analysis Plan</td>
<td>47</td>
</tr>
<tr>
<td>2. Descriptive Statistics Table</td>
<td>48</td>
</tr>
<tr>
<td>3. Within Subject # of Significant Findings Table</td>
<td>49</td>
</tr>
<tr>
<td>4. Between Subject # of Significant Findings Table</td>
<td>50</td>
</tr>
<tr>
<td>5. Within Subject Initial Contact Correlation Table</td>
<td>51</td>
</tr>
<tr>
<td>6. Within Subject Peak ATSF Correlation Table</td>
<td>52</td>
</tr>
<tr>
<td>7. Initial contact Regression Table</td>
<td>53</td>
</tr>
<tr>
<td>8. Initial contact Peak ATSF Table</td>
<td>54</td>
</tr>
<tr>
<td>9. Between Subject Correlation Table</td>
<td>55</td>
</tr>
<tr>
<td>10. Between Subject Regression Table</td>
<td>56</td>
</tr>
</tbody>
</table>
# LIST OF FIGURES

<table>
<thead>
<tr>
<th>Figure</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>1. Between Subjects Plot of Knee Flexion Angle at Peak ATSF.</td>
<td>58</td>
</tr>
</tbody>
</table>

CHAPTER 1
INTRODUCTION

The anterior cruciate ligament (ACL) is one of the most important ligaments for knee joint stability and yet is the most frequently ruptured ligament in the knee (Fleming, Renstrom et al. 2001; Darcy, Kilger et al. 2005). An estimated 80,000-100,000 ACL tears occur annually in the United States (Griffin, Agel et al. 2000; Huston, Greenfield et al. 2000). For the general population, there is an annual ACL injury incidence rate of approximately 1 per 3000 people (Miyasaka, Daniel et al. 1991). It is estimated that over 70% of total ACL injuries are sports-related (Colby, Francisco et al. 2000; Chappell, Yu et al. 2002). Because there is a high incidence of this devastating injury occurring during sports related activity, research is essential to determine all the possible causes and predisposing factors.

It has been found that female athletes have a much higher incidence of ACL injury than male athletes involved in the same sports. Epidemiological research has demonstrated that female athletes are two to eight times more likely to sustain an ACL injury than their male counterparts (Arendt and Dick 1995; Huston, Greenfield et al. 2000; Fagenbaum and Darling 2003). In addition to this increased risk, female participation in collegiate athletics has increased over the years (Arendt and Dick 1995) and it is estimated that 2,200 ACL ruptures occur each year in female collegiate athletes (Hewett, Lindenfeld et al. 1999).
Research suggests that the primary mechanism of ACL injury is non-contact in nature, as 70-80% of all ACL injuries involve non-contact injury mechanism (Agel, Arendt et al. 2005; Chappell, Herman et al. 2005; Kernozek, Torry et al. 2005; McLean, Huang et al. 2005). Non-contact injury mechanisms are most prevalent in sports such as women’s basketball and soccer (Arendt and Dick 1995; Agel, Arendt et al. 2005). Non-contact injuries typically occur during execution of jump-landing, sudden deceleration, pivoting or cutting movements (Agel, Arendt et al. 2005; McLean, Huang et al. 2005). Due to these findings, it is essential to examine the causative factors that lead to ACL injury during non-contact functional tasks.

A variety of intrinsic and extrinsic factors have been proposed that may contribute to the increased rate of non-contact ACL injury among female athletes. The intrinsic factors include a smaller cross-sectional area of the ACL in women, a narrower intercondylar notch, an increased Q angle, a greater amount of knee laxity, a greater amount of subtalar joint pronation, and even hormonal variations. The extrinsic factors that are believed to contribute to ACL injury include the level of conditioning of the athlete, level of muscular strength, altered motor control strategies, and altered lower extremity kinetics and kinematics during functional tasks. Many of these factors have been examined in previous research, yet two factors needing more focus are the kinematics and kinetics of athletes during functional sports tasks.

Gender differences have been found in motion patterns, positions, and forces generated from the hip and trunk to the knee (Griffin, Agel et al. 2000). These differences are important because of the kinetic chain relationship of the lower extremity. Hip position and motion may influence knee position and knee loads. The same may be said for kinetic chain patterns stemming from trunk motion, although extensive research in this area has yet to be
completed. Females have been shown to perform cutting and landing maneuvers in a more erect posture than males, displaying less hip and knee flexion (Griffin, Agel et al. 2000). Current research suggests that the ACL is most vulnerable to injury at or immediately following initial contact with the ground when coupled with an awkward body position during landing tasks (Lephart, Abt et al. 2002). Gender differences in landing technique may be a cause of the increased incidence of knee injury found in female athletes.

Biomechanics and lower extremity posture also affect loads at the knee joint. Knee flexion angles between fifty degrees and full extension have been shown to increase strain on the ACL (Beynnon and Fleming 1998). More specifically, extended knee positions place the ACL in a stretched and more vulnerable position for injury (Arendt and Dick 1995; Malinzak, Colby et al. 2001). Movement of the knee from a flexed to an extended position, either passively or through contraction of the leg muscles, produces an increase in ACL strain values (Beynnon and Fleming 1998). However, research is limited when looking at hip flexion angle or trunk flexion angles, and how each of these may impact the load on the ACL. The knee is only one part of the kinetic chain, therefore it is important to look at other anatomical sites such as the hip and trunk, and their potential roles in ACL injury.

Joint biomechanics associated with a jump-landing task could be a possible mechanism of ACL rupture. Landing from a jump has often been reported as a cause of non-contact anterior cruciate ligament injuries in competitive athletics (Yu, McClure et al. 2005; Yu, Lin et al. 2006). Jump landings are also a common and essential task in sports, especially female sports such as basketball, soccer, and volleyball. It has been found through research that female athletes tend to land from drop jumps with less knee flexion at initial contact than
male athletes (Huston, Greenfield et al. 2000; Malinzak, Colby et al. 2001; Chappell, Yu et al. 2002; Lephart, Ferris et al. 2002; Decker, Torry et al. 2003; Yu, Lin et al. 2006).

The cause of non-contact ACL injuries is speculative and can include many of the intrinsic and extrinsic factors discussed in previous research, but one factor not examined is increased anterior tibial shear force (ATSF) caused by motion at the knee, hip, and trunk. Previous research has suggested that an increased anterior shear load may correspond with an increase in ACL strain. Additionally, research has found that an increased ATSF is associated with jump-landing tasks (Fleming, Renstrom et al. 2001; Chappell, Yu et al. 2002; Kernozek, Torry et al. 2005; Withrow, Huston et al. 2006; Yu, Lin et al. 2006). It has also been found that females exhibit significantly greater proximal ATSF than males during the landing phase of a jump task. (Chappell, Yu et al. 2002; Chappell, Herman et al. 2005; Yu, McClure et al. 2005) Overall, few studies have examined the factors associated with the increased peak proximal ATSF in female athletes. Additionally, although some research has examined ATSF at the knee, research has not yet determined if ATSF is influenced by hip and trunk flexion angles.

**Purpose**

The purpose of this study is to determine if a relationship exists between ATSF and knee flexion, hip flexion, and trunk flexion angles during a jump-landing task.

**Independent/Predictor Variables:**

1. Knee flexion angle at initial ground contact
2. Hip flexion angle at initial ground contact
3. Trunk flexion (world) angle at initial ground contact
4. Trunk flexion (pelvis) angle at initial ground contact
5. Knee flexion angle at the time of peak ATSF
6. Hip flexion angle at the time of peak ATSF
7. Trunk flexion (world) angle at the time of peak ATSF
8. Trunk flexion (pelvis) angle at the time of peak ATSF

**Dependent/Criterion Variable:**

1. Peak anterior tibial shear force

**Research Questions:**

1. Is there a relationship between knee flexion angle and anterior tibial shear force at initial contact during a jump-landing task?
2. Is there a relationship between hip flexion angle and anterior tibial shear force at initial contact during a jump-landing task?
3. Is there a relationship between trunk flexion angle and anterior tibial shear force at initial contact during a jump-landing task?
4. Does the linear combination of knee flexion, hip flexion, and/or trunk flexion angle at initial contact of a jump-landing task predict anterior tibial shear force?
5. Is there a relationship between knee flexion angle and anterior tibial shear force at the time of peak ATSF during a jump-landing task?
6. Is there a relationship between hip flexion angle and anterior tibial shear force at the time of peak ATSF during a jump-landing task?
7. Is there a relationship between trunk flexion angle and anterior tibial shear force at the time of peak ATSF during a jump-landing task?

8. Does the linear combination of knee flexion, hip flexion, and/or trunk flexion angle at the time of peak ATSF of a jump-landing task predict anterior tibial shear force?

**Null Hypothesis**

1. $H_0$: There is no relationship between knee flexion angle at initial contact and anterior tibial shear force during a jump-landing task.

2. $H_0$: There is no relationship between hip flexion angle at initial contact and anterior tibial shear force during a jump-landing task.

3. $H_0$: There is no relationship between trunk flexion angle at initial contact and anterior tibial shear force during a jump-landing task.

4. $H_0$: Anterior tibial shear force is not predicted by the linear combination of trunk flexion, hip flexion, and/or knee flexion angle at initial contact of a jump-landing task.

5. $H_0$: There is no relationship between knee flexion angle at the time of peak ATSF and anterior tibial shear force during a jump-landing task.

6. $H_0$: There is no relationship between hip flexion angle at the time of peak ATSF and anterior tibial shear force during a jump-landing task.

7. $H_0$: There is no relationship between trunk flexion angle at the time of peak ATSF and anterior tibial shear force during a jump-landing task.
8. \( H_0 \): Anterior tibial shear force is not predicted by the linear combination of trunk flexion, hip flexion, and/or knee flexion angle at the time of peak ATSF during a jump-landing task.

**Alternate Hypothesis**

1. \( H_A \): There is a relationship between the knee flexion angle at initial contact and anterior tibial shear force.

2. \( H_A \): There is a relationship between the hip flexion angle at initial contact and anterior tibial shear force.

3. \( H_A \): There is a relationship between the trunk flexion angle at initial contact and anterior tibial shear force.

4. \( H_A \): Anterior tibial shear force is predicted by a combination of trunk flexion, hip flexion, and/or knee flexion angle at initial contact of a jump-landing task.

5. \( H_A \): There is a relationship between the knee flexion angle at the time of peak ATSF and anterior tibial shear force.

6. \( H_A \): There is a relationship between the hip flexion angle at the time of peak ATSF and anterior tibial shear force.

7. \( H_A \): There is a relationship between the trunk flexion angle at the time of peak ATSF and anterior tibial shear force.

8. \( H_A \): Anterior tibial shear force is predicted by a combination of trunk flexion, hip flexion, and/or knee flexion angle at the time of peak ATSF of a jump-landing task.
Research Hypothesis

1. There will be a negative relationship between knee flexion angle at initial contact and anterior tibial shear force.

2. There will be a negative relationship between hip flexion angle at initial contact and anterior tibial shear force.

3. There will be a negative relationship between trunk flexion angle at initial contact and anterior tibial shear force.

4. Anterior tibial shear force will be predicted by a combination of trunk flexion, hip flexion, and/or knee flexion angle at initial contact of a jump-landing task.

5. There will be a negative relationship between knee flexion angle at the time of peak ATSF and anterior tibial shear force.

6. There will be a negative relationship between hip flexion angle at the time of peak ATSF and anterior tibial shear force.

7. There will be a negative relationship between trunk flexion angle at the time of peak ATSF and anterior tibial shear force.

8. Anterior tibial shear force will be predicted by a combination of trunk flexion, hip flexion, and/or knee flexion angle at the time of peak ATSF of a jump-landing task.

Operational Definitions

Anterior tibial shear force (ATSF): anteriorly directed force on the tibia relative to the femur
Knee flexion angle: the angle measured between the longitudinal axes of the shank and thigh segments in the sagittal plane.

Hip flexion angle: the angle measured between the longitudinal axes of the thigh and pelvis segments in the sagittal plane.

Trunk flexion angle: the angle measured between the longitudinal axes of the sacrum and thorax segments in the sagittal plane.

Dominant leg: the leg each subject would use to kick a soccer ball for maximal distance.

Jump-landing task: An athlete performs a jump off a 30-cm high platform from 50% of his/her height away from a force plate. The athlete lands on the force plate, and then immediately jumps vertically for maximum height.

Jump landing initial contact: The time point following a jump-landing task as indicated by the first data point measuring vertical ground reaction force above ten volts.

Assumptions

1. The Flock of Birds instrumentation for force plate calculation of ATSF is a valid and reliable measure.
2. Subjects will be truthful about previous medical history of lower extremity injury.
3. Subjects will perform a jump landing in the lab setting as they would in their respective sports activities or on the field.
Delimitations

1. All subjects will be healthy and free from injury to the lower extremity in the six months prior to data collection.
2. All subjects will be free from prior ACL injury or any knee surgery.
3. All subjects will be female recreational athletes between the ages of 18-25.
4. All analyses will be completed on the dominant leg.

Limitations

1. Athletes will be tested in a laboratory setting instead of their sports setting.
2. Research will correctly differentiate between trunk and hip flexion kinematic measurements.
CHAPTER 2

LITERATURE REVIEW

Concern is still prevalent in the sports medicine community regarding the high incidence of anterior cruciate ligament (ACL) injury in athletics, especially in females. The ACL is one of the most important ligaments to knee joint stability, and yet, is the most frequently ruptured ligament in the knee (Fleming, Renstrom et al. 2001; Darcy, Kilger et al. 2005). This high injury incidence occurs despite increased research working to identify important risk factors in order to accurately modify them, as well as the increased efforts to implement preventative programs. Still, there are no clear answers to explain why there is a significant gender based difference observed in sport related non-contact ACL injury. An emerging theory for this gender disparity proposes that females perform high demand athletic maneuvers differently than males, and in a manner that predisposes them to higher knee joint stress (Colby, Francisco et al. 2000; Kirkendall and Garrett 2000; Huston, Vibert et al. 2001; Wojtys, Ashton-Miller et al. 2002; Decker, Torry et al. 2003). The purpose of this literature review is to investigate additional risk factors that may predispose athletes, especially females, to ACL injury.
Anatomy of the ACL

The ACL is an intra-articular structure that traverses the knee joint through the intercondylar notch of the femur, attaching to the tibia and the femur. The ligament inserts proximally on the medial aspect of the lateral femoral condyle and distally on the anterior tibial plateau (Fleming and Beynnon 2004). The ACL is a primary stabilizer of the knee (Beynnon and Fleming 1998). It is responsible for providing primary restraint to anterior translation of the tibia relative to the femur, as well as serving to limit varus-valgus motion and tibial rotations of the knee (Smith, Livesay et al. 1993). Lephart et al. found that the ACL resists 80-85% of anterior translatory loads at the knee joint (Lephart, Abt et al. 2002). The ACL is comprised of two bundles, the anteromedial and posterolateral. The restraint forces in the anteromedial and posterolateral portions of the ACL are different, and change with respect to knee flexion angle (Woo, Chan et al. 1997). The posterolateral portion of the ACL is strained as the knee nears full extension, while the anteromedial portion is taut in flexion (Heijne, Fleming et al. 2004; Woo, Abramowitch et al. 2006).

The soft tissues surrounding the knee joint that provide dynamic support are referred to as secondary stabilizers or restraints. These stabilizers include the quadriceps, hamstrings, and gastrocnemius muscles. The proximal attachments of the gastrocnemius and the distal attachments of the hamstrings are important to the stability of the knee because they provide a posterior translatory force on the tibia that counteracts anterior tibial translation (Lephart, Abt et al. 2002). The quadriceps muscles, on the other hand, provide an anterior translatory force to the tibia due to the extensor mechanism of the
knee and the patellar tendon-tibia angle. Dynamic muscle stabilization protects the knee joint, and allows the knee to withstand stress and strain (Nyland, Caborn et al. 1999).

Epidemiology of ACL Injuries

An estimated 80,000-100,000 ACL injuries occur annually in the United States alone (Griffin, Agel et al. 2000; Huston, Greenfield et al. 2000; McLean, Huang et al. 2005). DeMorat et al. increased that estimation and stated occurrence as high as 200,000 tears each year in the United States (DeMorat, Weinhold et al. 2004). Over 70% of all these ACL injuries occurred in sport activities involving both recreational and elite athletes (Smith, Livesay et al. 1993). Although ACL injuries are not gender specific, disparities are found and do occur at a significantly greater rate in female athletes. Approximately 13,000 knee injuries will occur in females who participate in athletics at the collegiate level during any given year, and 2,200 of these knee injuries will be ACL ruptures (Hewett, Lindenfeld et al. 1999; Huston, Greenfield et al. 2000). Additionally, females are two to eight times more likely to sustain injury to the ACL than their male counterparts in similar sports (Ferretti, Papandrea et al. 1992; Arendt and Dick 1995; Griffin, Agel et al. 2000; Huston, Greenfield et al. 2000; Fagenbaum and Darling 2003). Female collegiate soccer and basketball players were found to be three and four times, respectively, more likely to incur a non-contact ACL injury than their male counterparts (Arendt and Dick 1995; Agel, Arendt et al. 2005). Based on these findings, it may be concluded that there are gender differences in the incidence of ACL injuries.
Etiology of ACL Injuries

The exact etiology of ACL injury is a topic that has been researched extensively, but a definitive answer is still yet to be found. Both intrinsic and extrinsic factors are thought to contribute to the occurrence of ACL injury, especially in females. Intrinsic factors have long been proposed to contribute to the increased injury rate in females (Griffin, Agel et al. 2000; Chappell, Yu et al. 2002). An intrinsic factor refers to a factor that is within an individual. They are usually anatomical and inherent. Intrinsic factors that have been theorized to increase the risk of ACL injury in females include: smaller cross-sectional area of the ACL, narrower intercondylar notch, increased Q angle, increased knee laxity, increased subtalar joint pronation, and hormonal variations. Muneta et al have shown that the cross-sectional area of the ACL is smaller in females as compared to males (Muneta, Takakuda et al. 1997). Additionally, Davis et al. (1999) reported that the mean ACL width is significantly greater in males than in females. Anderson et al. (2001) also found that the ACL is relatively smaller in females than males, which led them to conclude that the ACL then probably is not able to compensate for the lack of strength and stiffness at the knee in females. Studies have concluded that total notch width and intercondylar notch width of the femur are significantly more narrow in females than males (Shelbourne, Davis et al. 1998; Shelburne, Pandy et al. 2004). Additionally, in both males and females, the notch width was narrower in those patients who sustained ACL tears when compared to an uninjured control group (Shelbourne, Davis et al. 1998).

A study by Shambaugh et al. (1991) found that the average Q angles of athletes sustaining knee injuries were significantly larger than the average angles for the players who were not injured. Women with high levels of joint laxity are found to have a
significantly increased risk for incurring non-contact ACL injuries relative to those without lax knee joints (Uhorchak, Scoville et al. 2003). Overall, studies have shown that joint laxity tends to be greater in females than in males (Huston and Wojtys 1996). Additionally, excessive subtalar joint pronation has been demonstrated to be a postural fault significantly different between ACL-injured female athletes and matched control subjects (Loudon, Jenkins et al. 1996). This postural fault may cause an increased valgus moment at the knee due to the compensatory tibial internal rotation that occurs with excessive subtalar joint pronation, leading to excessive stress placed on the knee and possibly an increased risk for ACL injury. Additionally, several researchers have attempted to link hormonal fluctuations during the menstrual cycle to the rate of ACL injury. It has been suggested that women not using oral contraceptives are more susceptible to ACL injury during the ovulatory phase of the menstrual cycle than any of the other phases (Wojtys, Huston et al. 2002). Females are at a greater risk of ACL injury, potentially due to these intrinsic factors, than their male counterparts.

Unfortunately, many of these intrinsic risk factors are not modifiable. Instead it seems that extrinsic factors should be a focus among researchers since these risk factors have the potential to be modified and corrected among individuals. An extrinsic factor refers to a factor with external origin. Such extrinsic factors, viewed as primary contributors to ACL injury, include the level of conditioning, level of muscular strength in the athlete, altered motor control strategies, and joint kinetics and kinematics during sports related tasks. Several researchers have documented that females have significantly less muscular strength in the quadriceps and hamstrings compared with males, even when normalized to body weight (Miller, MacDougall et al. 1993; Huston and Wojtys 1996; Huston,
Additionally, studies have found that when normalized to body weight, the hamstring muscles in female athletes were relatively weaker than in the male athletes when compared with the quadriceps muscles (Shelburne, Pandy et al. 2004). These differences may limit dynamic knee joint stability in females. Wojtyś et al. (2002, 2003) found that maximum co-contraction of the knee musculature significantly decreased mean anterior tibial translation in both males and females, yet males were able to increase shear stiffness of the knee significantly more than females could. During knee extension with an anterior pull on the tibia, the ACL is solely responsible for stabilization and for not allowing further translation unless offset by activation of the hamstring muscles. Due to a non-optimal line of pull, the hamstring musculature cannot reduce the strain generated by the quadriceps musculature on the ACL in knee positions ranging from 0-30 degrees of knee flexion (Renstrom, Arms et al. 1986; Nyland, Caborn et al. 1999; Colby, Francisco et al. 2000). It is necessary for the quadriceps and hamstrings to work in concert to stabilize the knee and minimize the shear forces at the knee.

Altered motor control strategies between genders have also been viewed as a risk factor. Previous research by Malinzak et al. has reported that females have increased quadriceps activity and decreased hamstring activity during running, cutting, and jumping tasks in comparison to males. Further, female subjects have been found to have a higher activation of their quadriceps muscles, but lower activation for their hamstring muscles, than male subjects in selected athletic tasks (Malinzak, Colby et al. 2001). This quadriceps dominance in female athletes may place significantly more strain on the ACL than those athletes who co-contract their quadriceps and hamstring muscles, or even
those who contract their hamstrings first. Studies have shown that the co-contraction of
the hamstrings muscles with the quadriceps muscles decreases the total knee ATSF and
thus the stress on the ACL (Smidt 1973).

Lower extremity kinetic and kinematic differences among males and females may also
be an extrinsic risk factor for ACL injury (Markolf, Burchfield et al. 1995; Griffin, Agel
et al. 2000; Huston, Vibert et al. 2001; Malinzak, Colby et al. 2001; Chappell, Yu et al.
2002; Leaphart, Ferris et al. 2002; Decker, Torry et al. 2003; Heijne, Fleming et al. 2004;
Yu, Lin et al. 2006). Interest in biomechanical research has increased due to a large
amount of ACL injuries occurring during non-contact movements. Approximately 78%
of all ACL injuries are non-contact in nature, and the injuries most often occur when
landing from a jump, while cutting, or with sudden deceleration movements (Arendt and
2005; Chappell, Herman et al. 2005; Kernozek, Torry et al. 2005). The exact cause of
non-contact ACL injuries is speculative, and can include many of the intrinsic and
extrinsic factors mentioned above. However, one cause needing more examination may
be an increased anterior tibial shear force, caused by motion at the knee, hip, and trunk.

Biomechanics and lower extremity posture differences among gender may affect knee
loads in addition to playing an important role in the mechanism of non-contact ACL
injury. One important biomechanical factor is the amount of anterior tibial shear force
(ATSF) in the knee. It has been found that an increased anterior shear load may
correspond with an increase in ACL strain (Fleming, Renstrom et al. 2001; Withrow,
Huston et al. 2006). ATSF is generated in vivo from both high quadriceps activity and by
inertial forces due to sudden accelerations and decelerations of the lower limb (Markolf,
Burchfield et al. 1995). Additionally, ATSF is arguably the most direct and important mechanism for loading the ACL. Markolf et al. (1995) found that the application of an ATSF to the knee increased the mean ACL force at all flexion angles. Researchers have reported that the ACL is loaded when the total shear force applied to the tibiofemoral joint is directed anteriorly (Shelburne, Pandy et al. 2004). It has been reported that the tensile strength of the ACL is approximately 2,100 N in healthy knees (Woo, Hollis et al. 1991). ATSF may be an important factor to analyze when examining causative biomechanical factors related to ACL injury.

Studies have also focused on how the ACL is strained during different loading scenarios. In cadaveric specimens, Shoemaker et al. (1993) reported that anterior displacement due to quadriceps muscle force was greatest when the knee was flexed approximately 30 degrees. Subsequent investigations found the ATSF increased significantly with knee extension, and tended to be greatest at the most extended knee positions tested, which was approximately 25 degrees (Osternig, Ferber et al. 2001). When in terminal knee extension, the ACL is in a more stretched position. On the contrary, a flexed knee position allows the knee joint to be in a more favorable position for control of rotation and anterior displacement.

When viewing gender differences regarding ATSF, females have exhibited greater proximal ATSF than males during the landing phase of a jump task (Chappell, Yu et al. 2002). This difference may likely be due to an increased quadriceps muscle force, a decreased hamstring muscle force, and/or a decreased knee flexion angle (Malinzak, Colby et al. 2001). ATSF generated by the quadriceps muscles increases as the knee flexion angle decreases. DeMorat et al. (2004) found that an aggressive quadriceps load,
with the knee near full extension, is strong enough to rupture the ACL, as it produces
significant anterior tibial translation that can cause injury. Decreased hamstring muscle
force doesn’t allow a sufficient counteractive posterior shear force to be applied against
the quadriceps. Anterior shear forces are greatest during extension exercises at less than
40 degrees of knee flexion or at the most extended knee joint positions tested (Kaufman,
(2002) found that the peak extension moment and the peak proximal ATSF for females
occurred essentially at the same time during a landing task. The extended knee joint
position findings along with the correlating increase in ATSF raise concern that lower
extremity posture may be a risk factor for ACL injury.

Gender differences have been found in motion patterns, positions, and forces generated
from the hip and trunk to the knee (Griffin, Agel et al. 2000; Malinzak, Colby et al.
2001). These differences are important because of the kinetic chain relationship between
the lower extremity and force transmission up and down this chain. Hip and trunk
position and motion may have an influence on knee positions and knee loads. Females
have been shown in previous literature to perform cutting and landing maneuvers in a
more erect posture than males, with less hip flexion and less knee flexion (Huston, Vibert
et al. 2001; Malinzak, Colby et al. 2001; Decker, Torry et al. 2003). Overall, compared
to males, females tend to execute high demand activities in a more erect sagittal plane
posture (Decker, Torry et al. 2003). This erect posture potentially predisposes the ACL
to greater loads and injury. A more erect landing strategy with a decrement in shock
absorption may provide a mechanical disadvantage to the hamstring muscles as described
previously. Additionally it allows the unopposed quadriceps muscles to pull the tibia anteriorly when contracting, resulting in larger values of ATSF.

Knee flexion angles and their affect on strain of the ACL have been examined frequently in previous literature. When the knee angle is between 50 degrees and full extension, the ACL is strained (Markolf, Burchfield et al. 1995; Beynnon and Fleming 1998). Movement of the knee from a flexed to an extended position, either passively or through contraction of the leg muscles, produces an increase in ACL strain values (Beynnon and Fleming 1998). Fagenbaum and Darling (2003) found that ACL forces progressively decreased as the knee was flexed to 15, 30, and 60 degrees. Research completed by Colby et al. (2000) found the average knee flexion angle at foot strike to be 23 degrees when landing from a jump. Coincidentally, it was also found that the average angle of knee flexion at the time of ACL injury during a jump is 21 degrees. Lephart et al. (2002) reported that females tend to land with less knee flexion after impact than males during a jump-landing task. This is all evidence that knee flexion angles affect the load placed on the knee. One study had contrasting findings, stating that females landed with greater knee flexion angles than their male counterparts (Fagenbaum and Darling 2003). However, this study examined elite male and female basketball players at the same college with similar training techniques. This finding may also imply that recreational female athletes land differently than more experienced elite athletes or it may imply that the training programs of both teams allowed females to perform these tasks more like males. These conflicting findings warrant additional study to determine how knee flexion angles affect ATSF.
It is also important to examine other anatomical sites such as the hip and trunk, as they may play a role in ACL injury. Due to the interconnected relationship of all the lower extremity joints in the kinetic chain, hip flexion angles may very well affect ACL strain values. However, not enough research has been completed in this area to understand the effect of hip angle during a jump-landing task. Many studies have established that females land with less hip flexion during a landing task than their male counterparts (Yu, McClure et al. 2005). Another study found that closed kinetic chain exercises strain the ACL less when the hip is in a more flexed position as compared with those exercises that required less hip flexion (Heijne, Fleming et al. 2004). Additionally, quadriceps activity and therefore possibly ATSF, have been reported to decrease when hip flexion angles are increased (Lephart, Abt et al. 2002). Lephart et al. (2002) also found that females tend to activate their quadriceps near full extension of the knee with little hamstring activity and that they also land with smaller angles of hip flexion compared with males. This combined effect suggests that smaller hip flexion angles may predispose female athletes to ACL injury. Trunk flexion angle and its role in ACL injury and amount of ATSF generated have not yet been extensively examined in previous research to date. Additional research in this area needs to be completed in order to determine how much lower extremity posture and landing technique affect ATSF and ACL strain.

Sports and ACL Injury

ACL injury is non-discriminatory to any specific sport; however its prevalence is higher in certain sports. Sports in which ACL injury is common include basketball, soccer, volleyball, field hockey, football, gymnastics, and skiing (Childs 2002). The rate
of ACL injury, regardless of mechanism of injury, continues to be significantly higher for female collegiate athletes than for male collegiate athletes in both soccer and basketball (Agel, Arendt et al. 2005). Additionally, Agel et al. (2005) also found that the rate of ACL injury remained stable for both males and females across the 13 year span of their study, and that the rates for all injuries in female players were significantly higher than male players regardless of participation in soccer or basketball. The rates for all non-contact ACL injuries in females were significantly higher than the rates for all non-contact ACL injuries for males, also regardless of the sport. The difference between males and females did not diminish over this lengthy time period despite vigorous efforts to address the concern (Arendt and Dick 1995; Agel, Arendt et al. 2005).

Jump-landing Task

As can be seen from discussions previously mentioned above, the jump-landing task could be a possible mechanism of ACL injury due to the type of task, lower extremity posture, and ATSF generated. Landing from a jump is a common and essential sports specific task. More specifically, basketball, volleyball, and soccer are all sports that implement landing from a jump on a consistent basis. Kirkendall and Garrett (2000) reported that ACL injuries occurring in competitive sports such as basketball and soccer were non-contact in nature 88% of the time, and a result of a deceleration type of movement. Landing from a jump was reported at least 41% of the time in these cases. It has also been reported that landing from a jump is one of the primary non-contact mechanisms for ACL injury in female basketball and volleyball players (Ferretti,
Based on the demanding nature of the jump-landing task, some researchers have concluded that this may be a mechanism for ACL injury.

The most common phase of the jump-landing task analyzed during research is initial ground contact. When examined in previous literature, hip and knee kinematics at initial foot contact with the ground have been demonstrated to affect the lower extremity kinetics of the landing in the jump task (Yu, McClure et al. 2005). The landing motion has also been associated with notable ATSF (Chappell, Yu et al. 2002; Chappell, Herman et al. 2005; Kernozek, Torry et al. 2005). All of these factors provide insight to the possible causes of ACL rupture in athletes, especially female participants.

Prevention strategies are necessary to reduce the risk of ACL injury found in the athletic population, especially among females. Since most intrinsic risk factors are non-modifiable, focusing on the extrinsic factors such as kinetics, kinematics, and lower extremity posture during sports specific activity should be moved to the forefront. Proper landing biomechanics may decrease the amount of ATSF in females, thus decreasing the risk for ACL injury. Teaching athletes to avoid vulnerable lower extremity joint angles upon ground contact from jumping activities or awkward landing positions may reduce the incidence of ACL injuries in females. Overall, this study will aid in the design of prevention programs to lower the incidence of ACL injury.

The equipment commonly used to study jump-landings and ATSF include the Flock of Birds and the force plate. The Flock of Birds is an electromagnetic motion analysis system that monitors limb segment position and orientation of a receiver with respect to a transmitter in 6 degrees of freedom, using pulsed direct current (DC) magnetic fields. Stokdijk et al. (2000) found that the Flock of Birds (FoB) was a reliable and valid
measure used to determine position and orientation of a joint. Due to the use of only two observers in this study, intraobserver reliability was found to be poor to excellent for position and fair to good for orientation. Interobserver reliability was poor for position but good to excellent for orientation. This showed no difference between inter- and intraobserver reliability, but found that the differences between position and orientation can be attributed to the small variation occurring between subjects. However, these authors concluded that the FoB system is accurate and the method is easy to perform for both the investigator and the subject being tested. Additionally, the results were found to be reproducible. Many other studies have agreed in their conclusions that the FoB system has been found to be accurate and easy to use (Milne, Chess et al. 1996; Meskers, Fraterman et al. 1999; Stokdijk, Biegstraaten et al. 2000). In a study performed by Milne et al. (1996) the FoB was found to have positional and rotational errors of less than 2% when utilized within its optimal operating range. The device is sensitive enough to read positional and rotational changes of 0.25mm and 0.1 degrees, respectively (Milne, Chess et al. 1996). Special care should be taken to measure the bony landmarks consistently, as this is the factor found to be the most important source of errors.

Force plate data are also collected in order to determine the ground reaction forces and peak ATSF upon landing from the jump. Literature shows that landing with a large ground reaction force (GRF) may be a risk factor for knee injuries, especially to the ACL (Malinzak, Colby et al. 2001; Chappell, Yu et al. 2002; Decker, Torry et al. 2003; Pflum, Shelburne et al. 2004). Pflum et al. (2004) determined that the ACL load during landing is largely explained by the magnitude and direction of the resultant GRF relative to the knee joint center. The GRF is commonly calculated during the first 20% of the stance
phase of a jump-landing task following initial contact. GRF reaching 3 to 14 times body weight have been measured for landing activities, suggesting that the body absorbs tremendous loads during these athletic tasks (Dufek and Bates 1991; Huston, Greenfield et al. 2000). Smaller degrees of knee flexion may be associated with increased peak vertical GRF. Dufek and Bates (1991) reported that the lower the GRF, the less chance of injury. Peak ATSF has also been found to occur at or just after initial contact, as the GRF is increasing to its peak. In a study by Pflum et al. (2004), peak shear force induced by the ground reaction force was 1,214 N, and occurred almost immediately after initial impact from a jump. Inverse dynamics are used to derive ATSF from the forces generated on the force plate.

**Purpose**

The purpose of this study is to identify if a relationship exists between anterior tibial shear force and knee flexion, hip flexion, and trunk flexion angles during a jump-landing task.
CHAPTER 3
METHODS

Subjects

Thirty-three female subjects (age =19.7 ± 2.0 years, height =165.23 ± 6.23 cm, weight = 61.7 ± 9.7 kg) were recruited from the University of North Carolina at Chapel Hill student population to participate in the study. To be eligible for participation, all subjects were required to be between the ages of 18-25 years. Subjects were recruited from physical education classes as well as from posted flyers. Exclusion criteria included any history of ACL injury, knee surgery in the past two years, and current lower extremity injury that limited participation in physical activity. Due to previously noted gender differences in jump-landing technique, and the increased risk of ACL injury found in female athletes, this study focused on female recreational athletes. Female recreational athletes were defined as those participating in sports activity 2-3 times per week for an average of 30 minutes per session. Each subject was required to have 2 years of varsity, club, or intramural experience in either basketball, volleyball, or soccer, as these sports implement a jump-landing task. In addition, to avoid possible fatigue, no testing took place within one hour of physical activity or any other strenuous activity.
Instrumentation

Motion Analysis

An electromagnetic tracking system, Flock of Birds®, (Ascension Technologies, Inc., Burlington, VT), and Motion Monitor Software® (Innovative Sports Training, Inc. Chicago, IL) were used to collect three dimensional lower extremity kinematic data at a sampling rate of 144Hz. The system employs a DC transmitter with three orthogonal coils to generate a magnetic field. Four sensors/receivers record the electromagnetic flux in the field generated by the transmitter, and convey those signals to a computer via hard wiring. The electromagnetic tracking system was calibrated prior to data collection. The transmitter is affixed to a stationary stand, 0.914 meters in height, to establish the global reference system used during testing. An embedded right-hand Cartesian coordinate system was defined for the shank, thigh, hip, and trunk to describe the three-dimensional positions and orientations of these segments. The reference system axes were defined so that the x-axis was positive in the anterior direction, the positive y-axis was directed to the left of the individual, and the positive z-axis was designated superiorly to the subject.

Force Plate

To measure ground reaction forces (GRF), subjects performed a jump-landing task on a non-conductive force platform (Bertec 4060-10, Chicago, IL), sampling at 1,440 Hz, and using the same right-hand Cartesian coordinate system (x, y, z) described above.
Procedure

Setting

All subjects reported to the Sports Medicine Research Laboratory for testing. Testing for each subject was performed during a single testing session lasting approximately 60 minutes. Prior to testing, all subjects read and signed an informed consent form, approved by the Biomedical Institutional Review Board at the University of North Carolina at Chapel Hill.

Subject Preparation

Each subject completed a health questionnaire form, and his/her anthropomorphic data were collected, including age(years), height(cm), mass(kg), and leg dominance prior to the start of the testing session. A distance equal to 50% of the subject’s body height was calculated prior to the start of testing to determine each subject’s jump distance from the force plate. The dominant leg of each subject was used as the test limb. Dominant leg was defined as the leg chosen by the subject to kick a ball for maximum distance. Each subject was instructed to wear athletic shoes, athletic shorts or spandex, and a t-shirt during the testing session. All subjects were allotted five minutes for warm-up on a stationary bike at moderate intensity. Following proper warm-up, the examiner demonstrated the jump-landing task, and the subjects were allowed to practice this task a maximum of five times.
Electromagnetic Sensor Placement

The final step in preparing the subject for data collection included placement and set-up of electromagnetic sensors for the kinematic analyses. A total of four electromagnetic sensors were used. Electromagnetic sensors were placed on the trunk, sacrum, thigh, and shank. The sensors were fixed to the subject with doubled-sided tape, pre-wrap, and athletic tape. The sacrum marker was placed at the midpoint of the sacrum between the PSIS. The thigh marker was placed on the flattest portion of the distal one-third of the iliotibial band in order to minimize movement due to muscle contraction. The marker for the shank was placed just distal to the pes anserine insertion on the flat portion of the medial side of the proximal tibia. The trunk marker was placed on the C7 vertebrae. This set-up allowed for data to be recorded for knee flexion angle, hip flexion angle, and trunk flexion angle.

With the markers in place, each joint was digitized with a moveable electromagnetic sensor. Subjects were asked to stand in a neutral position on the force plate while the following bony landmarks were digitized in order to define segments and joint centers: T12 spinous process, xyphoid process, medial femoral condyle, lateral femoral condyle, medial malleolus, and lateral malleolus. Hip joint center was calculated via the method recommended by Bell, from left ASIS to right ASIS (Bell, Pedersen et al. 1990). Data from digitizing each joint were stored in the computer and used to calibrate the subject before each jump-landing trial.
**Data Collection Procedure**

The jump-landing task performed in this study required the athlete to perform a jump off a 30-cm high platform over a horizontal distance equal to 50% of their height, and land with each foot on separate force plates placed side by side. Subjects were then instructed to immediately jump vertically for maximum height. Each subject performed 10 jump-landings with 30 seconds of rest time between trials to minimize the risk of fatigue. Trials in which the subject failed to land with the foot of the dominant leg on the force plate were removed, and a new trial was performed.

**Data Analysis**

**Data Processing**

Lower extremity sagittal plane angles were recorded at both initial contact and at the time of peak ATSF during the jump-landing task. Anterior tibial shear forces (ATSF) were normalized to body weight in Newtons (N) in order to run correlation and regression analyses on each subject. ATSF was recorded as the peak value occurring in the time period from initial contact (IC) to maximum knee flexion angle during the landing phase. This ensured that the peak ATSF value occurred when the subject was decelerating her body from the jumping task. Peak ATSF induced by GRF has been found to occur almost immediately after initial impact from a jump (Pflum, Shelburne et al. 2004). Initial contact for this study was defined as the first data point measuring vertical GRF above 10 N. Kinematic data were low pass filtered using a using a 4th order zero phase lag Butterworth filter at 14.5 Hz. (Yu, Gabriel et al. 1999). Euler angles were used to calculate the sagittal plane knee, hip, and trunk joint angles with a Y,X,Z order of
rotation. The flexion-extension angles were determined at the knee joint between the shank and thigh, at the hip joint between the thigh and pelvis, the trunk joint between the pelvis and thoracic spine, and the trunk relative to the world. The Flock of Birds was integrated with a force plate, and ATSF was calculated through the use of inverse dynamics.

Statistical Analysis

All data were analyzed using SPSS 14.0 statistical software (Chicago, IL). An *a priori* alpha-level of 0.05 was established prior to the beginning of data analysis. Additionally, an *a priori* statistical power analysis was performed based on pilot data collected previously by the principle investigator. It was found that to attain a statistical power of 0.80, an estimated 12 subjects would be needed for data collection. Within subject Pearson correlation analyses were completed to determine if a relationship exists between ATSF and knee, hip, or trunk sagittal plane joint angles for all trials. Additionally, the average of the 10 trials for each of the four variables was calculated in order to examine the correlation of the variables across all subjects. Following the correlation analyses, within subject forward step-wise multiple regression analyses were performed to determine if trunk flexion, hip flexion, and/or knee flexion angles at IC or at the time of peak ATSF significantly predict ATSF during a jump-landing task. We also performed a forward stepwise multiple regression for the averages of data across all subjects to see how well each variable predicts ATSF. Forward stepwise regressions were completed by putting the variable with the highest correlation to ATSF into the regression equation.
first. If this produced a significant finding, then the variable with the next highest correlation was added to the equation to determine any significant association.
CHAPTER 4
RESULTS

Between Subject Analyses

Thirty-three female subjects (age = 19.7 ± 2.0 years, height =165.23 ± 6.23 cm., weight = 61.7 ± 9.7 kg.) were tested; however the data for two subjects were not included for the analysis due to hardware problems with the force plate that occurred in all 10 trials. Additionally, four subjects had a minimal number of trials affected, but their data from unaffected trails was still usable. Subjects 9 and 19 had nine trials used for data analysis, subject 12 had seven trials, and subject 13 had eight trials used for data analysis. Of the thirty-one subjects analyzed, no significant correlations were found for knee flexion, hip flexion, trunk flexion, or trunk flexion world angles at IC with peak ATSF (p>0.05). Additionally, a combination of knee flexion, hip flexion, trunk flexion, and trunk flexion world at initial contact was not a significant predictor of peak ATSF. Knee flexion angle at the time of peak anterior tibial shear force was found to have a significant negative relationship with peak ATSF (p=0.021, r = -0.367). Additionally, trunk flexion world at time of peak ATSF was found to have a significant positive relationship with peak ATSF (p=0.046, r =0.309). There were no significant correlations found for hip flexion angle and trunk flexion angle at time of peak ATSF and peak ATSF (p>0.05). From the regression analysis, knee flexion angle at the time of peak ATSF was found to
predict approximately 13.5% of the variance in ATSF (p=0.042, R²=0.135), which may be seen in Figure 1.

**Within Subject Analyses**

**Initial Contact Angles**

Of the 31 female subjects analyzed, there were a few correlations that were significant between ATSF and each dependent variable (Table 5). Four significant within subject correlations were found between knee flexion angle at IC and peak ATSF, two significant within subject correlations were found between hip flexion angle at IC and peak ATSF, three significant within subject correlations were found between trunk flexion angle at IC and peak ATSF as well as trunk flexion angle relative to the world at IC and peak ATSF. Five significant regressions were found including 3 with knee flexion, 1 with hip flexion, and 1 with trunk flexion relative to the world. The significant p-values, r-values, and R²-values may be found in Table 5.

**ATSF Angles**

Of the 31 female subjects tested, 4 significant correlations were found between the knee flexion angle at the time of peak ATSF and peak ATSF. Five significant correlations were found between the hip flexion angle at the time of peak ATSF and peak ATSF. Three significant correlations each were found between the trunk flexion angle at the time of peak ATSF and peak ATSF as well as between the trunk flexion relative to the world angle at the time of peak ATSF and peak ATSF. Ten total significant regressions were found across 9 subjects including 2 with knee flexion angle at the time

34
of peak ATSF, 4 with hip flexion angle at the time of peak ATSF, 1 with trunk flexion angle at the time of peak ATSF, and 3 with trunk flexion angle relative to the world at the time of peak ATSF. The significant p-values, r-values, and $R^2$-values may be found in Table 6.
CHAPTER 5
DISCUSSION

The purpose of this study was to determine the relationships between sagittal plane knee, hip, and trunk angles with peak ATSF during a jump-landing task. The relationships were assessed using both within-subject and between-subject analyses. The within-subject analyses involved investigating the relationships between sagittal plane joint angles and peak ATSF for each trial across 10 jump-landing trials for each subject. The between-subject analyses involved investigating the relationships between the average sagittal plane joint angles and average peak ATSF during the 10 jump-landing trials across all subjects. The between subject correlation analyses revealed that only knee flexion ($r = -0.37$) and trunk flexion relative to the vertical world axis ($r = 0.31$) angles at the time of peak ATSF were significantly related to peak ATSF. The direction of the correlation indicates that less knee and more trunk (world) flexion resulted in larger peak ATSF values during a jump-landing task. While the relationship between knee flexion angles and ATSF were predicted, the positive correlation between trunk flexion relative to the world vertical axis and peak ATSF was unexpected. The fact that trunk flexion relative to the pelvis segment was not significant ($r = -0.09$, $p = 0.31$) is also unexpected. These results may be due to the way trunk flexion relative to the world vertical axis was defined. Trunk flexion relative to the world was measured as the angle between the longitudinal axis of the thorax segment and the vertical
axis of the world axis system. Trunk flexion was also defined as the angle between the longitudinal axis of the thorax relative to the vertical axis of the pelvis at the T12-L1 joint. Due to the difference in measurements, the trunk flexion angle relative to the world may be influenced by other body motions (hip and lumbar spine flexion). For instance, an individual who exhibits a large trunk flexion angle relative to the world axis may purely exhibit trunk flexion of the thorax at the T12-L1 joint, or they may be in a neutral trunk position but be in a flexed position at the hip. Trunk flexion relative to the pelvis should not be influenced by these other movements. Thus, the finding of a positive correlation between trunk flexion relative to the world and ATSF may not reflect true trunk flexion.

The positive correlation between trunk flexion relative to the world may also be a response to the relative contributions of the knee and hip to energy absorption during a landing task. Low knee flexion angles have already been found to increase peak ATSF, but it may also cause an increase in hip and trunk flexion angles in order to absorb the forces transmitted up the kinetic chain. If the subject were to land in an extended knee position, the hip and trunk may flex more in order to compensate and absorb the forces. Thus, while we found that increased trunk flexion relative to the world is associated with increased ATSF, it may be a product of the decreased knee flexion angle also found in this study.

Multiple regression analyses revealed that knee flexion angle at the time of peak ATSF was the only variable to significantly predict peak ATSF ($R^2 = 0.135$) as no other variables explained additional variability in peak ATSF and entered into the regression model. Thus, knee flexion angle at the time of peak ATSF appears to have the most prominent influence on peak ATSF values during a jump-landing task.
The within-subject analyses revealed that only 10 of 31 subjects (32%) demonstrated significant relationships between either sagittal plane joint angles at initial contact or at the time of peak ATSF with peak ATSF across the 10 jump-landing trials. The adjusted $R^2$ values in these 9 subjects ranged from 0.34 to 0.85, suggesting that in these individuals sagittal plane joint position had a moderate to large effect on peak ATSF during the 10 jump-landing trials. Knee flexion at initial contact and hip and trunk (world) flexion at the time of peak ATSF appear to be the most important factors that influence peak ATSF, as these were the variables that were most commonly found to be significantly related to peak ATSF. The within-subject analyses indicate that the influence of sagittal plane joint angles on peak ATSF varies across subjects, as not all subjects demonstrated significant relationships and the specific sagittal plane joint angles that were significantly related to peak ATSF were not consistent across subjects. Thus, the relationships between peak ATSF and sagittal plane joint angle differ within subjects, and cannot be generalized across subjects.

Knee flexion angles and their affect on strain of the ACL have been examined frequently in previous literature. Multiple studies found that when the knee angle is between 50 degrees and full extension, the ACL is placed under strain (Markolf, Burchfield et al. 1995; Beynnon and Fleming 1998). Subsequent investigations have also demonstrated that the ACL is under increasing strain as the knee moves into full extension, and that ATSF increases as the knee moves close to full extension (Osternig, Ferber et al. 2001). Fagenbaum and Darling (2003) found that ACL forces progressively decrease as the knee is flexed to 15, 30, and 60 degrees. Colby et al. (2000) reported the average knee flexion angle at foot strike to be 23 degrees when landing from a jump, and that the average angle of knee flexion at the time of ACL injury during a jump is 21 degrees.
Knee flexion angle at IC resulted in only four significant correlations with peak ATSF. However, the findings corresponded to those presented in previous research. Of these four subjects, the average angle of knee flexion for subject 15 was 11.15 degrees with a corresponding average of 483.13N of peak ATSF. Subject 11 had a knee flexion angle of 19.95 degrees at IC with an average peak ATSF of 178.82N. Subject 14 had 22.22 degrees of knee flexion at IC with a corresponding average peak ATSF of 533.33 N. Only subject 25 had a flexion angle of greater than 25 degrees, with an average of 27.27 degrees. The corresponding average peak ATSF was 34.72N. It is also important to note that the average ATSF values were much higher for the three subjects with angles under 20 degrees of knee flexion than the average ATSF found when the knee flexion angle was over 25 degrees. The three subjects who demonstrated a significant correlation between knee flexion angle at IC and ATSF had a knee flexion angle of less than 25 degrees, while the fourth had a knee flexion angle of under 30 degrees, which has been stated as a dangerous zone in previous research to increase risk of an ACL injury.

Similar results were found when examining the knee flexion angles at the time of peak ATSF. Knee flexion angle at the time of peak ATSF resulted in only four significant correlations and explained only 13.5% of the variance in ATSF. This indicates that knee kinematics alone do not adequately explain the presence of ATSF in most subjects. Subject 8 had a knee flexion angle of 22.37 degrees at the time of peak ATSF with an associated ATSF of 500.95N. Subject 33 had an average knee flexion angle of 23.52 degrees to go along with an average of 240.65 N of peak ATSF. Subject 16 had an average of 25.65 degrees of knee flexion at the time of peak ATSF to go along with 297.94 N of peak ATSF. The fourth significant correlation was found in subject 25, with a flexion angle average of 43.59 degrees.
and a corresponding average of 34.72 N of peak ATSF. These results show the trend of decreasing ATSF when the knee flexion angle increases. It is interesting to note the knee flexion angles at the time of peak ATSF as well, as three of the four significant correlations found had subjects with a knee flexion angle less than 25 degrees even past the IC point.

These data may indicate that knee flexion only has a strong relationship to ATSF at low flexion angles. In the literature, these low flexion angles are associated with an increased risk of ACL injury due to more risky landing strategies. Thus, for some individuals who exhibit knee flexion angles below 25 degrees, ATSF may be higher and create the increased risk of injury.

Hip and trunk position and motion may have an influence on knee positions and knee loads because of the kinetic chain relationship between the lower extremity and force transmission up and down this chain. Previous literature has shown that females perform cutting and landing maneuvers in a more erect posture than males, with less hip flexion and less knee flexion (Huston, Vibert et al. 2001; Malinzak, Colby et al. 2001; Decker, Torry et al. 2003). Additionally, Lephart et al. (2002) found that quadriceps activity and, therefore, possibly ATSF, have been reported to decrease when hip flexion angles are increased. However, females tend to land with smaller angles of hip flexion compared with males. This combined effect suggests that smaller hip flexion angles may very well affect ACL strain values and predispose female athletes to ACL injury. However, not enough research has been completed in this area to understand the effect of hip angle during a jump-landing task, thus the inclusion of this variable in our study.

Hip flexion angle at initial contact did not seem to be an important variable when examining any between or within-subject relationship with peak ATSF. Two significant
correlations were found in the within subject IC data. Subject 4 was found to have a significant correlation, displaying an average of 47.84 degrees of hip flexion corresponding with an average of 329.15N of ATSF, while subject 11 had an average of 26.89 degrees of flexion with a corresponding average of 178.82N of ATSF. When examining the hip flexion angles at the time of peak ATSF, the broad range of degrees of flexion with the correlated average peak ATSF also suggested that hip flexion angle may not be an important contributor to the peak ATSF during a jump-landing task as no consistent trends could be found. Further research should be completed as to the role the hip joint may have on ACL injury.

Trunk flexion angle and its role in ACL injury and amount of ATSF generated have not yet been extensively examined in previous research to date. Thus, our study included this variable in the lower extremity posture examination to determine if any such relationship exists. Kulas et al. compared the trunk relative to the pelvis, through a “joint” modeled at T12-L1, and the hip joint in order to determine whether sex differences in landing posture arise from the pelvis-femoral (hip joint) articulation or from the trunk relative to the pelvis. They concluded that females demonstrated greater trunk extension relative to the pelvis than males while there were no sex differences in hip flexion angle during landing. These results provide validity to the common assumption that females land with a more erect trunk posture than males and also provides the rationale for modeling the trunk and pelvis as two separate segments in order to differentiate between trunk-pelvis versus hip joint motion (Kulas AS and Henning JM 2007).

We included both the trunk flexion angle relative to the pelvis as well as the trunk flexion angle relative to the world, in addition to the hip joint motion, due to the uncertainty in current literature as to which variable is the best predictor of actual trunk motion. It is
possible that the trunk flexion relative to the pelvis is a more biomechanically correct assessor, while the trunk flexion relative to the world is one that can been seen more easily clinically. It is interesting to note that the values of the trunk flexion angle relative to pelvis and trunk flexion angle relative to the world data in this study were different, with no two numbers being the same. The values found with the trunk flexion angles relative to the world were almost always higher than the values of trunk flexion relative to pelvis in each subject. More research should be completed as to which of these angles is more accurate to the data of interest.

In the between subject data, the trunk flexion relative to the world angle at the time of peak ATSF was correlated significantly with peak ATSF. Trunk flexion angle at IC found three significant correlations each for trunk flexion relative to the pelvis and trunk flexion relative to the world. Two of the subjects had significant correlations for both trunk flexion and trunk flexion relative to the world and peak ATSF. Subject 9 had a trunk flexion angle of -4.12 degrees, trunk flexion relative to the world angle of 10.11 degrees and a peak ATSF of 421.97 N. Subject 22 was similar as she had 13.91 degrees of trunk flexion, 21.58 degrees of trunk flexion relative to the world, and a peak ATSF of 190.15 N. Subject 2 had 17.82 degrees of trunk flexion relative to the world with a peak ATSF of 426.03N while subject 11 had 19.71 degrees of trunk flexion with a corresponding peak ATSF of 178.82 N.

When the trunk flexion angles at the time of peak ATSF were examined, three significant correlations were found for both types of trunk flexion angles as well. Subject 9 had correlations for both types of flexion angle tested, with -3.86 degrees of trunk flexion relative to the pelvis and 10.64 degrees of trunk flexion relative to the world. The correlating average peak ATSF was 421.97 N. Subject 2 had an average of 16.66 degrees of trunk flexion
relative to the world with a corresponding average 426.03 N of peak ATSF. Subject 13 had an average trunk flexion relative to the world angle of 8.45 degrees with a corresponding average peak ATSF of 241.24 N. Subject 11 had 18.86 degrees of trunk flexion relative to the pelvis that was correlated with 178.82N of peak ATSF. Finally, subject 23 had 16.54 degrees of trunk flexion relative to the pelvis that was correlated with only an average of 16.78N of peak ATSF.

When looking at the trunk flexion relative to the pelvis values, the lowest flexion angles corresponded to the highest peak ATSF values, while the higher flexion angles resulted in a decrease in peak ATSF. These numbers show a trend of decreasing peak ATSF when the trunk flexion angle increases, and may give more evidence to the biomechanical accuracy of this trunk measurement. However, the values seen between the trunk flexion angles relative to the world and peak ATSF showed no consistent trends. Further studies should be completed to find supporting results.

Another note of interest is the mean flexion angle at the time of peak ATSF compared to the mean flexion angle of each joint at IC. Both the knee and hip flexion angles increased in degree from IC to the time of peak ATSF as would be expected, whereas both trunk flexion angle relative to the world and trunk flexion angle relative to the pelvis decreased in degrees from IC to the time of peak ATSF. Also, the influence of IC angles versus peak ATSF angles on peak ATSF values was an interesting finding of the study. The between subject analyses suggest that peak ATSF values are more sensitive to changes in sagittal plane joint angles at the time of peak ATSF than at IC. There were no significant correlation or regression analyses for joint angles at initial contact, but there were for joint angles at time of peak ATSF.
Limitations

A potential limitation of this study is that we only examined the flexion angles in association with a jump-landing task. The relationships between sagittal plane angles with peak ATSF may differ depending upon the nature of the task, so future research involving other non-contact athletic movements may be helpful in determining if a risk factor exists. Another limitation may be that muscle activity was not completed during the jump-landing task. It is possible that the activity of sagittal plane musculature, such as the gluteus maximus, hamstrings, quadriceps, and gastrocnemius muscles, is related to peak ATSF during a jump-landing task. Gaining an understanding of other factors that may influence ATSF during a jump-landing task may help decrease ACL injury risk.

Implications on ACL Injury Prevention

From these results, it may be concluded that the lower extremity joint angles during a jump-landing task at both IC and at the time of peak ATSF may not be the most important factors to examine for ACL injury prevention. This may have implications on ACL prevention training. The ultimate goal from this study as well as from athletic training clinicians in general is to reduce ACL injury risk in the athletic population. If this is to be done by decreasing ATSF, than the results of this study suggest that we need to work on getting athletes to flex their knees, hip, and trunk to a greater extent. Knee flexion angle is an important factor in decreasing ACL injury risk, however, just focusing on the knee will not necessarily work for all subjects as it does not incorporate all joints in the kinetic chain, so we also need to train individuals to focus on hip and trunk flexion as well in order to
decrease their individual risk for ACL injury. Also, since sagittal plane joint angles did not explain all the variability in peak ATSF, it suggests that there are other factors that are related to peak ATSF values. Some of these factors may possibly be EMG activity of the quadriceps and hamstring muscles during a jump-landing task, the quad/hamstring ratio, the ground reaction force when landing, and possibly even the moments and velocities associated with the lower extremity joints during a jump-landing task. Future research needs to be completed in order to investigate these factors so that we have a better understanding of all factors that may influence peak ATSF.

Conclusions

The current study examined the relationship between knee flexion, hip flexion, and trunk flexion angles and peak ATSF at both initial contact and the time of peak ATSF. Based on the results of this study, we conclude the following:

1. Knee flexion angle at the time of peak ATSF appears to have the most prominent influence on peak ATSF values during a jump-landing task for between subject data, with the direction of the correlation indicating that less knee and trunk (world) flexion results in larger peak ATSF values.

2. Knee flexion at IC and hip and trunk (world) flexion at the time of peak ATSF appear to be the most important factors that influenced peak ATSF for within subject data, however the influence of sagittal plane joint angles on peak ATSF is not consistent across subjects, therefore the relationships between peak ATSF and sagittal plane joint angles cannot be generalized across subjects.
3. The knee is an important factor, however just focusing on the knee will not necessarily work for all subjects, so we also need to train individuals to focus on hip and trunk flexion as well in order to decrease their risk for ACL injury.

4. Sagittal plane joint angles did not explain all the variability in peak ATSF, therefore there are other factors that are related to peak ATSF values that need to be examined.
Appendices

Appendix A: Tables
Table 1: Data Analysis Plan

<table>
<thead>
<tr>
<th>RQ</th>
<th>Description</th>
<th>Data Source</th>
<th>Method</th>
</tr>
</thead>
</table>
| 1  | Is there a relationship between ATSF and knee flexion angle during a jump-landing task? | Dependent Variables: ATSF  
Independent Variables: Knee flexion angle (IC) Knee flexion angle (pATSF) | Separate Pearson Product Correlations were performed between knee flexion angle at the time of peak ATSF and at IC with ATSF to determine if a relationship exists between the variables |
| 2  | Is there a relationship between ATSF and hip flexion angle during a jump-landing task? | Dependent Variables: ATSF  
Independent Variables: Hip flexion angle (IC) Hip flexion angle (pATSF) | Separate Pearson Product Correlations were performed between hip flexion angle at the time of peak ATSF and at IC with ATSF to determine if a relationship exists between the variables |
| 3  | Is there a relationship between ATSF and trunk flexion angle during a jump-landing task? | Dependent Variables: ATSF  
Independent Variables: Trunk flexion angle (IC) Trunk flexion angle (pATSF) | Separate Pearson Product Correlations were performed between each trunk flexion angle at the time of peak ATSF and at IC with ATSF to determine if a relationship exists between the variables |
| 4  | Does knee flexion, hip flexion, and/or trunk flexion angle predict ATSF during a jump-landing task? | Dependent Variables: ATSF  
Independent Variables: Knee flexion angle (IC) Knee flexion angle (pATSF) Hip flexion angle (IC) Hip flexion angle (pATSF) Trunk flexion angle (IC) Trunk flexion angle (pATSF) | Forward step-wise multiple regression analyses were performed to determine how well each variable predicts ATSF |
Table 2: Descriptive Statistics

<table>
<thead>
<tr>
<th>Variable</th>
<th>N</th>
<th>Mean</th>
<th>SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age</td>
<td>31</td>
<td>19.71</td>
<td>2.003</td>
</tr>
<tr>
<td>Height</td>
<td>31</td>
<td>165.226</td>
<td>6.233</td>
</tr>
<tr>
<td>Weight</td>
<td>31</td>
<td>61.703</td>
<td>9.692</td>
</tr>
<tr>
<td>Peak Anterior Tibial Shear Force (ATSF)</td>
<td>31</td>
<td>0.481</td>
<td>0.357</td>
</tr>
<tr>
<td>Knee Flexion Angle at IC</td>
<td>31</td>
<td>20.523</td>
<td>6.135</td>
</tr>
<tr>
<td>Hip Flexion Angle at IC</td>
<td>31</td>
<td>-40.263</td>
<td>11.453</td>
</tr>
<tr>
<td>Trunk Flexion Angle at IC (pelvis)</td>
<td>31</td>
<td>8.412</td>
<td>9.801</td>
</tr>
<tr>
<td>Trunk Flexion Angle at IC (world)</td>
<td>31</td>
<td>17.291</td>
<td>6.557</td>
</tr>
<tr>
<td>Knee Flexion Angle at peak ATSF</td>
<td>31</td>
<td>36.585</td>
<td>11.202</td>
</tr>
<tr>
<td>Hip Flexion Angle at peak ATSF</td>
<td>31</td>
<td>-46.985</td>
<td>13.652</td>
</tr>
<tr>
<td>Trunk Flexion Angle at peak ATSF (pelvis)</td>
<td>31</td>
<td>5.889</td>
<td>9.874</td>
</tr>
<tr>
<td>Trunk Flexion Angle at peak ATSF (world)</td>
<td>31</td>
<td>15.427</td>
<td>6.886</td>
</tr>
<tr>
<td>Joint</td>
<td>IC Correlation</td>
<td>IC Regression</td>
<td>pATSF Correlation</td>
</tr>
<tr>
<td>------------------------</td>
<td>----------------</td>
<td>---------------</td>
<td>-------------------</td>
</tr>
<tr>
<td>Knee flexion</td>
<td>4</td>
<td>3</td>
<td>4</td>
</tr>
<tr>
<td>Hip flexion</td>
<td>2</td>
<td>1</td>
<td>5</td>
</tr>
<tr>
<td>Trunk flexion (world)</td>
<td>3</td>
<td>1</td>
<td>3</td>
</tr>
<tr>
<td>Trunk flexion (pelvis)</td>
<td>3</td>
<td>0</td>
<td>3</td>
</tr>
</tbody>
</table>
Table 4: Between Subject Number of Significant Findings at Initial Contact (IC) and Peak Anterior Tibial Shear Force (pATSF)

<table>
<thead>
<tr>
<th>Joint</th>
<th>IC Correlation</th>
<th>IC Regression</th>
<th>pATSF Correlation</th>
<th>pATSF Regression</th>
</tr>
</thead>
<tbody>
<tr>
<td>Knee flexion</td>
<td>0</td>
<td>0</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td>Hip flexion</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>Trunk flexion (world)</td>
<td>0</td>
<td>0</td>
<td>1</td>
<td>0</td>
</tr>
<tr>
<td>Trunk flexion (pelvis)</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>Subject</td>
<td>Knee Flexion</td>
<td>Hip Flexion</td>
<td>Trunk Flexion (pelvis)</td>
<td>Trunk Flexion (world)</td>
</tr>
<tr>
<td>---------</td>
<td>--------------</td>
<td>-------------</td>
<td>------------------------</td>
<td>-----------------------</td>
</tr>
<tr>
<td></td>
<td>ATSF</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1</td>
<td>0.04 ± 0.20</td>
<td>24.04 ± 8.87</td>
<td>39.72 ± 3.54</td>
<td>11.20 ± 3.62</td>
</tr>
<tr>
<td>2</td>
<td>0.61 ± 0.41</td>
<td>45.25 ± 21.13</td>
<td>-60.23 ± 18.02</td>
<td>0.27 ± 9.60</td>
</tr>
<tr>
<td>3</td>
<td>0.29 ± 0.19</td>
<td>43.52 ± 16.04</td>
<td>-40.66 ± 5.88</td>
<td>19.57 ± 2.90</td>
</tr>
<tr>
<td>4</td>
<td>0.54 ± 0.39</td>
<td>20.98 ± 9.39</td>
<td>-49.43 ± 3.35</td>
<td>5.80 ± 4.00</td>
</tr>
<tr>
<td>5</td>
<td>0.26 ± 0.31</td>
<td>53.78 ± 3.15</td>
<td>-64.74 ± 6.93</td>
<td>-9.70 ± 6.09</td>
</tr>
<tr>
<td>6</td>
<td>1.42 ± 1.04</td>
<td>15.56 ± 3.38</td>
<td>-31.62 ± 4.80</td>
<td>13.56 ± 3.11</td>
</tr>
<tr>
<td>7</td>
<td>0.25 ± 0.27</td>
<td>47.02 ± 8.33</td>
<td>-33.36 ± 6.65</td>
<td>14.79 ± 3.93</td>
</tr>
<tr>
<td>8</td>
<td>1.02 ± 0.85</td>
<td>22.37 ± 10.18</td>
<td>-41.27 ± 9.15</td>
<td>11.36 ± 4.73</td>
</tr>
<tr>
<td>9</td>
<td>0.79 ± 0.50</td>
<td>27.99 ± 17.80</td>
<td>-50.83 ± 11.31</td>
<td>-3.13 ± 4.54</td>
</tr>
<tr>
<td>10</td>
<td>0.24 ± 0.20</td>
<td>39.62 ± 19.05</td>
<td>-33.16 ± 6.94</td>
<td>18.86 ± 7.61</td>
</tr>
<tr>
<td>11</td>
<td>0.29 ± 0.39</td>
<td>28.82 ± 10.31</td>
<td>-23.06 ± 16.21</td>
<td>-6.65 ± 9.70</td>
</tr>
<tr>
<td>12</td>
<td>0.46 ± 0.53</td>
<td>39.63 ± 12.40</td>
<td>-54.85 ± 7.78</td>
<td>-1.38 ± 4.70</td>
</tr>
<tr>
<td>13</td>
<td>0.62 ± 0.26</td>
<td>46.04 ± 23.56</td>
<td>-47.96 ± 14.19</td>
<td>2.10 ± 8.21</td>
</tr>
<tr>
<td>14</td>
<td>0.64 ± 0.33</td>
<td>37.01 ± 16.81</td>
<td>-42.04 ± 12.75</td>
<td>4.68 ± 7.87</td>
</tr>
<tr>
<td>15</td>
<td>0.59 ± 0.30</td>
<td>25.65 ± 8.67</td>
<td>-30.16 ± 5.62</td>
<td>18.62 ± 3.38</td>
</tr>
<tr>
<td>16</td>
<td>1.07 ± 0.64</td>
<td>36.59 ± 13.81</td>
<td>-54.68 ± 10.21</td>
<td>13.96 ± 5.44</td>
</tr>
<tr>
<td>17</td>
<td>0.79 ± 0.40</td>
<td>28.26 ± 9.79</td>
<td>-48.80 ± 6.83</td>
<td>-1.59 ± 2.97</td>
</tr>
<tr>
<td>18</td>
<td>0.46 ± 0.55</td>
<td>49.52 ± 17.30</td>
<td>-68.70 ± 15.04</td>
<td>-3.44 ± 5.51</td>
</tr>
<tr>
<td>19</td>
<td>0.04 ± 0.10</td>
<td>33.39 ± 15.03</td>
<td>-62.04 ± 8.28</td>
<td>-1.27 ± 6.66</td>
</tr>
<tr>
<td>20</td>
<td>0.02 ± 0.05</td>
<td>50.69 ± 5.14</td>
<td>-50.21 ± 4.52</td>
<td>17.41 ± 2.76</td>
</tr>
<tr>
<td>21</td>
<td>0.35 ± 0.17</td>
<td>49.05 ± 4.50</td>
<td>-45.71 ± 4.29</td>
<td>11.63 ± 5.10</td>
</tr>
<tr>
<td>22</td>
<td>0.04 ± 0.14</td>
<td>42.91 ± 13.08</td>
<td>-44.39 ± 10.78</td>
<td>16.54 ± 7.54</td>
</tr>
<tr>
<td>23</td>
<td>0.11 ± 0.13</td>
<td>25.80 ± 3.88</td>
<td>-33.06 ± 7.88</td>
<td>20.97 ± 9.19</td>
</tr>
<tr>
<td>24</td>
<td>0.06 ± 0.21</td>
<td>43.59 ± 18.47</td>
<td>-41.91 ± 7.48</td>
<td>12.10 ± 5.05</td>
</tr>
<tr>
<td>25</td>
<td>0.28 ± 0.52</td>
<td>52.38 ± 8.15</td>
<td>-33.53 ± 4.45</td>
<td>13.53 ± 5.60</td>
</tr>
<tr>
<td>26</td>
<td>0.34 ± 0.45</td>
<td>18.67 ± 10.30</td>
<td>-31.17 ± 5.64</td>
<td>-2.25 ± 4.64</td>
</tr>
<tr>
<td>27</td>
<td>0.23 ± 0.34</td>
<td>28.39 ± 3.39</td>
<td>-42.03 ± 4.43</td>
<td>0.04 ± 3.36</td>
</tr>
<tr>
<td>28</td>
<td>0.39 ± 0.32</td>
<td>43.57 ± 17.60</td>
<td>-57.22 ± 16.79</td>
<td>-1.41 ± 10.74</td>
</tr>
<tr>
<td>29</td>
<td>0.49 ± 0.34</td>
<td>46.19 ± 15.25</td>
<td>-81.93 ± 9.33</td>
<td>-13.64 ± 6.36</td>
</tr>
<tr>
<td>30</td>
<td>0.61 ± 0.49</td>
<td>43.41 ± 13.69</td>
<td>-45.62 ± 11.78</td>
<td>9.17 ± 6.20</td>
</tr>
<tr>
<td>31</td>
<td>0.39 ± 0.26</td>
<td>23.52 ± 19.25</td>
<td>-72.58 ± 10.06</td>
<td>-9.37 ± 6.52</td>
</tr>
</tbody>
</table>

* denotes a significant finding (p < 0.05)
<table>
<thead>
<tr>
<th>Subject</th>
<th>ATSF Mean ± SD</th>
<th>Knee Flexion Mean ± SD</th>
<th>Hip Flexion Mean ± SD</th>
<th>Trunk Flexion (pelvis) Mean ± SD</th>
<th>Trunk Flexion (world) Mean ± SD</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>r   p</td>
<td>r   p</td>
<td>r   p</td>
<td>r   p</td>
</tr>
<tr>
<td>1</td>
<td>0.04 ± 0.20</td>
<td>24.04 ± 8.87</td>
<td>-39.72 ± 3.54</td>
<td>11.20 ± 3.62</td>
<td>16.00 ± 4.41</td>
</tr>
<tr>
<td>2</td>
<td>0.61 ± 0.41</td>
<td>45.25 ± 21.13</td>
<td>-60.23 ± 18.02</td>
<td>0.27 ± 9.60</td>
<td>16.66 ± 3.94</td>
</tr>
<tr>
<td>3</td>
<td>0.29 ± 0.19</td>
<td>43.52 ± 16.04</td>
<td>-40.66 ± 5.88</td>
<td>19.57 ± 2.90</td>
<td>18.79 ± 1.78</td>
</tr>
<tr>
<td>4</td>
<td>0.54 ± 0.39</td>
<td>20.98 ± 9.39</td>
<td>-49.43 ± 3.35</td>
<td>5.80 ± 4.00</td>
<td>22.40 ± 2.90</td>
</tr>
<tr>
<td>5</td>
<td>0.26 ± 0.31</td>
<td>53.78 ± 3.15</td>
<td>-64.74 ± 6.93</td>
<td>-9.70 ± 6.09</td>
<td>12.94 ± 2.55</td>
</tr>
<tr>
<td>6</td>
<td>1.42 ± 1.04</td>
<td>15.56 ± 3.38</td>
<td>-31.62 ± 4.80</td>
<td>13.56 ± 3.11</td>
<td>19.09 ± 3.01</td>
</tr>
<tr>
<td>7</td>
<td>0.25 ± 0.27</td>
<td>47.02 ± 8.33</td>
<td>-33.36 ± 6.65</td>
<td>14.79 ± 3.93</td>
<td>10.74 ± 4.29</td>
</tr>
<tr>
<td>8</td>
<td>1.02 ± 0.85</td>
<td>22.37 ± 10.18</td>
<td>-41.27 ± 9.15</td>
<td>11.36 ± 4.73</td>
<td>22.76 ± 5.44</td>
</tr>
<tr>
<td>9</td>
<td>0.79 ± 0.50</td>
<td>27.99 ± 17.80</td>
<td>-50.83 ± 11.31</td>
<td>-3.13 ± 4.54</td>
<td>11.00 ± 4.44</td>
</tr>
<tr>
<td>10</td>
<td>0.24 ± 0.20</td>
<td>39.62 ± 19.05</td>
<td>-33.16 ± 6.94</td>
<td>18.86 ± 7.61</td>
<td>14.38 ± 4.13</td>
</tr>
<tr>
<td>11</td>
<td>0.29 ± 0.39</td>
<td>28.82 ± 10.31</td>
<td>-23.06 ± 16.21</td>
<td>-6.65 ± 9.70</td>
<td>-0.73 ± 10.55</td>
</tr>
<tr>
<td>12</td>
<td>0.46 ± 0.53</td>
<td>39.63 ± 12.40</td>
<td>-54.85 ± 7.78</td>
<td>-1.38 ± 4.70</td>
<td>9.03 ± 2.25</td>
</tr>
<tr>
<td>13</td>
<td>0.62 ± 0.26</td>
<td>46.04 ± 23.56</td>
<td>-47.96 ± 14.19</td>
<td>2.10 ± 8.21</td>
<td>8.68 ± 4.78</td>
</tr>
<tr>
<td>14</td>
<td>0.64 ± 0.33</td>
<td>37.01 ± 16.81</td>
<td>-42.04 ± 12.75</td>
<td>4.68 ± 7.87</td>
<td>15.53 ± 4.97</td>
</tr>
<tr>
<td>15</td>
<td>0.50 ± 0.30</td>
<td>25.65 ± 8.67</td>
<td>-30.16 ± 5.62</td>
<td>18.62 ± 3.38</td>
<td>22.70 ± 3.25</td>
</tr>
<tr>
<td>16</td>
<td>1.07 ± 0.64</td>
<td>36.59 ± 13.81</td>
<td>-54.68 ± 10.21</td>
<td>13.96 ± 5.44</td>
<td>30.11 ± 5.43</td>
</tr>
<tr>
<td>17</td>
<td>0.79 ± 0.40</td>
<td>28.26 ± 9.79</td>
<td>-88.60 ± 6.83</td>
<td>-1.59 ± 2.97</td>
<td>17.35 ± 5.47</td>
</tr>
<tr>
<td>18</td>
<td>0.46 ± 0.55</td>
<td>49.52 ± 17.30</td>
<td>-87.60 ± 15.04</td>
<td>-3.44 ± 5.51</td>
<td>21.39 ± 4.64</td>
</tr>
<tr>
<td>19</td>
<td>0.04 ± 0.10</td>
<td>33.39 ± 15.03</td>
<td>-62.64 ± 8.28</td>
<td>-2.17 ± 6.66</td>
<td>20.36 ± 6.20</td>
</tr>
<tr>
<td>20</td>
<td>0.02 ± 0.05</td>
<td>50.69 ± 5.14</td>
<td>-50.21 ± 4.52</td>
<td>17.41 ± 2.76</td>
<td>15.45 ± 4.34</td>
</tr>
<tr>
<td>21</td>
<td>0.35 ± 0.17</td>
<td>49.05 ± 4.50</td>
<td>-45.71 ± 4.29</td>
<td>11.63 ± 5.10</td>
<td>17.05 ± 4.77</td>
</tr>
<tr>
<td>22</td>
<td>0.04 ± 0.14</td>
<td>42.91 ± 13.08</td>
<td>-44.39 ± 10.78</td>
<td>16.54 ± 7.54</td>
<td>17.15 ± 3.42</td>
</tr>
<tr>
<td>23</td>
<td>0.11 ± 0.13</td>
<td>25.80 ± 3.88</td>
<td>-33.06 ± 7.88</td>
<td>20.97 ± 9.19</td>
<td>19.93 ± 2.25</td>
</tr>
<tr>
<td>24</td>
<td>0.06 ± 0.21</td>
<td>43.59 ± 18.47</td>
<td>-41.91 ± 7.48</td>
<td>12.10 ± 5.05</td>
<td>11.06 ± 4.27</td>
</tr>
<tr>
<td>25</td>
<td>0.28 ± 0.52</td>
<td>52.38 ± 8.15</td>
<td>-33.15 ± 4.45</td>
<td>13.53 ± 5.60</td>
<td>10.75 ± 3.97</td>
</tr>
<tr>
<td>26</td>
<td>0.34 ± 0.45</td>
<td>18.67 ± 10.30</td>
<td>-31.17 ± 5.64</td>
<td>-2.25 ± 4.64</td>
<td>-2.02 ± 3.79</td>
</tr>
<tr>
<td>27</td>
<td>0.23 ± 0.34</td>
<td>28.39 ± 3.39</td>
<td>-42.03 ± 4.43</td>
<td>0.04 ± 3.36</td>
<td>4.43 ± 1.83</td>
</tr>
<tr>
<td>28</td>
<td>0.39 ± 0.32</td>
<td>43.57 ± 17.60</td>
<td>-81.93 ± 9.33</td>
<td>-1.41 ± 10.74</td>
<td>17.05 ± 2.63</td>
</tr>
<tr>
<td>29</td>
<td>0.61 ± 0.49</td>
<td>43.41 ± 13.69</td>
<td>-45.62 ± 11.78</td>
<td>9.17 ± 6.20</td>
<td>14.09 ± 5.50</td>
</tr>
<tr>
<td>30</td>
<td>0.39 ± 0.46</td>
<td>23.52 ± 19.25</td>
<td>-72.56 ± 10.06</td>
<td>-9.37 ± 6.52</td>
<td>24.06 ± 4.79</td>
</tr>
</tbody>
</table>

* denotes a significant finding (p<0.05)
Table 7: Initial Contact Regression Table

<table>
<thead>
<tr>
<th>Subject</th>
<th>$R^2$</th>
<th>Adjusted $R^2$</th>
<th>$P$</th>
<th>Predictor Variables</th>
</tr>
</thead>
<tbody>
<tr>
<td>2</td>
<td>0.64</td>
<td>0.59</td>
<td>0.01</td>
<td>Trunk Flexion (world)</td>
</tr>
<tr>
<td>4</td>
<td>0.56</td>
<td>0.51</td>
<td>0.01</td>
<td>Hip Flexion</td>
</tr>
<tr>
<td>11</td>
<td>0.81</td>
<td>0.79</td>
<td>&lt; 0.001</td>
<td>Knee Flexion</td>
</tr>
<tr>
<td>14</td>
<td>0.58</td>
<td>0.53</td>
<td>0.01</td>
<td>Knee Flexion</td>
</tr>
<tr>
<td>25</td>
<td>0.50</td>
<td>0.44</td>
<td>0.02</td>
<td>Knee Flexion</td>
</tr>
<tr>
<td>Subject</td>
<td>$R^2$</td>
<td>Adjusted $R^2$</td>
<td>$P$</td>
<td>Predictor Variables</td>
</tr>
<tr>
<td>---------</td>
<td>-------</td>
<td>----------------</td>
<td>------</td>
<td>----------------------------------------</td>
</tr>
<tr>
<td>2</td>
<td>0.48</td>
<td>0.42</td>
<td>0.03</td>
<td>trunk flexion (world)</td>
</tr>
<tr>
<td>9</td>
<td>0.50</td>
<td>0.43</td>
<td>0.03</td>
<td>trunk flexion (world)</td>
</tr>
<tr>
<td>11</td>
<td>0.89</td>
<td>0.85</td>
<td>0.001</td>
<td>trunk flexion (pelvis), hip flexion</td>
</tr>
<tr>
<td>13</td>
<td>0.58</td>
<td>0.51</td>
<td>0.03</td>
<td>trunk flexion (world)</td>
</tr>
<tr>
<td>16</td>
<td>0.45</td>
<td>0.38</td>
<td>0.03</td>
<td>knee flexion</td>
</tr>
<tr>
<td>17</td>
<td>0.40</td>
<td>0.32</td>
<td>0.05</td>
<td>hip flexion</td>
</tr>
<tr>
<td>23</td>
<td>0.47</td>
<td>0.41</td>
<td>0.03</td>
<td>hip flexion</td>
</tr>
<tr>
<td>25</td>
<td>0.45</td>
<td>0.38</td>
<td>0.04</td>
<td>knee flexion</td>
</tr>
<tr>
<td>33</td>
<td>0.42</td>
<td>0.34</td>
<td>0.04</td>
<td>hip flexion</td>
</tr>
</tbody>
</table>
Table 9: Between Subject Correlation Table

<table>
<thead>
<tr>
<th></th>
<th>Knee Flexion</th>
<th>Hip Flexion</th>
<th>Trunk Flexion (pelvis)</th>
<th>Trunk Flexion (world)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>r</td>
<td>p</td>
<td>r</td>
<td>p</td>
</tr>
<tr>
<td><strong>Initial Contact</strong></td>
<td>0.25</td>
<td>0.09</td>
<td>-0.03</td>
<td>0.44</td>
</tr>
<tr>
<td><strong>Peak ATSF</strong></td>
<td>-0.37</td>
<td>0.02</td>
<td>*</td>
<td>-0.2</td>
</tr>
</tbody>
</table>
Table 10: Between Subject Regression Table

<table>
<thead>
<tr>
<th>Joint Angle</th>
<th>R²</th>
<th>Adj R²</th>
<th>p</th>
<th>Predictor Variables</th>
</tr>
</thead>
<tbody>
<tr>
<td>Knee Flexion</td>
<td>0.14</td>
<td>0.11</td>
<td>0.04</td>
<td>*</td>
</tr>
<tr>
<td>Hip Flexion</td>
<td>0.00</td>
<td>-0.07</td>
<td>0.99</td>
<td></td>
</tr>
<tr>
<td>Trunk Flexion (pelvis)</td>
<td>0.01</td>
<td>-0.06</td>
<td>0.85</td>
<td></td>
</tr>
<tr>
<td>Trunk Flexion (world)</td>
<td>0.10</td>
<td>0.04</td>
<td>0.23</td>
<td></td>
</tr>
</tbody>
</table>
Appendix B: Figures
Between Subjects Plot of Knee Flexion Angle at Peak ATSF

\[ y = -0.012x + 0.908 \]

\[ R^2 = 0.135 \]
ABSTRACT

Hollie Walusz: The relationship between knee flexion, hip flexion, and trunk flexion angles and anterior tibial shear force during a jump-landing task
(Under the direction of Darin Padua)

Objective: To evaluate the relationships between sagittal plane knee, hip, and trunk angles and peak anterior tibial shear force (ATSF) during a jump-landing task.

Subjects: Thirty-one healthy female recreational athletes with no prior history of ACL injury volunteered to participate in this study.

Measurements: Knee, hip, and trunk kinematic data and landing forces were sampled during a jump-landing task. Both within and between subject data were analyzed using correlation and multiple regression statistical analyses.

Results: Knee flexion angle at peak ATSF was found to be significantly correlated with peak ATSF (p=0.021, r = -0.367). Additionally, trunk flexion relative to the vertical world axis at the time of peak ATSF was found to have a significant positive relationship with peak ATSF (p=0.046, r =0.309). Knee flexion angle at peak ATSF was found to predict approximately 13.5% of the variance in peak ATSF (p=0.042, R²=0.135). Multiple within subject correlations and regressions were found, however they varied amongst all angles between subjects.

Conclusion: Knee flexion angles appear to predict ATSF in female athletes. While hip and trunk flexion angles did not predict ATSF in this study, training should emphasize increased flexion at all three joints in order to decrease ACL injury. Sagittal plane joint angles did not explain all the variability in peak ATSF, therefore other factors need to be examined.
Introduction

Non-contact injuries to the anterior cruciate ligament (ACL) are devastating for athletes. Researchers have estimated that 80,000 to 100,000 ACL injuries occur annually, and up to 80% are non-contact in nature. (Griffin, Agel et al. 2000; Agel, Arendt et al. 2005). Additionally, research supports that over 70% of all ACL injuries are sports-related and involve execution of jump-landing, sudden deceleration, pivoting or cutting movements (Agel, Arendt et al. 2005; McLean, Huang et al. 2005; Colby, Francisco et al. 2000; Chappell, Yu et al. 2002). Epidemiological research has demonstrated that female athletes are two to eight times more likely to sustain an ACL injury than their male counterparts (Arendt and Dick 1995; Huston, Greenfield et al. 2000; Fagenbaum and Darling 2003).

Faulty joint biomechanics associated with jump-landing tasks could be a possible mechanism for ACL rupture. Landing from a jump has often been reported as a cause of non-contact ACL injuries in competitive athletics (Yu, McClure et al. 2005; Yu, Lin et al. 2006). Jump landings are a common and essential task in sports, especially female sports such as basketball, soccer, and volleyball. Research demonstrates that female athletes land in a more erect sagittal plane posture, with less knee flexion and hip flexion than male athletes (Huston, Greenfield et al. 2000; Malinzak, Colby et al. 2001; Chappell, Yu et al. 2002; Lephart, Ferris et al. 2002; Decker, Torry et al. 2003; Yu, Lin et al. 2006). Based on these findings, researchers speculate that females tend to land with a more stiff posture than males. This erect posture potentially predisposes the ACL to greater loads and injury.

Previous research has suggested that an increased anterior tibial shear force (ATSF) at the knee may correspond with an increase in ACL strain. Additionally, research has shown that ATSF are high during jump-landing tasks (Fleming, Renstrom et al. 2001; Chappell, Yu et al.
It has also been observed that females exhibit significantly greater proximal ATSF than males during the landing phase of a jump task (Chappell, Yu et al. 2002; Chappell, Herman et al. 2005; Yu, McClure et al. 2005). We believe that females may experience greater ATSF compared to males due to differences in sagittal plane movement patterns during a landing task.

Griffin et al. (2000) reported gender differences in motion patterns, positions, and forces generated from the hip and trunk to the knee. They also reported females performed cutting and landing maneuvers in a more erect posture than males, displaying less hip and knee flexion. These results were also confirmed by Yu et al. (2006), whose study found that in addition to differences commonly found in sagittal plane knee angles, female subjects landed with smaller hip flexion angles than males. These findings in females are important, as current research also suggests that the ACL is most vulnerable to injury at or immediately following initial contact with the ground when coupled with an awkward body position during landing tasks (Lephart, Abt et al. 2002). Knee flexion angles between fifty degrees and full extension have been shown to increase strain on the ACL (Beynnon and Fleming 1998). More specifically, extended knee positions place the ACL in a stretched and more vulnerable position for injury (Arendt and Dick 1995; Malinzak, Colby et al. 2001). If females are landing in a more erect sagittal posture with extended lower extremity positions, they may be at a greater risk of injuring their ACL.

Limited research has investigated the relationship between sagittal plane kinematics and ATSF in females. Therefore, the purpose of this investigation was to determine if a relationship exists between sagittal plane kinematics of the trunk, hip, and knee, and ATSF.
during a jump-landing task. A secondary purpose was to determine if any linear combination of knee, hip, and/or trunk flexion angle during a jump-landing task predicts ATSF.

Methods

Subjects

Thirty-one female subjects (age = 19.7 ± 2.00 years, height = 165.23 ± 6.23 cm, weight = 61.7 ± 9.69 kg) were recruited from the University of North Carolina at Chapel Hill student population to participate in the study. To be eligible for participation, all subjects were required to be between the ages of 18-25 years. Subjects were verbally recruited from physical education classes as well as from flyers posted on campus. Exclusion criteria included any history of ACL injury, knee surgery in the past two years, and current lower extremity injury that limited participation in physical activity. Due to previously noted gender differences in jump-landing technique, and the increased risk of ACL injury found in female athletes, this study focused on female recreational athletes. Female recreational athletes were defined as those participating in sports activity 2-3 times per week for an average of 30 minutes per session. Each subject was required to have 2 years of varsity, club, or intramural experience in either basketball, volleyball, or soccer, as these sports implement a jump-landing task. In addition, to avoid possible fatigue, no testing took place within one hour of physical activity or any other strenuous activity.

Instrumentation

An electromagnetic tracking system, Flock of Birds®, (Ascension Technologies, Inc., Burlington, VT), and Motion Monitor Software® (Innovative Sports Training, Inc. Chicago, IL) were used to collect three dimensional lower extremity kinematic data at a sampling rate of
144Hz. An embedded right-hand Cartesian coordinate system was defined for the shank, thigh, hip, and trunk segments to describe the three-dimensional positions and orientations of these segments. The reference system axes were defined so that the x-axis was positive in the anterior direction, the positive y-axis was directed to the left of the individual, and the positive z-axis was designated superiorly to the subject. This system was interfused with a non-conductive force plate (Bertec Corp., Columbus, OH) which sampled ground reaction forces at 1,440 Hz. The same Cartesian axis system was used to define motion analysis and force plate environments.

**Procedure**

Setting

Testing for each subject was performed during a single testing session lasting approximately 60 minutes. Prior to testing, all subjects read and signed an informed consent form.

Subject Preparation

Each subject completed a health questionnaire form, and his/her anthropomorphic data were collected. The dominant leg of each subject was used as the test limb, defined as the leg chosen by the subject to kick a ball for maximum distance. Each subject was instructed to wear athletic shoes, athletic shorts or spandex, and a t-shirt during the testing session. All subjects were allotted five minutes for warm-up on a stationary bike at moderate intensity. Following proper warm-up, the examiner demonstrated the jump-landing task, and the subjects were allowed to practice this task a maximum of five times.
Electromagnetic Marker Placement

The final step in preparing the subject for data collection included placement and set-up of electromagnetic sensors and digitization of select bony landmarks to enable kinematic analyses. A total of four electromagnetic sensors were used and were placed on the trunk, sacrum, thigh, and shank using doubled-sided tape, pre-wrap, and athletic tape. The sacrum marker was placed at the midpoint of the sacrum between the PSIS. The thigh marker was placed on the flattest portion of the distal one-third of the iliotibial band in order to minimize movement due to muscle contraction. The marker for the shank was placed just distal to the pes anserine insertion on the flat portion of the medial side of the proximal tibia. The trunk marker was placed on the C7 vertebrae.

With the markers in place, each joint was digitized with an electromagnetic sensor. Subjects were asked to stand in a neutral position on the force plate while the following bony landmarks were digitized in order to define segment endpoints and joint centers: T12 spinous process, xyphoid process, medial femoral condyle, lateral femoral condyle, medial malleolus, and lateral malleolus. Hip joint center was calculated via the method recommended by Bell, from left ASIS to right ASIS (Bell, Pedersen et al. 1990).

Data Collection Procedure

The jump-landing task required the athlete to perform a jump off a 30-cm high platform from a horizontal distance equal to 50% of their height, and land with the dominant foot on a force plate. Subjects were then instructed to immediately jump vertically for maximum height. Each subject performed 10 jump-landings with 30 seconds of rest between trials to minimize the risk of fatigue. Trials in which the subject failed to land with the foot of the dominant leg on the force plate were removed, and a new trial was performed.
Data Analysis
Data Processing

Lower extremity sagittal plane angles were recorded at initial contact during the jump-landing task. Initial contact was defined as the time when the vertical ground reaction force exceeded 10 Newtons. ATSF was calculated via a standard inverse dynamics procedure and standardized to body weight. ATSF was defined as the peak value occurring in the time period from initial contact (IC) to maximum knee flexion angle, referred to as the landing phase of a jump-landing task. This ensured that the peak ATSF value occurred when the subject was decelerating her body from the jumping task. Kinematic data were low pass filtered using a using a 4th order zero phase lag Butterworth filter at 14.5 Hz. (Yu, Gabriel et al. 1999). Euler angles were used to calculate the sagittal plane knee, hip, and trunk angles with a Y,X,Z order of rotation. Sagittal plane angles were calculated at the knee joint between the shank and thigh, at the hip joint between the thigh and pelvis, at the trunk joint between the pelvis and thoracic spine and at the trunk joint relative to the world vertical axis. The Flock of Birds was integrated with a force plate, and ATSF was calculated through the use of inverse dynamics.

Statistical Analysis

All data were analyzed using SPSS 14.0 statistical software (Chicago, IL). An a priori alpha-level of 0.05 was established. The average of the 10 trials for each of the four variables was calculated in order to examine the correlation of the variables across all subjects. We also performed a forward stepwise multiple regression for the averages of the joint flexion angle in order to determine any linear combination of the data across all subjects that might predict ATSF. After the between subject analyses were performed, within subject
Pearson correlation analyses were completed to determine if a relationship exists between ATSF and knee, hip, or trunk sagittal plane joint angles for all trials. This was completed to determine if the factors that influenced peak ATSF varied from individual to individual. Following the correlation analyses, within subject forward step-wise multiple regression analyses were performed to determine if trunk flexion, hip flexion, and/or knee flexion angles at IC significantly predicted ATSF during a jump-landing task.

Results

Inter-Subject Analyses

Thirty-three subjects (age = 19.7 ± 2.00 years, height =165.23 ± 6.233 cm., weight = 61.7 ± 9.69 kg.) were tested; however the data for two subjects were not included for the analysis due to hardware problems with the force plate that occurred in all 10 trials. Additionally, four subjects had a minimal number of trials affected, but their data from unaffected trails was still usable. Subjects 9 and 19 had nine trials used for data analysis, subject 12 had seven trials, and subject 13 had eight trials used for data analysis. No significant correlations were found for knee flexion, hip flexion, trunk flexion, and trunk flexion world angles at IC with peak ATSF (p>0.05). Additionally, a combination of knee flexion, hip flexion, trunk flexion, and trunk flexion world at initial contact was not a significant predictor of peak ATSF. These findings suggest that knee, hip and trunk flexion angles at the time of initial
contact do not have a significant influence on peak ATSF values during a jump-landings task in female recreational athletes.

Knee flexion angle at the time of peak anterior tibial shear force was found to have a significant negative relationship with peak ATSF (p=.021, r = -.367). This finding indicates that greater knee flexion resulted in smaller peak ATSF values. Additionally, trunk flexion world at time of peak ATSF was found to have a significant positive relationship with peak ATSF (p=.046, r=.309). There were no significant correlations found for hip flexion angle and trunk flexion angle relative to the pelvis at time of peak ATSF and peak ATSF (p>0.05). From the regression analysis, only knee flexion angle at the time of peak ATSF was found to significantly explain 13.5% of the variance in peak ATSF (p=.042, $r^2=.135$).

*Intra-Subject Analyses*

*Initial Contact Angles*

Of the 31 female subjects analyzed, there were a few correlations that were significant between ATSF and each dependent variable (Table 5). Four significant within subject correlations were found between knee flexion angle at IC and peak ATSF, two significant within subject correlations were found between hip flexion angle at IC and peak ATSF, three significant within subject correlations each were found between trunk flexion angle at IC and peak ATSF as well as trunk flexion angle relative to the world at IC and peak ATSF. Five significant regressions were found including 3 with knee flexion, 1 with hip flexion, and 1 with trunk flexion relative to the world. The significant p-values, r-values, and $r^2$-values may be found in Table 5.
ATSF Angles

Of the 31 female subjects tested, 4 significant correlations were found between the knee flexion angle at the time of peak ATSF and peak ATSF. Five significant correlations were found between the hip flexion angle at the time of peak ATSF and peak ATSF. Three significant correlations each were found between the trunk flexion angle at the time of peak ATSF and peak ATSF as well as between the trunk flexion relative to the world angle at the time of peak ATSF and peak ATSF. Ten total significant regressions were found across 9 subjects including 2 with knee flexion angle at the time of peak ATSF, 4 with hip flexion angle at the time of peak ATSF, 1 with trunk flexion angle at the time of peak ATSF, and 3 with trunk flexion angle relative to the world at the time of peak ATSF. The significant p-values, r-values, and R²-values may be found in Table 6.

Discussion

The purpose of this study was to determine the relationships between sagittal plane knee, hip and trunk angles with peak ATSF during a jump-landing task. The relationships were assessed using both inter-subject and intra-subject analyses. The intra-subject analyses involved investigating the relationships between sagittal plane joint angles and peak ATSF for each trial across 10 jump-landing trials for each subject. The inter-subject analyses involved investigating the relationships between the average sagittal plane joint angles and average peak ATSF during the 10 jump-landing trials across all subjects. The inter-subject correlation analyses revealed that only knee flexion (r = -0.37) and trunk flexion relative to the vertical world axis (r = 0.31) angles at the time of peak ATSF were significantly related to peak ATSF. The direction of the correlation indicates that less knee and more trunk
(world) flexion resulted in larger peak ATSF values during a jump-landing task. While the relationship between knee flexion angles and ATSF were predicted, the positive correlation between trunk flexion relative to the world vertical axis and peak ATSF was unexpected. The fact that trunk flexion relative to the pelvis segment was not significant ($r = -0.09, p = 0.31$) is also unexpected. These results may be due to the way trunk flexion relative to the world vertical axis was defined. Trunk flexion relative to the world was measured as the angle between the longitudinal axis of the thorax segment and the vertical axis of the world axis system. Trunk flexion was also defined as the angle between the longitudinal axis of the thorax relative to the vertical axis of the pelvis at the T12-L1 joint. Due to the difference in measurements, the trunk flexion angle relative to the world may be influenced by other body motions (hip and lumbar spine flexion). For instance, an individual who exhibits a large trunk flexion angle relative to the world axis may purely exhibit trunk flexion of the thorax at the T12-L1 joint, or they may be in a neutral trunk position but be in a flexed position at the hip. Trunk flexion relative to the pelvis should not be influenced by these other movements. Thus, the finding of a positive correlation between trunk flexion relative to the world and ATSF may not reflect true trunk flexion.

The positive correlation between trunk flexion relative to the world may also be a response to the relative contributions of the knee and hip to energy absorption during a landing task. Low knee flexion angles have already been found to increase peak ATSF, but it may also cause an increase in hip and trunk flexion angles in order to absorb the forces transmitted up the kinetic chain. If the subject were to land in an extended knee position, the hip and trunk may flex more in order to compensate and absorb the forces. Thus, while we found that
increased trunk flexion relative to the world is associated with increased ATSF, it may be a product of the decreased knee flexion angle also found in this study.

Multiple regression analyses revealed that knee flexion angle at the time of peak ATSF was the only variable to significantly predict peak ATSF (adjusted $R^2 = 0.11$) as no other variables explained additional variability in peak ATSF and entered into the regression model. Thus, knee flexion angle at the time of peak ATSF appears to have the most prominent influence on peak ATSF values during a jump-landing task.

The intra-subject analyses revealed that only 10 of 31 subjects (32%) demonstrated significant relationships between either sagittal plane joint angles at initial contact or at the time of peak ATSF with peak ATSF across the 10 jump-landing trials. The adjusted $R^2$ values in these 9 subjects ranged from 0.34 to 0.85, suggesting that in these individuals sagittal plane joint position had a moderate to large effect on peak ATSF during the 10 jump-landing trials. Knee flexion at initial contact and hip and trunk (world) flexion at the time of peak ATSF appear to be the most important factors that influenced peak ATSF as these were the variables that were most commonly found to be significantly related to peak ATSF. The intra-subject analyses indicates that the influence of sagittal plane joint angles on peak ATSF varies across subjects as not all subjects demonstrated significant relationships and the specific sagittal plane joint angles that were significantly related to peak ATSF was not consistent across subjects. Thus, the relationships between peak ATSF and sagittal plane joint angle differ within subjects and cannot be generalized across subjects.

Knee flexion angles and their affect on strain of the ACL have been examined frequently in previous literature. Multiple studies found that when the knee angle is between 50 degrees and full extension, the ACL is placed under strain (Markolf, Burchfield et al. 1995; Beynnon
and Fleming 1998). Subsequent investigations have also demonstrated that the ACL is under increasing strain as the knee moves into full extension, and that ATSF increases as the knee moves close to full extension (Osternig, Ferber et al. 2001). Fagenbaum and Darling found that ACL forces progressively decrease as the knee is flexed to 15, 30, and 60 degrees (Fagenbaum and Darling 2003). Colby et al. found the average amount of knee flexion angle at foot strike to be 23 degrees when landing from a jump and that the average angle of knee flexion at the time of ACL injury during a jump is 21 degrees (Colby, Francisco et al. 2000).

Knee flexion angle at IC resulted in only 4 significant correlations with peak ATSF. However, the findings corresponded to those presented in previous research. The average angle of knee flexion for subject 15 was 11.15 degrees with a corresponding average of 483.13N of peak ATSF. Subject 11 had a knee flexion angle of 19.95 degrees at IC with an average peak ATSF of 178.82N. Subject 14 had 22.22 degrees of knee flexion at IC with a corresponding average peak ATSF of 533.33 N. Only subject 25 had a flexion angle of greater than 25 degrees, with an average of 27.27 degrees. The corresponding average peak ATSF was 34.72N. It is also important to notice the average ATSF values were much higher for the three subjects with angles under 20 degrees of knee flexion than the average ATSF found when the knee flexion angle was over 25 degrees. The 3 subjects who demonstrated a significant correlation between knee flexion angle at IC and ATSF had a knee flexion angle of less than 25 degrees, while the fourth had a knee flexion angle of under 30 degrees, which has been stated as a dangerous zone in previous research to increase risk of an ACL injury.

Similar results were found when examining the knee flexion angles at the time of peak ATSF. Knee flexion angle at the time of peak ATSF resulted in only 4 significant correlations and explained only 13.5% of the variance in ATSF. This indicates that knee
kinematics alone do not adequately explain the presence of ATSF in most subjects. Subject 8 had a knee flexion angle of 22.37 degrees at the time of peak ATSF with an associated ATSF of 500.95N. Subject 33 had an average knee flexion angle of 23.52 degrees to go along with an average of 240.65 N of peak ATSF. Subject 16 had an average of 25.65 degrees of knee flexion at the time of peak ATSF to go along with 297.94 N of peak ATSF. The fourth significant correlation was found in subject 25, with a flexion angle average of 43.59 degrees and a corresponding average of 34.72 N of peak ATSF. These results show the trend of decreasing ATSF when the knee flexion angle increases. It is interesting to note the knee flexion angles at the time of peak ATSF as well, as 3 of the 4 significant correlations found had subjects with a knee flexion angle less than 25 degrees even past the IC point.

This data may indicate that knee flexion only has a strong relationship to ATSF at low flexion angles. In literature, these low flexion angles are associated with an increased risk of ACL injury due to more risky landing strategies. Thus, overall, for some individuals who exhibit knee flexion angles below 25 degrees, ATSF may be higher and create the increased risk of injury.

Hip and trunk position and motion may have an influence on knee positions and knee loads because of the kinetic chain relationship between the lower extremity and force transmission up and down this chain. Previous literature has shown that females perform cutting and landing maneuvers in a more erect posture than males, with less hip flexion and less knee flexion (Huston, Vibert et al. 2001; Malinzak, Colby et al. 2001; Decker, Torry et al. 2003). Additionally, Lephart et al. found that quadriceps activity and therefore possibly ATSF, have been reported to decrease when hip flexion angles are increased. However again, females tend to land with smaller angles of hip flexion compared with males (Lephart,
Abt et al. (2002). This combined effect suggests that smaller hip flexion angles may very well affect ACL strain values and predispose female athletes to ACL injury. However, not enough research has been completed in this area to understand the effect of hip angle during a jump-landing task, thus the inclusion of this variable in our study.

Hip flexion angle at initial contact did not seem to be an important variable when examining any between or within-subject relationship with peak ATSF. Two significant correlations were found in the within subject IC data. Subject 4 was found to have a significant correlation, displaying an average of 47.84 degrees of hip flexion corresponding with an average of 329.15N of ATSF while subject 11 had an average of 26.89 degrees of flexion with a corresponding average of 178.82N of ATSF. When examining the hip flexion angles at the time of peak ATSF, the broad range of degrees of flexion with the correlated average peak ATSF also suggested that hip flexion angle may not be an important contributor to the peak ATSF during a jump-landing task as no consistent trends could be found. Further research should be completed as to the role the hip joint may have on ACL injury.

Trunk flexion angle and its role in ACL injury and amount of ATSF generated have not yet been extensively examined in previous research to date. Thus, our study included this variable in the lower extremity posture examination to determine if any such relationship exists. Kulas et al. compared the trunk relative to the pelvis, through a “joint” modeled at T12-L1, and the hip joint in order to determine whether sex differences in landing posture arise from the pelvis-femoral (hip joint) articulation or from the trunk relative to the pelvis. They concluded that females demonstrated greater trunk extension relative to the pelvis than males while there were no sex differences in hip flexion angle during landing. These results provide validity to the common assumption that females land with a more erect trunk posture.
than males and also provides the rationale for modeling the trunk and pelvis as two separate segments in order to differentiate between trunk-pelvis versus hip joint motion (Kulas AS and Henning JM 2007).

We included both the trunk flexion angle relative to the pelvis as well as the trunk flexion angle relative to the world, in addition to the hip joint motion, due to the uncertainty in current literature as to which variable is the best predictor of actual trunk motion. It has been thought that the trunk flexion relative to the pelvis is a more biomechanically correct assessor, while the trunk flexion relative to the world is one that can been seen more easily clinically. It is interesting to note that the values of the trunk flexion angle relative to pelvis and trunk flexion angle relative to the world data in this study were different, with no two numbers being the same. The values found with the trunk flexion angles relative to the world were almost always higher than the values of trunk flexion relative to pelvis in each subject. More research should be completed as to which of these angles is more accurate to the data of interest.

In the inter-subject data, the trunk flexion relative to the world angle at the time of peak ATSF was correlated significantly with peak ATSF. Trunk flexion angle at IC found three correlations for each of the types of trunk angle measurements. Two of the subjects had significant correlations for both trunk flexion and trunk flexion relative to the world and peak ATSF. Subject 9 had a trunk flexion angle of -4.12 degrees, trunk flexion relative to the world angle of 10.11 degrees and a peak ATSF of 421.97 N. Subject 22 was similar as she had 13.91 degrees of trunk flexion, 21.58 degrees of trunk flexion relative to the world, and a peak ATSF of 190.15 N. Subject 2 had 17.82 degrees of trunk flexion relative to the world
with a peak ATSF of 426.03N while subject 11 had 19.71 degrees of trunk flexion with a corresponding peak ATSF of 178.82 N.

When the trunk flexion angles at the time of peak ATSF were examined, 3 significant correlations were found for both types of trunk flexion angles as well. Subject 9 had correlations for both types of flexion angle tested, with -3.86 degrees of trunk flexion relative to the pelvis and 10.64 degrees of trunk flexion relative to the world. The correlating average peak ATSF was 421.97 N. Subject 2 had an average of 16.66 degrees of trunk flexion relative to the world with a corresponding average 426.03 N of peak ATSF. Subject 13 had an average trunk flexion relative to the world angle of 8.45 degrees with a corresponding average peak ATSF of 241.24 N. Subject 11 had 18.86 degrees of trunk flexion relative to the pelvis that was correlated with 178.82N of peak ATSF. Finally, subject 23 had 16.54 degrees of trunk flexion relative to the pelvis that was correlated with only an average of 16.78N of peak ATSF.

When looking at the trunk flexion relative to the pelvis values, the lowest flexion angles corresponded to the highest peak ATSF values, while the higher flexion angles resulted in a decrease in peak ATSF. These numbers show a trend of decreasing peak ATSF when the trunk flexion angle increases, and may give more evidence to the biomechanical accuracy of this trunk measurement. However, the values seen between the trunk flexion angles relative to the world and peak ATSF showed no consistent trends. Further studies should be completed to find supporting results.

Another note of interest is the mean flexion angle at the time of peak ATSF compared to the mean flexion angle of each joint at IC. Both the knee and hip flexion angles increased in degree from IC to the time of peak ATSF as would be expected, whereas both trunk flexion
angle relative to the world and trunk flexion angle relative to the pelvis decreased in degrees from IC to the time of peak ATSF. Also, the influence of IC angles versus peak ATSF angles on peak ATSF values was an interesting finding of the study. The between subject analyses suggest that peak ATSF values are more sensitive to changes in sagittal plane joint angles at the time of peak ATSF than at IC. There were no significant correlation or regression analyses for joint angles at initial contact, but there were for joint angles at time of peak ATSF.

Limitations

A potential limitation of this study is that we only examined the flexion angles in association with a jump-landing task. The relationships between sagittal plane angles with peak ATSF may differ depending upon the nature of the task, so future research involving other non-contact athletic movements may be helpful in determining if a risk factor exists. Another limitation may be that EMG was not completed during this study. It is possible that sagittal plane musculature, such as the gluteus maximus, hamstrings, quadriceps, and gastrocnemius muscles, is related to peak ATSF during a jump-landing task and determining if a kinetic relationship with peak ATSF exists may help decrease ACL injury risk.

Implications on ACL Injury Prevention

From these results, it may be concluded that the lower extremity joint angles during a jump-landing task at both IC and at the time of peak ATSF may not be the most important factors to examine for ACL injury prevention. This may have implications on ACL prevention training. The ultimate goal from this study as well as from athletic training
clinicians in general is to reduce ACL injury risk in the athletic population. If this is to be done by decreasing ATSF, than the results of this study suggest that we need to work on getting athletes to flex their knees, hip, and trunk to a greater extent. The knee is an important factor, however, just focusing on the knee will not necessarily work for all subjects, so we also need to train individuals to focus on hip and trunk flexion as well in order to decrease the risk for ACL injury. Also, since sagittal plane joint angles did not explain all the variability in peak ATSF, it suggests that there are other factors that are related to peak ATSF values. Some of these factors may possibly be EMG activity of the quadriceps and hamstring muscles during a jump-landing task, the quad/hamstring ratio, the ground reaction force when landing, and possibly even the moments and velocities associated with the lower extremity joints during a jump-landing task. Future research needs to be completed in order to investigate these factors so that we have a better understanding of all factors that may influence peak ATSF.

Conclusions

The current study examined the relationship between knee flexion, hip flexion, and trunk flexion angles and peak ATSF at both initial contact and the time of peak ATSF. Based on the results of this study, we conclude the following:

1. Knee flexion angle at the time of peak ATSF appears to have the most prominent influence on peak ATSF values during a jump-landing task for between subject data, with the direction of the correlation indicating that less knee and trunk (world) flexion results in larger peak ATSF values.
2. Knee flexion at IC and hip and trunk (world) flexion at the time of peak ATSF appear to be the most important factors that influenced peak ATSF for within subject data, however the influence of sagittal plane joint angles on peak ATSF is not consistent across subjects, therefore the relationships between peak ATSF and sagittal plane joint angles cannot be generalized across subjects.

3. The knee is an important factor, however just focusing on the knee will not necessarily work for all subjects, so we also need to train individuals to focus on hip and trunk flexion as well in order to decrease their risk for ACL injury.

4. Sagittal plane joint angles did not explain all the variability in peak ATSF, therefore there are other factors that are related to peak ATSF values that need to be examined.
Appendix D: Informed Consent Form
University of North Carolina-Chapel Hill
Consent to Participate in a Research Study
Adult Participants Female Recreational Athletes ages 18-25
Social Behavioral Form

IRB Study # 06-0543

Title of Study: The Relationship between Anterior Tibial Shear Force and Quadriceps/Hamstring Strength, Knee Flexion Angle, Hip Flexion Angle, and Trunk Flexion Angle during a Jump Landing Task

Principal Investigator: Douglas R. Bennett LAT, ATC
UNC-Chapel Hill Department: Exercise and Sport Science
UNC-Chapel Hill Phone number: 919-962-2067
Email Address: drb215@email.unc.edu
Co-Investigators: Hollie J. Walusz LAT, ATC; Darin Padua, PhD, ATC; Troy Blackburn, PhD, ATC; Michelle Boling, MS, ATC; Melanie McGrath, MS, ATC; Chris Hirth MS, PT, ATC
Faculty Advisor: Darin Padua, PhD, ATC
Funding Source:

Study Contact telephone number: 919-962-2067
Study Contact email: drb215@email.unc.edu

What are some general things you should know about research studies?

You are being asked to take part in a research study. To join the study is voluntary. You may refuse to join, or you may withdraw your consent to be in the study, for any reason.

Research studies are designed to obtain new knowledge that may help other people in the future. You may not receive any direct benefit from being in the research study. There also may be risks to being in research studies.

Deciding not to be in the study or leaving the study before it is done will not affect your relationship with the researcher, your health care provider, or the University of North Carolina-Chapel Hill. If you are a patient with an illness, you do not have to be in the research study in order to receive health care.

Details about this study are discussed below. It is important that you understand this information so that you can make an informed choice about being in this research study. You will be given a copy of this consent form. You should ask the researchers named above, or staff members who may assist them, any questions you have about this study at any time.
What is the purpose of this study?
The purpose of this study is to determine if a relationship exists between lower extremity motion and strength with forces at the knee during a jump landing task. The ACL provides as much as 86% of the static restraint to anterior tibial translation (ATT) on the femur. When anterior tibial shear forces are great enough, the tibia will translate anteriorly, and ACL rupture can occur. During the dynamic activities of sports such as jumping, cutting, and landing, these forces can greatly exceed the loading capacity of the ACL. Thus, there is a need for additional stability at the knee joint. This additional stability is derived via dynamic stabilizers (musculotendinous structures). Also, more erect sagittal plane postures have been found to increase strain on the ACL. Therefore, determining if such relationships exist may aid us in preventing ACL injury.

You are being asked to participate in the study because you are a female recreational athlete between the ages of 18 and 25 who may participate in sporting activities that involve a jump-landing.

Are there any reasons you should not be in this study?
You should not participate in this study if any of the following apply to you:

- You have a previous history of an anterior cruciate ligament (ACL) injury
- You have had ligamentous reconstruction or any knee surgery within the past two years
- You have had a current lower extremity injury that would affect your performance of a jumping task.

How many people will take part in this study?
If you decide to participate in this study, you will be one of approximately 40 females in this research study.

How long will your part in this study last?
All testing will be performed in one session lasting approximately 60 minutes

What will happen if you take part in the study?
During the course of this study, the following will occur:

- You will report to the Sports Medicine Research Laboratory in Fetzer Gymnasium for one 60-minute testing session. You will be asked to wear athletic shorts, a t-shirt, and your athletic shoes. You will then complete a health questionnaire, and data will be collected from you including your age(years), height(cm), mass(kg), and leg dominance prior to the start of the testing session. Next you will ride a stationary bicycle at moderate intensity for 5 minutes as part of a warm-up prior to the jumping activity. Demonstration of the jumping task will be shown to you and you will be able to practice prior to data collection.
• You will have electromagnetic motion-tracking sensors placed on your upper back, low back, thigh, and shin that are designed to measure the movement patterns of the lower extremity and trunk. A female examiner (Hollie Walusz) will perform all sensor placements. Once the electromagnetic sensors are attached, you will be asked to stand in a neutral position with your arms relaxed at your sides. We will then define the following bony landmarks for the motion tracking software: T12 spinous process (middle of back), xyphoid process (lower end of breast bone), medial femoral condyle (inside of knee), lateral femoral condyle (outside of knee), distal end of medial malleolus (inside of ankle), distal end of lateral malleolus (outside of ankle), left Anterior Superior Iliac Spine (ASIS) (hip bone at the top of your thigh) and right ASIS (hip bone at the top of your thigh).

• You will then be asked to perform a jump-landing task that involves jumping down from a 30-cm high box and landing with one foot on a force plate and the other off to the side on a carpeted surface. You will perform 10 trials with at least 30 seconds of rest between each trial.

• Next you will be set up on a dynamometer (measures force production) in order to test muscular strength. You will be positioned sitting upright and the female investigator (Hollie Walusz) will secure you using torso, hip, thigh, and shin stabilization straps. After an explanation of the strength testing task, you will be allowed to practice each task sub-maximally. You will then perform five maximal contractions of two different strength protocols, consisting of three different testing speeds, targeting your quadriceps and hamstring musculature.

What are the possible benefits from being in this study?
Research is designed to benefit society by gaining new knowledge. You will not benefit personally from being in this research study, however the benefits to society include gaining information that researchers can analyze to better understand how body movement and strength differences in females affect ACL injury. This may help to prevent ACL injury in the future.

What are the possible risks or discomforts involved from being in this study?
As with any physical activity, participation in this study carries the risk of injury. The motions that you will be asked to perform are performed regularly during sporting activities, therefore, you will be familiar with them and should be able to perform the tasks with minimal injury risk. Demonstration of jump-landing task will be shown to you prior to completing the task. Also, practice repetitions of both the jump landing task and strength testing procedures will be allowed for familiarization. In case of injury, medical personnel (certified athletic trainers) will be located in the same building as where the testing will take place, and ice will be available if needed. You will be free to cease participation at any time. In addition, there may be uncommon or previously unknown risks that might occur. You should report any problems to the researchers.

How will your privacy be protected?
No participants will be identified in any report or publication about this study. You will be assigned an identification number (ID) for data collection that will be matched to the
identifiers listed above in an excel document. This document will be stored on a separate CD apart from all other data that will be collected. These CDs will be stored in a locked cabinet with access only to members of the research team. Once all participants have completed the testing, identifiers will be deleted from the excel document. All data will be stored on CDs which will be kept in the Sports Medicine Research Laboratory. All data analysis will be performed on computers in the Sports Medicine Research Laboratory where a password is necessary for access to the computers. Only members performing research have access to these computers, therefore identification of any participants or data is very unlikely. If disclosure is ever required, UNC-CH will take all steps allowable by law to protect the privacy of personal information.

Personal privacy during testing sessions will be maintained by limiting the people within the research lab to current employees of the lab and the testers themselves. The only door to enter the lab is locked with key card access to ensure privacy. Although every effort will be made to keep research records private, there may be times when federal or state law requires the disclosure of such records, including personal information. This is very unlikely, but if disclosure is ever required, UNC-Chapel Hill will take steps allowable by law to protect the privacy of personal information. In some cases, your information in this research study could be reviewed by representatives of the University, research sponsors, or government agencies for purposes such as quality control or safety.

**What will happen if you are injured by this research?**
All research involves a chance that something bad might happen to you. This may include the risk of personal injury. In spite of all safety measures, you might develop a reaction or injury from being in this study. If such problems occur, the researchers will help you get medical care, but any costs for the medical care will be billed to you and/or your insurance company. The University of North Carolina at Chapel Hill has not set aside funds to pay you for any such reactions or injuries, or for the related medical care. However, by signing this form, you do not give up any of your legal rights. Certified Athletic Trainers will be present during all testing sessions in the rare possibility that an injury occurs.

**Will you receive anything for being in this study?**
You will not receive anything for taking part in this study.

**Will it cost you anything to be in this study?**
It will not cost you anything to be in this study. Each participant is only responsible for her own transportation to the Sports Medicine Research Laboratory for their one-hour testing session.

**What if you are a UNC student?**
You may choose not to be in the study or to stop being in the study before it is over at any time. This will not affect your class standing or grades at UNC-Chapel Hill. You will not be offered or receive any special consideration if you take part in this research.

**What if you are a UNC employee?**
Taking part in this research is not a part of your University duties, and refusing will not affect
your job. You will not be offered or receive any special job-related consideration if you take part in this research.

**What if you have questions about this study?**
You have the right to ask, and have answered, any questions you may have about this research. If you have questions, or if a research-related injury occurs, you should contact the researchers listed on the first page of this form.

**What if you have questions about your rights as a research participant?**
All research on human volunteers is reviewed by a committee that works to protect your rights and welfare. If you have questions or concerns about your rights as a research subject you may contact, anonymously if you wish, the Institutional Review Board at 919-966-3113 or by email to IRB_subjects@unc.edu.

---

**Title of Study:** The Relationship between Anterior Tibial Shear Force and Quadriceps/Hamstring Strength, Knee Flexion Angle, Hip Flexion Angle, and Trunk Flexion Angle during a Jump Landing Task

**Study # :** 06-0543

**Participant’s Agreement:**

I have read the information provided above. I have asked all the questions I have at this time. I voluntarily agree to participate in this research study.

_________________________   _____________________
Signature of Research Participant     Date

_________________________
Printed Name of Research Participant

_________________________   _____________________
Signature of Person Obtaining Consent     Date

_________________________
Printed Name of Person Obtaining Consent
Title of Study: The Relationship Between Anterior Tibial Shear Force and Quadriceps/Hamstring Strength, Knee Flexion Angle, Hip Flexion Angle, and Trunk Flexion Angle during a Jump Landing Task
Principal Investigator: Douglas R. Bennett, LAT, ATC
UNC-CH Department: EXSS
Phone Number: 919-962-7187
Co-Investigators: Hollie J. Walusz LAT, ATC; Darin Padua, PhD, ATC; Troy Blackburn, PhD, ATC; Michelle Boling, MS, ATC; Melanie McGrath, MS, ATC; Chris Hirth MS, PT, ATC
Sponsor: None

Name_________________________   Age__________________________
Height (cm) ____________________   Weight (kg)____________________

1. Are you currently in good general health?
   YES / NO

2. Do you currently have a lower extremity injury that has required days missed from physical activity?
   YES / NO

3. Do you have a prior history of ACL injury, ligamentous reconstruction, or any knee surgery within the past two years?
   YES / NO

4. Do you have any current symptoms of injury?
   YES / NO

5. How often do you exercise per week? ________________ Days

6. Approximately how many minutes do you exercise on those days? ______ Minutes

7. What type of exercise activity do you most often participate in (soccer, volleyball, basketball, etc.)?
VOLUNTEERS NEEDED FOR RESEARCH STUDY

Female volunteers who participate in recreational sporting activity are needed to participate in a research study.

**You should not volunteer in the study if you have...**

* prior history of knee surgery within the past 2 years
* prior history of ACL injury
* presence of other lower extremity injury

**If you volunteer for this study, you will...**

* report to the Sports Medicine Research Laboratory in Fetzer Gymnasium for one testing session lasting approximately 60 minutes
* perform 10 trials of a jump landing task while motion analysis data is collected
* perform 4 strength tests (concentric knee extension, eccentric knee extension, concentric knee flexion, eccentric knee flexion)
* participate in research that may help prevent ACL injury!!

Contact Doug Bennett or Hollie Walusz if you are interested in volunteering for this study.

Doug Bennett, LAT, ATC
Phone number: 814-244-2803
Email: drb215@email.unc.edu

Hollie Walusz, LAT, ATC
Phone number: 319-830-0796
Email: walusz@email.unc.edu
References


