Old Dominion University

ODU Digital Commons

Human Movement Sciences & Special Education Theses & Dissertations

Human Movement Sciences and Special Education

Spring 2009

The Effects of Isolated Hamstring Fatigue on Jump Landing in Healthy Recreational Athletes

Kristen Anne Phillips
Old Dominion University

Follow this and additional works at: https://digitalcommons.odu.edu/hms_etds

Part of the Exercise Science Commons, and the Sports Sciences Commons

Recommended Citation

Phillips, Kristen A.. "The Effects of Isolated Hamstring Fatigue on Jump Landing in Healthy Recreational Athletes" (2009). Master of Science in Education (MSEd), Thesis, Human Movement Sciences, Old Dominion University, DOI: 10.25777/1x4v-0b47
https://digitalcommons.odu.edu/hms_etds/86

This Thesis is brought to you for free and open access by the Human Movement Sciences and Special Education at ODU Digital Commons. It has been accepted for inclusion in Human Movement Sciences & Special Education Theses & Dissertations by an authorized administrator of ODU Digital Commons. For more information, please contact digitalcommons@odu.edu.

THE EFFECTS OF ISOLATED HAMSTRING FATIGUE ON JUMP LANDING IN HEATHLY RECREATIONAL ATHLETES

By

Kristen Anne Phillips, ATC
B.A. May 2007, Catawba College

A Thesis Submitted to the faculty of
Old Dominion University in Partial Fulfillment of the
Requirement for the Degree of

MASTER OF SCIENCE IN EDUCATION

ATHLETIC TRAINING

OLD DOMINION UNIVERSITY
May 2009

Approved by:

James Oñate (Director)

Bonnie Van Lunen (member)

Stacie Ringleb (member)

ABSTRACT

THE EFFECTS OF ISOLATED HAMSTRING FATIGUE ON JUMP LANDING IN HEATHLY RECREATIONAL ATHLETES

Kristen Phillips
Old Dominion University, 2009
Director: Dr. James A. Oñate

The musculature of the lower extremity provides dynamic stability during daily and athletic tasks. In landing tasks, the lower extremity absorbs forces and reduces excessive stresses and motions that are placed upon the body. Neuromuscular fatigue has been linked to an increase in injury and altered kinematics during jump landing tasks. The purpose of this study was to analyze the effects of hamstring fatigue on knee and hip kinematics pre and post hamstrings fatigue while landing from a box drop of 30cm height, during the stop-jump phase. The study aimed to quantify the kinematic data of the hip and knee in order to examine the lower extremity motion pattern variations following hamstring fatigue. Fourteen healthy female recreational athletes (age=21.4±1.45 yrs; height=165.14±8.26 cm; mass=65.56±16.43 kg) with no history of surgery, no ACL injury or current injury involving their lower extremity were subjects in the current study. Eccentric hamstring fatigue, a 30% eccentric isokinetic knee-flexor torque reduction, was induced on the dominant leg using Primus RS (BTE Technologies Inc). Kinematic data were processed using VICON Motion system during the box drop task both pre and post fatigue. Hip and knee kinematics were measured pre and post fatigue for maximum values, and at time instances of initial contact and maximum knee flexion values during the stop-jump phase of the bilateral box drop. Statistic tests included a univariate analysis of variance with repeated measures to analyze the hip and knee kinematics. A significant level of < .05 was set a priori for all analyses. Upon initial contact of the stance phase, hip flexion significantly increased post-fatigue (43.78±8.21, CI=39.03;48.52) than pre-fatigue (40.68±6.82, CI=36.71;44.62). No other significant difference was found at either the hip or knee for maximum values or at initial or maximum knee flexion. Our findings revealed fatigue of the hamstrings caused a disruption in hip kinematics at initial contact. Although fatigue adversely affects the muscular function, altered kinematics of the lower extremity during tasks may be highly dependent upon the fatigue task chosen. Subjects may have placed greater reliance on unfatigued musculature in the lower extremity to accomplish the box drop task.

Co-Directors of Advisory Committee: Dr. Bonnie VanLunen
Dr. Stacie Ringleb

© Copyright, 2009, by Kristen Phillips, All Rights Reserved

ACKNOWLEDGEMENTS

I would like to thank first and foremost, my family and friends who have supported me through this journey. Thank you for providing words of wisdom and encouragement during times of need. Specifically to my mother and grandmother, thank you for exemplifying and instilling within me a true thirst for knowledge throughout my life. Grandma, you have created great footsteps to follow but as you say, "knowledge is power", so hopefully I can come close. To my mother, thank you for always encouraging me to do anything I set my mind to and to simply be me along the way. To my dad, who has taught me first hand actions speak louder than words and to enjoy life and family, thank you. And finally Drew, thank you for your undying support and laughs along the way.

I would like to thank my thesis advisor Dr. Jimmy Oñate. Thank you for giving me direction through my thesis but also instilling in me an inquisitive perspective on things around me. I don't believe I will ever read another article without raising questions, even if it involves People magazine. You have an uncanny ability to make students want to do better and learn more than face value, even if its just to be prepared for one of those "Jimmy" questions that we know you're going to ask in class. Thank you for pushing me to accomplish things I could handle, even when I doubted myself.

Also, thank you to Dr. Bonnie VanLunen, for getting me started on this whole thesis writing business. You gave me direction early on in the process and made me realize I could do it. You also demonstrate to me nearly every time I see you, that I can also one day have both a great career and a family. And thanks for always laughing at my jokes, especially the ones about Jimmy.

I must also give a huge thanks to Nelson Cortes. I honestly don't know where I would be in this process without your guidance, advice and support. You were there for me during this whole process, from data collection to my discussion section. No matter if it was via text, email or phone, you always made me feel as if my concerns and questions were a priority. Somehow you always seemed like you were a step ahead of me, setting deadlines and checking on my progress. Thank you for pushing me through this process to not only meet my goals, but to accomplish them well and be proud of my finished product.

I would like to thank everyone that played a part in this process. Thank you to Jena Etnoyer for your help in the data collection process. We couldn't have done it without each other, and who knew we could hear me say "pull" so many times! To all my subjects and pilot test subjects, thank you for allowing me to ensue pain in your hamstring for days to follow. To think when I first arrived at graduate school I really had no idea what was involved to writing a thesis. I could not have done it without everyone's help and support. As they say it takes a village to raise a child, I think the same holds true for a thesis. Thank you all so much.

TABLE OF CONTENTS

LIST OF FIGURES

LIST OF TABLES

CHAPTER I
INTRODUCTION

Injury to the anterior cruciate ligament (ACL) has become a common occurrence in athletics and recreational activities, especially to the female population (Arendt EA, Agel J, & R, 1999; Arendt EA & Dick R, 1995; Gabbett TJ, 2000; Lindenfeld TN, Schmitt DJ, Hendy MP, Mangine RE, & Noyes FR, 1994; Messina DF, Farney WC, & DeLee JC, 1999; Myklebust G, Maehlum S, Holm I, & Bahr R, 1998; Stevenson H, Webster J, Johnson RJ, & Beynnon B, 1998). The associated high expense of treatment, complications of further injury, joint osteoarthritis (Frank CB & Jackson DW, 1997; Lohmander LS, Ostenberg A, Englund M, & Roos H, 2004) and the high rate of incidence, have raised interest in the underlying risk factors of ACL injury (Griffin LY et al., 2000; Smith BA, Livesay GA, & SL, 1993).

Approximately three-fourths of ACL injuries occur through a non-contact decelerating mechanism, such as landing and cutting, and are often seen in basketball, soccer and volleyball athletes (Arendt EA et al., 1999; Arendt EA & Dick R, 1995; Boden, Dean, Feagin, & Garrett, 2000; Kirkendall & Garrett, 2000). During dynamic movements, the lower extremity musculature generates both preparatory and reactive movements to counterbalance external forces and internal moments that may cause injury to the ACL (Cerulli, Benoit, Lamontagne, Caraffa, & Liti, 2003; G. Li et al., 1999; McNair, Marshall, & Matheson, 1990). The co-contraction of the quadriceps and hamstrings provide dynamic stability to limit these stresses placed on the ACL (Imran A & O'Connor JJ, 1997; Lloyd & Buchanan, 2001; Markolf, Bargar, Shoemaker, & Amstutz, 1981; Markolf, Graff-Radford, & Amstutz, 1978; Markolf, Mensch, &

Amstutz, 1976; Solomonow et al., 1987; Torzilli, Deng, & Warren, 1994). In particular, the hamstrings are thought to play a direct synergistic role to the ACL, limiting anterior tibial translation and internal rotation, a common mechanism of ACL injury (Aune, Nordsletten, Skjeldal, Madsen, & Ekeland, 1995; Baker BE, 1991; Barratta R et al., 1988; Dureselen L, Claes L, & H, 1995; MacWilliams, Wilson, DesJardins, Romero, & Chao, 1999).

Due to its importance, dynamic control provided by the neuromuscular system has been a focus of investigation in an effort to understand underlying causes of ACL injury. Consequently, numerous studies have revealed gender differences in neuromuscular recruitment and activation strategies during athletic tasks and in response to external perturbations (Chappell, Creighton, Giuliani, Yu, & Garrett, 2007; Hewett, Stroupe, Nance, & Noyes, 1996; Huston & Wojtys, 1996; Malinzak, Colby, Kirkendall, Yu, & Garrett, 2001; Urabe et al., 2005; Zazulak et al., 2005). Females utilize a quadriceps dominant strategy with greater quadriceps activation and less activation of the hamstring musculature. Thus, having a direct influence and implication on landing mechanics and harmful positioning at the knee (Borotikar, Newcomer, Koppes, & McLean, 2008; Chappell et al., 2007; Chappell, Yu, Kirkendall, & Garrett, 2002; Cowley HR, Ford KR, Myer GD, Kernozek TW, & Hewett TE, 2006; Decker, Torry, Wyland, Sterett, & Richard Steadman, 2003; Ford, Myer, & Hewett, 2003; Lephart SM, Ferris CM, Riemann BL, Myers JB, & FH, 2002; Malinzak et al., 2001; Pappas E, Hagins ME, Sheikhzadeh A, Nordin M, & Rose D, 2007; Salci Y, Kentel BB, Heycan C, Akin S, & F, 2004; Yu B, Lin CF, & Garrett WE, 2006), and hip (Chappell et al., 2007; Decker et al., 2003; Pappas E et al., 2007; Pollard CD, Davis LM, & Hamill J, 2004; Salci Y et al., 2004; Yu B et al., 2006), and the ground reaction forces subjected to the body during landing (Chappell et al., 2007; Decker et al., 2003; Salci Y et al., 2004).

Neuromuscular fatigue can further exacerbate these potentially harmful landing mechanics and potentially place greater risk upon the ACL (Chappell et al., 2005; Coventry, O'Connor, Hart, Earl, & Ebersole, 2006; Horita, Komi, Nicol, & Kyrolainen, 1996; Kellis & Kouvelioti, 2007; Kernozek, Torry, & Iwasaki, 2008; Madigan & Pidcoe, 2003; McLean et al., 2007; Moran KA & Marshall BM, 2006; Nyland, Caborn, Shapiro, & Johnson, 1999; Nyland, Shapiro, Stine, Horn, & Ireland, 1994; Orishimo KF & Kremenic IJ, 2006; Padua et al., 2006; Rozzi, Lephart, & Fu, 1999). The occurrence of injury has been recorded higher in later parts of competition, suggesting fatigue as one contributing factor to injury (Gabbett TJ, 2000, 2002; Hawkins RD & Fuller CW, 1999; Pinto, Kuhn, Greenfield, & Hawkins, 1999; Rahnama, Reilly, & Lees, 2002; Tuggy ML & Ong R, 2000). Vollestad (1997) defined fatigue as any exercise-induced reduction in the maximal capacity to generate force or power output (Vollestad NK, 1997). In conjunction to decreasing force production of musculature, fatigue has been shown to adversely affect reaction time (Hakkinen & Komi, 1986; McMorris & Keen, 1994; Melnyk M & Gollhofer A, 2007; Nelson & Hutton, 1985; Nicol, Komi, Horita, Kyrolainen, & Takala, 1996; Nyland et al., 1994; Rozzi, Lephart, & Fu, 1999), ligament laxity (Rozzi, Lephart, & Fu, 1999; Sakai, Tanaka, Kurosawa, & Masujima, 1992; Skinner, Wyatt, Stone, Hodgdon, & Barrack, 1986; Stoller, Markolf, Zager, & Shoemaker, 1983) and proprioceptive capabilities (Givoni NJ, Pham T, Allen TJ, & Proske U, 2007; Lattanzio, Petrella, Sproule, & Fowler, 1997; Skinner, Wyatt, Hodgdon, Conard, & Barrack, 1986; Walsh, Hesse, Morgan, & Proske, 2004), thus decreasing muscle performance. A decrease in muscular performance can markedly increase risk of injury because the protective capabilities of the musculature surrounding the knee are compromised.

Understanding the role fatigue plays on the ACL is important for preventative and rehabilitative purposes. Fatigue is associated with athletic competitions and joint kinematics during fatigue may give insight into the high incidence of non-contact ACL injuries in athletic activities. The purpose of this study is to demonstrate the effects of an isolated hamstring fatigue on hip and knee kinematics during a double-leg box drop jump at initial contact and at maximum knee flexion.

Statement of Purpose

The purpose of this study is to determine if isolated hamstring fatigue has an effect on landing mechanics from a drop jump in healthy female recreational subjects.

Research Questions

Null: There will be no significant difference in knee flexion between pre-fatigue and post fatigue at initial contact.

Research Hypothesis 1- Subjects will exhibit greater knee flexion post-fatigue at initial contact.

Secondary: (Coventry et al., 2006; Fagenbaum & Darling, 2003; Kellis & Kouvelioti, 2007; Madigan & Pidcoe, 2003; Nyland et al., 1999)

Null: There will be no significant difference in knee valgus between pre-fatigue and post fatigue at initial contact.

Research Hypothesis 2- Subjects will exhibit greater knee valgus post-fatigue at initial contact.

Primary: (McLean et al., 2007)

Secondary: (Borotikar et al., 2008; Chappell et al., 2005; Ford et al., 2003)

Null: There will be no significant difference in hip flexion between pre-fatigue and post fatigue at initial contact.

Research Hypothesis 3- Subjects will exhibit greater hip flexion post-fatigue at initial contact.

Primary: (Borotikar et al., 2008; Kernozek et al., 2008)

Null: There will be no significant difference in maximum knee flexion between pre-fatigue and post fatigue.

Research Hypothesis 4- Subjects will exhibit greater maximum knee flexion post-fatigue.

Primary: (Fagenbaum & Darling, 2003; Kellis & Kouvelioti, 2007; Madigan & Pidcoe, 2003; Orishimo KF & Kremenic IJ, 2006).

Null: There will be no significant difference in knee valgus between pre-fatigue and post fatigue at maximum knee flexion.

Research Hypothesis 5- Subjects will exhibit greater knee valgus post-fatigue at maximum knee flexion.

Secondary: (Borotikar et al., 2008; Chappell et al., 2005; Ford et al., 2003)

Null: There will be no significant difference in hip flexion between pre-fatigue and post fatigue at maximum knee flexion.

Research Hypothesis 6- Subjects will exhibit greater hip flexion post-fatigue at maximum knee flexion

Secondary:(Coventry et al., 2006; Kellis & Kouvelioti, 2007; Kernozek et al., 2008)

Null: There will be no significant difference in maximum hip flexion between pre-fatigue and post-fatigue.

Research Hypothesis 7- Subjects will exhibit greater maximum hip flexion post-fatigue.

Secondary: (Coventry et al., 2006; Kellis & Kouvelioti, 2007; Kernozek et al., 2008)

Independent Variable

The independent variable for this study is time (pre-fatigue, post-fatigue).

Dependent Variables

The dependent variables for this study are knee flexion, knee valgus, and hip flexion measured in degrees at initial contact and maximum knee flexion. Maximum knee flexion and maximum hip flexion will also be dependent variables.

Operational Definitions

1) Initial Contact- The point of contact when vertical ground reaction force first exceed 10 N. (Coventry et al., 2006; Myer GD, Ford KR, Brent JL, & Hewett TE, 2007)
2) Jump Landing Task- The jump landing task will start with the subject's feet positioned shoulder width apart and on a box height of 30 cm high. The subject will have arms crossed across their chest. The subjects will jump down from the box landing with one foot on a 40 cm x 60 cm forceplate and immediately jump up for maximum height. The subjects then land back on the forceplates with one foot on each forceplate (McLean et al., 2007; Pappas E et al., 2007).
3) Maximum Jump Height- With maximum effort, a subject will jump as high as he or she can. The subject is instructed upon contact with the forceplates, to immediately jump straight up in the air as high as they possibly can. No equipment or measuring device is used to record maximum jump height.

4) Physically active healthy female subjects- physically active females that participate in exercise a minimum of 3 times per week for a minimum of 20 minutes (Onate et al, 2005).
5) Fatigue- one repetition below 30% of maximum torque or a 70% reduction of maximum torque (Kellis & Kouvelioti, 2007)
6) Dominant Limb- limb that the subject preferred to kick a soccer ball (Wikstrom et al 2006)

Assumptions

1) All equipment used will be reliable, valid and accurately calibrated before each session.
2) Subjects will participate with full effort during the induction of fatigue and jump landing task.
3) Different footwear and clothing worn between subjects will have no impact on the results of the study.
4) Subjects will not alter natural landing patterns due to its isolation in nature.

Limitations

1) The jump landing test used in this study does not replicate functional tasks seen in many athletics that increases risk or results in lower-extremity injury.
2) Maximum jump task did not use any objective measurement device to determine if maximal effort was being achieved by the subjects. It was matter of convenience to verbally instruct subjects to exert maximum effort.
3) Caffeine intake, sleep patterns, prior activity and current hydration will not be assessed prior to testing.
4) The peripheral fatigue induced can not be isolated from central fatigue, therefore central fatigue may influence the results.

Delimitations

This study is delimited to 15 physically active females between the ages of 18-35 that exercise for a minimum of 3 times per week for a minimum of 20 minutes and have no previous lower extremity injury in the last six months, no history of ACL injury or surgery, and no previous lower extremity surgery in the last two years.

CHAPTER II

REVIEW OF LITERATURE

The following is a review of literature concerning the anterior cruciate ligament and subsequent issues involving injury risk factors, contribution of dynamic constraints, jump landing and effects of fatigue. These topics all play an important piece in gaining better understanding of the enigma that surrounds anterior cruciate ligament injury. While studies have examined jump landing in conjunction with fatigue, there is a need for more research to identify the factors that cause injury to rise among the onset of fatigue. Identification of kinematics, kinetics and neuromuscular strategies utilized during fatigue and subsequent correction, may provide better insight into prevention of anterior cruciate ligament injuries.

Anatomy of Anterior Cruciate Ligament

The anterior cruciate ligament (ACL) is a major ligamentous structure in the knee. It is the primary restraint to anterior tibial translation on the femur (Butler DL, Noyes FR, & ES, 1980; Ellis H, 2007; Fukubayshi T, Torzilli PA, Sherman MF, & Warren RF, 1982) as well as rotational movement of the tibia on the femur (Ellis H, 2007; Fleming BC et al., 2001; Kanamori A et al., 2002), varus and valgus forces (Marder RA, Raskind JR, & Carroll M, 1991; Withrow TJ, Huston LJ, Wojtys EM, & Ashton-Miller JA, 2006) and any combination of those forces (Markolf et al., 1995). Running through the intercondyles of the tibia, it originates from the anterior aspect of the tibial intercondylar space and inserts on the posteriormedial aspect of the lateral femoral condyle (Chhabra A, Elliott CC, & MD, 2001). The ACL is comprised of two bundles that compliment each other in providing stability through the full range of motion at the knee (Gabriel MT, Wong EK, Woo SLY, Yagi M, & Debski RE, 2004; Xerogeanes JW et al., 1995). Due to the anatomical alignment of the bundles, the degree of knee flexion dictates the contribution of each bundle in contributing to stability. The anteromedial bundle is more taught in knee flexion moments and the more convex posterolateral bundle, plays more of a restraint in extension and rotary moments (Feagin JA, 1988; Gabriel MT et al., 2004; Xerogeanes JW et al., 1995). Both bundles of the ACL are incased in a synovial membrane which ranges from 3 to 38mm in length and 10 to 12 mm in width (Smith BA et al., 1993). The ACL contains virtually no pain fibers, giving reason for minimal pain felt after an acute ACL tear (Shutte MJ, Dabesies EJ, & Zimney ML, 1987). Mechanoreceptors located within the ACL are innervated by the tibia nerve and assist in the ACL's proprioceptive role (Kennedy JC, Alexander IJ, & Hayes KC, 1982).

Anatomy of Musculature

Although the ACL provides up to 86% of anterior tibial translation (Butler DL et al., 1980), internal and external forces surrounding the knee during activities stress the ligament past its constraints, therefore demanding the surrounding musculature to maintain joint stability (Solomonow et al., 1987; Williams GN, Chmielewski T, Rudolph KS, Buchanan TS, & ML, 2001). The hamstrings are composed of three different muscles. The biceps femoris consists of two heads , short and long, and is innervated by two different nerves, peroneal branch and tibial branch of sciatic respectively (Clanton, 1998). The biceps femoris extends the hip, externally rotates the leg and has been shown to display increased activity in ACL deficient knees (Marshall JL, Girgis FG, & RR, 1972; Sebastianelli WJ, Hanks GA, & A, 1990). The long head originates from the ischial tuberosity and inferior aspect of the sacrotuberous ligament and then crosses medial to lateral to fuse with the short head that originates from the lateral lip of the linea apsera, proximal two-thirds of the supracondylar line and from the lateral intermuscular septum (Woodburne RT & Burkel WE, 1994). The conjoined muscle continues to insert into the head of the fibula, crural fascia and proximal tibia (Terry GC & LaPrade RF, 1996). Along with the long head of the biceps femoris, the semitendinosus and semimembranosus originate from the ischial tuberosity of the pelvis and track along the medial side of the femur (Clanton, 1998). The semimembranosus inserts on the oblique popliteal ligament, the posterior capsule, the posterior tibia, the politeus and the medial meniscus (Warren LF & Marchal JL, 1979). The semitendinosus inserts on the proximal medial tibial metaphysic overlying the distal insertion of the medial collateral ligament (Warren LF & Marchal JL, 1979).

The quadriceps, consisting of the rectus femoris, vastus lateralis, vastus medialis, and vastus intermedius, accompany the hamstring in providing dynamic stability of the knee. These four muscles extend the knee and all converge to form a common tendon that attaches to the superior border of the patella and continues to the tibial tuberosity. The rectus femoris originates superiorly to the anterior inferior iliac spine and superiorly to the acetabulum on the ilium. Along with knee extension, the rectus femoris also acts to provide flexion of the hip, given its origin crossing the hip joint. The vastus medialis and vastus lateralis muscles originate from the lateral and medial linea aspera of the femur, and the vastus intermedius muscle originates from the anterior and lateral portion of the femur. The quadriceps muscles are all innervated by the femoral nerve (Seeley RR, Stephens TD, & Tate P, 2003; Stone & Stone, 2006).

Epidemiology/ Etiology of Anterior Cruciate Ligament

The incidence in the general American population is an estimated one in three thousand to thirty five hundred people rupturing the ACL annually (Smith BA et al., 1993). The majority of ACL injuries occur in individuals between the ages of 15- 47, a demographic that includes approximately 47% of the United States population (Griffin LY et al., 2000). Yet the so called epidemic of ACL injury occurs more commonly in athletic competition and more specifically female participants. Females are 2 to 9.7 times more likely to sustain an injury to their ACL in athletics ranging from basketball, soccer, handball, rugby, alpine skiing and military training (Arendt EA et al., 1999; Arendt EA & Dick R, 1995; Gwinn DE, Wilckens JH, McDevitt ER, Rooss G, & Kao TC, 2000; Lindenfeld TN et al., 1994; Messina DF et al., 1999; Myklebust G et al., 1998; Stevenson H et al., 1998). Approximately 70% of all ACL injuries occur through non-contact mechanisms associated with foot contact, such as deceleration, poor landing, pivoting and any other uncontrolled movements in athletic competition (Boden et al., 2000; Ferretti, Papandrea, Conteduca, & Mariani, 1992; Griffin LY et al., 2000; Kirkendall & Garrett, 2000; McNair et al., 1990). An individual with a torn ACL can suffer from a variety of complications that may compromise the integrity of the knee. Untreated complete tears of the ACL have been shown to be functionally unstable leading to recurrent injuries, damage of menisci and articular cartilage, and osteoarthrosis (Frank CB & Jackson DW, 1997; Lohmander LS et al., 2004). These are in conjunction with the cost of repair, an approximate cost of 17,000 dollars per procedure; therefore nationally a financial cost of approximately 850,000,000 dollars is spent each year on ACL repair (Griffin LY et al., 2000).

Mechanism of Injury

Several mechanisms have been hypothesized as responsible for the damage of the ACL. A common noncontact mechanism involves the knee between 20 degrees of flexion and full extension in deceleration before a change in direction or landing (Boden et al., 2000; McNair et al., 1990). Ireland (1999) found the "point of no return", a mechanism for complete structural failure of the ACL, as a position of forward flexion, the hip in adduction, the femur in internal rotation, the knee in 20 to 30 degrees of flexion, the tibia in external rotation and the foot in pronation (Ireland ML, 1999). A strong quadriceps contraction between 50 degrees of flexion and full extension (DeMorat G, Weinhold P, Blackburn T, Chudik S, & Garrett W, 2004; Markolf, Gorek, Kabo, & Shapiro, 1990; McNair et al., 1990), causing excessive anterior tibial displacement, may also cause damage to the ACL (DeMorat G et al., 2004; Feagin JA & Lambert KL, 1985).

Excessive tibial rotation and valgus and varus stresses can cause ACL injury. According to Fu and Stone (1994), excessive tibial external rotation and a valgus stress on the knee has been identified as the most common mechanism for athletes participating in soccer, football and skiing (Fu FH & Stone DA, 1994). Hewett et al (2005) found that increased valgus and external knee abduction moments can predict ACL injury in the female athlete with 73% specificity and 78% sensitivity (Hewett, Myer, & Ford, 2005). Hyperextension or lack of flexion of the knee, in combination with tibial internal rotation, also contributes as a main mechanism of injury to the ACL (Fu FH & Stone DA, 1994; McNair et al., 1990).

Risk Factors for Injury

In an effort to fully understand ACL injuries, risk factors have been identified and classified for those who may be predisposed to an ACL injury. The risk factors are broken down into four categories: environmental, anatomical, hormonal and biomechanical (Griffin et al, 2000). Environmental risk factors are those that would be in an individual's external environment, such as bracing and shoe surface contact. Anatomical risk factors are concerned with the shapes, sizes and interaction between skeletal structures, such as Q angle and joint laxity. Hormonal risk factors are concerned with the levels of particular sex hormones, especially estrogen and progesterone, and the influence of those hormones on the ACL. Finally, biomechanical risk factors include problems or deficits associated with proprioception, neuromuscular control, and technique (Griffin et al, 2000).

As mentioned, the external environment surrounding active individuals can be risk factors in contributing to ACL injury. The use of knee braces has been a controversial issue as whether they contribute or prevent ACL injury. In the 1970s, reports indicated a decrease in the number of knee injuries in collegiate and high school athletes wearing braces (McNair et al., 1990; Randall F, Miller H, & D, 1983). Yet a position statement was released in 1984, by the American Academy of Orthopedics Surgeons, that there was no clear and definite evidence that the use of prophylactic knee braces prevented knee injuries, including injury to the ACL (Griffin LY et al., 2000; McNair et al., 1990). Playing surface and shoe type have also been investigated to determine their relationship to ACL injury. Although a dry surface may be most desirable for athletic performance, noncontact ACL injuries most commonly occur when a playing surface is dry (Scranton PE, Whitesel JP, & Powell JW, 1997). Myklebust et al (1997) found a positive relationship between high level of friction between shoes and playing surface with rate of non-contact ACL injury in team handball (Myklebust G, Maehlum S, Engebretsen L, Strand T, Solheim E, 1997). Similarly, football cleats that have a high level of friction, although they may enhance performance, are also associated with a higher injury risk of ACL (Lambson RB, Barnhill BS, & Higgins RW, 1996).

Anatomical risk factors have been investigated in correlation to their influence on the integrity of the ACL. These risk factors are mainly innate, meaning that they are structural variables and are uncontrollable to those at risk. Excessive foot pronation may be a predisposing factor as prolonged pronation causes further tibial internal rotation, a motion that causes the ACL to tighten (Beckett ME, Massie DL, Bowers KD, & Stoll DA, 1992). Beckett et al (1992) found a significant correlation as subjects with ACL injury had a significant increase in navicular drop in comparison to those uninjured (Beckett ME et al., 1992). Furthermore, Louden et al (1996) found that genu recurvatum in combination with excessive foot pronation can result in greater strain of the ACL (Louden JK, Jenkins W, & KL, 1996). Genu recurvatum causes the ACL to become impinged in the intercondylar notch and over time with repetitive stress produce tensile strain on the ACL (Ireland ML, Gaudette M, & Crook S, 1997). Intercondylar notch width and shape may also influence susceptibility of injury, as a more confined area may place undue stress upon the ACL. Griffin et al. (2000) concluded the notch width in patients with bilateral ACL injuries are less than that in patients with unilateral ACL injuries. Therefore suspecting the notch width in knees with both unilateral and bilateral ACL injuries may be less than in comparison to notch width of uninjured knees (Griffin

LY et al., 2000). Excessive pathological laxity, external tibial rotation, the Q-angle and other foot-pelvis alignments have also been found as anatomical risk factors for female athletes that could result in failure of the ACL (Hutchinson MR & Ireland ML, 1995). Overall, anatomical risk factors can predispose individuals to harmful positioning during activity that are associated to mechanisms of injury.

Hormonal risk factors have also been examined to evaluate the influence of hormones in susceptibility of ACL injuries. Liu et al (1996) found estrogen and progesterone receptor sites in human ACL cells, thus suggesting hormone levels may be a factor in the integrity of the ACL (Liu SH, Al-Shaikh R, & Panossian V, 1996). Wojitys et al (1998) reported a connection between the phases of the menstrual cycle and rate of injury. In the ovulatory phase, when the estrogen level is high, more injuries occurred, yet in the follicular phase, when estrogen and progesterone levels are low, fewer injuries occurred (Wojtys EM, Huston LJ, Lindenfeld TN, Heweet TE, & MLVH, 1998). Adachi et al (2007) also found a significant increase in ACL injuries in teenage females athletes during the ovulatory phase of the menstrual cycle (Adachi N, Nawata K, Maeta M, & Y, 2007). Yet, VanLunen et al (2003) found no significant relationship between the different phases of the menstrual cycle and laxity of the ACL (Van Lunen BL, Roberts J, Branch JD, & Dowling EA, 2003).

Biomechanical risk factors are generally controllable factors that have become the focal point of prevention strategies to reduce the risk of ACL injury. Prevention strategies vary in focus, ranging from muscular strength, plyometrics, flexibility, proprioception, agility and instruction (Hewett et al., 1996; Holm et al., 2004; Paterno, Myer, Ford, & Hewett, 2004; Wilderman, Ross, & Padua, 2009; Wilkerson et al., 2004; Wojtys EM, Huston LJ, Taylor PD, & Bastain SD, 1996). These different facets of prevention are assessed through objective measures such as muscular activation, strength, proprioception, kinematics, kinetics, and performance data (Chimera, Swanik, Swanik, & Straub, 2004; Hewett et al., 1996; Holm et al., 2004; Paterno et al., 2004; Wilkerson et al., 2004; Wojtys EM et al., 1996). Muscular activation refers to the interplay of the quadriceps, hamstrings and gastrocnemius musculature and the timing of the activation (Chimera et al., 2004; Wilderman et al., 2009; Wojtys EM et al., 1996). Similarly, muscular strength of theses muscles is also an area of interest due to the protection it can offer to the knee joint (Hewett et al., 1996; Holm et al., 2004; Wilkerson et al., 2004; Wojtys EM et al., 1996). Intervention programs also focus on improving balance and proprioception (Holm et al., 2004; Paterno et al., 2004). Proprioception is afferent information sent from mechanoreceptors in muscles, ligaments, tendons and capsular structures concerning the position and velocity of lower extremity joints (Griffin LY et al., 2000; Matthews PBC, 1982). The information from the mechanoreceptors is necessary for initiating and mediating neuromuscular control, and enhancing functional joint stability, especially during dynamic movements such as landing (Griffin LY et al., 2000; Shultz & Prerrin, 1999). Neuromuscular control refers to the efferent activation of the dynamic restraints surrounding the joint in response to afferent stimuli (Shultz & Prerrin, 1999). The activation of the musculature, or dynamic restraints, may be initiated automatically as a reflex in response to a stimuli or initiated consciously through voluntary contraction (Griffin LY et al., 2000). Proprioception and neuromuscular control are collective components of focus because they provide functional stability during athletic movements (Griffin LY et al., 2000; Shultz & Prerrin, 1999). Kinetic and kinematic data of the lower extremity are also used as objective feedback for plyometric and agility prevention training (Hewett et al., 1996; McNair, Prapavessis, & Callender, 2000; Wilkerson et al., 2004). As noncontact mechanisms account for the majority of ACL injury, kinetic and kinematic data relays potentially harmful body positioning in all planes during maneuvers.

Jump Landing Mechanism of Injury

Jump landing has been identified as a primary non-contact mechanism for ACL injury in female basketball, volleyball and soccer players (Arendt EA et al., 1999; Arendt EA & Dick R, 1995; Piasecki, Spindler, Warren, Andrish, & Parker, 2003). ACL injuries that occur during jump landing are also associated with an increased rate of meniscus tear which increases the likelihood of osteoarthritis (Lohmander LS et al., 2004; Paul JJ et al., 2003). Jump landing increases the risk of injury due to the potential increase of forces (McNair et al., 1990; Pappas E et al., 2007; Salci Y et al., 2004; Self BP & Paine D, 2001; Zhang SN, Bates BT, & Durek JS, 2000) and an amplified dependence of the neuromuscular system to coordinate the body in a protective and proficient position during landing (Cerulli et al., 2003; G. Li et al., 1999). Yet, females have been shown to land in athletic activities in a more erect and potentially harmful posture, thus putting the ACL at greater risk for injury (Chappell et al., 2007; Chappell et al., 2002; Cowley HR et al., 2006; Decker et al., 2003; Fagenbaum & Darling, 2003; Hewett et al., 2005; Huston, Vibert, Ashton-Miller, & Wojtys, 2001; Lephart SM et al., 2002; Malinzak et al., 2001; Pappas E et al., 2007; Salci Y et al., 2004; Yu B et al., 2006).

Types of Jump Landing in Research

Numerous types of jump landing tasks are utilized in research. Most often jump landing tasks are standardized in some fashion, as to keep the task uniform across all subjects and reduce individual differences. A popular jump landing method used in research is a box drop landing. The height of the box can be to a set height, such as 20cm (Lephart SM et al., 2002), 25 cm (Madigan & Pidcoe, 2003), 30cm (Fagenbaum & Darling, 2003; Ford et al., 2003; Kellis & Kouvelioti, 2007; Moran KA & Marshall BM, 2006), 40 cm (Sander TC et al., 2004), 50 cm(Kernozek et al., 2008; McLean et al., 2007; Moran KA & Marshall BM, 2006), 52 cm (Gehring, Melnyk, & Gollhofer, 2009) and 60cm (Fagenbaum & Darling, 2003). Drop distance can also be adjusted to a percentage of a maximal height jump of each subject (Coventry et al., 2006). Studies can include various heights, if landing adaptations to changes of height are of interest (Huston et al., 2001; McNitt-Gray, 1993). The height of the box may also be standardized to each subject by their height (Huston et al., 2001). The distance between the edge of the box and the designated area for landing may also be standardized, such as 30cm(Sander TC et al., 2004), 33cm (Madigan & Pidcoe, 2003).

Upon contact, studies have utilized double leg and single leg landings. Single leg landings require the subjects to land successfully within an outlined area (Kellis & Kouvelioti, 2007; Orishimo KF & Kremenic IJ, 2006; Sander TC et al., 2004). The single leg chosen for analysis were standardized for the dominant leg (Coventry et al., 2006; Kellis & Kouvelioti, 2007) or one particular side (Madigan & Pidcoe, 2003). Double leg landings require subjects to land successfully within an outlined area and subsequently perform a jump for maximum height. The maximum height jump can be standardized by a Vertec jump device (Wikstrom EA, Powers ME, & Tillman MD, 2004), measured by an overhead object (Jacobs, Uhl, Mattacola, Shapiro, & Rayens, 2007) or verbal instructions to reach personal maximum height (McLean et al., 2007).

More functional jump landing tasks are also used in research, in efforts to utilize tasks that are may be seen in athletics. The jump landing tasks may include: forward stop jump task (Benjaminse et al., 2008; Chappell et al., 2005; Orishimo KF & Kremenic IJ, 2006), countermovement jumps (Coventry et al., 2006), vertical stop jump task(Chappell et al., 2005), backward stop jump task(Chappell et al., 2005), and continuous hopping (Padua et al., 2006).

Jump Landing on ACL Research

Utilizing jump landing tasks, numerous studies have supported that females land with lower knee flexion (Chappell et al., 2007; Chappell et al., 2002; Decker et al., 2003; Lephart SM et al., 2002; Malinzak et al., 2001; Salci Y et al., 2004; Yu B et al., 2006). Salci et al (1994) investigated different landing patterns in collegiate volleyball players from different box heights and horizontal distances, and consistently found less knee flexion in female athletes (Salci Y et al., 2004). Similarly, Lephart et al (2002) also found less knee flexion in landing of female volleyball, basketball and soccer athletes in comparison to their male counterparts. Results revealed that females took significantly less time to reach their maximal knee flexion once landed, causing a more abrupt and less absorbent impact of landing (Lephart SM et al., 2002). Landing with increased knee flexion is a more protective position as it may reduce anterior load on the ACL during landing. Various studies have examined ACL loading in cadaver knees as a method to determine the direct influence of in-situ forces on the strain of the ACL (G. Li et al., 1999; Markolf et al., 1995; Markolf et al., 1990). Cadaver models demonstrate that the quadriceps musculature causes an anterior shear force at the proximal tibia, a major ACL loading mechanism. Landing in higher angles of knee flexion is more favorable because the force of the quadriceps inversely decreases to the protective force of the hamstrings on the tibia to create less anterior shear force on the ACL (G. Li et al., 1999; Markolf et al., 1995; Markolf et al., 1990). Using cadaver knees with simulated muscular forces, Li et al (1999) reported the protection of the hamstrings contraction is ineffective at knee flexion angles less than 30 degrees because it loses its mechanical advantage of pull to decrease anterior shear forces on the ACL (G. Li et al., 1999). Similarly, O'Connor et al (1993) found from full extension to 22 degrees of flexion the hamstrings contraction was unable to reduce the strain on the ACL (O'Connor, 1993).

Similarly, the ACL endures greater strain and is more prone to injury when the knee is placed under valgus stress. Knee valgus angles is often attributed to part of ACL mechanism of injury (Fu FH & Stone DA, 1994). Although valgus forces are often seen in conjunction with other forces, computer modeling has shown valgus moments alone are high enough to rupture the ACL (Cowley HR et al., 2006). Various studies have reported females landing with high knee valgus forces in comparison to male counterparts (Chappell et al., 2002; Cowley HR et al., 2006; Ford et al., 2003; Malinzak et al., 2001; Pappas E et al., 2007). Knee valgus angles and moments during drop landings have been found to predict noncontact ACL injury risk in female athletes with high sensitivity and specificity (Hewett et al., 2005). Unilateral landing exacerbate even more unfavorable knee valgus angles in comparison to bilateral landing. Pappas et al (2007) reported a mean valgus increase of 2.4 degrees in unilateral landings in comparison to bilateral landings in both male and female subjects. Overall, gender comparison revealed female subjects had 4.5 greater degrees of knee valgus than males during both landing types (Pappas E et al., 2007). Although the increase of degrees may appear minimal, Bendjaballah et al (1997) found that five degrees of knee valgus, in comparison to the knee aligned to the frontal plane, could increase the load on the ACL up to six times (Bendijaballah MZ, Shirazi-Adl A, & Zukor DJ, 1997).

Although the knee and ACL are the areas of interest, the hip influences knee position, loads and stiffness (Gregoire L, Veeger HE, Huijing PA, & GJ, 1991; Putnam CA, 1993) and moment produced at the hip characterizes moment developed at the knee (Bobbert MF & van Zandwijk JP, 1999). Palmitier et al (1991) concluded that eccentric contraction of the quadriceps maintains stability of the knee, while eccentric contraction of the hamstrings maintains hip stabilization (Palmitier RA, An KN, Scott SG, & EY, 1991). Similar to knee kinematics, hip flexion has shown to be less in females in comparison to males (Chappell et al., 2007; Decker et al., 2003; Pappas E et al., 2007; Salci Y et al., 2004; Yu B et al., 2006). Hip flexion is desired in landing, as similar to the knee, because it allows for the absorption of ground forces acting upon the body when landing. If decreased hip flexion is present, accompanied by the contraction of the quadriceps and large ground reaction force, the tibia may be forcefully translated anteriorly and create a common mechanism of injury to the ACL (Ball KA, 1999).

Hip extensor and abductor musculature have been reported as important contributors to lower extremity stability. McNitt-Gray et al (1993) documented peak hip extensor moment was 163% greater than peak knee extensor moment and 137% greater than peak ankle extensor moment in healthy male subjects landing from 72cm drop landing(McNitt-Gray, 1993). Similarly in 60 cm drop landings, Decker et al (2003) reported the peak hip extensor moment as 120% and 243% greater than peak knee extensor moment and peak ankle plantar flexion moment in male recreational athletes respectively (Decker et al., 2003). The authors report the hip extensor musculature is mainly responsible for the absorption and control of the lower extremity during the drop landings. Jacobs et al (2007) found a significant correlation between hip abductor strength and knee valgus when landing from jump. Women demonstrated a significant decrease in abductor strength to male counterparts and an average of 4 degrees greater knee valgus during a single leg landing (Jacobs et al., 2007). An increase knee valgus during landing is a main concern, as it is a mechanism of injury to the ACL (Bendijaballah MZ et al., 1997; Hewett et al., 2005).

Peak vertical ground reaction force may increase internal loads, increasing the risk of injury, if the musculoskeletal system does not distribute the forces sufficiently (Lephart SM et al., 2002). Peak ground reaction force to body weight measurements have varied from 1.0 to 6.7 times body weight (McNair et al., 1990; Pappas E et al., 2007; Salci Y et al., 2004; Self BP & Paine D, 2001; Zhang SN et al., 2000). Different tasks are likely to elicit these different ground reaction forces due to the requirements of the task at hand. Similarly, different demographics of people, whether distinguished by sex, activity level, or level of competition, may produce different forces during given tasks. In a study by Cowley et al (2006), high school female basketball players and soccer players elicited different ground reaction forces in both cutting and jump landing. The duration of the stance phase was an indication of the ground forces, as soccer players had a 20.5% decrease in stance phase to basketball players, accounting for the average increase of 15.6% in peak vertical ground reaction force (Cowley HR et al., 2006). The stance phase allows the body to absorb the ground reaction forces through the contribution of hip, knee and ankle range of motion. Salci et al (2004) found female volleyball players to land with higher ground reaction forces than males and attributed the results for the decreased knee flexion during landing (Salci Y et al., 2004). Kinematic alterations of the lower extremity can also yield similar ground reaction forces during different tasks. Double leg and single leg landings recorded comparable forces due to range of motion variations at the hip, knee and ankle (Pappas E et al., 2007). Studies have also reported no differences in the magnitude of vertical ground reaction forces between genders due to their landing strategies that utilize different range of motion at the lower extremity joints. Decker et al (2003) reported female recreational athletes landed with greater knee and ankle range of motion compared to male counter parts increasing the absorption time (Decker et al., 2003). Likewise, McNair and Prapavessis et al 1999 discovered that both genders of adolescents landed with similar landing forces due to increased ankle range of motion in females (McNair PJ & Prapavessis H, 1999). Although there is controversy on the degree of ground reaction forces absorbed during landing, animal studies have shown that repetitive impact forces may threaten the integrity of the lower extremity and cause joint degeneration due to overwhelming and cyclical stress (Radin et al, 1973).

Muscular Stability

Dynamic stabilization of the knee and absorption of ground reaction forces during landing is provided primarily through the coactivation of the quadriceps and hamstrings (Bonci CM, 1999; Fleming BC et al., 2001; Imran A & O'Connor JJ, 1997; Lloyd, Buchanan, & Besier, 2005). The quadriceps eccentrically contracts to control knee flexion and decelerate the land (Lephart SM et al., 2002), while the hamstrings are thought to stabilize the knee (Barratta R et al., 1988; Solomonow et al., 1987), by providing protection to the ACL by resisting anterior translation and internal rotation of the tibia (Aune et al., 1995; Baker BE, 1991; Barratta R et al., 1988; Dureselen L et al., 1995; MacWilliams et al., 1999; Solomonow et al., 1987). The co contraction of these muscles provides dynamic stability and stiffness to the knee that protect against excessive anterior translation and valgus and varus moments (Lloyd & Buchanan, 2001; Markolf et al., 1978). More specifically, the co-contraction compresses the joint which increases contact surface within the joint and prevents excess movement (Imran A & O'Connor JJ, 1997; Markolf et al., 1981; Markolf et al., 1976; Solomonow et al., 1987; Torzilli et al., 1994). Markolf et al (1978) displayed a three fold decrease in both valgus and varus laxity of the knee with the addition of muscular protection (Markolf et al., 1978). Muscular protection provided by the co-contraction was found to be significantly less in female athletes in comparison to male athletes in studies by Wojtys et al. 2002, and Wojtys et al. 2003. In response to torsion and shear perturbations, females displayed greater translation than the male counterparts, thus causing greater strain on the ligamentous support (Wojtys EM, Huston LJ, Schock HJ, Boylan JP, & Ashton-Miller JA, 2003; Wojtys, Ashton-Miller, & Huston, 2002).

Studies have revealed that female athletes have weaker quadriceps and hamstring muscles when normalized to body weight, thus possibly playing a role in susceptibility of an ACL injury (Huston & Wojtys, 1996; Lephart SM et al., 2002; Myer et al., 2009; Wojtys EM et al., 2003; Wojtys et al., 2002). Lephart et al (2002) suggests that quadriceps weakness causes females to land in a more extended knee position, a common mechanism of ACL injury, and maintain the extension with ground contact rather than absorbing the impact through knee flexion (Lephart SM et al., 2002). As hamstrings play a synergistic role to the ACL, Myer et al (2009) found a correlation with deficient hamstring strength and ACL injury (Myer et al., 2009). Using data collected prior to injury, hamstring strength of collegiate female athletes were significantly lower than hamstring strength of matched male control athletes, indicating a predisposed condition. Conversely, the control group of female athletes with no ACL injury had decreased quadriceps strength and similar hamstrings strength compared to match male athletes. Thus suggesting females who have a greater hamstrings and quadriceps ratio (H/Q ratio) or balance between these two opposing muscle groups, may be at lower risk to ACL injury(Myer et al., 2009). A low H/Q ratio indicates a low activity level of the hamstring muscles or hyperactivity of the quadriceps muscle.

Different methods have been used in research to establish safe parameters of strength ratios. Conventional ratios compare concentric strength of the quadriceps to the concentric strength of the hamstrings while functional ratios compare concentric strength of the quadriceps to the eccentric strength of the hamstrings (Aagaard, Simonsen, Magnusson, Larsson, & Dyhre-Poulsen, 1998; Aagaard, Simonsen, Trolle, Bangsbo, & Klausen, 1995; Clanton, 1998; Heiser TM, Weber J, Sullivan G, Clare P, & Jacobs RR, 1984; Knapik JJ, Bauman CL, Jones BH, Harris JM, & L, 1991; R. C. Li, Maffulli, Hsu, & Chan, 1996). The functional ratio assesses muscle action that is more functional and occurs during activities (Aagaard et al., 1998; Aagaard et al., 1995; R. C. Li et al., 1996). Knapik et al (1991) found female athletes with a conventional H/Q ratio of less than .75

were 1.6 times more likely to sustain a knee injury during a collegiate season(Knapik JJ et al., 1991). While a H:Q ratio of .5 and higher is recommended specifically for hamstring prevention. Considering an eccentric contraction will be higher than a concentric contraction, a healthy recommended functional H:Q ratio is 1:1 or 1.0 (Aagaard et al., 1998; Aagaard et al., 1995; R. C. Li et al., 1996). Li et al (1996) proposed an equal value ratio to be beneficial in limiting anterior tibial translation, an injury mechanism of the ACL (R. C. Li et al., 1996).

Neuromuscular activation to produce force and the pattern in which they are activated has been found to be different between males and females. Females have displayed a greater quadriceps dominant muscle activation strategy to low hamstring activation during athletic tasks, thus creating a greater H/Q ratio and placing higher risk upon the ACL (Chappell et al., 2007; Hewett et al., 1996; Sander TC et al., 2004; Zazulak et al., 2005). The quadriceps contributes to ACL loading through the anterior pull of the patellar tendon on the tibia when the knee flexion is less than forty-five degrees (Markolf et al., 1995; Myer GD, Ford KR, & Hewett TE, 2005). Increased quadriceps activation will result in increased quadriceps force that may increase ACL loading. To reduce quadriceps induced anterior translation, ligament and graft strain, the hamstring muscles serve as dynamic ACL synergists (Aune et al., 1995; Baker BE, 1991; Barratta R et al., 1988; Dureselen L et al., 1995; MacWilliams et al., 1999; Solomonow et al., 1987). Zazulak et al (2005) reported a quadriceps dominant strategy in landing tasks with an increase in quadriceps activation in preparation to landing, thus decreasing the H/Q ratio (Zazulak et al., 2005). Similarly the H/Q ratio may be altered by a decrease in hamstring activation. In a study of recreational athletes, Chappell et al (2007) found that muscle activation of hamstring and quadriceps in females was higher in preparation of landing, yet during the contact phase hamstring activation was lower than males (Chappell et al., 2007). Likewise, Hewett et al (1996) reported knee flexor moments three times lower in female athletes in comparison to male athletes during landing tasks (Hewett et al., 1996). Sander et al (2004) reported females demonstrate a quadriceps dominant landing strategy in single leg landing. Females pre-activated their quadriceps 49% more than male counterparts, resulting in a significant increase in anterior shear force and internal extension moments (Sander TC et al., 2004). Huston et al. (1996) compared neuromuscular activity of elite Division I female athletes to Division I football players and nonathlete males and females. The elite female athletes more often initiated the quadriceps muscles first, while the other groups more often initiated the hamstrings first during landing. Isokinetic testing was also performed and showed female athletes took longer than the male athletes to reach peak torque in their hamstrings at 60 and 240 degrees per second (Huston & Wojtys, 1996). The reduced and delayed activation of the hamstrings shown by the elite female athletes, demonstrates that neuromuscular recruitment of females, no matter skill level, puts the ACL at higher risk by limiting the synergist capabilities of the hamstrings.

Similar to the voluntary activation of the surrounding musculature, the reflex responses also play an important role in the stability of the knee. Mechanoreceptors, which are located in the muscles, tendons, ligaments and joint structures of the knee, detect change in position and acceleration of the knee, and initiate the reflex (Boyd, 1954; Johansson, Sjolander, & Sojka, 1991; Schultz, Miller, Kerr, & Micheli, 1984; Solomonow & Krogsgaard, 2001; Zimny, Schutte, & Dabezies, 1986). Whether classified as low or high threshold, once the mechanoreceptors surpass their threshold a protective involuntary reflexive muscular response is initiated (Sjolander, Johansson, Sojka, & Rehnholm, 1989).

Playing a direct protective mechanism, a reflect pathway between the mechanoreceptors in ACL and the hamstrings has been demonstrated in studies (Dyhre-Poulsen & Krogsgaard, 2000; Friemert et al., 2005; Fujita, Nishikawa, Kambic, Andrish, & Grabiner, 2000; Solomonow et al., 1987). Solomonow et al (1987) showed a responsive reflex hamstring contraction subsequent to anterior translation of the tibia, placing the ACL under stress limits, in healthy male subjects (Solomonow et al., 1987). Dhyre-Poulsen et al (2000) further investigated the protective reflex by placing wire electrodes through the substance of ACLs of healthy participants. Postoperatively, stimulation of electrodes at a high stimuli resulted in muscular contraction of the semitendinosus and an inhibition of the antagonist musculature, demonstrating the reflex pathway present in the ACL (Dyhre-Poulsen & Krogsgaard, 2000). The reflex activity may not purely prevent ligament failure due to the latency of the ligament-muscular reflex and the high threshold that is necessary (Dyhre-Poulsen & Krogsgaard, 2000; Fujita et al., 2000). Due to the neural integration with other mechanoreceptors around the knee, it may contribute to the overall stability during motion (Dyhre-Poulsen & Krogsgaard, 2000; Solomonow & Krogsgaard, 2001). Johansson et al (1991) suggests that low-threshold mechanoreceptors in the cruciate ligaments may increase the sensitivity of the muscle spindles. The increased sensitivity provides a heightened awareness of muscle lengthening in response to stresses, resulting in an earlier contraction of the musculature and joint stiffness (Johansson et al., 1991).

The mechanoreceptors in other joint structures can also elicit a protective muscular reflex. If a muscle is subjected to rapid change in length, the muscle spindles will inherently contract, as a protective mechanism, to resist damage. During anterior tibial translation of the tibia, the hamstrings are placed on rapid stretch, exciting the muscle spindles, and rapidly contract to counterbalance the movement (Solomonow & Krogsgaard, 2001). Likewise, the muscle spindles and joint capsule provide a reflex pathway of the hamstrings in ACL deficient subjects in response to tibia translation (Solomonow et al., 1987).

Fatigue on Incidence of Injury

Studies have shown that fatigue (Chappell et al., 2005; Coventry et al., 2006; Horita et al., 1996; Kellis & Kouvelioti, 2007; Kernozek et al., 2008; Madigan & Pidcoe, 2003; McLean et al., 2007; Moran KA & Marshall BM, 2006; Nyland et al., 1999; Nyland et al., 1994; Orishimo KF & Kremenic IJ, 2006; Padua et al., 2006; Rozzi, Lephart, & Fu, 1999) and decision making (Borotikar et al., 2008; Pollard CD et al., 2004) contribute directly to ACL injury risk because of high-risk kinematics and joint neuromuscular strategies. Altered knee joint biomechanics are common when individuals are exposed to either factor (Chappell et al., 2005; McLean et al., 2007). Although in a real game setting, neither one of these factors would be independent of each other therefore potentially providing an even greater risk of ACL injury. Suggesting the effects of fatigue, the occurrence of athletic injuries has been shown to be higher later in play. The incidence of injury to skiers has been reported higher in the afternoon (Tuggy ML & Ong R, 2000) and rugby injuries have been recorded significantly higher in the second half of play (Gabbett TJ, 2000, 2002). A great percentage of non-contact knee injuries in soccer occur in the last 15 minutes of the first half and last 30 minutes of the second half (Hawkins RD & Fuller CW, 1999). More specifically the last 15 minutes having the highest number of injuries with moderate injury potential (Rahnama et al., 2002). Hockey injuries have also occurred more often later in periods and toward the end of competition (Pinto et al., 1999).

Definition of Fatigue

Neuromuscular fatigue can be defined as a decrease in performance, under certain conditions (Edwards RHT, Hill DK, Jones DA, & Merton PA, 1977). For example, during a maximal contraction of a muscle, the force being produced will decline steadily and fatigue will be recordable. According to the given definition, fatigue is when the target force is no longer achievable, yet if submaximal force is produced, performance may be maintained at the target force for a long time (Vollestad NK, 1997). Therefore fatigue can further be defined as "a loss of maximal force generating capacity" (Gandevia SG & McKenzie DK, 1988). Although muscle weakness can also fall within the given definition, consequently Vollestad et al (1997) provides a clear distinction that fatigue is "any exercise-induced reduction in the maximal capacity to generate force or power output" (Vollestad NK, 1997) and fatigue, unlike weakness, is reversible with rest (Fitts, 1994).

Central Fatigue

The generation of voluntary force results from a sequence of events and reactions and each of these is a potential site of fatigue. The "chain of commands" involves the central nervous system, starting at the brain and ending at the excitation of motorneurons, and the peripheral mechanisms, starting at the excitation of muscle fibers through the neuromuscular junction and ending at the cross-bridges of the muscles themselves (Edwards, 1981; Gibson & Edwards, 1985; Vollestad NK, 1997). Central fatigue can be caused by lack of motivation, impaired transmission down the spinal cord and impaired recruitment of motor neurons (Gibson & Edwards, 1985; Vollestad NK, 1997). Asmussen and Mazin (1978) conducted two fundamental studies that demonstrated the central component of fatigue in male recreational athletes. The subjects fatigued the biceps of their dominant arm, and then following a two minute pause, the subjects continued the lifting. The study revealed that when the subjects participated in a mental distraction, mental arithmetic, or a physical distraction such as lifting with the opposing limb, the subjects produced greater work output in comparison to doing nothing during the two minute rest. The results demonstrate that the central nervous system does play a role on fatigue, and in this case the arousal of the central nervous system facilitated a motor unit recruitment to increase strength (Asmussen & Mazin, 1978a, 1978b). Since then, studies have demonstrated central fatigue by an increase in force as a result of nerve stimulation, thus bypassing the central input of muscle contraction (Gandevia, Allen, Butler, & Taylor, 1996; Herbert & Gandevia, 1996). In a study by Gandevia et al (1996), a progressive decline of force was documented in the biceps after repeated and sustained maximal voluntary contractions (MVCs). Following the decline of force, the motor cortex was electrically stimulated and an increase in force resulted. This can suggest that some motor units were not functioning optimally to produce maximal contractions at the time of stimulation (Gandevia et al., 1996).

Peripheral Fatigue

Fatigue may also occur due to dysfunction within the peripheral contributions of muscle contraction. Peripheral fatigue may involve impairment of the function of the peripheral nerves, neuromuscular junction transmission, electrical activity of muscle fibers or the processes of activation within the fiber (Edwards, 1981). The cross-bridges of the muscle fibers can be compromised during fatigue. To create a contraction, actin and myosin proteins transition from a weak bond to a strong high-force state and subsequent force production through a power stoke with the release of inorganic phosphate. Adenosine diphosphate (ADP) is then produced to repeat the cross-bridge cycle (Fitts, 2008). A decline in the force production of each cross bridge and/or a decline in the number of cross bridges in a strong high-force state can result in a reduction of peak force during fatigue (Fitts, 2008). Intense activities increase the rate of adenosine triphosphate (ATP) glycolysis which thus increases the production rate of hydrogen and inorganic phosphate. These two byproducts subsequently diminish sensitivity to Ca concentration needed for the contractile proteins (Fitts, 2008; MacIntosh BR & Rassier DE, 2002). Decreased force during fatigue can also result from a low free Ca concentration needed for the contractile proteins (MacIntosh BR & Rassier DE, 2002).

Effects of Fatigue

Fatigue is most commonly assessed and measured by the capacity to generate force. Maximal voluntary contractions (MVCs) are a method used by researchers. MVCs can be used as a tool to directly asses fatigue (Clark BC, Collier SR, Manini TM, & Ploutz-Snyder LL, 2005) and as benchmarks for muscular fatigue in electromyography (EMG) data (Kellis & Kouvelioti, 2007; Madigan & Pidcoe, 2003; Orishimo KF &

Kremenic IJ, 2006; Padua et al., 2006; Rozzi, Lephart, & Fu, 1999). Subjects are placed in a standard testing position and generate the highest possible force while the length of the muscle does not change (Vollestad, 1997). For instance, fatigue for the knee extensors can be evaluated in only one direction and the position of knee and hip flexion can be standardized for replication. If vocal or visual encouragement is supplemented, low variability of unfatigued MVCs of upper and lower limb joints is reported (Gandevia et al., 1996; Vollestad NK, 1997). Although it can prove to be difficult to attribute a decline in maximal force to a single muscle in particular muscular groups, such as the back (Vollestad, 1997). The static nature of MVCs also do not replicate dynamic movements and though they are used to assess the whole muscle being tested, less than 50% of the muscle mass is activated at a given time (Sargeant & Kernell, 1993).

Dynamic measurement variables, such as velocity and power, are also means of assessing fatigue (Cairns SP, Knicker AJ, Thompson MW, & Sjogaard G, 2005; Vollestad NK, 1997). In comparison, power output declines much more rapidly than generated isometric force (Cairns SP et al., 2005). This may be apparent because concentric contractions are more demanding of energy, in comparison to isometric contractions (Woledge, Curtin, & Homsher, 1985), and power is dictated by force and velocity, which both are affected during fatigue (Cairns SP et al., 2005). Power output or torque can be calculated at different isokinetic speeds and measured during each contraction to give insight into the exercise changes occurring at different stages of fatigue and phases of contraction (Vollestad, 1997).

Fatigue not only decreases force production, but also adversely affects other musculoskeletal functions. Studies have shown deterioration in proprioception as a result of physical fatigue (Givoni NJ et al., 2007; Lattanzio et al., 1997; Skinner, Wyatt, Hodgdon et al., 1986; Walsh et al., 2004). As the mechanoreceptors within the musculature around the knee play an important link in joint position sense, it had been hypothesized that if the musculature is fatigued, the afferent pathway is disrupted, reducing the joint's stability (Lattanzio et al., 1997; Skinner, Wyatt, Hodgdon et al., 1986). While other studies attribute the decline of position sense on central origin or the impaired judgment of muscle force and effort (Allen & Proske, 2006). Proprioception is often evaluated in studies using position matching, as one limb must match the position of the opposing limb. During position matching of fatigued musculature, subjects must exert more effort to attain a given position; therefore if they match the contralateral limb effort matching errors will occur (Allen & Proske, 2006; Walsh et al., 2004). Therefore it is suggested athletes may decrease proprioception by disrupted pathways and/or using effort as a gauge of joint position instead of feedback. Whether the origin of deficit is peripheral or central, decreased proprioception decreases the ability to detect body positioning during movements. Therefore reduced proprioception, with the onset of fatigue, will increase the likelihood of potentially harmful body positioning during athletic tasks.

Ligamentous laxity has also been shown to coincide with the onset of fatigue. Cyclical loading, or high repetitive stresses, of the ACL occurs during normal and athletic activities such as walking, running, bicycling, cutting, and jumping (Rozzi, Lephart, & Fu, 1999; Sbriccoli P, Solomonow M, Zhou BH, Lu Y, & Sellards R, 2005). Due to the protective viscoelastic characteristics of the ACL, repetitive loading will decrease the threshold of load or stress of the ligament causing laxity (Woo, Abramowitch, Kilger, & Liang, 2006). Chu et al (2003) recorded co-contraction changes after static ACL ligament creep. During both active knee flexion and extension, increased force and electromyography was recorded in the agonist musculature without compensation from the antagonist (Chu et al., 2003). Following a running fatigue protocol of 3.5 miles, Stoller et al (1983) demonstrated an increase in knee torsional laxity. The increase remained above baseline measures for an average of 52 minutes following the conclusion of the fatigue protocol (Stoller et al., 1983). Likewise, 90 minutes after basketball participation, Sakai et al (1992) reported an increased anterior knee laxity in female semiprofessional basketball players (Sakai et al., 1992). Anterior knee laxity was also found by Skinner et al (1986) following a fatigue protocol consisting of sprints and isokinetic testing (Skinner, Wyatt, Stone et al., 1986). As studies have reported females possess greater joint laxity than male counterparts (Huston & Wojtys, 1996; Rozzi, Lephart, Gear, & Fu, 1999), fatigue can further exacerbate this potential risk factor. Sbriccoli et al (2005) applied a passive anterior cyclic load, approximately 20 to 25% of body weight, for 10 minutes to healthy subjects in knee flexion angles of 35 and 90 degrees. Females displayed significantly greater anterior displacement following cyclic loading, suggesting the ACL experienced laxity, as it is the major restraint to anterior tibia translation at the testing angles (Sbriccoli P et al., 2005).

Studies have demonstrated that ligamentous creep due to static or cyclic loading is associated with a decrease in excitability of the reflex arcs initiated by mechanoreceptors within the ACL, causing impairment of the protective reflex response of the musculature (Solomonow et al., 2003; Solomonow, Zhou, Baratta, Lu, & Harris, 1999). Fatigue has been attributed to decrease reflex magnitude (Melnyk M & Gollhofer A, 2007; Nelson &

Hutton, 1985; Nicol et al., 1996) and increase muscle reaction time (Hakkinen & Komi, 1986; McMorris & Keen, 1994; Nyland et al., 1994; Rozzi, Lephart, & Fu, 1999). Whether initiated by the mechanoreceptors within the musculature, ligaments or joint capsule, reflex activity occurs before the onset of voluntary muscle contraction once a threshold of stress is met (Sjolander et al., 1989). Therefore a decrease in the magnitude of reflex or latency in reflex activity can compromise the dynamic stability of the knee. Subsequent to hamstring fatigue, the magnitude of the biphasic hamstring reflex significantly reduced in response due to anterior tibial translation in a study by Melynk et al (2007) (Melnyk M & Gollhofer A, 2007). Likewise, Nicole et al (1996) showed a decrease in magnitude of the stretch reflex in the lateral gastrocnemius, a synergist to the ACL, immediately after a fatiguing stretch shortening cycle exercise (Nicol et al., 1996). Other studies have focused on the latency of reflexes in response to fatigue to assess joint stability. Rozzi et al (1999) reported an increase of the onset of contraction time for the medial hamstring muscle and lateral gastrocnemius during a jump landing task after an isokinetic knee flexor and extensor fatigue protocol (Rozzi, Lephart, Gear et al., 1999). Wojtys et al (1996) also demonstrated delayed reaction time of the gastrocnemius, quadriceps and hamstrings musculature in response to anterior tibial translation after fatigue (Wojtys, Wylie, & Huston, 1996). The delayed or decreased magnitude of reflexes can place greater risk on the ACL by limiting the dynamic stability provided by the thigh musculature.

Fatigue Protocols in Research

The onset of fatigue around the knee is often replicated in research using either a general fatigue protocol or an isolated isokinetic fatigue protocol. General lower extremity protocols utilize different repetitive tasks such as squats (Borotikar et al., 2008; Coventry et al., 2006; Kernozek et al., 2008; Madigan & Pidcoe, 2003; Padua et al., 2006), vertical jumps (Chappell et al., 2005; Coventry et al., 2006), landings from various heights (Coventry et al., 2006; Madigan & Pidcoe, 2003), step-ups (McLean et al., 2007; Orishimo KF & Kremenic IJ, 2006), plyometric bounding (McLean et al., 2007), and sprinting or running (Benjaminse et al., 2008; Chappell et al., 2005; Miura KM et al., 2004; Moran KA & Marshall BM, 2006). Fatigue can be determined when the subject feels he/she can no longer successfully perform the tasks in the protocol (Borotikar et al., 2008; Chappell et al., 2005; Coventry et al., 2006), the subject's heart rate (Benjaminse et al., 2008; Miura KM et al., 2004), increase in time of completion of tasks (Wikstrom EA et al., 2004), or the subject reports the highest numerical value in a scale of rating of perceived exhaustion (RPE) (Benjaminse et al., 2008; Moran KA & Marshall BM, 2006).

Isolated isokinetic fatigue protocols induce fatigue through repetitive (Melnyk M & Gollhofer A, 2007; Miura KM et al., 2004; Nyland et al., 1999; Rozzi, Lephart, & Fu, 1999; Wikstrom EA et al., 2004) or sustained contractions (Clark BC et al., 2005). The onset of fatigue is then determined when the sustained contraction falls below a percentage of each subject's baseline (Clark BC et al., 2005) or three to five repetitions fall below a predetermined percentage of each subject's baseline peak torque (Melnyk M & Gollhofer A, 2007; Miura KM et al., 2004; Nyland et al., 1999; Rozzi, Lephart, & Fu, 1999; Wikstrom EA et al., 2004).

Fatigue Neuromuscular Strategies

Kinematic changes that occur subsequent to the onset of fatigue have been attributed to changes in neuromuscular strategies. Due to compromised muscular function, redistribution of work and muscular recruitment is altered to accomplish a given task. Although the redistribution can be classified as a shift from fatigued musculature to a greater reliance to less or non-fatigued musculature (Coventry et al., 2006; Kellis & Kouvelioti, 2007; Madigan & Pidcoe, 2003; Orishimo KF & Kremenic IJ, 2006). One strategy found in research is the ankle dominant strategy, which is an increased utilization of ankle musculature, mainly the gastrocnemius and soleus. Padua et al (2006) concluded due to the fatigue task chosen, consisting of squats, little stress was placed on the ankle musculature, thus causing the increased activation of the ankle musculature and less reliance placed upon the knee musculature, largely the hamstrings musculature (Padua et al., 2006). Orishimo et al (2006) recorded a proximal to distal redistribution of moments during a single leg forward hop. The plantar flexion moment at the ankle increased to the decreasing moments at the knee and hip, thus a greater reliance was placed on the ankle musculature. Conversely, Coventry et al (2006) reported increased range of motion and work at the hip, following a functional fatigue protocol of squats and landing, suggesting an increased reliance on the larger proximal muscles at the hip (Coventry et al., 2006). Although discrepancy in ankle mechanics in literature may be dependent upon the type of landing technique or task that is analyzed.

Quadriceps dominant strategy has also been found as a common neuromuscular strategy due to the onset of fatigue in jump landing (Kellis & Kouvelioti, 2007; Padua et al., 2006). As females have demonstrated this as a preferred landing strategy without the effects of fatigue (Chappell et al., 2007; Hewett et al., 1996; Huston & Wojtys, 1996; Sander TC et al., 2004; Zazulak et al., 2005), it raises concern for the female population. A high hamstrings to quadriceps ratio can cause direct strain on the ACL (Markolf et al., 1995; Myer GD et al., 2005). Due to quadriceps fatigue, Padua et al (2006) hypothesized the increased contraction on the quadriceps was due to a compensatory strategy in order to accomplish the hopping task (Padua et al., 2006). Similarly, Kellis et al (2007) reported increased quadriceps contraction in response to isolated quadriceps fatigue and isolated hamstrings fatigue (Kellis & Kouvelioti, 2007). Kellis et al (2007) theorized the increased quadriceps contraction in response to isolated hamstrings fatigue was due to an attempt for the bi-articulate quadriceps to control the knee and hip due to the impairment of the hamstrings.

Effects of Fatigue on Jump Landing Mechanics

As the function of musculature is compromised during fatigue, the likelihood and severity of potentially harmful landing mechanics and forces may increase the probability of injury. Fatigue may cause a decrease in shock absorption during landing. Forces placed on the body during landing are attenuated primarily through muscles because of their active contractile properties (Mizarhi and Susak 1992). Yet the onset of fatigue reduces that capacity of muscles fibers to absorb energy (Lepers, Hausswirth, Maffiuletti, Brisswalter, & van Hoecke, 2000). Voloshin et al (1998) and Mercer et al (2003) both reported less shock attenuation after the induction of fatigue at heel strike during fatigued running (Mercer, Bates, Dufek, & Hreljac, 2003; Voloshin A, Mizrani J, & Isakov E, 1998). It is hypothesized that the fatigued muscles are less effective in force absorption and caused increased, potentially dangerous, forces due to heel strike. An increase in forces during landing can cause increase stress through the kinetic chain and produce direct strain on the ACL (Lephart SM et al., 2002). Although other fatigue landing studies have reported no significant change in shock attenuation at heel strike for male athletes during running and single and double leg landing (Derrick et al 2002, Coventry et al 2006, Kernozek et al 2008, Orishimo et al 2006). Contrary, Kellis et al (2007) documented a significantly lower maximum GRF following a knee extensor fatigue protocol (Kellis & Kouvelioti, 2007). The greater attenuation in these studies can be attributed to the increased knee flexion causing a mechanical advantage for shock absorption (Coventry et al., 2006; Derrick, Dereu, & McLean, 2002; Kernozek et al., 2008; Madigan & Pidcoe, 2003; Orishimo KF & Kremenic IJ, 2006).

Knee joint landing kinematics after fatigue have also been reported with mixed results. Coventry et al (2006) reported an increase of knee flexion at initial contact post-fatigue during a maximal single-leg counter-movement jump. Male subjects performed a single leg fatigue, consisting of single leg landing, counter-movement jumps and squats (Coventry et al., 2006). Similarly, Kellis et al (2007) reported an increase in knee flexion at initial contact and peak knee flexion angles in a single leg landing subsequent to isolated single leg isokinetic hamstrings fatigue and isokinetic quadriceps fatigue (Kellis & Kouvelioti, 2007). Yet Benjaminse et al (2008) documented decreased knee flexion at initial contact from a single leg landing following a running fatigue protocol (Benjaminse et al., 2008). Increased knee flexion is a desirable landing position, as the hamstrings musculature can resist anterior tibial translation at higher knee flexion angles (G. Li et al., 1999; Markolf et al., 1995; Markolf et al., 1990). In a study by Kernozek et al (2008), both male and female subjects displayed increased anterior tibial translation force post-fatigue in comparison to baseline measures during a single leg landing. Yet males were more effective in resisting the maximum anterior shear force produced by reaching greater knee flexion (Kernozek et al., 2008).

Fatigue has also influenced knee kinematics in the frontal and transverse planes of motion. McLean et al (2007) induced fatigue on 10 male and female Division I athletes through a 4-minute general lower extremity fatigue consisting of step-downs and bounding drills. Pre-fatigue females displayed greater peak knee valgus angles and peak knee internal rotation during a 50 cm bilateral drop vertical jump, which is similar to other studies (Chappell et al., 2002; Cowley HR et al., 2006; Ford et al., 2003; Malinzak et al., 2001; Pappas E et al., 2007). With the onset of fatigue, both males and females peak knee valgus angles and knee internal rotation increased, while increases were statically more pronounced in females (McLean et al., 2007). Similarly, Nyland et al (1999) reported an increase in internal rotation at initial contact of a unilateral cutting maneuver following an isolated eccentric hamstring fatigue protocol (Nyland et al., 1999). Knee valgus has been reported as a common mechanism of injury to the ACL and predict ACL injuries with high sensitivity (Hewett et al., 2005). Similarly, internal rotation of the tibia has been shown to induce increased loading on the ACL (Kanamori A et al., 2002). These harmful stresses on the knee can be further exacerbated with unanticipated tasks, which is common in athletics. In a study by Borotikar et al (2008), female NCAA athletes endured a fatigue protocol consisting of single leg landings with cutting, forward double leg landings followed by maximum height jumps and double leg squats. Random series of anticipated and unanticipated of single leg landings were used for analysis, while the other tasks were intended to induce fatigue. Borotikar et al (2008) reported fatigue and decision making significantly increased peak knee valgus and knee internal rotation angles at peak stance of a single leg landing followed by a cutting motion (Borotikar et al., 2008). As both fatigue and decision making are synonymous with athletic competition, this may provide further insight into further risk for females for non-contact ACL injuries.

Similar to the knee, hip kinematics have also been affected by fatigue during athletic tasks. The hip musculature plays a primary role in the support and stabilization of the body during the absorption or contact phase of landing tasks (Decker et al., 2003; McNitt-Gray, 1993). Borotikar et al (2008) reported a more extended landing strategy of female athletes with less hip flexion at initial contact and increase of hip internal rotation. Authors suggested due to the squat fatigue protocol, athletes may have used a more upright strategy to insure landing during fatigue due to the compromised function of the hip and knee extensors. Yet this may place more strain on the non contractile structures of the knee by adopting a stiff strategy that limits the body's ability to absorb forces during landing (Lephart SM et al., 2002). Fatigue of the hip musculature can not only have an effect on the hip but also change other joints throughout the distal kinetic chain. As increased hip internal rotation at initial contact, has been correlated with peak knee abduction moments in landing activities (McLean, Huang, & van den Bogert, 2005). Likewise, Carcia et al (2005) documented increase knee valgus angles subsequent to an isometric fatigue on the hip abductor musculature (Carcia CR, Eggen JM, & Shultz SJ, 2005).

Hip flexion may also increase due to the onset of fatigue. Coventry et al (2006) reported an increase in hip flexion at initial contact and peak hip flexion angles. The subjects endure a lower extremity fatigue, consisting of single leg landings and single leg squats, and performed single leg landings pre and post fatigue. Coventry et al (2006) suggested that the onset of fatigue resulted in a greater dependence of the larger muscles in the lower extremity to decelerate the body at contact (Coventry et al., 2006). Due to inconsistency in fatigue studies, hip kinematics may be highly dependent upon the musculature that is fatigued during a chosen fatigue protocol. Kellis et al (2007) performed two separate isokinetic fatigue protocols of the hamstrings and quadriceps separately. The quadriceps fatigue resulted in increased maximum hip flexion values in contrast to the hamstring fatigue that did not affect hip flexion kinematics. They suggested that fatigue responses may differ corresponding to the musculature that becomes fatigued and the function, or agonist versus antagonist role, that the muscle acts on the joint (Kellis & Kouvelioti, 2007).

Summary

Jump landing has been identified as a primary mechanism of ACL injury, with females at greater risk. Fatigue, having a direct detrimental effect on muscular function, has shown to further affect the landing mechanics during jump landing tasks (Benjaminse et al., 2008; Borotikar et al., 2008; Chappell et al., 2005; Coventry et al., 2006; Kernozek et al., 2008; McLean et al., 2007; Orishimo KF & Kremenic IJ, 2006; Padua et al., 2006). Due to the dynamic stability the musculature around the knee provides, fatigue has been shown to modify neuromuscular strategies that may increase the stresses placed upon the ACL (Kellis & Kouvelioti, 2007; Padua et al., 2006). As the hamstrings act to assist in the stability of both the knee and hip and are aims of injury prevention programs (Chimera et al., 2004; Hewett et al., 1996; Wilkerson et al., 2004), it is important to understand how fatigue can affect their protective role (Aune et al., 1995; Baker BE, 1991; Barratta R et al., 1988; Dureselen L et al., 1995; MacWilliams et al., 1999;

Palmitier RA et al., 1991; Solomonow et al., 1987). Due to different tasks and types fatigue protocols, the effect of fatigue on injury still remains unclear.

CHAPTER III

METHODOLOGY

Design

An experimental design, of separate repeated measures analysis of variance, was used to determine the effects of fatigue on landing kinematics. Separate repeated-measures analyses of variance for each of the dependent variables to determine if instruction and fatigue had an influence on the variables tested. Knee and hip angles involved two with-in subject variables: time (pre-instruction/pre-fatigue, post-instruction/pre-fatigue, post-fatigue) and feedback group (self feedback, combination feedback, no feedback).

The control group, subjects who received no instruction, was used in the experimental design. The independent variable was time which has three levels: pre-instruction/pre-fatigue, post-instruction/pre-fatigue and post-fatigue. The post-instruction/pre-fatigue was chosen as pre-fatigue for analysis, as there was no significant difference between the two pre-fatigue times. The dependent variables were kinematic values, measured in degrees, of knee flexion, knee valgus, and hip flexion at maximum knee flexion and at initial contact. Maximum hip flexion was also a dependent variable.

Subject Characteristics

A sample size estimation was calculated using G* Power 3.0 for a minimal statistical power of 80% and p-value of .05. Sample size estimation from the different significant kinematic and kinetic variables ranged from 6 to 20 subjects, determining that 15 healthy female recreational athletes would be sufficient for this study (Decker et al., 2003; Ford et al., 2003; Kellis & Kouvelioti, 2007; Lephart SM et al., 2002). Female subjects were of interest as they are 2 to 9.7 times more likely to sustain an injury to their ACL in athletics ranging from basketball, soccer, handball, rugby, alpine skiing and military training (Arendt EA et al., 1999; Arendt EA & Dick R, 1995; Gwinn DE et al., 2000; Lindenfeld TN et al., 1994; Messina DF et al., 1999; Myklebust G et al., 1998; Stevenson H et al., 1998). Of the 15 subjects, 14 were classified as right leg dominant and 1 subject as left leg dominant. The left leg dominant subject was dropped from analyses to eliminate leg dominance as a covariate. This aids to eliminate leg dominance as a factor that may affect the outcomes of the results, as leg dominance was not a factor of interest. The 14 healthy female recreational athletes (age= 21.4 ± 1.45 yrs; height= 165.14 ± 8.26 cm; mass= 65.56 ± 16.43 kg) exercised a minimum of three times per week for a minimum of twenty minutes. They did not have a history of lower extremity injury in the past two months that limited them from activity for more than one day, no current self-reported history of lower extremity instability, or history of any lower extremity surgery within the past two years (Onate et al., 2005) . All subjects signed informed consent documents approved by the Old Dominion University Institutional Review Board.

Instrumentation

The Primus RS (BTE Technologies Inc, Hanover, MD) was used to induce isolated hamstring fatigue through a custom designed endurance fatigue protocol. The Primus RS measured the torque provided by each subject and numerical values were instantaneously visible on a computer-based software provided by Primus RS, operated on a Windows XP system (Microsoft Corp, Redman, WA). The Primus RS machine was calibrated prior to the start of data collection and completed on a bimonthly basis, as recommended by the company calibration protocol. The calibration protocol included a calibration tool attachment that was attached to the axis of rotation of the Primus RS and provided accurate calibration of position, force, and torque throughout testing.

The lower extremity kinematic data of the landing trials were collected with the VICON Motion System (VICON Motion Systems Ltd., Oxford England). The system consisted of eight cameras positioned in the laboratory at a sampling rate of 500 Hz. The reflective markers, 14mm in diameter, were placed on the specific body landmarks and were detected by at least two of the high speed cameras. The VICON Motion System has been found to be reliable, both with-in day and between day (Kadaba et al., 1989). To ensure high reliability during between day testing, marker placements should be standardized to minimize errors (Kadaba et al., 1989). Before each day of data collection, the VICON system was calibrated through a dynamic calibration by waving a 5-marker wand around the laboratory to fine-tune camera positions and a static calibration frame by placing a 5-marker wand in the right hand corner of the force plates to determine the volume origin.

Two Bertec 4060-NC (Bertec Corporation, Columbus, Ohio) forceplates were used to collect the ground reaction forces at a sampling rate of 500Hz. The forceplates were synchronized with the VICON motion analysis system and used to determine the point at which initial contact occurred. The two forceplates were secured in a wooden runway platform for a combined area of 60 cm x 80 cm. The analog signal was amplified and sent to an A to D board to be converted to a digital signal. Data reduction was performed with Visual 3D (C-Motion Inc.), where a kinematic model was built based on the subjects' static trial.

Box Drop Task

The box drop jump-landing task has been used by various researchers to replicate landings demonstrated in sports, such as basketball, soccer and volleyball (Cowley HR et al., 2006; Fagenbaum & Darling, 2003). The task started with the subject standing on a 30 cm high box with feet shoulder width apart and placed 30 cm back from the force plates (Cowley et al, 2006). Each subject was instructed to drop directly down off the box and land with one foot on each of the force plates. Upon foot contact and the loading phase of the landing, each subject subsequently jumped up for maximum vertical height while reaching both hands up (Cowley et al, 2006) and landed with each foot on a separate force plate. No verbal or visual reinforcement was provided during this task.

Testing Procedure

Subjects reported to testing wearing a sports bra or tight shirt, spandex shorts and athletic sneakers that they regularly used for exercise. Subjects completed a questionnaire to ensure subject inclusion and exclusion criteria before participating in any activity. Pre-testing measurements of height and weight were taken, and anthropometric measurements of leg length, knee width, and ankle width were also recorded.

Each subject was given ten minutes to warm up on a bike and self-directed stretch. During these ten minutes, the anthropometric measurements were inputted into the VICON motion system. Following the time allowed for stretch and warm-up, the reflective markers were placed on the subject by double-sided taped and reinforced with flexible tape.

The marker system consisted of sixteen reflective markers placed on bony landmarks of the body. The pelvis was defined by placing reflective markers on both anterior superior iliac crests and and posterior superior iliac crests. The lower extremity was defined by a series of reflective markers placed bilaterally. The lateral lower 1/3 surface of the thigh, the lateral epicondyle of the knee, the lower 1/3 of the shank, and the lateral malleolus along the imaginary line that passes through the transmalleolar axis were points of reflective marker placement. The feet bilaterally received reflective markers over the second metatarsal head, mid-foot side of the equines break between forefoot and mid-foot and on the calcaneous at the same height as the metatarsal marker.

Following the placement of all the markers, a static trial was taken of each subject. This required the subject to stand with each foot on the corresponding forceplate with arms crossed, ensuring all reflective markers were visible (Figure 1). Upon completion of the static trial, the box drop task was explained to each subject. The subjects were instructed to place their feet at the edge of the box, shoulder width apart and shift their weight forward to fall off the box (Figure 2a). The subjects were also instructed to land with their feet in the corresponding right and left force plate (Figure 2b) and, upon landing, jump straight up into the air to achieve maximal vertical jump height. After proper instruction of the box drop task, each subject was allowed three practice trials to become familiar with the box-drop task. Then five trials of the box-drop followed the familiarization for each subject. Once completed, the subjects were given ten minutes of rest with no instruction and performed five trials of the box drop task again. This second time instance of five trials was used as the pre-fatigue trials for analyses.

After performing the box-drop trials, the subjects completed an isolated hamstring fatigue protocol. Each subject was informed that upon completion of the fatigue protocol, they would be asked to attain another static trial and then immediately perform the same landing task.

Induction of Fatigue

Isolated fatigue was induced on the hamstrings on the dominant leg using a modified protocol initially designed by Kellis et al 2007. The dominant leg was operationally defined as the preferred leg to kick a ball (Orishimo KF & Kremenic IJ, 2006). The subject was positioned in a seated position with arms crossed and hip flexion at 90 degrees and stabilized at the thigh and chest with Velcro straps (Figure 3a). The axis of rotation was set at the posterior aspect of the lateral femoral condyle of the knee, the approximate center of rotation (Figure 3b) .

Maximum hamstring torque of each subject was determined from five eccentric maximal effort repetitions of the hamstrings (Figure 4). The maximal value was established for each subject to determine target levels of the fatigue. The isokinetic velocity was set at 120°/s and range of motion allowed at the knee was from 0 degrees to 90 degrees of knee flexion. The subjects were instructed to exert minimal or no effort during the concentric contraction and maximal effort during the eccentric contraction through the 90 degrees of range of motion.

During the fatigue testing, subjects performed continuous maximal effort repetitions until three continuous repetitions fall below 30% of the pre-determined peak eccentric hamstring torque (Kellis & Kouvelioti, 2007) (Figure 5). The subjects also received standardized verbal instruction of "pull, pull, pull, pull, pull" during each eccentric contraction to maintain maximal effort throughout testing. After fifteen consecutive repetitions, the Velcro straps were readjusted and the emphasis of effort on the eccentric contraction was reiterated to maintain proper positioning of the subject and reduce the effects of cardiovascular or central fatigue. Once a subject had two consecutive eccentric contractions under the 30% reduction of maximum torque, the instructions of maximum effort during the eccentric contraction were reiterated to ensure full effort of the subject. The average number of repetitions until fatigued was 86.93 ± 48.8 repetitions with time to completion lasting an average of 18.66 ± 10.14 minutes.

Upon completion of the fatigue protocol, the subject was unstrapped and was instructed to assume the static position on the force plates in the corresponding room. Each subject then performed five trials of the box jump task, lasting an average of 47.93 ± 18.64 seconds. The time pass between the completion of the fatigue protocol and the beginning of the first box drop post-fatigue was an average of 72.79 ± 18.39 seconds. This is in agreement with other fatigue studies as the average time of rest between fatigue protocol and task ranges between 90 to 120 seconds.

Data Analysis

Separate repeated measures analysis of variance were conducted for each one of the dependent variables, with time as the within subject effect. The dependent variables included maximum hip flexion, and knee flexion, knee valgus, and hip flexion at initial contact and maximum knee flexion. A significant level of < .05 was set a priori for all analyses. Means and standard deviations were calculated for all dependent measures. Statistical analyses were conducted in SPSS for Windows (version 16.0; SPSS Inc, Chicago, IL).

Figure 1: Static trial.

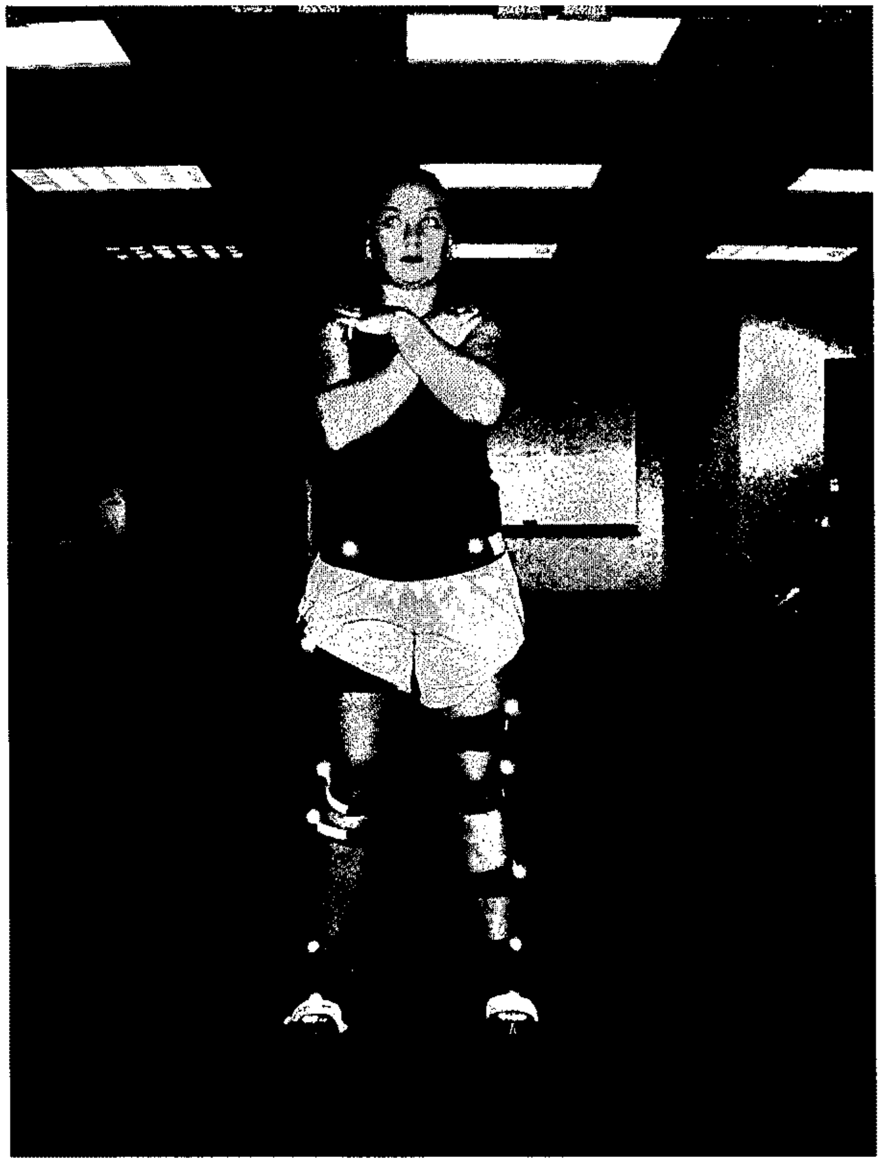
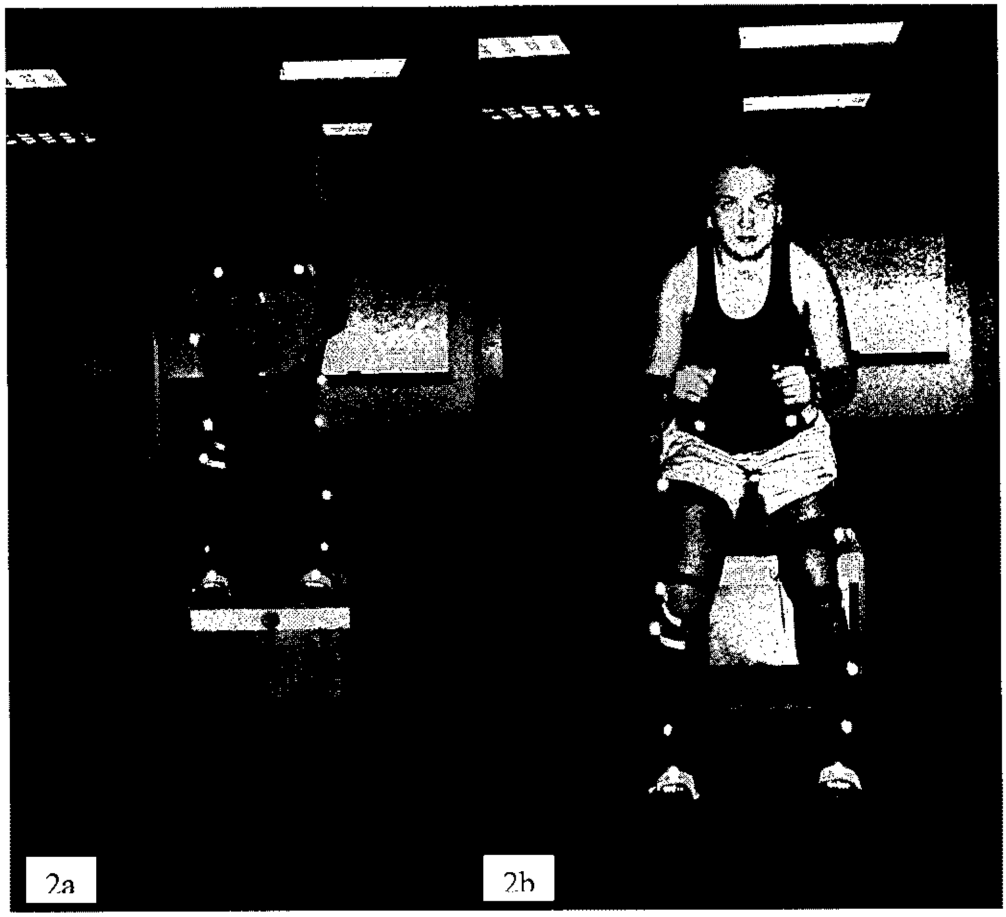


Figure 2: Box drop task.



2a. Subject on the box at the beginning of the box drop task.
2b. Subject on the force plates during the stop-jump phase.

Figure 3: Hamstring fatigue protocol.

3a. Frontal view
3b. Axis of rotation

Figure 4. Peak Eccentric Torque

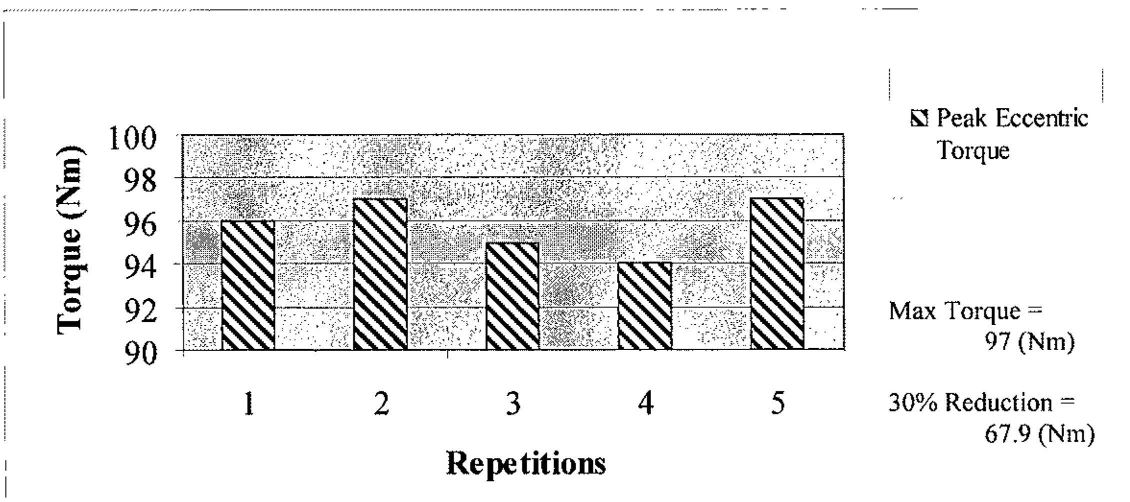
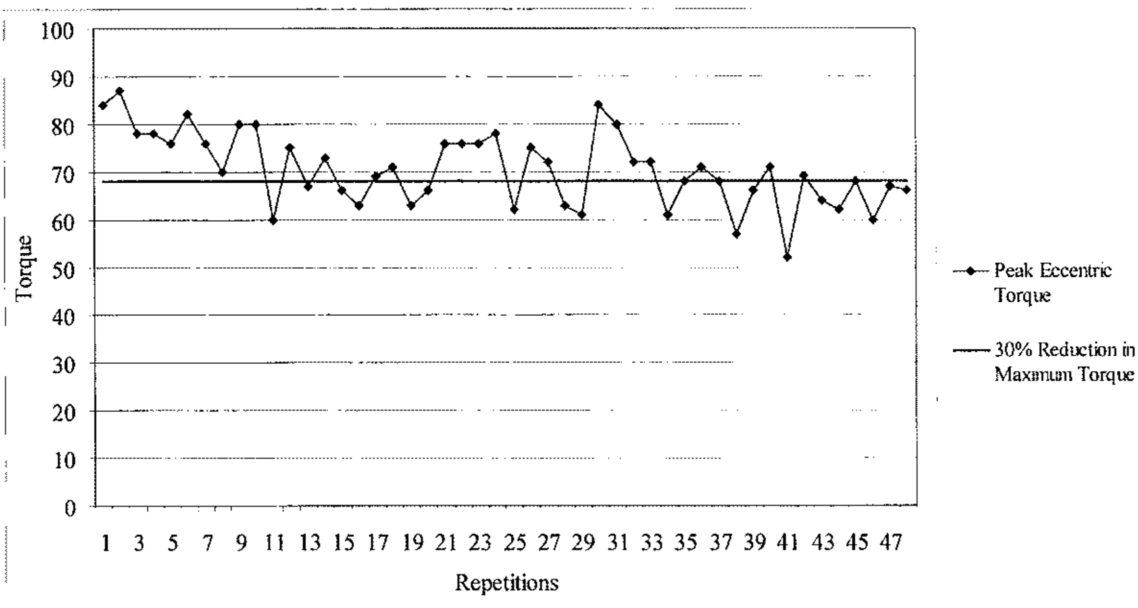


Figure 5: Example of subject peak eccentric torque decline during hamstring fatigue protocol.



CHAPTER IV

RESULTS

The means and standard deviations for all measures at initial contact are presented in Table 1. The means and standard deviations for all measures at maximum knee flexion are presented in Table 2. Finally, the means and standard deviations for maximum value measures are presented in Table 3.

Knee Flexion at Initial Contact

There was no statistical significant difference in knee flexion angles at initial contact due to fatigue ($F_{(2,13)}$ =.839, p=.444) (Table 1). There was no significant difference between pre (-22.25±9.2, CI= -27.58;-16.92) and post-fatigue (-23.15±1.1, CI= -29.6;-16.7) (Figure 6). The associated effect size was low d = .09. Therefore, the null hypothesis for knee flexion at initial contact was not rejected.

Knee Valgus at Initial Contact

There was no statistical difference in knee valgus angles at initial contact ($F_{(2,13)}$ =.701, p= .448) (Table 1). Subjects displayed no significant difference between pre (-2.44±4.09, CI= -4.8; -.07) and post- fatigue (-2.03±4.42, CI= -4.59; .519) (Figure 7). The associated effect size was low d= .09. The null hypothesis for knee valgus at initial contact was not rejected.

Hip Flexion at Initial Contact

There was a statistically significant difference in hip flexion angles at initial contact due to fatigue ($F_{(2,13)}$=4.226, p=.026) (Table 1). Subjects displayed significantly less hip flexion at initial contact pre-fatigue (40.68 ± 6.82, CI= 36.71; 44.62) than post-fatigue (43.78 ± 8.21, CI= 39.03; 48.52) (Figure 8). The associated effect size was moderate d= .38. The null hypothesis that there would be no significant difference post-fatigue, in hip flexion, was rejected.

Knee Valgus at Maximum Knee Flexion

Knee valgus at maximum knee flexion was not statistically significant due to fatigue ($F_{(2,13)}$ =.104, p=.902) (Table 2). There was no significance difference in knee valgus at maximum knee flexion between pre (2.29±8.35, CI= -2.52; 7.11) and post fatigue (2.51±8.78, CI= -2.55; 7.58) (Figure 9). The associated effect size was low d= .002. Therefore, the null hypothesis of maximum knee flexion was not rejected.

Hip Flexion at Maximum Knee Flexion

There was not a statistically significant difference in hip flexion at maximum knee flexion ($F_{(2,13)}$ = .910, p=.378) (Table 2). Subjects displayed no significant difference between pre (77.11± 11.11, CI= 70.69; 83.53) and post-fatigue (75.97± 1.14, CI= 69.35; 82.59) (Figure 10). The associated effect size was low d= .10. The null hypotheses postulating differences in hip flexion at maximum knee flexion due to fatigue was not rejected.

Maximum Knee Flexion

There was not a statistically significant difference in maximum knee flexion due to fatigue ($F_{(2,13)}$ =1.254, p=.302) (Table 3). There was no significant difference between pre (-81.71±11.89, CI= -88.57; -74.84) and post-fatigue (-80.86±11.67, CI= -87.6; -74.11) (Figure 11). The associated small effect size was d= .07. The null hypothesis that there would be no significant difference due to fatigue is not rejected.

Maximum Hip Flexion

Maximum hip flexion was not statistically significantly different due to fatigue ($F_{(2,13)}$=.118, p= .889) (Table 3). There was no significant difference in maximum hip flexion between pre (78.55± 11.22, CI= 72.07; 85.03) and post-fatigue (78.89± 1.08, CI= 72.61; 85.17) (Figure 12). The associated effect size was low d= .03. The hypothesis that there would be no significant difference due to fatigue at maximum hip flexion was not rejected.

Table 1: Results at initial contact for knee and hip kinematics pre- and post-fatigue (Mean± SD)

	Pre- Fatigue	Post-Fatigue	df[1]	*F*	*P*
Knee Flexion (°)	-22.25 ± 9.2	-23.15 ± 1.1	2,13	0.839	0.444
Knee Valgus (°)	-2.44 ± 4.09	-2.03 ± 4.42	2,13	0.701	0.448
Hip Flexion (°)	40.68 ± 6.82	43.78 ± 8.21	2,13	4.226	.026*

Knee flexion (+)
Knee extension (-)
Knee valgus (-)
Knee varus (+)
Hip flexion (+)
Hip extension (-)

* $p < .05$
[1] Degrees of freedom

Table 2: Results at maximum knee flexion for knee and hip kinematics pre- and post-fatigue (MEAN±SD).

	Pre- Fatigue	Post-Fatigue	*df*[1]	*F*	*P*
Knee Valgus (°)	2.29 ± 2.52	2.51 ± 8.78	2,13	0.104	0.902
Hip Flexion (°)	77.11 ± 11.1	75.97 ± 1.14	2,13	0.91	0.378

Knee valgus (-)
Knee varus (+)
Hip flexion (+)
Hip extension (-)

[1] Degrees of freedom

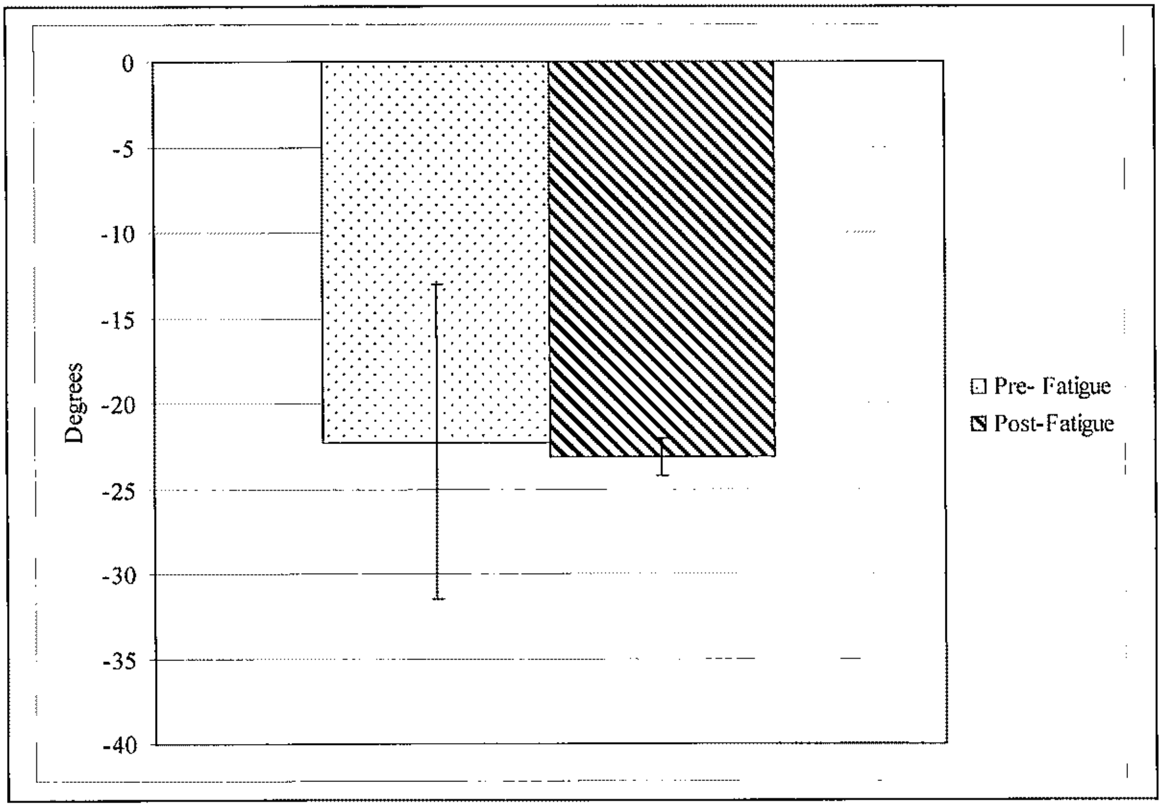
Table 3: Maximum results for knee and hip kinematics pre- and post-fatigue (MEAN±SD)

	Pre- Fatigue	Post-Fatigue	df[1]	*F*	*P*
Knee Flexion (°)	-81.71 ± 11.9	-80.86 ± 11.7	2, 13	1.254	0.302
Hip Flexion (°)	78.55 ± 11.1	78.89 ± 1.08	2, 13	0.118	0.889

Knee flexion (+)
Knee extension (-)
Knee valgus (-)
Knee varus (+)
Hip flexion (+)
Hip extension (-)

[1] Degrees of freedom

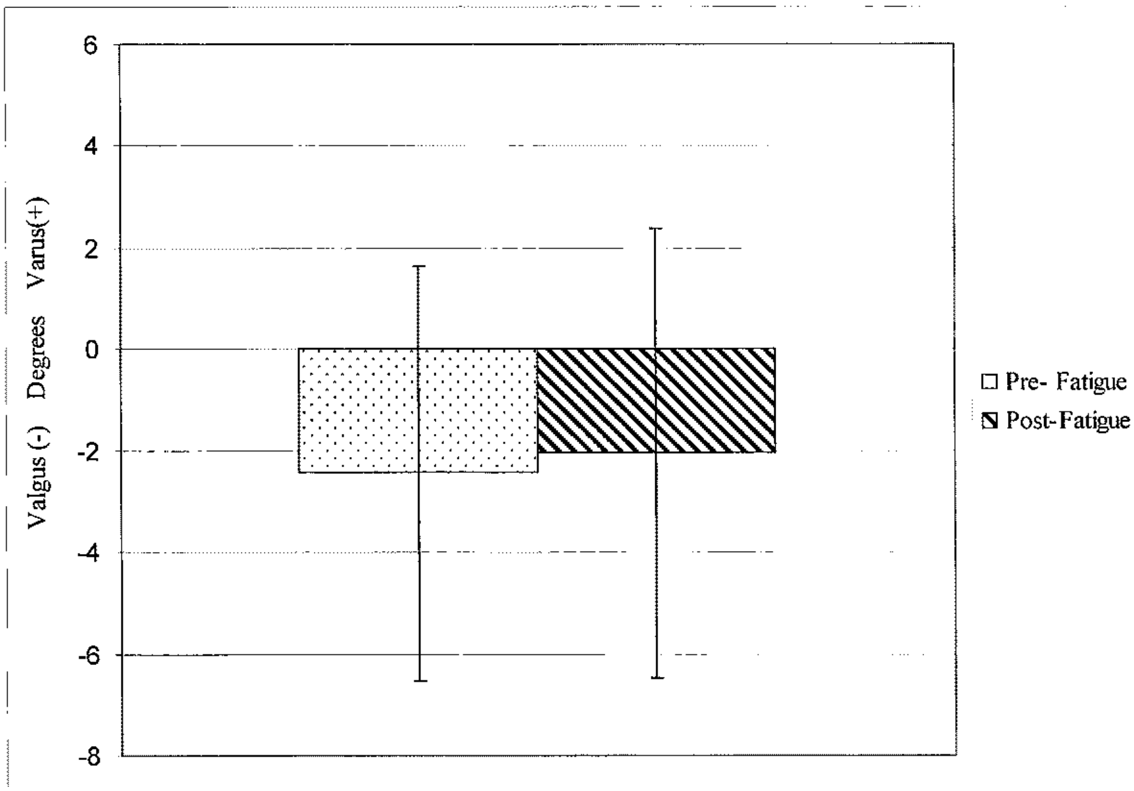
Figure 6. Knee flexion at initial contact.



Pre- Fatigue: (-22.25±9.2, CI= -27.58;-16.92)
Post-Fatigue: (-23.15±1.1, CI= -29.6;-16.7)

$F_{2,13}$ =.839, p =.444
d = .09

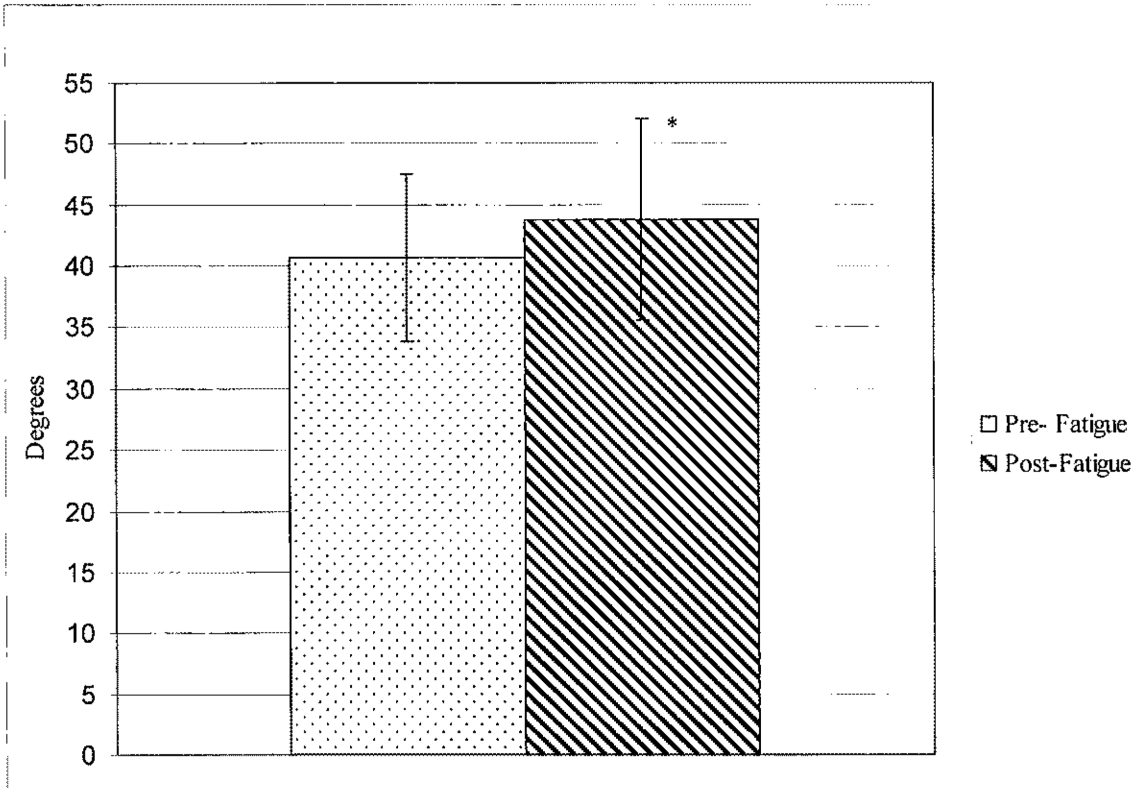
Figure 7. Knee valgus at initial contact.



Pre- Fatigue: (-2.44±4.09, CI= -4.8; -.07)
Post- Fatigue: (-2.03±4.42, CI= -4.59; .519)

$F_{2,13}$ =.701, p= .448
d = .09

Figure 8. Hip flexion at initial contact.

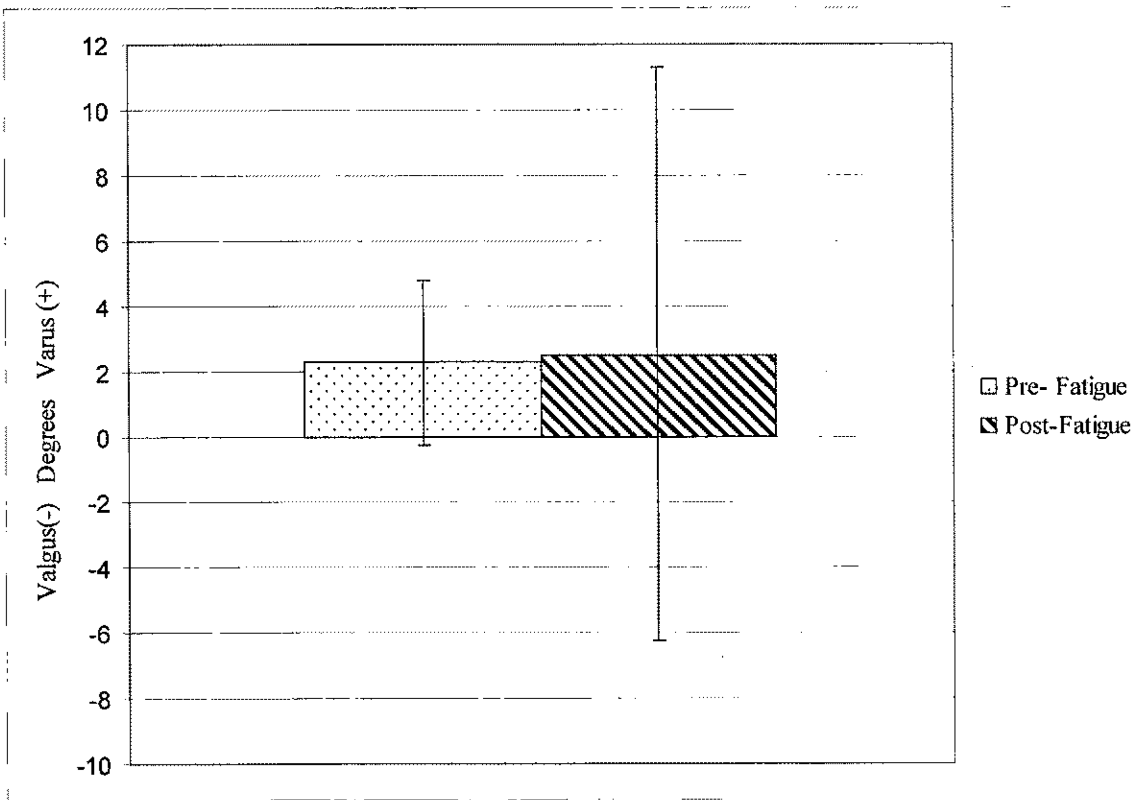


* $p \leq .05$

Pre- Fatigue: (40.68 ± 6.82, CI= 36.71; 44.62) $F_{2,13}$ =4.226, p=.026

Post- Fatigue: (43.78 ± 8.21, CI= 39.03; 48.52) d = .38

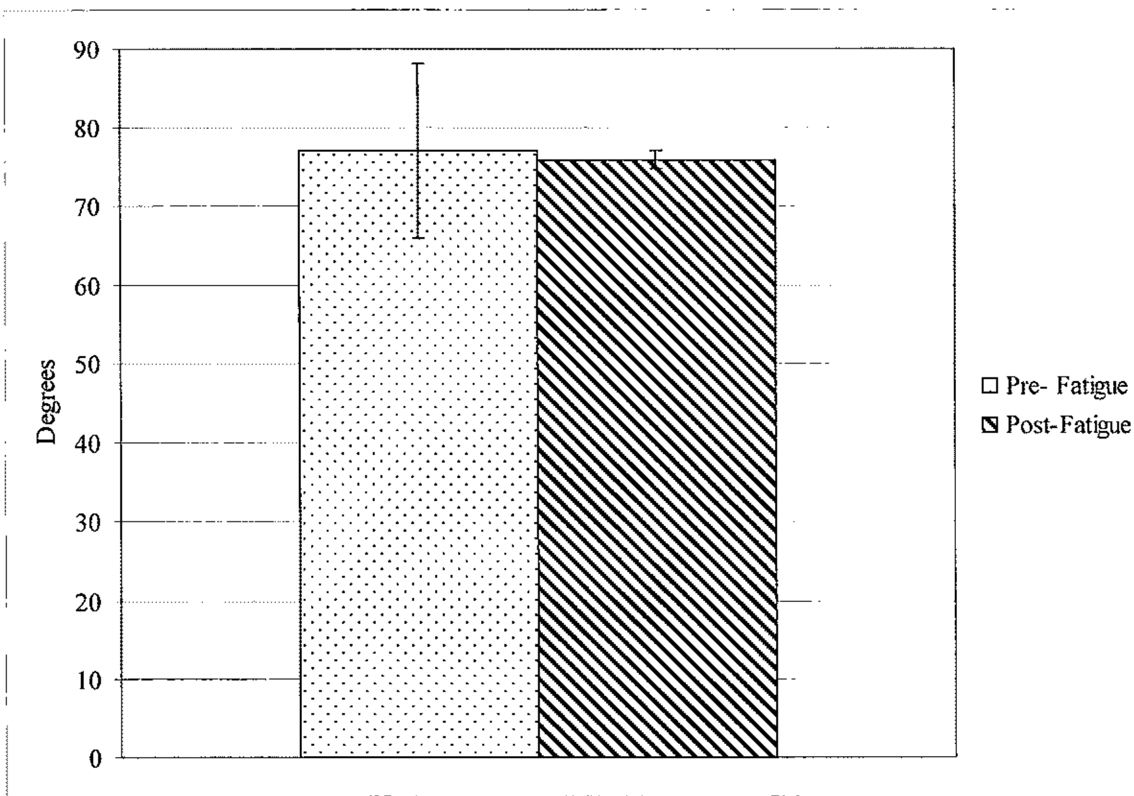
Figure 9. Knee valgus at maximum knee flexion.



Pre- Fatigue: (2.29±8.35, CI= -2.52; 7.11)
Post- Fatigue: (2.51±8.78, CI= -2.55; 7.58)

$F_{2,13}$ =.104, p=.902
d = .002

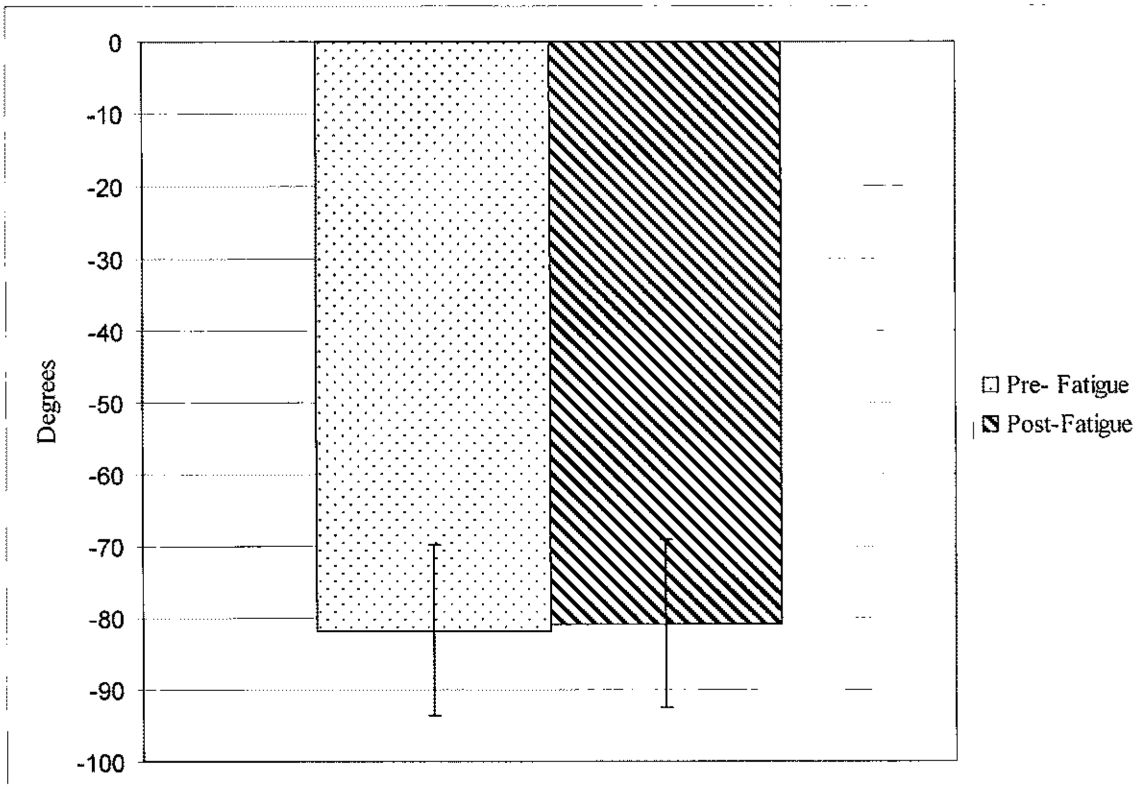
Figure 10. Hip flexion at maximum knee flexion.



Pre- Fatigue: (77.11± 11.11, CI= 70.69; 83.53)
Post- Fatigue: (75.97± 1.14, CI= 69.35; 82.59)

$F_{2,13} = .910, p=.378$
$d = .10$

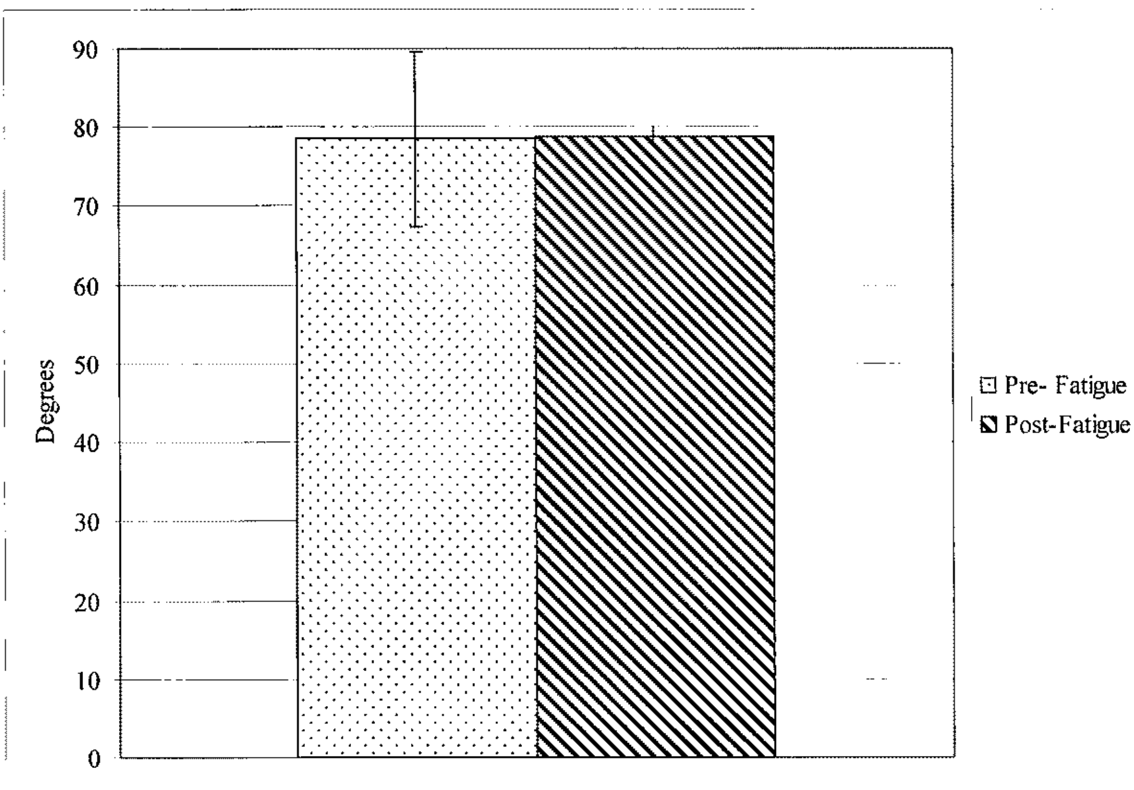
Figure 11. Maximum knee flexion.



Pre- Fatigue: (-81.71±11.89, CI= -88.57; -74.84)
Post- Fatigue: (-80.86±11.67, CI= -87.6; -74.11)

$F_{2,13}$ =1.254, p=.302
d= .07

Figure 12. Maximum hip flexion.



Pre- Fatigue: (78.55± 11.22, CI= 72.07; 85.03)
Post- Fatigue: (78.89± 1.08, CI= 72.61; 85.17)

$F_{2,13}$ =.118, p= .889
d= .03

CHAPTER V

DISCUSSION

The purpose of this study was to investigate the effects of isolated hamstring fatigue on maximum hip flexion, maximum knee flexion, knee valgus and hip flexion at maximum knee flexion and knee valgus, knee flexion and hip flexion at initial contact. We hypothesized that isolated hamstring fatigue would result in altered kinematics during a box drop task. The major finding indicated that isolated hamstring fatigue caused a significant increase in hip flexion at initial contact, while having no significant effect on maximum hip and knee flexion, knee flexion at initial contact, hip flexion at maximum knee flexion, and knee valgus at initial contact and maximum knee flexion.

Hip flexion at initial contact was hypothesized to increase following isolated hamstrings fatigue. It is important to understand the functions the hamstrings provide on the stability of the lower extremity. As the hamstrings play an important role in providing stability to the knee, the eccentric contraction of the hamstrings also provides stabilization to the hip (Palmitier RA et al., 1991) The function of the hamstrings serve as both hip extensors and knee flexors, as they are bi-articulate muscles (Seeley RR et al., 2003). Therefore it can be theorized if the hamstrings force production is reduced, due to fatigue, and the quadriceps ability to produce force is unaffected, mechanical efficiency of the quadriceps may have resulted in an increase in hip flexion. Padua et al (2006) reported an antagonist inhibition strategy of the hamstrings during a hopping task following a squat fatigue protocol. Padua et al (2006) concluded inhibition of the hamstrings may have represented a compensatory mechanism to maximize efficiency of the quadriceps necessary for the hopping task. As the current study fatigued the hamstrings, therefore deceasing force capabilities, it may have potentially increased the efficiency of the unfatigued quadriceps. The increased efficiency of the quadriceps, with decreased antagonist contribution at the hip joint may have resulted in increased hip flexion. However, electromyography (EMG) data is necessary to verify any neuromuscular recruitment strategies that may have occurred due to hamstring fatigue.

Increase of hip flexion at initial contact can also be theorized due to the neuromuscular quadriceps dominant strategy that females have shown in previous jump landing studies (Chappell et al., 2007; Hewett et al., 1996; Malinzak et al., 2001; Sander TC et al., 2004; Zazulak et al., 2005). The at risk neuromuscular quadriceps dominant strategy has been shown to become exacerbated with the addition of fatigue (Kellis & Kouvelioti, 2007; Padua et al., 2006). Similar to the hamstrings, the rectus femoris is a bi-articulate muscle, allowing movement at both the hip and knee (Seeley RR et al., 2003). Therefore a quadriceps dominant landing strategy in conjunction with decreased recruitment or reduced capabilities of the hamstrings, which serve as hip extensors, may result in increased hip flexion. Padua et al. (2006) recorded increased activation of the rectus femoris and vastus medialis in both genders during a continuous hopping task following a fatigue protocol consisting of squats, yet females demonstrated 2 to 4 times greater activation of the quadriceps in preparation to landing and immediately after contact respectively. Independent fatigue of both the quadriceps and hamstrings also resulted in increased activation of the quadriceps (Kellis & Kouvelioti, 2007). Fatigue of the quadriceps resulted in reduced activation of the biceps femoris and maintained quadriceps activation before and just after initial contact from a single leg landing from a 30cm box drop. Fatigue of the hamstrings, a protocol nearly identical to that used in this study, resulted in increased activation of the quadriceps prior to contact (Kellis & Kouvelioti, 2007). If EMG data was available in the current study, more objective data would be available to demonstrate the neuromuscular activation patterns.

The timing of the quadriceps activation may also provide insight into the significant increase of hip flexion at initial contact. As increased quadriceps activation was documented in studies by Padua et al (2005) and Kellis et al (2007), the enhanced activation occurred during the preparatory and initial loading phases of their tasks. Padua et al. (2006) operationally defined the preparatory phase as 50 milliseconds(ms) before initial contact and the initial loading phase as 50ms after initial contact (Padua et al., 2006). Kellis et al. (2007) operationally defined the preparatory phase as 100ms before initial contact and the initial loading phase as 100ms after initial contact (Kellis & Kouvelioti, 2007). The average time of initial contact for the subjects in this study post-fatigue was 43 ms; therefore it seems probable that subjects displayed greater hip flexion at initial contact due to the increased quadriceps activation. However, since we did not record EMG data it is difficult to draw a clear conclusion on neuromuscular activation.

Although researchers have demonstrated an increase in maximum hip flexion due to fatigue, the jump landing task may be a contributing factor for such outcomes. Researchers who have reported increased hip flexion have utilized single leg landing tasks for analysis (Coventry et al., 2006; Kellis & Kouvelioti, 2007; Kernozek et al., 2008), whereas the current study examined a double leg landing for analysis. Single leg landings utilize different ranges of motion at each joint to land successfully (Madigan & Pidcoe, 2003) and cannot gain as much range of motion at each joint due to the support of only one limb. Single leg landing tasks used in fatigue studies yield much smaller hip flexion angles in comparison to the current study both pre- and post-fatigue (Coventry et al., 2006; Kellis & Kouvelioti, 2007; Kernozek et al., 2008). Kernozek et al (2008) reported maximum hip flexion pre-fatigue mean was 40.72 degrees and post-fatigue was 48.02 degrees, while the current study resulted in 78.55 degrees pre-fatigue and 78.89 degrees post-fatigue of maximum hip flexion. The double leg landing used in the current study allowed subjects to gain greater maximum hip flexion pre-fatigue compared to the single leg task. To accomplish maximum jump height, subjects may have reached maximum hip flexion angles pre-fatigue. In accordance to McLean et al (2007), no significant increase in maximum hip flexion during a double leg box drop landing following a lower extremity fatigue protocol was recorded (McLean et al., 2007). Additionally, maximum hip flexion angles were approximately 16 degrees higher post-fatigue in the current study than maximum hip flexion angles post-fatigue in the study by McLean et al 2007. Therefore if subjects significantly increased hip flexion, their center of gravity may have been placed beyond their base of support and caused an unbalanced landing.

It can be theorized that different findings may have resulted post-fatigue if a single leg landings was chosen for analysis, as comparisons of previous studies reveal differences with different tasks. The double-leg landing was chosen as the task in the current study, as it is a task that is used in ACL research in both fatigue and non-fatigue jump landing studies. It also requires the subject to jump for vertical displacement subsequent to landing, as nearly all single leg landings only require subjects to land successfully with no subsequent jump. The double- leg landing requires greater exertion from the subjects and better replicates a game situation in sports, as athletes often land with a task followed. For example, an athlete may land from a soccer header but then immediately run in a given direction.

Similarly hip flexion at maximum knee flexion may not have significantly increased due to task and the fatigue protocol. The task following the double leg landing required the subjects to achieve a vertical jump, exerting maximum effort and power. As double leg landings have been found to have the greatest joint moment and power around the hip joint and the knee providing the greatest excursion (Decker et al., 2003; Zhang SN et al., 2000), maximum joint angles may have been reached to acquire optimal energy for the vertical jump pre-fatigue. A change in hip flexion at maximum knee flexion may alter muscle recruitment to decrease performance during a jump landing (Blackburn & Padua, 2008; Shimokochi, Yong Lee, Shultz, & Schmitz, 2009). Shimokochi et al (2009) and Blackburn et al (2008) suggest leaning forward during a land, by moving the center of pressure or gravity anteriorly, will decrease the demand of the quadriceps and increase or maintain the contraction demand of the hamstrings (Blackburn & Padua, 2008; Shimokochi et al., 2009). As hip flexion is in reference to the femur or joint center while trunk flexion is in reference to the vertical, greater hip flexion may yield comparable muscular demands. Although this is a protective landing strategy, subjects may not have increased hip flexion in order to limit greater reliance on the fatigued hamstrings. Also at the time instances of hip flexion at maximum knee flexion, the body is transitioning into the positive phase of the task (Rodacki, Fowler, & Bennett, 2001) The positive phase of the task begins once maximal knee joint flexion has been achieved and the body is converting efforts of deceleration to acceleration in the vertical direction (Rodacki, Fowler, & Bennett, 2002). At the time of maximum knee flexion, the body is preparing to propel for a vertical jump. Therefore an increase in hip flexion may decease the contraction demand on the quadriceps, the musculature group that propels the body during vertical jump (Rodacki et al., 2001). The reduced function of the hamstrings due to fatigue may have essentially maximized the efficiency of the quadriceps to produce the vertical jump (Padua et al., 2006), therefore not causing other compensatory movement patterns in the kinetic chain.

Knee flexion angle findings are in contrast with those of other authors who found significant differences between pre-fatigue and post-fatigue (Kellis & Kouvelioti, 2007; Madigan & Pidcoe, 2003; Orishimo KF & Kremenic IJ, 2006), whereas the current study found no significant difference. Researchers have suggested that fatigue causes an increase in knee flexion to reduce ground reaction forces at impact and increase absorption of forces that are placed on the body during landing (Coventry et al., 2006; Kellis & Kouvelioti, 2007; Madigan & Pidcoe, 2003; Orishimo KF & Kremenic IJ, 2006). Kellis et al (2007) found an increase in maximum knee flexion and knee flexion at initial contact after both independent hamstrings and quadriceps fatigues during a single leg drop landing (Kellis & Kouvelioti, 2007). It was hypothesized that due to the similar fatigue protocol used in the current study and that used by Kellis et al (2007), similar results would be yielded. Although Fagenbaum et al (2003) and Wikstrom et al (2004) also reported no significant differences in knee kinematics with fatigue during single leg landings (Fagenbaum & Darling, 2003; Wikstrom EA et al., 2004). Therefore the actual patterns of the knee joint angles following fatigue protocols can be difficult to compare when evaluating different jump landing tasks and fatigue tasks. Our study supports changes occurred in the proximal control of the jump-landing phase via increase in hip flexion, while knee flexion remained relatively unaltered due to isolated hamstring fatigue.

It may be argued that due to the quadriceps dominant strategy suggested in the current study, subjects would display less knee flexion at initial contact, as the quadriceps are knee extensors. Although given the subjects displayed low knee flexion angles pre-fatigue, 22.25 degrees, it is likely that subjects could not land successfully with even lower knee flexion angles. The subjects exhibited potentially at-risk knee flexion angles at initial contact patterns during pre-fatigue analysis, a common biomechanical mechanism theorized to increase risk of ACL injury when ranging from 20-25 degrees (Boden et al., 2000). Decker et al (2003) suggested the onset of fatigue can place greater risk of injury on individuals with erect position at initial contact (Decker et al., 2003). Although post-fatigue knee flexion at initial contact increased to 23.15 degrees, it was not significant and it is an erect landing with high risk association. It is also important to remember the quadriceps are contracting eccentrically to slow down the lowering the center of gravity, therefore the muscle is not shortening concentrically during the initial contact of the jump landing. It should be noted though that all of the subjects were female, most likely lending to lower knee flexion angles overall, nor did anyone have a previous history of ACL injury or incur an ACL injury during this study. Thus, knee flexion angle at initial contact patterns alone are most likely not the singular causative factor in ACL injury and are not affected by isolated hamstring fatigue; however, they may be affected by additional isolated lower extremity musculature fatigue (e.g., hip abductors) or overall cardiovascular fatigue protocols.

Knee valgus angles at initial contact and at peak knee flexion were predicted to increase due to the onset of the fatigue. Yet, unlike other fatigue studies that caused increases of knee valgus angles, the current study did not reveal any significant change in knee valgus angles (Benjaminse et al., 2008; Borotikar et al., 2008; McLean et al., 2007). During a similar task, McLean et al (2007) reported increased maximum knee valgus angles during a 50cm double leg box drop with a subsequent vertical jump. Yet the fatigue protocol utilized in the current study may have a direct impact on the knee valgus angles in the current study. McLean et al (2007) utilized a lower extremity fatigue protocol that consisted of step-ups and bounding drills, causing fatigue of the hamstrings in conjunction with the quadriceps, gluteal musculature, hip flexors and lower leg musculature. The fatigue protocol used in the current study only fatigued the hamstrings musculature with no involvement of the other hip musculature. As current literature has focused on the importance of the hip abductors in sagittal motion of the knee (Carcia CR et al., 2005; Jacobs et al., 2007), significant increases of knee valgus may not have been found due to the unaffected hip abductors. Jacobs et al (2007) attributes the weaker hip abductors to the increased knee valgus angles in females during a forward hop (Jacobs et al., 2007). Similarly, Carcia et al 2005 documented increased knee valgus angles subsequent to an isometric fatigue on the hip abductor musculature (Carcia CR et al., 2005).

Knee valgus at initial contact may not have significantly increased subsequent to fatigue due to knee flexion angles. As the subjects displayed small knee flexion angles at initial contact pre-fatigue and post-fatigue, the hamstrings may not have played a role during either condition at initial contact. In order for the hamstrings to play a protective role on the ACL, knee flexion angles must be above 30 degrees of flexion (G. Li et al., 1999; O'Connor, 1993). Therefore fatigue of the hamstrings may not have adversely decreased stability in the sagittal plane at the knee, if the hamstrings were not in a position to mechanically create a line of pull. Again, considering females naturally yield small knee flexion angles (Chappell et al., 2007; Chappell et al., 2002; Decker et al., 2003; Lephart SM et al., 2002; Malinzak et al., 2001; Salci Y et al., 2004; Yu B et al., 2006), different results may have resulted if the subject population had greater knee flexion angles. Subjects who have sustained neuromuscular or jump landing training to increase knee flexion angles (Chimera et al., 2004; Hewett et al., 1996; Wilkerson et al., 2004), thus increasing the recruitment of the hamstrings, significant differences may have been yielded post-fatigue.

Compensatory strategies have been reported due to the onset of fatigue and in an effort to accomplish a given task (Coventry et al., 2006; Kellis & Kouvelioti, 2007; Madigan & Pidcoe, 2003; Padua et al., 2006). As previous studies have found, fatigue responses may differ depending on musculature fatigued and the function it serves in the jump landing task (Kellis & Kouvelioti, 2007; Rodacki et al., 2002). Lower extremity fatigue protocols that utilize functional tasks, such as jumping, squats or step-ups, will fatigue all muscles involved in the task. The musculature that is affected by the fatigue protocol has been shown to re-distribute the work to other less fatigued musculature in the lower extremity kinetic chain to accomplish the task. Coventry et al (2006) documented a shift or redistribution of work toward greater dependence on the less fatigued ankle musculature (Coventry et al., 2006). Similarly, Madigan et al (2003), attributed a distal to proximal redistribution of extensor moment production subsequent to plantar flexor fatigue during a 25cm single leg landing(Madigan & Pidcoe, 2003). The redistribution of work or control of the body during landing may have been redistributed to the proximal hip musculature (e.g., hip abductors or external rotators), as they play an important role in knee control. The redistribution may have also been placed on the ankle musculature. Decker et al (2002) found reduced hamstrings strength in ACL reconstructed female athletes applied an adaptive landing strategy that utilized more ankle work in landing and less reliance on the hip extensor musculature, or hamstrings (Decker, Torry, Noonan, Riviere, & Sterett, 2002). Ankle kinematics were not chosen for analysis as the knee and hip are more commonly areas of interest for ACL research, although a recent shift in focus has been directed toward the ankle joint as insight into ACL injury (Hewett, Torg, & Boden, 2009). Although ankle kinematics were not measured in this study, the redistribution of muscular reliance should be considered as a possibility and area for future research. Future isolation and combined protocols of the various musculature and their effects on landing motion patterns should be investigated in the effort of increasing ACL injury prevention program effectiveness.

Conclusion

In conclusion, hamstring fatigue showed a significant increase of hip flexion at initial contact, but failed to show other altered kinematic movement patterns. We theorized that female subjects displayed greater dependence upon musculature unaffected by the hamstring fatigue protocol, thus displaying little difference between pre- and post-fatigue. Also, subjects utilized joint motions pre-fatigue that have been shown to mechanically limit the involvement of the hamstrings, therefore fatigue of only the hamstring musculature may not have affected subjects' natural landing. Future studies should include various subject populations with different natural landing patterns to further understand the effects of hamstring fatigue on jump landing.

The results of the current study may also provide further insight into the need of consistency among fatigue jump landing studies. It is important to analyze the effects of fatigue to various muscles or muscle groups, as the body's subsequent movement patterns are likely to change. Similarly, different jump landing tasks are of interest to replicate different demands seen in athletics. However some consistency must be maintained to analyze and compare different fatigue jump landing studies, in order to draw any predictable movement patterns that may lead us closer to preventing injury. With such discrepancies among studies, even the operational definition of fatigue, it is difficult to draw any predictable movement patterns that may lead us closer to preventing injury.

Limitations

The evaluation of isolated hamstrings fatigue protocols on lower extremity kinematic patterns is quite limited in this study. The comparison of various other kinematic features of the landing pattern (e.g., ankle position, trunk movement, and hip internal rotation) and individual analyses to show significant differences by subject would provide increased understanding on the effects of isolated hamstring fatigue on the distal kinetic chain. The number of variables evaluated in this study was limited by the data reduction time, and amount of information that could be provided in one research project. The information to evaluate other kinematic and kinetic variables remains to be analyzed, but the scope of this initial study report was to focus on a few of the most important kinematic factors generally considered to be important to the biomechanical causative mechanisms of ACL injury. Additionally, the current study did not collect

EMG data during the jump landing task to evaluate muscular activation patterns relative to the hamstring fatigue protocol. Future studies should include EMG information to provide more insight into the neuromuscular responses of isolated hamstrings fatigue on jump-landing motion. The maximum height jump subsequent to contact was also not recorded, as a matter of convenience, to insure subjects reached maximum distance, therefore future jump landing studies should utilize an objective tool to ensure maximum height is reached and provide objective data of jump height. Perhaps the most important limitation in evaluating the movement patterns relative to ACL injury are the lack of ecological validity of the task and also the lack of injury occurrence during movement pattern testing. None of the tasks currently evaluated in the biomechanical literature actually replicate all of the ecological aspects of ACL injury (e.g., surface, decision-making, opponent, etc…). The box-drop task is a controlled task that reflects a jump-landing motion pattern, but lacks ecological validity relative to an actual practice/game condition. Future experimental designs and equipment should be developed to further understand the actual movement patterns that occur in the field. Additionally, it must be noted that the conditions utilized to evaluate potential at-risk movement patterns did not result in actual ACL injury occurrence. Due to the unethical aspects of creating an injurious situation, it should be noted most biomechanical studies of ACL at-risk movement patterns are actually evaluated in individuals who have never injured their ACL and never will. This is an unanswerable dilemma since researchers do not wish to create an injury, but it also indicates that perhaps the movement tasks are not demanding enough to truly impose at-risk movement patterns and/or the populations studied are not actually at-risk for injuring their ACL's. Prospective studies utilizing pre-participation screening processes in the most at-risk individuals (e.g., collegiate female soccer athletes) and demanding movement tasks under various fatigue conditions should be instituted to allow evaluation of individuals across a time history to track movement patterns associated with ACL injury occurrence.

VITA

Kristen Anne Phillips

EDUCATION

2007- Current **Old Dominion University**
M.S.Ed in Athletic Training
Thesis: *Effects of Isolated Hamstring Fatigue on Jump Landing in Healthy Recreational Athletes*

2003- 2007 **Catawba College**
Summa Cum Laude
B.S. in Athletic Training
B.S. in Physical Education
Minor in Special Subjects Areas Education

PROFESSIONAL EXPERIENCE

August 2007- Current **Graduate Asst. Athletic Trainer**
Norfolk State University
Primary Sport Assignments: Womens Basketball, Football

August 2003- May 2007 **Athletic Training Student**
Catawba College
Primary Sport Assignments: Football, Baseball, Womens Basketball, Mens Lacrosse

TEACHING EXPERIENCE

January 2009- Current **Instructor**
Old Dominion University
Advanced First Aid and Emergency Care

AWARDS/HONORS

2003- 2007 First Family Scholarship
2003- 2007 Catawba College Dean's List
Whitley Physical Education Award
Junior Marshall

PROFESSIONAL PRESENTATIONS

Phillips KA, Yacobino NE. *Stress Fracture of the Medial Epicondyle after Jobe Reconstruction: A Case Report.* Student Presentation- Mid Atlantic Trainers Association Symposium; May 2007.

REFERENCES

Aagaard, P., Simonsen, E. B., Magnusson, S. P., Larsson, B., & Dyhre-Poulsen, P. (1998). A new concept for isokinetic hamstring: quadriceps muscle strength ratio. *Am J Sports Med, 26*(2), 231-237.

Aagaard, P., Simonsen, E. B., Trolle, M., Bangsbo, J., & Klausen, K. (1995). Isokinetic hamstring/quadriceps strength ratio: influence from joint angular velocity, gravity correction and contraction mode. *Acta Physiol Scand, 154*(4), 421-427.

Adachi N, Nawata K, Maeta M, & Y, K. (2007). Relationship of the menstrual cycle phase to anterior cruciate ligament injuries in teenaged female athletes. *Arch Orthop Trauma Surg*, Epub ahead of print.

Allen, T. J., & Proske, U. (2006). Effect of muscle fatigue on the sense of limb position and movement. *Exp Brain Res, 170*(1), 30-38.

Arendt EA, Agel J, & R, D. (1999). Anterior cruciate ligament injury patterns among collegiate men and women. *J Athl Train, 34*(2), 86-92.

Arendt EA, & Dick R. (1995). Knee injury patterns among men and women in collegiate basketball and soccer. *Am J Sports Med, 23*(6), 694-701.

Asmussen, E., & Mazin, B. (1978a). A central nervous component in local muscular fatigue. *Eur J Appl Physiol Occup Physiol, 38*(1), 9-15.

Asmussen, E., & Mazin, B. (1978b). Recuperation after muscular fatigue by "diverting activities". *Eur J Appl Physiol Occup Physiol, 38*(1), 1-7.

Aune, A. K., Nordsletten, L., Skjeldal, S., Madsen, J. E., & Ekeland, A. (1995). Hamstrings and gastrocnemius co-contraction protects the anterior cruciate ligament against failure: an in vivo study in the rat. *J Orthop Res, 13*(1), 147-150.

Baker BE. (1991). Prevention of ligament injuries to the knee. *Exerc SPort Sci Rev, 18*, 291-305.

Ball KA. (1999). Lack of hip flexion: a mechanical for ACL injury. *Med Sci Sports Exerc, 31*, S295.

Barratta R, Solomonow M, Zhou BH, Lestion D, Chinard R, & D'Ambrosia R. (1988). Muscular coactivation: the role of anatogonist musculature in maintaining knee stability. . *Am J Sports Med, 16*, 113-122.

Beckett ME, Massie DL, Bowers KD, & Stoll DA. (1992). Incidence of hyperpronation in the ACL injured knee: a clinical perspective. *J Athl Train 27*, 58-62.

Bendijaballah MZ, Shirazi-Adl A, & Zukor DJ. (1997). Finite element analysis of human knee joint in valgus-varus.. *Clin Biomech, 12*, 139-148.

Benjaminse, A., Habu, A., Sell, T. C., Abt, J. P., Fu, F. H., Myers, J. B., et al. (2008). Fatigue alters lower extremity kinematics during a single-leg stop-jump task. *Knee Surg Sports Traumatol Arthrosc, 16*(4), 400-407.

Blackburn, J. T., & Padua, D. A. (2008). Influence of trunk flexion on hip and knee joint kinematics during a controlled drop landing. *Clin Biomech (Bristol, Avon), 23*(3), 313-319.

Bobbert MF, & van Zandwijk JP. (1999). Dynamics of force and muscle stimulation in human vertical jumping. *Med Sci SPorts Exerc, 31*, 303-310.

Boden, B. P., Dean, G. S., Feagin, J. A., Jr., & Garrett, W. E., Jr. (2000). Mechanisms of anterior cruciate ligament injury. *Orthopedics, 23*(6), 573-578.

Bonci CM. (1999). Assessment and evaluation of predisposing factors to anterior cruciate ligament injury. *J Athl Train, 34*(2), 155-164.

Borotikar, B. S., Newcomer, R., Koppes, R., & McLean, S. G. (2008). Combined effects of fatigue and decision making on female lower limb landing postures: central and peripheral contributions to ACL injury risk. *Clin Biomech (Bristol, Avon), 23*(1), 81-92.

Boyd, I. A. (1954). The histological structure of the receptors in the knee-joint of the cat correlated with their physiological response. *J Physiol, 124*(3), 476-488.

Butler DL, Noyes FR, & ES, G. (1980). Ligamentous restraints to anterior-posterior drawer in the human knee. A biomechanical study. *J Bone Joint Surg [Am], 62*, 259-270.

Cairns SP, Knicker AJ, Thompson MW, & Sjogaard G. (2005). Evaluation of models used to study neuromuscular fatigue. *Exerc SPort Sci Rev, 33*(1), 9-16.

Carcia CR, Eggen JM, & Shultz SJ. (2005). Hip-abductor fatigue, frontal-plane landing angles, and excursion during a drop-jump. *J Sport Rehab, 14*(4), 321-331.

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

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

Chappell, Herman, Knight, Kirkendall, Garrett, & Yu. (2005). Effect of fatigue on knee kinetics and kinematics in stop-jump tasks. *Am J Sports Med, 33*(7), 1022-1029.

Chappell, Yu, Kirkendall, & Garrett. (2002). A comparison of knee kinetcis between male and female recreational athletes in stop-jump tasks. *Am J Sports Med, 30*(2), 261-267.

Chhabra A, Elliott CC, & MD, M. (2001). Normal anatomy and biomechanics of the knee. *Sport Medicine and Arthroscopy Review, 9*, 166-177.

Chimera, N. J., Swanik, K. A., Swanik, C. B., & Straub, S. J. (2004). Effects of Plyometric Training on Muscle-Activation Strategies and Performance in Female Athletes. *J Athl Train, 39*(1), 24-31.

Chu, D., LeBlanc, R., D'Ambrosia, P., D'Ambrosia, R., Baratta, R. V., & Solomonow, M. (2003). Neuromuscular disorder in response to anterior cruciate ligament creep. *Clin Biomech (Bristol, Avon), 18*(3), 222-230.

Clanton, T. O. (1998). Hamstring strains in athletes; diagnosis and treatment. *J Am Acad Orthop Surg, 6*(4), 237.

Clark BC, Collier SR, Manini TM, & Ploutz-Snyder LL. (2005). Sex differences in muscle fatigability and activation patterns of the human quadriceps femoris. *Eur J Appl Physiol, 94*, 196-206.

Coventry, E., O'Connor, K. M., Hart, B. A., Earl, J. E., & Ebersole, K. T. (2006). The effect of lower extremity fatigue on shock attenuation during single-leg landing. *Clin Biomech, 21*, 1090-1097.

Cowley HR, Ford KR, Myer GD, Kernozek TW, & Hewett TE. (2006). Differences in neuromuscular strategies between landing and cutting tasks in female basketball and soccer athletes. *J Athl Train, 41*(1), 67-73.

Decker, M. J., Torry, M. R., Noonan, T. J., Riviere, A., & Sterett, W. I. (2002). Landing adaptations after ACL reconstruction. *Med Sci Sports Exerc, 34*(9), 1408-1413.

Decker, M. J., Torry, M. R., Wyland, D. J., Sterett, W. I., & Richard Steadman, J. (2003). Gender differences in lower extremity kinematics, kinetics and energy absorption during landing. *Clin Biomech (Bristol, Avon), 18*(7), 662-669.

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

Derrick, T. R., Dereu, D., & McLean, S. P. (2002). Impacts and kinematic adjustments during an exhaustive run. *Med Sci Sports Exerc, 34*(6), 998-1002.

Dureselen L, Claes L, & H, K. (1995). The influence of muscle forces and external loads on cruciate ligament strain. *Am J Sports Med, 23*, 129-136.

Dyhre-Poulsen, P., & Krogsgaard, M. R. (2000). Muscular reflexes elicited by electrical stimulation of the anterior cruciate ligament in humans. *J Appl Physiol, 89*(6), 2191-2195.

Edwards, R. H. (1981). Human muscle function and fatigue. *Ciba Found Symp, 82*, 1-18.

Edwards RHT, Hill DK, Jones DA, & Merton PA. (1977). Fatigue of long duration in human skeletal muscle after exercise. *J Physiol Lond, 272*, 769-778.

Ellis H. (2007). The applied anatomy of examination of the knee. *British Journal of Hospital Medicine, 68*(4), M60-61.

Fagenbaum, R., & Darling, W. G. (2003). Jump landing strategies in male and female college athletes and the implications of such strategies for anterior cruciate ligament injury. *Am J Sports Med, 31*(2), 233-240.

Feagin JA. (1988). Isolated anterior cruciate injury. In Feagin JA (Ed.), *The Cruciate Ligaments* (pp. 15-23). New York: Churchill Livingstone.

Feagin JA, & Lambert KL. (1985). Mechanism of injury and pathology of anterior cruciate ligament injuries. *Orthop Clin North Am, 16*, 41-45.

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

Fitts, R. H. (1994). Cellular mechanisms of muscle fatigue. *Physiol Rev, 74*(1), 49-94.

Fitts, R. H. (2008). The cross-bridge cycle and skeletal muscle fatigue. *J Appl Physiol, 104*(2), 551-558.

Fleming BC, Renstrom PA, Beynnon BD, Engstrom B, Peura GD, Badger GJ, et al. (2001). The effect of weightbearing and external loading on anterior cruciate ligament strain. *J Biomech, 34*, 163-170.

Ford, K. R., Myer, G. D., & Hewett, T. E. (2003). Valgus knee motion during landing in high school female and male basketball players. *Med Sci Sports Exerc, 35*(10), 1745-1750.

Frank CB, & Jackson DW. (1997). Current concepts review- The science of reconstruction of the anterior cruciate ligament. *J Bone Joint Surg [Am], 79*, 1556-1576.

Friemert, B., Bumann-Melnyk, M., Faist, M., Schwarz, W., Gerngross, H., & Claes, L. (2005). Differentiation of hamstring short latency versus medium latency responses after tibia translation. *Exp Brain Res, 160*(1), 1-9.

Fu FH, & Stone DA. (1994). *Sports Injuries: Mechanisms, Prevention, Treatment.* Philadelphia: Williams & Wilkins.

Fujita, I., Nishikawa, T., Kambic, H. E., Andrish, J. T., & Grabiner, M. D. (2000). Characterization of hamstring reflexes during anterior cruciate ligament disruption: in vivo results from a goat model. *J Orthop Res, 18*(2), 183-189.

Fukubayshi T, Torzilli PA, Sherman MF, & Warren RF. (1982). An in Vitro biomechanical evaluation of anterior-posterior motion of the knee *J Bone Joint Surg 64*(2), 258-264.

Gabbett TJ. (2000). Incidence, site and nature of injruies in amateur rugby league over three consecutive seasons. *Br J Sports Med, 34*(2), 98-103.

Gabbett TJ. (2002). Incidence of injury in amateur rugby league sevens. *Br J Sports Med, 36*(1), 23-26.

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

Gandevia, S. C., Allen, G. M., Butler, J. E., & Taylor, J. L. (1996). Supraspinal factors in human muscle fatigue: evidence for suboptimal output from the motor cortex. *J Physiol, 490 (Pt 2)*, 529-536.

Gandevia SG, & McKenzie DK. (1988). Activation of human muscles at short muscle lengths during maximal static efforts. *J Physiol Lond, 407*, 599-613.

Gehring, D., Melnyk, M., & Gollhofer, A. (2009). Gender and fatigue have influence on knee joint control strategies during landing. *Clin Biomech (Bristol, Avon), 24*(1), 82-87.

Gibson, H., & Edwards, R. H. (1985). Muscular exercise and fatigue. *Sports Med, 2*(2), 120-132.

Givoni NJ, Pham T, Allen TJ, & Proske U. (2007). The effect of quadriceps muscle fatigue on position matching at the knee. *J Physiol, 584*(1), 111-119.

Gregoire L, Veeger HE, Huijing PA, & GJ, v. I. S. (1991). Dynamics of force and muscle stimulation in human vertical jumping. *Med Sci SPorts Exerc, 31*, 303-310.

Griffin LY, Agel J, Albohm MJ, Arendt EA, Dick RW, Garrett WE, et al. (2000). Noncontact anterior curciate ligament injuries: risk factors and prevention strategies. *J Am Acad Orthop Surg, 8*(3), 141-150.

Gwinn DE, Wilckens JH, McDevitt ER, Rooss G, & Kao TC. (2000). The relative incidence of anterior cruciate ligament injury in men and women at the United States Naval Academy. *Am J Sports Med, 28*, 98-102.

Hakkinen, K., & Komi, P. V. (1986). Effects of fatigue and recovery on electromyographic and isometric force- and relaxation-time characteristics of human skeletal muscle. *Eur J Appl Physiol Occup Physiol, 55*(6), 588-596.

Hawkins RD, & Fuller CW. (1999). A prospective epidemiological study of injuries in four English professional football clubs. *Br J Sports Med, 33*, 196-203.

Heiser TM, Weber J, Sullivan G, Clare P, & Jacobs RR. (1984). Prophylaxis and management of hamstring muscle injuries in intercollegiate football players. *Am J Sports Med, 12*(5), 368-370.

Herbert, R. D., & Gandevia, S. C. (1996). Muscle activation in unilateral and bilateral efforts assessed by motor nerve and cortical stimulation. *J Appl Physiol, 80*(4), 1351-1356.

Hewett, T. E., Myer, G. D., & Ford, K. R. (2005). Biomechanical measures of neuromuscular control and valgus loading of the knee predict ACL injury risk in female athletes: a prospective study. *Am J Sports Med, 33*, 492-501.

Hewett, T. E., Stroupe, A. L., Nance, T. A., & Noyes, F. R. (1996). Plyometric training in female athletes. Decreased impact forces and increased hamstring torques. *Am J Sports Med, 24*(6), 765-773.

Hewett, T. E., Torg, J. S., & Boden, B. P. (2009). Video Analysis of Trunk and Knee Motion during Non-Contact ACL injury in Female Athletes: Lateral Trunk and Knee Abduction Motion are Combined Components of the Injury Mechanism. *Br J Sports Med.*

Holm, I., Fosdahl, M. A., Friis, A., Risberg, M. A., Myklebust, G., & Steen, H. (2004). Effect of neuromuscular training on proprioception, balance, muscle strength, and lower limb function in female team handball players. *Clin J Sport Med, 14*(2), 88-94.

Horita, T., Komi, P. V., Nicol, C., & Kyrolainen, H. (1996). Stretch shortening cycle fatigue: interactions among joint stiffness, reflex, and muscle mechanical performance in the drop jump [corrected]. *Eur J Appl Physiol Occup Physiol, 73*(5), 393-403.

Huston, L. J., Vibert, B., Ashton-Miller, J. A., & Wojtys, E. M. (2001). Gender differences in knee angle when landing from a drop-jump. *Am J Knee Surg, 14*(4), 215-219; discussion 219-220.

Huston, L. J., & Wojtys, E. M. (1996). Neuromuscular performance characteristics in elite female athletes. *Clin Ortho Rel Res, 371*, 50-63.

Hutchinson MR, & Ireland ML. (1995). Knee injuries in female athletes. *Sports Med, 19*, 288-302.

Imran A, & O'Connor JJ. (1997). Theoretical estimates of cruciate ligaments forces: effects of tibial surface geometry and ligament orientations. . *Proc Inst Mech Eng [H], 211*, 425-439.

Ireland ML. (1999). Anterior cruciate ligament injury in female athelets: epidemiology. *J Athl Train, 34*(2), 150-154.

Ireland ML, Gaudette M, & Crook S. (1997). ACL injuries in the female athletes. *J Sport Rehab, 6*(2), 97-110.

Jacobs, C. A., Uhl, T. L., Mattacola, C. G., Shapiro, R., & Rayens, W. S. (2007). Hip abductor function and lower extremity landing kinematics: sex differences. *J Athl Train, 42*(1), 76-83.

Johansson, H., Sjolander, P., & Sojka, P. (1991). Receptors in the knee joint ligaments and their role in the biomechanics of the joint. *Crit Rev Biomed Eng, 18*(5), 341-368.

Kadaba, M. P., Ramakrishnan, H. K., Wootten, M. E., Gainey, J., Gorton, G., & Cochran, G. V. (1989). Repeatability of kinematic, kinetic, and electromyographic data in normal adult gait. *J Orthop Res, 7*(6), 849-860.

Kanamori A, Zeminski J, Rudy TW, Li G, Fu FH, & Woo SLY. (2002). The effect of axial tibial torque on the function of the anterior cruciate ligament; a biomechanical study of a simulated pivot shift test. *Arthroscopy, 18*(4), 394-398.

Kellis, E., & Kouvelioti, V. *Epub*(2007). Agonist versus antagonist muscle fatigue effects on thigh muscle activity and vertical ground reaction during drop landing. *J Electromyogr Kinesiol.*

Kennedy JC, Alexander IJ, & Hayes KC. (1982). Nerve supply of the human knee and its functional importance. *Am J Sports Med, 10*, 329-335.

Kernozek, T. W., Torry, M. R., & Iwasaki, M. (2008). Gender differences in lower extremity landing mechanics caused by neuromuscular fatigue. *Am J Sports Med, 36*(3), 554-565.

Kirkendall, D. T., & Garrett, W. E., Jr. (2000). The anterior cruciate ligament enigma. Injury mechanisms and prevention. *Clin Orthop Relat Res*(372), 64-68.

Knapik JJ, Bauman CL, Jones BH, Harris JM, & L, V. (1991). Preseason strength and flexibility imbalances associated with athletic injuries in female collegiate athletes. *Am J Sports Med, 1*, 76-81.

Lambson RB, Barnhill BS, & Higgins RW. (1996). Football cleat design and its effect on anterior cruciate ligament injuries: a three-year prospective study. *Am J Sports Med, 24*, 155-159.

Lattanzio, P. J., Petrella, R. J., Sproule, J. R., & Fowler, P. J. (1997). Effects of fatigue on knee proprioception. *Clin J Sport Med, 7*(1), 22-27.

Lepers, R., Hausswirth, C., Maffiuletti, N., Brisswalter, J., & van Hoecke, J. (2000). Evidence of neuromuscular fatigue after prolonged cycling exercise. *Med Sci Sports Exerc, 32*(11), 1880-1886.

Lephart SM, Ferris CM, Riemann BL, Myers JB, & FH, F. (2002). Gender differences in strength and lower extremity kinematics during landing. *Clinical Orthopaedics and Related Research, 401*, 162-169.

Li, G., Rudy, T. W., Sakane, M., Kanamori, A., Ma, C. B., & Woo, S. L. (1999). The importance of quadriceps and hamstring muscle loading on knee kinematics and in-situ forces in the ACL. *J Biomech, 32*(4), 395-400.

Li, R. C., Maffulli, N., Hsu, Y. C., & Chan, K. M. (1996). Isokinetic strength of the quadriceps and hamstrings and functional ability of anterior cruciate deficient knees in recreational athletes. *Br J Sports Med, 30*(2), 161-164.

Lindenfeld TN, Schmitt DJ, Hendy MP, Mangine RE, & Noyes FR. (1994). Incidence of injury in indoor soccer. *Am J Sports Med, 22*, 364-371.

Liu SH, Al-Shaikh R, & Panossian V. (1996). Primary immunolocalization of estrogen and progesterone target cells in the human anterior cruciate ligament. *J Orthop Res, 14*, 526-533.

Lloyd, D. G., & Buchanan, T. S. (2001). Strategies of muscular support of varus and valgus isometric loads at the human knee. *J Biomech, 34*(10), 1257-1267.

Lloyd, D. G., Buchanan, T. S., & Besier, T. F. (2005). Neuromuscular biomechanical modeling to understand knee ligament loading. *Med Sci Sports Exerc, 37*(11), 1939-1947.

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

Louden JK, Jenkins W, & KL, L. (1996). The relationship between static posture and ACL injury in female athletes. *J Orthop Sports Phys Ther, 24*, 91-97.

MacIntosh BR, & Rassier DE. (2002). What is fatigue? *Can J Appl Physiol, 27*(1), 42-55.

MacWilliams, B. A., Wilson, D. R., DesJardins, J. D., Romero, J., & Chao, E. Y. (1999). Hamstrings cocontraction reduces internal rotation, anterior translation, and anterior cruciate ligament load in weight-bearing flexion. *J Orthop Res, 17*(6), 817-822.

Madigan, M. L., & Pidcoe, P. E. (2003). Changes in landing biomechanics during a fatiguing landing activity. *J Electromyogr Kinesiol, 13*, 491-498.

Malinzak, R. A., Colby, S. M., Kirkendall, D. T., Yu, B., & Garrett, W. E. (2001). A comparison of knee joint motion patterns between men and women in selected athletic tasks. *Clin Biomech (Bristol, Avon), 16*(5), 438-445.

Marder RA, Raskind JR, & Carroll M. (1991). Prospective evaluation of arthroscopically assisted anterior ligament reconstruction. *Am J Sports Med, 19*, 478-484.

Markolf, K. L., Bargar, W. L., Shoemaker, S. C., & Amstutz, H. C. (1981). The role of joint load in knee stability. *J Bone Joint Surg Am, 63*(4), 570-585.

Markolf, K. L., Burchfield, D. M., Shapiro, M. M., Shepard, M. F., Finerman, G. A., & Slauterbeck, J. L. (1995). Combined knee loading states that generate high anterior cruciate ligament forces. *J Orthop Res, 13*(6), 930-935.

Markolf, K. L., Gorek, J. F., Kabo, J. M., & Shapiro, M. S. (1990). Direct measurement of resultant forces in the anterior cruciate ligament. An in vitro study performed with a new experimental technique. *J Bone Joint Surg Am, 72*(4), 557-567.

Markolf, K. L., Graff-Radford, A., & Amstutz, H. C. (1978). In vivo knee stability. A quantitative assessment using an instrumented clinical testing apparatus. *J Bone Joint Surg Am, 60*(5), 664-674.

Markolf, K. L., Mensch, J. S., & Amstutz, H. C. (1976). Stiffness and laxity of the knee--the contributions of the supporting structures. A quantitative in vitro study. *J Bone Joint Surg Am, 58*(5), 583-594.

Marshall JL, Girgis FG, & RR, Z. (1972). The biceps femoris tendon and its function significance. *J Bone Joint Surg [Am], 54*, 1444-1450.

Matthews PBC. (1982). Where does sherrington's "muscular sense" originiate? Muscles, joints, corollary discharges? *Annu Rev Neurosci, 5*, 189-218.

McLean, S. G., Fellin, R. E., Suedekum, N., Calabrese, G., Passerallo, A., & Joy, S. (2007). Impact of fatigue on gender-based high-risk landing strategies. *Med Sci Sports Exerc, 39*(3), 502-514.

McLean, S. G., Huang, X., & van den Bogert, A. J. (2005). Association between lower extremity posture at contact and peak knee valgus moment during sidestepping: implications for ACL injury. *Clin Biomech (Bristol, Avon), 20*(8), 863-870.

McMorris, T., & Keen, P. (1994). Effect of exercise on simple reaction times of recreational athletes. *Percept Mot Skills, 78*(1), 123-130.

McNair PJ, & Prapavessis H. (1999). Normative data of vertical ground reaction forced during landing from a jump. *J Sci Med Sport, 2*, 86-88.

McNair, P. J., Marshall, R. N., & Matheson, J. A. (1990). Important features associated with acute anterior cruciate ligament injury. *N Z Med J, 103*(901), 537-539.

McNair, P. J., Prapavessis, H., & Callender, K. (2000). Decreasing landing forces: effect of instruction. *Br J Sports Med, 34*(4), 293-296.

McNitt-Gray, J. L. (1993). Kinetics of the lower extremities during drop landings from three heights. *J Biomech, 26*(9), 1037-1046.

Melnyk M, & Gollhofer A. (2007). Submaximal fatigue of the hamstrings impairs specific reflex components and knee stability. *Knee Surg Sports Traumatol Arthrosc, 15*, 525-532.

Mercer, J. A., Bates, B. T., Dufek, J. S., & Hreljac, A. (2003). Characteristics of shock attenuation during fatigued running. *J Sports Sci, 21*(11), 911-919.

Messina DF, Farney WC, & DeLee JC. (1999). The incidence of injury in Texas high school basketball: a prospective study among male and female athletes. *Am J Sports Med, 27*, 294-299.

Miura KM, Ishibashi Y, Tsuda E, Okamura Y, Otsuka H, & Toh S. (2004). The effect of local and general fatigue on knee proprioception. *J Arthroscopic Related Surgery, 20*(4), 414-418.

Moran KA, & Marshall BM. (2006). Effect of fatigue on tibial impact accelerations and knee kinematics in drop jumps. *Med Sci SPorts Exerc, 38*(10), 1836-1842.

Myer GD, Ford KR, Brent JL, & Hewett TE. (2007). Differential neuromuscular training effect on ACL injury risk factors in "high-risk" versus "low risk" athletes. *BMC Musculoskeletal Disorders, 39*(8).

Myer GD, Ford KR, & Hewett TE. (2005). The effects of gender on quadriceps muscle activation strategies during a maneuver that mimics a high ACL injury risk position. . *J Electromyogr Kinesiol, 15*, 181-189.

Myer, G. D., Ford, K. R., Barber Foss, K. D., Liu, C., Nick, T. G., & Hewett, T. E. (2009). The relationship of hamstrings and quadriceps strength to anterior cruciate ligament injury in female athletes. *Clin J Sport Med, 19*(1), 3-8.

Myklebust G, Maehlum S, Holm I, & Bahr R. (1998). A prospective cohort study of anterior cruciate ligament injuries in elite Norwegian team handball. *Scand J Med Sci Sports, 8*, 149-153.

Nelson, D. L., & Hutton, R. S. (1985). Dynamic and static stretch responses in muscle spindle receptors in fatigued muscle. *Med Sci Sports Exerc, 17*(4), 445-450.

Nicol, C., Komi, P. V., Horita, T., Kyrolainen, H., & Takala, T. E. (1996). Reduced stretch-reflex sensitivity after exhausting stretch-shortening cycle exercise. *Eur J Appl Physiol Occup Physiol, 72*(5-6), 401-409.

Nyland, J. A., Caborn, D. N. M., Shapiro, R., & Johnson, D. L. (1999). Crossover cutting during hamstring fatigue produced transverse plane knee control deficits. *J Athl Train, 34*(2), 137-143.

Nyland, J. A., Shapiro, R., Stine, R. L., Horn, T. S., & Ireland, M. L. (1994). Relationship of fatigued run and rapid stop to ground reaction forces, lower extremity kinematics, and muscle activation. *J Orthop Sports Phys Ther, 20*(3), 132-137.

O'Connor, J. J. (1993). Can muscle co-contraction protect knee ligaments after injury or repair? *J Bone Joint Surg Br, 75*(1), 41-48.

Onate, J. A., Guskiewicz, K. M., Marshall, S. W., Giuliani, C., Yu, B., & Garrett, W. E. (2005). Instruction of jump-landing technique using videotape feedback: altering lower extremity motion patterns. *Am J Sports Med, 33*(6), 831-842.

Orishimo KF, & Kremenic IJ. (2006). Effect of fatigue on single-leg hop landing biomechanics. *J Appl Biomech, 22*, 245-254.

Padua, D. A., Arnold, B. L., Perrin, D. H., Gansneder, B. M., Carcia, C. R., & Granata, K. P. (2006). Fatigue, vertical leg stiffness, and stiffness control strategies in males and females. *J Athl Train, 41*(3), 294-304.

Palmitier RA, An KN, Scott SG, & EY, C. (1991). Kinetic chain exercise in rehabilitation. *Sports Med, 11*, 402-413.

Pappas E, Hagins ME, Sheikhzadeh A, Nordin M, & Rose D. (2007). Biomechanical differences between unilateral and bilateral landings from a jump: gender differences. *Clin J Sport Med, 17*(4), 263-267.

Paterno, M. V., Myer, G. D., Ford, K. R., & Hewett, T. E. (2004). Neuromuscular training improves single-limb stability in young female athletes. *J Orthop Sports Phys Ther, 34*(6), 305-316.

Paul JJ, Spindler KP, Andrish JT, Parker RD, Secic M, & JA, B. (2003). Jumping versus nonjumping anterior cruciate ligament injuries: a comparison of pathology. *Clin J Sport Med, 13*, 1-5.

Piasecki, D. P., Spindler, K. P., Warren, T. A., Andrish, J. T., & Parker, R. D. (2003). Intraarticular injuries associated with anterior cruciate ligament tear: findings at ligament reconstruction in high school and recreational athletes. An analysis of sex-based differences. *Am J Sports Med, 31*(4), 601-605.

Pinto, M., Kuhn, J. E., Greenfield, M. L., & Hawkins, R. J. (1999). Prospective analysis of ice hockey injuries at the Junior A level over the course of one season. *Clin J Sport Med, 9*(2), 70-74.

Pollard CD, Davis LM, & Hamill J. (2004). Influence of gender on hip and knee mechanics during a randomly cued cutting maneuver. *Clin Biomech, 19*(1022-1031).

Putnam CA. (1993). Sequential motions of body segments in striking and throwing skills: descriptions and explanations. *J Biomech, 26*, 125-135.

Rahnama, N., Reilly, T., & Lees, A. (2002). Injury risk associated with playing actions during competitive soccer. *Br J Sports Med, 36*(5), 354-359.

Randall F, Miller H, & D, S. (1983). The use of prophylactis knee orthoses at Iowa State University. *Orthot Prosthet, 37*, 54-57.

Rodacki, A. L., Fowler, N. E., & Bennett, S. J. (2001). Multi-segment coordination: fatigue effects. *Med Sci Sports Exerc, 33*(7), 1157-1167.

Rodacki, A. L., Fowler, N. E., & Bennett, S. J. (2002). Vertical jump coordination: fatigue effects. *Med Sci Sports Exerc, 34*(1), 105-116.

Rozzi, S. L., Lephart, S. M., & Fu, F. H. (1999). Effects of muscular fatigue on knee joint laxity and neuromuscular characteristics of male and female athletes. *J Athl Train, 34*(2), 106-114.

Rozzi, S. L., Lephart, S. M., Gear, W. S., & Fu, F. H. (1999). Knee joint laxity and neuromuscular characteristics of male and female soccer and basketball players. *Am J Sports Med, 27*(3), 312-319.

Sakai, H., Tanaka, S., Kurosawa, H., & Masujima, A. (1992). The effect of exercise on anterior knee laxity in female basketball players. *Int J Sports Med, 13*(7), 552-554.

Salci Y, Kentel BB, Heycan C, Akin S, & F, K. (2004). Comparison of landing maneuvers between male and female college volleyball players. *Clin Biomech, 19*, 622-628.

Sander TC, Gansneder BM, Bunker LK, Gaesser GA, Saliba EN, Perrin DH, et al. (2004). Females utilize quadriceps dominant landing strategies that increase knee shear forces and internal extension moments. *J Athl Train, 39*(2 S), S-110.

Sargeant, A. J., & Kernell, D. (1993). *Neuromuscular Fatigue*. Amsterdam: Royal Netherlands Academy of Arts and Sciences.

Sbriccoli P, Solomonow M, Zhou BH, Lu Y, & Sellards R. (2005). Neuromuscular response to cyclic loading of the anterior cruciate ligament. *Am J Sports Med, 33*(4), 543-551.

Schultz, R. A., Miller, D. C., Kerr, C. S., & Micheli, L. (1984). Mechanoreceptors in human cruciate ligaments. A histological study. *J Bone Joint Surg Am, 66*(7), 1072-1076.

Scranton PE, Whitesel JP, & Powell JW, e. a. (1997). A review of selected non-contact anterior cruciate ligament injuries in the National Football League. *Foot Ankle Int, 18*, 772-776.

Sebastianelli WJ, Hanks GA, & A, K. (1990). Isolated avulsion of the biceps femoris insertion. A case report. *Clin Ortho Rel Res, 259*, 200-203.

Seeley RR, Stephens TD, & Tate P. (2003). Muscular System In M. J. Lange (Ed.), *Anatomy & Physiology* (Vol. Sixth, pp. 313-362). Boston: McGraw-Hill.

Self BP, & Paine D. (2001). Ankle biomechanics during four landing techniques. *Med Sci Sports Exerc, 27*, 294-299.

Shimokochi, Y., Yong Lee, S., Shultz, S. J., & Schmitz, R. J. (2009). The relationships among sagittal-plane lower extremity moments: implications for landing strategy in anterior cruciate ligament injury prevention. *J Athl Train, 44*(1), 33-38.

Shultz, S. J., & Prerrin, D. H. (1999). Using surface electromyography to assess sex differenced in neuromuscular response characteristics. *J Athl Train, 34*(2), 165-176.

Shutte MJ, Dabesies EJ, & Zimney ML, e. a. (1987). Neural anatomy of the human anterior cruciate ligament. *J Bone Joint Surg [Am] 69*, 243-247.

Sjolander, P., Johansson, H., Sojka, P., & Rehnholm, A. (1989). Sensory nerve endings in the cat cruciate ligaments: a morphological investigation. *Neurosci Lett, 102*(1), 33-38.

Skinner, H. B., Wyatt, M. P., Hodgdon, J. A., Conard, D. W., & Barrack, R. L. (1986). Effect of fatigue on joint position sense of the knee. *J Orthop Res, 4*(1), 112-118.

Skinner, H. B., Wyatt, M. P., Stone, M. L., Hodgdon, J. A., & Barrack, R. L. (1986). Exercise-related knee joint laxity. *Am J Sports Med, 14*(1), 30-34.

Smith BA, Livesay GA, & SL, W. (1993). Biology and biomechanics of the anterior cruciate ligament. *Clin Sports Med, 12*, 637-670.

Solomonow, M., Baratta, R., Zhou, B. H., Shoji, H., Bose, W., Beck, C., et al. (1987). The synergistic action of the anterior cruciate ligament and thigh muscles in maintaining joint stability. *Am J Sports Med, 15*(3), 207-213.

Solomonow, M., Baratta, R. V., Zhou, B. H., Burger, E., Zieske, A., & Gedalia, A. (2003). Muscular dysfunction elicited by creep of lumbar viscoelastic tissue. *J Electromyogr Kinesiol, 13*(4), 381-396.

Solomonow, M., & Krogsgaard, M. (2001). Sensorimotor control of knee stability. A review. *Scand J Med Sci Sports, 11*(2), 64-80.

Solomonow, M., Zhou, B. H., Baratta, R. V., Lu, Y., & Harris, M. (1999). Biomechanics of increased exposure to lumbar injury caused by cyclic loading: Part 1. Loss of reflexive muscular stabilization. *Spine, 24*(23), 2426-2434.

Stevenson H, Webster J, Johnson RJ, & Beynnon B. (1998). Gender differences in knee injury epidemiology among competitive alpine ski racers. *Iowa Ortho J., 18*, 64-66.

Stoller, D. W., Markolf, K. L., Zager, S. A., & Shoemaker, S. C. (1983). The effects of exercise, ice, and ultrasonography on torsional laxity of the knee. *Clin Orthop Relat Res*(174), 172-180.

Stone, R. J., & Stone, J. A. (2006). *Atlas of Skeletal Muscles* (Fifth ed.). Boston: McGraw Hill

Terry GC, & LaPrade RF. (1996). The biceps femoris muscle complex at the knee. Its anatomy and injury patterns associated with acute anterolateral-anteromedial rotatory instability. *Am J Sports Med, 24*, 2-8.

Torzilli, P. A., Deng, X., & Warren, R. F. (1994). The effect of joint-compressive load and quadriceps muscle force on knee motion in the intact and anterior cruciate ligament-sectioned knee. *Am J Sports Med, 22*(1), 105-112.

Tuggy ML, & Ong R. (2000). Injury risk factors among telemark skiers. *Am J Sports Med, 28*, 83-89.

Urabe, Y., Kobayashi, R., Sumida, S., Tanaka, K., Yoshida, N., Nishiwaki, G. A., et al. (2005). Electromyographic analysis of the knee during jump landing in male and female athletes. *Knee, 12*(2), 129-134.

Van Lunen BL, Roberts J, Branch JD, & Dowling EA. (2003). Association of menstrual-cycle hormone changes with anterior cruciate ligament laxity measurements. *J Athl Train, 38*(4), 298-303.

Vollestad NK. (1997). Measurement of human muscle fatigue. *J Neurosci Meth, 74*, 219-227.

Voloshin A, Mizrani J, & Isakov E. (1998). Dynamic loading on the human musculoskeletal system-effect of fatigue. *Clin Biomech, 13*, 513-520.

Walsh, L. D., Hesse, C. W., Morgan, D. L., & Proske, U. (2004). Human forearm position sense after fatigue of elbow flexor muscles. *J Physiol, 558*(Pt 2), 705-715.

Warren LF, & Marchal JL. (1979). The supporting structure and layers of the medial side of the knee. *J Bone Joint Surg [Am], 61*, 56-62.

Wikstrom EA, Powers ME, & Tillman MD. (2004). Dynamic stabilization time after isokinetic and functional fatigue. *J Athl Train, 39*(3), 247-253.

Wilderman, D. R., Ross, S. E., & Padua, D. A. (2009). Thigh muscle activity, knee motion, and impact force during side-step pivoting in agility-trained female basketball players. *J Athl Train, 44*(1), 14-25.

Wilkerson, G. B., Colston, M. A., Short, N. I., Neal, K. L., Hoewischer, P. E., & Pixley, J. J. (2004). Neuromuscular Changes in Female Collegiate Athletes Resulting From a Plyometric Jump-Training Program. *J Athl Train, 39*(1), 17-23.

Williams GN, Chmielewski T, Rudolph KS, Buchanan TS, & ML, S. (2001). Dynamic knee stability: current theory and implications for clinicians and scientists. *J Orthop SPorts Phys Ther, 31*, 546.

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

Wojtys EM, Huston LJ, Lindenfeld TN, Heweet TE, & MLVH, G. (1998). Association between the menstrual cycle and anterior cruciate ligament injuries in female athletes. *Am J Sports Med, 26*, 614-619.

Wojtys EM, Huston LJ, Schock HJ, Boylan JP, & Ashton-Miller JA. (2003). Gender differences in muscular protection of the knee in torsion in size-matched athletes *J Bone Joint Surg [Am], 85-A*(5), 782-789.

Wojtys EM, Huston LJ, Taylor PD, & Bastain SD. (1996). Neuromuscular adaptations in isokinetic, isotonic, and agility training programs. *Am J Sports Med, 24*(2), 187-192.

Wojtys, E. M., Ashton-Miller, J. A., & Huston, L. J. (2002). A gender-related difference in the contribution of the knee musculature to sagittal-plane shear stiffness in subjects with similar knee laxity. *J Bone Joint Surg Am, 84-A*(1), 10-16.

Wojtys, E. M., Wylie, B. B., & Huston, L. J. (1996). The effects of muscle fatigue on neuromuscular function and anterior tibial translation in healthy knees. *Am J Sports Med, 24*(5), 615-621.

Woledge, R. C., Curtin, N. A., & Homsher, E. (1985). Energetic aspects of muscle contraction. *Monogr Physiol Soc, 41*, 1-357.

Woo, S. L., Abramowitch, S. D., Kilger, R., & Liang, R. (2006). Biomechanics of knee ligaments: injury, healing, and repair. *J Biomech, 39*(1), 1-20.

Woodburne RT, & Burkel WE. (1994). *Essentias of human anatomy* (9th ed.). New York: Oxford University Press.

Xerogeanes JW, Takeda Y, Livesay GA, Ishibashi Y, Kim HS, Fu FH, et al. (1995). Effect of knee flexion on the in situ force distribution in the human anterior cruciate ligament *Knee Surg Sports Traumatol Arthrosc, 3*, 9-13.

Yu B, Lin CF, & Garrett WE. (2006). Lower extremity biomechanics during the landing of a stop-jump task. *Clin Biomech, 21*, 297-305.

Zazulak, B. T., Ponce, P. L., Straub, S. J., Medvecky, M. J., Avedisian, L., & Hewett, T. E. (2005). Gender comparison of hip muscle activity during single-leg landing. *J Orthop Sports Phys Ther, 35*(5), 292-299.

Zhang SN, Bates BT, & Durek JS. (2000). Contributions of lower extremity joints to energy dissipation during landings. *Med Sci Sports Exerc, 32*, 812-819.

Zimny, M. L., Schutte, M., & Dabezies, E. (1986). Mechanoreceptors in the human anterior cruciate ligament. *Anat Rec, 214*(2), 204-209.

Dear Prospective Study Participant,

We are conducting a study that examines the effects of instruction and neuromuscular fatigue on jump landing kinematics on female recreational and varsity collegiate athletes. The attached "informed consent" form describes the study and asks for your permission to participate in the study.

After reviewing the attached information, please return a signed copy of the "Informed Consent" form to Kristen Phillips, ATC or Jena Etnoyer, ATC. Keep the additional copy of the form for you records.

If you have any questions pertaining to the attached form or to the research study, please feel free to contact James Onate, PhD, ATC, Kristen Phillips, ATC or Jena Etnoyer, ATC at the information listed below.

Thank you in advance for participating in this study.

Sincerely,

James A. Onate, PhD, ATC – Responsible Project Investigator
Director, ODU Sports Medicine Research
Laboratory
Old Dominion University
224 Spong Hall
Norfolk, VA 23509-0196
(O) 757-683-4351
Jonate@odu.edu

Jena Etnoyer, ATC
Masters Student,
Graduate Athletic Training Program
Old Dominion University
Norfolk, VA 23509-0196
(C) 717-991-5595
jetno001@odu.edu

Kristen Philips, ATC
Masters Student
Graduate Athletic Training
Old Dominion University
Norfolk, VA 23509-0196
(C) 540-446-3984
kphil015@odu.edu

INFORMED CONSENT DOCUMENT

OLD DOMINION UNIVERSITY

PROJECT TITLE: Effect of Instruction and Neuromuscular Fatigue on Jump Landing in Female Recreational and Collegiate Athletes

INTRODUCTION

The purposes of this form are to give you information that may affect your decision whether to say YES or NO to participation in this research, and to record the consent of those who say YES. The purpose of this project is to assess the effect of instruction and neuromuscular fatigue on jump landing.

RESEARCHERS

James Onate, PhD, ATC
Director of Sports Medicine Research Laboratory, Old Dominion University, ESPER Dept

Kristen Phillips, ATC
Masters Student, Graduate Athletic Training Program, Old Dominion University ESPER Dept

Jena Etnoyer, ATC
Masters Student, Graduate Athletic Training Program, Old Dominion University ESPER Dept

Bonnie Van Lunen, PhD, ATC
Director- Graduate Athletic Training Program and Doctorate in Human Movement Science Program, Old Dominion University, ESPER Dept.

DESCRIPTION OF RESEARCH STUDY

Several studies have been conducted looking into the effect of instruction or neuromuscular fatigue on jump landing to assess the risk of ACL injury. However none of these studies have determined the effect of jump landing feedback of box-drop jump, running-stop jump, and side-step maneuver, or isolated hamstring fatigue on double leg drop box landing.

This study will involve your participation in two testing sessions. The first session will be followed by a second session two to three months later. Approximately a total of 2.5 hours of your time will be required. Upon arriving in the Sports Medicine Research Laboratory, your height, weight and body dimensions will be measured. Following those measurements, you will complete a brief warm up on a stationary bicycle and a self-directed stretch. After the brief warm up time you will perform jump landing tasks, that include a box-drop jump, a running-stop jump and a sidestep maneuver, and a fatigue task of the hamstrings. The fatigue task will only be utilized on the first session of testing.

- **Jump Landing (Box-drop Jump, Running-stop Jump, Side-step Maneuver):** These tasks involve using Vicon, which entails reflective markers to be placed on various landmarks on the body. The cameras around the laboratory track the reflective marks to analyze movement during the tasks. Five test trials will be conducted for each test.
- **Hamstring Fatigue:** The Primus RS is used for this task to record the torque (force) produced during each repetition of knee flexion of the fatigue protocol. Continuous repetitions will be performed until hamstring fatigue is achieved.

Approximately 90 subjects will be participating in this study.

EXCLUSIONARY CRITERIA
In order for you to participate in this study, you must be a healthy female recreational or varsity collegiate athlete who exercises a minimum of three times per week for a minimum of 20 minutes. You must not have had a lower extremity injury in the past two months that limited you from activity for more than one day, no current self-reported history of lower extremity instability, or history of any lower extremity surgery within the past two years. You also have never had an ACL injury or reconstructive surgery.

RISKS AND BENEFITS
RISKS: If you decide to participate in this study, then you may face a risk of ligament injury, muscle strains or delayed onset muscle soreness due to the hamstring fatigue activity. The fatigue activity may decrease muscular control, therefore affecting the body positioning during the landing activity and increasing the risk of injury to ligaments of the knee. To reduce this risk, the fatigue protocol and jump landing task are performed in a controlled lab environment without external stimuli and are supervised by trained and practiced researchers. The hamstring fatigue also requires full effort over a series of repetitions to elicit a decrease in maximum peak toque, which increases the likelihood of a strained muscle or delayed onset muscle soreness following the completion of the initial testing session. The researchers try to reduce these risks specifically by allowing a 10 minute warm-up, consisting of a stationary bike ride and self-directed static stretching, to assist in properly increasing the body temperature and elastic properties of the hamstring musculature. There is also a possibility of slight discomfort of the Velcro strap at the testing station. Also, as with any research, there is some possibility that you may be subject to risks that have not yet been identified.

BENEFITS: If you decide to participate in this study, you will receive points *equivalent to one extra credit assignment* if you are registered for any of the designated ESPER Department courses. Equivalent points may be obtained in other ways. Another benefit to you for participating in this study is learning how to properly position yourself during a jump-landing activity. The results of this study may benefit others and expand the knowledge in the profession of athletic training by drawing more precise and clearer conclusions on risk factors that predispose female athletes to risk of ACL injury.

COSTS AND PAYMENTS

The researchers are unable to give you any payment for participating in this study.

NEW INFORMATION

If the researchers find new information during this study that would reasonably change your decision about participating, then they will give it to you.

CONFIDENTIALITY

All information obtained about you in this study is strictly confidential and will not be released to anyone unless disclosure is required by law. The results of this study may be used in reports, presentations and publications, but the researcher will not identify you.

WITHDRAWAL PRIVILEGE

It is OK for you to say NO. Even if you say YES now, you are free to say NO later, and walk away or withdraw from the study -- at any time. Your decision will not affect your relationship with Old Dominion University, or otherwise cause a loss of benefits to which you might otherwise be entitled. The researchers reserve the right to withdraw your participation in this study, at any time, if they observe potential problems with your continued participation.

COMPENSATION FOR ILLNESS AND INJURY

If you say YES, then your consent in this document does not waive any of your legal rights. However, in the event of harm arising from this study, neither Old Dominion University nor the researchers are able to give you any money, insurance coverage, free medical care, or any other compensation for such injury. In the event that you suffer injury as a result of participation in any research project, you may contact Dr. James A. Onate, ODU Sports Medicine Research Laboratory at 757-683-4351 or Dr. George Maihafer the current IRB chair at 757-683-4520 at Old Dominion University, who will be glad to review the matter with you.

VOLUNTARY CONSENT

By signing this form, you are saying several things. You are saying that you have read this form or have had it read to you, that you are satisfied that you understand this form, the research study, and its risks and benefits. The researchers should have answered any questions you may have had about the research. If you have any questions later on, then the researchers should be able to answer them: Dr. James A. Onate, ODU Sports Medicine Research Laboratory at 757-683-4351

If at any time you feel pressured to participate, or if you have any questions about your rights or this form, then you should call Dr. George Maihafer, the current IRB chair, at 757-683-4520, or the Old Dominion University Office of Research, at 757-683-3460.

And importantly, by signing below, you are telling the researcher YES, that you agree to participate in this study. The researcher should give you a copy of this form for your records.

Subject's Printed Name & Signature	**Date**

INVESTIGATOR'S STATEMENT

I certify that I have explained to this subject the nature and purpose of this research, including benefits, risks, costs, and any experimental procedures. I have described the rights and protections afforded to human subjects and have done nothing to pressure, coerce, or falsely entice this subject into participating. I am aware of my obligations under state and federal laws, and promise compliance. I have answered the subject's questions and have encouraged him/her to ask additional questions at any time during the course of this study. I have witnessed the above signature(s) on this consent form.

Investigator's Printed Name & Signature	**Date**

APPENDIX B

PARTICIPATION QUESTIONNAIRE

Subject # __________

Pre-Screening Questionnaire

Name: ______________________________ Date: ____________

Age: _________

Ethnicity: _____________

Varsity Athlete? YES NO

What leg would you kick a ball with? LEFT RIGHT

How many times do you engage in exercise per week? _________

On average, how long do you exercise per session? _______

Do you have any lower extremity instability (ankle or knee give way frequently)?

YES NO

Have you had an injury to the lower extremity in the past 2 months? YES NO

If yes, did this injury limit you from activity for more than one day? YES NO

Have you had any lower extremity surgeries within the past 2 years? YES NO

Do you have any history of ACL injury or reconstructive surgery? YES NO

Have you had any previous jump-landing technique training? YES NO

FATIGUE FORM

Subject ID________________________ **Date**__________________

Dominate Leg: L R

Positioning Check Off:

___ Axis match-up with knee jt line

___ Leg pad placed lower shaft of leg

___ Velcro straps tightened

___ Arms crossed

Lever Length: _________ cm

FATIGUE

Max Torque (MT): __________

Subject Fatigued: (MT) x .7 = _________

*** 3 consecutive reps***

Torque of last 3 reps: _____

_____

_____

Time of Fatigue _________

Time of Rest _________

Time of Jumps _________

Total Number of Reps: ___________

Explanation of Fatigue:

We are now going to induce fatigue on your hamstring. The starting position will be with your knee straight. The machine will then move back to allow your knee to bend and you pull your heel back with the machine. Once the machine starts to move in the other direction, I want to really resist it by trying to keep your knee bent. When the machine forces your leg straight to the starting position you can relax, and that will complete one repetition.

We will do three practice reps to allow you to feel comfortable.

Now we are doing 5 individual reps for test to establish your maximum strength.

Now you will do continuous individual reps as we just tested until I tell you to stop. Once I tell you we are done, we will unstrap you and you will immediately go to the next room and assume the starting position with one foot in each force plate. Are you ready?

Subject # ___ ______

Reps of Fatigue

1		41		81		121		161	
2		42		82		122		162	
3		43		83		123		163	
4		44		84		124		164	
5		45		85		125		165	
6		46		86		126		166	
7		47		87		127		167	
8		48		88		128		168	
9		49		89		129		169	
10		50		90		130		170	
11		51		91		131		171	
12		52		92		132		172	
13		53		93		133		173	
14		54		94		134		174	
15		55		95		135		175	
16		56		96		136		176	
17		57		97		137		177	
18		58		98		138		178	
19		59		99		139		179	
20		60		100		140		180	
21		61		101		141		181	
22		62		102		142		182	
23		63		103		143		183	
24		64		104		144		184	
25		65		105		145		185	
26		66		106		146		186	
27		67		107		147		187	
28		68		108		148		188	
29		69		109		149		189	
30		70		110		150		190	
31		71		111		151		191	
32		72		112		152		192	
33		73		113		153		193	
34		74		114		154		194	
35		75		115		155		195	
36		76		116		156		196	
37		77		117		157		197	
38		78		118		158		198	
39		79		119		159		199	
40		80		120		160		200	

Mean	

APPENDIX D

NORMATIVE DATA OF FATIGUE JUMP LANDING STUDIES

Peak Proximal Anterior Tibial Translation

	Pre-Fatigue	Post-Fatigue	Task	Fatigue Protocol	Fatigue	Sig
Chappell et al 2005	0.24	0.29	Forward, vertical, backward jump	30m Sprint, 5 Vertical Jumps	Failure	p<.05
Kernozek et al 2008	.95(.2)	.76(.15)	50cm single leg land	Squat, 60%RM	Failure	p<.05
Rozzi et al 1999	6.05(1.46)	6.06(1.23)	KT2000 supine, 20 knee flexion	Quadriceps & Hamstring fatigue	25% reduction	p<.05

Knee Flexion at Initial Contact

	Pre-Fatigue	Post-Fatigue	Task	Fatigue Protocol	Fatigue	Sig
Benjaminse et al 2008	13.6(5.3)	11.5(5.8)	Single leg stop jump	Running	HR, PRE	p<.05
Coventry et al 2006	-8.8(2.5)	-14.6(5.2)	Single leg counter movement jump	Single leg land, CMJ, 5 squats	Failure, RPM	p<.05
Kellis et al 2007	10.62(2.58)	18.5(4.09)	30 cm single leg land	Quadriceps	30% reduction	p<.05
Kellis et al 2007	9.75(2.29)	15(3.74)	30 cm single leg land	Hamstrings	30% reduction	p<.05
Gehring et al 2009	15.9(1.7)	16.5(1.5)	52cm bilateral land	Leg press with 50%rm	Failure	p>.05
Madigan et al 2003	8.5(5.1)	10.1(6.2)	25cm singe leg land	25cm single leg land, single squat	Failure	p>.05
Orishimo et al 2006	23.8(5.5)	21.9(6.7)	Single leg forward hop	30cm Step-ups	80% distance	p>.05

Maximum Knee Flexion

	Pre-Fatigue	Post-Fatigue	Task	Fatigue Protocol	Fatigue	Sig
Kellis et al 2007	47.17(9.77)	57.84(11.68)	30 cm single leg land	Hamstrings	30% reduction	p<.05
Kellis et al 2007	66.9(18.7)	80.9(19)	30 cm single leg land	Quadriceps	30% reduction	p<.05
Madigan et al 2003	42.6(7.1)	49.3(9.7)	25cm singe leg land	25cm single leg land, single squat	Failure	p>.05
Orishimo et al 2006	66.9 (18.7)	80.9(19)	Single leg forward hop	30cm Step-ups	80% distance	p<.05
Borotikar et al 2008	14.4(4.3)	12.5(3.6)	Dominant leg anticipated forward single land	Squats between jumps	Failure	p>.05
Borotikar et al 2008	13.4(3.2)	12.6(4.8)	Dominant leg unanticipated forward single land	Squats between jumps	Failure	p>.05
Coventry et al 2006	54.6(10.9)	54.3(9.9)	Single leg counter movement jump	Single leg land, CMJ, 5 squats	Failure, RPM	p>.05
Gehring et al 2009	15.9(1.7)	16.5(1.5)	52cm bilateral land	Leg press with 50%rm	Failure	p>.05

Knee Valgus at Initial Contact
(+) varus, (-) valgus

	Pre-Fatigue	Post-Fatigue	Task	Fatigue Protocol	Fatigue	Sig
Gehring et al 2009	3.1(1.1)	3.5(1.1)	52cm bilateral land	Leg press with 50%rm	Failure	p>.05
McLean et al 2007	2(4.3)	.4(3.7)	50cm bilateral land	Step-downs, bounding	4min, HR	p>.05

Maximum Knee Valgus
(+) varus, (-) valgus

	Pre-Fatigue	Post-Faigue	Task	Fatigue Protocol	Fatigue	Sig
Benjaminse et al 2008	-3.5(3.7)	-2.7(3.5)	Single leg stop jump	Running	HR, PRE	p<.05
Borotikar et al 2008	-3.5(3.2)	-4.3(3.1)	Dominant leg anticipated forward single land	Squats between jumps	Failure	p<.05
Borotikar et al 2008	-3.9(2.8)	-7.5(3.8)	Dominant leg unanticipated forward single land	Squats between jumps	Failure	p<.05
McLean et al 2007	-3.4(4.4)	-10.2(5)	50cm bilateral land, dominate	Step-downs, bounding	4min, HR	p<.05
McLean et al 2007	-4.5(4.4)	-12.3(4)	50cm bilateral land, non-dominate	Step-downs, bounding	4min, HR	p<.05
Gehring et al 2009	-4(3)	-4.3(3.1)	52cm bilateral land	Leg press with 50%rm	Failure	p>.05
Kernozek et al 2008	-3.86(4.37)	-3.91(4.63)	50cm single leg land, female	Squat 60%RM	Failure	p>.05

Hip Flexion at Initial Contact

	Pre-Fatigue	Post-Fatigue	Task	Fatigue Protocol	Fatigue	Sig
Coventry et al 2006	.09(3.5)	4.3(5.6)	Single leg counter movement jump	Single leg land, CMJ, 5 squats	Failure, RPM	p<.05
Madigan et al 2003	19.4(8.6)	20.7	25cm singe leg land	25cm single leg land, single squat	Failure	p>.05
McLean et al 2007	33.3(5.3)	32(3.9)	50cm bilateral land, dominate	Step-downs, bounding	4min, HR	p>.05

Maximum Hip Flexion

	Pre-Fatigue	Post-Fatigue	Task	Fatigue Protocol	Fatigue	Sig
Borotikar et al 2008	31(3)	27.2(2.4)	Dominant leg anticipated forward single land	Squats between jumps	Failure	p<.05
Borotikar et al 2008	25.8(2.4)	21.9(2.6)	Dominant leg unanticipated forward single land	Squats between jumps	Failure	p<.05
Coventry et al 2006	32.5(13.2)	43.7(19.7)	Single leg counter movement jump	Single leg land, CMJ, 5 squats	Failure, RPM	p<.05
Kellis et al 2007	32(7.97)	40.91(9.83)	30 cm single leg land	Quadriceps	30% reduction	p<.05
Kernozek et al 2008	40.72(9.57)	48.02(14.44)	50cm single leg land, female	Squat 60%RM	Failure	p<.05
Kellis et al 2007	33.29(5.93)	34.93(9.56)	30 cm single leg land	Hamstrings	30% reduction	p>.05
Madigan et al 2003	29(11.3)	32.9(16.2)	25cm singe leg land	25cm single leg land, single squat	Failure	p>.05
McLean et al 2007	62.7(13.1)	64.1(13.5)	50cm bilateral land, dominate	Step-downs, bounding	4min, HR	p>.05