


Spring 2012

# The Influence of Interlimb Differences on Anterior Cruciate Ligament Injury Risk Factors in Female Collegiate Soccer Athletes

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## Recommended Citation

Greska, Eric K.. "The Influence of Interlimb Differences on Anterior Cruciate Ligament Injury Risk Factors in Female Collegiate Soccer Athletes" (2012). Doctor of Philosophy (PhD), dissertation, , Old Dominion University, DOI: 10.25777/bcse-bx21  
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THE INFLUENCE OF INTERLIMB DIFFERENCES ON ANTERIOR CRUCIATE  
LIGAMENT INJURY RISK FACTORS IN FEMALE COLLEGIATE SOCCER  
ATHLETES

by

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A Dissertation Submitted to the Faculty of  
Old Dominion University in Partial Fulfillment of the  
Requirement for the Degree of

DOCTOR OF PHILOSOPHY

HUMAN MOVEMENT SCIENCE

OLD DOMINION UNIVERSITY  
May 2012

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## ABSTRACT

### THE INFLUENCE OF INTERLIMB DIFFERENCES ON ANTERIOR CRUCIATE LIGAMENT RISK FACTORS IN FEMALE COLLEGIATE SOCCER ATHLETES

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Old Dominion University, 2012  
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Female collegiate soccer athletes suffer anterior cruciate ligament (ACL) injuries at a rate of 0.31 per 1000 athlete-exposures, with 67% of injuries expected to be from a non-contact mechanism. As well, previous studies have demonstrated dissimilarity in non-contact ACL injuries relative to dominant and non-dominant legs; with the non-dominant leg being the involved side in 48-67% of reported incidents. The aim of this study was to examine lower extremity neuromechanics relative to leg dominance during an unanticipated sidestep cutting task, with differing states of fatigue and training.

Twenty female collegiate soccer players performed three valid trials for both the dominant and non-dominant leg of an unanticipated sidestep cutting task. Three-dimensional kinematics and kinetics were recorded. Participants performed tasks during pre-and post-fatigue states, and partook in an injury prevention intervention, repeating the testing procedure at the cessation. To assess differences in neuromechanical parameters, for experiment I, a multivariate analyses of variance was conducted to assess the effect of leg dominance, and experiments II and III conducted repeated measures analyses of variance to assess the effects of fatigue and training ( $p < 0.05$ ).

For experiment I, no significant differences were found between dominant and non-dominant legs, but each leg displayed a unique time occurrence for peak electromyographic activity prior to contact. Experiment II, no differences were found for leg dominance, but fatigue demonstrated significant decreases in gluteus medius activation during pre-contact ( $p < 0.001$ ) and initial contact ( $p = 0.002$ ). At post-fatigue, there was a significant change in hip flexion at initial contact, and hip and knee flexion at peak knee adduction moment and peak stance ( $p < 0.001$ ). Significant increases ( $p < 0.001$ ) were also noted at peak stance for vertical ground reaction force and hip flexion moment. For experiment III, training produced decreases ( $p < 0.001$ ) in hip flexion angle throughout the stance phase, creating a more erect position. As well, fatigue induced decreases ( $p < 0.006$ ) in hip flexion and abduction throughout the stance phase, as well as decreasing ( $p < 0.006$ ) hip flexion and abduction moments during peak knee adduction moment. The experiments demonstrate that lower extremity neuromechanics, in the absence of fatigue, do not differ between legs during an unanticipated sidestep cutting task.

## ACKNOWLEDGEMENTS

This Dissertation is dedicated to those who made this journey possible, my committee members, my family, my friends, and my love, Lauren. If it were not for their tolerance, my goals would still be dreams.

A special thank you is extended to the ladies and coaches of the Old Dominion University Women's soccer team.

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## CHAPTER I

### INTRODUCTION

It is estimated that between 80,000 and 250,000 anterior cruciate ligament (ACL) ruptures occur each year (Griffin, et al., 2000; Mandelbaum, et al., 2005; Owings & Kozak, 1998). Within the general population, ACL injuries occur at an approximate rate of 1 in 3000. In suffering an ACL injury, reconstruction surgery costs range from \$4,622 to \$8,754 dependant upon the ligament reconstruction technique utilized (Bonsell, 2000; Cole, et al., 2005; Gianotti, Marshall, Hume, & Bunt, 2009). This cost does not take into account the subsequent rehabilitation required after surgery, which can raise the overall cost of treatment to an estimated \$17,000 per ACL injury (Gottlob, Baker, Pellissier, & Colvin, 1999; Owen, Campbell, Falkner, Bialkowski, & Ward, 2006). Applying these expenses to the estimated yearly incidents of ACL injuries within the United States leaves an amassed annual cost between 1.3 and 4.25 billion dollars.

For college athletics, findings reinforce the fact that female athletes are at a greater disposition than their male counterparts to suffer a non-contact ACL injury (Agel, Arendt, & Bershadsky, 2005; Arendt, Agel, & Dick, 1999). For high school and college aged females participating in sports, ACL injuries occur at dramatic rates. In high school soccer and basketball for females, ACL injuries occur at a rate of 1 in 200 athletes, where as their male counterparts are at a rate of 1 in 1000 (Powell & Barber-Foss, 2000). At the collegiate level, this rate increases to 0.32 ACL injuries per 1000 athletic exposures, with an athletic exposure defined as participating in one practice or game session (Mihata, Beutler, & Boden, 2006). Putting this rate into perspective, a collegiate soccer team consisting of 25 athletes will

participate in over 120 practice and game sessions from pre-season to the spring, leaving a projected rate of approximately one ACL injury per season per team, with 67% of injuries expected to be from a non-contact mechanism (Agel, et al., 2005).

Over the past decade, studies have attempted to identify the mechanisms involved in perpetuating a non-contact ACL injury. The mechanism of injury has been theorized to occur by means of an interaction between internal and external risk factors (Bahr & Krosshaug, 2005; Meeuwisse, Tyreman, Hagel, & Emery, 2007). In Bahr and Krosshaug's comprehensive model for injury causation, an athlete is already predisposed to an injury by internal factors. Internal risk factors that have been identified for non-contact ACL injuries can be divided into four categories: anatomical and structural, hormonal, screening and prevention, and neuromuscular and biomechanical (Shultz, Schmitz, Nguyen, et al., 2010). Anatomical factors including the size and shape of the condylar surfaces and the cross-sectional area of the ligaments have been shown to have an influence on ACL injuries. The size and shape of the condylar surfaces focus on the overall width and the size of the intercondylar notch, along with its shape (Anderson, Dome, Gautam, Awh, & Rennirt, 2001; Anderson & Pandy, 2001; Shelbourne, Davis, & Klootwyk, 1998). An ACL with a greater cross-sectional area allows for a greater tensile stiffness, limiting anterior tibial translation. Females have a significantly smaller cross-sectional area in comparison to males, which translates to ruptures occurring at 30% less of a loading force on the ACL (Chandrashekar, Mansouri, Slauterbeck, & Hashemi, 2006). An excessive flexibility of the knee joint creates anterior knee laxity, which allows for greater anterior tibial translation, a negative factor (Shultz, Carcia, & Perrin, 2004; Shultz, et al., 2006). Hormonal factors have also been shown to influence the likelihood of suffering an ACL injury by altering ligament

compliance (Arendt, Bershadsky, & Agel, 2002; Myklebust, et al., 2003; Park, Stefanyshyn, Ramage, Hart, & Ronsky, 2009; Shultz, Schmitz, & Beynnon, 2010; van Lunen, Roberts, Branch, & Dowling, 2003; Wojtys, Huston, Boynton, Spindler, & Lindenfeld, 2002). ACL injury risk tends to be greater during the pre-ovulatory phase of the cycle than during the postovulatory phase. Joint laxity is further included in screening and prevention mechanisms, as screening tests such as the Beighton Joint Mobility Index have been correlated to an increased risk of incurring an ACL Injury (Hewett, et al., 2010; Quatman, Ford, Myer, Paterno, & Hewett, 2008). As all of these factors are not directly under human control, whereas neuromuscular and biomechanical are.

Neuromuscular and biomechanical factors are the main contributing factor to the loading of the ACL, regardless of the other factors (McLean, Huang, Su, & van den Bogert, 2004; Shin, Chaudhari, & Andriacchi, 2011). Neuromuscular control has been defined as an unconscious activation of the dynamic restraints surrounding a joint in response to sensory stimuli (Griffin, et al., 2000). As tasks during sports movements allocate conscious attention, the unconscious triggering of muscles to accomplish such movements is vital. Differences in movement patterns while performing such tasks as landing, cutting, and pivoting, have been demonstrated between males and females (Beaulieu, Lamontagne, & Xu, 2008, 2009; Bencke & Zebis, 2011; Carcia & Martin, 2007; Chappell, Creighton, Giuliani, Yu, & Garrett, 2007; Cowling & Steele, 2001; Ford, Myer, & Hewett, 2003; Ford, Myer, Toms, & Hewett, 2005; Hewett, Ford, Myer, Wanstrath, & Scheper, 2006; Joseph, et al., 2011; McLean, et al., 2007; McLean, Huang, & van den Bogert, 2005; McLean, Lipfert, & van den Bogert, 2004; Pollard, Davis, & Hamill, 2004). Females have been found to perform such tasks with increased hip adduction and internal rotation, decreased knee and hip flexion, increased knee

abduction, and decreased knee flexor increased knee extensor activations. To accompany these differences between genders, there is also the possibility of asymmetries existing between limbs.

There is a lack of consistency in the literature regarding asymmetries in leg dominance as an etiological factor for non-contact anterior cruciate ligament injuries, with injuries rates for the non-dominant leg ranging from 43-67% (Brophy, Silvers, Gonzales, & Mandelbaum, 2010; Matava, Freehill, Grutzner, & Shannon, 2002; Negrete & Schick, 2007). Some conflicting outcomes have been presented in the literature (Borotikar, Newcomer, Koppes, & Mclean, 2008; Brown, Palmieri-Smith, & McLean, 2009; Cowley, Ford, Myer, Kernozek, & Hewett, 2006; Mclean, et al., 2007) making it difficult to know which neuromechanical factors may contribute to risk of non-contact ACL injury. Studies using jump-to-cut (Borotikar, et al., 2008) and drop-jump tasks (Mclean, et al., 2007) found no differences for leg dominance in both female and male collegiate athletes. Contrary to the aforementioned findings, Brown et al. (2009) performed the same jump-cut task as Borotikar et al. (2008), using college-aged recreational athletes, and reported that hip joint angles and knee abduction at initial contact, as well as hip and knee internal rotation angles at peak stance, presented leg dominance differences. Additionally, the non-dominant leg displayed greater joint excursions compared to the dominant leg, of which could enhance the risk of incurring a non-contact ACL injury. In a younger population comprised of high school athletes, a difference was found between dominant and non-dominant legs for peak ground reaction force, as well as knee abduction moment at peak stance (Cowley et al. 2006), both of which have been implicated as influential factors in non-contact ACL injuries (Cerulli, Benoit, Lamontagne, Caraffa, & Liti, 2003; Ford, et al., 2005). The varied results of the



aforementioned studies leave the notion of leg dominance as a risk factor for suffering a non-contact ACL injury unknown.

Through cortical activation analyses, the concept of asymmetries existing due to leg dominance has been supported by studies that examined neural components relative to lower extremity movements (Huda, et al., 2008; Kapreli, et al., 2006; Luft, et al., 2002; Rocca & Filippi, 2010). Functional magnetic resonance imaging (fMRI) allows for the assessment of the role of cortical activity in motor control. fMRI studies of simple joint movements of the hip, knee, and ankle reported that the non-dominant leg produces a more bilateral cortical activation pattern, whereas the dominant leg produces a predominantly contralateral cortical activation pattern (Kapreli, et al., 2006; Luft, et al., 2002). These patterns of cortical activation, relative to leg dominance, enhance the notion that differing supraspinal activation patterns may propagate through neural pathways to the motor units, while still producing similar movements between legs. As fMRI usage requires a participants' head to remain in a static position, therefore the usage of fMRI in dynamic studies is not feasible. To date, one of the most valuable methods to quantify neural activity is electromyography (EMG). As surface EMG lends itself to practical usage during dynamic physical tasks, it has been suggested as an assessment tool for analyzing possible ACL injury risk factors in females (Shultz & Perrin, 1999). In activities such as walking gait, neural asymmetries did exist in EMG data at the individual level, but were masked when the data were grouped (Ounpuu & Winter, 1989). Currently, there is limited research regarding the use of EMG bilaterally during running and cutting tasks. EMG activation patterns of the dominant leg differ between genders for EMG amplitude and timing when adolescent male and female athletes performed multiple athletic maneuvers while examining biomechanical variables and EMG of the

dominant leg (Anderson, et al., 2001; Landry, Mckean, Hubley-Kozey, Stanish, & Deluzio, 2007a, 2007b, 2009). Similar results for the dominant leg have been seen in collegiate athletes when examining EMG activation patterns of the quadriceps and hamstrings muscles relative to timing and amplitude (Beaulieu, et al., 2008, 2009; Bencke & Zebis, 2011; McLean, Borotikar, & Lucey, 2010). Though the aforementioned studies demonstrate the existence of cortical asymmetries in simple motor movements, it is still unknown if such asymmetries perpetuate through neural pathways to differentiate dominant from non-dominant legs.

Neuromechanical factors are also influenced by physiological and psychological components that change dynamically during physical participation. The most noticeable of such alterations during sports participation is fatigue, which has been comprehensively defined as the sensation caused by a central governor within the brain, incorporating external environmental and internal physiological inputs (Eston, Faulkner, St Clair Gibson, Noakes, & Parfitt, 2007; Lambert, St Clair Gibson, & Noakes, 2005; Noakes, St Clair Gibson, & Lambert, 2005; St Clair Gibson & Noakes, 2004). Coupling fatigue with neuromechanical factors has been shown to alter movement and loading patterns that may increase the risk of incurring an ACL injury (Chappell, et al., 2005; Lucci, Cortes, Van Lunen, Ringleb, & Onate, 2011; Mclean, et al., 2007; Quammen, et al., 2012). Additionally, the use of an unanticipated task requires greater cognitive demand, and further influences the alterations to neuromechanical factors brought about by fatigue (Borotikar, et al., 2008; Mclean & Samorezov, 2009). Through the use of different tasks, fatigue has induced alterations to lower extremity neuromechanics in all three planes of motion (Borotikar, et al., 2008; Chappell, et al., 2005; Lucci, et al., 2011; Mclean, et al., 2007; Mclean & Samorezov, 2009;

Quammen, et al., 2012). As these studies demonstrate divergent effects of fatigue on planar alterations, it is important to gain an understanding of fatigues influence on neuromechanics relative to task requirements.

As previously mentioned, neuromuscular factors have the ability to be altered, and injury prevention programs have been developed to attempt to create such alterations. Such prevention programs aim to expose a participant within a limited threshold, allowing for adaptation through experience. These adaptations can be made through strengthening the supporting musculature, developing proprioception, developing neuromuscular coordination, developing correct motor patterns through feedback, or through a combination of any of these factors. Researchers have utilized multiple interventions to attempt to manipulate the controllable mechanisms of an ACL injury, including, but not limited to, altered strength, balance, augmented feedback, or neuromuscular abilities. Strength training has been a focus in order to increase the ability of the hamstrings to counteract the anterior tibial translation of the quadriceps (Gilchrist, et al., 2008; Herman, et al., 2008; Madhavan & Shields, 2007; Mandelbaum, et al., 2005; Pollard, Sigward, Ota, Langford, & Powers, 2006). Programs that have solely focused on strength training have been shown to decrease the co-activation of selected muscle groups about the knee in a controlled weight bearing movement and to allow for a maximal co-contraction to increase the stiffness about the knee during a dynamic movement (Madhavan & Shields, 2007). Biomechanical feedback can be presented in the form of direct verbal feedback from a knowledgeable coach while performing movements, or through the usage of video analysis, using direct observation of a self-performance, or a comparative observation of an expert performer. Training programs that have focused on augmented feedback have been shown to be effective in reducing non-contact ACL injuries

by correcting landing technique and body positioning during dynamic tasks (Greska, Cortes, Van Lunen, & Onate, 2011; Herman, et al., 2009; Hewett, Lindenfeld, Riccobene, & Noyes, 1999; Oñate, et al., 2005; Onate, Guskiewicz, & Sullivan, 2001). Using ballistic movements relative to sports participation, plyometric training has exhibited improvements in neuromuscular control, lower extremity musculature strength, and a reduced risk of suffering an ACL injury (Chimera, Swanik, Swanik, & Straub, 2004; Irmischer, et al., 2004; Lephart, et al., 2005; Myer, Ford, Brent, & Hewett, 2006; Wilkerson, et al., 2004). Balance training, which may or may not require specialized equipment, has also been shown to have a positive effect in reducing non-contact ACL injury risk factors (Caraffa, Cerulli, Progetti, Aisa, & Rizzo, 1996; Holm, et al., 2004; Myklebust, et al., 2003; Paterno, Myer, Ford, & Hewett, 2004; Petersen, et al., 2005; Zebis, et al., 2008), but some studies have also shown that balance training has no effect on preventing non-contact ACL injuries (Söderman, Werner, Pietilä, Engström, & Alfredson, 2000). Most effectively, ACL prevention programs that utilize all-encompassing training methods have been shown to positively alter risk factors related to ACL injuries (Chappell & Limpisvasti, 2008; Gilchrist, et al., 2008; Greska, et al., 2011; Heidt, Sweeterman, Carlonas, Traub, & Tekulve, 2000; Mandelbaum, et al., 2005). These programs have used differing combinations of strength, balance, proprioception, feedback and neuromuscular training. Though efforts have been made to reduce the injury occurrence rate through the implementation of ACL injury prevention programs, the injury rate has not yet demonstrated a decline (Hootman, Dick, & Agel, 2007). As ACL injuries can lead to long-term health impairments, including an early onset of osteoarthritis (Brown, Johnston, Saltzman, Marsh, & Buckwalter, 2006; Kessler, et al., 2008;

Lohmander, Ostenberg, Englund, & Roos, 2004), feasible methods to thwart these injuries are necessary.

## Experiment I

### *Statement of the Problem*

The purpose of this study is to evaluate lower extremity neuromechanical differences in leg dominance of female collegiate soccer athletes during the pre-contact, initial contact, and peak stance phases of an unanticipated sidestep cutting task. The study aims to quantify kinematic data (hip flexion, hip abduction, knee flexion, and knee abduction) and kinetic data (electromyography, vertical ground reaction forces, hip flexion moment, and knee abduction moment) in order to evaluate the lower extremity mechanics between the dominant and non-dominant leg while performing an unanticipated sidestep cutting task.

### *Null Hypothesis*

It is hypothesized that there will be no significant differences in the kinematic (hip flexion, hip abduction, knee flexion, and knee abduction) and kinetic data (electromyography, vertical ground reaction forces, hip flexion moment, and knee abduction moment) for lower extremity mechanics between the dominant and non-dominant leg while performing an unanticipated sidestep cutting task.

### *Research Hypothesis*

The dominant leg will demonstrate significantly higher knee and hip flexion angles, higher hip abduction angles, lower knee abduction angles, lower vertical ground reaction forces, higher hip flexion moments, and lower knee abduction moments than the non-dominant leg during initial contact and peak stance phases of the sidestep cutting task. It is also hypothesized that the dominant leg will display lower peak EMG activation amplitudes

than the non-dominant leg during the pre-contact phase and at initial contact for the unanticipated sidestep cutting task.

The independent variables in this study will be:

Leg dominance: dominant and non-dominant

The dependent variables in this study will be:

Kinematic variables:

Hip flexion angles ( $^{\circ}$ )

Hip abduction angles ( $^{\circ}$ )

Knee flexion angles ( $^{\circ}$ )

Knee abduction angles ( $^{\circ}$ )

Kinetic variables:

Electromyography:

Biceps femoris activation amplitudes (%MVC)

Semitendinosus activation amplitudes (%MVC)

Rectus femoris activation amplitudes (%MVC)

Vastus lateralis activation amplitudes (%MVC)

Vastus medialis activation amplitudes (%MVC)

Guteus medius activation amplitudes (%MVC)

Vertical ground reaction forces (MBw)

Hip flexion moments (Nm/kg·m)

## Knee abduction moments (Nm/kg·m)

The kinematic and kinetic variables will be analyzed at different time instances:

Pre-contact

Initial contact

Peak Stance

## Experiment II

### *Statement of the Problem*

The purpose of this study is to evaluate the effects of fatigue and leg dominance on lower extremity neuromechanics of female collegiate soccer athletes during pre-contact, initial contact, and peak stance during an unanticipated sidestep cutting task. The study aims to quantify kinematic data (hip flexion, hip abduction, hip rotation, knee flexion, knee abduction, and knee rotation) and kinetic data (electromyography, vertical ground reaction forces, hip flexion moment, knee flexion moment and knee adduction moment) pre- and post-fatigue, in order to evaluate lower extremity mechanics between the dominant and non-dominant leg while performing a sidestep cutting task.

### *Null Hypothesis*

It is hypothesized that there will be no significant differences in the pre- and post-fatigue kinematic (hip flexion, hip abduction, hip rotation, knee flexion, knee abduction, knee rotation) and kinetic data (electromyography, vertical ground reaction forces, hip flexion



moment, knee flexion moment and knee adduction moment) for lower extremity mechanics between the dominant and non-dominant leg while performing a sidestep cutting task.

### *Research Hypothesis*

The dominant and non-dominant legs will demonstrate significantly lower knee and hip flexion and external rotation angles, lower hip abduction angles, higher knee abduction angles, lower knee and hip flexion moments, higher internal knee adduction moment, higher vertical ground reaction force, and lower peak electromyographic activation amplitudes during post-fatigue at pre-contact, initial contact and peak stance of the unanticipated sidestep cutting task. The dominant leg will demonstrate significantly higher knee and hip flexion and external rotation angles, higher hip abduction angles, lower knee abduction angles, higher knee and hip flexion moments, lower internal knee adduction moment, lower vertical ground reaction force, and lower peak electromyographic activation amplitudes than the non-dominant leg during post-fatigue at pre-contact, initial contact and peak stance of the unanticipated sidestep cutting task.

The independent variables in this study will be:

Leg dominance: dominant and non-dominant

Fatigue state: pre-fatigue and post-fatigue

The dependent variables in this study will be:

Kinematic variables:

Hip flexion angles (°)

Hip abduction angles ( $^{\circ}$ )

Hip rotation angles ( $^{\circ}$ )

Knee flexion angles ( $^{\circ}$ )

Knee abduction angles ( $^{\circ}$ )

Knee rotation angles ( $^{\circ}$ )

Kinetic variables:

Electromyography:

Biceps femoris activation amplitudes (%MVC)

Semitendinosus activation amplitudes (%MVC)

Rectus femoris activation amplitudes (%MVC)

Vastus lateralis activation amplitudes (%MVC)

Vastus medialis activation amplitudes (%MVC)

Guteus medius activation amplitudes (%MVC)

Vertical ground reaction forces (MBw)

Hip flexion moments (Nm/kg·m)

Knee flexion moments (Nm/kg·m)

Knee abduction moments (Nm/kg·m)

The kinematic and kinetic variables will be analyzed at different time instances:

Pre-contact

Initial contact

Peak Stance

## Experiment III

### *Statement of the Problem*

The purpose of this study is to evaluate the effects of a neuromuscular training intervention on lower extremity mechanical differences in leg dominance of female collegiate soccer athletes during a sidestep cutting task. The study aims to quantify kinematic data (hip flexion, hip abduction, hip rotation, knee flexion, knee abduction, and knee rotation) and kinetic data (electromyography, vertical and posterior ground reaction forces, hip flexion moment, hip abduction moment, knee flexion moment and knee abduction moment) at pre-intervention, post-intervention, and retention, in order to evaluate lower extremity mechanics between the dominant and non-dominant leg while performing a sidestep cutting task.

### *Null Hypothesis*

It is hypothesized that there will be no significant differences in pre-intervention, post-intervention, and retention kinematic (hip flexion, hip abduction, hip rotation, knee flexion, knee abduction, knee rotation, and ankle flexion) and kinetic data (electromyography, vertical and posterior ground reaction forces, hip flexion moment, hip abduction moment, knee flexion moment and knee abduction moment) for lower extremity mechanics between groups for the dominant and non-dominant legs while performing a sidestep cutting task.

### *Research Hypothesis*

At post-intervention, significantly higher knee and hip flexion angles, higher hip abduction and external rotation angles, higher hip flexion moments, higher hip abduction moment, higher trunk anterior flexion and lower trunk lateral flexion compared to pre-intervention, in both the dominant and non-dominant leg at initial contact and peak stance of the unanticipated sidestep cutting task. Also at post-fatigue, significantly higher knee and hip flexion angles, lower hip abduction and external rotation angles, higher hip flexion moments, lower hip abduction moment, lower trunk anterior flexion and higher trunk lateral flexion compared to pre-fatigue, in both the dominant and non-dominant leg at initial contact and peak stance of the unanticipated sidestep cutting task.

The independent variables in this study will be:

Fatigue status: pre-fatigue and post-fatigue

Leg dominance: dominant and non-dominant

Training status: pre-intervention and post-intervention

The dependent variables in this study will be:

Kinematic variables:

Hip flexion angles ( $^{\circ}$ )

Hip abduction angles ( $^{\circ}$ )

Hip rotation angles ( $^{\circ}$ )

Knee flexion angles ( $^{\circ}$ )

Trunk anterior flexion angles ( $^{\circ}$ )

Trunk lateral flexion angles ( $^{\circ}$ )

Kinetic variables:

Hip flexion moments (Nm/kg·m)

Hip abduction moments (Nm/kg·m)

The kinematic and kinetic variables will be analyzed at different time instances:

Initial contact

Peak Stance

### *Operational Definitions*

- Initial contact is defined as the point in time where vertical ground reaction force caused by a foot strike is greater than 10 Newtons (Cortes, et al., 2007; Greska, et al., 2011; Lucci, et al., 2011);
- Pre-contact is defined as 100 ms prior to initial contact (Landry, et al., 2009);
- Peak stance is defined as the first 50% of the stance phase (Borotikar, et al., 2008; Mclean, et al., 2007);
- The dominant leg will be defined as the leg that the subject would use to kick a soccer ball as far as possible (Ford, et al., 2003; Hewett, et al., 2005);
- Sidestep cutting task is a change of direction to the contralateral side of the foot touching the force plate at an angle of approximately 45° (Colby, et al., 2000; Lucci, et al., 2011; Mclean, Huang, et al., 2004; McLean, et al., 2005);
- Fatigue is defined as the sensation caused by a central governor within the brain, incorporating external environmental and internal physiological inputs (Eston, et al., 2007; Lambert, et al., 2005; Noakes, et al., 2005; St Clair Gibson & Noakes, 2004).

*Assumptions*

- The eight high-speed cameras, Model MX-F40 (Vicon Motion Systems Ltd., Oxford, England) and two Bertec Force Plates, Model FP4060-10 (Bertec Corporation, Columbus OH, USA), will be accurately calibrated for each subject and be maintained throughout the experiments;
- The ZeroWire EMG system will maintain a clear signal during each trial and electrical interference will not corrupt the data transmitted;
- The Accutrend Lactate analyzer will maintain its calibration and will report accurate results throughout the experiments;
- The Viasys Vmax Encore metabolic cart will be accurately calibrated for each subject and be maintained throughout the experiments;
- The subjects will give a maximal effort during the graded exercise test;
- The subjects will perform the tasks to the best of their abilities;
- The subjects will not anticipate the randomly cued tasks;
- The subjects will give maximum effort on all components for each fatigue protocol task;
- The subjects will attain a true level of fatigue.

*Limitations*

- The subjects used for this study will be limited to female collegiate soccer athletes;
- Each subject will have varying degrees of sport experience and years of experience that may influence their reaction and cutting skills;

- The measurements will be done in a laboratory setting, and not in a "real life" situation (i.e., practice situation, game, etc.).

#### *Delimitations*

- This study will use collegiate female soccer athletes that must have been exercising at least 30 minutes per day, 3 times per week, for the past 6 months;
- The subjects will range in age from 18-22 years of age;
- Any subject who has had any type of surgery in the lower extremity, ankle and knees, resulting in missing one or more days of sport participation within the previous six months will be excluded from this study;
- Any subject that reports any physical impairment that will limit them in performing a sidestep cutting task will be excluded from this study;
- Any subject who is not cleared by the team physician to practice will be excluded from this study;
- Any subject who is presently pregnant will be excluded from this study;
- Any subject who does not complete both the physiologic and motion analysis testing sessions will be excluded from the study;
- Any subject who does not complete ninety percent (90%) of the training sessions will be excluded from the study.

## CHAPTER II

### REVIEW OF THE LITERATURE

The following review of literature will include anatomy of the ACL, epidemiology of ACL injuries, risk factors associated with non- contact ACL injuries, and prevention programs associated with ACL injuries. With a growth in female athletics participation since the inception of Title IX, there has been a subsequent increase in ACL injuries within this population. Because of this, much research has been undertaken to identify the factors that increase the likelihood of suffering an ACL injury, as well as research focused on the potential ability to reduce the occurrence of ACL injuries. Multiple risk factors have been identified, but others remain questionable, as does the effectiveness of specific injury prevention programs in altering such risk factors. Therefore, there is a need to better understand if proposed risk factors can be altered through injury prevention programs, and if so, can this be done so in differing fatigue states.

#### *Anterior Cruciate Ligament – Anatomy*

The ACL is a band of dense connective tissue that originates from the femur and inserts on the tibia creating stability in the knee, and contributes to anterior tibial translation and axial tibial rotation stability of the knee (Woo, Wu, Dede, Vercillo, & Noorani, 2006). The origin of the ACL initiates from a fossa on the posteromedial surface of the lateral femoral condyle, and runs anterior, medial, and distal to the insertion on the anterior portion of the central tibial plateau (Arnoczky, 1983; Girgis, Marshall, & Monajem, 1975). The ACL spirals laterally from the femur to the tibia, creating a fan-like structure, allowing the tibial



attachment to be wider, and theoretically stronger than the femoral attachment (Girgis, et al., 1975; Zantop, Petersen, Sekiya, Musahl, & Fu, 2006). The attachment sites at the femur and tibia have larger cross-sectional areas than the middle of the ACL (Woo, et al., 2006; Zantop, et al., 2006). The ACL appears as a single band, but is made up of two functional bundles, the anteriomedial and posteriolateral, consisting of fibers ranging from about 22-41 millimeters in length (Amis & Dawkins, 1991; Duthon, et al., 2006; Hara, et al., 2009; Woo, et al., 2006). The anteriomedial bundle originates at the proximal portion of the femur and attaches to the anterior medial portion at the tibial attachment, whereas the posteriolateral portion originates at the distal part of the femoral attachment, and inserts on the posteriolateral part of the tibial attachment. Due to the variation in the length of fibers, previous studies have shown that the ACL dynamically changes tension throughout its range of motion (Amis & Dawkins, 1991; Zantop, et al., 2006). The ACL is mainly vascularized by genicular artery, but the blood supply is not equal throughout the ligament, with three avascular areas (Arnoczky, Rubin, & Marshall, 1979; Petersen & Zantop, 2007). The ACL contains mechanoreceptors that aid in afferent signals during postural changes (Duthon, et al., 2006; Hogervorst & Brand, 1998; Kennedy, Alexander, & Hayes, 1982).

#### *Anterior Cruciate Ligament Injury Epidemiology*

The prevalence of ACL injuries within the general population of the United States is about 1 in 3000 (Griffin, et al., 2000; Owings & Kozak, 1998). Comparatively, the general population in Norway has a prevalence of ACL injuries at a rate of thirty-four people per 100,000, with the greatest affected group ranging from sixteen to thirty-nine years of age, suffering injuries at a rate of eighty-five people per 100,000 (Granán, Forssblad, Lind, &

Engebretsen, 2009). Gianotti and colleagues examined the ACL injury incidence rate within the general population of New Zealand, and revealed an incidence rate of 36.9 per 100,000 people (Gianotti, et al., 2009). Financially, ACL injury repair costs are high, with values ranging from \$17,000 from beginning to end, involving radiology, surgical, and rehabilitation costs (Bonsell, 2000; Cole, et al., 2005; Nagda, Altobelli, Bowdry, Brewster, & Lombardo, 2010).

As the majority of knee injuries occur within athletic tasks, the bulk of current ACL injury research has focused on collegiate athletes. The Injury Surveillance System, implemented by the National Collegiate Athletic Association (NCAA) within college athletics, is a central tracking system supported by the participating university's sports medicine team. Over the reporting period from 1990 to 2002, basketball and soccer athletes suffered 1268 ACL injuries, with male and female athletes incurring 360 and 908 ACL injuries, respectively (Agel, et al., 2005). From the same study, it was found that female soccer and basketball athletes incurred non-contact ACL injuries at rates of 1 in 7692 and 1 in 6250, respectively, whereas males in both sports suffered non-contact ACL injuries at the rate of 1 in 25,000. Overall, a female athlete is more likely to suffer an ACL injury 2.67 times greater than that of a male athlete (Prodromos, Han, Rogowski, Joyce, & Shi, 2007), with surgery rates being twice that of males (Fernandez, Yard, & Comstock, 2007). When looking at the inciting mechanism of ACL injuries, non-contact mechanisms account for seventy percent of all reported injuries (Arendt & Dick, 1995; Boden, Griffin, & Garrett, 2000; Griffin, et al., 2000), and are theorized to occur by means of an interaction between internal and external risk factors (Bahr & Krosshaug, 2005; Meeuwisse, et al., 2007).

*External and Internal Risk Factors.* External risk factors that affect the chance of an athlete incurring an injury include: the rules of the sport (Faunø & Wulff Jakobsen, 2006), coaching, protective equipment involved (McDevitt, et al., 2004), required equipment (Lambson, Barnhill, & Higgins, 1996), and the environment of the sport (Ford, Manson, et al., 2006; Orchard, Chivers, Aldous, Bennell, & Seward, 2005; Orchard, Seward, McGivern, & Hood, 1999, 2001). Internal factors include age, gender, anatomy, and neuromechanical influences. Age becomes a mechanism in relation to maturity level. Prepubescent females are equivalent to their male counterparts in relation to ACL injury rates, and neuromuscular landing patterns (Barber-Westin, Galloway, Noyes, Corbett, & Walsh, 2005; Hass, et al., 2005; Noyes, Barber-Westin, Fleckenstein, Walsh, & West, 2005; Yu, et al., 2005). Post menarche, females demonstrate a lower quadriceps-to-hamstrings ratio (Barber-Westin, et al., 2005; Holm & Vøllestad, 2008) than their male counterparts and retain a valgus knee landing pattern (Noyes, et al., 2005; Yu, et al., 2005) Anatomical factors including the size and shape of the condylar surfaces and the cross-sectional area of the ligaments have been shown to have an influence on ACL injuries. The size and shape of the condylar surfaces focus on the overall width and the size of the intercondylar notch, along with its shape (Anderson, et al., 2001; Shelbourne, et al., 1998). An ACL with a greater cross-sectional area allows for a greater tensile stiffness, limiting anterior tibial translation. Females have a significantly smaller cross-sectional area in comparison to males, which translates to ruptures occurring at 30% less of a loading force on the ACL (Chandrashekar, et al., 2006). Excess flexibility of the knee joint creates anterior knee laxity, allowing for greater anterior tibial translation, increasing the risk of a non-contact ACL injury factor (Shultz, et al., 2004; Shultz, et al., 2006). Though one cannot change their age, gender, or structural anatomy, neuromechanics

are adaptable and can be manipulated in order to reduce the risk of incurring a non-contact ACL injury.

### *Neuromechanical Influences To Anterior Cruciate Ligament Injuries*

*Joint Angles and Moments.* Female athletes have been reported to employ neuromechanical strategies that differ from their male counterparts, possibly placing them at higher risk for non-contact ACL injuries (Decker, Torry, Wyland, Sterett, & Steadman, 2003; Hewett, et al., 2005; Mclean, et al., 2007; Mclean, Lipfert, et al., 2004). Such neuromechanical strategies include smaller ranges of motion within the sagittal plane of the knee and increased vertical ground reaction forces during landing (Decker, et al., 2003; Hashemi, et al., 2010; Hewett, et al., 2005; Lephart, et al., 2005). It has been further argued that sagittal plane motion may not solely be able to rupture the ACL (Hewett, Torg, & Boden, 2009; Mclean, Huang, et al., 2004; Myer, Ford, Mclean, & Hewett, 2006), and that frontal plane knee motion may have a greater influence on non-contact ACL injury risk (Ford, Myer, et al., 2006; Kernozek, Torry, van Hoof, Cowley, & Tanner, 2005; Mclean, Lipfert, et al., 2004; Myer, Ford, Mclean, et al., 2006; Russell, Palmieri, Zinder, & Ingersoll, 2006). As the knee moves medially into an abducted position, the resultant force increases the loading of the ACL, which has been implicated as a non-contact ACL injury risk factor (McLean, et al., 2005; Mclean, Huang, & van den Bogert, 2008; Shin, et al., 2011; Sigward & Powers, 2007). Though the knee is the afflicted joint when a non-contact ACL injury occurs, proximal movement control about the hip has been found to influence risk factors.

Similarly to decreased knee flexion angles, females have also been shown to display decreased hip flexion angles during run-to-cut and stop-jump tasks (McLean, et al., 2005;

Mclean, Lipfert, et al., 2004; Pollard, et al., 2004; Yu, et al., 2005). Other hip motions that have been looked at between genders and thought to be a risk factor are hip abduction angles and hip internal rotation (Chappell, et al., 2007; Pollard, Sigward, & Powers, 2007). Chappell, et al., (2007) found that female recreational athletes have a decreased hip abduction angle and hip external rotation angle when compared to male recreational athletes during landing from a stop-jump task. Comparatively, females have also demonstrated significantly less peak hip abduction than males during cutting tasks in a few studies (Mclean, Lipfert, et al., 2004; Pollard, et al., 2004; Pollard, et al., 2007). As no single neuromechanical factor has been singled out as the main causative factor, it is theorized that a non-contact ACL injury occurs due to a multifactorial mechanism (Nguyen, Shultz, Schmitz, Luecht, & Perrin, 2011; Quatman, Quatman, & Hewett, 2009).

*Leg Dominance.* Relative to neuromechanical risk factors, the concept of leg dominance as an etiological factor for non-contact anterior cruciate ligament injuries has also been proposed, with injuries rates for the non-dominant leg ranging from 43-67% (Brophy, et al., 2010; Matava, et al., 2002; Negrete & Schick, 2007). Some conflicting outcomes have been presented in the literature (Borotikar, et al., 2008; Brown, et al., 2009; Cowley, et al., 2006; Mclean, et al., 2007) making it difficult to know which neuromechanical factors may contribute to risk of non-contact ACL injury. Studies using jump-to-cut (Borotikar, et al., 2008) and drop-jump tasks (Mclean, et al., 2007) found no differences for leg dominance in both female and male collegiate athletes; yet, Brown et al. (2009), performing the same jump-cut task as Borotikar et al. (2008), used college-aged recreational athletes and reported that hip joint angles and knee abduction at initial contact, as well as hip and knee internal

rotation angles at peak stance, presented leg dominance differences which could enhance the risk of incurring a non-contact ACL injury. In a younger population comprised of high school athletes, a difference was found between dominant and non-dominant legs for peak vertical ground reaction force, as well as knee abduction moment at peak stance (Cowley, et al., 2006), both of which are implicated as influential factors in non-contact ACL injuries (Cerulli, et al., 2003; Ford, et al., 2005). The varied results of the aforementioned studies may be a factor related to experience level of the participants chosen. Leg dominance has been supported by neural studies relative to lower extremity movements (Huda, et al., 2008; Kapreli, et al., 2006; Luft, et al., 2002; Rocca & Filippi, 2010). Functional magnetic resonance imaging studies of simple joint movements of the hip, knee, and ankle reported that the non-dominant leg produces a more bilateral cortical activation pattern, whereas the dominant leg produces a predominantly contralateral cortical activation pattern (Kapreli, et al., 2006; Luft, et al., 2002). Other neural studies using surface electromyography (EMG) have shown muscular activation patterns of the dominant leg differing between genders for activation amplitude and timing when adolescent male and female athletes performed multiple athletic maneuvers while examining biomechanical variables and EMG of the dominant leg (Landry, et al., 2007a, 2007b, 2009). As these activation patterns are occurring peripherally, it is currently unknown what cortical processes are occurring during these dynamic tasks.

#### *Physiological Contributions to Anterior Cruciate Ligament Injuries – Fatigue*

Fatigue has been defined in multiple ways as it can be applied with or without a physical component. As a combined theory, St. Clair Gibson et al. (St Clair Gibson &

Noakes, 2004) has defined fatigue as the sensation caused by a central governor within the brain, incorporating external environmental and internal physiological inputs. Coupling fatigue with neuromechanical factors has been shown to alter movement and loading patterns that may increase the risk of incurring an ACL injury (Chappell, et al., 2005; Lucci, et al., 2011; Mclean, et al., 2007; Quammen, et al., 2012). As well, the use of an unanticipated task requires greater cognitive demand, and further influences the alterations to neuromechanical factors brought about by fatigue (Borotikar, et al., 2008; Mclean & Samorezov, 2009).

Through the use of different tasks, fatigue has brought about alterations to lower extremity neuromechanics in all three planes of motion (Borotikar, et al., 2008; Chappell, et al., 2005; Lucci, et al., 2011; Mclean, et al., 2007; Mclean & Samorezov, 2009; Quammen, et al., 2012). Common fatigue effects shown across multiple studies while performing differing tasks include: increased knee valgus moments for females (Carcia, Eggen, & Shultz, 2005; Chappell, et al., 2005), increased peak proximal tibial anterior shear force (Moran & Marshall, 2006), decreased knee flexion angle (Chappell, et al., 2005); increased hip and knee internal rotation angles and moments (Mclean & Samorezov, 2009; Nyland, Caborn, Shapiro, & Johnson, 1999; Sanna & O'Connor, 2008; Tsai, Sigward, Pollard, Fletcher, & Powers, 2009); and increased ground reaction forces (Wikstrom, Powers, & Tillman, 2004).

Strategies witnessed during fatiguing tasks include the knee landing in a more extended position during a landing task, resulting in a decreased knee flexion moment. Other studies have viewed the effects of recovery after undergoing fatigue. In viewing a drop-jump task, altered frontal plane knee mechanics occurred after a bilateral hip abduction protocol, resulting in increased abduction at initial contact during the recovery phase compared to the post-fatigue phase (Carcia, et al., 2005). Similar results were seen in another study, but also

included increased peak knee flexion, abduction and internal rotation angles at initial contact after a 40-minute rest period (Tsai, et al., 2009).

Fatigue may also influence joint laxity and impair mechanoreceptors about the knee. Studies have demonstrated that prolonged physical exertion can significantly increase knee joint laxity (Skinner, Wyatt, Stone, Hodgdon, & Barrack, 1986; Wojtys, Wylie, & Huston, 1996). As well, fatigue has also been shown to decrease the proprioceptive ability of the knee, as well as muscle activation amplitudes and timing (Melnik & Gollhofer, 2007; Rozzi, Lephart, & Fu, 1999). Impairments to muscular activation amplitudes and timing have been exhibited in multiple studies, demonstrating decreased co-contraction for the hamstrings and quadriceps (Padua, et al., 2006; Wright, Ball, & Wood, 2009), as well as increased activation levels for the quadriceps (Kellis & Kouvelioti, 2009; White, Lee, Cutuk, Hargens, & Pedowitz, 2003).

#### *Anterior Cruciate Ligament Injury Prevention Interventions*

Researchers have utilized multiple interventions to attempt to manipulate the controllable mechanisms of an ACL injury and view the outcomes. Those interventions have attempted to alter the strength, balance, biomechanical feedback, or neuromuscular abilities of the participants being studied. Comprehensive training programs, which encompass multiple intervention strategies, have shown the most promise in preventing non-contact ACL injuries.

The study performed by Hewett and colleagues in 1999 (Hewett, et al., 1999), utilized a comprehensive program, consisting of stretching, plyometrics, and strength training, on high school females participating in soccer, basketball, and volleyball. They chose these



sports due to the nature of the sports requiring jumping and cutting maneuvers. The prevention program they utilized was implemented as a preseason-training program that occurred over a 6-week time period. The study utilized three groups, two female groups for intervention and control, and one male group as a population control. Both the female control group and the male control group did not receive any specialized training prior to the start of their respective sports season. The females in the intervention group participated in the injury prevention program three days per week, with each session occurring over a 60 to 90-minute time period. The program was split into three phases, with the initial first two-weeks focusing on proper jumping and landing technique, the second two-weeks focusing on strength, power, and agility, and the final two-weeks focusing on the development of maximal vertical jump height. The outcome of the study revealed that the females athletes in the intervention group had a decreased likelihood of incurring an ACL injury, though still 1.3 to 2.4 times greater than male athletes, compared to the control group of female athletes, who maintained a 4.8 to 5.8 greater likelihood of incurring an ACL injury compared to male athletes. A study by Wilkerson and colleagues (Wilkerson, et al., 2004), utilized the same 6-week injury prevention program developed by Hewett, and sought to examine the strength changes brought about by the program. In that study, the researchers observed a total of twenty-seven National Collegiate Athletics Association (NCAA) Division I female basketball players, with nineteen participating in the injury prevention program and eight serving as controls. The prevention program was once again utilized as a preseason-training program, and the researchers performed concentric isokinetic strength tests for the quadriceps and hamstrings, along with ground impact force tests utilizing a step-down motion and a forward lunging motion. An agility test was also performed to observe for any performance

enhancements due to the injury prevention program. There was a significant increase in hamstrings peak torque at a rate of 60 degrees per second in the prevention program group compared to the control group. This also led to a significant increase in hamstrings-to-quadriceps ratio at a rate of 60 degrees per second in the prevention program group compared to the control group. Wilkerson and colleagues concluded that a structured neuromuscular training program, which includes plyometrics and strength components, induces strength increases within the hamstring muscles.

In a study that only performed low-intensity plyometrics and was designed for the general public (Irmischer, et al., 2004), ground reaction forces were observed to see if the injury prevention program was effective in attenuating the impact force upon landing. The prevention program, which was termed KLIP for knee ligament injury prevention, occurred over nine weeks and used physically active females from a university population. The KLIP program is split into four phases, with an increase in volume occurring through each phase, along with an increase in the skill required to perform the plyometrics exercises. Prior to the engagement of the KLIP, both the KLIP group and the control group performed a step-land protocol from a 0.69-m platform to measure ground reaction forces upon landing. At the onset, there were no significant differences between groups; but upon post-testing, the KLIP group demonstrated significant decreases in peak vertical impact forces and rate of force development. These decreases seen in the KLIP group were also significantly lower than the decreases exhibited by the control group.

In 2005, Mandelbaum and colleagues (Mandelbaum, et al., 2005) utilized an ACL intervention program that was designed as a preactivity warm-up. The Prevent Injury and Enhance Performance (PEP) Program was developed by the Santa Monica Orthopedic and

Sports Medicine Research Foundation, and consists of five components that require a twenty minute period to perform. The five components consist of a warm-up, stretching, strengthening, plyometrics, and agilities. This longitudinal study took place over two years and utilized female soccer athletes ranging in age from 14 to 18 years. The study had 1946 participants during the first year, with 1041 athletes in the PEP group and 1905 athletes in the control group. For the second year, a total of 2757 participants were involved, with 844 athletes in the PEP group and 1913 in the control group. The PEP program was implemented through videotape and a literature packet sent to the participating soccer coach to perform with their team. In the first year, a significant difference was found between the PEP group and the control group in relation to the injury incidence rate per player and per team, with the control group demonstrating a higher incidence with thirty-two ACL tears compared to the PEP group with only two ACL tears. Year two demonstrated the same pattern with a significant difference found between the PEP group, with a total of four ACL tears, and the control group, with a total of thirty-five ACL tears, in relation to the injury incidence rate per player and per team. This study demonstrated the effectiveness of a warm-up geared towards preventing ACL injuries in reducing the occurrence of ACL injuries. A follow up study (Gilchrist, et al., 2008), that utilized the PEP program, was performed as a randomized control study that examined its effect during an entire 12-week NCAA Division I soccer season. The study had a total of 61 teams, with 583 female athletes participating in the PEP group and 852 female athletes participating in the control group. At the onset of the study, there were no significant differences between the two groups. At the conclusion of the study, ACL injuries occurring during practices were significantly greater in the control group compared to the PEP group. In viewing athletes that had a reoccurrence of a non-contact

ACL injury during the study, a significant difference was found between the groups as only the control group exhibited non-contact injuries in this category. The PEP group did not incur any ACL injuries between weeks 6 to 11, which was significantly better than the 5 incurred by the control group. This study reinforced the initial findings of Mandelbaum and colleagues as to the effectiveness of the PEP Program and its ability to reduce the occurrence of ACL injuries. Instead of viewing injury rates, another study (Pollard, et al., 2006) viewed the kinetic and kinematic changes brought about by the PEP Program over an entire soccer season. The participants performed a drop-jump task from a 30cm platform, with each foot landing on an individual force plate. During the soccer season, the PEP program was utilized two to three times per week. Post-season testing showed a significant decrease in hip internal rotation compared to pre-season values. A significant increase was also exhibited for hip abduction between the two testing sessions. This study provided evidence as to how the PEP Program and other neuromuscular intervention programs may alter mechanics to aid in the reduction of ACL injuries.

Another warm-up based ACL injury prevention program was studied by Olsen (Olsen, Myklebust, Engebretsen, Holme, & Bahr, 2005) and colleagues. For this study, the prevention program placed an emphasis on awareness of body positions relating to the knees and ankles and was geared towards team handball athletes. The warm-up intervention was designed to be completed within 15 to 20-minutes and consisted of dynamic warm-up movements, technique in planting, cutting, and jumping movements, balance exercises while on an Airex mat or wobble board, and strength and power exercises with body weight. The study was carried out over an entire handball season of eight months, and used 1837 team handball athletes between the ages of 15 to 17 years. For the intervention group there were

808 female athletes and 150 male athletes, and in the control group there were 778 female athletes and 101 male athletes. At the conclusion of the season, the intervention group had significantly lower occurrences for injuries overall, acute injuries overall, and acute injuries. During the study, 13 of the control clubs had already instituted similar warm-up movements into their programs, and in relation to the other control groups they exhibited a significantly lower incidence of injuries. The study demonstrated a reduction in injury rates in adolescent athletes by utilizing a warm-up injury prevention program. Holm and colleagues (Holm, et al., 2004) used the same warm-up injury prevention program to analyze physiological and biomechanical effects due to the prevention program. Twenty-seven female team handball athletes utilized the warm-up injury prevention program three times per week during the first 5 to 7-weeks, and then once per week through the rest of the season. Performance tests were performed prior to initiating the program, at 8-weeks, and after one full year. The researchers performed a clinical balance test, a knee kinesthesia test, isokinetic quadriceps and hamstrings testing at 60 degrees per second and 240 degrees per second, and three functional hop tests. A significant difference was found for dynamic balance in the athletes, with an increase in performance between the initial test to the test at 8-weeks, and the athletes were able to maintain the elevated level at the one year test. At a velocity of 240 degrees per second, a significant difference was found between the dominant and non-dominant legs in relation to hamstrings strength for both the initial testing and at the one-year testing. The increase in balance performance can be attributed to either a learning effect or the prevention program having a large concentration on balance exercises. As no other performance improvements were exhibited, no evidence exists that the warm-up injury prevention program improves physical performance parameters.

A similar study was performed on adult female handball players participating in an 8-week warm-up injury prevention program (Petersen, et al., 2005). The prevention program was performed three times per week, lasting 10-minutes per session. The prevention program was divided into six phases, with an emphasis at the beginning with an awareness of injury prevention strategies. The warm-up program consisted of balance board and plyometrics exercises and progressed in difficulty as the phases progressed. The researchers classified injury severity into three categories, and any injury that occurred during practice or competition was recorded. At the conclusion of the study, no significant differences were noted between the control group and the prevention program group. Though statistically not significant, it is important to note that the prevention program group did not incur any non-contact ACL injuries, where as the control group had 5 non-contact ACL injuries.

Steffen and colleagues (Steffen, Bakka, Myklebust, & Bahr, 2008) examined the effects of a 10-week intervention using the FIFA “11” lower extremity injury prevention program on performance variables in 17 adolescent female soccer athletes. The FIFA “11” consists of ten exercises that can be completed in fifteen minutes. The ten exercises focus on core stability, balance, plyometrics, and strength, with the 11<sup>th</sup> component focusing on fairplay. Performance tests were performed prior to the initiation of the prevention program and consisted of eccentric and concentric isokinetic strength tests and isometric strength tests for the quadriceps and hamstrings, isometric hip adduction and abduction strength, three vertical jumping tasks, a forty-meter sprint, two speed dribble tests, and a shooting distance test. A significant increase was seen in isometric quadriceps torque at 90 degrees from initial testing to post-testing for the control group. Both groups demonstrated a significant improvement in the straight speed dribble test. The lack of any performance increases can be

attributed to the fact that volume and intensity of the FIFA “11” is not enough to elicit a performance effect.

### CHAPTER III

#### **Experiment I – Bilateral Neuromechanics of the Hip and Knee During an Unanticipated Sidestep Cutting Task**

Title: Bilateral Neuromechanics of the Hip and Knee During an Unanticipated  
Sidestep Cutting Task

Authors: Eric Greska, Stacie Ringleb, Nelson Cortes, Bonnie Van Lunen

Submitting to: Clinical Biomechanics



## *Introduction*

Female collegiate soccer athletes suffer anterior cruciate ligament (ACL) injuries at a rate of 0.31 per 1000 athlete-exposures, with 67% of injuries expected to be from a non-contact mechanism (Agel, et al., 2005). Including surgery and rehabilitation, overall cost of treatment is estimated at \$17,000 per ACL injury (Gottlob, et al., 1999; Owen, et al., 2006). Over the past decade, studies have attempted to identify the biomechanical mechanisms involved in perpetuating a non-contact ACL injury. One of the critical issues that still needs to be addressed is how bilateral leg neuromechanical differences (dominant and non-dominant) may affect patterns during landing and the potential for non-contact ACL injury.

There is a lack of consistency in the literature regarding leg dominance as an etiological factor for non-contact anterior cruciate ligament injuries, with injuries rates for the non-dominant leg ranging from 43-67% (Brophy, et al., 2010; Matava, et al., 2002; Negrete & Schick, 2007). Clinically, it is important to understand if leg dominance is an etiological factor for ACL tears, as this would lead to further development of targeted interventions to address asymmetries between legs. Some conflicting outcomes have been presented in the literature (Borotikar, et al., 2008; Brown, et al., 2009; Cowley, et al., 2006; Mclean, et al., 2007) making it difficult to know which biomechanical factors may contribute to risk of non-contact ACL injury. Studies using jump-to-cut (Borotikar, et al., 2008) and drop-jump tasks (Mclean, et al., 2007) found no differences for leg dominance in both female and male collegiate athletes. Contrary to the aforementioned findings, Brown et al. (2009) performed the same jump-cut task as Borotikar et al. (2008), using college-aged recreational athletes, and reported that hip joint angles and knee abduction at initial contact, as well as hip and knee internal rotation angles at peak stance, presented leg dominance differences.

Additionally, the non-dominant leg displayed greater joint excursions compared to the dominant leg, of which could enhance the risk of incurring a non-contact ACL injury. In a younger population comprised of high school athletes, a difference was found between dominant and non-dominant legs for peak ground reaction force, as well as knee abduction moment at peak stance (Cowley et al. 2006), both of which have been implicated as influential factors in non-contact ACL injuries (Cerulli, et al., 2003; Ford, et al., 2005). The varied results of the aforementioned studies may be a factor related to experience level of the participants chosen, therefore it should be determined if experienced female athletes display leg dominance asymmetries.

The concept of leg dominance has been supported by studies that examined neural components relative to lower extremity movements (Huda, et al., 2008; Kapreli, et al., 2006; Luft, et al., 2002; Rocca & Filippi, 2010). Functional magnetic resonance imaging (fMRI) allows for the assessment of the role of cortical activity in motor control. fMRI studies of simple joint movements of the hip, knee, and ankle reported that the non-dominant leg produces a more bilateral cortical activation pattern, whereas the dominant leg produces a predominantly contralateral cortical activation pattern (Kapreli, et al., 2006; Luft, et al., 2002). These patterns of cortical activation, relative to leg dominance, enhance the notion that differing supraspinal activation patterns may propagate through neural pathways to the motor units, while still producing similar movements between legs. As fMRI usage requires a participants' head to remain in a static position, therefore the usage of fMRI in dynamic studies is not feasible. To date, one of the most valuable methods to quantify neural activity is electromyography (EMG). As surface EMG lends itself to practical usage during dynamic physical tasks, it has been suggested as an assessment tool for analyzing possible ACL injury

risk factors in females (Shultz & Perrin, 1999). In activities such as walking gait, neural asymmetries did exist in EMG data at the individual level, but were masked when the data were grouped (Ounpuu and Winter 1989). Currently, there is limited research regarding the use of EMG bilaterally during running and cutting tasks. EMG activation patterns of the dominant leg differ between genders for EMG amplitude and timing when adolescent male and female athletes performed multiple athletic maneuvers while examining biomechanical variables and EMG of the dominant leg (Landry, et al., 2007a, 2007b, 2009). These studies demonstrated that it is important to gain an understanding of neural activity relative to leg dominance while performing athletic activities, and how such neural activity may relate to injury risk.

There is a lack of evidence on the influence that leg dominance may have to non-contact ACL injury risk. It has been suggested that further research is needed to elucidate the potential contribution of leg dominance to the deleterious position for ACL tears (Shultz, Schmitz, Nguyen, et al., 2010). Therefore, the purpose of this study was to evaluate lower extremity neuromechanical differences in leg dominance of female collegiate soccer athletes during the pre-contact phase, initial contact, and peak stance of a sidestep cutting task. It was hypothesized that neuromechanical non-contact ACL injury risk factors would be more pronounced for the non-dominant leg compared to the dominant leg.

### *Methods*

*Participants.* An *a priori* power analysis, with an exploratory alpha level of 0.05, revealed that a minimum of twenty-four subjects would be required to achieve 80% statistical power (Borotikar, et al., 2008; Mclean, et al., 2007). Twenty-nine participants completed the

study, but nine were removed from analysis due to incomplete EMG data. The study consisted of twenty female collegiate soccer athletes ( $19.2 \pm 1.0$  years,  $1.67 \pm 0.05$  m,  $53.2 \pm 6.2$  kg), with  $13.6 \pm 2.7$  years of experience playing soccer, and at least one season of playing experience at the collegiate level. Inclusion for the study required the participants to be free from any lower extremity injury within the six months preceding the start of the initial testing session, as well as team physician clearance to practice and play. Prior to the testing session, each participant read and signed the informed consent form that was approved by the University's Institutional Review Board.

*Instrumentation.* An 8-camera Vicon motion analysis system (Vicon Motion Systems, Denver, CO), coupled with two Bertec force plates (Bertec Corp., Columbus, OH) and a ZeroWire EMG system (Aurion, Milano, Italy), were utilized for data collection. Kinematic and kinetic data were collected at 250 Hz and 2000 Hz, respectively. The ZeroWire EMG signal latency of 13 ms was preset in the Vicon Nexus software (Vicon Motion Systems, Denver, CO) according to the manufacturers specifications (Aurion, Milano, Italy) to synchronize with the timing of motion and force plate data. The Vicon motion analysis system was calibrated each day prior to testing, and force plates were zeroed prior to the testing of each participant.

*Neuromechanic Analyses.* Participants reported to the Motion Analysis Laboratory, signed the informed consent, and completed a demographic assessment. The participant's weight and height were assessed using a physician's scale with a height rod (Detecto Model #339, Webb City, MO). EMG was setup bilaterally by locating the specified landmarks for electrode placement according to standards set forth by the Surface Electromyography for the Non-Invasive Assessment of Muscles (SENIAM) project (Hermens, et al., 1999). In

particular, the biceps femoris, semitendinosus, rectus femoris, vastus lateralis, vastus medialis, and gluteus medius muscles were assessed. Prior to electrode placement, hair was removed from the site, skin was abraded using a coarse material, and cleaned with isopropyl alcohol. A self-adhesive pre-gelled silver/silver chloride dual electrode (Noraxon USA Inc., Scottsdale, AZ) was placed parallel to the direction of the muscle fibers, directly on the specified landmark. This electrode was utilized due to its preset 20mm spacing between electrode poles, as this distance is congruent with that recommended by SENIAM. The ZeroWire transceivers were attached to their respective electrode and secured to the participant with double-sided tape. Maximum voluntary contractions (MVC) were performed for five seconds according to the SENIAM recommendations for each muscle group to normalize EMG data.

After EMG preparation, forty-four retro-reflective markers were placed bilaterally on the participants' lower extremities. Markers locations were replicated from previous studies (Greska, et al., 2011; Lucci, et al., 2011), with markers added bilaterally to the tibial tuberosity. A dynamic calibration consisting of a "hula motion" performed for three consecutive circles was obtained to estimate hip joint centers (Schwartz & Rozumalski, 2005). Additionally, a static capture was obtained as a reference to calculate angles during dynamic movements. After calibration captures were completed, calibration markers were removed.

The unanticipated sidestep cutting task required the participant to successfully complete three trials using the dominant leg and three trials using the non-dominant leg. Leg dominance was determined by the designation of the leg with which the participant could kick a ball furthest (Ford, et al., 2003). The unanticipation factor of the tasks were controlled

through a custom designed computer program that randomized the trials, and provided a visual cue projected on the wall in front of the participant (Cortes, Blount, Ringleb, & Onate, 2011). A photoelectric gate, set 2 m prior to the front edge of the force plates, sent a signal, when physically occluded, to the computer running the randomization program to randomly display a sidestep cutting task to either the dominant or non-dominant side. Each participant was allotted a minimal approach distance of 6 m prior to the front edge of the force plates. The approach velocity was measured by two sets of digital timing gates (Speed Trap II, Brower Timing Systems, Draper, UT), set 1 m prior to the front edge of the force plates and inline with the front edge of the force plates. White floor tape was placed on the exit side of the force plates, designating an angle between  $35^\circ$  and  $55^\circ$ , in order for the participant to obtain an approximate cutting angle of  $45^\circ$  (Lucci, et al., 2011; Mclean, Huang, et al., 2004). To complete a successful trial of the sidestep cutting task, the participant had to meet three criteria: (1) make contact within the area of a single force plate utilizing the leg opposite of the directional cue given, (2) perform a  $45^\circ$  sidestep cut in the direction of the cue given, and (3) attain a velocity of at least 3.0 m/s prior to making contact with the force plate. All trials were recorded, and those in which the participant failed to meet the three criteria were marked rejected. Testing concluded once the participant attained three valid trials each for the dominant and non-dominant legs.

*Data Reduction.* Data were post-processed within the Vicon Nexus software and transferred to Visual 3D (C-Motion Inc., Rockville MD, USA) to reconstruct a kinematic model; which was used to resolve joint angles and moments from the marker data and ground reaction force. Power spectral density was calculated for kinematic and kinetic data from each trial to obtain optimum cutoff frequencies. Low-pass, fourth-order Butterworth filters

were utilized to filter marker trajectories (7 Hz) and ground reaction force (15 Hz). Three-dimensional hip and knee joint angles were calculated according to the Cardan/Euler rotation sequence. EMG data were also processed within Visual 3D, and prior to rectification, all EMG recordings were high-pass filtered at 30 Hz using a fourth order Butterworth digital filter with zero phase lag (Besier, Lloyd, & Ackland, 2003). The signal was then rectified, low-pass filtered at 6 Hz using a fourth order Butterworth digital filter with zero phase lag, and normalized to the MVC peak value for each muscle (Besier, et al., 2003; Landry, et al., 2009). The highest mean activation amplitude during a 100 ms window of the MVC EMGs was selected for normalization (Landry, et al., 2009).

Within Visual 3D, events were marked for pre-contact, initial contact, and toe-off, where pre-contact to initial contact defined the pre-contact phase and initial contact to toe-off defined the stance phase. Pre-contact was defined as the 100 ms preceding initial contact (Landry, et al., 2009), with initial contact defined as the point at which the vertical impact force on the force plate surpassed a threshold of 10 N, and peak stance was defined as the first 50% of the stance phase (Borotikar, et al., 2008; Brown, et al., 2009; Mclean & Samorezov, 2009). For all trials, data for the pre-contact phase and for the stance phase were each normalized to 101 data points prior to being exported. A custom written Matlab (Mathworks, Natick, MA) program was utilized to define the peak values of the dependent variables during pre-contact and peak stance. Joint angles were measured in degrees, whereas joint moments were normalized to the participants' height and body mass (Nm/Kgm), and ground reaction force data were normalized to body mass (N/Kg). EMG activity was represented as normalized amplitude of the MVC for each muscle (%MVC).

*Statistical Analysis.* Data were input into SPSS (IBM, Armonk, NY) and analyzed for normalcy using a Shapiro-Wilks test to ensure that the dependent measures met the assumption of sphericity. A multivariate analysis of variance (MANOVA) was performed for each of the peak time occurrences, hip and knee rotations and moments, ground reaction force, EMG amplitudes, stance time, and approach velocity to determine the main effect of leg dominance. Effect size was also calculated for each dependent variable (Cohen, 1988). From the numerous dependent variables, a Bonferroni adjusted alpha level of 0.004 was calculated *a priori* to determine significance (McLean, et al., 2007).

### *Results*

Stance time did not differ ( $p > 0.002$ ) between dominant ( $283 \pm 48$  ms) and non-dominant ( $265 \pm 39$  ms) legs, and approach velocity was also equivalent ( $3.26 \pm 0.18$  m/s and  $3.25 \pm 0.16$  m/s, respectively). Similarly, no statistically significant differences ( $p > 0.002$ ) were observed for joint angles (Figure 3-1), moments (Figure 3-2), or vertical ground reaction force between dominant and non-dominant legs at initial contact, peak knee abduction moment, and peak stance (Table 3-1). Though not significant ( $p > 0.002$ ), during peak stance the dominant leg displayed more knee flexion than the non-dominant leg ( $-57.0 \pm 7.1^\circ$  to  $-52.5 \pm 5.8^\circ$ ).

Peak EMG activation levels (%MVC) at pre-contact and initial contact demonstrated no statistically significant differences ( $p > 0.002$ ) between dominant and non-dominant legs (Table 3-2). Notably, the semitendinosus of the non-dominant leg displayed a peak activation that was 17% and 19% greater than that achieved by the dominant leg during the pre-contact phase and at initial contact, respectively. The vastus lateralis of the non-



dominant leg also displayed an EMG activation 19% greater than that of the dominant leg at initial contact. For all other muscles, the peak EMG activation for the dominant and non-dominant legs remained within 10% of one another.

Time occurrence, represented as a percent of the phase, for peak EMG activation during the pre-contact phase and for kinematic and kinetic components of the peak stance phase are reported (Table 3-3). No significant differences ( $p>0.002$ ) were observed between dominant and non-dominant legs for any time occurrence during pre-contact and peak stance. It was noted that the time occurrence of peak EMG activation did not significantly differ between the dominant and non-dominant legs, but the order of occurrence of peak EMG activation between muscles produced unique patterns for the dominant and non-dominant legs.

### *Discussion*

The purpose of this study was to examine neuromechanical differences in leg dominance of female collegiate soccer athletes during an unanticipated sidestep cutting task. Our primary finding was that collegiate female soccer athletes exhibit similar movement patterns between the dominant and non-dominant leg while performing a sidestep cutting task. Though minimal asymmetries were noted between legs, no differences were found to be significant between the dominant and non-dominant legs. It was also noted that peak EMG activation was similar between legs, with a differing pattern observed for the time occurrence of peak activation of muscles between legs. With regards to possible non-contact ACL injury risk factors, the current study did not demonstrate any differentiation of risk relative to leg dominance.

Our results are consistent with previous research examining differences between dominant and non-dominant legs (Borotikar, et al., 2008; Mclean, et al., 2007), which found no significant differences in relation to leg dominance. Kinematic differences were noted between the previous studies and the current study at initial contact and peak stance, but such differences can be attributed to the task chosen. The previous studies used similar female collegiate athletes, but differed in their tasks, utilizing a drop-jump task (Mclean, et al., 2007) and an unanticipated jump-cut task (Borotikar, et al., 2008). As the current task utilized an approach area, our participants could have employed a feed forward mechanism and prepared themselves during the penultimate step, allowing for a more flexed position at initial contact while running (Seyfarth, Geyer, & Herr, 2003). The two previous studies examined the jump-cut task, requiring a participant to land from a set horizontal distance and cut laterally, and the drop-jump task, requiring a participant to land from a set vertical distance and jump vertically, necessitating the participant to absorb an initial impact force prior to a change of direction. The lack of differences in leg dominance for kinematic and kinetic parameters within these studies may be a factor of experience and level of expertise. Supporting this concept, previous studies demonstrating differences in leg dominance during jump-landing and unanticipated cutting tasks utilized high school athletes (Cowley, et al., 2006) or recreationally active participants (Brown, et al., 2009). As an unanticipation factor was utilized in the majority of the studies discussed, it does not appear that such a factor influences differences in leg dominance. Further investigation to define the role of experience in negating differences in leg dominance is warranted.

If such asymmetries witnessed in knee flexion angle at peak stance were to reach a level of significance, the impact may not be substantial because previous studies (Kernozek

& Ragan, 2008; Koga, et al., 2010; Krosshaug, et al., 2007; Shin, Chaudhari, & Andriacchi, 2007) have shown that non-contact ACL injuries occur prior to the time of peak knee flexion within this study. The peak occurred around 40% of the stance phase (Figure 1), equating to approximately 100 ms, for both the dominant and non-dominant legs, with a  $4.5^\circ$  difference between them. Previous studies have demonstrated that a non-contact ACL occurs about 40 ms post-ground contact (Koga, et al., 2010; Krosshaug, et al., 2007), and loading of the ACL occurs over the first 40-60 ms of the landing phase (Kernozek & Ragan, 2008; Shin, et al., 2007). It has also been demonstrated that ACL strain is greatest at knee flexion angles less than  $30^\circ$  (Beynon, Howe, Pope, Johnson, & Fleming, 1992). With the current difference noted between legs occurring at greater than  $50^\circ$  of knee flexion, such a difference may not produce a substantial consequence relative to non-contact ACL injury risk. Though this difference may not seem impactful singularly, its effects when compounded with other potential risk factors likely of causing injury remains unknown and requires further research.

Joint angles and moments presented no asymmetries between legs during initial contact and peak stance phase, yet pre-contact peak EMG activation data displayed noteworthy asymmetries between the dominant and non-dominant legs, as well as a difference in peak EMG activation time occurrence between the muscles. At peak EMG activation during the pre-contact phase, the dominant leg had higher peaks for the biceps femoris and rectus femoris compared to the non-dominant leg, with the opposite being true for the gluteus medius, semitendinosus, vastus medialis, and vastus lateralis. As well, the difference of the time occurrence of the peak EMG activation demonstrates a possible alternate pattern of motor control between the dominant and non-dominant legs. The dominant leg produced a peak activation sequence of: rectus femoris, vastus lateralis, biceps

femoris, gluteus medius, semitendinosus, and vastus medialis; while the non-dominant leg produced a peak activation sequence of biceps femoris, vastus lateralis, semitendinosus, gluteus medius, rectus femoris, and vastus medialis. As motor patterns are controlled at the cortical level, previous imaging studies viewing cortical activity while performing lower extremity movements have demonstrated differing cortical activation patterns between the dominant and non-dominant legs, eluding such differences to filtering processes required to obtain the desired outcome (Kaprili, et al., 2006; Luft, et al., 2002). Such cortical activation processes could clarify the divergences in peak pre-contact EMG activation and occurrence time between legs, while still permitting the execution of similar joint angles and moment at initial contact and peak stance. As imaging technology progresses, future research should include cortical analyses to complement neuromechanical assessments in order to provide a comprehensive representation of the processes leading to the movement selection.

A lack of evidence exists relating EMG bilateral asymmetry to running and cutting activities; therefore previous studies viewing bilateral asymmetry during gait may provide insight into the present study. It has been noted from gait studies that averaging of the participants produces symmetrical outcomes bilaterally, but in turn negates possible individual variability produced during the task (Arsenault, Winter, & Marteniuk, 1986; Ounpuu & Winter, 1989; Yang & Winter, 1985). Though the peak EMG differences were not significant, they displayed large standard deviations, demonstrating possible intra- and inter-individual variability. It is plausible that the proposed intra-individual variability between legs could be caused by the task chosen, but it is reasonable to assume that the studied population is an expert performer of the task. As the pre-contact phase of the task was only analyzed from a period 100 ms prior to initial contact, it is possible that intra-individual

variability stems from temporal factors related to the toe-off from the previous foot strike. Further research is necessary to define the role of the penultimate step and its relation to the neuromechanics propagated during the cutting step.

The calculated low to moderate effect sizes exhibited for the dependent variables reveals that the current study may have been underpowered. With a lack of information regarding leg dominance, the *a priori* power calculation from the few previous studies may have influenced the lack of power, and therefore larger sample sizes may be a necessity in future studies. As well, a potential limitation to the current study is the application of the laboratory-based task to a real world environment. As the sidestep cutting task was performed in the laboratory, the required reaction was limited to two choices when cutting with either the dominant or non-dominant leg. However, the unanticipated nature of the task is more ecologically valid than the aforementioned studies examining differences between leg dominance. Moving such a task to the real world environment for a soccer athlete would increase task complexity as players interact with other players, playing surface, and the environment. Such interactions would multiply the possible movement choices exponentially, possibly bringing about different movement patterns via cortical filtering.

### *Conclusion*

Our primary purpose was to identify neuromechanical differences between dominant and non-dominant legs during a sidestep cutting task. We hypothesized that a difference would be noted for leg dominance, with the non-dominant leg displaying characteristics more apt to incurring a non-contact ACL injury. We found that minimal kinematic and kinetic differences exist between dominant and non-dominant legs for experienced female soccer

athletes, though notable asymmetries for peak EMG activation time occurrences were evident. This suggests that leg dominance within highly experienced athletes does not adversely influence known biomechanical non-contact ACL risk factors, and also demonstrates that similar joint angles and moments between legs may be accomplished through differing neural activity. With a lack of evidence in the literature regarding EMG and leg asymmetries during dynamic tasks, future studies should examine these elements and their role in non-contact ACL injuries. As fatigue has been implicated as a compounding factor that influences non-contact ACL injury risk factors, further studies should also examine neural activity related to the interaction of leg dominance and fatigue.

## CHAPTER IV

### **Experiment II – Influence of Fatigue on Bilateral Lower Extremity Neuromechanics During Sidestep Cutting**

Title: Influence of Fatigue on Bilateral Lower Extremity Neuromechanics During  
Sidestep Cutting

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Submitting to: Journal of Science and Medicine in Sport

### *Introduction*

Some findings reinforce the fact that female collegiate athletes are at a greater disposition than their male counterparts to suffer a non-contact anterior cruciate ligament (ACL) injury (Agel, et al., 2005; Arendt, et al., 1999). For high school and college aged females participating in sports, ACL injuries occur at dramatic rates. In high school soccer and basketball for females, ACL injuries occur at a rate of 1 in 200 athletes, where as their male counterparts are at a rate of 1 in 1000 (Powell & Barber-Foss, 2000). At the collegiate level, this rate increases to 0.32 ACL injuries per 1000 athletic exposures, with an athletic exposure defined as participating in one practice or game session (Mihata, et al., 2006). Putting this rate into perspective, a collegiate soccer team consisting of 25 athletes will participate in over 120 practice and game sessions from pre-season to the spring, leaving a projected rate of approximately one ACL injury per season per team. Though efforts have been made to reduce the injury occurrence rate through the implementation of ACL injury prevention programs, the injury rate has not demonstrated a decline (Hootman, et al., 2007).

ACL injury risk factors can be divided into four categories: neuromuscular and biomechanical, anatomical and structural, hormonal, and screening and prevention (Shultz, Schmitz, Nguyen, et al., 2010). Neuromuscular and biomechanical factors are the main contributing factor to the loading of the ACL, regardless of the other factors (McLean, Huang, et al., 2004; Shin, et al., 2011). Coupling fatigue with neuromechanical factors has been shown to alter movement and loading patterns that may increase the risk of incurring an ACL injury (Chappell, et al., 2005; Lucci, et al., 2011; Mclean, et al., 2007; Quammen, et al., 2012). Additionally, the use of an unanticipated task requires greater cognitive demand, and further influences the alterations to neuromechanical factors brought about by fatigue



(Borotikar, et al., 2008; Mclean & Samorezov, 2009). Through the use of different tasks, fatigue has induced alterations to lower extremity neuromechanics in all three planes of motion (Borotikar, et al., 2008; Chappell, et al., 2005; Lucci, et al., 2011; Mclean, et al., 2007; Mclean & Samorezov, 2009; Quammen, et al., 2012). As these studies demonstrate divergent effects of fatigue on planar alterations, it is important to gain an understanding of fatigues influence on neuromechanics relative to task requirements.

Of the aforementioned studies regarding the effects of fatigue on lower extremity neuromechanics, two studies (Borotikar, et al., 2008; Mclean, et al., 2007) examined the bilateral effects between the dominant and non-dominant legs, and found no differences at pre- and post-fatigue. Currently there is limited research regarding leg dominance and fatigue, but other studies solely viewing the effects of leg dominance have demonstrated conflicting findings (Brown, et al., 2009; Cowley, et al., 2006). Between the contradicting studies, the research performed by Borotikar et al. (Borotikar, et al., 2008) and Brown et al. (Brown, et al., 2009) shared similar tasks with differing collegiate populations. It is presently unknown if experience or skill level negates the possibility of exhibiting dominance in one leg over the other. The notion of leg dominance has been proposed as a possible ACL injury risk factor, but current epidemiological studies are not in agreement as to which limb may be more susceptible to suffering a non-contact ACL injury (Brophy, et al., 2010; Matava, et al., 2002; Negrete & Schick, 2007). As limited data exists regarding bilateral neuromechanics and the influence of fatigue, we propose to further investigate the effects of fatigue on leg dominance relative to ACL injury risk.

Therefore, it is the purpose of this study to evaluate the effects of fatigue and leg dominance on lower extremity neuromechanics of female collegiate soccer athletes during

pre-contact, initial contact, and peak stance during an USCT. It was hypothesized that both legs would demonstrate significantly lower knee and hip flexion and external rotation angles, lower hip abduction angles, higher knee abduction angles, lower knee and hip flexion moments, higher internal knee adduction moment, higher vertical ground reaction force, and lower peak electromyographic activation amplitudes during post-fatigue at pre-contact, initial contact and peak stance of the SCT. It was also hypothesized that the dominant leg would demonstrate significantly higher knee and hip flexion and external rotation angles, higher hip abduction angles, lower knee abduction angles, higher knee and hip flexion moments, lower internal knee adduction moment, lower vertical ground reaction force, and lower peak electromyographic activation amplitudes than the non-dominant leg during post-fatigue at pre-contact, initial contact and peak stance of the SCT.

### *Methods*

A non-randomized experiment was conducted in order to assess the effects of fatigue on neuromechanical properties of the dominant and non-dominant leg. Participants attended two separate laboratory-testing sessions, separated by at least forty-eight hours between sessions. The first session consisted of a graded exercise test (GXT), which was performed to assess the participants' physiological capacity; and the second session used three-dimensional motion analysis to assess the participants' limb neuromechanics while performing USCT's, interspersed with fatiguing tasks.

### *Participants*

An *a priori* power analysis, with an exploratory alpha level of 0.05, revealed that a minimum of twenty-four subjects would be required to achieve 80% statistical power (Borotikar, et al., 2008; Mclean, et al., 2007). Twenty female National Collegiate Athletic Association (NCAA) soccer athletes completed the study ( $19.2 \pm 1.0$  years,  $1.67 \pm 0.05$  m,  $53.2 \pm 6.2$  kg). Participants had  $13.6 \pm 2.7$  years of experience playing soccer, with at least one season of playing experience at the collegiate level. Inclusion for the study required the participants to be free from any lower extremity injury within the six months preceding the start of the initial testing session, as well as team physician clearance to practice and play. Participants were excluded from the study if they had undergone any surgical procedure to the lower extremity within the previous twenty-four months. To differentiate for leg dominance, the dominant leg was designated as the leg with which the participant could kick a ball furthest (Ford, et al., 2003). Prior to the start of each testing session, participants read and signed the informed consent form that was approved by the University's Institutional Review Board.

### *Procedures*

*Graded Exercise Testing Session.* Participants reported to the Human Performance Laboratory, completed a demographic assessment, and height and weight were assessed using a physician's scale with a height rod (Detecto Model #339, Webb City, MO). Participants were fitted with a Polar WearLink+ chest strap transmitter (W.I.N.D. Model, Polar Electro, Kempele, Finland) that telemetered continuous heart rate (HR) data at a rate of 1000 Hz to the Polar ProTrainer 5 software (Polar Electro, Kempele, Finland). A resting HR value was obtained by having participants lie supine for a 5-minute period. At the conclusion

of the resting HR period, a capillary blood sample was obtained via a finger prick for an initial resting lactate level. An approximate 25  $\mu$ l blood sample was required to assess blood lactate levels using an automated portable analyzer (Accutrend, Roche Diagnostics, Mannheim, Germany). Participants then performed a treadmill graded exercise test (GXT) to volitional exhaustion using a computerized system (VMax Encore Metabolic Cart with TrackMaster Treadmill TMX425cp, CareFusion, Waukegan, IL) that analyzed expired gases to determine oxygen consumption ( $\text{VO}_2$ ) and respiratory quotient (RQ). Prior to each test, the metabolic cart was calibrated according to the manufacturer's instructions, using a 2-L syringe and known-composition gases. The protocol for the GXT (Figure 4-1) was developed for the specific population from previous studies that performed incremental stages interspersed with blood draws (Chmura & Nazar, 2010; Denadai, Gomide, & Greco, 2005; Ziogas, Patras, Stergiou, & Georgoulis, 2011). The initial stage of the GXT protocol required the participant to begin at a speed of 5.5 mph with a  $0^\circ$  incline for 3 minutes. At the completion of the first stage, a blood draw stage occurred in which the treadmill slowed to a speed of 4.0 mph for 30 seconds. During the blood draw stage, the participant continued to walk as a researcher performed a finger prick to obtain a blood sample. Concurrently during the blood draw stage, the participant was asked to physically point to a number (1-10) on a color-coded rate of perceived exertion (RPE) scale placed on the front panel of the treadmill. For each blood draw stage, HR, RPE, and blood lactate level were recorded. The cycle of moving between running and blood draw stages continued, with running stages increasing in intensity. Stages 2 and 3 required the participant to run at 7.0 mph and 8.5 mph, respectively, with a  $0^\circ$  incline for 2 minutes. Stages 4 through 8 required the subject to maintain a speed of 8.5 mph, with a 3% grade increase in incline occurring for each 2-minute stage.

Participants were instructed to grab the handrails when they felt that they could no longer continue performing the test. When this occurred, the treadmill was stopped and another blood sample was obtained immediately, as well as RPE and HR measures. The GXT was considered a valid assessment if at least two of the following criteria were met: (a) a plateau in  $\text{VO}_2$  albeit increasing speeds, (b) a HR of at least 90% of age-predicted maximal heart rate ( $207 - 0.7 * \text{age}$ ) (Cleary, et al., 2011; Gellish, et al., 2007), or (c) an RQ above 1.10.  $\text{VO}_{2\text{max}}$  and RQ were determined by averaging the three highest consecutive 20-second interval measurements.

*Motion Analysis Session.* No less than 48-hours after the completion of the GXT session, participants reported to the Motion Analysis Laboratory, were fitted with a Polar WearLink+ chest strap transmitter, and instructed to lie supine for a five-minute period to obtain a resting HR. Bilateral EMG setup was initiated by locating the specified landmarks for electrode placement according to standards set forth by the Surface Electromyography for the Non-Invasive Assessment of Muscles (SENIAM) project (Hermens, et al., 1999) for the biceps femoris (BF), semitendinosus (ST), rectus femoris (RF), vastus lateralis (VL), vastus medialis (VM), and the gluteus medius (GM) muscles. Prior to electrode placement, hair was removed from the site (if needed), the skin was abraded using a coarse material, and then cleaned with isopropyl alcohol. A self-adhesive pre-gelled silver/silver chloride dual electrode (Noraxon USA Inc., Scottsdale, AZ) was then placed parallel to the direction of the muscle fibers directly on the specified landmark. This electrode was utilized due to its preset 20mm spacing between electrode poles, as this distance is congruent with that recommended by SENIAM. The ZeroWire transceivers were then attached to their respective electrode and secured to the subject with double-sided tape. To amplitude normalize the EMG data for the

USCT trials, three maximal voluntary isometric contractions (MVIC) were performed for five seconds according to the SENIAM recommendations for each muscle group.

Upon completion of the EMG preparation, participants were readied for three-dimensional motion analysis by placing forty-four retro-reflective markers bilaterally on the participants' lower extremities. Marker locations were similar to those used in previous studies (Greska, et al., 2011; Lucci, et al., 2011; Quammen, et al., 2012), with additional markers added bilaterally to the tibial tuberosity. Clusters were attached to participants by applying a layer of pre-wrap directly to the skin, placing the cluster on the lateral midline of the segment, and securing the cluster with PowerFlex tape (Andover Healthcare, Salisbury, MA). Individual markers were attached to participants using non-toxic contact cement. Two calibration captures were obtained prior to the initiation of testing. The first calibration capture was a dynamic "hula motion" performed for three consecutive circles and was utilized to estimate internal location of the hip joint centers (Schwartz & Rozumalski, 2005). The second calibration capture was a static frame obtained as a reference to calculate angles during dynamic movements. After calibration captures were completed, calibration markers were removed. Using a countermovement jump, maximal vertical jump height was assessed with a Vertec (Sports Import, Hilliard OH) (Chappell, et al., 2005).

*Unanticipated Sidestep Cutting Task.* The unanticipated SCT required the participant to successfully complete three trials using the dominant leg and three trials using the non-dominant leg. The unanticipation factor of the tasks were controlled through a custom designed computer program that randomized the trials, and provided a visual cue projected on the wall in front of the participant (Cortes, et al., 2011). A photoelectric gate, set 2 m prior to the front edge of the force plates, sent a signal, when physically occluded, to the

computer running the randomization program to randomly display a USCT to either the dominant or non-dominant side. Participants were allotted a minimal approach distance of 6 m prior to the front edge of the force plates. The participants velocity was measured by two sets of digital timing gates (Speed Trap II, Brower Timing Systems, Draper, UT) set 1 m prior to, and inline with, the front edge of the force plates. White floor tape was placed on the exit side of the force plates, designating an lane between 35° and 55°, in order for the participant to obtain an approximate cutting angle of 45° (Mclean, Huang, et al., 2004). To complete a successful trial of the USCT, the participant had to meet three criteria: (1) make contact within the area of a single force plate utilizing the leg opposite of the directional cue given, (2) perform a 45° sidestep cut in the direction of the cue given, and (3) attain a velocity of at least 3.0 m/s prior to making contact with the force plate. All trials were recorded, and those in which the participant failed to meet the three criteria were marked as rejected. Once participants obtained three valid trials for the dominant and non-dominant legs, they transitioned to the dynamic fatigue protocol.

*Dynamic Fatigue Battery Protocol.* The dynamic fatigue battery was a general fatigue model consisting of anaerobic tasks familiar to the participants, designed to bring about volitional exhaustion through repetition. Each set of the dynamic fatigue battery consisted of: five countermovement jumps at 90% of their maximum vertical jump, five squats, a 5-10-5 agility drill (Jones, Matthews, Murray, van Raalte, & Jensen, 2010), and step-ups on a 30 cm box. Participants began the protocol by performing five countermovement jumps at a pace of one jump per second, followed by five squats that required their thighs to reach a level parallel to the floor. The participant then completed the 5-10-5 agility drill, followed by step-ups on a 30-cm for 20 seconds, performing a cadence of up-up-down-down paced

with a metronome set at 240 bpm. After fulfilling the step-ups, the participant was asked to physically point to a number (1-10) on the same color-coded RPE scale utilized during the GXT session.

Upon completion of one set of the dynamic fatigue battery, participants transitioned back to completing another round of six valid unanticipated trials (three dominant and three non-dominant). The cycle between USCTs and the dynamic fatigue battery continued until the participant attained physiological fatigue. Physiological fatigue was determined from the participants' criterion RPE and HR data acquired during the GXT. Upon attaining the same, or higher, RPE from the GXT, and the HR either (a) met the same maximal value from the GXT, or (b) reached a plateau for three consecutive dynamic fatigue battery bouts, the participant was deemed physiologically fatigued. This method of assessing fatigue was chosen as previous studies have shown relationships between  $VO_{2max}$ , blood lactate, HR, and RPE (Coutts, Rampinini, Marcora, Castagna, & Impellizzeri, 2009; Davies, Rowlands, & Eston, 2008; Hetzler, et al., 1991), and their implications on fatigue (Eston, et al., 2007; Joseph, et al., 2008).

*Instrumentation and Data Processing.* An 8-camera Vicon motion analysis system (Vicon Motion Systems, Denver, CO), coupled with two Bertec force plates (Model FP4060-NC, Bertec Corp., Columbus, OH) and a ZeroWire EMG system (Aurion, Milano, Italy) were utilized for data collection. Kinematic and kinetic data were collected at 250 Hz and 2000 Hz, respectively. ZeroWire EMG signal latency of 13 ms was preset in the Vicon Nexus software (Vicon Motion Systems, Denver, CO) according to the manufacturers specifications, (Aurion, Milano, Italy) to synchronize with the timing of the motion and force plate data.



Data were post-processed in Vicon Nexus software and transferred to Visual 3D (C-Motion Inc., Rockville MD, USA ) to reconstruct a model and resolve joint angles and moments from the marker data. Power spectral densities were calculated to determine cutoff frequencies for kinematic, kinetic, and EMG data. Marker trajectories from each trial were filtered at a cutoff frequency of 7 Hz (low-pass, fourth-order Butterworth filter), and three-dimensional hip and knee joint angles were calculated according to the Cardan/Euler rotation sequence. The ground reaction force was filtered through a low-pass, fourth-order Butterworth filter at a cutoff frequency of 15 Hz in order to minimize possible impact peak errors. EMG data were also processed within Visual 3D, and prior to rectification, all EMG recordings were high-pass filtered at 30 Hz using a fourth order Butterworth digital filter with zero phase lag (Besier, et al., 2003). The signal was then rectified, low-pass filtered at 6 Hz using a fourth order Butterworth digital filter with zero phase lag, and normalized to the MVIC peak value for each muscle (Besier, et al., 2003; Landry, et al., 2009). The highest mean activation amplitude during a 100 ms window of the MVIC EMGs was selected for normalization (Landry, et al., 2009).

Within Visual 3D, events were marked for pre-contact, initial contact, and toe-off, with pre-contact to initial contact defining the pre-contact phase and initial contact to toe-off defining the stance phase. Initial contact was defined as the point at which the vertical impact force on the force plate surpassed a threshold of 10 N, with pre-contact defined as the 100 ms preceding initial contact (Landry, et al., 2009), and peak stance defined as the first 50% of the stance phase (Borotikar, et al., 2008; Brown, et al., 2009; Mclean & Samorezov, 2009). For all trials, data for the pre-contact phase and for the stance phase were each normalized to 101 data points prior to being exported. Custom written MatLab (MathWorks,

Natick, MA) code was utilized to define the peak values of the dependent variables during the pre-contact phase and peak stance. Data were then averaged for the three trials of the dominant leg and non-dominant legs. Joint angles were measured in degrees, whereas internal joint moments were normalized to body mass and height (Nm/Kgm), and ground reaction force data was normalized to the participants' body mass (N/Kg). EMG activity was represented as a normalized amplitude of the MVIC for each muscle (%MVIC). In viewing criterion HR and RPE data, agreement ratios were calculated between the GXT and dynamic fatigue battery sessions.

*Statistical Analysis.* Data were input into PASW (IBM, Armonk, NY) and analyzed descriptively for normalcy using a Shapiro-Wilks test. Separate 2 (leg dominance) x 2 (fatigue status) repeated measures ANOVA's were used to assess the effect of fatigue for approach velocity, stance time, vertical ground reaction force, and each of the hip, knee, and EMG dependent variables in relation to the main and interaction effects of fatigue and leg dominance. For the peak time occurrence of the dependent variables, separate 2 (leg dominance) x 2 (fatigue status) repeated measures ANOVA's were also used to assess the effect of fatigue and leg dominance. To determine significance, a Bonferroni corrected alpha level of 0.003 was utilized to account for the extent of dependent measures, and minimize the probability of type I error. When a significant difference was exhibited, post-hoc analysis using Tukey's honestly significant difference test was performed.

### *Results*

Descriptive physiologic data for the GXT and dynamic fatigue battery are presented in Table 4-1. Similar criterion HR and RPE values were obtained between testing sessions

for the graded exercise test and dynamic fatigue battery ( $p > 0.003$ ), with agreement ratios demonstrating equivalent individual HR demands and RPE scores between the two sessions.

To obtain the required number of successful trials, at pre-fatigue the participants performed an average of  $19 \pm 7$  USCTs, and an average of  $13 \pm 5$  post-fatigue. On average, participants completed  $5 \pm 1$  bouts of the dynamic fatigue battery, while performing  $69 \pm 15$  total USCTs from pre- to post-fatigue, with a completion time of  $19.2 \pm 3.8$  minutes. Though not significant ( $p > 0.003$ ), approach velocity increased from pre- to post-fatigue for the dominant ( $3.25 \pm 0.16$  m/s to  $3.27 \pm 0.11$  m/s) and non-dominant ( $3.26 \pm 0.18$  m/s to  $3.27 \pm 0.13$  m/s) legs; with a subsequent decrease occurring for the stance time of the dominant ( $283 \pm 48$  ms to  $260 \pm 39$  ms) and non-dominant ( $265 \pm 39$  ms to  $248 \pm 22$  ms) legs.

Descriptive statistics with associated  $p$ -values for peak electromyographic activation during the pre-contact phase are presented in Table 4-2. The peak EMG activation (%MVIC) for the GM demonstrated a significant decrease ( $p < 0.001$ ) from pre- to post fatigue for both the dominant and non-dominant legs. All other findings for the pre-contact phase were not significant ( $p > 0.003$ ).

EMG activation levels at initial contact displayed similar alterations relative to activation patterns seen during the pre-contact phase, with the GM demonstrating the only significant decrease ( $p = 0.002$ ) from pre- to post fatigue for both the dominant and non-dominant legs (Table 4-3). For joint angles, significant differences were demonstrated for the main effect of fatigue at initial contact (Table 4-4). A significant decrease occurred from pre- to post-fatigue for hip flexion ( $p < 0.001$ ), resulting in a more extended position for both the dominant and non-dominant leg (Figure 4-2). No other significant differences were exhibited

for the main effect of leg dominance or for the interaction effect of leg dominance and fatigue at initial contact ( $p > 0.003$ ).

Kinematics and kinetics of the hip and knee at peak internal knee adduction moment are presented in Table 4-5. The main effect of fatigue brought about significant differences for sagittal hip and knee joint angles, displaying a significant decrease ( $p = 0.001$ ) in the hip flexion angle from pre- to post-fatigue produced  $6^\circ$  and  $3^\circ$  alterations for the dominant and non-dominant legs, respectively. The decrease in knee flexion angle was significant ( $p = 0.001$ ) for the dominant leg from pre- to post-fatigue. No other significant differences were displayed for the main effect of leg dominance or for the interaction effect of leg dominance and fatigue at peak internal knee adduction moment ( $p > 0.003$ ).

At peak stance (Table 4-6), significant differences were exhibited for the main effect of fatigue with decreases in peak knee flexion ( $p < 0.001$ ) and peak hip flexion ( $p < 0.001$ ) angles following fatigue. From pre- to post-fatigue, the peak hip flexion moment displayed significant differences for both legs ( $p < 0.001$ ), with the dominant leg and non-dominant legs undergoing 0.25 Nm/Kgm and 0.15 Nm/Kgm increases, respectively (Figure 4-3). Both legs also demonstrated significant increases ( $p < 0.001$ ) for the peak vertical ground reaction force. No other significant differences were exhibited for the main effect of leg dominance or for the interaction effect of leg dominance and fatigue ( $p > 0.003$ ).

### *Discussion*

The purpose of this study was to evaluate the effects of fatigue and leg dominance on lower extremity neuromechanics in female collegiate soccer athletes during a USCT. The primary finding was that fatigue brought about significant some changes for hip and knee

sagittal neuromechanics in female collegiate soccer athletes when performing an unanticipated sidestep task. Regardless of fatigue status, it was also observed that female collegiate soccer athletes display similar movement patterns between the dominant and non-dominant legs. Though movement patterns were similar, minimal asymmetries were observed for joint angles and moments between the dominant and non-dominant legs. EMG activation also displayed minimal asymmetries between legs, but demonstrated different patterns of time occurrence for peak activation between dominant and non-dominant legs, and produced further alterations to such patterns due to the influence of fatigue.

For the current study, a general fatigue model similar to previous studies (Borotikar, et al., 2008; Mclean, et al., 2007) was chosen to simulate movements relevant to soccer participation that could bring about fatigue through peripheral and central fatigue mechanisms. As the perceptual and physical impact of fatigue demonstrate variability between participants (Eston, et al., 2007), both objective and subjective measures of fatigue were utilized within the current study. Relating the collected criterion HR and RPE values from the GXT to those attained during the dynamic fatigue battery allowed for quantification of volitional exhaustion. The criterion HR agreement ratio demonstrated objectively similar values achieved during both testing sessions, with the criterion RPE agreement ratio slightly higher for the dynamic fatigue battery, suggesting it as a more demanding task than the GXT. Using this approach possibly negates fitness level differences between participants and provides an accurate assumption of volitional exhaustion. Though this method may provide an ecologically valid approach to inducing fatigue and analyzing concurrent movement patterns, it is not possible to discern the degree to which peripheral and central fatigue mechanisms influenced the movement alterations observed.

The decreases constituted by fatigue in GM muscle activation at pre-contact and initial contact, ~20% and ~15% respectively, may influence the risk of injury. As the main function of the GM is to stabilize frontal plane motion of the femur and pelvis (Neumann, 2010), such decrements in neural activation could allow for undesired movements within the frontal plane. However, we have not found altered frontal movements suggesting that our athletes may have been able to compensate for the decreased muscle activation. McMullen et al. (2011) found that a decreased EMG activation level of the GM, induced by fatigue, impairs postural control, as well as movement quality, during single limb balance tasks. Though the current study utilized a cutting task, such modifications to postural control could affect the participants' stance limb as they perform a change of direction. Relative to the GM muscle, decreased strength has been related to increased knee abduction angles in females (Jacobs, Uhl, Mattacola, Shapiro, & Rayens, 2007), as well as an increased risk of injury for both genders (Leetun, Ireland, Willson, Ballantyne, & Davis, 2004). As force production cannot be directly inferred from surface EMG activation due to motor unit depth and recruitment strategies (Farina, Fosci, & Merletti, 2002), it is unclear as to the extent the decrement in GM muscle activation impacts lower extremity risk within the current study. It is unclear if fatigue impaired the muscles ability to achieve greater activation amplitudes, or if our observed decrease is a result of an altered recruitment strategy selected to maintain movement homeostasis.

Relative to a change in recruitment strategy, it was noted that during the pre-contact phase the time occurrence of peak EMG activation for each muscle differed between legs, and further altered the time occurrence pattern post-fatigue. The dominant leg displayed a peak EMG activation pattern of RF, VL, BF, GM, ST, and VM pre-fatigue, and GM, BF,

VL, VM, ST, and RF post-fatigue; whereas the non-dominant leg displayed a pattern of BF, VL, ST, GM, RF, and VM pre-fatigue, and VL, RF, BF, ST, VM, and GM post-fatigue. These alterations within peak activation patterns arose within a 20 ms time epoch that occurred approximately 50 ms prior to initial contact for both pre- and post-fatigue. It is not possible to determine the degree of how such neural alterations would impact movement patterns as only peak EMG activation was examined, and other factors such as motor unit conduction velocity, total muscle activation time, and spatial positioning of the participant prior to initial contact must be accounted for. Further analysis is warranted to better understand the influence of fatigue on neural components of the lower extremity and the relationship to injury risk.

Accompanying the neural decrements seen about the hip at initial contact, hip flexion also decreased from pre- to post-fatigue. Similar adaptations occurred for both legs, leaving the participants in a more extended position within the sagittal plane. These findings are in agreement with previous studies examining the effects of fatigue during cutting (Borotikar, et al., 2008; Lucci, et al., 2011) and landing (Kernozek, Torry, & Iwasaki, 2008; Quammen, et al., 2012) tasks. It is possible that the decrease in hip flexion is relative to trunk control in order to maintain postural stability while performing a change of direction. In viewing video data relative to non-contact ACL injuries, Boden et al. (2009), found that participants who sustained non-contact injuries incurred increased hip flexion directly after initial contact. The researchers postulated that the increased hip flexion may impair the ability of the hip musculature to attenuate forces transferred down through the upper body. It has been further theorized that increased trunk flexion at initial contact during a cutting task would cause the trunk to rotate forward, shifting the center of pressure under the stance foot in an anterior and

lateral direction, further increasing frontal plane knee joint loads (Kipp, Mclean, & Palmieri-Smith, 2011; McLean, et al., 2005). As trunk flexion was not analyzed in the current study, it is uncertain if the decrease in hip flexion post-fatigue was brought about by performance limiting factors of fatigue, or if the decrease was due to a behavior modification enacted to maintain trunk control.

Alternatively, an extended position about the hip may increase non-contact ACL injury risk. Previous studies (Hashemi, et al., 2010; Shultz, Nguyen, Leonard, & Schmitz, 2009) have proposed that an extended hip, flexing slower relative to a co-flexing knee, would cause anterior tibial translation, directly loading the ACL. Hashemi et al. (2010) further proposed that coupling this with possible impaired or delayed activation of the quadriceps and hamstring muscles would most likely cause the ACL to fail. Though the current study did not demonstrate significant decrements in quadriceps and hamstrings muscles activation post-fatigue, other studies (Gehring, Melnyk, & Gollhofer, 2009; Zebis, et al., 2010) have shown that fatigue is capable of bringing about such impairments. In further discussion of the quadriceps and hamstrings muscles relative to hip flexion, a decrease in hip flexion shortens the biceps femoris (Visser, Hoogkamer, Bobbert, & Huijing, 1990), limiting its effective tension (Withrow, Huston, Wojtys, & Ashton-Miller, 2008); whereas increased hip flexion would have a similar effect on the rectus femoris. It has been previously demonstrated that decreased hamstring strength relative to quadriceps strength may be a risk factor in female athletes (Myer, et al., 2009), and such an alteration to the effective tension, due to an extended position at the hips, may create a functional strength imbalance. Considering the possible mechanisms presented for loading the ACL due to an extended position in the sagittal plane, it is plausible that our participants experienced an increased



load on the ACL. As non-contact ACL injuries have been proposed to occur through multifactorial mechanisms, the neuromechanical control displayed in other planes by our participants may have enabled the attenuation of the speculated sagittal loading.

It is noteworthy that the primary changes occurred in the sagittal plane while no changes were noted for the transverse and frontal plane. Previous research reported that fatigue primarily affects frontal and transverse plane (Borotikar, et al., 2008; Mclean, et al., 2007; Mclean & Samorezov, 2009). However, there is a body of literature that demonstrates the effects of fatigue mainly on the sagittal plane (Chappell, et al., 2005; Lucci, et al., 2011; Quammen, et al., 2012). The contradiction in the literature may derive from several factors; i) sample, ii) fatigue protocol, and iii) task requirements. Population samples for the aforementioned studies were composed of participants who participated in soccer, volleyball, and/or basketball, with Chappell et al. (2005) using recreationally active college students, and the rest of the studies using NCAA athletes. The fatigue protocols varied between studies, with exception to those selected by Lucci et al. (2011) and Quammen et al. (Quammen, et al., 2012). The studies that demonstrated frontal and transverse plane changes utilized predominantly sagittal plane movements requiring deep squatting positions. Borotikar et al. (2008) recognized that such a task may fatigue the gluteal muscles, leading to decreased transverse plane control about the hip. The subsequent increased internal rotation of the hip has been correlated with a resultant increase in knee abduction load (McLean, et al., 2005), which could possibly explain the alterations seen in frontal and transverse planes brought about by the squat-based fatigue protocols. Similar to the aforementioned protocols, the procedures chosen for the studies (Lucci, et al., 2011; Quammen, et al., 2012) that mainly displayed sagittal plane alterations also involved specific movements that required the

participants to performing squatting motions. Knowing this, one could conclude that similar outcomes relative to the frontal and transverse plane alterations should have been displayed, and therefore the fatigue protocol chosen does not appear to be the most influential factor affecting plane alterations. It must also be noted that Lucci et al. (Lucci, et al., 2011) and Quammen et al. (Quammen, et al., 2012) performed two fatigue protocols, an aerobically based linear run and an anaerobically based protocol consisting of squatting movements coupled with agility and sprint drills, yet similar alterations incurred through fatigue were demonstrated by both protocols.

In viewing the task requirements, the studies demonstrating alterations due to fatigue within the frontal and transverse planes utilized jump-to-cut and drop-jump tasks, while studies demonstrating sagittal plane alterations utilized sidestep cutting and stop-jump tasks. Tasks requiring a landing from a jump, necessitate the participant to attenuate the ground reaction force generated upon landing, prior to applying an action force to create a change of direction. Comparatively for tasks using an initial running approach, increased hip flexion may cause decreased frontal and transverse torque production from the gluteal muscles (Ward, Winters, & Blemker, 2010); therefore it stands to reason that an optimal level of hip flexion must be maintained in order to sustain efficient frontal and transverse plane control, as well as sagittal plane force production, about the hip when performing change of direction tasks while running. Relative to the previously mentioned studies of Hashemi et al. (2010) and Shultz et al. (2009), such optimal positioning of the hip while running may dictate a necessitated decrement in knee flexion, as to avoid direct loading of the knee. Another point to contemplate is the fact that based on computer modeling (Mclean, Huang, et al., 2004), and prospective (Hewett, et al., 2005) studies, the sagittal plane cannot tear the ACL, and that

knee valgus angle and loading are the main predictors of injury. Therefore, we would suggest that despite the changes produced by fatigue on the sagittal plane, our sample was not at increased risk of injury. Fatigue may negatively influence lower extremity biomechanics, but not to the threshold for rupture in highly trained individuals.

In viewing leg dominance in relation to fatigue, the findings of the current study are in agreement with previous research (Borotikar, et al., 2008; Mclean, et al., 2007). Though slight neuromechanical asymmetries were present between the dominant and non-dominant legs at both pre- and post-fatigue, the lack of significant differences within these studies may be related to experience or level of expertise. The previous studies, as well as the current study, utilized collegiate athletes, whereas those studies that identified differences in leg dominance examined recreationally active college students (Brown, et al., 2009) and high school athletes (Cowley, et al., 2006). If leg dominance is related to experience, it is important to define the threshold at which it is negated. In doing so, injury prevention programs can be tailored to meet the differing needs between novice and expert athletes.

The actual number of participants examined in the current study was less than the number calculated in the *a priori* power analysis, and *post hoc* power analyses subsequently reveal that the study was moderately underpowered relative to the frontal and transverse plane variables previously seen to demonstrate significant changes. As well, we cannot directly determine the level of fatigue at which alterations actually occurred, due to the study's design (pre- and post-fatigue). As previous studies (Borotikar, et al., 2008) have analyzed movement alterations in tandem with the evolution of fatigue, such insight, coupled with a continuous measure of HR, could possibly allow for the determination of an objective physiological threshold at which neuromechanical alterations occur. The laboratory setting

may have also limited the current study, as the participants interacted with a finite virtual environment. As the number of possible choices would multiply exponentially during a real world practice or game setting for the participants, cognitive resources may be allocated differently, and the subsequent neuromechanical alterations may differ from those displayed in the current study. Future research should seek alternative methods to assess neuromechanics during actual practice or game settings.

### *Conclusion*

The primary purpose of the current study was to determine the effects of fatigue and leg dominance on lower extremity neuromechanics of female collegiate soccer athletes during a USCT. It was hypothesized that fatigue would alter neuromechanical factors, with differences also being displayed between the dominant and non-dominant legs at pre- and post-fatigue, increasing the participants risk of incurring a non-contact ACL injury. We found that fatigue decreased gluteus medius muscle activation at pre-contact and initial contact, and caused sagittal plane alterations about the hip and knee throughout the stance phase in both legs. As these findings agree with previous studies using similar tasks, this suggests that alterations in neuromechanics brought about by fatigue are task specific. Implementing this information into ACL injury prevention programs would be useful, allowing the coach to emphasize specific planar movement cues during discrete tasks as an athlete becomes fatigued (i.e. focusing on sagittal plane movement during running tasks, or frontal and transverse plane movement during plyometric tasks). ACL injury prevention programs should also attempt to maintain pre-fatigue movement patterns as the athlete undergoes varying levels of fatigue. As minimal asymmetries were found between the

dominant and non-dominant legs for our female collegiate soccer athletes, it is not likely that leg dominance influences ACL injury risk in highly experienced athletes. In order to understand the impact of leg asymmetries on possible ACL injury risk factors, future studies should examine the role that experience plays in negating differences in leg dominance.

**CHAPTER V****EXPERIMENT III– The Effects Of An Injury Prevention Program Implemented Post-Fatigue On Lower Extremity And Trunk Neuromechanics**

Title: The Effects Of An Injury Prevention Program Implemented Post-Fatigue On Lower Extremity And Trunk Neuromechanics

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Submitting to: Journal of Strength and Conditioning Research

### *Introduction*

Female collegiate soccer athletes suffer anterior cruciate ligament (ACL) injuries at a rate of 0.31 per 1000 athlete-exposures, with 67% of injuries expected to be from a non-contact mechanism (Agel, et al., 2005). Studies have reported that a high percentage of overall injuries occur towards the mid to late stages of games and practices (Ostenberg & Roos, 2000; Rahnama, Reilly, & Lees, 2002; Rampinini, Impellizzeri, Castagna, Coutts, & Wisloff, 2009). Thus, it is speculated that athletes should present a high level of fatigue that is declining the neuromechanical control. Lately, research has focused on understanding the effects that fatigue may have in increasing the risk of injury (Borotikar, et al., 2008; Chappell, et al., 2005; Lucci, et al., 2011). To counteract injury risk factors, injury prevention programs have demonstrated efficacy in altering neuromechanical control (Greska, et al., 2011; Myer, Ford, Palumbo, & Hewett, 2005; Myers & Hawkins, 2010), and therefore may provide a mechanism to attenuate the effects of fatigue.

The majority of injury prevention programs are designed as either warm-up programs (Gilchrist, et al., 2008; Mandelbaum, et al., 2005), or independent training sessions (Hewett, et al., 1999), and have shown success in reducing non-contact ACL injuries. These prevention programs aim to alter the participants' neuromechanical traits by improving strength, motor control, and proprioception. These are all factors that are susceptible to the detrimental effect of fatigue, and it has been suggested that injury prevention programs incorporate methods to circumvent such detrimental effects (Borotikar, et al., 2008; Lucci, et al., 2011; Mclean, et al., 2007). Though it is questionable if such a concept is conceivable, as previous studies have demonstrated positive (Arnett, DeLuccia, & Gilmartin, 2000; Williams, Daniell-Smith, & Gunson, 1976) and negative (Anshel & Novak, 1989; Berger &

Smith-Hale, 1991) effects of skill transfer in performing motor tasks while physically fatigued. According to the specificity hypothesis (Henry, 1968), replicating the desired skill and context should lead to the most effective outcome. The current study aims to implement an injury prevention program, intervening post-fatigue, to assess the effectiveness of motor learning and skill transfer while fatigued. Little evidence exists to support or denounce such an application of training in an attempt to curtail the unfavorable effects of fatigue, therefore it is necessary to obtain an understanding of motor skill acquisition under a state of fatigue, with implications on reducing ACL risk factors associated with fatigue.

Neuromuscular and biomechanical factors are the main contributing factors to the loading of the ACL (McClean, Huang, et al., 2004; Shin, et al., 2011), and when coupled with fatigue, altered movement and loading patterns have been demonstrated that may increase the risk of incurring an ACL injury (Chappell, et al., 2005; Lucci, et al., 2011; McClean, et al., 2007). Fatigue has brought about alterations to lower extremity neuromechanics in all three planes of motion (Borotikar, et al., 2008; Lucci, et al., 2011; McClean, et al., 2007) through the use of differing tasks, and therefore, it is important to gain an understanding of the discrete influence fatigue imparts on neuromechanics relative to task requirements. Implementing a higher cognitive demand, by utilizing an unanticipated task, further influences the alterations to neuromechanical factors brought about by fatigue (Borotikar, et al., 2008; McClean & Samorezov, 2009). As well, the effect of fatigue on bilateral neuromechanics is not well understood, as few articles have examined this notion (Borotikar, et al., 2008; Brown, et al., 2009; Cowley, et al., 2006). Regardless of fatigue, such studies have demonstrated a lack of differentiation between legs in collegiate athlete populations (Borotikar, et al., 2008), (McClean, et al., 2007), with other studies contradicting such findings in differing active



populations (Brown, et al., 2009), (Cowley, et al., 2006). Therefore, it is important to understand the role of leg dominance relative to fatigue status with regards to ACL injury risk.

Therefore, the purpose of this study was to evaluate the effects of a neuromuscular training intervention on lower extremity neuromechanics relative to fatigue and leg dominance in female collegiate soccer athletes during an unanticipated sidestep cutting task (USCT). It was hypothesized that at post-intervention there would be significantly higher knee and hip flexion angles, higher hip abduction and external rotation angles, higher hip flexion moments, higher hip abduction moment, higher trunk anterior flexion and lower trunk lateral flexion compared to pre-intervention, in both the dominant and non-dominant leg at initial contact and peak stance of the USCT. It was also hypothesized that at post-fatigue there would be significantly higher knee and hip flexion angles, lower hip abduction and external rotation angles, higher hip flexion moments, lower hip abduction moment, lower trunk anterior flexion and higher trunk lateral flexion compared to pre-fatigue, in both the dominant and non-dominant leg at initial contact and peak stance of the USCT.

### *Methods*

A non-randomized quasi-experiment was conducted in order to assess the effects of a training intervention on neuromechanical properties of the dominant and non-dominant leg. The study occurred over a twelve-week period during the 4-month collegiate soccer off-season. Participants partook in a 10-week injury prevention intervention, with laboratory pre-testing occurring one week prior to the initial intervention session, and post-testing occurring within one week of concluding the final session of the intervention. A National Strength and

Conditioning Association certified strength and conditioning specialist (CSCS) performed all sessions of the injury prevention intervention.

### *Participants*

Initially, seventeen female collegiate soccer athletes from a National Collegiate Athletic Association (NCAA) Division I university were recruited for the study, with three being removed due to incurring lower extremity injuries during the study time period, which included an ankle sprain, medial meniscal tear and a lateral collateral ligament sprain. Fourteen athletes ( $19.1 \pm 1.0$  years,  $1.67 \pm 0.05$  m,  $63.6 \pm 6.0$  kg) completed the study, having  $13.6 \pm 2.2$  years of experience playing soccer, with at least one season of playing experience at the collegiate level. Inclusion for the study required the participants to be free from any lower extremity injury six months prior to and throughout the study, as well as team physician clearance to practice and play. Participants were also required to participate in 100% of the training sessions to be included in the study. Prior to the start of each testing session, participants read and signed the informed consent form that was approved by the University's Institutional Review Board.

### *Procedures*

*Graded Exercise Testing Session.* Participants reported to the Human Performance Laboratory, completed a demographic assessment, and height and weight were assessed using a physician's scale with a height rod (Detecto, Webb City, MO). Participants were fitted with a Polar WearLink+ chest strap transmitter (Polar Electro, Kempele, Finland) that telemetered continuous heart rate (HR) data at a rate of 1000 Hz to the Polar ProTrainer 5

software (Polar Electro, Kempele, Finland). A resting HR value was obtained by having participants lie supine for a 5-minute period. At the conclusion of the resting HR period, a capillary blood sample was obtained via a finger prick for an initial resting lactate level. An approximate 25  $\mu$ l blood sample was required to assess blood lactate levels using an automated portable analyzer (Accutrend, Roche Diagnostics, Mannheim, Germany). Participants then performed a treadmill graded exercise test (GXT) to volitional exhaustion using a computerized system (VMax Encore Metabolic Cart and TrackMaster Treadmill, CareFusion, Waukegan, IL) that analyzed expired gases to determine oxygen consumption ( $\text{VO}_2$ ) and respiratory quotient (RQ). Prior to each test, the Viasys VMax Encore Metabolic Cart was calibrated according to the manufacturer's instructions, using a 2-L syringe and known-composition gases. The protocol for the GXT (Figure 5-1) was developed for this specific population from previous studies that performed incremental stages interspersed with blood draws (Chmura & Nazar, 2010; Denadai, et al., 2005; Ziogas, et al., 2011). The initial stage of the GXT protocol required the participant to begin at a speed of 5.5 mph with a  $0^\circ$  incline for 3 minutes. At the completion of the first stage, a blood draw stage occurred in which the treadmill slowed to a speed of 4.0 mph for 30 seconds. During the blood draw stage, the participant continued to walk as a researcher performed a finger prick to obtain a blood sample. Concurrently during the blood draw stage, the participant was asked to physically point to a number (1-10) on a color-coded rate of perceived exertion (RPE) scale placed on the front panel of the treadmill. For each blood draw stage, HR, RPE, and blood lactate level were recorded. The cycle of moving between running and blood draw stages continued, with running stages increasing in intensity. Stages 2 and 3 required the participant to run at 7.0 mph and 8.5 mph, respectively, with a  $0^\circ$  incline for 2 minutes. Stages 4

through 8 required the subject to maintain a speed of 8.5 mph, with a 3% grade increase in incline occurring for each 2-minute stage. Participants were instructed to grab the handrails when they felt that they could no longer continue performing the test. When this occurred, the treadmill was stopped and another blood sample was obtained immediately, as well as criterion RPE and HR measures. The GXT was considered a valid assessment if at least two of the following criteria were met: (a) a plateau in  $VO_2$  albeit increasing speeds, (b) a HR of at least 90% of age-predicted maximal heart rate ( $207 - 0.7 * \text{age}$ ) (Cleary, et al., 2011; Gellish, et al., 2007), or (c) an RQ above 1.10.  $VO_{2\max}$  and RQ were determined by averaging the three highest consecutive 20-second interval measurements.

*Motion Analysis Session.* Participants reported to the Motion Analysis Laboratory after a minimum of 48-hours post-GXT session, were fitted with a Polar WearLink+ chest strap transmitter, and instructed to lie supine for a five-minute period to obtain a resting HR. Sixty-three retro-reflective markers were placed bilaterally on the participants' upper and lower extremities to ready them for three-dimensional motion analysis. Upper extremity markers were placed at the medial and lateral joint line of the wrist, lateral joint line of the elbow, acromio-clavicular joint, forearm, and upper arm for both arms, as well as the manubrium, xyphoid process, right scapular spine, 7<sup>th</sup> cervical vertebrae, 10<sup>th</sup> thoracic vertebrae, 1<sup>st</sup> lumbar vertebrae, 5<sup>th</sup> lumbar vertebrae, and left and right transverse processes of the 3<sup>rd</sup> lumbar spine to define the torso. Lower extremity marker location landmarks included the posterior superior iliac spine, iliac crest, greater trochanter, thigh, medial joint line of the knee, lateral joint line of the knee, shank, tibial tuberosity, medial malleolus, lateral calcaneus, heel, head of the first metatarsal, head of the fifth metatarsal, intermediate

cuneiform, and the peroneal trochlea of the calcaneus. Clusters of non-collinear markers attached to rigid backing plates were utilized to designate the thigh and shank, with the thigh composed of five markers and the shank composed of four markers. Clusters were attached to participants by applying a layer of pre-wrap directly to the skin, placing the cluster on the lateral midline of the segment, and securing the cluster with PowerFlex tape (Andover Healthcare, Salisbury, MA). Individual markers were attached to participants using non-toxic contact cement. Two calibration captures were obtained prior to the initiation of testing. The first calibration capture was a dynamic “hula motion” performed for three consecutive circles and was utilized to calculate an estimated internal location of the hip joint centers (Schwartz & Rozumalski, 2005). The second calibration capture was a static frame obtained as a reference to calculate angles during dynamic movements. After calibration captures were completed, calibration markers were removed.

*Anaerobic Power Tasks.* To assess anaerobic performance adaptations, maximal vertical jump height and bilateral maximal horizontal single leg hop distances were acquired. Maximal vertical jump height was assessed during a countermovement jump using a Vertec (Sports Import, Hilliard OH) (Chappell, et al., 2005). Prior to performing the jump, the participants reach height was assessed by having the participant stand under the Vertec while extending the dominant arm overhead. Participants were instructed to keep their shoulders parallel to the ground while reaching overhead. Participants performed three maximal countermovement jumps and the highest value was recorded in centimeters. A single leg hop task similar to previous studies (Myer, et al., 2005) was utilized to assess unilateral leg power. To control for upper extremity acceleration, participants were instructed to keep their

arms crossed in front of their sternum. If the landing was not held for 3 seconds, or the participants arms came uncrossed, the trial was not counted. Participants completed 3 valid attempts for each leg, and the highest value was recorded in centimeters.

*Unanticipated Sidestep Cutting Task.* The USCT required the participant to successfully complete three trials using the dominant leg and three trials using the non-dominant leg. Leg dominance was determined by the designation of the leg with which the participant could kick a ball furthest (Ford, et al., 2003). The unanticipation factor of the tasks were controlled through a custom designed computer program (Cortes, et al., 2011) that randomized the trials, and provided a visual cue projected on the wall in front of the participant. A photoelectric gate, set 2 m prior to the front edge of the force plates, sent a signal, when physically occluded, to the computer running the randomization program to randomly display a USCT to either the dominant or non-dominant side. Participants were allotted a minimal approach distance of 6 m prior to the front edge of the force plates. The participants velocity was measured by two sets of digital timing gates (Speed Trap II, Brower Timing Systems, Draper, UT) set 1 m prior to, and inline with, the front edge of the force plates. White floor tape was placed on the exit side of the force plates, designating an lane between 35° and 55°, in order for the participant to obtain an approximate cutting angle of 45° (McClean, Huang, et al., 2004). To complete a successful trial of the USCT, the participant had to meet three criteria: (1) make contact within the area of a single force plate utilizing the leg opposite of the directional cue given, (2) perform a 45° sidestep cut in the direction of the cue given, and (3) attain a velocity of at least 3.0 m/s prior to making contact with the force plate. All trials were recorded, and those in which the participant failed to

meet the three criteria were marked as rejected. Once participants obtained three valid trials for the dominant and non-dominant legs, they transitioned to the dynamic fatigue protocol.

*Dynamic Fatigue Battery Protocol.* The dynamic fatigue battery was a general fatigue model consisting of anaerobic tasks familiar to the participants, designed to bring about volitional exhaustion through repetition. Each set of the dynamic fatigue battery consisted of: five countermovement jumps at 90% of their maximum vertical jump, five squats, a 5-10-5 agility drill (Jones, et al., 2010), and step-ups on a 30 cm box. Participants began the protocol by performing five countermovement jumps at a pace of one jump per second, followed by five squats that required their thighs to reach a level parallel to the floor. The participant then completed the 5-10-5 agility drill, followed by step-ups on a 30-cm for 20 seconds, performing a cadence of up-up-down-down paced with a metronome set at 240 bpm. After fulfilling the step-ups, the participant was asked to physically point to a number (1-10) on the same color-coded RPE scale utilized during the GXT session.

Upon completion of the first set of the dynamic fatigue battery, participants transitioned back to completing another round of six valid unanticipated trials (three dominant and three non-dominant). The cycle between USCTs and the dynamic fatigue battery continued until the participant attained physiological fatigue. Physiological fatigue was determined from the participants' criterion RPE and HR data acquired during the GXT. Upon attaining the same, or higher, RPE from the GXT, and the HR either (a) met the same maximal value from the GXT, or (b) reached a plateau for three consecutive dynamic fatigue battery bouts, the participant was deemed physiologically fatigued. This method of assessing fatigue was chosen as previous studies have shown relationships between  $\text{VO}_{2\text{max}}$ , blood

lactate, HR, and RPE (Coutts, et al., 2009; Davies, et al., 2008; Hetzler, et al., 1991), and their implications on fatigue (Eston, et al., 2007; Joseph, et al., 2008).

*Injury Prevention Intervention.* The injury prevention intervention occurred over a 10-week period during the collegiate soccer off-season. The soccer team maintained its normal off-season training schedule, which consisted of four training sessions per week comprised of two weight room sessions and two team practice sessions. The team practice sessions focused on skill and tactical components, with no discrete physical performance drills performed. To maximize the coach to athlete ratio, participants were split into two groups for the intervention sessions, with each group participating in one session per week, performing the same drills with no discrepancy between groups. All intervention sessions were performed at the end of the team practice session, preceded by a conditioning session purposefully implemented to bring about fatigue. The conditioning sessions lasted approximately 20 minutes and consisted of agility and sprint drills, emphasizing physical exertion. The intervention session immediately followed the conditioning session and lasted approximately 20 minutes as well, and consisted of plyometric drills (Greska, et al., 2011; Mandelbaum, et al., 2005; McCann, Cortes, van Lunen, Greska, & Onate, 2011) and agility drills used in previous studies (Lephart, et al., 2005; Mandelbaum, et al., 2005) (Table 5-1). The intervention protocol emphasizes maximum effort and power while maintaining proper body angles during acceleration and jumping, along with body positioning and force attenuation during deceleration and landing. Agility drills were performed with interspersed change of direction commands, visually cued through hand signals provided by the CSCS. This was done to mimic game-like decision-making tasks that rely solely upon visual stimuli.



Throughout the intervention training sessions, the CSCS performed visual demonstrations and verbalized cues for performing the drills properly, as well as providing constant augmented feedback to the participants and group concerning their performances.

### *Instrumentation and Data Processing*

An 8-camera Vicon motion analysis system (Vicon Motion Systems, Denver, CO) was coupled with two Bertec force plates (Bertec Corp., Columbus, OH) for data collection. Kinematic and kinetic data were collected at 250 Hz and 2000 Hz, respectively. The Vicon motion analysis system was calibrated each day prior to testing, and the force plates were zeroed prior to the testing of each participant.

Data were post-processed in Vicon Nexus software and transferred to Visual 3D (C-Motion Inc., Rockville MD, USA) to reconstruct a model and resolve joint angles and moments from the marker data. Angles for trunk anterior and lateral flexion were calculated relative to the pelvis. Power spectral densities were calculated to determine cutoff frequencies for kinematic and kinetic data. Marker trajectories from each trial were filtered at a cutoff frequency of 7 Hz (low-pass, fourth-order Butterworth filter), and three-dimensional hip and knee joint angles were calculated according to the Cardan/Euler rotation sequence. The ground reaction force was filtered through a low-pass, fourth-order Butterworth filter at a cutoff frequency of 15 Hz in order to minimize possible impact peak errors.

Within Visual 3D, events were marked for initial contact and toe-off, with initial contact to toe-off defining the stance phase. Initial contact was defined as the point at which the vertical impact force on the force plate surpassed a threshold of 10 N, and peak stance

was defined as the first 50% of the stance phase (Borotikar, et al., 2008; Brown, et al., 2009; Mclean & Samorezov, 2009). For all trials, data for the stance phase were normalized to 101 data points prior to being exported. Custom written MatLab (The Mathworks, Natick, MA) code was utilized to define the peak values of the dependent variables during peak stance. Data were then averaged for the three trials of the dominant leg and non-dominant legs. Joint angles were measured in degrees, whereas internal joint moments were normalized to body mass and height (Nm/Kgm), and ground reaction force data was normalized to the participants' body mass (N/Kg). Angles for trunk anterior and lateral flexion were calculated relative to the pelvis

#### *Statistical Analysis*

Data was input into PASW (IBM, Armonk, NY) and analyzed descriptively for normalcy to ensure that the participants met the assumption of sphericity. Separate 2 (training status) x 2 (fatigue status) x 2 (leg dominance) repeated measures ANOVA's were used to assess the effects of training for each of the hip, knee, and trunk dependent variables in relation to the main and interaction effects of training, fatigue, and leg dominance. To determine significance, a Bonferroni corrected alpha level of 0.006 was utilized to account for the extent of dependent measures. Tukey's HSD test was used *post hoc* for existences of statistically significant differences to determine exactly where they occurred.

#### *Results*

Though increases for body composition and physical performance measures were displayed from pre- to post-training, none were found to be significant ( $p > 0.006$ ).

Participants increased body mass from pre- to post-training ( $63.6 \pm 6.0$  kg to  $65.1 \pm 5.8$  kg), with a subsequent decrease in body fat percentage ( $21.4 \pm 4.0$  % to  $20.0 \pm 4.0$  %), resulting in an average increase in  $2.1 \pm 1.2$  kg of lean body mass. For the GXT,  $\text{VO}_{2\text{max}}$  increased from pre-to post-training ( $50.0 \pm 3.1$  ml/kg/min to  $50.9 \pm 3.2$  ml/kg/min), and blood lactate concentrations decreased for the criterion ( $11.9 \pm 1.3$  mmol/L to  $10.8 \pm 3.1$  mmol/L) and the recovery ( $10.7 \pm 2.3$  mmol/L to  $9.6 \pm 2.7$  mmol/L) measures. Anaerobic power tests demonstrated increases from pre- to post-training for the vertical jump ( $46.4 \pm 7.9$  cm to  $48.3 \pm 5.9$  cm), and single leg hop for both the dominant ( $151.6 \pm 18.7$  cm to  $159.6 \pm 13.5$  cm) and non-dominant ( $146.6 \pm 16.8$  cm to  $159.2 \pm 15.7$  cm) legs.

Descriptive statistics for kinematic and kinetic data during the USCT are presented in Table 5-2. At initial contact, alterations in hip angles and moments demonstrated significant differences ( $p < 0.006$ ). There was a main effect for fatigue with significant decreases in hip flexion ( $p = 0.002$ ) and abduction ( $p = 0.006$ ) angles following the protocol (Figure 5-2). Training also demonstrated a significant main effect for hip flexion angle ( $p < 0.001$ ), causing decreased flexion post-training. A significant interaction effect for dominance and training was exhibited for hip abduction moment ( $p = 0.003$ ), with pre-training measures of the dominant leg shifting to an abduction moment ( $0.11 \pm 0.11$  Nm/kgm to  $-0.01 \pm 0.10$  Nm/kgm), and the non-dominant leg shifting to an increased adduction moment ( $0.04 \pm 0.12$  Nm/kgm to  $0.09 \pm 0.09$  Nm/kgm) post-training (Figure 5-3). No significant differences ( $p > 0.006$ ) were noted for knee or trunk angles.

At peak internal knee adduction moment, significant differences ( $p < 0.006$ ) were noted for knee and hip angles, as well as hip moments. Hip ( $p = 0.004$ ) and knee ( $p = 0.001$ ) flexion and hip abduction ( $p = 0.006$ ) angles displayed significant decreases brought about by

fatigue, at both pre- and post-training. As at initial contact, participants also demonstrated a significant decrease in hip flexion angle ( $p = 0.003$ ) from pre- to post-training. The main effect of fatigue also demonstrated significant differences for hip flexion ( $p = 0.005$ ) and abduction ( $p = 0.005$ ) moments. Hip flexion moment displayed a decreased extension moment from pre- to post-fatigue, whereas hip abduction moment displayed a decreased adduction moment from pre- to post-fatigue.

At peak stance, significant differences were noted for the main effects of fatigue and training. A significant decrease ( $p = 0.001$ ) was exhibited for hip flexion from pre- to post-training, as well as for the main effect of fatigue. Knee flexion ( $p < 0.001$ ) and hip abduction ( $p = 0.002$ ) angles exhibited significant decreases from pre- to post-fatigue, with similar values maintained from pre- to post-training. From pre- to post-fatigue, peak stance hip internal rotation angle significantly decreased ( $p = 0.003$ ) at pre- and post-training, whereas peak stance hip external rotation angle shifted significantly ( $p = 0.006$ ) shifted from an internal to an external rotation angle post-training. For both legs, hip flexion ( $p < 0.001$ ) moment displayed a significant increase from pre- to post-fatigue, with similar values for pre- and post-training attained at post-fatigue. The main effect of training demonstrated significant decreases in hip flexion ( $p = 0.001$ ) angle, with similar alterations for both the dominant and non-dominant leg. No significant differences ( $p > 0.006$ ) were presented for trunk angles.

### *Discussion*

The purpose of this study was to evaluate the influence of a fatigue-based training intervention relative to fatigue and leg dominance for lower extremity neuromechanics in

female collegiate soccer athletes during an USCT. Though post-intervention measurements of body composition and physical performance displayed improvements, the effectiveness of the training intervention to alter neuromechanics relative to non-contact ACL injury risk factors remains questionable. Alterations to sagittal hip angles, induced by the effect of training, demonstrated decreased hip flexion angles at initial contact and peak internal knee adduction moment, which may potentially increase non-contact ACL injury risk. It is possible that direct loading of the ACL could occur from such a position of extension about the hips, if the coupled knee were to flex at a relatively faster rate, causing anterior tibial translation (Hashemi, et al., 2010; Shultz, et al., 2009). Within the current study, post-training knee flexion angles decreased at initial contact, but remained relatively unchanged at peak internal knee adduction moment. As the change in hip flexion angle from initial contact to peak internal knee adduction moment remained fairly unchanged, this gives rise to the possibility that our participants experienced a co-flexing knee moving at a faster rate than the hips, increasing the possibility of incurring a non-contact ACL injury.

Contrary to such theory, it is possible that the decrease in hip flexion was relative to trunk control in order to maintain postural stability while performing a change of direction. In a previous study (Boden, et al., 2009) viewing video footage of non-contact ACL injuries, increased hip flexion angle directly after initial contact was seen in participants who sustained non-contact injuries. Kipp, et al. (2011) further theorized that during a cutting task, a relationship exists between increased hip flexion angle and an increased frontal load at the knee, caused by forward flexion and rotation of the trunk away from the direction of the cut. Though alterations to trunk flexion angles were not found to be significant within the current study, the alterations exhibited may support the concept of reduced hip flexion to maintain

trunk control. Frontal plane trunk angles consistently demonstrated movement favoring the direction of the USCT and remained unchanged from pre- to post-training, but sagittal plane trunk angles moved to a more anterior direction post-training. The current study examined a sidestep cutting task, but in viewing landing strategies during a single leg landing, it has been shown that slight anterior flexion of the trunk may lead to a decreased knee-extensor moment, with a subsequent increased hip-extensor moment, more effectively decelerating the body and increasing stability of the knee (Shimokochi, Yong Lee, Shultz, & Schmitz, 2009). As well, the consistent contralateral lean, relative to the stance limb, during the USCT demonstrates adequate frontal plane postural control of the hip, potentially decreasing loading of the knee (Winter, 2009). It is possible that the minor changes in anterior trunk positioning were induced by the training intervention, since visually cued movement perturbations were implemented during the agility drills. While previous studies focused on training static core control have not produced trunk control alterations (Jamison 2012), the use of unanticipated sport specific movements may provide a viable stimulus to develop necessary trunk control and warrants further research.

To corroborate the control about the hip aiding in unloading forces at the knee, McLean et al. (McLean, et al., 2005) demonstrated that increased hip flexion and/or internal rotation during sidestep cutting are directly related to increased knee abduction loads. As well, it has been demonstrated that the sole predictor of a knee abduction motion during an USCT is hip adduction (Imwalle, Myer, Ford, & Hewett, 2009). Though changes in hip internal rotation angle were only found to be significant at peak stance in the current study, decreases were noted throughout the stance phase and frontal plane hip movement consistently remained in a hip abducted position. It is possible that the alterations brought

about by the main effect of training allowed for a more protective positioning within all planes of the hip, resulting in decreased loading at the knee and potentially reducing the likelihood of sustaining a non-contact ACL injury.

The training intervention demonstrated minor success moving toward mitigating the detrimental effects initiated by fatigue (Borotikar, et al., 2008; Lucci, et al., 2011; Mclean, et al., 2007). At initial contact, divergent alterations occurred between legs for the hip abduction moment post-training. When subjected to fatigue, the dominant limb maintained its hip abduction moment, whereas the non-dominant limb moved toward a less adducted moment. It is unclear as to why the dominant limb displayed alterations that favor protective mechanisms relative to non-contact ACL injury risk factors, while the non-dominant limb shifted towards an unfavorable position post-training, as both limbs received the same volume of training. The favorable shift from a greater hip adduction moment to a lesser hip adduction moment or hip abduction moment was replicated throughout the stance phase for both legs from pre-to post-fatigue. Contradictory to the aforementioned favorable change, the main effect of fatigue produced alterations, similar to previous studies (Borotikar, et al., 2008; Lucci, et al., 2011; Mclean, et al., 2007), in all planes about the hip, as well as decreased knee flexion, throughout the stance phase. As these detrimental alterations have been linked to possible non-contact ACL injury risk factors, it is important to discover techniques that may be utilized in injury prevention programs to hinder such decrements.

Relative to the intended outcomes sought by the implementation of the injury prevention program in attenuating fatigue-induced alterations, frontal plane hip angles did not produce anticipated results. It was believed that hip abduction would remain similar during pre- and post-fatigue after the prevention program, yet similar effects induced by

fatigue and training were seen for both legs. In agreement with previous studies (Sanna & O'Connor, 2008), the hip abduction angle decreased post-fatigue, yet remained in an abducted position. It is possible that this may represent a compensatory mechanism to control frontal plane loading of the hips, and maintain proper movement mechanics in light of fatigue. The hip abduction moment demonstrates a decrease in the peak internal hip adduction moment post-fatigue (Fig 5-3), and it is possible that the relative decrease in hip abduction angle may permit for such a loading decrement. If this theory were to hold true, attempting to alter such fatigue-induced movement patterns through training may inadvertently increase undesired loading at the hip, which could possibly translate into an increased ACL injury risk. Further research is necessary to determine the role of specific neuromechanical adaptations while fatigued, as well as their implications to non-contact ACL injury risk.

The current intervention sought to execute performance tasks while in a state of fatigue, while providing augmented feedback to the participants regarding movement patterns and body positioning. As previous studies have demonstrated success in motor skill transfer while training under fatigue (Arnett, et al., 2000; Williams, et al., 1976), it was believed that the participants would be able to replicate the movement patterns attained while performing the drills of the prevention program, during dynamic tasks when fatigued. Such a notion may have been hindered during the application of the injury prevention program, as no prior knowledge of performance was provided during a non-fatigued state. Receiving instruction and feedback during a non-fatigue state may have been a more optimal learning environment, allowing the participants to gain an understanding of proper and improper movement patterns. Due to the fatigued status when learning was supposedly taking place,



the allocation of cognitive processes may have been impeded, not allowing for movement acquisition. This may have directly affected the participant's ability to alter their movement pattern during the drills, or the ability to retain the altered pattern beyond the duration of the individual injury prevention program sessions. It is also possible that the lack of an altered movement pattern post-training may be attributed to a deficiency within the amount of stimulus. As the intervention was only performed one time per week, the stimulus may not have been enough to elicit alterations in lower extremity neuromechanics over the 10-week period. Prior injury prevention programs that have demonstrated success in altering neuromechanics during non-fatigued states, used implementation schemes of three sessions per week, over a shorter duration of six weeks (Myer, Ford, Brent, et al., 2006; Myer, et al., 2005; Paterno, et al., 2004). Such a training scheme may result in optimized learning and retention, whereas the currently employed scheme possibly impaired retention between learning sessions. Though it is currently unknown as to the optimal amount of stimulus required to induce neuromechanical alterations through a prevention program, with regards to the outcomes of the current study, it is recommended that future studies increase the amount of stimuli applied in order to establish a minimum threshold required to elicit change.

To assist in the allocation of altered movement patterns while fatigued, participants were being provided with augmented feedback via the CSCS directing the prevention program. Such feedback was directed at correcting improper movement patterns and reinforcing proper movement patterns, with the feedback provided globally and locally. Relative to such cues, the sagittal plane alterations produced post-training are counterintuitive to what was being taught, with positions of greater hip and knee flexion emphasized throughout the plyometric and agility tasks. As reported by the CSCS, the

participants exhibited alterations relative to the cueing during the prevention program, performing agility and plyometric drills in positions of greater flexion. It is possible that these changes were noted due to the knowledge of performance being provided to the participants from the CSCS via augmented feedback, and that was the only corrective factor being employed, resulting in a lack of acquisition, retention, and transfer. It has been previously demonstrated (Weeks & Kordus, 1998; Wulf, Chiviawowsky, Schiller, & Avila, 2010) that an abundance of feedback, relative to performance of body movement, can lead to decreased acquisition, retention, or transfer, which may explain the lack of, or divergent, changes witnessed during the USCT post-training.

It should also be noted that the cues provided by the CSCS were of an internal focus, directing the participants to focus on their own actions. It has been shown that an internal focus of attention may be inferior, or even detrimental, relative to an external focus, in which attention is focused on the effects of the participants' movements (Shea & Wulf, 1999; Wulf, Hob, & Prinz, 1998). Though the participant needs to be aware of their movement patterns in relation to optimal positioning, using an internal focus of attention employs explicit learning, and could prove to be detrimental as the participant is consciously attending to the control of their mechanics (Gray, 2004). As many of the previous injury prevention programs (Gilchrist, et al., 2008; Greska, et al., 2011; Myer, Ford, Brent, et al., 2006) have utilized explicit learning to produce successful outcomes in altering potential neuromechanical ACL risk factors, their use should not be discounted, but permanent long-term outcomes have yet to be retained (McCann, et al., 2011; Padua, et al., 2012). In developing movement pattern changes relative to reducing ACL injury risk factors, Benjaminse and Otten (Benjaminse & Otten, 2010) proposed the use of implicit learning

techniques. Such techniques utilize an external focus of attention, allowing for self-discovery in acquiring a motor pattern and the development of such motor skills have been shown to remain stable under various states of fatigue (Masters, Poolton, & Maxwell, 2008; Poolton, Masters, & Maxwell, 2007). Future studies should consider the use of such alternative techniques and assess their efficacy in bringing about alterations in motor patterns that may possibly reduce the likelihood of a non-contact ACL injury.

Limitations within the current study warrant reservations when interpreting the results. The lack of a control group in the study also limited the ability to quantify the changes solely brought about by the intervention. As well, since the athletes participated in the regular off-season team practices, it is not possible to differentiate whether the changes were brought about by the intervention program or standard practice sessions. The manner in which the participants were assessed in the laboratory may also limit the application of the study. The subjects were required to react to one of two choices during the USCT, limiting the amount of information that needed to be processed. Compared to a real-time game or practice setting, in which the participant must interact with other players and the environment, the amount of information needing to be processed may increase exponentially, possibly resulting in the selection of an alternative movement pattern.

### *Conclusion*

The implementation of an injury prevention program, coupled with regular off-season training sessions, affected sagittal plane hip neuromechanics during an USCT. Such alterations exhibit a possible increase in postural stability about the hips, demonstrating potentially protective mechanisms that may reduce risk factors related to non-contact ACL

injuries. The current intervention did not demonstrate major success in inhibiting unfavorable effects of fatigue, and therefore it is recommended that future studies investigate the use of increased stimuli, as well as alternative techniques in elucidating alterations in motor patterns while in a fatigued state.

## CHAPTER VI

### CONCLUSIONS

The three experiments presented provide enhanced understanding to the concept of leg dominance and its role in non-contact ACL injuries. The first experiment demonstrated that minor asymmetries do exist in neuromechanics between legs when performing an unanticipated sidestep cutting task, but they are not enough to define a difference between the dominant and non-dominant legs. It was found that joint angles and moments were similar throughout the stance phase, but prior to contact differing patterns of peak electromyographic activity were displayed between the dominant and non-dominant legs. Our second experiment determined the influence of fatigue and leg dominance on lower extremity neuromechanics during the unanticipated sidestep cutting task. Fatigue demonstrated similar decrements to both legs, particularly about the sagittal plane for the hip and knee. Fatigue also decreased the electromyographic amplitude of the gluteus medius muscles for both legs, and altered the peak activity in differing orders for the dominant and non-dominant legs. The third experiment implemented an injury prevention intervention to attempt to attenuate the detrimental effects of fatigue on neuromechanics. Post-intervention, the effects of fatigue were still present, altering hip moments and angles in all planes, and knee flexion angle also. The intervention brought about decreases in hip flexion and internal rotation angles for both legs, as well as divergent changes in the hip abduction moment between legs.

These experiments demonstrated that leg dominance is likely not a factor that leads to increased risk of suffering a non-contact ACL injury in a collegiate female soccer population. The experiments also demonstrated that fatigue did not bring about any alterations in

mechanics relative to leg dominance, but it did influence control about the hips. As the population of interest in these experiments can be regarded as experts, it is possible that previous experiences have influenced the lack of differentiation between legs, as well as the strategies adopted during the sidestep cutting task once fatigued. The extended position seen about the hips and knees post-fatigue has implications for loading the ACL, but it also represents a pattern of control to stabilize the trunk. Adopting such a position to allow for proximal control about the hips can lessen the loading of the ACL in other planes. As this erect positioning about the hips was increased after the intervention, it stands to reason that dynamic control of ACL loading may best be accomplished at the hips during a sidestep cutting task.

As the effects of fatigue were still present after the injury prevention intervention, other techniques should be investigated to attempt to off-set the ensuing detrimental effects. If neuromechanical adaptations are to be the focus of reducing non-contact ACL injuries, such alterations must remain consistent no matter the athletes' physical or psychological condition. Future studies defining relationships between cortical processes and movement outcomes need to be explored to understand the full impact of fatigue and how it relates to injury risk.

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APPENDIX I - Table 3-1. The effects of leg dominance on hip and knee joint angles and moments (mean  $\pm$  SD) across movement

	Dominant	Non-Dominant	<i>P</i>	Cohen's <i>d</i>
Initial Contact				
Hip Flexion, deg	53.3 $\pm$ 10.0	52.9 $\pm$ 7.4	.877	0.05
Hip Abduction, deg	-11.0 $\pm$ 6.7	-11.6 $\pm$ 6.5	.798	0.09
Knee Flexion, deg	-28.2 $\pm$ 6.4	-28.6 $\pm$ 7.0	.859	0.06
Knee Abduction, deg	-0.6 $\pm$ 4.4	-1.6 $\pm$ 4.4	.469	0.23
Vertical GRF <sup>a</sup> , N/kg	0.3 $\pm$ 0.2	0.3 $\pm$ 0.1	.760	0.10
Peak Knee Adduction Moment				
Hip Flexion, deg	51.9 $\pm$ 10.2	50.7 $\pm$ 8.4	.677	0.13
Knee Flexion, deg	-35.2 $\pm$ 8.4	-32.7 $\pm$ 7.7	.319	0.32
Knee Abduction, deg	-1.5 $\pm$ 5.9	-2.5 $\pm$ 5.2	.560	0.19
Knee Adduction Moment, Nm/Kgm	0.3 $\pm$ 0.1	0.2 $\pm$ 0.1	.415	0.26
Peak Stance				
Hip Flexion, deg	56.4 $\pm$ 8.8	54.5 $\pm$ 7.6	.479	0.23
Knee Flexion, deg	-57.0 $\pm$ 7.1	-52.5 $\pm$ 5.8	.034	0.69
Knee Abduction, deg	-4.7 $\pm$ 5.8	-5.9 $\pm$ 5.8	.486	0.21
Hip Flexion Moment, Nm/Kgm	-0.1 $\pm$ 0.3	0.1 $\pm$ 0.2	.713	0.11
Vertical GRF <sup>a</sup> , N/kg	2.2 $\pm$ 0.3	2.1 $\pm$ 0.2	.485	0.39

Joint rotations for hip and knee abduction, hip extension, and knee flexion are negative

<sup>a</sup> - GRF, Ground Reaction Force

APPENDIX II - Table 3-2. The effects of leg dominance on muscle EMG activation (expressed as %MVC) from pre-contact (100ms prior to initial contact) and at initial contact (mean  $\pm$  SD)

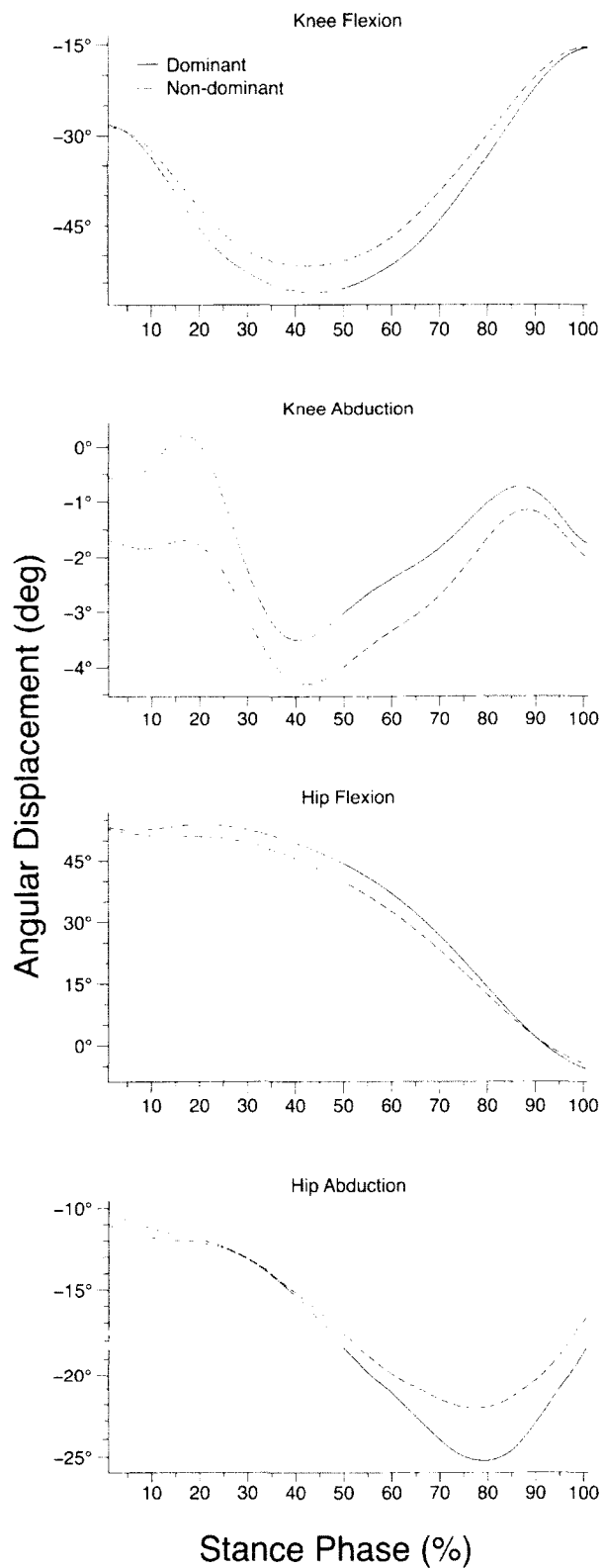
	Dominant	Non-Dominant	<i>P</i>	Cohen's <i>d</i>
Precontact Peak				
Gluteus Medius	50.5 $\pm$ 24.5	57.5 $\pm$ 36.0	.477	0.22
Biceps Femoris	67.3 $\pm$ 52.9	62.0 $\pm$ 49.1	.677	0.13
Semitendinosus	51.4 $\pm$ 37.7	68.4 $\pm$ 43.4	.089	0.55
Rectus Femoris	61.5 $\pm$ 37.9	57.8 $\pm$ 56.9	.735	0.11
Vastus Medialis	60.5 $\pm$ 58.2	70.4 $\pm$ 49.3	.957	0.02
Vastus Lateralis	53.0 $\pm$ 48.9	69.2 $\pm$ 54.7	.679	0.13
Initial Contact				
Gluteus Medius	36.2 $\pm$ 22.0	40.1 $\pm$ 30.7	.649	0.15
Biceps Femoris	41.2 $\pm$ 28.5	47.4 $\pm$ 44.5	.888	0.04
Semitendinosus	35.4 $\pm$ 26.3	54.4 $\pm$ 34.6	.211	0.40
Rectus Femoris	36.8 $\pm$ 29.3	44.4 $\pm$ 55.2	.567	0.18
Vastus Medialis	43.9 $\pm$ 49.2	50.3 $\pm$ 44.6	.702	0.12
Vastus Lateralis	25.5 $\pm$ 22.1	44.0 $\pm$ 53.0	.546	0.19

APPENDIX III - Table 3-3. The effects of leg dominance on time occurrence (expressed as % of phase) of peak muscle EMG activation during pre-contact, and hip and knee kinematics and kinetics during peak stance (mean  $\pm$  SD)

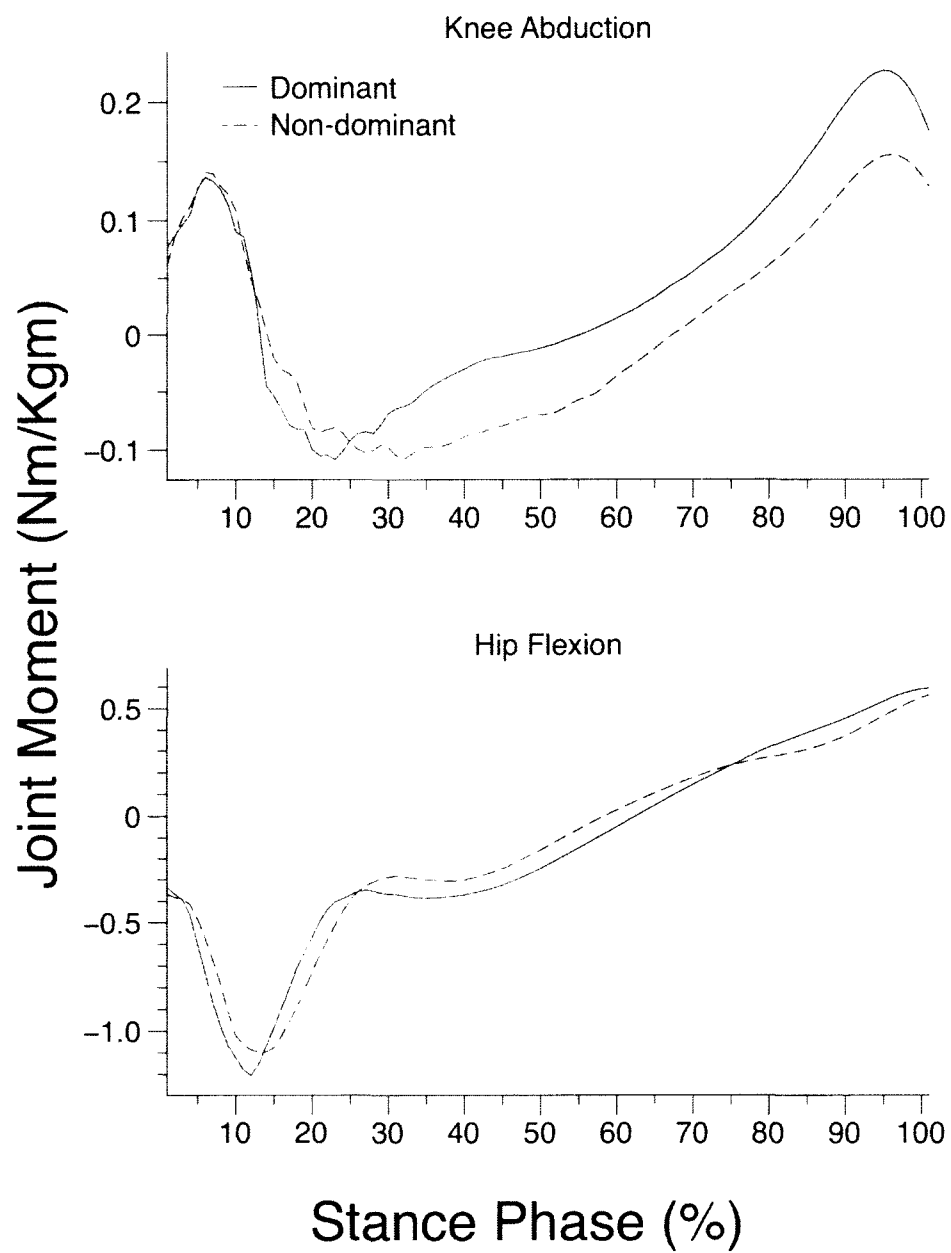
	Dominant	Non-Dominant	<i>P</i>	Cohen's <i>d</i>
Pre-contact EMG Peak				
Gluteus Medius	54.2 $\pm$ 29.3	58.6 $\pm$ 28.4	.634	0.15
Biceps Femoris	52.5 $\pm$ 30.3	52.1 $\pm$ 31.1	.560	0.19
Semitendinosus	55.1 $\pm$ 35.9	56.7 $\pm$ 24.0	.562	0.18
Rectus Femoris	47.8 $\pm$ 36.0	59.3 $\pm$ 32.3	.343	0.30
Vastus Medialis	58.6 $\pm$ 30.4	62.5 $\pm$ 28.2	.407	0.27
Vastus Lateralis	52.4 $\pm$ 30.2	56.6 $\pm$ 29.6	.683	0.13
Peak Stance				
Hip Flexion	13.1 $\pm$ 9.1	9.2 $\pm$ 8.8	.185	0.44
Knee Flexion	43.3 $\pm$ 3.1	41.3 $\pm$ 4.9	.121	0.49
Knee Abduction	36.3 $\pm$ 12.1	35.1 $\pm$ 11.7	.756	0.10
Hip Flexion Moment	23.8 $\pm$ 14.5	26.1 $\pm$ 14.2	.616	0.16
Knee Abduction Moment	20.2 $\pm$ 8.7	25.2 $\pm$ 9.0	.081	0.56
Vertical GRF <sup>a</sup>	24.0 $\pm$ 12.7	25.7 $\pm$ 10.2	.643	0.15

<sup>a</sup> - GRF, Ground Reaction Force

APPENDIX IV - Figure 3-1. Effect of leg dominance on mean hip and knee kinematic parameters during the stance phase of the sidestep cutting task



APPENDIX V - Figure 3-2. Effect of leg dominance on mean normalized hip and knee kinetic parameters during the stance phase of the sidestep cutting task



APPENDIX VI - Table 4-1. Descriptive physiological data from fatigue tasks

<b>Variable</b>	<b>Mean</b>
Graded Exercise Test	
VO <sub>2</sub> max (ml kg <sup>-1</sup> min <sup>-1</sup> )	51.4 ± 4.6
Blood Lactate (mmol L <sup>-1</sup> )	11.1 ± 1.7
Criterion Heart Rate (bpm)	198 ± 8
Criterion RPE	9 ± 1
Dynamic Fatigue Battery	
Criterion Heart Rate (bpm)	193 ± 9
Criterion RPE	9 ± 1
Agreement Ratio	
Criterion Heart Rate (bpm)	0.98 ± 0.03
Criterion RPE	1.07 ± 0.15

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RPE – Rating of Perceived Exertion

Variable	Fatigue	Leg		p-value		
		Dominant	Non-Dominant	Leg	Fatigue	Leg x Fatigue
Gluteus Medius	Pre-Fatigue	50.5±24.5	57.5±36.0	0.207	<b>&lt; 0.001</b>	0.974
	Post-Fatigue	30.4±17.5	37.7±24.8			
Time Occurrence	Pre-Fatigue	54.2±29.3	58.6±28.4	0.270	0.368	0.592
	Post-Fatigue	44.9±28.6	55.6±29.3			
Biceps Femoris	Pre-Fatigue	67.3±52.9	62.0±49.1	0.857	0.105	0.618
	Post-Fatigue	50.4±42.6	51.0±27.3			
Time Occurrence	Pre-Fatigue	52.5±30.3	52.1±31.1	0.556	0.618	0.622
	Post-Fatigue	45.2±32.8	51.1±20.3			
Semitendinosus	Pre-Fatigue	51.4±37.7	68.4±43.4	0.076	0.945	0.288
	Post-Fatigue	33.3±29.3	84.2±130.9			
Time Occurrence	Pre-Fatigue	55.1±35.9	56.7±24.0	0.887	0.515	0.890
	Post-Fatigue	51.7±24.3	51.4±29.7			
Rectus Femoris	Pre-Fatigue	61.5±37.9	57.8±56.9	0.499	0.434	0.618
	Post-Fatigue	56.4±61.2	44.7±38.1			
Time Occurrence	Pre-Fatigue	47.8±36.0	59.3±32.3	0.617	0.992	0.073
	Post-Fatigue	56.6±21.0	50.8±23.5			
Vastus Medialis	Pre-Fatigue	60.5±58.2	70.4±49.3	0.752	0.578	0.273
	Post-Fatigue	66.6±65.4	49.8±29.3			
Time Occurrence	Pre-Fatigue	58.6±30.4	62.5±28.2	0.517	0.295	0.995
	Post-Fatigue	50.9±28.4	54.7±28.3			
Vastus Lateralis	Pre-Fatigue	53.0±48.9	69.2±54.7	0.074	0.355	0.883
	Post-Fatigue	41.7±30.0	61.3±53.7			
Time Occurrence	Pre-Fatigue	52.4±30.2	56.6±29.6	0.851	0.206	0.605
	Post-Fatigue	46.7±25.6	44.8±29.4			

APPENDIX VII - Table 4-2. Peak muscle EMG activation (%MVIC) and time occurrence (% phase) during pre-contact phase relative to fatigue status and leg dominance (mean ± SD)



APPENDIX VIII - Table 4-3. Peak muscle EMG activation (%MVIC) and time occurrence (% phase) at initial contact relative to fatigue status and leg dominance (mean  $\pm$  SD)

Variable	Fatigue		Leg		p-value	
	Dominant	Non-Dominant	Leg	Fatigue	Leg x Fatigue	
Gluteus Medius	Pre-Fatigue	40.1 $\pm$ 30.7	0.592	<b>0.002</b>	0.816	
	Post-Fatigue	24.5 $\pm$ 17.0				
Biceps Femoris	Pre-Fatigue	47.4 $\pm$ 44.5	0.780	0.308	0.525	
	Post-Fatigue	36.5 $\pm$ 22.3				
Semitendinosus	Pre-Fatigue	54.4 $\pm$ 34.6	0.054	0.010	0.926	
	Post-Fatigue	39.9 $\pm$ 42.5				
Rectus Femoris	Pre-Fatigue	44.4 $\pm$ 55.2	0.492	0.773	0.850	
	Post-Fatigue	43.5 $\pm$ 64.2				
Vastus Medialis	Pre-Fatigue	50.3 $\pm$ 44.6	0.823	0.840	0.704	
	Post-Fatigue	44.4 $\pm$ 58.2				
Vastus Lateralis	Pre-Fatigue	46.8 $\pm$ 40.4	0.095	0.391	0.626	
	Post-Fatigue	44.0 $\pm$ 53.0				

APPENDIX IX - Table 4-4. Hip and knee joint angles at initial contact relative to fatigue status and leg dominance (mean  $\pm$  SD)

Variable	Fatigue	Leg		p-value		
		Dominant	Non-Dominant	Leg	Fatigue	Leg x Fatigue
Knee Flexion Angle (°)	Pre-Fatigue	-28.2 $\pm$ 6.4	-28.6 $\pm$ 7.0	0.355	0.012	0.392
	Post-Fatigue	-23.5 $\pm$ 8.4	-26.2 $\pm$ 8.6			
Knee Abduction Angle (°)	Pre-Fatigue	-0.6 $\pm$ 4.4	-1.6 $\pm$ 4.4	0.089	0.907	0.516
	Post-Fatigue	-0.3 $\pm$ 3.7	-1.8 $\pm$ 4.3			
Knee Internal Rotation Angle (°)	Pre-Fatigue	9.5 $\pm$ 6.4	9.0 $\pm$ 5.1	0.763	0.081	0.373
	Post-Fatigue	7.7 $\pm$ 4.8	8.7 $\pm$ 5.3			
Hip Flexion Angle (°)	Pre-Fatigue	53.3 $\pm$ 10.0	52.9 $\pm$ 7.4	0.563	<b>&lt; 0.001</b>	0.141
	Post-Fatigue	46.5 $\pm$ 10.5	48.7 $\pm$ 10.5			
Hip Abduction Angle (°)	Pre-Fatigue	-11.0 $\pm$ 6.7	-11.6 $\pm$ 6.5	0.435	0.017	0.224
	Post-Fatigue	-7.2 $\pm$ 7.6	-9.6 $\pm$ 6.2			
Hip Internal Rotation Angle (°)	Pre-Fatigue	16.3 $\pm$ 5.5	16.9 $\pm$ 7.7	0.918	0.010	0.618
	Post-Fatigue	14.4 $\pm$ 5.4	14.2 $\pm$ 9.2			

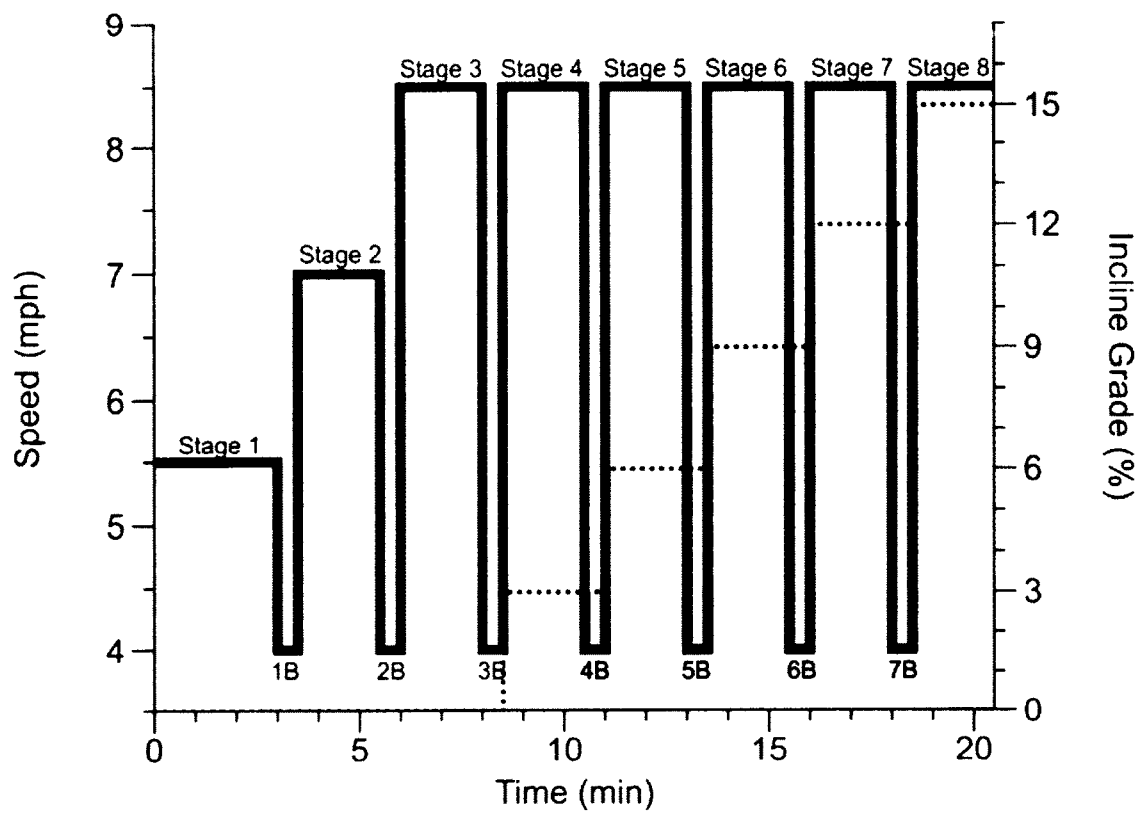
APPENDIX X - Table 4-5. Hip and knee joint angles, knee adduction moment, and vertical ground reaction force (GRF) at peak internal knee adduction moment relative to fatigue status and leg dominance (mean ± SD)

Variable	Fatigue	Leg		p-value		
		Dominant	Non-Dominant	Leg	Fatigue	Leg x Fatigue
Time Occurrence	Pre-Fatigue	13.3±10.2	10.5±7.6	0.395	0.057	0.079
	Post-Fatigue	9.3±8.4	10.0±6.7			
Knee Flexion Angle (°)	Pre-Fatigue	-35.2±8.4	-32.7±7.7	0.741	<b>0.001</b>	0.089
	Post-Fatigue	-28.2±9.1	-29.9±7.5			
Knee Abduction Angle (°)	Pre-Fatigue	-1.5±5.9	-2.5±5.2	0.146	0.243	0.516
	Post-Fatigue	-0.8±4.5	-2.2±4.8			
Knee Internal Rotation Angle (°)	Pre-Fatigue	11.8±6.4	11.2±5.1	0.913	0.047	0.278
	Post-Fatigue	9.8±4.3	10.7±4.2			
Hip Flexion Angle (°)	Pre-Fatigue	51.9±10.2	50.7±8.4	0.911	<b>0.001</b>	0.138
	Post-Fatigue	45.7±10.8	47.3±11.4			
Hip Abduction Angle (°)	Pre-Fatigue	-12.6±6.6	-12.7±6.0	0.470	0.008	0.113
	Post-Fatigue	-8.2±8.0	-10.8±5.7			
Hip Internal Rotation Angle (°)	Pre-Fatigue	15.5±8.3	15.6±9.0	0.982	0.111	0.897
	Post-Fatigue	13.9±8.7	13.9±10.0			
Knee Adduction Moment (Nm/Kgm)	Pre-Fatigue	0.27±0.12	0.24±0.13	0.372	0.006	0.723
	Post-Fatigue	0.21±0.12	0.19±0.12			
Vertical GRF (N/Kg)	Pre-Fatigue	1.46±0.61	1.23±0.51	0.118	0.239	0.321
	Post-Fatigue	1.29±0.65	1.19±0.54			

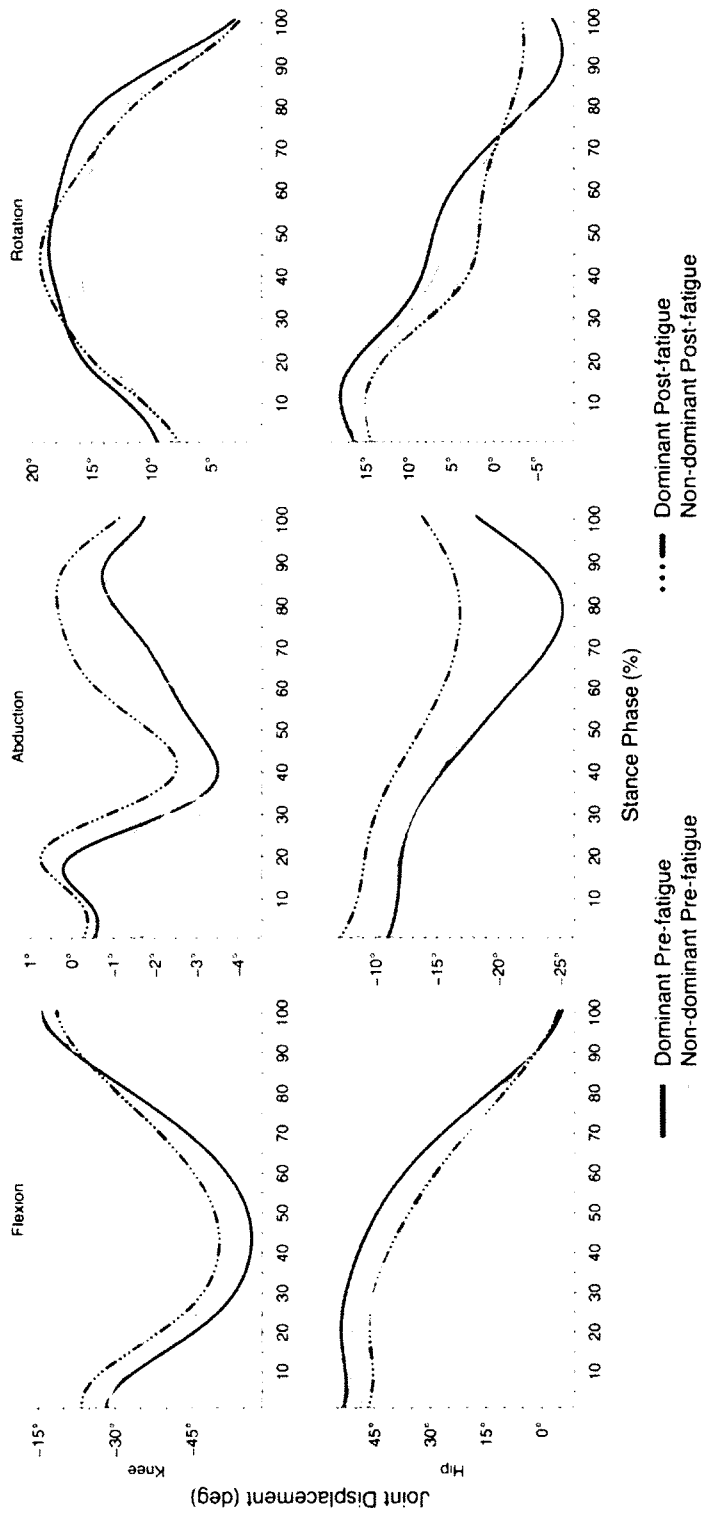
APPENDIX XI - Table 4-6. Peak values and time occurrence (% phase) for hip and knee joint angles, hip flexion moment, and vertical ground reaction force (GRF) during stance relative to fatigue status and leg dominance (mean ± SD)

Variable	Fatigue	Leg		p-value		
		Dominant	Non-Dominant	Leg	Fatigue	Leg x Fatigue
Knee Flexion Angle (°)	Pre-Fatigue	-57.0±7.1	-52.5±5.8	0.014	<b>&lt; 0.001</b>	0.381
	Post-Fatigue	-50.8±7.0	-48.0±5.7			
Time Occurrence	Pre-Fatigue	43.3±3.1	41.3±4.9	0.168	0.326	0.330
	Post-Fatigue	42.0±4.1	41.4±2.8			
Knee Abduction Angle (°)	Pre-Fatigue	-4.7±5.8	-5.9±5.8	0.146	0.266	0.589
	Post-Fatigue	-4.0±5.0	-5.6±4.7			
Time Occurrence	Pre-Fatigue	36.2±12.1	35.1±11.7	0.266	0.812	0.075
	Post-Fatigue	31.9±14.1	38.3±10.1			
Hip Flexion Angle (°)	Pre-Fatigue	56.4±8.8	54.5±7.6	0.899	<b>&lt; 0.001</b>	0.063
	Post-Fatigue	48.2±10.7	49.7±10.7			
Time Occurrence	Pre-Fatigue	13.0±9.1	9.2±8.8	0.043	0.847	0.838
	Post-Fatigue	13.7±8.7	9.3±7.5			
Hip Internal Rotation Angle (°)	Pre-Fatigue	19.5±5.2	18.5±7.1	0.620	0.004	0.750
	Post-Fatigue	16.4±5.6	15.8±9.1			
Time Occurrence	Pre-Fatigue	12.6±10.0	9.1±6.9	0.305	0.234	0.144
	Post-Fatigue	8.8±5.4	8.9±5.1			
Hip Flexion Moment (Nm/Kgm)	Pre-Fatigue	-0.01±0.26	0.01±0.20	0.698	<b>&lt; 0.001</b>	0.084
	Post-Fatigue	0.24±0.35	0.16±0.23			
Time Occurrence	Pre-Fatigue	23.8±14.5	26.1±14.2	0.924	0.007	0.127
	Post-Fatigue	26.1±14.2	29.8±12.6			
Vertical GRF (N/Kg)	Pre-Fatigue	2.16±0.31	2.10±0.23	0.565	<b>&lt; 0.001</b>	0.541
	Post-Fatigue	2.25±0.30	2.25±0.30			
Time Occurrence	Pre-Fatigue	24.0±12.7	25.7±10.2	0.874	0.939	0.560
	Post-Fatigue	25.2±9.8	24.3±9.8			

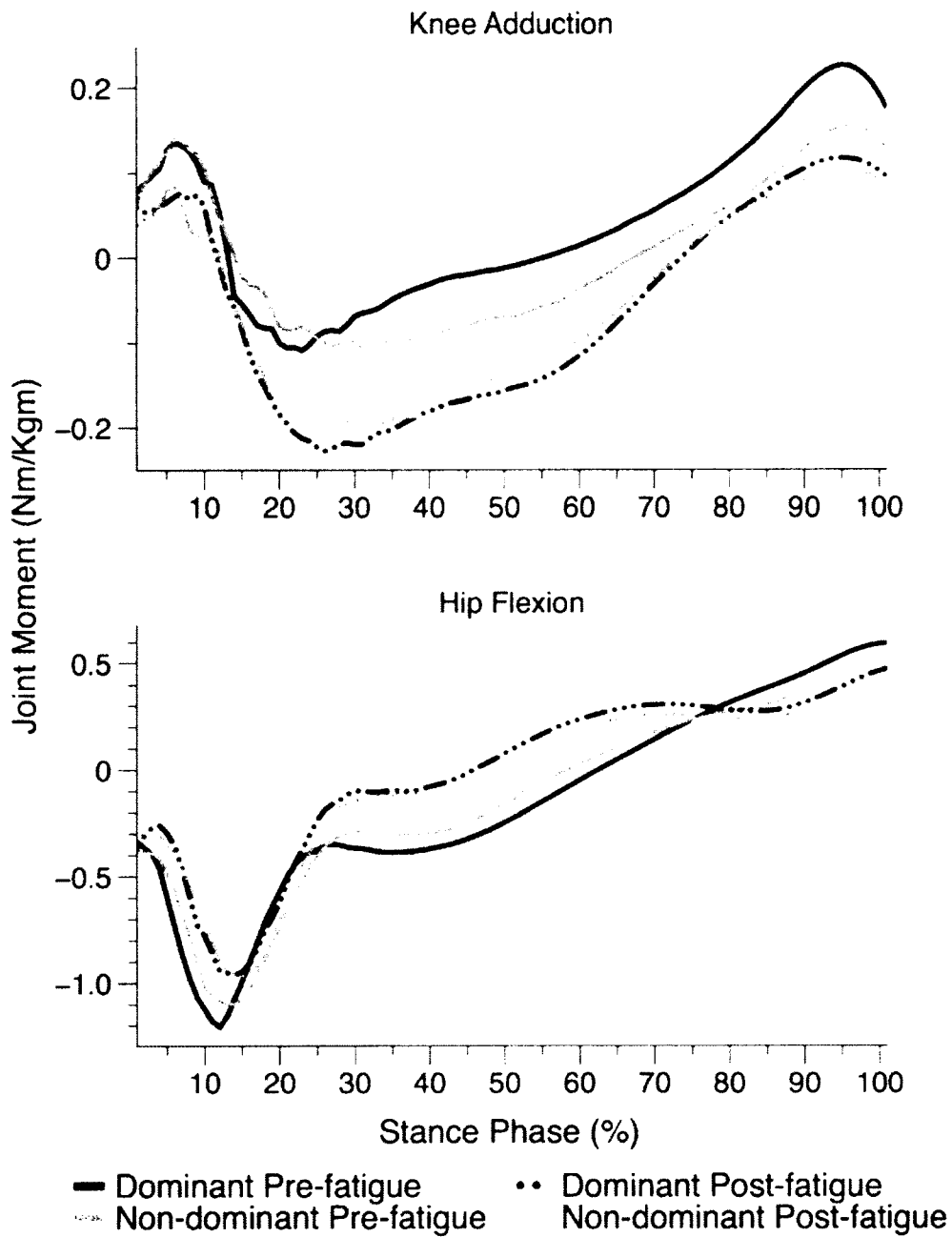
APPENDIX XII - Figure 4-1. Schematic of Graded Exercise Test (GXT) protocol



APPENDIX XIII - Figure 4-2. Mean stance phase knee and hip joint angular displacement relative to leg dominance and fatigue status during unanticipated sidestep cutting.



APPENDIX XIV - Figure 4-3. Mean stance phase internal knee adduction and hip flexion moments relative to leg dominance and fatigue status during unanticipated sidestep cutting.



APPENDIX XV - Table 5-1. List of weekly drills performed while in a state of fatigue

Drills	Sets	Repetitions	Rest
Tuck Jump	4	6	15 sec
Tuck Jump with 180° turn	4	6	15 sec
Single Leg Triple Hop	4	3/leg	15 sec
Single Leg Lateral Hop and Hold	4	6/leg	30 sec
10m Box Drill with Visual Cues			
Shuffle-Sprint-Shuffle-Backpedal	6	3/direction	15 sec
Diagonal Sprint-Backpedal-			
Diagonal Sprint-Backpedal	6	3/direction	15 sec



APPENDIX XVI - Table 5-2. Descriptive statistics for joint angles (degrees) and moments (Nm/Kgm) relative to leg dominance, training, and fatigue at initial contact, peak internal knee adduction moment, and peak stance (mean ± SD)

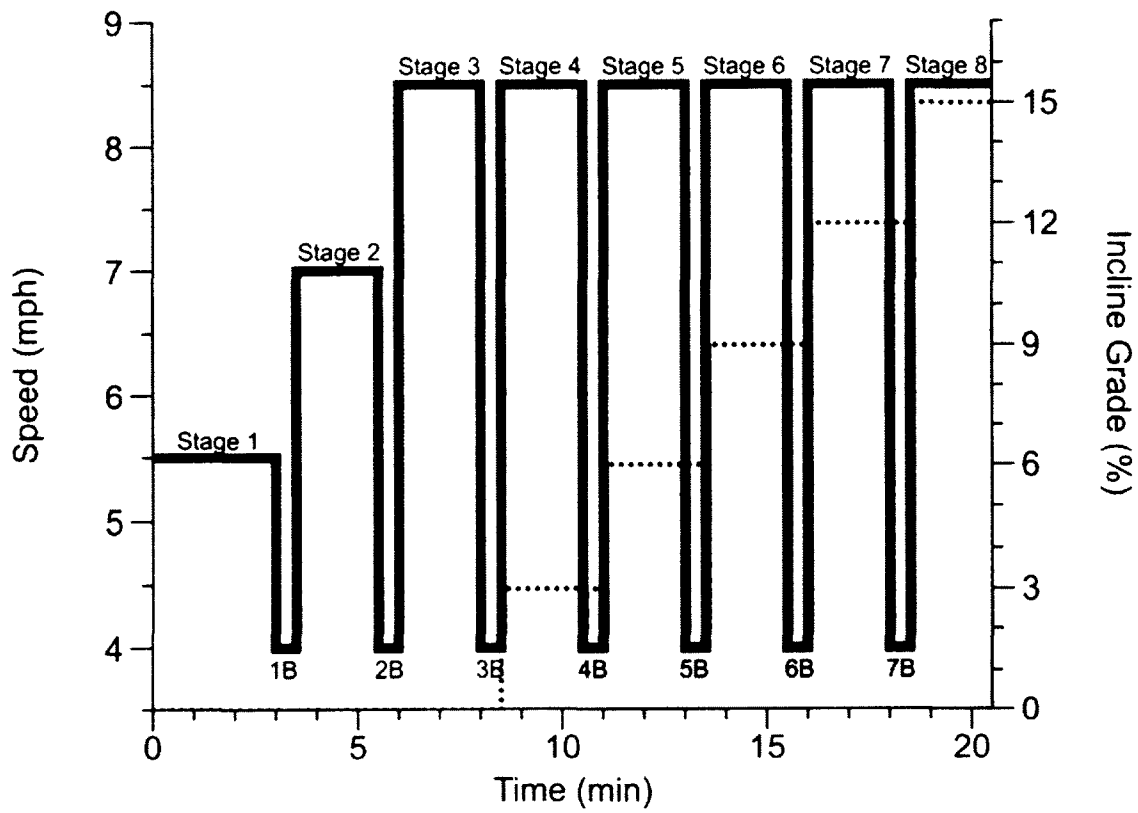
	Dominant				Non-dominant			
	Pre-Intervention		Post-Intervention		Pre-Intervention		Post-Intervention	
	Pre-Fatigue	Post-Fatigue	Pre-Fatigue	Post-Fatigue	Pre-Fatigue	Post-Fatigue	Pre-Fatigue	Post-Fatigue
<b>Initial Contact</b>								
Knee Flexion Angle	-26.0 ± 7.9	-21.9 ± 9.0	-19.6 ± 8.4	-18.5 ± 7.3	-28.0 ± 8.1	-23.4 ± 8.9	-26.6 ± 7.2	-22.5 ± 11.0
Hip Flexion Angle <sup>a,b</sup>	48.8 ± 10.8	43.4 ± 10.7	41.7 ± 6.4	37.1 ± 7.9	48.9 ± 9.5	44.6 ± 11.0	43.3 ± 8.1	39.3 ± 9.1
Hip Abduction Angle <sup>a</sup>	-12.6 ± 5.4	-8.2 ± 7.5	-11.3 ± 8.2	-9.0 ± 7.8	-9.5 ± 7.5	-7.6 ± 7.4	-11.0 ± 7.0	-8.3 ± 6.6
Hip Rotation Angle	16.8 ± 5.1	15.9 ± 4.7	12.2 ± 7.8	9.8 ± 6.9	17.9 ± 6.1	15.7 ± 5.4	15.1 ± 5.9	12.9 ± 8.4
Hip Flexion Moment	-0.2 ± 0.2	-0.2 ± 0.23	-0.29 ± 0.28	-0.29 ± 0.37	-0.34 ± 0.22	-0.29 ± 0.22	-0.26 ± 0.26	-0.29 ± 0.31
Hip Abduction Moment <sup>c</sup>	0.11 ± 0.11	0.06 ± 0.12	-0.01 ± 0.10	-0.01 ± 0.11	0.04 ± 0.12	0.01 ± 0.13	0.09 ± 0.09	0.00 ± 0.10
Trunk Anterior Flexion Angle	5.4 ± 10.6	4.2 ± 9.3	2.8 ± 13.5	0.1 ± 12.9	3.3 ± 11.2	3.9 ± 10.2	-0.9 ± 11.4	-1.6 ± 13.6
Trunk Lateral Flexion Angle	8.3 ± 8.2	8.4 ± 5.6	7.8 ± 5.2	6.9 ± 5.1	6.5 ± 7.9	6.8 ± 6.4	6.6 ± 7.8	5.5 ± 6.3
<b>Peak Internal Knee Adduction Moment</b>								
Knee Flexion Angle <sup>a</sup>	-31.8 ± 8.8	-28.2 ± 10.2	-30.7 ± 13.2	-27.4 ± 12.6	-32.8 ± 7.9	-27.7 ± 7.5	-33.8 ± 9.9	-28.3 ± 10.8
Hip Flexion Angle <sup>a,b</sup>	47.8 ± 10.8	41.8 ± 10.9	41.2 ± 6.8	36.7 ± 8.8	47.1 ± 9.8	43.7 ± 11.8	41.7 ± 8.3	37.7 ± 8.4
Hip Abduction Angle <sup>a</sup>	-13.4 ± 6.0	-9.1 ± 7.7	-12.9 ± 9.1	-9.7 ± 8.6	-11.0 ± 7.0	-8.4 ± 7.1	-12.8 ± 7.8	-10.1 ± 7.0
Hip Rotation Angle	17.0 ± 5.1	15.0 ± 5.4	11.1 ± 9.4	9.2 ± 8.8	17.0 ± 5.6	14.1 ± 7.2	15.8 ± 6.8	11.7 ± 8.9
Hip Flexion Moment <sup>a</sup>	-0.83 ± 0.78	-0.50 ± 0.71	-0.85 ± 0.70	-0.64 ± 0.56	-0.77 ± 0.54	-0.54 ± 0.50	-0.89 ± 0.55	-0.75 ± 0.58
Hip Abduction Moment <sup>a</sup>	0.14 ± 0.18	0.04 ± 0.19	0.08 ± 0.26	-0.01 ± 0.18	0.07 ± 0.22	0.04 ± 0.20	0.10 ± 0.18	0.05 ± 0.16
Trunk Anterior Flexion Angle	4.2 ± 11.5	2.0 ± 9.9	-0.5 ± 14.2	-2.9 ± 15.2	1.8 ± 11.2	2.7 ± 11.3	-3.0 ± 12.8	-2.9 ± 14.1
Trunk Lateral Flexion Angle	9.8 ± 8.4	8.8 ± 5.8	8.9 ± 6.9	8.5 ± 6.1	7.1 ± 8.3	7.2 ± 6.6	7.3 ± 7.6	6.1 ± 6.3
<b>Peak Stance</b>								
Knee Flexion Angle <sup>a</sup>	-55.1 ± 8.9	-50.0 ± 6.7	-51.7 ± 7.0	-48.1 ± 7.0	-52.6 ± 6.6	-47.0 ± 5.7	-51.8 ± 6.2	-48.7 ± 6.3
Hip Flexion Angle <sup>a,b</sup>	52.0 ± 10.1	45.1 ± 10.6	45.3 ± 6.7	39.7 ± 7.9	50.5 ± 9.3	45.9 ± 11.2	45.1 ± 8.0	41.4 ± 8.2
Hip Abduction Angle <sup>a</sup>	-19.0 ± 5.3	-13.6 ± 7.8	-17.7 ± 8.9	-13.5 ± 8.4	-18.5 ± 7.6	-15.0 ± 6.1	-19.0 ± 6.4	-15.6 ± 6.7
Hip Internal Rotation Angle <sup>a,b</sup>	19.9 ± 4.6	17.7 ± 4.0	14.4 ± 7.3	11.7 ± 6.2	19.6 ± 5.0	17.4 ± 6.1	17.1 ± 5.8	14.8 ± 7.5
Hip External Rotation Angle <sup>a</sup>	6.4 ± 6.3	2.9 ± 7.0	2.0 ± 7.3	-0.7 ± 6.9	4.7 ± 6.3	2.6 ± 7.6	4.0 ± 6.7	-0.4 ± 8.0
Hip Flexion Moment <sup>a</sup>	0.06 ± 0.24	0.23 ± 0.32	0.07 ± 0.21	0.25 ± 0.27	0.00 ± 0.23	0.20 ± 0.19	0.09 ± 0.20	0.20 ± 0.19
Hip Abduction Moment <sup>a</sup>	-0.22 ± 0.17	-0.32 ± 0.20	-0.35 ± 0.25	-0.44 ± 0.26	-0.32 ± 0.24	-0.45 ± 0.25	-0.39 ± 0.22	-0.52 ± 0.28
Trunk Anterior Flexion Angle	-2.4 ± 11.2	-4.1 ± 11.4	-7.2 ± 14.5	-8.6 ± 15.3	-3.9 ± 10.2	-3.5 ± 11.2	-8.1 ± 12.2	-9.2 ± 14.4
Trunk Lateral Flexion Angle	13.7 ± 8.8	12.6 ± 5.9	13.8 ± 6.4	13.0 ± 5.3	9.7 ± 8.5	10.5 ± 7.3	11.3 ± 8.3	10.2 ± 7.3

<sup>a</sup> - Denotes statistically significant effect of fatigue

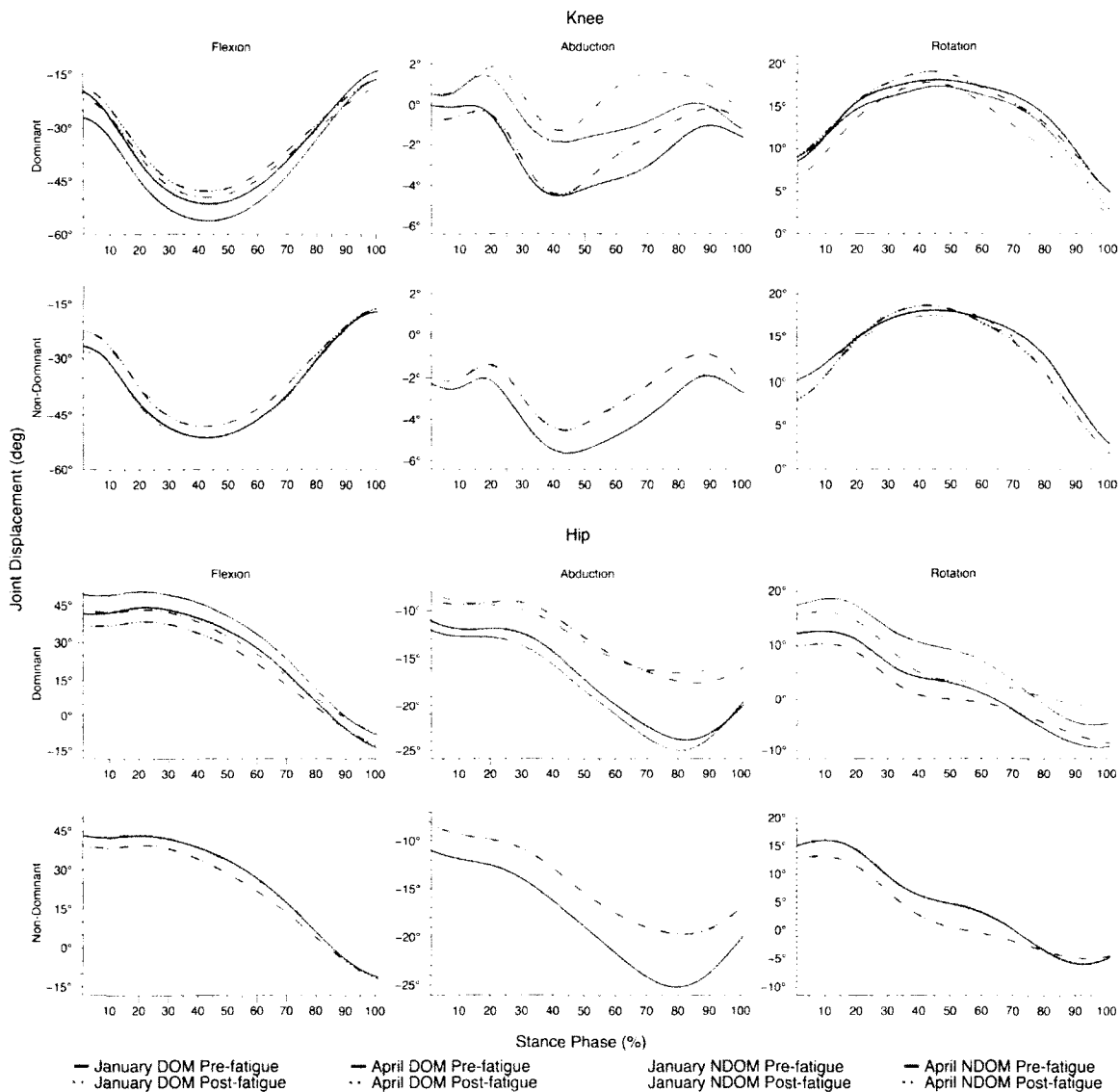
<sup>b</sup> - Denotes statistically significant effect of training

<sup>c</sup> - Denotes statistically significant interaction effect between training and dominance

APPENDIX XVII - Figure 5-1. Schematic of Graded Exercise Test (GXT) protocol

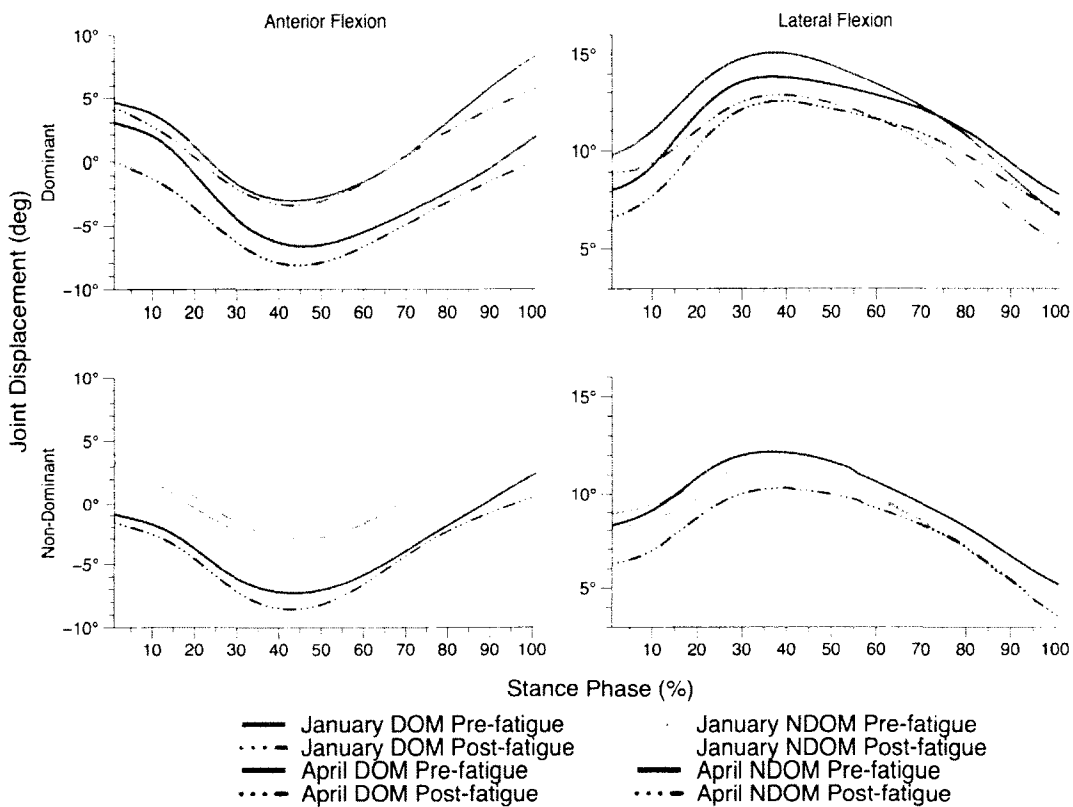


APPENDIX XVIII - Figure 5-2. Effect of training, fatigue, and leg dominance on mean hip and knee kinematic parameters during the stance phase of the sidestep cutting task





APPENDIX XX - Figure 5-4. Effect of training, fatigue, and leg dominance on mean anterior and lateral trunk flexion angles during the stance phase of the sidestep cutting task



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