AMBEGAONKAR, JATIN P., Ph.D. A Comparison of Knee Muscle Activation and Knee Joint Stiffness between Female Dancers and Basketball Players during Drop Jumps. (2007)
Co-directed by Dr. Sandra J. Shultz and Dr. David H. Perrin. 166 pp.

This dissertation compared knee muscle activation of the lateral gastrocnemius, medial and lateral hamstrings, and lateral quadriceps (LG, MH, LH, and LQ) and knee joint stiffness(KJS) between female dancers(D) and basketball players(B) during the initial landing of a double-leg drop jump. The purpose was to examine possible neuromechanical strategies dancers employ that might protect them from Anterior Cruciate Ligament (ACL) injuries during a potentially high ACL-injury risk activity. Fifty-five females (D=35, 20.7±2.3yrs, 164.3±6.7cm,62.2±1.9kg, B=20,20.1±2.0yrs, 170.5±6.1cm,72.6±11.4kg) performed 5 double-leg drop jumps from a 45cm box. Muscle activity was recorded via surface electromyography (sEMG). A force plate and three-dimensional electromagnetic tracking system were used to record kinetic and kinematic data and calculate KJS (ratio of change in sagittal knee moment to sagittal knee flexion angle from ground contact to maximum knee flexion). sEMG data were normalized to maximum volitional isometric contractions(%MVIC), and joint moments to body weight (Nm/kg). Separate 2x4 ANOVAs compared D and B on muscle onsets (ms) and mean RMS amplitudes (%MVIC) before ($_{PRE}$=150 ms) and after ($_{POST}$=50 ms) ground contact. A one-way ANOVA examined group differences in KJS (Nm/kg°), with a stepwise regression model examining prediction of KJS. No significant group differences were observed in muscle onsets (D=133.4±53.2ms, B=121.6±50.2ms;P=.22), activation amplitudes ($_{PRE}$: D=28.1±8.7%MVIC, B=27.7±10.5%MVIC;P=.60; $_{POST}$:

D=51±17.3%MVIC, B=49.6±21.4%MVIC;P=.78), or KJS (D=.0163±.009Nm/kg°, B=.0185±.011Nm/kg°;P=.44). Due to recruitment challenges the proposed full complement of participants (N=70;D=35,B=35) was not achieved. Moderate effect sizes (ES) between-groups indicated a trend towards higher muscle activation levels in dancers in MH both pre (34vs.26%MVIC;ES=.55) and post (38vs.25%MVIC;ES=.41) contact, and in LG post contact (45vs.35%MVIC;ES=.33). The exception was $LQ_{POST}$ (90vs.109%MVIC;ES=.30) where dancers had a tendency for lower muscle activation levels. Prelanding muscle activation amplitudes and group membership were not able to predict changes in KJS. These results suggest that the lack of findings may in-part be due to low statistical power. Further, although KJS did not differ between groups, between-group effect sizes noted in $LG_{POST}$, $MH_{PRE, POST}$, and $LQ_{POST}$ suggest possible differences in neuromechanical strategies over other lower extremity joints. Additional research is necessary to determine possible ACL-injury protective mechanisms employed by dancers during other high ACL-injury risk activities.

# A COMPARISON OF KNEE MUSCLE ACTIVATION AND KNEE JOINT STIFFNESS BETWEEN FEMALE DANCERS AND BASKETBALL PLAYERS DURING DROP JUMPS

by

Jatin P. Ambegaonkar

A Dissertation Submitted to
the Faculty at The Graduate School at
The University of North Carolina at Greensboro
In Partial Fulfillment
of the Requirements for the Degree
Doctor of Philosophy

Greensboro
2007

Approved by

_____
Committee Co-chair

_____
Committee Co-chair

In loving memory of Aai …. Dr. Jyoti P. Ambegaonkar

In honor of Baba …. Mr. Prafull G. Ambegaonkar

APPROVAL PAGE

This dissertation has been approved by the following committee of the

Faculty of The Graduate School at the University of North Carolina at Greensboro.

Committee Co-chair _____

Committee Co-chair _____

Committee Members _____

_____

_____

_____
Date of Acceptance by Committee

_____
Date of Final Oral Examination

ACKNOWLEDGEMENTS

I would like to thank my committee members, Drs David Perrin, Sandra Shultz, Randy Schmitz, Terry Ackerman, and Mark Schulz for your time and patience during this process. I want to offer special acknowledgments to the following exceptional people:

My wife Shruti

THANK YOU for your unending love and support through each and every morning, afternoon, evening, and night. Thank you for reading my endless drafts and listening to me ramble on late through the nights on occasion. What a journey we are having, from the North to the South to the Middle of the Land, after coming halfway across the world!! Thank you for all these years.... and many more to come

My Elder Brother Dr. Gautam P. Ambegaonkar and his family: Thank you for your unvarying support and encouragement throughout my life

My Twin Brother Capt. Kunal P. Ambegaonkar and Pooja: I feel blessed to have a twin brother to have shared my childhood and parallel life paths; now with the beginnings of a family

Dr. David H. Perrin: Dr Perrin, I would like to express my gratitude to you for your constant positive reinforcement all through my doctoral career. Your personal and

professional qualities are what I imagine an ideal mentor, teacher, and educator should possess, and I will strive to live up to them

Dr. Sandra J. Shultz: Sandy, I will be forever indebted to you for teaching me so much and being such a good tutor. Thank you for looking at those endless drafts and having the patience to go over them again and again, I am privileged to call you a friend as well as a teacher

Dr. Randy J. Schmitz: Randy, thank you for constantly reminding me the value of curiosity during research and perseverance during doctoral school and life in general

Dr. Jolene Henning; Jolene, thank you for lending an ear whenever I needed one. At all the critical times, you helped bolster my self-confidence with all those pep talks and advice

My colleagues at the Applied Neuromechanics Research Laboratory: Thank you for living through these years with me. It has been a long and eventful journey and I hope it continues into our future professional careers and lives. I would especially like to acknowledge Dr. Yohei Shimokochi: Yohei, thank you for your insightful comments and thoughts. You make me think differently and due to that I am a better researcher

TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

Page

# CHAPTER I

## INTRODUCTION

Anterior cruciate ligament (ACL) injuries often occur in individuals participating in landing and jumping sports (Arendt, Agel, & Dick, 1999; Arendt & Dick, 1995). A vast majority of these ACL injuries are non-contact in nature, where there is no physical contact of the injured individual with another individual (Boden, Dean, Feagin, & Garrett, 2000; Griffin et al., 2000). Landing and plant-and-cut activities are among the most common activities associated with non-contact ACL injuries (Boden et al., 2000; Ferretti, Papandrea, Conteduca, & Mariani, 1992; Hutchinson & Ireland, 1995; Olsen, Myklebust, Engebretsen, & Bahr, 2004). Females in selected sports have been reported to have a 3 to 8 times greater risk for non-contact ACL injury as compared to males during these activities (Arendt et al., 1999; Arendt & Dick, 1995; Gray et al., 1985; Gwinn, Wilckens, McDevitt, Ross, & Kao, 2000; Malone, Hardaker, Garrett, Feagin, & Bassett, 1993; Moeller & Lamb, 1997). This gender bias has been reported in various physical activity settings including collegiate athletics (Arendt et al., 1999; Arendt & Dick, 1995; Malone et al., 1993) and in the military (Gwinn et al., 2000). Gender differences in neuromuscular and biomechanical parameters during dynamic activity are thought to be a potential explanation for this injury bias (Decker, Torry, Wyland, Sterett, & Steadman, 2003; Griffin et al., 2000; Myer, Ford, & Hewett, 2005; Noyes, Barber

Westin, Fleckenstein, Walsh, & West, 2005; Shultz et al., 2001). Critical gender differences during physical activity include observations that females have higher levels of quadriceps and lower levels of hamstrings muscle activity (Malinzak, Colby, Kirkendall, Yu, & Garrett, 2001; Sander et al., 2004; Zazulak et al., 2005), and increased knee extensor moments (Chappell, Yu, Kirkendall, & Garrett, 2002) during functional activities, including landing.

Of interest, female dancers who also regularly perform landing and jumping activities do not appear to injure their ACL as frequently as other female athletes in sports that involve jumping and landing. The reasons for this discrepancy in ACL injury rates between dancers and other at-risk athletes are unknown. However, differences in neuromuscular parameters have been found between dancers and other athletes. Specifically, female dancers have been reported to have smaller H-reflexes than athletes participating in landing and jumping type activities (Mynark & Koceja, 1997; Nielson, Crone, & Hultborn, 1993), which are indicative of increased levels of cocontraction (Koceja, Trimble, & Earles, 1993; Llewellyn, Yang, & Prochazka, 1990).

Cocontraction of the knee musculature has been suggested to be protective of the knee during dynamic activity (Baratta et al., 1988; Draganich, Jaeger, & Kralj, 1989; Shields et al., 2005; Solomonow et al., 1987). Increased muscle activation increases joint stiffness and has been theorized to improve dynamic joint stability by reducing the chance of buckling following joint loading (Arampatzis, Bruggemann, & Klapsing, 2001; Goldfuss, Morehouse, & LeVeau, 1973). Further, increased stiffness via greater muscle contraction has been suggested to be a possible protective mechanism against ACL injury

through reductions in the proportion of external force to be resisted by the ACL and other knee structures during activity (Wojtys, Ashton-Miller, & Huston, 2002). Whether dancers, who undergo years of disciplined jump training (Harley et al., 2002) use muscle activation strategies that increase knee joint stiffness and stabilization during landing is unknown.

## Statement of the Problem

A large number of ACL injuries are known to occur during landing and/or jumping activities (Boden et al., 2000; Ferretti et al., 1992; Olsen et al., 2004). Although both basketball players and dancers regularly jump and land during practice and performance, the incidence of ACL injuries in dancers is much lower than that in basketball players. It is possible that female dancers use neuromuscular and biomechanical strategies that are more protective of the knee during landing, an activity commonly performed at the time when ACL injury occurs. The purpose of this study therefore, was to compare knee muscle activation patterns and knee joint stiffness in female dancers and basketball players in the initial landing phase of a double-leg drop jump.

## Objectives and Hypotheses

1. To compare knee muscle activation strategies between female dancers and basketball players during the initial landing phase of a drop jump.

Hypothesis 1: Compared to basketball players, female dancers would land with earlier onsets of muscle activity in the lateral quadriceps (LQ), medial (MH) and lateral (LH) hamstrings, and the lateral gastrocnemius (LG) muscles.

Hypothesis 2a: Compared to basketball players, female dancers would land with higher muscle activation amplitudes in the LQ, MH, LH, and LG muscles, pre ground contact.

Hypothesis 2b: Compared to basketball players, female dancers would land with higher muscle activation amplitudes in the LQ, MH, LH, and LG muscles, post ground contact.

2. To compare knee joint stiffness between female dancers and basketball players during the initial landing phase of a drop jump.

   Hypothesis 3: Compared to basketball players, female dancers would land with greater knee joint stiffness.

3. To determine if variations in muscle activation amplitudes and group membership explain differences in knee joint stiffness

   Hypothesis 4: Greater gastrocnemius, hamstrings, quadriceps muscle activation amplitudes and membership in the dance group would predict greater knee joint stiffness.

## Independent Variables

1. Group at two levels: Female basketball players and Female Dancers
2. Muscles at four levels: Lateral Quadriceps (LQ), Medial Hamstrings (MH), Lateral Hamstring (LH), and Lateral Gastrocnemius (LG)
3. Impact phase at two levels: $_{PRE}$ (150 ms prior to ground contact) and $_{POST}$ (50 ms after ground contact).

## Dependent Variables

1. Muscle onset time in milliseconds (ms)
2. Mean muscle amplitudes, expressed as a percentage of a maximal contraction (% MVIC)
3. Knee Joint Stiffness (Nm/kg°)

## Operational Definitions

1. Dancer – A female participant between the ages 18-30 whose primary form of physical activity had been dancing for at least the past two years and who had been dancing at least 3 times a week for at least 30 minutes for the last 3 months.
2. Basketball Player – A female participant between the ages 18-30 whose primary form of physical activity had been basketball for at least the past two years and who had been playing basketball at least 3 times a week at least 30 minutes for the last 3 months.

3. Preferred Leg – The lower extremity that the participant used to land on 2 out of 3 times when performing 3 single-leg landings from a 45 cm box.
4. Drop Jump – A task in which participants stood on a 45 cm box, extended their non-preferred leg and dropped off the box performing a double-leg landing followed by a maximum vertical jump immediately on ground contact.
5. Ground Contact – The time point at which the body contacted the forceplate with a force equal to or greater than 10 N.
6. Muscle Onset (ms) – The time prior to ground contact when muscle activity first exceeded 5 standard deviations above its baseline activity recorded in quiet standing for at least 25 ms or longer.
7. Maximum Voluntary Isometric Contraction (MVIC) – The peak of the peak mean muscle activation amplitude collected from three trials for each muscle group.
8. Muscle Amplitude (%MVIC) – The mean root mean square activation amplitude of a muscle over a specified time period, normalized to each individual's MVIC for that muscle.
9. Knee Joint Stiffness (Nm/kg°) – The ratio of the change in the net knee moment (Nm/kg) to the change in sagittal knee flexion angle (in °) from initial ground contact to peak sagittal knee flexion angle (Farley, Houdijk, Van Strien, & Louie, 1998; Farley & Morgenroth, 1999).
10. Onset of the Menstrual Period – The first day of menstrual bleeding per self-report.

## Limitations and Assumptions

The following limitations and assumptions were accepted for this study:

1. All participants provided an honest and accurate health and injury history and maximum effort during testing.
2. Surface Electromyography (sEMG) was a reliable and valid method of measuring muscle activity during dynamic activity and muscle force was not directly measured.
3. sEMG obtained over the electrode placements for each muscle was adequately representative of the muscle as a whole.
4. A double-leg drop jump represented only one activity that may be potentially harmful to the knee.
5. Measurements collected on the preferred leg are representative of both legs.
6. The findings from this study would be limited to only double-leg drop jumps.

## Delimitations

1. Fifty-five female participants (35 dancers, 20 basketball players) ranging in age from 18 - 30 were recruited from the local university community.
2. Only participants who were healthy with no musculoskeletal injury to either lower extremity for the past 6 months, did not have surgery on either lower extremity, had no history of cardiovascular or neurological problems, and no physical activity restrictions participated.
3. All measurements were obtained from the preferred leg.

4. Measures of muscle activity and knee joint stiffness were restricted to the initial landing phase of a double leg drop jump.
5. The ensemble average of five consecutive trials represented a participant's landing strategy.
6. Muscle activity was measured via sEMG over the lateral quadriceps (vastus lateralis), medial hamstrings (semimembranosis, semitendinosis), lateral hamstring (biceps femoris) and lateral gastrocnemius muscles.
7. The lateral quadriceps was representative of the whole quadriceps muscle, and the lateral gastrocnemius was representative of the whole gastrocnemius muscle.
8. Results from this dissertation are limited to female dancers and basketball players and cannot be generalized across genders, or to injured populations.

# CHAPTER II

# REVIEW OF THE LITERATURE

## Introduction

The purpose of this review was to support the theoretical framework that increased muscle activation increases knee joint stiffness, thereby indicating a possible protective mechanism against ACL injury during dynamic activity. This model was tested in part by examining selected lower extremity neuromuscular ( knee muscle activation) and biomechanical (knee joint stiffness) patterns during a drop jump between two groups; one that has been reported to have high rates of ACL injury (female basketball players) and the other with reported low rates of ACL injury (female dancers). This review of literature addresses what is known about: 1) ACL injury, 2) control of joint stability, 3) landing as a model to examine joint control strategies, and 4) measurement of neuromuscular and biomechanical factors involved in joint stability.

## ACL Injury

The following sections provide an overview of ACL injuries including the incidence rates, the financial implications, the mechanisms of ACL injury, the gender bias in ACL injury rates, and the effectiveness of training programs designed to reduce incidence rates of this injury.

### *Incidence and Long-term Implications*

About 100,000 ACL injuries of the knee occur every year in the United States (Griffin et al., 2000; Huston, Greenfield, & Wojtys, 2000), with a 1 in 3000 incidence rate in the general population between 18-45 years of age (Miyasaka, Daniel, Stone, & Hirshman, 1991). The majority of ACL injuries are sustained in sports that involve landing, jumping, pivoting, and cutting (Boden et al., 2000; Ferretti et al., 1992; Gray et al., 1985; Olsen et al., 2004). Although time lost from sports participation is unfortunate for the ACL injured athlete, the consequences of ACL injuries are not limited to the individual's athletic career. Longer lasting and more serious implications of ACL injury exist due to the subsequent potential for post-traumatic arthritis and other joint problems over the lifetime of the injured individual (Jomha, Borton, Clingleffer, & Pinczewski, 1999; Sommerlath, Lysholm, & Gillquist, 1991).

### *Costs Associated with ACL Injury*

The approximate cost of each ACL surgery has been estimated to be around 17,000 dollars, with the annual surgical costs alone in the United States being reported to be around 850 million dollars (Griffin et al., 2000). Additional costs exist with the need for post surgical care and rehabilitation for the ACL injured individual, as well as those associated with treating the secondary complications that often result from the primary ACL injury (Noyes, Mooar, Matthews, & Butler, 1983).

*Mechanisms of ACL Injury*

The majority of ACL injuries are non-contact in nature (NCACL) with researchers over the past decade having identified some common activities associated with these NCACL injuries as being planting/cutting (sidestep cutting) (29%), landing from a jump, either with a straight knee (28%) or with a hyperextended knee (26% ), and deceleration of the body with the knee in an extended position (Boden et al., 2000; Gray et al., 1985; Hutchinson & Ireland, 1995; Olsen et al., 2004).

In a study examining knee ligament injuries in volleyball players, the most frequent mechanism of knee injury was reported to be landing from a jump (Ferretti et al., 1992). In another study where the researchers analyzed videos of 20 ACL injuries and interviews of 32 ACL-injured players over a period of 12 seasons in team handball (Olsen et al., 2004), the two main mechanisms of ACL injury were identified as either being : (a) a landing from a jump shot associated with valgus and external rotation with the knee close to full extension, or (b) a plant-and-cut movement associated with a valgus and external or internal rotation with the knee again close to full extension. Ireland (2002) specifically identified the 'position of no return' as a typical position related to ACL injury. During basketball, this position was noted to occur at some stage during an awkward landing, when the player was trying to shoot, rebound, or attempting to stay in bounds (Ireland, 2002). The profile includes the individual being relatively upright (i.e. straight back) with low hip and knee flexion angles, excessive valgus of the knee, and tibial external rotation (Ireland, 2002). Collectively, these studies suggest that landing is

one of the primary NCACL injury mechanisms, with the other mechanism being cutting and planting activities.

### *ACL Injury Rates*

ACL injury rates are 3 to 8 times higher in female athletes as compared to male athletes (Arendt et al., 1999; Arendt & Dick, 1995; Gwinn et al., 2000; Malone et al., 1993). This trend has been noted across various practice settings and in different populations. Arendt and Dick (1995) identified the gender disparity in the rates of ACL injuries at the collegiate level. The researchers analyzed incidence data obtained through the National Collegiate Athletic Association (NCAA) Injury Surveillance System from 1988-93 and found that ACL injury rates in soccer and basketball athletes, when expressed as injuries per 1000 athlete-exposures were significantly higher in female (0.31 and 0.29 respectively) as compared to male athletes (0.13 and 0.07 respectively). A follow-up study that analyzed injury data over an additional five year period indicated that these ACL injury rates have remained consistently higher in soccer (females: 0.33 vs. males: 0.12) and basketball (females: 0.29 vs. males 0.10) (Arendt et al., 1999).

Similarly, Malone et al. (1993) compared the prevalence and frequency of ACL injuries in intercollegiate basketball players at the Division I level from 29 institutions. ACL injury prevalence data were based on using the total number of reported ACL injuries and the number of participants from the previous 5 years. For ACL injury frequency, data were obtained from the NCAA Injury Surveillance report for the 1988-89 and 1989-90 seasons. While only 2.2% (9 out of 402) male athletes sustained ACL

injuries, 16.1% (62 out of 385) of the female athletes sustained ACL injuries. Female basketball players were therefore found to be eight times more likely to suffer an ACL injury (Malone et al., 1993). In a sports medicine clinical practice survey of basketball-related injuries occurring over a 30-month period, 25% of all injuries in female basketball players were noted to be ACL injuries as compared to only 2.6% ACL injuries in male basketball players over the same time period (Gray et al., 1985). The gender disparity in ACL injury rates has also been found at the military level across different activities. Gwinn et al. (2000) found that compared to males, female midshipmen had a relative injury risk of 1.40 in coed soccer, basketball, softball, and volleyball, 3.96 in intercollegiate soccer, basketball, and rugby, and 9.74 during actual military training. Across all activities, females had a 2.44 relative risk of ACL injury compared to males (Gwinn et al., 2000).

Collectively these data suggest that across a variety of physically active populations, females have a higher incidence of ACL injury as compared to similarly trained males. However, when comparing female athletes in landing sports and female dancers, dancers appear to have lower rates of knee and ACL injuries. While knee injuries have been reported to account for 15-21 % of all injuries in basketball and soccer (Arendt & Dick, 1995; Kujula et al., 1995) they have been reported to account for only 6.8-15% of all injuries seen in dancers (Evans, Evans, & Carvajal, 1996; Garrick & Requa, 1993). Furthermore, the majority of these knee injuries in dancers have been reported to be overuse and patellar problems (Liederbach, 2000; Rovere, Webb, Gristina, & Vogel, 1983). Other researchers (Scioscia, Griffin, & Fu, 2001) have also commented

on the lower incidence of ACL injuries in dancers as compared to other sports such as basketball and soccer, although no detailed statistical validation and support could be found in the current literature.

### *Training Programs and ACL Injury*

To try and reduce the incidence of ACL injury, training programs have been developed in an effort to intervene on the functional mechanics and strength of the lower extremity musculature in female athletes involved in high ACL risk sports (e.g. basketball, soccer, handball) (Hewett, Stroupe, Nance, & Noyes, 1996; Myklebust et al., 2003). One of these programs was designed to decrease landing forces by teaching neuromuscular control of the lower limb during landing, and increasing joint stability by maximizing the strength of the knee musculature (Hewett et al., 1996). After 6 weeks of participation in the training program, participants decreased peak landing forces by 22%, increased maximal knee flexion angles at landing from 69° ± 14° to 72° ± 9°, and decreased knee adduction and abduction moments at landing approximately by 50% (Hewett et al., 1996). The program also led to increased hamstrings power (33%) and strength (20%), increased hamstrings to quadriceps peak torque ratios (20%), and improved side-to-side hamstring strength imbalances in the participants when compared to pre-training measures.

In a follow-up study, decreased incidence of serious knee injury rates were noted after this training program was administered to high school female athletes (basketball, volleyball, and soccer) (Hewett, Lindenfeld, Riccobene, & Noyes, 1999). After one

season, knee injury incidence rates were 0.12 per 1000 athlete-exposures in female athletes who had undergone the training program versus 0.43 per 1000 athlete-exposures in untrained female athletes (Hewett et al., 1999). The reduction in the injury rates in this study was suggested to be due to improvements in technique and strength, primarily through decreases in the magnitudes of knee abduction/adduction moments and improved hamstrings/quadriceps strength ratios as a result of improved neuromuscular strategies (Hewett, 2000; Hewett et al., 1996).

Another prospective study investigated the efficacy of a 15 minute, five-phase progression program administered to Division I-III female team handball players over three seasons (Myklebust et al., 2003). The program included balancing, cutting, jumping and landing exercises performed on the floor, on a wobble board, and on a mat for approximately five minutes on each surface. Players were trained three times a week for the first 5-7 weeks before the season and then once a week over the course of the season. The researchers noted a decrease in the number of ACL injuries over three seasons from 29 in the control season to 23 and 17 in the first and second intervention seasons respectively. In the elite division, the ACL injury rate was 13 in the control season and 6, and 5 respectively in the first and second intervention seasons. Although the reductions in ACL injury rates were not statistically significant ($p=0.15$ for all levels, $p=0.06$ for elite levels), the authors suggested a trend towards injury reduction after administering the training program (Myklebust et al., 2003).

Collectively these results are encouraging in that participation in these programs seems to result in decreases in ACL injury rates. However, it is still unknown whether a

specific portion of these programs or the program as a whole is effective. Thus these programs might be taking an excessive 'shotgun' approach without specifically targeting one possible problem area, resulting in unnecessary lost time and resources in the administration of these programs. Continued research is therefore needed to delineate the exact causes that might place an individual at risk for ACL injury. This knowledge would eventually result in the development of specific and targeted training programs that would be highly efficient in terms of resources for both health care professionals and clients.

### *Summary*

Occurrence of ACL injuries can have costly and long-terms health implications, especially in the physically active population. A clear gender bias exists in the incidence of ACL injuries, with females being injured at much higher rates than males. One of the most common activities associated with ACL injuries appears to be landing from a jump. Although female dancers also regularly perform landing activities, they seem to have lower rates of ACL injury compared to female athletes in landing and jumping sports like basketball. While some encouraging results from comprehensive training programs have been noted in reducing ACL injury rates, it is still unknown whether one or multiple components of these programs are truly effective. Research needs to continue to uncover possible ACL injury risk factors; answers for which may lie in the further appreciation of factors that control joint stability during landing.

## Control of Joint Stability

Knee joint stability is provided through the presence of several factors including bone/cartilage forces, ligament and capsular forces, intrinsic muscle stiffness, and reflexive or voluntary muscle mediated stiffness (Dhaher, Tsoumanis, Houle, & Rymer, 2004; Dhaher, Tsoumanis, & Rymer, 2003; H. Johansson, 1991). Bone/cartilage alignments, ligaments, and joint capsules comprise the passive components, while muscles are the active structures. These active and passive structures act in unison to provide protection to a joint when it is exposed to potentially damaging forces (Dhaher et al., 2004; Dhaher et al., 2003). The initial forces are absorbed via the passive structures and the active structures deal with the subsequent forces (Schot & Dufek, 1993). Preparatory muscle activity (preactivity) also plays an important role in assisting with initial load absorption in landings (Santello, 2005).

The following sections will provide overviews of the structures and mechanisms involved in the control and maintenance of knee joint stability. Thereafter the role of muscle activation on ACL loading patterns will be discussed.

### *Passive Joint Stability*

Passive joint stability at the knee is maintained by the anatomical arrangement of the tibio-femoral joint, with added congruence provided by the lateral and medial menisci (Snell, 1995). Further stability is provided by the intracapsular cruciate ligaments, the ACL and the posterior cruciate ligament (PCL), the extracapsular collateral ligaments (medial and lateral), and the joint capsular structures (Snell, 1995). The PCL is

responsible for the majority of the restraint to posterior displacement of the tibia on the femur (Snell, 1995). The ACL has two bundles of fibers: the anteromedial and the posterolateral (Woo et al., 1998). The ACL is responsible for the majority of the restraint to anterior displacement of the tibia on the femur as well as internal and external rotation of the tibia with respect to the femur (D. L. Butler, Noyes, & Grood, 1980; Furman, Marshall, & Girgis, 1976; Markolf, Gorek, Kabo, & Shapiro, 1990; Woo et al., 1998). As a result of their location on the medial and lateral aspects of the knee joint, the medial and lateral collateral ligaments provide additional support to the knee joint, resisting valgus and varus forces respectively (Snell, 1995). Supplementary support is provided via other non-contractile structures including joint capsules and retinacular structures (Snell, 1995).

Although the term 'passive' has been traditionally employed to describe ligaments and other non-contractile structures, the existence of afferents from these ligaments and capsular structures has been conclusively confirmed (Fujita, Nishikawa, Kambic, Andrish, & Grabiner, 2000; H. Johansson, 1991; Solomonow et al., 1987; Tsuda, Okamura, Otsuka, Komatsu, & Tokuya, 2001). Mechanoreceptor afferents on the ACL are capable of affecting sensitivity of the muscle spindles through gamma motorneuron activation and subsequently heighten muscular reflexes (Dyhre-Poulsen & Krogsgaard, 2000; H. Johansson, Sjolander, & Sojka, 1990). Researchers have found direct connections from the ACL to the hamstrings in an ACL-to-hamstrings reflex arc pattern (Tsuda et al., 2001). Additionally, reflexive muscle contractions that serve to increase knee abduction stiffness secondary to stretch of the periarticular tissues have been

reported after valgus perturbations at the knee (Dhaher et al., 2003). Thus although the ligaments and other non-contractile tissues might not contract themselves, they play an important sensory role in the ability of the joint to prepare for expected loads.

### *Active Joint Stability*

Active joint stability is provided by the activation of the musculature surrounding the joint. At the knee, this primarily involves the quadriceps, hamstrings, and gastrocnemius muscles (Schot & Dufek, 1993; Wojtys & Huston, 1994). Neuromuscular control of joint stability is the activation of dynamic restraints occurring in preparation for, and in response to joint loads to maintain and restore joint stability (Reimann & Lephart, 2002a). This control is mediated through both feedforward and feedback systems (H. Johansson, 1991; Reimann & Lephart, 2002a).

Feedforward systems are anticipatory control systems that are activated before the actual imposition of joint destabilizing forces (Reimann & Lephart, 2002a). Feedback systems on the other hand, are the responses produced in response to sensory detection of these destabilizing forces (R. Johansson & Magnusson, 1991; Reimann & Lephart, 2002a). Because both these systems work so closely together, it is often times complicated to distinguish when one or both systems are active (Reimann & Lephart, 2002a). The use of the term “feedforward systems” has thus been suggested to encompass actions taking place at the beginning, or before the impending loads, with the term “feedback systems” referring to events in response to afferent inputs after load imposition (Reimann & Lephart, 2002a).

Preceding expected joint loads, there is preparatory muscle activation (preactivity) which assists with the short-range stiffness already present within the muscle. Upon actual load imposition, afferent receptors arising from the somatosensory, visual, and vestibular systems send sensory inputs to the central nervous system (Hewett & Paterno, 2002; H. Johansson, Sjolander, & Sojka, 1991; Reimann & Lephart, 2002b). These afferent inputs are processed at several levels and lead to a variety of motor responses, starting with short loop responses, then long loop responses, and finally with voluntary responses (Bennett, Gorassini, & Prochazka, 1994; Enoka, 1994; Lephart & Henry, 1995). The following sections will provide a synopsis of these responses.

*Short-range Stiffness*

In a normal muscle, there is a baseline amount of stiffness present due to the existence of active cross bridges that help maintain baseline muscle tone. This intrinsic or short-range stiffness represents the initial resistance to joint loads (Hoffer & Andreasson, 1981). This short-range stiffness is effective for the first 10 ms after joint load imposition and provides protection until reflex responses can take place (Kerney, Stein, & Parameswaran, 1997; Rack & Westbury, 1974).

*Preactivity*

Prior to actual imposition of joint loads, there is preparatory muscle activity in anticipation of the expected load (Horita, Komi, Nicol, & Kyrolainen, 2002; Santello, 2005). This preactivity is a feedforward mechanism contributing to joint stability

(Schmidt & Lee, 1999; Stokes, Gardner-Morse, Henry, & Badger, 2000). Memories from previous experiences are suggested to be involved with the production of this preparatory baseline level of muscle activity (Dhaher et al., 2004; Enoka, 1994; Wolpaw & Carp, 1990). Preactivity enhances the effectiveness of short-range stiffness already present in the muscle through active action-myosin cross bridges and consequently augments joint stiffness (Hoffer & Andreasson, 1981; Kerney et al., 1997; Rack & Westbury, 1974). Preactivity is important in that it assists the initial short-range stiffness in maintaining joint integrity until protective reflex and voluntary muscle contractions are elicited in response to joint loading (Dyhre-Poulsen & Mosfeldt Laursen, 1984; Greenwood & Hopkins, 1976).

*Short loop Responses*

Short loop reflex responses are the first responses from the spinal cord level. These responses represent a fast response through the monosynaptic reflex arc and do not require inputs from the central nervous system (Darton, Lippold, Shahani, & Shahani, 1985). Occurring around 30 ms after load imposition, these responses are important in protective reflexive joint stabilization (Darton et al., 1985; Enoka, 1994; Hewett & Paterno, 2002). Short loop responses enhance the stiffness of the joint after the cross bridges responsible for the intrinsic stiffness can no longer resist the imposed loads (Hoffer & Andreasson, 1981).

*Long loop Responses*

The subsequent level of motor control is via the lower brain (basal ganglia, brainstem, and cerebellum), which acts as a long loop latency response and represents the first line of commands from the higher centers (Hewett & Paterno, 2002). These responses typically occur around 50-60 ms after load imposition (Enoka, 1994). Long loop responses are also involved in the timing of motor activities and the learning of planned movements over time (Enoka, 1994).

*Voluntary Responses*

Finally, processing occurs at the cerebral motor cortex level, resulting in voluntary motor responses. These represent the slowest neural response because of the presence of multiple synapses and increased distance of impulse propagation (Hewett & Paterno, 2002). In the lower extremity, the earliest that this voluntary response can be initiated is around 145-157 ms (Chan, Jones, Kearney, & Watt, 1979), which might be too late when considered alone to protect from ligamentous injury (Konradsen, Voight, & Hojsgaard, 1997; Pope, Johnson, Brown, & Tighe, 1979).

*Muscle Coactivation and Knee Joint Stability*

Voluntary responses represent muscle activation under conscious control. In human movement, voluntary muscle responses can be broadly divided into two types of activation patterns: reciprocal activation, and coactivation (or cocontraction) (Humphrey & Reed, 1983; Smith, 1981). Reciprocal activation occurs in rhythmic motor processes

requiring alternate contraction and relaxation of agonists and antagonists; when external resistance prevents displacement or muscle shortening by the prime movers, relaxing the antagonist muscles; and in low velocity voluntary limb displacements without load (Smith, 1981). Coactivation on the other hand occurs in activities that either require high limb displacement velocities or are performed under loaded conditions, needing the agonists and antagonists to cocontract strongly to decelerate the limb (e.g. landing) (Smith, 1981). During these type of activities muscle tension needs to be precisely monitored (e.g. during the initial phase of learning a motor skill), assisting with joint stabilization to allow for precise movements (Smith, 1981).

Coactivation has been theorized to assist in joint stabilization and reduce ligament strain by allowing the musculature to absorb joint loads (Baratta et al., 1988; Da Fonseca et al., 2004; Doorenbosch & Harlaar, 2003; Noyes, Butler, & Malek, 1980). In a study investigating the role of muscular coactivation in maintaining knee joint stability (Baratta et al., 1988), muscle activation data were simultaneously recorded from the hamstrings and quadriceps of: a) normal healthy subjects, b) athletes in predominantly jumping sports who did not perform hamstrings exercises, and c) athletes who routinely performed other exercises including hamstring curls. Hamstrings muscle activity was significantly depressed in athletes who did not regularly perform hamstrings exercises as compared to healthy normal individuals and athletes who regularly performed hamstring exercises (Baratta et al., 1988). The results emphasize the role of muscular balance in joint stability in that those individuals with increased quadriceps, but lower hamstring activation (suggesting lower levels of coactivation) possibly could have reduced stabilizing

muscular forces available to attenuate forces during joint loading (Baratta et al., 1988). These lower levels of coactivation would then expose the ligaments to the majority of joint loading and subsequent chances of injury.

Higher levels of knee muscle coactivation (hamstrings and quadriceps) have been found to reduce external loads imposed on knee ligaments in the form of valgus/varus moments by as much as 90% during cutting maneuvers (Beiser, Lloyd, Ackland, & Cochrane, 2001). Quadriceps, hamstrings, and gastrocnemius muscle contractions have been noted to increase knee joint stiffness up to 48-400% (Goldfuss et al., 1973; Louie & Mote, 1987; Markolf, Graff-Radford, & Amstutz, 1978; Wojtys et al., 2002), and improve joint congruence (Baratta et al., 1988; Draganich et al., 1989; Markolf, Bargar, Shoemaker, & Amstutz, 1981; Markolf et al., 1978). Coactivation also has been noted to reduce reflex activity to help in the maintenance of balance and allow for precise movements via reduction in transmission from Ia afferents to motor neurons as tasks become more demanding (Llewellyn et al., 1990).

In a study examining dynamic knee muscle cocontraction levels, as compared to healthy subjects, individuals with ACL injury (range 3-36 months) had lower cocontraction levels both pre- and post- perturbation in a walking task (Da Fonseca et al., 2004). In this study, subjects walked across a platform which upon the investigator's manipulation instantaneously caused a 20° medial-lateral perturbation in the frontal plane during walking, leading to a slight varus and external rotation at the knee and femur respectively. While lower cocontraction levels were noted, the researchers could not speculate as to whether they were a predisposing factor to, or a result of the ACL injury

(Da Fonseca et al., 2004). In a related study comparing knee muscle cocontraction levels across genders and activity levels, no gender differences were noted in cocontraction levels during landings from 30 cm (Da Fonseca, Vaz, De Aquino, & Bricio, 2005). Further, vastus lateralis (quadriceps) and biceps femoris (hamstrings) cocontraction levels did not appear to correlate with ligament laxity, flexor/extensor torque ratio or flexor and extensor work values (Da Fonseca et al., 2005).

*Muscle Coactivation and the H-reflex*

The H-reflex is the electrical analog of the stretch reflex, but bypasses the effects of the gamma motorneurons and muscle spindle discharge (Schieppati, 1987). A H-reflex is recorded when the electrical stimulation of the nerve is above threshold for activation of the Ia afferents, and the afferent terminals are sufficiently depolarized to cause neurotransmitter release at the Ia afferent/ alpha-motoneuron synapse (Zehr, 2002). This reflex is evoked by the electrical stimulation of a mixed nerve, containing both sensory and motor neurons (Zehr, 2002). This stimulation involves both afferent sensory (from the point of stimulation to spinal cord) and efferent motor (from Ia motorneurons in the spinal cord to the neuromuscular junction) arcs, and also a direct (from the point of stimulation to the neuromuscular junction) efferent motor response (M wave) (Aagaard, Simonsen, Anderson, Magnusson, & Dyhre-Poulsen, 2002; Zehr, 2002). Thus, the H-reflex can be recorded with and or without an M wave (Zehr, 2002). An H-reflex is recorded if the electrical stimulation of the nerve is above threshold for activation of the Ia afferents and the afferent terminals are sufficiently depolarized to cause

neurotransmitter release at the Ia afferent/ alpha-motoneuron synapse (Zehr, 2002). The release of the neurotransmitter from primary afferents causes postsynaptic depolarization of the Ia motorneurons, which leads to propagation of action potentials causing neurotransmitter release at the neuromuscular junction resulting in depolarization and contraction of muscle fibers, and is consequently recorded as an H-reflex (Schieppati, 1987; Zehr, 2002). The H-reflexes therefore test the excitability of the motorneuron pool from the Ia monosynaptic pathway, whereas the stretch reflex tests the whole reflex pathway (Llewellyn et al., 1990; Nielson et al., 1993; Nielson & Kagamihara, 1992; Zehr, 2002). The H-reflex can be utilized to evaluate the changes in human reflex pathways and adaptive plasticity that takes place due to various stimuli like training (Zehr, 2002). The changes and modulation in the amount of the H-reflex therefore can provide information regarding the body's modifications and responses to different task demands and stimuli, and its consequent effects on task performance.

*Modulation of the H-reflex.*

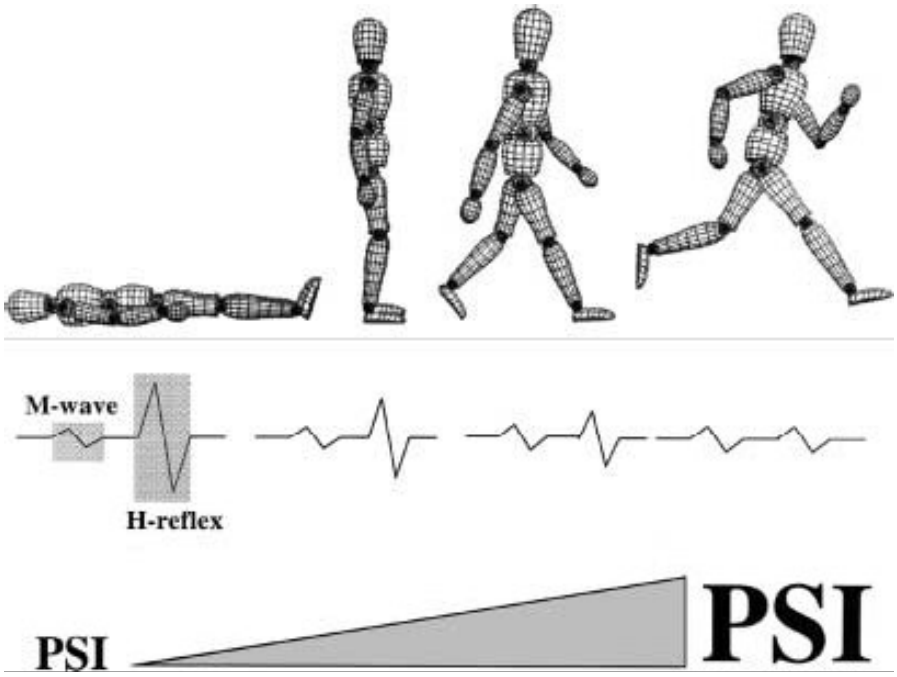
The amplitude of the H-reflex and the gains in the H-reflex can be affected substantially by movement and task performance (Zehr, 2002). With an increase in the level of difficulty of a motor task, proprioceptive sensitivity is elevated providing supraspinal areas with increased feedback gain and resolution (Llewellyn et al., 1990). In the segmental reflex arc however, this increased gain could lead to instability due to reflex modulated contractions (Aagaard et al., 2002; Llewellyn et al., 1990). Thus, when the need is to provide fine-tuning and motor control, the body tries to compensate for

possible instability produced by the reflex contractions by attenuating Ia motorneuron transmission via increasing presynaptic inhibition (PSI) (Llewellyn et al., 1990; Trimble & Koceja, 1994). This progressively increasing PSI on the alpha motoneuron pool consequently leads to a reduction in H-reflex amplitude (See Figure 1) and also allows for increased levels of cocontraction (Latash, 1998; Llewellyn et al., 1990; Zehr, 2002).

During normal activity, antagonist musculature opposing a contraction is reflexively relaxed for speed and efficacy of movement (Pearson & Gordon, 2000). This mechanism is known as reciprocal inhibition, and is mediated through the Ia motorneurons (Pearson & Gordon, 2000). In tasks requiring joint stabilization and precise control however, it is helpful to reduce reciprocal inhibition and simultaneously increase muscle cocontraction, allowing for contraction of both agonist and antagonist muscles together (Palmieri et al., 2004). Increased amounts of coactivation have been consistently noted to be associated with reductions in H-reflex activity (Llewellyn et al., 1990; Nielson et al., 1993).

In tasks that require greater stability and control, the H-reflexes are suppressed (Edamura, Yang, & Stein, 1991; Koceja et al., 1993; Llewellyn et al., 1990; Stein & Capaday, 1988; Trimble & Koceja, 1994). For example, smaller H-reflexes have been found in beam walking, as compared to walking (Llewellyn et al., 1990). Lower H-reflex gains have also been reported during dynamic activity as compared to resting conditions (Stein & Capaday, 1988). In attempts to reduce postural perturbations and allow maintenance of balance, neurologically sound individuals have been noted to successfully reduce the gain of their soleus H-reflex with training (Trimble & Koceja, 1994).

Figure 1: Decreases in H-reflex Amplitude with activity due to Increasing Presynaptic Inhibition




Adapted from Zehr, E. P. (2002). Considerations for use of the Hoffmann reflex in exercise studies. European Journal of Applied Physiology, 86, 455-468

Functional adaptations of the H-reflex have also been noted in difficult tasks that require increased balance (Trimble & Koceja, 1994; Wolpaw & O'Keefe, 1984; Zehr, 2002). Similar results have been noted in primates, where long-term functional plasticity of the sensorimotor system has been demonstrated to result in alteration of the amplitude of the H-reflexes (Meyer-Lohmann, Christakos, & Wolf, 1986; Wolpaw & Carp, 1990; Wolpaw & O'Keefe, 1984).

Interestingly, smaller H-reflexes have been noted in dancers as compared to other physically active groups (Mynark & Koceja, 1997; Nielson et al., 1993). In a study comparing H-reflexes of dancers from the Royal Danish Ballet with well-trained athletes (volleyball, soccer, handball, and basketball players), dancers were found to have significantly lower H-reflexes than the athletes at rest in a sitting position (Nielson et al., 1993). Similar results also have been noted in the H-reflex gain from when changing from a prone to a standing position between dancers and active controls (Mynark & Koceja, 1997). Although no H-reflex differences were noted between the groups in the prone condition, controls showed a consistent H-reflex gain in the standing condition, and dancers showed a notable decrease in the H-reflex gain (Mynark & Koceja, 1997). These results suggest that dancers probably have higher levels of coactivation, (as evidenced by lower H-reflexes) during functional activity as compared to other athletes and active control subjects (Llewellyn et al., 1990). However, whether these decreased H-reflexes truly indicate higher levels of coactivation is unknown, and no specific works were found in the current literature examining this relationship directly. Theoretically, if dancers do have decreased H-reflexes and consequently increased coactivation during activity, it

may imply a potential protective neuromuscular mechanism which may protect them from ACL injury and aid in the maintenance of joint stability.

*Summary*

Overall, joint stability is maintained by a combination of passive and active systems. Passive stability is maintained by bone and cartilage contact and ligamentous and capsular restraints. Active joint stability is primarily provided by activation of musculature around the joint. This activation occurs across a continuum which is based on specific external demands on the joint. Components of this continuum include short range stiffness, preactivity, short loop responses, long loop responses, and finally voluntary muscle activation. All these components collectively contribute through muscle activation (whether reflexive or voluntary) and positively add to joint stability by increasing joint stiffness in preparation for, or in response to joint loading (Chmielewski, Rudolph, & Synder-Mackler, 2002; Da Fonseca et al., 2004; Konradsen et al., 1997; Lark, Buckley, Bennett, Jones, & Sargeant, 2003; Shultz & Perrin, 1999). Muscle coactivation assists in maintaining joint stability by improving joint congruence and increasing joint stiffness. Higher levels of muscle coactivation prior to loading would theoretically allow the joint to be better prepared and assist with joint protection upon loading through increased joint stiffness (Bennett et al., 1994; Wojtys et al., 2002). Decreased levels of H-reflexes (possibly indicating higher levels of coactivation) have been noted in dancers as compared to other female athletes, suggesting a promising neuromuscular mechanism that may be protecting dancers from ACL injuries.

*Muscle Coactivation and ACL Loading*

The ability of muscle activation to provide joint stability has led researchers to investigate the effects of muscle activation on ACL loading patterns. Beynnon and colleagues have examined ACL strain in vivo through a series of studies (Beynnon & Fleming, 1998; Beynnon et al., 1995; Beynnon, Howe, Pope, Johnson, & Fleming, 1992; Beynnon et al., 1997; Fleming et al., 2003; Fleming et al., 2001). In 1998, they detailed a review of their previous work where they implanted transducers onto the anterior bundle of the intact ACL's of volunteers who already were candidates for diagnostic arthroscopic surgery (Beynnon & Fleming, 1998). Following implantation, the researchers measured ACL strain while volunteers performed various commonly prescribed rehabilitation exercises, and isolated and combined contractions of the quadriceps and hamstring muscles. The ACL was strained at low knee flexion angles (< 30°), irrespective of the type of contractions. At 15° of knee flexion, the highest ACL strain was produced with isometric quadriceps muscle contraction (4.4%) followed by decreased ACL strain values upon cocontraction of the hamstrings and quadriceps muscles (2.8%). The lowest ACL strain values were associated with isolated isometric hamstrings muscle contraction (0.6%). Similarly, at 30° knee flexion the highest ACL strain was produced with isometric quadriceps muscle contraction (2.7%), then with hamstrings and quadriceps muscle cocontraction (0.4%) and lastly by isometric hamstrings muscle contraction (0%). In another study, ACL strain was studied in 10° increments from 0-90 degrees using simulated joint compression (Draganich & Vahey, 1990). Isolated isometric quadriceps contractions and an equal force

quadriceps/hamstring cocontraction were applied in vitro to the knee. ACL strain was found to be reduced in the simultaneous cocontraction condition as compared to the isolated quadriceps contraction condition at 10, 20, and 90 degrees of knee flexion (Draganich & Vahey, 1990). Together, the results from studies investigating the effects of knee muscle activation and ACL loading suggest that contraction of the quadriceps, especially at low flexion angles (around 30°), and possibly in combination with rotatory loads causes increases in ACL strain values (Beynnon & Fleming, 1998; Draganich & Vahey, 1990). Further, hamstrings contraction progressively decreases the ACL strain with increasing knee flexion angles (Draganich & Vahey, 1990).

Despite the gastrocnemius crossing the knee joint, relatively lesser work (Fleming et al., 2001; Houck, 2003; Nyland, Caborn, Shapiro, & Johnson, 1997) has examined the coactivation of this muscle with the thigh muscles in controlling ACL loading and knee joint stability. Simultaneous contraction of the gastrocnemius muscle with the quadriceps, or with the hamstrings, has been reported to produce greater ACL strain than that produced either by the quadriceps or the hamstrings muscles acting alone (Fleming et al., 2001). In closed kinetic chain function, fatigue of the quadriceps muscles has been noted to lead to earlier activation of the gastrocnemius muscle, probably to compensate for the fatigued quadriceps muscles (Nyland et al., 1997). As the gastrocnemius is a synergist of the quadriceps, it has been suggested to be an antagonist of the ACL (Fleming et al., 2001). These studies underscore the role of the gastrocnemius muscle in influencing ACL strain and knee muscle activation patterns in dynamic activity.

Collectively, while isolated hamstrings contraction is the least threatening to the ACL, this is unfeasible when performing functional activities such as landing, as it is impossible to selectively isolate and contract the hamstrings to decelerate the body. Thus, coactivation of knee musculature is a safer, yet realistic muscle activation pattern to provide maximal joint stability during landing. This coactivation would be protective of the knee joint and the ACL through improvements in joint congruence (Baratta et al., 1988; Markolf et al., 1981) and increases in joint stiffness (Draganich et al., 1989; Hirokawa, Solomonow, Luo, & Lu, 1991; Humphrey & Reed, 1983; Solomonow et al., 1987). Increased knee joint stiffness via greater muscle contraction (Goldfuss et al., 1973; Louie & Mote, 1987; Markolf et al., 1978; Wojtys et al., 2002) demonstrates a possible protective mechanism against ACL injury by reducing the percentage of external force to be resisted by the ACL and other knee structures during movement (Wojtys et al., 2002).

### *Joint Stiffness*

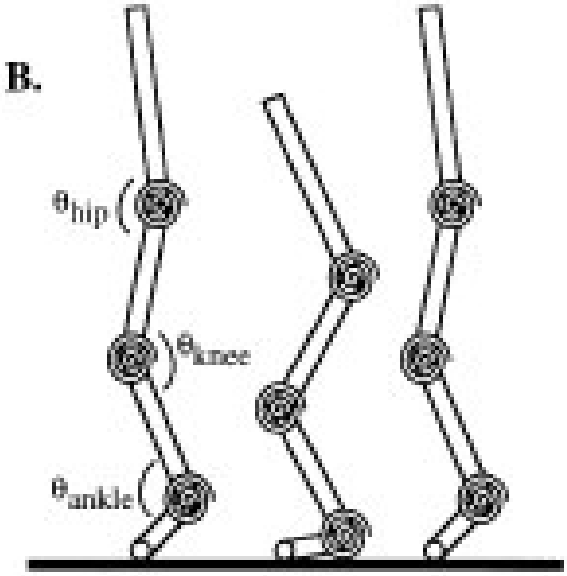
Joint stiffness (also known as torsional or rotational stiffness) is the resistance that a joint offers in response to an applied moment (Davis & DeLuca, 1996). Joint stiffness is expressed as the ratio of the change in moment to the change in joint displacement, and depends on the stiffness of each muscle-tendon unit crossing the joint (Farley et al., 1998; Farley & Morgenroth, 1999). The stiffness of each muscle-tendon unit in turn is dependent on several factors including the level of muscle activation which determines the number of active cross bridges, the cross-sectional area and angle of pennation of the muscle, the amount of passive connective tissue in the muscle, the velocity and volume of

load imposition, and tendon stiffness (Wojtys et al., 2002). Farley et al. (1999) have applied this definition of joint stiffness to a hopping model (See Figure 2) and it can be similarly utilized for a landing model.

If the load imposition is rapid and the muscles are completely relaxed, the voluntary responses initiated through feedback loops are too late and therefore incapable by themselves in protecting a ligamentous injury (Pope et al., 1979; Santello, 2005). However, if the muscles are already activated through feedfoward mechanisms based on previous memories, then the inherent increase in stiffness that is associated with muscle contraction would help protect the joint upon load imposition (Stokes et al., 2000; Wojtys et al., 2002). During a dynamic activity such as landing, this increased joint rotational stiffness via increased muscle preactivity prior to ground impact is necessary to prevent buckling and possible injury following ground contact (Dyhre-Poulsen & Mosfeldt Laursen, 1984; Lees, 1981; Santello, 2005; Santello & McDonagh, 1998). This pattern of muscle activity in preparation for expected joint loads in landing has been noted by several researchers (Duncan & McDonagh, 2000; Horita et al., 2002; Santello & McDonagh, 1998; Viitasalo, Salo, & Lahtinen, 1998).

While onset of muscular activity is thought to be timed to the expected time of ground contact (Duncan & McDonagh, 2000; Santello & McDonagh, 1998; Santello, McDonagh, & Challis, 2001; Viitasalo et al., 1998), the amplitude of muscle activation appears to be modulated by the expected demands of the task (Hass et al., 2005; McNitt-Gray, Hester, Mathiyakom, & Munkasy, 2001).

Figure 2 : Schematic Depiction of Joint Stiffness




Adapted from Farley, C. T., & Morgenroth, D. C. (1999). Leg stiffness primarily depends on ankle stiffness during human hopping. Journal of Biomechanics, 32, 267-273.

In landings, a distal-to-proximal timing pattern of muscle activation before ground contact has been noted ranging from ~90-192 ms in the gastrocnemius, to ~30-127 ms in the hamstrings, and ~43-132 ms in the quadriceps muscles (Arampatzis, Bruggemann et al., 2001; Liebermann & Goodman, In Press; McKinley & Pedotti, 1992; Viitasalo et al., 1998).

When an external perturbation in the form of a force is applied to a joint, muscles surrounding the joint are activated in order to increase joint stiffness and maintain dynamic stability (Hortobagyi & DeVita, 2000; Nichols, 1994). Higher muscle activation levels result in higher levels of stiffness and concomitant decreases in joint excursions that assist with joint stability (Arampatzis, Bruggemann et al., 2001; Arampatzis, Schade, Walsh, & Bruggemann, 2001; Horita et al., 2002). Although increased stiffness may theoretically reduce the potential for ligamentous and soft tissue injury by reducing the chances of joint buckling, too much stiffness may place an individual at risk for bony injury due to increased shock absorption via the skeletal system secondary to increased joint stiffness (DeVita & Hortobagyi, 2000; DeVita & Skelly, 1992). Conversely, lower levels of stiffness have been proposed to explain higher incidences of ligamentous injuries in females as compared to males (Granata, Wilson, & Padua, 2002b). There probably exists an optimal zone of stiffness, between which lower extremity injury risk may be minimized. However this optimal zone has not yet been determined (R. J. Butler, Crowell, & Davis, 2003). Therefore, until the ranges of this optimal zone are found, muscle coactivation theoretically seems to be an effective activation pattern to achieve dynamic stability in functional activities like landing.

### *Summary*

While control of joint stability is maintained by a combination of passive (e.g. ligaments) and active structures (muscles), voluntary responses represent the only form of muscle activation under conscious control. In expectation of, or in response to joint perturbations, the muscles around a joint coactivate to reduce the chances of buckling on joint loading. Decreased H-reflexes, associated with this coactivation have been noted in dancers as compared to other athletes. Increases in muscle activation levels have been noted to increase joint stiffness levels. Additionally, while isolated hamstrings contraction causes least strain on the ACL, this is not feasible in functional activity. Hence, coactivation of knee musculature (quadriceps, hamstrings, and gastrocnemius) might be the better option to maintain joint stability and reduce ACL strain during a functional activity like landing.

## Landing as a Model to Examine Joint Control Strategies

With landing being a frequent activity in human movement (Lees, 1981; Santello & McDonagh, 1998) and also one of the most common activities involved with ACL injury (Boden et al., 2000; Hutchinson & Ireland, 1995), it is important to understand the underlying neuromuscular patterns controlling impact absorption during landing (Santello, 2005). During landing, controlling the body segments presents significant challenges to the neuromuscular system to absorb and distribute the impact loads produced on ground contact (i.e. ground reaction forces) (DeVita & Skelly, 1992; Dufek & Bates, 1990; Lees, 1981; McNitt-Gray et al., 2001; Seegmiller & McCaw, 2003).

These impact forces on the body may be as high as 8.2 – 11.6 times body weight (Ozguven & Berme, 1988), and must be absorbed primarily by the lower extremity. If loads become excessive for the body to accommodate or if impact absorption fails, injury may occur (Dufek & Bates, 1990; James & Bates, 2003). Ground reaction forces reaching multiples of body weight have been measured for different kinds of landing activities (e.g. basketball rebounding, block landing in volleyball) indicative of high impact energy absorption over the lower extremity joints during these activities. Several researchers (DeVita & Skelly, 1992; Dufek & Bates, 1990; Lees, 1981; McNitt-Gray et al., 2001; Seegmiller & McCaw, 2003) have therefore used landings as a model to examine lower extremity injury risk factors.

### *Gender Differences in Landing*

Given that ACL injuries often take place during landing and occur at higher rates in females, several researchers (Decker et al., 2003; Kernozek, Torry, Van Hoof, Cowley, & Tanner, 2005; Lephart, Ferris, Riemann, Myers, & Fu, 2002; Sander et al., 2004; Urabe et al., 2005; Zazulak et al., 2005) have used the landing model to investigate gender differences in the neuromuscular and biomechanical control of the body during landing. While most of these researchers have found that gender differences exist in neuromuscular and biomechanical parameters, some (Fagenbaum & Darling, 2003; Garrison, Hart, Palmieri, Kerrigan, & Ingersoll, 2005) have not.

In a study investigating gender differences in intercollegiate basketball athletes during jump landing, females were noted to have higher activation levels in the vastus

medialis (quadriceps) from 15-45° of knee flexion and lower hamstrings activity at 15, 20, and 25° of knee flexion (Urabe et al., 2005). A comparison of electromyographic activity in the hip and thigh musculature between intercollegiate soccer and track (jumping events) athletes (13 female, 9 male) revealed that females had decreased gluteus maximus and increased rectus femoris (quadriceps) muscle activity than males during single-leg landings (Zazulak et al., 2005). Similarly, higher levels of quadriceps muscle activity in females have also been noted during landing by other researchers (Sander et al., 2004).

Obvious gender differences have also been noted in biomechanical parameters during landing. In a study evaluating kinematic, kinetic, and strength variables between collegiate basketball players (n=15) and matched male (n=15) subjects during a single-leg landing, females were noted to have lesser knee and hip flexion displacements, lower leg internal rotation angular displacement, and shorter times to maximum displacement (indicating more abrupt increases in force) (Lephart et al., 2002). However, in this study while sagittal knee angular excursions were found to be lower in females, initial knee flexion angles at ground contact was not reported. Thus while clear gender differences were found, whether this decreased excursion is beneficial or harmful is still unclear. Female college volleyball players have also been reported to have significantly lower knee and hip flexion angles, higher knee extensor moments, and lower knee varus moments than males during landings, and to land in more erect postures than males (Decker et al., 2003; Kernozek et al., 2005; Salci, Kentel, Heycan, Akin, & Korkusuz, 2004).

In the few studies that did not observe gender differences in landing performance, possible methodological concerns exist in data assimilation and statistical analyses. In their sample of varsity basketball players (females=8, males=6), Fagenbaum et al. (2003) used the averages of the medial and lateral quadriceps and hamstrings muscles to describe total muscle activity of knee extensors and flexors respectively. However, if artifacts were noted in the activity of one of the knee flexors or extensors, the activity level of the other muscle of the group was used to represent the combined activity instead of the averaged activation signal (Fagenbaum & Darling, 2003) possibly biasing the results. In another study, landing strategies were compared between intercollegiate soccer players (females=8, males=8), with muscle activation data recorded only around two time periods during the landing: 80 ms around initial contact (40 ms each before and after), and 40 ms around peak knee internal rotation moment (20ms each before and after) (Garrison et al., 2005). The researchers did not find any differences in muscle activity between genders. While this novel method of analyzing data around specific biomechanical events is interesting, the temporal sequence of these two time points might have differed between participants (intra-individual variability) and how these possible differences may have affected the results and subsequent interpretations made from the data are unknown.

### *Gender Differences in Other Dynamic Activities*

Gender differences have also been found during performance of various other functional tasks including hopping, running, cutting, and postural perturbations (Chappell

et al., 2002; Decker et al., 2003; Granata, Wilson, & Padua, 2002a; Lephart et al., 2002; Malinzak et al., 2001; Sander et al., 2004; Shultz et al., 2001). Malinzak et al. (2001) found that as compared to males, normalized integrated EMG amplitudes at initial contact prior to beginning cutting activities were 17 – 40% greater in the quadriceps and 20% lesser in the hamstrings in recreationally active females. Further, females had 11° greater knee valgus and 8 – 15° less knee flexion angle ranges during running, side-cutting and cross-cutting maneuvers (Malinzak et al., 2001). In a follow-up study comparing knee kinetics in three stop-jump tasks (forward, vertical and backward jumping), females (n=10) had greater knee extension and valgus moments, and greater anterior proximal shear force in the backward stop-jump task than males (n=10) (Chappell et al., 2002). The researchers consequently suggested that female athletes on average, may have knee motion patterns that bring them close more often to positions supposedly at high-risk for ACL injuries than males (Malinzak et al., 2001).

Collectively, studies investigating gender differences in landing performance suggest that when compared to males, females have higher levels of quadriceps (Sander et al., 2004; Urabe et al., 2005; Zazulak et al., 2005) and lower levels of hamstrings (Urabe et al., 2005) activation in landing tasks. Females have also have been noted to possibly demonstrate decreased gluteus maximus (hip extensor and hamstrings synergist) activity in landings as compared to males (Zazulak et al., 2005). Similarly, with regard to biomechanical parameters, females perform functional tasks with increased knee extensor moments (Chappell et al., 2002), and possibly lesser knee flexion and more knee valgus and hip rotation angles than males.

*Landings and Drop Jumps – Appropriateness of Model*

Although researchers often times make inferences about landings and injury based on data obtained from drop landing studies, in actual sport situations athletes more often perform jumping, running, or directional change activities immediately after the landing (e.g. rebounding in basketball). Thus, landing followed by a subsequent activity (e.g. drop jumps) might be a more functional model to examine lower extremity injury risk factors. Performing drop jumps involves landing from a height and immediately performing a maximal vertical jump upon ground contact (Viitasalo et al., 1998). In drop jumps therefore, muscle activity is needed not only to reduce the body's momentum to zero after ground contact, but also to accelerate the body against gravity to efficiently execute a subsequent jump (Arampatzis, Bruggemann et al., 2001; Bosco, Viitasalo, Komi, & Luhtanen, 1982; Gollhofer & Kyrolainen, 1991; Gollhofer & Schmidtbleicher, 1988; Viitasalo et al., 1998). Due to the differing functional outcomes following the initial landing in drop landings and drop jumps, dissimilarities in muscle activation patterns would be expected.

Prelanding muscle activity has been previously reported during landing activities (Duncan & McDonagh, 2000; Gollhofer & Schmidtbleicher, 1988; Kellis, Arabatzi, & Papadopoulos, 2003; McKinley & Pedotti, 1992; Viitasalo et al., 1998). Prelanding muscle activity reflects a strategy to prepare the body to smoothly absorb the impact of landing (Duncan & McDonagh, 2000; Horita et al., 2002; Viitasalo et al., 1998). Further, this muscle activity is important to prepare the musculo-tendinous complex for the rapid and forceful stretch occurring upon ground contact and throughout the following joint

rotations (Santello, 2005; Santello & McDonagh, 1998). The temporal and amplitude characteristics of prelanding muscle activity have been suggested to be modulated based on various factors including the drop height and the stiffness of the landing surface (Arampatzis, Bruggemann et al., 2001; Santello, 2005).

Postlanding muscle activity on the other hand, has been suggested to consist of a combination of preprogrammed central control overlapping with contributions from reflex mechanisms (Duncan & McDonagh, 2000; McDonagh & Duncan, 2002; Santello, 2005). While the preprogrammed component arises from prior experiences (Liebermann & Goodman, In Press), the reflex component arises after stretching of the musculo-tendinous complex upon landing (Duncan & McDonagh, 2000; McDonagh & Duncan, 2002). However, the exact interaction of the predictive and reflex components of neuromuscular activity is still not completely understood (Santello, 2005). During drop jumps, higher postlanding muscle activation amplitudes serve two functions: firstly leading to higher muscle stiffness allowing for a damping shock-absorber effect, and secondly preparing the body for the subsequent jump (Arampatzis, Schade et al., 2001; Viitasalo et al., 1998).

### *Summary*

With landing being implicated as a major activity associated with ACL injury, understanding of the neuromuscular control strategies underlying impact absorption in landing is imperative. Although gender differences have been identified in landing patterns, the exact mechanisms by which the body controls itself via selective

modulations of neuromuscular and biomechanical patterns leading to safe landings are unclear. Some research suggests that higher levels and earlier onsets of muscular activation might positively contribute to joint stability through increased joint stiffness during landing. Overall, the landing model, specifically the drop jump model appears to be a viable model to study the neuromuscular and biomechanical factors that control joint stability.

## Measurement of Neuromuscular and Biomechanical Parameters

To acquire a comprehensive understanding of the neuromuscular and biomechanical factors involved in landing, it is necessary to study how these factors are modulated throughout the landing task. Neuromuscular patterns in the form of muscle activation can be studied through the use of electromyography, while biomechanical patterns in the form of joint stiffness can be investigated using a combination of inverse dynamics solution and position data obtained from a three-dimensional electromagnetic motion tracking system.

### *Surface Electromyography*

Surface electromyography (sEMG) is a non-invasive measurement tool used to monitor electrical activity of muscle action potentials from a relatively large area of neuromuscular discharge (Basmaijian & De Luca, 1985; Enoka, 1994). sEMG is a popular tool used in the study of human movement (DeLuca, 1997; Yang & Winter, 1983), with researchers using it for various purposes including understanding of clinical pathologies (Bennis, Roby-Brami, Dufosse, & Bussel, 1996), performance related issues

(Arampatzis, Bruggemann et al., 2001; Gollhofer & Schmidtbleicher, 1988; Viitasalo et al., 1998), effects of training and conditioning programs (Chmielewski et al., 2002; Hewett et al., 1999), and in the study of injury risk and prevention factors (Baratta et al., 1988; Shultz, Carcia, & Perrin, 2004; Shultz et al., 2001).

sEMG can reveal information regarding the timing and amplitude of dynamic stabilizers during sport-specific activities that are associated with common NCACL injury mechanisms, and has been extensively used to quantify muscle activation about the knee (Chappell et al., 2002; Malinzak et al., 2001; Viitasalo et al., 1998; Zazulak et al., 2005). During landings, lower extremity muscles are activated to assist the body in attenuating impact forces encountered on ground contact (McNitt-Gray et al., 2001; Seegmiller & McCaw, 2003). sEMG can offer insights as to how the body adjusts muscle activation patterns, allowing for injury-free landings. The most common variables of interest from the sEMG signals are those of muscle timing (onset and duration) and activation amplitudes. Hence sEMG is a useful tool to assess possible differences in the timing, recruitment order and force produced by the muscles in response to a potentially injurious force (Huston et al., 2000; Shultz & Perrin, 1999).

The raw sEMG signal is constructed from multiple, bipolar waveforms formed from motor unit action potentials (Hillstrom & Triolo, 1995). The signal obtained is produced by the voltage differences between two electrodes as the action potential passes under two electrodes (Basmaijian & De Luca, 1985). When collecting sEMG data in a dynamic activity like landing, it is paramount to ensure that the signals are accurately detected and recorded, as the collection can be prone to unnecessary errors that can be

easily corrected with utilization of appropriate methods and instrumentation. Some of the potential sources of error have been identified by previous researchers (DeLuca, 1997, , 2002) and by the International Society of Electrophysiology and Kinesiology (Merletti, 1999) . These errors and their solutions are addressed when appropriate in the following discussion.

*Electrodes*

Accurate recording of sEMG data starts with the use of appropriate electrodes. The typical electrodes used in sEMG data collection are Ag/Ag-Cl electrodes, with an inter-electrode distance around 1-2 cm (DeLuca, 2002). The set up consists of two recording electrodes placed along the length of the muscle perpendicular to the direction of the muscle fibers. The electrode locations for individual muscles have been described in the literature (Rainoldi, Melchiori, & Caruso, 2004). An additional reference electrode is placed away from the site of the muscle recording, over a non-contractile area (Basmaijian & De Luca, 1985). This electrode configuration is known as a differential setting. Prior to application of electrodes, the overlying skin is prepared to remove hair, dead cells, oils and other substances thus reducing skin impedance and ensure reliable recording of sEMG signals.

*Amplification and Sampling*

When recording the sEMG signal, there is a possibility of recording signals that are not due to actual physiological events (noise) along with the signal (Winter, 1990). To minimize the amount of noise in the signal, it is important to limit cross talk from surrounding muscles, movement artifacts due to the dynamic nature of landings, and any other sources of electrical interference like power line hums (~50-60 Hz) (Basmaijian & De Luca, 1985; DeLuca, 1997; Winter, 1990). A differential setting as detailed above is used to reduce the potential noise. In this setting, any common signals between the two recording electrodes and the ground electrode are removed, and the remaining signals that are different at the two sites result in a 'differential' voltage that is recorded and amplified (DeLuca, 2002). The accuracy of how well the differential amplification can subtract the signals is measured by the Common Mode Rejection Ratio (CMMR). In theory, a perfect machine would have a CMRR of infinity. In practice however, a CMRR of > 80 dB, 10,000:1 (meaning that all but 1/ 10,000$^{th}$ of the noise will be rejected) is generally adequate. Also, the input impedance of the instrumentation is generally set at least 10 times higher than that of skin impedance to around 1 MΩ so that it effectively stabilizes the voltage changes due to skin impedance, and reduces chances of the noisy signals being amplified (DeLuca, 1997).

The raw signal then needs to be digitally sampled into the computer. This is usually done by using an Analog-Digital board. Most of the physiological information contained in a surface EMG signal is within the 5-500 Hz range (Winter, 1990). Thus the sEMG sampling rate needs to be at least 1000 Hz, which would prevent aliasing under

the concept of the Nyquist theorem. This theorem suggests that the data sampled be at least twice the frequency of the expected rate at which the raw data are produced to reduce possible distortion of the signal when it is digitally sampled (Basmaijian & De Luca, 1985).

*Signal Filtering*

To make clinically meaningful comparisons between groups, the signal then needs to be filtered and processed appropriately (Basmaijian & De Luca, 1985; DeLuca, 1997; Shultz & Perrin, 1999). It is important to be aware of the effects that different signal conditioning and processing techniques can have on the actual sEMG signals before processing any raw data. This is necessary because it has been demonstrated that different techniques can affect statistical results and interpretations of sEMG signals (Gabel & Brand, 1994).

Signal filtering is performed via digital filters which selectively reject or attenuate certain frequencies within the physiological frequency spectrum (Winter, 1990). These filters are described based on which frequencies they allow and which frequencies they reject. As mentioned earlier, the majority of the EMG signal lies between 5-500 Hz; frequencies on either ends of this range (i.e. below 5 Hz and above 500 Hz) would have more noise and lesser EMG signal, and are consequently filtered out. Additionally there may be noise within the signal range (e.g. @ 60 Hz – Power Line Hums), which can be removed through the use of a notch filter set at a specified frequency. Various types of filters are used to filter these data, with low-pass filters removing signals above the pre-

selected frequency, high-pass filters removing signals below the pre-selected frequency, band-pass filters removing signals above and below the pre-selected frequency, and notch filters removing signals at a pre-selected frequency (DeLuca, 2002).

It is important to remember that no filter is ideal and that there will be some true physiological signal lost with filtering, especially near the cut off ranges, possibly causing slight signal distortion (DeLuca, 2002; Winter, 1990). Thus, a compromise is needed when filtering between the cut-off frequency and the roll-off of the filter (i.e. how sharp the transition from the 'accept' to the 'reject' range is). Additionally there might be a phase-lag due to the inability of the filter to act instantaneously on the inputted signal (Delsys-Inc., 2002). The filter then would cause a systematic phase shift on all the data. To counter this systematic phase shift of the data, a reverse filter needs to be conducted at the same frequency to correct this phase lag. This type of filter is known as a zero-lag filter, with a Butterworth filter being commonly employed for this purpose.

*Signal Processing*

After signals are smoothed, they need to be processed to determine the overall, representative muscle activity. Selected processing techniques (Gerleman & Cook, 1990; Hillstrom & Triolo, 1995; Winter, 1990) are detailed below.

*Rectification.*

Rectification refers to converting all signals to a single polarity by obtaining an absolute value of the signal (Basmaijian & De Luca, 1985; Winter, 1990). Full wave

rectification inverses one polarity (usually negative), whereas half wave rectification refers to the elimination of one polarity (e.g. negative polarity), but is not usually preferred as it involves loss of signals.

*Root Mean Square (RMS).*

This is a processing technique that is popular among engineers to quantify errors and is becoming increasingly popular when examining sEMG during dynamic activities (Garrison et al., 2005; Myer et al., 2005; Shultz et al., 2005; Zazulak et al., 2005). The RMS value is a measure of the power of the signal, thus it has a clear physical meaning and is recommended for EMG data processing (Basmaijian & De Luca, 1985; DeLuca, 2002). Obtaining the RMS value is a three step process; first, the signal is squared; second, the mean of the squared signal is obtained over a specified time period; and third, the root of the quantity is taken (Hillstrom & Triolo, 1995). With increases in the time interval of the averaging, fewer data points are produced resulting in a smoother signal and vice versa (Shultz & Perrin, 1999). Thus, the time constants used for this process need to be reflective of the research questions being asked, as smoothing with higher time constants may mask or filter out subtle phase changes (Shiavi, Frigo, & Pedotti, 1998). Thus wide time constants should be only used when the variable of interest is mean amplitude and not timing interactions between events (Merletti, 1999).

*Ensemble Averaging of Signals.*

Ensemble averaging involves superimposing several trials on one another to produce a combination or averaged signal that is representative of the EMG activity of all trials for that individual (Shultz & Perrin, 1999). The advantage of this process is that over a succession of trials, possibly aberrant and erroneous spikes in myoelectric activity are eliminated to obtain an 'overall' measure of an individual's movement patterns. Newer software programs allow for this processing electronically. When ensemble averaging trials, it is important to obtain all trials over the same exact time so that all data points can line up equally. If the exact same time periods are not obtained, an alternative method is to linearly interpolate or extrapolate the raw data and divide it into the required number of data points before ensemble averaging the data. However, this procedure should be used with caution as it may introduce errors if the raw data does not follow a linear pattern.

*Measuring Temporal Differences*

One of the most basic pieces of information obtained from the sEMG signal is whether the muscle was active or not at a particular point in time. To determine whether a muscle was truly active, it needs to exceed a pre-defined threshold (Gerleman & Cook, 1990; Shultz & Perrin, 1999). Pre-defining this threshold value helps to accurately determine muscle activation onset times by clearly identifying when the muscle activity deviated from baseline EMG activity. While DeLuca (1997; 2002) has suggested that a $\pm$ 2 SD (standard deviations) from baseline would allow one to be 95% confident that the

muscle onset time has been accurately recorded, other researchers (Gabel & Brand, 1994) have used ± 3 SD as their threshold to increase their procedural rigor. In a study examining the role of skill in a landing task, onsets were determined by the maintenance of the EMG signal above the 95% confidence interval (i.e. 2 SD) from baseline activity for more than 10ms (McKinley & Pedotti, 1992). In gait studies, muscle onsets have been defined as muscle activity from a range of at least ± 1 SD above baseline for at least 30 ms (Tang, Woollacott, & Chong, 1998) to ± 3 SD above baseline from the quietest 100 ms period during each EMG trace (Mickelborough, van der Linden, Tallis, & Ennos, 2004). As baseline activity might differ across muscles, it may be necessary to individually set the threshold value for each muscle (Shultz & Perrin, 1999). Further, the muscle activation needs to be sustained over a certain time period (usually 25 ms) to ensure that the recorded time truly signifies the actual onset of muscle activity and not just an erroneous spike in the activity.

The width of the RMS smoothing window needs to be small to detect subtle yet clinically significant onset time differences between groups. Wider time windows can lead to oversmoothing, causing subtle phasic changes in muscle activation to be washed out from the original signal (Merletti, 1999). Additionally wider windows may also cause time and phase shifts in the data, actually changing temporal parameters (Merletti, 1999). A compromise needs to be therefore made by keeping the window wide enough to get adequate data smoothing, while not making this window too wide as to over smooth the data. In previous research examining preactivity onset times during landing, smoothing window durations have ranged from 3-25 ms (Garrison et al., 2005; Urabe et al., 2005).

RMS window lengths between these ranges would be adequate to reach an acceptable conciliation between oversmoothing the data and enough filtering.

*Measuring Amplitude Differences*

Differences in muscle amplitude levels is another important factor to consider when studying landing strategies, as it offers an indirect insight into the forces and torques produced by the muscle (Basmaijian & De Luca, 1985; DeLuca, 1997; Winter, 1990). Although the relationship between muscle activation and force produced varies over different muscles, ranges of motion and types of contraction, this information about activation amplitudes is useful in attaining an overall idea of the force produced by a particular muscle during activity.

To properly compare muscle activation levels between groups and with other studies, sEMG signals need to be normalized and expressed as a percentage of an individual's own maximum voluntary contraction (MVC) (Knutson, Soderberg, Ballantyne, & Clarke, 1994). Several researchers have attempted to ascertain the most reliable method to obtain this maximum activity. Knutson et al.(1994) compared three normalization techniques: a) MVIC (maximum voluntary isometric contraction), b) the peak dynamic value (peak EMG during the activity), and c) the mean dynamic value (mean EMG during activity) and found that reproducibility using the MVIC was the best. Similarly, Burden et al. (2003) compared four methods of normalizing gait EMGs; the three mentioned above, with the fourth being an isokinetic MVC. The MVIC normalization method was again found to yield EMG values that would be accurate

representations of the muscle activity required (Burden et al., 2003). Thus, the MVIC method appears to be the most stable and valid method to study amplitude differences in landing. When measuring this MVIC, the recommendations suggested by the International Society of Electrophysiology and Kinesiology (Merletti, 1999) should also be followed.

For a landing study, important information regarding amplitude of muscle activity includes information of the amount of muscle activity present before ground contact. This provides a measure of how the body is preparing for the impending load. Further, the calculation of muscle activity over a predetermined period of time after ground contact (e.g. 50 ms post contact) would shed light on the body's response to the impact forces and joint loading through the time period where the body would be in the load absorption (descent) phase. As suggested earlier, mean RMS amplitudes, previously used by other researchers (Garrison et al., 2005) is a good option to study these responses of the lower extremity muscles in landing tasks. With very subtle phasic changes in muscle activation of lesser interest than the overall mean activation amplitudes, wider time windows (~100 ms) would suffice when filtering amplitude measures. These wider windows would provide a good smooth signal to study overall mean differences in muscle activation amplitudes between groups.

### *Calculation of Kinetics and Kinematics*

The comprehensive biomechanical analysis of human movement during dynamic activity requires the study of movements associated with individual body segments and

the forces and moments that cause or impede those movements. The description of human movement is known as kinematics and is not concerned with the forces (either internal or external) that cause the movement, but rather with the details of the movement itself (Winter, 1990). The analysis of the forces causing the movement on the other hand is known as kinetics (Winter, 1990).

*Electromagnetic Tracking Systems*

Electromagnetic tracking systems (e.g. Flock of Birds® ) utilize individual position sensors that are attached to bony segments such as the foot, tibia, femur, and sacrum to acquire position data (Madigan & Pidcoe, 2003). Recording position data of the human segments is made possible by establishing two coordinate systems: global and local (Wu, 1995). The global or fixed coordinate system is defined by an orthogonal (X, Y, Z) axis system and provides the three-dimensional environment in which the movement takes place (Wu, 1995). A local coordinate system for each body segment is used to establish the segment's position in three planes ($X_i$, $Y_i$, $Z_i$) and orientation about three axes (rotation around each X, Y, and Z axis) within the global coordinate system's environment (Wu, 1995). The system is therefore a six degree-of-freedom measurement device (the 3 position data points: $X_i$, $Y_i$, $Z_i$, and the 3 orientation data points about the X, Y and Z axes). In the case of electromagnetic tracking systems, the position and orientation of the sensors are measured with respect to the transmitting antenna that is fixed in space (metal mapped), and emits a three-dimensional magnetic field through a pulsed DC (direct current) signal.

*Calculation of Position and Orientation Data.*

A multiple step process is used to calculate position and orientation data. First, the individual's anthropometrics are measured. Next, each segment of the individual's body is assigned its own unique mass and length based on the person's height and weight. Normative anthropometric data charts (Dempster, 1955) are already present within the computer software. These data charts are used to estimate segment mass, length, and radius of gyration using predetermined criteria according to a particular individual's mass and height (LeVeau, 1991). Kinematic raw data (position data) are determined by using motion tracking systems such as video or electromagnetic tracking systems. These data are then used to calculate the angular displacements using the local coordinate system, and can determine joint excursions during activity (Winter, 1990).

*Calculation of Joint Moments using Inverse Dynamics.*

During activities that require the ground contacting the floor, estimates of the individual joint forces and moments are calculated through an inverse dynamics solution (Seliktar & Bo, 1995; Winter, 1990). In this case, three pieces of information are necessary to make calculations: 1) anthropometric data, 2) raw kinematic (position) data, and 3) raw force data. The anthropometric and kinematic data are obtained as explained above. Force data are acquired through forceplates on which the individual lands (Seliktar & Bo, 1995; Winter, 1990). Once the anthropometric, position, and force data have been obtained, joint moments are calculated as the product of the force produced by the muscle and the perpendicular distance from the muscle's line of action applied to the

joint axis or the instantaneous center of rotation of the joint (if the joint is polycentric, i.e. where the center varies with joint angles e.g. the knee) (Baratta et al., 1988; LeVeau, 1991). This joint moment represents the combined moment required by both the active (muscles) and passive (ligaments, capsule) structures of the joint to overcome all the external forces imposed on the joint (Winter, 1990). As anthropometrics differ between individuals, the moments produced also differ. Therefore to allow for inter-subject comparisons, these moments need to be normalized. Different methods have been employed to normalize these moments, including (a) normalizing moments by each individual's own body mass, and (b) by a product of body mass and height. In a study comparing these two methods of normalization of moments during gait, Moisio, Sumner, Shott, and Hurwitz (2003) found that while both methods of normalizing moments were effective, using the body mass times height method could have potential for overcorrection. Thus, normalizing moments by an individual's own body weight seems to be the preferred method for normalizing moments during activity.

*Calculation of Joint Stiffness.*

From the measures of net moments, and the changes in the angular displacement, joint stiffness can be calculated. Joint stiffness has been defined as the ratio of the change in net moment to the change in angular displacement from the beginning of the ground contact phase and the instant when the joint is maximally flexed (Farley et al., 1998; Farley & Morgenroth, 1999) (See Figure 2).

*Summary*

Surface electromyography is a valuable tool in appreciating temporal and amplitude differences in neuromuscular patterns during landing, provided appropriate procedures are followed. Similarly, electromagnetic tracking systems in association with forceplates offer insights into how kinetics and kinematics in terms of joint stiffness are affected due to changes in neuromuscular patterns during activity. Together, both these techniques can help provide a comprehensive view of the movement patterns of the body during landing.

## Overall Summary

ACL injuries account for almost one billion dollars in healthcare costs annually. For reasons not completely understood, females have a 3-8 times higher risk of these injuries than their male counterparts in selected activities. Although several differences in neuromuscular and biomechanical parameters have been identified in the recent years in activities that are considered high risk for ACL injuries, no conclusive solution or 'silver bullet' has been found to modify the incidence of ACL injuries.

We are unaware of any studies that have examined differences in neuromuscular and biomechanical patterns between female basketball players and dancers when performing drop jumps. This is in spite of the knowledge that this activity, (a possible high-risk activity for ACL injury) is regularly performed by both populations, but that dancers seem to suffer far fewer ACL injuries as a result of this activity. Diminished H-reflexes suggest that dancers may have higher levels of coactivation during activity.

Evaluating how neuromechanical patterns differ between these populations would further our understanding of how the body prepares for different tasks through selective adjustments of lower extremity musculature. This information may help health care professionals protect athletes from these devastating injuries by designing, and if necessary, changing strength and conditioning training programs to improve neuromuscular function and dynamic stability. While sEMG has been shown to be a useful tool to assess gender differences in functional tasks through the assessment of neuromuscular response characteristics, it offers limited information on the resulting joint motions and forces in response to these joint control strategies. Similarly, although kinematic and kinetic data provide information regarding the forces and moments experienced at the joint upon landing, and the displacements of various body segments, they do not provide information as to the neuromuscular factors that produce those biomechanical patterns. In order to obtain a holistic view of the body's responses to potentially high ACL injury risk activities therefore, it is necessary and desirable to assess both neuromuscular and biomechanical function, by examining how neuromuscular function (i.e. muscle activation) can lead to biomechanical changes (i.e. joint stiffness).

# CHAPTER III

# METHODS

## Design

This study followed a single-session quasi-experimental design. Two groups of participants (female dancers and basketball players) performed 5 double-leg drop jumps from a 45cm box while neuromuscular and biomechanical parameters on the preferred landing leg were measured. Muscle activity was acquired via surface electromyography to assess neuromuscular differences in muscle activity onsets and amplitudes in the lower extremity. Knee joint stiffness was calculated from kinetic and kinematic data acquired through a Bertec non-conducting forceplate interfaced with a three-dimensional electromagnetic tracking device. Groups were then compared on muscle activity onset times, muscle activity amplitudes by way of separate 2 x 4 ANOVAs, with a one-way ANOVA comparing group differences in knee joint stiffness. A stepwise linear regression examined the ability of prelanding muscle activation amplitudes and group membership to predict knee joint stiffness.

## Participants

Healthy females (35 dancers, 20 basketball players) between the ages of 18-30 years, who had no musculoskeletal injury to either lower extremity for the past 6 months,

no previous surgery on either lower extremity, no history of cardiovascular or neurological problems, and no pre-existing conditions that would have detracted from the ability to land or jump participated. The age group criterion was used as a large majority of ACL injuries have been noted to occur in this age group (Miyasaka et al., 1991). Only those whose primary form of physical activity was dance or basketball for at least the past two years, and who were involved in dance or basketball at least 3 days/week for at least 30 minutes/day were recruited to participate in the study. The reason for recruiting only physically active participants was that most ACL injuries occur during participation in physical activity (Boden et al., 2000; Ferretti et al., 1992; Gray et al., 1985; Miyasaka et al., 1991; Olsen et al., 2004). All data were collected within the first 10 days of the participants' menstrual cycle, as previous data suggest that sex hormones may affect knee laxity (Shultz, Sander, Kirk, & Johnson, 2004), which in turn may affect muscle activation patterns at the knee (Shultz, Carcia et al., 2004).

## Instrumentation

All sEMG data were collected using a 16-Channel Myopac surface EMG unit (Run Technologies, Mission Viejo, CA). The sEMG unit has an amplification of 1mV/V with a frequency bandwidth of 10 to 1000Hz, a common mode rejection ratio of 90dB min at 60Hz, an input resistance of 1 MHz and an internal sampling rate of 8 KHz. Bipolar, Ag/Ag-Cl surface electrodes (Blue Sensor N-00-S; Ambu Products, Ølstykke, Denmark; skin contact size 30x22mm) with a center-to-center distance of 20 mm were used to collect the sEMG data. A Biodex System 3 isokinetic dynamometer (Biodex

Medical Systems Inc.; Shirley, NY) was used to position the participant at a fixed knee flexion angle of 30 ° during the maximum voluntary isometric contraction (MVIC) trials. sEMG data were acquired, stored, and analyzed using the Datapac 2K2 Lab Application Software (Run Technologies; Mission Viejo, CA). sEMG activity were synchronized with a type 4060 non-conducting forceplate (Bertec Corporation; Columbus, OH) using a trigger sweep mode. The sEMG signal was synchronized with the forceplate, so that foot contact with the forceplate that resulted in a force higher than 10 N triggered simultaneous collection of sEMG data. All sEMG were sampled at 1000 Hz.

Kinematic data were collected at 120 Hz using a three-dimensional electromagnetic tracking system (Motion Star Hardware, Ascension Technology, Burlington, VT, USA; Motion Monitor software, Innovative Sports Training; Chicago, IL) and kinetic data were collected at 1000 Hz using a Type 4060 non-conducting forceplate. Kinematic set up included the attachment of six-degree-of-freedom position sensors (Ascension Technologies; Burlington, VT) on the lower extremity of the preferred landing leg to record the movement of the lower extremity during the drop jumps.

## Procedures

### *Informed Consent and Demographics*

Participants reported to the Applied Neuromechanics Research Laboratory and signed an informed consent form approved by the Institutional Review Board at the University of North Carolina at Greensboro (Appendix A). Demographics including age,

height, weight, and years of experience in dance or basketball were recorded using a Demographics Data Sheet (Appendix B). To demonstrate day-to-day reliability of the sEMG and biomechanical data, 10 randomly chosen participants (5 dancers and basketball players each) returned for a second testing session, where the exact same procedures were repeated.

### *Body Composition Measurements*

As body composition may potentially affect the amount of weight that has to be balanced by the lower extremity muscles upon landing, body fat percentages were measured using Jackson and Pollock's 3-site skin fold measurement technique (Jackson & Pollock, 1978). A Lange Caliper (Beta Technology Inc., Santa Cruz, CA) was used to measure the thickness of the skin folds over the following sites: (a) triceps, (b) supra iliac, and (c) thigh. The exact procedures for each body part are detailed in Appendix C. The formula used for measuring body fat percentages for females was based on that suggested by Pollock, Schmidt, and Jackson (1980) as: Percent Body Fat = [(4.95/Db)-4.5] X 100, where Db = Body Density and is calculated as: Db = 1.099421 - .0009929 (Sum of three sites) + .0000023 (Sum of three sites) $^{2}$ -.0001392 (Age in years).

### *Determination of Preferred Leg*

The preferred leg was determined by asking participants to perform 3 single-leg landings from a 45 cm box placed above two forceplates. The leg that the participant used to land 2 out of 3 times was chosen as the preferred landing extremity. All subsequent

recordings were done on this extremity. While basketball players wear athletic footwear during playing, depending on the type of dance form, dancers may either use some footwear (e.g. tap dance, ballet) or may dance barefoot (e.g. modern dance, folk dance). Hence, to reduce the possible confounding effects due to variable foot-shoe-ground interfaces present amongst differing types of footwear, all testing was done barefoot.

### *Task Familiarization*

The same investigator then demonstrated the drop jump task for all participants. The participants were asked to stand on a 45 cm box, extend their preferred leg and then drop off the box, performing a double-leg landing, and with the preferred leg landing on the forceplate synchronized to the sEMG trigger mechanism (See Figure 3). As soon as they made contact with the ground, participants were instructed to immediately perform a maximal vertical jump and land back again onto the forceplate. Throughout the landing trial, they were asked to look forward at a marker placed at eye level in front of them, and keep their hands on their hips at all times. Looking forward at all times simulated real life functional activity as both dancers and basketball players do not always look at the ground during landing from a jump, and keeping hands on the hips standardized the task performance procedure. Participants were also asked to maintain their balance upon landing and not move off of the forceplates until told to do so by the investigator. Sufficient practice was allowed for participants to become comfortable with the task, and the number of practice trials was recorded. The 45 cm height was used as this height has been used previously (Duncan & McDonagh, 2000; Zazulak et al., 2005). Further, this

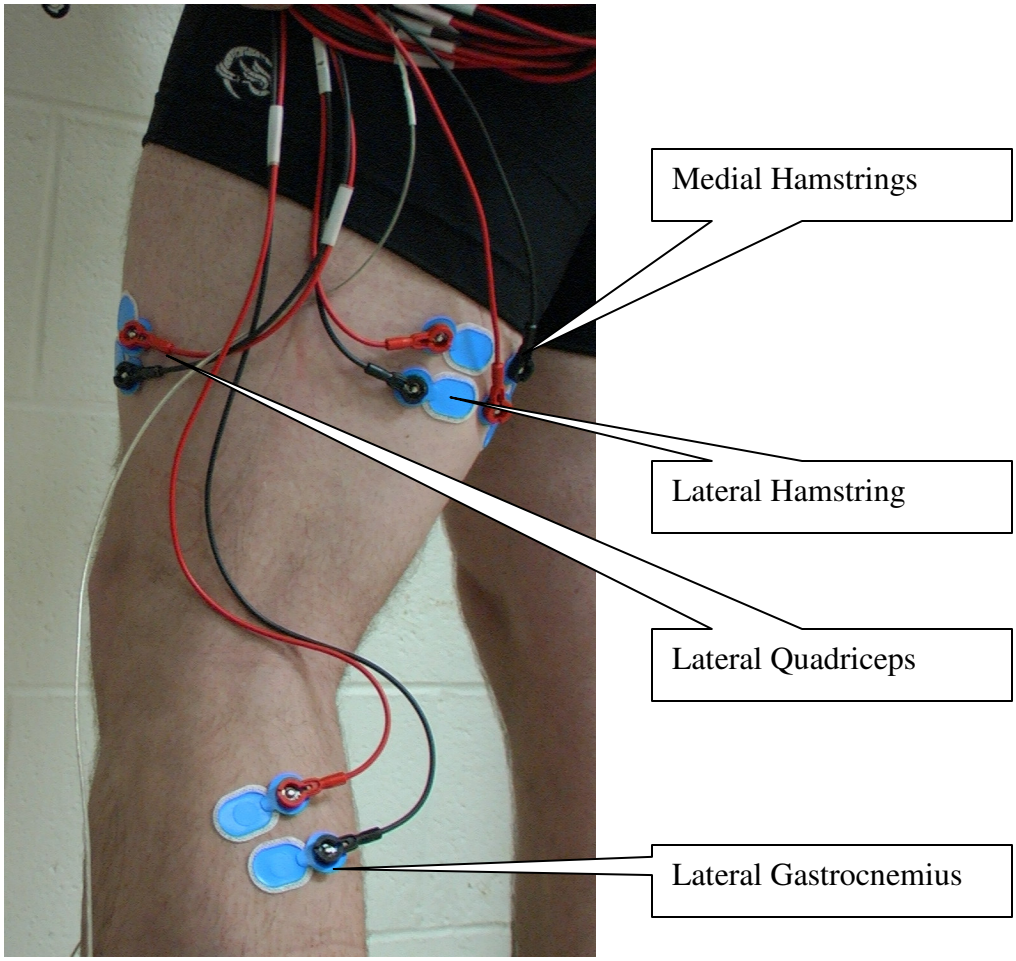
Figure 3 : Actual Performance of the Drop Jump Task

height also has been reported to be the average maximum vertical jump height for women (Chu, 1996; Patterson & Peterson, 2004). In the current study, both groups were physically active, and thus were assumed to comfortably be able to land from this height, allowing for functional mimicking of the jumping and landing demands in their own respective activities.

*Electrode Placement*

Prior to electrode placement, the skin over the muscle bellies of the lateral quadriceps (LQ), medial hamstrings (MH) and lateral hamstring (LH), and the lateral gastrocnemius (LG) muscles and the anteromedial aspect of the tibia (reference electrode) of the preferred leg was shaved and wiped with alcohol swabs. The electrodes were oriented perpendicular to the length of the muscle fibers and placed midway between the motor point and the distal muscle tendon of the LQ, MH, LH, and the LG (Shultz et al., 2005). The reference electrode was attached over the flat anteromedial bony aspect of the tibia, midway between the tibial tuberosity and the intermalleolar point. A typical electrode set up for data collection can be seen in Figure 4. Absence of crosstalk between the electrodes was then visually confirmed with manual muscle testing using the scope mode of the data acquisition software. To prevent any pulling or twisting of the wires during activity that potentially could affect the sEMG signal, the electrodes and wires were secured to the skin using stress loops with pre-wrap and regular white athletic tape.

Figure 4 : Placement of Surface Electrodes



*Collection of Maximal Voluntary Isometric Contraction (MVIC) sEMG Signals*

Participants then performed maximal voluntary isometric contractions of each muscle for normalization purposes while seated in a Biodex dynamometer. The preferred lower extremity was secured at 90° of hip flexion and 30° of knee flexion for all trials, with the resistance pad placed along the shaft of the tibia, taking care to see that the reference electrode was not impinged. Participants were told to keep their arms crossed over their chest, holding the shoulder pads at all times. The quadricep muscles were tested by asking participants to kick out with their leg as hard as possible for period of 5s, trying to extend their knee that was secured at 30° of knee flexion. For the hamstring muscles, participants were asked to bend their knee as hard as possible for a period of 5s, trying to flex their knee while it was secured at 30° of knee flexion. For the gastrocnemius muscle, participants were asked to plantar flex their foot as hard as possible for 5s into the hands of the investigator, who offered isometric manual resistance in the direction of dorsiflexion. A 30 s rest interval was given between each trial.

*Attachment of Position Sensors*

Participants were then prepared for collection of kinematic data by attaching position sensors over each of the following locations: 1) the dorsal aspect of the foot over the mid-shaft of the 3$^{rd}$ metatarsal bone; 2) the flat anteromedial aspect of the tibia, just above the reference electrode; 3) the lateral aspect of the femur, midway between the greater trochanter and the lateral epicondyle of the femur; and 4) over the flattest part on

the midline over the sacral bone, just below the level of the posterior superior iliac spines. A typical sensor set up for data collection can be seen in Figure 5.

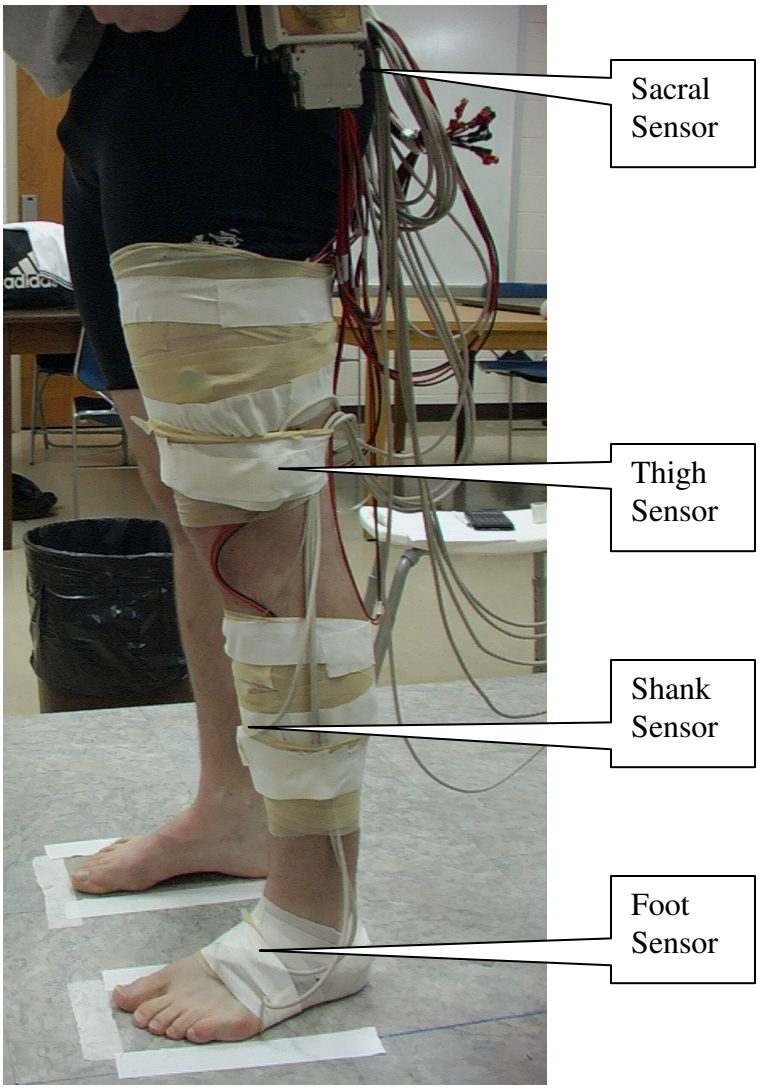
### *Digitization*

Digitization procedures were then performed by identifying and placing the movable sensor systematically on selected anatomical landmarks as required by the software to calculate kinematic data. A segmental reference system was defined for all body segments with the positive Z-axis defined as the medial to lateral axis; the positive Y-axis defined as the distal to proximal longitudinal axis; and the positive X-axis defined as the posterior to anterior axis. Knee flexion angles were calculated using Euler angle definitions with a rotational sequence of Z Y'X". Estimation of the joint centers were calculated based on the midpoint between two points on the medial and lateral aspects of the ankle and knee joints while a series of thigh positions relative to the sacrum were used to estimate the hip joint center (Leardini et al., 1999; Madigan & Pidcoe, 2003). Kinetic measures were recorded from the forceplate using the motion monitor software.

### *Task Performance*

Participants then performed 5 double leg drop jumps from the 45 cm box. They were asked to drop off the box and perform a vertical jump as high as possible immediately upon ground contact, and land back onto the forceplate while keeping their hands on their hips and looking forward at all times. A graphical representation of the drop jump task and actual participant positioning on ground contact can be seen in

Figure 5 : Placement of Position Sensors



figures 6 and 3 respectively. A rest interval of 10 seconds was provided between each trial. A trial was discarded and participants were asked to repeat the trial if they lost their balance, if their hands came off of their hips at any point during the trial, or if they failed to land back onto the forceplates. A pilot study was conducted to examine the reliability of using 5 trials to represent characteristic movement patterns of an individual. Participants in this pilot study performed 10 double-leg drop jumps in a similar manner as in the current investigation. High Intra-Class Coefficient (ICC) values were obtained when ensemble averaged signals from trials 1-10 were compared with 1-5, indicating that 5 trials were sufficient to obtain representative data (See Appendix D). Further, high ICC values were also obtained between ensemble averages of trials 1-2 and 9-10 (See Appendix E). These values suggested that there would be no systematic shifts in the data due to fatigue or a learning effect.

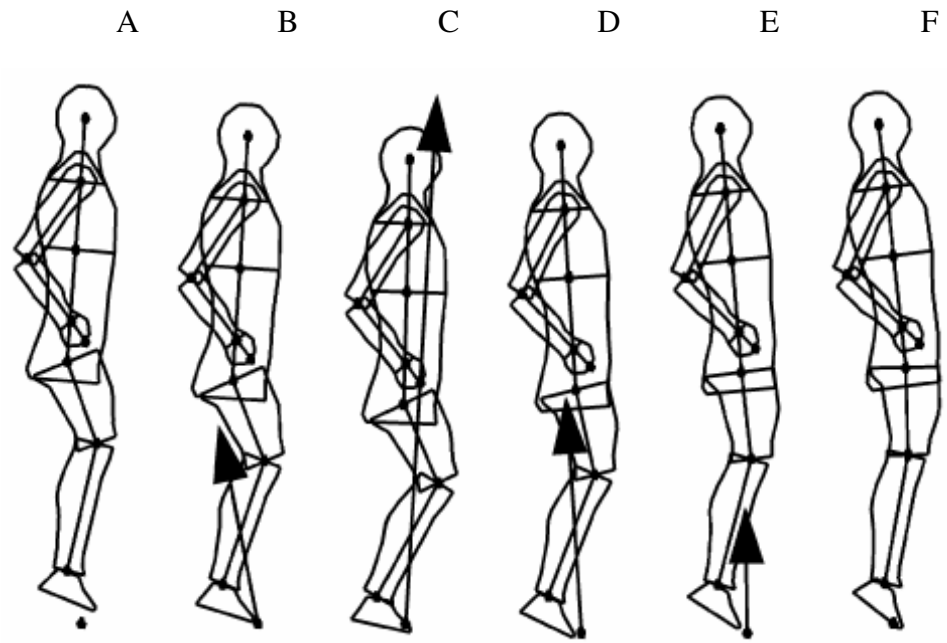
### *Post Testing*

All sensors and electrodes were carefully removed from the participants' body, and they were instructed to contact the investigator in case of any further questions or concerns. The participants were also informed that the results of the study would be made available to them if they so desired.

## Data Processing

For the MVIC data, the first and last second of each trial were discarded prior to analysis to assure steady state results. Then, the MVIC trials were digitally processed with a

Figure 6 : Graphical Representation of the Drop Jump Task



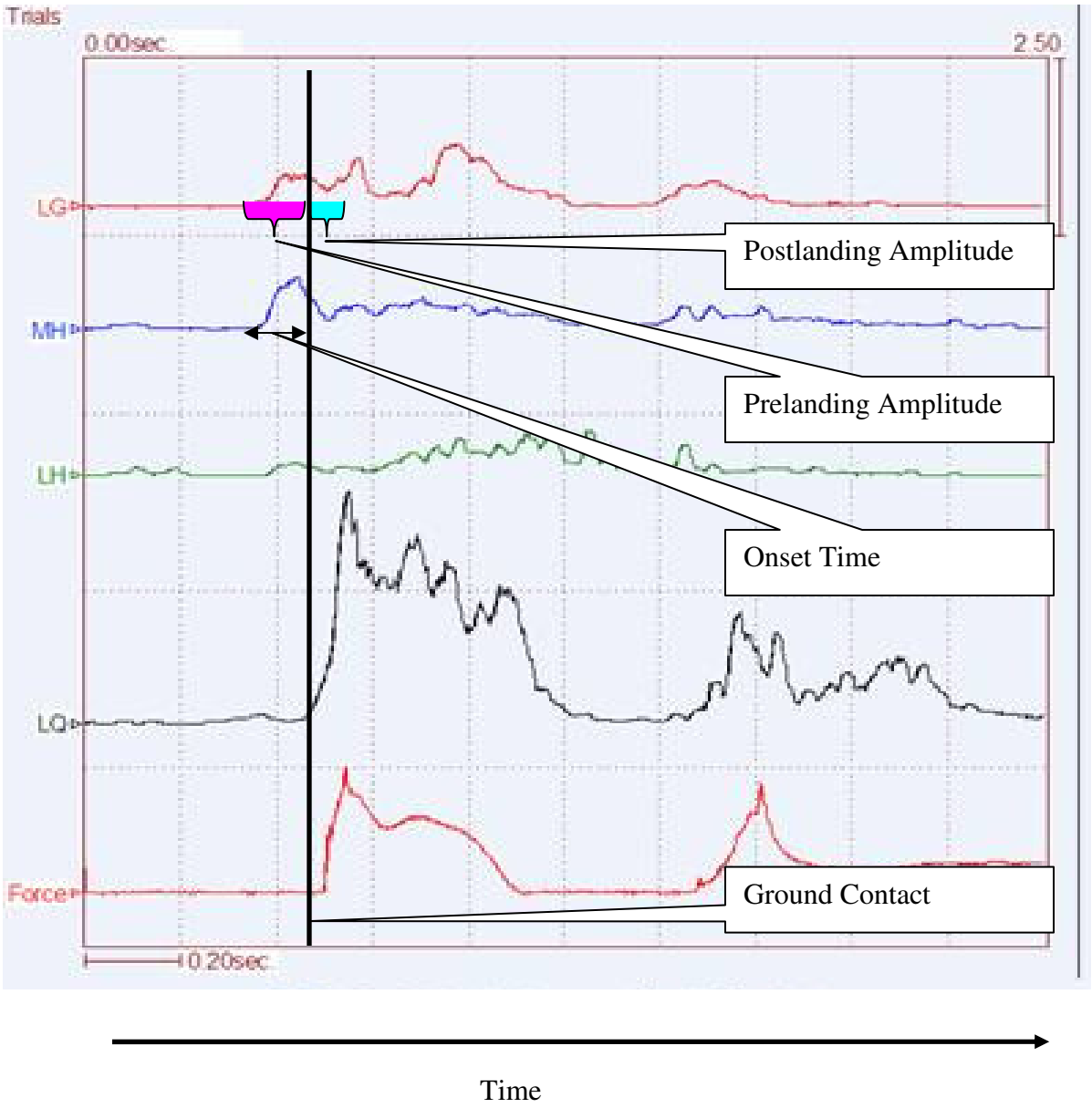

A: Flight
B: Ground Contact
C: Peak Knee Flexion (Lowest Point of Braking Phase before upward movement)
D: Beginning of Upward Movement
E: Leaving Ground for Vertical Jump
F: In air for the following Vertical Jump
B-to-C: Braking Phase
C-to-D: Propulsion Phase

Adapted from Arampatzis, A., Bruggemann, G.-P., & Klapsing, G. M. (2001). Leg stiffness and mechanical energetic processes during jumping on a sprung surface. Medicine and Science in Sports and Exercise, 33(6), 923-931

band pass filter from 10 Hz to 350 Hz, using a fourth-order, zero-lag Butterworth filter and a centered RMS algorithm with 100 ms time constant. The peak RMS amplitude (in V) obtained over three trials was then used to normalize the EMG data during the drop jumps.

A representative trial showing neuromuscular dependent variables during the drop jump is presented in Figure 7.The sEMG signals during the tasks were digitally processed with a band pass filter from 10 Hz to 350 Hz, using a fourth-order, zero-lag Butterworth filter and a centered RMS algorithm with a 25 ms time constant. The 5 trials were then ensemble averaged to obtain one representative trial for each participant. A means $\pm$ SD interval event buffer was then set to extract the onset times (in ms). Muscle activity onset times were defined as the time point when the muscle activity first exceeded five SDs above the baseline activity of the muscle for at least 25 ms or longer. Baseline data were collected for 2 seconds in quiet standing prior to task performance after digitization procedures were completed. In this quiet stance position, the participant stood in a neutral stance with feet pointing forward and shoulder width apart while keeping the arms relaxed. For the muscle activation amplitude data, a time interval buffer was set to extract the mean amplitudes, collected over a time period of 150 ms before ($_{PRE}$), and 50 ms after ($_{POST}$) ground contact in the initial landing of the drop jumps. Mean RMS amplitudes were normalized to each participant's peak RMS value obtained previously during the MVIC trials, and are reported as a percentage of the MVIC (%MVIC).

Figure 7: Representative Trial showing Neuromuscular Dependent Variables during the Drop Jump

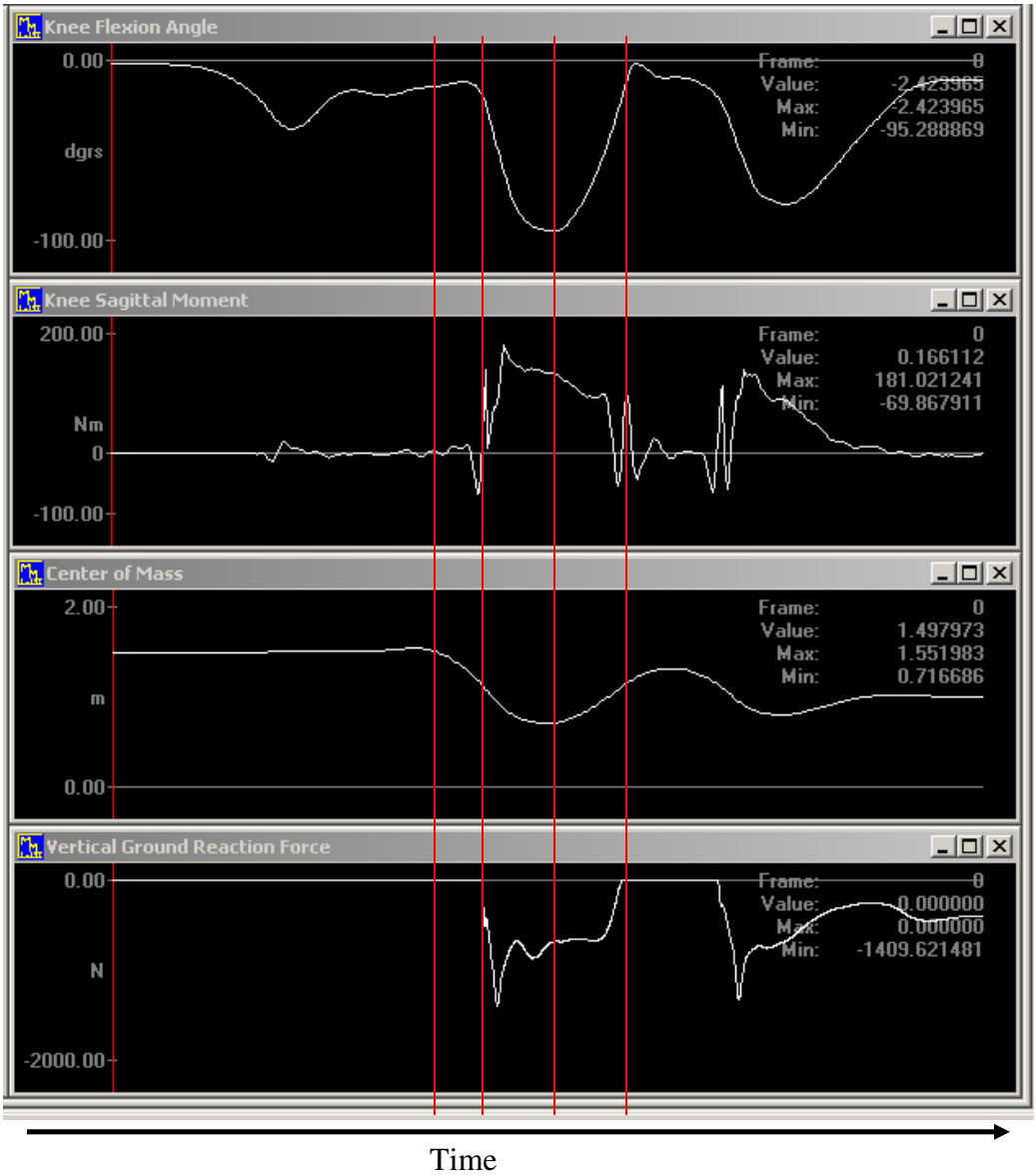


The law of constant acceleration guaranteed that the estimated time of free fall from the box (45 cm) was ~262.5 ms. This ensured that the participants were not in contact with the box 150 ms before initial contact (Hall, 2003). In drop jumps, the total ground contact time can be separated into a braking phase, where the body's momentum is reduced, and a propulsive phase where after reaching the lowest displacement of the body's center of mass, the direction of the momentum of the body is reversed upwards against gravity to push the body up into the air (Kellis et al., 2003; Viitasalo et al., 1998). Additionally a portion of the contact time is used for the coupling phase while moving from the braking to the propulsive phase (Bosco et al., 1982). Individuals have different contact times, with previous research suggesting that these contact times range from 165 ms to 226 ms with differing periods of the braking and propulsive phases (Arampatzis, Bruggemann et al., 2001; Horita et al., 2002; Walsh, Arampatzis, Schade, & Bruggemann, 2004). Thus, the first 50 ms after ground contact were examined in the current study, as it ensured that participants were still in the braking or load absorption phase during this time period.

A representative trial showing biomechanical dependent variables during the drop jump is presented in Figure 8. Force and position data used to calculate knee joint stiffness were filtered using a fourth-order, zero-lag low-pass Butterworth filter at 60 Hz and 12 Hz respectively. The net sagittal knee moment data at the point of ground contact and at the point of maximum knee flexion angle were extracted to record the change in knee joint moment ($\Delta M$ ; Nm). The knee moment data were then normalized to each

Figure 8: Representative Trial showing the Temporal Sequence of Biomechanical Events during the Drop Jump

A B C D




A: Take Off from Box (First Change in Center of Mass)
B: Initial Ground Contact (First Change in Vertical Ground Reaction Force)
C: Peak Knee Flexion (Lowest Point of Braking Phase before upward movement)
D: Leaving Ground for Vertical Jump (Absence of Vertical Ground Reaction Force)
B-to-C: Braking Phase
C-to-D: Propulsive Phase

participants' mass (Nm/kg). Similarly, the sagittal knee flexion angle were recorded both at initial ground contact and at maximum knee flexion angle and recorded as the change in knee joint angle ($\Delta\Theta$). The knee flexion angles are reported in degrees (°). Knee joint stiffness was then calculated by dividing the change in the net knee moment by the change in the knee flexion angle (Nm/kg°) (Farley & Morgenroth, 1999).

## Statistical Analyses

1. To test hypothesis 1, a 2 x 4 ANOVA with one between subjects factor (group: dance, basketball) and one within subjects factor (muscle: LQ, MH, LH, LG) compared groups on differences in the muscle activity onset times.
2. To test hypothesis,
    a. 2a: A 2 x 4 ANOVA with one between subjects factor (group: dance, basketball) and one within subjects factor (muscle activity prelanding: $LQ_{PRE}$, $MH_{PRE}$, $LH_{PRE}$, $LG_{PRE}$) compared groups on differences in mean muscle activation amplitudes prelanding.
    b. 2b: A 2 x 4 ANOVA with one between subjects factor (group: dance, basketball) and one within subjects factor (muscle activity postlanding: $LQ_{POST}$, $MH_{POST}$, $LH_{POST}$, $LG_{POST}$) compared groups on differences in mean muscle activation amplitudes postlanding.
3. To test hypothesis 3, a one-way ANOVA was used to determine group differences in knee joint stiffness.

4. To test hypothesis 4, a stepwise linear regression analysis was conducted using prelanding muscle activation amplitudes of all four muscles ($LQ_{PRE}$, $MH_{PRE}$, $LH_{PRE}$, $LG_{PRE}$), and group membership (dance or basketball) to predict knee joint stiffness.

An alpha level of 0.05 was set a priori for all tests. All analyses were conducted using the SPSS 14.0 version for Windows software (Statistical package for Social Sciences, Chicago, IL). If significant interactions were noted, simple main effects testing with Bonferroni's correction were used to determine where significant differences existed.

# CHAPTER IV

# RESULTS

## Descriptive Data

Of the 70 participants (35 per group) originally proposed, difficulties in recruitment resulted in 35 dancers (age = 20.7 ± 2.3 yrs, height = 164.3 ± 6.7 cm, weight = 62.2 ± 1.9 kg, experience = 13.9 ± 5.2 yrs, body fat = 21.4 ± 5.1 %) and 20 basketball players (age = 20.1 ± 2.0 yrs, height = 170.5 ± 6.1 cm, weight = 72.6 ± 11.4 kg, experience = 10.7 ± 3.5 yrs, body fat = 24.3 ± 5.0 %) successfully completing the study. Complete raw data for all participants including demographics, muscle onsets, muscle amplitudes, and kinetic and kinematic data can be found in Appendices F-I respectively. Given the unbalanced design, adjustments were computed automatically by the statistical software, maintaining the robustness of the model. In cases where Mauchly's assumptions for sphericity were not met for the ANOVA, the Huynh-Feldt correction was used for group comparisons. Further, because the full complement of subjects were not recruited as determined a-priori via power analyses, simple effect sizes between groups are also presented.

## Reliability of Neuromuscular and Biomechanical Parameters

Day-to-day reliability of the muscle activation and biomechanical variables were assessed on 10 randomly selected participants (5 dancers, age = 21.2 $\pm$ 3.1 yrs, height = 164.6 $\pm$ 4.7 cm, weight = 61.2 $\pm$ 8.4 kg, experience = 16.0 $\pm$ 5.0 yrs, body fat = 23.5 $\pm$ 5.2 %; 5 basketball players, age = 20.6 $\pm$ 3.0 yrs, height = 171.4 $\pm$ 6.0 cm, weight = 74.4 $\pm$ 6.0 kg, experience = 10.6 $\pm$ 3.0 yrs, body fat = 25.4 $\pm$ 5.4 %). Intraclass Correlation Coefficients ($ICC_{2, k}$), and Standard Error of the Measurement (SEM) for muscle onset times, muscle activation amplitudes and stiffness variables are presented in Table 1. The raw data and mean squares used to calculate these values are presented in Appendix J. With the exception of $LH_{POST}$ (.65), muscle activation amplitudes and biomechanical measures showed generally good day-day reliability (.71 – .86). Muscle onset timings tended to be less reliable (~.61 – .76).

## Hypothesis 1: Comparison of Groups based on Muscle Activity Onset Times

Means, standard deviations and effect sizes for muscle activity onset times by muscle and group are presented in Table 2. No differences were found in muscle activity onsets between groups ($F_{1, 53} = 1.56$, P =.22, ES = .03, 1-β =.23) or groups by muscle ($F_{2.8, 150.4} = .61$, P = .60, ES = .01, 1-β = .17). However, significant differences in muscle onset times revealed a consistent recruitment order ($F_{2.8, 150.4} = 25.19$, P < .001). Post-hoc pairwise comparisons with Bonferroni's correction revealed that the lateral gastrocnemius, lateral hamstring and medial hamstrings muscles activated at statistically similar times and were followed by the activation of the lateral quadriceps muscle (See

Table 1: Day-to-Day Reliability Estimates ($ICC_{2,k}$ and SEM) for Study Variables

| | $\bar{X}_{Day1}$ | $SD_{Day1}$ | $\bar{X}_{Day2}$ | $SD_{Day2}$ | N | $ICC_{2,k}$ | SEM |
|---|---|---|---|---|---|---|---|
| $LG_{ON\ (ms)}$ | 161.1 | 54.5 | 155.4 | 48.1 | 10 | .76 | 26.6 |
| $MH_{ON\ (ms)}$ | 131.8 | 44.6 | 151.7 | 52.2 | 10 | .62 | 32.3 |
| $LH_{ON\ (ms)}$ | 125.1 | 51.9 | 117.9 | 33.8 | 10 | .62 | 32.2 |
| $LQ_{ON\ (ms)}$ | 101.2 | 73.1 | 97.2 | 58.3 | 10 | .61 | 45.9 |
| $LG_{PRE\ (\%MVIC)}$ | 44.7 | 21.2 | 43.1 | 19.0 | 10 | .77 | 1.3 |
| $MH_{PRE\ (\%MVIC)}$ | 25.2 | 9.9 | 24.5 | 11.4 | 10 | .79 | 5.2 |
| $LH_{PRE\ (\%MVIC)}$ | 19.9 | 8.6 | 2.1 | 9.8 | 10 | .83 | 4.1 |
| $LQ_{PRE\ (\%MVIC)}$ | 16.4 | 9.0 | 2.9 | 12.0 | 10 | .81 | 5.2 |
| $LG_{POST\ (\%MVIC)}$ | 32.0 | 7.9 | 35.7 | 11.3 | 10 | .71 | 6.0 |
| $MH_{POST\ (\%MVIC)}$ | 19.3 | 9.6 | 2.2 | 1.8 | 10 | .86 | 4.1 |
| $LH_{POST\ (\%MVIC)}$ | 24.1 | 15.6 | 39.5 | 33.3 | 10 | .65 | 19.8 |
| $LQ_{POST\ (\%MVIC)}$ | 83.2 | 33.3 | 111.4 | 53.9 | 10 | .73 | 28.2 |
| $KFA_{CHANGE\ (°)}$ | 71.0 | 11.8 | 71.8 | 9.4 | 10 | .84 | 4.71 |
| $MOM_{CHANGE\ (Nm/kg)}$ | 1.08 | .62 | .93 | .54 | 10 | .84 | .25 |
| $STIFFNESS_{(Nm/kg.\ °)}$ | .02 | .01 | .01 | .01 | 10 | .83 | .00 |

LG = Lateral Gastrocnemius; MH = Medial Hamstrings; LH = Lateral Hamstring; LQ = Lateral Quadriceps; $_{ON}$ = Onset $_{PRE}$ = 150 ms before ground contact; $_{POST}$ = 50 ms after ground contact; $KFA_{CHANGE}$ = Change in Knee Flexion Angle from Ground Contact to Maximum Knee Flexion; $MOM_{CHANGE}$ = Change in Sagittal Knee Net Moment from Ground Contact to Maximum Knee Flexion; $\bar{X}$ = Means; SD = Standard Deviations; SEM Standard Error of the Measurement

Table 2: Muscle Activation Onset Times (ms) (Means ± SD)

| | Dance | Basketball | Totals | Between Group Effect Sizes |
|---|---|---|---|---|
| LG | 146.5 ± 52.1 | 140.9 ± 52.6 | 144.5 ± 51.9 * | .11 |
| MH | 158.2 ± 50.2 | 131.4 ± 44.2 | 148.4 ± 49.4 * | .53 |
| LH | 132.1 ± 41.5 | 127.4 ± 50.8 | 130.4 ± 44.7 * | .09 |
| LQ | 96.8 ± 69.0 | 87.0 ± 53.2 | 93.2 ± 63.3 | .14 |
| Totals | 133.4 ± 53.2 | 121.6 ± 50.2 | 129.1 ± 33.7 | |

LG = Lateral Gastrocnemius; MH = Medial Hamstrings; LH = Lateral Hamstring; LQ = Lateral Quadricep; Bonferroni's adjustment for multiple comparisons at P =.05
* Indicates significantly greater than LQ

Table 2). The ANOVA summary results are presented in Appendix K.

Hypothesis 2: Comparison of Groups based on Muscle Activation Amplitudes

Means, standard deviations, and effect sizes for muscle activation amplitudes during pre and post ground contact are presented in Tables 3 and 4 respectively. Muscle activation amplitudes did not differ between groups either pre ($F_{1, 53} = .28$, $P = .60$, ES = 01, $1-\beta = .08$) or post ($F_{1, 53} = .07$, $P = .78$, ES = .00, $1-\beta = .06$) landing. However, trends toward differences in muscles by group were present both pre ($F_{2.6, 130.2} = 2.04$, $P = .12$, ES = .02, $1-\beta = .47$), and post ($F_{2.3, 121.6} = 2.50$, $P = .08$, ES = .05, $1-\beta = .53$) landing. Moderate effect sizes (.30-.55) were noted for several muscle activation parameters, with higher muscle activation levels seen in dancers in $MH_{PRE}$ (34 vs. 26.3 % MVIC), $LG_{POST}$ (45.1 vs. 35.5 % MVIC), and $MH_{POST}$ (38.2 vs. 24.9 % MVIC). The exception to this was the $LQ_{POST}$, where higher muscle activation levels were noted in basketball players than dancers (108.5 vs. 89.6% MVIC) (See Tables 3, and 4).

When all participants were considered together, significant differences were found between individual muscle amplitudes both pre ($F_{2.5, 130.2} = 35.20$, $P < .001$) and post ($F_{2.3, 121.6} = 52.13$, $P < .001$) landing. Post-hoc Bonferroni's pairwise comparisons revealed that prelanding, the LG activated at higher levels than all other muscles, and the MH activated at higher levels than the LH and LQ. Postlanding, the LQ activated at higher levels than all other muscles (See Tables 2, 3, and 4). The ANOVA summary results for pre and postlanding muscle activity amplitudes are presented in Appendices L and M respectively.

Table 3: Prelanding Muscle Activation Amplitudes (%MVIC) (Means ± SD)

| | Dance | Basketball | Totals | Between Group Effect Sizes |
|---|---|---|---|---|
| LG | 39.8 ± 18.1 | 38.8 ± 19.8 | 39.4 ± 18.5 * † # | .05 |
| MH | 34.0 ± 14.2 | 26.3 ± 10.9 | 31.2 ± 13.5 † # | .55 |
| LH | 20.6 ± 7.3 | 21.7 ± 11.4 | 21.0 ± 8.9 | .10 |
| LQ | 18.1 ± 9.6 | 20.3 ± 12.7 | 18.9 ± 10.7 | .17 |
| Totals | 28.1 ± 8.7 | 27.7 ± 10.5 | 27.6 ± 9.3 | |

LG = Lateral Gastrocnemius; MH = Medial Hamstrings; LH = Lateral Hamstring; LQ = Lateral Quadricep; Bonferroni's adjustment for multiple comparisons at P =.05;
* Indicates significantly greater than MH, LH and LQ; † Indicates significantly greater than LH and LQ; # Indicates significantly greater than LQ

Table 4: Postlanding Muscle Activation Amplitudes (%MVIC) (Means ± SD)

| | Dance | Basketball | Totals | Between Group Effect Sizes |
|---|---|---|---|---|
| LG | 45.1 ± 29.2 | 35.5 ± 10.9 | 41.6 ± 24.5 | .33 |
| MH | 38.2 ± 32.7 | 24.9 ± 13.7 | 33.4 ± 28.0 | .41 |
| LH | 31.0 ± 25.4 | 29.3 ± 25.6 | 30.4 ± 25.3 | .07 |
| LQ | 89.6 ± 44.0 | 108.5 ± 63.7 | 96.5 ± 51.2 * | .30 |
| Totals | 51.0 ± 17.3 | 49.6 ± 21.4 | 50.5 ± 18.7 | |

LG = Lateral Gastrocnemius; MH = Medial Hamstrings; LH = Lateral Hamstring; LQ = Lateral Quadricep; Bonferroni's adjustment for multiple comparisons at P =.05;
* Indicates significantly greater than LG, MH and LH.

## Hypothesis 3: Comparison of Groups based on Knee Joint Stiffness

No differences were found in knee joint stiffness between dancers (.0163 $\pm$ .009 Nm/kg°) and basketball players (.0185 $\pm$ .012 Nm/kg°) ($F_{1, 53}$ = .61, P =.44, ES = .19, 1-$\beta$ =.12). However, the small effect size (ES = .19) noted indicate weak trends towards higher knee joint stiffness in basketball players. The ANOVA summary results are presented in Appendix N. On average, participants landed with knee flexion angles of 16.5 $\pm$ 5.7 ° at ground contact and reached peak knee flexion angles ($KFA_{PK}$) of 87.3 $\pm$ 11.5 ° about 235.4 $\pm$ 47.3 ms after contact ($KFA_{TPK}$) (indicating the deceleration or shock absorption phase). Average sagittal plane net knee moments were .02 $\pm$ .04 Nm/kg at initial ground contact increasing to .14 $\pm$ .05 Nm/kg at $KFA_{PK}$. The biomechanical variables of interest are presented in Table 5.

## Hypothesis 4: Prediction of Knee Joint Stiffness Based On Prelanding Muscle Activation Amplitudes and Group Membership

Bivariate correlations and stepwise linear regression summary results are listed in Tables 6, and 7 respectively. Weak, non-significant relationships were noted between $LG_{PRE}$, $LQ_{PRE}$ and knee joint stiffness. However, neither prelanding muscle activation amplitudes nor group membership, either alone or in combination were significant predictors of knee joint stiffness. The complete results of the linear regression model are presented in Appendix O.

Table 5: Biomechanical Variables of Interest (Means ± SD)

| Variable | Dance | Basketball | Totals |
|---|---|---|---|
| $KFA_{IN}$ (°) | 16.0 ± 5.7 | 17.3 ± 5.7 | 16.5 ± 5.7 |
| $KFA_{PK}$ (°) | 88.6 ± 11.2 | 84.3 ± 11.6 | 87.3 ± 11.5 |
| $KFA_{CH}$ (°) | 72.6 ± 11.9 | 67.0 ± 11.8 | 70.6 ± 12.0 |
| MOM @ $KFA_{IN}$ (Nm/kg) | .01 ± .04 | .02 ± .04 | .02 ± .04 |
| MOM @ $KFA_{PK}$ (Nm/kg) | .13 ± .05 | .14 ± .05 | .14 ± .05 |
| $MOM_{CH}$ (Nm/kg) | .11 ± .06 | .12 ± .07 | .11 ± .06 |
| $KFA_{TPK}$ (ms) | 233.3 ± 40.7 | 239.0 ± 58.0 | 235.4 ± 47.3 |
| Contact Time (ms) | 510.1 ± 82.6 | 500.4 ± 118.7 | 506.1 ± 96.3 |
| $GRF_{PK}$ (BW) | 2.1 ± .4 | 1.8 ± .3 | 2.0 ± .4 |
| One Leg $GRF_{TPK}$ (ms) | 52.7 ± 10.8 | 53.8 ± 10.6 | 53.1 ± 10.7 |

KFA = Knee Flexion Angle; $_{IN}$ = Initial Ground Contact; $_{PK}$ = Peak; $_{CH}$ = Change from Peak to Initial; MOM = Net Sagittal Knee Moment; $_{TPK}$ = Time to Peak; GRF = Ground Reaction Force; BW = Times Body Weight

Table 6: Correlations between Knee Joint Stiffness, Prelanding Muscle Amplitudes and Group Membership

| Correlations | | | | | | | |
|---|---|---|---|---|---|---|---|
| | | STIFF | GROUP | $LG_{PRE}$ | $MH_{PRE}$ | $LH_{PRE}$ | $LQ_{PRE}$ |
| Pearson Cor. | STIFF | 1 | .11 | .18 | .00 | .12 | .18 |
| | GROUP | | 1 | -.03 | -.28 | .06 | .09 |
| | $LG_{PRE}$ | | | 1 | .47 | .20 | .30 |
| | $MH_{PRE}$ | | | | 1 | .39 | .29 |
| | $LH_{PRE}$ | | | | | 1 | .35 |
| | $LQ_{PRE}$ | | | | | | 1 |
| Sig.(1-tail) | STIFF | | .22 | .10 | .49 | .19 | .10 |
| | GROUP | | | .42 | .02 | .33 | .25 |
| | $LG_{PRE}$ | | | | .00 | .08 | .01 |
| | $MH_{PRE}$ | | | | | .00 | .02 |
| | $LH_{PRE}$ | | | | | | .00 |
| | $LQ_{PRE}$ | | | | | | |

STIFF = Knee Joint Stiffness; LG = Lateral Gastrocnemius; MH = Medial Hamstrings; LH = Lateral Hamstring; LQ = Lateral Quadricep; $_{PRE}$ = Prelanding Muscle Activation Amplitudes

Table 7: Model Summary for the Prediction of Knee Joint Stiffness based on Prelanding Muscle Amplitudes and Group Membership

| Model | R | $R^2$ | Adj.$R^2$ | SEE | Change | | | | |
|---|---|---|---|---|---|---|---|---|---|
| | | | | | $R^2$ Ch. | F Ch. | df1 | df2 | Sig. F Ch. |
| 1.00 | .179 | .032 | .014 | .01 | .03 | 1.76 | 1 | 53.00 | .19 |
| 2.00 | .221 | .049 | .012 | .01 | .02 | .90 | 1 | 52.00 | .35 |
| 3.00 | .249 | .062 | .007 | .01 | .01 | .74 | 1 | 51.00 | .40 |

1.00 = Predictors: (Constant), $LG_{PRE}$
2.00 = Predictors: (Constant), $LG_{PRE}$, $LQ_{PRE}$
3.00 = Predictors: (Constant), $LG_{PRE}$, $LQ_{PRE}$, $MH_{PRE}$

LG = Lateral Gastrocnemius; MH = Medial Hamstrings; LQ = Lateral Quadricep; $_{PRE}$ = Prelanding Muscle Amplitudes

# CHAPTER V

# DISCUSSION

The main purpose of this dissertation was to compare knee muscle activation patterns and knee joint stiffness during the initial phase of a drop jump landing between two groups; one with reported high (female basketball players) and the other with reported low (female dancers) rates of ACL injury. Female dancers have been noted to have smaller H-reflexes than other athletes (Mynark & Koceja, 1997; Nielson et al., 1993), which are considered indicative of increased muscle cocontraction levels (Llewellyn et al., 1990; Nielson et al., 1993). Increased muscle cocontraction increases knee joint stiffness (Goldfuss et al., 1973; Louie & Mote, 1987; Markolf et al., 1978; Wojtys et al., 2002), thereby indicating a possible protective mechanism against ACL injury during activity (Wojtys et al., 2002). However, no prior published literature had directly compared either knee muscle activation or knee joint stiffness between dancers and basketball players. An additional purpose of this study was to examine the ability of prelanding knee muscle activation amplitudes and group membership to predict changes in knee joint stiffness.

The primary findings of the current study were that dancers and basketball players did not have different knee muscle onset times, muscle activation amplitudes, or knee

joint stiffness levels during the initial landing in a double-leg drop jump. Further, prelanding muscle activation amplitudes and group membership were not significant predictors of changes in knee joint stiffness. However, due to recruitment problems, data were collected from only 55 subjects (35 dancers and 20 basketball players) rather than the proposed sample size of 70 subjects (35 dancers and basketball players each). Because the full complement of participants proposed based on a-priori power analyses were not achieved, careful consideration will be given to the potential for low statistical power. The following discussion will address how the current findings reflect on existing theory and research relative to pre and postlanding knee muscle activation strategies and knee joint stiffness, and will consider alternative explanations for the lack of significant findings between groups.

## Muscle Activation Parameters

Dancers were hypothesized to have higher muscle activation amplitudes both pre and postlanding as compared to basketball players. This theory was proposed as dancers have been reported to have lower H-reflexes than other athletes (Mynark & Koceja, 1997; Nielson et al., 1993) and lower H-reflexes are indicative of increased muscle cocontraction (Koceja et al., 1993; Llewellyn et al., 1990). With muscle onset timings and amplitudes suggested to be controlled as a unit during landings (Santello, 2005), dancers were also expected to have earlier muscle onsets. However, no statistically significant group differences were found for either muscle onset times or amplitudes (pre or postlanding). After comparing muscle parameter values from the current study with those reported in previous literature, measurement issues, low statistical power and effect

sizes, and alternative theories were considered and will be discussed to explain in part, the lack of statistically significant findings.

*Comparison with Previous Literature*

Muscle activity onset times were consistent with those noted previously during drop jumps and drop landings (Arampatzis, Bruggemann et al., 2001; Liebermann & Goodman, In Press; McKinley & Pedotti, 1992; Viitasalo et al., 1998) (See Table 8). The overall temporal patterns of muscle onsets in the current study with a greater sample size (N = 55) are also comparable to previous research (Range of N's = 4-10; See Table 8). In agreement with previously reported results (Liebermann & Goodman, In Press; Liebermann & Hoffman, 2005) the gastrocnemius and hamstrings muscles activated at similar times (145 $\pm$ 52 ms, 148 $\pm$ 49 ms and 130 $\pm$ 45 ms respectively), and earlier than the quadriceps (93 $\pm$ 63 ms). This pattern of muscle activation was expected during landing as it prepares the body for impact absorption upon ground contact.

The impact forces encountered upon ground contact need to be absorbed across the three major lower extremity joints (hip, knee and the ankle) (Decker et al., 2003; Kulas, Schmitz, Shultz, Watson, & Perrin, 2006; Schot & Dufek, 1993; Zhang, Bates, & Dufek, 2000). To allow for the safe absorption of these impact forces, the knee joint would need to flex during the flight phase before the initial landing, via the preactivation of the knee flexor muscles (gastrocnemius and hamstring muscles). To avoid complete buckling of the knee joint upon ground contact however, preactivation of the knee extensor muscles (quadriceps muscles) would need to follow. This pattern of muscle

Table 8: Comparison of Knee Muscle Onset Times to Previous Research

| Study | N | Task | Gastrocnemius | Hamstrings | Quadriceps |
|---|---|---|---|---|---|
| (McKinley & Pedotti, 1992) | 4 | DJ 45cm | 175 ± 30 ms | 132 ± 41 ms | 97 ± 53 ms |
| (Viitasalo et al., 1998) | 7 | DJ 40cm | 145 ± 7 ms | 31 ± 4 ms | 43 ± 7 ms |
| (Arampatzis, Bruggemann et al., 2001) | 10 | DJ 40cm | ~ 90 ± 40 ms | ~ 115 ± 40 ms | ~ 60 ± 50 ms |
| (Liebermann & Goodman, In Press) | 6 | DL 45cm | 121 ± 28 ms | NA | 75 ± 29 ms |
| Current Study | 55 | DJ 45cm | LG 145 ± 52 ms | MH 148 ± 49 ms LH 130 ± 45 ms | LQ 93 ± 63 ms |

DJ = Drop Jump; DL = Drop Landing

preactivity was clearly noticeable in the current study. Accompanying this activation timing patterns were greater activation of the gastrocnemius muscle (~39%MVIC) compared to all other muscles (medial hamstrings ~ 31%MVIC, lateral hamstring ~21%MVIC, lateral quadriceps ~19%MVIC) prelanding, and greater activation of the quadriceps muscle (97% MVIC) compared to all other muscles (lateral gastrocnemius = 42%MVIC, medial hamstrings = 33%MVIC, lateral hamstrings = 30%MVIC) post landing. The high prelanding amplitudes noted in the gastrocnemius muscle are understandable, given its more distal anatomical location in the lower extremity and thus its activation in preparation for the impact forces encountered upon landing. Similarly, the rapid and considerable increase seen in the quadriceps muscle activation level post landing is also understandable, as it allows for the successful execution of the subsequent jump.

Comparisons of muscle activation amplitudes from the current study with those reported in previous literature are difficult due to differing normalization procedures and time periods of interest (Colby et al., 2000; Duncan & McDonagh, 2000; Garrison et al., 2005; Hortobagyi & DeVita, 2000; Kovacs et al., 1999; Santello et al., 2001; Urabe et al., 2005; Viitasalo et al., 1998; Zazulak et al., 2005) (See Table 9). However, overall patterns of increased muscle amplitudes across all muscles postlanding (50.5 $\pm$ 18.7 %MVIC) as compared to prelanding (27.6 $\pm$ 9.3 %MVIC) observed in the current study are similar to those noted in previous reports (McNair, Prapavessis, & Callender, 2000; Onate et al., 2005). Recording muscle activity 150 ms before ground contact was acceptable in the current study as it ensured that a majority of muscle activation before

Table 9: Comparison of Knee Muscle Amplitudes to Previous Research

| Study | N | Task | Type of sEMG | Normalization Procedure Used | Time Periods Used | Gastrocnemius (Gas) | Hamstrings (Hams) | Quadriceps (Quads) |
|---|---|---|---|---|---|---|---|---|
| (Viitasalo et al., 1998) * | 7 | DJ 40cm | Mean Rectified | Arbitrary units – estimated | Pre – 2 periods of 50 ms<br>Post – Braking period | Pre ~ 40<br>Post ~ 60 | Pre ~ 30<br>Post ~ 60 | Pre ~ 60<br>Post ~ 180 |
| (Kovacs et al., 1999) | 10 | DJ 40cm | Peak RMS | % of average in eccentric phase | Pre – 80 ms<br>Post – Braking phase (80-120 ms) | Pre – 47 ± 48<br>Post – 100 ± 42 | - | Pre – 27 ± 11<br>Post – 100 ± 27 |
| | 15 | Land 40 cm, then pivot to contralateral side | Mean Rectified, integrated; Quads $_{Max}$; Hams $_{Min}$ | % MVC over trial (dynamic peak) | 500 ms pre to when pivoting started (not defined) | - | 14 ± 15 | 126 ± 52 |
| (Duncan & McDonagh, 2000) | 10 | DL 45cm | Peak Rectified | 80 ms preceding contact % | Post –200ms | 186 ± 158 | 1063 ± 692 | 287 ± 282 |
| (Hortobagyi & DeVita, 2000) | 11 | Step Down 33cm | Peak RMS (%);(µV for Gas) | Max MVC Isokinetic Eccentric 100°/s | Pre – 200ms<br>Post – 100ms | Pre – 364 ± 170<br>Post – 534 ± 137 | Pre – 19 ± 13<br>Post – 20 ± 12 | Pre – 18 ± 13<br>Post –52 ± 30 |
| (Santello et al., 2001) * | 8 | DL 40cm | Mean Rectified | % MVIC | Pre –100ms<br>Post –100ms | - | Pre ~ 110<br>Post ~ 130 | Pre ~ 120<br>Post ~ 160 |

| | | | | | | | | |
|---|---|---|---|---|---|---|---|---|
| (Zazulak et al., 2005) † | 13 | Single leg Landings Average of 30 & 45cm | Peak RMS | % MVIC | Pre – 200ms Post –250ms | - | NA | Pre – 34 ± 19 Post – 66± 32 |
| (Urabe et al., 2005) † | 8 | Land from max vertical jump | Peak integrated over 5° increments | % MVC over trial (dynamic peak) | Post – from 15 to 55° of knee flexion | - | 45 ± 20 | 220 ± 75 |
| (Garrison et al., 2005) † | 8 | Single leg Land, L 60 cm | RMS | MVIC over static single leg stance | 80 ms (40ms pre + 40ms post ground contact) | - | 6.1 ± 3.5 | 9.7 ± 3.6 |
| Current Study | 55 | DJ 45cm | Mean RMS | % Peak MVIC | Pre –150ms Post –50ms | Pre – LG = 39 ± 18 Post – LG = 42 ± 25 | Pre – MH = 31 ± 14, LH = 21 ± 9 Post – MH = 33 ± 28, LH = 30 ± 25 | Pre – LQ = 19 ± 10 Post – LQ = 97 ± 51 |

DJ = Drop Jump, DL = Drop Landing, sEMG = Type of surface Electromyography recorded; RMS = Root Mean Square, MVC = Maximum Voluntary Contraction, MVIC = Maximum Isometric Voluntary Contraction, Pre = prelanding muscle amplitudes, Post = postlanding muscle amplitudes, LG = Lateral Gastrocnemius; MH = Medial Hamstrings; LH = Lateral Hamstring; LQ = Lateral Quadricep. * Estimated from graphs in article, actual values not reported. † Where reported values were separated by gender, values obtained for females are displayed

the landing was included in the data analyses. This is because on average, all muscles across all participants activated 129 ms before ground contact. Similarly, recording muscle activity

50 ms postlanding ensured that the participants were still in the braking or shock absorption phase, as the time required to reach peak knee flexion (and thus indicative of the braking phase) across all participants was 235 ms after initial ground contact.

Hence, the overall muscle activation onset and amplitude patterns noted in the current study appear to be in agreement with those noted by previous researchers in studies examining muscle parameters during landing. The time periods for extracting muscle activation amplitude data appear to be appropriate and reasonable both pre (given the onset times across all muscles), and postlanding (all in the shock absorption phase). With the muscle parameter data appearing to be valid, recording issues were probably not the reason for the lack of significant group differences in muscle parameters. The following sections will thus examine other possible reasons for this absence of significant group differences noted in the results.

### *Measurement Issues*

Although day-day reliability values for muscle activity onset times were only fair to good (ICC =.61-.76), they were considered acceptable since similar reliability values (ICC = .63-.81) have been previously reported for onset times during isometric knee extension (Wong & Ng, 2005), which is a much more restricted activity than was used in the current study. To reduce measurement error in the form of trial-to-trial variability,

sEMG signals from all five drop jump trials of each participant were ensemble averaged to obtain a single representative trial for amplitudes.

Further, to allow for a more precise extraction of muscle onset timings, a much higher and stringent criterion was used to define onset timings in the current study (5 standard deviations above baseline), than those used previously (2-3 standard deviations above baseline) (Gabel & Brand, 1994; McKinley & Pedotti, 1992; Mickelborough et al., 2004; Tang et al., 1998). However, inter-individual variability is inherent in landing tasks (James & Bates, 2003; James, Dufek, & Bates, 2000; Schot, Hart, & Mueller, 2002) and thus an important factor to consider when interpreting results. While the possibility of day-day variability in task performance was accounted for by retesting a subset of the participants a second time, whether participants' performance varied during different times of the day was not examined in the current study. Nonetheless it appears that while issues of measurement error might not be the only explanation, the inherent variability of sEMG data when considered along with low statistical power (see next section) may in part contribute to the lack of group differences noted in neuromuscular parameters in the current study.

### *Statistical Power and Effect Sizes*

In light of the variability in sEMG parameters, it is imperative that a sufficient number of participants be recruited to have a sufficiently powered study to identify clinically meaningful differences in neuromuscular strategies during landing. Considering the reduced final cohort size, moderate effect sizes (ES) were found between dancers and

basketball players for several muscle activation parameters including muscle onset times and pre and postlanding amplitudes (See Tables 3 and 4).

As compared to basketball players, dancers demonstrated trends towards earlier onsets ($MH_{ON}$: 158.2 ms vs. 131.4 ms; ES = .53), and higher amplitudes both pre ($MH_{PRE}$: 34 vs. 26 % MVIC; ES = .55) and post ($LG_{POST}$: 45 vs. 35 % MVIC; ES = .33; $MH_{POST}$: 38 vs. 25 % MVIC; ES = .41) landing, lending partial support to the proposed research hypotheses of earlier onsets and higher activation amplitudes in dancers. The exception to the increased muscle amplitudes noted in dancers was the $LQ_{POST}$ (90 vs. 109%MVIC; ES=.30) where dancers had a tendency for lower muscle activation levels than basketball players. The possible explanation for this finding may lie in the differences in the nature of the activities performed by the two groups.

Basketball is a jumping-intensive sport, where the result of the game may be directly affected by the ability of a player to jump higher than the opponent. Basketball players therefore regularly perform maximal jumps during both practice and play (using high quadriceps activation levels). On the other hand, due to the aesthetic component involved in dance, dancers infrequently perform repeated maximal jumps and thus do not often use maximal quadriceps activation levels. As a result, basketball players may be able to increase the activation levels of the quadriceps muscle more rapidly and to higher levels than dancers during activity. Support for this explanation can be found in previous literature.

Baratta et al. (1988) reported depressed hamstrings sEMG activity in athletes who participated in predominantly jumping sports but did not perform hamstrings exercises as

compared to athletes who regularly performed hamstring exercises. Further, in a study comparing isometric quadriceps muscle strength and sEMG activity between semi-professional female dancers and matched physically active controls with no dance training, Harley et al. (2002) found that that while dancers produced greater quadriceps muscle force output during maximal voluntary isometric contractions, their vertical jump was not higher than the controls. Dancers also generated similar quadriceps force outputs as the controls, but had lesser sEMG activity when performing several functional activities including squat jumps, counter-movement jumps, and drop jumps (Harley et al., 2002). The researchers suggested that this occurrence was a result of subconscious modulation of jump height by the dancers for aesthetics and possibly a result of long-term disciplined jump training (Harley et al., 2002). The researchers also speculated that dancers may be recruiting more hip and calf muscles during jumping (Harley et al., 2002).

Accordingly, the dancers in the current study may not be accustomed to maximally activating the quadriceps muscles repeatedly during their rehearsals and performance. On the other hand, the basketball players in the current study probably did activate their quadriceps muscles at high levels consistently during both practice and play, and hence were able to produce greater quadriceps muscle activation levels post contact than the dancers. As a result, although the dancers did increase the activation of the quadriceps muscles from pre to post contact (Pre-to-Post: 19-to-90%MVIC), the magnitude of this increase was lesser than that seen in the basketball players (Pre-to-Post: 20-to-109%MVIC). The trends towards statistically significant group-by-muscle

amplitude interactions both pre (P =.12), and post (P =.08) landing (See Appendices L and M for the full ANOVA summary results), indicative of differing activation levels of muscles from pre to post landings are likely a result of the patterns noted in the moderate effect sizes discussed above (i.e. higher quadriceps and lower hamstrings muscle activation levels in basketball players as compared to dancers).

Interestingly, these patterns of muscle activation differences between basketball players and dancers in the current study bear resemblance to gender differences noted by previous researchers examining lower extremity neuromechanics during functional activity (Malinzak et al., 2001; Sander et al., 2004; Zazulak et al., 2005). These gender differences have included higher levels of quadriceps and lower levels of hamstrings muscle activity in females as compared to males during functional activity. Similarly, basketball players in the current study also appeared to show trends towards higher levels of quadriceps and lower levels of hamstrings muscle activity when compared to the dancers. These trends lend partial support to the originally proposed hypothesis and supposition of possible ACL protection in dancers during activity. With a larger sample size and adequate power therefore, the hypothesis may have been upheld.

Support of this hypothesis would lend credence to the theory that differences in neuromuscular patterns may partially explain higher ACL injury rates seen in females than males. In other words, irrespective of gender, an individual with a neuromuscular strategy dangerous to the ACL (i.e. higher quadriceps and lower hamstrings muscle activation amplitudes) when performing a task would be at a greater risk for ACL injury than another individual with a more balanced neuromuscular strategy (lesser mismatch

between quadriceps and hamstrings muscle activation amplitudes) performing the same task. These neuromuscular differences would also vary based on the actual demands of the task to be performed by the individual.

*Task Demands*

Accordingly, understanding the demands of the task performed in the current study might offer an alternative explanation for the lack of group differences in muscle activation parameters. This explanation is based on the premise that muscle activation during landing is primarily modulated to drop height and stiffness of the landing surface (McKinley & Pedotti, 1992). In the current study, both groups dropped off from the same height (45 cm). According to the laws of constant acceleration, each participant would therefore have equal flight times and have contacted the ground at similar times. Also, both groups landed on the same surface (wooden forceplate) during the drop jumps. As these two factors were the same for both groups, they probably affected task performance equally in both groups. Muscle activity during landings has been suggested to be modulated similarly across different individuals provided they have prior knowledge of the surroundings and sufficient practice on the task (Liebermann & Goodman, In Press). In the current study, all participants were able to observe how high they were prior to take off, and had equal familiarization practice trials (D = 3.9; B = 3.9, P = .88, ES = .00, 1-$\beta$ = .05).

Further, previous research suggests that landing performance can be augmented using specific instructions (Arampatzis, Bruggemann et al., 2001; McNair et al., 2000;

Onate et al., 2005). Both groups in the current study were given the same standardized set of instructions when performing the drop jumps. Both groups were also highly trained in their respective activities (10-14 years of experience), and were familiar with receiving and following instructions as part of their normal practice routines. Therefore both groups were probably very receptive to the instructions given and responded by modulating neuromuscular patterns in similar ways.

From the above discussion it is probable that these similar task demands may have suppressed group differences in task performance to some extent. Still, the drop jump task was consciously chosen in order to specifically examine whether differences would exist in neuromuscular parameters between the groups in the presence of similar task demands that are commonly performed by both groups (drop jump). If significant group differences would have been noted in this task despite all these similarities, it would have provided strong evidence in support of possible ACL-injury risk reduction in dancers via protective neuromuscular mechanisms. In retrospect however, it appears that the task demands may have been too similar, suppressing possible subtle group differences to a degree of non-significance. Accordingly, future researchers should consider limiting prior knowledge of the task and have participants perform novel tasks when examining neuromuscular responses between basketball players and dancers during activity.

In summary, low statistical power and similar task demands partially explain the minimal group differences observed in muscle onset times and activation amplitudes in knee musculature in the current study. Despite attempting to control for sEMG

measurement error and variability via more stringent criterion and ensemble averaging sEMG signals, it appears that when individuals have prior experience, sufficient practice, and receive the same instructions when performing a drop jump task, their knee muscle activation parameters would be similar. The height of the drop off and the landing surface also seem to be among the more important factors determining modulation of knee musculature during a known task (e.g. landing).

Still, the presence of trends towards increased muscle activation levels in ACL-protective muscles (i.e. hamstrings) and decreased muscle activation levels in ACL-antagonist muscles (i.e. quadriceps) in dancers as compared to basketball players is encouraging and partially supports the proposed research theory of ACL-protective neuromuscular strategies employed by dancers during functional activity. To summarize, although statistical between-group differences were not found in the current cohort, the presence of moderate effect sizes noted in several variables advocates for the need for additional work to examine possible neuromuscular differences between dancers and basketball players during a variety of functional activities.

## Knee Joint Stiffness

Dancers were hypothesized to have higher knee joint stiffness than basketball players. This theory was proposed as dancers were expected to have increased muscle activity via decreased H-reflexes, and increased muscle activity has been found to increase joint stiffness (Goldfuss et al., 1973; Louie & Mote, 1987; Markolf et al., 1978; Wojtys et al., 2002). However, knee joint stiffness did not statistically differ between the

groups. The following sections will compare the biomechanical data values from the current study with previously published data. Thereafter, possible explanations for lack of differences in knee joint stiffness will be discussed. These explanations will include an analysis of the measurement of knee joint stiffness as defined in this study, followed by a discussion of the effect sizes for group differences in energy absorption strategies across all lower extremity joints.

*Comparison with Previous Literature*

The measure of knee joint stiffness as defined in the current study was chosen as a biomechanical parameter to describe the combined kinetics (sagittal knee moment) and kinematics (sagittal knee flexion excursion) through the complete shock absorption phase of a landing (Δ moment/ Δ angle). Knee flexion angle and moment values noted in the current study are similar to those observed in previous landing studies (Decker et al., 2003; Zhang et al., 2000). Overall, knee flexion excursion values noted in the current study during the 45 cm landing (70.4 °) are in close agreement to those reported by previous researchers in landings from 32 cm, 40 cm, and 60 cm (52.2 °, 70.5°, and 75.8 °) respectively (Decker et al., 2003; Kovacs et al., 1999; Zhang et al., 2000). Similarly, overall peak net knee moment values observed in the current study during the 45 cm landing (~ 1.8 Nm/kg) are in agreement with previously values reported during landings from 32 cm, 40 cm, and 60 cm (~ 1.7-2.0 Nm/kg, ~1.8 Nm/kg, and ~ 2.6 Nm/kg) respectively (Decker et al., 2003; Kovacs et al., 1999; Zhang et al., 2000). Thus, the biomechanical data appear to be valid, and data errors can be excluded as a possible

cause of the lack of significant group differences noted in knee joint stiffness. The explanations for this absence of group differences in knee joint stiffness could thus lie in the actual calculation of knee joint stiffness values, based on the definition chosen in the current study.

*Calculation of Knee Joint Stiffness Values*

Post hoc analyses of knee flexion angle values indicated that basketball players showed trends towards lower sagittal knee excursions than dancers (B: 67.0 °, D= 72.6 °, p =.09, ES = 0.47, 1-β = .38). This decreased knee excursion (the denominator term of the knee joint stiffness measure) would mathematically increase knee joint stiffness values in basketball players. Knee moments were defined in the current study as the net resultant moments of all structures (muscles, ligaments, joint reaction forces) acting across the knee joint. The mathematical convention used was to give positive values to extensor moments (e.g. those produced by the quadriceps) and negative values to flexor moments (e.g. those produced by the hamstrings and gastrocnemius) at the knee joint. Thus, an increased knee extensor moment (quadriceps) would mean a net increase in recorded knee moment, but an increased knee flexor moment (hamstrings, gastrocnemius) would mean a net decrease in the recorded knee moment.

Post-hoc effect size calculations revealed that basketball players had trends towards higher quadriceps muscle activation levels (109 vs. 90%MVIC; ES=.30) and trends towards lower medial hamstring (25 vs.38 %MVIC; ES=.41) and lateral gastrocnemius (35 vs. 45%MVIC; ES=.33) muscle activation post landing compared to

dancers. As the quadriceps muscle produces an extensor moment, it would result in an increase in the numerator term (knee joint moment) of the knee joint stiffness measure, mathematically increasing knee joint stiffness values in basketball players. Similarly, the trend toward lower activation levels in the knee flexor muscles in basketball players would produce lower knee flexion moments, and therefore in spite of these muscle activation levels being lower in basketball players, the mathematical result would still be an increase in the recorded net knee moment in basketball players compared to dancers. This increase in the numerator term value (knee joint moment) of the knee joint stiffness measure would then result in higher knee joint stiffness values in basketball players compared to dancers, despite dancers having higher hamstrings and gastrocnemius muscle activation levels. Thus, the effect of muscle activation amplitude levels on knee joint stiffness (as defined in the current study) would depend on their contribution as a knee flexor or extender.

To overcome this inconsistency, the originally proposed hypothesis of greater knee joint stiffness in dancers (secondary to overall higher knee muscle activation levels) needs to be redefined and be muscle-specific. In other words, this revised hypothesis would propose that dancers will have lower quadriceps and higher hamstrings muscle activation amplitudes (both ACL protective patterns), therefore resulting in lower knee joint stiffness (as currently defined) than basketball players during landing. This modified hypothesis (in contrast of the originally proposed hypothesis) would also be consistent with previous observations examining gender differences in knee biomechanics during landing which indicate lower knee extensor moments in males than females (Chappell et

al., 2002; Salci et al., 2004). Clinically therefore, and similar to muscle parameter findings, dancers in the current study appear to exhibit biomechanical patterns noted previously in males, and basketball players show patterns like those seen in females in studies examining gender differences during functional activity. Partial support for this explanation is found in post-hoc analyses of the knee joint stiffness data which revealed weak trends (ES = .19) towards higher knee joint stiffness in basketball players (.0185 $\pm$ .012 Nm/kg°) than dancers (.0163 $\pm$ .009 Nm/kg°).

Although this definition of knee stiffness was consciously chosen as it was specific to the knee joint, due to this very factor it did not take into account joint stiffnesses across other lower extremity joints. As the lower extremity is a multi-joint linked system, differences in biomechanical patterns at one joint can have an effect across other joints (McNitt-Gray et al., 2001). By its definition, knee joint stiffness can not explain differences occurring at other joints. The appreciation of energy absorption strategies across the lower extremity joints during the landing may thus offer clues as to the absence of statistical group differences noted in knee joint stiffness.

*Joint Energy Absorption Strategies*

Knee joint stiffness in this study was measured over the complete deceleration phase of the initial landing of the drop jump. The stiffness of the whole leg during this damping or deceleration phase has been referred to as leg impedance (Kulas et al., 2006). Kulas et al (2006) investigated the relationship between leg impedance and energy absorption at the ankle, knee, and hip during the early (impact) and late (stabilization)

phases of double leg landings in highly trained female dancers across preferred, stiff, and soft landing conditions. The researchers found that ankle and knee energy absorption during the impact phase (100 ms after ground contact), and knee and hip energy absorption during the stabilization phase (100 ms to the body's maximal center of mass displacement) of landing explained 75.5% of the variance in leg impedance (Kulas et al., 2006). Further, knee energy absorption during the stabilization phase independently accounted for 55% of the variance in leg impedance. In the current study, the total deceleration phase across which knee joint stiffness was recorded lasted for 235 ms after ground contact, and thus encompasses both the impact and stabilization phases as defined by Kulas et al. (2006), where the total landing impulse time across all conditions was 217.5 ms. The results noted by Kulas et al. (2006) also demonstrate the close relationship between leg stiffness (as leg impedance is the damping portion of this variable) and joint energy absorption. Since leg stiffness is a combination of joint stiffness across the lower extremity joints (Farley & Morgenroth, 1999; Hortobagyi & DeVita, 2000; McMahon & Cheng, 1990), it can be reasonably assumed that joint stiffness and energy absorption patterns would also be closely related to each other.

While the knee joint is the primary contributor in controlling the descent during the shock absorption phase of landing, the ankle and hip joints also play important roles in shock absorption (Decker et al., 2003; Kulas et al., 2006; Schot, Dufek, & Bates, 1991; Zhang et al., 2000). In females, the ratio of energy absorption across the lower extremity joints (ankle, knee, and hip) have been found to average around 39%, 43% and 18% respectively (Decker et al., 2003; DeVita & Skelly, 1992; Schot et al., 1991). In the

current study, although the measure of knee joint stiffness could provide an understanding of possible group differences in impact force absorption at the knee, it could not provide information regarding possible group differences across the other joints (hip, and ankle).

Farley and Morgenroth (1999) have reported that during hopping, changes in leg stiffness values are primarily modulated at the ankle joint to accommodate for changes in surface and height, highlighting the importance of the ankle joint in differential absorption of impact forces on ground contact during activity. In the current study, secondary to repeated practice of dance moves like being en pointe (standing on the toes), dancers may have the ability to produce higher activation levels in the gastrocnemius muscles (across the ankle joint) than the basketball players. This position requires the dancer to sustain muscle contraction of the gastrocnemius and keep the ankle in plantar flexion, while balancing the body over the toes. Ankle joint energy absorption patterns may therefore be different between groups, with dancers having more ankle joint stiffness than basketball players, a variable that was not measured in the current study. In support of this explanation, Harley et al. (2002) also suggested that dancers may be recruiting more hip and calf muscles during jumping, possibly indicating that dancers may have energy absorption patterns different from other athletes. The potential for higher ankle muscle activation in dancers is partially supported in the current study via trends towards higher gastrocnemius muscle activation in dancers compared to basketball players ($LG_{POST}$: D = 45%MVIC vs. B = 35%MVIC; ES=.33).

Additional support to the possibility of differing joint energy absorption patterns across the ankle and/or hip joint is provided in secondary kinetic analyses, where dancers had significantly greater peak ground reaction forces (GRFs) (4.26 $\pm$ .9 BW) than basketball players (3.68 $\pm$ .6 BW) ($F_{1, 53} = 6.49$, P =.01). Since no observable group differences were found in measured knee joint mechanics, it can be reasonably assumed that dancers and basketball players dissipated these higher GRFs differently over other lower extremity joints (i.e. ankle and/or hip). To allow for these differing energy absorption patterns, dancers and basketball players may have also used different muscle activation patterns across these other joints. Thus, while knee muscle activation (quadriceps, hamstrings and gastrocnemius) was not appreciably different between the groups, differences may have existed in the ankle and/or hip muscle activation patterns. Accordingly, future researchers examining neuromechanical differences between dancers and basketball players should include analyses of joint energy absorption patterns and muscle activation patterns across all lower extremity joints.

Overall, it appears that knee joint stiffness as defined in the current study was not different across the groups. In fact, in opposition to the proposed hypothesis of greater knee joint stiffness in dancers, weak trends indicated higher levels of knee joint stiffness in basketball players secondary to lower knee joint sagittal excursions, and higher quadriceps and lower hamstrings muscle activation levels. Perhaps then, the use of the knee joint stiffness as defined and employed in the current study is inadequate. Future researchers should explore alternative means of measuring biomechanical parameters when comparing neuromechanical strategies between dancers and basketball players

during functional activity (see future recommendations). Additionally, taking into account the higher GRFs and the potential that dancers may rely more heavily on the ankle joint musculature poses encouraging possibilities of differing energy absorption and joint stiffness patterns across other lower extremity joints between groups. Finally, understanding the compensations over the total lower extremity, and not just across the knee joint, may provide a better understanding as to how these differences may in some measure relate to possible ACL-injury protection in dancers during activity.

## Prediction of Knee Joint Stiffness

For a comprehensive analysis of the body's responses to potentially high ACL injury risk activities (e.g. landing), it is essential to understand the sequence of events by linking neuromuscular function (i.e. muscle activation) to resulting biomechanical changes (i.e. joint stiffness). This analysis is particularly important during the shock absorption phase of a landing, as in this phase the body absorbs forces imposed on it following ground contact. In the current study, increased prelanding muscle amplitudes and dance group membership were hypothesized to predict increased knee joint stiffness. However, neither amplitudes nor group membership were able to predict changes in knee joint stiffness. The following sections will attempt to explain why the relationships between muscle activation, group membership, and knee joint stiffness were not supported by examining the possible effects of muscle cocontraction levels, and the time periods of muscle activity used to predict changes in knee joint stiffness levels.

In agreement with current findings, Da Fonseca et al. (2005) found no correlations between quadriceps: hamstrings muscle cocontraction levels and knee flexor/extensor moment ratios (a component of knee joint stiffness as defined in the current study) during landings in females. However, other previous researchers have found differing results. Although Arampatzis et al. (2001) did not find any relationships between knee muscle activation and knee joint stiffness in drop jumps from 40 cm, they did find significant positive correlations between integrated prelanding quadriceps and hamstrings amplitudes and leg stiffness. The researchers calculated knee joint stiffness by modeling the knee as a rotational linear spring and leg stiffness by modeling the lower leg as a linear spring, and used linear regression equations to obtain their results (Arampatzis, Bruggemann et al., 2001). Further, the relationships were examined using integrated peak muscle amplitudes over the whole combined flight and ground support phase, and the muscle activity data were normalized to the amplitude during the trial in which the participant demonstrated the highest leg stiffness, possibly biasing the data towards obtaining the noted relationships.

Hortobagyi and DeVita (2000) also reported that un-normalized vastus lateralis and lateral gastrocnemius peak amplitudes (in μV) recorded 200 ms prior to ground contact explained 60 % and 45% of the variance respectively in the un-normalized leg stiffness (kN/m) when comparing muscle activity between older and younger women in a step down task. Further, hamstring preactivity was reported to be positively correlated with leg stiffness (Hortobagyi & DeVita, 2000). However, the definitions of leg stiffness were again different than those in the study by Arampatzis et al. (2001), with Hortobagyi

et al. (2000) defining leg stiffness as the ratio of the maximum ground reaction force to the leg displacement. Leg displacement was computed as the shortening of the limb during the period from ground contact to occurrence of the maximum ground reaction force (Hortobagyi & DeVita, 2000). Due to the differing definitions of the variables examined extrapolations of these previous results to that of the current study are difficult.

### *Muscle Cocontraction Levels*

In the current study there was an attempt to control for inter-participant variability by using muscle amplitudes normalized to participants' own MVICs, and using normalized knee joint stiffness values. Little information is present in the current literature however, examining the relationship between individual joint stiffness and muscle cocontraction levels. While it is understandable that overall muscle cocontraction levels might explain variations in leg stiffness, measures of cocontraction are relative measures typically based on agonist-antagonist ratios (Kellis et al., 2003). Therefore, they cannot provide information regarding how activation of an individual muscle might affect joint or total leg stiffness levels. Further, different researchers have used differing definitions of muscle cocontraction (Da Fonseca et al., 2004; Da Fonseca et al., 2005; Hirokawa et al., 1991; Kellis et al., 2003). In an attempt to partially account for this drawback, absolute MVICs for each muscle were entered individually into the stepwise linear regression model in the current study in order to account for both magnitude and relative differences between muscles. Using this procedure allowed for examination of the effect of each individual muscle as well as its interaction with other muscles and

group membership on knee joint stiffness. Despite these methodological adjustments, prelanding amplitudes or group membership still did not predict knee joint stiffness.

### *Appropriate Time Periods*

Another possible explanation for the lack of relationships found might be the time period over which the muscle activity used to predict knee joint stiffness was recorded. The prelanding amplitudes were used as the muscle activity to predict knee joint stiffness, as it was assumed that this muscle activity would best determine the state of the knee joint upon ground contact. However, while muscle activity prior to ground contact prepares the body for landing, changes in muscle activity upon ground contact are also important in protecting the body from the rapid and forceful joint decelerations upon ground contact and throughout the shock absorption phase (Santello, 2005). Thus, muscle activity in the postlanding phase would also contribute to the production and regulation of knee joint stiffness. This shock absorption period lasted ~235 ms in the current study. Therefore, using only prelanding muscle activation amplitudes might not have been adequate to accurately predict changes in knee joint stiffness levels.

Instead, using combined time periods that include portions of both pre and postlanding muscle activation amplitudes may better explain how knee joint stiffness is modulated from initial ground contact through the shock absorption phase. This method of examining time periods around significant events has been used previously by Garrison et al. (2005) who used time periods around ground contact (40 ms pre and 40 ms post) and around peak knee internal rotation moment (20 ms pre and 20 ms post) to

compare average muscle activity between genders in lower extremity musculature. Further, as ligament injuries have been suggested to occur almost immediately after joint loading (within 73ms) (Pope et al., 1979), examining muscle activity during this time period is imperative, as it would provide information regarding the state of the joint at this time.

With regards to group membership, from these results it does not appear that membership in a particular group (dance or basketball) predicts changes in knee joint stiffness levels. As groups did not differ in knee joint stiffness, it is understandable that membership in a specific group would not predict changes in knee joint stiffness. However, as mentioned earlier it is feasible that groups may differ across other lower extremity joints, and therefore these relationships should be further explored while also accounting for the ankle and/or hip joints.

Overall, it appears that despite using more controlled definitions of knee joint stiffness, normalized individual knee muscle amplitudes, and a more inclusive prediction model, knee joint stiffness could not be predicted by prelanding muscle activation amplitudes and group membership in the current study. Alternative strategies to predict knee joint stiffness could include alternative definitions of knee joint stiffness and possibly using other time periods (part of post landing amplitudes) to examine changes in knee joint stiffness.

## Limitations

Despite the previously mentioned methodological adjustments employed in the current study, several limitations were still present. A noteworthy limitation of the current study is the reduced final sample size from that originally proposed, which may have led to low statistical power and the rejection of the research hypotheses. This possible influence was evident upon calculation of effect sizes, where moderate effect sizes were noted in several of the parameters in directions that support the research hypotheses.

Another limitation was the lack of footwear use when performing the drop jump task in the study. Basketball players always use footwear when participating in sport, whereas dancers may or may not use footwear depending on the type of dance form. Hence, landing without footwear in the current study might possibly have changed the way basketball players and some dancers landed and absorbed forces imposed upon ground contact. Although there was an attempt to control this factor across all participants by having them perform the task barefoot, how this actually affected functional landing performance in both groups could not be evaluated. While it is necessary to control variables during a laboratory study, the more controlled a study is, the less it mimics functional and real-life settings. Thus, while dancers and basketball players did not differ appreciably in neuromuscular activation patterns in this controlled laboratory setting, it also remains unclear whether they would respond differently in their own regular activity settings.

Further, muscle activation and biomechanics of the hip and ankle, which might influence lower extremity neuromechanics as described above were not examined in the

current study. Simultaneous collection of kinematic, muscle activity and kinetic data of the complete lower extremity needs to be performed as it can allow for comprehensive analyses of landing strategies. Previous research has examined these collective factors for stop-jump (Chappell et al., 2002), and cutting maneuvers (Malinzak et al., 2001), and future researchers need to do the same for landing activities.

## Future Recommendations

Although the proposed research hypotheses were not accepted, several promising directions for future research were obtained from these results that would allow for stronger research designs. These research designs include the examination of additional variables to determine possible neuromechanical differences between dancers and basketball players and verify whether dancers employ ACL protective strategies during activity. The following sections will expand on these areas and offer specific recommendations for future researchers.

### *Neuromuscular Parameters*

Whereas it appears that knee muscle activation is similarly modulated across female dancers and basketball players in landing, whether group differences exist in plant-and-cut type activities (the other major activity type associated with ACL injury) is still unknown and needs to be examined by future researchers. Additionally although the knee musculature did not statistically differ in this familiar task, the presence of moderate effect sizes noted suggests the need for further investigations into whether these potential

differences become more pronounced in other potentially high ACL injury risk activities that are more novel for both dancers and basketball players (e.g. reactive multi-directional cut-and-plant movements). In these novel and unfamiliar tasks, muscle cocontraction levels may be higher as the body would attempt to protect itself from possible injury (Humphrey & Reed, 1983; Smith, 1981). As a result, any minimal differences in neuromechanical parameters existing between the groups in a familiar task would become exacerbated in this unfamiliar task. Increased contraction levels in both groups would likely make it easier to identify possible group differences. Future study designs should also include examination of group differences in reactive tasks. Such investigations will allow researchers to answer the question whether the ACL injury risk differential between dancers an basketball players exists merely due to differences in the nature of activity performed by the two groups, i.e. planned (in dancers) versus unplanned (in basketball) or is due to actual differences in neuromechanical strategies between the groups.

Although moderate effect sizes were noted in several muscle parameters, the actual clinical significance of these differences it is still unknown. It is understandable that earlier activation of the knee musculature would allow the body to be better prepared to protect the knee joint (and consequently the ACL) during activity. Similarly, balanced cocontraction of the knee musculature via decreased quadriceps and increased hamstring muscle activation would be protective of the ACL during activity. However, the actual onset times and activation levels of various muscles that separate neuromechanical strategies from being potentially harmful to being safe are still unclear. For example,

whether the 27 ms earlier onset times noted in the medial hamstrings in dancers as compared to basketball players is actually protective to the ACL is unknown. Similarly, it is not clear if the 20% MVIC increase in lateral quadriceps activation amplitudes post landing in basketball players is clinically problematic for ACL injury. Future researchers should investigate the actual effects of these differences by means of cadaver or biomechanical modeling studies. These research designs could be used to activate the muscles that cross the knee joint and varying times and amplitudes, mimicking the action of the knee musculature during sport activity and examine their effects on knee joint stability and ACL strain.

*Biomechanical Parameters*

The use of only two discrete points to define knee joint stiffness in the current study may miss important information occurring across the landing phase. An alternative method would be to examine the complete time continuum across the shock absorption phase and compare instantaneous knee joint stiffness between groups. Further, the weak trends towards higher knee joint stiffness in basketball players than dancers indicate the need for continued examinations of knee joint stiffness with appropriate sample sizes to confirm these trends. These investigations also need to examine the lower extremity as a whole by comparing overall leg stiffness between the groups. The possibility of differing energy absorption strategies, as evidenced by differing ground reaction forces between the two groups is also very likely and needs to be examined further. While some researchers have examined energy absorption patterns during landing separately in

intercollegiate females (Decker et al., 2003; DeVita & Skelly, 1992), and in dancers (Kulas et al., 2006), additional work comparing energy absorption strategies between dancers and basketball players, and other high ACL–risk athletes (e.g. soccer players) is necessary and warranted in the future.

### *Prediction of Knee Joint Stiffness*

Prelanding muscle activation amplitudes were not able to predict changes in knee joint stiffness values in the current study. Future studies need to examine other time periods of muscle activation including those that encompass part of the postlanding muscle activity period and check their relationships with stiffness measures. Using these more inclusive time periods might allow for a better prediction of knee joint stiffness during the initial landing in a drop jump task. Clarification of this link between neuromuscular and biomechanical parameters will allow healthcare professionals to understand how modification of one parameter (e.g. muscle activation via a training program) would affect another parameter (e.g. landing biomechanics).

As mentioned previously, a possible limitation of the study was that all participants performed the task barefoot, likely altering landing muscle activation patterns and biomechanics. Future researchers could thus consider employing a repeated-measures research design examining the effects of footwear/no footwear on the neuromechanics of the lower extremity in landings in these groups. Finally, given that the hypotheses for this study were based on the premise that increased muscle activation

levels would be a result of decreased H-reflexes, researchers need to further investigate if this link remains viable during functional and high ACL-injury risk activities.

## Conclusions

This study represents a first step in the process of examining differences in lower extremity neuromechanics between female dancers and basketball players during functional activity. Collectively, these findings along with the previous literature suggest that both dancers and basketball players have similar knee muscle activation patterns and joint stiffness in the initial landing of a drop jump task. However, these findings are limited to a familiar task under similar environments, which may have suppressed group differences to some extent. While the primary hypotheses proposed in this study were not supported, low statistical power is likely responsible, as encouraging signs in the form of moderate effect sizes were found between groups across several variables. Additional research in this line of work is warranted to conclusively determine possible ACL injury protective mechanisms employed by dancers during dynamic activity.

This information will help bolster the body of literature examining ACL injury risk factors. Healthcare professionals including athletic trainers and strength and conditioning specialists will also find this information extremely useful, as it will inform them of possible training differences in basketball players and dancers that potentially allow for acute knee injury protection during activity. The ultimate goal of this line of research is to help health professionals by allowing them to design and if necessary,

change training programs to improve neuromuscular function and dynamic stability to reduce the incidence of ACL injury.

REFERENCES

Aagaard, P., Simonsen, E. B., Anderson, J. L., Magnusson, S. P., & Dyhre-Poulsen, P. (2002). Neural adaptation to resistance training: changes in evoked V-wave and H-reflex responses. Journal of Applied Physiology, 92, 2309-2318.

Arampatzis, A., Bruggemann, G.-P., & Klapsing, G. M. (2001). Leg stiffness and mechanical energetic processes during jumping on a sprung surface. Medicine and Science in Sports and Exercise, 33(6), 923-931.

Arampatzis, A., Schade, F., Walsh, M., & Bruggemann, G.-P. (2001). Influence of leg stiffness and its effect on myodynamic jumping performance. Journal of Electromyography and Kinesiology, 11, 355-364.

Arendt, E. A., Agel, J., & Dick, R. (1999). Anterior cruciate ligament injury patterns among collegiate men and women. Journal of Athletic Training, 34(2), 86-92.

Arendt, E. A., & Dick, R. (1995). Knee injury patterns among men and women in collegiate basketball and soccer: NCAA data and review of literature. American Journal of Sports Medicine, 23(6), 694-701.

Baratta, R., Solomonow, M., Zhou, B. H., Leston, D., Chuinard, R., & D'Amrosia, R. (1988). Muscular coactivation: The role of the antagonist musculature in maintaining knee stability. American Journal of Sports Medicine, 16(2), 113-122.

Basmaijian, J. V., & De Luca, C. J. (1985). Muscles Alive (5th ed.). Baltimore: Williams & Wilkins.

Beiser, T. F., Lloyd, D. G., Ackland, T. R., & Cochrane, J. L. (2001). Anticipatory effects on knee joint loading during running and cutting maneuvers. Medicine and Science in Sports and Exercise, 33(7), 1176-1181.

Bennett, D. J., Gorassini, M., & Prochazka, A. (1994). Catching a ball: Contributions of intrinsic muscle stiffness, reflexes, and higher order responses. Canadian Journal of Physiology and Pharmacology, 72, 525-534.

Bennis, N., Roby-Brami, A., Dufosse, M., & Bussel, B. (1996). Anticipation responses to a self-applied load in normal subjects and hemiparetic patients. Journal of Physiology-Paris, 90, 27-42.

Beynnon, B. D., & Fleming, B. C. (1998). Anterior cruciate ligament strain in vivo: A review of previous work. Journal of Biomechanics, 31, 519-525.

Beynnon, B. D., Fleming, B. C., Johnson, R. J., Nichols, C. E., Renstrom, P. A., & Pope, M. H. (1995). Anterior cruciate ligament strain behavior during rehabilitation exercises in vivo. American Journal of Sports Medicine, 23(1), 24-34.

Beynnon, B. D., Howe, J. G., Pope, M. H., Johnson, R. J., & Fleming, B. C. (1992). The measurement of anterior cruciate ligament strain in vivo. International Orthopedics, 16(1), 1-12.

Beynnon, B. D., Johnson, R. J., Fleming, B. C., Peura, G. D., Renstrom, P. A., Nichols, C. E., et al. (1997). The effect of functional knee bracing on the anterior cruciate ligament in the weightbearing and nonweightbearing knee. American Journal of Sports Medicine, 25(3), 353-359.

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

Bosco, C., Viitasalo, J. T., Komi, P. V., & Luhtanen, P. (1982). Combined effect of elastic energy and myoelectrical potentiation during stretch-shortening cycle. Acta Physiologica Scandinavica, 114, 557-565.

Burden, A. M., Trew, M., & Baltzopoulos, V. (2003). Normalisation of gait EMGs: a re-examination. Journal of Electromyography and Kinesiology, 13, 519-532.

Butler, D. L., Noyes, F. R., & Grood, E. S. (1980). Ligamentous restraints to anterior-posterior drawer in the human knee. A biomechanical study. Journal of Bone and Joint Surgery America, 62, 259-270.

Butler, R. J., Crowell, I. H. P., & Davis, I. M. (2003). Lower extremity stiffness: implications for performance and injury. Clinical Biomechanics, 18, 511-517.

Chan, C. W. Y., Jones, G. M., Kearney, R. E., & Watt, D. G. D. (1979). The 'late' electromyographical response to limb displacment in man. I. Evidence for supraspinal contribution. Electroencephalography and Clinical Neurophysiology, 46, 173-181.

Chappell, J. D., Yu, B., Kirkendall, D. T., & Garrett, W. E. (2002). Male and female recreational athletes in stop-jump tasks. American Journal of Sports Medicine, 30(2), 261-267.

Chmielewski, T. L., Rudolph, K. S., & Synder-Mackler, L. (2002). Development of dynamic knee stability after acute ACL injury. Journal of Electromyography and Kinesiology, 12, 267-274.

Chu, D. A. (1996). Explosive Power and Strength. Champaign, Il: Human Kinetics.

Colby, S. M., Francisco, A., Yu, B., Kirkendall, D. T., Finch, M., & Garrett, W. E. J. (2000). Electromyographic and kinematic analysis of cutting maneuvers: implications for anterior cruciate ligament injury. American Journal of Sports Medicine, 28(234-240).

Da Fonseca, S. T., Silva, P. L. P., Ocarino, J. M., Guimaraes, R. B., Oliveira, M. T. C., & Lage, C. A. (2004). Analyses of dynamic co-contraction level in individuals with anterior cruciate ligament injury. Journal of Electromyography and Kinesiology, 14, 239-247.

Da Fonseca, S. T., Vaz, D. V., De Aquino, C. F., & Bricio, R. S. (2005). Muscular co-contraction during walking and landing from a jump: comparison between genders and influence of activity level. Journal of Electromyography and Kinesiology, 15, 1-8.

Darton, K., Lippold, O. C., Shahani, M., & Shahani, U. (1985). Long-latency spinal reflexes in humans. Journal of Neurophysiology, 53(6), 1604-1618.

Davis, R. B., & DeLuca, P. A. (1996). Gait characterization via dynamic joint stiffness. Gait and Posture, 4, 24-231.

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

Delsys-Inc. (2002). Fundamental concepts in EMG signal acquisition: Delsys User Manual at htp://delsys.com/library/tutorials.htm by Delsys Incorporated.

DeLuca, C. J. (1997). The use of surface electromyography in biomechanics. Journal of Applied Biomechanics, 13, 135-163.

DeLuca, C. J. (2002). Surface electromyography: detection and recording: Delsys User Manual at htp://delsys.com/library/tutorials.htm by Delsys Incorporated.

Dempster, W. T. (1955). Space requirements for the seated operator. Ohio: Wright-Patterson Airforce Base :WADCTR pp 55-159.

DeVita, P., & Hortobagyi, T. (2000). Age increases skeletal versus muscular component of lower extremity stiffness during stepping down. Journal of Gerontology, 55A(12), B593-B600.

DeVita, P., & Skelly, W. A. (1992). Effect of landing stiffness on joint kinetics and energetics in the lower extremity. Medicine and Science in Sports and Exercise, 24(1), 108-115.

Dhaher, Y. Y., Tsoumanis, A. D., Houle, T. T., & Rymer, W. Z. (2004). Neuromuscular reflexes contribute to knee stiffness in valgus loading. Journal of Neurophysiology.

Dhaher, Y. Y., Tsoumanis, A. D., & Rymer, W. Z. (2003). Reflex muscle contractions can be elicited by valgus positional perturbations of the human knee. Journal of Biomechanics, 36, 199-209.

Doorenbosch, C. A. M., & Harlaar, J. (2003). A clinically applicable EMG-force model to quantify active stabilization of the knee after a lesion of the anterior cruciate ligament. Clinical Biomechanics, 18, 142-149.

Draganich, L. F., Jaeger, R. J., & Kralj, A. R. (1989). Coactivation of the hamstrings and quadriceps during extension of the knee. Journal of Bone and Joint Surgery, 71(A), 1075-1081.

Draganich, L. F., & Vahey, J. W. (1990). An in vitro study of anterior cruciate ligament strain induced by quadriceps and hamstrings forces. Journal of Orthopaedic Research, 8(1), 57-63.

Dufek, J. S., & Bates, B. T. (1990). The evaluation and prediction of impact force during landings. Medicine and Science in Sports and Exercise, 22(3), 370-377.

Duncan, A., & McDonagh, M. J. N. (2000). Stretch reflex distinguished from pre-programmed muscle activations following landing impacts in man. Journal of Physiology, 526(2), 457-468.

Dyhre-Poulsen, P., & Krogsgaard, M. R. (2000). Muscular reflexes elicited by electrical stimulation of the anterior cruciate ligament in humans. Journal of Applied Physiology, 89, 2191-2195.

Dyhre-Poulsen, P., & Mosfeldt Laursen, A. (1984). Programmed electromyographic activity and negative incremental muscle stiffness in monkeys jumping downward. Journal of Physiology, 350, 121-136.

Edamura, M., Yang, J. F., & Stein, R. B. (1991). Factors that determine the magnitude and time course of human H-reflexes in locomotion. Journal of Neuroscience, 11, 420-427.

Enoka, R. M. (1994). Neuromechanical Basis of Kinesiology (2nd ed.). Champaign, Ill: Human Kinetics.

Evans, R. W., Evans, R. I., & Carvajal, S. (1996). A survey of injuries among broadway performers: types of injuries, treatments, and perceptions of performers. Medical Problems of Performing Artists, 11, 15-19.

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

Farley, C. T., Houdijk, H. H. P., Van Strien, C., & Louie, M. (1998). Mechanism of leg stiffness adjustment for hopping on surfaces of different stiffnesses. Journal of Applied Physiology, 85, 1044-1055.

Farley, C. T., & Morgenroth, D. C. (1999). Leg stiffness primarily depends on ankle stiffness during human hopping. Journal of Biomechanics, 32, 267-273.

Ferretti, A., Papandrea, P., Conteduca, F., & Mariani, P. P. (1992). Knee ligament injuries in volleyball players. American Journal of Sports Medicine, 20(2), 203-207.

Fleming, B. C., Ohlen, G., Renstrom, P. A., Peura, G. D., Beynnon, B. D., & Badger, G. J. (2003). The effects of compressive load and knee joint torque on peak anterior cruciate ligament strains. American Journal of Sports Medicine, 31(5), 701-707.

Fleming, B. C., Renstrom, P. A., Ohlen, G., Johnson, R. J., Peura, G. D., Beynnon, B. D., et al. (2001). The gastrocnemius muscle is an antagonist of the anterior cruciate ligament. Journal of Orthopaedic Research, 19, 1178-1184.

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

Furman, W., Marshall, J. L., & Girgis, F. G. (1976). The anterior cruciate ligament. a functional analysis based on postmortem studies. Journal of Bone and Joint Surgery, 58(A), 179-185.

Gabel, R. H., & Brand, R. A. (1994). The effects of signal conditioning on the statistical analyses of gait EMG. Electroencephalography and Clinical Neurophysiology, 93, 188-201.

Garrick, J. G., & Requa, R. K. (1993). Ballet injuries. American Journal of Sports Medicine, 21(4), 586-590.

Garrison, J. C., Hart, J. M., Palmieri, R. M., Kerrigan, D. C., & Ingersoll, C. D. (2005). Lower extremity EMG in male and female college soccer players during single-leg landing. Journal of Sports Rehabilitation, 14, 48-57.

Gerleman, D. G., & Cook, T. M. (1990). Instrumentation: National Institute for Occupational Safety and Health.

Goldfuss, A. J., Morehouse, C. A., & LeVeau, B. F. (1973). Effect of muscular tension on knee stability. Medicine and Science in Sports and Exercise, 5, 267-271.

Gollhofer, A., & Kyrolainen, H. (1991). Neuromuscular control of the human leg extensor muscles in jump exercises under various stretch-load conditions. International Journal of Sports Medicine, 12(1).

Gollhofer, A., & Schmidtbleicher, D. (1988). Muscle activation patterns of human leg extensors and force-time-characteristics in jumping exercises under increased stretching loads. In G. Groot, A. P. Hollander, P. A. Huijing & G. J. v. I. Schenau (Eds.), International Series on Biomechanics (Vol. XI-A, pp. 143-148). Amsterdam: Free Uni Press.

Granata, K. P., Wilson, S. E., & Padua, D. A. (2002a). Gender differences in active musculoskeletal stiffness. Part I : Quantification in controlled measurements of knee joint dynamics. Journal of Electromyography and Kinesiology, 12(2), 119-126.

Granata, K. P., Wilson, S. E., & Padua, D. A. (2002b). Gender differences in active musculoskeletal stiffness. Part II: Quantification of leg stiffness during functional hopping tasks. Journal of Electromyography and Kinesiology, 12(2), 127-135.

Gray, J., Taunton, J. E., McKenzie, D. C., Clement, D. B., McConkey, J. P., & Davidson, R. G. (1985). A survey of injuries to the anterior cruciate ligament of the knee in female basketball players. International Journal of Sports Medicine, 6(6), 314-316.

Greenwood, R., & Hopkins, A. (1976). Landings from an unexpected fall and a voluntary step. Brain, 99, 375-386.

Griffin, L. Y., Agel, J., Albohm, M. J., Arendt, E. A., Dick, R. W., Garrett, W. E., et al. (2000). Noncontact anterior cruciate ligament injuries: risk factors and prevention strategies. Journal of the American Academy of Orthopedic Surgeons, 8(3), 141-150.

Gwinn, D. E., Wilckens, J. H., McDevitt, E. R., Ross, G., & Kao, T. C. (2000). The relative incidence of anterior cruciate ligament injury in men and women at the United States Naval Academy. American Journal of Sports Medicine, 28(1), 98-102.

Hall, S. (2003). Basic Biomechanics. New York, NY: McGraw-Hill.

Harley, Y. X., Gibson, A. S., Harley, E. H., Lambert, M. I., L., V. C., & Noakes, T. D. (2002). Quadriceps strength and jumping efficiency in dancers. Journal of Dance Medicine and Science, 6(3), 87-94.

Hass, C. J., Schick, E. A., Tillman, M. D., Chow, J. W., Brunt, D., & Cauraugh, J. H. (2005). Knee biomechanics during landings: comparison of pre- and post pubescent females. Medicine and Science in Sports and Exercise, 37(1), 100-107.

Hewett, T. E. (2000). Neuromuscular and hormonal factors associated with knee injuries in female athletes: Strategies for intervention. Sports Medicine, 29(5), 313-327.

Hewett, T. E., Lindenfeld, T. N., Riccobene, J. V., & Noyes, F. R. (1999). The effect of neuromuscular training on the incidence of knee injury in female athletes: A prospective study. American Journal of Sports Medicine, 27(6), 699-705.

Hewett, T. E., & Paterno, M. V. (2002). Strategies for enhancing proprioception and neuromuscular control of the knee. Clinical Orthopaedics and Related Research, 402, 76-94.

Hewett, T. E., Stroupe, A. L., Nance, T. A., & Noyes, F. R. (1996). Plyometric training in female athletes: Decreased impact forces and increased hamstring torques. American Journal of Sports Medicine, 24(6), 765-773.

Hillstrom, H. J., & Triolo, R. J. (1995). EMG Theory. In R. L. Craik & C. A. Oatis (Eds.), Gait Analysis: theory and application (pp. 271-292). St. Loius, Missouri: Mosby Year-Book.

Hirokawa, S., Solomonow, M., Luo, Z., & Lu, Y. (1991). Muscular co-contraction and control of knee stability. Journal of Electromyography and Kinesiology, 1(3), 199-208.

Hoffer, A. S., & Andreasson, S. (1981). Regulation of soleus muscle stiffness in premamillary cats: intrinsic and reflex components. Journal of Neurophysiology, 45(2), 267-285.

Horita, T., Komi, P. V., Nicol, C., & Kyrolainen, H. (2002). Interaction between pre-landing activities and stiffness regulation of the knee joint musculoskeletal system in the drop jump: implications to performance. European Journal of Applied Physiology, 88, 76-84.

Hortobagyi, T., & DeVita, P. (2000). Muscle pre- and coactivity during downward stepping are associated with leg stiffness in aging. Journal of Electromyography and Kinesiology, 10, 117-126.

Houck, J. (2003). Muscle activation patterns of selected lower extremity muscles during stepping and cutting tasks. Journal of Electromyography and Kinesiology, 13, 545-554.

Humphrey, D. R., & Reed, D. J. (1983). Separate cortical systems for control of joint movement and joint stiffness: reciprocal activation and coactivation of antagonist muscles. Advances in Neurology, 39, 347-372.

Huston, L. J., Greenfield, M. L., & Wojtys, E. M. (2000). Anterior cruciate ligament injuries in the female athlete. Clinical Orthopaedics, 372, 50-63.

Hutchinson, M. R., & Ireland, M. L. (1995). Knee injuries in female athletes. Sports Medicine, 19, 288-302.

Ireland, M. L. (2002). The female ACL: Why is it more prone to injury. Orthopedic Clinics of North America, 33, 637-651.

Jackson, A. S., & Pollock, M. L. (1978). Generalized Equations for predicating body density of men. British Journal of Nutrition, 40, 497- 504.

James, C. R., & Bates, B. T. (2003). Classification and comparison of biomechanical response strategies for accommodating landing impact. Journal of Applied Biomechanics, 19, 106-118.

James, C. R., Dufek, J. S., & Bates, B. T. (2000). Effects of injury proneness and task difficulty on joint kinetic variability. Medicine and Science in Sports and Exercise, 32(11), 1833-1844.

Johansson, H. (1991). Role of knee ligaments in proprioception and regulation of muscle stiffness. Journal of Electromyography and Kinesiology, 1(3), 158-179.

Johansson, H., Sjolander, P., & Sojka, P. (1990). Activity in receptor afferents from the anterior cruciate ligament evokes reflex effects on fusimotor neurones. Neuroscience Research, 8, 54-59.

Johansson, H., Sjolander, P., & Sojka, P. (1991). A sensory role for the cruciate ligaments. Clinical Orthopaedics and Related Research, 268, 161-179.

Johansson, R., & Magnusson, M. (1991). Human postural dynamics. Critical Reviews in Biomechanical Engineering, 18, 413-437.

Jomha, N. M., Borton, D. C., Clingleffer, A. J., & Pinczewski, L. A. (1999). Long Term Osteoarthritic Changes in Anterior Cruciate Ligament Reconstructed Knees. Clinical Orthopaedics and Related Research, 358, 188-193.

Kellis, E., Arabatzi, F., & Papadopoulos, C. (2003). Muscle co-activation around the knee in drop jumping using the co-contraction index. Journal of Electromyography and Kinesiology, 13, 229-238.

Kerney, R. E., Stein, R. B., & Parameswaran, L. (1997). Identification of intrinsic and reflex contributions to human ankle stiffness dynamics. IEEE Transactions on Biomedical Engineering, 44, 493-504.

Kernozek, T. W., Torry, M. R., Van Hoof, H., Cowley, H., & Tanner, S. (2005). Gender differences in frontal and sagittal plane biomechanics during drop landings. Medicine and Science in Sports and Exercise, 37(6), 1003-1012.

Knutson, L. M., Soderberg, G. L., Ballantyne, B. T., & Clarke, W. R. (1994). A study of various normalization procedures for within day electromyographic data. Journal of Electromyography and Kinesiology, 4(1), 47-59.

Koceja, D. M., Trimble, M. H., & Earles, D. R. (1993). Inhibition of the soleus H-reflex in standing man. Brain Research, 629, 155-158.

Konradsen, L., Voight, M., & Hojsgaard, C. (1997). Ankle inversion injuries: the role of the dynamic defense mechanism. American Journal of Sports Medicine, 25, 54-58.

Kovacs, I., Tihanyi, J., DeVita, P., Racz, L., Barrier, J., & Hortobagyi, T. (1999). Foot placement modifies kinematics and kinetics during drop jumping. Medicine and Science in Sports and Exercise, 31(5), 708-716.

Kujula, M., Taimela, S., Antti-Poika, I., Orava, S., Tuominen, R., & Myllynen, P. (1995). Acute injuries in soccer, ice hockey, volleyball, basketball, judo, and karate: analysis of national registry data. British Medical Journal, 311, 1465-1468.

Kulas, A. S., Schmitz, R. J., Shultz, S. J., Watson, M. A., & Perrin, D. H. (2006). Energy absorption as a predictor of leg impedence in highly trained females. Journal of Applied Biomechanics, 22(3).

Lark, S. D., Buckley, J. G., Bennett, S., Jones, D., & Sargeant, A. J. (2003). Joint torques and dynamic joint stiffness in elderly and young men stepping down. Clinical Biomechanics, 18, 848-855.

Latash, M. L. (1998). Neurophysiological Basis of Movement. Champaign, Il: Human Kinetics.

Leardini, A., Cappozzo, A., Catani, F., Toksvig-Larsen, S., Petitto, A., Sforza, V., et al. (1999). Validation of a functional method for the estimation of hip joint centre location. Journal of Biomechanics, 32, 99-103.

Lees, A. (1981). Methods of impact absorption when landing from a jump. Engineering in Medicine, 10(4), 207-211.

Lephart, S. M., Ferris, C. M., Riemann, B. L., Myers, J. B., & Fu, F. H. (2002). Gender differences in strength and lower extremity kinematics during landing. Clinical Orthopaedics and Related Research, 401, 162-169.

Lephart, S. M., & Henry, T. J. (1995). Functional rehabilitation for the upper and lower extremity. Orthopedic Clinics of North America, 26, 579-592.

LeVeau, B. F. (1991). Williams and Lissner's Biomechanics of Human Motion (3rd ed.). Philadelphia, PA: W. B. Saunders Company.

Liebermann, D. G., & Goodman, D. (In Press). Pre-landing muscle timing and post-landing effects of falling with continuous vision and in blindfold conditions. Journal of Electromyography and Kinesiology.

Liebermann, D. G., & Hoffman, J. R. (2005). Timing of preparatory landing responses as a function of availability of optic flow information. Journal of Electromyography and Kinesiology, 15, 120-130.

Liederbach, M. (2000). General considerations for guiding dance injury rehabilitation. Journal of Dance Medicine and Science, 4(2), 54-65.

Llewellyn, M., Yang, J. F., & Prochazka, A. (1990). Human H-reflexes are smaller in difficult beam walking than in normal treadmill walking. Experimental Brain Research, 83, 22-28.

Louie, J. K., & Mote, C. D. J. (1987). Contribution of the musculature to rotatory laxity and torsional stiffness at the knee. Journal of Biomechanics, 20(3), 281-300.

Madigan, M. L., & Pidcoe, P. E. (2003). Changes in landing biomechanics during a fatiguing landing activity. Journal of Electromyography and Kinesiology, 13, 491-498.

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

Malone, T. R., Hardaker, W. T., Garrett, W. E., Feagin, J. A., & Bassett, F. H. (1993). Relationship of gender to anterior cruciate ligament injuries in intercollegiate basketball players. Journal of the Southern Orthopedic Association, 2, 36-39.

Markolf, K. L., Bargar, W. L., Shoemaker, S. C., & Amstutz, H. C. (1981). The role of joint load in knee stability. Journal of Bone and Joint Surgery America, 63(4), 570-585.

Markolf, K. L., Gorek, J. F., Kabo, J. M., & Shapiro, M. S. (1990). Direct measurement of resultant forces in the anterior cruciate ligament. Journal of Bone and Joint Surgery, 72-A(4), 55-567.

Markolf, K. L., Graff-Radford, A., & Amstutz, H. C. (1978). In vivo stability. A quantitative assessment using an instrumented clinical testing apparatus. Journal of Bone and Joint Surgery America, 60(5), 664-674.

McDonagh, M. J. N., & Duncan, A. (2002). Interaction of pre-programmed control and natural stretch reflexes in human landing movements. Journal of Physiology, 544, 985-994.

McKinley, P., & Pedotti, A. (1992). Motor strategies in landing from a jump: the role of skill in task execution. Experimental Brain Research, 90, 427-440.

McMahon, T. A., & Cheng, G. C. (1990). The mechanics of running: how does stiffness couple with speed? Journal of Biomechanics, 23(Supplement 1), 65-78.

McNair, P. J., Prapavessis, H., & Callender, K. (2000). Decreasing landing forces: effect of instruction. British Journal of Sports Medicine, 34, 293-296.

McNitt-Gray, J., Hester, D. M. E., Mathiyakom, W., & Munkasy, B. A. (2001). Mechanical demand and multijoint control during landing depend on orientation of the body segments relative to the reaction force. Journal of Biomechanics, 34, 1471-1482.

Merletti, R. (1999). Standards for reporting EMG data: International Society of Electrophysiology and Kinesiology at htp://isek.bu.edu/publications/pdf/ISEK_EMG-Standards.pdf.

Meyer-Lohmann, J., Christakos, C. N., & Wolf, H. (1986). Dominance of the short-latency component in perturbation induced electromyographic responses of long-trained monkeys. Experimental Brain Research, 64(393-399).

Mickelborough, J., van der Linden, M. L., Tallis, R. C., & Ennos, A. R. (2004). Muscle activity during gait initiation in normal elderly people. Gait and Posture, 19, 50-57.

Miyasaka, K. C., Daniel, D. M., Stone, M. L., & Hirshman, P. (1991). The incidence of knee ligament injuries in the general population. American Journal of Knee Surgery, 4, 3-8.

Moeller, J. L., & Lamb, M. M. (1997). Anterior cruciate ligament injuries in female athletes: why are women more susceptible? Physician and Sportsmedicine, 25(4).

Moisio, K. C., Sumner, D. R., Shott, S., & Hurwitz, D. E. (2003). Normalization of joint moments during gait: a comparison of two techniques. Journal of Biomechanics, 36, 599-603.

Myer, G. D., Ford, K. R., & Hewett, T. E. (2005). The effects of gender on quadriceps muscle activation strategies during a maneuver that mimics a high ACL injury risk position. Journal of Electromyography and Kinesiology, 15, 181-189.

Myklebust, G., Engebretsen, L., Braekken, I. H., Skjolberg, A., Olsen, O.-E., & Bahr, R. (2003). Prevention of anterior cruciate ligament injuries in female team handball players: a prospective study over three seasons. Clinical Journal of Sport Medicine, 13.

Mynark, R. G., & Koceja, D. M. (1997). Comparison of soleus H-reflex gain from prone to standing in dancers and controls. Electroencephalography and Clinical Neurophysiology, 10(5), 135-140.

Nichols, T. R. (1994). A biomechanical perspective on spinal mechanisms of coordinated muscular action: an architectural principle. Acta Anatomica, 151, 1-13.

Nielson, J., Crone, C., & Hultborn, H. (1993). H-reflexes are smaller in dancers from the Royal Danish Ballet than in well-trained athletes. European Journal of Applied Physiology, 66, 116-121.

Nielson, J., & Kagamihara, Y. (1992). The regulation of disynaptic reciprocal Ia inhibition during co-contraction of antagonistic muscles in man. Journal of Physiology, 45(6), 373-391.

Noyes, F. R., Barber-Westin, S. D., Fleckenstein, C., Walsh, C., & West, J. (2005). The Drop-Jump screening test: difference in lower limb control by gender and effect on neuromuscular training in female athletes. American Journal of Sports Medicine, 33(2), 197-207.

Noyes, F. R., Butler, D. L., & Malek, M. (1980). Clinical laxity tests and functional stability of the knee. Clinical Orthopaedics and Related Research, 146, 84-89.

Noyes, F. R., Mooar, P. A., Matthews, D. S., & Butler, D. L. (1983). The symptomatic anterior cruciate-deficient knee. Part I: the long-term functional disability in athletically active individuals. Journal of Bone and Joint Surgery, 65(2), 154-162.

Nyland, J. A., Caborn, D. N. M., Shapiro, R., & Johnson, D. L. (1997). Fatigue after eccentric quadriceps femoris work produces earlier gastrocnemius and delayed quadriceps femoris activation during crossover cutting among normal athletic women. Knee Surgery, Sports Traumatology, Arthroscopy, 5, 162-167.

Olsen, O.-E., Myklebust, G., Engebretsen, L., & Bahr, R. (2004). Injury mechanisms for anterior cruciate ligament injuries in team volleyball. American Journal of Sports Medicine, 32(4), 1002-1012.

Onate, J. A., Guskiewicz, K. M., Marshall, S. W., Giuliani, C., Yu, B., & Garrett, W. E. (2005). Instruction of Jump-Landing Technique Using Videotape Feedback: Altering Lower Extremity Motion Patterns. American Journal of Sports Medicine, 33(6), 831-842.

Ozguven, H. N., & Berme, N. (1988). An experimental and analytical study of impact forces during human jumping. Journal of Biomechanics, 21(12), 1061-1066.

Palmieri, R. M., Ingersoll, C. D., Hoffman, M. A., Cordova, M. L., Porter, D. A., Edwards, J. P., et al. (2004). Arthrogenic muscle response to a simulated ankle joint effusion. British Journal of Sports Medicine, 38(1), 26-30.

Patterson, D. D., & Peterson, D. F. (2004). Vertical Jump and Leg Power Norms for Young Adults. Measurement in Physical Education and Exercise Science, 8(1), 33-41.

Pearson, K., & Gordon, J. (2000). Fundamentals of Neuroscience. In T. M. Jessell (Ed.), Spinal Reflexes (pp. 713-735). New York: McGraw Hill.

Pollock, M. L., Schmidt, A. S., & Jackson, A. S. (1980). Measurement of cardiorespiratory fitness and body composition in the clinical setting. Comprehensive Therapy, 6, 12-27.

Pope, M. H., Johnson, R. J., Brown, D. W., & Tighe, C. (1979). The role of musculature in injuries to the medial collateral ligament. Journal of Bone and Joint Surgery America, 61.

Rack, P. M. H., & Westbury, D. R. (1974). The short range stiffness of active mammalian muscle and its effect on mechanical properties. Journal of Physiology, 240, 331-350.

Rainoldi, A., Melchiori, G., & Caruso, I. (2004). A method of positioning electrodes during surface EMG recordings in lower limb muscles. Journal of Neuroscience Methods, 134, 37-43.

Reimann, B. L., & Lephart, S. M. (2002a). The sensorimotor system, Part I: The physiological basis of functional joint stability. Journal of Athletic Training, 37(1), 71-79.

Reimann, B. L., & Lephart, S. M. (2002b). The sensorimotor system, Part II: The role of proprioception in motor control and physiological basis of functional joint stability. Journal of Athletic Training, 37(1), 80-84.

Rovere, G. D., Webb, L., Gristina, A. G., & Vogel, J. M. (1983). Musculoskeletal injuries in theatrical dance students. American Journal of Sports Medicine, 11(4), 195-199.

Salci, Y., Kentel, B. B., Heycan, C., Akin, S., & Korkusuz, F. (2004). Comparison of landing maneuvers between male and female college volleyball players. Clinical Biomechanics, 19, 622-628.

Sander, T. C., Gansneder, B. M., Bunker, L. K., Gaesser, G. A., Saliba, E. N., Perrin, D. H., et al. (2004). Females utilize quadriceps dominant landing strategies that increase anterior tibial shear forces at the knee. Journal of Athletic Training, 39(2 (Suppl.)), S-110.

Santello, M. (2005). Review of motor control mechanisms underlying impact absorption from falls. Gait and Posture, 21, 85-94.

Santello, M., & McDonagh, M. J. N. (1998). The control of timing and amplitude of EMG activity in landing movements in humans. Experimental Physiology, 83, 857-874.

Santello, M., McDonagh, M. J. N., & Challis, J. H. (2001). Visual and non-visual control of landing movements in humans. Journal of Physiology, 537(1), 313-327.

Schieppati, M. (1987). The Hoffmann reflex: A means of assessing spinal reflex excitability and its descending control in man. Progress in Neurobiology, 28, 345-376.

Schmidt, R. A., & Lee, T. D. (1999). Motor control and learning (3rd ed.). Champaign, IL: Human Kinetics.

Schot, P. K., & Dufek, J. S. (1993). Landing performance, part 1: kinematic, kinetic, and neuromuscular aspects. Medicine, Exercise, Nutrition, and Health, 2, 69-83.

Schot, P. K., Dufek, J. S., & Bates, B. T. (1991). Individual joint contributions to shock absorption during vertical drop landing. Proceeds of the 15th Annual Society of Biomechanics, 224-225.

Schot, P. K., Hart, B. A., & Mueller, M. (2002). Within-participant variation in landing kinetics: Movement behavior trait or transient ? Research Quarterly for Exercise and Sport, 73(4), 450-456.

Scioscia, T. N., Griffin, J. R., & Fu, F. H. (2001). Knee ligament and meniscal injuries in dancers. Journal of Dance Medicine and Science, 5(1), 11-15.

Seegmiller, J. G., & McCaw, S. T. (2003). Ground reaction forces among gymnasts and recreational athletes in drop landings. Journal of Athletic Training, 38(4), 311-314.

Seliktar, R., & Bo, L. (1995). The theory of kinetic analysis in human gait. In R. L. Craik & C. A. Oatis (Eds.), Gait Analysis: theory and application. St. Louis, Missouri: Mosby Year-Book.

Shiavi, R., Frigo, C., & Pedotti, A. (1998). Electromyographic signals during gait: criteria for envelope filtering and number of strides. Medical and Biological Engineering and Computing, 35, 171-178.

Shields, R. K., Madhavan, S., Gregg, E., Leitch, J., Peterson, B., Salata, S., et al. (2005). Neuromuscular control of the knee during resisted single-limb squat exercise. American Journal of Sports Medicine, 33(10), 1520-1526.

Shultz, S. J., Carcia, C. R., & Perrin, D. H. (2004). Knee joint laxity affects muscle activation patterns in the healthy knee. Journal of Electromyography and Kinesiology, 14(4), 475-483.

Shultz, S. J., & Perrin, D. H. (1999). Using surface electromyography to assess gender differences in neuromuscular response characteristics. Journal of Athletic Training, 34, 165-176.

Shultz, S. J., Perrin, D. H., Adams, J. M., Arnold, B. L., Gansneder, B. M., & Granata, K. P. (2001). Neuromuscular response characteristics in males and females following knee perturbation in a single leg weight bearing stance. Journal of Athletic Training, 36(1), 37-43.

Shultz, S. J., Sander, T. C., Kirk, S. E., & Johnson, M. L. (2004). Relationship between sex hormones and anterior knee laxity across the menstrual cycle. Medicine and Science in Sports and Exercise, 36(7), 1165-1174.

Shultz, S. J., Windley, T. C., Kulas, A. S., Schmitz, R. J., Valovich-McLeod, T. C., & Perrin, D. H. (2005). Low levels of anterior tibial loading enhance knee extensor reflex response characteristics. Journal of Electromyography and Kinesiology, 15, 61-71.

Smith, A. M. (1981). The co activation of antagonist muscles. Canadian Journal of Physiology and Pharmacology, 59, 733-747.

Snell, R. S. (1995). Clinical Anatomy for Medical Students (5 ed.). Boston, MA, 02108: Little, Brown, and Company.

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

Sommerlath, K., Lysholm, J., & Gillquist, J. (1991). The long-term course after treatment of acute anterior cruciate ligament ruptures. A 9 to 16 year followup. American Journal of Sports Medicine, 19(2), 156-162.

Stein, R. B., & Capaday, C. (1988). The modulation of human reflexes during functional motor tasks. Trends in Neurosciences, 11, 328-332.

Stokes, I. A., Gardner-Morse, M., Henry, S. M., & Badger, G. J. (2000). Decrease in trunk muscular response to perturbation with preactivation of lumbar spine musculature. Spine, 25(12), 1957-1964.

Tang, P.-F., Woollacott, M. H., & Chong, R. K. Y. (1998). Control of reactive balance adjustments in perturbed human walking: roles of proximal and distal postural muscle activity. Experimental Brain Research, 119, 141-152.

Trimble, M. H., & Koceja, D. M. (1994). Modulation of the triceps surae H-reflex with training. International Journal of Neuroscience, 76, 293-303.

Tsuda, E., Okamura, Y., Otsuka, H., Komatsu, T., & Tokuya, S. (2001). Direct evidence of the anterior cruciate ligament-hamstring reflex arc in humans. American Journal of Sports Medicine, 29(1), 83-87.

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

Viitasalo, J. T., Salo, A., & Lahtinen, J. (1998). Neuromuscular functioning of athletes and non-athletes in the drop jump. European Journal of Applied Physiology, 78(5), 432-440.

Walsh, M., Arampatzis, A., Schade, F., & Bruggemann, G.-P. (2004). The effect of drop jump starting height and contact time on power, work performed, and moment of force. Journal of Strength and Conditioning Research, 18(3), 561-566.

Winter, D. A. (1990). Biomechanics and motor control of human movement (2 ed.). New York: John Wiley & Sons Inc.

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

Wojtys, E. M., & Huston, L. (1994). Neuromuscular performance in normal and anterior cruciate ligament-deficient lower extremities. American Journal of Sports Medicine, 22(1), 89-104.

Wolpaw, J. R., & Carp, J. S. (1990). Memory traces in the spinal cord. Trends in Neurosciences, 13, 137-142.

Wolpaw, J. R., & O'Keefe, J. A. (1984). Adaptive plasticity in the primate spinal stretch reflex: evidence for a two-phase process. Journal of Neuroscience, 4, 2718-2724.

Wong, Y. M., & Ng, G. Y. (2005). The double peak-to-peak analysis for determining EMG onset of muscle contraction. Electromyography and Clinical Neurophysiology, 45(5), 267-271.

Woo, S.-Y., Fox, R. J., Sakane, M., Livesay, G. A., Rudy, T. W., & Fu, F. H. (1998). Biomechanics of the ACL: Measurements of in situ force in the ACL and knee kinematics. Knee, 5, 267-288.

Wu, G. (1995). Kinematics Theory. In R. L. Craik & C. A. Oatis (Eds.), Gait Analysis: theory and application (pp. 159-182). St. Louis, Missouri: Mosby Year-Book.

Yang, J. F., & Winter, D. A. (1983). Electromyography reliability in maximal and submaximal isometric contractions. Archives of Physical Medicine and Rehabilitation, 34(9), 41-420.

Zazulak, B. T., Ponce, P. L., Straub, S. J., Medvecky, M. J., Avedisian, L., & Hewett, T. E. (2005). Gender comparison of hip muscle activity during single-leg landing. Journal of Orthopaedic and Sports Physical Therapy, 35, 292-299.

Zehr, E. P. (2002). Considerations for use of the Hoffmann reflex in exercise studies. European Journal of Applied Physiology, 86, 455-468.

Zhang, S.-N., Bates, B. T., & Dufek, J. S. (2000). Contributions of lower extremity joints to energy dissipation during landings. Medicine and Science in Sports and Exercise, 32(4), 812-819.

APPENDIX A: CONSENT TO ACT AS A HUMAN PARTICIPANT: LONG FORM

THE UNIVERSITY OF NORTH CAROLINA

# GREENSBORO

CONSENT TO ACT AS A HUMAN PARTICIPANT: Long Form

Project Title: Differences in Muscle Activation and Knee Joint Stiffness between Female Dancers and Basketball Players during Drop Jumps

Project Director: Jatin P. Ambegaonkar, MS ATC OT CSCS

Participant's Name:

Date of Consent: ____/____/____

DESCRIPTION AND EXPLANATION OF PROCEDURES:

PURPOSE OF THE RESEARCH:
The primary purpose of this project is to determine the differences in the timing and activation of the quadriceps (front of thigh), hamstrings (back of thigh), and lateral gastrocnemius (lateral upper calf) muscles and knee joint stiffness between female dancers and female basketball players during drop jumping

THIS STUDY WILL EXAMINE how basketball players and dancers activate the muscles of their lower extremities when performing drop jumps. We will examine both the timing and magnitude at which you activate your muscles around your knee produce both prior to and following the drop jumps. The amount of stiffening that muscle around your knee produce during the landing (knee joint stiffness), will also be measured.

PARTICIPANT SELECTION:
By agreeing to participate in the study, you are indicating you are a female dancer or basketball player between the ages of 18-30 years, are within 10 days of the onset of your menstrual cycle have no history of recent surgery, injury or chronic pain in your lower extremity, do not have any cardiovascular or neurological problems, have not been asked by your physician to not perform landing and jumping activities, and are otherwise healthy. If you are a dancer, you are indicating that you are involved in dance at least 3 days/ week for at least 30 minutes. If you are a basketball player, you are indicating that you are involved in recreational club or collegiate basketball at least 3 days/ week for the past 3 months for at least 30 minutes. Further, dancing and basketball respectively have been your primary form of physical activity over the past 2 years.

WHAT YOU WILL DO IN THE STUDY:
Prior to testing, your age, height, weight, and years of experience will be recorded. Thereafter, selected anthropometric measurements will be performed including girth, length, and skin fold

measurements will be taken over the back of the arm (triceps), the waist (supra iliac), and the mid thigh Your preferred leg will then determined by asking you to perform 3 double-leg landings from a box 45 cm above two forceplates onto the forceplates. The leg that makes first contact with the forceplates 2 out of 3 times will be referred to as your preferred leg. All measurements will then be taken on this leg.

You will then again stand on the 45 cm box and drop off the box, perform a double-leg drop onto the forceplates, and immediately upon ground contact, will jump up vertically as high as possible and land back onto the forceplate. You will be asked to maintain your balance until told to step off the forceplates by the investigator. Prior to actual testing, you will be prepared or placement of surface electrodes. The preferred leg will be shaved and wiped with alcohol swabs in preparation for placement of two electrodes on the front of the thigh, four on the back of the thigh, two on the upper calf, and one on the flat part of your shin bone. The electrodes will be placed over the muscle bellies of the lateral quadriceps (front of thigh), medial and lateral Hamstring (back of thigh), lateral gastrocnemius (back of calf ), and on the tibia (shin bone). The electrodes will be connected to wires that lead to a computer that measures muscle activity. You will then be asked to perform maximal voluntary contractions of these muscles for normalization purposes while seated in a dynamometer. Your preferred leg will be secured at 30 degrees of knee flexion for these trials. For the quadriceps, you will be asked to kick out with your knee as hard as possible; for the hamstrings, you will be asked to bend your knee as hard as possible; and for the gastrocnemius you will be asked to push your ankle, like you are pushing a gas pedal into the hands of an assistant, who will be resisting your foot motion. Three trials, each lasting for 5 s will be conducted for each of the three muscle groups (quadriceps, hamstrings, gastrocnemius)

Next, four position will be attached on your body (one each for the foot, the shin, the thigh of the preferred leg, and one on the sacrum). Digitization procedures will then be performed to obtain a digital picture of your leg on the computer. You will then perform 10 drop jumps as described above. You may be asked to repeat a trial if your hands come off your hips, you lose balance during the drop jumps, or you do not land back onto the forceplates.

If you are one of the first 5 dancers or basketball players, you may be asked to come back for testing on a second day with the same protocol being followed on the second testing session.

TIME REQUIRED: About 1 1/2 hour.

CONFIDENTIALITY:

The information that you give in the study will be handled confidentially. Your information will be assigned a code number. The list connecting your name to this number will be kept in a locked file. When the study is completed and the data have been analyzed, this list will be destroyed. Your name will not be used in any report.

VOLUNTARY PARTICIPATION:

Your participation in the study is completely voluntary.

RISKS AND DISCOMFORTS:

There is a possibility that you may land awkwardly during the testing and subsequently suffer a strain, sprain, or contusion. If at any time the testing causes you any discomfort or concern, please notify the investigator immediately. Please contact Dr. Eric Allen at 336-334-5878 about any research related injuries.

POTENTIAL BENEFITS:
There are no direct benefits to you for participating in the study. The study may help us better understand the risk factors associated with injuries in the lower extremity.

CONSENT:
By signing this consent form, you agree that you understand the procedures and any risks and benefits involved in this research. You are free to refuse to participate or to withdraw your consent to participate in this research at any time without penalty or prejudice; your participation is entirely voluntary. Your privacy will be protected because you will not be identified by name as a participant in this project.

The research and this consent form have been approved by the University of North Carolina at Greensboro Institutional Review Board, which insures that research involving people follows federal regulations. Questions regarding your rights as a participant in this project can be answered by calling Mr. Eric Allen at 336-334-5878. Questions regarding the research itself will be answered by Jatin P. Ambegaonkar by calling 336-334-3039 or by David H. Perrin at 336-334-5644. Any new information that develops during the project will be provided to you if the information might affect your willingness to continue participation in the project.

By signing this form, you are agreeing to participate in the project described to you by Jatin P. Ambegaonkar.

_____
Participant's Signature*

APPENDIX B: DEMOGRAPHICS DATA SHEET AND ACTIVITY HISTORY QUESTIONNAIRE

Participant ID: _____ Date: _____

Name: _____ Preferred Leg: Right Left

Age: _____ Height: _____ cm Weight: _____ kg

1. Primary form of physical activity for the past two years: Basketball Dance
2. For how long have you been involved in Basketball or Dance? _____ yrs ___ mnths
3. Did you have any surgery on your lower extremity or any lower extremity musculoskeletal injury in the past six months? Yes No
4. Are you physically active at least 3 days/week for at least 30 mins/day? Yes No
5. Do you regularly perform any other physical activity? Yes No

If yes, Please specify

Type: _____ Frequency: _____ Duration: _____

6. Do you have any history of cardiovascular or neurological problems that not allow you to participate in landing/jumping activities? Yes No
7. Have been told by your physician to avoid landing/jumping activities? Yes No
8. Are you in within the first ten days of your menstrual cycle (Day 1 = First day of menstrual bleeding)? Yes No
9. Jackson and Pollock 3 Site Skinfold Measurements for Females (Right Side)
    a. Triceps = ____ , ____ , ____ mm
    b. Supra Iliac = ____ , ____ , ____ mm
    c. Thigh = ____ , ____ , ____ mm
10. Number of Trials needed for familiarization: ____

## APPENDIX C: INSTRUCTIONS FOR THREE-SITE SKINFOLD MEASUREMENTS

A. TRICEPS: Between the tip of the olecranon process of the ulna (elbow) and the acromion process of the scapula (shoulder).

   1. Mark the point of the back of the arm midway between the tip of the elbow and the shoulder.
   2. Pick up skinfold with thumb and forefinger of the left hand.
   3. Apply jaws of the caliper to the skinfold so that the mark is midway between the jaws.
   4. Release your thumb from the caliper handle, so that the tips of the caliper have full exertion on the skinfold. Take reading immediately after the first rapid fall.
   5. Repeat steps 2 through 4 three times. Average of the 3 is the measurement of this site.

B. SUPRA ILIAC: Above ilac crest in mid-axillary line. (approximately 2.5 cm above hip bone.)

   1. Pick up skinfold following the natural fold of the skin (horizontal).
   2. Mark midway the fold. While holding the skinfold approximately 1 inch from the mark, proceed with steps 3, 4, and 5; in Triceps section

C. THIGH: Midway between the patella and the inguinal crease

   1. Pick up skinfold following the natural fold of the skin (vertical).
   2. Mark midway the fold. While holding the skinfold approximately 1 inch from the mark, proceed with steps 3, 4, and 5; in Triceps section

Source: Beta Technology (2005). Lange Skinfold Caliper Operator's Manual.

APPENDIX D: MEANS, STANDARD DEVIATIONS, INTRACLASS CORRELATION COEFFICIENTS ($ICC_{2,K}$) AND STANDARD ERROR OF MEASUREMENTS (SEM) COMPARING ENSEMBLE AVERAGES OF TRIALS 1-5 WITH 1-10

| LQ | | $\bar{X}$ 1-5 | Sd 1-5 | $\bar{X}$ 1-10 | Sd 1-10 | $ICC_{2,K}$ | $SEM_{2,K}$ | N | TMS | EMS | BMS |
|---|---|---|---|---|---|---|---|---|---|---|---|
| | Onset (ms) | 63.00 | 91.21 | 50.64 | 71.31 | 0.97 | 15.26 | 11 | 840.73 | 320.13 | 13085.53 |
| | $_{PRE}$(v) Raw | 107.14 | 55.08 | 1111.16 | 54.89 | 0.97 | 9.10 | 11 | 89.20 | 167.28 | 5879.13 |
| | $_{POST}$ (v) Raw | 624.11 | 323.02 | 630.21 | 320.36 | 0.98 | 42.77 | 11 | 205.26 | 3887.39 | 203003.49 |

| MH | | $\bar{X}$ 1-5 | Sd 1-5 | $\bar{X}$ 1-10 | Sd 1-10 | $ICC_{2,K}$ | $SEM_{2,K}$ | N | TMS | EMS | BMS |
|---|---|---|---|---|---|---|---|---|---|---|---|
| | Onset (ms) | 123.45 | 39.98 | 117.45 | 37.79 | 0.96 | 7.49 | 11 | 198.00 | 96.50 | 2851.64 |
| | $_{PRE}$(v) Raw | 230.51 | 150.14 | 222.60 | 147.92 | 0.98 | 18.85 | 11 | 344.05 | 722.50 | 43702.44 |
| | $_{POST}$ (v) Raw | 287.27 | 260.40 | 276.17 | 261.84 | 1.00 | 11.07 | 11 | 677.66 | 199.81 | 136165.57 |

| LH | | $\bar{X}$ 1-5 | Sd 1-5 | $\bar{X}$ 1-10 | Sd 1-10 | $ICC_{2,K}$ | $SEM_{2,K}$ | N | TMS | EMS | BMS |
|---|---|---|---|---|---|---|---|---|---|---|---|
| | Onset (m) | 96.64 | 46.52 | 97.73 | 46.06 | 0.88 | 33.23 | 11 | 6.55 | 531.45 | 4232.13 |
| | $_{PRE}$(v) Raw | 112.80 | 65.86 | 107.16 | 57.55 | 0.98 | 10.30 | 11 | 174.73 | 174.58 | 7139.62 |
| | $_{POST}$ (v) Raw | 197.12 | 85.45 | 192.26 | 78.21 | 0.99 | 9.82 | 11 | 130.10 | 179.35 | 13239.02 |

| LG | | $\bar{X}$ 1-5 | Sd 1-5 | $\bar{X}$ 1-10 | Sd 1-10 | $ICC_{2,K}$ | $SEM_{2,K}$ | N | TMS | EMS | BMS |
|---|---|---|---|---|---|---|---|---|---|---|---|
| | Onset (ms) | 155.36 | 91.45 | 153.45 | 81.52 | 0.99 | 6.80 | 11 | 20.05 | 88.65 | 14920.48 |
| | $_{PRE}$(v) Raw | 190.81 | 130.81 | 189.80 | 132.28 | 1.00 | 4.46 | 11 | 5.70 | 42.62 | 34569.75 |
| | $_{POST}$ (v) Raw | 187.78 | 97.13 | 192.35 | 102.90 | 0.99 | 10.65 | 11 | 114.55 | 221.77 | 19800.84 |

LQ = Lateral Quadricep; MH = Medial Hamstrings; LH = Lateral Hamstring; LG = Lateral Gastrocnemius; $_{PRE}$ Raw = Pre-landing muscle activation amplitude raw voltage; $_{POST}$ Raw = Postlanding Muscle Activation Amplitude raw voltage; TMS = Total Mean Square; EMS = Error Mean Square; BMS = Between Mean Square Error; N = Number of Subjects

APPENDIX E: MEANS, STANDARD DEVIATIONS (SD), INTRACLASS CORRELATION COEFFICIENTS (ICC) AND STANDARD ERROR OF MEASUREMENTS(SEM) WHEN COMPARING ENSEMBLE AVERAGES OF TRIALS 1-2WITH 9-10

| LQ | | $\bar{X}$ 1-5 | Sd 1-5 | $\bar{X}$ 1-10 | Sd 1-10 | ICC $_{2,K}$ | SEM $_{2,K}$ | N | TMS | EMS | BMS |
|---|---|---|---|---|---|---|---|---|---|---|---|
| | | 52.55 | 51.54 | 61.18 | 77.53 | 0.93 | 20.60 | # | 410.23 | 585.27 | 8079.51 |
| | Onset (ms) | 29.92 | 15.73 | 31.57 | 16.50 | 0.94 | 4.13 | 11 | 14.56 | 32.05 | 487.76 |
| | $_{PRE}$(v) Raw | 138.56 | 95.21 | 128.72 | 88.20 | 0.94 | 23.11 | 11 | 532.15 | 972.35 | 15870.54 |
| | $_{POST}$ (v) Raw | | | | | | | | | | |
| MH | | $\bar{X}$ 1-5 | Sd 1-5 | $\bar{X}$ 1-10 | Sd 1-10 | ICC $_{2,K}$ | SEM $_{2,K}$ | N | TMS | EMS | BMS |
| | | 144.91 | 49.87 | 134.09 | 41.54 | 0.89 | 16.62 | 11 | 643.68 | 403.88 | 3807.90 |
| | Onset (ms) | 56.36 | 39.52 | 49.51 | 38.62 | 0.88 | 13.45 | 11 | 257.73 | 321.43 | 2732.37 |
| | $_{PRE}$(v) Raw | 88.22 | 118.19 | 71.67 | 45.45 | 0.83 | 48.45 | 11 | 12158.90 | 3163.50 | 22870.98 |
| | $_{POST}$ (v) Raw | | | | | | | | | | |
| LH | | $\bar{X}$ 1-5 | Sd 1-5 | $\bar{X}$ 1-10 | Sd 1-10 | ICC $_{2,K}$ | SEM $_{2,K}$ | N | TMS | EMS | BMS |
| | | 108.36 | 47.26 | 107.91 | 42.82 | 0.96 | 9.87 | 11 | 1.14 | 185.36 | 3881.31 |
| | Onset (m) | 31.73 | 23.97 | 28.73 | 25.98 | 0.90 | 8.31 | 11 | 49.50 | 121.43 | 1128.39 |
| | $_{PRE}$(v) Raw | 59.99 | 68.72 | 58.46 | 54.28 | 0.93 | 18.24 | 11 | 12.98 | 546.73 | 7121.60 |
| | $_{POST}$ (v) Raw | | | | | | | | | | |
| LG | | $\bar{X}$ 1-5 | Sd 1-5 | $\bar{X}$ 1-10 | Sd 1-10 | ICC $_{2,K}$ | SEM $_{2,K}$ | N | TMS | EMS | BMS |
| | | 152.36 | 92.34 | 167.55 | 90.08 | 0.84 | 36.53 | 11 | 1267.68 | 2321.48 | 14318.25 |
| | Onset (ms) | 25.22 | 16.23 | 29.79 | 31.71 | 0.86 | 12.06 | 11 | 115.00 | 163.65 | 1105.29 |
| | $_{PRE}$(v) Raw | 47.06 | 31.60 | 58.27 | 50.93 | 0.78 | 24.11 | 11 | 692.16 | 655.91 | 2937.39 |

LQ = Lateral Quadricep; MH = Medial Hamstrings; LH = Lateral Hamstring; LG = Lateral Gastrocnemius; $_{PRE}$ Raw = Pre-landing muscle activation amplitude raw voltage; $_{POST}$ Raw = Postlanding Muscle Activation Amplitude raw voltage; TMS = Total Mean Square; EMS = Error Mean Square; BMS = Between Mean Square Error; N = Number of Subject

APPENDIX F: RAW DEMOGRAPHIC DATA

| # | Code | Age (yrs) | Height (cm) | Weight (kg) | Experience (yrs) | Preferred Leg | Body Fat Percentage | Familiarization Trials Needed |
|---|---|---|---|---|---|---|---|---|
| S01 | Dance-01 | 22 | 174 | 74 | 20 | Right | 20.63 | 5 |
| S02 | Dance-02 | 19 | 152 | 54 | 8 | Right | 21.67 | 4 |
| S03 | Dance-03 | 22 | 165 | 59 | 8 | Left | 21.59 | 5 |
| S04 | Dance-04 | 27 | 164 | 61 | 20 | Left | 25.32 | 7 |
| S05 | Dance-05 | 25 | 164 | 60 | 22 | Left | 20.74 | 3 |
| S06 | Dance-06 | 20 | 153 | 54 | 12 | Right | 24.31 | 7 |
| S07 | Dance-07 | 20 | 160 | 55 | 5 | Left | 18.93 | 4 |
| S08 | Dance-08 | 19 | 155 | 50 | 10 | Right | 20.73 | 6 |
| S09 | Dance-09 | 20 | 162 | 53 | 10 | Right | 17.35 | 4 |
| S10 | Dance-10 | 25 | 163 | 58 | 21 | Right | 19.77 | 2 |
| S11 | Dance-11 | 20 | 168 | 52 | 17 | Right | 17.64 | 2 |
| S12 | Dance-12 | 24 | 165 | 67 | 19 | Left | 19.23 | 3 |
| S13 | Dance-13 | 21 | 161 | 54 | 7 | Left | 18.75 | 5 |
| S14 | Dance-14 | 19 | 170 | 56 | 15 | Right | 22.06 | 6 |
| S15 | Dance-15 | 19 | 174 | 69 | 15 | Left | 25.70 | 7 |
| S16 | Dance-16 | 22 | 171 | 91 | 20 | Left | 31.49 | 4 |
| S17 | Dance-17 | 20 | 173 | 67 | 18 | Left | 23.53 | 3 |
| S18 | Dance-18 | 19 | 168 | 78 | 14 | Left | 31.84 | 3 |
| S19 | Dance-19 | 19 | 160 | 67 | 17 | Left | 29.58 | 4 |
| S20 | Dance-20 | 18 | 176 | 69 | 15 | Left | 23.56 | 6 |

| # | Code | Age (yrs) | Height (cm) | Weight (kg) | Experience (yrs) | Preferred Leg | Body Fat Percentage | Familiarization Trials Needed |
|---|---|---|---|---|---|---|---|---|
| S21 | Dance-21 | 19 | 160 | 60 | 16 | Right | 22.83 | 4 |
| S22 | Dance-22 | 22 | 168 | 58 | 5 | Right | 21.47 | 3 |
| S23 | Dance-23 | 23 | 166 | 77 | 17 | Right | 24.52 | 7 |
| S24 | Dance-24 | 20 | 165 | 66 | 7 | Right | 22.93 | 2 |
| S25 | Dance-25 | 18 | 154 | 47 | 12 | Right | 19.30 | 2 |
| S26 | Dance-26 | 22 | 160 | 61 | 6 | Right | 21.22 | 2 |
| S27 | Dance-27 | 19 | 172 | 72 | 16 | Right | 25.52 | 4 |
| S28 | Dance-28 | 19 | 158 | 80 | 17 | Left | 34.95 | 2 |
| S29 | Dance-29 | 19 | 165 | 32 | 6 | Right | 24.57 | 2 |
| S30 | Dance-30 | 19 | 169 | 61 | 17 | Left | 27.83 | 5 |
| S31 | Dance-31 | 23 | 149 | 56 | 20 | Left | 18.73 | 2 |
| S32 | Dance-32 | 18 | 171 | 68 | 13 | Left | 26.49 | 5 |
| S33 | Dance-33 | 19 | 170 | 61 | 13 | Right | 23.49 | 4 |
| S34 | Dance-34 | 20 | 165 | 65 | 10 | Right | 28.59 | 3 |
| S35 | Dance-35 | 24 | 161 | 69 | 21 | Left | 31.08 | 2 |
| S36 | Basketball-01 | 19 | 173 | 71 | 7 | Right | 24.60 | 4 |
| S37 | Basketball-02 | 19 | 177 | 66 | 15 | Right | 24.95 | 2 |
| S38 | Basketball-03 | 19 | 171 | 73 | 12 | Left | 26.51 | 2 |
| S39 | Basketball-04 | 19 | 178 | 76 | 13 | Left | 28.47 | 3 |
| S40 | Basketball-05 | 23 | 165 | 71 | 10 | Left | 20.62 | 4 |

| # | Code | Age (yrs) | Height (cm) | Weight (kg) | Experience (yrs) | Preferred Leg | Body Fat Percentage | Familiarization Trials Needed |
|---|---|---|---|---|---|---|---|---|
| S41 | Basketball-06 | 20 | 177 | 76 | 6 | Left | 23.37 | 7 |
| S42 | Basketball-07 | 18 | 169 | 71 | 5 | Right | 26.40 | 2 |
| S43 | Basketball-08 | 18 | 176 | 70 | 8 | Left | 27.92 | 2 |
| S44 | Basketball-09 | 18 | 174 | 98 | 11 | Left | 35.71 | 4 |
| S45 | Basketball-10 | 20 | 180 | 103 | 12 | Left | 33.11 | 5 |
| S46 | Basketball-11 | 26 | 178 | 84 | 14 | Left | 33.76 | 6 |
| S47 | Basketball-12 | 19 | 167 | 70 | 11 | Left | 24.10 | 5 |
| S48 | Basketball-13 | 19 | 164 | 69 | 10 | Left | 19.16 | 6 |
| S49 | Basketball-14 | 20 | 158 | 54 | 6 | Left | 26.29 | 8 |
| S50 | Basketball-15 | 22 | 165 | 59 | 8 | Left | 21.59 | 5 |
| S51 | Basketball-16 | 20 | 165 | 66 | 7 | Right | 22.93 | 2 |
| S52 | Basketball-17 | 20 | 168 | 66 | 13 | Left | 24.87 | 3 |
| S53 | Basketball-18 | 19 | 172 | 68 | 13 | Left | 27.38 | 2 |
| S54 | Basketball-19 | 21 | 162 | 71 | 15 | Left | 26.97 | 4 |
| S55 | Basketball-20 | 23 | 170 | 71 | 17 | Left | 25.47 | 2 |

APPENDIX G: RAW MUSCLE ONSET DATA (MS)

| # | Code | $LG_{ON}$ | $MH_{ON}$ | $LH_{ON}$ | $LQ_{ON}$ |
|---|---|---|---|---|---|
| S01 | Dance-01 | 145 | 171 | 110 | 31 |
| S02 | Dance-02 | 151 | 189 | 188 | 67 |
| S03 | Dance-03 | 157 | 120 | 88 | 59 |
| S04 | Dance-04 | 252 | 103 | 102 | 82 |
| S05 | Dance-05 | 224 | 97 | 93 | 77 |
| S06 | Dance-06 | 94 | 88 | 99 | 96 |
| S07 | Dance-07 | 132 | 163 | 135 | 72 |
| S08 | Dance-08 | 71 | 244 | 142 | 186 |
| S09 | Dance-09 | 158 | 227 | 168 | 81 |
| S10 | Dance-10 | 125 | 217 | 111 | 47 |
| S11 | Dance-11 | 178 | 128 | 211 | 210 |
| S12 | Dance-12 | 133 | 133 | 150 | 184 |
| S13 | Dance-13 | 177 | 162 | 136 | 241 |
| S14 | Dance-14 | 141 | 224 | 177 | 194 |
| S15 | Dance-15 | 160 | 230 | 165 | 174 |
| S16 | Dance-16 | 131 | 168 | 211 | 96 |
| S17 | Dance-17 | 163 | 228 | 144 | 116 |
| S18 | Dance-18 | 148 | 107 | 192 | 153 |
| S19 | Dance-19 | 148 | 138 | 66 | 265 |
| S20 | Dance-20 | 90 | 178 | 120 | 13 |

LG = Lateral Gastrocnemius; MH = Medial Hamstrings; LH = Lateral Hamstring; LQ = Lateral Quadriceps; $_{ON}$ = Muscle Activity Onsets

| # | Code | $LG_{ON}$ | $MH_{ON}$ | $LH_{ON}$ | $LQ_{ON}$ |
|---|---|---|---|---|---|
| S21 | Dance-21 | 72 | 53 | 90 | 29 |
| S22 | Dance-22 | 96 | 165 | 170 | 74 |
| S23 | Dance-23 | 136 | 162 | 140 | 164 |
| S24 | Dance-24 | 206 | 265 | 165 | 176 |
| S25 | Dance-25 | 245 | 181 | 129 | 41 |
| S26 | Dance-26 | 121 | 104 | 52 | 47 |
| S27 | Dance-27 | 277 | 137 | 90 | 40 |
| S28 | Dance-28 | 85 | 116 | 109 | 38 |
| S29 | Dance-29 | 100 | 144 | 106 | 37 |
| S30 | Dance-30 | 93 | 124 | 68 | 59 |
| S31 | Dance-31 | 187 | 159 | 139 | 14 |
| S32 | Dance-32 | 98 | 84 | 89 | 71 |
| S33 | Dance-33 | 217 | 168 | 181 | 55 |
| S34 | Dance-34 | 103 | 185 | 120 | 32 |
| S35 | Dance-35 | 115 | 174 | 166 | 65 |
| S36 | Basketball-01 | 145 | 145 | 55 | 159 |
| S37 | Basketball-02 | 122 | 217 | 156 | 201 |
| S38 | Basketball-03 | 98 | 103 | 53 | 52 |
| S39 | Basketball-04 | 131 | 185 | 129 | 141 |
| S40 | Basketball-05 | 79 | 60 | 64 | 39 |

LG = Lateral Gastrocnemius; MH = Medial Hamstrings; LH = Lateral Hamstring; LQ = Lateral Quadriceps; $_{ON}$ = Muscle Activity Onsets

| # | Code | $LG_{ON}$ | $MH_{ON}$ | $LH_{ON}$ | $LQ_{ON}$ |
|---|---|---|---|---|---|
| S41 | Basketball-06 | 129 | 103 | 214 | 90 |
| S42 | Basketball-07 | 177 | 149 | 161 | 188 |
| S43 | Basketball-08 | 163 | 169 | 99 | 41 |
| S44 | Basketball-09 | 217 | 67 | 128 | 58 |
| S45 | Basketball-10 | 35 | 86 | 31 | 82 |
| S46 | Basketball-11 | 118 | 85 | 85 | 107 |
| S47 | Basketball-12 | 220 | 206 | 161 | 114 |
| S48 | Basketball-13 | 193 | 87 | 180 | 50 |
| S49 | Basketball-14 | 151 | 110 | 163 | 136 |
| S50 | Basketball-15 | 116 | 150 | 170 | 59 |
| S51 | Basketball-16 | 116 | 150 | 170 | 59 |
| S52 | Basketball-17 | 118 | 159 | 174 | 44 |
| S53 | Basketball-18 | 106 | 139 | 118 | 63 |
| S54 | Basketball-19 | 261 | 143 | 89 | 22 |
| S55 | Basketball - 20 | 122 | 114 | 147 | 34 |

LG = Lateral Gastrocnemius; MH = Medial Hamstrings; LH = Lateral Hamstring; LQ = Lateral Quadriceps; $_{ON}$ = Muscle Activity Onsets

APPENDIX H: RAW MUSCLE ACTIVATION DATA (% MVIC)

| # | Code | $LG_{PRE}$ | $MH_{PRE}$ | $LH_{PRE}$ | $LQ_{PRE}$ | $LG_{POST}$ | $MH_{POST}$ | $LH_{POST}$ | $LQ_{POST}$ |
|---|---|---|---|---|---|---|---|---|---|
| S01 | Dance-01 | 74.31 | 68.77 | 21.47 | 37.53 | 31.40 | 29.13 | 24.30 | 178.01 |
| S02 | Dance-02 | 51.37 | 35.14 | 20.46 | 9.33 | 47.93 | 38.24 | 28.91 | 71.09 |
| S03 | Dance-03 | 38.38 | 40.12 | 16.23 | 10.91 | 67.97 | 134.28 | 21.95 | 82.44 |
| S04 | Dance-04 | 44.41 | 46.25 | 23.51 | 14.77 | 72.22 | 104.92 | 47.66 | 93.35 |
| S05 | Dance-05 | 44.82 | 36.47 | 23.97 | 5.21 | 33.17 | 37.94 | 62.27 | 19.65 |
| S06 | Dance-06 | 26.89 | 19.99 | 23.40 | 16.28 | 27.38 | 48.45 | 135.75 | 98.96 |
| S07 | Dance-07 | 47.72 | 61.54 | 27.56 | 16.36 | 38.50 | 168.76 | 45.52 | 94.25 |
| S08 | Dance-08 | 16.50 | 26.54 | 21.07 | 20.16 | 36.51 | 35.08 | 62.88 | 103.84 |
| S09 | Dance-09 | 19.90 | 19.98 | 14.93 | 14.08 | 33.65 | 23.31 | 8.80 | 124.35 |
| S10 | Dance-10 | 46.13 | 34.66 | 30.76 | 18.48 | 55.98 | 18.66 | 16.11 | 124.21 |
| S11 | Dance-11 | 36.64 | 28.22 | 19.24 | 7.44 | 27.23 | 20.87 | 16.57 | 30.73 |
| S12 | Dance-12 | 73.41 | 29.66 | 24.86 | 28.99 | 37.06 | 25.53 | 27.26 | 100.31 |
| S13 | Dance-13 | 47.06 | 45.27 | 25.71 | 12.89 | 38.51 | 52.71 | 51.65 | 55.22 |
| S14 | Dance-14 | 41.85 | 57.59 | 29.52 | 13.39 | 38.93 | 33.83 | 23.18 | 69.05 |
| S15 | Dance-15 | 31.30 | 33.68 | 24.19 | 14.19 | 37.64 | 24.88 | 38.37 | 98.71 |
| S16 | Dance-16 | 52.66 | 41.44 | 27.95 | 18.96 | 50.11 | 26.00 | 25.06 | 48.95 |
| S17 | Dance-17 | 40.88 | 51.37 | 17.96 | 52.78 | 45.87 | 30.72 | 13.96 | 116.20 |
| S18 | Dance-18 | 37.07 | 53.37 | 34.38 | 6.01 | 51.26 | 35.12 | 29.01 | 25.02 |
| S19 | Dance-19 | 28.77 | 26.16 | 6.65 | 14.37 | 37.02 | 18.23 | 11.55 | 112.60 |
| S20 | Dance-20 | 24.53 | 22.70 | 14.77 | 18.23 | 29.42 | 19.63 | 13.18 | 71.29 |

LG = Lateral Gastrocnemius; MH = Medial Hamstrings; LH = Lateral Hamstring; LQ = Lateral Quadriceps; $_{PRE}$ = 150 ms before ground contact; $_{POST}$ = 50 ms after ground contact

| # | Code | $LG_{PRE}$ | $MH_{PRE}$ | $LH_{PRE}$ | $LQ_{PRE}$ | $LG_{POST}$ | $MH_{POST}$ | $LH_{POST}$ | $LQ_{POST}$ |
|---|---|---|---|---|---|---|---|---|---|
| S21 | Dance-21 | 22.22 | 17.47 | 13.17 | 17.48 | 29.60 | 28.99 | 22.18 | 42.06 |
| S22 | Dance-22 | 62.87 | 33.89 | 18.90 | 27.00 | 76.05 | 32.95 | 24.40 | 168.44 |
| S23 | Dance-23 | 24.75 | 21.26 | 14.37 | 7.21 | 34.92 | 23.54 | 14.55 | 32.11 |
| S24 | Dance-24 | 102.26 | 30.22 | 19.08 | 28.91 | 55.47 | 19.56 | 13.78 | 172.09 |
| S25 | Dance-25 | 35.44 | 24.43 | 15.75 | 17.40 | 19.98 | 19.84 | 11.07 | 57.37 |
| S26 | Dance-26 | 29.54 | 17.09 | 11.12 | 8.35 | 42.62 | 21.61 | 18.80 | 49.57 |
| S27 | Dance-27 | 36.32 | 18.17 | 10.80 | 30.82 | 40.91 | 19.30 | 15.76 | 147.04 |
| S28 | Dance-28 | 21.17 | 15.90 | 32.31 | 23.35 | 25.64 | 32.33 | 85.31 | 130.32 |
| S29 | Dance-29 | 25.42 | 16.18 | 7.20 | 16.87 | 45.13 | 18.86 | 7.73 | 68.17 |
| S30 | Dance-30 | 20.65 | 19.62 | 13.87 | 23.45 | 16.19 | 33.78 | 18.59 | 132.70 |
| S31 | Dance-31 | 40.92 | 52.23 | 17.21 | 16.16 | 31.07 | 39.76 | 50.58 | 122.07 |
| S32 | Dance-32 | 34.80 | 49.04 | 16.52 | 11.76 | 48.44 | 46.55 | 20.14 | 83.45 |
| S33 | Dance-33 | 57.70 | 38.02 | 32.31 | 14.88 | 40.79 | 39.97 | 35.75 | 75.49 |
| S34 | Dance-34 | 32.16 | 27.43 | 31.82 | 16.52 | 38.57 | 15.66 | 27.85 | 47.90 |
| S35 | Dance-35 | 22.34 | 29.71 | 18.76 | 25.41 | 194.54 | 19.49 | 15.77 | 90.25 |
| S36 | Basketball-01 | 26.12 | 10.33 | 7.41 | 6.41 | 32.24 | 9.22 | 16.75 | 92.05 |
| S37 | Basketball-02 | 29.24 | 23.06 | 12.29 | 4.51 | 39.52 | 13.83 | 11.36 | 27.77 |
| S38 | Basketball-03 | 26.77 | 6.71 | 10.75 | 7.67 | 23.44 | 7.24 | 10.95 | 76.15 |
| S39 | Basketball-04 | 43.39 | 31.22 | 11.06 | 22.53 | 47.19 | 18.51 | 10.60 | 240.16 |
| S40 | Basketball-05 | 18.86 | 27.81 | 18.42 | 6.80 | 14.03 | 12.01 | 17.67 | 63.96 |

LG = Lateral Gastrocnemius; MH = Medial Hamstrings; LH = Lateral Hamstring; LQ = Lateral Quadriceps; $_{PRE}$ = 150 ms before ground contact; $_{POST}$ = 50 ms after ground contact

| # | Code | $LG_{PRE}$ | $MH_{PRE}$ | $LH_{PRE}$ | $LQ_{PRE}$ | $LG_{POST}$ | $MH_{POST}$ | $LH_{POST}$ | $LQ_{POST}$ |
|---|---|---|---|---|---|---|---|---|---|
| S41 | Basketball-06 | 23.82 | 14.46 | 12.93 | 11.46 | 26.67 | 10.02 | 15.23 | 105.61 |
| S42 | Basketball-07 | 66.65 | 32.73 | 13.44 | 20.37 | 48.17 | 39.50 | 10.85 | 50.73 |
| S43 | Basketball-08 | 29.88 | 26.54 | 17.02 | 11.65 | 26.83 | 18.14 | 17.39 | 58.42 |
| S44 | Basketball-09 | 53.26 | 23.35 | 13.38 | 25.76 | 42.67 | 36.73 | 23.34 | 117.75 |
| S45 | Basketball-10 | 23.56 | 18.17 | 30.96 | 23.60 | 20.25 | 22.53 | 58.27 | 122.41 |
| S46 | Basketball-11 | 33.98 | 22.00 | 16.94 | 15.28 | 34.91 | 30.03 | 22.95 | 121.87 |
| S47 | Basketball-12 | 65.77 | 41.01 | 31.59 | 34.42 | 44.65 | 17.12 | 18.02 | 111.50 |
| S48 | Basketball-13 | 82.44 | 23.16 | 23.92 | 13.39 | 24.96 | 11.03 | 33.61 | 79.18 |
| S49 | Basketball-14 | 35.13 | 37.84 | 36.42 | 40.83 | 41.21 | 41.69 | 90.93 | 239.79 |
| S50 | Basketball-15 | 63.17 | 46.05 | 31.07 | 33.61 | 42.22 | 25.52 | 16.83 | 100.38 |
| S51 | Basketball-16 | 63.17 | 46.05 | 31.07 | 33.61 | 42.22 | 25.52 | 16.83 | 100.38 |
| S52 | Basketball-17 | 23.00 | 15.48 | 37.93 | 9.93 | 48.83 | 45.22 | 93.55 | 20.84 |
| S53 | Basketball-18 | 11.50 | 24.98 | 13.29 | 21.61 | 24.51 | 39.59 | 23.38 | 69.25 |
| S54 | Basketball-19 | 31.38 | 22.62 | 16.36 | 12.34 | 51.50 | 20.74 | 16.45 | 139.86 |
| S55 | Basketball-20 | 24.47 | 31.42 | 48.55 | 49.54 | 34.55 | 54.21 | 61.13 | 232.00 |

LG = Lateral Gastrocnemius; MH = Medial Hamstrings; LH = Lateral Hamstring; LQ = Lateral Quadriceps; $_{PRE}$ = 150 ms before ground contact; $_{POST}$ = 50 ms after ground contact

APPENDIX I: RAW KINETIC AND KINEMATIC DATA

| # | Code | $KFA_{IN}$ ° | $KFA_{PK}$° | $KFA_{CHANGE}$ ° | Normalized Moment @ $KFA_{IN}$ Nm/kg | Normalized Moment @ $KFA_{PK}$ Nm/kg | Change in Normalized Moment Nm/kg | Stiffness Nm/kg° |
|---|---|---|---|---|---|---|---|---|
| S01 | Dance-01 | 21.60 | 83.51 | 61.91 | 0.02 | 0.07 | 0.50 | 0.0080 |
| S02 | Dance-02 | 16.40 | 98.69 | 82.28 | 0.09 | 0.06 | -0.26 | -0.0031 |
| S03 | Dance-03 | 23.24 | 101.26 | 78.02 | 0.00 | 0.22 | 2.11 | 0.0270 |
| S04 | Dance-04 | 17.68 | 78.65 | 60.97 | 0.01 | 0.14 | 1.32 | 0.0217 |
| S05 | Dance-05 | 19.99 | 95.13 | 75.14 | 0.08 | 0.08 | -0.07 | -0.0010 |
| S06 | Dance-06 | 18.69 | 89.82 | 71.13 | -0.01 | 0.12 | 1.30 | 0.0182 |
| S07 | Dance-07 | 16.23 | 103.83 | 87.60 | 0.00 | 0.10 | 0.96 | 0.0110 |
| S08 | Dance-08 | 16.77 | 103.54 | 86.77 | 0.05 | 0.09 | 0.42 | 0.0048 |
| S09 | Dance-09 | 24.28 | 128.36 | 104.08 | 0.05 | 0.22 | 1.68 | 0.0162 |
| S10 | Dance-10 | 14.57 | 91.46 | 76.89 | 0.02 | 0.17 | 1.48 | 0.0192 |
| S11 | Dance-11 | 29.91 | 92.62 | 62.71 | -0.01 | 0.16 | 1.65 | 0.0263 |
| S12 | Dance-12 | 21.47 | 79.58 | 58.11 | -0.03 | 0.10 | 1.27 | 0.0219 |
| S13 | Dance-13 | 8.50 | 85.54 | 77.04 | -0.01 | 0.12 | 1.26 | 0.0163 |
| S14 | Dance-14 | 19.04 | 71.70 | 52.65 | 0.01 | 0.16 | 1.46 | 0.0278 |
| S15 | Dance-15 | 19.65 | 71.58 | 51.93 | 0.01 | 0.17 | 1.47 | 0.0284 |
| S16 | Dance-16 | 20.93 | 80.62 | 59.69 | -0.02 | 0.15 | 1.72 | 0.0289 |
| S17 | Dance-17 | 23.47 | 70.39 | 46.92 | 0.03 | 0.16 | 1.34 | 0.0285 |
| S18 | Dance-18 | 5.33 | 90.25 | 84.92 | -0.03 | 0.04 | 0.71 | 0.0084 |
| S19 | Dance-19 | 7.15 | 93.39 | 86.24 | 0.02 | 0.10 | 0.85 | 0.0099 |
| S20 | Dance-20 | 12.39 | 79.44 | 67.05 | -0.03 | 0.12 | 1.46 | 0.0218 |

KFA = Knee Flexion Angle; $_{IN}$ = at Ground Contact; $_{PK}$ = at Maximum Knee Flexion
$KFA_{CHANGE}$ = Change in Knee Flexion Angle from Ground Contact to Maximum Knee Flexion

| # | Code | $KFA_{IN}$ ° | $KFA_{PK}$° | $KFA_{CHANGE}$ ° | Normalized Moment @ $KFA_{IN}$ Nm/kg | Normalized Moment @ $KFA_{PK}$ Nm/kg | Change in Normalized Moment Nm/kg | Stiffness Nm/kg° |
|---|---|---|---|---|---|---|---|---|
| S21 | Dance-21 | 13.43 | 87.10 | 73.68 | 0.04 | 0.14 | 0.96 | 0.0130 |
| S22 | Dance-22 | 22.15 | 98.10 | 75.95 | 0.04 | 0.16 | 1.18 | 0.0155 |
| S23 | Dance-23 | 17.67 | 82.19 | 64.51 | -0.01 | 0.06 | 0.72 | 0.0111 |
| S24 | Dance-24 | 10.43 | 82.33 | 71.90 | 0.03 | 0.19 | 1.61 | 0.0224 |
| S25 | Dance-25 | 10.82 | 87.51 | 76.69 | 0.01 | 0.09 | 0.75 | 0.0098 |
| S26 | Dance-26 | 14.33 | 93.38 | 79.04 | 0.02 | 0.12 | 1.05 | 0.0133 |
| S27 | Dance-27 | 14.12 | 81.14 | 67.03 | 0.00 | 0.13 | 1.25 | 0.0187 |
| S28 | Dance-28 | 11.88 | 84.20 | 72.32 | 0.08 | 0.13 | 0.52 | 0.0072 |
| S29 | Dance-29 | 3.85 | 91.13 | 87.28 | -0.03 | 0.24 | 2.62 | 0.0301 |
| S30 | Dance-30 | 13.49 | 88.50 | 75.01 | 0.10 | 0.12 | 0.17 | 0.0023 |
| S31 | Dance-31 | 13.86 | 95.12 | 81.27 | 0.03 | 0.10 | 0.64 | 0.0079 |
| S32 | Dance-32 | 13.05 | 80.58 | 67.53 | -0.02 | 0.17 | 1.91 | 0.0282 |
| S33 | Dance-33 | 11.27 | 74.56 | 63.30 | -0.03 | 0.08 | 1.10 | 0.0174 |
| S34 | Dance-34 | 14.08 | 96.92 | 82.84 | 0.01 | 0.12 | 1.06 | 0.0128 |
| S35 | Dance-35 | 17.45 | 88.47 | 71.02 | -0.01 | 0.15 | 1.65 | 0.0232 |
| S36 | Basketball-01 | 16.09 | 79.62 | 63.53 | 0.04 | 0.22 | 1.81 | 0.0285 |
| S37 | Basketball-02 | 16.32 | 98.68 | 82.36 | 0.07 | 0.05 | -0.21 | -0.0026 |
| S38 | Basketball-03 | 22.20 | 94.42 | 72.22 | -0.01 | 0.14 | 1.84 | 0.0254 |
| S39 | Basketball-04 | 17.70 | 104.45 | 86.74 | -0.01 | 0.12 | 1.24 | 0.0143 |
| S40 | Basketball-05 | 17.16 | 88.27 | 71.12 | 0.04 | 0.09 | 0.46 | 0.0064 |

KFA = Knee Flexion Angle; $_{IN}$ = at Ground Contact; $_{PK}$ = at Maximum Knee Flexion
$KFA_{CHANGE}$ = Change in Knee Flexion Angle from Ground Contact to Maximum Knee Flexion

| # | Code | $KFA_{IN}$ ° | $KFA_{PK}$° | $KFA_{CHANGE}$ ° | Normalized Moment @ $KFA_{IN}$ Nm/kg | Normalized Moment @ $KFA_{PK}$ Nm/kg | Change in Normalized Moment Nm/kg | Stiffness Nm/kg° |
|---|---|---|---|---|---|---|---|---|
| S41 | Basketball-06 | 19.62 | 88.80 | 69.18 | 0.03 | 0.17 | 1.42 | 0.0206 |
| S42 | Basketball-07 | 11.28 | 81.42 | 70.14 | -0.02 | 0.16 | 1.74 | 0.0248 |
| S43 | Basketball-08 | 18.23 | 91.20 | 72.97 | 0.01 | 0.05 | 0.38 | 0.0052 |
| S44 | Basketball-09 | 14.61 | 59.95 | 45.35 | -0.03 | 0.12 | 1.41 | 0.0311 |
| S45 | Basketball-10 | 10.88 | 81.49 | 70.61 | 0.09 | 0.15 | 0.58 | 0.0082 |
| S46 | Basketball-11 | 32.44 | 80.25 | 47.81 | 0.08 | 0.12 | 0.39 | 0.0083 |
| S47 | Basketball-12 | 12.17 | 75.07 | 62.90 | 0.00 | 0.19 | 1.93 | 0.0307 |
| S48 | Basketball-13 | 7.98 | 86.92 | 78.93 | -0.01 | 0.13 | 1.38 | 0.0175 |
| S49 | Basketball-14 | 26.85 | 95.63 | 68.77 | -0.03 | 0.10 | 1.25 | 0.0182 |
| S50 | Basketball-15 | 12.92 | 94.58 | 81.66 | 0.01 | 0.20 | 1.88 | 0.0231 |
| S51 | Basketball-16 | 16.46 | 72.35 | 55.89 | 0.03 | 0.16 | 1.28 | 0.0229 |
| S52 | Basketball-17 | 19.43 | 66.70 | 47.27 | 0.01 | 0.24 | 2.25 | 0.0475 |
| S53 | Basketball-18 | 21.91 | 93.30 | 71.39 | 0.10 | 0.14 | 0.38 | 0.0053 |
| S54 | Basketball-19 | 18.15 | 84.43 | 66.29 | 0.02 | 0.14 | 1.21 | 0.0183 |
| S55 | Basketball-20 | 13.21 | 68.33 | 55.13 | 0.07 | 0.17 | 0.99 | 0.0179 |

KFA = Knee Flexion Angle; $_{IN}$ = at Ground Contact; $_{PK}$ = at Maximum Knee Flexion
$KFA_{CHANGE}$ = Change in Knee Flexion Angle from Ground Contact to Maximum Knee Flexion

APPENDIX J: MEAN SQUARES COMPARING DAY-TO-DAY VALUES FOR THE DEPENDENT VARIABLES

| | TMS | EMS | BMS |
|---|---|---|---|
| $LG_{PRE}$ | 13.63 | 163.73 | 649.39 |
| $MH_{PRE}$ | 2.24 | 42.03 | 183.74 |
| $LH_{PRE}$ | .20 | 22.01 | 118.82 |
| $LQ_{PRE}$ | 95.78 | 81.79 | 443.33 |
| $LG_{POST}$ | 68.17 | 4.49 | 148.43 |
| $MH_{POST}$ | 112.80 | 4.07 | 94.94 |
| $LH_{POST}$ | 1183.97 | 173.38 | 679.88 |
| $LQ_{POST}$ | 3962.48 | 1129.15 | 4881.10 |
| $LG_{ON}$ | 162.45 | 1071.01 | 4205.36 |
| $MH_{ON}$ | 198.05 | 1856.16 | 485.03 |
| $LH_{ON}$ | 259.20 | 1102.98 | 273.11 |
| $LQ_{ON}$ | 80.00 | 3764.11 | 8972.69 |
| $KFA_{CHANGE}$ | 2.83 | 33.38 | 194.42 |
| $MOM_{CHANGE}$ | .12 | .09 | .58 |
| STIFFNESS | .00 | .00 | .00 |

LG = Lateral Gastrocnemius; MH = Medial Hamstrings; LH = Lateral Hamstring; LQ = Lateral Quadricep; $_{PRE}$ = 150 ms before ground contact; $_{POST}$ = 50 ms after ground contact; $KFA_{CHANGE}$ = Change in Knee Flexion Angle from Ground Contact to Maximum Knee Flexion; $MOM_{CHANGE}$ = Change in Sagittal Knee Net Moment from Ground Contact to Maximum Knee Flexion

APPENDIX K: 2 X 4 REPEATED MEASURES ANOVA COMPARING GROUPS (BASKETBALL, DANCE) ON MUSCLE ACTIVATION ONSETS (LQ, MH, LH, LG)

| Source | | SS | df | MS | F | Sig. | ES | 1 – βa |
|---|---|---|---|---|---|---|---|---|
| Between | | | | | | | | |
| | Group | 7032.91 | 1 | 7032.91 | 1.56 | .22 | 0.03 | .23 |
| | Error (Group) | 238528.44 | 53 | 4500.54 | | | | |
| Within | | | | | | | | |
| | Onset | 93483.83 | 2.84 | 32938.44 | 14.08 | .00 | 0.21 | 1.00 |
| | Onset * Group | 4037.97 | 2.84 | 1422.75 | .61 | .60 | 0.01 | .17 |
| | Error (Onset) | 351862.39 | 150.42 | 2339.18 | | | | |

a Computed using alpha = .05

APPENDIX L: 2 X 4 ANOVA COMPARING GROUPS (BASKETBALL, DANCE) ON PRELANDING MUSCLE ACTIVATION AMPLITUDES (LQ, MH, LH, LG)

| Source | | SS | df | MS | F | Sig. | ES | $1 - \beta^a$ |
|---|---|---|---|---|---|---|---|---|
| Between | | | | | | | | |
| | Group | 97.99 | 1 | 97.99 | .28 | .60 | .01 | .08 |
| | Error (Group) | 238528.44 | 53 | 4500.54 | | | | |
| Within | | | | | | | | |
| | PRE | 12950.4 | 2.5 | 5272.4 | 35.20 | .000 | 0.40 | 1.00 |
| | PRE*Group | 749.7 | 2.5 | 305.2 | 2.04 | .12 | 0.03 | .46 |
| | Error (Onset) | 19500.2 | 130.2 | 367.9 | | | | |

a Computed using alpha = .05

PRE = Prelanding Amplitudes

APPENDIX M: 2 X 4 ANOVA COMPARING GROUPS (BASKETBALL, DANCE) ON POSTLANDING MUSCLE ACTIVATION AMPLITUDES (LQ, MH, LH, LG)

| Source | | SS | df | MS | F | Sig. | ES | $1 - \beta^a$ |
|---|---|---|---|---|---|---|---|---|
| Between | | | | | | | | |
| | Group | 104.63 | 1 | 104.631 | .07 | .787 | .07 | .06 |
| | Error (Group) | 238528.44 | 53 | 4500.54 | | | | |
| Within | | | | | | | | |
| | POST | 164638.74 | 2.30 | 71730.99 | 52.13 | .000 | .50 | 1.00 |
| | POST*Group | 7882.30 | 2.30 | 3434.22 | 2.50 | .08 | .05 | .53 |
| | Error (Onset) | 167402.88 | 121.65 | 1376.14 | | | | |

a Computed using alpha = .05

POST = Postlanding Amplitudes

APPENDIX N: ONE-WAY ANOVA COMPARING GROUPS (BASKETBALL, DANCE) ON KNEE JOINT STIFFNESS

| Source | SS | df | MS | F | Sig. | ES | 1 – βa |
|---|---|---|---|---|---|---|---|
| Group | .0006 | 1 | .00 | .61 | .44 | .01 | .12 |
| Error | .005 | 53 | .00 | | | | |
| Total | .005 | 54 | | | | | |

a Computed using alpha = .05
b R2= .01 (Adjusted R2 = -.00)

APPENDIX O: ANOVA, COEFFICIENT TABLES, AND EXCLUDED VARIABLES FOR PREDICTING KNEE JOINT STIFFNESS

ANOVA

| Model | | SS | df | MS | F | Sig. |
|---|---|---|---|---|---|---|
| 1.00 | Regression | .000 | 1 | .00 | 1.76 | .191 |
| | Residual | .005 | 53 | .00 | | |
| | Total | .005 | 54 | | | |
| 2.00 | Regression | .000 | 2 | .00 | 1.33 | .274 |
| | Residual | .005 | 52 | .00 | | |
| | Total | .005 | 54 | | | |
| 3.00 | Regression | .000 | 3 | .00 | 1.13 | .347 |
| | Residual | .005 | 51 | .00 | | |
| | Total | .005 | 54 | | | |

a Predictors: (Constant), LGPRE
b Predictors: (Constant), LGPRE, LQPRE
c Predictors: (Constant), LGPRE, LQPRE, MHPRE
d Dependent Variable: STIFFNESS

Coefficient Tables

| Model | | Unstandardized Coefficients | | Standardized Coefficients | t | Sig. |
|---|---|---|---|---|---|---|
| | | B | Std. Error | Beta | | |
| 1.00 | (Constant) | .01 | .00 | | 4.22 | .000 |
| | LGPRE | .00 | .00 | .18 | 1.33 | .191 |
| 2.00 | (Constant) | .01 | .00 | | 3.34 | .002 |
| | LGPRE | .00 | .00 | .14 | .98 | .334 |
| | LQPRE | .00 | .00 | .13 | .95 | .346 |
| 3.00 | (Constant) | .01 | .00 | | 3.37 | .001 |
| | LGPRE | .00 | .00 | .19 | 1.24 | .220 |
| | LQPRE | .00 | .00 | .16 | 1.08 | .285 |
| | MHPRE | .00 | .00 | -.13 | -.86 | .395 |
| a | Dependent Variable: STIFFNESS | | | | | |

Excluded Variables

| Model | | Beta In | t | Sig. | Partial Correlation | Collinearity Statistics |
|---|---|---|---|---|---|---|
| | | | | | | Tolerance |
| 1.00 | GROUP | .11a | .83 | .41 | .11 | 1.00 |
| | MHPRE | -.10a | -.68 | .50 | -.09 | .78 |
| | LHPRE | .09a | .66 | .51 | .09 | .96 |
| | LQPRE | .13b | .95 | .35 | .13 | .91 |
| 2.00 | GROUP | .10b | .73 | .47 | .10 | .99 |
| | MHPRE | -.13b | -.86 | .40 | -.12 | .76 |
| | LHPRE | .06c | .39 | .70 | .05 | .87 |
| 3.00 | GROUP | .07c | .47 | .64 | .07 | .88 |
| | LHPRE | .11c | .69 | .49 | .10 | .79 |
| a | Predictors in the Model: (Constant), LGPRE | | | | | |
| b | Predictors in the Model: (Constant), LGPRE, LQPRE | | | | | |
| c | Predictors in the Model: (Constant), LGPRE, LQPRE, MHPRE | | | | | |
| d | Dependent Variable: STIFFNESS | | | | | |

GROUP = Group Membership; LG = Lateral Gastrocnemius; MH = Medial Hamstrings; LH = Lateral Hamstring; LQ = Lateral Quadricep; PRE = Prelanding Muscle Activation Amplitude