PYE, MICHELE LYNETT, Ph.D. Comparison of Neuromuscular Control Strategies between Collegiate Female Dancers and Athletes (2014)
Directed by Dr. Sandra J. Shultz. 206 pp.

High risk neuromuscular control strategies during landing and cutting maneuvers are thought to be a major contributing factor to the 6 times greater risk of ACL injury in female athletes compared to male athletes. However, female dancers who have similar fitness capabilities and perform many of the same cutting and landing tasks as female athletes are less likely to display high risk neuromuscular strategies, have similar neuromuscular control strategies as male dancers, and are 3- 5 times less likely to suffer an ACL injury compared to female athletes. While multiple theories have been proposed to explain this protection in female dancers, preliminary research suggests female dancers may adopt a more protective neuromuscular control strategies as a result of their training practices. Therefore, the primary objective of this study was to comprehensively compare neuromuscular control strategies in collegiate female dancers and collegiate female field athletes to determine if female dancers demonstrate more protective neuromuscular control patterns during functional tasks as characterized by 1) decreased vertical ground reaction forces (vGRF), 2) quicker stabilization of the anterior-posterior and medial-lateral ground reaction force, 3) smaller distance between center of mass (COM) and location of center of pressure (COP), 4) decreased knee valgus, 5) increased ankle plantar flexion, 6) decreased knee extensor moment, and 7) quicker muscular activation. Forty collegiate females, 20 dancers (age= 20.4 ± 1.9 yrs, height= 164.8 ± 6.1 cm, weight= 63.5 ± 8.8kg, experience= 14.3 ± 3.9 yrs) and 20 athletes (age= 19.4± .9 yrs, height= 169.3 ±

7.1 cm, weight= 69.8 ± 13.0 kg, experience= 12.2 ± 2.9 yrs) matched on year of experience were measured for postural control during a dynamic forward hop stabilization task; hip, knee and ankle joint neuromechanics during a planned double leg drop landing; and reflex response characteristics during an unplanned lower extremity perturbation. Results revealed no significant differences between female athletes and dancers on muscle reflex time following a functional perturbation or in their time to stabilization during the dynamic balance test. During the drop jump landing, dancers versus athletes landed with lower vGRF [F (3, 33) = 3.44, p = .03, ES = .24], position their COM more anteriorly [F (1,38) = 4.8, p=.03], moved through a greater sagittal plane ROM [F (3, 36) = 4.6, p=.008] primarily driven by greater ankle joint excursion, and move through equal frontal plane motion at the hip and knee [F (2, 37) = 1.6 p=.23, Partial Eta Squared (η_p^2)=.08]. The greater sagittal plane excursions values were largely a product of a more extended posture at ground contact and did not result in larger peak values. These findings suggest that dancers and athletes may have similar abilities to respond to postural perturbations, but that female dancers may demonstrate some elements of more protective neuromuscular control strategies during planned movements as a result of their training practices. Investigation of dance training may assist in the development of more protective strategies in dancers and inform our future prevention efforts in female athletics.

COMPARISON OF NEUROMUSCULAR CONTROL STRATEGIES BETWEEN COLLEGIATE FEMALE DANCERS AND ATHLETES

by

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

Greensboro
2014

Approved by

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Committee Chair

To Mom and Dad:

I am who you made me to be ~ Determined
I am who you thought I was ~ Hard-Headed
I am who you told me I was ~ Capable
I am, because you love me
Thank you

To John and Robyn:

Thank you for being my God-Given best friends and understanding my internal struggle with stress throughout this process and only ever wanting to keep me focused on what you knew I could accomplish.

To Ashley and Tori:

Your sacrifice did not go unnoticed. Well this is what I was working on…but I hope this also serves to show you both that with sacrifice anything you want can be accomplished, and I will always be there to help you achieve whatever you dream of.

APPROVAL PAGE

This dissertation written by Michele Lynett Pye has been approved by the following committee if the Faculty of The Graduate School at The University of North Carolina at Greensboro

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ACKNOWLEDGEMENTS

Dr. Sandra J. Shultz, you have been a very influential figure in my academic as well as personal life. Thank you for your patience, determination, and faith in my ability.

My Committee, thank you for the opportunity to learn from a team of successful professionals; furthermore, thank you for always being willing to meet, offer advice when I was unsure of what direction to go.

My Lab Colleagues, to all lab members, current and past, thank you for your assistance, thought provoking discussions, and all the laughs over the last 4 years.

Funding for this project provided by the National Athletic Trainers' Association Research & Education Foundation – Doctoral Research Grant

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CHAPTER I

INTRODUCTION

Anterior cruciate ligament injuries are estimated to affect 100,000 individuals a year in the United States alone (Prodromos, Han et al. 2007). Due to reconstructive surgery, extended rehabilitation, and increased risk of secondary injuries, an ACL tear remains one of the most costly injuries to treat (Lohmander, Ostenberg et al. 2004). The majority of these ACL injuries occur during non-contact mechanisms (Boden, Dean et al. 2000), which occur in the absence of physical contact with another individual or object at time of injury (Walden, Hagglund et al. 2011). Deceleration movements, such as landing from a jump or changing directions, are common examples of non-contact injury mechanisms (Ferretti, Papandrea et al. 1992; Boden, Dean et al. 2000; Olsen, Myklebust et al. 2003; Fauno and Jakobsen 2006), where females are up to 6 times more likely to suffer an ACL injury than males (Arendt and Dick 1995; Deitch, Starkey et al. 2006).

Although multiple sex specific risk factors (i.e., hormones, structural alignment, body composition, training, etc.) are proposed to contribute to this increased risk in females, many experts believe that sex differences in neuromuscular control patterns is a main contributor to the greater risk in females (Griffin, Agel et al. 2000; Agel, Arendt et al. 2005; Hootman, Dick et al. 2007; Renstrom, Ljungqvist et al. 2008). During deceleration movements, females typically display a “stiff” landing strategy (Boden, Dean et al. 2000), that coincides with larger vertical ground reaction forces and smaller knee and hip flexion angles (Devita and Skelly 1992; Decker, Torry et al. 2003; Ford, Myer et al. 2003; Kernozek, Torry et al. 2005; Pappas, Hagins et al. 2007; Kernozek, Torry et al. 2008), which has been associated with larger quadriceps forces and peak knee extensor moments (McNittgray 1993; Yu, Lin et al. 2006; Blackburn and Padua 2009) that are known to strain the ACL (Berns, Hull et al. 1992). Females are also more likely to demonstrate a “dynamic knee valgus” position, identified as increased hip adduction and internal rotation, knee valgus, and tibial rotation (Quatman and Hewett 2009). The combination of frontal and or transverse plane knee motion with a shallow knee flexion angle has been shown to place the greatest amount of strain on the ACL (Berns, Hull et al. 1992; Woo, Fox et al. 1998; Meyer and Haut 2008), inferring that this neuromuscular control profile of female athletes places the lower extremity in a position of greatest risk of injury. While prospective research has yet to confirm whether these higher risk neuromuscular profiles commonly observed in females are actually predictive of their greater risk for ACL injury, improving these motion patterns are the primary focus of current ACL prevention efforts, which have yet to result in a significant reduction in overall injury risk in females (Arendt and Dick 1995; Agel, Arendt et al. 2005; Hootman, Dick et al. 2007; Benjaminse and Otten 2011).

Female dancers who require similar fitness capabilities and who perform numerous deceleration movements during activity are 3- 5 times less likely to injure their ACL compared to female field athletes (soccer, basketball) (Liederbach, Dilgen et al.

2008; Meuffels and Verhaar 2008). The prevailing theories as to why dancers display a lower risk of ACL injury include: rehearsed choreography (Orishimo, Kremenic et al. 2009), controlled toe to heel landing techniques (McNitt-Gray, Koff et al. 1992; Liederbach, Dilgen et al. 2008; Orishimo, Kremenic et al. 2009), a more neutral alignment during jumping tasks (Liederbach, Dilgen et al. 2008; Ambegaonkar, Shultz et al. 2009; Orishimo, Kremenic et al. 2009), improved postural control ability (Liederbach, Dilgen et al. 2008; Orishimo, Kremenic et al. 2009; Ambegaonkar, Caswell et al. 2013), and years of training (Orishimo, Kremenic et al. 2009). These theories largely center on training practices, as dance training includes many of the same balance, stretching, plyometric, agility, landing and strengthening exercises as ACL injury prevention programs. These activities are directly incorporated into their daily training over many years and are subject to continual visual and augmented feedback (Liederbach, Dilgen et al. 2008; Orishimo, Kremenic et al. 2009). However, few investigations have characterized the neuromuscular control pattern in female dancers, or compared the neuromuscular control patterns to other physically active populations to determine if dancers indeed develop more protective strategies that lower their injury risk. Among these investigations, female dancers are more stable during a single leg balance task (Crotts, Thompson et al. 1996; Simmons 2005; Gerbino, Griffin et al. 2007; Ambegaonkar, Caswell et al. 2013), and have shorter reflex times following an unanticipated perturbation (Simmons 2005). Further, no sex differences in landing mechanics have been identified in male and female dancers who both perform “soft” landings in a neutral position (Orishimo, Kremenic et al. 2009; Orishimo, Liederbach et al. 2014). However, this research is largely limited to comparisons to non-athletic populations (McNitt-Gray, Koff et al. 1992; Simmons 2005; Simmons 2005) or recreational athletes with less training experience (Crotts, Thompson et al. 1996; Ambegaonkar, Caswell et al. 2013), which alone may explain the superior findings in dancers. Comprehensive comparisons of neuromuscular profiles of female athletes (at high risk for ACL injury) and dancers (at low risk for ACL injury) from similar training intensity and experience backgrounds are needed to better understand the potential protective strategies that dancers may utilize to lower their risk. If differences do exist in these populations, this will pave the way for investigators to focus future research on understanding the specific training practices that promote and retain safe movement patterns in dancers, and develop more effective intervention strategies to lower female athletes risk for ACL injury

Statement of Problem

Despite wide implementation of neuromuscular training programs, epidemiology studies show no reduction in ACL injury incidence or the gender disparity in female athletes over the last 15-20 years (Arendt and Dick 1995; Agel, Arendt et al. 2005; Hootman, Dick et al. 2007). This finding suggests that protective landing mechanics developed during neuromuscular training programs are not retained or transferred to sport (Benjaminse and Otten 2011). Dance represents a population that performs the same type of landing and cutting maneuvers associated with ACL injuries in female athletes (Orishimo, Kremenic et al. 2009), yet report 3 to 5 times lower risk of ACL injury compared to female athletes and no difference in injury rate to male dancers (Liederbach, Dilgen et al. 2008; Meuffels and Verhaar 2008). Dance practice includes neuromuscular training in combination with visual augmented feedback, a technique used to increase the retention of complex multi limb movements (Sigrist, Schellenberg et al. 2011), during activity specific tasks. While this training is theorized to result in more effective transference and retention of these movement patterns to actual skilled movements, research to date has not comprehensively compared neuromuscular control strategies in (landing mechanics, postural control and muscular activation) in female dancers and female athletes of similar training intensity and experiences to determine if females dancers do in fact demonstrate more protective neuromuscular control strategies than female athletes. These protective neuromuscular control strategies are considered to be 1) decreased vertical ground reaction forces, 2) quicker stabilization of the anterior-posterior and medial-lateral ground reaction force, 3) anteriorly positioned center of mass 4) decreased knee valgus, 5) increased ankle plantar flexion, 6) decreased knee extensor moment, and 7) quicker hamstring muscular activation.

Objective and Hypothesis

The primary objective of this study was to characterize and compare the neuromuscular control profiles of female dancers who are at low risk for ACL injury with female athletes at high risk for ACL injury. Our approach was to assess neuromuscular patterns during a dynamic balance task (postural control), an anticipated task (double leg drop jump) and an unanticipated task (lower extremity perturbation). Due to different footwear during dance and athletic activity, the postural control, and double leg drop jump tasks (the tasks requiring a jumping action) were tested in both shod and barefoot conditions. Specifically we tested the following questions:

Question 1: When compared to athletes, do dancers demonstrate more stable postural control, as assessed by time to stabilization (TTS) during a forward hop task?

> *Hypothesis 1*: Dancers will require significantly less time to stabilize the ground reaction force following a hopping task

To test hypothesis 1, a 2 (group) x 2 (plane) x 2 (limb) x 2 (footwear) repeated measures ANOVA was used to assess differences in TTS (dependent variable) in the anterior-posterior (A-P) and medial-lateral (M-L) plane (independent variable – plane) on the dominant and non-dominant limb (independent variable – limb) when shod and barefoot (independent variable – footwear) between dancers and athletes (independent variable – group).

Question 2: Are there neuromuscular control differences between dancers and athletes during a drop jump task?

> *Hypothesis 2a*: Dancers will position their center of mass (COM) closer to the location of the center of pressure (COP) at initial ground contact following a drop jump task compared to athletes.

To test Hypothesis 2a, a 2 (group) x 2 (footwear) repeated measures ANOVA was used to compare the COM to COP displacement in the A-P plane (dependent variable) between dancers and athletes (independent –group) in the shod and barefoot condition (independent –footwear).

> *Hypothesis 2b:* Dancers will land from a drop jump with greater ankle plantar flexion, and similar hip and knee flexion compared to athletes

To test hypothesis 2b, two separate multivariate ANOVA's were used to assess differences in hip, knee, and ankle kinematics (dependent variables = ankle plantar flexion, knee flexion, and hip flexion) between dancers and athletes (independent variable) during shod and barefoot conditions (independent variable), where kinematics were measured at initial ground contact and for total joint excursions (initial ground contact to peak center of mass displacement).

> *Hypothesis 2c:* Dancers will land from a drop jump with less frontal plane hip and knee motion compared to athletes.

To test hypothesis 2c, two separate multivariate ANOVA were used to assess differences in frontal plane hip and knee kinematics (dependent variables = knee valgus and hip adduction) between dancers and athletes (independent variable) during shod and barefoot conditions (independent variable), where kinematics were measured at initial ground contact and for total joint excursions (initial ground contact to peak center of mass displacement).

> *Hypothesis 2d:* Dancers will demonstrate lower vGRF values and peak knee extensor moments compared to athletes.

To test hypothesis 2d, a repeated measures ANOVA was used to assess differences in vGRF between dancers and athletes (independent variable) during shod and barefoot conditions (independent variable). A separate multivariate ANOVA was used to assess differences in and hip, knee, and ankle peak extensor moments (dependent variables) between dancers and athletes (independent variable) during shod and barefoot conditions (independent variable).

Hypothesis 2e: Dancers versus athletes will absorb a larger relative amount of total energy at the ankle joint compared to the knee joint.

To test hypothesis 2e, a multivariate ANOVA was used to assess differences in relative energy absorption across the hip, knee and ankle (dependent variables) between dancers and athletes (independent variable) during shod and barefoot conditions (independent variable).

Hypothesis 2f: Dancers will demonstrate higher hamstring amplitude prior to ground contact during a drop jump task compared to athletes.

To test hypothesis 2f, a 2 (group) x 2 (footwear) x 6 (muscle) ANOVA was used to assess differences in pre-landing activation amplitude (dependent variable) during the double leg drop jump task between dancers and athletes (independent variable – group) during shod and barefoot conditions (independent variable – footwear) across the medial and lateral gastrocnemius, quadriceps, and hamstring muscles (independent variable – muscle).

Question 3: Is there a difference in reflexive muscular activation between dancers and athletes following an unanticipated lower extremity perturbation?

<u>Hypothesis 3:</u> Dancers will activate musculature significantly quicker than athletes

To test hypothesis 3, a 2 (group) x 6 (muscle) x 2 (perturbation direction) ANOVA was used to assess differences in muscular onset time (dependent variable) between dancers and athletes (independent variable) during internal and external cable releases (independent variable – perturbation direction) across the medial and lateral gastrocnemius, quadriceps, and hamstring muscles (independent variable – muscle).

Limitations and Assumptions

1. Results from this dissertation cannot be generalized to populations other than the collegiate female dancers and athletes studied when performing a forward hop stabilization, double leg drop jump and lower extremity perturbation task.
2. The Phase Space IMPULSE motion analysis system and Bertec Force platforms are valid and reliable devices for kinematic and kinetic measurements, respectively.
3. Electromyography analysis by surface electrode with the Delsys system is a valid and reliable device for the assessment of muscular activation timing and amplitude.
4. The muscular activity obtained at each muscle site is representative of the total muscle activity.
5. Inverse dynamics calculations represent the total moments occurring at the joint.
6. Participants exerted maximal effort during all testing procedures.

7. This work did not account for anatomical and hormonal risk factors that are potentially associated with high-risk knee joint neuromechanics.
8. The representation of the foot, shank, and thigh as a rigid segment are accurate depictions of the motion occurring in the lower extremity during athletic movements.

Delimitations

1. Participants were limited to females who have a minimum of 5 years of experience in dance (ballet, modern, or contemporary styles) or field sports (soccer, basketball, volleyball, rugby, field hockey, lacrosse, or tennis).
2. Participants did not participated in both dance and field sports.
3. Participants were considered healthy as defined by no lower extremity injury or vestibular or balance disorder in the last 6 months.
4. Participants were able to successfully and consistently complete all tasks following familiarization to participate.
5. Kinematic and kinetic data were only obtained from the left leg.
6. All participants wore standardized shoes during the shod condition.

Operational Definitions

Base of Support (BOS): The portion of the foot segment that is in direct contact with the ground.

Baseline Muscle Activity: The mean electromyography (EMG) activity 100ms prior to perturbation.

Center of Pressure (COP): The planar point location of the vertical ground reaction force vector.

Collegiate Dancer (herein dancer): Current participation in a minimum of 120 minutes of a ballet or contemporary dance per week within a University level dance program.

Collegiate Field Athlete (herein athlete): Current participation in a field sport for a minimum of 120 minutes per week within a University's Athletic Department.

Dominant Limb: The self-selected stance leg when kicking a ball for maximum distance.

Field Sport: Soccer, Basketball, Volleyball, Rugby, Tennis, Field Hockey, Lacrosse.

Ground Contact: the first frame when the ground reaction force reaches or exceeds 10 newtons (N).

Healthy: No history of lower extremity injury in the past 6 months. No vestibular or balance disorders, no history of cardiac disease.

Landing Phase: The period between foot contact and peak center of mass (COM) displacement.

Perturbation: An unanticipated disturbance of postural control initiated by a cable release mechanism resulting in an internal or external rotation of the trunk and femur on a weight bearing tibia.

Pre-Landing Phase: 150ms before ground contact.

Range of Variation: the peak variation in the ground reaction force during the final 5 seconds of a trial.

Time to Stabilization (TTS): the point at which an unbound 3rd order polynomial fit to the ground reaction force crosses below the range of variation.

Independent Variables

Activity: Dancers and Field Athletes.

Limb: Dominant and Non-Dominant Limb.

Muscles: Medial and Lateral Gastrocnemius, Quadriceps, and Hamstrings.

Joint: Hip, Knee, and Ankle.

Perturbation Release: Internal and External Rotation.

Shoe Condition: Barefoot, Shod.

Dependent Variables

A-P Time to Stabilization (sec) – time to stabilization of the ground reaction force in the anterior-posterior direction.

Ankle Plantar flexion (°) – flexion angle of the foot segment relative to the tibia at initial ground contact and excursion (peak-initial).

COM to COP displacement (cm) – the anterior-posterior distance between the position of the center of mass relative to the center of pressure.

Energy Absorption (Joules x BW^{-1} x Ht^{-1}) – The integration of the negative portion of the joint power curve (the product of the normalized joint moment and joint angular velocity at each time point), normalized to body weight and height.

Hip Adduction (°) – adduction angle of the femur relative to the pelvis at initial ground contact and excursion (peak-initial).

Hip Flexion (°) – flexion angle of the femur relative to the pelvis at initial ground contact and excursion (peak-initial).

Knee Extensor Moment (Nm/kg) – the angular force which causes a rotation about the knee joint axis calculated as the product of force and moment arm.

Knee Valgus (°) – abduction angle of the tibia relative to the femur at initial ground contact and excursion (peak-initial).

Knee Flexion (°) – flexion angle of the tibia relative to the femur at initial ground contact, peak displacement, and excursion (peak-initial).

Mean Muscular Amplitude (%MVIC) – average EMG activity during pre-landing or post-landing for each muscle that has been normalized to a maximal voluntary isometric contraction for that respective muscle, and averaged over multiple trials.

M-L Time to Stabilization (sec) – time to stabilization of the ground reaction force in the medial-lateral direction.

Muscular Onset Time (ms) – the first frame where the muscular activity is 1 (hamstrings and gastrocnemius) or 2 (quadriceps) standard deviations above baseline muscle activity for 10ms or longer.

Relative Joint Energy Absorption (%) – The percentage of work of the individual joints (hip, knee, and ankle) to the total work produced (hip work + knee work + ankle work).

vGRF (%Bodyweight) – Ground reaction force in the vertical direction divided by body weight.

CHAPTER II

REVIEW OF THE LITERATURE

Introduction

The purpose of this literature review is to support the theoretical framework that the neuromuscular control patterns of collegiate female dancers who are at low risk for anterior cruciate ligament (ACL) injury may differ from that of collegiate female athletes who are at higher risk for ACL injuries. Specifically, collegiate female dancers may demonstrate more protective neuromuscular control patterns during functional movements commonly associated with ACL injuries (i.e., jumping and cutting) thereby reducing their risk of ACL injury compared to collegiate female athletes. To support this theoretical framework, this review will briefly discuss what is currently known about ACL injuries, the neuromuscular control patterns thought to be predictive of increased ACL injury risk, the differences in neuromuscular control strategies demonstrated by athletes and dancers, and how these differences may contribute to the lower risk of ACL injury in the female dancer population.

Anterior Cruciate Ligament Injury

The anterior cruciate ligament is the primary static stabilizer of the knee and injury to this ligament can cause costly short term and long term debilitation (Starkey and Ryan 2002; Lohmander, Ostenberg et al. 2004). The following sections will provide an overview of ACL injury including the structure and function of the ACL, potential mechanisms of injury, and the incidence of ACL injury in relation to sex and type of activity.

Structure and Function of the Anterior Cruciate Ligament

The anterior cruciate ligament originates from the posterior medial aspect of the lateral femoral epicondyle and inserts anterior to the tibial spine and blends with the anterior horn of the medial meniscus (Arnoczky 1983). As a whole, the ACL resists the following motions: 1) anterior translation of the tibia on the femur, 2) internal rotation of the tibia on the femur, and 3) hyperextension of the tibiofemoral joint (Starkey and Ryan 2002). The ACL has two distinct bundles, the anteromedial bundle and the posterolateral bundle, where their directional names specify the insertion site on the tibial plateau (Arnoczky 1983; Woo, Fox et al. 1998). The arrangement of the double bundles allows for different portions of the ACL to be taut throughout the range of motion (Arnoczky 1983; Woo, Fox et al. 1998). Specifically the anteromedial bundle has been found to provide the majority of resistance to an anterior tibial load when the knee is in greater than 45° of flexion (Takai, Woo et al. 1993). When the knee is near full extension both the anterior and posterior portions of the ACL resists anterior tibial loading (Takai, Woo et al. 1993). Understanding the structure and function of the ACL is imperative to understanding what mechanisms result in ligament strain and failure, which is discussed in the following section.

Mechanism of Anterior Cruciate Ligament Injury

Approximately 100,000 anterior cruciate ligament injuries occur each year within the United States of America (Griffin, Agel et al. 2000; Huston, Greenfield et al. 2000). Early research documenting injury incidence reported that 70%-83% of all ACL injuries were due to non-contact mechanisms (Chick and Jackson 1978; McNair, Marshall et al. 1990; Boden, Dean et al. 2000). Non-contact ACL injury has been defined as an injury to the ACL in the absence of any physical contact with another player or object at the time of injury (Walden, Hagglund et al. 2011). While the etiology of a non-contact ACL injury is only partially understood, retrospective interviews, video analysis, as well as in vivo and in vitro force assessments provide plausible maneuvers, joint angles, and loads that can increase strain on the ACL and lead to a non-contact ACL injury.

<u>Retrospective Interview</u>

Retrospective interviews of individual accounts of ACL injury has provided the initial insight into possible mechanisms of an ACL injury (Ferretti, Papandrea et al. 1992; Boden, Dean et al. 2000; Olsen, Myklebust et al. 2003; Fauno and Jakobsen 2006). From these interviews, researchers determined that up to 70% of ACL injuries were noncontact in nature (McNair, Marshall et al. 1990; Ferretti, Papandrea et al. 1992; Boden, Dean et al. 2000; Fauno and Jakobsen 2006). Of these noncontact ACL injuries, the most common movements leading up to an ACL injury were decelerating movements with or without a change in direction (Boden, Dean et al. 2000). These decelerating movements were most often identified as landing from a jump (Ferretti, Papandrea et al. 1992; Boden, Dean et al. 2000; Olsen, Myklebust et al. 2003; Fauno and Jakobsen 2006) or a plant and cut maneuver (Olsen, Myklebust et al. 2003).

While these studies identified the movements that prelude an ACL injury, other retrospective interview studies attempted to identify the joint position that the subject was in when the ACL tear occurred. These interviews revealed that the knee joint was commonly reported to be positioned near full extension or in hyperextension (McNair, Marshall et al. 1990; Boden, Dean et al. 2000), in a valgus position (Boden, Dean et al. 2000), in a valgus position combined with either internal or external tibial rotation (Ferretti, Papandrea et al. 1992), or tibial internal rotation without frontal plane motion (McNair, Marshall et al. 1990). Furthermore, 99% of the injuries occurred while the foot was in contact with the ground (Fauno and Jakobsen 2006). However, retrospective interviews are subject to inaccuracies, as they are dependent on subject memory recall (Krosshaug, Nakamae et al. 2007).

Video Analysis

ACL injuries were subsequently analyzed using video footage of the injury event to eliminate inaccuracies due to memory recall. These video analyses revealed mechanisms of ACL injuries that parallel those reported during retrospective interviews (Olsen, Myklebust et al. 2004; Cochrane, Lloyd et al. 2007; Krosshaug, Nakamae et al. 2007; Boden, Torg et al. 2009; Hewett, Torg et al. 2009; Koga, Nakamae et al. 2010). Inspection of video footage from ACL injuries confirmed that the majority of ACL injuries occurred during a change of direction movement or when landing from a jump (Olsen, Myklebust et al. 2004; Krosshaug, Nakamae et al. 2007). While, Olsen et al (Olsen, Myklebust et al. 2004), noted all ACL injuries that occurred from landing a jump occurred during a single-leg landing maneuver, Krosshaug et al (Krosshaug, Nakamae et al. 2007), approximated that only 43% of ACL injuries that occurred when landing from a jump were single-leg landing maneuvers. Despite the discrepancies on whether ACL injuries more commonly occur during single versus double leg landings, these findings support the results from Fauno et al (Fauno and Jakobsen 2006), which show 99% of ACL injuries occur during ground contact.

Joint positioning observed through video analysis, also coincided with findings from retrospective interviews. During a cutting maneuver that resulted in failure of the ACL, the knee was commonly positioned in approximately 14^0 of valgus with either internal or external rotation of the tibia, and near full extension (approximately 13^0 of flexion) (Olsen, Myklebust et al. 2004). Joint positioning at time of ACL injury was similar when landing from a jump except that the tibia was consistently positioned in external rotation (approximately 10^0) (Olsen, Myklebust et al. 2004).

Inspection of video recordings as used in these studies were later shown to have poor accuracy when determining hip and knee joint angles as compared to 3D motion capture (Krosshaug, Nakamae et al. 2007). For example, hip and knee flexion angles as determined by a 3D motion capture system were 7 and 19 degrees higher than what was estimated through visual inspection of video recordings, respectively (Krosshaug, Nakamae et al. 2007). Thus, modeling of the lower extremity as determined by visual inspection should be taking cautiously.

To improve the reliability of video analysis, Koga et al (Koga, Nakamae et al. 2010), developed a model based imaging matching (MBIM) technique that developed a skeletal model from the video. The skeletal image can then be used to model and measure the joint angles prior to and following ACL injury (Koga, Nakamae et al. 2010). Analysis of ACL injury events with the MBIM technique consistently reported a valgus position of the knee that was commonly combined with internal rotation of the tibia during cutting or single leg landing maneuver 40 milliseconds after initial contact with the ground. However at initial contact, the knee valgus angle was at approximately 0°, while knee rotation was commonly position in 5° of external rotation. The precise moment of ligament failure was unknown; therefore, it is possible this positioning 40 milliseconds after ground contact was the result of an ACL tear, rather than the cause of ligament failure.

In Vitro and *In Vivo* Analysis

In vitro and *in vivo* studies have been conducted to gain a clearer understanding of joint positions that stress the ACL and thereby may increase risk for injury. *In vitro* studies provide insight on the forces endured by the ACL with the added benefit of controlling for knee joint angles at initiation of load acceptance. Although the initial position can be constrained, the load application systems used in these investigations also allowed for natural movement of the joint following acceptance of a load (Berns, Hull et al. 1992).

Woo et al (Woo, Fox et al. 1998) reported that the ACL resists 80% of an anteriorly directed load when the knee is positioned in less than 30° of knee flexion. This can imply that the ACL is the primary restraint to pure anterior translation when the knee is near full extension. The strain on the ACL with only an external flexion moment created approximately 2.76% relative strain on the ACL, however, with the addition of valgus loading, the relative strain on the ACL increased to 3.12% (Withrow, Huston et al. 2006). This finding coincides with other investigations that have noted up to a 30% increase in ACL strain when an external flexion moment is combined with an external valgus moment while the knee positioned in shallow knee flexion (Berns, Hull et al. 1992; Woo, Fox et al. 1998; Withrow, Huston et al. 2006). This suggests that a shallow knee flexion position in combination with knee abduction can place a significantly larger strain on the ACL compared to just landing near full knee extension (Berns, Hull et al. 1992). Transverse plane loads on a nearly extended knee joint further increase the strain on the ACL when combined with valgus loads (Meyer and Haut 2008).

A limitation to *in vitro* research is the inability to replicate the protective forces of muscular contraction that occur during dynamic activity. Although quadriceps and hamstring stiffness were applied to the cadaveric load application systems, the magnitude of resistance provided by the musculature is nearly impossible to replicate. Therefore *in vivo* research, such as the study conducted by Flemings et al (Fleming, Renstrom et al. 2001), allow for a more realistic representation of the loads applied to the ACL. In this study, a strain gauge transducer was surgically implanted into the ACL of subjects, who were positioned within a knee joint loading device that allowed for compressive forces to mimic a weight bearing position. Strain on the ACL was greater during weight-bearing conditions compared to non-weight bearing conditions during low anterior shear loads, with no significant difference in strain at anterior loads greater than 40N (Fleming, Renstrom et al. 2001). In the frontal plane, the strain on the ACL was greater in the weight bearing conditions across 20Nm of varus torque to 15 Nm of valgus torque. While in the transverse plane, the strain on the ACL was greater during weight bearing with low internal rotation and all external torques applied. Although there was no difference between the weight bearing and non-weight bearing conditions when loads were applied in the anterior, or internal rotation directions, this study identified that the ACL is strained in all planes when in a weight bearing position (Fleming, Renstrom et al. 2001). This would suggest that a weight bearing knee subjected to greater out of plane motions and moments (e.g., excessive knee valgus, excessive rotation) would be at greater risk of ACL strain and rupture. This provides the basis for examining lower extremity kinematics and kinetics during functional tasks near the time of ground contact, and comparing these biomechanics between athletes and dancers at high and low risk for ACL injury respectively.

Summary

Anterior cruciate ligament fibers are aligned to protect against anterior translation and internal rotation of the tibia relative to the femur; therefore, it is logical that a rupture of the ACL may be multi-planar phenomenon (Quatman and Hewett 2009). *In vitro* and *in vivo* analysis reported that the strain placed on the ACL is larger when the limb is weight-bearing (Berns, Hull et al. 1992; Fleming, Renstrom et al. 2001; Withrow, Huston et al.

2006); coinciding with retrospective interviews and video analysis that reported ACL injuries typically occur when the foot is in contact with the ground (Woo, Fox et al. 1998; Boden, Dean et al. 2000; Olsen, Myklebust et al. 2004; Fauno and Jakobsen 2006; Krosshaug, Nakamae et al. 2007; Boden, Torg et al. 2009; Koga, Nakamae et al. 2010). When the knee is positioned in less than 30° of knee flexion, contraction of the quadriceps results in an anteriorly directed force of the proximal aspect of the tibia through the patellar tendon (Renstrom, Ljungqvist et al. 2008). Retrospective interviews and video analysis report the majority of ACL injuries occur while in this shallow knee flexion (Boden, Dean et al. 2000; Teitz 2001; Olsen, Myklebust et al. 2004; Krosshaug, Nakamae et al. 2007; Boden, Torg et al. 2009; Koga, Nakamae et al. 2010). Additionally, ACL loads are increased when a shallow knee flexion is combined with knee valgus or tibial rotation (Berns, Hull et al. 1992; Woo, Fox et al. 1998; Meyer and Haut 2008). This is consistent with interviews and video analysis that frequently report an extended and valgus knee position at the time of an ACL injury (Boden, Dean et al. 2000; Olsen, Myklebust et al. 2004; Krosshaug, Nakamae et al. 2007; Boden, Torg et al. 2009; Koga, Nakamae et al. 2010). Although video analysis have reported both internal and external rotation of the tibia at time of injury, *in vitro* investigations suggests that internal tibial rotation in particular increases the strain on the ACL when combined with a valgus and extended knee (Meyer and Haut 2008). This data suggests that when the knee is positioned near full extension and larger amounts of knee valgus and tibial rotation, this places the largest strain on the ACL, thereby increasing the risk for ligament failure.

Throughout the remainder of this review, we will consider this as an "at-risk" positioning of the knee joint (Figure 1).

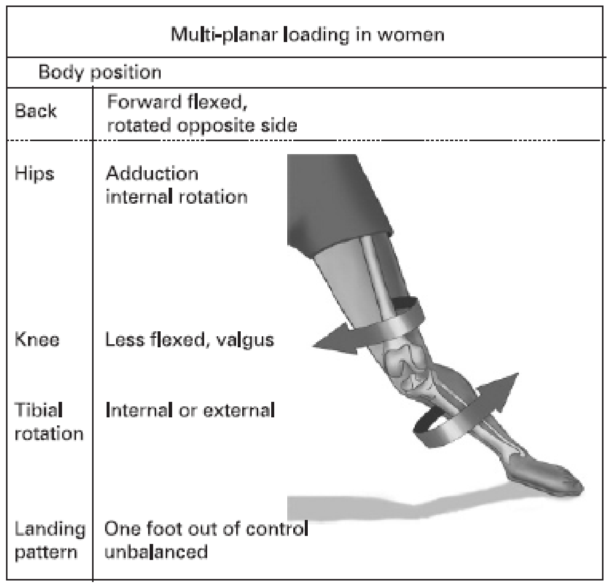

Figure 1. "At-Risk" positioning of the knee joint (Quatman & Hewett 2009)

Occurrence of Anterior Cruciate Ligament Injury

Epidemiology studies have identified that the occurrence of anterior cruciate ligament injuries vary dependent on the type of activity and the individual. Research has also identified different rates of injury between sexes. The following section will review the epidemiology of ACL injury.

<u>Injury Rates by Activity Type</u>

Epidemiology studies indicate that ACL injuries are more common in sports that perform planting/cutting maneuvers or landing from a jump (Arendt and Dick 1995). Specifically, the highest reported ACL injury rate per 1000 athletic exposures occur in soccer (.09-.33) (Arendt and Dick 1995; Arendt, Agel et al. 1999; Agel, Arendt et al. 2005; Hootman, Dick et al. 2007; Prodromos, Han et al. 2007), wrestling (.11-.77) (Hootman, Dick et al. 2007; Prodromos, Han et al. 2007), football (.18-.33) (Hootman, Dick et al. 2007), gymnastics (.33) (Hootman, Dick et al. 2007), basketball (.07-.29) (Arendt and Dick 1995; Agel, Arendt et al. 2005; Hootman, Dick et al. 2007; Prodromos, Han et al. 2007), and lacrosse (.12-.17) (Hootman, Dick et al. 2007). Sports that do not perform these movements as frequently, such as baseball (.02) (Hootman, Dick et al. 2007) and softball (.08) (Hootman, Dick et al. 2007), have much lower ACL injury rates.

It is important to note that the injury rates listed above represent both contact and non-contact ACL injuries; however, it has been shown that 70% of ACL injuries are non-contact in nature (Boden, Dean et al. 2000). Furthermore, sports listed as “contact sports” (i.e. football) do not have a significantly higher rate of ACL injuries, suggesting that non-contact ACL injuries remain more prevalent than contact ACL injuries. The wide range of ACL injury rates for the sports such as soccer and basketball can be attributed to the fact that both males and females participate in these sports, where the injury rates are markedly higher in females.

Sex Rates

Much attention has focused on the sex disparity in ACL injury rates. Data obtained from the National Collegiate Athletic Association from 1988 – 2004 showed male athlete incurred 3,285 ACL injuries while female athletes only suffered 1,515 ACL injuries over the same time period (Hootman, Dick et al. 2007). However if sex specific sports are removed (football, wrestling, gymnastics), the occurrence of ACL injuries for male athletes drops substantially to 600 ACL tears, while incidence of ACL injury in female athletes remains relatively high at 1,381 ACL tears (Hootman, Dick et al. 2007). This suggests that the overall occurrence of ACL injuries is greater in male athletes solely due to a greater number of male athletes participating in sports compared to female athletes. Because of this, ACL injury data is typically reported as injury rate (calculated as the number of ACL injuries per 1000 athlete exposures), thereby controlling for the number of participants. When data is analyzed in this manner, females are at a 3-4 fold greater risk of injury than males (Arendt and Dick 1995; Arendt, Agel et al. 1999). Specifically, non-contact ACL injury rates for female basketball and soccer athletes are reported to be .16 and .13 respectively, as compared to .04 for male basketball and soccer athletes (Agel, Arendt et al. 2005; Moses and Orchard 2012).

Anterior Cruciate Ligament Injury Summary

The anterior cruciate ligament provides multi-planar stability at the knee joint thereby resisting forces on the tibia relative to the femur in the anterior, valgus, and tibial rotation directions (Starkey and Ryan 2002). Although there is yet no clear consensus on the mechanism of injury, retrospective interviews, video analysis and *in-vitro* and *in-vivo* studies of knee load applications previously described suggest that decelerating forces with the knee positioned near full extension with valgus and internal or external rotation of the tibia increases the strain on the ACL (Berns, Hull et al. 1992; Boden, Dean et al. 2000; Olsen, Myklebust et al. 2004; Fauno and Jakobsen 2006; Withrow, Huston et al. 2006; Krosshaug, Nakamae et al. 2007; Boden, Torg et al. 2009; Hewett, Torg et al. 2009; Koga, Nakamae et al. 2010). This positioning of the knee joint during functional activity has been described as a “higher-risk” because the increase strain on the ACL is thought to also increase the likelihood for ligament failure (Quatman and Hewett 2009). This “higher-risk” positioning of the knee is commonly seen during plant-and-cut maneuvers or while landing a jump, and is more frequently demonstrated by female athletes (Boden, Dean et al. 2000; Olsen, Myklebust et al. 2004; Krosshaug, Nakamae et al. 2007; Quatman and Hewett 2009). The fact that these neuromuscular control patterns are more observed in female compared to male athletes has led to the prevailing theory among researchers that this “higher-risk” positioning in females is the likely cause of their greater risk for ACL Injury.

Neuromuscular Control Patterns

The specific reasons or risk factors that explain the sex disparity in ACL injury rates is still unknown. However, there are three main areas that have been investigated that are known to differ considerably between males and females: 1) hormonal, 2) structural alignment, and or 3) neuromuscular control differences (Hewett, Myer et al. 2005). Although there is evidence that the hormonal and alignment differences between sexes may increase risk for ACL injury (Huston, Greenfield et al. 2000; Shultz, Schmitz et al. 2012), these are considered non-modifiable risk factors because they cannot be altered through preventative training. Moreover, hormonal and alignment differences would likely not explain the large disparity in ACL injury between dancers and athletes of the same sex (which will be addressed later in this review). Therefore, the next section will focus on studies that have examined and compared neuromuscular control strategies in male and females athletes' through the assessment of postural control, landing mechanics and muscular activation patterns. This information highlights the neuromuscular differences between sexes that have been observed in the athletic population that to date has not been observed in the dance population. This will set the stage for comparisons between female athletes at high risk for ACL injury and female dancers who are at lower risk of ACL injury.

Neuromuscular Control Patterns

Neuromuscular control can be defined as the conscious and unconscious activation of dynamic restraints in preparation for or in response to a joint motion with the purpose of providing joint stability (Riemann and Lephart 2002). While the musculature surrounding the joint serve as primary dynamic restraints, it is important to note that the muscles also rely on input from non-contractile tissues surrounding a joint such as the joint capsule, skin, and ligaments. Within both contractile and non-contractile tissues, mechanoreceptors provide sensory information regarding movement, force, and stretch to various afferent pathways that ultimately result in activation of the muscle thereby providing appropriate joint stabilization strategies (Riemann and Lephart 2002).

Despite no known structural differences between sexes in the central or peripheral neuromuscular pathways, research has demonstrated various differences in the neuromuscular control strategies exhibited by men and women. These sex differences have been noted in postural control, biomechanical movement patterns, and in muscular activation patterns (relative to timing and amplitude) during functional activity.

Sex Differences in Postural Control

Postural control is achieved by central nervous system (CNS) processing of the combined inputs of our vestibular, visual, and somatosensory systems to initiate the proper neuromuscular response to maintain a stable upright position (Riemann and Lephart 2002; Wikstrom, Tillman et al. 2006). Specifically, afferent information obtained from these sensory receptors is integrated by the CNS to produce a motor command to the muscles to provide stabilizing or corrective contractions to maintain postural stability (Riemann and Lephart 2002). Accurate and timely sensory information allows for rapid activation of the stabilizing muscles which in turn decrease the sway of an individual's center of mass (COM). Smaller movements of the COM are typically thought to represent improved postural control.

Postural control has been associated with the risk for ACL injury from the standpoint that video analysis of ACL injuries have shown that individuals who suffered an ACL tear landed with a posterior positioning of the COM (Griffin, Agel et al. 2000; Teitz 2001; Sheehan, Sipprell et al. 2012). In addition, it has been suggested that a lateral positioning of the COM, resulting from lateral trunk motion, may create a longer lever arm relative to the knee joint which has the potential to increase the knee abduction moment (Hewett, Torg et al. 2009). Further, a prospective study reported that female athletes that went on to sustain an ACL injury reported balance index scores significantly higher than non-injured subjects, with higher balance index scores signifying larger movements of the COM (Vrbanic, Ravlic-Gulan et al. 2007). These balance index scores were a composite of dynamic and static balance assessments using the SportKat 2000 (a circular platform on a pivot that can provide varying degrees of stability), and are based on the accurate positioning of the center of mass in reference to a moving target (dynamic balance) or the stability of the center of mass (static balance) (Vrbanic, Ravlic-Gulan et al. 2007). Therefore it has been suggested that imprecise movement of the center of mass (e.g., inability to keep the COM over the anterior section of the base of support) during functional tasks may represent a neuromuscular control pattern that may increase the risk for ACL injury.

Sex Differences in Static Postural Control

There are mixed reports in the literature regarding sex differences in static postural control (Hellenbrandt and Braun 1939; Black, Wall et al. 1982; Hewett, Paterno et al. 1999; Sullivan, Rose et al. 2009). While some researchers found no significant difference between sexes in postural sway (Hellenbrandt and Braun 1939; Black, Wall et al. 1982), others noted that females were more stable during a single limb (Hewett, Paterno et al. 1999) or double limb stance (Sullivan, Rose et al. 2009). The difference in findings could be due participants' age range. Specifically, Black et al (Black, Wall et al. 1982) noted no postural control difference between men and women between the age of 20-49 during double limb static standing assessments; whereas Sullivan et al (Sullivan, Rose et al. 2009) assessed men and women from 30-74 years of age in one group. It is possible that the difference found in this study was driven by the subjects above the age of 49 years old as previous studies have identified greater postural instability in individuals over 50 years old (Sheldon 1963). Hewett et al (Hewett, Paterno et al. 1999) is the only study that assessed postural control during a single limb stance, which is considerably more challenging.

Activity type has been reported to affect postural control within healthy participants, with female gymnasts more stable during a single leg stance compared to female basketball athletes, and female soccer athletes having larger limits of stability compared to female basketball athletes (Bressel, Yonker et al. 2007). Despite literature reporting that healthy female subjects are more stable than males (Hewett, Paterno et al. 2002), the ability for the type of activity to alter postural control supports the need for further investigation comparing female dancers and female athletes, the purpose of this review.

Sex Differences in Dynamic Postural Control

The majority of the previous mentioned investigations were conducted using static balance assessments which may not be an accurate depiction of postural control during dynamic activity. Fewer investigations have compared dynamic postural control strategies between sex (Wikstrom, Tillman et al. 2006; Gribble, Robinson et al. 2009; Ericksen and Gribble 2012). Ericksen et al (Ericksen and Gribble 2012), assessed postural control using the star excursion balance test (SEBT) which assesses how far an individual can move their center of mass to the edge of the base of support while maintaining an upright posture. Reach distances were normalized to leg length to control for height differences. From this investigation, male subjects were able to reach further in the posteromedial direction compared to female subjects (Ericksen and Gribble 2012). Research has also shown that the posteromedial reach direction is compromised following an ACL injury and may challenge the knee musculature greater than other reach directions. (Earl and Hertel 2001; Herrington, Hatcher et al. 2009). As such this reach direction may more accurately estimate the ability of the dynamic restraints to stabilize the knee joint in healthy individuals (Herrington, Hatcher et al. 2009). While Gribble et al (Gribble, Robinson et al. 2009), noted that female subjects were able to reach further compared to males; the posteromedial reach direction was not assessed in this investigation (Gribble, Robinson et al. 2009). Therefore it is unknown if this study would of also identified a sex difference in the posteromedial reach direction that is suggested to challenge the knee musculature the greatest.

The dynamic postural stability index (DPSI) is a measure that has been used to compare males and females on postural control. This index calculates a composite score of the time required to stabilize the ground reaction force in all three coordinates, and thus, a more functional task such as a forward hop can be used to challenge the neuromuscular system (Wikstrom, Tillman et al. 2005). Wikstrom et al (Wikstrom, Tillman et al. 2006) reported that females had significantly higher dynamic postural stability index scores compared to male subjects, with higher scores indicating longer time required to stabilize the ground reaction force. The authors acknowledged that the DPSI is a new stability measure and the composite score is heavily influenced by the vertical ground reaction force due to the task being a predominantly vertical jumping task as compared to a horizontal task. Although the authors normalized vertical ground reaction force to body weight, the female subjects landed with significantly greater vertical ground reaction forces (Wikstrom, Tillman et al. 2006). Prospective research has suggested that greater vertical ground reaction force is associated with increased risk for ACL injury (Hewett, Myer et al. 2005). Therefore, the large vertical ground reaction forces in females during the dynamic postural control assessments that led to higher DPSI values further support the potential that this neuromuscular control strategies may increase the risk for ACL injury in female athletes.

Time to Stabilization (TTS) is another dynamic postural control measure that quantifies the body's ability to minimize postural sway when transitioning from a dynamic to static state (Colby, Hintermeister et al. 1999). The TTS score represents the time required to integrate afferent inputs such as proprioceptive and kinesthetic, and the efferent output of reflexive and voluntary muscle responses, and return the system to a static state (Wikstrom, Tillman et al. 2006). Instead of examining actual muscle activation to measure the time to complete the afferent and efferent response to a perturbation of functional task, the GRF in the vertical, anterior-posterior, and medial-lateral directions are utilized. When performing a static stance, there are small fluctuations in the directional GRF; however following a functional task or perturbation, these fluctuations are increased ranging away from the static stance overall GRF mean. The increase in GRF following movement is expected, but what was unknown is the time required to return to the static stance GRF values. TTS is a measure designed to utilize previous static stance fore platform measures with a dynamic task.

TTS has consistently identified neuromuscular control deficits in injured individuals compared to healthy,(Ross and Guskiewicz 2003; Brown, Ross et al. 2004; Ross and Guskiewicz 2004; Ross, Guskiewicz et al. 2005; Wikstrom, Tillman et al. 2005;

Ross and Guskieivicz 2006; Brown and Mynark 2007; Ross, Guskiewicz et al. 2008; Gribble and Robinson 2009; Marshall, McKee et al. 2009; Ross, Guskiewicz et al. 2009). The majority of this literature pertains to injury at the ankle joint with limited research distinguishing between healthy and ACL deficient (ALCd) or ACL reconstructed (ACLr) (Colby, Hintermeister et al. 1999; Phillips and van Deursen 2008; Webster and Gribble 2010). Consistent reports of decreased TTS values for ACLd and ACLr individuals supports the use of TTS for identification of postural control deficits from neuromuscular deficiencies at the knee joint.

Sex Differences in Landing Mechanics

Due to the high occurrence of ACL injuries during jumping and landing maneuvers and the difference in injury rates between males and females, sex differences in knee joint biomechanics during a landing task has been extensively studied (Zhang, Bates et al. 2000; Lephart, Ferris et al. 2002; Decker, Torry et al. 2003; Fagenbaum and Darling 2003; Kernozek, Torry et al. 2005; Ford, Myer et al. 2006; Pappas, Hagins et al. 2007; Pappas, Sheikhzadeh et al. 2007; Schmitz, Kulas et al. 2007; Hughes, Watkins et al. 2008; Brown, Palmieri-Smith et al. 2009; Kiriyama, Sato et al. 2009; Orishimo, Kremenic et al. 2009; Shultz, Nguyen et al. 2009; Sigward, Pollard et al. 2012). Investigators have examined these landing mechanics in sagittal as well as frontal and transverse planes.

Sagittal Plane Landing Mechanics

Large vertical ground reaction forces have been associated with landing in shallow knee flexion angles (Devita and Skelly 1992). During a drop jump, female athletes typically perform a stiffer landing (increased vertical ground reaction forces) (Ford, Myer et al. 2003; Kernozek, Torry et al. 2005; Pappas, Hagins et al. 2007; Kernozek, Torry et al. 2008) in a more erect or upright position (decreased hip and knee flexion) compared to males (Decker, Torry et al. 2003; Kernozek, Torry et al. 2008). As previously stated, the contraction of the quadriceps produces an anteriorly directed force on the proximal aspect of the tibia through the patellar tendon (Renstrom, Ljungqvist et al. 2008), which is accentuated in shallow knee flexion angles where the ACL resists approximately 85% of the anterior tibial shear force (Renstrom, Ljungqvist et al. 2008). Despite research reporting that female athletes perform a stiffer more erect landing strategy, other investigations report no difference between sexes on the vertical ground reaction force (McNair and Prapavessis 1999; Decker, Torry et al. 2003; Blackburn and Padua 2009), or hip and knee flexion (Cowling and Steele 2001; Kernozek, Torry et al. 2005). However, it is important to note that no identified studies to date have reported that male subjects land in less knee flexion or with larger vGRF values than female athletes.

The inconsistent findings between sexes may be due to methodological differences. The two studies that did not report a sex difference in hip and knee flexion performed a single leg landing task (Cowling and Steele 2001; Kernozek, Torry et al. 2005), while the study conducted by Decker et al that reported more upright landings in females was based on a double leg landing (Decker, Torry et al. 2003). Landing height may also influence findings. In studies where subjects performed a double leg landing task from 60cm, there was no sex difference in vertical ground reaction force (Decker, Torry et al. 2003; Blackburn and Padua 2009), whereas differences were noted when performing a double leg landing from 31, and 40cm (Ford, Myer et al. 2003; Pappas, Hagins et al. 2007). It is possible that tasks that are of higher difficulty (higher landing heights or single leg landings) are equally challenging for females and males (thereby eliminating sex differences), while lower difficulty tasks are more challenging for female subjects (thus accentuating sex differences).

Frontal and Transverse Plane Landing Mechanics

Although sex differences in sagittal plane drop jump landing biomechanics are not conclusive, sex differences in frontal and transverse plane landing mechanics are more unified (Ford, Myer et al. 2003; Kernozek, Torry et al. 2005; Noyes, Barber-Westin et al. 2005; Pappas, Hagins et al. 2007; Hughes, Watkins et al. 2008; Haines, McBride et al. 2011). In the frontal plane, female athletes tend to land with greater hip adduction, hip internal rotation, knee valgus, and tibial rotation compared to male athletes (Ford, Myer et al. 2003; Kernozek, Torry et al. 2005; Earl, Monteiro et al. 2007; Pappas, Hagins et al. 2007). These landing mechanics of female athletes mirror the self-reports and video analysis of ACL injury events (Boden, Dean et al. 2000; Olsen, Myklebust et al. 2004; Krosshaug, Nakamae et al. 2007; Kobayashi, Kanamura et al. 2010), and are motions that have been shown to increase the strain on the ACL (Woo, Fox et al. 1998; Withrow, Huston et al. 2006). These combined motions of hip adduction, knee valgus, and tibial rotation are often described as “dynamic valgus” or “valgus collapse” (Olsen, Myklebust et al. 2004; Hewett, Myer et al. 2005; Krosshaug, Nakamae et al. 2007; Quatman and Hewett 2009), and female basketball athletes are reported to be 5.3 times more likely to demonstrate these combined motions during ACL injury compared to male basketball players (Krosshaug, Nakamae et al. 2007).

In summary, sex comparisons in landing biomechanics support that females are more likely to land with an erect landing posture and demonstrate "dynamic knee valgus" or "valgus collapse" compared to males. Results from load application studies suggest that the greater prevalence of these combined landing patterns in females (upright landing and dynamic valgus) are more likely to strain the ACL than either landing pattern alone (Woo, Fox et al. 1998; Sakane, Livesay et al. 1999; Fleming, Renstrom et al. 2001; Withrow, Huston et al. 2006). This has led to the widely held theory that female athletes demonstrate higher risk knee biomechanics during landing that are thought to increase their risk of ACL injury.

Sex Differences in Muscular Activation

Lower extremity muscular activation can also influence strain on the ACL as muscle recruitment order and timing of the quadriceps, hamstrings and gastrocnemius muscles has been shown to affect dynamic knee stability, thus joint motion and loads during a landing task (Renstrom, Arms et al. 1986; Markolf, Burchfield et al. 1995; Hewett, Stroupe et al. 1996; Shultz, Perrin et al. 2000; Shultz, Perrin et al. 2001; Hewett, Zazulak et al. 2005; Myer, Ford et al. 2005; Zazulak, Ponce et al. 2005; Palmieri-Smith, Woitys et al. 2008; Palmieri-Smith, McLean et al. 2009; Brown, McLean et al. 2013). Increased quadriceps activation, while the knee is positioned in less than 30° of flexion, has been suggested to increase anterior tibial shear forces and the strain placed on the ACL and thereby may increase the risk for ligament failure (Renstrom, Arms et al. 1986;

Zazulak, Ponce et al. 2005; Withrow, Huston et al. 2006). This is supported by studies noting that greater pre-activation amplitude of the quadriceps muscle has a small but significant association with increased peak anterior tibial shear force during a drop jump leg landing (Shultz, Nguyen et al. 2009; Brown, McLean et al. 2013). Research has also shown that when the knee joint is flexed to greater than 60°, activation of the hamstring muscles can effectively decrease the strain on the ACL by counteracting the anterior tibial shear forces (Renstrom, Arms et al. 1986). Therefore a mechanism of higher relative quadriceps to hamstring activation with an extended knee may increase the strain on the ACL, thereby creating a greater risk of injury.

Thigh Muscle Activation Amplitude

Research has repeatedly shown that female athletes activate the quadriceps muscles greater than male athletes during a variety of athletic movements such as a side step cut (Sigward and Powers 2006), a single leg squat maneuver (Myer, Ford et al. 2005) and a double leg drop jump landing (Shultz, Nguyen et al. 2009). There were no reported difference between sex in hamstring activation during a side step cut (Sigward and Powers 2006); however, during a landing task, female athletes activated the hamstring musculature greater than male counterparts (Chappell, Creighton et al. 2007; Sell, Ferris et al. 2007; Shultz, Nguyen et al. 2009). Although Renstrom et al (Renstrom, Arms et al. 1986) reported that the activation of hamstrings can decrease the strain on the ACL, it is important to note that muscle activation levels are not linearly related to force of contraction (Woods and Biglandritchie 1983) and the relative contribution of the hamstring muscles can be dependent on position of the COM relative to the foot and hip and knee flexion angles.

Unbalanced lateral versus medial muscular contraction of the quadriceps and hamstring musculature has also been suggested to increase ACL loading. Researchers have demonstrated that female athletes have significantly higher peak amplitude of lateral quadriceps and hamstrings compared to the medial musculature (Rozzi, Lephart et al. 1999; Myer, Ford et al. 2005; Palmieri-Smith, Woitys et al. 2008), which has the potential to open the medial knee joint space, thus increasing the potential for valgus positioning of the knee joint and increasing strain on the ACL (Markolf, Burchfield et al. 1995; Rozzi, Lephart et al. 1999; Palmieri-Smith, Woitys et al. 2008). In support of this premise, Palmieri-Smith et al (Palmieri-Smith, Woitys et al. 2008), reported that the higher preparatory amplitude of the vastus lateralis and biceps femoris in women were associated with larger knee valgus angles during landing.

Hip Muscle Activation Amplitude

Muscular activation at the hip joint has also been investigated in males and females relative to ACL injury risk potential (Zazulak, Ponce et al. 2005; Nguyen, Shultz et al. 2011). It has been postulated that the hip joint musculature controls the positioning of the distal segments as well as assists in the absorption of landing forces. The eccentric contraction of the hip extensors (gluteus maximus) assists in the deceleration of the body while the gluteus medius plays a critical role in the frontal and transverse plane positioning of the hip joint (Hewett, Zazulak et al. 2005; Zazulak, Ponce et al. 2005). Lower gluteus maximus activation has been observed in females compared to males during a single leg landing task (Zazulak, Ponce et al. 2005), and decreased gluteus maximus activation predicted greater hip internal rotation excursion during a single leg squat (Nguyen, Shultz et al. 2011). This may increase the risk of injury as internal rotation of the thigh contributes to the “valgus collapse” commonly demonstrated in females (Olsen, Myklebust et al. 2004; Hewett, Myer et al. 2005; Krosshaug, Nakamae et al. 2007; Quatman and Hewett 2009). Decreased gluteus maximus activation has also been associated with decreased valgus excursion at the knee joint, yet peak valgus angles were not evaluated in this study (Nguyen, Shultz et al. 2011). It is also important to note that an inverse relationship has been reported between gluteal strength and activation, with individuals with lower hip extension and abduction strength requiring greater gluteus maximus and medius activation during a single leg squat respectively (Nguyen, Shultz et al. 2011). This suggests that greater activation of a gluteal muscle group may not directly correlate to improved hip control or safer positioning of the lower limb, but rather may signal a need for greater activation levels to stabilize the joint.

Reflex Response to Unanticipated Perturbation

Although it is yet unclear whether reflexive muscular activation can generate enough force to protect a joint against a sudden externally applied load, sex differences in the timing of muscular activation during athletic movements have been reported, with females having slower hamstring reflex responses following an anterior tibial stress (Wojtys, Ashton-Miller et al. 2002), and faster quadriceps reflex responses following a sudden rotational perturbation of the knee joint (Shultz, Perrin et al. 2001). The early quadriceps with or without delayed hamstring activation may inhibit the hamstrings ability to generate adequate force to control anterior tibial translation and protect the ACL from excessive strain. However, to date, no one has directly examined how these strategies affect ACL loading.

Summary of Neuromuscular Control Patterns

The preponderance of literature suggests that the neuromuscular control strategies of female athletes during functional tasks are more likely to increase ACL strain and loading than those observed in males. Sex differences in neuromuscular control patterns through assessments of postural control, landing mechanics, as well as muscular activation patterns support the greater likelihood of females displaying "high-risk" strategies. During dynamic balance tasks, female athletes demonstrate larger balance index scores, and position the center of mass more outside the base of support compared to male athletes which has to the potential to increase knee extensor moments (posterior positioning of the center of mass) or knee abduction moment (lateral positioning of the center of mass) in order to maintain an upright posture (Griffin, Agel et al. 2000; Teitz 2001; Vrbanic, Ravlic-Gulan et al. 2007; Hewett, Torg et al. 2009; Sheehan, Sipprell et al. 2012). During drop landing maneuvers, female athletes are more often observed to land with larger vertical ground reaction forces (Ford, Myer et al. 2003; Kernozek, Torry et al. 2005; Pappas, Hagins et al. 2007; Kernozek, Torry et al. 2008), a more extended hip and knee posture (Decker, Torry et al. 2003; Kernozek, Torry et al. 2008), and greater hip adduction (Earl, Monteiro et al. 2007) and knee valgus (Ford, Myer et al. 2003; Pappas, Sheikhzadeh et al. 2007), each of which can independently increase the load on the ACL, and when performed collectively may increase ACL strain further (Berns, Hull et al. 1992). Associated with these higher risk landing biomechanics are greater relative quadriceps to hamstring activation (Myer, Ford et al. 2005), and greater lateral to medial quadriceps and hamstring muscle activation (Markolf, Burchfield et al. 1995; Rozzi, Lephart et al. 1999; Palmieri-Smith, Woitys et al. 2008), and decreased gluteus maximus activation in females compared to male athletes (Zazulak, Ponce et al. 2005). The compilation of these findings has led to a growing consensus that the neuromuscular control patterns of female athletes are a significant contributing factor to the gender disparity in ACL injury rates.

Neuromuscular Control Strategies in Dancers

Thus far, this literature review has identified that neuromuscular control patterns commonly displayed by female athletes may contribute to their higher rate of ACL injury. This section will highlight the neuromuscular control patterns commonly displayed by female dancers who, despite performing decelerating movements (jumping and change of direction), are at a decreased risk of ACL injury (Liederbach, Dilgen et al. 2008; Meuffels and Verhaar 2008). We will first review the physical demands and characteristics of dance, followed by a theoretical rationale for the decreased rate of ACL injury in this population based on their training.

Physical Demands and Characteristics of Dance

The physical activity of dance has been described as “quick bursts of energy interspersed with steady state activity”, which is similar to other sports such as soccer and volleyball (Cohen 1984). Further, the type of athletic maneuvers associated with ACL injury (e.g., jumping and changing direction) are also performed by dancers (Figure 2).

Figure 2. Female dancer landing a jump in knee valgus (Meuffels and Verhaar 2008)

Despite the comparable intensities and maneuvers in athletic and dance activities, the physical condition of dancers has been questioned when making comparisons to an athletic population. However, an investigation conducted by Angioi et al (Angioi, Metsios et al. 2009) showed the aerobic capacity (as measured by VO_2 max) of professional contemporary dancers (49.1 ml*kg/min) is comparable to volleyball athletes (46.5 1 ml*kg/min), gymnasts (49.61 ml*kg/min), and even football players (50 1 ml*kg/min). In regards to physical strength, dancers were reported to have five times greater quadriceps mean maximal voluntary isometric force compared to physically active individuals matched on age (Harley 2002). Although dancers reported greater quadriceps strength compared to physically active individuals, there were no statistical differences in power between the two groups as measured by vertical jump height (Harley 2002). But, while the literature indicates similar physical demands of the activity as well as physical condition and performance of participants in dance and athletic field sports, there is a noteworthy difference in ACL injury occurrence between the two populations.

Rationale for Low Anterior Cruciate Ligament Injury Rate

Although dancers also perform plant-and-cut maneuvers and numerous jumps, they are 3-5 times less likely to suffer an ACL injury compared to female field athletes (i.e., soccer and basketball athletes) (Liederbach, Dilgen et al. 2008; Meuffels and Verhaar 2008). Moreover, there is no sex difference in ACL injury within the dance population (Liederbach, Dilgen et al. 2008). Researches have theorized potential reasons for decreased ACL injury risk in dances, which include rehearsed choreography (Orishimo, Kremenic et al. 2009), controlled toe to heel landing techniques (McNitt-Gray, Koff et al. 1992; Liederbach, Dilgen et al. 2008; Orishimo, Kremenic et al. 2009), a more neutral alignment during jumping tasks (Liederbach, Dilgen et al. 2008; Ambegaonkar, Shultz et al. 2009; Orishimo, Kremenic et al. 2009), improved balance ability (Liederbach, Dilgen et al. 2008; Orishimo, Kremenic et al. 2009), and years of training (Orishimo, Kremenic et al. 2009). However, few investigations have directly compared the dance and athletic populations to test these theories and better understand the cause for the lower rate of injury in dancers. The next section will examine the literature associated with each of the above theories to support a plausible theoretical rationale for the low rates of ACL injury in dancers compared to athletes.

Planned versus Reactive Movements

One commonly proposed theory for the lower injury rate in the dance population is the performance of choreographed, or planned, movements. Anticipating a movement has been shown to change an individual's reflex response and postural adjustments to maintain appropriate posture (Besier, Lloyd et al. 2001). Biomechanical differences are also noted when performing an unplanned compared to a planned cutting task, with a decrease in knee flexion moment during an unplanned cut (Besier, Lloyd et al. 2001). However, knee valgus and internal rotation moments were significantly increased and generalized muscle activation was 20% higher during the unplanned task which produce 70% greater external forces as compared to the planned task (Besier, Lloyd et al. 2001). These findings would suggests that performing unplanned tasks are more likely to place greater internal and external loads on the ACL thereby increasing the vulnerability for failure.

While this evidence suggests that the unplanned tasks of athletes may increase their risk of ACL injury compared to dancers who typically perform choreographed movements, there is conflicting research regarding injury rates in populations that perform planned movements. This is exemplified in two populations that performed choreographed or anticipated movements but have very difference ACL injury rates. Specifically female dancers injury their ACL at a rate of .009 per 1000 exposures

(Liederbach, Dilgen et al. 2008); however, gymnast who also performs choreographed routines injure their ACL at a rate of .33 rate per 1000 exposures (Hootman, Dick et al. 2007), a rate similar to that of women's soccer (.28/1000 exposures) and women's basketball (.23/1000 exposures). Despite all four of these activities performing functional tasks associated with ACL injury (jumping and change of directions), only the dance population reports a significantly lower rate of injury. Thus there are likely other explanations for the lower risk of ACL injury within the dance population beyond the choreographed nature of the activity.

Postural Control

The majority of research conducted on dancers has focused on their postural control, and report that dancers are more stable compared to healthy individuals, recreational athletes, and collegiate athletes (Crotts, Thompson et al. 1996; Golomer, Cremieux et al. 1999; Hugel, Cadopi et al. 1999; Schmit, Regis et al. 2005; Simmons 2005; Gerbino, Griffin et al. 2007). However, the majority of this research has been done through static assessments, which as previously noted, may not be representative of postural control requirements during functional activities.

Research has shown that dancers and healthy controls can maintain a single leg stance for 20-30 seconds when accurate somatosensory, visual and vestibular information is provided (Crotts, Thompson et al. 1996). When comparing healthy populations, differences are not always observed without challenging the sensory inputs; therefore, the somatosensory, vision and/or vestibular information is selectively challenged through the use of a foam mat and a visual dome. In conditions where the sensory inputs were challenged, female dancers were able to maintain a single leg stance significantly longer than healthy controls (Crotts, Thompson et al. 1996).

Other postural control assessments have quantified postural sway in dancers by tracking movement of the center of pressure underneath the foot (Goldie, Bach et al. 1989). Dancers were reported to have decreased movement of the center of pressure while maintaining a single leg stance on a firm and foam surface with eyes open compared to collegiate female soccer athletes (Gerbino, Griffin et al. 2007). During a functional forward hop task where subjects land on a single leg and hold that position for 10 seconds, female dancers again demonstrated decreased postural sway compared to soccer athletes (Gerbino, Griffin et al. 2007). As previously stated, increased variability the center of mass (COM), or increased postural sway, during static and dynamic tasks has been associated with increased risk for ACL injury (Griffin, Agel et al. 2000; Teitz 2001; Vrbanic, Ravlic-Gulan et al. 2007; Hewett, Torg et al. 2009; Sheehan, Sipprell et al. 2012). Therefore, the findings from Gerbino et al (Gerbino, Griffin et al. 2007), suggest dancers may have a more accurate and stable positioning of their center of mass compared to female athletes, which in turn may be protective of the ACL and may contribute to the lower rate of injury.

Landing Mechanics

As previously mentioned, research assessing neuromuscular control in females athletes has documented an extended and valgus knee position when landing a jump (Decker, Torry et al. 2003; Ford, Myer et al. 2003; Kernozek, Torry et al. 2005; Earl, Monteiro et al. 2007; Pappas, Hagins et al. 2007; Kernozek, Torry et al. 2008), which *in-*

vitro studies suggests may increase strain placed on the ACL (Berns, Hull et al. 1992; Woo, Fox et al. 1998; Sakane, Livesay et al. 1999; Fleming, Renstrom et al. 2001; Withrow, Huston et al. 2006).

Sagittal Plane Landing Mechanics

Compared to healthy controls matched on age and weight, female dancers landed in greater plantar flexion at the ankle, and displayed greater peak hip and knee flexion and during a countermovement jump task (McNitt-Gray, Koff et al. 1992). Further, landing phase time (from ground contact to minimum vertical position of total body center of gravity) was significantly longer in female dancers compared to healthy controls (McNitt-Gray, Koff et al. 1992), suggesting that dancers utilize more range of motion at the joints of the lower extremity, which may assist in absorbing ground reaction force and thereby decreasing external loads placed on passive knee structure. Despite these kinematic differences between dancers and non-dancers, there was no significant difference in ground reaction force. However, there was a trend for a smaller ground reaction force in dancers, and the lack of significant difference may be due to the study being inadequately powered secondary to a relatively small sample size (N=12). More work is needed to determine if female dancers exhibit more protective kinematics and ground reaction forces during a landing compared to athletes.

Of particular note, sex differences in landing mechanics have not been observed in the dance population (Orishimo, Kremenic et al. 2009), which is contrary to sex differences noted in most sports (Decker, Torry et al. 2003; Pappas, Hagins et al. 2007; Schmitz, Kulas et al. 2007). Specifically, during single limb landing tasks, male and female dancers were found to land similarly on a single limb in 59.2° ± 12.5° and 58.7° ± 5.5° of peak knee flexion respectively (Orishimo, Kremenic et al. 2009), which was greater than the knee flexion observed in male (51.8°) and females (50.8°) recreational athletes (Schmitz, Kulas et al. 2007). Additionally, while male and female recreational athletes had similar peak knee flexion angles, total joint displacement was significantly less in the female (8.3° ± 5.9°) compared to male recreational athletes (12.9° ± 6.9°) (Schmitz, Kulas et al. 2007). This sex disparity was not observed in the dance population (males = 58.2° ±8.7°; females = 55.1° ± 5.1) (Orishimo, Kremenic et al. 2009). Also notable is that the knee flexion at initial contact for male and female dancers [1° ± (7.0°) and 3.5° ± (4.4°) respectively] is considerably smaller than that reported for male and female athletes. Although this extended knee posture upon ground contact is considered to be of higher risk for the ACL, it is also a goal of dancers to maintain the "artistic line of the leg" during flight, and landing in more extended knee may provide a greater range of motion over which to decelerate the landing. Further investigation is needed to determine if these kinematic strategies in dancers are able to offset externally applied loads through a more absorptive landing upon ground contact.

Frontal Plane Landing Mechanics

Frontal plane motion is also reported to be similar in male and females dancers who demonstrated 3.2° ± 4.3° and 1.7° ± 11.1° of peak knee valgus respectively (Orishimo, Kremenic et al. 2009). As previously noted, knee valgus can contribute to increase strain on the ACL (Berns, Hull et al. 1992; Woo, Fox et al. 1998; Withrow, Huston et al. 2006). The lack of sex difference in frontal plane knee motion, where both male and female dancers display knee valgus angles more similar to male athletes, further supports the idea that dancers display more protective neuromuscular control strategies. This finding has recently been supported as female athletes were shown to have greater peak knee valgus angles during a single limb landing task compared to female dancers, and male athletes and dancers (Orishimo, Liederbach et al. 2014). Future studies to further elucidate the extent to which dances display biomechanical strategies that may be more protective of the ACL.

Neuromuscular Training

The type of training dancers undergo has also been proposed as a reason for the decreased rate of ACL injury in the dance population, and may contribute to the differences in neuromuscular strategies reported. While dance requires physical fitness, its focus is on the artistic quality of movement (Brown, Wells et al. 2007). The athletic demands within ballet include having a full range of motion of the lower extremities, power to perform jumping movements, and the strength to control the limb at the end ranges of flexibility (Hamilton, Hamilton et al. 1992). Motions such as a développé requires the strength to slowly control the entire lower limb while moving across the end range of motion (Brown, Wells et al. 2007). Whereas a tour jete requires power to jump high enough to switch the position on the lower legs while in air (Brown, Wells et al. 2007). Due to these demands, dance training typically includes a variety of training techniques such as balance training, stretching, plyometrics, agility and strengthening exercises that are all performed in an activity specific manner that is fully integrated into their daily training.

While the training techniques used in dance practice are similar to what is currently being used for female athletes in ACL prevention programs, there are distinct differences that may contribute to more protective neuromuscular control patterns in female dancers. Prior to a depiction of these differences, we will briefly discuss the current techniques in neuromuscular training programs designed to reduce the risk and occurrence of ACL injury. This will be followed by specific differences between the ACL prevention programs and dance practice, supplemented with literature to suggest how these difference training practices may allow dancers to better retain protective neuromuscular control strategies.

ACL Prevention Programs

Neuromuscular training programs designed with the intent to reduce the occurrence of ACL injuries began in the late 1990s and continues today (Donnelly, Elliott et al. 2012). Specifically the goal of neuromuscular training programs is to improve joint positioning by: 1) increasing hip and knee flexion (Lephart, Abt et al. 2005; Myer, Ford et al. 2005; Herman, Onate et al. 2009; Lim, Lee et al. 2009; Cochrane, Lloyd et al. 2010), 2) decreasing hip adduction and knee valgus (Hewett, Stroupe et al. 1996; Myer, Ford et al. 2006; Pollard, Sigward et al. 2006; Herman, Onate et al. 2009), and or, 3) decreasing hip internal rotation (Pollard, Sigward et al. 2006)]; improve muscular activation by 1) increasing hamstring or gluteus medius activation prior to ground contact (Lephart, Abt et al. 2005; Zebis, Bencke et al. 2008), and or, 2) increasing hamstring: quadriceps MVIC ratio (Lim, Lee et al. 2009)]; and improve postural control by decreasing ground reaction force (Hewett, Stroupe et al. 1996; Onate, Guskiewicz et al.

2005; Vescovi, Canavan et al. 2008; Herman, Onate et al. 2009)]. Attempts to develop these protective neuromuscular control patterns have been through a combination of strength, balance, plyometrics, landing technique, risk-awareness, agility, stretching, and proprioceptive exercises (Alentorn-Geli, Myer et al. 2009).

Many programs report a reduction in high risk biomechanics as well as actual injury incidence (Caraffa, Cerulli et al. 1996; Hewett, Stroupe et al. 1996; Hewett, Lindenfeld et al. 1999; Lephart, Abt et al. 2005; Mandelbaum, Silvers et al. 2005; Myer, Ford et al. 2005; Vescovi, Canavan et al. 2008; Lim, Lee et al. 2009). However, due to the shotgun nature of current neuromuscular training programs, it is unknown what specific aspects of programs developed the desired improvements in neuromuscular control and thereby decrease the incidence of ACL injury. Studies have shown biomechanical improvements or a reduction in injury occurrence using specific neuromuscular training technique such as educational information (Iversen and Friden 2009), balance training (Caraffa, Cerulli et al. 1996), and plyometric training (Hewett, Stroupe et al. 1996; Myer, Ford et al. 2006; Vescovi, Canavan et al. 2008; Zebis, Bencke et al. 2008). For example, plyometric training studies lasting 20 minutes or more for 6 weeks reported decreased landing forces (Hewett, Stroupe et al. 1996; Vescovi, Canavan et al. 2008), decreased knee valgus angles (Hewett, Stroupe et al. 1996; Myer, Ford et al. 2006) and hip adduction angles (Myer, Ford et al. 2006), and increased knee flexion (Myer, Ford et al. 2006) and hamstring activation (Zebis, Bencke et al. 2008). A progressive balance training program was also shown to decrease occurrence of ACL injury in male athletes (Caraffa, Cerulli et al. 1996). Finally Iversen and Frieda (Iversen and Friden 2009), were able to increase knee flexion at contact during a drop jump landing task simply from instructions on what is considered to be at risk landing maneuvers.

Multi-component interventions are more frequently investigated and according to a meta-analysis conducted by Yoo et al (Yoo, Lim et al. 2010), are more likely to develop protective landing mechanics if the multi-component intervention includes plyometrics or strength training. These neuromuscular programs have been reported to increase knee flexion angles (Lephart, Abt et al. 2005; Myer, Ford et al. 2005; Lim, Lee et al. 2009), decreased hip internal rotation (Pollard, Sigward et al. 2006), decreased hip adduction (Pollard, Sigward et al. 2006), increased hip abduction muscle activation (Lephart, Abt et al. 2005), and decreased hamstring to quadriceps ratio (Lim, Lee et al. 2009). Findings from the meta-analysis also revealed that neuromuscular training programs are more likely to develop protective landing mechanics when implemented in females under the age of 18 and span from pre-season to the end of regular season (Yoo, Lim et al. 2010). The length of the intervention may also be important, as a systematic review noted that neuromuscular training programs that lasted a minimum of 6 weeks with a minimum of 3 training session a week for 20 to 90 minutes were more likely to alter the neuromuscular control patterns (Dai, Herman et al. 2012).

Despite the wide implementation of neuromuscular training programs and their success in improving neuromuscular control patterns and decreasing ACL injury rates, epidemiology data report no overall decrease in the rate of ACL injury in the female population over a 12 year span (Arendt and Dick 1995; Hootman, Dick et al. 2007). The continued higher rate of ACL injury in female athletes and the lack of decrease in the sex disparity suggest that the neuromuscular training programs may not be retained or transferred to the sport.

Few articles have examined the retention of protective neuromuscular control following the cessation of the intervention (Onate, Guskiewicz et al. 2001; Prapavessis, McNair et al. 2003; Padua, DiStefano et al. 2012). These investigations found protective neuromuscular control patterns can last up to 3 months after cessation of training following a 9 month program (Padua, DiStefano et al. 2012). Research has yet to determine how long protective neuromuscular control strategies are retained following cessation of a neuromuscular training program. The investigations that have assessed retention provided augmented feedback to subjects during landing technique training through visual (Onate, Guskiewicz et al. 2001), and or verbal (Onate, Guskiewicz et al. 2001; Padua, DiStefano et al. 2012) mechanisms; however, it is important to note this technique is not consistently used in traditional ACL prevention neuromuscular training programs. The next section will discuss how dance training routinely uses augmented feedback, which may lead to improved retention and transfer of protective neuromuscular control patterns to activity.

Difference between Dance Training and Neuromuscular Control Programs

As previously mentioned, a typical dance practice includes plyometric, agility, strength, balance, and flexibility training along with education on soft, neutral alignment landing strategies; very similar to the techniques used during ACL prevention programs. Yet, the neuromuscular training dancers undergo occurs during activity specific movements within dance practice, rather than as a separate training component performed outside of regular practice. Thus, the majority of dancers receive neuromuscular training throughout the entire technique (similar to pre-season) and performance season (similar to regular season). As suggested by Yoo et al (Yoo, Lim et al. 2010), this may be the optimal time for the development of protective neuromuscular control patterns. Furthermore the average starting age of ballet training is 7.3 ± 3.9 years (Hamilton, Aronsen et al. 2006), which may afford dancers the opportunity to incorporate neuromuscular training prior to the age of 18, which has been suggested as the ideal time for the development of protective movement patterns from neuromuscular training programs (Yoo, Lim et al. 2010). Thus, dance training implements the neuromuscular techniques that are most commonly successful in the development of safer movement patterns, and initiates this training at the ideal age and time of season thought to be most effective in ACL prevention research (Yoo, Lim et al. 2010). In addition, dance training also incorporates motor learning techniques that have been shown to improve retention.

Verbal feedback is consistently provided by dance instructors and choreographers on the quality and alignment of movement (Ramsay and Riddoch 2001). Visual feedback is also provided during dance practice (which inherently consists of neuromuscular training interventions) from the mirrored walls of the dance studios (Ramsay and Riddoch 2001). The visual and verbal feedback (i.e., augmented feedback) is consistently provided in dance practice but not in ACL prevention programs. Specifically augmented feedback provides additional information provided about a movement or skill that cannot be detected from the individuals intrinsic senses (Maas, Robin et al. 2008); rather it takes advantage of extrinsic information provided through auditory, visual, or tactile sensation. Augmented feedback is a tool commonly used to reach the final stage of motor learning represented by the relatively permanent change in the skill practiced (Schmidt and Lee 2005).

Based on these findings, the use of augmented feedback assists with the acquisition and retention of complex motor skills (Clarkson, James et al. 1986; Broker, Gregor et al. 1993; Maas, Robin et al. 2008; Sigrist, Schellenberg et al. 2011). Thus, if visual augmented feedback improves retention of complex movement, the neuromuscular training implemented in dance training that incorporates visual and verbal augmented feedback on body positioning during landing may assist female dancers in performing and retaining softer landing strategies and more neutral positions when performing functional tasks.

Summary

Dancers are physically active artists that require similar fitness capabilities of athletes to perform risky maneuvers such as jumping and planting-and cutting. Despite the performance of movements that are associated with ACL injury, dancers are 3-5 times less likely to suffer an ACL tear compared to female field athletes (Liederbach, Dilgen et al. 2008; Meuffels and Verhaar 2008). Based on this review, more protective neuromuscular control patterns may underlie the low rate of ACL injury in female dancers (Liederbach, Dilgen et al. 2008; Meuffels and Verhaar 2008; Orishimo, Kremenic et al. 2009). Moreover, the dance literature shows a lack of gender disparity in postural control and landing mechanics in the dance population, which is present in the athletic population. However there remains a large gap in the literature assessing the neuromuscular control patterns in female dancers compared to female athletes. If differences do exist between these populations, this will pave the way for investigators to focus future research on understanding how dance specific neuromuscular training programs may protect female dancers from ACL injury.

Neuromuscular training programs include various techniques such as plyometrics, balance, agility, strength and flexibility training (Yoo, Lim et al. 2010; Dai, Herman et al. 2012). In athletic populations, neuromuscular training is typically done in addition to athletic practice and not during sport specific activities, whereas neuromuscular training is inherent to dance practices due to the constant focus on the quality of movement and alignment of lower extremity during the movement. Augmented feedback, a motor learning tool commonly used to create a permanent change (Schmidt and Lee 2005), is also provided to dancers through verbal (instructor) and visual means (mirrored walls) (Ramsay and Riddoch 2001), where only a few ACL prevention programs include augmented feedback (Prapavessis and McNair 1999; Onate, Guskiewicz et al. 2001; Prapavessis, McNair et al. 2003; Padua, DiStefano et al. 2012). Dance practices include the same neuromuscular training principles that have been shown to develop protective landing mechanics and decrease the risk of ACL injury; moreover, these training principles are implemented prior to and during the performance season, as well as before the age of 18, which was suggested as the opportune time for neuromuscular training (Yoo, Lim et al. 2010). Because of these training differences and preliminary comparisons, it is expected that female dancers may develop and retain more protective neuromuscular control patterns compared to female athletes which may in part explain their lower risk of ACL injury.

Summary

The anterior cruciate ligament resists movement in multiple planes and injury to this ligament can cause costly long term disabilities (Lohmander, Ostenberg et al. 2004). Neuromuscular control strategies are thought to be a major contributing factor to the risk of ACL injury in female athletes as they commonly display mechanics that are thought to increase the strain on the ACL (Quatman and Hewett 2009). However, preliminary studies suggest female dancers are less likely to display high risk neuromuscular strategies (Crotts, Thompson et al. 1996; Gerbino, Griffin et al. 2007; Orishimo, Kremenic et al. 2009), have similar movement profile to male dancers (Orishimo, Kremenic et al. 2009), and have a much lower incidence of ACL injury compared to female field athletes (Liederbach, Dilgen et al. 2008; Meuffels and Verhaar 2008). Multiple theories have been proposed to explain this lower incidence in dancers, which may include rehearsed choreography (Orishimo, Kremenic et al. 2009), controlled toe to heel landing techniques (McNitt-Gray, Koff et al. 1992; Liederbach, Dilgen et al. 2008; Orishimo, Kremenic et al. 2009), a more neutral alignment during jumping tasks (Liederbach, Dilgen et al. 2008; Ambegaonkar, Shultz et al. 2009; Orishimo, Kremenic et al. 2009), improved balance ability (Liederbach, Dilgen et al. 2008; Orishimo, Kremenic et al. 2009), and years of training (Orishimo, Kremenic et al. 2009). Currently, there is a substantial gap in the literature comparing the neuromuscular profile of female dancers and athletes to determine if they indeed demonstrate different neuromuscular control strategies as a result of fundamental differences in their training. Should female dancers be found to perform functional tasks in a more protective manner, this would suggest that the neuromuscular training program used by female dancers may better develop and retain protective neuromuscular control strategies, and may serve as a model for the development and retention of protective neuromuscular control patterns in female athletes.

To that end, the first step is to comprehensively characterize and compare the neuromuscular control strategies used by female dancers and athletes during planned and unplanned functional tasks. Successful completion of this work will allow for the direct comparison of the neuromuscular control strategies of a female population at low (dance) and high (athletic) risk for ACL injury. Limiting comparisons to a female population will allow us to effectively control for hormonal or bony alignment differences that commonly confound sex comparison investigations. The findings from this study will also encourage future research to assess the benefit of visual augmented feedback during ACL prevention programs, which is imperative for the development of evidence based neuromuscular training programs that lead to protective neuromuscular control patterns that are retained and transferred to sport.

CHAPTER III

METHODS

The primary objective of this research was to characterize the neuromuscular control strategies in collegiate female dancers and athletes during planned and unplanned functional tasks to determine if dancers demonstrate more protective neuromuscular control patterns. To achieve this objective a comparative study design was conducted where female dancers and athletes were paired on years of experience in their respective activities, and the neuromuscular control patterns were measured during three functional tasks: 1) forward hop, 2) anticipated double-leg drop jump, and 3) an unanticipated lower extremity perturbation.

During the forward hop task, postural control was assessed using the dynamic balance measure of time to stabilization (TTS) in the anterior-posterior (A-P) and medial-lateral (M-L) plane in which we expected female dancers to stabilize the ground reaction force (GRF) faster compared to athletes. During the anticipated double-leg drop jump, we assessed postural control, kinematic, kinetic, and muscular amplitude variables. The postural control measure assessed the A-P positioning of the center of mass (COM) at ground contact relative to the position of the center of pressure (COP), and we anticipated dancers to position their COM more anterior compared to athletes. The kinematic variables assessed were hip, knee, and ankle flexion at initial ground contact and excursion, as well as, hip and knee frontal plane motion at initial ground contact and excursion. We hypothesized that dancers would demonstrate greater ankle plantar flexion and equal hip and knee flexion compared to athletes, while expecting less frontal plane motion at the hip and knee in dancers compared to athletes. Vertical ground reaction force (vGRF), peak hip, knee, and ankle extensor moment, and relative hip, knee and ankle energy absorption were also assessed during the double-leg drop jump. For these kinetic variables, we expected dancers to show lower vGRF values, and peak knee extensor moment, while absorbing more relative energy at the ankle compared to the knee joint compared to athletes. Finally for this task we also assessed the muscular amplitude 150ms prior to ground contact and we expected dancers to demonstrate a higher hamstring activation amplitude compared to athletes. The last task, the unanticipated lower extremity perturbation, we assessed muscular onset timing, and expected fast muscular onset in dancers compared to athletes

Participants

Forty female participants (20 dancers, 20 athletes), between the ages of 18-30 years, were recruited from local universities to participate in this study. Collegiate female athletes were recruited from university sport teams that required running, cutting and or landing maneuvers (e.g., basketball, soccer, volleyball rugby, tennis). Collegiate female dancers were recruited from local University Dance Departments. Inclusion criteria were: 1) a minimum of 5 years' experience in their respective sport or activity, and 2) currently participating in a minimum of 120 minutes per week in their respective sport or activity.

Subjects were excluded if they had, 1) a lower extremity injury in the last 6 months; 2) any vestibular or balance disorder that could cause them to lose their balance during functional tasks, 3) cardiovascular disease, or 4) participation in both dance and field sports. All participants read and signed an informed consent form approved by the University of North Carolina at Greensboro's Institutional Review Board for the Protection of Human Subjects (Appendix A). Each participant attended a single testing session consisting of a familiarization to all study procedures, a standardized warm-up, a strength assessment, and neuromuscular assessment during the three functional tasks.

Procedures

All testing took place on the University of North Carolina at Greensboro's campus in the Applied Neuromechanics Laboratory. Upon arrival, subjects provided written consent, and subject demographics (age, sex, height, and mass) were recorded. A standard laboratory scale (Seca, Hamburg, Germany) was used to measure participants mass. Next, participants completed a physical activity (type, duration, intensity, and years of experience) and injury history questionnaire (Appendix B). Following the completion of the questionnaires, subjects were then outfitted in standardized compression shorts that allow for the attachment of motion analysis markers. Standardized athletic shoes (Adidas, Uraha 2, Adidas North America, Portland, OR) were provided to all subjects. The forward hop stabilization and double leg drop jump task were completed during both shod and barefoot conditions, with the order counterbalanced across subjects (Appendix C).

Familiarization

Prior to instrumentation and data collection, subjects were familiarized to the three functional tasks.

Forward Hop Stabilization

For this task, subjects stood 40% of their height behind two non-conducting force platforms (Type 4060-130, Bertec Corporation, Columbus OH), jumping forward off of two feet, clearing a 25 cm foam barrier, and then landing on a pre-determined single leg in the center of one of the force plates. Subjects were instructed to maintain their arms across the chest for the duration of the task, and hold the single leg stance immediately upon landing for 10 seconds. Subjects were provided a minimum of 5 consecutive successful trials to become comfortable with the task and further trials were provided if needed. A trial was considered successful if the participant: 1) cleared the foam barrier; 2) landed on a single leg without jumping, hopping or shifting the foot upon landing; 3) maintained a single leg stance for 10 seconds following landing; and 4) maintained arms across the chest throughout the entire trial.

Double Leg Drop Jump

The drop jump is one of the most commonly assessed movement patterns for the identification of ACL injury risk and has been used in previous research in the Applied Neuromechanics Research Laboratory (Shultz, Nguyen et al. 2009; Schmitz and Shultz 2010). This task was chosen because of the frequency of injury that occurs during a jumping task as well as the ability to standardize the height of the task within a laboratory setting.

To perform the task, participants stood atop a .45m box in front of the two force platforms with feet shoulder width apart, their toes off the edge of the box and hands at ear level. Participants were instructed to gradually lean forward through their hips so that they fell straight down without having to take a step off the box. Participants were instructed to land evenly on both feet (one foot on each force platform), immediately perform a maximal vertical jump, and land again on the force platform evenly on both feet. Subjects were considered comfortable with the task after completing 5 consecutive successful trials and further trials were provided if needed. A trial was considered successful if the participant: 1) slid off the box without jumping or stepping; 2) landed with one foot on each force plate both prior to and following maximal vertical jump; and 3) maintained hands at ear level throughout entire trial.

Lower Extremity Perturbation

This task use of a lower extremity perturbation device (LEPD) to assess reactive postural and lower extremity muscle reflex response times (Shultz, Perrin et al. 2000; Shultz, Perrin et al. 2001) (Figure 3). The task allows for the assessment of reactive muscular reflexes during an unanticipated perturbation that mimics a change of direction (side cut and cross over cut) maneuver. Because performance of planned movements is a prevailing theory as to why dancers have a lower rate of injury, it was important to also compare dancers and athletes during an reactive task that was relatively novel for both groups to further discern the effects of dance training on potential protective neuromuscular strategies.

The LEPD consists of a restraining belt worn by the participant at the level of the ASIS, and attached to force transducers (WMC-1000, Interface, Scottsdale, AZ) imbedded in to each of the two Kevlar cables connected to the wall through quick release trigger mechanisms. The release triggers are mounted to a height adjustable wall mount to ensure the Kevlar cables remained parallel to the floor regardless of the height of the participant. When released from the wall mount, it causes an unanticipated forward and internal (right cable release) or external rotation (left cable release) of the trunk and femur on a weight bearing tibia.

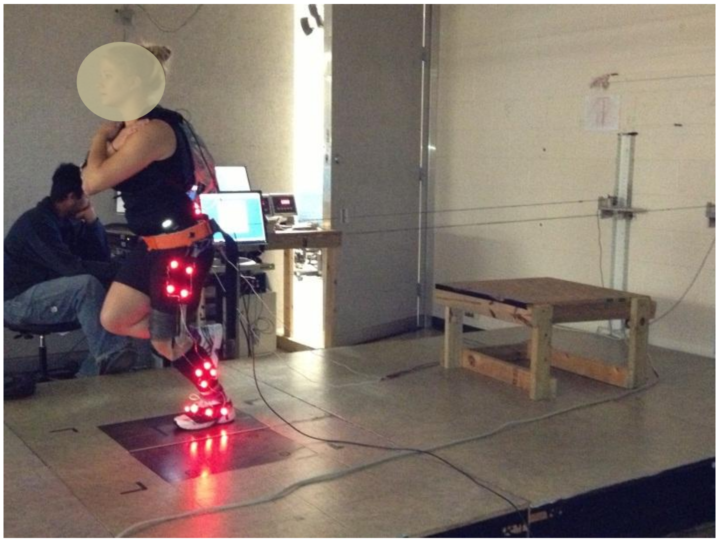

Figure 3. Subject instrumented in the lower extremity perturbation device (LEPD)

To perform the task, subjects were harnessed into the LEPD belt with the cables taut. Participants then assumed a single leg position on the dominant limb with the knee flexed to approximately 30º, arms across the chest, and leaning forward into the belt evenly with both hips. The instructions provided were to react to the perturbation, by attempting to maintain their single leg stance upon cable release. During the familiarization, two anticipated perturbations were provided by telling the participant which cable would be released, followed by 5 unanticipated perturbations in each direction. Participants were considered comfortable with the task when they performed a minimum of 5 trials successfully and were provided more trials if needed. A trial was considered successful when the participant, 1) leaned equally through both hips, as confirmed with force scale readings attached to each cable; 2) the center of pressure remained between the 5th metatarsophalangeal joint and the navicular prior to cable release; 3) maintained a knee flexion angle between 25º-35º prior to the cable release; and 4) participant did not take a step with the dominant stance limb.

Subject Instrumentation

Following familiarization, participants were instrumented for the collection of neuromuscular and biomechanical data. First the skin was prepped with shaving (if necessary) and alcohol wipes prior to the attachment of surface EMG electrodes to the medial and lateral aspects of the gastrocnemius, quadriceps, and hamstring musculature. Six double differential surface electrodes (Trigno Wireless Sensors, Delsys, Boston, MA) were placed in a parallel arrangement to the muscle fibers at the mid belly of each of the muscle sites. All electrode placements were confirmed with manual muscle testing. The sEMG electrodes were then secured with double sided tape and pre-wrap to prevent movement artifact during the functional tasks. Once the sEMG sensors were in place, participants were instrumented with clusters of 4 LED marker sets attached to the foot, shank, thigh, and sacrum (Phase Space, San Leandro, CA). Shank and thigh marker sets were attached using hook and loop material, while the sacrum marker set was attached directly to the skin using double sided tape and the foot marker set was secured with tape to the shoe or foot. Joint centers were determined as the midpoint of the medial and lateral malleoli (ankle), femoral epicondyles (knee), and using the Bell method (hip) (Bell, Brand et al. 1989). A segmental reference system was used for kinematic data with an Euler angle rotational sequence of Z (flexion/extension) Y' (internal/external rotation) X" (abduction/adduction). Once the subject was instrumented, they performed a 5 minute bike warm-up at a self-selected pace.

Strength Testing

Maximal voluntary isometric contraction (MVIC) of the quadriceps, hamstring, and gastrocnemius was assessed with a dynamometer (model 3; Biodex Medical Inc., Shirley NY). Participants were seated and positioned at a fixed knee flexion angle of 25°. The axis of the dynamometer was aligned with the lateral femoral condyle and resistance pads placed at the distal tibia. Participants were verbally encouraged to maximally extend their knee (quadriceps) or flex their knee (hamstring) for 3 seconds. Each participant completed 3 trials with 30 second rest in between each trial. To assess the MVIC of the gastrocnemius, participants maintained a position of 25° of knee flexion and were strapped into 10° of dorsiflexion. Participants were verbally encouraged to maximally point their foot for 3 seconds, and 3 trials were collected with 30 second rest in between each trial.

The strength testing was conducted to obtain maximal sEMG signal amplitudes for later normalization of the peak muscle amplitudes during the drop landing, as well as to compare dancers and athletes on baseline strength values. The average of the peak torque for each muscle group across the 3 MVIC trials was then used to normalize the mean peak amplitudes of muscle activation obtained during the planned double leg drop jump task. The average peak torque values were also used for demographic comparisons between groups.

Forward Hop Stabilization

Next, participants completed three successful trials of the forward hop stabilization on each limb in both the shod and barefoot condition for a total of 12 trials. A minimum of 30 seconds was provided between trials to reduce the chance of fatigue, but more time was provided if needed. Ground reaction force was analyzed using the time to stabilization technique which has been found to be sensitive and reliable at detecting dynamic postural differences (Ross and Guskiewicz 2003; Ross, Guskiewicz et al. 2005). We chose to examine postural control with this measure because A-P TTS has been shown to be one of the most accurate balance measures to identify group differences between uninjured and injured populations (Ross, Guskiewicz et al. 2009). Specifically, time to stabilization (seconds) was defined as the point at which an unbound third order polynomial fit to the ground reaction force crosses below the range of variation. The range of variation is the max ground reaction force value during the final 5 seconds of the stabilization trial.

Data Reduction

Ground reaction force data was collected in the anterior-posterior and medial-lateral direction using two non-conducting force platforms (Type 4060-130, Bertec Corporation, Columbus OH) at 1000 Hz over a 10 second period and interfaced with Motion Monitor software (Innovative Sports Training, Chicago, IL). The average TTS value from the 3 trials in each condition for the A-P and M-L was used for further analysis.

Double Leg Drop Jump

Five successful trials of the double leg drop jump were performed in shod and barefoot conditions for a total of 10 trials. A minimum of 30 seconds was provided between trials to reduce the chance of fatigue, but more time was provided if needed. We choose to assess postural control through the positioning of the COM at initial ground contact relative to the anterior-posterior positioning of the COP, which represented positioning of the base of support. Examination of kinematics, and kinetics has previously identified ACL injury risk during planned double leg drop jump task (Hewett, Myer et al. 2005); therefore we assessed sagittal and frontal plane kinematics, vertical ground reaction force, as well as sagittal plane extensor moments. Relative energy absorption was assessed as it is a biomechanical measure that quantifies the entire landing phase rather than discrete time points such as the kinematic and kinetic variables (Norcross, Lewek et al. 2013). Pre-activity muscular amplitude was assessed to describe the preparatory neuromuscular control patterns during planned activity between dancers and athletes.

Data Reduction

Over a 3-second interval (.5 seconds prior to ground contact and 2.5 seconds following ground contact) kinematic data were collected at 240Hz using a 8-camera IMPULSE motion tracking system (Phase Space, San Leandro, CA) and kinetic and postural control (COM-COP position) data were collected at 1000Hz with the two non-conducting force platforms as previously described (Shultz, Nguyen et al. 2009; Schmitz and Shultz 2010). Kinematic, kinetic, and postural control data were interfaced with Motion Monitor software (Innovative Sports Training, Chicago, IL) using a fourth-order, zero-lag low-pass Butterworth filter at 12 Hz. Regardless of joint, the following motions were defined as positive: flexion, internal rotation, and adduction. Surface EMG data was collected with Trigno Wireless System (Delsys, Boston, MA) at 1000 Hz, and interfaced with Motion Monitor software. sEMG signals were rectified and filtered using a root mean square algorithm (10-millisecond time constant).

The lab convention is set so that the COM-COP value of 0 represents the COM being directly over the base of support, with a positive number representing the COM being more posterior to the COP, and vice-versa. COM-COP values were extracted at initial ground contact (mm). Hip, knee, and ankle joint sagittal plane, as well as hip and knee frontal plane angles were extracted at initial ground contact (vGRF >10N) and excursions, calculated from initial ground contact to peak vertical center of mass displacement. Intersegmental kinetic data were calculated via an inverse dynamic approach and resultant hip, knee, and ankle internal moments were normalized to weight and height (Nm x BW^{-1} x Ht^{-1}). Net joint powers were calculated as the product of the normalized joint moment and joint angular velocity at each time point. Energy absorption (work done on the extensor muscles) was then calculated by integrating the negative portion of the joint power curve, and reported as normalized to body weight and height (Joules x BW^{-1} x Ht^{-1}). To calculate relative energy absorption at each joint the absolute energy for each individual joint was divided by the total energy absorption across the hip, knee and ankle joint (% of total energy absorption). Muscle activation amplitudes (expressed as %MVIC) acquired from each of the 6 muscle sites during a 150ms time window prior to ground contact, were normalized to the MVIC amplitude for each respective muscle. All variables were calculated as the average value obtained across five trials for the shod and barefoot conditions.

Lower Extremity Perturbation

Participants then underwent assessment of reflex responses during ten (5 internal, 5 external) unanticipated perturbations delivered in a randomized order to eliminate anticipatory responses. The lower extremity perturbation device used in this study was chosen over the model used by Simmons (Simmons 2005), which only stretched the ankle musculature, to increase the functionality of the task. The order of LEPD releases for each subject can be found in the counterbalance table in Appendix C.

Data Reduction

Surface EMG data (1000Hz) was acquired for 100ms prior and 500ms after cable release, with the onset of perturbation determined by a voltage signal at trigger release. sEMG signals were rectified and filtered using a root mean square algorithm (10-millisecond time constant) and then ensemble averaged across the 5 trials each for internal rotation and external rotation perturbations. Muscular onset times (ms) were then calculated from the ensemble averaged signal as the time point where EMG activity remained 2SD (gastrocnemius, hamstrings) or 1SD (quadriceps) above the mean EMG activity acquired during the 100ms prior to cable release, for 10 milliseconds or longer, and confirmed with visual recognition. If a muscular onset time could not be determined from the ensemble average, the onset time from each individual trial that produced a clear muscular onset was used and then averaged together.

Statistical Approach

All dependent and predictor variables were entered into Excel then transferred to SPSS for analysis. The following statistical approaches were used to test each of the following hypotheses.

Hypothesis 1: Dancers will require significantly less time to stabilize the ground reaction force following a hopping task

To test hypothesis 1, a 2 (group) x 2 (plane) x 2 (limb) x 2 (footwear) repeated measures ANOVA compared dancers and athletes on TTS in the anterior-posterior (A-P) and medial-lateral (M-L) plane on the dominant and non-dominant limbs during both shod and barefoot conditions.

> <u>*Hypothesis 2a*</u>: *Dancers will position their center of mass (COM) closer to the location of the center of pressure (COP) at initial ground contact following a drop jump task compared to athletes.*

To test Hypothesis 2a, a 2 (group) x 2 (footwear) repeated measures ANOVA compared dancers and athletes on COM to COP displacement in the A-P plane during both shod and barefoot condition.

> <u>*Hypothesis 2b:*</u> *Dancers will land from a drop jump with greater ankle plantar flexion, and similar hip and knee flexion compared to athletes*

To test hypothesis 2b, multivariate ANOVA's compared dancers and athletes on hip, knee, and ankle flexion during both shod and barefoot conditions. Separate MANOVAs examined joint flexion at 1) initial ground contact and 2) for total joint excursions (initial ground contact to peak center of mass displacement).

> <u>*Hypothesis 2c:*</u> *Dancers will land from a drop jump with less frontal plane hip and knee motion compared to athletes.*

To test hypothesis 2c, multivariate ANOVAs compared dancers and athletes on frontal plane hip and knee kinematics (knee valgus and hip adduction) during shod and barefoot conditions. Separate MANOVAs examined joint angles at 1) initial ground contact and 2) for total joint excursions (initial ground contact to peak center of mass displacement).

> <u>*Hypothesis 2d:*</u> *Dancers will demonstrate lower vGRF values and peak knee extensor moments compared to athletes.*

To test hypothesis 2d, a repeated measures ANOVA's compared dancers and athletes on vGRF during both shod and barefoot conditions. A separate multivariate ANOVA compared dancers and athletes on hip, knee, and ankle peak extensor moments during both shod and barefoot conditions).

<u>*Hypothesis 2e:*</u> *Dancers will absorb a larger relative amount of total energy at the ankle joint compared to the knee joint than female athletes.*

To test hypothesis 2e, a multivariate ANOVA compared dancers and athletes on hip, knee and ankle energy absorption during both shod and barefoot conditions).

<u>*Hypothesis 2f:*</u> *Dancers will demonstrate higher hamstring amplitude prior to ground contact during a drop jump task compared to athletes.*

To test hypothesis 2f, a 2 (group) x 2 (footwear) x 6 (muscle) ANOVA compared dancers and athletes on pre-landing activation amplitude of the medial and lateral quadriceps, hamstring and gastrocnemius muscles during the double leg drop jump during shod and barefoot conditions.

<u>*Hypothesis 3:*</u> *Dancers will activate musculature significantly quicker than athletes*

To test hypothesis 3, a 2 (group) x 6 (muscle) x 2 (perturbation direction) ANOVA compared dancers and athletes on muscular onset time of the medial and lateral quadriceps, hamstring and gastrocnemius muscles during both internal rotation and external rotation perturbations.

Power Analysis

All analyses were evaluated at $p \leq .05$. Preliminary data comparing AP TTS between dancers and recreational athletes reported a large effect size (d=1.6), with dancers stabilizing the ground reaction force significantly faster than recreational athletes. Another pilot study examining the muscular activation timing of the medial and lateral hamstrings in dancers compared to athletes following an unanticipated lower extremity perturbation revealed a large effect size (d=.82) for dancers activating their muscles quicker than recreational athletes. Based on the smallest effect size from the preliminary data, our power analysis revealed that a sample size of 28 total participants (14 participants per group) would achieve a value of .80 power. Since the preliminary data is based on comparisons between dancers and recreational athletes, and all variables were not assessed in preliminary data, a sample size of 40 participants (20 participants per group) was used to ensure adequate power.

CHAPTER IV

RESULTS

Forty collegiate females, 20 dancers (age= 20.4 ± 1.9 yrs, height= 164.8 ± 6.1 cm, weight= 63.5 ± 8.8kg, experience= 14.3 ± 3.9 yrs) and 20 athletes (age= 19.4± .9 yrs, height= 169.3 ± 7.1 cm, weight= 69.8 ± 13.0 kg, experience= 12.2 ± 2.9 yrs), participated and completed all aspects of the study. Comparative demographic data for each group are presented in Table 1. Histograms graphically depicting the distributions for all dependent variables with measures of central tendencies can be found in Appendix D. Long latency reflex is dependent on the length or distance an action potential must travel to reach the motor unit, and in taller individuals, this distance is longer (Basmajian and De Luca 1985). To ensure differences in muscle reflex between the two groups were not due to the group difference in height, this variable was included as a covariate in the statistical model for hypothesis 3. Height was not included as a covariate in statistical analyses for the forward hop or drop jump task as there is no literature to suggest the height affects kinematics, postural control, vGRF or muscular amplitude. Height is already accounted for in the drop jump task for kinetics (peak extensor moment, relative energy absorption) through standard normalization procedures. Complete results for all analyses can be found in Appendix E

Table 1. Mean (SD) of Demographics Variables and the associate *p*-value

Demographic Data	**Dancer**	**Athlete**	***p*-value**
Age (yrs)	20.4 (1.9)	19.4 (.9)	.55
Height (cm)	164.8 (6.1)	169.3 (7.1)	.04
Weight (kg)	63.5 (8.8)	67.1 (7.6)	.17
Experience (yrs)	14.3 (3.9)	12.2 (2.9)	.06
Quadriceps Strength (Nm/kg)	2.28 (.33)	2.58 (.50)	.03
Hamstring Strength (Nm/kg)	1.20 (.21)	1.46 (.33)	.01

Hypothesis 1: Dancers will Require Significantly Less Time to Stabilize the Ground Reaction Force Following a Hopping Task

The means and standard deviations for each time to stabilization measure stratified by group, footwear and limb dominance can be found in Table 2. The full ANOVA model can be found in Appendix E.1. There was no main effect for group [F (1, 38) = 3.1, *p*=.08, Partial Eta Squared (η_p^2) =.08], but there was a significant Plane*Footwear* Group interaction [F (1, 38) = 5.5, *p*=.03, η_p^2 = .13] (Means and standard deviations shown in Table 3). The means show that dancers require shorter time to stabilize in both planes under both footwear conditions; however, dancers and athletes TTS values are influenced in the A-P and M-L plane differently by footwear. Specifically athletes stabilize 20% faster when barefoot in the A-P plane yet 11% quicker in shoes in the M-L plane. Differences in stabilization times are not as pronounced in dancers, as they stabilize 5% quicker in the A-P plane while barefoot and 4% quicker in the M-L plane while in shoes. To further analyze this interaction we calculated the delta score from footwear in each plane (AP Shod – AP Barefoot and ML Shod – ML Barefoot) to create a variable that encompassed the difference in footwear across both planes. A repeated measure 2 (AP_Delta/ML_Delta) x 2 (Dancer/Athlete) ANOVA was run and revealed group*delta_score interaction [F (1, 78) = 3.81 *p*=.05, η_p^2=.05]. A graphical representation of the 3-way interaction is shown in Figure 4.

Table 2. Mean (SD) of TTS between dancers and athletes in the AP and ML direction

Footwear	**Limb**	**AP (secs)**		**ML (secs)**	
		Dancer	**Athlete**	**Dancer**	**Athlete**
Shod	Dominant	1.95 (.86)	2.77 (1.29)	1.50 (.92)	1.80 (1.34)
	Non-Dominant	2.40 (1.19)	2.76 (1.18)	1.22 (.82)	1.64 (1.07)
Barefoot	Dominant	1.85 (.61)	1.96 (.48)	1.80 (1.38)	2.47 (1.89)
	Non-Dominant	2.32 (1.22)	2.48 (1.11)	1.02 (.69)	1.38 (.78)
Ensemble		2.13 (1.00)	2.49 (1.12)	1.39 (1.00)	1.82 (1.36)

Table 3. Mean (SD) of Footwear-Plane Difference Scores between dancers and athletes

*Indicates dancers ≠ athletes (P<.05)

Footwear by Plane Difference Score	**Dancer**	**Athlete**
AP_Delta	.09 (1.39)	.54 (1.01)
ML_Delta	.05 (1.11)	.21 (1.27)
Ensemble*	.02 (1.31)	.17 (1.29)

Hypothesis 2a: Dancers will Position their Center of Mass Closer to the Base of Support Following a Drop Jump Task Compared to Athletes

The following results describe the group differences between the anterior-posterior (A-P) positioning of the center of mass in relation to the center of pressure (COP) which at initial ground contacts represents the location of our base of support (BOS). Means, standard deviations and effect sizes can be found in Table 4. The full ANOVA model can be found in Appendix E.2. The 2x2 ANOVA revealed a main effect for group [F (1, 38) = 4.84, *p*=.03, η_p^2 = .113]. While both groups landed with the COM posterior to the COP, the COM in dancers was positioned closer to neutral alignment compared to the athletes. This group difference was not affected by shoe condition [F (1, 38) = .68, *p*=.42, η_p^2 = .017].

Table 4. Mean (SD) of sagittal plane COM positioning relative to the COP position

Condition	**COM –COP Position (m)**	
	Dancer	**Athlete**
Shod	.18 (.03)	.19 (.03)
Barefoot	.18 (.04)	.20 (.03)
Ensemble*	.18 (.03)	.20(.03)

*Indicates dancers ≠ athletes (P<.05)

Hypothesis 2b: Dancers will Land from a Drop Jump with Greater Ankle Plantar Flexion, and Similar Hip and Knee Flexion Compared to Athletes

Sagittal plane joint angles at initial contact and total excursions stratified by group and footwear can be found in Table 5. Full MANOVA comparing dancers and athletes on joint flexion at initial ground contact and joint flexion excursions can be found in

Appendix E.3and Appendix E.4, respectively. Multivariate statistics revealed no group main effects [F (3,36) = 2.21, *p*=.104, η_p^2=.16] or group by footwear interactions [F (3,36) = 1.34, *p*=.277, η_p^2=.10] for joint angles at initial contact. There was a significant group main effect for joint excursions [F (3,36) = 4.6, *p*=.01, η_p^2=.28]. Follow up univariate analyses revealed that dancers moved through 12% greater ankle dorsiflexion compared to athletes (63.0 ± 8.8°, 55.6 ± 8.5° respectively) [F (1,38) = 12.1, *p*=.001, η_p^2 =.24] but went through similar knee (dancer = 72.6± 12.6°, athlete = 67.4 ± 10.6°) [F (1,38) = 2.57, *p*=.117, η_p^2=.06] and hip (dancer = 54.1 ± 14.5°, athlete = 52.4 ± 19.2°) [F (1,38) = .11, *p*=.748, η_p^2=.00] motion. Group differences in joint excursions were not affected by footwear [Group*Footwear interaction = F (1,36) = 1.2, *p*=.33, η_p^2=.09].

Table 5. Mean (SD) of sagittal plane hip knee and ankle position at ground contact and total joint excursion stratified by group, joint, and shod condition

Footwear	**Flexion (°)**	**Ground Contact**		**Excursion***	
		Dancer	**Athlete**	**Dancer**	**Athlete**
Shod	Hip	17.1 (12.7)	25.6 (13.6)	53.5 (13.9)	51.9 (21.2)
	Knee	10.6 (6.1)	13.2 (6.3)	71.9 (8.9)	67.0 (12.4)
	Ankle	43.4 (6.8)	48.6 (7.1)	65.5 (6.0)	60.5 (6.1)
Barefoot	Hip	22.2 (16.4)	21.8 (13.6)	54.6 (15.3)	52.9 (18.2)
	Knee	9.8 (6.4)	13.8 (5.5)	73.3 (16.0)	67.8 (9.1)
	Ankle	43.5 (6.7)	47.2 (8.3)	60.4 (10.6)	50.6 (7.9)
Ensemble	Hip	19.6 (14.5)	23.7 (13.4)	54.05 (14.5)	52.4 (19.2)
	Knee	10.2 (6.1)	13.5 (5.8)	72.60 (12.6)	67.4 (10.6)
	Ankle	43.5 (6.6)	48.0 (7.4)	63.0 (8.8)*	55.6 (8.5)

*Indicates dancers ≠ athletes (P<.05)

Hypothesis 2c: Dancers will Land from a Drop Jump with Less Motion in the Frontal Plane at the Knee and Hip Compared to Athletes

Frontal plane joint angles for initial contact and total excursions stratified by group and footwear can be found in Table 6. Full MANOVA statistics comparing dancers and athletes on frontal plane position at ground contact can be found in Appendix E.5. The full MANOVA statistics for frontal plane excursions can be found in Appendix E.6. Multivariate statistics revealed no group main effects [F (2, 37) = 2.1 p=.14, η_p^2=.10] or group*footwear interactions [F (2, 37) = .18 p=.84, η_p^2=.01] for initial contact position. Similarly there was no group main effects [F (2, 37) = 1.6 p=.23, η_p^2=.08] or group*footwear interactions [F (2, 37) = .003 p=1.0, η_p^2<.001] for excursion.

Table 6. Mean (SD) of frontal plane hip and knee position at ground contact and total joint excursion. (+) = Adduction; (-) = Abduction; (+) = Varus; (-) = Valgus

Footwear	**Abduction(°)**	**Ground Contact**		**Excursion**	
		Dancer	**Athlete**	**Dancer**	**Athlete**
Shod	Hip	-8.8 (3.6)	-10.8 (3.1)	7.2 (2.4)	8.2 (2.8)
	Knee	1.4 (4.1)	2.8 (5.5)	10.7 (6.8)	9.6 (6.2)
Barefoot	Hip	-8.3 (5.2)	-10.7 (4.4)	7.32 (4.4)	8.4 (3.3)
	Knee	1.8 (4.6)	2.5 (6.3)	9.3 (4.9)	8.1 (4.1)
Ensemble	Hip	-8.5 (4.3)	-10.7 (3.7)	7.3 (3.8)	8.3 (4.1)
	Knee	1.6 (4.2)	2.7 (5.8)	10.0 (5.6)	8.8 (5.3)

Hypothesis 2d: Dancers will Demonstrate Lower vGRF Values and Knee Extensor Moments Compared to Athletes

Descriptive statistics for vGRF stratified by group and footwear can be found in Table 7. Complete analyses can be found in Appendix E.7. There was a significant group main effect [F (1, 38) = 5.18 p=.03, η_p^2=.12] with female dancers landing with 15% less vGRF than athletes. This group difference was not affected by an interaction with footwear [F (1, 38) = 2.64, p=.11, η_p^2=.07].

Table 7. Mean (SD) of vGRF with and without shoes

vGRF(%BW)	Dancer	Athlete
Shod	2.09 (.52)	2.56 (.66)
Barefoot	2.08 (.45)	2.36 (.56)
Ensemble*	2.09 (.48)	2.46 (.61)

*Indicates dancers ≠ athletes (P<.05)

Means and standard deviations for peak joint extensor moments stratified by group and footwear condition are listed in Table 8. The full MANOVA statistics can be found in Appendix E.8. Multivariate statistics revealed no main effect for group [F (3, 36) = 1.80, p = .17, η_p^2 = .16] and no Group*Footwear interaction [F (3, 36) = .14, p = .94, η_p^2 = .01]. Despite no significant group differences, it is worth noting that dancers had 19% lower peak ankle joint extensor moments compared to athletes.

Table 8. Mean (SD) of peak internal extensor moment at the hip knee and ankle with and without shoes

Footwear	**Joint**	**Peak Moment (Nm/BW*Ht)**	
		Dancer	**Athlete**
Shod	Hip	.086 (.027)	.093 (.031)
	Knee	.116 (.021)	.122 (.031)
	Ankle	.065 (.167)	.081 (.024)
Barefoot	Hip	.084 (.030)	.094 (.036)
	Knee	.102 (.023)	.104 (.028)
	Ankle	.067 (.022)	.081 (.024)
Ensemble	Hip	.085 (.028)	.094 (.033)
	Knee	.109 (.023)	.113 (.030)
	Ankle	.066 (.020)	.081 (.023)

Hypothesis 2e: Dancers will Absorb a Larger Relative Amount of Total Energy at the Ankle Joint Compared to the Knee Joint than Female Athletes

Means and standard deviations for work absorption values stratified by group, joint and footwear are listed in Table 9. The full MANOVA statistics comparing dancers and athletes on hip, knee and ankle energy absorption can be found in Appendix E.9. Multivariate statistics revealed no significant group main effect [F (2, 37) = .37, p = .69, η_p^2 = .02] or group by footwear interactions [F (2, 37) = 1.45, p = .25, η_p^2 = .07].

Table 9. Mean (SD) of relative energy absorption (EA) at the hip knee and ankle with and without shoes

Footwear	Joint	Relative EA (%)	
		Dancer	Athlete
Shod	Hip	26.3 (12.3)	25.6 (14.8)
	Knee	28.1 (10.1)	26.9 (11.5)
	Ankle	45.6 (10.6)	47.5 (15.7)
Barefoot	Hip	24.0 (9.7)	30.6 (16.7)
	Knee	26.6 (15.3)	24.2 (10.2)
	Ankle	49.4 (2.9)	45.2 (14.5)
Ensemble	Hip	25.1 (10.9)	28.1 (15.6)
	Knee	27.4 (12.7)	25.5 (10.7)
	Ankle	47.5 (11.7)	46.3 (14.8)

Hypothesis 2f: Dancers will Demonstrate Higher Hamstring Amplitude prior to Ground Contact during a Drop Jump Task Compared to Athletes

Means and standard deviations for pre-contact muscular amplitude values stratified by group, joint and footwear are listed in Table 10. The full MANOVA statistics compared dancers and athletes on medial and lateral gastrocnemius, hamstrings, and quadriceps pre-landing muscle activation amplitude 150ms prior to ground contact can be found in Appendix E.10. During data collection, EMG signals were not obtained on 2 subjects, and therefore the sample size for this analysis is N=38 (both subjects from the dancer group). Multivariate statistics revealed no significant group main effect [F (1, 36) = .26, p = .61, $\eta_p^2 < .01$] or group by footwear interactions [F (1, 36) = .01, p = .91, $\eta_p^2 < .01$].

Table 10. Mean (SD) of pre-contact muscular activation (%MVIC) at the LG = Lateral Gastrocnemius, MG = Medial Gastrocnemius, LH = Lateral Hamstrings, MH = Medial Hamstrings, LQ = Lateral Quadriceps, MQ = Medial Quadriceps

	Shod		**Barefoot**	
Muscle	**Dancer**	**Athlete**	**Dancer**	**Athlete**
LG	5.62 (3.5)	5.49 (2.8)	6.05 (4.4)	4.87(6.8)
MG	5.17 (3.0)	5.57 (3.7)	4.82 (3.0)	4.16 (2.2)
LH	4.36 (2.7)	4.02 (2.8)	4.38 (2.7)	4.93 (2.9)
MH	4.92 (2.3)	3.81 (2.4)	5.84 (2.9)	3.97 (2.4)
LQ	12.17 (5.9)	10.34 (5.3)	12.96 (5.7)	11.49 (5.8)
MQ	13.51 (9.7)	13.92 (8.9)	13.60 (7.1)	15.21 (8.8)

Hypothesis 3: Dancers will Activate the Hamstring Musculature Significantly Quicker than Athletes

Means and standard deviations for muscular onset time stratified by group, perturbation direction, and muscle site are listed in Table 11. The full MANOVA statistics comparing dancers and athletes on reflex onset times for the medial and lateral gastrocnemius, hamstrings and quadriceps can be found in Appendix E.11. Including height as a covariate did not have a significant effect on the model, and was therefore excluded. Multivariate statistics revealed no significant group main effect [F (1, 38) = .05, p = .83, $\eta_p^2 < .01$], group by perturbation direction interaction [F (1, 38) = .02, p = .89, $\eta_p^2 < .01$], group by muscle interactions [F (5, 34) = 1.54, p = .20, $\eta_p^2 = .19$], or group by perturbation direction by muscle interaction [F (5, 34) = .18, p = .97, $\eta_p^2 = .03$].

Table 11. Mean (SD) of muscular onset (msec) at the LG = Lateral Gastrocnemius, MG = Medial Gastrocnemius, LH = Lateral Hamstrings, MH = Medial Hamstrings, LQ = Lateral Quadriceps, MQ = Medial Quadriceps

	Internal Rotation		**External Rotation**	
Muscle	**Dancer**	**Athlete**	**Dancer**	**Athlete**
LG	46.1 (11.4)	41.6 (5.8)	48.0 (9.6)	42.1 (6.8)
MG	45.5 (11.2)	44.8 (7.1)	49.5 (8.6)	46.8 (9.2)
LH	75.8 (12.4)	77.7 (10.0)	74.0 (10.6)	75.7 (9.3)
MH	75.3 (11.6)	79.1 (9.7)	76.0 (11.3)	80.2 (12.5)
LQ	95.3 (11.8)	95.2 (11.4)	97.2 (16.7)	96.3 (10.3)
MQ	98.5 (12.7)	95.9(11.7)	99.0 (16.5)	101.2 (2.2)

CHAPTER V

DISCUSSION

Preliminary research suggests that as a function of their training experience female dancers may use neuromuscular control patterns that allow for more protection around the knee joint during functional activities, and subsequently lower their risk for ACL injury, compared to female athletes. Specifically, the decreased ACL injury risk in dancers is thought to result from their softer landings (McNitt-Gray, Koff et al. 1992; Liederbach, Dilgen et al. 2008; Orishimo, Kremenic et al. 2009), more neutral alignment during jumping (Liederbach, Dilgen et al. 2008; Ambegaonkar, Shultz et al. 2009; Orishimo, Kremenic et al. 2009), decreased postural sway (Crotts, Thompson et al. 1996; Simmons 2005; Gerbino, Griffin et al. 2007), and greater years of training experience (Orishimo, Kremenic et al. 2009). Additionally, dancers consistently use augmented feedback during dance training which has been shown to not only assists in the development, but also the retention of protective movement patterns (Sigrist, Schellenberg et al. 2011). However, no prior published literature has provided a comprehensive comparison of the neuromuscular profiles between these two populations to determine if training practices in dancers should be further investigated for its potential to inform ACL prevention efforts. Therefore, the purpose of this study was to comprehensively compare neuromuscular control profiles in female dancers and athletesduring both planned (time to stabilization dynamic landing task; drop jump landing) and unplanned (lower extremity perturbation) functional tasks. The primary findings were that dancers performed a drop jump landing with 15% lower ground reaction forces, more anterior positioning of the COM, and greater sagittal plane ankle range of motion than female athletes. However we observed no significant differences in their ability to dynamically stabilize during a single leg landing or in reflex response characteristics to a functional perturbation. Therefore, the research hypotheses were only partially supported, although a number of non-significant trends were noted in the data that were consistent with expected findings. The discussion will summarize the influence of footwear on the overall study findings, discuss the findings for each task, and then consider the overall findings as to how they advance theory and clinical practice.

Influence of Footwear

This study examined neuromuscular control patterns in both shod and barefoot conditions to ensure that different familiarity to footwear did not bias the findings. This is because dancers typically perform their activities barefoot while athletes wear athletic shoes, and footwear has been shown to influence lower extremity mechanics during landing, specifically, increased ankle joint stiffness, lower vGRF and increased knee flexion excursion when barefoot compared to shod during a drop jump (Shultz, Schmitz et al. 2012).

Our results showed that footwear did not interact with the neuromuscular control patterns of the groups during the planned drop jump task. However, during the planned forward hop task, footwear decreased athletes' ability to stabilize GRF by 20% when wearing shoes in the A-P plane, yet increased athletes' ability to stabilize GRF in the M-L plane by 11% when wearing shoes. Differences in stabilization times in the A-P and M-L plane between shod and barefoot conditions only ranged between 4-5% in dancers. This is possibly because dancers may or may not use footwear depending on the style of dance, whereas athletes always use athletic shoes. This suggests that the unfamiliarity with landing and balancing while barefoot for athletes caused a larger variability in their dynamic balance results, whereas dancers were not as affected by footwear condition due to exposure to jumping and landing in both conditions.

The lack of footwear influence during the planned drop jump task suggests that the neuromuscular differences identified between these two populations are not due to footwear condition in their respective sport/activity. During this task future studies could compare both groups in either footwear condition without concern of it altering their natural movement patterns. However, when testing a forward hop dynamic balance task, group differences can occur based on the shoe condition, with athletes results being more variable across footwear condition. Due to the limited difference in performance across footwear condition in the dance group, there is rationale that future studies examine both groups in the shod condition. This is because dancers' postural control will not be influenced by footwear; therefore, the use of the shod condition will reflect the dynamic balance of athletes and dancers during activity to allow for the best comparison between groups.

Forward Hop Task

Previous research has reported that individuals with decreased postural control are more likely to go on to sustain an ACL tear (Vrbanic, Ravlic-Gulan et al. 2007). Further the posterior or lateral positioning of the COM has also been shown to be linked to increased risk for injury (Griffin, Agel et al. 2000; Hewett, Torg et al. 2009; Sheehan, Sipprell et al. 2012). Because prior research suggests that static and dynamic postural control is far more stable in dancers compared to non-dancers and athletes (Crotts, Thompson et al. 1996; Hugel, Cadopi et al. 1999; Schmit, Regis et al. 2005; Simmons 2005; Gerbino, Griffin et al. 2007; Ambegaonkar, Caswell et al. 2013), our expectation was that female dancers would stabilize the A-P and M-L GRFs significantly faster than athletes as this would represent a more stable postural control system.

The TTS values obtained in the current investigation are comparable to other results published using an unbound 3rd order polynomial calculation which range from 1.35 –2.33 seconds in the A-P plane and 1.56- 2.00 seconds in the M-L plane (Ross and Guskiewicz 2004; Ross, Guskiewicz et al. 2005). However, in contrast to previous research identifying more stable postural control in dancers compared to athletes, our hypotheses were not supported. While dancers generally displayed a trend toward 14% and 24% faster stabilization times in the AP and ML planes compared to athletes, these differences did not reach a level of statistical significance (p= .08). The lack of significant difference we observed compared to previous research may be due to the matching of our groups on activity level and experience.

Because balance has been shown to be affected by activity levels (Ferreira, Sherrington et al. 2012), with higher levels of physical activity being associated with a more stable posture, we advanced the literature by ensuring our groups were better matched on activity levels than previous studies who typically have compared dancers to non-dancers. For example, Crotts et al (Crotts, Thompson et al. 1996), who compared adolescent dancers to age matched non-dancers reported that the dance group could maintain a single limb stance longer than non-dancers. Similarly, Ambegaonkar et al (Ambegaonkar, Caswell et al. 2013), compared dancers to non-dancers reporting that dancers had significantly fewer errors on the Balance Error Scoring System (BESS) as well as greater reach distances in the medial and posterior-medial direction on the star excursion balance test (SEBT). Ambegaonkar et al (Ambegaonkar, Caswell et al. 2013), also noted that the SEBT reach distances reported for dancers were comparable to previous studies who examined the SEBT on athletes. Thus, it is possible that previously reported postural control differences in dancers reflect a difference in activity or skill level between groups. Further studies examining dynamic balance of dancers should match on physical activity and skill level to ensure statistical differences are due to neuromuscular control patterns and not varying levels of physical activity.

The current investigation controlled for level of physical activity by recruiting collegiate athletes or dance majors or minors. Collegiate athletes were recruited from Division I and III Universities in the Greensboro area. Collegiate dance majors are required to audition to confirm their skill and technique is acceptable at the collegiate level; while minors are required to take a minimum of 6 credit hours, providing a wide range of skill at the collegiate level for both athletes and dancers alike. Furthermore, all participants were required to have a minimum of 5 years' experience in their respective activity as well as currently participating a minimum of 90 minutes a week. Demographic data confirms that our dancers and athletes had similar years of experience and activity level.

Many researchers have suggested enhanced postural control in dancers contributes to their lower injury rate compared to athletes (Liederbach, Dilgen et al. 2008; Orishimo, Kremenic et al. 2009; Ambegaonkar, Caswell et al. 2013; Orishimo, Liederbach et al. 2014). However, the lack of difference between these two populations implies that alone may not explain why dancers exhibit a lower ACL injury rate compared to athletes. This is partially supported by the lack of balance training interventions that have shown to decrease ACL injury occurrence in female athletes (Yoo, Lim et al. 2010). This suggests that the examination of other neuromuscular factors may shed light on the injury rate difference between these female populations.

Planned Drop Jump Landing

Despite the literature suggesting that landing with the knee and hip extended is more risky for the ACL, we hypothesized that dancers would initially land more extended but would better absorb the forces of landing by positioning the COM more anteriorly (thus increasing ankle extensor moments and energy absorption while decreasing knee extensor moments and energy absorption), and moving through a greater range of motion.

Our results in part support the original hypothesis that female dancers landed with similar lower extremity positioning at initial ground contact, greater relative anterior positioning of their center of mass, moved through greater range of motion, particularly at the ankle joint, and had 15% lower vGRF compared to athletes. Although we expected a more neutral alignment in the frontal plane, our results did not report a significant difference at the hip or knee for initial contact or excursion in the frontal plane between groups. Our results also did not identify significant differences between groups on peak extensor moments with only an 18%, 4% and 10% lower peak ankle, knee and hip extensor moments, respectively, in dancers versus athletes. Further, approximately 47.5% of energy absorbed in dancers occurred at the ankle joint with the knee and hip contributing 27.4% and 25.1% respectively. However there were no group differences in energy absorption as the athletes also attenuated nearly half of the landing force at the ankle joint (46.3%) with the knee and hip absorbing 25.5% and 28.1% respectively. Finally there were no differences in pre-ground contact muscular activation between the groups. Despite not fully supporting each individual hypothesis, the results for planned motion during a drop jump task generally support the theory that dancers demonstrate a more protective neuromuscular control pattern compared to female athletes. Further, this protective movement pattern is driven by decreasing landing forces, rather than decreasing forces solely about the knee joint. The following section will provide an integrative view on a neuromuscular control pattern exhibited by female dancers that may contribute to their low risk of injury.

Neuromuscular Control Patterns in Dancers

A forward positioned COM has been related to a decrease in quadriceps activation, increase in hamstring activation, lower vGRF, as well as, lower knee extensor moments with an increase in ankle and hip moments (Blackburn and Padua 2009; Shimokochi, Lee et al. 2009; Kulas, Hortobagyi et al. 2010). Similar to a recent publication comparing dancers and athletes on trunk positioning during a single limb landing, the dance population in the current investigation positioned their trunk closer to neutral, yet the COM was still posterior to the COP position (Orishimo, Liederbach et al. 2014). Orishimo et al (Orishimo, Liederbach et al. 2014), also reported equal pre-ground contact muscular activation amplitudes between female dancers and athletes similar to the current investigation.

A possible rationale for the lack of increased hamstring muscular amplitude in the dance population is the relationship between muscular activation and lower extremity positioning. Activation of the lower extremity musculature prior to ground contact directly influences the positioning of the hip, knee, and ankle joint (Palmieri-Smith, Woitys et al. 2008). We hypothesized that dancers and athletes would land in a similar upright position. This initial positioning with the knee near full extension would suggest great quadriceps activation to extend the knee rather than hamstring activation that would flex the knee. Since our results showed no difference in initial ground contact positioning between the groups in the sagittal plane, it is also plausible that the lack of muscular activation difference between groups is due to the similar initial positioning of the lower extremity at ground contact.

An upright landing is thought to be a "higher risk" position; however, the dance population also demonstrated this landing position and has a lower risk of injury, suggesting this may not be the primary contributor to the injury rate disparity in female athletes. Consider the alternative, that a near full extension landing position was utilized by dancers to allow for a greater range of motion to decelerate the landing, thus lowering the loads applied to the system. This relationship was driven by 11% greater motion at the ankle joint. Future research should investigate the temporal aspect of the landing phase between these two populations to determine if dancers increase stance time during the landing phase as they move through a greater range of motion. This will be important to examine as a stance time during landing phase has been prospectively linked to ACL injury risk (Hewett, Myer et al. 2005).

Vertical ground reaction force has been suggested as an injury risk predictor as subjects who reported a 20% higher vGRF went on to sustain an ACL tear (Hewett, Myer et al. 2005). Joint range of motion directly affects impact landing, with stiff landings characterized by a more erect final position, rather than initial positioning (Devita and Skelly 1992; McCaw and Cerullo 1999; Zhang, Bates et al. 2000). In this study, female dancers had a 15% lower vGRF compared to athletes which may be attributed to increased sagittal plane joint excursions specifically at the ankle joint.

The lower vGRF in dancers compared to athletes may suggest a decrease in peak extensor moment across all joints, as there is a lower impact force to counteract. The more neutral COM alignment, as well as, erect initial contact position in dancers also could contribute to a lower peak extensor moment as this body positioning in theory should decrease the moment arm length as the segments move closer to the joint centers (Derrick 2004). Our results did not support these theories, although there was a trend for decreased peak extensor moments across the hip (10%), knee (4%), and ankle (18%) joints in the dancers compared to athletes.

Kinematic and kinetic variables have complex interactions in that there are different methods by which the lower extremity can attenuate forces during landing. Despite the lower vGRF and more anterior COM, dancers had an insignificant difference in peak extensor moment compared to athletes. Hewett et al (Hewett, Stroupe et al. 1996), noted similar results following a plyometric training program that lowered vGRF yet resulted in no change in knee extensor moment. The female subjects in the aforementioned study did significantly reduce their knee adduction moment which was significantly associated with the peak landing force. Although we did not examine frontal plane moments in the current investigation, future investigations should look into these variables to determine if the decrease in vGRF are associated with a decrease in frontal plane moments.

Energy absorption is a biomechanical measure that quantifies the eccentric action of musculature during the deceleration phase of landing and it is a product of joint moments and joint velocity (Devita and Skelly 1992; McNittgray 1993; Zhang, Bates et al. 2000; Schmitz, Kulas et al. 2007). The dancers in the current investigation increased joint range of motion yet trended towards a decrease in joint moments, thus it is plausible that the slight decrease in joint moments and increase in joint excursions resulted in no difference in the relative energy absorption between the groups.

As we measured relative rather than absolute energy absorption, our findings suggest dancers and athletes use similar patterns for attenuating the landing force. The lower vGRF, which infers a lower total force exerted on the body in the dance group compared to athletes, may imply that although the relative energy absorption is the same, the absolute energy absorbed at each joint may be in lower in dancers. This study did not investigate the total work done by each joint, however, our findings indicate this may be a focus of future research as the lower vGRF, greater sagittal plane excursion and trends for lower peak extensor moments all suggest there may be a difference between groups in the total work of the lower extremity.

Finally, dancers and athletes demonstrated similar frontal plane positioning at the hip and knee at initial contact and through excursion. While variables such as vGRF, knee valgus at initial contact, and knee abduction moment have been prospectively linked to ACL injury (Hewett, Myer et al. 2005), Smith et al (Smith, Johnson et al. 2012), found no relationship between an increase in lower extremity scoring system (LESS) and ACL injury. The LESS is a scoring construct that counts landing technique errors such as stiff landings, valgus positioning or tibial rotation (Padua, Boling et al. 2011). A limitation of the LESS is there are a large number of items and includes some items that are infrequently endorsed (Padua, Boling et al. 2011). When assessing knee valgus using the LESS, 5 points can be assessed for the knee valgus motion (2 points for identifying a large frontal plane motion at the knee, 2 points for overall landing impression which states to deem landing as poor if there are large frontal plane motions at the knee, and 1 point for foot rotation which could be demonstrated in combination with the valgus positioning). In the study by Smith et al (Smith, Johnson et al. 2012), LESS scores ranged from 0-11, meaning that if a subject demonstrated knee valgus and received all 5 possible points for this motion, nearly half of their score was weighted on one joint motion. If this were the case, and the LESS was unable to predict ACL injury, it would question the predictability of knee valgus positioning. Although this must be taking with caution as the actual scoring for each subject was not published. Therefore we do not know if the scoring was heavily weighted on knee valgus or other items. Thus more research is needed to determine if the frontal plane motion demonstrated by female dancers and athletes alike is a critical factor in predicting injury risk, or if it is a function of anatomical differences between males and females.

Collectively our results show small differences in individual variables where each alone may not reach significance, but overall combine to describe a movement pattern that is more protective. Specifically, dancers had 15% lower vGRF; while this did not translate to significantly lower peak extensor moments, dancers did demonstrate a 19% decrease in ankle extensor moment and 12% greater sagittal plane ankle excursion. A single neuromuscular variable alone does not result in ACL injury as it is a multi-planar phenomenon (Quatman and Hewett 2009); therefore, the isolated assessment of neuromuscular variables may minimize the impact of subtle changes across multiple variables leading to an overall change in the gross movement pattern. From our findings we suggest that despite the lack of statistical significant differences in isolated variables, there is evidence that dancers make subtle adjustments in their neuromuscular control patterns that collectively combine to reduce landing forces.

Several implications can be derived from these findings. First it is likely that the extensive landing technique training that dancers undergo, may be partially responsible for the protective neuromuscular control patterns exhibited during a planned double leg drop jump task. Second, the collective neuromuscular pattern demonstrated by collegiate female dancers appears to lower the overall loading of the lower extremity rather than specific decreases of joint loading at the knee. Thus, a driving component of the injury rate difference between dancers and athletes is likely the ability to lower landing force.

The limited literature comparing dancers to athletes has identified neuromuscular differences that suggest protective movement patterns in dancers. Similar to our results, the differences noted focus on overall load reduction rather than decreasing loading specific to the knee joint. Specifically, a study identified dancers as having higher leg spring stiffness, but similar knee joint stiffness compared to basketball athletes (Ambegaonkar, Shultz et al. 2011). The higher leg spring stiffness is suggested to decrease soft tissue injuries as this represents the resistance of the limb to compression by the load (Farley and Morgenroth 1999). In another study dancers exhibited an erect trunk position compared to male and female athletes who positioned their trunk more lateral and posterior during a single leg landing; however, they also found no difference between groups in quadriceps/hamstring ratio, and peak knee flexion moment between the groups (Orishimo, Liederbach et al. 2014).

Similar to the current investigation, not all neuromuscular variables that are suggested to be “high-risk” movements differ between female dancers and female athletes. Hence the variables previously thought to be driving the high rate of ACL injury in female athletes, may not be as critical as variables assessing overall loading. Future investigations need to assess if athletes have decreased stance time compared to dancers as this may be another overall lower extremity loading variable driving the injury rate difference as this has been prospectively linked to increased injury risk (Hewett, Myer et al. 2005).

Unanticipated Functional Perturbation

Early lateral hamstring activation is thought to play a role in stabilizing the tibia by preventing excessive anterior tibial translation and ACL strain once the quadriceps are fully active (Huston and Wojtys 1996; Fujii, Sato et al. 2012). Given the strength differential between the quadriceps and hamstring musculature, a longer delay in quadriceps activation may provide the weaker hamstrings more time to reach peak amplitude for maximum force production to stabilize the tibia before an anterior translational force is created by the quadriceps (Huston and Wojtys 1996). Previous literature has shown that female dancers had quicker gastrocnemius long latency reflex compared to non-dancers during an unanticipated ankle perturbation (Simmons 2005). It was proposed that dance training and its focus on postural stability and forward upright posture allowed for quicker activation of the posterior gastrocnemius musculature. Their findings are supported by other studies who have identified that stability training can decrease the reflex time of ankle musculature (Lloyd 2001). From this theory of balance training producing quicker activation of musculature, we anticipated dancers to demonstrate quicker muscular activation onset times across all muscles of the lower extremity.

No difference in muscular onset times was noted between dancers and athletes in either rotation direction. Unlike the work by Simmons (Simmons 2005), in the current investigation, dancers were matched to athletes who had similar levels of physical fitness. Further it has also been shown that agility and plyometric training can increase reflex speed (Wojtys, Huston et al. 1996). Therefore, the equal physical activity level, and both groups participating in training that has been shown to reduce reflex time may have led to the insignificant findings in the current study.

Inclusion of an unplanned task was critical in the assessment of neuromuscular differences between these two populations since a prevailing theory for the low rate of injury in dancers is the performance of choreographed movement (Liederbach, Dilgen et al. 2008; Orishimo, Kremenic et al. 2009). Besier et al (Besier, Lloyd et al. 2001), noted that when individuals are able to prepare for a cutting task, the neuromuscular patterns demonstrated are more protective than when the same task is unplanned. Specifically, females increased frontal and transverse plane moments at the knee during an unplanned cutting task (Besier, Lloyd et al. 2001), indicating that the unplanned tasks load the ACL greater, and thereby increases the risk of injury.

Dancers rarely improvise during performances as even spacing on the floor is tightly choreographed (Liederbach, Dilgen et al. 2008). With no difference in muscular onset time following an unanticipated lower extremity perturbation, our results supports the prevailing theory that lower injury rates in dancers is partly due to the performance of choreographed movements. This study is limited to the evaluation of muscular onset times during an unplanned task to build upon previous work that suggested the proprioceptive training would elicit fasted reflex in dancers. A more comprehensive assessment of movement patterns during a unplanned landing task is need to determine if the dancers maintain their protective neuromuscular pattern of decreased loading when preparation to the landing is eliminated.

Dance Training Contribution to Protective Neuromuscular Control Pattern

Dance training has been shown to improve proprioception (Marmeleira, Pereira et al. 2009), balance ability (Shick, Stoner et al. 1983), as well as increase peak knee and hip flexion during landing (McNitt-Gray, Koff et al. 1992). Female dancers typically start dance training at an early age (6-8 years old) (Liederbach, Dilgen et al. 2008), with strong focus on flexibility (Hamilton, Aronsen et al. 2006), balance (Shick, Stoner et al. 1983; Crotts, Thompson et al. 1996), and landing technique (McNitt-Gray, Koff et al. 1992; Orishimo, Kremenic et al. 2009). Dancer practice uses techniques similar to current ACL prevention programs which focus on plyometrics and landing technique, balance, strength, and agility (Alentorn-Geli, Myer et al. 2009; Yoo, Lim et al. 2010; Dai, Herman et al. 2012), which may contribute to the safe movement patterns during the planned landing task demonstrated by our dance population.

Dancers practice hundreds of jumps daily focusing a square upright torso, as well as, a soft toe-to-heel landing during the jump technique (McNitt-Gray, Koff et al. 1992). Dance jumps do not only focus on the lines of the lower extremity, but they are typically choreographed positions for the upper extremities. Years of jump training with an awareness of the lower extremity, and precise upper extremity positioning may have developed control of the position of the COM during landings shown by the erect landing posture. Soft landings are emphasized in dance practices, specifically teaching dancers to “roll through the foot” to achieve a quiet landing as the heel touches down (Orishimo, Kremenic et al. 2009), similar to what recent ACL prevention programs have begun to focus on during landing training (Prapavessis, McNair et al. 2003; Pollard, Sigward et al. 2006; Padua, DiStefano et al. 2012). As higher landing force are associated with increased risk of ACL injury it is possible that focusing on an upright torso as well as implementation of soft toe-to-heel landings during neuromuscular training programs may explain the lower vGRF in dancers and lower rate of injury.

A distinction of dance training is the inclusion of visual augmented feedback through mirrors in the dance studios. Recently ACL prevention programs that have implemented augmented feedback have found improvements in biomechanics as well as retention of the movement patterns (Prapavessis and McNair 1999; Onate, Guskiewicz et al. 2001; Herman, Onate et al. 2009; Padua, DiStefano et al. 2012; Myer, Stroube et al. 2013). A benefit to the implementation of augmented feedback during dance training compared to ACL prevention programs is its integration into normal practice. Because dance studios are built with mirrors, augmented feedback is provided during technique portions of class as well as choreography. This would translate into providing basketball players augmented feedback during shooting and dribbling drills, as well as during a scrimmage or game. Although impractical to expect athletes to utilize the augmented feedback during a game, the point to make is that dancers do not only receive feedback during neuromuscular training, but also during rehearsals of the movements to be performed in concert. More work needs to done to identify if augmented feedback enhances the neuromuscular benefits of dance training.

Limitations

The current investigation serves as a foundational building block into the comparison between female dancers and athletes. The results from this study are limited to the collegiate athletic and dance populations that represent a wide range of skill level as we recruited athletes from DI – DIII Universities in the Greensboro area, as well as dance majors, who were required to audition into the Department of Dance, and dance minors, who only need to be currently taking 6 credit hours of dance class. The dance minors still upheld the requirement of 5 years' experience, yet the inclusion of this group into the dance population allowed for a wider technical skill range that matched the athletic level across the collegiate divisions.

When interpreting these results it is important to keep in mind that while dancers and athletes were matched on training, they were not matched on demographics. Specifically dancers were weaker in the thigh muscles and were shorter than athletes, which suggest differences in body composition between groups. This may in part have confounded our findings, as Montgomery et al (Montgomery, Shultz et al. 2012) concluded that the maximum strength of the thigh musculature significantly affect the ability to absorb energy during a drop jump. Prior to this study, Harley (Harley 2002), reported similar quadriceps strength output between dancers and non-dancers; therefore, we did not consider matching dancers and athletes on strength values. Harley (Harley 2002), conducted his study on teenage dancers and non-dancers which may have led to the inconsistency with our current findings. We attempted to control for this by initially including strength in all of our statistical models, but ultimately found that strength had no effect on any of the tested hypotheses. Therefore it does not appear that the strength differences influence the reported results. Height was also significantly different between the groups and included as a covariate in the analysis of hypothesis 3 to ensure the length of muscle due to height, did not also affect the reflex response; however, height did not significantly influence the reported results. Future studies should be aware of the potential differences in body composition and strength in these populations, and consider controlling for these factors in their study designs.

Conclusion

ACL prevention programs have been shown to improve protective landing neuromechanics through neuromuscular training (Dai, Herman et al. 2012; Donnelly, Elliott et al. 2012); however, recent evidence suggests that protective neuromechanics are not being retained (Prapavessis and McNair 1999; Benjaminse and Otten 2011). While motor learning literature recommends visual augmented feedback to improve the retention of complex multi limb movements (Clarkson, James et al. 1986; Broker, Gregor et al. 1993; Maas, Robin et al. 2008; Sigrist, Schellenberg et al. 2011), this technique is rarely used in ACL neuromuscular training programs (Prapavessis and McNair 1999; Onate, Guskiewicz et al. 2001; Prapavessis, McNair et al. 2003; Onate, Guskiewicz et al. 2005; Herman, Onate et al. 2009; Myer, Stroube et al. 2013). Female dancers who routinely utilize visual augmented feedback during neuromuscular training are 3 to 5 times less likely to suffer an ACL injury compared to female athletes (Liederbach, Dilgen et al. 2008; Meuffels and Verhaar 2008). We are unsure if the dance training alone results in protective landing patterns that are thought to contribute to the lower rate of injury, or if augmented feedback enhances these benefits. Future research should investigate dance training techniques, with and without the inclusion of augmented feedback, to determine what aspects of dance training may assist in the development of protective landing mechanics.

From this investigation we identified that collegiate female dancers when matched to athletes with similar physical activity and experience levels, do not demonstrate quicker muscular onset times following an unanticipated perturbation, nor stabilize their ground reaction force significantly faster following a forward hop task. Neuromuscular differences were noted during a planned double leg drop jump, in that dancers land with 15% lower peak vGRF, positioned their COM more anteriorly, and moved through greater sagittal plane range of motion, particularly at the ankle joint, compared to female athletes. There were no group differences on sagittal plane initial contact position, frontal plane hip and knee motion, peak extensor moment, relative energy absorption patterns or, pre-ground contact muscular activation.

Examining multiple aspects of the neuromuscular control system has provided insight on subtle differences in variables that collectively assist in an overall protective movement pattern. Specifically, that dancers utilize a movement pattern to lower impact forces of the entire lower extremity rather than specific mechanics to decrease the loading at the knee joint. With similar energy absorption patterns across all joints, the lower landing force might result in a lower overall loading at each joint. Further, the frontal plane motion which has been frequently suspected as a risk factor did not differentiate between these groups, and given evidence from prior prospective studies (Smith, Johnson et al. 2012), may not represent critical neuromuscular differences that contribute to the injury risk.

The findings from this study have provided insight on the neuromuscular control patterns of a female population reported to have a lower rate of ACL injury compared to female athletes. Our findings are also supported by previous literature suggesting that dance training can create a lower loading neuromuscular control pattern. Finally, we have highlighted the importance of integrative assessment of the neuromuscular control profile when attempting to identify what “high-risk” mechanics contribute to the injury disparity in female athletes. Variables such as frontal plane motion and sagittal plane extensor moment may not significantly contribute to the high injury rate in female athletes since dancers demonstrate similar motions and moments with a much lower injury rate.

More research on the neuromuscular differences between female dancers and athletes is warranted to conclusively determine possible ACL injury protective mechanisms employed by dancers during functional movements. Additionally, future research should investigate the components of dance training that contribute to a soft landing technique. This line of research will aid healthcare professionals as it will provide insight into training differences between athletes and dancers that may produce lower loading forces during activity, which can then be implemented into neuromuscular training programs to reduce the occurrence of ACL injury in female athletes.

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APPENDIX A

UNIVERSITY APPROVED CONSENT FORM

UNIVERSITY OF NORTH CAROLINA AT GREENSBORO
CONSENT TO ACT AS A HUMAN PARTICIPANT: LONG FORM

Project Title: Comparison of Neuromuscular Control Strategies in Collegiate Female Dancers and Athletes
Project Director: Michele Pye
Faculty Advisor: Dr. Sandra Shultz

Participant's Name: ____

What is the study about?
Your participation is voluntary. The purpose of this research project is to measure, joint laxity (how loose your joint is), (how well you balance), and muscular activation (when your muscles contract) during a forward hop, a drop jump landing and a lower extremity perturbation in female collegiate athletes and dancers.

Why are you asking me?
You are being asked to participate in this study because you are 18-25 years of age and are a collegiate athlete or dancer who has had at least 5 years of training in your respective activity. You should not participate in this study if you have had a lower extremity injury in the last six months, have cardiac disease, or if you have any head trauma or vestibular disorders in the last six months that may affect your balance.

What will you ask me to do if I agree to be in the study?
You will be asked to attend one testing session that will last approximately 3 hours. We will provide you a physical activity questionnaire and knee outcome survey to be filled out. Then we will measure anthropometrics (height and weight), and joint laxity. Next you will be familiarized and tested on three functional tasks that you will be required to perform. These tasks are:

1) Forward Hop: Hop forward and immediately hold a single leg stance for 10 seconds
2) Drop Jump: Drop off a box from a height of .45m (approximately 1.5 feet) and immediately jump as high as you can upon landing
3) Perturbation Task: With two taut cables attached at the waist, you will be asked to maintain balance on a single limb when one cable is released.

The forward hop task will be performed 3 times, while the drop jump will be performed 5 times. The perturbation task will be performed 10 times (5 causing inward rotation of your trunk, 5 causing outward rotation of your trunk, in a randomized order, and without forewarning of the balance disturbance). These tasks will allow us to measure your movement strategies during a planned, anticipated task, and in reaction to a sudden, unanticipated body movement. We will measure your muscular activation, and joint movement through sensors that will be placed on your thigh and lower leg.

Is there any audio/video recording?
There will be no audio/video recording during this research project.

What are the dangers to me?
The Institutional Review Board at the University of North Carolina at Greensboro has determined that participation in this study poses minimal risk to participants. During the functional tasks, there is a minimal risk that you may lose your balance and possibly pull or strain a muscle leading to muscle soreness. If this occurs, you should stop and tell the tester immediately. If you have any concerns about your rights, how you

are being treated or if you have questions, want more information or have suggestions, please contact Eric Allen in the Office of Research Compliance at UNCG at (336) 256-1482. Questions, concerns or complaints about this project or benefits or risks associated with being in this study can be answered by Dr. Sandra Shultz who may be contacted at 336 334-3027 or sjshultz@uncg.edu.

Are there any benefits to me for taking part in this research study?
There are no direct benefits for your participation in this research study.

Are there any benefits to society as a result of me taking part in this research?
There are no direct benefits to society. The results from this study will improve our understanding of the differences in movement strategies between dancers who are at low risk for knee injuries compared to other physically active populations at higher risk for knee injuries.

Will I get paid for being in the study? Will it cost me anything?
Yes. You will be compensated $30 in a Visa gift card at the completion of the study.

How will you keep my information confidential?
All information in this study is strictly confidential unless disclosure is required by law. Your information will be assigned a code number and the form that has identifiable data and the code number will be kept in a locked file cabinet separate from all data. The file connecting the participant code number to their data will be stored indefinitely in a locked file cabinet within the Applied Neuromechanics Research Laboratory in HHP Room 239. No identifiable data will be used in any report. All consent forms will be maintained in a confidential file only accessible by the investigator. The consent forms will be kept in a file in a locked room for 3 years after completion of the study at which time they will be destroyed by shredding. All de-identified data will be stored on a password protected computer or hardcopies in a locked file cabinet. De-identified data will be kept indefinitely on a password protected computer and in a locked file cabinet within the Applied Neuromechanics Laboratory. A photocopy of this original consent form will be provided to you for your records.

What if I want to leave the study?
You have the right to refuse to participate or to withdraw at any time, without penalty. If you do withdraw, it will not affect you in any way. If you choose to withdraw, you may request that any of your data which has been collected be destroyed unless it is in a de-identifiable state.

What about new information/changes in the study?
If significant new information relating to the study becomes available which may relate to your willingness to continue to participate, this information will be provided to you.

Voluntary Consent by Participant:
By signing this consent form you are agreeing that you read, or it has been read to you, and you fully understand the contents of this document and are openly willing consent to take part in this study. All of your questions concerning this study have been answered. By signing this form, you are agreeing that you are 18 years of age or older and are agreeing to participate, or have the individual specified above as a participant participate, in this study described to you by _________________________.

Signature: ________________________ Date: ______________

APPENDIX B

PARTICIPANT INTAKE SURVEYS

PHYSICAL ACTIVITY AND HEALTH HISTORY

Do you have any General Health Problems or Illnesses? (e.g. diabetes, respiratory disease) Yes____ No____

Do you have any vestibular (inner ear) or balance disorders? Yes____ No____

Do you smoke? Yes____ No____

Do you drink alcohol? Yes____ No____ If yes, how often?______________________________

Do you have any history of connective tissue disease or disorders? (e.g. Ehlers-Danlos, Marfan's Syndrome, Rheumatoid Arthritis) Yes____ No____

Has a family member of yours ever been diagnosed with breast cancer? Yes____ No____ (if no, please skip next question.)

If yes, please put a check next to the types of relatives that have been diagnosed. You may check more than one box:

Mother______ Sister___Grandmother ___Aunt ___.
Male relative (father, brother, grandfather, or uncle) ______________.
Other type of relative (please write in) ___________________.

Please list any medications you take regularly: ____
____

Please list any previous injuries to your lower extremities. Please include a description of the injury (e.g. ligament sprain, muscle strain), severity of the injury, date of the injury, and whether it was on the left or right side.

Body Part	Description	Severity	Date of Injury	L or R
Hip				
Thigh				
Knee				
Lower Leg				
Ankle				
Foot				

Please list any previous surgery to your lower extremities (Include a description of the surgery, the date of the surgery, and whether it was on the left or right side)

Body Part Description Date of Surgery L or R

Please list all physical activities that you are currently engaged in. For each activity, please indicate how much time you spend each week in this activity, the intensity of the activity (i.e. competitive or recreational) and for how long you have been regularly participating in the activity.

Activity #Days/week #Minutes/Day Intensity Experience in this Activity (# of years)

What time of day do you generally engage in the above activities?______________________________

Please list other conditions / concerns that you feel we should be aware of:______________________

PAR-Q

The Activity Rating Scale

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

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

Investigator Comments:

Knee Outcome Survey Activities of Daily Living Scale (ADLS).

Symptoms: To what degree does each of the following symptoms affect your level of activity? (check one answer on each line)

	I do not have the symptom	I have the symptom, but it does not affect my activity	The symptom affects my activity slightly	The symptom affects my activity moderately	The symptom affects my activity severely	The symptom prevents me from all daily activity
Pain	( )	( )	( )	( )	( )	( )
Stiffness	( )	( )	( )	( )	( )	( )
Swelling	( )	( )	( )	( )	( )	( )
Giving way, buckling, or shifting of the knee	( )	( )	( )	( )	( )	( )
Weakness	( )	( )	( )	( )	( )	( )
Limping	( )	( )	( )	( )	( )	( )

Functional Limitations With Activities of Daily Living: How does your knee affect your ability to: (check one answer on each line)

	Activity is not difficult	Activity is minimally difficult	Activity is somewhat difficult	Activity is fairly difficult	Activity is very difficult	I am unable to do the activity
Walk	( )	( )	( )	( )	( )	( )
Go up stairs	( )	( )	( )	( )	( )	( )
Go down stairs	( )	( )	( )	( )	( )	( )
Stand	( )	( )	( )	( )	( )	( )
Kneel on front of your knee	( )	( )	( )	( )	( )	( )
Squat	( )	( )	( )	( )	( )	( )
Sit with your knee bent	( )	( )	( )	( )	( )	( )
Rise from a chair	( )	( )	( )	( )	( )	( )

Scoring: The first column is scored 5 points for each item, followed in successive columns by scores of 4, 3, 2, 1, and 0 for the last column. The total points from all items are summed, then divided by 70 and multiplied by 100 for the ADLS score. For example, if the individual places marks for 12 items in the first column, and 2 items in the second column the total points would be 12x5 = 60 points, plus 2 x 4 = 8 points, for a total of 68 points. The ADLS score would then be 68/70 x 100 = 97%.

Knee Outcome Survey Sports Activities Scale (SAS).

Symptoms: To what degree does each of the following symptoms affect your level of sports activity? (check one answer on each line)

	Never have	Have, but does not affect my sports activity	Affects sports activity slightly	Affects sports activity moderately	Affects sports activity severely	Prevents me from all sports activity
Pain	( )	( )	( )	( )	( )	( )
Grinding or grating	( )	( )	( )	( )	( )	( )
Stiffness	( )	( )	( )	( )	( )	( )
Swelling	( )	( )	( )	( )	( )	( )
Slipping or partial giving way of knee	( )	( )	( )	( )	( )	( )
Buckling or full giving way of knee	( )	( )	( )	( )	( )	( )
Weakness	( )	( )	( )	( )	( )	( )

Functional Limitations With Sports Activities: How does your knee affect your ability to: (check one answer on each line)

	Not difficult at all	Minimally difficult	Somewhat difficult	Fairly difficult	Very difficult	Unable to do
Run straight ahead	( )	( )	( )	( )	( )	( )
Jump and land on your involved leg	( )	( )	( )	( )	( )	( )
Stop and start quickly	( )	( )	( )	( )	( )	( )
Cut and pivot on your involved leg	( )	( )	( )	( )	( )	( )

Scoring: The first column is scored 5 points for each item, followed in successive columns by scores of 4, 3, 2, 1, and 0 for the last column. The total points from all items are summed, then divided by 55 and multiplied by 100 for the SAS score. For example, if the individual places marks for 9 items in the first column, and 2 items in the second column the total points would be 9x5 = 45 points, plus 2 x 4 = 8 points, for a total of 53 points. The SAS score would then be 53/55 x 100 = 96

APPENDIX C

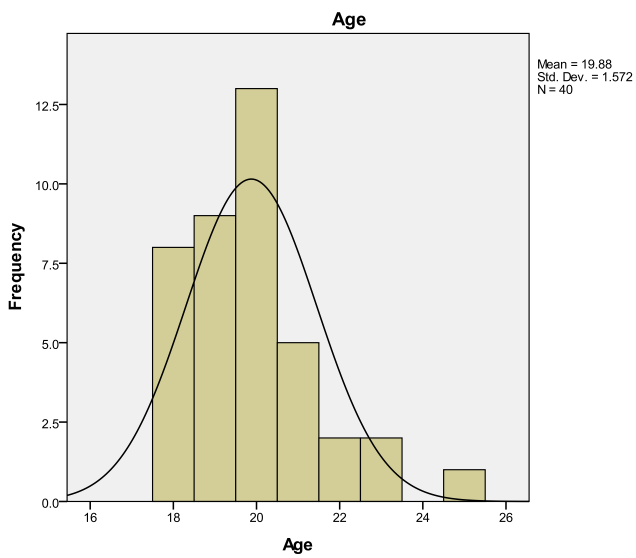
COUNTERBALANCE ORDER

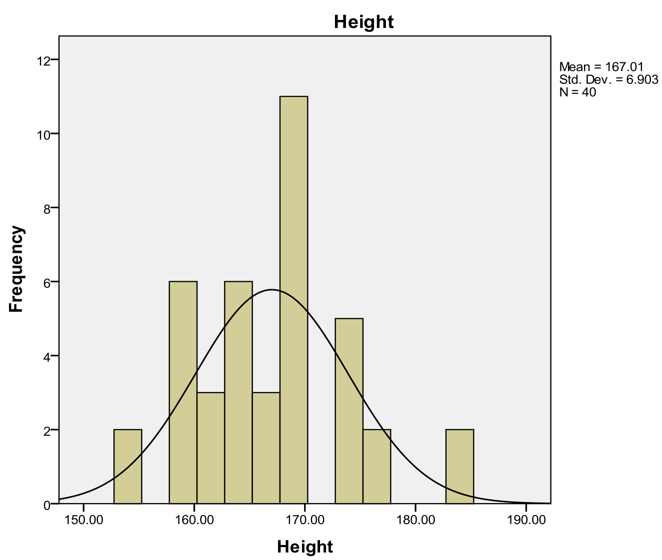
Subject	Shoe Condition	Limb Dominance	LEPD Order									
1	Shod	Dominant	IR	IR	ER	ER	ER	ER	IR	IR	ER	IR
2	Shod	Non-Dominant	IR	ER	IR	IR	ER	IR	ER	IR	ER	ER
3	Barefoot	Dominant	ER	ER	IR	IR	ER	IR	ER	IR	IR	ER
4	Barefoot	Non-Dominant	IR	ER	IR	IR	ER	ER	ER	IR	IR	ER
5	Shod	Dominant	ER	IR	IR	ER	ER	IR	ER	IR	IR	ER
6	Shod	Non-Dominant	IR	IR	IR	ER	ER	IR	ER	ER	ER	IR
7	Barefoot	Dominant	IR	IR	ER	IR	ER	ER	IR	IR	ER	ER
8	Barefoot	Non-Dominant	ER	IR	IR	ER	ER	ER	IR	ER	IR	IR
9	Shod	Dominant	ER	IR	ER	IR	ER	IR	ER	IR	IR	ER
10	Shod	Non-Dominant	IR	IR	IR	ER	ER	ER	IR	ER	IR	ER
11	Barefoot	Dominant	ER	ER	IR	ER	IR	ER	IR	IR	ER	IR
12	Barefoot	Non-Dominant	IR	ER	IR	ER	ER	ER	ER	IR	IR	IR
13	Shod	Dominant	ER	ER	ER	IR	IR	IR	IR	IR	ER	ER
14	Shod	Non-Dominant	ER	IR	ER	ER	IR	IR	IR	IR	ER	ER
15	Barefoot	Dominant	ER	IR	IR	IR	IR	ER	IR	ER	ER	ER
16	Barefoot	Non-Dominant	IR	IR	ER	ER	ER	IR	IR	ER	IR	ER
17	Shod	Dominant	ER	ER	ER	ER	ER	IR	IR	IR	IR	IR
18	Shod	Non-Dominant	IR	IR	IR	ER	ER	ER	ER	IR	ER	IR
19	Barefoot	Dominant	IR	IR	ER	IR	ER	ER	ER	IR	ER	IR
20	Barefoot	Non-Dominant	ER	ER	IR	IR	ER	IR	ER	ER	IR	IR
21	Shod	Dominant	ER	IR	IR	ER	ER	ER	ER	IR	IR	IR
22	Shod	Non-	ER	ER	IR	ER	ER	IR	IR	IR	IR	ER

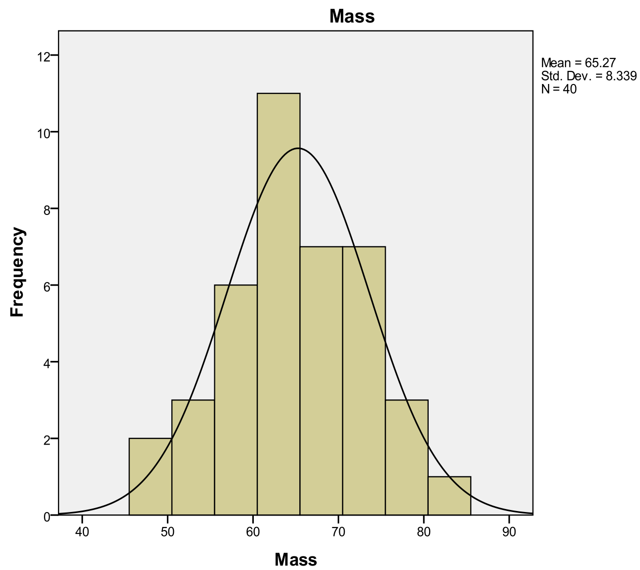
		Dominant										
23	Barefoot	Dominant	IR	IR	ER	IR	IR	IR	ER	ER	ER	ER
24	Barefoot	Non-Dominant	IR	ER	IR	ER	IR	ER	IR	ER	IR	ER
25	Shod	Dominant	ER	ER	IR	IR	ER	IR	ER	ER	IR	IR
26	Shod	Non-Dominant	ER	IR	ER	ER	IR	ER	IR	IR	ER	IR
27	Barefoot	Dominant	ER	IR	IR	ER	IR	IR	ER	ER	ER	IR
28	Barefoot	Non-Dominant	IR	IR	IR	ER	ER	IR	ER	ER	IR	ER
29	Shod	Dominant	IR	ER	ER	IR	ER	IR	IR	IR	ER	ER
30	Shod	Non-Dominant	ER	ER	IR	IR	IR	ER	IR	ER	IR	ER
31	Barefoot	Dominant	IR	ER	ER	IR	IR	IR	ER	IR	ER	ER
32	Barefoot	Non-Dominant	IR	IR	ER	ER	ER	ER	IR	IR	IR	ER
33	Shod	Dominant	ER	IR	IR	IR	ER	ER	IR	IR	ER	ER
34	Shod	Non-Dominant	ER	IR	ER	ER	ER	IR	IR	ER	IR	IR
35	Barefoot	Dominant	ER	IR	ER	IR	IR	ER	ER	IR	ER	IR
36	Barefoot	Non-Dominant	IR	IR	IR	ER	ER	IR	ER	ER	ER	IR
37	Shod	Dominant	ER	ER	IR	ER	IR	ER	IR	IR	ER	IR
38	Shod	Non-Dominant	ER	IR	ER	IR	ER	IR	ER	IR	ER	IR
39	Barefoot	Dominant	ER	IR	IR	IR	ER	ER	ER	IR	IR	ER
40	Barefoot	Non-Dominant	IR	IR	ER	IR	ER	ER	ER	IR	IR	ER

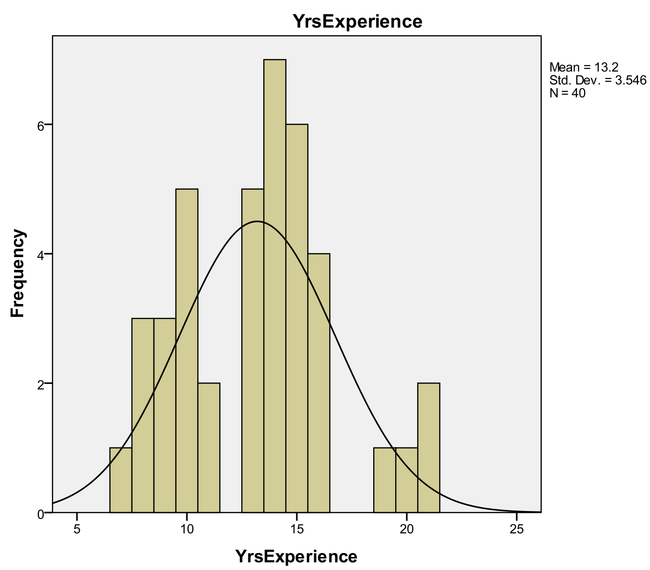
APPENDIX D

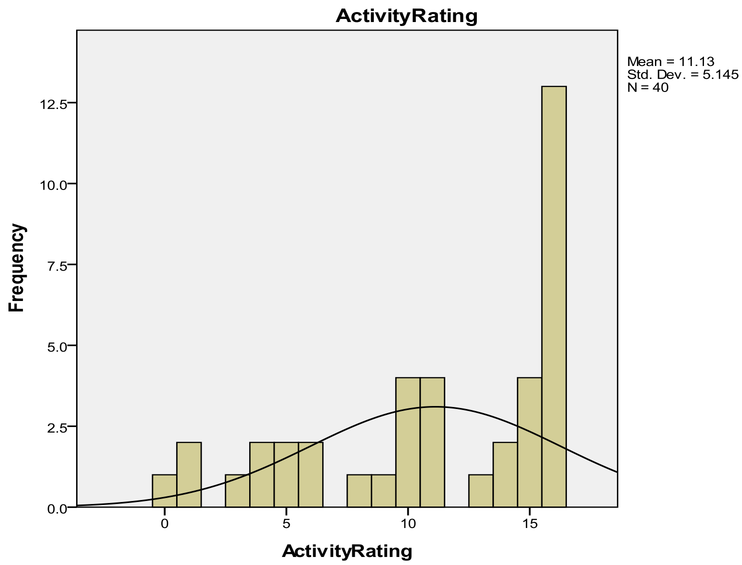
HISTOGRAMS OF ALL DEPENDENT VARIABLES



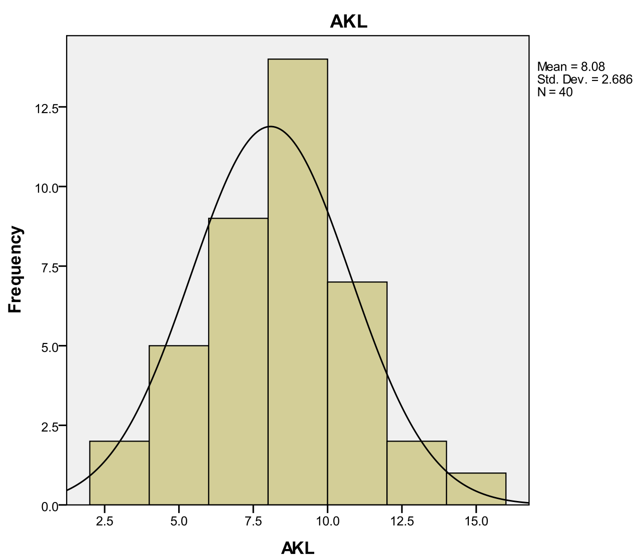



Mass
Mean = 65.27
Std. Dev. = 8.339
N = 40
Frequency
Mass
0
2
4
6
8
10
12
40
50
60
70
80
90

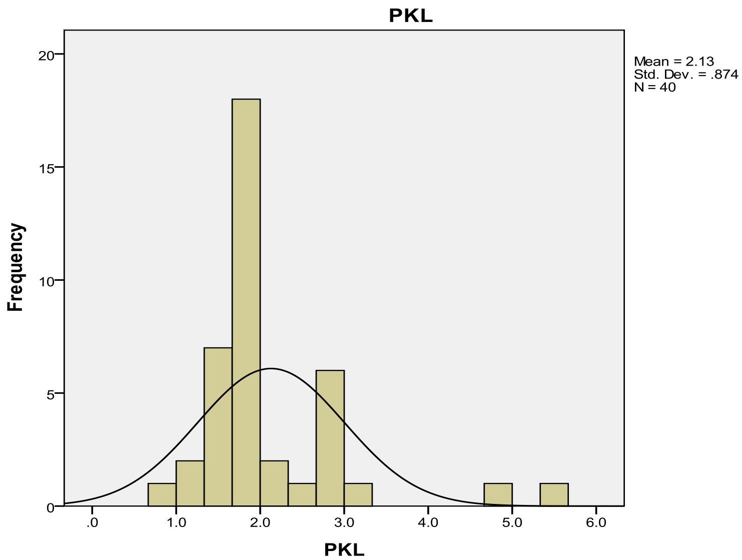

YrsExperience
Mean = 13.2
Std. Dev. = 3.546
N = 40
Frequency
YrsExperience
0
2
4
6
5
10
15
20
25

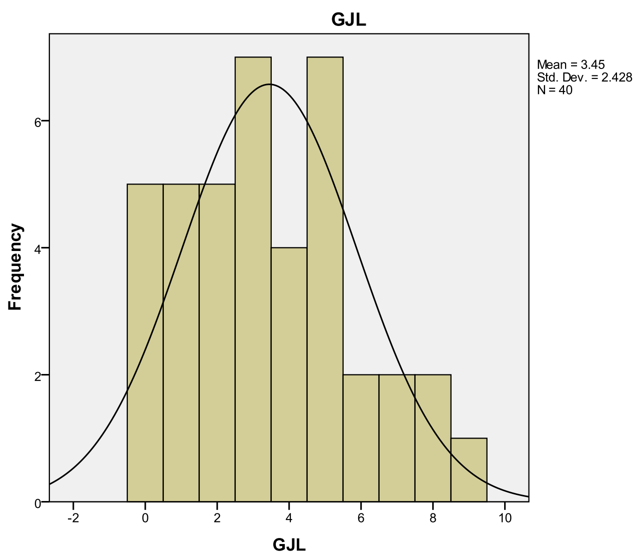


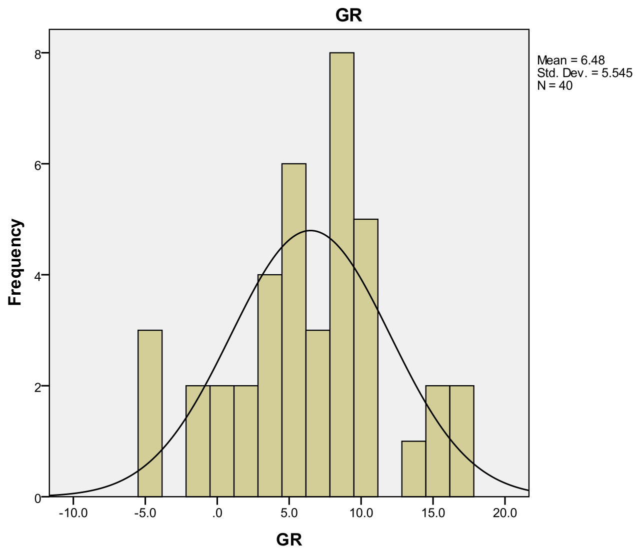
ActivityRating
Mean = 11.13
Std. Dev. = 5.145
N = 40
Frequency
0.0
2.5
5.0
7.5
10.0
12.5
0
5
10
15
ActivityRating

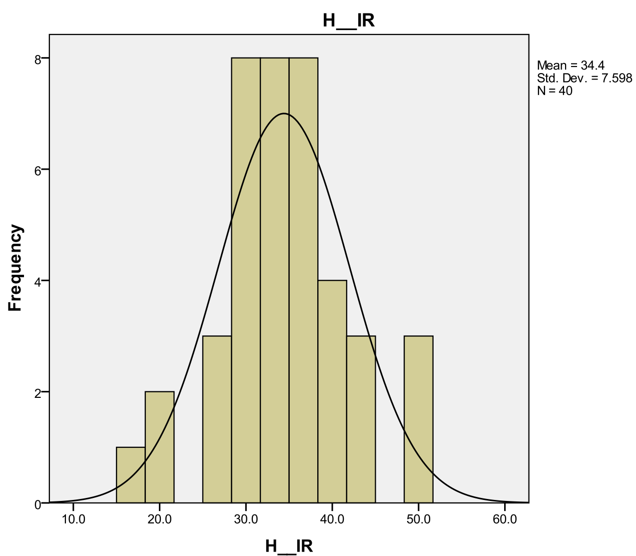


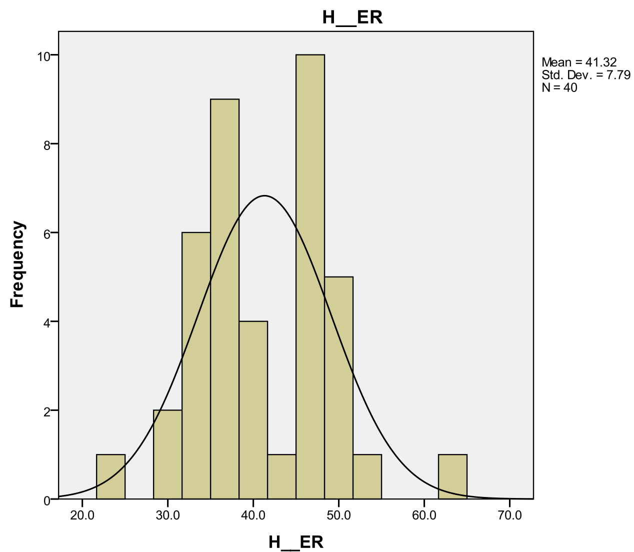
AKL
Mean = 8.08
Std. Dev. = 2.686
N = 40
Frequency
0.0
2.5
5.0
7.5
10.0
12.5
2.5
5.0
7.5
10.0
12.5
15.0
AKL

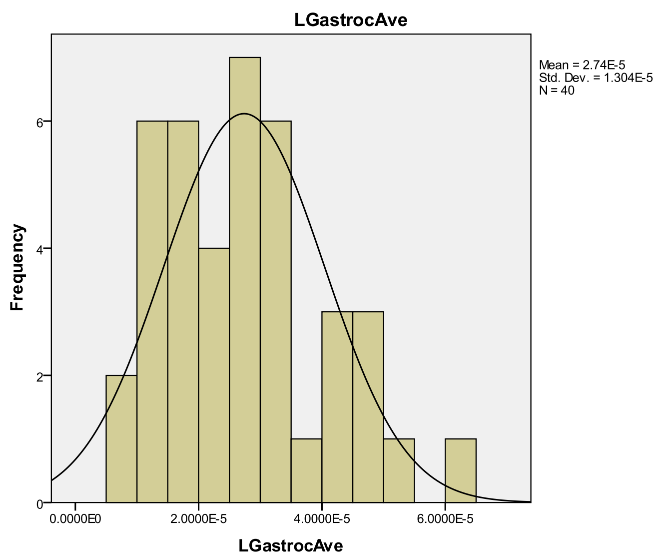

PKL
Mean = 2.13
Std. Dev. = .874
N = 40
Frequency
20
15
10
5
0
.0
1.0
2.0
3.0
4.0
5.0
6.0
PKL

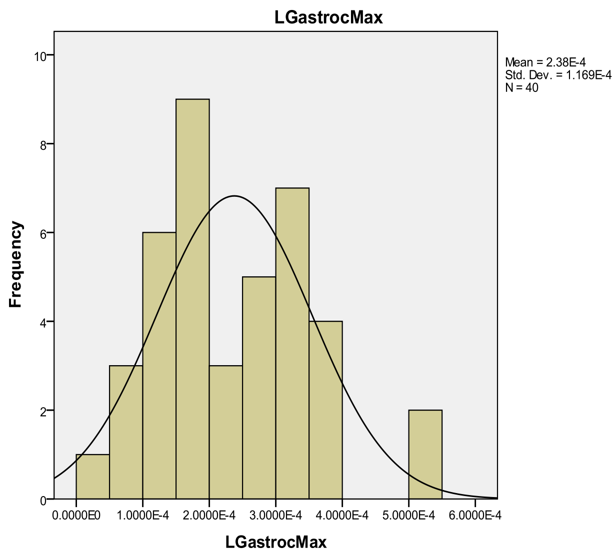

GJL
Mean = 3.45
Std. Dev. = 2.428
N = 40
Frequency
6
4
2
0
-2
0
2
4
6
8
10
GJL


GR
Mean = 6.48
Std. Dev. = 5.545
N = 40
Frequency
GR

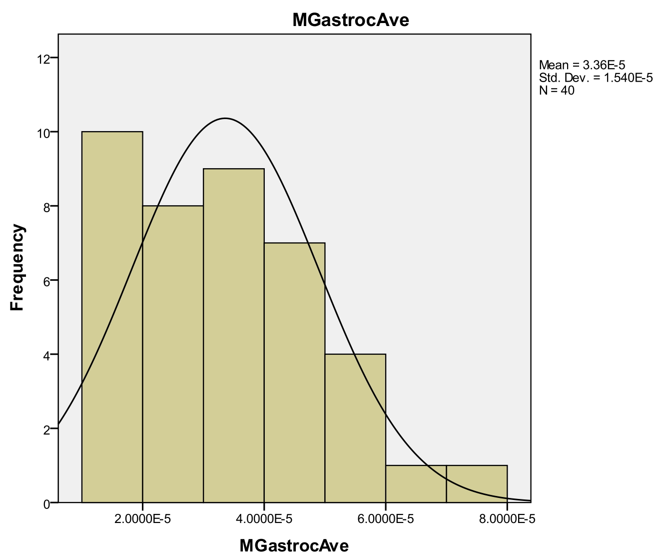

H__IR
Mean = 34.4
Std. Dev. = 7.598
N = 40
Frequency
H__IR


H__ER
Mean = 41.32
Std. Dev. = 7.79
N = 40
Frequency
H__ER

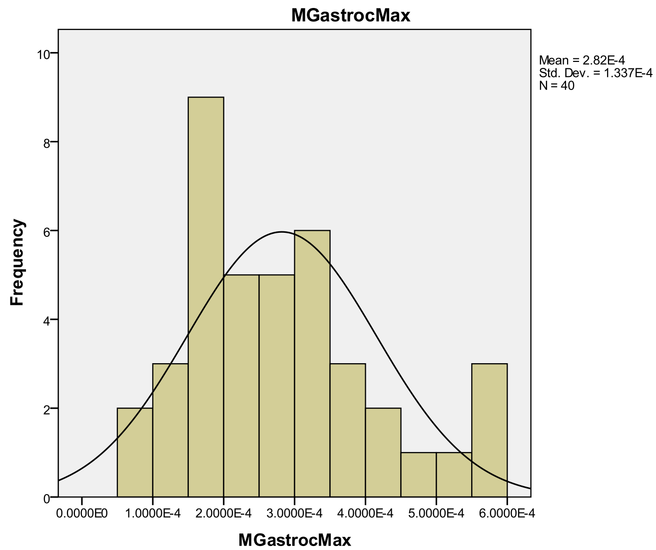

LGastrocAve
Mean = 2.74E-5
Std. Dev. = 1.304E-5
N = 40
Frequency
LGastrocAve

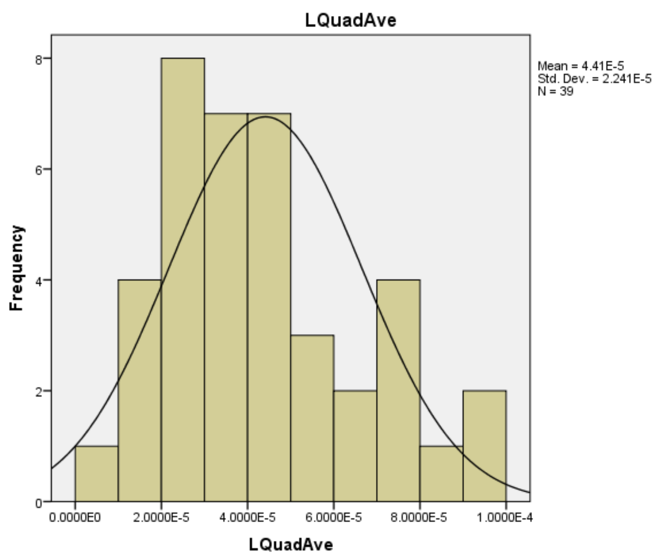


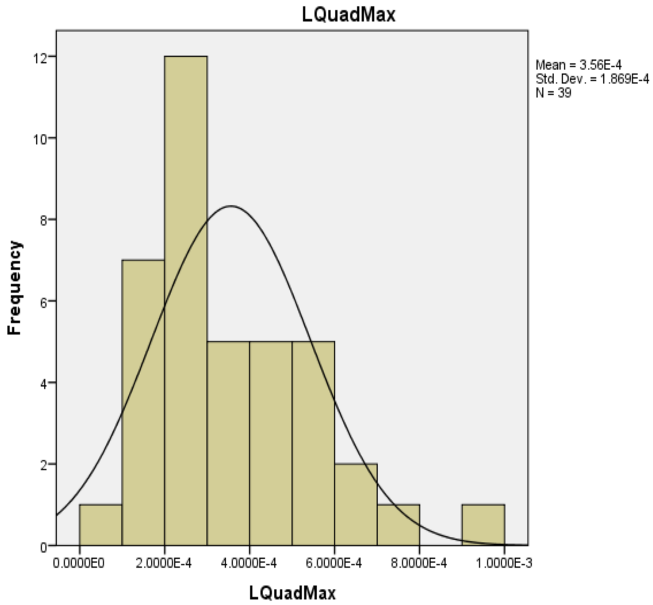
LGastrocMax
Mean = 2.38E-4
Std. Dev. = 1.169E-4
N = 40
Frequency
0
2
4
6
8
10
0.0000E0
1.0000E-4
2.0000E-4
3.0000E-4
4.0000E-4
5.0000E-4
6.0000E-4
LGastrocMax

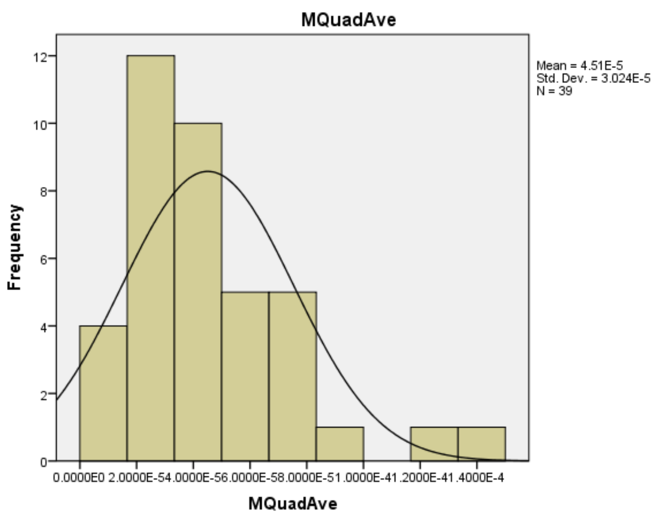


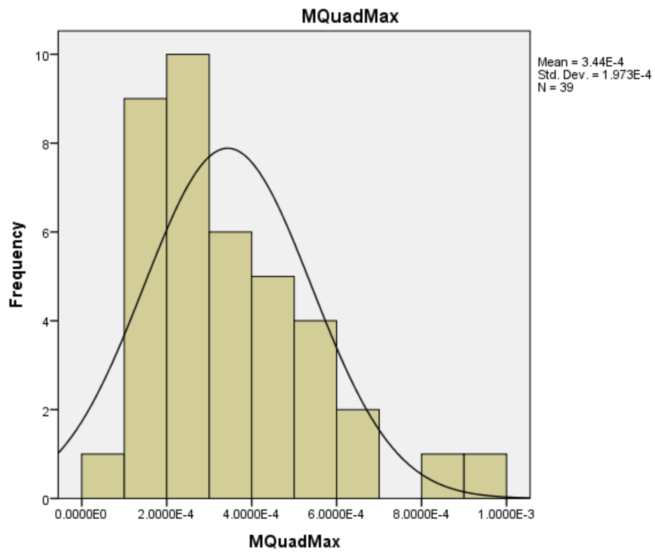
MGastrocAve
Mean = 3.36E-5
Std. Dev. = 1.540E-5
N = 40
Frequency
0
2
4
6
8
10
12
2.0000E-5
4.0000E-5
6.0000E-5
8.0000E-5
MGastrocAve

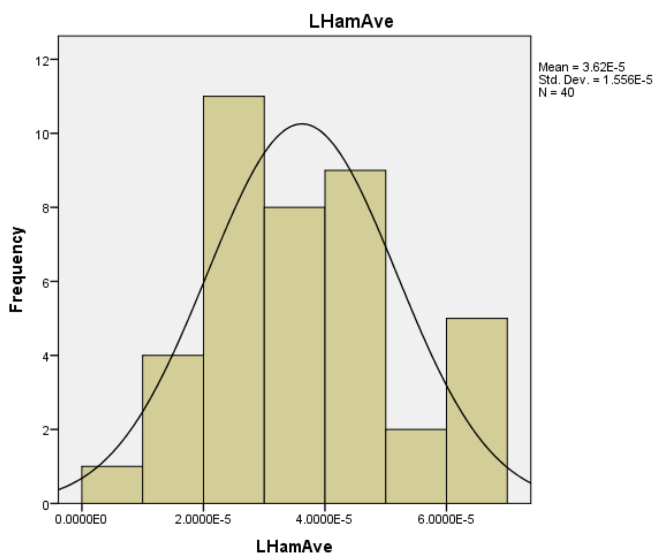

MGastrocMax
Mean = 2.82E-4
Std. Dev. = 1.337E-4
N = 40
Frequency
0
2
4
6
8
10
0.0000E0
1.0000E-4
2.0000E-4
3.0000E-4
4.0000E-4
5.0000E-4
6.0000E-4
MGastrocMax

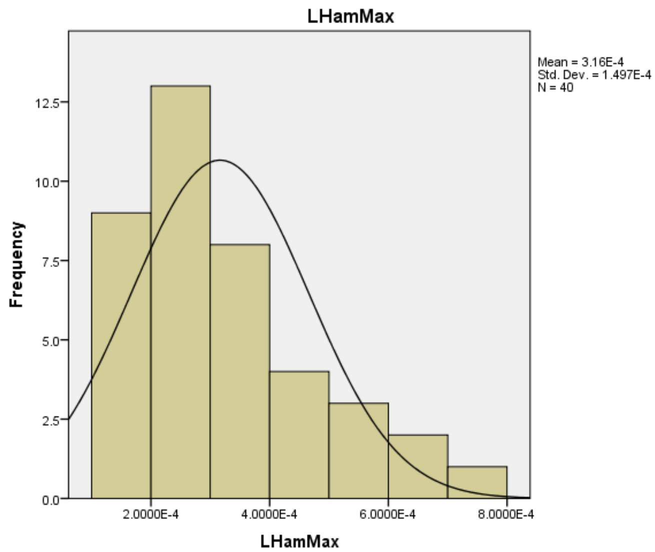

LQuadAve
Mean = 4.41E-5
Std. Dev. = 2.241E-5
N = 39
Frequency
0
2
4
6
8
0.0000E0
2.0000E-5
4.0000E-5
6.0000E-5
8.0000E-5
1.0000E-4
LQuadAve

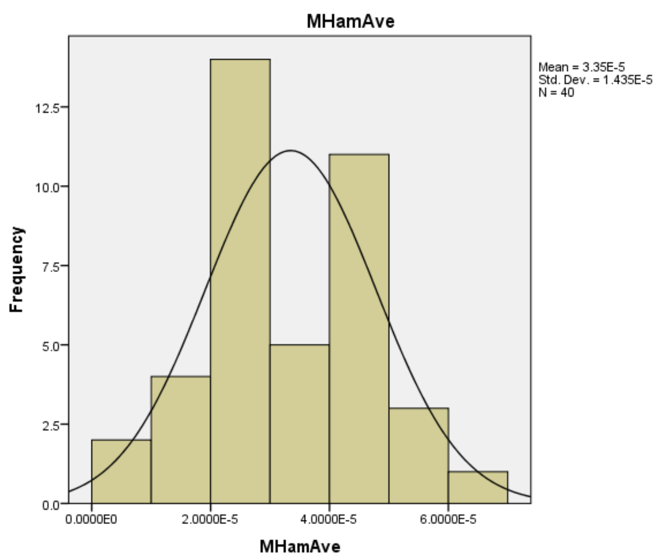

LQuadMax
Mean = 3.56E-4
Std. Dev. = 1.869E-4
N = 39
Frequency
0
2
4
6
8
10
12
0.0000E0
2.0000E-4
4.0000E-4
6.0000E-4
8.0000E-4
1.0000E-3
LQuadMax

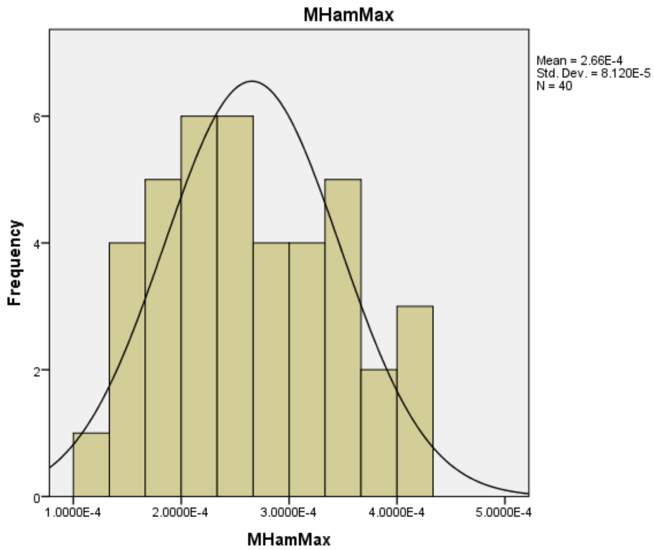

MQuadAve
Mean = 4.51E-5
Std. Dev. = 3.024E-5
N = 39
Frequency
0
2
4
6
8
10
12
0.0000E0 2.0000E-54.0000E-56.0000E-58.0000E-51.0000E-41.2000E-41.4000E-4
MQuadAve

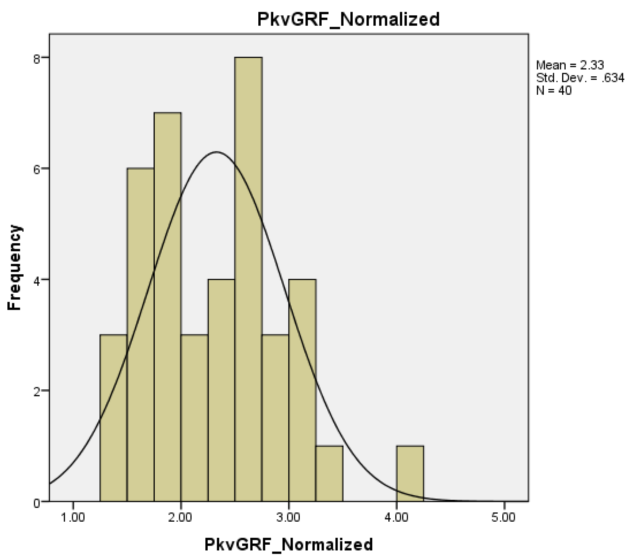

MQuadMax
Mean = 3.44E-4
Std. Dev. = 1.973E-4
N = 39
Frequency
0
2
4
6
8
10
0.0000E0
2.0000E-4
4.0000E-4
6.0000E-4
8.0000E-4
1.0000E-3
MQuadMax

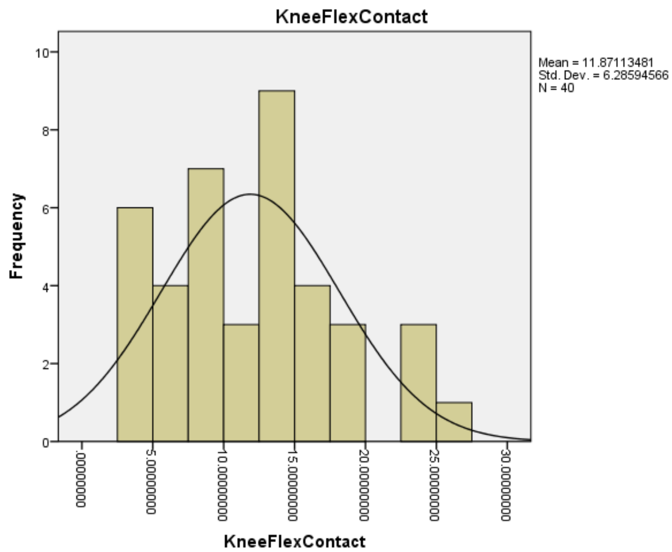

LHamAve
Mean = 3.62E-5
Std. Dev. = 1.556E-5
N = 40
Frequency
0
2
4
6
8
10
12
0.0000E0
2.0000E-5
4.0000E-5
6.0000E-5
LHamAve

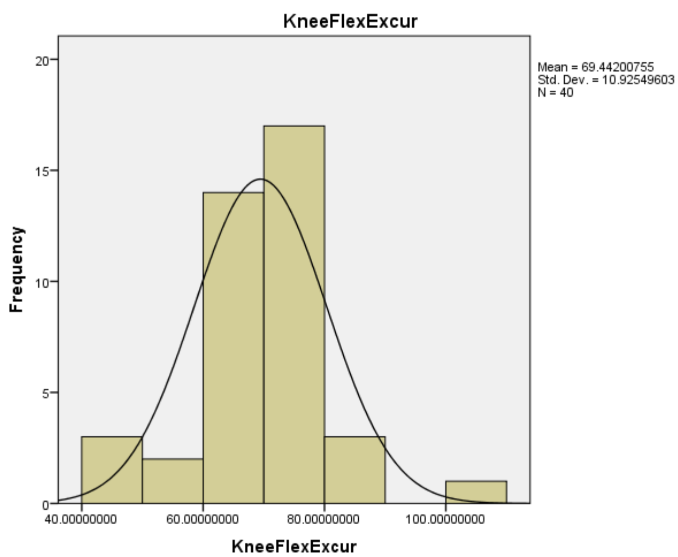

LHamMax
Mean = 3.16E-4
Std. Dev. = 1.497E-4
N = 40
Frequency
12.5
10.0
7.5
5.0
2.5
0.0
2.0000E-4
4.0000E-4
6.0000E-4
8.0000E-4
LHamMax

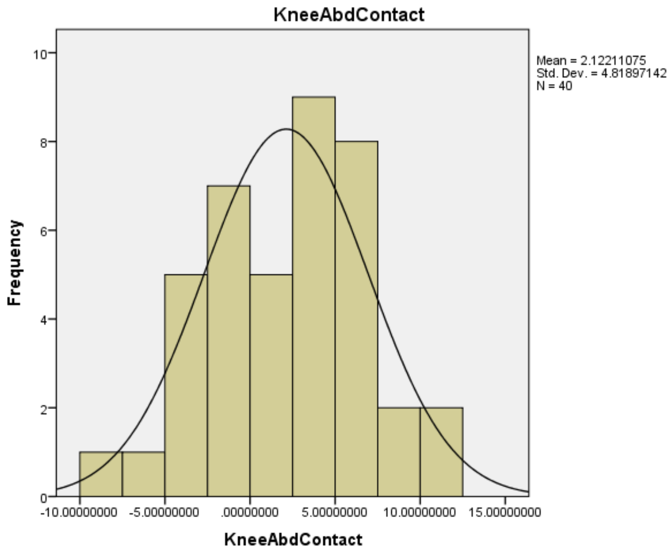

MHamAve
Mean = 3.35E-5
Std. Dev. = 1.435E-5
N = 40
Frequency
12.5
10.0
7.5
5.0
2.5
0.0
0.0000E0
2.0000E-5
4.0000E-5
6.0000E-5
MHamAve

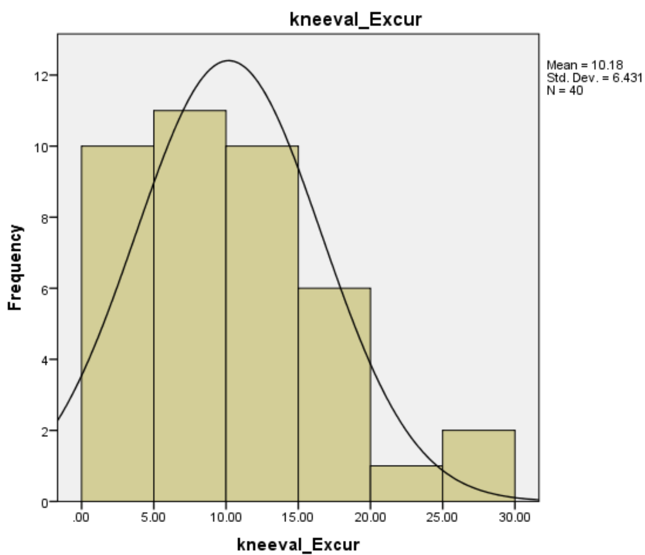

MHamMax
Mean = 2.66E-4
Std. Dev. = 8.120E-5
N = 40
Frequency
0
2
4
6
1.0000E-4
2.0000E-4
3.0000E-4
4.0000E-4
5.0000E-4
MHamMax

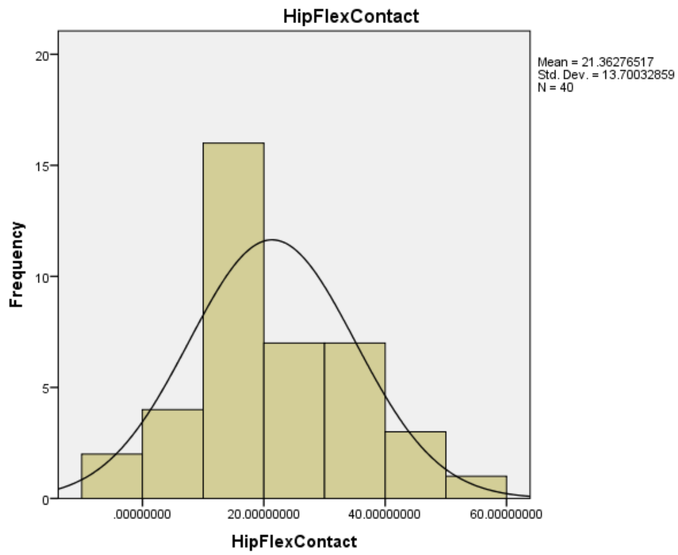

PkvGRF_Normalized
Mean = 2.33
Std. Dev. = .634
N = 40
Frequency
0
2
4
6
8
1.00
2.00
3.00
4.00
5.00
PkvGRF_Normalized

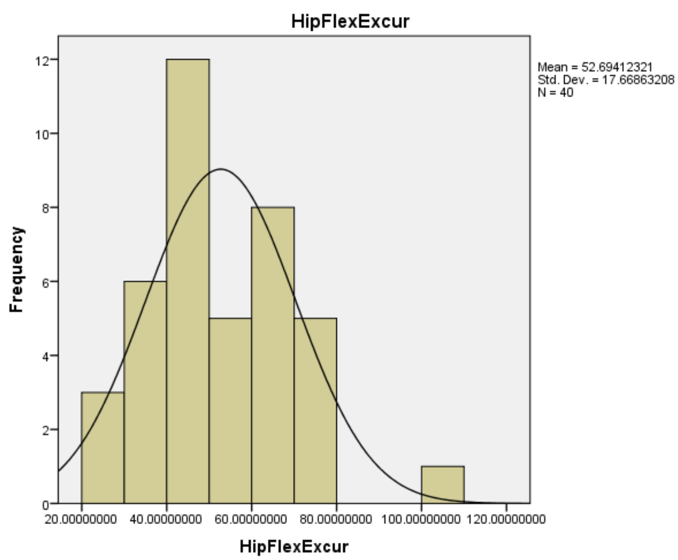

KneeFlexContact
Mean = 11.87113481
Std. Dev. = 6.28594566
N = 40
Frequency
KneeFlexContact

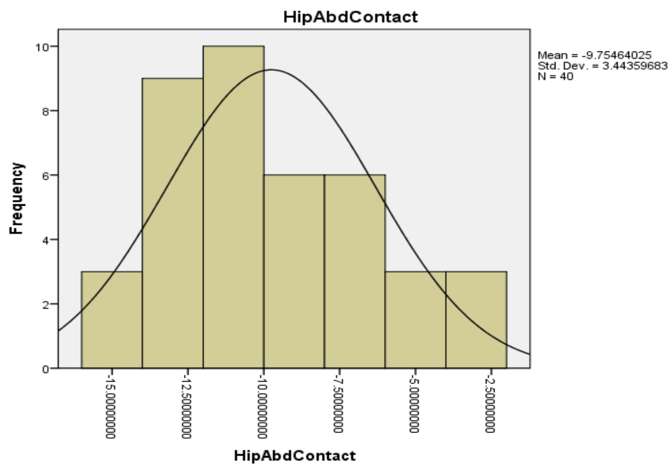

KneeFlexExcur
Mean = 69.44200755
Std. Dev. = 10.92549603
N = 40
Frequency
KneeFlexExcur

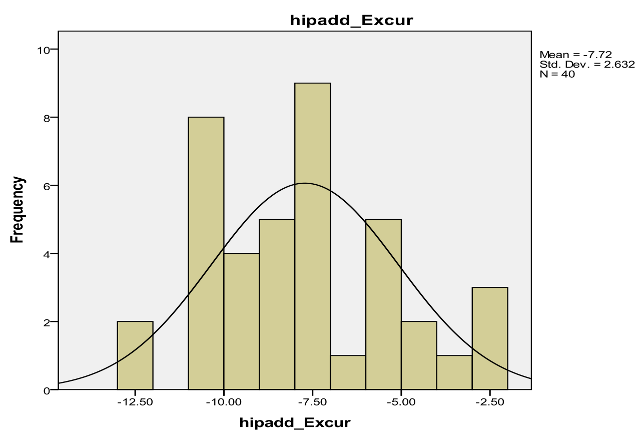

KneeAbdContact
Mean = 2.12211075
Std. Dev. = 4.81897142
N = 40
Frequency
0
2
4
6
8
10
-10.00000000
-5.00000000
.00000000
5.00000000
10.00000000
15.00000000
KneeAbdContact

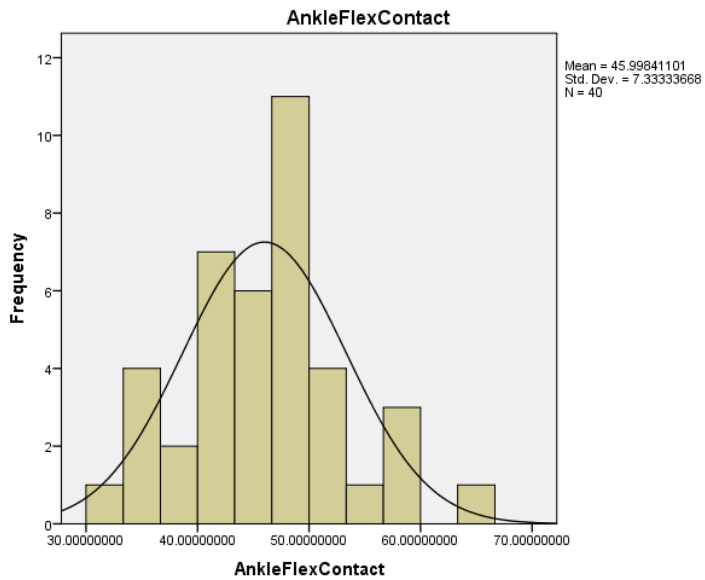

kneeval_Excur
Mean = 10.18
Std. Dev. = 6.431
N = 40
Frequency
0
2
4
6
8
10
12
.00
5.00
10.00
15.00
20.00
25.00
30.00
kneeval_Excur

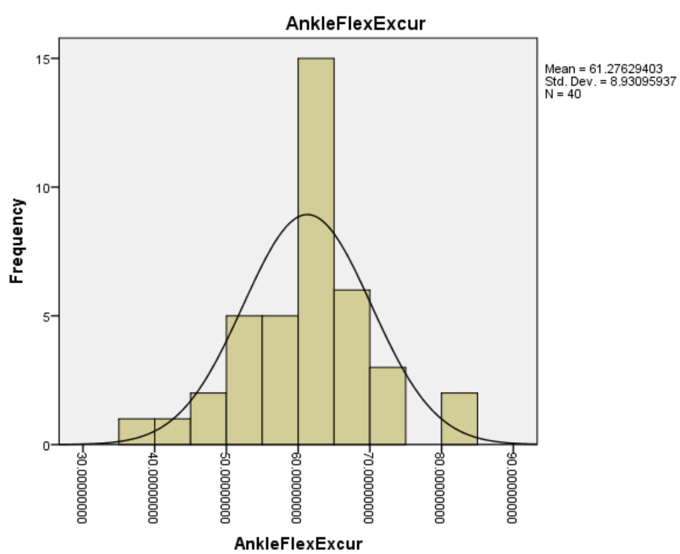

HipFlexContact
Mean = 21.36276517
Std. Dev. = 13.70032859
N = 40
Frequency
20
15
10
5
0
.00000000
20.00000000
40.00000000
60.00000000
HipFlexContact

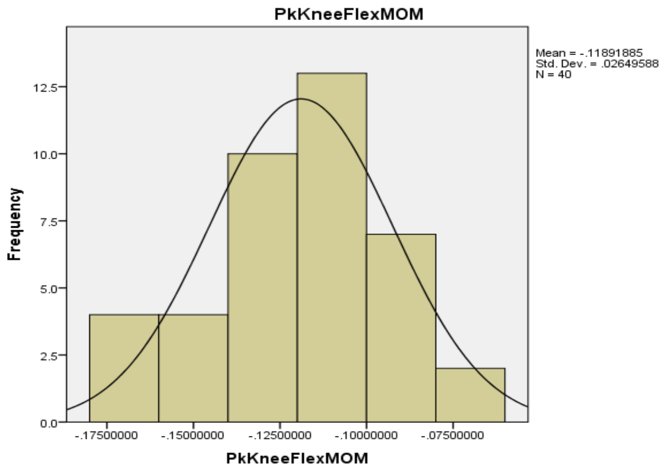

HipFlexExcur
Mean = 52.69412321
Std. Dev. = 17.66863208
N = 40
Frequency
12
10
8
6
4
2
0
20.00000000
40.00000000
60.00000000
80.00000000
100.00000000
120.00000000
HipFlexExcur

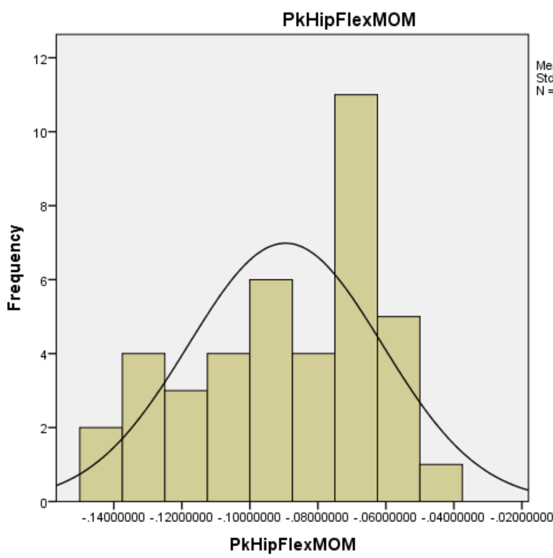

HipAbdContact
Mean = -9.75464025
Std. Dev. = 3.44359683
N = 40
Frequency
0
2
4
6
8
10
-15.00000000
-12.50000000
-10.00000000
-7.50000000
-5.00000000
-2.50000000
HipAbdContact

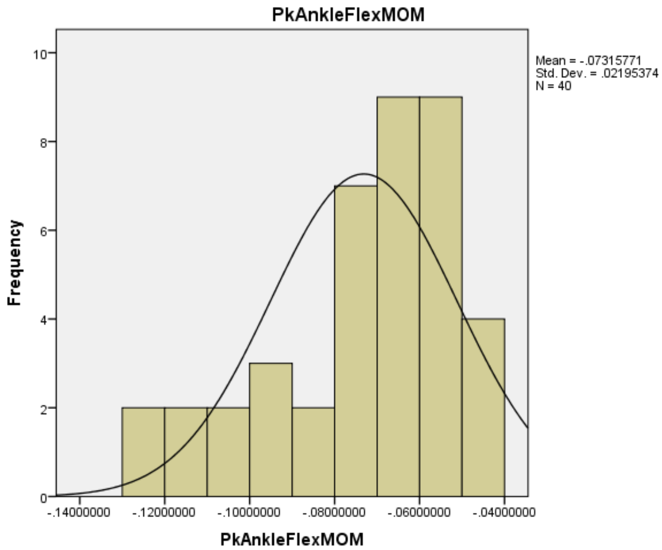

hipadd_Excur
Mean = -7.72
Std. Dev. = 2.632
N = 40
Frequency
0
2
4
6
8
10
-12.50
-10.00
-7.50
-5.00
-2.50
hipadd_Excur

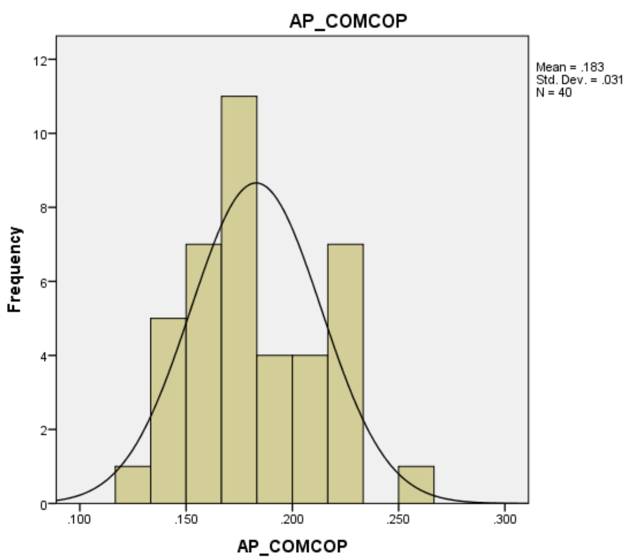

AnkleFlexContact
Mean = 45.99841101
Std. Dev. = 7.33333668
N = 40
Frequency
12
10
8
6
4
2
0
30.00000000
40.00000000
50.00000000
60.00000000
70.00000000
AnkleFlexContact

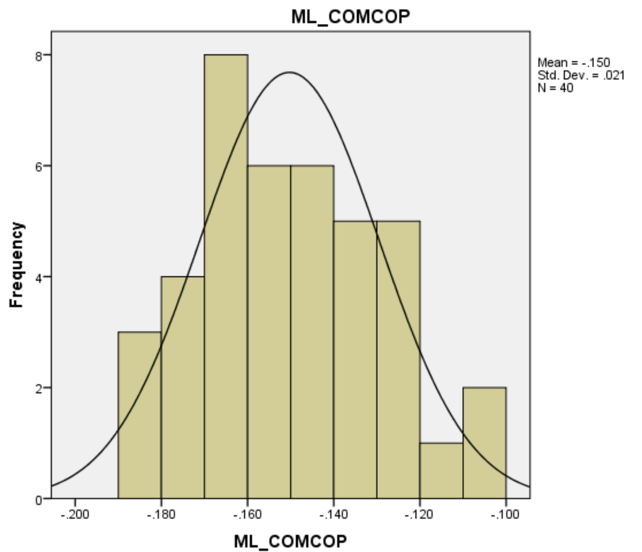

AnkleFlexExcur
Mean = 61.27629403
Std. Dev. = 8.93095937
N = 40
Frequency
15
10
5
0
30.00000000
40.00000000
50.00000000
60.00000000
70.00000000
80.00000000
90.00000000
AnkleFlexExcur

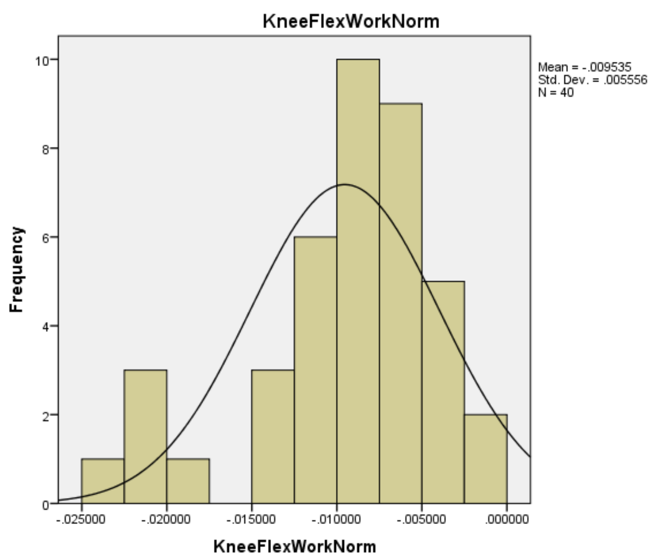

PkKneeFlexMOM
Mean = -.11891885
Std. Dev. = .02649588
N = 40
Frequency
0.0
2.5
5.0
7.5
10.0
12.5
-.17500000
-.15000000
-.12500000
-.10000000
-.07500000
PkKneeFlexMOM

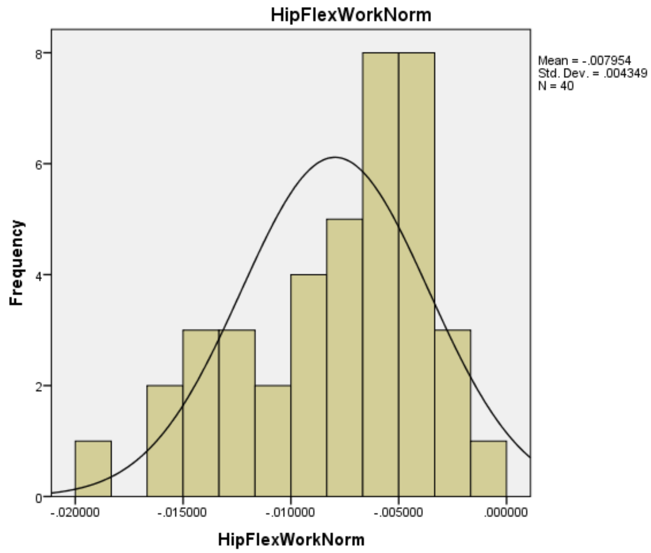

PkHipFlexMOM
Mean = -.08952963
Std. Dev. = .02855478
N = 40
Frequency
0
2
4
6
8
10
12
-.14000000
-.12000000
-.10000000
-.08000000
-.06000000
-.04000000
-.02000000
PkHipFlexMOM

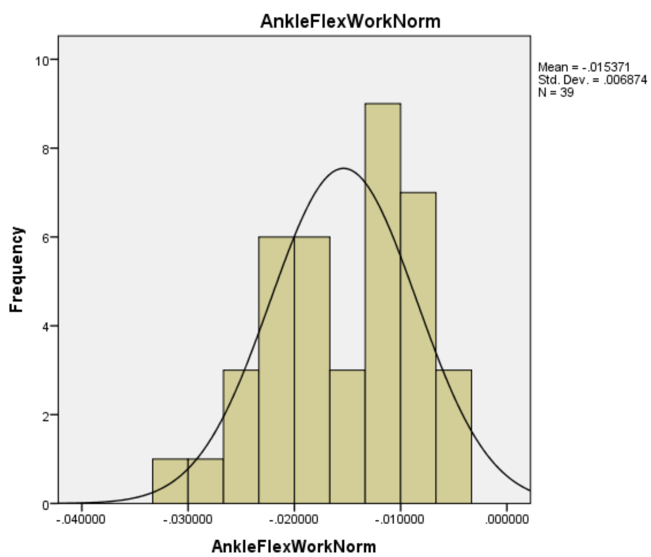

PkAnkleFlexMOM
Mean = -.07315771
Std. Dev. = .02195374
N = 40
Frequency
0
2
4
6
8
10
-.14000000
-.12000000
-.10000000
-.08000000
-.06000000
-.04000000
PkAnkleFlexMOM

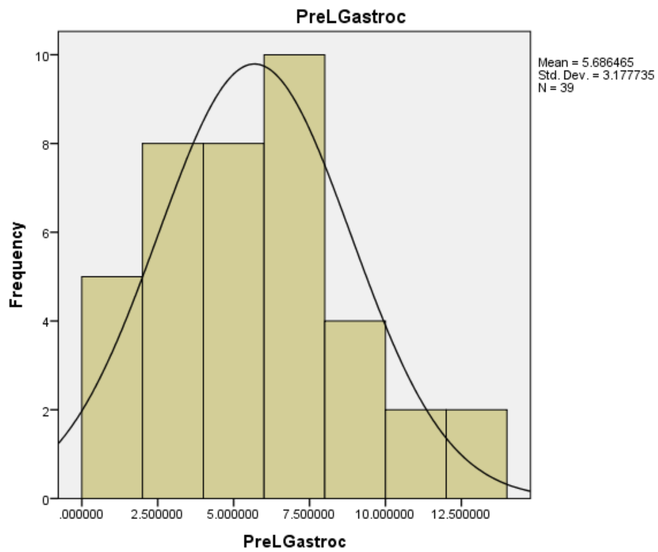

AP_COMCOP
Mean = .183
Std. Dev. = .031
N = 40
Frequency
0
2
4
6
8
10
12
.100
.150
.200
.250
.300
AP_COMCOP

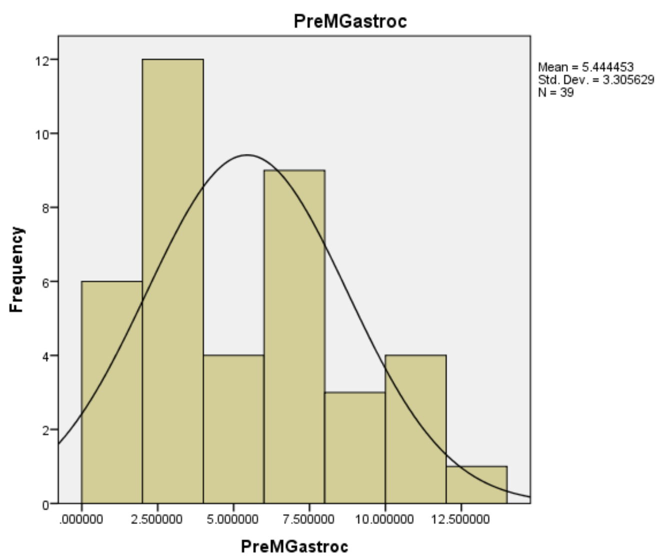

ML_COMCOP
Mean = -.150
Std. Dev. = .021
N = 40
Frequency
0
2
4
6
8
-.200
-.180
-.160
-.140
-.120
-.100
ML_COMCOP

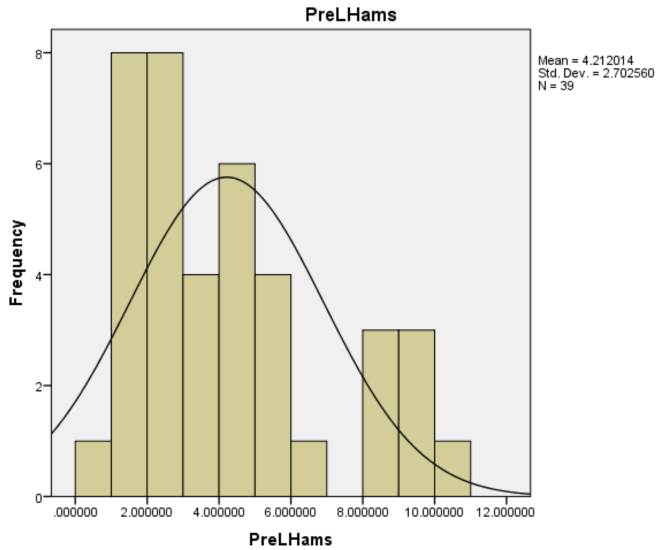

KneeFlexWorkNorm
Mean = -.009535
Std. Dev. = .005556
N = 40
Frequency
0
2
4
6
8
10
-.025000
-.020000
-.015000
-.010000
-.005000
.000000
KneeFlexWorkNorm

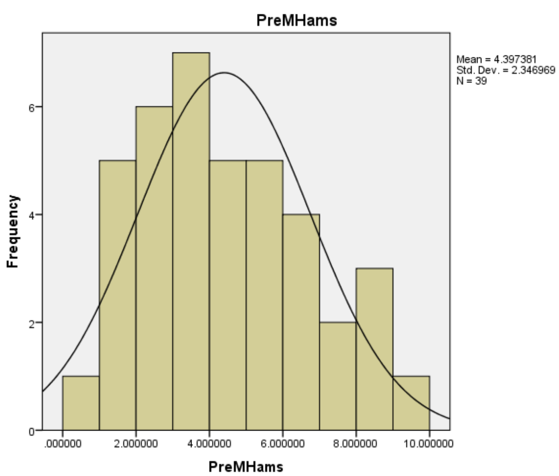

HipFlexWorkNorm
Mean = -.007954
Std. Dev. = .004349
N = 40
Frequency
0
2
4
6
8
-.020000
-.015000
-.010000
-.005000
.000000
HipFlexWorkNorm

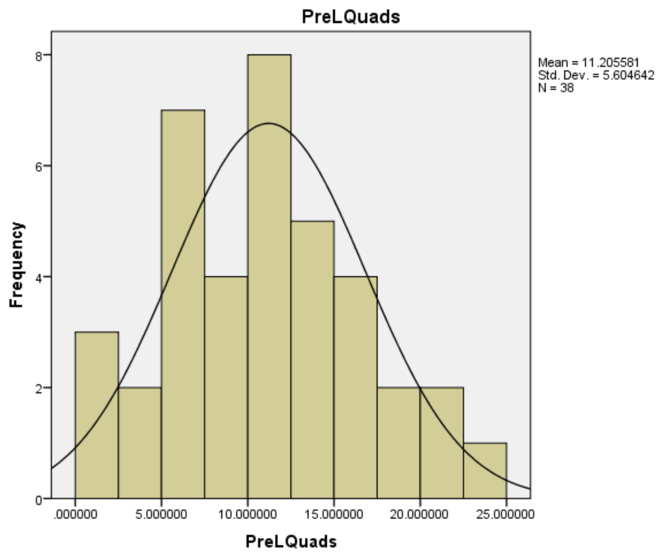

AnkleFlexWorkNorm
Mean = -.015371
Std. Dev. = .006874
N = 39
Frequency
0
2
4
6
8
10
-.040000
-.030000
-.020000
-.010000
.000000
AnkleFlexWorkNorm

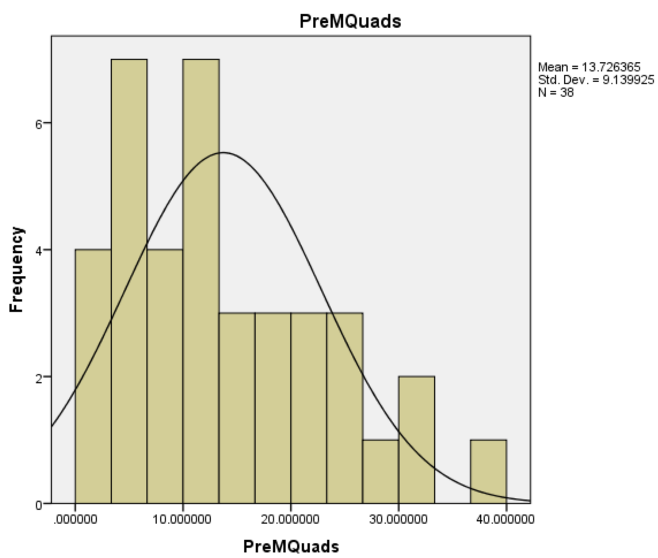

PreLGastroc
Mean = 5.686465
Std. Dev. = 3.177735
N = 39
Frequency
10
8
6
4
2
0
.000000
2.500000
5.000000
7.500000
10.000000
12.500000
PreLGastroc

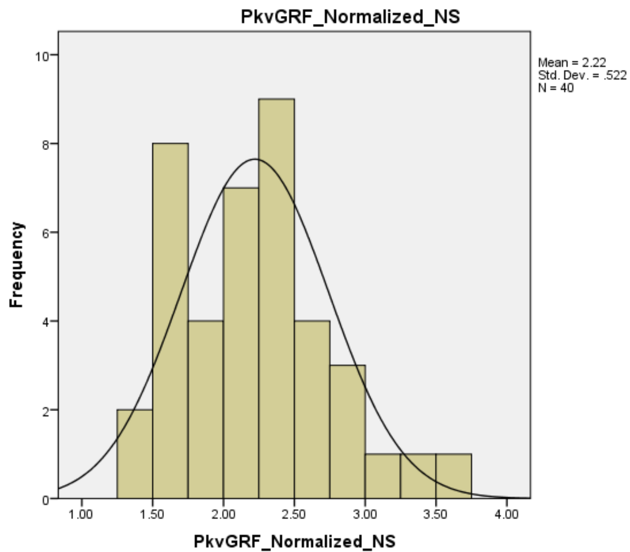

PreMGastroc
Mean = 5.444453
Std. Dev. = 3.305629
N = 39
Frequency
12
10
8
6
4
2
0
.000000
2.500000
5.000000
7.500000
10.000000
12.500000
PreMGastroc

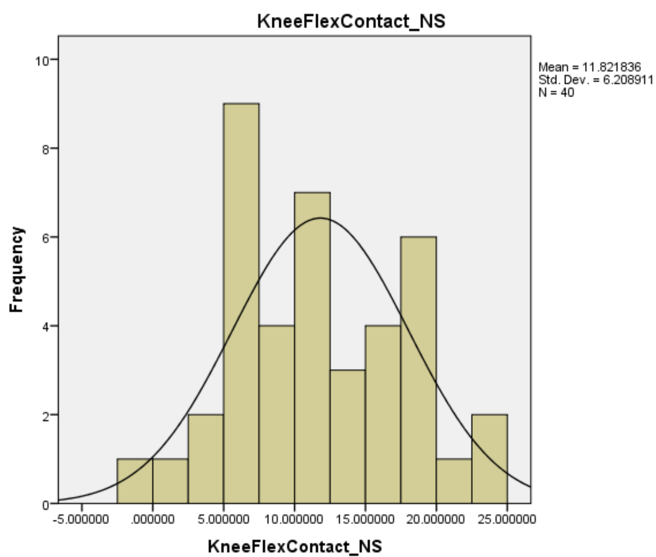

PreLHams
Mean = 4.212014
Std. Dev. = 2.702560
N = 39
Frequency
PreLHams

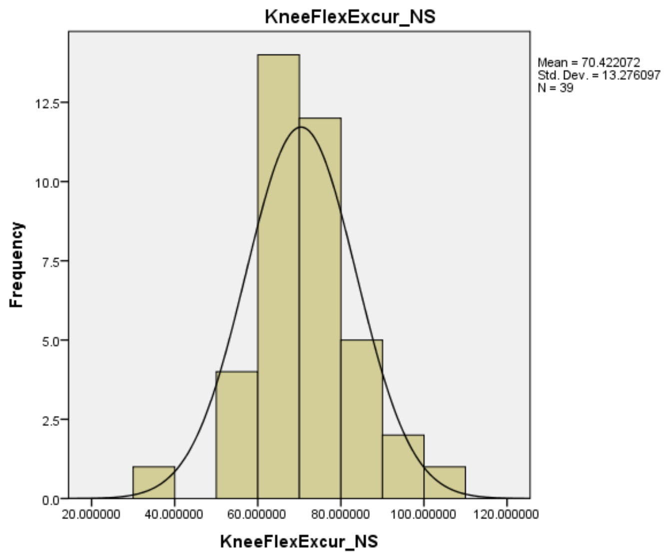

PreMHams
Mean = 4.397381
Std. Dev. = 2.346969
N = 39
Frequency
PreMHams

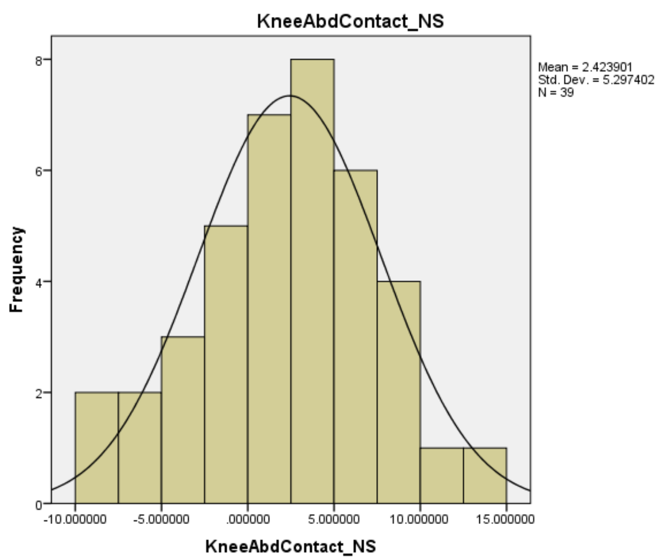

PreLQuads
Mean = 11.205581
Std. Dev. = 5.604642
N = 38
Frequency
0
2
4
6
8
.000000
5.000000
10.000000
15.000000
20.000000
25.000000
PreLQuads

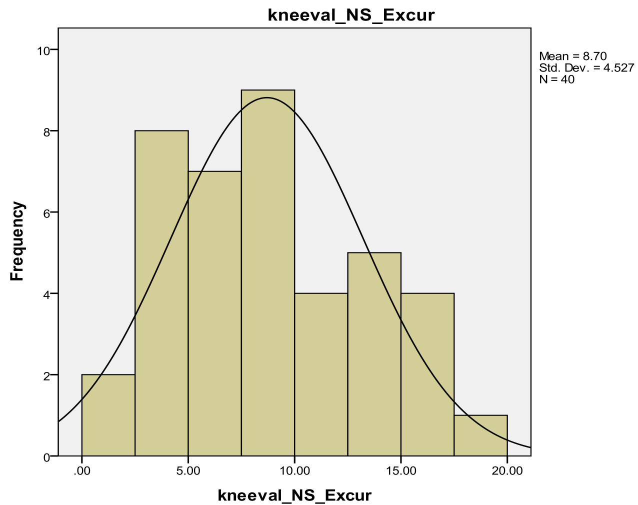

PreMQuads
Mean = 13.726365
Std. Dev. = 9.139925
N = 38
Frequency
0
2
4
6
.000000
10.000000
20.000000
30.000000
40.000000
PreMQuads

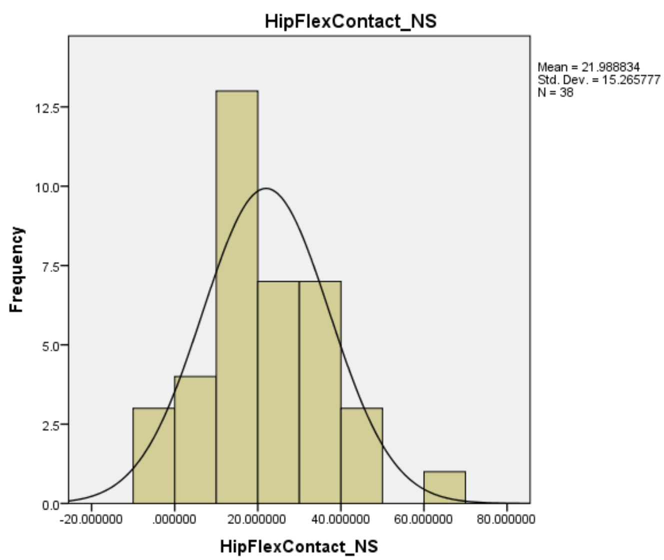

PkvGRF_Normalized_NS
Mean = 2.22
Std. Dev. = .522
N = 40
Frequency
0
2
4
6
8
10
1.00
1.50
2.00
2.50
3.00
3.50
4.00
PkvGRF_Normalized_NS

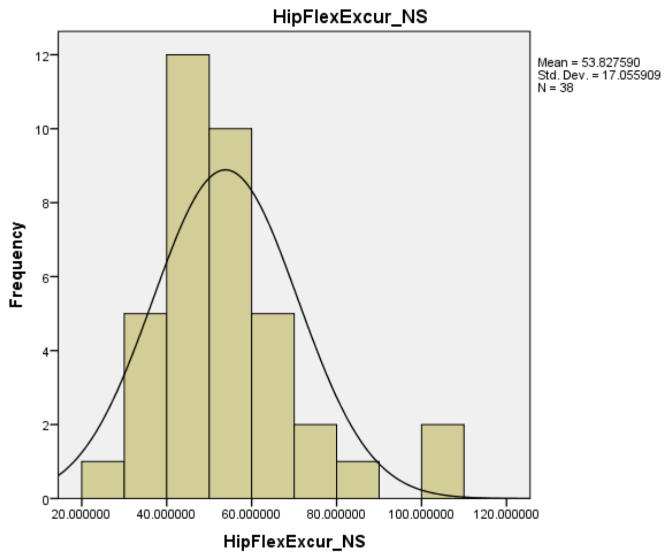

KneeFlexContact_NS
Mean = 11.821836
Std. Dev. = 6.208911
N = 40
Frequency
0
2
4
6
8
10
-5.000000
.000000
5.000000
10.000000
15.000000
20.000000
25.000000
KneeFlexContact_NS

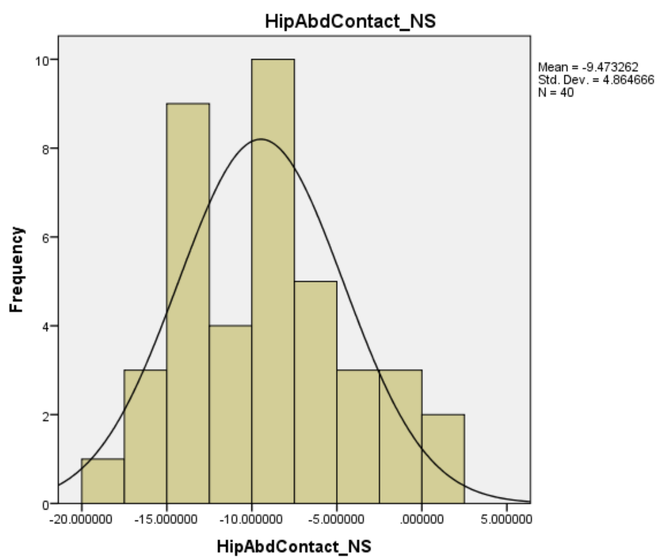

KneeFlexExcur_NS
Mean = 70.422072
Std. Dev. = 13.276097
N = 39
Frequency
0.0
2.5
5.0
7.5
10.0
12.5
20.000000
40.000000
60.000000
80.000000
100.000000
120.000000
KneeFlexExcur_NS

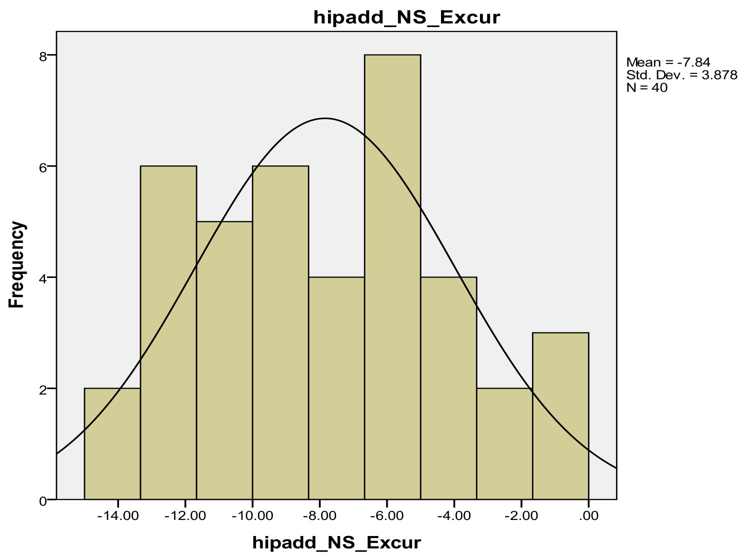

KneeAbdContact_NS
Mean = 2.423901
Std. Dev. = 5.297402
N = 39
Frequency
0
2
4
6
8
-10.000000
-5.000000
.000000
5.000000
10.000000
15.000000
KneeAbdContact_NS

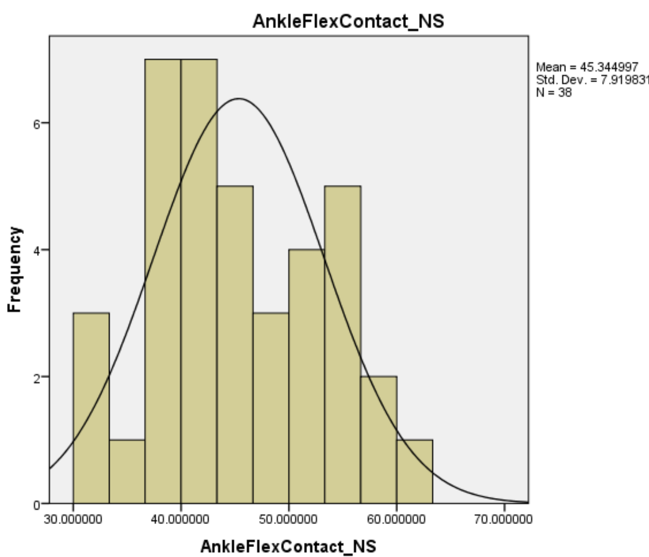

kneeval_NS_Excur
Mean = 8.70
Std. Dev. = 4.527
N = 40
Frequency
10
8
6
4
2
0
.00
5.00
10.00
15.00
20.00
kneeval_NS_Excur

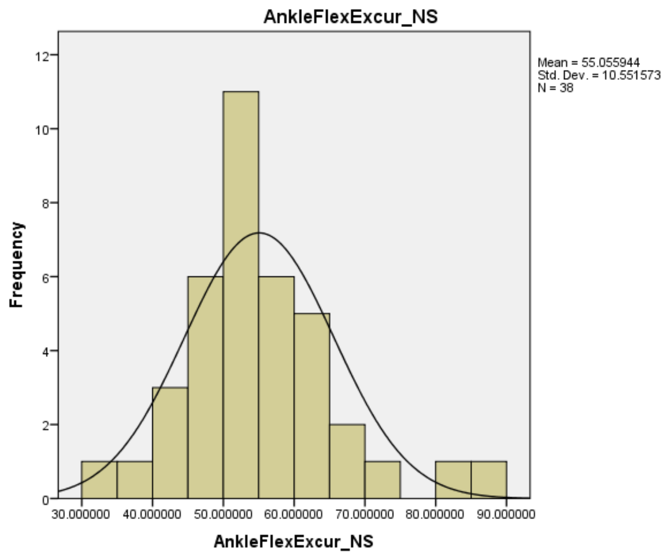

HipFlexContact_NS
Mean = 21.988834
Std. Dev. = 15.265777
N = 38
Frequency
12.5
10.0
7.5
5.0
2.5
0.0
-20.000000
.000000
20.000000
40.000000
60.000000
80.000000
HipFlexContact_NS

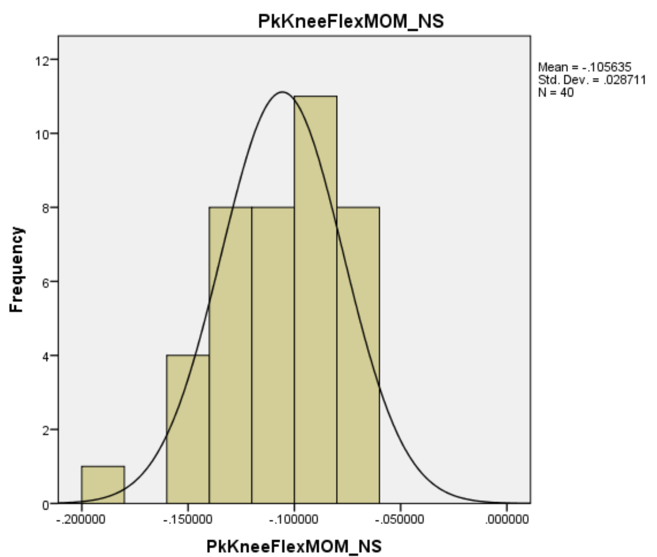

HipFlexExcur_NS
Mean = 53.827590
Std. Dev. = 17.055909
N = 38
Frequency
0
2
4
6
8
10
12
20.000000
40.000000
60.000000
80.000000
100.000000
120.000000
HipFlexExcur_NS

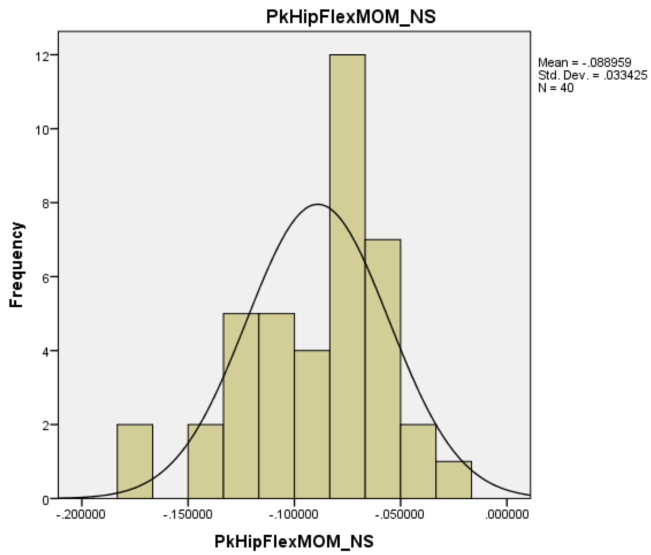

HipAbdContact_NS
Mean = -9.473262
Std. Dev. = 4.864666
N = 40
Frequency
0
2
4
6
8
10
-20.000000
-15.000000
-10.000000
-5.000000
.000000
5.000000
HipAbdContact_NS

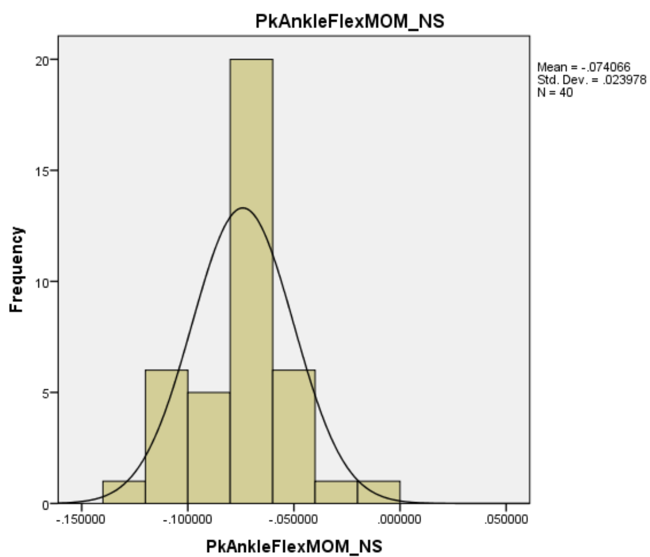

hipadd_NS_Excur
Mean = -7.84
Std. Dev. = 3.878
N = 40
Frequency
0
2
4
6
8
-14.00
-12.00
-10.00
-8.00
-6.00
-4.00
-2.00
.00
hipadd_NS_Excur

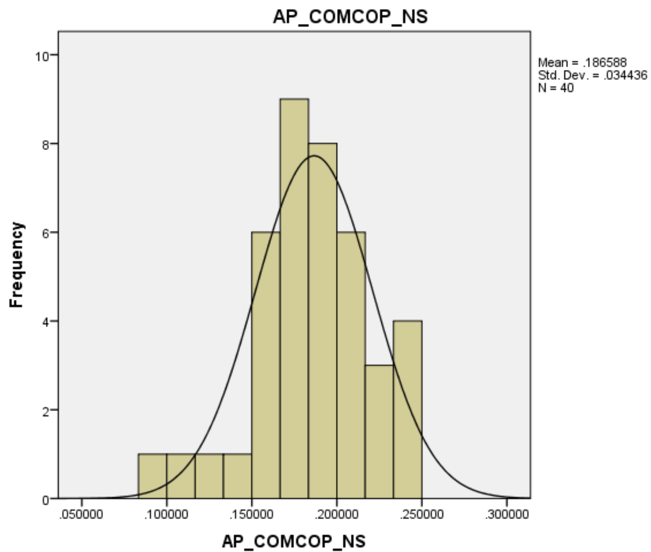

AnkleFlexContact_NS
Mean = 45.344997
Std. Dev. = 7.919831
N = 38
Frequency
0
2
4
6
30.000000
40.000000
50.000000
60.000000
70.000000
AnkleFlexContact_NS


AnkleFlexExcur_NS
Mean = 55.055944
Std. Dev. = 10.551573
N = 38
Frequency
AnkleFlexExcur_NS

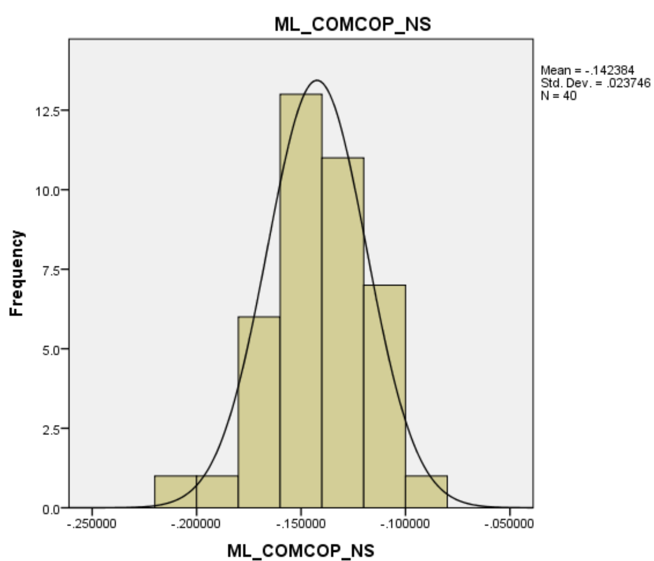

PkKneeFlexMOM_NS
Mean = -.105635
Std. Dev. = .028711
N = 40
Frequency
PkKneeFlexMOM_NS


PkHipFlexMOM_NS
Mean = -.088959
Std. Dev. = .033425
N = 40
Frequency
0
2
4
6
8
10
12
-.200000
-.150000
-.100000
-.050000
.000000
PkHipFlexMOM_NS

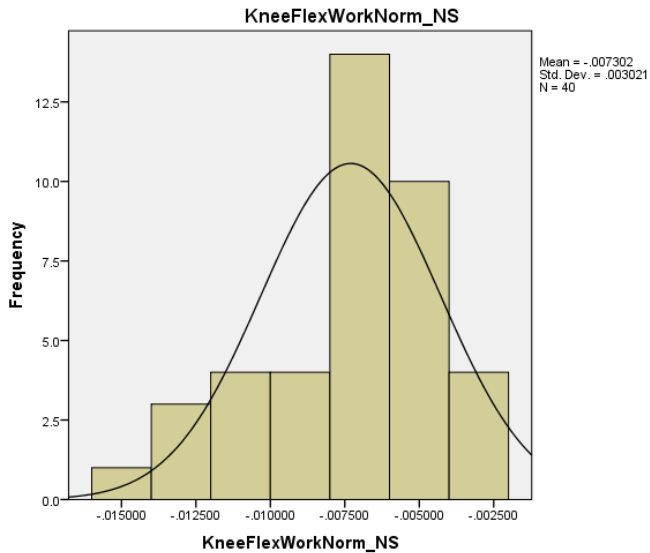

PkAnkleFlexMOM_NS
Mean = -.074066
Std. Dev. = .023978
N = 40
Frequency
0
5
10
15
20
-.150000
-.100000
-.050000
.000000
.050000
PkAnkleFlexMOM_NS



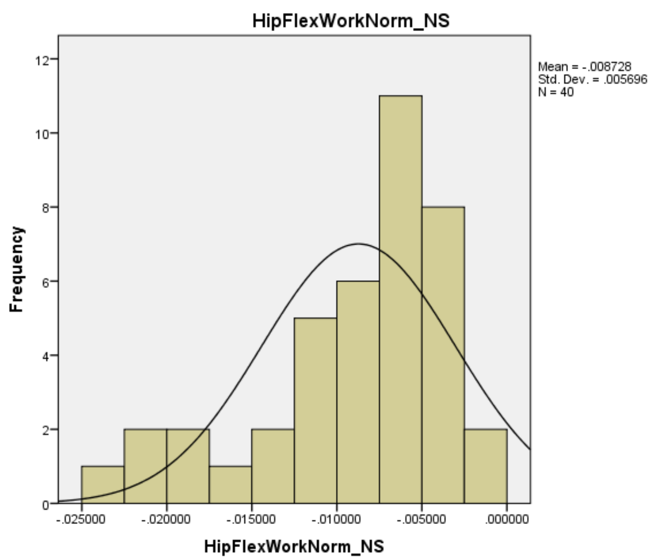
AP_COMCOP_NS
Mean = .186588
Std. Dev. = .034436
N = 40
Frequency
AP_COMCOP_NS



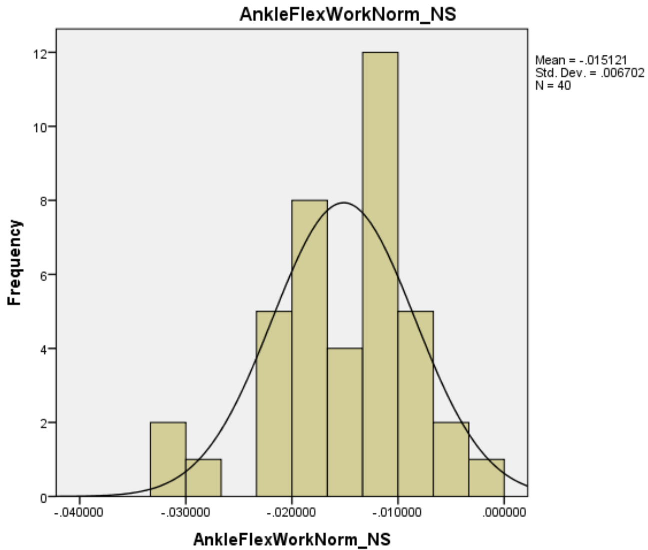
ML_COMCOP_NS
Mean = -.142384
Std. Dev. = .023746
N = 40
Frequency
ML_COMCOP_NS

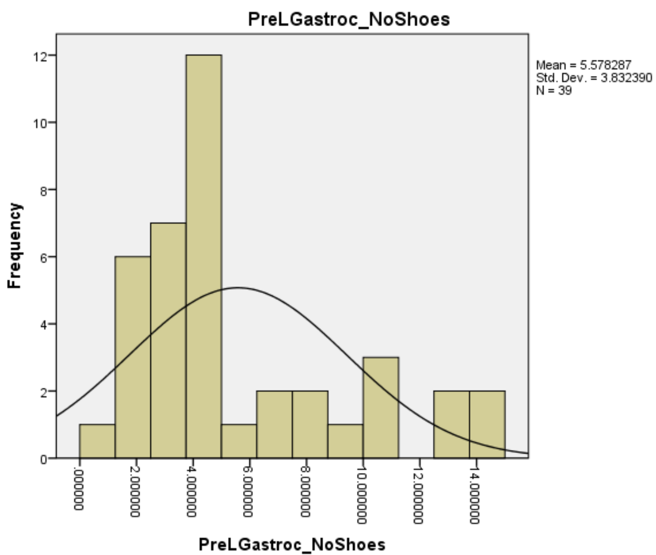


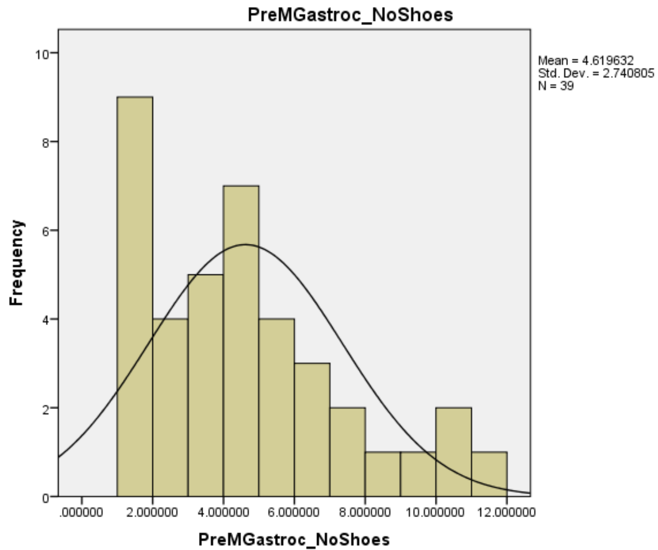
KneeFlexWorkNorm_NS
Mean = -.007302
Std. Dev. = .003021
N = 40
Frequency
0.0
2.5
5.0
7.5
10.0
12.5
-.015000
-.012500
-.010000
-.007500
-.005000
-.002500
KneeFlexWorkNorm_NS

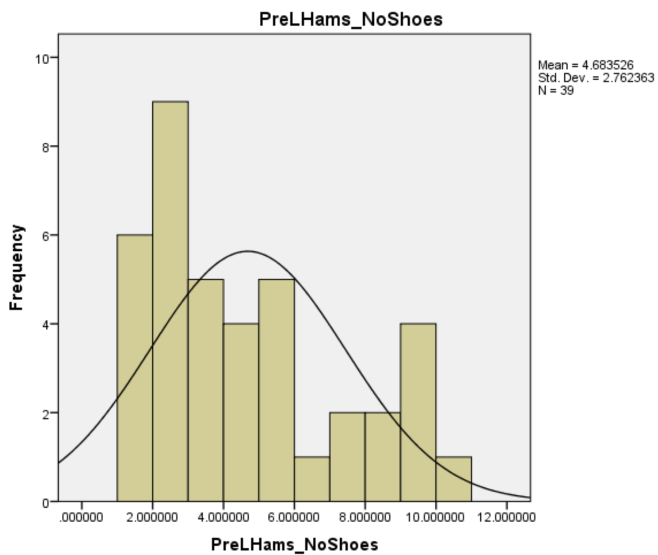


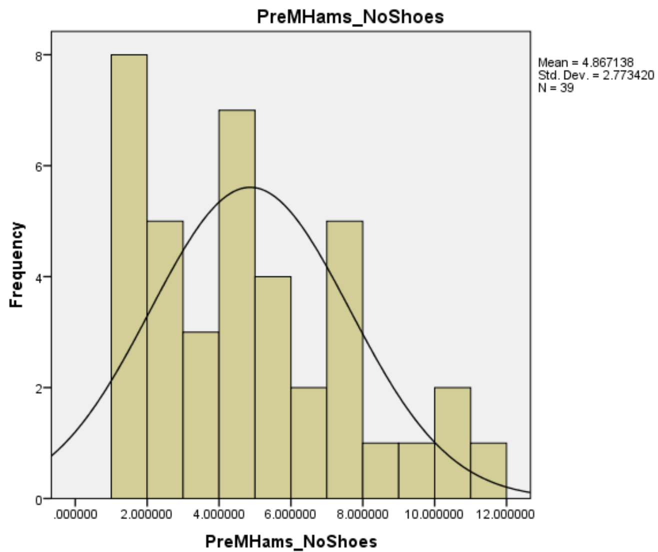
HipFlexWorkNorm_NS
Mean = -.008728
Std. Dev. = .005696
N = 40
Frequency
0
2
4
6
8
10
12
-.025000
-.020000
-.015000
-.010000
-.005000
.000000
HipFlexWorkNorm_NS

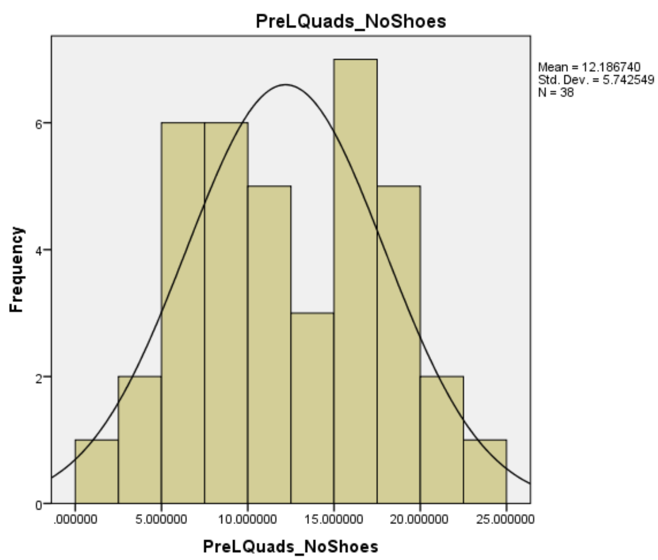

AnkleFlexWorkNorm_NS
Mean = -.015121
Std. Dev. = .006702
N = 40
Frequency
0
2
4
6
8
10
12
-.040000
-.030000
-.020000
-.010000
.000000
AnkleFlexWorkNorm_NS

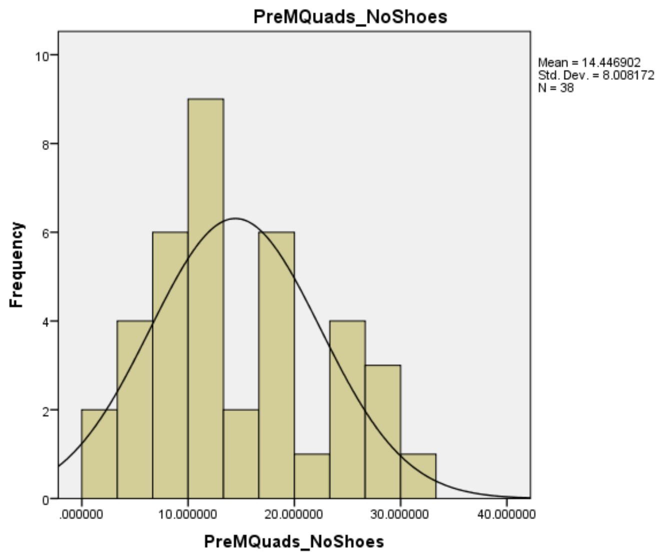

PreLGastroc_NoShoes
Mean = 5.578287
Std. Dev. = 3.832390
N = 39
Frequency
0
2
4
6
8
10
12
.000000
2.000000
4.000000
6.000000
8.000000
10.000000
12.000000
14.000000
PreLGastroc_NoShoes

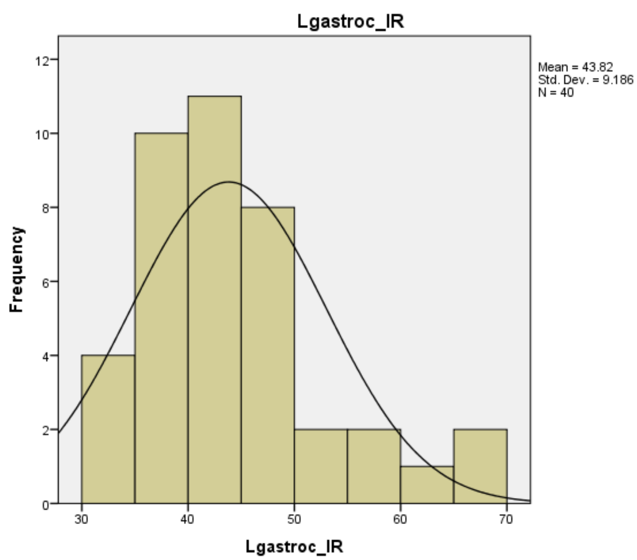

PreMGastroc_NoShoes
Mean = 4.619632
Std. Dev. = 2.740805
N = 39
Frequency
10
8
6
4
2
0
.000000
2.000000
4.000000
6.000000
8.000000
10.000000
12.000000
PreMGastroc_NoShoes

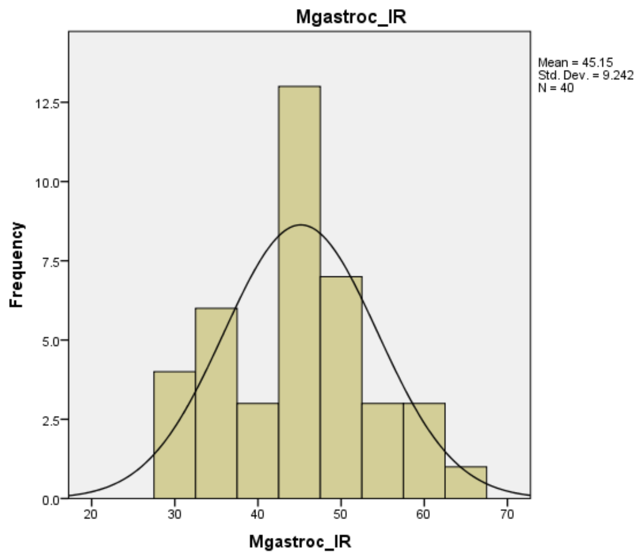

PreLHams_NoShoes
Mean = 4.683526
Std. Dev. = 2.762363
N = 39
Frequency
10
8
6
4
2
0
.000000
2.000000
4.000000
6.000000
8.000000
10.000000
12.000000
PreLHams_NoShoes


PreMHams_NoShoes
Mean = 4.867138
Std. Dev. = 2.773420
N = 39
Frequency
0
2
4
6
8
.000000
2.000000
4.000000
6.000000
8.000000
10.000000
12.000000
PreMHams_NoShoes

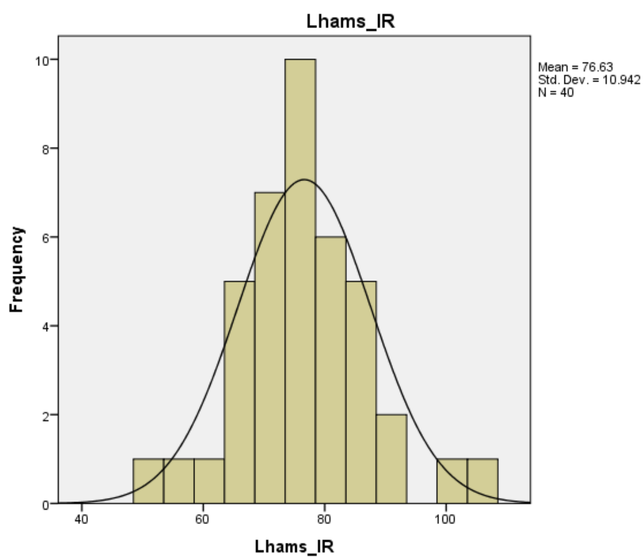

PreLQuads_NoShoes
Mean = 12.186740
Std. Dev. = 5.742549
N = 38
Frequency
0
2
4
6
.000000
5.000000
10.000000
15.000000
20.000000
25.000000
PreLQuads_NoShoes


PreMQuads_NoShoes
Mean = 14.446902
Std. Dev. = 8.008172
N = 38
Frequency
0
2
4
6
8
10
.000000
10.000000
20.000000
30.000000
40.000000
PreMQuads_NoShoes

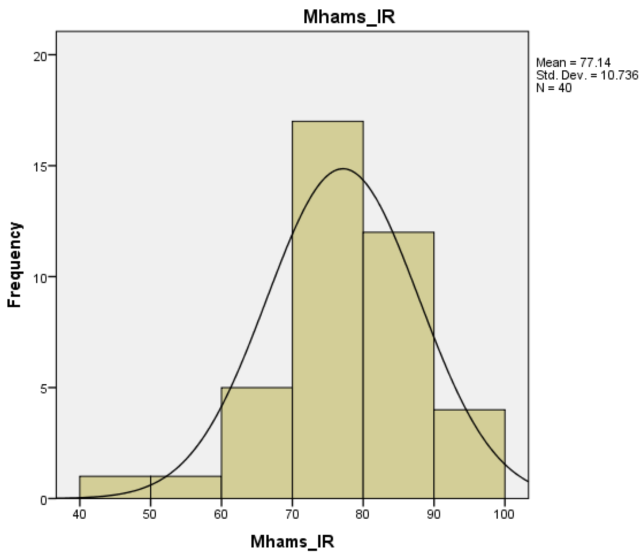

Lgastroc_IR
Mean = 43.82
Std. Dev. = 9.186
N = 40
Frequency
0
2
4
6
8
10
12
30
40
50
60
70
Lgastroc_IR

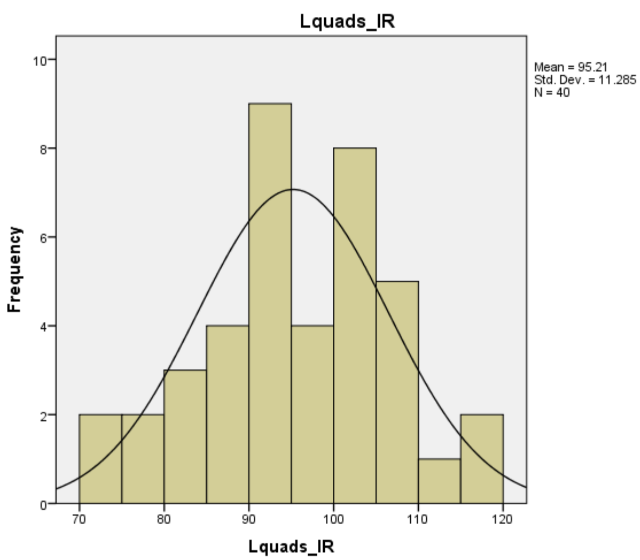


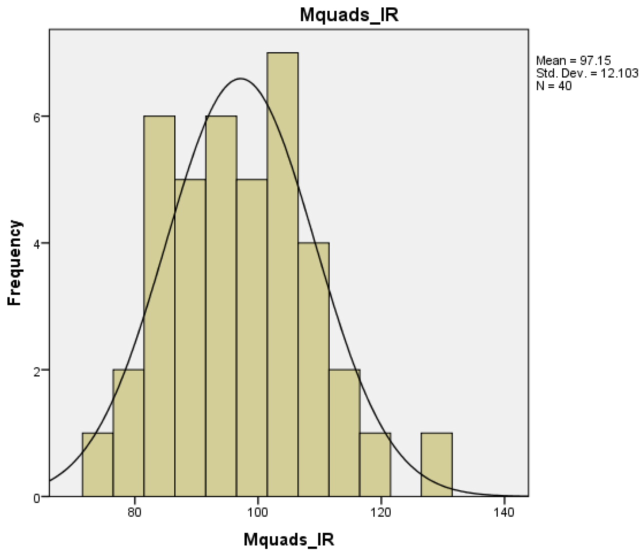
Mgastroc_IR
Mean = 45.15
Std. Dev. = 9.242
N = 40
Frequency
12.5
10.0
7.5
5.0
2.5
0.0
20
30
40
50
60
70
Mgastroc_IR

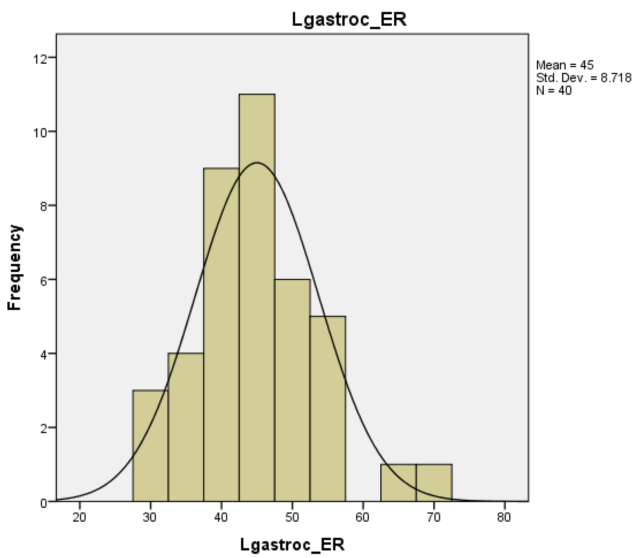


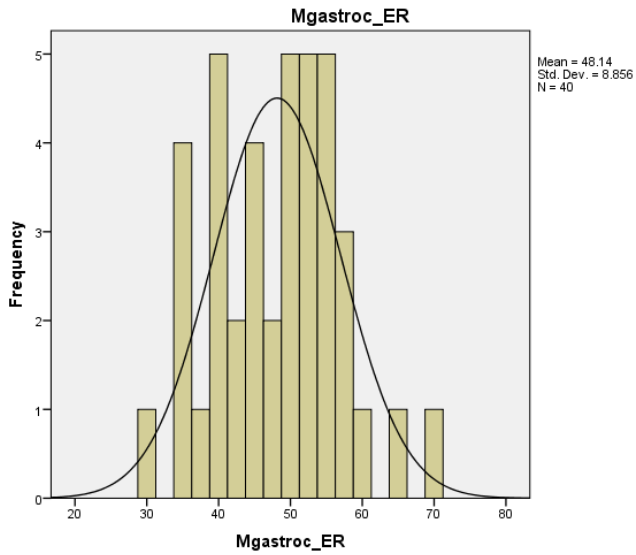
Lhams_IR
Mean = 76.63
Std. Dev. = 10.942
N = 40
Frequency
10
8
6
4
2
0
40
60
80
100
Lhams_IR

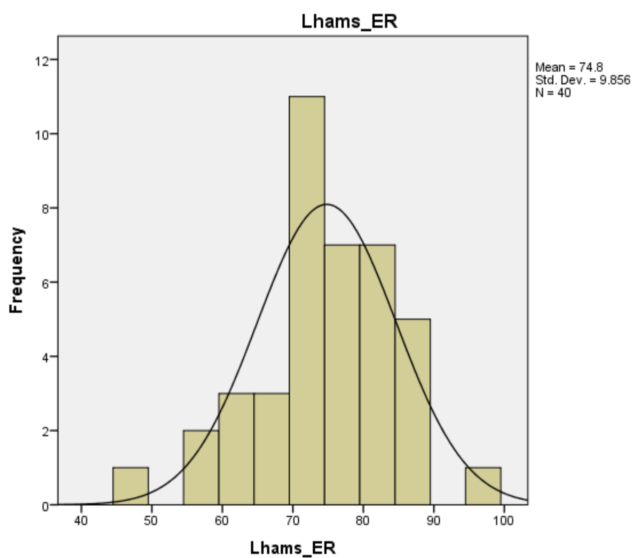

Mhams_IR
Mean = 77.14
Std. Dev. = 10.736
N = 40
Frequency
20
15
10
5
0
40
50
60
70
80
90
100
Mhams_IR

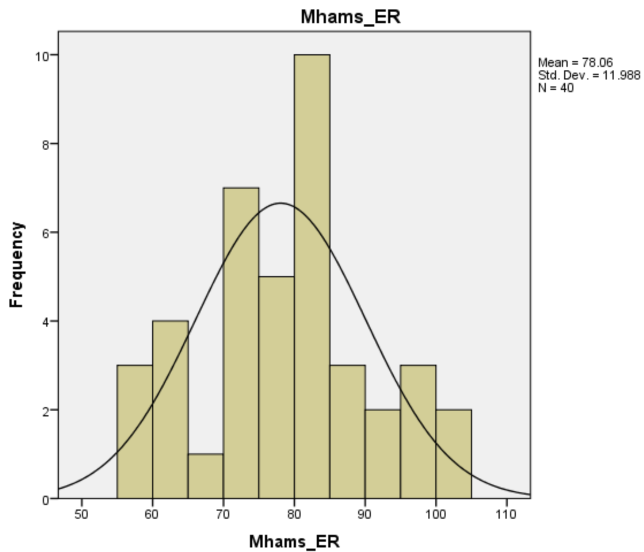

Lquads_IR
Mean = 95.21
Std. Dev. = 11.285
N = 40
Frequency
10
8
6
4
2
0
70
80
90
100
110
120
Lquads_IR

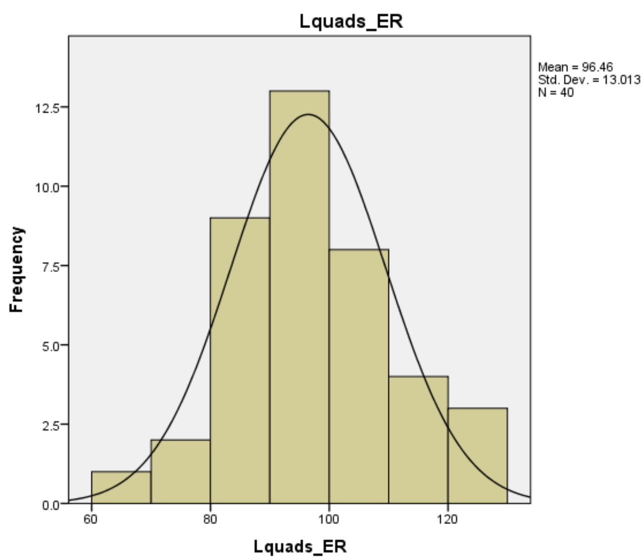

Mquads_IR
Mean = 97.15
Std. Dev. = 12.103
N = 40
Frequency
0
2
4
6
80
100
120
140
Mquads_IR

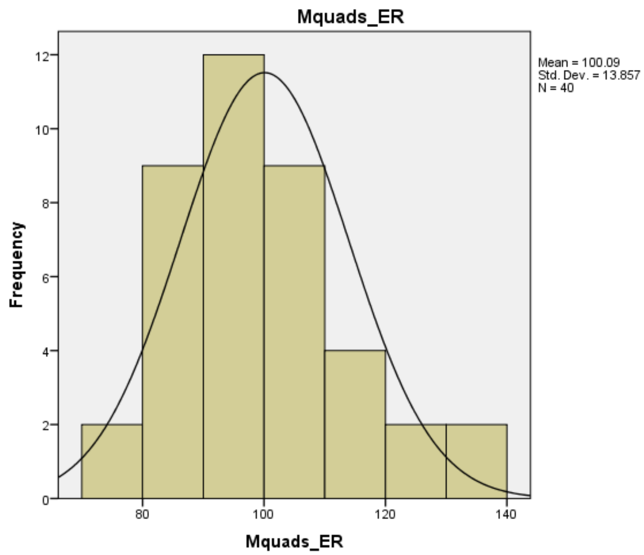

Lgastroc_ER
Mean = 45
Std. Dev. = 8.718
N = 40
Frequency
0
2
4
6
8
10
12
20
30
40
50
60
70
80
Lgastroc_ER

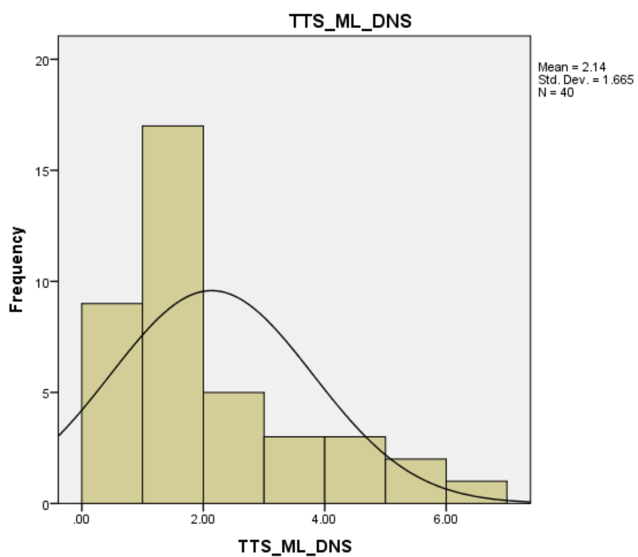

Mgastroc_ER
Mean = 48.14
Std. Dev. = 8.856
N = 40
Frequency
Mgastroc_ER

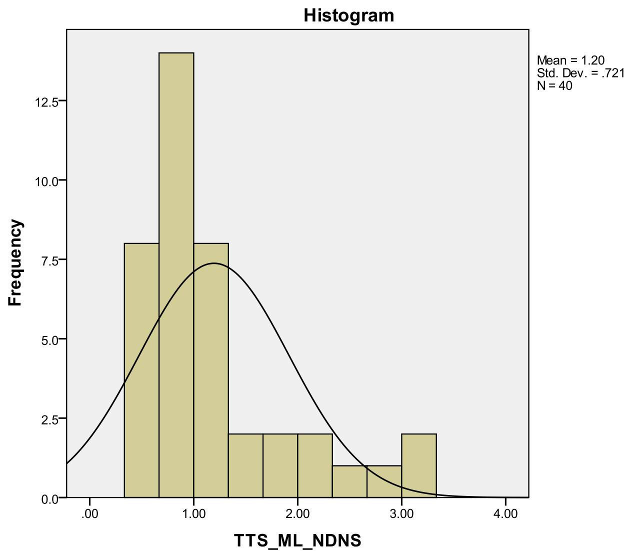

Lhams_ER
Mean = 74.8
Std. Dev. = 9.856
N = 40
Frequency
Lhams_ER

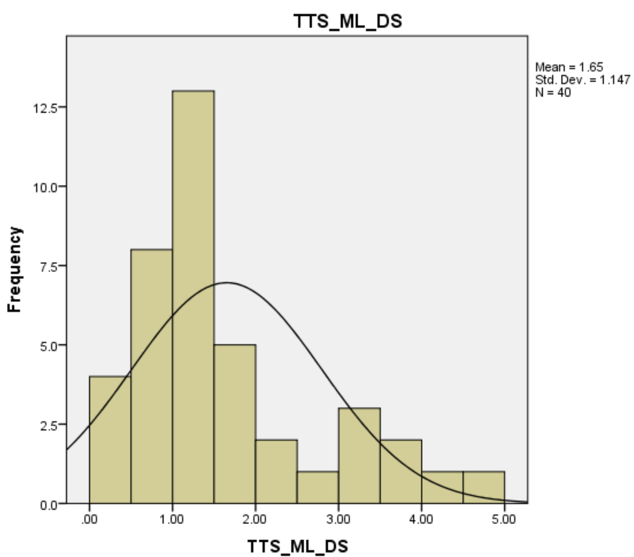

Mhams_ER
Mean = 78.06
Std. Dev. = 11.988
N = 40
Frequency
Mhams_ER

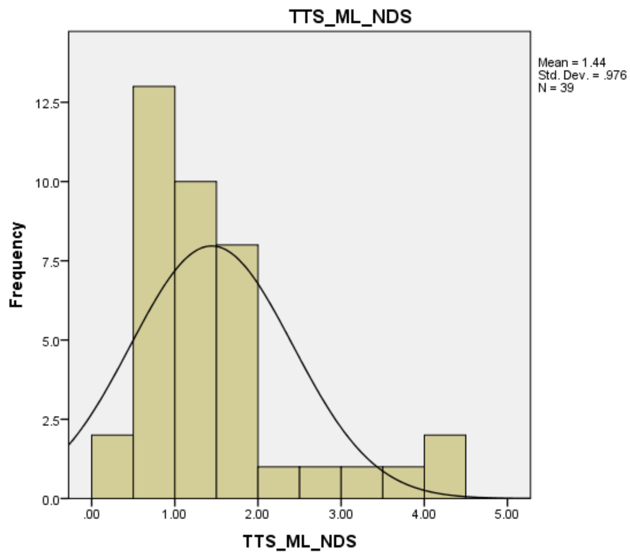

Lquads_ER
Mean = 96.46
Std. Dev. = 13.013
N = 40
Frequency
Lquads_ER

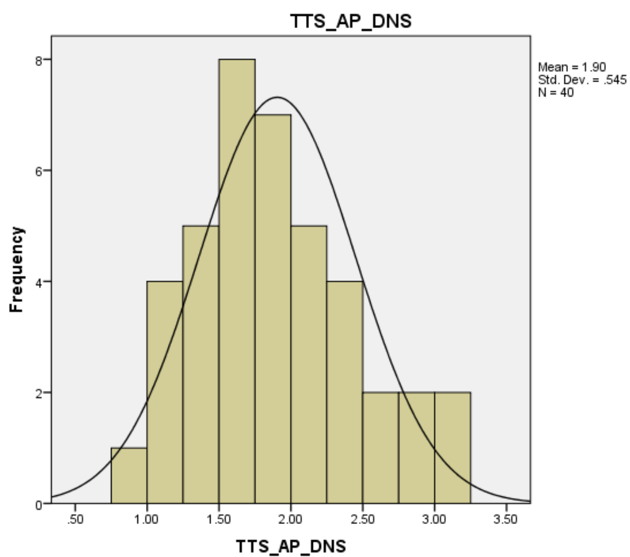

Mquads_ER
Mean = 100.09
Std. Dev. = 13.857
N = 40
Frequency
0
2
4
6
8
10
12
80
100
120
140
Mquads_ER

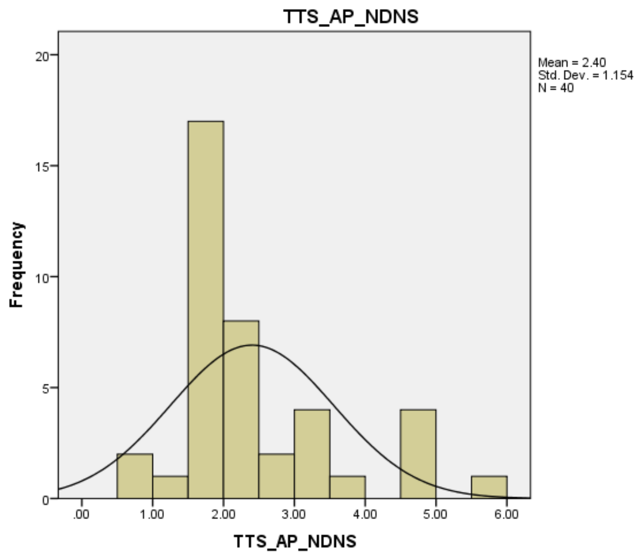

TTS_ML_DNS
Mean = 2.14
Std. Dev. = 1.665
N = 40
Frequency
0
5
10
15
20
.00
2.00
4.00
6.00
TTS_ML_DNS

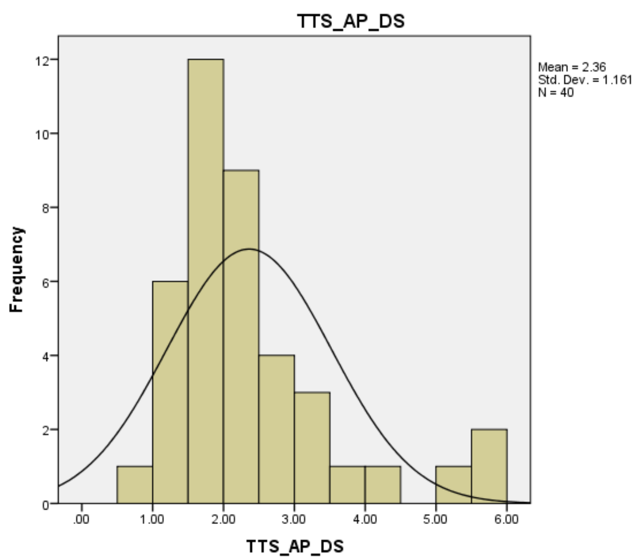

Histogram
Mean = 1.20
Std. Dev. = .721
N = 40
Frequency
0.0
2.5
5.0
7.5
10.0
12.5
.00
1.00
2.00
3.00
4.00
TTS_ML_NDNS

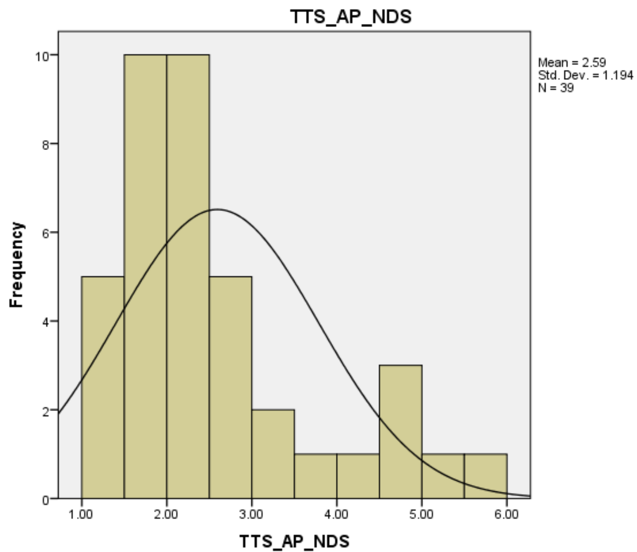

TTS_ML_DS
Mean = 1.65
Std. Dev. = 1.147
N = 40
Frequency
0.0
2.5
5.0
7.5
10.0
12.5
.00
1.00
2.00
3.00
4.00
5.00
TTS_ML_DS


TTS_ML_NDS
Mean = 1.44
Std. Dev. = .976
N = 39
Frequency
12.5
10.0
7.5
5.0
2.5
0.0
.00
1.00
2.00
3.00
4.00
5.00
TTS_ML_NDS


TTS_AP_DNS
Mean = 1.90
Std. Dev. = .545
N = 40
Frequency
8
6
4
2
0
.50
1.00
1.50
2.00
2.50
3.00
3.50
TTS_AP_DNS


TTS_AP_NDNS
Mean = 2.40
Std. Dev. = 1.154
N = 40
Frequency
TTS_AP_NDNS


TTS_AP_DS
Mean = 2.36
Std. Dev. = 1.161
N = 40
Frequency
TTS_AP_DS


TTS_AP_NDS
Mean = 2.59
Std. Dev. = 1.194
N = 39
Frequency
0
2
4
6
8
10
1.00
2.00
3.00
4.00
5.00
6.00
TTS_AP_NDS

APPENDIX E

SPSS OUTPUT OF STATISTICAL ANALYSES

Appendix E.1. 2 (group) x 2 (plane) x 2 (limb) x 2 (footwear) Repeated Measures ANOVA to assess TTS.

Within-Subjects Factors

Measure: MEASURE_1

Plane	Footwear	Limb	Dependent Variable
1	1	1	TTS_AP_DNS
		2	TTS_AP_NDNS
	2	1	TTS_AP_DS
		2	TTS_AP_NDS
2	1	1	TTS_ML_DNS
		2	TTS_ML_NDNS
	2	1	TTS_ML_DS
		2	TTS_ML_NDS

Descriptive Statistics

	Group	Mean	Std. Deviation	N
TTS_AP_DNS	Dance	1.8500	.61061	20
	Athlete	1.9592	.48084	20
	Total	1.9046	.54529	40
TTS_AP_NDNS	Dance	2.3186	1.21668	20
	Athlete	2.4843	1.11308	20
	Total	2.4015	1.15404	40
TTS_AP_DS	Dance	1.9510	.85830	20
	Athlete	2.7713	1.29436	20
	Total	2.3611	1.16087	40
TTS_AP_NDS	Dance	2.3995	1.18690	20
	Athlete	2.7599	1.18037	20
	Total	2.5797	1.18253	40
TTS_ML_DNS	Dance	1.8035	1.37535	20
	Athlete	2.4720	1.88764	20
	Total	2.1378	1.66495	40
TTS_ML_NDNS	Dance	1.0238	.68850	20
	Athlete	1.3840	.78052	20
	Total	1.2039	.74900	40
TTS_ML_DS	Dance	1.5025	.92072	20
	Athlete	1.8013	1.34329	20
	Total	1.6519	1.14672	40
TTS_ML_NDS	Dance	1.2220	.82037	20
	Athlete	1.6400	1.07412	20
	Total	1.4310	.96682	40

Multivariate Tests[a]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power[c]
Plane	Pillai's Trace	.463	32.787[b]	1.000	38.000	.000	.463	32.787	1.000
	Wilks' Lambda	.537	32.787[b]	1.000	38.000	.000	.463	32.787	1.000
	Hotelling's Trace	.863	32.787[b]	1.000	38.000	.000	.463	32.787	1.000
	Roy's Largest Root	.863	32.787[b]	1.000	38.000	.000	.463	32.787	1.000
Plane * Group	Pillai's Trace	.002	.087[b]	1.000	38.000	.770	.002	.087	.059
	Wilks' Lambda	.998	.087[b]	1.000	38.000	.770	.002	.087	.059
	Hotelling's Trace	.002	.087[b]	1.000	38.000	.770	.002	.087	.059
	Roy's Largest Root	.002	.087[b]	1.000	38.000	.770	.002	.087	.059
Footwear	Pillai's Trace	.020	.756[b]	1.000	38.000	.390	.020	.756	.135
	Wilks' Lambda	.980	.756[b]	1.000	38.000	.390	.020	.756	.135
	Hotelling's Trace	.020	.756[b]	1.000	38.000	.390	.020	.756	.135
	Roy's Largest Root	.020	.756[b]	1.000	38.000	.390	.020	.756	.135
Footwear * Group	Pillai's Trace	.012	.471[b]	1.000	38.000	.497	.012	.471	.103
	Wilks' Lambda	.988	.471[b]	1.000	38.000	.497	.012	.471	.103
	Hotelling's Trace	.012	.471[b]	1.000	38.000	.497	.012	.471	.103
	Roy's Largest Root	.012	.471[b]	1.000	38.000	.497	.012	.471	.103
Limb	Pillai's Trace	.025	.955[b]	1.000	38.000	.335	.025	.955	.159
	Wilks' Lambda	.975	.955[b]	1.000	38.000	.335	.025	.955	.159
	Hotelling's Trace	.025	.955[b]	1.000	38.000	.335	.025	.955	.159
	Roy's Largest Root	.025	.955[b]	1.000	38.000	.335	.025	.955	.159
Limb * Group	Pillai's Trace	.011	.434[b]	1.000	38.000	.514	.011	.434	.098
	Wilks' Lambda	.989	.434[b]	1.000	38.000	.514	.011	.434	.098
	Hotelling's Trace	.011	.434[b]	1.000	38.000	.514	.011	.434	.098
	Roy's Largest Root	.011	.434[b]	1.000	38.000	.514	.011	.434	.098
Plane * Footwear	Pillai's Trace	.238	11.867[b]	1.000	38.000	.001	.238	11.867	.919
	Wilks' Lambda	.762	11.867[b]	1.000	38.000	.001	.238	11.867	.919
	Hotelling's Trace	.312	11.867[b]	1.000	38.000	.001	.238	11.867	.919
	Roy's Largest Root	.312	11.867[b]	1.000	38.000	.001	.238	11.867	.919
Plane * Footwear * Group	Pillai's Trace	.127	5.510[b]	1.000	38.000	.024	.127	5.510	.629
	Wilks' Lambda	.873	5.510[b]	1.000	38.000	.024	.127	5.510	.629
	Hotelling's Trace	.145	5.510[b]	1.000	38.000	.024	.127	5.510	.629
	Roy's Largest Root	.145	5.510[b]	1.000	38.000	.024	.127	5.510	.629
Plane * Limb	Pillai's Trace	.314	17.363[b]	1.000	38.000	.000	.314	17.363	.982
	Wilks' Lambda	.686	17.363[b]	1.000	38.000	.000	.314	17.363	.982
	Hotelling's Trace	.457	17.363[b]	1.000	38.000	.000	.314	17.363	.982
	Roy's Largest Root	.457	17.363[b]	1.000	38.000	.000	.314	17.363	.982
Plane * Limb * Group	Pillai's Trace	.001	.057[b]	1.000	38.000	.813	.001	.057	.056
	Wilks' Lambda	.999	.057[b]	1.000	38.000	.813	.001	.057	.056
	Hotelling's Trace	.001	.057[b]	1.000	38.000	.813	.001	.057	.056
	Roy's Largest Root	.001	.057[b]	1.000	38.000	.813	.001	.057	.056
Footwear * Limb	Pillai's Trace	.034	1.357[b]	1.000	38.000	.251	.034	1.357	.206
	Wilks' Lambda	.966	1.357[b]	1.000	38.000	.251	.034	1.357	.206
	Hotelling's Trace	.036	1.357[b]	1.000	38.000	.251	.034	1.357	.206
	Roy's Largest Root	.036	1.357[b]	1.000	38.000	.251	.034	1.357	.206
Footwear * Limb * Group	Pillai's Trace	.000	.014[b]	1.000	38.000	.906	.000	.014	.052
	Wilks' Lambda	1.000	.014[b]	1.000	38.000	.906	.000	.014	.052
	Hotelling's Trace	.000	.014[b]	1.000	38.000	.906	.000	.014	.052
	Roy's Largest Root	.000	.014[b]	1.000	38.000	.906	.000	.014	.052
Plane * Footwear * Limb	Pillai's Trace	.205	9.773[b]	1.000	38.000	.003	.205	9.773	.861
	Wilks' Lambda	.795	9.773[b]	1.000	38.000	.003	.205	9.773	.861
	Hotelling's Trace	.257	9.773[b]	1.000	38.000	.003	.205	9.773	.861
	Roy's Largest Root	.257	9.773[b]	1.000	38.000	.003	.205	9.773	.861
Plane * Footwear * Limb * Group	Pillai's Trace	.055	2.215[b]	1.000	38.000	.145	.055	2.215	.306
	Wilks' Lambda	.945	2.215[b]	1.000	38.000	.145	.055	2.215	.306
	Hotelling's Trace	.058	2.215[b]	1.000	38.000	.145	.055	2.215	.306
	Roy's Largest Root	.058	2.215[b]	1.000	38.000	.145	.055	2.215	.306

a. Design: Intercept + Group
Within Subjects Design: Plane + Footwear + Limb + Plane * Footwear + Plane * Limb + Footwear * Limb + Plane * Footwear * Limb

b. Exact statistic

c. Computed using alpha = .05

Tests of Between-Subjects Effects

Measure: MEASURE_1 ...

	Tvpe III Sum					Partial Eta	Noncent.	Observed
Intercept	1227.961	1	1227.961	301.347	.000	.888	301.347	1.000
Group	12.808	1	12.808	3.143	.084	.076	3.143	.408

a. Computed using alpha = .05

Appendix E.2. 2 (group) x 2 (footwear) Repeated Measures ANOVA to assess COM to COP displacement.

Within-Subjects Factors

Measure: MEASURE_1

Footwear	Dependent Variable
1	AP_COMCOP_NS
2	AP_COMCOP

Descriptive Statistics

	Group	Mean	Std. Deviation	N
AP_COMCOP_NS	Dance	.17496965	.037940991	20
	Athlete	.19820680	.026651816	20
	Total	.18658823	.034435588	40
AP_COMCOP	Dance	.17486	.028619	20
	Athlete	.19099	.031308	20
	Total	.18293	.030712	40

Multivariate Tests[a]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power[c]
Footwear	Pillai's Trace	.018	.716[b]	1.000	38.000	.403	.018	.716	.131
	Wilks' Lambda	.982	.716[b]	1.000	38.000	.403	.018	.716	.131
	Hotelling's Trace	.019	.716[b]	1.000	38.000	.403	.018	.716	.131
	Roy's Largest Root	.019	.716[b]	1.000	38.000	.403	.018	.716	.131
Footwear * Group	Pillai's Trace	.017	.675[b]	1.000	38.000	.417	.017	.675	.126
	Wilks' Lambda	.983	.675[b]	1.000	38.000	.417	.017	.675	.126
	Hotelling's Trace	.018	.675[b]	1.000	38.000	.417	.017	.675	.126
	Roy's Largest Root	.018	.675[b]	1.000	38.000	.417	.017	.675	.126

a. Design: Intercept + Group
Within Subjects Design: Footwear

b. Exact statistic

c. Computed using alpha = .05

Tests of Between-Subjects Effects

Measure: MEASURE_1 ...

	Type III Sum					Partial Eta	Noncent.	Observed
Intercept	2.731	1	2.731	1706.709	.000	.978	1706.709	1.000
Group	.008	1	.008	4.843	.034	.113	4.843	.573

a. Computed using alpha = .05

Appendix E.3. 2 (group) x 2 (footwear) Repeated Measures Multivariate ANOVA to assess hip, knee, and ankle flexion at initial ground contact.

Within-Subjects Factors

Measure	footwear	Dependent Variable
Hip	1	HipFlexContact
	2	HipFlexContact_NS
Knee	1	KneeFlexContact
	2	KneeFlexContact_NS
Ankle	1	AnkleFlexContact
	2	AnkleFlexContact_NS

Descriptive Statistics

	Group	Mean	Std. Deviation	N
HipFlexContact	Dance	17.08664033	12.68180272	20
	Athlete	25.63889002	13.63647798	20
	Total	21.36276517	13.70032859	40
HipFlexContact_NS	Dance	22.17230838	16.43149398	20
	Athlete	21.80637269	13.55587919	20
	Total	21.98934053	14.86925958	40
KneeFlexContact	Dance	10.56852013	6.106426686	20
	Athlete	13.17374949	6.343910098	20
	Total	11.87113481	6.285945662	40
KneeFlexContact_NS	Dance	9.82058928	6.353078767	20
	Athlete	13.82308328	5.507899086	20
	Total	11.82183628	6.208910601	40
AnkleFlexContact	Dance	43.44364362	6.809834509	20
	Athlete	48.76623330	6.712242455	20
	Total	46.10493846	7.197635759	40
AnkleFlexContact_NS	Dance	43.45968411	6.748772759	20
	Athlete	47.21989418	8.340981207	20
	Total	45.33978915	7.727128238	40

Multivariate Tests[a]

Effect			Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power[c]
Between Subjects	Intercept	Pillai's Trace	.984	743.732[b]	3.000	36.000	.000	.984	2231.195	1.000
		Wilks' Lambda	.016	743.732[b]	3.000	36.000	.000	.984	2231.195	1.000
		Hotelling's Trace	61.978	743.732[b]	3.000	36.000	.000	.984	2231.195	1.000
		Roy's Largest Root	61.978	743.732[b]	3.000	36.000	.000	.984	2231.195	1.000
	Group	Pillai's Trace	.156	2.211[b]	3.000	36.000	.104	.156	6.633	.515
		Wilks' Lambda	.844	2.211[b]	3.000	36.000	.104	.156	6.633	.515
		Hotelling's Trace	.184	2.211[b]	3.000	36.000	.104	.156	6.633	.515
		Roy's Largest Root	.184	2.211[b]	3.000	36.000	.104	.156	6.633	.515
Within Subjects	footwear	Pillai's Trace	.014	.176[b]	3.000	36.000	.912	.014	.527	.079
		Wilks' Lambda	.986	.176[b]	3.000	36.000	.912	.014	.527	.079
		Hotelling's Trace	.015	.176[b]	3.000	36.000	.912	.014	.527	.079
		Roy's Largest Root	.015	.176[b]	3.000	36.000	.912	.014	.527	.079
	footwear * Group	Pillai's Trace	.100	1.340[b]	3.000	36.000	.277	.100	4.020	.326
		Wilks' Lambda	.900	1.340[b]	3.000	36.000	.277	.100	4.020	.326
		Hotelling's Trace	.112	1.340[b]	3.000	36.000	.277	.100	4.020	.326
		Roy's Largest Root	.112	1.340[b]	3.000	36.000	.277	.100	4.020	.326

a. Design: Intercept + Group
Within Subjects Design: footwear

b. Exact statistic

c. Computed using alpha = .05

Tests of Between-Subjects Effects

Transformed Variable: Average

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power[a]
Intercept	Hip	37588.101	1	37588.101	154.053	.000	.802	154.053	1.000
	Knee	11227.138	1	11227.138	176.337	.000	.823	176.337	1.000
	Ankle	167242.764	1	167242.764	2202.346	.000	.983	2202.346	1.000
Group	Hip	335.079	1	335.079	1.373	.249	.035	1.373	.208
	Knee	218.310	1	218.310	3.429	.072	.083	3.429	.438
	Ankle	412.486	1	412.486	5.432	.025	.125	5.432	.622
Error	Hip	9271.820	38	243.995					
	Knee	2419.415	38	63.669					
	Ankle	2885.661	38	75.938					

a. Computed using alpha = .05

Appendix E.4. 2 (group) x 2 (footwear) Repeated Measures Multivariate ANOVA to assess hip, knee, and ankle flexion excursion.

Within-Subjects Factors

Measure	footwear	Dependent Variable
Hip	1	HipFlexExcur
	2	HipFlexExcur_NS
Knee	1	KneeFlexExcur
	2	KneeFlexExcur_NS
Ankle	1	AnkleFlexExcur
	2	AnkleFlexExcur_NS

Descriptive Statistics

	Group	Mean	Std. Deviation	N
HipFlexExcur	Dance	53.46072639	13.85477030	20
	Athlete	51.92752002	21.15655437	20
	Total	52.69412321	17.66863208	40
HipFlexExcur_NS	Dance	54.63858981	15.30063861	20
	Athlete	52.90660992	18.19181909	20
	Total	53.77259986	16.61477732	40
KneeFlexExcur	Dance	71.85407222	8.861662368	20
	Athlete	67.02994289	12.41925012	20
	Total	69.44200755	10.92549603	40
KneeFlexExcur_NS	Dance	73.33504758	15.96707031	20
	Athlete	67.81548768	9.135350041	20
	Total	70.57526763	13.14055433	40
AnkleFlexExcur	Dance	65.52235504	5.976904782	20
	Athlete	60.50281957	6.142073383	20
	Total	63.01258731	6.499463552	40
AnkleFlexExcur_NS	Dance	60.39210535	10.59582473	20
	Athlete	50.60345693	7.949523430	20
	Total	55.49778114	10.49057995	40

Multivariate Tests[a]

Effect			Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power[c]
Between Subjects	Intercept	Pillai's Trace	.992	1525.539[b]	3.000	36.000	.000	.992	4576.616	1.000
		Wilks' Lambda	.008	1525.539[b]	3.000	36.000	.000	.992	4576.616	1.000
		Hotelling's Trace	127.128	1525.539[b]	3.000	36.000	.000	.992	4576.616	1.000
		Roy's Largest Root	127.128	1525.539[b]	3.000	36.000	.000	.992	4576.616	1.000
	Group	Pillai's Trace	.276	4.576[b]	3.000	36.000	.008	.276	13.727	.850
		Wilks' Lambda	.724	4.576[b]	3.000	36.000	.008	.276	13.727	.850
		Hotelling's Trace	.381	4.576[b]	3.000	36.000	.008	.276	13.727	.850
		Roy's Largest Root	.381	4.576[b]	3.000	36.000	.008	.276	13.727	.850
Within Subjects	footwear	Pillai's Trace	.509	12.457[b]	3.000	36.000	.000	.509	37.371	.999
		Wilks' Lambda	.491	12.457[b]	3.000	36.000	.000	.509	37.371	.999
		Hotelling's Trace	1.038	12.457[b]	3.000	36.000	.000	.509	37.371	.999
		Roy's Largest Root	1.038	12.457[b]	3.000	36.000	.000	.509	37.371	.999
	footwear * Group	Pillai's Trace	.089	1.172[b]	3.000	36.000	.334	.089	3.515	.288
		Wilks' Lambda	.911	1.172[b]	3.000	36.000	.334	.089	3.515	.288
		Hotelling's Trace	.098	1.172[b]	3.000	36.000	.334	.089	3.515	.288
		Roy's Largest Root	.098	1.172[b]	3.000	36.000	.334	.089	3.515	.288

a. Design: Intercept + Group
Within Subjects Design: footwear

b. Exact statistic

c. Computed using alpha = .05

Tests of Between-Subjects Effects

Transformed Variable: Average

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power[a]
Intercept	Hip	226703.262	1	226703.262	444.766	.000	.921	444.766	1.000
	Knee	392096.747	1	392096.747	1888.859	.000	.980	1888.859	1.000
	Ankle	280894.149	1	280894.149	3106.058	.000	.988	3106.058	1.000
Group	Hip	53.307	1	53.307	.105	.748	.003	.105	.061
	Knee	534.960	1	534.960	2.577	.117	.064	2.577	.347
	Ankle	1096.412	1	1096.412	12.124	.001	.242	12.124	.924
Error	Hip	19369.136	38	509.714					
	Knee	7888.187	38	207.584					
	Ankle	3436.503	38	90.434					

a. Computed using alpha = .05

Appendix E.5. 2 (group) x 2 (footwear) Repeated Measures Multivariate ANOVA to assess hip, and knee frontal plane motion at initial ground contact.

Within-Subjects Factors

Measure	footwear	Dependent Variable
Hip	1	HipAbdContact
	2	HipAbdContact_NS
Knee	1	KneeAbdContact
	2	KneeAbdContact_NS

Descriptive Statistics

	Group	Mean	Std. Deviation	N
HipAbdContact	Dance	-8.75348355	3.562117183	20
	Athlete	-10.7557970	3.089017204	20
	Total	-9.75464025	3.443596830	40
HipAbdContact_NS	Dance	-8.27682434	5.157725669	20
	Athlete	-10.66969991	4.354277370	20
	Total	-9.47326213	4.864665850	40
KneeAbdContact	Dance	1.446739740	4.072990610	20
	Athlete	2.797481751	5.487958584	20
	Total	2.122110745	4.818971420	40
KneeAbdContact_NS	Dance	1.84523276	4.602846135	20
	Athlete	2.50729156	6.304216355	20
	Total	2.17626216	5.458563279	40

Multivariate Tests[a]

Effect			Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power[c]
Between Subjects	Intercept	Pillai's Trace	.904	173.330[b]	2.000	37.000	.000	.904	346.659	1.000
		Wilks' Lambda	.096	173.330[b]	2.000	37.000	.000	.904	346.659	1.000
		Hotelling's Trace	9.369	173.330[b]	2.000	37.000	.000	.904	346.659	1.000
		Roy's Largest Root	9.369	173.330[b]	2.000	37.000	.000	.904	346.659	1.000
	Group	Pillai's Trace	.102	2.092[b]	2.000	37.000	.138	.102	4.184	.402
		Wilks' Lambda	.898	2.092[b]	2.000	37.000	.138	.102	4.184	.402
		Hotelling's Trace	.113	2.092[b]	2.000	37.000	.138	.102	4.184	.402
		Roy's Largest Root	.113	2.092[b]	2.000	37.000	.138	.102	4.184	.402
Within Subjects	footwear	Pillai's Trace	.005	.097[b]	2.000	37.000	.908	.005	.194	.064
		Wilks' Lambda	.995	.097[b]	2.000	37.000	.908	.005	.194	.064
		Hotelling's Trace	.005	.097[b]	2.000	37.000	.908	.005	.194	.064
		Roy's Largest Root	.005	.097[b]	2.000	37.000	.908	.005	.194	.064
	footwear * Group	Pillai's Trace	.010	.181[b]	2.000	37.000	.835	.010	.362	.076
		Wilks' Lambda	.990	.181[b]	2.000	37.000	.835	.010	.362	.076
		Hotelling's Trace	.010	.181[b]	2.000	37.000	.835	.010	.362	.076
		Roy's Largest Root	.010	.181[b]	2.000	37.000	.835	.010	.362	.076

a. Design: Intercept + Group
Within Subjects Design: footwear

b. Exact statistic

c. Computed using alpha = .05

Tests of Between-Subjects Effects

Transformed Variable: Average

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power[a]
Intercept	Hip	7394.245	1	7394.245	325.236	.000	.895	325.236	1.000
	Knee	369.520	1	369.520	9.583	.004	.201	9.583	.855
Group	Hip	96.588	1	96.588	4.248	.046	.101	4.248	.520
	Knee	20.257	1	20.257	.525	.473	.014	.525	.109
Error	Hip	863.931	38	22.735					
	Knee	1465.276	38	38.560					

a. Computed using alpha = .05

Appendix E.6. 2 (group) x 2 (footwear) Repeated Measures Multivariate ANOVA to assess hip, and knee frontal plane excursion.

Within-Subjects Factors

Measure	footwear	Dependent Variable
hip	1	hipadd_Excur
	2	hipadd_NS_Excur
knee	1	kneeval_Excur
	2	kneeval_NS_Excur

Descriptive Statistics

	Group	Mean	Std. Deviation	N
hipadd_Excur	Dance	-7.2212	2.37940	20
	Athlete	-8.2170	2.83568	20
	Total	-7.7191	2.63248	40
hipadd_NS_Excur	Dance	-7.3278	4.40680	20
	Athlete	-8.3588	3.30009	20
	Total	-7.8433	3.87804	40
kneeval_Excur	Dance	10.7429	6.79908	20
	Athlete	9.6164	6.16456	20
	Total	10.1797	6.43119	40
kneeval_NS_Excur	Dance	9.3269	4.90589	20
	Athlete	8.0716	4.14260	20
	Total	8.6993	4.52657	40

Multivariate Tests[a]

Effect			Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power[c]
Between Subjects	Intercept	Pillai's Trace	.894	156.186[b]	2.000	37.000	.000	.894	312.373	1.000
		Wilks' Lambda	.106	156.186[b]	2.000	37.000	.000	.894	312.373	1.000
		Hotelling's Trace	8.443	156.186[b]	2.000	37.000	.000	.894	312.373	1.000
		Roy's Largest Root	8.443	156.186[b]	2.000	37.000	.000	.894	312.373	1.000
	Group	Pillai's Trace	.077	1.551[b]	2.000	37.000	.225	.077	3.103	.308
		Wilks' Lambda	.923	1.551[b]	2.000	37.000	.225	.077	3.103	.308
		Hotelling's Trace	.084	1.551[b]	2.000	37.000	.225	.077	3.103	.308
		Roy's Largest Root	.084	1.551[b]	2.000	37.000	.225	.077	3.103	.308
Within Subjects	footwear	Pillai's Trace	.064	1.266[b]	2.000	37.000	.294	.064	2.532	.258
		Wilks' Lambda	.936	1.266[b]	2.000	37.000	.294	.064	2.532	.258
		Hotelling's Trace	.068	1.266[b]	2.000	37.000	.294	.064	2.532	.258
		Roy's Largest Root	.068	1.266[b]	2.000	37.000	.294	.064	2.532	.258
	footwear * Group	Pillai's Trace	.000	.003[b]	2.000	37.000	.997	.000	.006	.050
		Wilks' Lambda	1.000	.003[b]	2.000	37.000	.997	.000	.006	.050
		Hotelling's Trace	.000	.003[b]	2.000	37.000	.997	.000	.006	.050
		Roy's Largest Root	.000	.003[b]	2.000	37.000	.997	.000	.006	.050

a. Design: Intercept + Group
Within Subjects Design: footwear

b. Exact statistic

c. Computed using alpha = .05

Tests of Between-Subjects Effects

Transformed Variable: Average

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power[a]
Intercept	hip	4843.750	1	4843.750	284.946	.000	.882	284.946	1.000
	knee	7128.277	1	7128.277	160.177	.000	.808	160.177	1.000
Group	hip	20.541	1	20.541	1.208	.279	.031	1.208	.188
	knee	28.362	1	28.362	.637	.430	.016	.637	.122
Error	hip	645.956	38	16.999					
	knee	1691.091	38	44.502					

a. Computed using alpha = .05

Appendix E.7. 2 (group) x 2 (footwear) Repeated Measures Multivariate ANOVA to assess peak vGRF.

Within-Subjects Factors

Measure: vGRF

footwear	Dependent Variable
1	PkvGRF_Normalized
2	PkvGRF_Normalized_NS

Descriptive Statistics

	Group	Mean	Std. Deviation	N
PkvGRF_Normalized	Dance	2.0915	.51986	20
	Athlete	2.5645	.66105	20
	Total	2.3280	.63397	40
PkvGRF_Normalized_NS	Dance	2.0845	.45368	20
	Athlete	2.3579	.55991	20
	Total	2.2212	.52170	40

Multivariate Tests[a]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power[c]
footwear	Pillai's Trace	.074	3.026[b]	1.000	38.000	.090	.074	3.026	.396
	Wilks' Lambda	.926	3.026[b]	1.000	38.000	.090	.074	3.026	.396
	Hotelling's Trace	.080	3.026[b]	1.000	38.000	.090	.074	3.026	.396
	Roy's Largest Root	.080	3.026[b]	1.000	38.000	.090	.074	3.026	.396
footwear * Group	Pillai's Trace	.065	2.641[b]	1.000	38.000	.112	.065	2.641	.354
	Wilks' Lambda	.935	2.641[b]	1.000	38.000	.112	.065	2.641	.354
	Hotelling's Trace	.069	2.641[b]	1.000	38.000	.112	.065	2.641	.354
	Roy's Largest Root	.069	2.641[b]	1.000	38.000	.112	.065	2.641	.354

a. Design: Intercept + Group
Within Subjects Design: footwear

b. Exact statistic

c. Computed using alpha = .05

Tests of Between-Subjects Effects

Measure: vGRF ...

	Type III Sum					Partial Eta	Noncent.	Observed
Intercept	413.914	1	413.914	769.599	.000	.953	769.599	1.000
Group	2.785	1	2.785	5.179	.029	.120	5.179	.602

a. Computed using alpha = .05

Appendix E.8. 2 (group) x 2 (footwear) Repeated Measures Multivariate ANOVA to assess hip, knee, and ankle peak extensor moment.

Within-Subjects Factors

Measure	footwear	Dependent Variable
Hip	1	PkHipFlexMOM
	2	PkHipFlexMOM_NS
Knee	1	PkKneeFlexMOM
	2	PkKneeFlexMOM_NS
Ankle	1	PkAnkleFlexMOM
	2	PkAnkleFlexMOM_NS

Descriptive Statistics

	Group	Mean	Std. Deviation	N
PkHipFlexMOM	Dance	-.0861051564	.0266622270	20
	Athlete	-.0929541034	.0306284597	20
	Total	-.0895296299	.0285547795	40
PkHipFlexMOM_NS	Dance	-.08364020	.030410752	20
	Athlete	-.09427752	.036178638	20
	Total	-.08895886	.033424991	40
PkKneeFlexMOM	Dance	-.1159820481	.0208118836	20
	Athlete	-.1218556584	.0314598203	20
	Total	-.1189188533	.0264958797	40
PkKneeFlexMOM_NS	Dance	-.10203985	.023190698	20
	Athlete	-.10482417	.027900383	20
	Total	-.10343201	.025362036	40
PkAnkleFlexMOM	Dance	-.0653565512	.0166653672	20
	Athlete	-.0809588712	.0241545621	20
	Total	-.0731577112	.0219537353	40
PkAnkleFlexMOM_NS	Dance	-.06743749	.022985808	20
	Athlete	-.08069378	.023650364	20
	Total	-.07406564	.023978273	40

Multivariate Tests[a]

Effect			Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power[c]
Between Subjects	Intercept	Pillai's Trace	.975	476.728[b]	3.000	36.000	.000	.975	1430.183	1.000
		Wilks' Lambda	.025	476.728[b]	3.000	36.000	.000	.975	1430.183	1.000
		Hotelling's Trace	39.727	476.728[b]	3.000	36.000	.000	.975	1430.183	1.000
		Roy's Largest Root	39.727	476.728[b]	3.000	36.000	.000	.975	1430.183	1.000
	Group	Pillai's Trace	.130	1.796[b]	3.000	36.000	.165	.130	5.387	.427
		Wilks' Lambda	.870	1.796[b]	3.000	36.000	.165	.130	5.387	.427
		Hotelling's Trace	.150	1.796[b]	3.000	36.000	.165	.130	5.387	.427
		Roy's Largest Root	.150	1.796[b]	3.000	36.000	.165	.130	5.387	.427
Within Subjects	footwear	Pillai's Trace	.370	7.049[b]	3.000	36.000	.001	.370	21.146	.967
		Wilks' Lambda	.630	7.049[b]	3.000	36.000	.001	.370	21.146	.967
		Hotelling's Trace	.587	7.049[b]	3.000	36.000	.001	.370	21.146	.967
		Roy's Largest Root	.587	7.049[b]	3.000	36.000	.001	.370	21.146	.967
	footwear * Group	Pillai's Trace	.011	.138[b]	3.000	36.000	.937	.011	.414	.073
		Wilks' Lambda	.989	.138[b]	3.000	36.000	.937	.011	.414	.073
		Hotelling's Trace	.012	.138[b]	3.000	36.000	.937	.011	.414	.073
		Roy's Largest Root	.012	.138[b]	3.000	36.000	.937	.011	.414	.073

a. Design: Intercept + Group
Within Subjects Design: footwear

b. Exact statistic

c. Computed using alpha = .05

Tests of Between-Subjects Effects

Transformed Variable: Average

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power[a]
Intercept	Hip	.637	1	.637	470.565	.000	.925	470.565	1.000
	Knee	.989	1	.989	922.503	.000	.960	922.503	1.000
	Ankle	.433	1	.433	519.099	.000	.932	519.099	1.000
Group	Hip	.002	1	.002	1.129	.295	.029	1.129	.179
	Knee	.000	1	.000	.350	.558	.009	.350	.089
	Ankle	.004	1	.004	4.986	.032	.116	4.986	.586
Error	Hip	.051	38	.001					
	Knee	.041	38	.001					
	Ankle	.032	38	.001					

a. Computed using alpha = .05

Appendix E.9. 2 (group) x 2 (footwear) Repeated Measures Multivariate ANOVA to assess hip, knee, and ankle relative energy absorption.

Within-Subjects Factors

Measure	footwear	Dependent Variable
HipWork	1	RelativeHipWork
	2	RelativeHipWork_NS
KneeWork	1	RelativeKneeWork
	2	RelativeKneeWork_NS
AnkleWork	1	RelativeAnkleWork
	2	RelativeAnkleWork_NS

Descriptive Statistics

	Group	Mean	Std. Deviation	N
RelativeHipWork	Dance	26.2658	12.34970	20
	Athlete	25.6268	14.84491	20
	Total	25.9463	13.48211	40
RelativeHipWork_NS	Dance	24.0122	9.70986	20
	Athlete	30.6365	16.67249	20
	Total	27.3243	13.87824	40
RelativeKneeWork	Dance	28.1371	10.10503	20
	Athlete	26.8776	11.49589	20
	Total	27.5074	10.70219	40
RelativeKneeWork_NS	Dance	26.5802	15.28617	20
	Athlete	24.1765	10.19622	20
	Total	25.3783	12.88285	40
RelativeAnkleWork	Dance	45.5970	10.57019	20
	Athlete	47.4956	15.68241	20
	Total	46.5463	13.23526	40
RelativeAnkleWork_NS	Dance	49.4076	12.93556	20
	Athlete	45.1870	14.49090	20
	Total	47.2973	13.72544	40

Multivariate Tests[a]

Effect			Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power[c]
Between Subjects	Intercept	Pillai's Trace	.953	376.146[b]	2.000	37.000	.000	.953	752.291	1.000
		Wilks' Lambda	.047	376.146[b]	2.000	37.000	.000	.953	752.291	1.000
		Hotelling's Trace	20.332	376.146[b]	2.000	37.000	.000	.953	752.291	1.000
		Roy's Largest Root	20.332	376.146[b]	2.000	37.000	.000	.953	752.291	1.000
	Group	Pillai's Trace	.020	.368[b]	2.000	37.000	.694	.020	.737	.105
		Wilks' Lambda	.980	.368[b]	2.000	37.000	.694	.020	.737	.105
		Hotelling's Trace	.020	.368[b]	2.000	37.000	.694	.020	.737	.105
		Roy's Largest Root	.020	.368[b]	2.000	37.000	.694	.020	.737	.105
Within Subjects	footwear	Pillai's Trace	.034	.650[b]	2.000	37.000	.528	.034	1.300	.151
		Wilks' Lambda	.966	.650[b]	2.000	37.000	.528	.034	1.300	.151
		Hotelling's Trace	.035	.650[b]	2.000	37.000	.528	.034	1.300	.151
		Roy's Largest Root	.035	.650[b]	2.000	37.000	.528	.034	1.300	.151
	footwear * Group	Pillai's Trace	.073	1.453[b]	2.000	37.000	.247	.073	2.905	.291
		Wilks' Lambda	.927	1.453[b]	2.000	37.000	.247	.073	2.905	.291
		Hotelling's Trace	.079	1.453[b]	2.000	37.000	.247	.073	2.905	.291
		Roy's Largest Root	.079	1.453[b]	2.000	37.000	.247	.073	2.905	.291

a. Design: Intercept + Group
Within Subjects Design: footwear

b. Exact statistic

c. Computed using alpha = .05

Tests of Between-Subjects Effects

Transformed Variable: Average

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power[a]
Intercept	HipWork	56755.344	1	56755.344	218.705	.000	.852	218.705	1.000
	KneeWork	55937.962	1	55937.962	258.134	.000	.872	258.134	1.000
	AnkleWork	176132.406	1	176132.406	598.356	.000	.940	598.356	1.000
Group	HipWork	179.115	1	179.115	.690	.411	.018	.690	.128
	KneeWork	67.094	1	67.094	.310	.581	.008	.310	.084
	AnkleWork	26.960	1	26.960	.092	.764	.002	.092	.060
Error	HipWork	9861.242	38	259.506					
	KneeWork	8234.639	38	216.701					
	AnkleWork	11185.704	38	294.361					

a. Computed using alpha = .05

Appendix E.10. 2 (group) x 2 (footwear) x 6 (muscle) Repeated Measures ANOVA to assess pre-landing activation amplitude.

Within-Subjects Factors

Measure: MEASURE_1

footwear	Muscle	Dependent Variable
1	1	PreLGastroc
	2	PreMGastroc
	3	PreLHams
	4	PreMHams
	5	PreLQuads
	6	PreMQuads
2	1	PreLGastroc_NoShoes
	2	PreMGastroc_NoShoes
	3	PreLHams_NoShoes
	4	PreMHams_NoShoes
	5	PreLQuads_NoShoes
	6	PreMQuads_NoShoes

Descriptive Statistics

	Group	Mean	Std. Deviation	N
PreLGastroc	Dance	5.61580304	3.453254880	18
	Athlete	5.48755219	2.833579218	20
	Total	5.54830259	3.099409516	38
PreMGastroc	Dance	5.16954984	2.996003168	18
	Athlete	5.56616677	3.661059536	20
	Total	5.37829559	3.323733793	38
PreLHams	Dance	4.35564353	2.745243531	18
	Athlete	4.02321863	2.780814840	20
	Total	4.18068306	2.731650075	38
PreMHams	Dance	4.91515671	2.256882217	18
	Athlete	3.81066324	2.352876819	20
	Total	4.33384436	2.344235913	38
PreLQuads	Dance	12.17088357	5.921883071	18
	Athlete	10.33680831	5.302475349	20
	Total	11.20558080	5.604641945	38
PreMQuads	Dance	13.50899570	9.700588288	18
	Athlete	13.92199729	8.854301632	20
	Total	13.72636496	9.139924955	38
PreLGastroc_NoShoes	Dance	6.05341854	4.379470783	18
	Athlete	4.86848459	3.129379038	20
	Total	5.42976909	3.768378192	38
PreMGastroc_NoShoes	Dance	4.81915090	3.023666847	18
	Athlete	4.15953779	2.219583005	20
	Total	4.47198611	2.615700341	38
PreLHams_NoShoes	Dance	4.38492588	2.685410925	18
	Athlete	4.93078037	2.941160064	20
	Total	4.67221772	2.798528188	38
PreMHams_NoShoes	Dance	5.84416189	2.944086231	18
	Athlete	3.96657840	2.420936593	20
	Total	4.85596005	2.809758359	38
PreLQuads_NoShoes	Dance	12.95841970	5.749301803	18
	Athlete	11.49222919	5.794063988	20
	Total	12.18674049	5.742548876	38
PreMQuads_NoShoes	Dance	13.60369546	7.124941281	18
	Athlete	15.20578836	8.842251560	20
	Total	14.44690225	8.008172112	38

Multivariate Tests[a]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power[c]
footwear	Pillai's Trace	.019	.713[b]	1.000	36.000	.404	.019	.713	.130
	Wilks' Lambda	.981	.713[b]	1.000	36.000	.404	.019	.713	.130
	Hotelling's Trace	.020	.713[b]	1.000	36.000	.404	.019	.713	.130
	Roy's Largest Root	.020	.713[b]	1.000	36.000	.404	.019	.713	.130
footwear * Group	Pillai's Trace	.000	.012[b]	1.000	36.000	.912	.000	.012	.051
	Wilks' Lambda	1.000	.012[b]	1.000	36.000	.912	.000	.012	.051
	Hotelling's Trace	.000	.012[b]	1.000	36.000	.912	.000	.012	.051
	Roy's Largest Root	.000	.012[b]	1.000	36.000	.912	.000	.012	.051
Muscle	Pillai's Trace	.679	13.509[b]	5.000	32.000	.000	.679	67.545	1.000
	Wilks' Lambda	.321	13.509[b]	5.000	32.000	.000	.679	67.545	1.000
	Hotelling's Trace	2.111	13.509[b]	5.000	32.000	.000	.679	67.545	1.000
	Roy's Largest Root	2.111	13.509[b]	5.000	32.000	.000	.679	67.545	1.000
Muscle * Group	Pillai's Trace	.160	1.217[b]	5.000	32.000	.324	.160	6.087	.374
	Wilks' Lambda	.840	1.217[b]	5.000	32.000	.324	.160	6.087	.374
	Hotelling's Trace	.190	1.217[b]	5.000	32.000	.324	.160	6.087	.374
	Roy's Largest Root	.190	1.217[b]	5.000	32.000	.324	.160	6.087	.374
footwear * Muscle	Pillai's Trace	.363	3.649[b]	5.000	32.000	.010	.363	18.247	.876
	Wilks' Lambda	.637	3.649[b]	5.000	32.000	.010	.363	18.247	.876
	Hotelling's Trace	.570	3.649[b]	5.000	32.000	.010	.363	18.247	.876
	Roy's Largest Root	.570	3.649[b]	5.000	32.000	.010	.363	18.247	.876
footwear * Muscle * Group	Pillai's Trace	.168	1.290[b]	5.000	32.000	.293	.168	6.448	.395
	Wilks' Lambda	.832	1.290[b]	5.000	32.000	.293	.168	6.448	.395
	Hotelling's Trace	.201	1.290[b]	5.000	32.000	.293	.168	6.448	.395
	Roy's Largest Root	.201	1.290[b]	5.000	32.000	.293	.168	6.448	.395

a. Design: Intercept + Group
Within Subjects Design: footwear + Muscle + footwear * Muscle

b. Exact statistic

c. Computed using alpha = .05

Tests of Between-Subjects Effects

Measure: MEASURE_1 ...

	Tvpe III Sum					Partial Eta	Noncent.	Observed
Intercept	25912.443	1	25912.443	268.555	.000	.882	268.555	1.000
Group	25.024	1	25.024	.259	.614	.007	.259	.079

a. Computed using alpha = .05

Appendix E.11. 2 (dancer/athlete) x 6 (muscle) x 2 (rotation) ANOVA was used to assess differences in muscular onset time.

Within-Subjects Factors

Measure: MEASURE_1

Muscle	CableRelease	Dependent Variable
1	1	Lgastroc_IR
	2	Lgastroc_ER
2	1	Mgastroc_IR
	2	Mgastroc_ER
3	1	Lhams_IR
	2	Lhams_ER
4	1	Mhams_IR
	2	Mhams_ER
5	1	Lquads_IR
	2	Lquads_ER
6	1	Mquads_IR
	2	Mquads_ER

Descriptive Statistics

	Group	Mean	Std. Deviation	N
Lgastroc_IR	Dance	46.03	11.392	20
	Athlete	41.60	5.753	20
	Total	43.82	9.186	40
Lgastroc_ER	Dance	47.95	9.567	20
	Athlete	42.05	6.794	20
	Total	45.00	8.718	40
Mgastroc_IR	Dance	45.50	11.152	20
	Athlete	44.80	7.120	20
	Total	45.15	9.242	40
Mgastroc_ER	Dance	49.48	8.565	20
	Athlete	46.80	9.157	20
	Total	48.14	8.856	40
Lhams_IR	Dance	75.75	12.443	20
	Athlete	77.50	9.451	20
	Total	76.63	10.942	40
Lhams_ER	Dance	73.95	10.595	20
	Athlete	75.65	9.252	20
	Total	74.80	9.856	40
Mhams_IR	Dance	75.23	11.573	20
	Athlete	79.05	9.747	20
	Total	77.14	10.736	40
Mhams_ER	Dance	75.93	11.335	20
	Athlete	80.20	12.526	20
	Total	78.06	11.988	40
Lquads_IR	Dance	95.26	11.485	20
	Athlete	95.15	11.380	20
	Total	95.21	11.285	40
Lquads_ER	Dance	96.63	15.549	20
	Athlete	96.30	10.286	20
	Total	96.46	13.013	40
Mquads_IR	Dance	98.45	12.697	20
	Athlete	95.85	11.659	20
	Total	97.15	12.103	40
Mquads_ER	Dance	98.98	16.518	20
	Athlete	101.20	10.895	20
	Total	100.09	13.857	40

Multivariate Tests[a]

Effect		Value	F	Hypothesis df	Error df	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power[c]
Muscle	Pillai's Trace	.173	1.382[b]	5.000	33.000	.256	.173	6.910	.424
	Wilks' Lambda	.827	1.382[b]	5.000	33.000	.256	.173	6.910	.424
	Hotelling's Trace	.209	1.382[b]	5.000	33.000	.256	.173	6.910	.424
	Roy's Largest Root	.209	1.382[b]	5.000	33.000	.256	.173	6.910	.424
Muscle * Height	Pillai's Trace	.114	.849[b]	5.000	33.000	.525	.114	4.247	.265
	Wilks' Lambda	.886	.849[b]	5.000	33.000	.525	.114	4.247	.265
	Hotelling's Trace	.129	.849[b]	5.000	33.000	.525	.114	4.247	.265
	Roy's Largest Root	.129	.849[b]	5.000	33.000	.525	.114	4.247	.265
Muscle * Group	Pillai's Trace	.213	1.783[b]	5.000	33.000	.144	.213	8.915	.538
	Wilks' Lambda	.787	1.783[b]	5.000	33.000	.144	.213	8.915	.538
	Hotelling's Trace	.270	1.783[b]	5.000	33.000	.144	.213	8.915	.538
	Roy's Largest Root	.270	1.783[b]	5.000	33.000	.144	.213	8.915	.538
CableRelease	Pillai's Trace	.115	4.788[b]	1.000	37.000	.035	.115	4.788	.568
	Wilks' Lambda	.885	4.788[b]	1.000	37.000	.035	.115	4.788	.568
	Hotelling's Trace	.129	4.788[b]	1.000	37.000	.035	.115	4.788	.568
	Roy's Largest Root	.129	4.788[b]	1.000	37.000	.035	.115	4.788	.568
CableRelease * Height	Pillai's Trace	.110	4.570[b]	1.000	37.000	.039	.110	4.570	.549
	Wilks' Lambda	.890	4.570[b]	1.000	37.000	.039	.110	4.570	.549
	Hotelling's Trace	.124	4.570[b]	1.000	37.000	.039	.110	4.570	.549
	Roy's Largest Root	.124	4.570[b]	1.000	37.000	.039	.110	4.570	.549
CableRelease * Group	Pillai's Trace	.019	.700[b]	1.000	37.000	.408	.019	.700	.129
	Wilks' Lambda	.981	.700[b]	1.000	37.000	.408	.019	.700	.129
	Hotelling's Trace	.019	.700[b]	1.000	37.000	.408	.019	.700	.129
	Roy's Largest Root	.019	.700[b]	1.000	37.000	.408	.019	.700	.129
Muscle * CableRelease	Pillai's Trace	.089	.643[b]	5.000	33.000	.669	.089	3.213	.205
	Wilks' Lambda	.911	.643[b]	5.000	33.000	.669	.089	3.213	.205
	Hotelling's Trace	.097	.643[b]	5.000	33.000	.669	.089	3.213	.205
	Roy's Largest Root	.097	.643[b]	5.000	33.000	.669	.089	3.213	.205
Muscle * CableRelease * Height	Pillai's Trace	.091	.664[b]	5.000	33.000	.654	.091	3.318	.211
	Wilks' Lambda	.909	.664[b]	5.000	33.000	.654	.091	3.318	.211
	Hotelling's Trace	.101	.664[b]	5.000	33.000	.654	.091	3.318	.211
	Roy's Largest Root	.101	.664[b]	5.000	33.000	.654	.091	3.318	.211
Muscle * CableRelease * Group	Pillai's Trace	.023	.155[b]	5.000	33.000	.977	.023	.773	.081
	Wilks' Lambda	.977	.155[b]	5.000	33.000	.977	.023	.773	.081
	Hotelling's Trace	.023	.155[b]	5.000	33.000	.977	.023	.773	.081
	Roy's Largest Root	.023	.155[b]	5.000	33.000	.977	.023	.773	.081

a. Design: Intercept + Height + Group
Within Subjects Design: Muscle + CableRelease + Muscle * CableRelease

b. Exact statistic

c. Computed using alpha = .05

Tests of Between-Subjects Effects

Measure: MEASURE_1 ...

	Type III Sum					Partial Eta	Noncent.	Observed
Intercept	2639.321	1	2639.321	16.642	.000	.310	16.642	.978
Height	107.248	1	107.248	.676	.416	.018	.676	.126
Group	35.623	1	35.623	.225	.638	.006	.225	.075

a. Computed using alpha = .05